University of Strathclyde
Department of Bioengineering

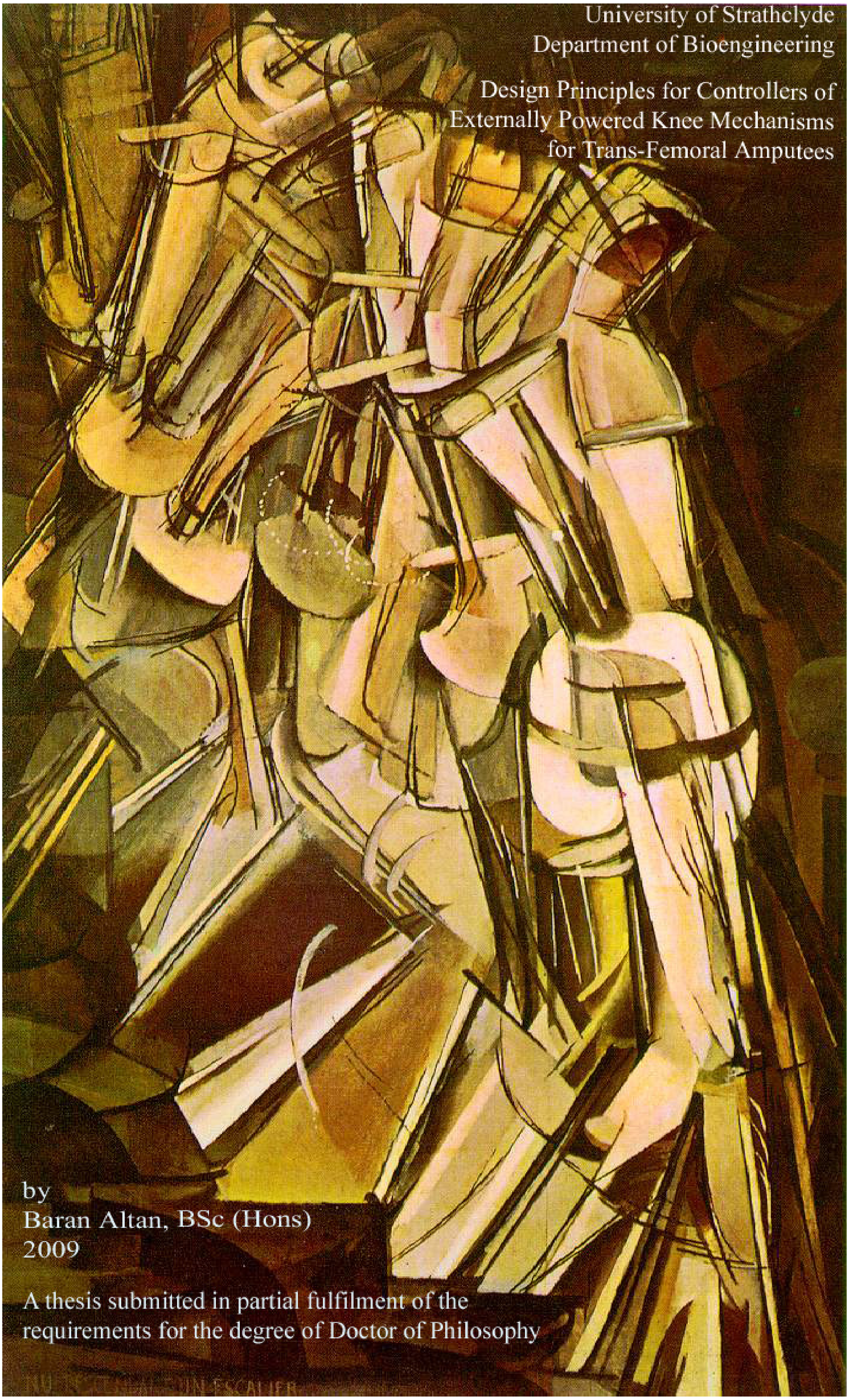
Design Principles for Controllers of Externally Powered Knee Mechanisms for Trans-Femoral Amputees

by

Baran Altan, BSc (Hons)

A thesis submitted in partial fulfilment
of the requirements for the
degree of Doctor of Philosophy

2009



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The image on the front cover is the painting by Marcel Duchamp, "*Nude Descending a Staircase*", drawn in the cubist style. It attracted much controversy in 1912 because - where the fourth dimension (time) was traditionally portrayed in this style as "*one moment, many views*" - he took the unprecedented step of painting "*one view, many moments*". His inspiration came from the breakthrough in cinematographic encapsulation of movement by Étienne-Jules Marey in 1884 [127, 106] and subsequently by Muybridge [144] in 1887. Duchamp wanted the viewer to interpret not only the figure, but also the motion. To this day, human motion and its cause remain enigmatic, and is the source of much discussion within this thesis.

Baran Altan

I ABSTRACT

Design principles were realised for artificial control of powered knee mechanisms to be used by trans-femoral amputees. Towards this, an extensive examination of pertinent issues was undertaken, ranging from natural control systems to artificial control systems, and including basics such as what constitutes a movement task. Considering all these ideas as a whole, the role of such a controller was specified and consequently a system proposed and developed to meet these needs. Fundamentally, it was to assist with "autonomic" movement tasks, which ordinarily are performed subconsciously rather than through "high level" conscious thought. In doing so, it had to be predictable and not require high levels of concentration. Practicalities also required the controller to be generic in nature for easy adaptation to future practical applications, and modular to allow control tasks to be extended.

To demonstrate, a simulation was undertaken to show how the task of walking might be assisted. A series of artificial neural networks was incorporated, each being individually trained to recognise their own specific *control state*. Data was gathered from trans-tibial amputees over their full walking speed range using the "VICON" camera data acquisition system. Subsequent analysis yielded a multitude of gait variables that could form a sensory array of inputs to the neural networks.

Critically and uniquely, the definition of control phases did not rely on external cues such as heel strike – instead they were determined according to the topology of the inter-segment power variable relating to the knee on the prosthetic side. After scrutinising all examples of this variable, it was possible to identify nine topological features common between the trans-tibial amputees; these features were thought potential candidates for control states.

Over 600 neural networks were trained and tested for each of the control states, in which different combinations of sensory inputs, network structures, and training algorithms were examined. It was possible to obtain networks with good results for most of the control states, while two of the states required more tuning or even a redefinition.

II ACKNOWLEDGEMENTS

During the course of this research I have been helped by many individuals. First and foremost, I would like to thank my family for their enduring patience, love and support they have shown me.

I would like to recognise the help given by many people within the department. In particular I would like to also give a special thanks to the following:

Bill Spence whom without his approachable manner, willingness to help, and many years of cutting edge experience as a prosthetist, this project would not have been possible. He has given much of his time facilitating experimental work, including contacting amputees and assisting with data acquisition.

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Finally I must give thanks to all the amputees whom contributed their time to this work.

III CONTENTS

1 INTRODUCTION

1.1 ORIGINS OF THE PROJECT

During his first year at the Bioengineering unit, the author attended the MSc course, which included classes in Prosthetics and Orthotics that helped to familiarize the author with trans-femoral prostheses. The principles for designing conventional trans-femoral prostheses were explained and it was evident that development had taken place mainly through improvements in socket design, using high-tech materials of lower mass and greater strength, and through shank designs with greater energy storing capabilities. However, it seemed to the author that improvements in design for knee mechanisms were only being made by enhancing certain passive mechanical properties. Conventional prostheses are considered to react *passively* because the mechanisms only respond to the external forces exerted upon them. The resulting patterns of movement thus are largely dependent on the mechanical and geometric properties of the knees, such as friction coefficients, fluid viscosity if hydraulic devices are used, and alignment configurations. When this project commenced, only one knee device was commercially available on the UK market that made use of an electronic controlling system to improve the function of the knee. The device was somewhat grandiosely labelled "Intelligent Prosthesis" in recognition of this unique feature, even though still largely reliant on conventional principles.

Despite some clever mechanical designs with conventional prostheses, these fulfil only a limited set of functions - the most important being prevention of collapse when supporting the body's weight, and enabling the shank to swing when non-weight bearing. However, because these prostheses rely on passive mechanical designs to achieve such functions, it was evident that the amputee was severely constrained. The natural joint is very complex, and subtly controlled by non-pathological persons to achieve a whole variety of movement patterns involving a wide range of knee angular velocities. In contrast, with conventional trans-femoral knee devices, swinging occurs much like a pendulum with a fixed damping resistance. This means that only a very narrow range of knee angular velocities is attainable by the amputee, the focus of the range being set by the prosthetist, e.g. by adjustments to valves in hydraulic devices, or to the frictional properties of mechanical devices. Under such restrictions, amputees are unable to produce the desired angular velocity profiles needed for most tasks they may wish to perform. Similarly, the methods that have been used to allow the prosthesis to transmit forces (i.e. support the body's mass) have mainly been directed at ensuring safety, which is often considered the main objective. With safety being of paramount importance, other aspects of prosthetic design have been of secondary consideration. Many devices are mechanically locked whenever the prosthesis is used for weight bearing, for example via a brake mechanism, or by aligning/positioning the knee centre so that a significant flexion moment would be required to cause flexion when the leg is straight. It could be argued that such methods impose severe limitations on the functional capability of the prosthesis so that it can hardly be expected to approximate the flexibility of the natural system.

Therefore, it was the author's desire to see how trans-femoral prostheses might be improved using *non-passive* methods whereby the prosthetic knee better mimics the natural systems.

1.2 DEFINING THE PROJECT

Overall, it seemed to the author that there was significant redundancy in the use of modern technology to enhance the capabilities of the prosthesis. In particular, the use of artificial knee controlling systems towards recreating the natural motion was lacking.

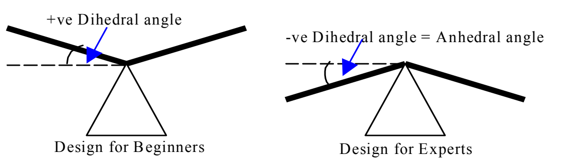

Figure 1-1; The dihedral angle will affect wing roll stability.

In many moving systems, greater performance can be achieved by approaching the limits of stability, although this is often accompanied by a trade off requiring a higher degree of control. To illustrate the principle, one needs look no further than the design of machines intended for beginners and experts. For example, the design of hang-gliders that are intended for beginners incorporate wings that take on a V-shaped profile as viewed from the back [Figure 1-1]. Technically each wing is said to have a positive dihedral angle[1]. This gives the hang-glider aerodynamically induced roll stability, which makes the glider naturally seek to level itself if the pilot releases the controls. As pilots improve, they tend to favour higher performance hang gliders in which the dihedral angle becomes smaller and smaller eventually forming into a negative angle (or anhedral angle). With this anhedral shape, the hang-glider lacks stability, allowing it to roll more easily, so that a roll input from the pilot is effectively amplified by the aerodynamics. This instability provides for much greater *manoeuvrability*, although at the expense of requiring a much greater *level of control* from the pilot for safe flying. This increased control requirement can only be mastered by experienced pilots and is the reason beginners learn with dihedral wings.

The same principles apply when choosing an amputee's prosthesis. The prosthetist will prescribe a particular prosthesis according to the level of control that the amputee can impose on the knee mechanism. Fit, well co-ordinated amputees, for example, are able to use their hip musculature to exert forces that can stabilise the knee mechanism when required. In such cases, the prosthetist can afford to prescribe a knee that does not lock during weight bearing, and the knee joint centre can be aligned to be much more unstable, i.e. positioned more anteriorly. This enhances the performance of the prosthesis where the amputee is able to flex the knee with greater ease when required. A weaker amputee, who is not able to provide adequate control, must rely on inherently stable prostheses. The drawback is that such a prosthesis will behave less and less like a natural knee, compromising performance aspects such as the smoothness of gait. Stabilised trans-femoral amputees are unable to flex their prosthesis whilst they bear weight so that vaulting effects can result.

Continuing with the analogy from the field of aviation, fly-by-wire systems employed in modern fighter planes might be of next consideration. Both high-performance and large aircraft originally used hydraulic-mechanical flight control systems, by which the pilot directly controlled the rudders and ailerons. These

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[1] The dihedral angle is the angle that each wing makes with the horizontal, when this angle is negative, it is known as anhedral.

conventional flight control systems restricted designers in the configuration and design of aircraft because of the need for flight stability since pilots were themselves unable to provide enough direct control to compensate for the instabilities that designers desired. With fly-by-wire systems, direct inputs via hydraulic-mechanical links were replaced by electric cables that activated actuators. Once fly-by-wire was introduced, it was possible to have some form of processing in addition to the actions of the pilot; this enabled the construction of much more responsive planes with higher performance as the artificial controller greatly enhanced the level of control that could be achieved with just the pilot alone. Designs were no longer so limited where rudders and ailerons could be controlled for much greater loads, and much more unstable aerodynamic designs could be implemented. A dramatic demonstration of such control systems manifests in the Nighthawk Stealth fighter plane (made infamous in the Gulf war). According to Hollway, "The Nighthawk's fly-by-wire system compensated for its instability so well that when one jet lost an entire rudder at high speed, the pilot had to be informed of the fact by the chase plane, after which he landed safely" [104]. The artificial control systems in aviation have replaced many of the actions that would have originally been performed by a pilot, allowing them to concentrate on other tasks such as navigation rather than the intricacies of the full aerodynamic necessities.

Returning now to the topic of prosthetic knee design; the author would argue that for prostheses to truly improve, designers must emulate the fighter plane approach where structural properties such as extreme instability can be controlled in a suitable fashion by using artificial control systems to aid the user. Such developments should benefit not only weaker amputees but also fitter amputees who currently use awkward movements to provide the increased control necessary for unstable knee structures. A control system can be developed hopefully to regulate all movements required by the knee including controlled flexion of unstable knees.

An artificial controlled prosthesis promises to assist the amputee by giving a greater degree of control over the knee unit, so that desirable movement patterns can be achieved with *less difficulty* than with conventional systems and consequently a more *natural movement style* promoted. A controlled system is fundamentally different from the passive system in that the system is able to produce its own forces via the actuators to influence knee movement patterns whereas the passive mechanism is only able to react to external forces. It has the potential to improve some of the shortcomings of conventional prostheses, such as locking whilst weight-bearing and providing a constant resistance when not bearing weight. For example, the original muscles that spanned the knee joint would have produced much more graded responses to control the knee pattern than conventional methods could hope to realise. Recovery of the natural performance seems much more attainable with artificial control of knee actuators.

At the beginning, the complexity of the project was not fully apparent. For example, it was not obvious what aspect of performance the controller should best be trying to improve. In the fighter plane example, the controller is likely to contribute to clearly defined objectives such as enabling the plane to turn as quickly as possible at high speeds without falling out of the sky, or keeping the plane level and steady while the pilot is engaged in another task. Such objectives can be arrived at through iteration of progressive designs. With knee motion however, the performance criteria were not so easily defined. For example, if one wants to replicate the actions of the natural knee, one must take into account the almost limitless combination of movement patterns that can be performed by utilising complex controlling mechanisms involving both the central nervous system (CNS) and the muscular mechanical properties. For the purposes of this project it was necessary to take a major decision regarding what aspect of movement the prosthetic knee should try to assist: What had started as a simple desire to improve the functional performance of trans-femoral prostheses required some serious thinking as to how to limit the scope of this project. The study of control strategies would be given higher priority than the design and construction of the mechanical system to implement the control commands. As such, it was felt that this research should be generic in nature, and the role of a “knee controller” considered systematically so that concepts and issues most pertinent to their design might be established. In this manner, the work might more readily be taken up by commercial organisations because the ideas can be moulded to fit their own agendas with greater ease.

The author envisaged the following basic objectives:

1. Discovering the best design *process* for an artificial control system of the prosthetic knee to aid movement so that it resembles the natural knee.
2. Designing a generic controller, assumed to be operating some non-specific actuator such as brakes, motors, etc., but able to exert the intended moments across the knee joint. No attempt, however, will be made to design the actuator system including its construction, location, and power source.
3. The envisaged design would be self-contained so that all components of the controller could be located within or in close proximity to the prosthesis; sensors placed on the sound leg would be avoided being a possible inconvenience to the amputee.

Such an approach would involve establishing knee movement patterns that best improve functional performance, i.e. determining what actions the controller should aid.

These aims are admittedly non-specific at this stage and drawn up to convey the overall direction of the project as seen at its inception. A major part of the research process was directed at establishing objectives that are more specific and as such form part of the design template. These specific objectives could only be ascertained after reviewing other research pertinent to this project.

Returning once again to the plane analogy to extract basic control issues, the question of informational flow between systems becomes an important one. There is an obvious interface between a fly-by-wire system and a pilot in the form of joysticks, pedals, and buttons in order to convey the pilot's intention to the mechanism. The options available to the amputee are far less obvious and the possibility of conflict between these options and the actual task being performed arises. When a pilot signifies intent e.g. via a joystick, the hand moves solely for this purpose, and is not involved in some other conflicting activity – the pilot is thus free to give as much attention as required to this task and after training can control the plane in a voluntary manner. The interface for the amputee, by inference from the above statement regarding containment of the controller, is likely to involve the amputee's stump. This situation fundamentally differs to the pilots because the stump not only has to convey the amputee's intention for the knee controller but also execute its role in the task (e.g. walking). It is possible that such separate roles will be incompatible or result in unusual motions that require more concentration. Unlike the fly-by-wire systems however where high levels of concentration were acceptable to produce the voluntary motions, this was undesirable for the amputee. This is because many basic motions undertaken by the lower limbs can be performed at the subconscious level with little thought. Such tasks, for example walking, that need little voluntary input were termed "autonomic" actions for the purposes of this project. Similarly therefore, the amputees contribution to controlling the knee should be as subconscious as possible and is likely to require some consideration of the interaction of the central nervous system with the artificial controller.

The author felt that the design process was likely to be almost the inverse of how fighter planes may be designed for example. For the amputee, the close-to-ideal performance is already obvious in natural walking. By scrutinising this performance, it should be possible in principle to design an artificial control system replicating this process (or performance). The goals of the controller are thus indicated by the natural system. This contrasts with fighter plane design where there is no ideal model to be copied and development depends on improving particular performance criteria. Such a process of iterative development may however be applied in the future once the fundamental principles that should govern a knee controller have been attained.

1.3 GETTING STARTED

As mentioned, establishing exactly *what* the controller should be trying to do was the first order of business. To this end, information was gathered regarding *biomechanics of movement*, and *natural control of movement*. It was hoped that some of the performance aspects regarding human movement might be gleaned from literature on biomechanical studies of human movement such as kinematic and kinetic movement patterns observed in non-pathological subjects.

The review on movement control literature was important to reach some understanding of how the natural system may integrate with the artificial controller.

Without this knowledge, it may not be possible to predict incompatibilities between these two controlling systems. It was expected that theories on movement control would be far from consensual, but even controversial theories that were not necessarily correct but approximated observed outcomes could still be potentially helpful to the project.

The ultimate objective was to develop the concepts and principles required of a generic knee controller. A demonstration of its application would be the most practical measure of its value. Controlling the knee for a *single task* was considered sufficient for a demonstration, as long as the potential to expand the controller's capabilities was implicit within its conception. The task of *walking* was an obvious choice, being autonomic, of sufficient complexity to challenge a controller, and is a fundamental task that an amputee may wish to accomplish. To be more specific, it was also hoped that the controller would be adaptable to the amputee's desired walking speed.

There now follows an extensive review of the pertinent literature regarding the walking gait and its quantification.

2 REVIEW 1: HUMAN MOVEMENT - GAIT PATTERNS

2.1 INTRODUCTION

This project has been undertaken on the belief that a significant improvement can be made for trans-femoral amputees if their knee mechanisms incorporate an active control mechanism that reacts appropriately to their functional needs. Walking has been selected as the demonstration task and therefore this review will examine the walking gait and will dwell less on other gaits such as running. The term gait in this sense may be considered a method of movement whereby the feet are alternately moved to different locations and a translation progression of the body achieved via rotational movements of the limbs.

By way of introduction to the complexities involved with a task such as walking, this review outlines the literature describing this act. It gives some insight into what may be required of a control system, and how difficult it was to develop a concise definition of the task for the controller.

2.1.1 GAIT STUDIES (HISTORY)

“It is logical that in order to understand the pathological effects of amputation, the functional significance of the intact leg in non-amputees must first be realised”. This statement by Eberhart [71] is worth repeating because the controller should seek to restore this function in the amputee. Therefore, focus was given to studies on gait with the intact knee.

The study of human movement has been underway for several centuries. As early as 1632, René Descartes proposed that both structure and function of living organisms could be described, in principle, by the laws of physics [58]. Inspired by this, Borelli in 1682 made a significant contribution to gait studies with his concept of muscle action [29]. He realised the importance of calculating the position of the centre of gravity of the body. He observed “… the forward displacement of the centre of gravity beyond the supporting area and the manner in which the forward swinging of the limbs saves the body from losing balance”; this is the foundation of the concept of propulsion and restraint in gait. Borelli's work was only continued much later on in the mid nineteenth century by the Weber brothers, who used observational analysis to begin an objective assessment of gait [215].

At the end of the 19th century, Marey et al [129] introduced a technique involving chronophotography to assess gait. This led to the introduction of photographic methods in 1887 by Muybridge [146], allowing gait to be measured much more precisely. However, Braune and Fischer [30] went on to develop this technique to the extent that they were able to obtain three-dimensional trajectories of individual segments of the body (e.g. shank, thigh, and torso). Although employing a laborious technique they were able to calculate the angles, angular velocities, and angular accelerations between adjoining segments along any chosen axis. In addition to gathering kinematic parameters, they were also able to measure ground reaction forces, which subsequently allowed them to calculate kinetic parameters such as moments across the joints. Consequently, their work is considered a classic in its field for the extent of its measurements. After this great advancement in gait studies, the focus was now on how the individual muscles behaved to produce the resulting patterns described by Braune and Fischer. This was done initially by using palpation techniques to define the sequence of muscle activation throughout the gait cycle. The method was much refined by the introduction of electromyograph (EMG) studies, in which the electrical activity produced by muscles during contraction was picked up by electrodes. With all this data available, researchers have tried to come up with key functional requirements for successful walking, and have tested their theories against observational results.

2.2 ASYMPTOMATIC WALKING GAIT PATTERNS

2.2.1 OBSERVATIONAL

2.2.1.1 TYPICAL GAIT

As early as 1881 von Meyer concluded, "There can be no such thing as a typical gait" [212]. Despite this Braune and Fisher observed that although people can of course exhibit many styles of gait, for example walking on tiptoes, walking with a small step, parade marching etc. people revert to a very similar style of gait if they are walking briskly or hiking long distances [30]. Other styles were found to be difficult to sustain. Among their observations on soldiers, "individual differences in gait which were obvious in slow walking in the streets of a garrison subsided during walking on country roads", and they were "… not able to distinguish any differences between the gait of a soldier and that of a hiker". For the purposes of this project, it was likewise assumed that "normal level walking" is the style which people revert to if they expect to hike some distance.

Although this description of gait is simplistic, it is insightful in that it implies some common strategy that is at work for most people, a thought that was carried throughout this work.

Given this generic view of gait, physical therapists have developed techniques allowing them to analyse pathologies of gait simply from observation. Some of the bigger exponents of clinical gait analysis by observation are Perry et al. [158] who devised a systematic method for gait analysis and interpretation [62]. Pathological gait could be analysed by using normal function as a reference, with "deviations from the normal pattern defining the functional error needing correction".

2.2.1.2 GAIT AS A CYCLICAL ACTIVITY

The act of walking is clearly a complex task, yet a task which is accomplished with little conscious thought and repeated cyclically. This observation has led researchers to divide walking into those observable features that repeat consistently. For example, if the leg under study is supporting the body's mass then it is in the stance period. On the other hand, when the leg is in the air it is in its swing period. Thus, the cycle for *each* leg can be divided into two distinct intervals. When both legs are considered simultaneously then it may be observed that there is an interval when the body's mass is supported by both legs, this period is known as "double support". Conversely, there is a period when only one leg weight bears while the other is in swing phase that is known as "single support". During the walking act, support is always present since there is no stage in which both feet are airborne, e.g. as in running. Other observable features within the gait cycle have also been categorised by researchers. The Rancho Los Amigos gait analysis committee [62] have proposed standardisations, which are sufficiently general to take into account peculiarities between individuals and allow most pathological gaits to be assessed. They reference for each limb the following functional phases of gait [Table 2-A]:

Stance Phase	*Initial contact* (IC): formerly Heel Strike[2] (HS). This is the point when one foot first makes contact with the ground and starts the double support period. It is often used as the starting point of the gait cycle.
	Loading response: during the double support period when body weight is transferred to the limb that has just made contact with the ground.
	Mid Stance: When the limb begins its single support period as the contra-lateral foot is lifted for swing phase.
	Terminal Stance: The ipsi-lateral heel begins to rise and single support ends as the contra-lateral limb reaches IC.
	Pre-Swing: This is the second double support interval and the ipsi-lateral foot begins lifting ready for swing phase.
Swing Phase	*Initial Swing:* This is the start of the swing phase and its end is defined when the foot swings in line with the contra-lateral foot.
	Mid-Swing: This describes the period when the swinging limb is advanced forward of the body and finishes when the tibia is vertical
	Terminal Swing: The hip stops flexing and the knee reaches full extension.

Table 2-A; Functional phases of gait according to Perry's team [62].

Other researchers also refer to slightly different events even within the above periods; Winter [222] for example, observes at the beginning of the loading response that the foot drops so that both the heel and the forefoot are in contact with the floor. He refers to this as "foot flat" (FF). Similarly, at the beginning of terminal stance, he refers to the heel rising as "heel off" (HO) and to the end of pre-swing as "toe off" (TO). "Push off" (PO) also occurs in late stance when the limb is used to accelerate the body up and forwards, mainly using the plantar flexors.

Throughout all such periods just described, an interaction of forces will occur resulting in the accelerations of the body's centre of mass. Therefore, when one

[2] In some pathologies heel strike does not occur.

makes observations of the walking cycle it is useful to consider the forces and their effects on limb segments and how they may influence the observable patterns.

An object may be considered stable if it can resist large perturbing forces that act to topple it over by forcing the centre of gravity outside the base of support. The lower limb comprises several segments that articulate at the ankle, knee, and hip joints where stability of the limb is maintained through both passive and active methods. Two forces allow for passive stability: ligament tension and that provided by the accelerating body mass (as indicated by the ground reaction force vector during stance). Active stability is generated by muscles, which in contrast to passive mechanisms can be regulated. Ligaments are positioned across the knee and hip joint to produce tensile forces as the knee or hip is hyper-extended. This stabilises the thigh relative to the shank, and the body relative to the thigh in the extension direction (i.e. prevents hyperextension). The stability via the ligaments is effective during certain alignments of the ground reaction force relative to the knee and hip joint. If the force vector passes anterior to the knee and posterior to the hip joint, the tendency will be to hyper-extend the joints and therefore gain stability from the action of the ligaments; this situation occurs while standing where the knee may effectively be locked in position. If the ground reaction force shifts behind the knee joint then a moment tending to flex the knee will gradually be induced. Since the surfaces between the tibial condyles and the femoral epicondyles form a small support base, and they are rounded, only a slight shift in the ground reaction force is required for instability. This shift is easy to achieve since the high centre of mass of the body tends to give a long lever arm for toppling forces to have effect. Therefore, the lower limbs may be considered inherently unstable which has the advantage of providing mobility but the demands of fine active control. The necessary control is transmitted by the appropriate activation of 57 muscles in the lower limb, producing forces across the joints to counter instabilities. This can be illustrated by considering the action of the ground reaction force at various phases of the walking gait cycle, and how muscular action is used in conjunction [Figure 2-1].

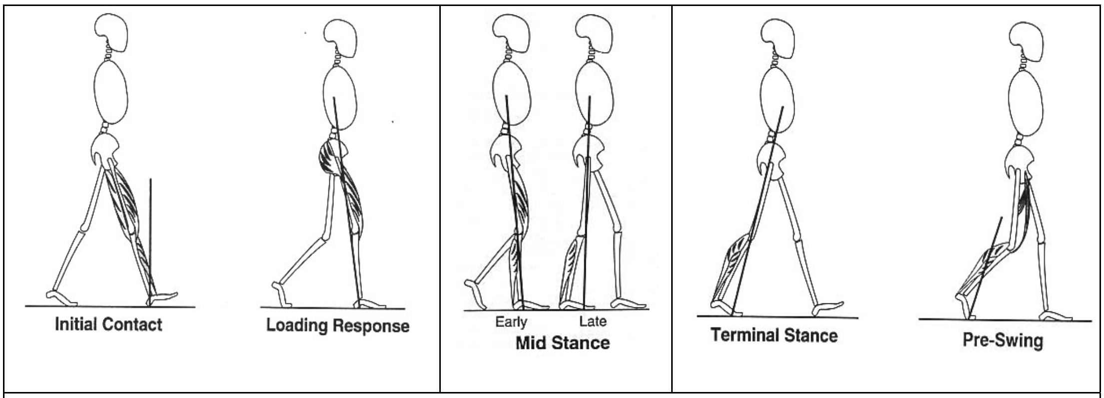


Figure 2-1: Adapted from Perry [158], Alignment of ground reaction force relative to joints throughout stance phase.

2.2.1.2.1 Initial Contact

At Initial Contact (IC) the hip is flexed, the knee is extended and the ankle in neutral position. As the ground reaction force vector lies anterior to both the hip and knee, and posterior to the ankle, a large flexion moment at the hip is created which requires hip extensors to be active to prevent collapse. There is an extending moment on the knee, which is countered by ligament forces and thus needs little muscle activity for stability (although some residual muscle activity may be present from the swing phase). The ground reaction force vector on the foot tends to produce a plantar flexion (foot extension) moment, which is controlled by the dorsi-flexor muscles acting across the ankle. The heel acts as a pivot, which because the ground reaction force passes posterior to the ankle, pushes the foot into plantar flexion. Dorsi-flexors are used to control this motion and prevent slapping as foot flat is reached.

2.2.1.2.2 Loading Response

As foot flat occurs, the ground reaction force vector now lies posterior to the knee, producing an external flexion moment that would tend to collapse the leg if muscles were not used to resist this force and control the flexion. Some flexion is allowed that is believed to aid shock absorption and help to smooth the trajectory of the body's centre of gravity, which would rise without any flexion. The activity of the hip extensors is reduced, since the force vector passes closer to the hip joint centre. The ankle can aid forward momentum of the body dorsiflexing and allowing the body to continue forward; plantar flexors may alternatively be used at this point if the person wishes to decelerate or is going down an incline.

2.2.1.2.3 Mid Stance

As the body progresses forward the knee extends and the force vector lies anterior to the knee allowing for passive stabilisation. Similarly, the hip gains some passive stability during this period. The total resultant forces are now supported by the limb under observation, since the contra-lateral limb has begun swing phase.

2.2.1.2.4 Terminal Stance

The body's centre of gravity is allowed to fall forward by the rolling motion over the ankle joint. The knee remains locked in extension and the hip reaches hyperextension. The foot reaches its maximum dorsiflexion as the rolling action continues, at which point the heel rises, and motion continues on the ball of the foot with the knee ending slightly anterior to the force vector. A flexing moment is thus generated across the knee.

2.2.1.2.5 Pre-Swing

The contra-lateral limb continues to support more and more of the load so that there is load transference between the limbs with the ipsi-lateral limb therefore needing less active force for stability. The force vector before transference is complete lies posterior to the ipsi-lateral knee, so that there is a flexing moment. The plantar flexor muscles also help prepare for swing by providing some push, which acts to swing the limb rather than accelerate the body as most of the load is now supported by the contra-lateral limb. The knee extensors may also become active if the knee flexion moment becomes too great and threatens to cause excessive heel rise. The flexors are also active in pushing the thigh forward.

2.2.1.2.6 Initial Swing

Rapid flexion of the hip encourages the limb to swing forward. The knee is still flexed following the actions of heel rise and will be controlled by the knee extensors.

2.2.1.2.7 Mid Swing

During this period, it is important that the foot clears the ground, usually by as little as 1cm; a movement facilitated by the action of ankle dorsiflexors and by hip flexion.

The knee passively extends because of energy imparted earlier in pre swing and initial swing.

2.2.1.2.8 Terminal Swing

The thigh is decelerated by the hip extensors, which allows the knee to continue extending. There is some knee flexor activity to prevent sudden hyperextension so that the knee is aligned for passive stability when Initial contact (IC) occurs.

2.2.2 BIOMECHANICAL ANALYSIS

The description of walking gait given above should allow readers to acquaint themselves with the basic cycle. However, for the purposes of this project, the biomechanics of these patterns is also required, which is the synthesis of biology and mechanics in order to understand and explain human movement. The biomechanical approach involves the gathering of quantitative data, which can then enable comparisons to be made between movement patterns on both an intra-subject or inter-subject basis.

When applying a biomechanical analysis to the study of human movement, one must be clear about exactly what information one is trying to extract from the data. The human musculoskeletal system comprises numerous bones that are linked by joints, which allow for up to 6 degrees of freedom (DOF) of movement of each bone. It is therefore possible to obtain a substantial amount of quantitative data relating to many aspects of the movement pattern dependent on many possible viewpoints or frames of reference. For example, bones move with respect to each other and a local reference frame can be defined for each bone thus allowing one relative description of the movement. Clinicians have their own convention for describing movement in terms of such local reference frames, for example, terms like anterior, flexion, abduction and proximal describe position or movement with respect to some anatomical reference. These conventions are often related to the body's reference frame, which consists of three orthogonal planes [Figure 2-2], namely the sagittal plane, the transverse plane, and the frontal (or coronal plane).

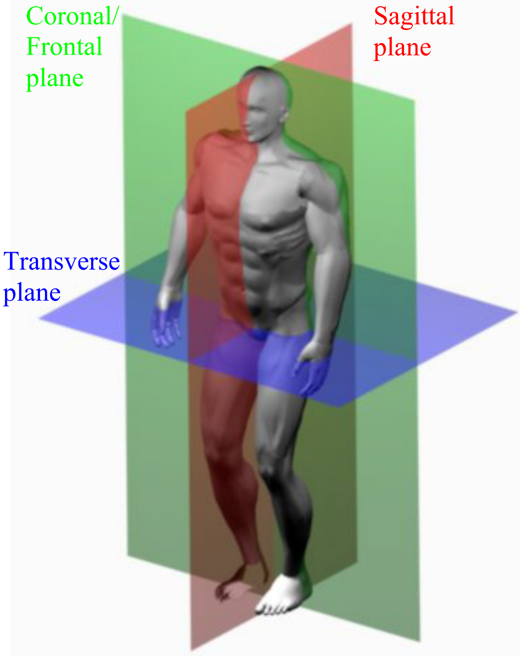


Figure 2-2; Clinical Spatial Co-ordinate system.

The direction of progression is perpendicular to the frontal plane; lateral movement is defined perpendicular to the sagittal plane while the vertical direction is perpendicular to the transverse plane. However, any description using a local frame does not describe the movement in space, for example, relative to the gravity vector or to the ground. For this, movements must be measured with respect to a global reference frame that remains fixed; this is normally the laboratory reference frame.

Movement can always be described with respect to some frame of reference, what aspect of movement is portrayed however depends on whether the outcome of the movement is to be expressed or the cause of the movement pattern. Outcome patterns can be described purely in spatial terms where parameters are derived according to the changes in spatial co-ordinates with time. These are termed ***kinematic*** parameters and can be given in either a local reference frame (for example joint angles, joint angular velocities, and angular accelerations) or a global frame (e.g. walking speed, stride length or the absolute velocity of limb segments translational and rotational). The analysis of the underlying forces that generate these kinematic patterns is known as ***kinetics*** and is associated with the cause of the movement. Kinetic parameters can include ground reaction forces, inter-segmental forces, moments, energy, and power. In addition, because muscle activity generates forces, EMG recordings are often considered like a kinetic parameter.

Most human movement patterns are exceedingly complex, and once one finally decided how the movement should be described then one has to take into account a whole variety of influences. These include subjective differences, external influences (e.g. floor surface, inclination), walking style and speed. A control system will need to make sense of all these differences or find some point of commonality between them.

It is suspected that people with different body types may generate different walking patterns. Amongst the various methods proposed to quantify bodily characteristics are:

- **Anthropometry:** This involves making measurements of the human body so that various ratios or proportions can be established. These ratios are then used to identify various shapes and sizes. For example, from measurements of segment lengths, widths, and diameters Braune and Fisher [30] were able to estimate the position of the centre of gravity of the body segments. Drillis and Contini [63, 48, 49] have also provided a substantial effort on quantifying such parameters, and Clauser [45, 97] additionally performed a large statistical study, based on air force crew.
- **Kinanthropometry:** This measures size, shape, proportion, composition, and other physical measurements of the human body as they relate to gross function such as growth, exercise, performance, and nutrition e.g. as suggested by Ross et al [172].
- **Somatotyping:** Originally introduced by Sheldon et al [183]; he observed after arranging 4000 standardised photos of men, that only three extreme body forms emerged from all the possibilities. He was then able to classify all subjects as having some proportion of each one of these traits. The extreme body types were:

    1. *ECTOMORPH (1,1,7)*: thin, flat chest, delicate build, young appearance, tall, lightly muscled, stoop-shouldered, has trouble gaining weight, muscle growth takes longer, i.e. the slender lean person, it is associated with how much the body is stretched out.
    2. *MESOMORPH (1,7,1)*: hard, muscular body, overly mature appearance, rectangular shaped, gains or loses weight easily, grows muscle quickly, i.e. muscle bound

3. *ENDOMORPH (7,1,1)*: soft body with excess adipose tissue, underdeveloped muscles, round shaped, over-developed digestive, system, trouble losing weight, i.e. is generally fat.

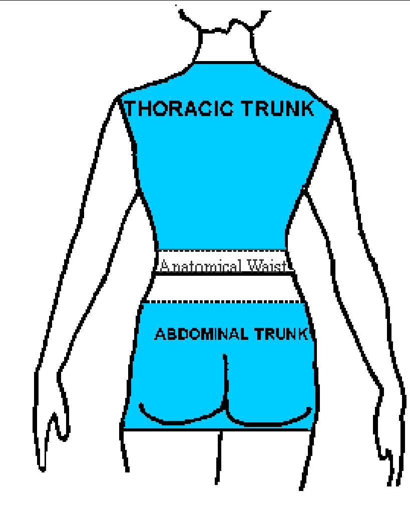


Figure 2-3; Areas used to measure the trunk index.

The numbers in brackets refer to the 7-point based scale indicating the proportion of each characteristic; the average person would be rated (4, 4, 4,) respectively (endo-, meso-, ecto-). Although one can roughly assess each type by eye, Sheldon also devised a measurement scheme that depended on height, weight history, and of most significance the trunk index (TI). TI was determined by dividing the cross sectional area of the Thoracic Trunk (upper area defined by the rib-cage) by the area of the Abdominal Trunk as measured from standardised somatotype photos. The lower the index - the greater is the degree of Endomorphy; the higher the index - the greater is the degree of Mesomorphy. The height of the person is the main factor in assessing Ectomorphy.

- **Ponderal Index (PI):** This is a measure of stature, or broadness as determined by a ratio of height and weight;

$$PI = 10^3 \times \frac{\sqrt[3]{M}}{H}$$

Where M = body mass (kg) and H = body height (m).

- **Body Mass Index (BMI):** This is another variation for stature although more emphasis is given to height in this case.

$$BMI = \frac{M}{H^2}$$

- **Relative Segment Proportions:** Although the overall height of a person is possibly more important in contributing to the likely performance of various tasks, the relative lengths of body segments making up the height will also have an effect. For example, someone with short legs but a long body may perform differently than someone with long legs and a short body even though they can have the same height. Another example includes the ratio of the length of the shank to that of the thigh, which is known as the ***crural index***.

Many studies, however, commonly use variables that appear intuitive such as subject height, weight, age, and gender. Possibly, because of all these different methods of reporting data, Hof has attempted to standardise many gait parameters by normalising them with respect to variables such as body height, or mass; this was intended to make inter-subject comparisons more meaningful [99]. The choice of normalisation was based on trying to form dimensionless parameters and minimising variability. For example, he proposes normalising walking speed by dividing $\sqrt{g * lleg}$ where g is the gravitational acceleration and lleg is the leg length, so that after normalisation a dimensionless quantity results. Similarly, force is divided by the product of body mass and g, and angular acceleration by g/lleg (see Table A-A in the Appendix for a full listing).

Pierrynowski and Galea [161] tried to show the efficacy of various normalisation studies on the gait parameters. They sought to develop "an overall scaling strategy to minimize the inter-subject variation of gait data outcomes, collected from ten thin and heavy-set, and short and tall individuals". The objective was to maximize the ability of statistical tests to differentiate between patients and comparative reference groups. Seven scaling strategies were compared against no scaling at all (strategy I); the strategies are listed in *Table A-B* and the results shown in Table A-C. The strategies were organised into the following three sets:

- The first set (strategy II) was based on ad hoc methods as widely utilised e.g. by Winter [222, 224, 225, 226], in which units of force, moments or power were divided by body weight or mass.
- The second set (strategy III) based on the dimensionless numbers as suggested by Hof [99].
- The third set (strategies IV_P where p=0, ½, 1, $^3/_2$ and 2), are what Pierrynowski and Galea termed the connected set of strategies. These are formulated on four assumptions based on physical similarities and how muscle works:

1. The first assumption is that different sized individuals are just “similar” versions of others. In this case, segments can be modelled as cylinders of length L, and diameter D. Because of similarity, it can be supposed that $D \propto L^P$. Therefore if p is assigned the five values shown above then the strategies proposed by Duncan [65], Günther [92], and McMahon [136, 137, 135] are included as well as an additional strategy with p= ½.
2. The second assumption is that all biomechanical measurements can be expressed in terms of the fundamental units of Mass (M), Length (L), and Time (T); this is similar to Hof’s methodology, but stipulates that the Ls should be broken up into their L and D constituents. Angles are mainly affected, which are thus scaled as LD^{-1} according to the ratio of the sides for a right angled triangle (note in the transverse plane this would be DD^{-1}). Similarly, moments should be converted because of their relationship with angular acceleration and inertia (the difference in the transverse plane was however ignored by Pierrynowski and Galea).
3. Force output is assumed to be proportional to the cross-sectional area of muscle ($\frac{\pi D^2}{4}$), which corresponds to the fundamental units of the kinetic variables (force, moment and power).
4. If the natural frequency of a system of joints, links and muscles is assumed to be proportional to a beam in its lowest mode of vibration, then the frequency is proportional to D/L^2. Therefore, time, which is the inverse of frequency, can be stated in terms of length.

After comparison of all these strategies with the non-scaled, Pierrynowski and Galea found strategies II, III and VI were the most effective at lowering the variability where they reduced the variability by up to 44% of the unscaled value [Table A-C].

This problem of attempting subjective comparisons was tackled slightly differently by Santambroglio [176], who sorted data into homogenous groups by subdividing parameters including gender, age, height, weight, and cadence into discrete intervals. The results of different subjects could then specifically be placed within this matrix. The problem was that he ended up with 2352 different possible categories of which he could only account for 44, indeed it would be a daunting task in terms of getting a sufficient number of subjects to fully represent these categories. His method does however exemplify the nature of this problem.

Before moving on, Pierrynowski and Galea showed that normalisation of the speed of progression by dividing by the square root of leg length produced the least variability of the inter-subject data. Whilst this was the goal, in that it allowed inter-subject comparisons of different speed values - could this allow true measures of the performance of the subject? Presumably, when the normalised velocities are compared, the slowest speeds should be comparable from one subject to another, as should the fastest speeds. It appeared to the author that although these methods seem to minimise the effects of mechanical advantage there remains a question as to the effects caused by the "skill" of the subject. A skilful walker should find it "easier" (however that may be measured) to walk at a particular normalised speed than a less skilful person should. Skill may therefore be a reflection of the ability of the CNS to utilise the musculoskeletal system (e.g. in terms of coordination). It is however difficult to incorporate such skill aspects into the normalisation of the speed and debatable whether it should be done.

Following this thought, Carollo [41] made investigations on gait performance according to measures of strength, balance, and coordination (according to neuromotor channel capacity). The measure of strength may be regarded as a mechanical factor whereas balance and coordination a skill factor.

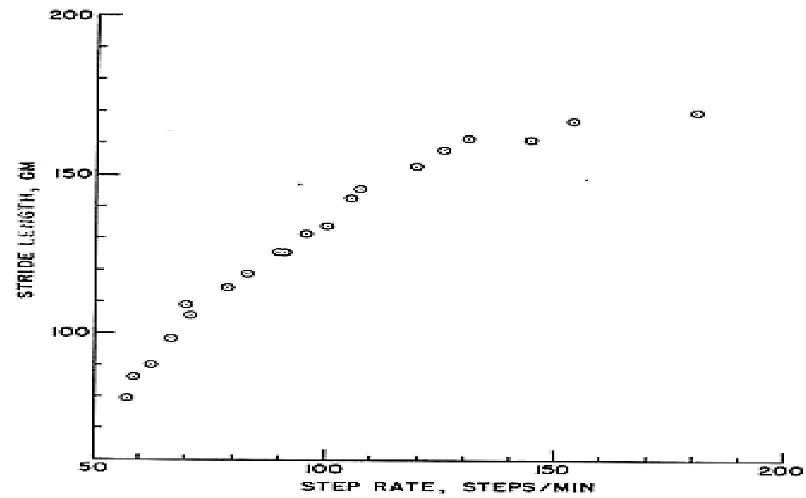
To continue, the rest of this review section outlines the results of many studies on gait; it should be noted that they do not conform to any particular standardisation described above making direct comparisons difficult.

2.2.2.1 KINEMATICS

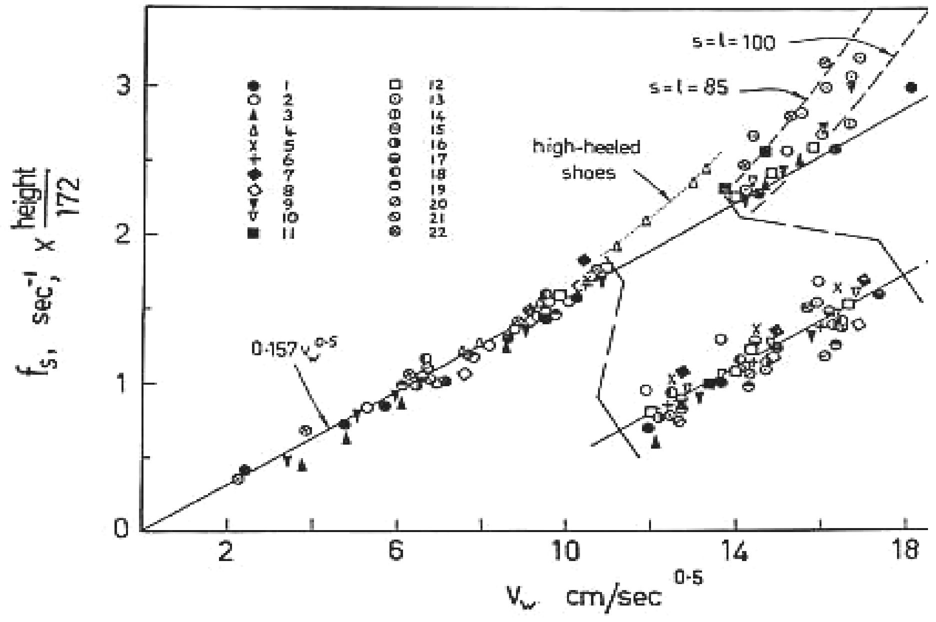
2.2.2.1.1 Spatiotemporal Measurements

The measurement of distance or time with regard to gait analysis can reveal a multitude of spatiotemporal parameters, reflecting different aspects of the movement pattern. For example, in the temporal domain, these include stride time or its reciprocal cadence (stride frequency), stance and swing time. Correspondingly, in the spatial domain, stride length may be measured and a parameter such as speed formed from a combination of these types.

"Walking speed" may be considered a fundamental functional measure of gait since the main purpose of walking is to move the body at some velocity. Walking speed can be determined from the product of stride length and cadence; however it was of interest whether individuals would opt to use particular combinations of these two variables to walk at various speeds. The Weber brothers [215] as early as 1836 were able to show that individuals did indeed seem to demonstrate some preferred combination of stride length and cadence to achieve different speeds while walking on level ground [Figure 2-4a]. One sees that the stride length is closely related to cadence, where there is an almost linear increase of stride length with cadence until eventually the stride length levels off, this increase corresponds to an increase in speed. This levelling off in stride length may be expected since a natural physical limitation is imposed of twice the leg length. Dean observed for normal walkers that the increase in cadence compared to stride length actually began once the stride length attained one leg length [56]. Lamoreux also comments, "the levelling off is accompanied by slipping on the floor" [125].



a) Figure from Weber [215], republished by Inman [110].



b) Figure from Dean [56].

Figure 2-4 a) Relationship of stride length to cadence or step rate. b) Gives a best fit of such data, relating speed with step frequency and includes data from many researchers.

In a similar vein, a limitation to the fastest attainable walking speed was also proposed by Alexander according to a simplistic model [2]. He assumes two segments; a rigid trunk, and a straight rigid leg. When the foot is in contact with the ground, the hip must move in a circular arc if the leg remains straight, the radius r being equal to the length of the leg. Therefore, if the body remains vertical, then the centre of mass located within the trunk, must also move in this arc of radius r. Any body moving in a circular motion must have its acceleration equal to v^2/r where v is the tangential velocity (or in this case the speed of the trunk). When the leg is vertical this acceleration is directed straight downwards and cannot be greater than that supplied by the gravitational force g, so that $\frac{v^2}{r} \leq g$. If an average leg length of 0.9m were assumed then the equality implies that the fastest possible walking speed is about $3ms^{-1}$, which is only a little faster than the speed people would normally start to run at.

To exactly quantify the stride length/cadence relationship however requires a study of both the intra-subject variability (e.g. from day to day) and the inter-subject variability.

Regarding intra-subject variability, several researchers have found the repeatability of various spatiotemporal parameters to be high. For example, Sekiya and Nagasaki demonstrated a high repeatability of the step-length/step rate ratio after a 3-month gap [181]. In this study, only the linear portion of the stride length - cadence relationship was used to enable comparisons. Stolze et al judged the retest reliability of stride length, step length, cadence, step width, foot angles, stance phase time, swing phase time, double support time, and cycle time on two different days to be high [195]. Their comparisons were however limited to "free" walking speed[3], which is itself subjective and thus contains variability.

Murray et al [143] tried to quantify inter-subject effects such as age and height on the spatiotemporal parameters when compared at a particular speed (free walking speed). The main effect of this work was to demonstrate that people chose to walk at slower free speeds with age, so that stride length also inevitably decreased. Maybe because of this in that like with like is not being compared, she was not able to "relate any of her measurements systematically with age". She did however report that only the mean step lengths and stride lengths related systematically with height (for 60 normal men).

[3] This is the speed that a person will naturally elect to walk at.

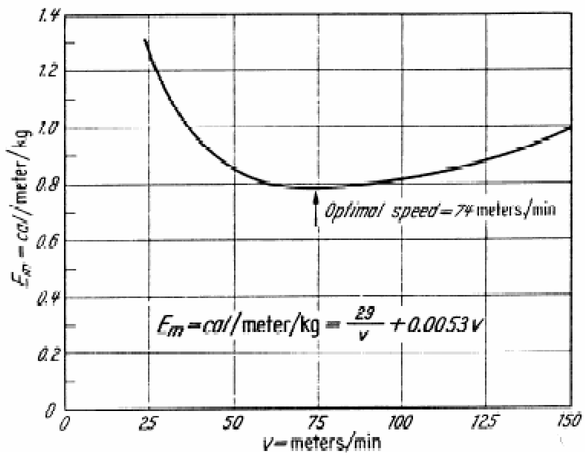


Figure 2-5 - Relationship between energy expenditure and speed for a particular subject, from Ralston [167]. The optimal speed or preferred speed corresponds with the minima.

Ralston tried to study the underlying reasons why people adopt a preferred speed of walking [167]. He measured the energy expenditure per metre per kilogram of subjects as they walked at different velocities, by monitoring the concentration of oxygen in respired air. He found that the energy expenditure (when un-normalised) was linearly related to the square of velocity [Figure 2-5]. The minimum of this relationship corresponded to the natural walking speed, implying that it is adopted because it is the most efficient speed for walking.

Other researchers have not limited their research to free speed walking and have tried a more detailed assessment of inter-subject variability. For example, Dean collated the results from several studies [56]. Although he was aware that cadence varied approximately as the square root of speed, he also tried to develop improved empirical formulae. He first reasoned that leg length is inversely proportional to the step frequency but concludes that using height is sufficient since it is proportional to leg length. Therefore, after collating both his results and that of several researchers he was able to use the data in Figure 2-4b to make a best fit of the data:

$$\frac{f_s \times H}{172} = 0.157v^{0.5}$$

where f_s = step frequency (s^{-1}) note, step frequency is half the cadence

H = body height (cm)

and v = walking velocity (cms^{-1})

Except for high speeds, there is good agreement between observations where, for speeds below 200 $cm.s^{-1}$ the standard deviation is $0.062s^{-1}$.

Grieve [87, 86] went further to try and quantify the effects of age on this ratio. Similar to Dean he also tried to fit the data obtained according to two models, linear and logarithmic which he referred to as type I and type II respectively

type I gives $f_s = \alpha_1.v' + \beta_1$

type II gives $f_s = \alpha_2.v'^{\beta_2}$

where v' = relative walking speed as measured with respect to height. (s^{-1})

α is the slope of the linear and log graphs

and β is the intercept of the linear and log graphs.

This relationship assumes first that the variability is reduced by normalisation with respect to height since the velocity is given as a relative velocity where the stride length is divided by height. For the subjects who had a type II as the best fit, Grieve found the average group equation to be

$$f = 64.8\ v'^{0.57}$$

where f, in this case, is the number of complete cycles per minute.

This equality compares favourably with Dean's, which when written in this form using a stature of 1.65m gives: $f = 63\ v'^{0.5}$

The difference lies within the deviation experienced between Grieve's subjects, although Grieve's subjects also walked in bare feet.

It is interesting to note that both researchers have an approximate square root of relative velocity relationship with step frequency, and since $v = f \times$ *stride-length* the stride length must also be proportional to the square root of the velocity.

Although the stride length is measured over the whole gait cycle, it should be remembered that it consists of both a stance and swing period. The contribution to stride length of these periods has been investigated by several researchers including Grieve, Murray, and Andriacchi [87, 144, 8].

As expected when speed increases the stance and swing periods decrease, however Murray observed that this decrease was not linear and best approximated by quadratic polynomials. She noted that stance duration decreased 3.5 times as rapidly as swing implying that they will make up differing amounts of the gait cycle depending on speed. This implication is demonstrated in Holden's study [102], where she analysed data gathered at five specific speeds from 18 subjects. These speeds were incremental at 25, 50, 75, 100, and 125% of a scaled natural speed (0.785 stature.s^{-1}) as suggested by Rosenrot et al [171]. After averaging this data of three trials per subject at each speed, he obtained the results shown in Table 2-B. The temporal contribution of stance diminishes from 73% to 59%, and swing contribution reciprocally increases from 27 to 41%.

	Speed				
(% of 0.785 stature.s^{-1})	25	50	75	100	125
Stride length (m)	0.75(0.09)	1.03(0.11)	1.21(0.10)	1.35(0.11)	1.51(0.1)
Stride time (s)	2.27(0.30)	1.55(0.12)	1.22(0.07)	1.02(0.06)	0.91
Speed (m.s^{-1})	0.33(0.02)	0.67(0.04)	0.99(0.05)	1.32(0.07)	1.56
Cadence (steps.min^{-1})	54(7)	79(7)	99(6)	118(7)	132(10)
Stance/Swing ratio	73/27	66/34	63/37	60/40	59/41

Table 2-B - Adapted from Holden [102]. Mean temporal and distance measures for 18 subjects.

Paradoxically, even though the percentage of stance time becomes less as speed increases, Grieve [87] showed that it actually contributes a greater distance to the relative stride length (i.e. normalised with respect to stature). These results are plotted in Figure 2-6. To accommodate increasing stride length (and therefore usually increasing speed), the contribution to stride length by support was seen to increase until both swing and stance were roughly equal. For the range of data that Grieve gathered, the stance phase contribution to the stride length increased from about 30% to about 50%, corresponding to the increase in relative stride length.

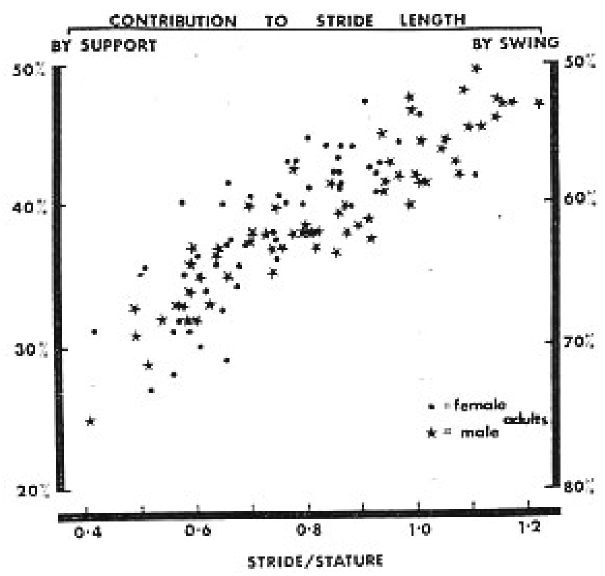


*Figure 2-6 – Gait chart showing swing and stance contribution vs relative stride length (**Figure from Grieve [87].***

As mentioned at the beginning of the chapter, temporal measures can be indicative of the overall movement. Schmidt has even argued that it is possible to establish the stability and skill involved with a movement by analysing the repeatability of the patterns [179]. Such conjectures using variability have led to the hypothesis that faster movements are more stable than slower ones. This has been demonstrated using spatiotemporal parameters. Newell et al showed that the variability in movement time decreases with speed [150]. Maruyama et al [131] went on to specifically quantify the variability in the phase durations of the walking gait cycle, and used it to support the speed/stability theory. He found that not only did the variance decrease with speed, but also that particular styles of gait produced less variability. They examined how varying the proportion of stride length and cadence for a particular speed would affect the variability of the spatiotemporal parameters. They noted that the minima in variability for each speed occurred when subjects walked at the same step rate that they would subconsciously assume for a particular speed. This is possibly a reflection of why a stride length/cadence relationship can be observed with speed for many subjects.

In summary, measurements of fundamental spatiotemporal parameters such as cadence, step length, swing time, and walking speed can serve to indicate the particular state (i.e. performance level) of the movement pattern and even with what skill the subject is able to accomplish it. Deviations from a normal non-pathological pattern can readily be observed by comparison with what is considered normal levels. For example, Grieve even considers the fact that if a subject is not able to demonstrate the normal range in speed, or if they have an unusual combination of cadence and stride length for their speed range, then it is likely to be indicative of some pathology.

Therefore, it is of interest to see if this predictability of walking is also evident on a more localised level.

2.2.2.1.2 Displacement of body segments

One set of parameters that have been used to give an overall exposition of the function of walking concerns the kinematic motion of the pelvis, as this is where the legs join the rest of the body. Since the motion of the pelvis will reflect the overall action of the legs, this motion can be a sensitive indicator of asymmetries between the two legs; the patterns would be expected to repeat twice during the cycle. This can be illustrated by looking at the forward displacement of the pelvis relative to the average displacement of the body. For example, Inman and Lamoreux [125] have obtained such figures for a range of speeds (see Figure 2-7). Interestingly, as speed increases it can be seen that the size of the progressional deviation **decreases**, i.e. the pelvis progression becomes smoother and approaches that of a constant average velocity. At the slowest speed, there is a maximum deviation of about 2cm either side of the constant velocity line (marked as the zero point) whereas at the faster speed it is only about 1cm.

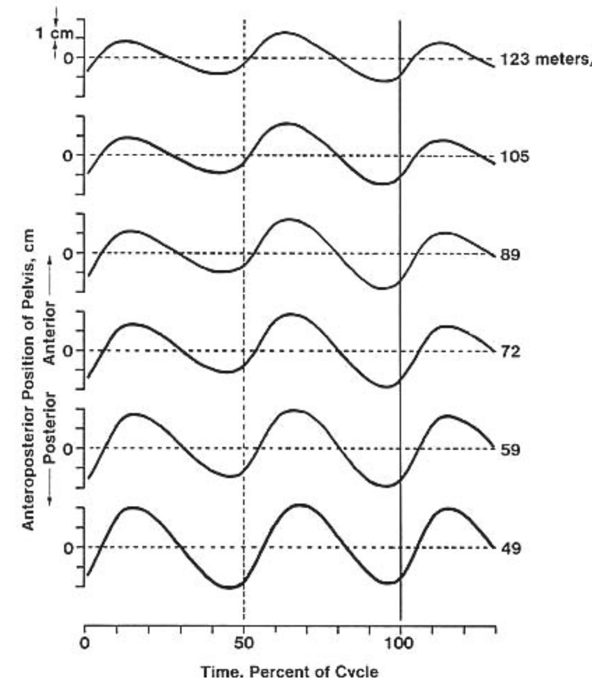
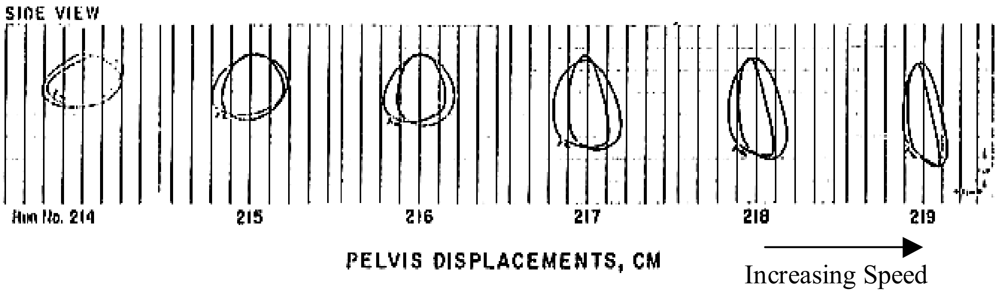
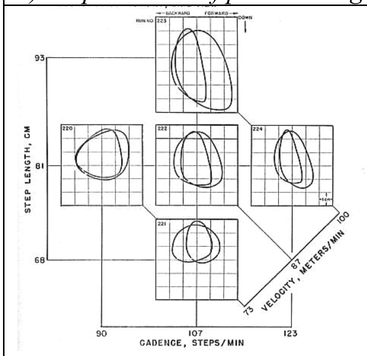


Figure 2-7 –Taken from Inman [110], Displacement of pelvis with speed along anterior-posterior direction

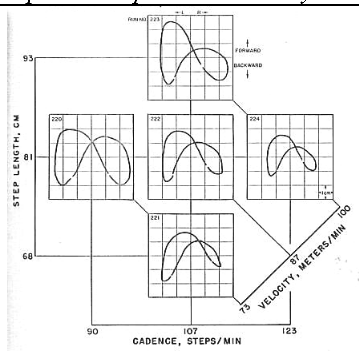
Similarly, displacements of the body's centre of gravity tend to follow the displacement of the pelvis. The amplitudes become less as displacements are taken closer towards the head. The pelvis reaches its maximum forward displacement just after toe off and the minimum just before IC. Therefore the body tends to exceed the average velocity during the double support period when the body is tilting forward. It is slower during the remaining period. The vertical displacement of the pelvis also shows similar speed dependent oscillations. As the speed is increased, the vertical position of the pelvis seems to get lower but the size of the total displacement larger. The highest point occurs at mid stance during single support; the lowest point is during double support. Therefore, in the sagittal plane the overall pelvis motion is to rotate in a circular motion relative to an average displacement reference-frame [Figure 2-7]. The actual shape tends to depend on speed where the ellipsoid becomes thinner and more elongated in the vertical direction as speed increases.



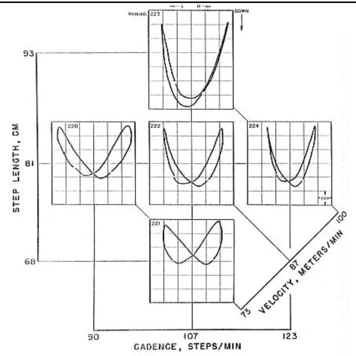
a) Displacement of pelvis in Sagittal plane as speed is naturally increased.



b) Sagittal plane



c) Transverse plane



d) Coronal plane

Figure 2-8: Pelvis Displacement from Lamoreux [125]; Figure a) shows the variation as speed is increased and stride length and cadence is freely chosen. Figures b), c) and d) show the effects of only increasing cadence and only increasing stride length

Asymmetries between both legs are also made highly visible by these figures since the patterns caused by both legs are imposed upon each other. The morphologies of these patterns could be observed more extensively because Lamoreux systematically varied cadence and step length, rather than just speed. In this artificial situation, the cadence was first kept constant whilst stride length was increased, and then the stride length was kept constant whilst the cadence was increased [Figure 2-8 b, c, and d]. In the sagittal plane, the main effect of only increasing cadence appeared to be to reduce the width of the ellipse (i.e. progressional displacement) whereas the effect of only increasing stride length was to increase the height of the ellipse (i.e. the vertical amplitude). In the transverse plane, as cadence increases medial-lateral excursions seem to decrease as well as the progressional. The effect of increasing only stride length however was to mostly increase the progressional displacement.

2.2.2.1.3 Relative rotations between segments (Joint angles) and variability

As mentioned earlier the patterns produced by the kinematic data depend on the relative viewpoint, so care must be taken when making comparisons between various researchers' work. For example, one could measure the rotation of a particular segment, such as the thigh, relative to the global frame or a local frame such as the shank (this would allow the knee joint angle to be measured). Due to the more clinical relevance of joint angles, more emphasis is given to them in this section, although with a choice of 80 kinematic variables for a 7-segment model [222], it is somewhat arbitrary.

2.2.2.1.3.1 Intra and Inter subject variability of Joint Angulation patterns

It has already been established that subjects have a very repeatable stride length/cadence relationship; therefore, one might expect a similar predictability with the joint angulation patterns for the individual. If, alternatively, the stride-length/cadence consistency is achieved by adaptive compensation at each joint, then the joint angulation patterns are likely to have a large variance associated with them. To ascertain which condition may be more correct, a review of some of the relevant data collected by other researchers is given. Once again, care must be taken in comparing results and attaching meaning because of the variety in techniques and the amount of statistical information given, nevertheless there is a wealth of useful information.

One of the first questions to ask is whether there is some predictable desired joint trajectory, which a control system should be trying to achieve. Extensive statistical analysis of the hip joint motion by Johnston and Smidt [114] established how well subjects reproduced their hip-joint motion, while walking at their free speeds, on two different occasions. Hip joint-motion was measured by determining the average maximum excursion of the hip. They found that for all the subjects there was a difference in the test and retest results of 1.9 degrees for the sagittal hip rotations, (the range of difference was only 0 - 4°). This difference was insignificant according to the t-test at the 0.01 level, and signifies there is little intra-subject variability of the hip-joint motion. Similarly, Johnston and Smidt also looked at the inter-subject variability of the whole group, which he termed objectivity of the data. It was shown that the mean difference in hip angles for the whole group was only 3.8° ranging from 0 - 10°.

More recently, Dujardin et al [64] also obtained such data on 55 normal subjects (25 male and 30 Female); an attempt was made to classify the range of motion during free speed walking. They were able to show that the variation in the hip joint flexion-extension range between all the subjects was within ±40% of the mean and had a normal distribution. This variation might be anticipated since the values for the free walking speed also varied within ±40% of the mean. They also found by multiple regressions that the "free" walking speed was related to height, cadence and step length (the latter two hardly surprising since walking speed is defined by a multiple of cadence and step length). The free walking speed was however not related to rotational components of the hip joint motion which is possibly more an indication of the previous result which quoted very little variation in the hip angles at free speed. No regression was however attempted with weight.

As previously discussed, Holden [102] was able to gather data at five different speeds [Table 2-B]. He used this data to calculate the variations in the knee angle patterns at these different speeds. He found, for each speed group, that the inter-subject standard deviation for the knee joint angle remained much the same at around 3.6 ranging from 3.3 to 3.8. This resulted in a small coefficient of variation and promoting the notion that joint angles are controlled in a specific manner where the movement pattern is accurately reproduced time after time in order to achieve the task of walking.

So far, a broad view of walking has been taken, where it has been observed that walking patterns are very similar at comparable speeds, and even between different subjects. However, more explanation is needed of the actual patterns and how they vary (in particular with speed).

Lamoreux, Winter, Inman and Holden include those researchers that have obtained such kinematic data over a range of speeds. Their findings are in much agreement, where recognition of a particular kinematic pattern for walking is possible e.g. the hip angles. This implies that they have common features between them, which can be identified. In an attempt to standardise the description of these features, Benedetti in 1998 [16] introduced what he described as gait parameters associated with each joint; for example, he identified 12 hip angle parameters labelled H1-H12 [Figure B-1, Appendix B]. Some of these could be linked with the clinical observations described earlier, for example H1=flexion at heel strike, H2= maximum flexion at loading response etc. These features can now be examined in more detail for the effects of walking speed upon them.

Before this, as an addendum, it should be noted that when examining gait patterns in more detail, many researchers normalise the time axis with respect to the whole gait cycle. This is usually done with respect to the heel strike where a cycle of data is presented from heel strike to heel strike referenced as 0 – 100% of the gait cycle. It is then often the case that ensemble averages are taken of this normalised data as such a procedure means temporal diffusion of the information is likely. For example, some of the gait parameters suggested by Benedetti (see Appendix B), will appear to occur at one specific phase on the % gait cycle scale; any variation along the temporal axis is often lost because of the averaging.

In effect, the same periodicity between the gait parameters does not have to hold for every walking trial and it may be possible for them to vary according to different conditions such as speed of walking; hence, a linear relationship of the timing of gait parameters should not automatically be assumed. In fact, considering the change in the stance/swing phase contribution, a non-linear relationship may be envisaged. Therefore care must be taken when assessing data. For example, the correlation of patterns on an ensemble average basis would possibly show a greater deviation simply because like with like is not being compared. This concept has dominated much of the thinking towards devising a control system because it applies to all gait profiles. Some of the author's own caution regarding this matter was justified by an article published in 2000 by Sadeghi et al [174]. In this study, they try to reduce such temporal variability by applying a technique known as Curve registration to warp the temporal scale between corresponding gait profiles, in effect similar peaks are aligned. Using this technique enabled them to "reduce inter-subject variability without perturbing the curve characteristics".

Nevertheless, it is almost standard practice for researchers to present their results in the normalised time format, and as such does have the advantage of enabling the data to be readily viewed.

2.2.2.1.3.2 Effect of speed on joint patterns

2.2.2.1.3.2.a Hip Kinematics

Many researchers have obtained very similar patterns for the hip angle as subjects walk; Figure 2-9 obtained from Lamoreux [125], shows one such example in which data is shown for a subject walking at six different speeds. Starting at initial contact, the hip is in flexion where it extends during stance phase and then starts to flex again throughout the swing phase until a maximum is reached just before the next heel strike.

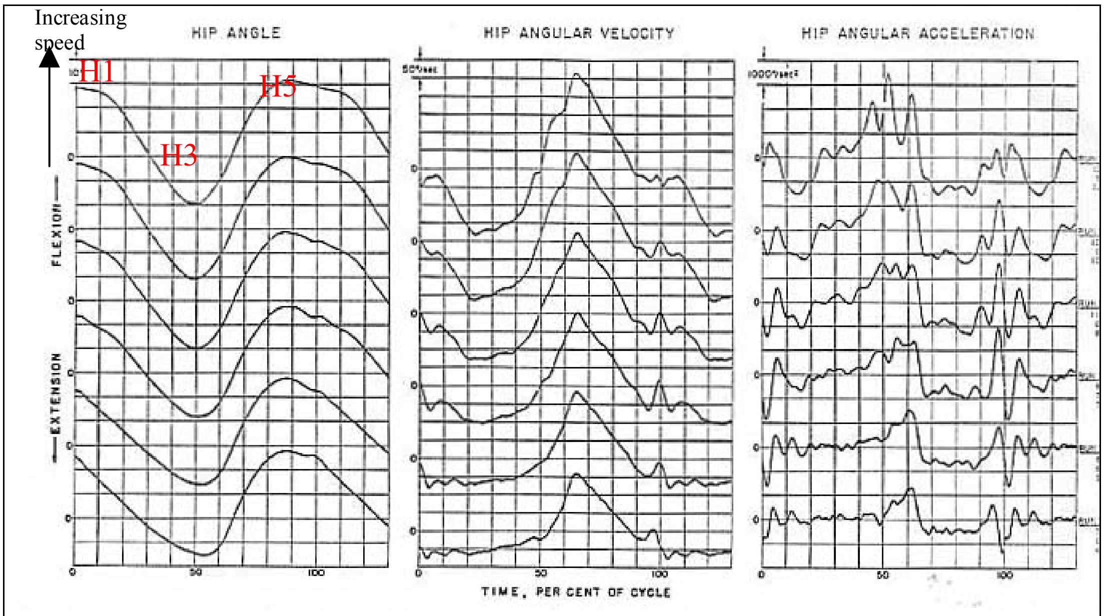


Figure 2-9: Figure from Lamoreux [125], The hip angles, and their derivatives (angular velocity and acceleration) are shown for a subject walking at different speeds. Waveforms are ordered with respect to speed where the fastest is at the top.

The pattern is much like a sinusoidal function apart from the slight asymmetry between the first half of the gait cycle and the second, therefore behaving much like a pendulum. The thigh movement is mostly responsible for the hip angle pattern since the trunk only rotates a relatively small amount.

Although the overall shape of patterns remains much the same, as speed changes, the amplitudes of the hip gait parameters exhibit the greatest change. The flexion range increases much more than the extension range and is possibly an indication of the ease of gaining extra flexion from the hips whereas extension does not deviate as far from the standing position. Considering temporal changes, the H3 flexion peak suggested by Benedetti [Appendix B] appears to occur earlier with speed whilst the maximum extension H5 remains relatively constant. The 3-5% shift in the H3 can be demonstrated more visibly by the gradient of the hip angle namely hip angular velocity crossing the zero line. Interestingly the peak angular velocity, similar to the H5 peak, also seems to occur at the same point as a percentage of cycle. It should be noted that when results are taken over an ensemble average, such as by Winter [222, 225] it is more difficult to distinguish these temporal changes with speed.

One might have expected a greater temporal variation in the angles than just described if one considers that the proportion of stance phase and swing phase making up the gait cycle changes as walking speed is changed. As speed increases, the contribution of swing phase to the gait cycle increases to almost 50%. If the patterns did truly superimpose each other then one might have expected the real time relationships of various features to vary similarly to the cadence / velocity relationships described earlier. The angular velocity and angular accelerations are however able to convey some of the lost time information since they are time differentials. These patterns are more representative of the changes in the walking pattern that are necessary to walk at different speeds. The other advantage of looking at angular velocities is that it removes systematic errors associated with drifting and inaccurate calibration of joint angles since the amplitude component is lost in the differentiation process, this means comparisons should be easier between different sets of data.

The sagittal hip angle chart to the eye is rather uneventful, however from the velocity chart it can be seen that as walking speed increases, various features dramatically change. For slow walking, just after heel strike (between H2 and H3), there is a small extension velocity, for faster walking speeds the angular velocity forms a trough of increasing magnitude. Similarly, the peak angular velocity also increases with walking speed.

2.2.2.1.3.2.b Knee kinematics

Generally, knee kinematic patterns are more complex than the hip, possibly because only the thigh contributes significantly to hip patterns since the body is relatively motionless relative to the global reference frame; the knee patterns additionally combine the shank motion with the thigh. In the sagittal plane, the knee angle is more phasic than the hip, where two troughs instead of one can be observed (compare Figure 2-10 with Figure 2-9). Flexion occurs just after heel strike reaching a peak K2; this peak increases with increasing walking speed although at very slow speeds it almost disappears. It is often stated that this flexion enables shock absorption at IC, and Saunders et al claimed that it enabled energy conservation by smoothing the body's trajectory [177] although this was disputed by Winter [226].

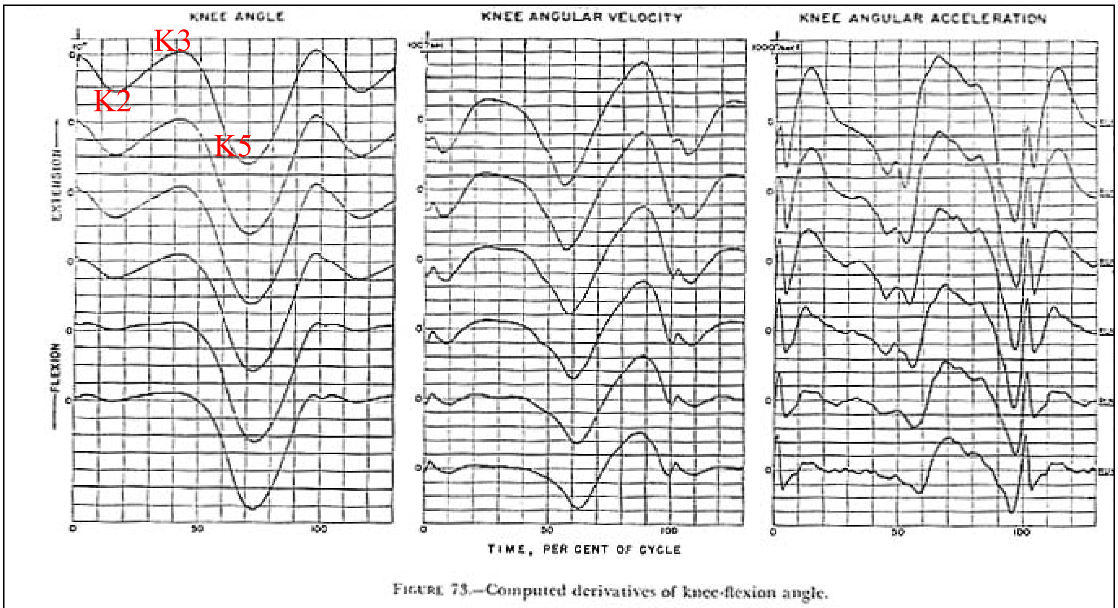


Figure 2-10 – Knee angles, knee angular velocity Knee acceleration

The leg extends once again after this peak flexion until Benedetti's K3 parameter is reached. It can be seen from Lamoreux's data that this K3 point seems to occur slightly earlier as speed is increased. It is not so obvious with Holden's data [102], which specifically investigated the knee joint function over a wide speed range. His results were however normalised over the stance phase, which may be expected to show less temporal variability because it is over a shorter interval than the whole gait cycle, but mostly because it might be said that the normalisation takes into account the change in stance/swing ratio that accompanies a change in speed.

The angular velocity of this absorbing flexion is also seen to increase with increasing speed starting at zero for very slow speeds, the angular acceleration shows when the necessary speed changes begin to take effect, peaking just before K2. Therefore, as speed increases the K2 bump (on the normalised time scale) effectively becomes taller and narrower whilst the angular velocity also goes through a similar transformation. The K5 flexion peak, which occurs during swing phase, does not appear to undergo as dramatic an increase with speed as the K2 parameter. In some respects, a constant pattern is possibly an indication of greater controlling influences being applied to maintain this despite extra energies with speed. The velocity changes for the flexion pattern show roughly two equal but opposite peaks either side of K5, the first one showing a flexion velocity accompanying the start of swing phase and heel rise, and the second an extension velocity. These two velocity peaks also increase with speed.

2.2.2.1.3.2.c Ankle Kinematics

The ankle pattern tends to be more variable than the hip or knee pattern. For example, for the subjects in Winter's results, when walking at natural cadence the ankle pattern had the greatest coefficient of variation at 68% whereas the knee and hip had 25% and 51% respectively [225]. There is also visibly more variation between each of the subjects presented by Inman [110, pp. 46]. There is less symmetry of features than can be observed for hip or knee patterns.

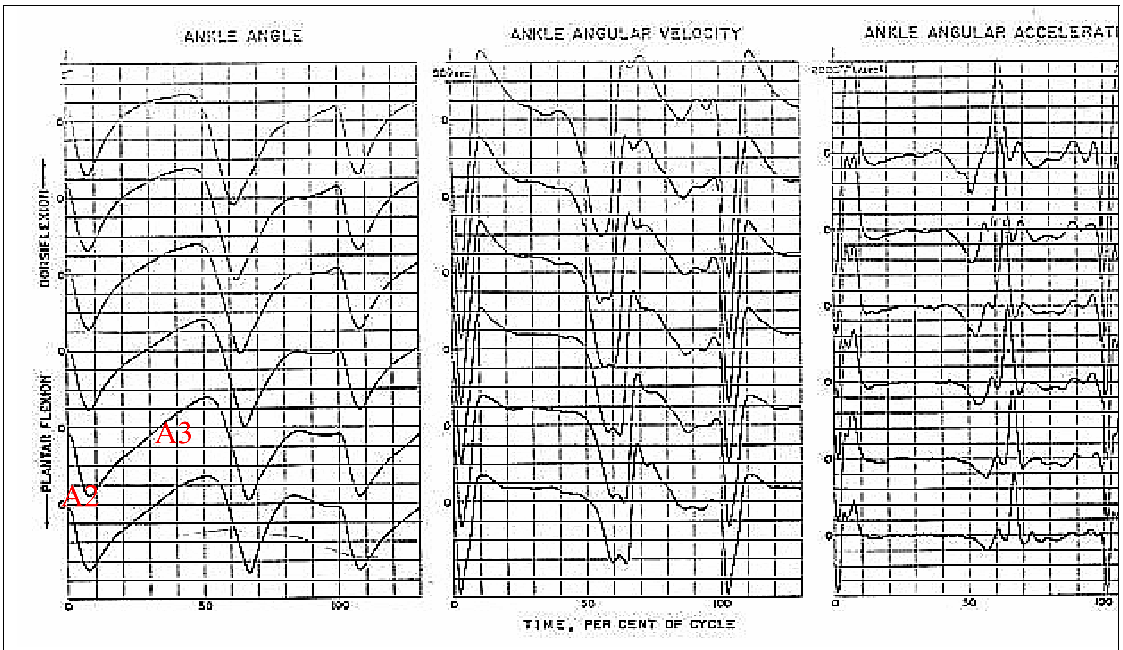


Figure 2-11 – Ankle angles, angular velocity, acceleration (from Lamoreux pp. 72)

It can be seen in Figure 2-11 that just after IC, a plantar flexion peak (labelled A2 by Benedetti [Appendix B]) occurs which is regulated by dorsiflexors. This plantar flexion angle at A2 increases with walking speed whilst remaining relatively fixed in terms of the phase of gait cycle. The shank then gradually rotates forward over the ankle causing a dorsi-flexion that peaks at A3. As walking speed increases there is only a slight increase in the dorsi-flexion angle depending on whose results one is examining. For example, hardly any difference can be detected in the peak amplitudes of Lamoreux's results whereas an increase of about 2.5° between slow and fast is visible in Winter's results. What is more commonly observed between different results is how the shape of the angle pattern leading up to A3 changes with speed. At slow speed the ankle rotates from A2 to A3 almost at a linear rate, however as speed increases there is a quicker initial increase in dorsiflexion. In fact, at the fastest speeds, this increase is fast enough to allow the dorsiflexion angle to plateau just before A3. This situation can be explained more easily if one observes the angular velocity during this period. After A2 during the foot flat period the angular velocity seems to end up at a plateau velocity of about 50°/s. The variation with walking speed seems to occur at the beginning just before the angular velocity plateau is reached. There is a bump in the dorsiflexion direction that is only just visible at slow speeds, however as speed increases the amplitude of this angular velocity bump dramatically increases. This would explain the initial increase in the angle pattern, which is observed at the faster speeds. It should be noted that despite these differences between slow and fast walking speeds there always seems a smooth transition between them.

The next interesting point in the gait cycle occurs just after toe off when there is a large plantar flexion peak labelled A5. This plantar flexion period is often described as a push off as explained in the observational section, where the rapid plantar flexion that can be seen is supposed to propel the body forward. Surprisingly there is not a large increase in this peak plantar flexion angle, however the large velocity peak associated with it is more decisive where an increase of about 100% in the angular velocity going from slow to fast speeds can be seen. After the plantar flexion, the ankle rapidly dorsiflexes into a neutral position in order to leave room for foot clearance during the swing phase.

2.2.2.1.4 Kinematic Summary

We have seen very consistent angle patterns both on an intra and inter-subject basis for all of the joints (especially the knee and hip). Winter for example correlates the angle patterns for various cadence groups and finds them to be high [222, pp. 24]. For example, the correlation for the slow/natural cadence group was 0.975, 0.995, and 0.99 for the ankle, knee and hip respectively, and for fast/natural cadence group 0.95, 0.99, and 0.99.

There seems to be a variety of strategies at different stages in the gait cycle to allow for changing speeds. The angle patterns do not retain exactly a proportional shape with increasing frequency; there is some transformation to the shapes in the angle patterns. In particular, the most striking effect is the elongation of various peaks according to increases in walking speed, e.g. the K2 peak almost disappears for some subjects at slow speeds. The phasic changes that occur with speed are not as prominent, although still visible with careful inspection, for example, the K5 peak appears to occur earlier in the gait cycle as speed increases. This observation is possibly a reflection of the fact that the proportion of stance and swing phase duration changes with speed.

The angular velocity at joints as might be expected is a better indicator of changes as walking speed is increased. Possibly an important point is that there is no abrupt change in the angle patterns as speed is changed but there is a gradual transformation of the patterns.

The above describes general patterns of the joints that the human system seems to try to achieve in order to facilitate a walking action. These patterns however do not describe how the human system is achieving these patterns, such as when and how much force should be applied to result in these joint trajectories. This will essentially be the subject of the next section, which reviews kinetic data in order to get some idea, from a control point of view, what is going on.

2.2.2.2 KINETICS

It is logical to begin with a review of the *forces* that are developed throughout the walking cycle as other kinetic variables such as *moments* and *power* can be derived from these forces. The resultant force exerted on the body by the ground can be measured using force plates and is known as ground reaction forces (GRF). This force has been studied extensively by many researchers as it is the algebraic sum of all mass × acceleration products of all body segments and thereby reflects the effect of all net muscle and gravitational forces acting at each instant of time. The ground reaction force is often used to gain an understanding of what is happening at joints since its effect is usually much larger than the other forces and apparent. Perry et al [158], as already discussed, used such methods (see Figure 2-1) where for example, if the force vector passed directly through a joint centre, then the external moment developed at the joint would be minimal.

The ground reaction forces, which will be transmitted through bone and tissue, must be kept under restriction, otherwise injuries can result. Similar to crumple zones in a car that help to increase the period over which forces act, the body has two such methods of force control: 1) by structure and 2) by movement. The first method incorporates a level of flexibility into the load bearing structures, for example via soft tissue cushioning (as with the soles of the feet) and structural design (as with the ability of the tibia to bend). The second method uses the joint articulations to effectively allow the body to crumple if stresses are likely to be too high. Take for example a person who jumps up a small distance and lands with very rigid legs, they will certainly notice the resulting stresses, whereas a slight bending of knees on landing will be much more comfortable. It would seem that because of the short time scales involved, the rigidity at the joint would have to be prepped before hand by the central nervous system and this can include using pre-established movement patterns.

For completeness, structural strength is also critical to how internal stresses generated by the ground reaction forces may be regulated. Increased strength, for example via increased bone thickness and density, will enable the body to cope with higher levels of stress.

Therefore, analysis of ground reaction forces during walking may help to establish acceptable levels for consideration by a knee controller where excessive forces in the amputee should be avoided partially by strategic movement patterns.

2.2.2.2.1 Ground Reaction Force

The figures produced by Benedetti [16,] show the general ground reaction force pattern for non-pathological subjects as in Appendix B [Figure B-1], split into its three orthogonal components; in the text that follows, Benedetti's nomenclature for the ground reaction force parameters will be adhered i.e. F1 – F9 [Table B-A].

The *Fore/Aft* force component shows whether there is an overall translational acceleration or deceleration of the body in the direction of progression. It can be seen that just after initial contact there is a large decelerating force acting to slow the body down (labelled F5 by Benedetti), however by halfway through the gait cycle this has been reversed into a propulsive force (F6). When walking with a steady velocity, the acceleration component practically cancels the deceleration component, which is logical, as a steady state velocity implies no overall acceleration. It should be noted that the negative F5 peak occurs just after the F6 propulsive peak on the contra-lateral limb possibly resulting in a smoother velocity path.

The *vertical* ground reaction force tends to show the forces required to balance the body's centre of mass and prevent it from falling to the ground, i.e. by mechanical support provided by the legs. It can be seen that the force rapidly rises after initial contact until it reaches a peak (F1), which is greater than body weight. The legs are clearly structurally strong enough to resist at least this force F1 and will often experience higher forces in other activities such as jumping, therefore these peak values do not ordinarily over stress the system. This peak occurs around foot flat during double support. To get a perspective on the amplitude of this peak, if the subject were to stand still in this foot flat position, gravitational forces would result in a ground reaction force of about half the body weight (assuming even weight distribution between the legs). Therefore, the deficit during walking must be due to inertial forces that must hence contribute over half body weight to the ground reaction force during this period. As the cycle continues on to mid stance, the vertical ground reaction force reduces to below body weight, despite it being in the single support period with forces due to gravity being equivalent to body weight. Thereby inertial forces must help to reduce the ground reaction force during this period and is consistent with upward accelerations of the body's centre of gravity at this point.

Another peak (F3) that is greater than body weight occurs half way through the gait cycle during the push off period. As this is also a double support period, inertial forces once again contribute significantly to the GRF, as might be expected during an active movement.

Several researchers have assessed the variability of the ground reaction force patterns in an attempt to identify pathologies according to significant differences from the normal range of patterns. Hamil and McNiven concluded that a stable mean pattern was only achievable after about 10 trials, which implies quite a large intra-subject variability [93]. However, Smith indicated that a four-trial mean was adequate to produce the normal pattern for a small group of subjects [187]. All this has been investigated in more detail by Giakas et al [82] who make an additional analysis, in which data comparisons are made in the frequency domain. Fast Fourier transforms (FFT) are used to find the harmonic coefficients from which the amplitudes of each harmonic can be calculated. Spatial and temporal comparison could thus be made on the force parameters[4] F1, F2, F3, F5, F6, F7, and F9 and their corresponding times T1, T2 … T9 although these are normalised to the stance phase (rather than the whole gait cycle as suggested by Benedetti). They found that the frequency content

[4] Benedettis' convention is referenced in this text although Giakas uses slightly different labels.

of the time domain parameters revealed the greatest information as to the stability of gait and allowed analysis of the symmetry between the two legs. The intra and inter-subject variability was also studied. It should be noted that they compared 200 trials (10 subjects × 10 trials × 2 sides) for subjects walking at their chosen free speed. They assessed that each subject had a fairly constant individual free speed and that this type of comparison was necessary because differences induced by body dimensions would make direct velocity comparisons unsuitable (i.e. same speed different subjects). No significant difference was found between the patterns for left, and right legs. Force components with high amplitudes had correspondingly higher standard deviations; however, when the coefficient of variation (CV) was used to assess the relative intra-subject variability only F8 and F9 in the medial-lateral direction had high variability reaching about 20% CV. The force parameters in the other directions had CVs less than half this value, which was considered to meet acceptable variability limits (<10%). The temporal variability was assessed by normalising the standard deviation by the stance duration. It was found that once again the medial-lateral parameters had the greatest variability, in particular for T9, which was significantly higher than the other variations. They also tried to test the hypothesis that the variability increases because of the impact due to heel strike. This was however only shown to be the case for the fore/aft force where the CV for F5 was greater than F6; in the vertical direction F1 and F3 were similar to each other as were F7, F8 and F9 in the medial-lateral. As a comparison, Winter calculated the CV for an ensemble average of ground reaction force data, which underwent temporal normalisation with respect to the whole gait cycle and the force with respect to weight [222]. He found that the intra-subject CV for the whole gait cycle for the vertical component was 10%, and the fore/aft component was 26% (much greater than Giakas) for trials taken days apart. He did not analyse the med-lat component. These results would be expected to be higher than those quoted above because the whole gait cycle is being compared and the temporal variance will affect an ensemble average as described previously, e.g. the standard deviation of the fore-aft temporal parameters in Giakas' work, were about 2% of stance phase. Nevertheless, the two sets of results do also demonstrate that the CV for the anterior-posterior force components is larger than the vertical indicating a greater amount of variability. Winter also finds that the CV representing the inter-subject variability is roughly double the intra-subject.

The frequency analysis by Giakas showed that the medial-lateral force component had the highest average frequency content. For the vertical direction 95% of the signal amplitude is contained within the first 20Hz of the frequency content, for the fore-aft direction about 19Hz and the medial-lateral component was within 26Hz.

It was thought that such information on the variability of ground reaction forces would be useful to distinguish regular asymptomatic walking patterns. Ultimately, if a knee control design is working well then it would be hoped that some of these invariant aspects in the ground reaction force would also be reproduced in the amputee.

Although the effects of walking speed were not investigated by Giakas, other researchers such as Andriacchi et al [8] have examined the ground reaction force parameters with respect to speed. They found that for most of the force parameters except F9 a linear relationship with velocity was adequate, where the vertical force parameters had a greater rate of change with speed than the other directions, Z2 being the largest. The amplitudes of F4 and F9 did not seem to show any meaningful changes with walking speed.

In conclusion, the general ground reaction force pattern can be easily recognised. However, differences do occur from trial to trial and from subject to subject, therefore if comparisons are to be made with amputees then parameters with high variability such as the medial lateral force parameters should be avoided. The question to answer from a control point of view however is, are these differences significant and do they represent subtle adaptations to environmental perturbations, walking speed and direction or indeed to previous control mistakes. If this is the case, should a controller of the knee try to help replicate any of these adaptations or would it be sufficient to replicate the mean pattern (assuming this is the most suitable for AKs, see later). Care must also be taken about the variability as once again thought needs to be given as to the walking speeds that are compared.

2.2.2.2.2 Moments

Since it is of concern how control of human movement is accomplished, a starting point is to see the direct effects of this control in the form of the net forces produced by all the muscles acting across a joint. This will be a reflection of the output of the neuromuscular system, as it actualises movement for a particular segment. Unfortunately, it is difficult/impractical to directly measure the individual muscle forces and even specify their origin and direction of action on the live subject so that a resultant force vector can be calculated. The inter-segment moments can however reveal much information of this nature. An inverse dynamic process can be used to eventually calculate the moment about the ankle joint, using ground reaction forces, gravitational forces, and the segment kinematics (details are given later [section 7.2.3, pp. 233]). Assuming joints have negligible friction forces, this moment about the joint axis is that developed by muscles and tendons (tissue), and is equivalent to the moment developed by "external" force actions due to linear and angular accelerations, gravity and ground reaction force (in the case of the foot). Similarly, calculations of moments can progress to the proximal segments as each unknown variable is solved gradually up the chain.

This net view using moments was of sufficient detail, as reproduction of the activity of individual muscles is not the goal. It is better to understand the combined effect of all of the muscles acting across the knee joint since any knee actuators will similarly act to give the gross function of the prosthetic knee.

Inter-segment muscle moments therefore highlight whether muscles are working to either extend or flex a joint. Winter suggests that the sign of moments conventionally is positive for counter clockwise moments acting at the proximal end of a segment [222, 224, 225, 227]. Thereby knee extension, hip flexion and ankle dorsi-flexion is positive. It should be noted however that in Benedetti's [16] definition of moments, the moments due to the external moments are presented which are equal but opposite in direction to the muscle moments, e.g. so that external extensor moments are countered by flexor muscle moments [Figure B-1]. In addition, Benedetti uses a positive sign for external flexing moments.

Inter-segment moments have thus been calculated for walking by many researchers; classically these include Bresler and Berry [31], Winter, [222, 224, 225, 226] and

Inman et al [110]. Although similar morphologies between corresponding moment patterns can be observed, the variability is somewhat greater than for kinematic variables such as joint angles. Caution must however be taken when interpreting variability of inter-segmental moments. Bresler found that a greater maximum variation results with more proximal joints because errors increase cumulatively, the number of parameters for proximal sections in effect increases. Bresler estimates the maximum value of these errors at the hip, knee, and ankle to be 10%, 8%, and 2% respectively, during stance and 7%, 5%, and 5% during swing.

The method of normalisation to enable inter-subject comparison differs according to researcher [section 2.2.2 and Appendix A], where for example Winter presents his data normalised to Body weight, whereas Hof [99] suggests body weight*leg length, which leaves a dimensionless parameter and in some respects is the purpose of normalisation. Pierrynowski [161] however considers Winter's method as ad hoc in terms of logic (see section 2.2.2) although he did find that it was equally as good at reducing the inter-subject variability of gait data.

Moisio et al. more recently specifically examined the effects of these two normalisation procedures on moment patterns [140]. They analysed the patterns of 158 subjects and tried to determine which "normalisation technique most reduced the effect of height and weight on peak, hip, knee and ankle moments during walking in a group of normal subjects". They gathered walking trials over a speed range of 0.9-1.4m/s so that they could also minimize the effects of walking speed. A "forward stepwise multiple regression" model was thus used to calculate the contribution of height, weight, and speed to the variance in the moment parameters. They found first that without any normalization, height or weight had significant effects on the variance of all the moment parameters, where 7-82% of the variance could be accounted for depending on the parameter (see Appendix A [Table A-D] for a full summary of results). They also found that both normalisation techniques were effective in reducing the variability; the similarity in results was possibly a reflection of the fact that weight was quite highly correlated with height ($r=0.603$ $p<0.001$). Therefore, they were able to show that, except for the ankle dorsiflexion moment and hip adduction moments, the variability after normalisation due to weight and height was reduced to at most 6%.

Similarly, Sum et al. tried to verify which normalisation technique was most effective at reducing variability; they concluded that weight*height was most effective [196].

It should be noted that the above variability's were made by comparison in the temporal dimension (albeit normalised to stride-length). Frigo et al. were however also able to glean further information by combining the inter-segmental moments with kinematics [79]. They were able to directly observe the relationship between joint angles and moments by plotting them. The variability in both dimensions was represented by plotting the standard deviations (SD) of both the angles and moments as an ellipse taken at each comparable stride time (%). The temporal dimension in these plots is thus only perceptible from the density of the ellipses along the curve and from specific labelling of direction and gait events such as HS and TO. The pure angle-moment plot without these SD ellipses superimposed does however enable a good visualisation of any changes that may take place in the moment patterns, for example with speed. They allow one to see if particular moment features can be linked to joint angles as opposed to the phase of the gait cycle. In this respect, the group reflect that the use of such diagrams may be indicative of a neuromechanical relationship because angles must comply with muscle kinetics such as force/length, force/velocity, and kinaesthetic feedback. In this regard, it is a shame they were not able to plot the direct standard deviations of the angle-moment curves (i.e. comparison of moment at discrete angles & vice versa) rather than deviations calculated on a temporal basis. To do this may require breaking the data into two or more sections to avoid problems associated with having two or more routes.

Frigo suggests that the morphology of the angle-moment plots may contain some information regarding the motion. For example, they identify separate phases of a gait cycle by finding linear portions of the profile. The linearity of these phases can be thought of as an indication of how closely an angular feature may match its moment counterpart; in effect, the angle profile is proportional to the moment profile over this linear duration. The slope of a linear phase would thus reflect a scaling factor between the two variables and a change in this slope, with speed for example, would highlight where moment adjustments may be made.

2.2.2.2.2.1 Ankle Moment

The most discernible ankle moment parameter, in the sagittal plane, is the peak plantar flexing muscle moment (AM2 according to Benedetti's labelling [Table B-C]). Benedetti also suggests AM1 as a parameter although it is small in amplitude and can be ambiguous in direction. AM1 occurs just after IC while AM2 is just before TO. Exemplar data, gathered by Winter [222] is presented in Figure 2-12, in which data from up to 19 subjects, at three different cadences was ensemble averaged; in addition data from Bresler [31] is shown, which shows data at four different speeds although each from a different subject.

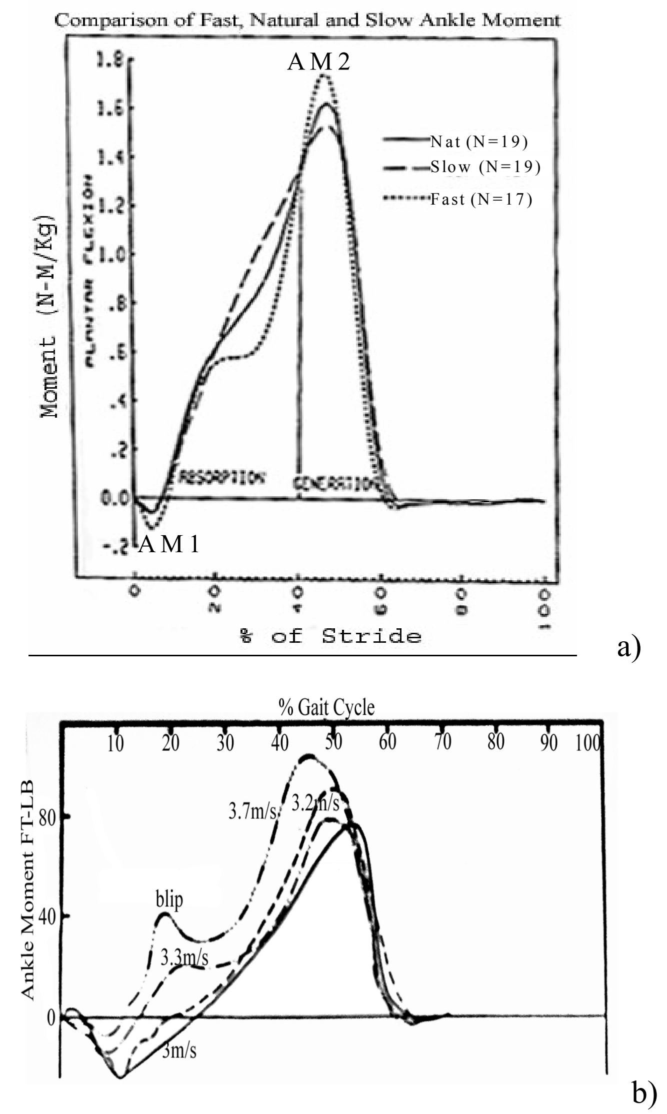


Figure 2-12 - Ankle moment patterns showing speed variation adapted from A) Winter [222],b) Bresler [31].

The intra and inter-subject variability for the ensemble averaged sagittal moment patterns were examined by Winter. He found that, for natural cadence, the intra-subject coefficient of variation (CV) of data taken minutes apart was 24%, whereas it reduced to 16% when measured days apart. There is almost double the variability for the inter-subject data, taken for 16 subjects, where the CV was 45%.

The angle-moment diagrams produced by Frigo et al [79] show the extent of the variation associated with both variables using ellipses superimposed onto the graph. Three phases labelled A, B, and C, were identified that correspond to linear segments within this loop. The first phase A, tallies with early dorsiflexion, phase B with late stance phase involving the plantar flexors for push off, and phase C the swing phase where ankle moments are negligible. It can be seen, from the sizes of the ellipses, that the intra-subject variability during phase B is greater than phase A.

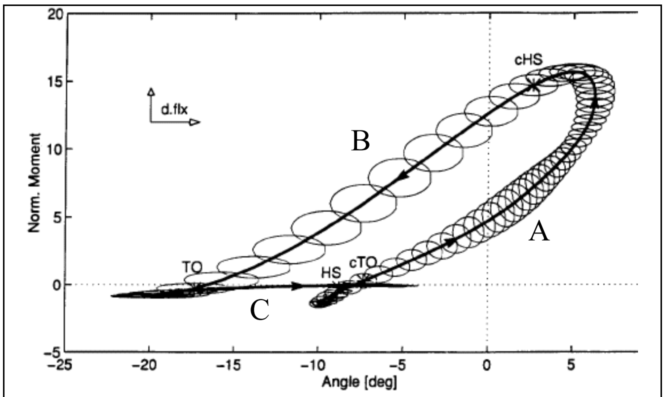


Figure 2-13 – Ankle angle-moment plot from Frigo [79]. Ellipses show SD. of both variables measured in stride time (%) base. Moments are normalised with respect to weight.

From Winter's inter-subject data Figure 2-12, it can be seen that general trends are evident in the moment profile as speed increases. For example, both AM1 and AM2 seem to increase in amplitude with speed; AM1 however retards in normalised time, whereas AM2 progressively shifts to an earlier percentage of the gait cycle (this effect is more pronounced in Bresler's data [31]). The effect of speed is also visible with angle-moment plots [Figure 2-14]. At lower speeds, the loop is smaller than for higher speeds, where a more pointed apex at the end of phase A is apparent. The apex seems to mostly shift left along the angle axis as speed increases. The drift along the moment axis is comparatively modest meaning there is a greater relative change in ankle angles with speed. At slow speeds, the apex is at its sharpest and positioned furthest to the top and right implying that the maximum moment (AM2) coincides to a greater degree with the maximum angle (A3); for the fastest speeds, AM2 however occurs after A3. In fact, as Frigo comments, only at the fastest speed, an abrupt change in the slope of phase A is perceptible; whether it reflects a sharp transition needs a more continuous assessment of speed. The ground reaction forces should however, with all these assessments, not be forgotten at this point.

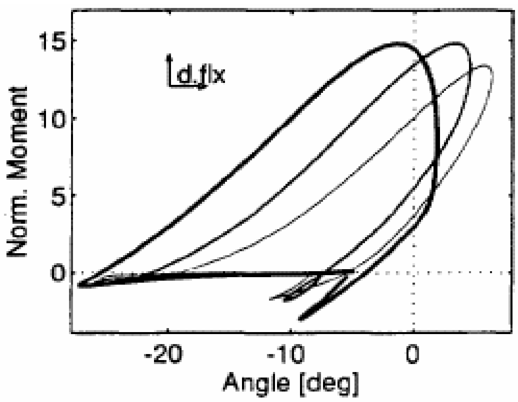


Figure 2-14 – Effect of speed on ankle angle-moment plots, as made by Frigo [79]. Thin, medium and thick lines show slow, natural and fast walking.

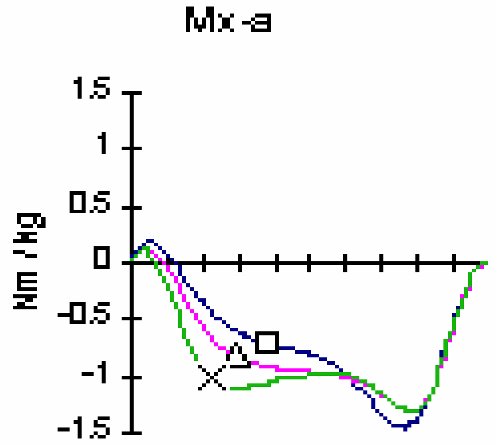
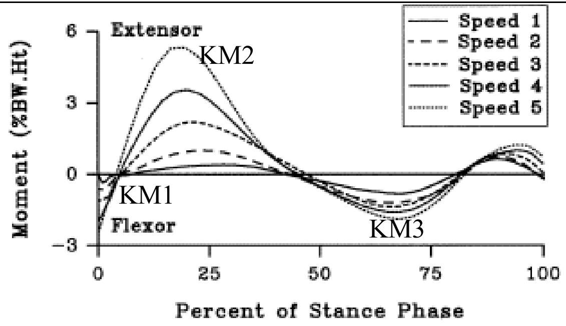


Figure 2-15 - Adapted from Xu [230]. Effect of changing direction on ankle moment: □ = 0 degrees change, ∇= 45 degrees, X = 90 degrees.

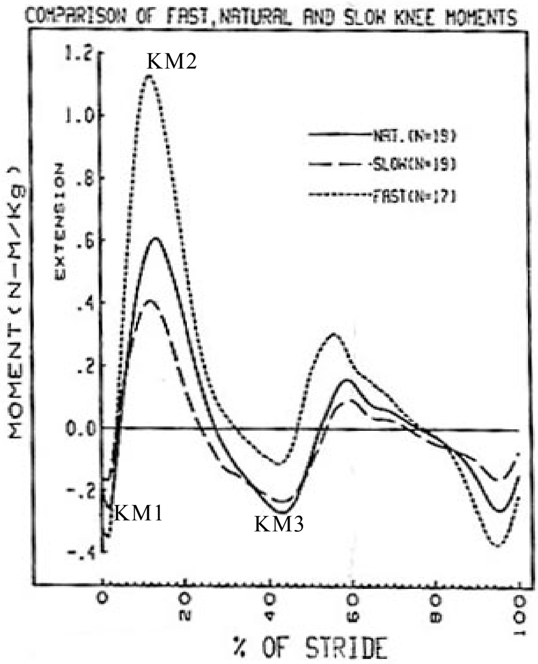
With the temporal plot of moment, the formation of an additional plantar flexion blip between AM1 and AM2 also seems to occur with increasing speeds (this is most pronounced in Bresler's plot [Figure 2-12b]). Intriguingly, some of the moment profiles presented by Xu [230] also form a similar blip, but in this case, it seems in response to changing the direction of travel [Figure 2-15]. The role of the blip may therefore possibly be to further decelerate the body by slowing down the rate at which the leg rolls over the ankle.

2.2.2.2.2.2 Knee Moment

Examples of knee moment curves of variation with normalised time, from two separate research groups, are shown in Figure 2-16. Although the profiles are somewhat more complex than Benedetti's example [16], the three gait parameters he suggests are distinguishable at all speeds. The intra-subject variability taken days apart was 37%, according to Winter's measure of CV, which is twice as high as the ankle moment. The inter-subject variability however was much higher at 135% CV for natural cadence. Holden however reports that the maximum CV for the inter-subject variability in his data was only 46.8% taken at the lowest speed and decreased slightly as speed increased [102, 101]. Holden only used about half as many subjects as Winter though, and strictly regulated speed according to the scaled walking speed system [section 2.2.2.1.1].



a) From Holden [102].



b) From Winter [222].

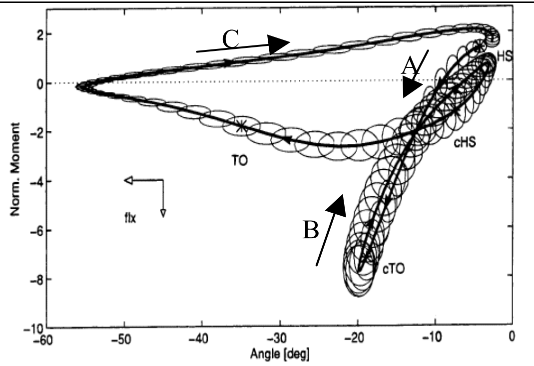
Figure 2-16 – Examples of knee moment patterns from two separate groups. Both include effects of speed.

Despite this apparent high variability, the ensemble-averaged data is able to smooth out the stride-specific features leaving a basic pattern. There are two extensor moments. The first (KM2) occurs just after IC, as the knee yields and reaches its first flexion peak. This requires muscular effort since the GRF vector lies posterior to the knee joint at this time. The other extensor moment is not labelled by Benedetti, but occurs around push off and possibly prevents buckling because of the flexing moment created by the AM2 plantar flexion. Thereby it gives a possible mechanism for controlling heel rise during early swing phase.

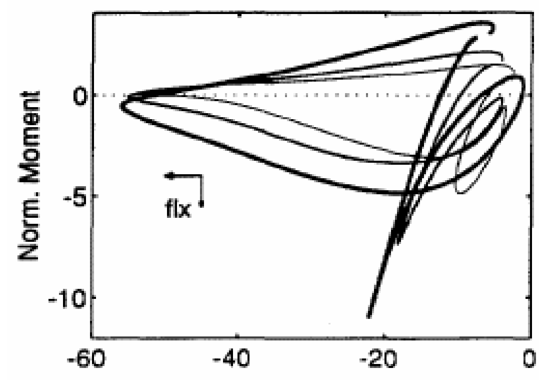
Before this regulation by the unlabelled extending moment, a flexing moment KM3, around heel off, is required to initiate knee bending since the leg is straight and the ground reaction force passes in front of the knee.

At the end of swing phase as IC is reached, there is another flexion moment visible in Winter's data [Figure 2-16b] (although not shown in Benedetti's). This flexing moment is possibly used to control the deceleration of the shank before it is fully straightened for IC.

Considering now the influence of speed on the knee moment pattern, KM1 undergoes the most noticeable change. Holden's data, for example, shows that the KM1 amplitude for fast speed increases by 19 times compared to the slow, even though the fastest cadence was only 2.4 times faster than the slowest. Winter's data however, does not show such a dramatic increase. KM1 has a 270% increase from slow to fast speed, although the cadence change in this case was just 147%.



a) Ellipses represent standard deviation.



b) Thin medium and thick lines represent slow, medium and fast walking..

Figure 2-17 – Knee angle-moment plot and its intra-subject variation at a) one speed and b) three different speeds from Frigo [79]

Frigo et al [79] have also made angle-moment plots for the knee [Figure 2-17]. It should be noted external moments, rather than muscle moments are plotted. These angle-moment diagrams show two high slope phases (A and B) corresponding to the knee yielding. The fact that phase B almost rebounds along the same path as phase A reflects the symmetry in the morphologies of both the moment peak (KM2) and the angular yielding peak (K2). Furthermore, because these phases are linear, the moment changes proportionally with the angle, so that these separate parameters have a close temporal link. This phenomenon is logical where a greater knee flexion at this point means a greater offset in the ground reaction force vector from the knee joint. Possibly, because of this straightforward cause and effect, the slope of these two phases shows little significant change with speed [Figure 2-14] implying no additional gain factor is imposed by other mechanisms, i.e. the angle morphology roughly increases proportionally to the moment morphology during this phase.

The next loop after phase B corresponds to the large flexion of the knee during the swing phase; this portion cannot be linearised. However, as the knee re-extends before IC, there is a linear portion of the graph labelled phase C. The slope of this phase showed a linear dependency on walking speed so that the flexing moment controlling the deceleration of the shank is more significant at faster speeds.

2.2.2.2.2.3 Hip Moment

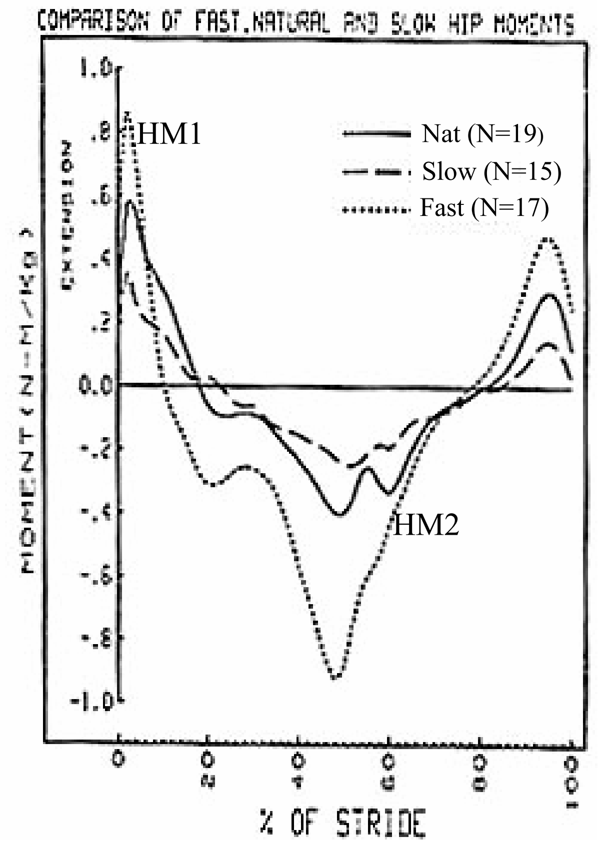


Figure 2-18 – Hip Moment profile, from Winter [222]; includes effect of different speeds

The ensemble averaged patterns for slow, natural and fast cadence gathered by Winter [222], is presented [Figure 2-18]. An extending moment peak (HM1) emerges just after initial contact driving the thigh from its flexed position towards extension thereby bringing the body forward relative to the thigh. Transmission of this moment to the knee joint can also provide additional extensor moments about the knee axis to control knee stability. The moment decreases during stance until a flexion peak (HM2) is reached just before toe off. This flexion moment will help initiate the swing phase by driving the thigh forward.

Although Benedetti does not include the next extension peak as a gait parameter, it can clearly be seen from Winter's graph at the end of swing phase. This extension moment is probably used to bring a controlled deceleration of the leg just before IC.

Hip moments also exhibit a similar variability to knee moments e.g. the intra-subject CV was 66% for trials taken on different days as opposed to 37% for the knee. The inter-subject variability was much higher as expected, where for fast, natural, and slow speeds the coefficients were 207%, 140%, and 76% respectively. It was only for fast speed that the hip had less variability than the knee.

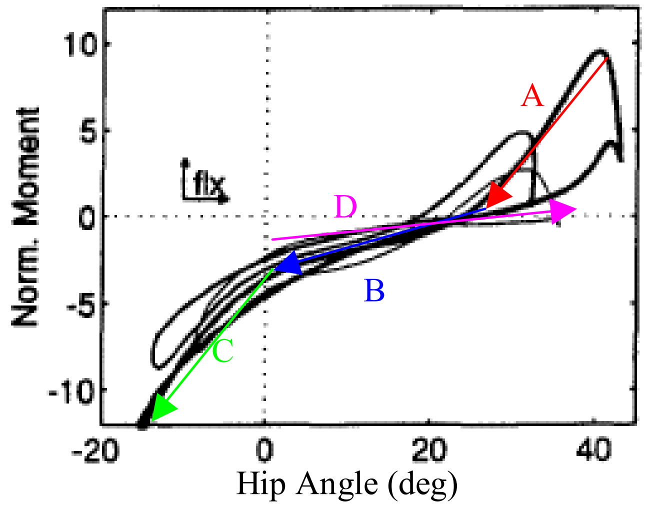


Figure 2-19 – Hip angle-moment plot for fast, natural and slow speeds, adapted from Frigo [79].

The angle-moment figures produced by Frigo et al. [79] of the hip show dramatically thinner loops than for the knee and ankle [Figure 2-19]. A very thin figure of eight is formed in which the first (top) loop is counter clockwise giving knee flexion and the second is a clockwise extensor moment. The two loops closely overlap each other, demonstrating symmetry of the hip movement. Frigo identified four linear phases within the loop (labelled A-D); such linear phases over a large portion of the gait cycle indicate the moment morphology is proportional to the angle morphology[5]. Phases A, B and C cover the hip as it extends and phase D corresponds to flexion during swing phase.

Although it can be seen from Winter's graphs that all the moment peaks exhibit variation with speed, Frigos plot showed that both phases B and C change proportionally with the change in hip angle. Only the slopes of phases A and D showed a significant linear dependence on speed.

2.2.2.2.2.4 Support Moment

As mentioned, both the knee and hip moments exhibit much greater variability than the ankle moment, which is to be expected because of the greater offset of the ground reaction forces from these proximal joints than the ankle. Winter however goes on to hypothesise that the hip and knee compensate each other's actions so that an interaction between the knee and hip variability occurs giving an overall repeatability of the behaviour of the whole leg. He tried to measure this phenomenon by trying to

[5] Such correlations between different variables has significance later on for control purposes

calculate the overall contribution of the joints towards support. He reasoned that extensor moments would contribute to what he termed support moment (M_s):

$M_s = M_k - M_h - M_a$ where k, h and a denote knee, hip and ankle respectively. Note Winter's convention of counter clockwise moments being positive for the above equation to be valid [section 2.2.2.2.2, pp. 46].

Remarkably, when M_s was calculated, its coefficient of variation was much less than the high variability experienced individually for the knee and hip components. He found for walking at natural cadence that the support moment CV was reduced to 56% as opposed to the hip and knee contributions of 144% and 150%; the ankle had little interaction where its CV was lowest at 45% [225]. Similarly, for patients with knee replacements, he was able to demonstrate how the hip compensated so that despite the abnormal knee pattern a relatively normal support pattern was produced [227]. In fact, in the Winter and White [223] report, a one to one trade off during stance is claimed, so that if the knee extensor reduces its contribution to support by 10Nm then the hip will increase its contribution by 10Nm. This compensation effect results in a much lower covariance for the support moment when compared to the hip and knee moment.

2.2.2.2.3 Power

From a control point of view, a very pertinent parameter is the power transmitted across joints. This variable represents the rate at which energy is absorbed, generated, or transferred across a joint. For example, if energy is absorbed, this can lead to a reduction in the relative rotation between the proximal segment and the distal segment, i.e. musculature at the inter-segment junction will act like a brake. Energy generation represents the ability of the muscles/tissues that surround the joint to generate movement.

Inter-segment power can be calculated by the product of the muscle joint moment and the relative angular velocity. If these variables are defined so that the same signs represent the same directions, then positive power values will indicate increasing limb rotation, i.e. muscle acts in the same direction as the limb rotation. The converse is true for negative powers where the muscle moment opposes the joint rotation so that a braking effect is applied. In terms of muscle biomechanics, concentric muscle contractions may thus provide positive joint powers whereas eccentric contractions may be related to negative powers. Isometric contractions, in which the muscle length remains the same for uni-articulate muscles, would not absorb or generate power, although they would enable forces/moments to be transmitted across joints.

Knowing the rate at which energy is generated or absorbed enables one to determine/estimate, in control terms, when key actions may occur to produce a movement. It is akin to knowing when to apply the brake or the throttle, although it does not directly show how much they should be applied.

Although, electromyographic (EMG) recordings may show when muscle groups are being activated, it was felt that the power was a better variable for analysing the overall activity of a limb for control purposes. EMG suffers from signal ambiguity and would be very difficult to decipher and quantify (i.e. which muscles are at work, and how are the signals related to function etc?). Winter also claims, "diagnostically he has found that joint mechanical powers to be the most discriminating in all their assessment of pathological gait" [224, pp. 103].

2.2.2.2.3.1 Ankle Power:

The general ankle power trend for non-pathological subjects shows that the greatest ankle activity takes place as push off occurs just before TO. At this point, there is a large energy generation peak referred to as A2 by Winter [Figure 2-20]. This happens to be the largest peak out of all of the joint powers. Meinders et. al. made a more in-depth study of the role of the ankle during this period [138]. They showed that although the plantar flexors generated a lot of energy during push off (aproximately. 23.1 joules) only 4.2J of this was transferred into the trunk. This meant that the power at the ankle was primarily used to accelerate the leg ready for the swing phase, rather than raising the trunk against gravity. Essentially, at this juncture, the ankle contributes mostly to the forward momentum of the body for walking. This is much in agreement with the observational findings of Perry described earlier on for terminal stance.

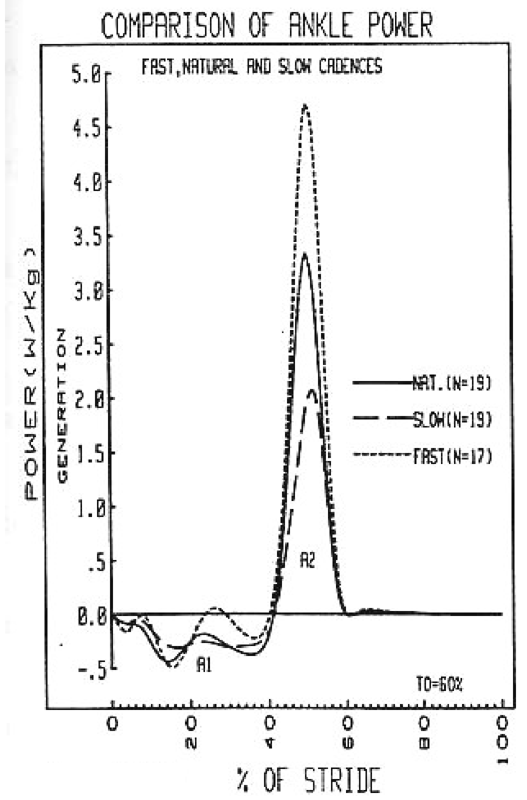


Figure 2-20 - Ensemble averages giving the Ankle Power for three speeds from Winter [222].

Preceding the A2 concentric burst of activity, there is a region of negative eccentric activity labelled A1 by Winter. Its amplitude was relatively small in comparison to A2, and is more variable and ambiguous; a variety of morphologies from subject to subject can be demonstrated during this period. For some subjects even positive power elements can occur within this A1 region.

For completeness, it should be mentioned that during swing phase, activity at the ankle is small since only small inertial and gravitational moments contribute to the external moments. Therefore, because the external moment is relative small, only a small internal muscle moment about the ankle is required to maintain control.

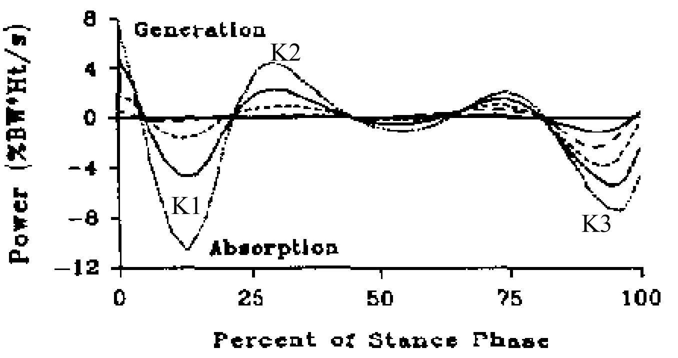
As might be expected, the variability of ankle power is higher than the moments about the ankle axis, because an additional variable is taken into account. For example, Winter calculates the intra-subject variability as measured by CV to be about 50% at natural cadence and the inter-subject variability about 100% which are roughly double those for the corresponding ankle moments.

Only a few reports, including Winter, Andriacchi and Holden, examine how the muscle joint powers vary with walking speed. Winter presents some of his data grouped into three cadences, slow, natural, and fast as shown in Figure 2-20. This averaged data clearly shows that the A2 peak is dependent on speed where an increase in the amplitude with speed is apparent. Winter quantifies this further by plotting the peak power amplitudes against speed. He finds a strong correlation and derives a linear regression model where $r=0.7725$, $p< 0.0005$ [225]. This result described by Winter however was not repeatable on children (ages 5-11 years) as investigated by Chen et al. [42]. They grouped their data into three speeds (slow: < 0.9m/s, normal: 0.9-1.2 m/s, and fast: > 1.2m/s) and no statistically significant relationship of the positive ankle peak power with walking speed could be demonstrated. In fact, results suggested a negative correlation, i.e. the A2 burst decreased with speed. This was further emphasised when the contribution of work (i.e. the areas under power profiles) made by the ankle to the total energy of the leg during this period was calculated. They found that at the slowest speeds ($0.63ms^{-1}$), the ankle contributes 62% of the required energy, whereas at the fastest speeds ($1.8ms^{-1}$), only 18% of the energy was produced by the ankle. He reasoned that the increasing energy demand required for faster walking was progressively met by larger muscle groups where they could work at a lower percentage of their maximum capacity. The statistical validity of both groups was similar where Chen et al. took 58 measurements whereas Winter used 43 trials to determine a speed relationship. Therefore, the difference may be attributed to inaccuracies, or discrepancies in methodologies or subjective differences especially considering children were studied by Chen whereas Winter used adults.

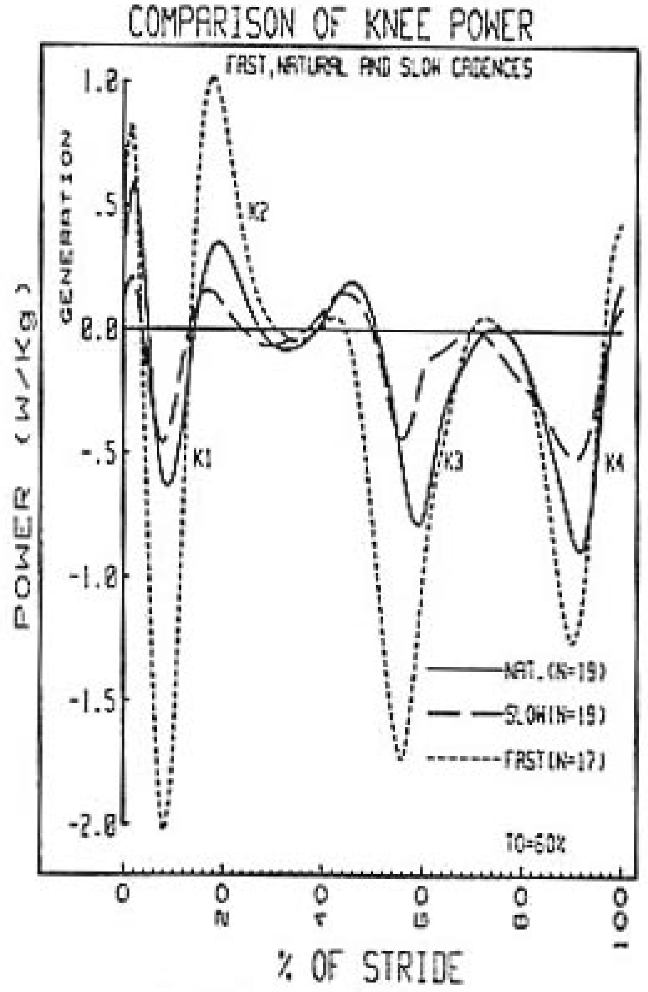
Considering now the temporal variance, it would appear that this is small for the A2 peaks of the averaged data presented by Winter, i.e. the A2 peak seems to occur at roughly the same phase whatever the walking speed. Meinders recorded the start of the A2 burst for self selected speeds as being at 44.3% ± 2.8 % of stride, i.e. it has a standard deviation which can lead to an error of about 0.03 seconds (assuming a stride time of 1.02s, from Holden's scaled natural speed [Table 2-B]). It is hard to know from these results the true speed dependency of the power features in the temporal dimension.

2.2.2.2.3.2 Knee Power

The knee power pattern can be seen to have more phases during the gait cycle than any of the other knee variables such as moments, angles, or angular velocities [Figure 2-21]. It has a mean value that is negative and tends to indicate that its major function is to absorb/transfer energy during the walking gait cycle so that it mostly has a braking action. The overall pattern published by researchers including Bressler, Winter, Inman and Holden [31, 222, 110, 102] appear to be very similar to each other in shape although there are slight differences in amplitudes. Winter has given labels to the main features common between subjects in the knee power patterns [Figure 2-21]. Just after IC there is a power absorption phase labelled K1. This is linked with controlling knee flexion during the loading response. It prevents the knee from completely buckling whilst permitting some flexion. It is thought that this enables functions such as shock absorption as



a) from Holden [102].



b) from Winter [222].

Figure 2-21 – Ensemble average of Knee powers taken at different speed groupings.

the knee yields and reduction of the vertical excursions of the body's centre of mass. The peak of this power occurs before the corresponding knee moment peak and the peak flexion angle and is therefore possibly in response to a peak in the angular velocity. For control purposes, it demonstrates that the greatest activity (in terms of rate of energy transfer) is not necessarily coincidental with the moment activity.

The next phase is K2 and is a main contributor to power generation at the knee joint. It is often thought to give the knee additional assistance to overcome the flexion moment caused by ground reaction forces (GRF) aligning posterior to the knee joint. As the knee re-extends, the GRF vector aligns through the knee joint centre and there is no more requirement for this concentric burst. The K2 phase, therefore, occurs just after the flexion peak, and before the leg is straightened again.

The next eccentric activity is the K3 burst that is attributed to controlling the amount of knee flexion for the swing phase. As mentioned earlier there is a lot of kinetic energy imparted to the leg during push off by the ankle A2 phase and by the action of the hip. This causes the shank to swing back and upwards according to the amount of kinetic energy imparted. Therefore the K3 burst can be used to control how far back the shank will swing. If this is not accurately controlled, it can have knock on effects for the remaining trajectory of the shank throughout swing phase, e.g. the timing of IC may be disrupted. Therefore, once swing has been initiated the knee remains relatively passive until the end of the swing phase where there is another absorption peak (K4). This peak once again precedes the maximum extension encountered just before initial contact. Therefore, the action of K4 may be to sufficiently reduce the angular velocity between the shank and thigh and prevent a clunking at maximum extension. The K4 phase finishes by rising slightly above zero around initial contact, maybe as a final controlling touch so that the leg is positioned exactly ready for initial contact.

Generally as speed increases the amplitudes of various features identified in the previous section increase. The study by Holden illustrates this very clearly for the stance phase of walking where mean data for five speed groups is presented [Figure 2-21a)]. These increases seem to relate to general increases in knee flexion angles although it is hard to say whether they are in proportion to each other since at very slow speeds the peak flexion angle is very small as are the peak knee powers.

However, comparing slightly faster speeds, the knee angle increases from -10^o to about -25^o an increase of 250% whereas the K1 power goes from -1.5 (%BW.Ht/s) to -11(%BW.Ht/s) for the same speed range, i.e. an increase of 730% and even the K2 peak increases by about 300%. These general observations are confirmed by Winter where he correlates both the eccentric K1 and K4 peaks with velocity; a negative linear relation was found with high correlations of 0.83 and 0.76 respectively. In addition Chen's study also points to this opinion where they examine the work done at the joints with speed, where they calculated the absolute area under the power curves. They indeed found that the work generated at the knee had the greatest increase with velocity of all the joints.

Another effect of increased speed is the appearance of various bumps/peaks in some subjects which are not present for slower speeds. For example, just before the K3 phase there is also another slight positive peak which becomes visible. The graphs presented by Winter for 3 speed groups also seem to exhibit the same behaviour with speed as given by Holden. However, the positive peak, which is not labelled and appears just before K3 for higher speeds in Holden's graphs, does not behave so well in Winter's. In Winter's example data, the natural speed has the largest peak value whereas the fastest has the lowest so that there is no longer a nice positive correlation with speed. This could possibly highlight the reason why this phase wasn't labelled by Winter in the first place, i.e. because of its inconsistency.

Once again when trying to assess inter/intra subject variability of power, care must be taken of the variability due to speed and subject dimension differences. Holden tries to minimise these factors by only comparing speeds which fell within ± 0.02 statures.s^{-1}. This uses assumptions previously discussed that subject height is the most influential subject parameter on speed/cadence relations. For the 5 speed groups she tested she showed that the coefficient of variability of mean knee power measured on 18 different subjects decreased with speed from 17.8% to 11.3%. This compares to the knee mean knee angle which did not seem to alter its variability much with speed. For example, the range of variability was from 7.6% to 9.7% and this was not correlated, i.e. speed 3 had the lowest value and speed 5 the highest. In other words the knee angle is controlled with similar precision for all speeds, however the variability in knee power required to do this between all subjects' increases as the walking speed reduces. Winter's data also shows similar findings in that the power variability decreased with speed whilst the angle variability remained much the same. It should be noted however that the actual CV values are much greater than Holden's possibly because Winter looked at data over the whole gait cycle and was not as strict with his selection of speeds. Also Winter normalised Power with respect to weight whereas Holden did so with respect to the product of weight and height. Therefore the results possibly indicate that subjects choose a greater variety of ways of producing similar knee angles at slow speeds than for fast speeds. Indeed Holden's report claims that at slow speeds the gait patterns could not be classified with confidence as "normal". They conclude the possible existence of different gait patterns at slower speeds.

From the ensemble average of power at different speed groupings, it is difficult to see a large temporal variation in the peak bursts. This matter possibly needs closer inspection.

2.2.2.2.3.3 Hip Power

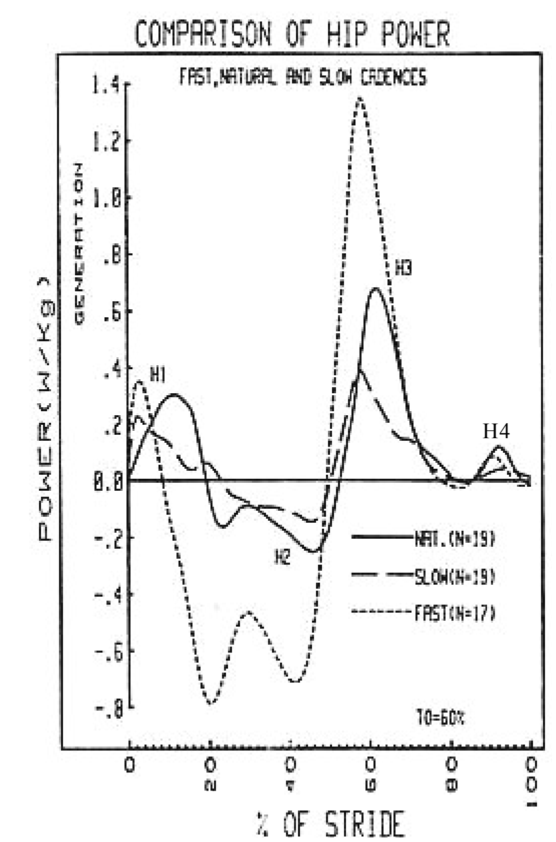


Figure 2-22 - Ensemble average of Hip powers taken at different speed groupings, adapted to include the H4 phase from Winter [222].

The hip power pattern [Figure 2-22] for non-pathological subjects has four main phases, which Winter labels H1-H4 [225]. He does however omit H4 in his later publication in 1987 [222]. This omission may be due to its unreliability especially at slower speeds.

The first phase (H1) generally consists of a small power generation burst, which occurs just after initial contact. Although this is visible in the data presented by Winter, and Chen [42], it is not so well defined in other studies such as by Bresler [31]. As mentioned in the kinematic section the hip reaches its maximum flexion (H5 as labelled by Benedetti and not to be confused with Winter's notation) just before initial contact. The hip is in a process of re-extension after this period. On initial contact, the ground reaction force passes just anterior to the hip joint creating a flexion moment across the hip [Figure 2-1]; therefore, it is likely that the power-generating burst H1 is used to counteract this action.

After the H1 burst initiating the hip extension, there is an absorption phase (H2) in which the extension of the hip appears to be regulated. The shape of this power phase seems to be fairly inconsistent even between data taken on the same subjects for example at different speeds. In fact, the only point of consistency seems to be that this is indeed an absorbing phase; there is no consistent feature which is readily distinguishable. The H2 phase seems to end just before the maximum hip extension about the start of double support where it is rapidly replaced by a positive burst of power. This is the most significant and obvious phase in the hip power and is labelled as H3. Its function is to possibly assist with the energy required for push off since the hip undergoes a large amount of angular acceleration at this time [Figure 2-9, pp. 35]. This ultimately acts to drive the leg forward although there is a short period before the shank catches up with the rotation of the thigh. After this large burst of energy, the hip is mostly left to passively swing although there is a small positive power burst at the end of the swing phase observed in some subjects.

Only the H3 phase shows a very obvious correlation with speed where there is a large increase in the peak with increasing speed. The increase according to Winter's data is only slight for the H1 phase and there is only a change in H2 for the fastest speed. The relationship with speed for the individual phases is therefore not very well documented. Only a relationship of the mean hip power with speed can be drawn up e.g. Chen et. al. [42] who showed a positive correlation with r=0.72 (Y= -0.21 + 0.55X).

2.2.3 SPECIFIC OBJECTIVES OF WALKING

Once gait had been observed and measured as described above, researchers wanted to *specify* the objective of these gait patterns. If a controller that enables walking is to be designed, then these objectives should also be fulfilled.

For example, Winter described walking as an integration of three sub-tasks which were:

- support against collapse during stance.
- dynamic balance of upper body during stance
- anticipatory control of the foot trajectory to achieve safe ground clearance and a gentle foot contact.

Another description of gait is made by Cappozzo [36] who states that 'the objective of walking is to displace the body through space in the most reliable way, obeying the relevant aesthetic canons, while satisfying a set of mechanical constraints typical of this locomotor act'. The author goes on to say that during walking the head and trunk are thus subjected to mechanical stimuli of a vibrational nature, which can be connected to the following hierarchical problems:

1. maintenance of balance.
2. maintenance of strain on tissues below a suitable level.
3. preventing energy expenditure becoming too high for the task.

He then concludes that if these are deemed to hold true in normal conditions, then the adaptations to abnormal conditions (pathological gait, amputee gait) may also obey these criteria.

The third criterion was studied by Saunders et al, who tried to show how energy is optimised through energy transfer and minimisation of the displacement of the centre of gravity [177]. These optimisations for the centre of gravity were described by the six determinants of gait (note 3, 4, and 5 act to adjust the effective leg length):

1. **Pelvic Rotation**: When the hip flexes or extends, the trunk will be lowered or raised. Therefore, if the hip flexion/extension can be minimised then the trunk's vertical displacement will also be minimised. This can be achieved by rotating the pelvis about the vertical axis so that one leg is brought further forward and the other further backward hence the hip will not need to flex or extend respectively as much for the same stride length.
2. **Pelvic Tilt**: Flexion and extension of the hip will cause the hip joint to rise or fall as mentioned above. If the pelvis were to keep level the trunk would follow this movement up and down. Therefore, to minimise this, the pelvis tilts side to side so that the swing leg is lowered at the highest point of the stance leg. The average height of the two hip joints therefore describes the height of the trunk; the displacement of this height is therefore reduced. The effect of the tilt however means that the swing phase leg has a shorter height clearance which has to be accommodated usually by bending the leg more. A control system that is devised must also allow for this tilt by allowing the knee to sufficiently bend at the appropriate time.
3. **Knee flexion in stance**: After heel strike as the hip goes from flexion into extension, the hip joint would rise and fall if the leg remained straight. Therefore, by flexing the knee the height at the apex of the hip trajectory is reduced.

4. **Ankle Mechanism**: At heel strike the effective length of the leg is increased because the heel juts out beyond the ankle. This raises the curve at the beginning thus slightly flattening the trajectory of the joint.
5. **Foot Mechanism**: Similarly to the ankle mechanism, the leg is lengthened at the end of the stance phase by using the forefoot and thus raising the trajectory at the end.
6. **Lateral displacement of body**: To reduce the lateral displacements of the centre of gravity whilst walking a narrow base can be used which requires little lateral movement to preserve balance. This is achieved by the thigh being angled medially so that the width from knee to knee is narrower than that from hip joint to hip joint. Such adjustments are also often made by prosthetists.

The combined effect of all these six determinants is to produce a much smoother trajectory for the centre of gravity and thereby a reduced energy expenditure.

Rose and Gamble cite two basic requisites for walking [169]:

1) Continuing ground reaction forces that support the body
2) Periodic movement of each foot from one position of support to the next, in the direction of progression.

These elements are necessary even in pathological gait and would have to be maintained if prosthetic and orthotic devices are used

2.3 SUMMARY

To conclude - there are steps in a design process that can be derived from this literature. Fundamentally, *stable patterns* are required if one hopes to establish from a control point of view what the objective is. Is there a generic pattern that represents a task for control e.g., what makes walking a unique movement pattern from all other patterns[6]? Indeed, at the observational level, something intrinsic allows this to be recognised as a repeated behavioural pattern known as walking.

[6] This question is so fundamental that it will pervade into the following chapter

The overall behaviour can even be defined by consistent relationships such as the cadence/stride-length ratio with speed. However, for control purposes the underlying principles need to be established so that a model can more fully describe the whole movement. Although clues are available such as the free speed of walking seeming to coincide with minimum energy expenditure, there is no consensus as to exactly what is going on in order to produce a full accurate model.

Therefore, it remained necessary to somehow identify stable patterns and pick a suitable set of variables to describe the behaviour. It was recognised that a systematic approach was needed to make comparisons between data sets – i.e. so that "like with like," was compared. Therefore, normalisations in both the amplitude space and the temporal space must be carefully considered to ensure appropriate intra-subject and inter-subject comparisons.

In control terms, this meant that stable patterns most representative of neural behaviour were required whilst effects due to mechanical advantage reduced. Therefore, normalisation should reduce variance due to differences such as height and not to skill related factors. Towards this, Pierrynowski and Galea [161] showed which normalisation methods were able to reduce variance in the amplitude of data. Unfortunately, the literature rarely adheres to this convention so that more awareness of intra-subject and inter-subject variability was needed. In contrast, the temporal dimension has been conventionally normalised according to the repetition of an observational event (commonly heel strike). The assumption then is that all other actions within the gait cycle must occur at the same proportion relative to this event. This author however could not advocate this method for control purposes because it could not be assumed that everything is linearly dependent on this scale. Indeed, it is most unlikely when considering, for example, that the stance/swing ratio changes as walking speed changes. In control terms, it would be better to make normalisations according to kinetic events than kinematic events since kinetics variables are associated with how the movement is caused.

Despite these concerns, the literature has shown consistent biomechanical patterns for walking. Angulations exhibit sufficient low variability for repeatable inter-subject gait parameters to be identified by various researchers such as Benedetti. The ankle patterns exhibited the greatest variability whilst the knee and hip were consistent. Some morphological changes (both amplitudinal and temporally) would occur with speed, although the basic features were still present. Kinetic data in contrast exhibited far more variation. The hip and knee in particular had the largest variations especially for inter-subject data at varying speeds. This was thought to be indicative of the neural system trying to control movement by varying the moments and powers in order to provide well behaved kinematic patterns. Despite this, the profiles were not chaotic and features were still evident in the kinetic data.

Overall, the literature gave optimism that there was sufficient structure in human movement to exploit for control purposes, and careful thought has been given as to how this should be assessed.

3 REVIEW 2-CONTROL THEORIES: FROM *NATURAL* TO ARTIFICIAL

3.1 INTRODUCTION

If an artificial knee mechanism is to be "actively controlled," there will be two controlling systems at work, the natural, and the artificial. The design process of the artificial controller should therefore take into account both of these controlling mechanisms so that they work in harmony rather than oppose each other's actions. An appreciation of the amputee's natural control system is thus essential so that a complementary control strategy can be developed. As an aid to development, a review of the main topics in biological control relevant to this project has been conducted, including neurophysiological views on movement control and schemes based on observation of kinematical and kinetic data.

Inevitably, therefore, a comprehensive review of Nicholas Bernstein's work has been made, as he pioneered many issues in the field of human movement control, and influenced subsequent researchers. His approach has been to hypothesize various methods of control and then apply deduction from experimental evidence to elucidate the most likely method and why.

Other studies pertinent to human movement control are also reviewed, including the role of the central nervous system and arguments for the possible existence of central pattern generators that may create the basic rhythmic nervous pulses for muscle control. Finally, other theories such as the "Equilibrium End Control" try to ascertain how a motor plan for movement may be carried out, and its rationale of control. For example whether there is a coding explicitly describing joint position, velocity, or maybe muscle tension, etc.

3.1.1 WHY DO WE NEED TO KNOW ANYTHING ABOUT HOW THE NATURAL SYSTEM COORDINATES MOVEMENT?

It has been possible to build mechanically controlled prostheses without much in-depth thought as to how the musculoskeletal system is being controlled to produce particular behavioural patterns. New designs are usually based on incremental improvements of previous designs, which are often instigated by practical considerations such as when amputees seek better performance of a particular characteristic. For example, the leg doesn't swing through fast enough to accommodate the preferred speed of walking, or it is awkward to flex during critical periods, etc. Practically speaking, one could continually try to solve each problem ad hoc, and hope one day to arrive at a prosthetic design that closely emulates the behaviour of a natural leg.

In contrast to what may be termed a trial and error approach, this project will concentrate on ultimate objectives by making as much use as possible of the previous work on co-ordination of movement.

3.1.2 WHAT ARE THE KEY QUESTIONS IN MOVEMENT CONTROL?

A person is able to achieve a behavioural goal under variable environmental conditions by using an "equivalence class" of movements; this was defined as an "action" by Green in 1988 [85]. Specific movement patterns can be observed in order to achieve particular tasks (this is especially the case for automated movements such as walking, which largely occur through subconscious control). Researchers continually speculate how the central nervous system is able to produce such a family of movements that take into account variations such as the speed of the movement, along with external perturbation forces.

The problem has been approached from different levels; for example, using downstream research strategies where the problems of motor control are defined at the higher levels, and upstream strategies that use a reductionist approach where understanding is gained from the functioning of neuronal circuits. At the higher levels, there exists the possibility of motor engrams[7] within the spinal circuits and their interaction with reflex mechanisms. At lower levels, researchers have considered the musculoskeletal response to central nervous innervations – how much of the observed movement is attributable to the dynamics of the musculoskeletal system and nervous system (e.g. rates of conduction and action potential frequency

[7] Pre-established motor patterns that are stored like a memory.

limits). This review will try to bring together some of these ideas to show how and why the control strategy for this project was developed.

3.2 BERNSTEIN'S THEORIES:

3.2.1 SUMMARY OF BERNSTEIN'S WORK ON MOVEMENT CONTROL

Nicholas A. Bernstein, during the period 1934 to 1966 addressed many of the topics faced in motor control. Because of the extent of his work, the results of which remain relevant to this day, only a flavour of his ideas can be reviewed although they give good foundation to the design process of a prosthetic knee control system.

He struggled throughout his life with trying to bring a mathematical construct to the science of movement. In the field of bioscience, this is particularly difficult because unlike other fields, such as physics, where it is sufficient to ask "… *How does a phenomenon occur,* and secondly *Why does it occur in exactly this way";* biology requires a third question "*to what purpose [24]?"* Such formulations within the biological field are thus very rare. For example, Bernstein singled out Darwin's stochastic model in which survival always works as a universal control and steering mechanism in the evolutionary process; this formulation lends to automatic selection of beneficial hereditary carriers.

Bernstein in this vein concerns himself with notions of what constitutes biological activity, what are its Goals? He suggests – "the key functions are the **co-ordination** of movement, the **planning** of movement and the **exploration** of better optimal movement patterns." He does not see Darwin's model as a goal for biological activity because, as he puts it, "Animals don't do things because they want to survive, but because they are hungry, thirsty, sleepy, and so on." On a fundamental level, he asserts that instead the organism strives (by reproducing successful experiences) to minimise **entropy**. This is fundamentally different to how inanimate objects are governed where the second law of thermodynamics works to increase entropy, i.e. an isolated system tends towards disorder, whereas a living system works against this by diminishing entropy within itself whilst increasing it within the environment. Such notions lead to the idea that the organism **conquers** the environment and does not merely passively follow the environment. Using the simplest example of action, he contrasts his view of the role of reflexes to Pavlovian theories [156] with deterministic theories on cause and effect (i.e. the mechanical stimuli being the cause, and the observable reaction the effect). In the Pavlovian sense, the organism will end in the "situation where the environment *pulls the ropes,* the actions of the organism forming an uninterrupted chain of responses to the stream of influences from the environment". Therefore, life would be in equilibrium with the environment, which is wholly different to conquering the environment, " … the organism is rather controlled by the model of the needed future". One must view "to what purpose" is the reflex, in order to understand the actual basis/genesis of the phenomenon.

He expands upon these arguments by proposing that the goal may be seen as a **coded model** in the brain for the desired future of the organism. This allows for motivation and the goal directed mechanisms, which can all be encompassed as "activity". Such coding then also moves into the realms of bio-cybernetics in which the theory of communication and control within such systems (living or machines) is studied.

In trying to resolve such issues, Bernstein spent much effort on coordination and localization, trying to understand the relationship between the central command and the peripheral effect. He examined whether co-ordination could be explained in terms of localisation of central processes, whether peripheral effects can be determinately linked to localised central processes.

Such investigations have led Bernstein to develop many theories and constructs, so that he is associated with the following areas:

The degrees of freedom problem: this is a basic statement alluding to the overall problem of motor control for the central nervous system. The number of degrees of freedom of a system may be defined as the number of parameters that are necessary to fully describe its state. The musculoskeletal system is composed of many linked segments that have even more muscles spanning across them, and each muscle group consists of many motor units each one of which is innervated by its own motor neurone. This complexity means that it is possible to have many ways of achieving a movement task and there is often no unique solution. To move one's hand from one place to another can be achieved using a variety of combinations of shoulders, elbow and wrist rotations, not to mention recruitment of different motor units within a particular muscle group, thereby a large variation in activation patterns results. This problem was effectively defined by Bernstein where he stated, “that the central nervous system has, at its disposal, redundant[8] degrees of freedom to achieve a particular task” [21]. There are effectively more degrees of freedom to be specified than are observed by the movement. How does the CNS therefore choose from this multitude of possibilities? Bernstein tried to see how the CNS might simplify this problem by reducing the number of degrees of freedom it needs to control. This reduction was imposed using constraints.

The effect of constraints on Movement dynamics: Skeletomotor controllability and observability may be achieved through constraints. Bernstein described agencies in the CNS whose task was to establish known dynamical regimes for different motor tasks. Regimes would have to be known in advance of the commands being sent to produce the movement. They will also take advantage of biomechanical external events e.g. reaction couples, and will act to minimize the number of equivalent degrees of freedom... Constraints may be mediated by:

- *Structured efference:* in which efferent signals directly constrain groups of muscles for particular tasks. This simple premise can encompass concepts such as motor synergies and oscillators, pattern and function generators, and other coordinative structures.
- *Structured afferents:* in which reflexes may directly seek to limit the movement to specific patterns. The structure and positioning of the sensors will also constrain the reflex response.
- *Performance criteria:* where the ideas of cost and achievement of a movement are considered. The existence of cost functions of the skeletomotor state was posed, which would also require searching

[8] The Russian word izbytochnost can mean redundance or abundance; Gelfand and Tsetlin (1966) considered that motor abundance was a more apt description since DOF are not really eliminated and were able to further develop Bernstein’s ideas from this point of view [81].

mechanisms to achieve these cost functions. The searching mechanism may itself act to constrain the movement dynamics.

Models of future sensory outcomes of an action may therefore be utilized to refine motor activity.

Hierarchical theories of motor coordination: because the computational task of producing appropriate innervational signals to the periphery for every motor pattern including the effects of external disturbances was considered so immense, hierarchy was used to try to simplify the problem. If one accepts the ideas of localisation in that mutual activity of "interneurones" (which in this case may be viewed as sub-processes) can act to produce a particular function e.g. flexion at a joint, then the whole movement must consist of the co-ordination of all such sub-processes. Therefore, instead of trying to bridge the gap between the motor problem and the innervational signals to muscles in one goes, he proposed progressively narrowing the gap by hierarchical levels. The top level may formulate and translate the motor problem in simpler terms ready for the next level, all the way down to the muscles. This has the advantage that the top level no longer needs to specify what each muscle is doing. This acts more as a conductor to the whole movement without requiring knowledge of how to perform each element. For more information, see section 3.2.3, pp. 82.

Topological motor/visual fields and Theory of Equal Simplicity: Bernstein observed that the output representation of many motor patterns was best described from a topological point of view, rather than in the metrical sense. Topological measures refer to the qualitative peculiarities of a movement, whereas metrical properties will give quantitative descriptions according to size, length, angles etc of the movement in space. A more detailed description of topology is available in section 3.2.4 pp. 86.

Bernstein's idea on the theory of equal simplicity is also felt worthy of mention - if nothing else but to encourage thoughts on the problem of movement control. The theory of equal simplicity, to this author, was Bernstein's way of analytically trying to unravel what may occur within a "black box", as it were, without seeing its contents; only the output of the box is observable. He illustrates the concept by using a circle of radius R drawn on paper as an output sample from some black box; he questioned how this might have been achieved. Bernstein thinks of three methods with which the circle could have been drawn – 1) using a template, 2) a compass, and 3) an ellipsograph. All instruments are therefore capable of drawing a circle, which is a curve of second order described by radius (r) and eccentricity (e). With the first method however, neither radius nor eccentricity can be varied, but it is a very quick method of drawing a circle of radius R. For the second method, only the radius can be varied but is accordingly a more complex process; and finally the third method allows both radius and eccentricity to be varied but is the most complex process to set up and use. Bernstein designated the ease of the process according to its "degree of simplicity" (S) e.g. as measured by time to complete the task. Then, for all of the structures, simplicity could be written as a function of r and e, so that $S = F(r, e)$. For example, for the template $F(r, e) = 0$; and $F(R, 0) \neq 0$. In this case, simplicity will change abruptly when the radius or eccentricity becomes R and zero respectively.

All of the three structures are capable of producing the same operation; the difference between them "*is in the form of the function S*". If there is a sudden jump in S as transitions are made from one element of the set of possible tasks to the next - then the structure is unlikely to be capable of much adaptation. For example as the radius is increased incrementally (as with the template). Conversely, if the degree of simplicity is maintained throughout a particular set of transitions, then it is likely that these transitions are closely related to the structural scheme of the device. Bernstein refers to such periods as "lines of equal simplicity" and describes them as "… corresponding to those directions (within the range of possible elementary processes) along which movement does not involve any change either in the structural principle or in the principles of operation".

Bernstein conceived these ideas as a tool to probe the workings of the central nervous system where the variable S may be speed, accuracy, variation etc. He gives, as an example, a circle described with arm directly to the front, then directly out to the side, and then some intermediate axis [19]. For each situation, the innervational schemes and the muscles will be different. Each movement is however made with roughly the same difficulty, accuracy, and variation. Thereby, Bernstein concludes, "… the structure of the central complex governing a given series of movements will be more closely related to spatial form than to muscle scheme". All the variations with spatial form had the same degree of simplicity whereas muscle innervation did not since one does not learn a new schema for every possible variation of the circle. This spatial form is once again better expressed in terms of topological properties.

In the simplest sense, one could construe from these ideas that an artificial controller of knee behaviour should ideally also enable differing movement tasks with the same degree of simplicity. In effect, the control system should be expandable to other tasks other than the demonstration task of walking by using the same methods (see aim 5 later on [section 4.2]). The deeper objectives desired through the theory of equal simplicity remain elusive however since the possible method of operation (used by the central nervous system presumably) has not been fully conceived. Bernstein however has left a tool for researchers with which to test models to see if they obey such observations of equal simplicity. Until this comes to fruition, there is no obvious controlling system to use for this project, however a general feeling of the problem is realised.

More detail is now given on some of the issues just raised.

3.2.2 THEORETICAL CONSTRUCT FOR CENTRAL NERVOUS SYSTEM CONTROL OF MOVEMENT; (BERNSTEIN'S PRINCIPLE OF INNERVATION).

Bernstein speculated on what was possible for an unknown control system and then from experimental data tried to narrow these possibilities.

He theorized that the degree of *tension* in muscle (F), which generates a moment about a *single* joint that it spans, is a function of:

- Its level of innervation[9] (E), based upon its (tonic[10] and tetanic[11]) condition

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[9] Innervation refers to the level of nervous excitation in order to recruit muscle groups.

[10] Tonic condition refers to continuous muscular contraction.

[11] Tetanic condition is the fusion of a number of spasms into an apparently smooth and continuous contraction.

- Its length at a given instant, which is itself dependent on joint angle (α)
- The rate of change in length (i.e. rate of muscle contraction), which must also be some function of $\frac{d\alpha}{dt}$.

So that

$$F = F\left(E, \alpha, \frac{d\alpha}{dt}\right)$$

For rotating bodies, the total moment M is proportional to the angular acceleration, so that

$$M = I \times \frac{d^2\alpha}{dt^2}$$

where *I* is the moment of inertia of the body

and $\frac{d^2\alpha}{dt^2}$ is the angular acceleration of the body

If the effect of gravity on the limb $G = G(\alpha)$ is also included, the total moments caused by muscle tension and gravity will lead to the following equation:

Equation 1 $$I \times \frac{d^2\alpha}{dt^2} = F\left(E, \alpha, \frac{d\alpha}{dt}\right) + G(\alpha)$$

This equation is fundamental for describing the movement of a single limb in a gravitational field under the influence of a single muscle group. The key to Bernstein's hypotheses was to try to understand what governs *E*, what is it dependent upon? For example, is it dependent on α or $d\alpha/dt$, or is it completely independent of α and simply changes as a function of time? Bernstein hence uses various mathematical constructs to hypothesise possible roles of *E* and how the central nervous system can produce such results.

Since Equation 1 is a second order differential, it can be solved by integration as long as F and G are known. However, a unique solution requires initial conditions governed by the angle and angular velocity, i.e. α_0 and $d\alpha_0/dt$. Different initial conditions will result in different patterns even though the governing rule is identical. This implies that knowledge of initial conditions is necessary for control of movement to be accurately maintained; otherwise, the movement pattern cannot be determined.

Considering now the first form of the function E, where it is a function of α and $d\alpha/dt$, Equation 1 can be written as $I \times \frac{d^2\alpha}{dt^2} = F\left(E\left(\alpha, \frac{d\alpha}{dt}\right), \alpha, \frac{d\alpha}{dt}\right) + G(\alpha)$, so that the solution will depend only on the initial conditions. Once these conditions have been met, a movement must occur and will continue unremittingly. There is no need for CNS interference as the amount of innervation will simply depend on the current state of the limb as described by α and $d\alpha/dt$. The innervation E, therefore, provides nothing more than what a system obeying a set of mechanical laws may do. This kind of response is analogous to the condition of central paralysis of the limb, where only reflex reactions may affect movement.

The other extreme view is to assume that E is a function of time only and once initiated will elicit a predetermined time-variant innervational pattern. The innervation E(t) can not react to any unwanted positional changes, as it is independent of such information although there will be some passive mechanical response due to muscle tension, which is said to be some function of α and $d\alpha/dt$. Bernstein notes that such a condition where the innervation is only a function of time may resemble proprioceptive ataxia, where an individual is unable to hit targets and staggers almost in a drunken fashion.

Because of the physiological implausibility, for non-pathological individuals, of the two extreme positions described above, Bernstein hypothesised that the "excitation (E) of a muscle must be both a function of time and a function of position and velocity" resulting in an equation of form:

Equation 2 $$I \times \frac{d^2\alpha}{dt^2} = F\left(E\left(t, \alpha, \frac{d\alpha}{dt}\right), \alpha, \frac{d\alpha}{dt}\right) + G(\alpha)$$

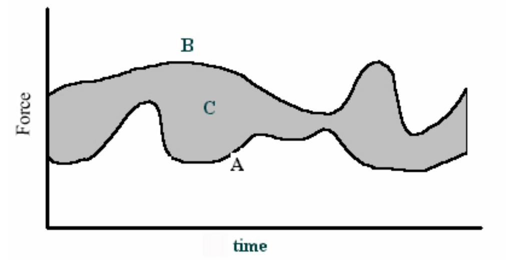

Figure 3-1; Diagram showing the force created by an innervational pulse that is necessary to bridge the gap(region C) between the force needed to produce a movement (B) and the resultant external forces (A) applied at the joint [220]

If *E* is a function of time only, then it is not possible to map the movement following an innervation *E(t),* because of the interplay of external forces and variations in initial conditions. However, for the combined case, a determinant effect is possible since E will vary with initial conditions and external perturbations. Bernstein uses a simple diagram to show one curve (B) that represents the forces required to produce a particular limb movement and another curve (A) representing the resultant external forces on the joint [Figure 3-1]. The vertical distances between the two curves A and B therefore represents the additional force required to make up the deficit of curve A and will be generated according to innervation by the central nervous system. A purely determinant E cannot be known without knowledge of the external perturbations since the innervation will have to complement the external influences. An artificial knee system will therefore also affect this dynamic and consideration must eventually be given to its contribution.

During rhythmical movements, such as walking or striking with a hammer, Bernstein claims that the course of the movement can be accurately described to within a few millimetres by a Fourier series of the third or 4^{th} order [22, 23, 20]. He notes “the external expression of co-ordinational activity provides a picture of such a high degree of reciprocity … it is therefore possible to reconstruct from a small section of such movement the entire movement as a whole with the same order of accuracy as the sum of four sinusoids …”. Take striking a hammer for example: if the period of the movement is 1 second, then the whole movement must have been organized at least one second beforehand and it is possible to predict the remaining movement from only a small section of the movement. Furthermore, if the hammer is struck from a different angle, the accuracy of the spatial interaction remains, while the innervation has changed slightly. Therefore, differences in innervation may also be caused by slight changes in the movement task variables as well as external influences. The question remains: How is the central nervous system able to create such different patterns of innervation for apparently the same task and how does it (or how might it) decide what curve B should be?

Such arguments form the basis of natural control theories on movement, which try to determine how much of a movement is pre-emptive, (i.e. in a feed forward sense) and how much is reflexive adaptation to the environment, (i.e. in a feedback sense). The following sections discuss some of the theories that have been developed - sharing more details of these feed forward / feedback mechanisms.

3.2.3 CONTROL STRUCTURES, HIERARCHICAL OR NOT?

Only the structure of the final innervation pulse E to a group of muscles has been discussed. However, it has been known from early days that the innervation does not result from just one centre within the central nervous system. Rather, there are several centres capable of producing innervational signals that can result in the same movement [25, 77]. For example, Bernstein notes that flexion and extension of a joint can be achieved through both the pyramidal systems (corticospinal tract) and the striopallidal group of nuclei[12]. A nerve impulse E can therefore be the result of many pulses from different parts of the brain, in other words, an individual muscle is not controlled by just one pool of interneurones; there being many different regions capable of producing flexion and extension movements using the same group of muscles. Bernstein states that: "all the clinical observations agree that these central nervous subsystems have one and the same object of excitation at the periphery, it is not the objective, but the manner of excitation that is specific." Therefore, the question is not so much whether individual areas can produce innervational pulses (as demonstrated in many pathologies) but how they all interact with each other in a normal situation to produce the observed movement patterns. Very complex coordination would be required between the different centres in order to produce the responses that are observed. Such considerations will have direct relevance to this project since there must also be a large degree of coordination between the artificial and natural controllers.

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[12] Including striatum-gl. pallidus-nucleus, rubber-tr., and rubro spinalis

Bernstein speculates upon the complexity of such co-ordination and how each innervational scenario might arise, i.e. when E is a function of time, α, and $d\alpha/dt$. If the final innervating pulse E is simply a function of time, then it is not difficult to envisage each centre producing its own innervation $E_n(t)$, and if these innervations are closely co-ordinated, they could unite and merge into the motor nerve to give E(t). Under this scenario, all the sources would have to be closely synchronized.

The construct for including the α and $d\alpha/dt$ dependence however was more difficult, and he could only envisage a simple solution in this situation if the proprioceptive input is referred to just one of the effector centres of the brain, e.g. the cerebellum; the other centres only responding to E(t). Then the output innervating pulse could be expressed as

$$E\left(t,\alpha,\frac{\mathrm{d}\alpha}{\mathrm{d}t}\right)=E_1(t)+E_2(t)+...+E_n\left(t,\alpha,\frac{\mathrm{d}\alpha}{\mathrm{d}t}\right).$$

Bernstein dismisses this equality, however, because there is no guarantee that it would be a simple series summation and not some other function of $E_1 \dots E_n$. It also appears that the afferent pathways directly reach two central nuclei (cerebella and thalamic), raising the possibility of the existence of proprioceptive multiplicity to the centres, which *would require the closest possible co-ordination between them* (an observation that should not be missed when conceptualising the structure of the artificial controller).

As a result of such reasoning, Bernstein thinks that such simple schema are not effective in describing the workings of the central nervous system, and that the problem is further complicated by the likeliness for multiplicity of efferent pathways. What this implies is that the localized response, as suggested by Foerster [77], who stimulated various regions of the brain and reported localised peripheral responses to them in the form of specific motor synergies, is very difficult to envisage - although still possible. Bernstein sees far more interaction than would be suggested by these experiments, and holds the view that if localization does exist then the categories that are assigned to various centres would be far more fundamental.

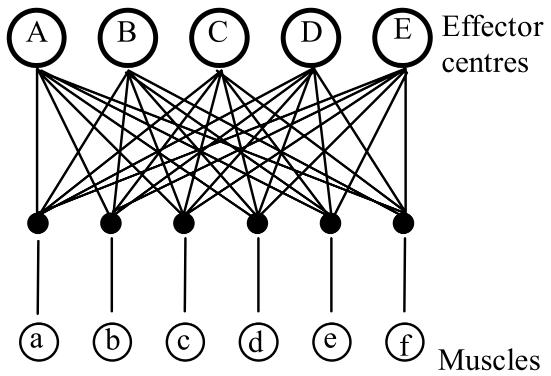

Figure 3-2 – Adapted from Bernstein [21]. Diagram to illustrate the Multiplicity of efferent pathways. Effector centres do not just produce a localised response (i.e. to one muscle group), but somehow combine their outputs to result in a coordinated movement.

The multiplicity problem can be illustrated with reference to Figure 3-2. Effector centres do not simply produce localised responses (e.g. a centre controls flexion of one muscle group), but its output will combine with other centres to facilitate several muscle groups. Co-ordination must occur so that all the impulses taken together produce a common action. The individual effects of each pulse Aa, Ab, … Ba, Bb, are less important than the combined effect of all the effector centres.

Having highlighted the efferent multiplicity problem where the central nervous system has to combine all effector centre signals to produce the final signal E, Bernstein was then faced with complications on how to handle the time dependence of the signals. His solution was to refer to what he called *engrams of movement*. His initial observation was that movement in the spatial sense is homogeneous, i.e. there is a similarity in the structure of movements. He then reasoned that this should also be true in the temporal sense and therefore not a result of external factors, but originate in the central nervous system. In other words, the accurate description of rhythmic movement patterns by Fourier series means they cannot be derived from any external influences, as these do not have the same/regular harmonics, which leaves the central nervous system as the controlling factor. The central nervous system must already have an exact plan or programmes of movement, which Bernstein refers to as an **engram**. The engram must contain the entire course in time of the movement. This is supported by the existence of habits and automated movement that must be the result of some pre-programmed plan and not just created on the spot as it were. If a particular movement were composed of a number of engrams, these would necessarily have to appear in their preset order, and should not occur simultaneously. Similarly, their onset and duration would have to be controlled, so that the whole movement obeyed a particular tempo (rate) and rhythm.

There would have to be determinant time intervals between the activation of one engram and the next. The periods do not necessarily have to be the same between all successive engrams, although a relationship is likely with their duration. For example, if the period of the entire movement increases, then the duration of each of the different engrams may increase, although not necessarily by the same proportion as each other.

Having postulated the existence of engrams, the possible method of coordination between them could have two structural possibilities consonant with the aforementioned logic. Figure 3-3 illustrates the two basic structures, which are described below:

Chain Hypothesis: Here, each engram follows the previous one in a serial manner, the trigger for the start of each engram effectively being the end of the previous one (for example by the use of proprioceptive signals).

Comb Hypothesis: Here the engrams are not successively linked to each other, but rather have some external controller regulating when they should be activated and their duration.

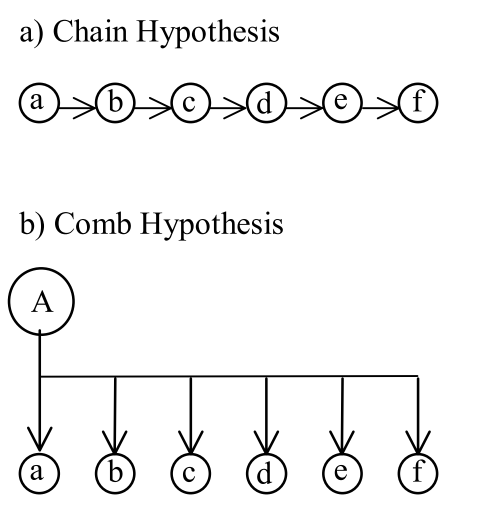

Figure 3-3: Diagram showing two possible methods of coordinating/linking engrams using: a) the chain hypothesis, and b) the comb hypothesis.

The chain hypothesis is reliant on peripheral signals so that the α and $d\alpha/dt$ dependency of the innervational signal play the dominant role. The temporal dependence of E, which must also be included, will be a natural consequence of the time taken for completion of each engram. The tempo and rhythm will in effect be determined according to events at the periphery. The chain hypothesis also implies that the order of engrams must be maintained as an engram can only be triggered by its immediate predecessor.

Whichever hypothesis may be correct, they must both, in some way, produce the whole movement by generating the entirety of the force, e.g. curve B [Figure 3-1].

The chain hypothesis however, does not easily account for how the innervational pulse E adjusts to external perturbations and maintains a consistent movement pattern. For this to occur, the *whole* movement should possibly be known, but there is no guiding principle for the whole movement with the chain hypothesis. The comb hypothesis, although more complicated, circumvents such a problem because it puts more emphasis on the E(t) dependence. Then proprioceptive influences can act as "correctors to the general whole". They will not have such an influence on the temporal aspects of the whole movement as in the chain hypothesis.

An engram may be considered an image of the final movement to be performed and it represents what should eventually occur at the periphery. The innervation pulse (shaded area C), according to Bernstein, must therefore be coded for, at a lower level than the engram. The final movement pattern is then decoded once again at the muscular level. Under this paradigm, there is some hierarchical order to the central nervous system that codes the original representation of the movement, before it is decoded once again to produce the movement. Bernstein expands upon this theory for the comb hypothesis by postulating that the engrams would only be time dependent E(t). Whilst encoding to produce the innervational pulse there will be an influence of proprioceptive signals to complement the pulse with α and $d\alpha/dt$ dependencies too i.e. to give $E(t, \alpha, d\alpha/dt)$.

3.2.4 TOPOLOGY FOR INSIGHTS INTO MOVEMENT CONTROL

Bernstein also recognized the importance of topology[13] in movement control, by arguing that the central nervous system is more adapted to respond to topological information than to metrical information[14]. Topological properties of a two dimensional figure may include things such as whether the object is an open or closed loop, or whether it has intersecting lines; such distinctions may arbitrarily be classed as zero order properties as they are so fundamental. Other topological

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[13] The study of qualitative geometrical properties and spatial relations that are unaffected by size or a continuous change of shape. It concerns features that can be assigned various topological classes, and allows for recognition based on such classes.

[14] In this context, refers to measurements based on size, length angles, etc.

properties may be related to the numbers of something (note not measurements). A star, for example, may be said to have five vertices. Such additional distinctions may be represented by higher order topological properties describing increasing detail. For example, individual letters may be classed separately and are recognisable from each other. Because symbols are simplistic, the distinction between letters may exist within the first order topological properties. In this case, all versions of a particular letter, e.g. capital "A", will be segregated by this first order property. All As, **irrespective** of size, width, font, symmetry, neatness etc. should then be recognisable.

It is believed that the central nervous system must also act on such topological information, for example as demonstrated by a keen ability to recognize objects, regardless of their metrical properties (we do not measure the lengths of the sides of the letter to see if it is the letter A). In fact, we are capable of making much more complex recognition tasks involving topological properties of much higher order. Bernstein cites an example of a child's ability to differentiate between a cat and dog, not so much according to "anatomical differences such as the structure of teeth and claws but on the general appearance", i.e. topological differences almost certainly of higher orders.

There is ample proof of the significance of topological properties on perception. The relevance to this project is that aspects of movement control are also accomplished with regard to topology. We are all capable of writing the letter "A" in various degrees of penmanship (all of which are recognizable as "A"). However, it would be very difficult to reproduce an exact match, for any particular one (all attempts at writing the letter A will be slightly different in size for example). That other people are capable of writing patterns that can be perceived as the letter "A" implies a replication of at least the first order topological properties of the letter via neuromuscular control. Higher order topological properties may similarly enable recognition of different individuals' handwriting.

What is further intriguing about the neuromuscular production of patterns with specific topological properties is that they do not depend entirely on position or orientation. For example, whether writing on a desk or a blackboard the pattern is largely unaffected and still distinguishable as the individuals' handwriting.

This can be observed for CNS control in the motor field by our ability to write with the same distinguishable handwriting at most practical orientations e.g. at a desk or on a blackboard. This means that the central nervous system must be able to control different sets of muscles (as is required by the different positions) to produce the same topological outcome. These ideas can be extended to other automated movements, for example, piano playing whereby the keys can be struck accurately whether sitting or standing.

3.3 FROM OBSERVATIONS OF RHYTHMIC MOVEMENT PATTERNS TO CENTRAL PATTERN GENERATORS AND AFFERENT INFLUENCE

3.3.1 WHAT CONTROLS RHYTHMIC MOVEMENT?

Even before the 20th century, researchers have speculated on how the central nervous system controls movement and for what specific purpose. It was not even obvious that the brain controls such actions; observations of decapitated animals (including humans during executions) revealed that some movement persisted within the body despite the brain having been detached. Pioneering researchers such as Brown and Sherrington [35, 33, 34] conducting thoughtful experiments on animals began to delve into these mysteries. Using myographical techniques[15] to investigate the movement of muscles in response to stimuli, they concluded "that the rhythm of progression is of central origin" and that movements are determined by a balance of flexors and extensors. The focus of Brown and Sherrington's work became the investigation of how the central nervous system controls these muscle pairs and at what level. Brown's experiments [34] looked at the response that remained when different levels of the central nervous system were cut off. For example, *decerebrate* preparations were made where everything above the cerebellum was transected (leaving the cerebellum, brainstem and spinal cord intact for movement). Similarly, *spinal* preparations were carried out where only the spinal cord was capable of

[15] Tracings of muscle movement could be recorded on a rotating drum (kymograph) by attaching the muscle tendons to levers that were connected to the drum.

producing any reaction in the rest of the body. Higher-level *Decorticate* preparations above the cerebellum additionally included the basal ganglia, whilst the cortex was transected.

Brown stimulated the ipsi-lateral and contra-lateral sapheneous nerves[16] in varying combinations to elicit reflex responses in the tibialis anticus and the gastrocnemius (ankle flexors and extensors). Movement was observed not only with the decerebrate preparations but also with the **spinal preparations**. A normal agonist/antagonist response was evident; the flexors would contract while the extensors would relax (or vice versa). If both ipsi-lateral and contra-lateral nerves were stimulated simultaneously, rhythmic movements occurred in either the flexors or extensors; the antagonist seemed to remain relatively relaxed while the agonist was flexing. This led to what Brown termed the half-centre model in which one half of the centre induced activity in flexors, and the other in extensors; the opposing antagonists were simultaneously inhibited. Table 3-A summarizes such general observations on movement taken from experiments involving transection of the various neural substrates. Comments for higher-level transections are additional to those made for lower levels.

[16] Arises from the femoral nerve

Central Nervous System capability after transection at different Levels (Afferent feedback still available)			
Spinal Preparation	Decerebrate preparation	Decorticate Preparation	Intact system
Near Normal inter / intra limb rhythmic coordination. Functionally Modulate reflex actions Can do other rhythmic movements concurrently			
	Improved coordination of activation patterns Weight Support Active Propulsion		
		Dynamic Stability Initiate reasonable normal goal directed behaviour in neonatal decorticated animal	
			Adaptable locomotor control system to meet goals of the animal in any environment

Table 3-A Adapted from Patla [155, pp. 7] shows what motor behaviour can be observed after transection of different levels of the neuroaxi (usually in response to stimuli).

Although movement patterns are possible with the spinal preparations, much coordination is lacking including many reflex type functions that require higher levels such as weight support. For example, Magnus showed for a decerebrated animal that reflex standing is not possible, although many other spinal reflexes remain such as movements in reaction to stimuli such as pinching the foot [128]. He comments that: “the centres of the spinal cord can indeed cause and regulate very complicated combinations of movement, but they are unable to give to muscles that steady and enduring tone which is necessary for simple standing”. The foremost part of the brain it seems is necessary to produce the normal muscles tones observed in standing. Continuing with this line, Shik showed that by stimulating various points on the cerebrum, locomotion could be initiated [185]. The experiments of Brown used afferent feedback in the form of reflexes as the stimulus for such movement patterns.

3.3.2 CENTRAL PATTERN GENERATORS

What is one to make of the observation of rhythmical movement that does **not** require higher brain centres to coordinate the movement? Does this represent the possibility of the movement pattern being stored and invoked at the lower levels much as Bernstein theorised with the use of engrams for feed-forward control (*see section 3.2.3),* or is it acting like a clock to regulate the temporal aspects of the movement*?* In the intact animal, there is clearly interplay between afferent feedback mechanisms and the descending signals from the higher centres. To remove this ambiguity between the feedback effects of afferent signals and feed-forward motor patterns, experiments have been conducted with the peripheral influence minimised by either deafferentiation or prevention of musculoskeletal movement[17]. The best examples of such experiments involve what is known as *fictive locomotion* in which an attempt is made to suppress the movement related afferent signals even though the central nervous system is trying to produce these movement patterns. Suppression is achieved by either injecting neuromuscular relaxants or cutting the efferent nerves innervating the muscles and preventing movement; only afferent feedback relating to the static position of the animal will be available. In experiments with the afferent movement feedback eliminated and the descending control signal to the spine also transected, *rhythmic* activity in the efferent nerves of the ventral root was recorded. The activity was reciprocally organized to give coordination between agonist and antagonist. This formed some of the leading evidence for the existence of circuitry in the *spinal network* that can generate the rhythm and activation profiles necessary for locomotion. These neuronal circuits or networks have been more recently referred to as *central pattern generators* (CPG), and superseded the half-centre model (although Lundberg still argues for a refined half-centre model [126] that could still be encompassed within the CPG). Central pattern generators are thought

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[17] If the muscles cannot produce a movement despite a nervous stimulus being invoked, then feedback mechanisms will not have a movement to report on. Although not mentioned in such work, one limitation of such methods that occurs to this author is that, a zero signal is still a signal and has some meaning for the central nervous system.

to exist mainly in the spinal cord and there would be at least one central pattern generator to control each limb.

For researchers who speculate on the existence of these central pattern generators (see review by Duysens and Van der Crommert who explore such evidence [69]), they believe that such inter-neurones largely simplify the computation required by the higher levels to perform automated rhythmical movements such as mastication, locomotion, and respiration. The higher levels need only to initiate the central pattern generators and set the speed of their rhythmic output. For example, Shik, observed how stimuli of various strengths to the midbrain region could alter the speed of locomotion and the style of the gait pattern i.e. walking, trotting and galloping [185]. Jordan later established that it was specifically the mesenphalic region of the midbrain that could induce the locomotion [116].

Therefore, investigators have sought to untangle the mysteries of this information flow and find out how descending and afferent signals may effect the activation profiles produced by such mechanisms.

3.3.3 EFFECT OF AFFERENT SIGNALS FOR MODULATING AND RESETTING LOCOMOTOR ACTIVITY

Central pattern generators thus far may be described as neural circuits within the spinal cord that produce self-sustaining patterns of behaviour that can occur independently of sensory inputs and only require relatively simple supra-spinal signals for initiation. In the intact animal, they do not work in isolation, and in particular, the effects of afferent signals must be considered.

Direct influence on movement via reflexes: Many movement patterns can be created purely according to afferent information in the form of reflexes. A case in point is the motion of the lamprey, an eel like invertebrate, that is able to swim using wave after wave of muscle contractions. Experiments showed that if the spinal cord of the lamprey is cut into short pieces along its length, it could still produce the wavelike motion in a coordinated fashion along the length of the cut section. Was this the result of lots of separate central pattern generators along the spinal cord, or some other cause? Since the lamprey is a primitive vertebrate with about 1000 nervous cells in each segment (up to 100) of the spinal cord, Grillner was able to map out the neural circuits responsible for this wave like swimming motion [88]. Circuits were established that contracted one side of a short segment while inhibiting the other side; such circuits occupied the whole length of the body.

He found that these circuits could be activated in response to afferent signals from stretch receptors. If the spine bent because one side of a segment was contracted, then the other side would be stretched. The stretch receptors, on the stretched side, would thereby induce ipsi-lateral contraction of muscles, whilst muscles on the other side were simultaneously relaxed via inhibiting nerve cells. Thus alternating contractions of the segment flanks could be generated simply through a reflex response. How then was the rest of the body coordinated with this reflex response of a section, and timing achieved between all of them? Grillner and his team discovered axons extending down the body that coupled adjacent segments. These axons conducted inhibitory signals down to the adjacent segments, which delayed their activation sufficiently to give a smooth wave along the length of the body. The brain of the animal was therefore only required to initiate and set the tempo of the wave and did so via axons extending to only the first segment; the rest of the body could be controlled through reflexes. Once the wave motion had been instigated at the head, the rest of the body was sure to follow using reflex type responses.

What then of the very complex circuitry in mammals and more specifically in humans? What role does such afferent input play for movement control and coordination? How much of the movement pattern can be explained, and can rhythms be sustained by such reflex type behaviour? Although the circuitry in humans and for that matter mammals is prohibitively large and complex for such blueprints of the circuitry to be formed, nevertheless, the effects of peripheral afferent input have long been observed, e.g. by Sherrington [184] and Magnus [128, 127], and indeed the simplest forms such as withdrawal from pain are clearly apparent. Other basic reflexes also assist movements by ensuring that they are performed within certain tolerances, a prime example being the classic patellar reflex. A hammer tap across the patellar tendon causes a stretching of the quadriceps muscle fibres, which in turn fires off the stretch receptors within the muscle spindles and invokes a stretch reflex response that contracts the quadriceps. This helps to avoid injury by automatically increasing resistance to the stretch and can be so rapid as to give a kicking effect.

Magnus concerned himself with the physiology of posture as he thought it significant for studies on muscular motion since "every movement starts with and ends in some posture". Of particular relevance was the contribution of reflex for posture; as already mentioned the 'standing muscles' should have a certain degree of enduring tone to prevent the body from falling to the ground. Magnus showed this could occur through reflexes where the output of the central pattern generators on their own had been insufficient. Already by the early 19th century, they were aware that posture was influenced by afferent inputs from the labyrinths reacting to changes in position and to accelerations, from the proprioceptive sensors in the muscles, joints and tendons, and from teleceptors such as eyes and ears. Numerous observations were made of attitudinal reflexes. For example, when a standing decerebrate animal was pinched on its hind foot, the limb that was pinched retracted. Simultaneously, the crossed extension reflex invoked, in the opposite limb, a strong extension sufficient to carry the weight of the hind body portion on its own. This particular reflex was a segmental attitudinal reflex, but there were also many more examples in which the whole body was influenced.

Whole body reflexes generally involved a combination of the labyrinthine and neck receptors; they could be invoked by positioning the head in different poses. Thereby, it was possible, in the decerebrate animal, to generate numerous body attitudes resembling those of the intact animal simply by orientating the head relative to the global axis or relative to the body. For example, the labyrinthine reflexes, which are sensitive to the global point of reference, only produce a maximal muscle standing tone when the animal's head is held in the supine position with the snout a little above the horizontal plane. The tonic neck reflexes however respond to changes of the head position relative to the body, e.g. by flexing, bending, or twisting the neck. Complex reactions are possible with the neck reflexes in which one half of the body reacts in the opposite way to the other e.g. fore-limbs extending and hind-limbs relaxing, and vice versa as the head is tilted up or down, or as the head is turned to the side, one forelimb is extended whilst the other flexed. The combination of both reflexes further complicates the movement patterns, where the effects of each stimulus may be combined at the interneurone synapses. The functional significance of some of these reflexes could be seen in intact animals. For example, if a piece of meat is held low to the ground in front of a sitting cat, the animal fixes on the food with its eyes and thereby bends its neck down. This evokes the labyrinthine and neck reflexes, which tend to flex the fore limbs so that the cat goes down towards the food. In humans and primates, many such effects are suppressed by higher levels, although there is still some evidence of them, particularly in early development of infants. Magnus also quotes a situation involving golf players in which, at the top of the golf swing, the head should be fixated down toward the ball. The shoulders are turned 180^0 relative to the head so that there is a reflex tendency to extend the arm and twist the body to the left (as if the body were trying to unravel). The required movement, in this case, is reinforced by the positioning of the head; instructors, without necessarily knowing this reflex role, will insist upon this posture.

The reflexes described above generally account for *static* situations and act to position the body favourably; indeed there is a passiveness to these actions because of a predictable neuromuscular response almost mechanical in nature. The interplay of central pattern generators and higher levels with afferent input however is not as obvious during these rhythmic movements. Nevertheless there are studies that have examined the role of reflexes during dynamic (in the temporal sense) behaviour, such as walking.

Influence of afferents in rhythmic movement: Overall, most reflexes that have been studied during rhythmic behaviour have been observed in response to perturbations to the movement. This author has found no evidence, in the literature, of reflexes in mammals acting to coordinate the whole course of the movement as with the lamprey described previously. It seems they play a regulatory role, ensuring everything is as it should be, or anticipatory role that prepares the body for quick adaptation to likely perturbations.

An indication of this intrinsic regulatory role has however been measured for the muscle stretch reflexes by applying a selective nerve block to the fusimotor nerves and preventing afferent feedback. One such experiment, regarding the effect on plantar flexor muscles during walking, showed a 50% reduction in their EMG activity when the reflex was removed [182]. Another study by Bennet showed that by applying a 5Hz cyclic movement to the ankle of a spinal cat, the stretch reflexes could induce a further 25% force production by the muscle [17]. The gain of this reflex was dependent on the phase of the step cycle where the force response was most significant during the first half of the step cycle and in contrast, for the last portion of the step cycle, the reflex added little to the intrinsic force. Similarly, Anderson and Sinkjaer perturbed human locomotion by mechanically inducing stretch across the ankle joint [6]. They also observed a phase modulation of the stretch reflex with a maximum during stance, approaching zero at the stance to swing transition and increasing to 50% of maximum during late swing.

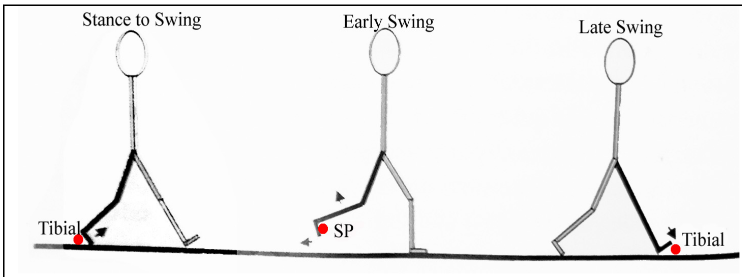

Figure 3-4: from Zehr et al [233, 231] showing functional response to stimulation of the foot surface at different gait phases. Depending on the phase either the tibial nerve (plantar surface) or the superficial peroneal (SP) nerve was stimulated. Red dot indicates stimulation point, and arrows give direction of response.

Although stretch receptors allow reactions to perturbation, striking examples are apparent with cutaneous reflexes during walking. Cutaneous reflexes are able to facilitate fast responses in order to correct for stumbles over unexpected obstacles. For example by mechanically or electrically stimulating the dorsal surface of a cat's paw, Forssberg demonstrated a coordinated reflex response that acted to correct stumbling [78]. An increase in knee and ankle flexion was observed only during the swing phase giving once again a phase-modulated reflex. Similarly, studies by Zehr et al. on humans also showed functionally relevant responses to perturbations during various phases to correct limb trajectory [232, 231]. Figure 3-4 shows the responses when the foot skin surface was stimulated at non-noxious intensities during treadmill walking. Tibial nerve stimulation at the stance to swing transition caused a withdrawal of the foot away from stimulus, whereas at late swing caused increased plantar flexion ready to place the foot. Stimulation of the superficial peroneal nerve during early swing elicited plantar flexion of the foot and increased knee flexion as if trying to reduce the chances of being caught on an obstacle by lifting the shank and flattening the foot.

Direct access of afferents to CPGs': The responses just described attenuate the basic activation profiles generated by the central pattern generators through reinforcement or inhibition of extensor and flexor activity. Phase modulation of these reflexes ensures their effect on the basic pattern will be at the appropriate times. There however remained a question of whether afferent input could have direct access to the central pattern generators? That is, to influence what output signal is produced by the central pattern generators rather than shape it afterwards. Such direct access would be expected to influence the rhythm/ timing of the movement.

Some confirmation for such direct interaction of reflexes was established through experiments that demonstrated a link with afferent input and stance phase duration. In particular, three sensory types were identified. The first two receptors react to the effects of load whereas the third to position; these were proprioceptors within extensors, mechanoreceptors in the foot and afferents relating to hip joint position. It was possible to selectively stimulate (either mechanically like Duysens et al [66], or electrically as by Whelan et al [218]), the afferent nerves relating to the force and length of the ankle extensors while walking on a treadmill. It was found that by either increasing the stretch on the Achilles tendon, or stimulating the group I afferents of the related muscles, there was an increase in the duration and amplitude of the rhythmic ankle extensor EMG burst. Correspondingly, the EMG burst of the flexors was reduced until they were eventually inactive. Neither technique could however differentiate how much the force related receptors (golgi tendon organs /group Ib afferents) or the length related receptors (muscle spindles / group Ia afferents) contributed to the reflex as they were activated at the same time. Selectivity was however improved by weakly stimulating the plantaris nerve, which activated Ib afferents only. Using this nerve, Conway et al [50], and more recently Pearson et al [157] showed that hip extensor bursts (in the medial gastrocnemius) were prolonged and the onset of flexor bursts could be delayed. In addition, Pearson found that if the same stimuli were applied during periods outside the locomotor state then the medial gastrocnemius was suppressed. It was generally concluded that Ib afferents (extensor load receptors) could inhibit the flexor half centre, whilst reinforcing the extensors. The same effects were not as apparent with the Ia afferents (extensor stretch receptors). Functionally, this means that the load must decrease before the flexors can be fully activated. The reflex therefore improves/assists the timing of such events rather than relying purely on the central pattern generators.

In a similar manner, cutaneous afferents from the foot are able to sense limb loading according to pressure on the skin [66, 67]. These signals were able to prolong the extensor burst throughout stance phase, and delay the subsequent flexion phase. Withdrawing this stimulus caused early onset of the flexors.

Finally, the hip stretch receptors were also shown to cause changes in the locomotor rhythm. Anderson and Grillner [7],for example, were able to change the tempo observed in fictive motion by applying small sinusoidal hip movements to a partially denervated cat (i.e. so that the limbs' muscles could not be activated). In another similar experiment by this group, they noted that manual flexion of the hip caused the stepping movement to disappear. Conversely, when the hip was extended to about the angle seen at the end of stance phase, stepping movements could once again be initiated. This led them to believe that hip afferents were important for initiating the swing phase.

Such ideas of central pattern generators and the reflexes that influence them have also become evident in patients suffering from spinal cord injuries [59, 61]. It has been possible, using a harness for partial weight support, to induce stepping motions that were seen to improve with training (e.g., EMG patterns exhibited an improved modulation. Dimitrivic et al. were also able to show rhythmic EMG patterns and stepping movements by electrically stimulating the upper segments of the lumbosacral cord [60].

3.4 EQUILIBRIUM END POINT CONTROL (λ, γ OR α MODEL)

The previous section gave an examination of the information flow used by the nervous system to produce movement patterns. In particular, a review was made of the feedback pathways via reflex responses, feed-forward mechanism of central pattern generators, and their interactions (see section 4.3.3 for more detail on feedback and feed-forward systems). However, these observations only describe a tenuous link between the outward motor patterns and the source of their activation;

they do not explain what and how the nervous system was trying to execute. What is the language of the nervous system, how does it talk about the movement so it can generate the commands for the muscles? For example, is the desired motor behaviour expressed in some form of extrinsic spatial coordinates within higher levels? Bernstein also had such concerns; he thought it likely that complex signals innervating each muscle fibre would be encoded for at higher levels. It was perceivable for only the image (engram) of the movement to be considered at these levels [section 3.2.3].

To this end, one interesting proposal was the equilibrium-point control hypothesis (originally proposed by Feldman [73, 74] although based on Merton's work [139]). It was inconceivable that the central nervous system would perform complex inverse dynamic computations in order to calculate the muscle forces needed to actualise movement plans[18]. The equilibrium point hypothesis was thus developed to circumvent this need, and positioned at a level between movement plans and the final innervations to muscles.

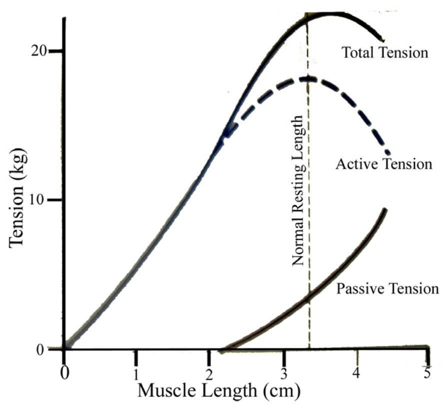

Figure 3-5; Taken from Martini [130], the tension (according to attached weights) developed by a muscle is related to its length. It is the sum of the passive (un-stimulated) tension and the active tension.

The premise was based on the elastic properties of muscles and the length-tension curves as related to the formation of cross bridges between actin and myosin filaments within the muscle [Figure 3-5], the overall slope of such curves being equivalent to stiffness. Co-contraction of opposing muscle pairs, across a joint, generates two opposing moments that subtract, and the impedances add. Since the joint angle will be a function of the lengths of the two opposing muscles, one can similarly also produce angle-moment curves. These however differ in that many curves can be created according to the levels of co-contraction, which give rise to

[18] Movement plans were likely to be operating at yet higher levels within the central nervous system, for example in the form of a topological representation of movement, which Bernstein conjectured using the theory of equal simplicity [section3.2.1, pp. 73].

different joint stiffness values. It is the implication of the equilibrium point hypothesis that each one of these curves can relate to a different equilibrium position, and these can be selectively activated by the nervous system. Defining a position in space according to equilibrium points has several implications:

- it obviates the need to consider muscle forces,
- it is robust to perturbation since the limb will rebound to the equilibrium position because of elastic properties.
- When an equilibrium point is set, the limb can directly go there.

Hence, it was not a great leap from this static situation to go onto the dynamic, in which a series of equilibrium points could define a desired trajectory (usually of the end-point e.g. wrist for elbow motions). The nervous system thereby only needs to specify a vector of equilibrium points in order to produce movements without any reference to forces.

Controversy however still rages concerning how exactly the nervous system may specify these equilibrium positions (e.g. see critiques given by a number of commentaries [51]). Two models were proposed, the first by Feldman known as the lambda (λ) model, and the second by Bizzi et al known as the alpha (α) model [27]. In the alpha model, the nervous system controls muscle activation through α motorneurones and consequentially muscle stiffness, whereas in Feldman's model the cause for shifts in the equilibrium point lay with what he termed the recruitment threshold (λ). This corresponded to a threshold length at which motorneurones would become active, i.e. when muscle length (x) exceeds λ. It thus followed that muscles could be activated either by stretching muscle or by shifts in λ via central commands. The lambda model is therefore viewed by some as the feedback version because of the stretching component (although Feldman himself disputes this [75]), and the alpha as the feed-forward.

The arguments advocating and criticising each model are too numerous and exceed the space available within this thesis. However, key concerns should be noted of the general equilibrium point hypothesis (as admitted by Bizzi et al) that equilibrium point trajectories are most suitable for large slow sweeping movements where accuracy is not critical. The high transients present during fast movements result in a dynamic form of stiffness that is not modelled within this structure but is significant; and furthermore, when accuracy is a requirement of the task, additional systems appear to be needed.

In addition, this theory has generally been applicable to pure voluntary movement such as arm motion. Little has been said regarding its applicability to walking. It may however be suspected that as the theory operates at a level between muscle activation and motor planning, the processes governing walking (e.g. central pattern generators etc.) could also utilise this scheme and control either activation of the alpha motorneurones or the lambda thresholds.

3.5 DYNAMIC PATTERN THEORY - ATTRACTOR STATES & CONTROL VARIABLES

The last two sections have reviewed issues relating to control, from what may be described as an internal perspective, trying to elucidate the workings of the nervous system. This particular review looks at a more global view to movement and its control descriptors. The work that is highlighted takes an unusual approach to examining the dynamics of behavioural patterns; stability measures of patterns are used to gain behavioural information that can result in key parameters related to control of the movement and can help define the task or action being performed.

The dynamic pattern theory owes much of its genesis to the exhaustive work of Erich von Holst, a behavioural physiologist [210], who developed the following rules (paraphrased by Jeka and Kelso [112]):

1. Of the extremely wide possible options available, the behavioural forms that are actually observed are distinguished from the others by their greater stability.
2. A stable pattern manifests in a smooth/gradual alteration of internal or external conditions and *periodic* forms maintain themselves until a critical *limiting* condition is reached. At that point, transference to another stable condition occurs (usually abruptly).

3. The stability that characterises the overall periodic form does not apply to temporal subdivisions, but is balanced out over the duration of the whole period.
4. There's a tendency towards states of increasing stability. Stability increases with the simplicity of frequency relationships, so that increasing complexity gives rise to less stability.

These ideas have been further developed by researchers such as Jeka, Kelso, and Schoner in order to formulate laws of coordinated behaviour. Evolving patterns of behaviour may be best understood in terms of their functional significance that can be described by the low-dimensional space of *"order parameters"(*or collective variables pertinent to the behaviour). Furthermore, the quest for these parameters corresponds to the identification of relevant degrees of freedom (which is generally unknown in biological systems). These degrees of freedom relate to phase transitions[19] within the patterns. Once such phase transitions are discerned, it then enables the identification of both order parameters and control variables as will be explained by the subsequent example.

Kelso made the following observations from a set of simple experiments regarding rhythmic movement [118, 119]. Subjects were asked to rhythmically move their index fingers or hands (as if drumming) under two separate starting conditions: 1) both appendages moving in-phase with each other, and 2) they move in anti-phase of each other. They found that as the frequency was gradually increased, as set by a pacing metronome, the anti-phase movement pattern would switch to in-phase, whereas no transition occurred for the in-phase movement. During the transition period, there would be fluctuations between in-phase and anti-phase, although this soon diminished once out of the sharp transition period. If the frequency was gradually reduced again, the pattern remained in-phase, and no switch back to anti-phase would occur. It seemed the in-phase pattern was more robust than the anti-phase and could occur at all frequencies, whereas the anti-phase pattern was more restricted to the lower range of frequencies. Although this is a relatively simple

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[19] Qualitative changes in the system's behaviour.

example of coordinated behaviour, it contains all the key components that should be sought in behaviours that are more complex.

The relevant degrees of freedom that are sought or the order parameter should have the following properties:

1. Can be used to characterise all observations
2. Changes abruptly from one state to another
3. Remains relatively stable whilst in a particular state, i.e. does not show much dependence until the transition point is reached. The stable periods will correspond to an *attractor.*

In the behaviour example given above, the relative phase (ϕ) between the two appendages had exactly these properties, where the transition from anti-phase to in-phase occurred abruptly over a small frequency range, and the pattern was constant otherwise. Therefore ϕ was treated as an order parameter and the *control parameter* was frequency since it drove the order parameter through all of its states. Control parameters carry no information about the patterns that will emerge and as such can arise spontaneously. They have no ordering influence on the patterns as these emerge solely as a function of the dynamics of the system.

Although the determination of such parameters is useful to objectively define and describe a motor task i.e., according to the state of the order parameter and the control variable, Kelso went further and used them to model the behaviour. In particular, a dynamical model (referring to the temporal evolution) of the order parameter is formulated. Denoting x as the order parameter, x = x(t) and obeys a dynamical law so that

Equation 3

$$\dot{x} = f(x_t, parameters, noise)$$

Special solutions to Equation 3, known as *attractors,* will exist and relate to the stable states of the order parameter. All neighbouring solutions will converge, in time, to the attractor solution, as would most independent trajectories with different initial conditions. The simplest attractor forms are constant solutions (point attractors) to Equation 3, or stable periodic solutions. The behaviour of the ordered parameter may therefore be mapped onto attractors and will be dependent on the control parameter. In the above example, the relative phases $\phi = 0^0$ and $\pm 180^0$ will act as point attractors within the space of the order parameter.

A model of this experiment must also account for the bi-stability of the behaviour in which two patterns are available below a critical frequency, and only one above that frequency. It will also be symmetric about $-\phi$ *to* ϕ and 2π periodic. Kelso suggests the simplest form of Equation 3 that would obey the experimental conditions is to have a so called potential function (V) which is the superposition of two cosines so that

$$\dot{\phi} = \frac{\partial V}{\partial \phi} \text{ where}$$

Equation 4

$$V = -a\cos(\phi) - b\cos(2\phi)$$

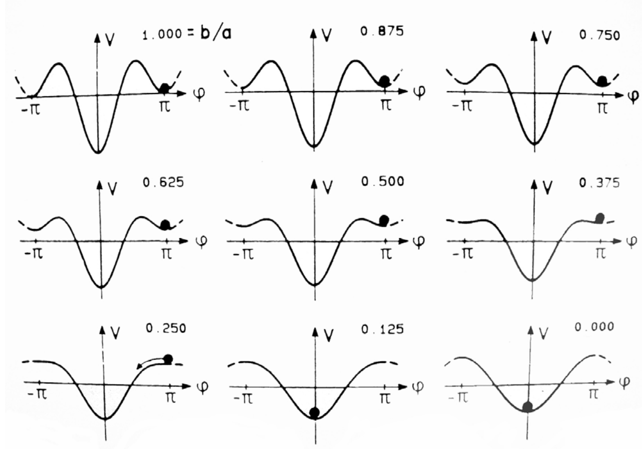

Figure 3-6: (taken from Jeka et al. [112]) Model representing the behaviour of the ordered parameter ϕ in terms of a potential function V. The system is prepared in anti-phase $\phi = 180^0$. Each graph is a different iteration of the ratio b/a that corresponds to frequency (the control parameter). Minima give the solutions to the derivatives known as attractors and reflex stable conditions.

The behaviour of the system can therefore be described by Equation 4 and its derivative. A whole series of potential fields can therefore be drawn by changing the ratio b/a, which corresponds to frequency [Figure 3-6]. The order parameter is therefore most stable when the rate of change is zero, i.e. the minima of Equation 4. The figure shows what happens when the system is initially started in anti-phase. For low frequencies with the ratio b/a >0.5, the system will simply remain at the local minima at $\phi = 180^0$. However, for values of b/a below 0.375 only the global minimum remains so that the system is forced into the in-phase mode. It can be seen how the system might be less stable around b/a = 0.375 because of the flattened profile of the local minima. Any small fluctuations will push the system into the in-phase attractor state, whereas for stable situations such b/a =1 a large perturbation is required to push the system out of the local minima.

3.6 SUMMARY

Although research in movement control is inconclusive, certain opinions seem to predominate. To some extent, this review has tried to link the various topics together. Beginning from a high-level viewpoint, various ideas are given relating to how the nervous system may conceptualise the goal of the movement. One useful high-level description was given by Bernstein where he states, "the key functions for movement are co-ordination, planning, and exploration," and these are attained through reproduction of successful experiences. To this author, reproduction of successful experiences implied learning by either iterative improvements (requiring a performance measure), or by imitation (requiring an idealised model). This clearly gives two paths along which the controller may develop and will dictate the criteria by which to judge its performance.

Lower level theories are related to how the movement goal may be achieved. The extent of the problem for the central nervous system was posed according to the "degrees of freedom problem." An escalation of states is needed to describe the system, ranging from joints to muscle groups, and down to motor units. This complexity provides many ways of achieving the same movement goal. What is the link between these high-level goals and lower level responses, how is the movement constrained to reveal any one particular solution?

Various researchers have tackled such issues, but it was useful to contextualise their investigations with reference to the pioneering work and insightful thoughts of Bernstein. If a hierarchical system can be assumed[20], the problem can be broken down into steps that are more manageable whereby a high-level movement goal is gradually decoded by lower and lower levels until finally motor units are innervated. This gave rise to ideas on whether the innervations were a function of time, muscle length, or its derivative, and how should the interneurones (or control centres) be coordinated. It was logically shown that the innervation signal was most likely to be a function of all three variables, which in other words, implied both feed forward elements and feedback elements had to be in use (as might be expected). This

[20] Whereby interneurone's perform sub-processes.

however led to a key issue where it was extremely difficult to conceptualise the coordination of many control centres if they had a feedback dependence as well as feed-forward[21]. It was only possible to conceive a solution if either one control centre responds alone to feedback signals, or there is a guiding influence much like a conductor for all the centres. This issue is highly relevant since the system to be designed will effectively have two control centres at work. To complicate matters yet further, because of the multiplicity of efferent pathways, control centres were likely to also innervate more than one motor unit, often in different muscle groups and even limb segments. To address this, Bernstein conceptualised the use of engrams (motor images/plans) that would describe the movement in some unknown space (most probably related to topological properties). These engrams could then be linked either serially as in the “chain hypothesis,” or through some external control structure as in the “comb hypothesis.” These topics are important to remember when designing a controller because at some level the controller must also be able to link different movement patterns.

Such ideas were further enforced by experimental evidence gained on central pattern generators (CPG) where rhythmic movements are achieved without participation of the higher centres. CPG theories propose the existence of interneurones that provide a guiding rhythm to enable basic coordination of many automated movement tasks. These CPG rhythms may well be the guiding influence to coordinate interneurone activity. However, it was useful to think of CPGs’ as generating feed forward patterns to govern the overall movement. Furthermore, it had been possible to see that these patterns could be sculpted via feedback reflexes even to the extent of direct influence of the CPG patterns. Overall, such ideas will pertain to the timing of critical movement events including initiation, duration and their amplitude gains.

As a final note on such ideas involving feed forward and feedback control, the Equilibrium point hypothesis was also mentioned as it related to some of the lower level work examining how engrams/plans/CPG patterns can be decoded into a movement. Interestingly, it showed how the actual dynamics of the muscular system

[21] Coordination of many centres for a system with only time dependent signals is much simpler in concept.

gave further scope for feedback control, whereby external perturbations could quickly be countered due to the spring-like properties of muscles.

The final discussion focuses in on the dynamic pattern theory to try and give structure to how the observable movement may be organised. Once again, there is a parallel with these thoughts and those asked in the previous section - what makes walking a unique motor pattern, and how can it be modelled?

Movement behaviours that are commonly observed tend to be more stable and can be characterised according to a low-dimensional set of variables relating to phase transitions. This process involves identifying a control variable[22] , and an order parameter[23] that describes its effect. Of interest to this project was that whilst the order parameter remained constant this indicated a stable movement pattern that related to a particular movement behaviour (or attractor state as mathematically labelled). When such theories are developed, they will allow the boundaries for movement control systems to be more readily defined such as the concise definition of walking.

[22] Control variable: acts as the underlying driving force, pushing the movement through various transitions

[23] Order parameter; condensing the overall behaviour of the pattern into a key descriptor.

4 WHAT IS THE OBJECTIVE FOR THE CONTROLLER, & WHAT METHODS ARE AVAILABLE TO ACHIEVE THIS?

4.1 INTRODUCTION

It is important to note that when designing anything, a well-defined objective is necessary. If this is not apparent then it must also be contrived. For this project, these objectives have been defined in very broad terms. For example, in the introductory section, it was mentioned that the author undertook this project with the belief that a control system, for the knee device of the prosthesis, should allow less stable systems with the benefit of greater performance to be used. The control system should aid the amputee to produce movement patterns that are closer to non-pathological subjects. There was an aspiration to find the best ideological approach for developing such a control system so that general principles of design may be established. Such ideas may then give the groundings for commercial applications. It was also anticipated that the task of walking at different speeds (which was considered a fundamental activity that the prostheses should aid) would be used to demonstrate the viability of the control system. Therefore, if the intended controller is to be successful, then the prosthesis should assist the amputee to correct the deficiencies that have been observed when walking with conventional knee devices so that a more natural gait results.

In this section therefore, the author has attempted to compile a list of more specific objectives that were in agreement with the broader statements made above and in the introduction chapter. For example, to aid the amputee to walk in a similar way to non-pathological subjects is an over-generalization, which conceals many complex interactions. It does not specifically describe the movement patterns that are produced by the intact natural system, and it does not specify how and when these patterns are produced. Therefore, such ideas were carefully considered by the author when compiling the list, although there were many controversies.

For walking, it is not even obvious how this pattern should be assessed. The choice of outcome measures that are often used to judge many sporting performances are not so obvious when looking at intrinsic movement patterns such as walking. To elaborate, the success of a competing sportsman may be judged by his ability to achieve a particular task, for example how consistently and accurately a golf player may aim a ball at different holes as averaged over 18 holes. However, for walking, the goal of the CNS may be to reduce energy consumption, or joint stress, or maintain a steady head movement etc. (see reviews), although nothing conclusive has been determined.

To move away from such a global perspective of a task, the skill of the golfer may be described in terms of more *local* actions. The golfer's motor objective may thus be to control the golf club to follow a smooth trajectory that consistently hits the ball at the optimal velocity. This may be broken down to activities such as keeping the arm straight, being balanced etc as reflected upon by a golf coach. Can such objectives be determined for the controller to emulate or is an even more in-depth analysis required that tries to associate the CNS response with such movement patterns? For example, Bernstein observes the ability of the CNS to consistently reproduce local movement patterns that are recognisable, even though the position, orientation, and amplitude of the movement may differ. Take, for instance, the ability to reproduce recognisable signatures and handwriting styles, regardless of the relative orientation of the plane of the paper with respect to the person [section 3.2.4]. In this example, the elbow and wrist dynamics will change with each orientation; the CNS cannot simply reproduce specific joint dynamics, e.g. particular elbow angle trajectories, to produce this invariant writing pattern (as approximated by the hand or end point trajectory)[24]. In this case, it is believed that the CNS is reproducing aspects of a movement pattern that can be better described by topology rather than by metrical measurements of trajectories for example. These topological features will therefore be independent of the orientation of the paper and will allow for changes in shape and size (such as height and width of letters).

Of course, identifying such descriptor variables will not be so easy, and will possibly have to be indirectly achieved; otherwise, it will be a matter of searching for invariant properties, which are distinct from other movement patterns. In this

[24] Lacquaniti also discusses such ideas on the affects of orientation in more depth, [123]

context, the act of walking is likely to have many such features, even when generalised to an inter-subject level. In the review section, for example, it was described how Braune & Fisher observed that they were not able to distinguish any differences between the gait of a soldier and that of a hiker [chapter 2.2.1.1, page 10]. These are both examples of situations where the subjects have settled into an optimum style of walking which turns out to be very similar for all normal subjects. Such regularity therefore enables one to analyse these common features with the hope of replicating them using the controlled knee mechanism.

This section therefore uses such arguments, inspired from the review sections, to outline the preliminary objectives for the controller. It describes the purpose of the controller in terms that are more specific than the general direction set during the introduction. As implied above, the section on movement control was particularly relevant towards this, since it tried to describe how the natural system might control movement. Such ideas were extrapolated towards what the artificial controller should try to do in order to aid amputee – these were in terms of the overall biomechanical patterns and in terms of complementing the natural control system.

It was thought that these aims would be used to set tangible limits as to what the controller should be expected to do. For example, objectives can be used to imply restrictions on the options available to the controller; these boundaries in keeping with the nomenclature used by Bernstein [Chapter 3.2, pp. 75] will be referred to as constraints. For natural movement, such constraints were used to describe how the CNS might manage the "number of degrees of freedom" problem, in which many different combinations of muscle fibres can be activated to result in the same movement patterns. Constraints enable choices to be made from the vast range of possibilities. Similarly, these choices must also be reduced for the artificial system; this can partially be achieved by applying suitable constraints although inevitably some final decision must still somehow be made.

Once the intended movement pattern that the amputee wishes has been established e.g. through constraints, the controller (in combination with the amputee) must have a strategy to produce this pattern. Therefore, a systematic method was required to assess all the possible interactions and incorporate issues that were raised in the review sections. Towards this, a general control structure was developed which was an estimate, using engineering approaches, of the likely interactions between the natural control system and the artificial. This generalised system could then be used to assess options for a variety of control strategies and picking one that most adheres to the constraints. The approach also allowed a standardised way of reviewing control systems designed by other researchers, and assessing how closely they obey the particular constraints set for this project. This was used to aid in generating ideas for the controller.

Finally, to clarify the approach that was used in this project, there were two methods available in designing a controller. The first was to consider the whole project from scratch and realise all the processes that can be influenced by design so that these can be constrained by the objectives of the project until finally the best logical solution was derived. The second was by a process of iteration. Design processes adopted by manufacturers would generally follow the second method where existing products are improved upon, such as with the design of the "Intelligent Prosthesis" which extends the usefulness of prior designs, as mentioned in the introductory chapter. It was however felt that the first approach was more suitable for this thesis in that the author did not want to be clouded by the practical problems associated with any random prior design. A more generic control system was desirable so that the ideas can be easily extracted and customised for use in a commercial solution.

4.2 PROJECT CONSTRAINTS AS DETERMINED FROM THE AIMS

A holistic approach was used by the author to decide what the controller should be trying to do. The list of these objectives for the controller was hierarchically ordered starting with the most general encompassing aims. The higher levels described the most general aims that the controller should fulfil, lower levels were subsets of these higher levels so that, gradually, a more focused set of criteria was developed whilst the previous levels still held true. Therefore when one sees the first aim for example, it should not be seen in the normal sense of an aim where no additional limits would bind it. These aims thus immediately implied restrictions to the controller, which are subsequently set out in more detail.

4.2.1 INITIAL AIMS

To design a controller that**:**

1. Behaves in a predictable manner that does not surprise the amputee. It responds according to the amputee's intention. This aim was devised because it was observed that prostheses that, for example, unexpectedly collapse make the user extremely wary so that they anticipate problems and move very conservatively. In terms of the rules for adaptive movement by Bernstein [section 3.2.3], the subject will not be able to anticipate its response if the prosthesis is not predictable.
2. Allows the amputee to perform autonomic/automatic actions, i.e. as subconsciously as possible. Coordination of many lower limb movements occurs with little conscious thought and enables consistent movement patterns whilst freeing the subject to think of other tasks. Similarly, the prosthesis should not require too much mental concentration in its use.
3. Responds in real time; the amputee should not be aware that there is a lag between them having an intention and the controller producing the suitable response.
4. Can be interchangeably used by most reasonably fit amputees (after training) without the controller having to be specifically reprogrammed.
5. Has a modular structure, so that its repertoire of controllable movement patterns can easily be extended to include more tasks. The system should be equally capable of achieving several movement tasks, as is the case with the natural system (see review on theory of equal simplicity [section 3.2.1]).
6. As a primary (demonstration) task, achieves improved walking at any reasonable speed (slow to fast). Too fast may become a new pattern i.e. jogging or running (also see review on dynamic pattern approach to coordination e.g. by von Holts and Kelso [210,211,112]). As already mentioned, there may be several ways to measure the success of the controller e.g. in terms of reduced energy cost, more shock absorption, increased range of walking speeds without having to dramatically adapt in

order to accommodate the prosthesis limitations, or simply reproducing particular knee trajectory patterns.

7. Controls a knee unit that is inherently unstable so that the controller is easily able to influence the knee pattern by electronically adjusting the torque produced by the actuator and hence the applied moment to the knee. It would be preferable if the structural template of the knee prosthesis were similar to basic conventional designs so that little modification is required to accommodate the control system, although allowance would eventually have to be made for the actuator. The power demands for the actuator would not be expected to exceed that experienced across the natural knee. It would also be desirable that the system is self-contained and located on the prosthesis as opposed to, for example, having sensors placed obtrusively on the amputee.

4.2.2 CONSTRAINTS AS DERIVED FROM THE ABOVE AIMS:

1. Reacting to the amputee's intention gives an extremely broad constraint that is simply meant to imply that it should not act against the amputee's will. For example, collapsing when the amputee is expecting support lest it prejudice safety. The controller should not confuse different tasks thereby creating the wrong results. This therefore leaves an extremely large number of possible tasks that could be controlled, however this will become sharply focused by the other aims. This aim also possibly limits the amount of adaptation that the controller may require to allow for external perturbations. It may be better to have a predictable response from the controller so that the amputee can instead respond to accommodate environmental differences. This problem is recognised by Bernstein in the context of multiplicity of afferent pathways [section 3.2.3].
2. Autonomic actions in this sense are considered movement patterns, which do not require the continuous involvement of higher-level systems so that they are independent of cognitive and perceptual control [142]. Therefore, this aim immediately imposes fundamental constraints relating to the CNS method of motor control. If the action is to be as subconscious as possible

there needs to a be a decision as to whether a transition, during learning, of the movement patterns from a conscious deep thought state to an autonomic state is possible. In other words, will the behaviour of the knee once "learned and mastered" eventually allow the amputee to use it without intense concentration that may be needed because the movement patterns feel "unnatural" or unpredictable. The restriction to autonomic actions may also simplify the controller's problem of accurately recognising the intent of the amputee [constraint 1] since these movement patterns should be more stable and regular. If, however, constraint 1 is not obeyed and the amputee has to provide correction, then the amputee may have to shift from an autonomic level of control to a much more conscious voluntary level thereby disobeying this constraint as well.

3. "Controller that works in real time" means that it must at least be responding to changing conditions sufficiently fast so that the amputee won't notice that the device has not responded to their new (autonomic) intention. Therefore, ideally it should be responding as quickly as might a reflex reaction, for example, which would be in the order of one hundredth of a second. However, for practicalities this shall be doubled to about 50th of second to fit in with the 50 Hz sampling frequency that is often encountered. This constraint will also complement the previous constraints.

4. The controller must be able to cope with quite a large variance of signals since amputees will not walk exactly the same way nor will they walk consistently. Therefore, the controller should not focus on any characteristic that is particular to the individual, but should concentrate on general features observable in the population. Referring to Bernstein's notion on topological properties [chapter 3.2.4], this is similar to finding topological features that are of sufficiently low order as to characterise/define most amputees.

5. The design of the controller should be such that it allows more functionality to be implemented at a later stage. Therefore, such a system is more likely to have a parallel architecture that will allow additional routines to be tagged on without adversely affecting previous ones.

6. Aim 6 gives a very global description as to the action of the controller; it is the resultant effect of the whole system and not just the controller. However, this global view must be conveyed in local terms so that it is more meaningful for a controller. In other words, the constraints imposed by this global aim must be represented by constraints that the controller will ultimately have direct control over. It was thought that such local constraints may take on several different forms so as to guide the knee during walking:

    - The outcome knee movement could be constrained to model a pattern based on non-pathological subjects during walking.
    - A superior pattern to non-pathological patterns could be used that is more suited to the circumstance of trans-femoral amputees. In particular, non-pathological subjects have full function of the natural foot, whereas this function is severely undermined for amputees. Therefore, an alternative model pattern to non-pathological subjects may be of greater benefit, which allows for this difference.
    - Fundamental constraints may exist which pertain to the knee, i.e. if the intrinsic purpose of the knee is known then the controller will simply obey these “rules”. This would hopefully by default produce a suitable knee pattern despite limitation of foot function. Such constraints may include limitation of shock transmitted to the stump (similar to optimal control ideas [165, 162]), focusing on the timing of critical gait factors such as the shank being straightened in time for the stance phase after swing, etc.
    - Additional factors may also act to dynamically change the constraints imposed upon the knee. For example, if speed were one of these factors then the boundaries imposed upon the walking pattern would be expected to change. Only those factors which have the most significant effect should however be regarded for the sake of practicality.

7. The constraint imposed by using knees of a similar structure to basic conventional prostheses was intended to reduce the degrees of freedom of movement that would have to be controlled. Most conventional knee

mechanisms only allow movement in the sagittal plane, which implies that any rotational movements that may occur in the frontal and transverse planes of non-pathological subjects, for example, cannot be recreated. This will ultimately have direct consequences on constraint 6, although it is hoped that this will be limited since the greatest proportion of motion during walking is in the sagittal plane. Some compensation may however be possible if torque-absorbers are used within the pylon e.g. as made by Blatchfords or the Torsion Adaptor by Otto Bock. A choice is also available between uni-axial, polycentric or condylar knees. The polycentric mechanisms were ruled out because they were mostly developed to gain geometric stability, which contradicts the postulate outlined in the introduction where it was desired that most of the stability should be controlled by an actuator. It should be noted that the desire for instability does not preclude replication of passive resistive forces as experienced in a relaxed knee joint. This may be beneficial even for an artificial system. The difference between the uni-axial and condylar type mechanisms is thought to be small, although because of the simplicity of uni-axial mechanisms these may gain favour. Finally, to touch on the consequence of the self-containment principle, this implies that all sensors used for control would be unobtrusively situated on the prosthesis. It is thus unlikely that information as to the status of the sound leg will be available since sensors cannot be attached to the sound leg and echolocation devices suffer from complexity.

This list of aims and constraints is what has been considered the most desirable fundamental characteristics that a controller should try to achieve. Other more practical considerations may also occur because of these characteristics. For example, the issue of safety will be dealt with by constraint 1, in which the controller should not behave in an unpredictable manner. It should further be noticed that the constraints imposed by commercial limitations have not been forced on this project as it was felt that these would be of higher order that should be left to manufacturers. Technological constraints such as the type of actuator, its power source and weight have also not been included as the control design was to be generic and it was felt better for manufacturers to implement the latest developments in the technologies.

Already, for example, since the inception of this project there has been considerable advancement in actuators such as those based on electrorheological fluids [1] or those that try to behave more like muscles [197]. More focus can therefore be given towards control schemes rather than trying to produce the final practical application.

4.3 General Control Systems applied to Project

The author has now specifically defined the purpose of the controller, but to proceed, it was necessary to identify all the factors to be tuned and sculpted through design to best fulfil this purpose. To elucidate, it was necessary to describe all the components likely to make up the system and how they may interact with each other. Using this approach, a general structure of the problem was gradually developed which enabled a good understanding so that various options could be explored towards achieving the project objectives. Ultimately, it was hoped that knowledge gained during the review sections could be used in conjunction with this remit to decide on the most appropriate course of action.

4.3.1 General principles of Control

In engineering terms, a system which acts under some control process may be broken down into the following components: a *control unit* that drives an *actuator*, (whether natural or artificial), which in turn affects a *plant* or *process*. The resultant effect on that plant can be measured using a variety of *sensors,* each of which detects a different variable; these variables will reflect the *outcome* of the whole system. The particular outcome variable that is utilised by the control system is often referred to as the *controlled variable* (see Figure 4-1, which shows the basic components making up a system under control). The author considered that the controlled variable is normally chosen because it is the most appropriate variable for describing the purpose of the controller. (The word "appropriate" is used in the previous sentence because it gives provision for not only the best sensor but also others that pose other practical advantages such as placement, and complexity.) Once the controlled variable has been chosen, the purpose of the controller can be specified by assigning values to an input variable that will represent the desired outcome of the plant. Engineers usually refer to this input variable as the "set point" (although the possibility for time dependency does exist). Therefore, if the controller has been effective, the controlled variable should end up being equivalent to the set point.

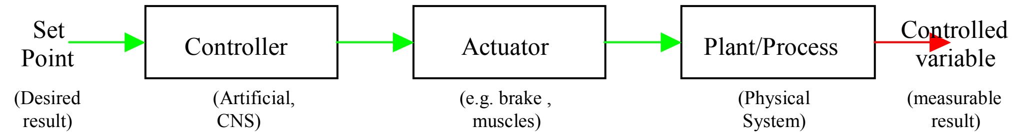

Figure 4-1 - General block diagram showing basic components of a control system

The key components, in the above diagram, act to transfer energy/information from one form to another. For example, actuators often convert an input voltage into a kinetic energy output such as with motors. This conversion process may be represented by a common language, which allows the effects of the various components to be examined in combination. In mathematics, transfer functions, transformations or operators can readily be used to represent such properties. These essentially give a relationship between the input and the output of the component in question so that the **transfer function** = *output / input*. Therefore it may be possible to represent each component in the above diagram by a transformation. All such transformations will be denoted by a bold type symbol. One of the useful properties of these transfer functions is that they allow a system to be simplified. A series of linked components each with its own transfer function can thus be represented by a single transfer function. Take for instance, a motor that is controlled by some voltage input so that its output drives gears. These in turn may rotate a winch. The whole process may thus be represented by a single transfer function. This function can directly give the relationship between the input voltage and the winch velocity rather than having to calculate the motor output, then the gear output, and finally the winch output. This is beneficial because much like driving a car one does not primarily consider all the details of what one is controlling (e.g. piston speeds, gear speeds) although these can be applicable. Instead, one is interested in the outcome, as possibly measured by the car speed or acceleration.

4.3.2 COMPONENTS IN TRANS-FEMORAL SYSTEM

Using these concepts it is now possible to start building up a picture of what possibilities exist for the control system sought in this project.

4.3.2.1 EFFECTORS

The plant in question has to be the knee mechanism, although more specifically from a control viewpoint, this was regarded as the knee dynamics since it was the forces that produce a motion that were of actual concern. Therefore, the resultant forces/moment (e_{knee}) at the knee joint was considered the input variable to the plant $\mathbf{P_{knee}}$. Some thought was given as to how these forces/moments could be generated. It was realised that these forces could be classed into different categories according to their origins; the source and cause of each of these forces/moments was labelled **effector**[25]. For this project, three basic effectors were identified which were the knee, stump, and external effectors; these were able to apply forces/moments to act on the knee mechanism in the following ways:

Knee/(Internal) – Actuators on the knee unit that generate internal forces/moments applied at the knee joint. This could include devices such as brakes that create frictional forces and thus moments about the knee axis or motors that directly induce torque. In addition, some internal force may also be caused by inherent properties of the mechanism such as unwanted friction (e.g. bearings) and subsequent forces produced from stored energy sources e.g. viscosity in hydraulic type mechanisms.

Stump/(Proximal) – This effector really refers to forces and moments that are generated proximally to the knee joint; this would have to be based on the overall acceleration of the stump and its inertial moment.

External/(Distal) – The cause of the resultant external forces on the knee unit could be, for example, by unpredictable perturbing forces from knocks on the shank, to more predictable ground reaction and inertial forces on the shank, and finally from the gravitational effects. All of this was really a representation of the distal forces on

[25] Derived from the word affect meaning “to act on; produce an effect or change in: Cold weather affected the crops.[see dictionary.com]

the knee unit, since external forces that act on the stump were lumped in with the stump effector.

4.3.2.2 OUTCOME VARIABLES

The plant thereby acts to transform the resultant force/moments e_{knee} into a knee motion pattern. As mentioned, such an outcome can be represented by many variables and detected by a variety of sensors. It will be a process of design as to which variables will best be suited for control purposes so that the constraints outlined previously are adhered. To furnish a more informed choice it was recognised that distinctions could be made between these outcome variables. Since the motion results from some form of load action at the knee, the primary effect of this is to cause accelerations of bodies. Therefore, the principal outcome variables were considered angular accelerations at the knee joint and the consequential integrals, i.e. knee angular velocity and knee angles. Because of their close relation to the forces, they were termed *dynamic variables*. In addition it was argued that many other variables could also some how be related to these dynamic variables, and it was plausible that these offered controlling possibilities. The argument relating these additional variables takes three forms:

1. Since acceleration, velocity, and position give a complete description of the motion of the centre of mass of a body in space, other variables may be derived by simply applying transformation cosine matrices. This process effectively allows the original dynamic variables to be viewed from other perspectives or frames of reference, which in practice require different sensory set-ups. In addition, given this complete description of a body, other variables may also subsequently be derived from relationships that are more obscure. The process does mean that it is possible for replication of information because of non-orthogonal systems, but for control purposes redundancy can be beneficial.

2. In natural movement the interdependencies that exist between various segments is substantial because of co-ordination. The hip dynamics ordinarily will closely be related to the knee dynamics so that an overall whole movement pattern is achieved. This co-ordination of body parts

therefore provides the possibility of utilising segmental variables not directly connected to the knee dynamics.

3. Since both the stump effector and the external effector will influence the knee, and the knee in turn will affect their dynamics, measurements regarding their actions are also likely to be related to the knee outcome. This allows many other variables to be used that are derived from some combination of the external effector, the stump effector, and the knee output. For example, energy transfers and mechanical stresses may hence justifiably be used for control purposes. The use of such variables is thought to have significance by many researchers looking at how the CNS may control movement. For example, proposals have been made that certain movements are performed in order to be energy efficient, or prevent excessive stress in musculoskeletal structures.

Such considerations give a wide array for the choice of a controlled variable, since all of these variables can be used to constrain the knee dynamics and hence influence the movement pattern. In fact, the author realised that given the complexities of human movement, a multiple selection of such variables was most likely required to adequately describe the intended motion that the controller would help to achieve. Therefore, maintaining the generality of this section, a vector Ψ was assigned to represent this possible multitude of controlled variables. It was termed **task vector** and represents the set of variables that best describe, for the controller, the outcome motion that results whilst trying to perform some task. It is akin to the controlled variable depicted in Figure 4-1. Therefore, in the particular case for the knee output, this vector may be written Ψ_{knee}^{out} and called the ***knee output* task vector**. This vector can therefore be composed of the variables just discussed. A trade off may occur between the use of orthogonal sets of variables to maximise the information content, and non-orthogonal sets to increase the robustness of the information. Ψ_{knee}^{out} thus may represent some vector space, of n dimensions (where n is the number of variables used by the control system).

The output of the system described so far can be said to be the result of the plant transformation ($\mathbf{P}_{\mathbf{knee}}$) on e_{knee}, so that $\Psi_{knee}^{out} = \mathbf{P}_{\mathbf{knee}} e_{knee}$ where e_{knee} is a function of the moments/forces (τ_{knee}), (τ_{stump}), and (τ_{ext}), produced by the knee, stump and external effectors respectively. This is shown schematically in the flow chart below [Figure 4-2].

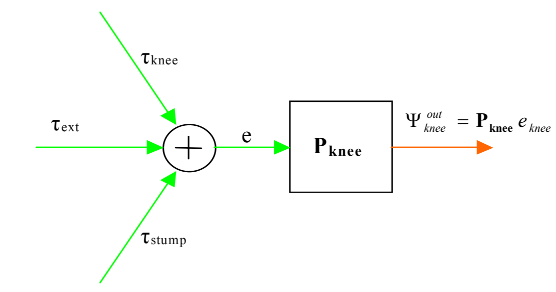

Figure 4-2 - General diagram showing how the plant (the prosthetic knee unit) transforms the combined inputs from 3 effector sources to yield the resultant task vector ψ_{knee}.

Note that to maintain generality, $\mathbf{P_{knee}}$ may be thought of as a non-linear operator that obeys the usual rules governing such operators. This technical detail is useful as it accommodates for the possibilities of functions such as memory, for instant human motion is often non-linear in that it allows whole time sequences to be mapped into one-another, e.g. so that delays between initial torque and response can be accommodated.

It will be noticed that, unlike Figure 4-1, this diagram does not yet include any of the components in front of the plant that generate e. To reduce confusion and to enable a good grasp of this problem some thought was given to the effectors that produce τ_{knee} and τ_{stump} in particular as these can be controlled.

4.3.2.3 CONTROL MECHANISMS FOR EFFECTORS

As mentioned, τ_{knee} is derived from the actions of an actuator on the prosthetic knee mechanism. However, more care must be made as to the stump effector, since these moments are generated by the angular and linear accelerations of the stump. Such motions are generated by the combined effects of the muscles, tendons, and other soft tissue spanning the **hip joint,** which primarily move the stump and generate distal forces by the stump. Confusion may thus arise since these muscles are really affecting another plant, namely the dynamics of the hip joint $\mathbf{P_{hip}}$. Therefore, in a similar fashion, a vector Ψ^{out}_{hip} was also assigned to represent the output from $\mathbf{P_{hip}}$. The moment/force (τ_{stump}) therefore happens to be one of the variables that occur in consequence to the output from $\mathbf{P_{hip}}$. It is therefore also quite possible and desirable that τ_{stump} could be included as one of the controlled variables of the vector Ψ^{out}_{hip}. If this were to be the case, it would imply that the control process for $\mathbf{P}_{\mathbf{hip}}$, i.e. the CNS, must give due regard to τ_{stump} or rather its effects on $\mathbf{P}_{\mathbf{knee}}$.

The question of what controls these effectors, and how, now arises. Referring back to the diagram in Figure 4-1, the knee effector could be described as the combination of the controller and the actuator. This may be represented by a single transformation as previously explained and will be advantageous to do so as it was not desired to constrain design to any particular actuator [section 4.2]. Instead, for this project, emphasis regarding the input (set point), output (τ_{knee}), and the overall transfer function for the effector will be given. Since a vector, Ψ^{out}_{knee}, is used to describe the outcome of the plant, a corresponding vector of set points must also be used as an input to the effector. This is the *knee target task vector* (Ψ^{T}_{knee}), and contains the target values of each controlled variable. These values should be chosen in order to comply with the constraints that were outlined earlier. Therefore, when the output task vector matches this target task vector so that $\Psi^{out}_{knee} = \Psi^{T}_{knee}$, then the knee will be producing the desired motor pattern (assuming the correct choice Ψ^{T}_{knee} has been made for the task being undertaken). Obtaining this match will therefore be the **object of control** or more specifically will be a matter of designing a suitable transfer function for the knee effector. The selection of the output task variables will be based on those variables that can best describe the intended movement pattern for the knee. This therefore highlights some of the major requirements for designing this control system. These are:

- Which target variables are most relevant towards describing the intention of the amputee and the project?
- What are the target patterns for each of the variables in the target task vector?
- How is the controller able to determine which task the amputee has decided to perform?
- How is co-ordination of the knee and hip motion managed? As mentioned the knee task vector may include variables that are related to the stump effector, which may help with co-ordination.
- How should the target task vector be transformed into an output task vector with as small an error as possible?

To answer such questions, consideration that is more detailed was made regarding what control structures may be added to the diagram of Figure 4-2.

Assuming first that a knee target vector has been chosen then the objective is to make $\Psi_{knee}^{out} = \Psi_{knee}^{T}$. Since $\Psi_{knee}^{out} = \mathbf{P_{knee}} e_{knee}$, this leaves two options where manipulation may be applied to produce the desired outcome. The first is obviously on the resultant effector output (e_{knee}) since this is how the original postulate was defined. The second is possibly more unexpected by changing the knee plant to try and influence the outcome. The author however rejected this second option and assumed it was not feasible to change $\mathbf{P_{knee}}$ during movement tasks[26], although it may be possible to tune various properties beforehand to suit individual subjects (e.g. alignment, weight). This therefore left e_{knee} as the only factor capable of influencing Ψ_{knee}^{out} during the task. Of course, since e_{knee} is actually a function of the moments/forces from the effectors, i.e. $\tau_{knee,}$ τ_{stump} or $\tau_{ext,}$ any one of these may be used to alter the behaviour of Ψ_{knee}^{out}. Therefore, each effector is considered separately.

- The external forces are influenced by the combination of activity of the whole body and gravity in the form of ground reaction forces. There is no direct generation of force as for example by the foot complex as in non-pathological subjects. Therefore, the best one can do in terms of control is to possibly incorporate the desired forces into the target task vectors for the knee, hip and general body motion. This means, if there were to be no other controlling influence from other effectors then the only possibility for designing a prosthesis that is capable of producing some target pattern is to tune the plant, i.e. make sure that the plant transformation results in a favourable reaction to probable inputs.
- Control of the stump is regulated by the CNS and the musculoskeletal system; in the following nomenclature, these processes are collectively denoted by

[26] **P** will be an invariant transformation because the transformation of the moments into dynamic variables represents factors intrinsic to the knee mechanism being used such as alignment, weight, and inertial properties.

AMP. Although, theoretically, it may also be possible to have an artificial influence on the stump movement, for example by electrical stimulation[27], it is unlikely that such a system would gain acceptance by amputees.

- There are two possibilities for controlling τ_{knee}, that is either by using some artificial controller that decides on the appropriate target pattern and the corresponding τ_{knee}, or by direct interaction of the CNS, for example as conveyed through EEG or EMG signals. In the second option, the CNS makes all the decisions as to how the knee should behave. This would still require an element of artificial control to decode such signals into a moment τ_{knee} from the actuator; devising the decoder would be a two way process in which the CNS may also have to get used to the artificial system.

Finally, the interplay between these three effectors will also be critical in order to produce a suitable e_{knee} pattern. The artificial controller which will only directly act upon τ_{knee} must not produce conflicting responses to the other two effectors.

These ideas correspond to some of the views expressed by Bernstein where he realises the possibility of multiplicity of efferent pathways arising from various localised control centres within the CNS to different muscle groups (see chapter 3.2.3 page 82). In the natural system, he postulates that there must be communication between all these centres that enables them to act as one, their common activity is organised much as a conductor may command an orchestra. The same must be true for this prosthetic system, where the two controlling centres (artificial and CNS) must also be co-ordinated to behave as a whole unit. One notable difference is that, unlike the natural system, where different control centres can act on the same group of muscles, it has been decided that the artificial controller will not control the thigh.

This requirement for co-ordination means that the target patterns Ψ_{hip}^{T} and Ψ_{knee}^{T} should be closely associated with each other and their results simultaneously

[27] The finer points of this argument however, may allow for some direct interaction with the CNS, in the sense of proprioceptive assistance (this type of work is more prevalent in hand prostheses).

realised. However, because of the restriction for autonomic behaviour [constraint 2] there is little scope to drastically vary the hip target pattern, which in some ways also reduces the choice of patterns for the knee. If amputees have to adapt their hip movement pattern too much to accommodate the prosthesis then the object of control will not be attained.

4.3.3 GENERIC STRUCTURE FOR TRANS-FEMORAL PROSTHETIC SYSTEMS

It has thus far been shown that to produce an effective target task vector pattern, co-ordination of the stump and the knee effectors must occur so that an appropriate e_{knee} pattern for the particular task is generated. What has not been discussed so far is the processes that may control these effectors, i.e. how might, Ψ^{T}_{knee} and Ψ^{T}_{hip} be transformed into τ_{knee} and τ_{stump} respectively.

In general, control systems may be composed of feed-forward and/or feedback configurations. Although the goal is the same, in that they are both trying to produce the same output as defined by the target task vector, their approach differs. Feedback does this by correction of the previous task vector output, whereas feed-forward does it according to prior knowledge of how a particular time sample of the task variable should be produced. Feed-forward control relies purely on the target task vector to produce a response whereas feedback control additionally bases its actions on the state of the output task vector produced by the plant. That is why feed-forward which has no link with the output from the plant is often referred to as open loop, and feedback is closed loop control.

The diagram in Figure 4-2 will be expanded to include the feed forward and feedback components denoted by the letters **A** and **R** respectively. For example, if the knee effector utilises a feed-forward transformation then it may be signified by $\mathbf{A_{knee}}$, and the output from this transformation would be τ^{A}_{knee}. It is therefore possible that the knee actuator and the hip actuators (i.e. muscles, tendons etc.) can be controlled by a combination of feed-forward and feedback systems; a similar strategy for the elbow joint was proposed by Hollerbach et al [103]. The diagram in Figure 4-2 has been expanded to include the effectors and details how feedback and feed-forward components may be represented; the overall structure is shown in Figure 4-3.

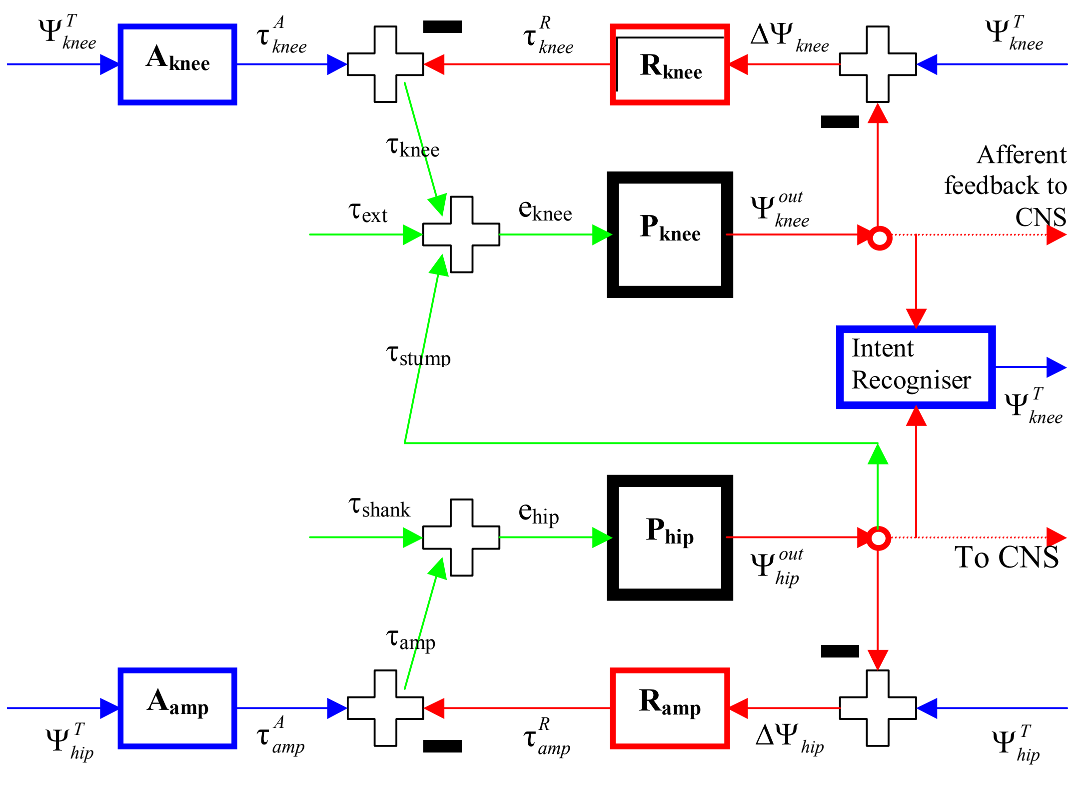

Figure 4-3; Flow diagram showing feed-forward and feedback components of a general control system for the prosthetic knee.

To aid visualisation, the feed forward components have been coloured blue whilst the feedback components are red and portions that result from the combination of both feed-forward and feedback are green. It will be noticed that to maintain generality feed-forward and feedback methods have been included for both the internal moments/forces being applied to the hip and knee plant, i.e. τ_{knee} and τ_{amp}. The outcome reactions from the hip plant (in the form of τ_{stump}) are then linked as an additional input to $\mathbf{P_{knee}}$. Similarly, it is conceivable that the shank can apply forces to the stump because of accelerations generated by the knee and through ground reaction forces; this input to $\mathbf{P_{hip}}$ is represented by τ_{shank}.

The patterns that Ψ^T_{hip} may represent will be somewhat abstract as there is debate as to exactly how the CNS controls movement (e.g. is it trying to minimise energy consumption, produce particular trajectory patterns or something else). Nevertheless the concept will be useful in order to understand what is required of an artificial controller especially because of the demands of co-ordination. It is therefore likely and preferable that some of the task variables that are used for the knee will be a reflection of the hip task variables. In fact, it would be a point of good design to include such a relationship so that the same final movement outcome is produced by the combination of both controllers.

The demands of co-ordination must also be maintained as the amputee decides to change the task being performed; this would have to be reflected in both a new Ψ^T_{knee} and a Ψ^T_{hip} pattern. Therefore there is a separate requirement to decide what the current objective of the controller is. This must always be linked to the amputee because of constraint 1. In essence one can not have a situation where the prosthesis decides what the amputee should be doing. Therefore, an intent recogniser has also been incorporated into the diagram of Figure 4-3; its sole purpose is to decide what the target pattern should be. It will be noticed that the intent recogniser monitors outcomes that the amputee can directly influence. To try to abide by the principle of self-containment [constraint 7], these signals are derived from the hip plant.

As one of the main aims for this project is to control the knee during autonomic tasks, it is important that the intent recogniser does not require high-level conscious input from the amputee.

Fortunately, the CNS should still have some afferent feedback e.g. via visual and proprioceptive sensors to be able to implement appropriate tasks according to the state of Ψ^{out}_{knee}. This feedback as mentioned will inevitably influence the output Ψ^{out}_{hip}. For instance, if Ψ^{out}_{knee} is not well behaved then the amputee will also have to adapt the Ψ^T_{hip} pattern from the preferred autonomic one. This sort of relationship also makes it clear why it is important to achieve the constraint set by aim 1 where the controller should behave in a predictable manner. If the controller is unreliable, the amputee will not be able to learn what the expected Ψ^T_{knee} pattern is likely to be so that they are not free to produce their ideal Ψ^T_{hip} pattern but instead will be heavily dependent on conscious high-level control.

The diagram has been constructed with the premise that both feed forward and feedback control can be used to control τ_{knee} and τ_{stump}, or at least to allow these possibilities to be explored. An explanation is now given how these methods may individually work.

4.3.3.1 FEEDBACK COMPONENT

In this section, only feedback is considered so that the output of the feed-forward components is made zero. In addition, to appreciate how feedback works, a simplified case is needed in which there is only one controlling influence on the knee. The problem of trying to combine two controllers to achieve one goal is dealt with at the end.

A feedback system performs a transformation of the error between the desired goal and the actual outcome into a correction. From a fundamental base the actual goal can be described by the target dynamic variables e.g. knee angle θ^{T}_{knee}, although eventually it shall be shown how the task variables can be related to these variables. It may be said that the feedback moment is some function of the difference between the target variable and the actual variable, so that

Equation 5 $$\tau^{R} = \boldsymbol{R_1}[\theta^{T} - \theta^{out}] = \boldsymbol{R_1}[\Delta\theta]$$

where the superscripts "T" and "out" signify the target and output variable, and $\Delta\theta$ gives the trajectory error.

However as discussed above it is preferable to have the option of using task variables, therefore we assume some linear relationship between the dynamic variable and the task variable

$$\theta = f(\Psi)$$

Since it is hoped that the target variables will be constraining the movement pattern, then

$$\Delta\theta = f\left(\Psi^{T}\right) - f\left(\Psi^{out}\right)$$

which gives for linear functions

$$\Delta\theta = f\left(\Psi^{T} - \Psi^{out}\right) = f\left(\Delta\Psi\right)$$

Therefore substituting this into Equation 5

$$\tau^R = \mathbf{R_1}[f(\Delta\Psi)]$$

and if a new feedback transformation **R** is used which incorporates the function f then

$$\tau^R = \mathbf{R}[\Delta\Psi]$$

Therefore as a first step, it has been shown that feedback methods can just as easily be used for task variables as dynamic variables.

The way feedback may track the target task variable can be shown mathematically.

Assuming there is only feedback to the knee actuator then

$$\Psi_{knee}^{out} = \mathbf{P_{knee}} e_{knee} = \mathbf{P_{knee}}\left[\tau_{\text{ext}} - \mathbf{R_{knee}}\left(\Psi_{knee}^{out} - \Psi_{knee}^{T}\right)\right]$$

And rearranging to get Ψ_{knee} on its own gives

Equation 6 $$\Psi_{knee}^{out} = \frac{\mathbf{P}_{\text{knee}}\tau_{\text{ext}} + \mathbf{P}_{\text{knee}}\mathbf{R}_{\text{knee}}\Psi_{\text{knee}}^{\text{T}}}{1 + \mathbf{P}_{\text{knee}}\mathbf{R}_{\text{knee}}}$$

Letting the numerator $\cancelto{0}{\mathbf{P}_{\text{knee}}}\tau_{\text{ext}} + \mathbf{P}_{\text{knee}}\mathbf{R}_{\text{knee}}\Psi_{\text{knee}}^{\text{T}} = \mathbf{P}_{\text{knee}}\mathbf{R}_{\text{knee}}\Psi_{\text{knee}}^{\text{T}}$ *condition 1*

Then Equation 6 becomes

$$\Psi_{knee}^{out} = \frac{\mathbf{P_{knee}}\mathbf{R_{knee}}\Psi_{knee}^{T}}{1 + \mathbf{P_{knee}}\mathbf{R_{knee}}}$$

and if $\mathbf{P_{knee}}\mathbf{R_{knee}} >> 1$ *condition 2*

then, the desired result $\Psi_{knee}^{out} \approx \Psi_{knee}^{T}$ should be produced.

Therefore, two conditions must be met to produce this desired result. For *condition 1* to be true, the result of $\mathbf{P_{knee}}\tau_{\text{ext}}$ must be negligible, otherwise as logic would dictate the feedback controller cannot react to large perturbations that have just occurred. It will only be able to react subsequently to the errors caused by the perturbation. Interestingly it does present the possibility of designing $\mathbf{P_{knee}}$ so that it acts to minimise the outcome caused by τ_{ext}. In fact, this would be analogous to the equilibrium hypothesis [section 3.4], which is based on the muscles' elastic properties to give such a feedback affect. This process is much quicker than anything dependent on the nervous system.

Condition 2 relies on the product $\mathbf{P_{knee}R_{knee}}$ being very large to create what is known as a high-gain feedback control where accurate control is possible without any knowledge of the plant dynamics. Of course, there are practical limits as to the extent of this gain but in general the larger the gain the more quickly the error will be reduced. A large gain however does not come without some drawbacks; the main one is the tendency for the reaction to be so large that it can overshoot its mark and cause oscillations (such ideas are explored further in Appendix C where the various types of feedback configurations are described in more detail).

The correction moment calculated at $t=t_1$ should bring Ψ_{knee}^{out} closer to Ψ_{knee}^{T} at $t=t_2$. The interval between t_1 and t_2 will depend on the system's ability to sample signals, process them, and effect a mechanical response; this is the delay in producing a correction moment. This should satisfy constraint 3, i.e. a desired ideal response time of 0.01s (although practical considerations may allow this to be increased).

As can be seen from the above it is important to decide how the error is to be transformed into a correction moment since if the conditions are not met then the result is more likely to be in error. Engineers have developed several approaches for the choice of the transformation type made by **R**. The various approaches are outlined in Appendix C.

4.3.3.1.1 Pros and Cons of Feedback control systems

Feedback control systems are often used because of their simplicity and robustness. This stems from the fact that for many situations it is possible to achieve accurate control without any knowledge of the plant dynamics. This is especially the case if high gain can be achieved where accuracy should be achieved as quickly as possible. A feedback system can however cause problems because there are physical limits as to what practical systems can achieve; some of these limitations are due to actuators, which are listed below:

- **Saturation:** A feedback system relies on the size of the gain **PR** it can produce for accurate control. However all actuators have a maximum moment which they can exert, therefore a large gain cannot always be realised in order to fully and speedily correct an error. This gain can therefore require unrealistically high moments.

- **Bandwidth:** There is a limit as to the frequency response of actuators which means instabilities can occur particularly if this puts the response 180^0 out of phase with the target pattern (see section on PID controllers).

Otherwise, real feedback systems must always suffer from delays, which will limit the size of the gain (see section on PID controllers). Finally feedback systems are error driven which always means that there will a dead zone that must be overcome somehow possibly with other side effects on response times so that a speed/accuracy trade off can ensue (see section on proportional controllers and also tuning PID controllers).

4.3.3.1.2 Possible Feedback problems associated with project.

All of the transformations mentioned in Appendix C, in effect, define the path back to the correct task variable trajectory, e.g. directly with a straight line or using a curved approach that slows down as the target is neared. Quite often the intention of such techniques is to get back to the target point as quickly as possible without causing overshoot. This however may not always be an appropriate solution for this project since too much acceleration may be generated which the amputee may find more disturbing than slowly getting back to goal. Therefore, if this were the case then acceleration may also have to be included within the task vector describing what should be constrained.

All of the above feedback methods, except two-point, require a bi-directional knowledge of how the knee should respond. A problem which may be faced by this project is deciding exactly what the direction is since the error in the task variable does not have to relate directly to the direction that the system is going.

It is also important to have a sufficiently high sampling rate to avoid aliasing. This must be at least twice the highest sampling frequency being monitored. For practical purposes, this is usually at least 10 times the highest (visually apparent) frequency in the variable being monitored.

The feedback transformation for the CNS under normal circumstances will be based on a much larger task vector set than an artificial knee controller can be based upon. For example, there will be feedback not only from proprioceptive and cutaneous systems but also from visual cues. However, because of amputation it must be recognised that some of the efferent signals required for the natural feedback control will be lost due to missing sensors in muscles, tendons, joints. There will also be distortions of signals from the stump due to new insertions, and damaged tissue. Some useful feedback is still possible however, as it is believed that most of the proprioceptive information used for balancing is derived from the hip and trunk as described by Allum et al. [4], similarly the vestibular and visual systems will help the CNS compensate for the sensory deficits. To further aid the CNS, some researchers such as de Vries [55] and O'Riain et al [151] feel that trying to artificially replace some of the missing afferent signals would be very beneficial in helping the amputee improve the control of the prosthesis. The idea has gained particular credit for hand prostheses where tactile information has been shown to be beneficial, e.g. by Pfeiffer, and Kyberd et al. [160, 121]. It would be possible to include this work at a later stage as it will not effect the control strategy.

4.3.3.1.3 Combined feedback

Figure 4-3 combines two feedback systems, i.e. the stump controller and the knee controller. If the role of both is to reduce the difference between Ψ_{knee}^{out} and Ψ_{knee}^{T}, then some conflict may arise between the effects of τ^{R}_{amp} and $\tau^{R}_{knee.}$ Either the two controllers must behave in a co-ordinated manner, or they must not react to the same things otherwise the combined outputs τ_{knee} and τ_{stump} may not produce a suitable enervation (e) pattern. For the two systems to be co-ordinated, they must both have some knowledge as to how each other will react to the error. This immediately presents a problem in that we cannot change the action of the CNS (although training may permit some adaptation[28]). Therefore, the artificial controller must have

[28] The role of learning is always key when one interacts with any artificial system, e.g. riding a bike, where learning of the new system can reduce the effects of inherent reflex actions (i.e. over reaction to slight wobbles is reduced through learning).

extensive knowledge as to how the CNS will react so that it complements the amputee's own corrective action. An imaginary example situation may be at the end of the swing phase, where the lower leg has not extended enough for the foot to arrive properly for initial contact. The amputee may react via reflexes to increase the motion of the stump and thereby impart more kinetic energy to the shank, however the knee controller may have also reacted to this situation by reducing the knee resistance thereby resulting in an overcompensating effect that is likely to result in the leg abruptly snapping into hyper-extension. The other suggested alternative may make for simpler design of a controller in that it would only have to react purely to situations to which the amputee does not. In some sense this is however an extreme case of co-ordination where each system produces zero correction when the other is working.

A feedback artificial controller in this situation therefore must have some feedback not only to the status of Ψ_{knee} but also on τ_{stump} so that a suitable combined input e_{knee} can be generated.

4.3.3.2 FEED-FORWARD COMPONENT

Feed-forward components of a controller are deprived of any information fed back to them reflecting the current state of the plant and so cannot know whether the output task vector has accurately followed the amputee's intention. A feed-forward control system only produces the desired response according to *predetermined* knowledge of how the desired movement pattern should be produced; this knowledge is defined over a short period by $\left[\Psi_{\text{knee}}^{\text{T}}(t)\right]_{t=0}^{t_1}$. Feedback tries to achieve set points by correction of the previous plant output; therefore, the controller is only concerned with the current error, which is independent of time. The time dependency in feed-forward control is much more significant since for a particular action the whole sequence of the movement pattern must be known/planned in advance for the duration of that movement. This motor plan can therefore be thought of like a tape or model describing events for the short future. The feed-forward controller will diligently follow this and repeat it exactly if the same movement pattern is required again; Bernstein refers to this mapping of the future as **engrams** of movement [section 3.2.3]. For the CNS, this may be in the form of mental images of the desired movement or something lower level such as in the CPG, whatever it is, some artificial version would be required for a feed-forward knee controller.

4.3.3.2.1 Feed-forward Engrams (motor plans)

If feed-forward control were to be the preferred system, the desired patterns for the knee would have to be produced from a model of the required task (namely walking at a range of speeds). These models may extend for the entire duration of the task or simply perform a segment of the task. In either case they must be linked with the previous engrams. The possible options for these motor plans are that either:

1. Separate engrams are used which represent the entire duration of the task (i.e. gait cycle) for each and every required walking speed.
2. or one engram, which is speed dependent, specifies the pattern for all the gait cycles.
3. or the gait cycle for each speed is composed of several engrams so that there is no speed dependent modelling and each individual feed-forward pattern is relatively short.
4. or several speed dependent engrams are defined for various portions of the gait cycle.

Even without considering the variability that different terrains may introduce (e.g. inclines) it can be seen that if an engram is required for each circumstance (e.g. different walking speeds) then there is likely to be a proliferation of engrams. Therefore, to moderate this, models that are dependent on these variable factors would be advantageous. According to reviews, the CPG hypothesis also displays to some extent such properties in that, for example, the basic output pattern is believed to stay much the same whilst the rhythm is set by higher levels, which allows the adjustments for different speeds.

The variety of methods with which engrams may be linked has been addressed early on by Bernstein where he postulated either the comb hypothesis or the chain hypothesis (see review 3.2.3, pp. 82). Essentially, for the artificial system, how will precise initiation and timing of these engrams be implemented?

In Bernstein's view of the chain hypothesis, the activation of one engram was brought on by the end of the previous engram. The initiation of different events can be detected at the periphery by proprioceptive feedback. Therefore, in a corresponding manner, for the artificial system, sensors could detect events that are related to the end of a particular engram. For example, the last column in the target task matrix may be used as an initiation factor (assuming it is a unique set of values). This method has further advantages in that it enables a correction of motor patterns, especially if a task is split into several engrams. If for example there is a delay in achieving a portion of the gait cycle then the chain hypothesis will wait until the appropriate time.

The comb hypothesis uses more of a feed-forward principle to initiate engrams because it hypothesises some higher control, which sets which engram is to be initiated, and when. It will essentially have a timing plan for the whole movement so that, possibly, trigger signals are sent to start the relevant engrams. Of course, this would once again leave the problem that something must initiate this commanding timer. This idea of an overall controlling timer also leaves the ability to control some variable aspects of the pattern such as speed.

Once the desired movement pattern has been established the target task variable engram must be transformed into a moment response from the knee mechanism. Unfortunately, there can be no one-to-one correspondence with the movement pattern and the forces required to produce this pattern. This is because any relationship between force and movement must be at least of second order, therefore solutions to this will require two constants of integration namely the initial position and initial velocity. For the artificial controller this is possibly a reinforcement of the previous statements regarding the comb or chain hypothesis although it does also emphasize the importance of using the initial velocity as an additional requisite. In effect without the same initial conditions, the same pattern of forces will not produce the same movement pattern. The problem is further complicated in the natural control system where it is possible to produce similar movement patterns even with different groups of muscles (part of the degrees of freedom problem). Although for the natural system, this question is answered to a limited extent by the equilibrium point hypothesis. In this context a mental image (engram) of performing a task is possibly used (which would represent $\psi^T(t)$). The equilibrium theory then tries to help to explain how this image may be transformed into signals that enervate the muscles to produce the required movement pattern.

4.3.3.2.2 The perfect feed-forward transformation

The question still remains however of what the transformation **A** should be, assuming that $\psi^T(t)$ is known for the period $t_0=t<t_1$. If there is to be no external perturbations then we may be able to calculate the perfect transformation to produce the target pattern. Take for example a case where the feed-forward controller has sole influence on the knee dynamics so that there is no feedback or other external sources. Then the perfect transformation **A** would be the inverse of the plant dynamics **P** since;

$$e = \mathbf{A}\Psi^T \qquad \text{if there is no feedback or external force}$$

but $\quad \Psi = \mathbf{P}e$

therefore $\quad \Psi = \Psi^T$ if $\mathbf{A} = \mathbf{P}^{-1}$, i.e. the inverse of the plant.

This makes sense since if one knows what the dynamic properties of the knee are (such as inertial moments, mass etc.) then it should be possible to calculate what forces are required to produce a certain trajectory as defined by the task variable.

4.3.3.2.3 Global and Local models for the transformation

If one could directly calculate these forces according to a target task vector then **A** would represent a **global** model. These types of models can be advantageous because they can often include variable factors such as speed of walking and other applied loads such as walking with a backpack.

The other possibility is to use a known activation pattern for the actuator that has already been shown to be useful for that task; this would form the basis of a **local** model. In the natural system, this can be achieved by a process of learning and memorisation of what is likely to be the most effective activation patterns to the different muscle groups (this agenda may be defined by various types of constraints that are beneficial such as energy expenditure and stress absorption). In our artificial situation, it must however be a process of design to establish what the learning process should be to find the ideal moment profile.

4.3.3.2.4 Coordination of the two feed-forward controllers:

Similar to the problem of co-ordination described for feedback some consideration must also be given as to how feed-forward controllers may act together to achieve the desired task. Unlike feedback systems, no ability is available to correct for discrepancies that may be caused through conflict. There are three places within the control structure where the effects of the other system on the overall movement may be considered. From the point of view of the amputee's controlling system there has to be some consideration as to whether the feed-forward pattern used to control the stump will be compatible with a new artificial pattern at the knee. Therefore, the compatibility for the feed-forward systems depends on whether any adaptation to the effects of amputation has taken place or whether the stump still receives the same feed-forward pattern as the non-amputee but discrepancies are accounted for by the feedback system (i.e. has a new CPG pattern been created?). If a different feed-forward stump pattern exists that produces an altered $\tau_{stump}(t)$ pattern then either compensation is required by the artificial controller, or subsequent adaptation by the amputee has to take place so that the stump resumes more of a normal pattern. Of course, this project is based on this second conjecture, i.e. that an actively controlled knee that behaves more as a normal knee will *restore* much of the normal gait pattern and reduce the problems associated with the abnormal gait. This would be more in line with the mentioned notion that normal feed-forward innervations can once again be produced by an amputee if there is a normal pattern being maintained for the knee.

In summary with a feed-forward system, co-ordination of the natural with an artificial controller should not be as much of an issue as using a feedback system because the role of the artificial controller would not counteract the natural system. A feedback system would compete with the natural feedback element resulting in unstable solutions. Therefore, as long both the natural system and the artificial have accurate models of how to produce their bit of the desired movement and, critically, they are able to initiate these at the correct times there should be little co-ordination problem.

4.3.3.2.5 Pros and Cons of feed-forward control systems

As already demonstrated for feed-forward control, if one is able to find the inverse of the plant dynamics then one can achieve perfect control so that the desired knee trajectory will be closely followed provided there are no significant external perturbations/disturbances. Therefore, the problems with solely feed-forward systems arise from the fact that they are unable to correct for errors however caused. The causes of these errors can stem from the following conditions:

- **Imperfect global models or defective local profiles:** Obviously, if one is unable to accurately determine the plant dynamics then the perfect trajectory cannot be achieved. Since there is no feedback then correction of these systematic errors cannot be made so that they are likely to accumulate and make the task variable drift further and further from the target.

  In addition, because of the complexity of modelling human movement, practical considerations may be made, such as the processing speed and producing a response that still relates to the initial conditions. Since this response time is measurable, some compensation may however be built into the control system.

  Imperfect local models are likely to yield similar problems except that they are less likely to be systematic. Some portions of the moment profile will have less impact on the gait cycle than others.

- **Unexpected perturbations:** Since there is no feedback as to the state of the system, there can be no knowledge that the plant dynamics have been forced off track by some external perturbation. Therefore whether global or local models are used this error will be carried for the rest of the engram until new appropriate initial conditions trigger another engram.

- **Mistaken Initial conditions:** Although, as indicated in the previous paragraph, initial conditions may have the ability to reset the feed-forward patterns, errors are likely to result if for some reason the initial conditions are inaccurate. The reasons why such initial conditions are necessary have been expressed earlier in this chapter and this can be seen intuitively where for example if the knee angle is not at its expected initial position then the model will not compensate for this error. The error will once again be propagated throughout the engram much like unexpected perturbations. A point of design will be to decide the acceptable tolerance for the detection of initial conditions.

The question that must be answered if a feed-forward system is to be used is what accuracy is required from it, can a model be developed which can achieve this and will possible errors be too significant without correction and if so - is additional correction possible.

4.3.3.3 HYBRID FEED FORWARD AND FEEDBACK CONTROL

So far, feedback and feed-forward components have only been considered individually, however at least for the natural system they act in unison and complement each other. For example, reviews were given of how the feed-forward patterns from CPGs may be altered by afferent feedback signals [section 3.3.3]. The majority of these reflexes are thought to affect the feed-forward pattern after it has been generated by the CPG although there were a few examples where afferent signals could also directly influence the CPG. The importance of such inter relationships between these two systems is further brought to light when one considers how adaptation occurs when either of the systems is altered. For new amputees with their afferent system significantly disturbed, it has been shown that there is a serious shift toward cognitive control of posture and gait [142]. The experiments showed that for new amputees "the speed of walking collapsed under cognitive load". The effect was also not due to the prosthesis because Heller et al [96] found that the intelligent prosthesis was no more cognitively demanding than a conventional prosthesis. Over time however there is a recovery in automaticity where the remaining afferent signal is integrated into a new motor plan; this is particularly true for young subjects who are more capable of adaptation and learning new patterns of movement.

Combining feedback and feed-forward systems may have many benefits including overcoming some of the disadvantages that can arise by only using either technique on its own. The price however would be increasing complexity of the whole system.

5 OTHER ARTIFICIAL KNEE CONTROLLERS

In this chapter, descriptions of knee controllers developed by other investigators are given. Some attempt is made to examine the controllers in light of the concepts discussed within chapter 4 (see section 4.3.2 for nomenclature).

5.1 CONVENTIONAL PROSTHESES (INCLUDING MAUCH)

5.1.1 BASIC CONVENTIONAL PROSTHESIS

Although prostheses have been in existence for many centuries[29], it is only recently that technology is sufficient to see the advent of prostheses, in a few cases, with artificially controlled knee units. However, in general today's conventional prostheses rely purely upon passive knee mechanisms to try to recreate some of the original function of a natural knee. From a control point of view, these systems limit the complexity they can afford to the target task variable pattern (Ψ_{knee}^{T}). Such target patterns can only be assisted by tuning the dynamic properties of the plant $\mathbf{P_{knee}}$ since active control of the knee effector and therefore adjustment of τ_{knee} was not possible [Figure 4-2]. Of the three possible inputs to $\mathbf{P_{knee}}$, the amputee is only able to control τ_{stump}, and must incorporate it with the fixed/predictable effects of τ_{knee} and τ_{ext} (when unperturbed) and generate a resultant output (Ψ_{knee}^{out}). Appropriate knee plant dynamic properties are sought that make useful output patterns possible for the amputee. This needs consideration of how τ_{stump} will interact and be controlled by adaptation of the hip target task vector Ψ_{hip}^{T} [Figure 4-3]. A well-designed knee mechanism with good intrinsic mechanical properties should therefore require less adaptation of the Ψ_{hip}^{T} pattern to yield appropriate τ_{stump} patterns to result in the desired knee motion.

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[29] It has always been possible to fashion a simple wooden stump with no articulation.

Unfortunately, it has not been possible to achieve the ideal $\mathbf{P_{knee}}$ dynamics, which enables amputees to produce the natural Ψ_{knee}^{out} pattern. Therefore, amputees have to adapt and generate limited target patterns. For example, tuning of the dynamics can be made via knee damping mechanisms that adjust the ballistic properties of the shank; such methods are unable to produce the ideal swing trajectory according to Whittlesey et al. and Jin et al. [221, 113]. Even more profoundly, many conventional prostheses during stance transform the τ_{ext} input (in the form of ground reaction forces) into a locking action at the knee mechanism. Although it fulfils the most basic requirement for the stance phase, i.e. support, it does not come close to matching the sophistication that is observed for the natural system. Similarly, the knee joint can be aligned behind the normal position thereby altering the plant dynamics to achieve increased stability because the ground reaction force is more likely to pass in front of the displaced knee joint. The amputee can also aid stability by modifying their τ_{stump} pattern by pushing into the posterior wall of the socket so that an overall extension moment at the knee is created by this hip extension. This also means that the ground reaction force can aid with stability as it now goes further in front of the knee joint. Prosthetists therefore have a key role by influencing the plant dynamics according to the fitness of the amputee, e.g. a stronger amputee that can exert greater moments to extend the knee will require less alignment stability.

As a final note, the importance of the transformation $\mathbf{P_{knee}}$ should not be neglected even if a fully controllable system is implemented. For example, its importance is evident even with the natural system where despite the highly sophisticated CNS much of the smoothness of movement can be attributed to the mechanical properties of muscle according to Myers et al [149].

The best way to address this deficiency with $\mathbf{P_{knee}}$ design is to generate a controlled input τ_{knee}, and this indeed has been argued in various forms by many researchers. Classically for example, Flowers and Grimes [76] developed a prosthesis simulator that used an external hydraulic power source to produce any desired moment inputs to the knee, and test different control schemes. Using this system, they were able to reproduce the action of a normal knee joint during level walking by forcing the knee to produce normal knee angle patterns as previously recorded on non-amputees.

Amputees found these one off simulations a definite improvement that gave validation for the desire to actively control a prosthesis, although an autonomous implementation of a control system would be required.

Jin [113] further supported this viewpoint through modelling and showed that constant frictional moments (generated in conventional knee designs) are incapable of resulting in the desired knee angle patterns. Similarly, Whitlesey et al. [221] also demonstrated how neglecting inter-segmental moments during the swing phase could be very limiting since simulations required muscular control to be accurate. There was no correlation with unaided pendulum swing times and human walking periods.

The following therefore reviews some of the control schemes that have been attempted for τ_{knee}.

5.2 EMG CONTROLLERS

Several researchers have attempted to use myoelectric (EMG) signals produced by stump muscles to elicit some control over the knee mechanism. The idea has shown relative success for upper limb prosthetics where voluntary intent could be established according to contraction of various muscle groups; however, it is more problematic for use with the lower limbs.

As a foreword, it should be mentioned that because of the high frequency content of raw EMG signals and their bipolar nature, signal processing is usually undertaken. The result is a signal that represents the envelope of the signal or its energy content; strong contractions can thus be distinguished from weak. In addition, some researchers have observed that information can be gained from the frequency content of the EMG signal. Therefore, when reference is made to a processed EMG signal it will imply that the signal has been amplified, possibly filtered and subsequent processing applied to reveal the information content of the signal.

5.2.1 SAXENA'S & HORN'S SYSTEM DESCRIPTION:

Horn [106], and Saxena and Mukkopadhyay [178] used EMG signals to enable the locking and unlocking of the knee mechanism. Horn used a clutch type mechanism to prevent flexion only; it had limited voluntary control because once EMG signals had triggered locking, full extension was required to mechanically release the lock.

EMG **relaxed = 0** **contraction = 1**	**Heel Pressure Transducer**	**Knee State; unlocked=0, locked=1**
1	0 (swing)	0
0	0 (swing)	1 (or same as previous state)
1	1 (stance)	1
0	1 (stance)	1

Table 5-A; Logic for either locking or unlocking Saxena's knee mechanism. EMG signals are from the Vastus Lateralis muscles and the pressure transducer is on the heel of the prosthetic foot.

Saxena's system was slightly more sophisticated as an additional pressure transducer placed on the heel of the prosthetic foot was used to give some artificial feedback control. The logic presented in Table 5-A was used: The knee remained ***locked - unless*** a ***high EMG*** signal **AND** a ***low-pressure signal*** was present.

Since the knee would be locked if a signal were present from the pressure transducer, this acted like an electronic implementation of a weight activated mechanical locking device. This would act as two-point feedback [Appendix C], which forces the system towards the locked state if the pressure signal is high; in this case, τ_{ext} does not act directly on the knee plant as in the conventional weight activated prosthesis but is incorporated into the τ_{knee} response.

A high EMG signal derived from the Vastus Lateralis muscle group was required to unlock the knee for the swing phase. This muscle group was supposedly chosen because it is not used in the locomotion cycle so that the amputee may independently control it to produce a desired signal. Once unlocked, the knee would remain in this state regardless of muscle relaxation until the stance phase occurred, allowing the amputee to only control unlocking for the swing phase and no more. Horn's prosthesis differed in that the amputee could lock the knee at any time desired, although unlocking could only be achieved when the knee was extended.

Discussion of Saxena's system: Although the basic idea of Saxena's system sounds quite simple it is more complicated to understand exactly what is happening in terms of the general control structure that was previously outlined [see Figure 4-3]. For this system, the value of τ_{knee} can take on one of two values, i.e. a very low value (unlocked) or an extremely high over riding value (locked), so how might feed-forward and/or feedback components (τ^{A}_{knee} and τ^{R}_{knee}) be used to yield the desired states?

The state of the pressure transducer may be used for feedback control by making it an element of the output task vector Ψ^{out}_{knee} and setting its corresponding target element in Ψ^{T}_{knee} to one. The error $\Delta\Psi$ would then be zero if a pressure signal were present and one otherwise. It is thus possible to envisage how such error signals may be transformed using a simple step function for $\mathbf{R_{knee}}$ to dictate if τ_{knee} has to be in the locked state or not.

A high EMG signal signifies the unlocked state, and low the locked state. This will represent the **intent** of the amputee. A simple feed forward transformation within the controller may convert this target state to a response $\tau^{\mathbf{A}}_{knee}$, which in this case will be selecting full moment or no moment. The Boolean operator "**OR"** may be applied to the feed-forward and feedback components to produce the resultant τ_{knee} effector output.

Maintaining conventions, the task vector may be written $\Psi_{knee}\begin{pmatrix} pressure \\ kneestate \end{pmatrix}$, where *knee state=1* is locked; the target vector may therefore take two forms state1 = $\Psi^{T}_{knee}\begin{pmatrix} 1 \\ 1 \end{pmatrix}$ or state2 = $\Psi^{T}_{knee}\begin{pmatrix} 1 \\ 0 \end{pmatrix}$ depending on the amputee's implied intent. All of the logic states for the response of the knee actuator τ_{knee} can therefore be derived according to the outcome Ψ^{out}_{knee}. For example, if the amputee indicates state2 and the current outcome is $\Psi^{out}_{knee}\begin{pmatrix} 1 \\ 1 \end{pmatrix}$, then τ_{knee} can only force the knee to lock, whereas $\Psi^{out}_{knee}\begin{pmatrix} 0 \\ 1 \end{pmatrix}$ causes it to unlock.

Although a little convoluted compared to logic states, the process of representing control systems in the generalised format discussed in section 4.3 was a powerful tool to understand what might be happening and an attempt has been made to fully illustrate the control scenario [Figure 5-1].

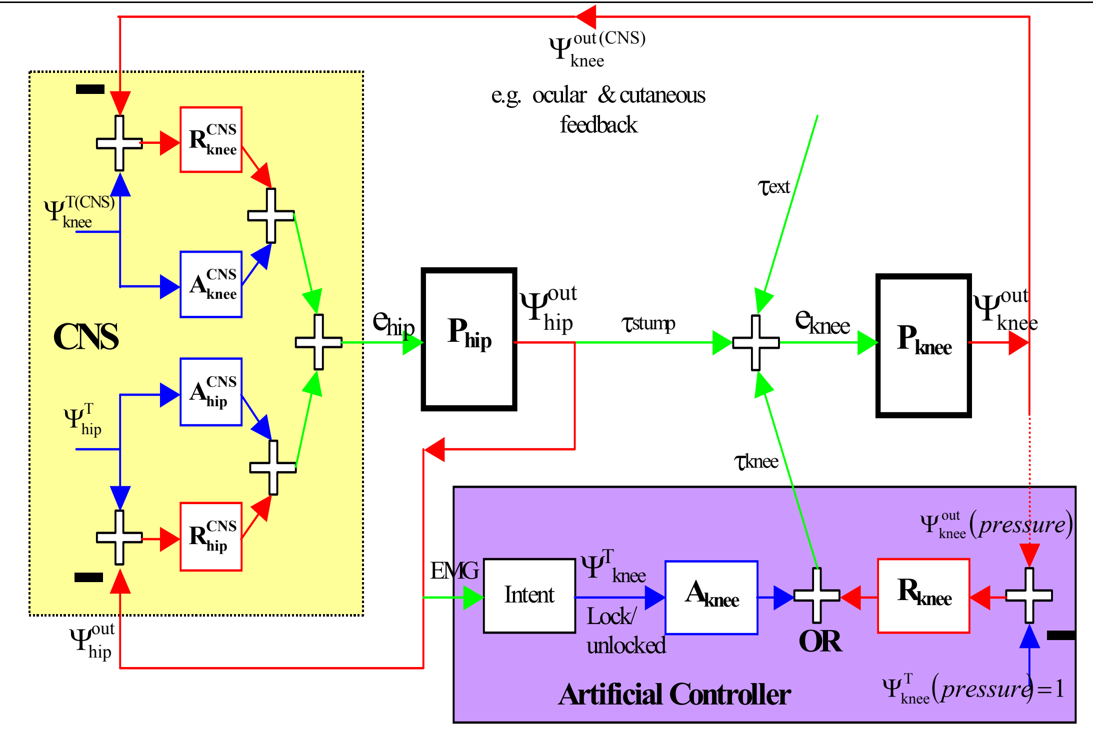

Figure 5-1; Envisaged control system for controller proposed by Saxena [178]

The output of the knee effector τ_{knee} is not very sophisticated as it's limited to two states; the CNS must however be highly involved in order to produce any meaningful/useful result. It must convey to the artificial controller what the state of Ψ^{T}_{knee} should be (i.e. indicate locked or unlocked), according to the incremental demands of the task being undertaken e.g. walking. This problem requires the CNS to order and link these knee states (by learning), a process similar to Bernstein's control structures for motor engrams of movement [section 3.2.3]. The assumption is that the brain will be able to use its problem solving capabilities to decide the optimal movement patterns for the prosthesis given its limitations; this movement pattern may be represented $\Psi^{T(CNS)}_{knee}$, (i.e. the knee target vector used by the CNS). This means the CNS needs to create feedback ($\mathbf{R}^{CNS}_{knee}$) and feed-forward ($\mathbf{A}^{CNS}_{knee}$) transformations that act to make the desired knee movement more likely, i.e. by tensing the Vastus Lateralis at the appropriate times. Consequential to tensing, the hip plant ($\mathbf{P}_{hip}$) must be influenced, which needs care as its output τ_{stump} will also be affected, and fed into the knee mechanism. The diagram includes such issues where it can be seen that the control schema for the hip dynamics must be combined with the knee dynamics, a situation reminiscent of Bernstein's multiplicity problem where many control centres could influence many effectors but their result had to be co-ordinated [section 3.2.3, Figure 3-2].

Such an additional schema for knee control within the CNS should therefore ideally act without compromising the output of the hip dynamics (Ψ_{hip}^{out}); a demanding undertaking for the CNS if a novel use of muscles is required. Saxena may have fortuitously avoided such problems by choosing the Vastus Lateralis because it solely acts as a knee extensor with no effect on hip action. In addition they claim these muscles are not used during the gait cycle, however it comes to mind that the process is likely to be very cognitively demanding especially considering the anticipation that is required for the feed-forward component of the CNS control to be effective, i.e. $\mathbf{A}_{\mathbf{knee}}^{\mathbf{CNS}}$.

As mentioned, the CNS may also partially use feedback control to dictate the intent to the controller via Ψ_{hip}^{out}. It must therefore monitor its manifestation on the knee output, hip dynamics and more generally on the body dynamics, e.g. using ocular and cutaneous feedback routes. This method of feedback differs from the intact system [section 3.3.3], which cannot depend upon *fast* - hardwired reflexes and full proprioception. The transformation $\mathbf{R}_{\mathbf{knee}}^{\mathbf{CNS}}$ must be developed and learnt over time - relying more on clues than an integrated feedback system so that it is likely to be much slower than the "normal" system. Therefore, to save time and concentration, it would be beneficial for the artificial controller to take over much of this processing required by the CNS.

As can be seen from this description of the possible control methods, there would be a tremendous amount of work involved for the amputee to become skilful at using the prosthesis if complex interactions of locking and unlocking are required. It is therefore more likely that the prosthesis will be used as simply as possibly otherwise it is believed that this method would be too mentally demanding with errors creeping in as the amputee becomes fatigued. It is possibly for this reason that the group constrain voluntary control to the swing phase period. The idea does however possibly have greater potential for control of non-autonomic responses that can represent random movements the amputee may wish to perform on a whim, or slow static events such as when standing.

5.2.2 EMG THRESHOLDS AND PROPORTIONAL CONTROL:

The EMG signal in the example by Saxena et al was used to indicate one of two output states of the actuator. Other researchers have attempted improvements by defining more states according to threshold values. The natural system has two mechanisms to increase the forces that it generates; this is by increasing the firing rate of muscles and by recruiting more motor units. The first method will be detected by the temporal patterns of an EMG signal whereas the second method will cause a spatial distribution of the EMG signal. Therefore, single electrode EMG devices will always be limited in the accuracy to which they can represent the forces produced by muscles. This has meant that, to achieve reliable operation, most researchers have had to limit the number of states per EMG channel to three. The technical problems in trying to produce such states have been researched by several researchers e.g. Daley et al, and Morin et al [52, 141]. They describe how the magnitude of the operator error can exceed the system error associated with the controller and such ideas can be important in optimising the decision boundaries to distinguish different actuator states. They describe for example, an operator relying on visual feedback (for hand movement) that can overshoot the response if there is a delay of 66ms in viewing the response of the hand.

Similarly for lower limb control, Dyke et al [70] encountered this problem when they first tried to implement proportional EMG control because they had to eventually limit control to three states of knee resistance. Having an increasing number of states also increases the complexity of the transformations required by the CNS because of a greater degree of freedom, and smaller margins for error given the diminishing intervals between decision boundaries.

The possible conflicting activities of controlling the hip plant and the knee discussed for Saxena’s system were apparent in Dyke’s system. When they used the Rectus Femoris as the source muscle group, an inadvertent (unintentional) locking would appear when they tried to walk, although the amputee was able to select the required resistance whilst sitting. Although the situation was improved by using the semibranosus, it gave little confidence that other similar problems would not arise during other untested tasks.

The general principles outlined in the discussion of Saxena's controller may also be applied to this system, although because the pressure transducer is not used, the methodology applies to the entire task (rather than just swing phase).

Although proportional EMG control had not been possible using one muscle group, researchers such as Myers and Moskowitz and Triola and Moskowitz [148, 147, 203, 202] looked at using multiple EMG channels usually recording different muscle groups to gain improved resolution of the intent.

5.2.2.1 MYERS AND MOSKOWITZ'S SYSTEM

Myers and Moskwitz went onto to create a sophisticated proportional[30] EMG controller by combining the signals from initially nine different muscle sources. They build two classifiers that categorised the combined hip and thigh EMG signals into signifying either a knee flexing moment, a knee extending moment, or no knee activity. Input samples, from model (non-pathological) subjects holding their leg in three static positions, were clustered according to whether external loads forced knee flexors or extensors to be active. By obtaining data from three positions they tried to ensure that classifiers would distinguish knee activity irrespective of hip activity during the swing phase.

It was also possible to correlate the positive values of the classifier output with the applied knee moment, while all negative values represented no knee activity, therefore giving a means of proportional control. The output of the flexion and extension classifiers could be combined to give an output corresponding to the direction and magnitude of the moment applied at the knee [Figure 5-2]. It should be emphasised that the summed output of the classifier derives from the external

[30] In this context **proportional** implies that the output was not limited to discrete states (i.e. it is proportional to the classifier response which directly relates to force). This matter is mentioned in order to distinguish from the output being proportional to EMG activity levels.

moments that the normal subject was asked to resist, therefore some conversion of this was necessary to make it compatible with the knee actuator.

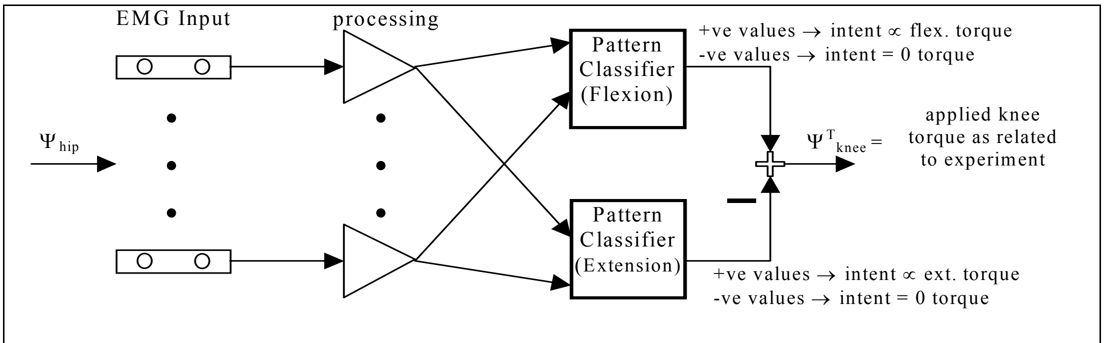

Figure 5-2; Myers & Moskowitz EMG Intent Recogniser

Myers and Moskowitz were eventually able to reduce the number of electrodes required to determine the intention of a non-pathological subject to 7 by examining which electrodes were least useful to the classifier. A simulation was tested on a trans-femoral amputee by asking him to *imagine* performing flexing and extending actions since direct external forces could not be applied to the amputee. The amputee reported that the output of the classifiers was proportional to his intended moment at the knee.

Although such a method is a distinct improvement over the two states EMG controllers of Saxena et al, it will still suffer their fundamental concerns involving cognitive demands, feedback speed issues, and multiplicity of control centres as discussed [section 5.2.1]. Its greatest benefit therefore possibly remains for slow, less complex actions of a purely voluntary nature.

5.3 MULTIMODE CONTROL SYSTEMS

Rather than using the amputee to directly control rudimentary variables such as the knee resistance, some pre-processing can determine when and how much resistance to apply. The controller "merely" needs to be told, or recognise what the intended task is so that it can implement the relevant knee behaviour. There appears to be a class of such prosthetic controllers, in which each task is considered a mode, and these are further divided into functionally significant sub-modes, thereby the general term multimode control systems. The sub-modes were arbitrarily defined e.g. sometimes to include as broad a scope as swing phase and stance phase, and other times reduced as single support and double support.

5.3.1 GRIMES & FLOWERS

As already mentioned, Grimes and Flowers developed a simulator whose angular velocity could be controlled using an externally powered hydraulic knee mechanism. They experimented with various control schemes with a variety of sensory inputs whilst using the amputee for interactive feedback of the controller. In this way, "human flexibility could be encompassed into the control model, and provide subjective feedback about the prosthesis performance".

In one of their investigations [89], they try to replicate the normal knee trajectory by separating the gait cycle into two control sub-modes according to swing phase and stance phase as inferred from foot switches. Stance phase mode was controlled using position feedback, whereas moment feedback control was used for the swing phase.

Positional control was specifically aimed at generating the knee-yielding (flexion/re-extension) pattern observed just after heel strike in asymptomatic gait. A target value of 0.17 radians was set for the controller and a high gain transformation applied. Thereby at heel-strike with full extension reached, the controller would flex the knee in order to try to attain the target value. However, because of the high gain there was a tendency to overshoot the target value, which required further correction by extending the knee again. Results were encouraging where amputees preferred this ability to flex the knee to a maximum of 16 to 20 degrees.

Swing phase mode was initiated as the plantar switch was closed (Heel Off to Toe Off). Moment feedback could be adjusted according to the gain; for example, high gains produced a low damping and fast response whereas low gains produced high damping and a sluggish response. As the mechanism used hydraulics with viscous damping forces that were proportional to velocity, only a constant gain was required for most of swing phase, although to prevent impact at full extension, additional damping was provided by dividing the gain by a factor proportional to the knee angle (max at full extension).

Although the system was basic, the resulting gait-patterns were still an improvement on conventional systems and highlighted the advantages of being able to electronically tune the system to suit the amputee (a factor specifically exploited in the intelligent prosthesis [section 5.4.1]).

The group now also including Stein and Flowers [191] went on to refine and improve their control system by trying to replicate the trajectory of the sound knee during the stance phase. Echolocation was used to obtain the sound side trajectory, although modification of the pattern was required because the sound side knee is flexed more than in non-pathological subjects. A cadence dependent gain was sufficient to correct the trajectory and match the maximum knee angles observed in non-pathological gait. Therefore, by forcing the knee to follow trajectories during stance similar to non-pathological, they found that the maximum vertical height of the hip intriguingly increased (as compared to conventional prostheses) although loading forces on the prosthetic foot differed. They concluded that the increase in height was due to the particular combination of geometry, alignment, and compliance of the SACH[31] foot. At the maximum hip height, the shank was angled forwards so that a bent knee allowed the thigh to be aligned more vertically than a straight-legged conventional prosthesis could. It is therefore not necessarily the case that trans-femoral amputees should replicate the non-pathological knee trajectories since much depends on the foot.

Grimes et al [90] advanced their controller by incorporating modes additional to walking including standing, sitting, running, stair climbing, and ramp climbing, although the sound side remains the template for the knee trajectory. More sub-modes are also included. For example, swing phase is now divided into three sub-modes to control knee buckling at the end of stance, prevention of excessive heel rise, and extension damping. Each one of these sub-modes used moment feedback to simulate pendulum damping although with different damping coefficients to produce the desired effect. Only two sub-modes were required for stair climbing in which stance was controlled using position feedback although the echolocation remained unmodified. For the swing phase, the knee was actively flexed using a spring with damping to prevent excessive flexion in order to clear the step. Although not counted as another sub-mode, extension control was applied once maximum flexion with zero velocity was achieved. Schemes similar to stair climbing were employed for ramp climbing. Finally, they also developed supplemental modes, which were

[31] Solid-Ankle Cushion Heel

not part of a cyclic task including standing still and stumbling. These supplemental modes were accessible from all other modes.

Before any transitions to sub-modes could be made, an intent recogniser was required to detect the particular mode the amputee wished to perform. Locomotion modes were recognised according to echolocation of the sound leg and were updated once per cycle. Specifically, they used the sound knee angle at sound foot contact to indicate whether level, ramp or stair climbing walking was to be the next locomotion mode. Knee angles less than 30^0 indicated level walking, between 30^0 and 62^0 gave ramp climbing, and greater than 62^0 indicated the next mode would be stair climbing. All locomotion modes therefore had to be initiated by the sound side at initial contact. When a new mode is detected, the controller only implements changes and activates a new swing phase sub-mode once the swing phase transition occurs. Prosthetic foot contact is used as the condition for stance phase sub-mode transition. The swing phase transitions are more varied however, dependent on the mode; for example, the transition condition during level walking is for the prosthetic knee moment to become flexile during double support, and for stair climbing the condition was prosthetic foot off at the end of double support.

It can be seen that transition conditions are chosen somewhat arbitrarily, their derivation based on trial and error or experience rather than governing principles. There is no good reasoning as to why one particular transition condition is used over another, and whether indeed it does represent a unique event pertinent to all amputees and movement patterns. It would be hoped that for this project, a more systematic approach could be developed. The reliance on echolocation was also a point of concern, because of reliability issues, complexity, and most pertinently lack of direct control. The third point is a concern because it means the control system is effectively led by the activity of the sound leg so that movements cannot be started with the prosthetic limb. The resultant delay in switching modes is also unlikely to be acceptable.

5.3.2 BELGRADE PROSTHESIS

Popovic et al [163, 162, 164] designed a more complex multimode control system based on what they termed Artificial Reflex Control (ARC), which, similarly to natural reflexes, implements mappings of sensory states to motor actions. They incorporate ARC into an expert system containing the rules (learned through "tricks of the trade") necessary to carry out the specified artificial reflex responses according to the specific (gait) mode.

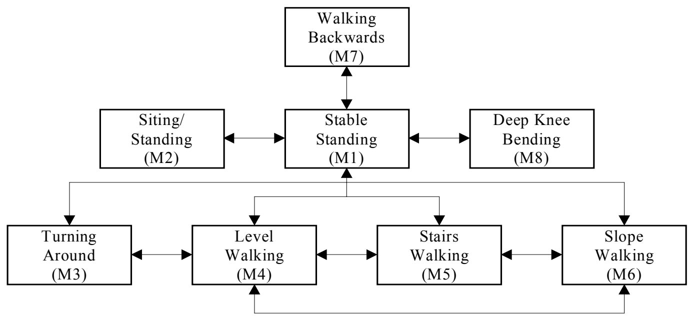

Figure 5-3 External Mode Rule Base indicating legal transitions. Adapted from Popovic et al [164].

The control system was composed of an inference engine, a database, a mode rule base, an operating rule base, and a regular rule base. Both the incoming set of sensory inputs and the previous set were stored within the database and these patterns used to activate a particular mode within the mode rule base. The mode rule base comprised:

- *External* modes [Figure 5-3], which were the basic activities such as walking, and stair climbing,
- And *internal* mode reflexes that included rules for gait initiation and termination by the prosthetic or sound leg, and adaptation to speed[32] and slopes.

Regular reflexes were also formulated which essentially contained the rules for the sub-modes; for instance walking mode was divided into push off, initial flexion, terminal flexion, initial extension, terminal extension, heel contact, knee bounce (flexion-extension), middle stance, and heel off. In addition to regular reflexes they included hazard reflexes to react to obstacles and sensory or actuator failure.

[32] Speed was determined according to the time interval between the maximal forces experienced by load sensors at the toe, and the heel. A linear correlation was found of this interval with both speed and terminal knee joint flexion angle.

Therefore, the inference engine would find a match for the sensory data in the database, and then accordingly update the set of rules within the operating rule base. The rules took the form of **IF-THEN** statements, which first allowed the state to be recognised and then a suitable response to be executed. For example, the organisation of the terminal flexion reflex was as follows:

IF knee angle < terminal value **AND** knee angular velocity < 0 **AND** prosthetic shank angle is behind the vertical **AND** the toe force < threshold force **AND** the heel force < heel threshold force **AND**

IF the previous reflex was initial reflex, **AND** initial flexion has been executed

THEN unlock the joint, and turn off the flexion motor unit.

In order to derive how much current is needed to drive the actuators during particular modes, they recently used machine-learning techniques to generate mappings between the gait pattern required for a particular mode and the controller output (i.e. current intensity) [163]. Simulation results from a fully customised biomechanical model were used to calculate the current intensities for particular knee angle trajectories. The trajectories were taken from an able-bodied subject with ankle-foot orthoses, which were chosen because it was considered a better trajectory for trans-femoral amputees than a purely non-pathological gait. They do however admit that there are many elements still to be improved in both modelling and mapping.

It is clear that the group have utilised a lot of expertise to develop these rules, however similar to Grimes and Flowers' controller, the ad hoc development of such systems does not allow for easy formalism of strategies and further expansion. As mentioned before, a design process that is more systematic was sought, rather than an arbitrary collection of outcome events to trigger some rule. Like other multimode control systems it would seem that the precise value of thresholds would be critical for the system robustness especially in the natural environment where amputees would not be the same as data gathered in the lab, and inter-subject variations could easily play havoc. Poor threshold values or indeed a poor choice of sub-modes might lead to situations where amputees become conscious of moving in such a way as to achieve necessary thresholds to trigger a desired mode.

5.4 COMMERCIALLY AVAILABLE CONTROL SYSTEMS

When this project was envisaged, the intelligent prosthesis manufactured by Blatchfords had just become available on the UK market. As the project progressed, another actively controlled prosthesis by Otto Bock was also introduced called the C-Leg, although because of its cost, it was generally not provided by the NHS. Both mechanisms use servomotors to adjust hydraulic or pneumatic valves and apply a level of control over the knee dynamics; the mechanisms are unable to act as a positive power source at the knee. As in multimode systems, they monitor a few key sensors to detect critical transition points. A brief description is given of both systems.

5.4.1 INTELLIGENT PROSTHESIS (PLUS)

This is an adaptation of a conventional prosthesis with ground force activated stance stability and pneumatic damping of heel rise during the swing phase. Unlike, the conventional prosthesis that requires the prosthetist to preset the pneumatic valves to one optimal value that is suitable for the amputee's "preferred" walking speed, this system automatically adjusts the valve settings during each stride to accommodate for speed changes. Since walking speed is calculated over the stride period, the response to speed changes is not immediate and takes one step which is a bit undesirable. The prosthetist has to first calibrate the valve presets to suit up to five different speed intervals. Amputees can thereby choose the range of speeds at which they would wish to walk, and preset the most appropriate damping characteristics for each speed setting.

Despite the system being quite simplistic because the controller does not truly shape the knee pattern, it still greatly increases the flexibility of a conventional prosthesis, and many amputees, as publicized by the questionnaire of Datta and Howitt [53], enjoy its advantages and certainly prefer it. However, there were still many aspects that remained the same as conventional prostheses; for example, subjects found no improvement for stair walking, and in another study, by the same group, the cognitive demands were shown to be the same [96].

Nevertheless, it must be said that the intelligent prosthesis is very restricted in terms of control, since it only tackles the swing phase and even this relies on pendulum type properties. An article by Whittlesey et al specifically concludes that swing phase is not a passive movement where a pendulum style swing does not suffice [221].

5.4.2 C-LEG

The C-leg uses hydraulics for both the stance and swing phase with the valves operated by servomotors. The stance phase behaviour is similar to the Mauch knee [133], which incorporates a high-yielding resistance to assist the amputee in stumble recovery, descending stairs, sitting down, and allows for slight flexion during the stance phase. However, unlike the reliance of the Mauch knee on mechanical cues, such as knee hyperextension to disengage stance resistance, the C-leg uses sensory information relating to knee angles/angular velocities, and ankle moments sampled at 50Hz to detect transition points. After discussions with Brian Wade and Wayne Jones from Dorsett Orthopaedic who were the principal C-leg providers to the UK, they comment that the active controlling device has the following advantages over the Mauch knee: "*increased stability* and *reliability,* and *less tiring".* For instance, to prepare the Mauch knee for the swing phase, a slight kick back was required to disengage stance resistance and allow swing flexion; unfortunately this would sometimes occur inadvertently, thereby startling the amputee, reducing their confidence and resulting in their reverting to safer, less efficient gait styles. The C-Leg is able to make this transition more certain by using pre-determined threshold values, related to the foot load, to trigger a change to swing phase hydraulics [154]. The value (heel rise threshold) was set during fitting by getting the amputee to stand and put as much force as possible on the toe; when this threshold was attained during the gait cycle, the stance resistance would be decreased ready for swing phase damping. The software also allowed for dynamic fine-tuning of this threshold to try to ensure good timing of this transition during the gait cycle.

The C-Leg could further be customised for the amputee by tuning settings for *swing phase extension damping, stance phase yielding damping,* and *extension damping.* These settings are intended for walking and stair climbing and are accessed when the system is in the 1st mode, however a 2nd mode is also available in which it is possible to set up a knee angle dependent flexion damping, with settings for the start angle and the angle dependent gain. The second mode makes the prosthesis more suitable for other activities such as biking (no damping), and standing around (locked in slight flexion). To switch modes however requires performing an unusual voluntary movement involving rocking three times on the forefoot.

The author was fortunate enough to witness an amputee evaluating the C-Leg at the Bioengineering Unit during which a few issues were highlighted. The main one it seemed was that she was unable to achieve sufficient heel rise since swing phase damping could not be set any lower. It was possible that she was either too light or not powerful enough for this prosthesis. Since Otto Bock do not advise on minimum weight restrictions and the C-Leg is intended for moderate to high functional level amputees, the latter reason seemed more likely although this was surprising as she normally walked well. For the same reasons possibly, it was also difficult to set an appropriate heel rise threshold level, which made the swing phase difficult to initiate properly. If the threshold for example was set too high then the resistance would remain too high, whereas a low threshold would flex the knee too early causing collapse; this is an example of the earlier criticism made for the Belgrade prosthesis regarding thresholds.

In general, however, amputees have reported positively about the finite control that the C-Leg gives, especially over stability, cadence, and gait [98]. One person comments that both the C-Leg and the intelligent prosthesis reacted similarly to speed changes during the swing phase. This is possible because, like the intelligent prosthesis, a continual time variant control pattern is not generated. Instead specific dynamic properties of empirically determined phases are adjusted. Both commercial systems tune the dynamics of the knee according to pre-established principles based on conventional prosthetic systems.

Overall, it can be seen that active control is certainly preferable to the passive conventional devices. The author's original conviction that active control should lead to a more natural behaviour of the prosthesis is reinforced [chapter 1] but there remains the systematic design process to truly give this behaviour. Much more thought is required of all the likely interactions and the best way to proceed.

5.5 SUMMARY

Several styles of prosthetic control systems have been reviewed and categorised as conventional, EMG control, and multimode controllers. An examination of these systems was made in light of the control structures developed in chapter 4. With this basic structure, it was possible to perceive the complexity of the interactions even for the conventional prosthesis that had no active control element. A conventional prosthesis has no effector source to generate τ_{knee} so that only the stump effector can exert a controlling influence τ_{stump}. Prosthetic design is therefore aimed at tuning the knee dynamics ($\mathbf{P_{knee}}$) so that an amputee can easily produce a τ_{stump} profile that results in a desirable outcome pattern for the task. The prosthesis in this situation is used as a tool; much like a tennis player may control a tennis racquet. The amputee learns the dynamics of the prosthesis and using a combination of forward planning and direct mechanical feedback generates a control strategy for the stump movement. Tuning of the plant dynamics however has not been able to replace the lost functionality of the natural knee. The method is hampered by the implied rigidity of only having a fixed transformation $\mathbf{P_{knee}}$ where normally much variability and flexibility is required.

EMG systems in contrast can additionally influence the outcome pattern through control of the τ_{knee} profile generated by the knee mechanism. Unfortunately, this influence is inferred through muscle activity usually at the stump. This can mean that the amputee has to generate a strategy not only to once again deal with the knee dynamics but also to deal with producing a τ_{knee} pattern. In effect, there may be a conflict with the muscle activity to generate τ_{knee} and the τ_{stump} profile. The complexity for the CNS was illustrated with a flow diagram that showed how various elements might interact. It was concluded that EMG systems incur much delay with feedback[33] and they ultimately suffer from high cognitive demands. They are however very useful to indicate slow highly volitional (impromptu) modes of behaviour.

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[33] The CNS must first receive feedback as to the state of the system (e.g. through visual or cutaneous) sensors, then it must develop a feedback and feed-forward control strategy based on the feedback information.

Finally, various researchers have been working on Multi-mode controllers, all of which come in different forms and guises. Generally, there is an increased level of strategic processing undertaken so that demands on the CNS are potentially reduced, and a greater degree of influence is exerted over the knee behaviour. In general, they all need to determine "*what* response and *when* it is required", and toward this, they seem to share the following requisites:

1. Dividing behaviour into tasks or functional modes and sub modes
2. Recognition of different modes and their onset.
3. Deriving appropriate responses for each particular mode[34].
4. Employing rules to govern the interaction of modes thereby enabling the execution of the appropriate response

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For example, many researchers such as Grimes and Flowers have clung to the convention of breaking walking into stance and swing phase to set their sub-modes. Others have extended modes to also include activities such as standing, sitting, running, stair and ramp climbing such as with the Belgrade prosthesis. The modes in general have been defined according to conditional thresholds being reached using a variety of sensors ranging from foot switches, knee angle sensors to echolocation devices and force sensors. Responses have included moment and positional feedback, where even the dynamics of the feedback controllers have been used to benefit such as high gains causing overshoot and mimicking knee yielding just after heel strike. Feed-forward control has been undertaken with the Belgrade prosthesis in which its response was based on able-bodied movement patterns. Generally, the systems employed rules, utilising conditional statements, to determine dynamic interactions between mode changes and mapping of responses with variables such as cadence.

The Intelligent prosthesis and the C-leg were the only commercially available systems. They were both largely based on conventional design but had the ability to tune dynamic properties on the fly, which normally would be permanently set by the

[34] Unlike EMG systems in which the CNS directly dictates the τ_{kee} profile, these control systems can be set up to produce any output profile as a response.

prosthetist. In control terms, they both seem more related to multimode controllers and given their conventional mechanics, they naturally adopt stance and swing modes of behaviour.

The C-leg however does additionally offer other modes including standing and bike riding.

Overall, for multimode controllers, much expertise was required to determine modes and iterations, it would be beneficial to derive a systematic approach to determine such events so that a greater extent of non-pathological behaviour can be captured and mimicked more accurately.

6 THE CONTROLLER

6.1 DESIGN VARIABLES (DV)

During the preceding sections, reviews have been given on topics that were inspirational towards understanding what is required for this project. A generic structure was also derived [chapter 4], which for the author simplified many of the circular arguments that could take place with respect to control. It allowed a clear way of seeing various interactions that are likely to occur when different methods are employed. It highlighted what were the most critical factors that must be harnessed when trying to design a controller. To summarise, it was realised that to design a control system that produces knee patterns suitable for the aims of this project, the following criteria or design variables (DV) were required:

DV 1. The definition of specific **tasks** or actions to be accomplished, so that the full range of motor patterns that represent a particular task may be identified. This involves knowing the extent of the motor patterns that a task may encompass. For instance, if the motor pattern for a task has a frequency dependency, then a completely new motor pattern may become preferable when a threshold frequency is reached. In terms of Kelso's nomenclature, [section 3.5], each task/motor pattern may be linked to different **attractor states.** These states will exist for periods along the dependent variable continuum (e.g. frequency), this dependent variable is known as the **control variable** and is chosen because it is key to inducing a change to the attractor state (this should not be confused with "controlled variable" used in control systems terminology)

DV 2. The designation of suitable target patterns Ψ^{T}_{knee} (see section 4.3 for definitions), to encompass the demands of each particular task (as defined above) that the controller is to assist. This requires first that appropriate outcome variables are selected as elements of the vector Ψ^{T}_{knee}, which will enable the constraints listed in section 4.2 to be realised, i.e. so a compatible outcome results. Thereby for each individual task, a separate set of patterns that will be portrayed by the vector Ψ^{T}_{knee} will have to be established. These target patterns must include any dependencies (e.g. on control variables or external influences) that the task may have so that the correct spatial and temporal patterns are accounted for. This may be achieved much like a look up table or through modelling of its effect.

DV 3. An **Intent Recogniser** in which the controller monitors the amputee's movement patterns so that:

a. A task may be associated with the movement pattern.

b. The correct Ψ^{T}_{knee} pattern is associated/selected for the particular task that is to be performed. This pattern should include any dependencies the particular task may have, for example on speed, gradient etc. In effect, during intent recognition, the value of the control variable must also be realised so that it can be used to model or lookup the final target pattern.

c. The initiation of the selected target pattern is accurately timed thereby preserving the co-ordination that is necessary between the Ψ^{out}_{hip} output and the selected Ψ^{T}_{knee} pattern.

DV 4. The definition of **transformations** necessary to make the output Ψ^{out}_{knee}, closely resemble the selected input Ψ^{T}_{knee}. This will include considerations of:

a. The plant transformation $\mathbf{P_{knee}}$

b. The control transformations, whether Feed-forward ($\mathbf{A_{knee}}$), Feedback ($\mathbf{R_{knee}}$), or a combination is more suited to the problem.

6.2 HOW TO FIND SOLUTIONS FOR THE DESIGN VARIABLES

In resolving the above issues, one is trying to find out precisely what the knee is supposed to do and how it should be doing it. The approach taken by the author was to imitate the patterns observed from the natural system and hopefully from this determine the elements outlined previously. Although there is much controversy with regard to how motor control is achieved in the natural system, it was hoped that sufficient information could be gained in order to answer the aforementioned criteria. Therefore, assessment of natural motor patterns will be used to find rudimentary requirements of an artificial trans-femoral knee controller.

The following will now describe the approach that was taken by the author in answering each of the issues that have been raised to make the control elements. This section is intended to show what the author considered the optimal approach towards satisfying all aspects of this project, i.e. solving each of the above elements whilst keeping within the bounds of the constraints.

6.2.1 (DV1) HOW CAN TASKS BE DEFINED

The first requisite was to define the knee movement patterns needed to perform different tasks. The tasks that the controller would be required to deal with were those that resulted from, what is loosely termed, autonomic control processes. Rather than the strict definition of autonomic control, the term is used here to emphasise tasks that require little conscious intervention [constraint 2, pp. 113]. It is suspected that such motor patterns rely to a certain extent on central pattern generators [section 3.3] in which neural circuits produce self-sustaining patterns of behaviour that are often based on pre-established feed-forward patterns. This differentiation in the tasks that the controller was to undertake was felt justified for the lower-limb because most useful motor patterns are of this nature, e.g. various forms of gait. If the knee is to be controlled by purely voluntary methods then this creates problems of concentration for the amputee, especially for tasks that would ordinarily be achieved by autonomic processes. Such disadvantages have been described in the review on other control systems that respond to higher-level conscious processes, e.g. the EMG based controllers [section 5.2]. A hybrid system however should not be ruled out, where the EMG type systems may complement the feed-forward approach if the amputee wishes to perform non-autonomic actions.

The restriction to performing autonomic type of movements, [constraint 2, page 113], alludes that some statistical repeatability with regard to various features of a task that a subject is performing may be expected. Although the CNS may be trying to repeat some unknown abstract topological feature, other characteristics may also consequently be repeated when the task is performed. For instance, experience enables one to easily recognise that a subject is performing a particular autonomic task such as walking; this indicates that there are features of walking which are consistent and unique. The observer is even able to distinguish if there is a deficiency in the motor pattern simply by noting differences from a regular pattern. It was therefore hoped that, in a similar manner, some of these repeating features would also be reflected in the biomechanical data that could be gathered for the movement. Such invariant features, irrespective of task dependencies such as speed, would be beneficial towards deciding the patterns that the controller should produce.

Therefore, experience dictates that walking is an applicable autonomic task for which knee control is desirable since one does not think about it as it is performed. The review on gait [section 2.2.2] even describes the high repeatability of different outcome variables that were measured whilst performing this task. In fact, it was possible to derive various empirical relations based on these outcome patterns, which remained relatively constant despite differences in who performed the task, when they performed it, and the speed they achieved [section 2.2.2.1.1]. For instance, the ratio of the relative[35] stride length to the cadence (taken over the linear portion of the plot) appeared to be very repeatable from day to day and even from subject to subject.

Such simple observations immediately indicate that the task of walking may be dependent on both morphological properties and the speed of walking (which also has a morphological dependency since it is equivalent to the product of stride-length and cadence). Although such dependence was derived for a global view of the overall movement, these variables may also be pertinent to the knee motor patterns because of the requirements for co-ordination.

The speed of walking has a further repercussion on the task that is defined because it cannot physically exceed about 3m/s [section 2.2.2.1.1]. Just before this threshold, it

[35] Normalised with respect to height

appears that subjects would naturally start to jog. Jogging may therefore be classed as a different task to walking, as it appears to utilise a different motor pattern. The CNS may in reality use another variable rather than speed with which to set a threshold value to decide when to use the jogging motor pattern (e.g. a discomfort factor). However, unless this is discovered, speed will possibly suffice. This situation may also be viewed according to Kelso's theories, in which, according to his terminology [section 3.5], walking and jogging may be the distinction between two different stable attractor states where speed may act as the control variable. Therefore, for low speeds a potential function is formed in which both attractor states may co-exist, i.e. walking or jogging, whereas for high speeds only jogging can exist. For example, Collins observes loss of stability at the preferred transition speed (PTS), in which there is a switch back and forth between jogging and walking [47]. Others such as Hreljac have tried to identify such criteria [107, 108] using both kinematic and kinetic parameters to look for a sharp transition at the PTS. Although their results were inconclusive, there seemed to be a large dependency on discomfort factors relating to the ankle. The main dependencies were with angular accelerations and velocities, which possibly relate to the safe stress limits of tendons/tissues at the ankle as the PTS is approached.

As a final note, although the morphological parameters of a task cannot really be classed as control variables, their effects nevertheless have to be taken into account. This is because it should not be necessary to customise the controller for each subject but only to set the values of such parameters [constraint 4, page 114].

To summarise, it was important to have some method of encapsulating all motor patterns representative of a particular task, and ensuring that transitions to new tasks with separate motor patterns did not cause confusion for the controller. This principle entails finding the control variable of the task, and the consequential thresholds of this variable before a new task is adopted. In particular, attention was focussed on walking gait, as this was to be the demonstration task. It has been realised that the term walking encompasses many variations in the movement patterns and these will all have to be accounted for when the target knee task vector pattern is selected.

6.2.2 (DV2) DESIGNATING TARGET OUTCOME PATTERNS (Ψ^{T}_{KNEE}) REPRESENTATIVE OF TASKS

It may thus be expected that the “ideal” outcome pattern would be to imitate biomechanical patterns of non-pathological subjects as they can perform the selected autonomic tasks (walking in this case) proficiently using a good level of control. Unfortunately, there is a problem with this because the circumstance experienced by non-pathological subjects will be dissimilar to the situation of a trans-femoral amputee with a controllable knee device. In particular, the forces/moments, τ_{ext}, applied by the external/distal effector of the amputee will not match the corresponding forces/moments produced by the non-pathological person. Since τ_{ext} is produced by the transformation of ground reaction forces, moments of inertia and forces caused by muscles (for the natural system) this leads to striking differences between the non-pathological subject and the amputee, for example:

- If $\mathbf{P_{ankle}}$ represents the dynamics at the prosthetic ankle joint, then this transformation is unlikely to exactly match that which exists for the natural ankle. The complexities of the natural system have not been fully reproduced despite numerous prosthetic foot, ankle, and pylon designs, which attempt to replicate the natural transformation (e.g. by incorporating various shock absorbing materials). Therefore given the same inputs to $\mathbf{P_{ankle}}$ (e.g. ground reaction forces), the outputs from the natural and artificial $\mathbf{P_{ankle}}$ cannot be identical, which is significant because this output partially contributes to the knee input τ_{ext} as derived from distal sources.
- Since this project is intended for controlling an active knee mechanism, only this device will be powered. The prosthetic foot and pylon mechanisms will not contain an internal power source with which to try to actively produce controlling forces at the ankle joint. The non-pathological system in contrast has muscles with which to exert such internal forces across the ankle joint and thereby ultimately influence the knee input τ_{ext} in a way that the prosthetic system cannot. The natural foot, for example, can plantar flex and generate considerable energy, usually transferred to proximal segments. Prosthetic feet can only vaguely approximate this by using energy storing mechanisms that release the energy

when the loading becomes less and not really as it is necessary. This response is the consequence of a purposely-designed $\mathbf{P_{ankle}}$ transformation and not the addition of a new torque source as found in the natural system.

Therefore, as the trans-femoral system to be controlled will experience a different τ_{ext} input to non-pathological subjects, there is little point in replicating the natural τ_{knee} pattern since this will still result in a different outcome Ψ_{knee}^{out} for the amputee. It cannot be assumed that this different knee outcome pattern would still be beneficial.

If non-pathological subjects were to be used to provide model data, the likely differences in τ_{ext} should ideally be compensated, i.e. the non-pathological subject should move as if they were walking with a prosthetic foot. The author did not feel it was practical or accurate to make subjects walk in this way, for example by using an orthosis to restrict the effects of muscles across the ankle joint (e.g. as attempted by Popovic [section 5.3.2]). It was also thought that trying to compensate for these differences by modelling would be futile because adaptations in the motor output of the knee and hip would also be expected to occur. Such adaptations would be very difficult to predict, making it difficult to calculate a new target knee pattern.

Therefore, an alternative solution was required which could include either modelling a solution according to a definitive objective of walking, or by finding an alternative more appropriate resource to imitate. In order to model an ideal solution many assumptions would have to be made as to the objective of walking, e.g. is the purpose to optimise energy efficiency, reduce stress, maintain a particular COG movement pattern etc? Then the whole movement pattern may be calculated from this requirement given the limits of a prosthetic foot. However, even if it were possible to define this purpose, it would be presumptuous to consider that the amputee would conform to this simulation, where the reaction of the stump to the knee pattern may be unpredictable.

Therefore, an alternative source for the trans-femoral motor pattern was identified which was thought to be more appropriate than the use of non-pathological subjects. It was hoped that Ψ_{knee}^{T} could be defined for various tasks using this source.

6.2.2.1 TRANS-TIBIAL AMPUTEE USED TO PROVIDE MOTOR PATTERNS FOR CONTROL TASKS

It was *postulated that* ***trans-tibial amputees*** *could perform tasks in a manner that was most suitable for the active trans-femoral knee device*. These amputees are likely to experience an input to the knee (τ_{ext}), similar to that of the trans-femoral amputee, but they still have control of their knee with the remaining musculature. Consequently, they have developed knee control strategies to account for the difference in τ_{ext}, whilst still maintaining the natural style despite the limitations of the prosthetic foot. Their CNS has facilitated such complex adaptations and possibly produced movement solutions for the task that will be far superior to artificially created ones. This is apparent where trans-tibial amputees are able to perform almost as well as non-pathological subjects, with many of the gait anomalies experienced by trans-femoral amputees alleviated. Furthermore, after some rehabilitation, trans-tibial amputees are able to move without too much conscious effort, indicating that the adaptation has become instilled and possibly integrated with lower level neural systems (e.g. incorporating central pattern generators with appropriate reflex corrections). This will be important when trying to identify motor patterns that the artificial system should reproduce. It is even plausible that the fundamental CPG patterns (i.e. as in non-pathological subjects) may still be in use, although modification for example via reflex responses will be required to cope with the effects of using a prosthetic foot and pylon. Whatever is the actual scenario, it remains important for control purposes that a regular motor pattern can be observed in order to ascertain the movement goals.

Therefore, it was considered a wise strategy to try to replicate the motor pattern encapsulated by the trans-tibial amputee rather than conjuring new movement patterns for the knee to perform. The second option would be extremely complex, especially as there is no consensus as to what is the low-level purpose of walking. In addition, even if this were known, a unique solution for the knee behaviour is not guaranteed (especially if the nervous system works to several different agendas depending on circumstance, and/or motivation).

6.2.2.2 WHAT TASK VARIABLES TO USE AND WHAT OUTCOME PATTERNS?

For this project, the chosen/demonstration task was to be walking, which as can be seen in the review is not expressly defined by any one single parameter. Despite this uncertainty about what best defines walking, something had to be used to describe walking in terms that the controller can utilise: this was already anticipated when formulating constraint 6. What are the essential variables that embody this movement for the knee? Once this is known it may be possible to assign values to them so that the Ψ_{knee}^{T} pattern can be defined. Therefore, as it is desirable to replicate the trans-tibial knee motion, control data must be obtained from these amputees. The experimental phase of this project will consist of gathering data, which is most appropriate for control purposes and that comply with the constraints. Since the variables and the patterns were not obvious, a method was required to extract this information in order to formulate the target task vector.

Because of the aspiration for inter-subject capability dictated by constraint 4, the controller will have to be robust enough to handle a large variance of signals. Therefore, sufficient data must be gathered to account for the range of signals that the controller is likely to face. Ideally, a morphological parameter should also be used for normalisation purposes to reduce the subject dependent deviation encountered in the data. Such normalised outcome variables that show little variation for both intra-subject and inter-subject comparisons are likely to be good candidates for the knee target vector.

The advantage of discovering such generic knee patterns that apply to most trans-tibial amputees is that it gives something for the controller to try to repeat in all cases (i.e. does not require customisation to the individual). The use of generic patterns also makes it more plausible that the trans-femoral amputee will revert to a similar hip pattern as the trans-tibial amputee if the artificial knee behaves similarly to the trans-tibial knee.

This assumption is based on the trans-femoral amputee's ability to regain hip motor patterns that are similar to pre-amputation by reusing the original central pattern generators (CPG) [section 3.3.2]. If this is the case then co-ordination may be an inevitable consequence as long as the controller is able to produce these desired patterns reasonably accurately and on time. The use of former CPG networks for hip motion thereby implies the involvement of more subconscious processes and thus facilitates constraint 2.

The morphological normalisation should enable closer comparisons of the amplitudes of different plant output variables, however there will still be an additional temporal and spatial variation due to speed. To overcome this, care must be taken; it cannot be assumed that it is possible to compare data of similar speeds for different subjects. According to the review sections, the walking speed is itself actually dependent on morphological parameters. To illustrate, a short person may use a higher cadence than a taller person walking at the same speed may; their stride-lengths must be different. Therefore, a morphological normalisation of the speed of walking may also be beneficial. The cadence may afford such an impartial comparison as it is, effectively, the walking speed normalised with respect to stride-length, or alternatively the square of the cadence is proportional to the speed normalised with respect to height [section 2.2.2.1.1].

Before any invariant outcome patterns could be presumed as precursors to the patterns required by control processes, phase lags between the control signal and the outcome must be considered. This is evident with the natural system where neural delays and muscle lag characteristics result in phasic relationships between the EMG and muscle tension profile [80]. Such lags will be even more pronounced when the effects are observed on outcome variables. Therefore, for control purposes it would be desirable to know what control events result in what outcome patterns. Since the CNS may be trying to produce movement patterns that are defined by abstract topological features [20] as Bernstein conjectured, it is very difficult to second-guess these features, especially in the temporal dimension. For example, although it is possible to compare data simply by looking at the variance of amplitudes, these amplitudes for repeating comparisons must be temporally matched. To try to clarify with an extreme example there is no point in comparing data relating to heel strike and toe off as they are functionally different parts of the gait cycle. This statement exemplifies the futility of comparing outcome patterns that occur during different phases of a task. The problem is further exacerbated because "phases" relevant to control processes are not readily defined. It was decided however, that for this project this indeed had to be defined. Phases should somehow refer to the control processes that may occur within the CNS rather than some feature of the outcome pattern, e.g. heel strike as is often the case with multimode controllers [section 5.3].

6.2.2.2.1 Definition of Control Phases

A method was required which could analyse the outcome variables with respect to the resultant control output of the CNS. Since the CNS is innervating muscles to create forces, the control output, for the purposes of this project, could be measured by the activity of the muscles. However, because it was better to know what effect the combination of all the relevant muscles and tendons produces, it was thought that some kinetic measure of the movement across the knee joint would be applicable. It was decided that the most appropriate measure of this activity was the **power calculated across the knee joint** [section 2.2.2.2.3]. Unusually the definition of this variable in gait analysis allows for positive and negative power, reflecting the rate the knee generates or absorbs kinetic energy respectively. This can form a useful controlling concept because it will show when the actuator of the knee will need to act as a motor or as a brake. Such periods may then provide a better definition of the various phases of the gait cycle especially since there is a greater informational content within this variable since it is the product of moment and angular velocity.

Ideally, for autonomic movement patterns, a regular consistent control pattern should exist that is repeated between strides and is similar for most subjects. This should be consequential for autonomic tasks where the underlying feed-forward patterns of the CPGs' are more likely to be discernible. The influence of feedback corrections may be averaged out because of their stochastic nature, unless a systematic error is encountered. Such a persistent feedback correction to systematic errors may be anticipated and pooled with the feed-forward approach since it would show up as a regular pattern.

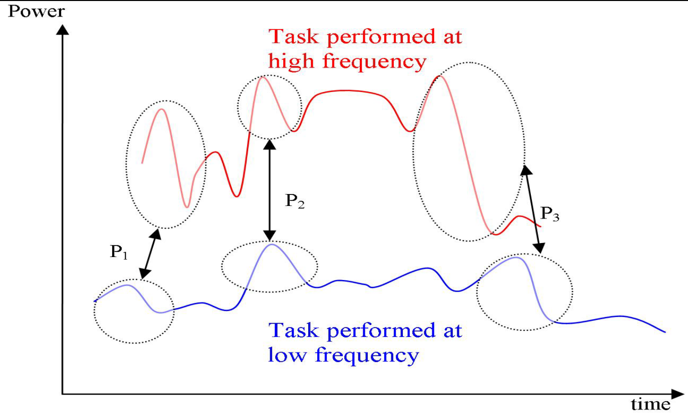
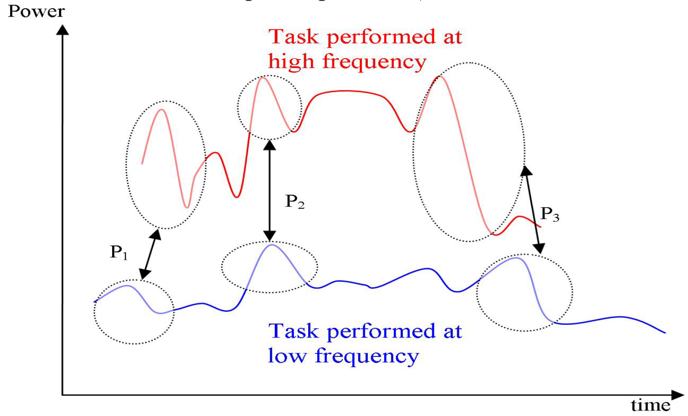


Figure 6-1. Topological similarities may occur in knee power patterns despite differences in amplitude and temporal duration. Such similar features may be the manifestation of a repetitive feed-forward enervation pattern to muscles produced by CPGs. The bubbles show similar topological features that occur during the task as it is performed at different frequencies. These will represent the different phases that may occur during the gait cycle.

If such feed-forward patterns do exist, then they may manifest possibly as recurrent topological features with the knee power patterns (see Figure 6-1, for an illustration of this concept). In this illustration, three features distinguishable by eye have been labelled, (P_1, P_2, and P_3). The eye is able to observe similarities that occur between the two profiles despite differences in the amplitudes and temporal duration. Such features can reflect the true phases of the



task as related to control. Similarly, it was hoped to use the trans-tibial knee power patterns to identify **phases of gait** that would be relevant to the trans-femoral amputee.

To clarify, it should be mentioned that this method of identifying the phases of gait was thought to be necessary because of the uncertainties of other methods. For much of the literature on gait analysis, the phases of gait are mapped according to a particular outcome event, usually heel strike (HS). Therefore, the whole gait cycle is viewed as HS-to-HS, and other events recorded as a percentage of this. Unfortunately, control phenomena do not necessarily have to occur at the same point relative to HS. For instance, as speed is increased, it may be necessary for muscles to act as a brake earlier on in this HS-HS % scale to produce the observed outcome.

For such a scale to be appropriate for control purposes, the timing of various control events would always have to occur at the same point on this scale irrespective of dependencies on speed and morphology. In the example just given, if braking does not occur earlier then the force would have to increase to compensate; otherwise, the body would stop too late. Therefore, it can be seen how critical timing issues are, so presumptions regarding temporal normalisations should not be made; in fact, reviews on gait have not definitively shown (or even examined) if the HS-HS normalisation is appropriate (especially with regard to control).

If some temporal normalisation procedure were to be used however, then the knee power features that are observed during different walking speeds should overlap. Therefore, it was felt more prudent not to prejudge what outcome events, if any, could define control events, especially for the lesser-studied trans-tibial amputee. This stance was also confirmed by the work of Sadeghi et al [174], where they avoided the problems of temporal normalisation when trying to analyse variability; they instead made comparisons using Curve Registration by aligning similar peaks (see also section 2.2.2.1.3.1). They note that as a criterion, the chosen landmarks for curve alignment should be clearly visible and identifiable.

Therefore, *it was proposed that the phases of gait, and the phases of other tasks for that matter, could be defined according to the topological features of the knee power patterns during the task. These phases would then be used as the basis to examine outcome events*. Defining phases according to knee power provides a temporal reference with which to examine the outcome variables and determine which gives the least variation for a particular phase. Such invariant outcome variables (for a particular phase) may then be associated with the target task vector so that patterns can be found that represent the autonomic task to be performed, i.e. provides Ψ^{T}_{knee} for the controller.

Therefore, in the experimental stage of this project, data will be collected with such thoughts uppermost so that the knee task can be defined in terms that also take into account the morphological parameters as well as speed.

It transpires that such ideas can be thought of in conjunction with the intent recogniser. Although the controller does not necessarily have to directly detect knee power, it should be able to recognise when events related to these power features occur; it needs to know which target pattern to *initiate* and *when*.

6.2.3 (DV3) INTENT RECOGNITION

6.2.3.1 THE ROLE OF THE INTENT RECOGNISER

Ideas have been proposed exploring how tasks should be defined for the controller, and indeed what motor patterns they should represent. However, the controller needs to know which particular target pattern is required and when, i.e. the appropriate Ψ_{knee}^{T} pattern needs to be selected and initiated precisely on time so that it is co-ordinated with the Ψ_{hip}^{out} pattern.

For intent recognition to be possible, an element of anticipation on the part of the controller is required. This can be achieved for movements that are autonomic in nature, as they are likely to utilise, to a certain degree, feed-forward control to produce the motor patterns. Indeed, it was the purpose of analysing knee power data to ascertain the nature of these feed-forward patterns. Since the target patterns should be associated with such feed-forward features in the power profile, it was thought they might also be applicable for intent recognition purposes.

The role of the intent recogniser was to determine which task the amputee is *about* to perform. It is thus vital that the initial stages of a task are recognised without any confusion arising between states representing other periods of that task or indeed of other tasks. If the beginning of the task has been detected sufficiently quickly, then the appropriate Ψ_{knee}^{T} pattern can be applied to tell the controller what it should be trying to do. Of course, if the task that Ψ_{knee}^{T} is trying to define also has a dependency on some control variable such as speed, then this variable must also be monitored so that the target pattern can duly be attenuated.

As the artificial controller must co-ordinate its response with the amputee, the intended knee movement should be compatible with the stump movement. It would thus be beneficial to partially derive the intent from sensors that directly measure the amputee's stump movement. This requirement does not prohibit the use of task variables because, as explained [section 4.3.3], it is possible for variables that are associated with the stump to also be used to define knee target patterns. It is suspected that in addition, the use of variables more directly linked to the knee joint will be required. This is to ensure that the initial state of the shank is appropriate for the intended knee motor pattern to take place. Although the amputee may stipulate their intention through stump movement, it is conceivable for the amputee to make mistakes. Thereby, if only stump variables are monitored, the controller may misconstrue the amputees' intention by not taking into account the shank position. To avoid such mistakes the amputee would have to compensate and ensure that the wrong intention is not conveyed. This would necessitate a significant reliance on proprioception related to the prosthetic shank and foot, which may be difficult given the limited sensory feedback to the CNS of the state of the prosthesis, and therefore higher-level systems may have to be used. Furthermore, if the controller is reacting without knowing the state of the shank then it is likely to defy constraints 1 and 2, i.e. by producing unexpected results that need higher-level intervention from the amputee. The intent should thus be derived from a combination of knee and hip joint variables so that the controller can produce a robust response. The inclusion of such task variables should thereby aid safety issues.

The method proposed to recognise the intent of the amputee relies on using outcome variables that can uniquely correspond to the beginning of a task as defined by the knee power profile. The big gamble in this project, as has been implied by other sections, is that autonomic tasks should provide discernible power characteristics that are generic for most reasonably fit trans-tibial amputees.

Consequently, if some knee power feature can be associated with the first phase of the walking cycle, irrespective of speed or subject, then this state should be indicative of the amputee's intention. It is postulated that this statement is true at least for a short period because of the feed-forward nature of autonomic motor patterns. It would be preferable that task variables could be used to recognise this initiation state, as they would specifically be chosen because they present a low variability when normalised with respect to speed and morphology. The only other criterion that needs to be checked is how such variables can unequivocally represent the intent state. If movement patterns were very simple, it might be that just two outcome variables would suffice, e.g. the hip and knee angles. When they reach a certain value, this may exclusively correspond with the beginning of the particular task. Unfortunately, considering the vast range of movement patterns that the natural system is capable of, two outcome variables are unlikely to be sufficient for these purposes. It was thought that a solution was more probable if a multitude of outcome variables are simultaneous measured, although it would be preferable to optimise the final number of variables.

Experimental data needed to be gathered and analysed to find relevant variables for identifying the initiation states. It was envisaged that this process should also influence the choice of variables that make up the task vector. To exemplify the demands of this problem, the scenario of a possible solution is painted so that the same situation may be sought experimentally.

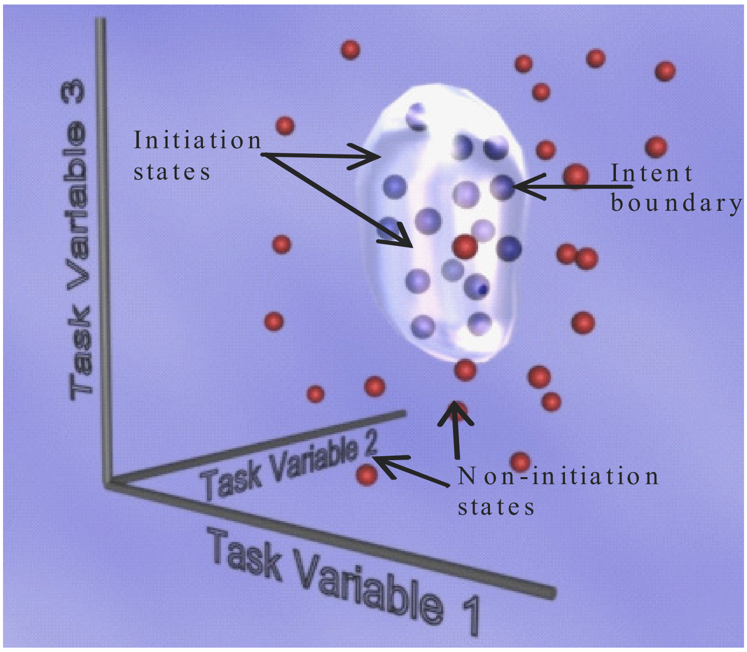


Figure 6-2 – Plots of 3-dimensional output task vectors as they occur throughout different motor states. Each axis represents one of the three chosen outcome task variables. The blue spheres (initiation states) are segregated from all other states (red spheres) by the virtual bubble. This ideal situation shows how, in this particular 3-D space, an individual motor state can be distinguished from all other motor states.

It was conjectured that a solution exists if the task's initial state can be differentiated from all other data when mapped in task vector space. A hypothetical solution is constructed which only uses three task variables (for the purposes of graphical presentation). With these three dimensions, it is possible to visualise any motor state that the amputee may create throughout their general movement range. These motor states can be displayed as discrete points within the task vector space (see Figure 6-2). An imaginary sample of data, representative of many different activities, is plotted for trans-tibial amputees. The following colour code is used to differentiate various states: blue spheres represent initiation states that were sampled at the initial transition point of a selected task of interest and red all other possible states within that task and other tasks. If the set of task variables have been well chosen and a solution exists, then the blue spheres should appear to cluster in a small volume whilst being isolated from red spheres. The diagram epitomises such a situation, where it has been possible to exclusively enclose the blue spheres in a bubble. The choice of task variables is critical to be able to generate such a solution since not all task variables may enable such a bubble to be drawn; indeed, it is not guaranteed that a set may even be found that results in an acceptable solution. However, if such a bubble can be established then it would define a boundary region that can be used for intent recognition. If the task vector associated with the bubble space is monitored then it should be possible to detect when a vector lies within the boundaries of the bubble. This would indicate, to the controller, the intent so that a response can be made according to the appropriate Ψ_{knee}^{T} pattern, which corresponds to that initiation state.

The dimensions of the bubble will be peculiar to the variations that were present in the original sample of initiation state vectors. Therefore, ideally the sample should be diverse enough so that a bubble can be defined which still encompasses any subsequent initiation states that it may encounter, i.e. when being used by the trans-femoral amputee. The size of the bubble should however not be over exaggerated either, as then it is likely to include some non-initiation states. A fine balancing act may be required in order to define the optimum bubble size. It can be seen that the size of the bubble will reflect the generalization ability of the controller, where if it is too small then initiation states may be missed for individuals that differ too much from the mean of the sample cluster. Since this contravenes constraint 4 it should be avoided, and instead all reasonable occurrences of the initiation state included.

In practice, one has to first find the vector space in which a reasonable solution may exist and then find a method to define the bubble that represents this recognition region. Note that the vector space need not be limited to only three dimensions, although the eventual shape of the bubble must still obey the above points on size and generality. This entails gathering sufficient data to fully represent the population and if possible try to broaden the diversity because it cannot be expected that trans-femoral amputees will walk with the same predictability. Therefore, the data must not only be representative of morphological differences but should also include other dependencies such as speed and slopes. Furthermore, sufficient samples of non-initiation data should be gathered to ensure that a vector space has truly been chosen in which a bubble can be established that prevents intermingling of initiation and non-initiation states. Otherwise, erroneous detection of initiation states could result when presented with live data.

It would be undesirable if an amputee had to use unusual movement patterns in order to establish a bubble that is able to segregate the initiation states (e.g. like the rhythmic tapping required to switch between modes of the C-Leg [section 5.4.2]). All that should be required of the trans-tibial amputee is to walk as naturally as possible whilst data is gathered. This data can then be used to determine when trans-femoral amputees, in combination with the actively controlled prosthesis, produce motor patterns that resemble those of the trans-tibial amputee. This process requires the amputee to reinstate, to some extent, elements of their pre-amputation motor patterns for the hip [section 6.2.2]. The greater the *generalisation* capability of the intent recogniser the more likely it is to classify *atypical* trans-femoral patterns (resulting from a lack of thigh muscle strength relative to trans-tibial amputees). The use of trans-tibial amputees, in this context, is advantageous because they are able to provide model data that encompasses a broader scope of movement than trans-femoral amputees may be able to fully utilise, e.g. in terms of speed; it is better to have more than less in this case. The weaker muscles in trans-femoral amputees may result in a reduction in the top speed; however, the proposed control system would not be the limiting factor with data available to contend with slower speeds. It is envisioned however that if such a knee control system were to be regularly used, then the trans-femoral amputee is also likely to strengthen their thigh/hip musculature since the knee device will enable and require greater movement diversity from these muscles. If such gains were to be made, the controller would be well placed to cope with increasing demands for performance since the controller is based on the superior abilities of trans-tibial amputees.

If it is not possible to establish a reliable cluster of initiation states using natural movement patterns, then the inferior approach of making unusual movement patterns may have to be considered. The amputee would have to make distinguishable movement patterns that can unequivocally be differentiated from all motor tasks. This, however, immediately compromises the prerequisite that the prosthetic knee should enable movement patterns that are maintained under autonomic control [constraint 2].

6.2.3.2 ARTIFICIAL NEURAL NETWORKS TO AID WITH INTENT RECOGNITION

It can be seen from the above that the problem of intent recognition is largely of a statistical nature. There was a variety of options available to tackle such matters.

For instance, it was thought that techniques such as Principal Component Analysis (PCA) might aid in analysing the suitability of different outcome variables for control purposes. PCA allows the overall data to be expressed according to the most relevant axes, thereby providing a useful tool to illuminate the choice of task variables used to recognise intent. The method is effective in data reduction and explanation and involves finding the eigenvectors (**U**) for the n by m matrix of data (**X**), where n is the number of samples and m the number of variables. The principal components (**Z**) can be calculated from the eigenvectors of the sample data, so that $\mathbf{Z} = \mathbf{U}^t\mathbf{X}$. The PCs are typically ordered according to decreasing variance, so that the first few should explain the majority of the data.

For instance, the method has been used by Deluzio et al to enable the modelling of knee kinematics and kinetics [57]. This group used PCA to develop a model of normal subjects whilst walking. They established a reference set of eigenvectors, from a set of sample data that was supposed to encompass reasonable variations in the standard gait pattern. The technique enabled them to find the most important variables in which the majority of variation could be explained by the first few principal components (the remaining components were associated with noise). They applied the reference eigenvector to the patient data and thereby modelled the patients' PCs. The reference model was then used to emphasise comparisons between the osteoarthritic patients' PC values. The method allowed subjects to be statistically discriminated and classified according to the entire gait waveform, as opposed to comparing individual arbitrary parameters.

The method of PCA, although useful in reducing and explaining the data, does not easily allow automatic classification for intent recognition. Instead, an alternative method was considered.

It was decided that the use of **Artificial Neural Networks (ANN)** would be most preferable towards solving the above issues as opposed to PCA or statistical methods such as linear discriminant analysis (LDA). ANNs' try, mathematically, to emulate some of the observed properties of the biological nervous system. As the existence of neurones within the brain became apparent through the work of neuroanatomists', other researchers tried to theorise the functioning of such structures. It was not long before models were developed that tried to show how such a labyrinth of nerve fibres might produce some of the complex processes associated with the brain. As early as 1873, Bain tried to explain how memory and movement control might result from the interlocking neurones [11]. Although the prime motivation for this work was to research the workings of the brain, it was later realised that these models, although not accurate descriptions of the brain had powerful mathematical properties in their own right. Once Bain theorised adaptive rules for associations, this became the precursor for the field of intelligent machines. By the 1950s, models that are more famous were proposed for example, in chronological order, by McCulloch and Pitts [134], Hebb [95], and Rosenblatt [170], all of which stimulated this new field of study.

Such work has led to the development of ANNs' for practical applications. These consist of processing elements that are analogous to neurones, which are highly interconnected by weighted links that derive from the synapses separating neurones. The power of such networks lay in the fact that by altering the weights of the links and assigning different functions to the processing elements, it was possible to make different network structures that were capable of **adaptation**. Thereby, various techniques were developed to adjust the networks, until a useful response could consistently be produced given a particular situation. Commonly, ANNs tend to *learn* how to respond according to *prior experience* that has been gained from training data. Neural computing has thus become "the study of adaptable nodes which, through a process of learning from task examples, store experiential knowledge and make it available for use" [3].

Since it is desirable to imitate the neural processes that result in trans-tibial knee motor patterns, it was thought the use of ANNs might be a natural choice for this project. In practice however, ANNs will be regarded as statistical tools whilst the close analogies with natural control systems will be consequential.

So what can these networks actually do? Why are they suitable for the intent recognition problem?

ANNs were thought to be a convenient tool because they are able to:

1. *Associate*, which involves the linkage of information with other information. The most common network architectures act as a pattern transformer in which input patterns are transformed into output patterns. The dynamics of the transformation depend on the weights and structure of the network, these are usually adjusted during training.
2. *Generalise* for a given set of data. As the networks are usually trained through experience, i.e. presentation of prior examples of input and corresponding output sets of data, it is possible to present the full diversity of what it may expect in the future.
3. *Solve non-linear problems*, which is more than likely to be the case regarding movement issues. They can thus conveniently tackle such problems rather than using what would otherwise be complex statistical methods.
4. *Respond quickly*, they can readily be incorporated into IC chips, and because of their parallel nature, work efficiently and quickly to transform an input set into an output.
5. *Solve without knowledge of the plant dynamics*. Since the training process adjusts the weights of the network to behave according to prior examples, then this bypasses the requirement for modelling the plant. Because of this ability to mimic, the ANN has itself been trained to act as a model of the plant dynamics, in some examples.
6. *Be implemented easily*. There is no requirement to know complex statistical techniques as this method effectively implements such ideas by default.

All of the above features mean that ANNs are capable of classifying input data quickly and efficiently into arbitrary sets that depend on the complexity and training of the network.

There are numerous examples of practical applications using ANN, ranging from vehicle control, to face recognition. Because of the pattern recognition ability of ANN, examples are even available for analyses of gait for clinical applications [83, 13, 105, 124].

Holzreiter and Kohle [105] trained ANNs to distinguish 'healthy' from 'pathological' gait, as caused by calcaneus fractures, amputations, and diseases, for example. The network's output was designed to give a probability, which indicated the degree of pathology, or rather how different it may be from 'healthy' samples. This response was also validated against conventional clinical assessment. The hope was to produce a consistent method of assessing patients and reduce the subjective differences present in most clinical analysis.

In an attempt to manage the vast quantities of data that can be gathered on gait, and try to determine the extent of pathology, ANNs were also employed by Lafuente et al [124]. They differentiated pathological gait from age matched healthy control subjects; in this case, they focused on one type of ailment, which was lower limb arthrosis. They found that the discrimination rate of the ANN was significantly greater than a Bayes quadratic classifier, which is an alternative statistical approach.

Gioftos et al. [83] used ANN to recognise various conditions of gait, including normal walking, walking with 3.5kg mass, and walking with a knee brace. Separate networks were also trained to detect the walking speed during each of the conditions, as will be discussed later [section 6.2.3.3]. These networks were sufficiently generalised to be applicable to most subjects. Similarly, Barton and Lees [13] only used hip-knee joint angle patterns to distinguish three conditions (normal gait, simulated leg length difference and leg weight difference). They were able to correctly classify 20 patterns out of 24.

6.2.3.2.1 Network structures

The problem of recognising intent, as summarised by Figure 6-2, embodies pattern recognition when presented through the context of ANNs. Pattern recognition involves classifying objects, usually presented as vectors, into categories. The trick however is to be able to compute the classification function that maps objects to categories, given that the input representations of an object can vary. This statement is very similar to the situation described in Figure 6-2, in which different bubbles may represent particular categories and the axes of the vector space will simply be the inputs to the networks. This reiteration however is useful because there are many examples of ANN that are used for pattern recognition. Several types of architecture are available to act as a template for the final design of the network. These include perceptrons, adalines, multi-layer networks, radial basis networks, etc., all of which rely on a supervised technique of training. The final choice may in fact even be a hybrid of these and must be designed in the context of the problem. It may also be possible that different architectures are capable of providing solutions, so that it is important to validate each network by thorough testing and choose one that behaves most adequately. During testing, it is important to distinguish data used for testing from training. This is to avoid choosing networks that recognise anomalies only particular to the training set (e.g. if training is based on one subject). Such design issues for networks are contemplated more thoroughly in section 8.3.

6.2.3.3 DETECTION OF CONTROL VARIABLES (E.G. SPEED AND INCLINE)

So far, intent recognition has concentrated on identifying the initiation state of a task, however, if the task is dependent on some control variable, then it is also likely that both the target pattern and the required effector output τ_{knee} will be dependent on this variable. Consequently, the control variable must be monitored; in particular, it was postulated that speed (or some related variable) would be such an important factor within the task of walking.

Fortunately, various researchers have already shown the feasibility of being able to detect such parameters. As has been mentioned [section 6.2.3.2], Gioftsos et al [83] used neural networks to predict the speed of walking, under three different walking conditions. These networks only required temporal gait parameters as inputs, including single support and double support duration, which although not instantaneous should enable speed measurements at least once per gait cycle (and up to three times per cycle after completion of the first cycle). After training 21 separate networks they were able to successfully categorise 64% of the test cases into seven different speed classes under the various walking conditions. Although this success rate was at least as good as Linear Discriminant Analysis (LDA), a statistical technique, it would require improvement for reliable operation.

Aminian et al [5] also used artificial neural networks to assess both the speed of walking and the incline of the slope being walked. Accelerometers were placed on a waist belt to give three orthogonal components of acceleration for the body and one on the heel to give the forward heel acceleration. These signals were parameterised to form 20 parameters according to: the mean and median taken over the gait cycle, the variance of the four accelerations, the covariance between each pair of acceleration, and the peak heel deceleration. It was possible to reduce the number of candidate input parameters to ten according to those that exhibited the greatest correlation with speed and incline. Using these ten inputs, two networks were trained to predict the speed and slope with an rms error of less than 5%. The generalisation ability of these networks was demonstrated using six different subjects although for a conclusive demonstration, more may have been desirable. In addition, because some of the input values had to be determined over the gait cycle, the system would only be able to respond at the end of a gait cycle, which is not ideal.

For this project, it would be more beneficial if the variable could be detected continuously (or at least at a sufficient frequency not to produce noticeable delays in response). It would also be valuable to detect the particular speed of the amputated leg rather than the overall speed of the subject. Then local adaptations can be possible, for example to walk round corners requiring different stride lengths and hence speeds for each leg [14]. Thereby if the controller were able to react immediately to these speed perturbations then the amputee should be better able to cope with such manoeuvres. Although the above has shown the viability of assessing speed, e.g. using neural networks, they mostly rely on data that is sampled over the whole gait cycle e.g. mean acceleration or gait cycle duration. Therefore, for the controller to be truly adaptive these techniques should be improved upon or alternatives found so that continuous speed assessment can be made.

Towards this, there are several possibilities. For example, there is a surge of development with mobile communication systems where there is a desire to make devices that are more "context" driven, i.e. that detect external situations [152, 26]. Such devices track the location of a person in order to implement context driven software. This has led to the use of Global Positioning Systems (GPS) to directly track the location of a person through a network of satellites and work out positions to sub-centimetre accuracy. The use of GPS has subsequently been suggested for biomechanical study where data can be collected outside of the laboratory [199], and variables such as velocity calculated. However, such systems suffer from blackouts where reception is lost because of buildings for example. These blank spots would be unacceptable for the amputee who would not be able to walk during these periods. To compensate for black spots there is also development of devices which also partially derive their location by measuring speed and direction using gyroscopes for inertial sensing and hence velocity calculations. However, because they are designed to measure the overall speed of the body, these devices may not relate sufficiently well to the individual speed of the prosthetic leg as required for going around corners.

Therefore, two alternative solutions were also conceptualised by the author. The first would rely on measuring the period between phases (as defined by subtask initiation states [section 6.2.4.2]) and the second using ANNs to predict the speed at particular instants during the gait cycle.

Assuming some relation exists for the interval between two subtasks and speed, then by measuring the time of this interval it should be possible to estimate the speed of the leg. The greater the number of intervals per gait cycle that can be timed the greater will be the continuous assessment of walking speed. This implies that more subtasks should be detected if possible. With this method, the speed will relate to the previous phase. This means that the initiation of the task must be carefully considered since it will not be possible to have a reading for the first phase. However, if this is problematic, it may be better to develop a separate task for the controller that initiates walking.

The second proposed method utilising ANN should eliminate the initiation problem as it is hoped that the data taken at any instant can be directly related to speed.

However, to increase the ANN's chance of success it would be better for the network to know when in the gait cycle the data has been sampled. This knowledge however should be available from the intent recogniser, which can be extended to also detect the initiation of subtasks [section 6.2.4.2]. Therefore, it is conceivable to train ANNs to map sensory information to the walking speed at specific points during the gait cycle (these points may relate to the control phases that the intent recogniser will recognise).

Whatever method is used, the eventual performance will depend on how rapidly the speed can be measured. Although delayed responses will not be the end of the world, as demonstrated by current active mechanisms such as the intelligent prosthesis [section 5.4], the eventual goal would be to give continuous speed assessments.

6.2.4 (DV4) FEED-FORWARD/BACK TRANSFORMATIONS REQUIRED BY CONTROLLER

So far, ideas have been proposed explaining what is necessary to define target task patterns, in particular, for walking. In addition, issues concerning how the controller may detect the task to be performed have been raised, i.e. the requirements for an intent recogniser that selects the appropriate Ψ^T_{knee} pattern on time have been acknowledged. What remains is for the controller to transform this desired pattern into a mechanical response from the actuator; this should result in a similar outcome to the target pattern. The control options available to do this have been reviewed in a generic manner during section 4.3. The purpose of this section is to emphasise the solution that was thought most applicable for this project.

For the author, the use of feed-forward techniques seemed very compelling given the pre-planned feed-forward nature of automated movement patterns [section 3.3]. Feedback strategies were less promising because of likely conflict issues with the natural system. It was however thought that if necessary they could be incorporated at a later stage to only correct the feed forward patterns.

For feed-forward techniques to work, the transformation $\mathbf{A_{knee}}$ has to convert the target vector $\left[\Psi^{\mathrm{T}}_{\mathrm{knee}}(t)\right]_{t=0}^{t_1}$ into a temporal knee moment pattern $\left[\tau_{\mathrm{knee}}(t)\right]_{t=\partial}^{t_1+\partial}$, where the duration of the task is denoted by t_1, and ∂ represents the delay in producing this response. When τ_{knee} is combined with the moments/forces from the other effectors, this should result in a close approximation of the target pattern. If the plant dynamics were fully known then it might be possible to model the required τ_{knee} pattern with respect to Ψ^T_{knee}, however this process is complicated by the fact that the contribution of the other two effectors must also be predicted. Therefore, an alternative technique was sought.

6.2.4.1 TRANS-TIBIAL AMPUTEE KNEE MOMENTS USED FOR FEED-FORWARD CONTROL.

Once intent is detected, the actuator contribution (τ_{knee}), towards the resultant knee moment (e_{knee}), entails a prediction of how the amputee may simultaneously move their stump. Therefore, it seemed more straightforward to directly establish the knee moment from prior examples exhibited by trans-tibial amputees during similar tasks. This naturally follows on from the method used for intent recognition, which utilises data from trans-tibial amputees to signify a particular control phase and thereby a particular target pattern Ψ^T_{knee}.

Implicit in this direct method is that the τ_{knee} pattern produced by trans-tibial amputees is the best available (most appropriate) for trans-femoral amputees, who share, in common, the foot prosthesis (this is a reiteration of the arguments made in section 6.2.2.1). For this statement to be most applicable, both τ_{stump} and τ_{ext} should ideally be similar to trans-tibial amputees. As discussed [section 6.2.2.2 and 6.2.3.1], it was anticipated that trans-femoral amputees should have sufficient control over their stump (albeit weakened) to generate the required τ_{stump} (even if it is only for slower speeds initially). However for τ_{ext} to be comparable, the dynamics of both the trans-tibial and trans-femoral shank/foot should ideally be equivalent, e.g. due to moments of inertia, and centres of rotation. The effect of such differences was examined by Bach et al., who showed, through computer simulations, how the inertial characteristics of the limb are important determinants of the gait pattern [9]. They even go on to demonstrate this experimentally where the walking pattern of trans-femoral amputees was improved by adjusting their inertias to be similar to the normal [10] (or in this case, they can be adjusted to be approximate trans-tibial amputees[36]). Therefore, the inertial properties of the trans-tibial shank should be matched, as closely as possible, by the active trans-femoral prosthesis. On a practical note, this gives provision for the installation of various components such as batteries within the shank for both extra ballast and power that the knee unit will require. If the prosthesis to be controlled has different inertial moments that result in significantly different τ_{ext} patterns, then the moment patterns attained from trans-tibial amputees cannot directly be used for τ_{knee} and some adjustment must be made. However, in such a case, the differences are also likely to propagate to the thigh segment so that the τ_{stump} pattern would also be much less predictable which makes modification of trans-tibial moment patterns even more complex if not impossible. In such a situation, it cannot immediately be presumed that the trans-tibial amputee will display model data to base control events upon.

Therefore, assuming the effects due to differences in inertial moments are small, what was required was the ability to gather data relating to the kinetic activity of the trans-tibial knee. Specifically, the inter-segmental moment about the trans-tibial knee joint was a good candidate as it represents the overall contribution of the muscles, tendons, and soft tissues spanning the knee joint. The action of the actuator and prosthetic knee mechanism should reproduce this contribution. Although an actuator may only directly produce a force/moment, its output may easily be derived by transformation of the trans-tibial moment according to the configuration of the actuator.

Given that such a transformation can be calculated according to the specific actuator mechanism that may be chosen, it was better at this stage to maintain generality and only try to define the overall moment pattern to be transformed. This will form what is termed the *reference profile*. Statistical techniques can be used to form such profiles, where dependencies on control variables such as speed, and morphological parameters may be included, perhaps through modelling. Although the dependencies

[36] If the concept of making the moment of inertia as close as possible to normal is eventually accepted for trans-tibial amputees, then the same will ultimately be desirable for the trans-femoral knee controller.

may be derived statistically, it does not mean that the reference profile has to be the average of all the trans-tibial moment data. It may be more advantageous to form this profile using only data from a few of the "better" walkers so that the controller also recreates the "better" style.

As an alternative to the use of knee moment patterns, a slightly more adaptive response may be possible by basing the reference profile on trans-tibial knee powers. Since power is the product of moment and angular velocity, the required knee moment may then be calculated from the reference profile and the real time knee angular velocity. This will then allow some variation of the response to slight perturbations in the angular velocity during the gait cycle, since the mechanism will now be trying to reproduce power patterns rather than moments.

The adaptability afforded by the system even if using a power as a reference profile is quite restricted. Whichever kinetic variable is used, the fact that the reference profile will be based on some statistical procedure such as averaging means that the same pattern is used for every step walked at a particular speed by an individual. To allow more adaptability from step to step, for example to environmental perturbations, would require a greater element of feedback control.

It was thought the most effective contribution of feedback control to the knee is through the stump effector. The human CNS, even given the pathology of amputation, is more adept at adaptation than any artificial system. This concept was realised during section 4.3.3, and is possibly the reason that amputees are able to use conventional prostheses at all [section 5.1].

Therefore, if the controller provides a good stable actuator profile that's suitable for each task, the amputee should be able to make the minor adaptations in response to perturbations. This avoids potential problems of co-ordinating two feedback controllers [section 4.3.3.1]. If, however, the reference profile produces systematic errors necessitating constant adaptation by the amputee, then the problem is most likely to be associated with morphological differences not being sufficiently modelled. A possible resolution is to implement a gain switch, to be adjusted by a prosthetist, so that the reference profile can be attenuated and fine-tuned to the individual.

As can be seen, artificial feedback has been avoided as this could make the amputee unsure as to what the knee is trying to do. However, it was thought that the very act of regularly detecting intention provides an element of feedback to the system.

6.2.4.2 THE DETECTION OF SHORTER PHASES TO IMPART MORE FEEDBACK

Although the intent recogniser was designed to recognise individual tasks, there is no reason why the concept cannot be extended to also include subtasks. Ordinarily, in control terms, an entire task may last for some considerable duration. If this is to be implemented using feed-forward techniques then the likelihood of errors propagating and attenuating increases as the task proceeds. Therefore, if a task is to be split into several subtasks that can be detected, then this provides an opportunity for resetting the feed-forward pattern. This resetting will work primarily on a temporal basis, although if each subtask is speed dependent then the amplitude may also be modified accordingly. This will consequently enable the amputee to instigate changes throughout the task, which should overcome some of the problems mentioned for the intelligent prosthesis where the speed could only be changed one gait cycle later [section 5.4] and comply with constraint 3, which necessitates real time operation.

The method employed to split a task into subtasks will be the same as that outlined in section 6.2.2, in which topological properties of the knee power will be used to label various phases of the task. The beginning of each feature will represent the initiation point of a feed-forward reference profile.

Therefore, a point has now been arrived at where the concepts described in chapter 4.3.1 have been turned upside down. It may be argued that the initial pattern described by the target task vector can now be used for intent recognition purposes by the neural networks. The intent recogniser then directly selects a predetermined τ_{knee} profile relevant to the intended target task, rather than having to calculate by using the transformation $\mathbf{A}_{\mathbf{knee}}$ on the target vector. According to Bernstein's terminology, it could be said that the intent recogniser is selecting engrams [section 3.2.3]. He even commented on the importance of accurately initiating such feed-forward engrams to avoid large errors. In a sense, these control concepts try to minimise errors by maintaining an element of feedback similar to the "chain hypotheses" of Bernstein where each engram is linked to the next via sensory events [section 3.2.3, pp. 82].

The actual profiles that are selected must now directly relate to the phase as determined from the topological properties of the knee power. Their duration should be sufficient to overlap the next two phases in case the intent recogniser misses a phase.

6.2.5 MODULAR STRUCTURE OF CONTROLLER

It was postulated that if it is possible to train a single network to recognise one task such as walking, then it should also be possible to train other networks to recognise other movement tasks. It would be a simple matter of extending the same techniques to these other patterns so that a variety of networks could be trained to recognise different intents. The results of these different networks may then easily be combined using a parallel structure so that the ambition for modularity (aim 5) would be realised. In order to combine such networks, the outputs of each network should refer to the probability that a particular task is currently occurring. A decision process can then be made using these probabilities so that the relevant target pattern can be determined. The decision process may be as simple as using the network that outputs the largest probability, or more complex, where for example fuzzy logic systems may decide (possibly by also referring to previous decisions so that the more likely of two similarly high probabilities is chosen).

Just as the walking cycle may be split into phases dependent on the power topology, it is expected that other tasks may also be broken down using this technique. The initiation of each phase may then be detected using similar intent recognition methods to that described above. It was thus envisaged that separate artificial neural networks (ANN) would be trained to specifically recognise each one of any number of possible phases related to autonomic movement. In this way, any phase of any task can be implemented as soon as the intent recogniser gets feedback signals corresponding to that particular state. Each network can then be arranged in parallel so that if the system needs to be subsequently extended a new network can easily be added, which should hopefully be in keeping with constraint 5 (see Figure 6-3).

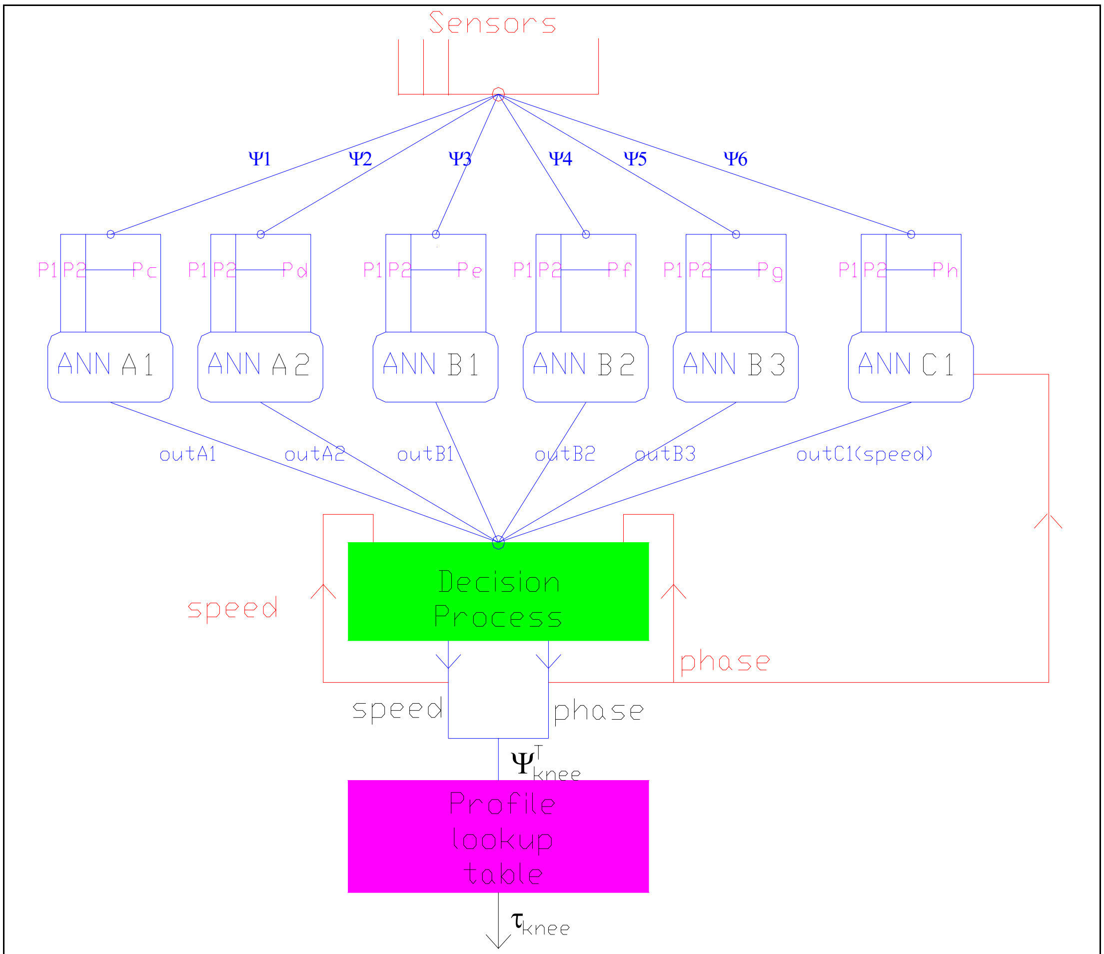


Figure 6-3 - Modular structure of Intent Recogniser. Several task vectors are formed from different combinations of sensor input. Each task vector supplies a separate neural network. In this example, there is a neural network to detect each phase of tasks A and B. Also included is Neural network C1 which is used to detect the speed of walking. With the exception of C1, the output of each network gives the probability that the current state of the amputee is the same as that which the network has been chosen to detect. Using these outputs it should be possible to decide what the current state of the amputee is in terms of phase of a task and speed of walking. To aid the decision process and prevent impossible combinations of phases being produced, some feedback of the previous phase and speed may also be desirable. Finally, once the state of the amputee is known, it is a simple matter of using this to select the reference profile suited to the particular task, phase, and speed.

Some safety considerations may have to be imposed over the top of this structure, for example if there is ambiguity in the choice of two networks, i.e. if their outputs give equal probabilities of being the current state. Such ideas can be incorporated into the decision process, for example by giving safer actions greater weighting. This stage of the work is possibly beyond the scope of this thesis as it would be easier to modify once a prototype has been constructed. The important point however is the ability to recognise phases according to the power topology of any particular task, and then to derive the feed-forward moment profiles (τ_{knee}) for each phase, from data gathered on trans-tibial amputees during that task.

Therefore, most of the system will involve simple, quick parallel computation, which directly links to look up tables or models to extract the relevant profiles. Although it is not an emphasis in this thesis to deliberate on practicalities, all of these computations, even given current technology, can be handled.

7 METHODOLOGY

Ideas have been formulated, during chapter 6, as to what was considered a reasonable solution to the problem posed by this thesis. It was hoped that experiments could be set up which demonstrate the viability of different aspects of the proposed control concepts. In particular, it was desired to show whether artificial neural networks were suitable for intent recognition of tasks and phases; could data based on trans-tibial amputees yield robust and accurate recognition strategies? With regard to this, methods were developed to gather and analyse data, which were appropriate and sufficient for the purposes outlined in section 6.2.3 and 6.2.4. In essence, they must yield data that is useable for neural networks to learn the phases of the walking gait cycle and for derivation of the subsequent reference profiles concerning these phases. Ethical approval for these methods was obtained from the Ethical approval committee at the Southern General Hospital [Appendix D].

The use of the gathered data for designing and training neural networks to fulfil the undertaking will be dealt with in chapter 8; it highlights the importance of gathering appropriate data.

7.1 EXPERIMENTATION

7.1.1 DEFINING TASKS

The influence of control variables [chapter 6] on an individual task may significantly increase the variability of the motor pattern associated with that task. As the value of the control variable is graduated, many aspects of the motor pattern, in terms of general outcome variables, will also incrementally change in response. However, according to Kelso's theories [112] discussed in section 3.5, there should be a particular variable known as the order parameter. It will have particular *stable* states (known as attractors) that the system will seek; different intervals along the control variable spectrum may therefore be favoured by different attractor states. The author considers that during such intervals, the motor patterns will uniquely represent a particular task, despite the variances that may be experienced in the motor pattern.

As the control variable is adjusted, a point will eventually be reached when a transition occurs from one attractor state to another as reflected by a sudden jump in the order parameter. It is possible, at this transition point, for instability to be observed where flipping between two states is common. For example, alternation between jogging and walking may be observed at what may be supposed a transition point in the control variable spectrum. This point is often described with respect to speed, although caution must be used if it is to be assumed that speed is the actual control variable. A more fundamental factor common between subjects may exist which triggers such transitions e.g. bone stress[37], or muscle force limits. Physiological measures such as relative heart rates (i.e. normalised with respect to resting rate) were however unlikely to be good candidates for being control variables, mainly because there would be a delayed response, i.e. exertion is felt after a speed increase. Therefore, although it is likely to limit the capacity for speed, it is unlikely to produce definitive transitions. Therefore, ideally whatever the control variable is to be, it should consistently be applicable to most subjects and the same principle be applied when finding other control variables. To summarise, it would be desirable to find out the actual control variables and the corresponding order parameters with their specific attractor states so that the full breadth of the task (i.e. walking) can systematically be defined for all subjects.

Unfortunately, control and attractor variables are somewhat abstract and thus difficult to elucidate. The ideas however remain useful as a guide for designing experiments to gather data that is representative of the full extent of the task in question. This is important, because in the proposed control strategy, artificial neural networks (ANN) have to be trained to recognise specific tasks or intent states. Therefore, regardless of the data acquisition method, it was important that the training set would be composed of what could unequivocally be described as similar samples of the task.

[37] It has empirically been observed that for a whole variety of different species, the behaviour is adjusted to limit stress through bones to one third of the breaking stress. Naturally walking speed is closely related to this factor given that the energetics of the system will increase with speed.

Strangely enough, intuition possibly gives an extremely useful tool for aiding in such assessments. The very fact that one immediately understands what is meant by the term "walking" indicates, at a subconscious level, the recognition of intrinsic factors related to this act. It is with this thought that the author hoped to find the finer details of how this task is modulated.

The following steps were therefore taken to regulate the demonstration task of walking.

7.1.1.1 DEFINITION OF WALKING STYLE

A contradiction may exist between the requirements of data needed to produce a reference profile [section 6.2.4] and those used to develop an intent recogniser [section 6.2.3]. The reference profile should ideally be based on data that represents the *standard* gait pattern, (i.e. an idealised sample if such a thing exists), whereas the intent data needs to include all the variations that might be encountered by the eventual amputee users.

However, before considering these intricacies, the general style of walking that the controller may be expected to cope with must be stated. It is possible for subjects to override any normal gait pattern and produce a diverse range of gaits given the flexibility of the central nervous system (CNS). For this project, only patterns that were under "autonomic" control were of interest. Therefore, the challenge was to collect data on subjects whilst "voluntary" regulation of their walking style was minimised. In essence, all that the subject had to do was walk as naturally and subconsciously as possible. Therefore, various experimental precautions were used to make this situation more likely. For example before acquiring any data, subjects had numerous practice attempts to become accustomed to their surroundings and not over focus on the task of walking so that autonomic systems had an opportunity to prevail. Subjects were also asked to walk for a short distance, before and after the data collection periods. This ensured that they were not preoccupied with initiating or stopping the task of walking; such periods may be sufficiently different as to represent a separate attractor state to normal walking. It was suspected that initiation and stopping may, therefore, better be recognised by additional neural networks than the ones used for level walking; to train such additional networks was however considered excessive for demonstration purposes.

To reiterate, trans-tibial subjects were required to "get into their stride" and walk as naturally as possible whilst data was collected. The following section describes procedures that were undertaken so that subjects would exhibit their entire display of autonomic walking patterns that the controller was expected to aid.

7.1.1.2 CONTROL VARIABLES TO SET THE EXTENT OF THE SAMPLE DATA

Intuitively two variables came to mind, which formed initial candidates for the actual control variables, namely these were speed and inclination of the ground. Of course, other factors may also be present e.g. the coefficient of friction of the walking surface, but it was felt that these two variables would necessitate the most significant motor variability for the walking attractor state. Therefore, it would be desirable to present to the ANN the effects of, at least, these two control variables in order to design the definitive control system. This entails asking subjects to provide sample gait patterns for the full range of each control variable that still manifests as the walking attractor state. An interesting point to note and be careful of in this multivariate control variable analysis is that each control variable may affect the extent of the other control variables. For instance, the range corresponding to speed dependency may increase for walking down slopes, whilst decreasing for going up slopes. This phenomenon however may also be because of an inappropriate choice of control variables that is not sufficiently generic.

Considering speed as the first of these possible control variables, the variable may not be suitable to allow true comparisons of movement patterns given its morphological dependencies. Thereby it was not considered appropriate to compel amputees to walk according to predefined absolute speeds, as this may force the amputee to perform a different task. Furthermore, although cadence might be more appropriate given its morphological normalisation of speed according to step length, it was not desired to force amputees to walk at a particular cadence either, for example by using a metronome. Such an attempt may inadvertently necessitate voluntary enervation and disrupt the autonomic pattern of walking.

Therefore, to avoid such dilemmas and enable the gathering of data throughout the control variable range, amputees were given instructions that hopefully would elicit the full range of walking motor patterns for each subject:

They were first asked to walk at their preferred **natural** speed, which should represent the most autonomic state. The amputee was then asked to repeat this so that a spread of data around the natural speed could be gathered.

The subjects were then asked to walk as **fast** as they could without jogging, and once again repeat this several times over.

Similarly, they were instructed to walk very **slowly** over several occasions.

They then had to walk at speeds between the **natural** and **fastest** speeds, and the **natural** and **slowest** speeds.

Overall, it was planned to gather at least 50 trials of walking data per subject, which should include the full range of subject dependent walking speeds.

This method avoided the dilemma of seeking the true control variable, which should be morphologically independent. It ensured that the task of walking was maintained throughout the experiment since subjects should naturally restrain themselves to whatever are the control variable limits associated with walking; this is underscored by the fact that subjects are asked to walk at their fastest speed without breaking into a jog. More care had to be taken for the slowest region because the walking style can easily break down at extremely slow speeds into numerous dawdling patterns that may manifest as different attractor states.

This method of maintaining the walking attractor state with respect to a "subjective speed" is possibly justified because a speed dependent control variable has to be set by the CNS. "Simple" high-level input to central patterns generators (CPG) will set the rhythm of the motor pattern; thereby a speed dependent control variable is more or less self-imposed, and given an individual's intuitive knowledge of walking they will subconsciously be aware of the boundaries for the walking attractor state. A more in depth discussion of such gait transition factors has been made by Collins [47].

In contrast, the second control variable, that is dependent on inclination, is an imposition on the CNS forced by the environment. Although the CNS may react by a combination of feedback and anticipatory feed-forward approaches, it is not something the subjects instigate, as for example, the tempo of the movement. Therefore, subjects cannot limit the situation to the boundaries representative of the walking attractor state except by avoidance.

It would seem logical that, for very shallow slopes, the motor pattern should not differ greatly from the level-walking patterns, and subsequently a gradual change in these patterns may occur as the slope is increased. This is the basis of the work by Aminian et al [5] where they were able to assess the slope incline from gait data. Therefore, to avoid reaching the critical transition points the attractor variable has to be found, or the slope angle should not be pushed to the extremes of inclination. If the attractor variable is known then it is a simple matter to check that the subject has not jumped to a new attractor state. Otherwise, one must accept that the controller will not aid at the extreme limits, i.e. it will not be perfect for steep mountain walking - a situation which might be remedied in future by combining additional networks which are only fed with extreme situations (the problem of which reference profile to employ at the transition will however remain).

Unfortunately, given the limitations of time on this project, only walking in level conditions was assessed. The controller was thereby only expected to demonstrate the control system for level walking conditions although it is hoped that the technique can be extended to gather data on slope walking in the future.

So far, consideration has been given as to how the task of walking should be performed by each subject, and how many samples are required to obtain sufficient representation of the spread of walking patterns for an individual. Similarly, a sufficient spread of data must also be acquired regarding subjective variability; the control system has to cope with almost any amputee wishing to use it [constraint 4, pp. 114].

The basis for subjective differences in gait patterns can be numerous, for example because of age (in particular the segregation of children and adults), gender, and morphology (as described, for example, by Kinanthropometry, Anthropometry, Somatyping, and Ponderal Index [section 2.2.2]). Furthermore, for amputees, the prosthetic components that are used will also impose some variety to the data. Finally, even subjects with the same body build, age, and gender may also display slight differences in muscle activation patterns and thereby gait style. Therefore, if it were practical, it might be thought best to use the whole walking trans-tibial population so that neural networks have an opportunity to experience all possible variations. Clearly though, this is not practical and may even lead to overgeneralization where the bubble is so large that it includes too many non-initiation states [section 6.2.3]. Instead a sample of the amputation population will be used that represents "reasonable" styles of gait. These should represent a sufficient range of data to allow for speed and morphological differences.

To further simplify matters for demonstration purposes, it was decided that the controller would focus on specific groups within the trans-tibial population without compromising its wide applicability too much.

These arbitrary categories were composed of the following, and were largely determined by default according to the amputees that were available for testing:

- *Age/Experience* - this limitation was used in a broad context simply to exclude children. It also avoids possible problems due to children having different motor patterns because their walking style is still developing. In this sense, recent amputees were also avoided.
- *Gender* - The majority of amputees that were available for testing were male, thereby it seemed natural to impose this restriction rather than trying to find a similar number of females.
- *Amputation side* - The side of amputation need not actually be a concern because it would be simple to transform left and right sides into common parameters as suggested by Lafuente [124]. However, it coincidentally transpired that all the available amputees had amputations on the right side, which meant that no conversion of the data was necessary to make them comparable.

Therefore even given these customisation restrictions, data was still required that would represent as much as possible the morphological spectrum. In particular, variances dependent on the subject's mass and height was thought to have a significant bearing on gait patterns. In addition, amputees that use a variety of standard prosthetic components such as feet rather than just one type of foot should be sampled.

Ideally, the sample of subjects should be sufficient to enable normalisation of data according to such relevant subjective parameters.

7.1.2 THE ACQUISITION OF DATA

The problem posed by this project was that it was not possible to know which choice of outcome variables was most suitable as task variables prior to analysis. Although the importance of dynamic variables has been described [section 4.3.2], the possibility of other variables being pertinent was also recognised. Thus, it was necessary to measure as many variables as possible, particularly those of a kinematic nature. In addition, variables related to control processes were also required, i.e. kinetic variables in order to establish phases of a gait cycle and the eventual feed-forward reference profiles. Therefore, a holistic approach was required to simultaneously gather both kinematic and kinetic data.

7.1.2.1 KINEMATIC DATA ACQUISITION

As already described in the review chapters [section 2.2.2], the gathering of movement data has been practiced for many years. The problem of simultaneously measuring as many variables as possible was also encountered by many researchers who wanted to explicitly portray movement patterns as fully as possible. One of the most significant advances was the use of camera systems that could capture multiple frames whilst subjects performed; subsequent analysis would then yield many kinematic variables associated with the segment dynamics. This approach seemed more apt for the purposes of this project than the alternative of having to design many different individual sensors and fit them to the subjects.

The camera system that was available in the bioengineering unit was the Vicon 370 [209]; the validity of this system has been reported in independent tests where it was shown to be second only to the “Elite plus” camera system in terms of accuracy and noise [72]. Apart from availability, the Vicon system also had the advantage of enabling simultaneous data acquisition from other devices so that, for example, signals derived from force plates could simultaneously be taken and linked with the camera data so that both kinematic and kinetic variables can be calculated.

7.1.2.1.1 Introduction to Vicon 370 "Camera" System

Essentially, the Vicon data acquisition system has up to six cameras that strobe infrared light out, so that the subsequent reflected infrared image from strategically positioned retro-reflective markers can be captured. The simultaneous images from each camera are processed to yield the 3D co-ordinates of any markers that are in the field of view. The efficacy of this, for biomechanical analysis, is that markers may be positioned on a subject thereby allowing their point of attachment to be tracked. If markers are appropriately sited, their locus can be related to specific anatomical structures common among subjects. Given a sufficient number of markers, (minimum of three), this then provides the opportunity to establish the relative orientation and position of any rigid body that the markers are placed upon, which for biomechanical applications enables the tracking of various segments associated with the skeletal structure. The illustration in Figure 7-1 shows the overall arrangement that was used to gather data; the following sections will describe this arrangement in more detail.

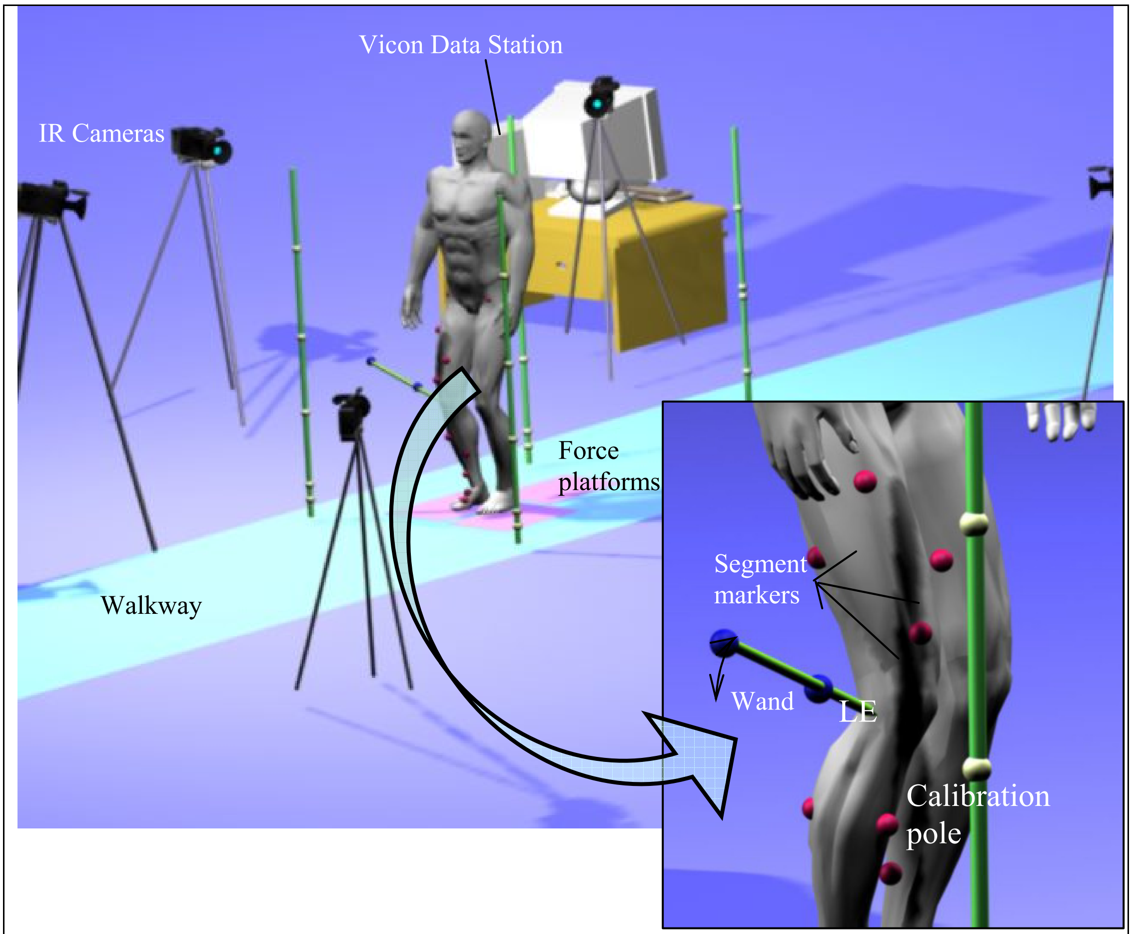


Figure 7-1 - Illustration of experimental procedure for acquiring movement data. Note that the force platforms and walkway have been coloured for clarity; they are not ordinarily obvious to the subject. The inset picture is a zoom up of a portion of the larger picture, although rotated to try to exemplify the IR reflective marker protocol that was used. See text for detailed explanation.

7.1.2.1.2 Camera set up

The set up of the cameras requires consideration of where the markers will be in space at all times when data is being recorded. Therefore, a walkway was identified within the lab in which the subject would perform the task. This area was more or less imposed because of the need to use the force plates, which were embedded into the floor; this region thus formed the focus for the cameras (see Figure 7-1).

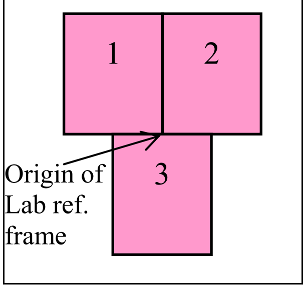
7.1.2.1.2.1 Camera Calibration

The purpose of camera calibration is to determine the position and orientation of the cameras with respect to the global or lab reference frame.

7.1.2.1.2.1.a Calibration Reference Object

Static calibration was undertaken which utilised a **calibration reference object** that is accurately positioned to pinpoint the origin of the lab reference frame as well as define the X, Y, and Z-lab axes. Bearing in mind the dimensions of the walking area, which should be suitable for the experiments described in section 7.1.1, a calibration object was chosen of a similar order size. This occupied a volume of about $8m^3$ around the force plates, as opposed to a few cm^3 as might be required for recording hand motion for example. The standard reference object that was used consisted of poles, which were accurately suspended from the ceiling. Four poles were used each of which had three markers in various arrangements along their length thus forming a solid shape [Figure 7-1]. The origin of the lab reference frame was situated (in this case) at the intercept between the three force plates [Figure 7-2]; its location, once defined, was assumed to remain constant relative to the calibration reference object for all subsequent experiments utilising this calibration object.

Figure 7-2 Force plates



7.1.2.1.2.1.b Camera Positioning

The calibration object was thus used to calibrate the position and orientation of each camera. The advantage of using this static reference frame was that it provided a point of focus for each camera that related to the experimental field of view; each camera was therefore positioned using live feed to a monitor. Six cameras were distributed evenly around the data collection region whilst ensuring that they did not impede the subject. They were placed sufficiently far to be able to view the whole calibration object (about 2-3m from the origin), and were set at different heights that were thought best for capturing markers attached to the lower limbs. The sensitivity of each camera was adjusted so that the markers were just visible; this helped to reduce unwanted noise that could be problematic later on.

When positioning the cameras it was also preferable not to directly view IR beams formed by the other cameras, otherwise blobs were seen in the image, which interfered with the marker reflections. This was best avoided by adjusting the height or position of cameras; however, as cameras were individually adjusted it was necessary to recheck the earlier cameras, to ensure that one of the latter IR sources had not come into view. Finally, spurious specula reflections from the floor for example were encountered which were also undesirable. These could be reduced by placing non-reflective coverings over the problem area.

Once the cameras had been satisfactorily positioned and aimed, the Vicon was used to calibrate the system; a resultant calibration residual[38] indicates the success and accuracy of the system [209]. For example, if residuals were greater than 5mm then the calibration was rejected. To try to improve this, camera positions were slightly adjusted and then the calibration performed again until satisfactory residuals were obtained (sometimes all this was easier said than done).

7.1.2.1.3 Marker Placement

The calculation of many kinematical variables in biomechanics is based on the assumption that they are performed on a solid object, i.e. one that is non-deformable; it would therefore be preferable to track anatomical structures which preserve this assumption as closely as possible to ensure accuracy. Although individual bones may be said to approximate such a solid structure, they are not however directly accessible given all of the surrounding soft tissues (skeletons do not walk unassisted alas). It is also not feasible to overcome this by attaching the markers through the soft tissue and on to the bone, as this requires invasive procedures that were inappropriate for the purposes of this project. To further compromise the assumption that the segments of interest are solid, individual segments such as limbs, may be composed of more than one bone thereby allowing a greater extent of deformation. Given these problems, the placement of markers to track the position and orientation of a body segment must be carefully considered.

Furthermore, another criterion also had to be considered before choosing a protocol for marker placement. Although any three points on a solid object can be used to define a set of three mutually orthogonal axes, e.g. as described by Berme et al [18],

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[38] Calibration Residual for each camera is the average distance by which the reconstructed rays from the calibrated camera to each reference point miss the location of the reference point.

it is often preferable to align and position such a local frame of reference according to clinical relevance. Such methods are advantageous because they standardise sets of data on an inter and intra subject basis, whilst also giving perceptive clinical meaning to the variables that can readily be calculated, such as the joint dynamics.

7.1.2.1.3.1 Standardised Local Reference Frames (Anatomical frame)

Fortunately, such ideas are not new, and various standards and methods have been proposed in the field of biomechanical study. For example, Wu et al. [229] recommend a convention for the global and segmental local centre of mass reference frame. However, the proposal that was felt to be most pertinent to this project was the convention suggested by Cappozzo et al [37]; they contrived a clinically relevant set of reference frames for the lower limbs. A bone embedded local frame of reference is described, which can be located mostly by palpation of the relevant structures[39]. The reference frames, which they suggest, have their axes so that “both rotations and translations of the joint may be defined (joint axes)”. Such a frame also enables the centre of mass frame to be easily located.

Using Cappozzo’s protocol, anatomical frames were defined for the pelvis, thigh, shank, and foot using the anatomical landmarks listed in Table 7-A.

Anatomical Frame	Anatomical Landmarks	Abbreviations
Foot	Dorsal Aspect of 1st Metatarsal	FM
	Dorsal Aspect of 2nd Metatarsal	SM
	Dorsal Aspect of 5th Metatarsal	VM
	Upper ridge of Calcaneus posterior surface	CA
Shank	Distal apex of the Lateral Malleolus	LM
	Apex of Head of the Fibula	HF
	Distal apex of the Medial Malleolus	MM
	Prominence of the Tibial Tuberosity	TT
Thigh	Lateral Epicondyle	LE
	Medial Epicondyle	ME
	Centre of the Femoral Head[39]	FH
Pelvis	Anterior Superior Iliac Spine	AS
	Posterior Superior Iliac Spine	PS

Table 7-A Anatomical bone embedded landmarks that are used to define the local/anatomical frames of reference.

The anatomical frames for each segment are thus constructed according to the guidance of Table 7-B.

[39] The femoral head (FH), unlike other landmarks, cannot be found by palpation. However, since other palpable landmarks (left and right AS & PS) exhibit consistent relative proportions with the FH, between subjects, these were used to estimate the position of FH [180]

Anatomical Frame	Origin of frame	Plane A	Plane B	Axes
Foot	CA	**Transverse plane** Defined by CA, FM & VM	**Sagittal Plane** Orthogonal to plane A going through CA & SM	**y axis** Intersection of plane A & B **z axis** lies in the transverse plane orthogonal, to the sagittal plane. **x axis** orthogonal to the yz plane.
Shank	Midpoint of LM & MM	**Frontal Plane** Defined by LM, MM & HF	**Sagittal Plane** Orthogonal to plane A going through TT & origin.	**y axis** Intersection of plane A & B **z axis** lies in the frontal plane, orthogonal to the sagittal plane. **x axis** orthogonal to the yz plane.
Thigh	Midpoint of LE & ME	**Frontal Plane** Defined by LE, ME & FH		y axis Line through origin and FH **z axis** lies in the frontal plane, orthogonal to the y axis. **x axis** orthogonal to the yz plane.
Pelvis	Midpoint of left & right AS	**Transverse Plane** Defined by left & right AS and midpoint of left and right PS.		**z axis** Line through left & right AS **x axis** lies in the transverse plane, orthogonal to the z axis. **y axis** orthogonal to the xz plane.

Table 7-B Construction of local frames of reference using anatomical landmarks listed in Table 7-A.

This method of construction presented in Table 7-B is illustrated in Figure 7-3. The various anatomical landmarks and the construction planes are presented and the colouring scheme is consistent for plane A and B and the axes definitions.

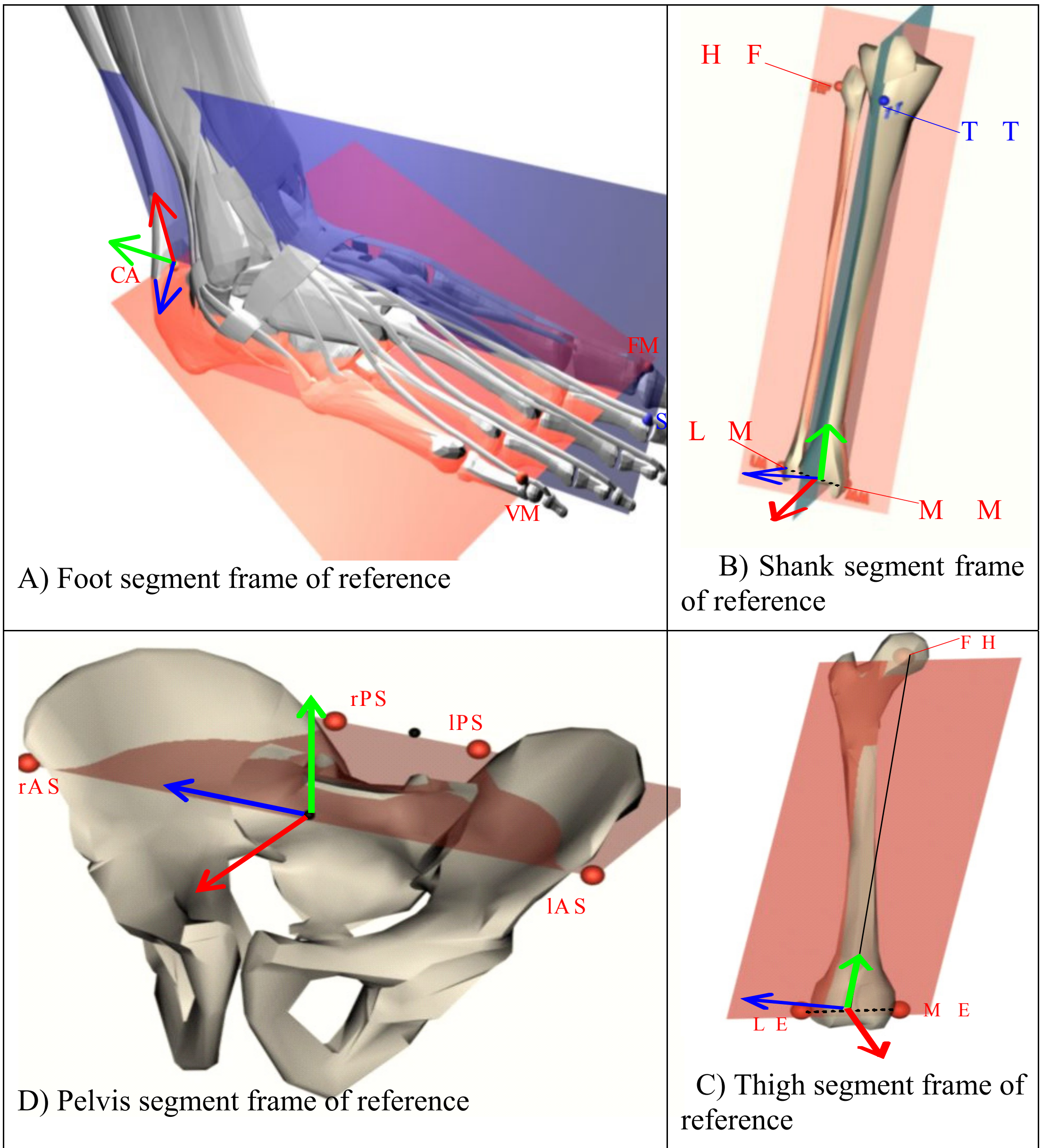


Figure 7-3: Bone Embedded anatomical frames as redrawn by the author according to the diagrams of Cappozzo [36]. The axes were constructed from the planes described in Table 7-B, where plane A is coloured red, and plane B blue. Hence red arrows are used to represent the x-axis, green the y-axis and blue the z-axis, all of which start at the specified origin of the anatomical frame of reference.

7.1.2.1.3.2 Locating Landmarks used to define Local Frames of Reference.

It is most convenient to position reflective markers directly over the anatomical landmarks of interest, i.e. those chosen to construct the segmental coordinate systems. Then, by neglecting the thickness of soft tissue and any tangential movements between the marker and the bone site[40], it can be assumed that the markers track the desired anatomical landmarks. However, apart from these inaccuracies, the method was fallible in the context of this project because anatomical landmarks do not exist for segments that lie distal to the trans-tibial stump. A method was consequently required, which was compatible for both the remaining anatomical structure and prosthetic components of the trans-tibial amputee.

The chosen technique utilised an alternative method of pinpointing anatomical landmarks as also suggested by Cappozzo [36]. With this method there is no requirement to position the reflective markers directly over the anatomical landmarks. Instead, they (a minimum of three markers) can be sited any where on the segment of interest, so long as they are not co-linear to each other. The idea is based on the assumption that the position of landmarks (anatomical or otherwise) should maintain the same geometric position with respect to the randomly positioned markers if the body is a solid and the markers do not move relative to this solid. Therefore, by measuring this geometric relation, the coordinates of the landmarks can subsequently be calculated simply from the coordinates of the three random markers. The practical implication of this was that it became possible to define "**Virtual landmarks"** anywhere in space relative to the reflective markers; this idea was used to locate prosthetic landmarks.

7.1.2.1.3.2.a Placement of random markers on lower limb segments

As mentioned, at least three markers must be positioned on each limb segment whether it is prosthetic or natural. The following guidelines were developed for placing these markers. The most important criteria were to ensure that the Vicon system would be able to view each marker with at least two cameras, and that the system was able to resolve the identity of markers that are close to each other. In addition, for the sake of practicality and convenience, areas where the markers can

[40] The difference in soft tissue thickness between most subjects should be comparatively small, so that placing the marker on skin should still enable a standardised protocol. Of course, the possibility of errors due to relative movement between skin and bone is however greater.

easily be knocked off or simply fall off were avoided. For example, a swinging arm can easily obscure markers or even knock them off if they are in its path; similarly, the medial surface of the leg was left alone and the flexing region of shoes (i.e. the metatarsal crease) was avoided. Finally, for increased accuracy less fleshy areas were also desired which did not create too much unwanted tissue movement.

To further enhance the prospect of the Vicon capturing at least three markers per segment, throughout the period of data gathering, it was decided that four markers per segment should be attached. In this manner, geometric relationships could be found between the anatomical landmarks and each set of three markers so that even if one marker was missed for an interval during the trial, the anatomical landmark position could still be calculated from the remaining three markers.

Two methods of attaching markers to the segment were available, using an elastic cuff with pre-attached markers or by simply sticking the markers directly onto the skin or the prosthesis. Some researchers suggest that an elastic cuff, worn by the subject, is advantageous because it is convenient, and in some respects restricts unwanted cutaneous movement as the soft tissue is restrained by the elasticity of the cuff. However, although the tissue movement may be reduced, the drawback is that it is more sensitive to relative movement effects between bone and skin as all the markers will be correlated with each other because they move as one entity with the cuff. The alternative, of directly sticking markers to the skin, can reduce such skin movement artefacts as each marker will move independently so that the combination will smooth the noise and pinpoint the segment reference frame more accurately. Attaching individual markers also makes it easier to customise the layout for each subject and each camera set up. Therefore, during experimentation, individual markers were attached to subjects using double-sided sticky tape. Care was taken to position markers where they would be clearly visible and not too close to each other so that the Vicon was still able to resolve them. This resolution limit was also a factor in limiting the number of markers per segment to four, as it would not be possible to squeeze in more markers whilst leaving adequate spacing between them. Finally despite the above discussions, it was thought that because of the distribution of the pelvis' anatomical landmarks and the fact that these were fairly bony areas, it was reasonable to place the markers directly over them rather than using the random marker method.

An example of such a marker protocol can be seen in Figure 7-1 (see zoomed up inset for clearer view) where the subject has four markers attached to each segment of interest, which in this case relates to the right leg and pelvis.

7.1.2.1.3.2.b Pinpointing Anatomical landmark positions relative to random segment markers (Calibration Trials)

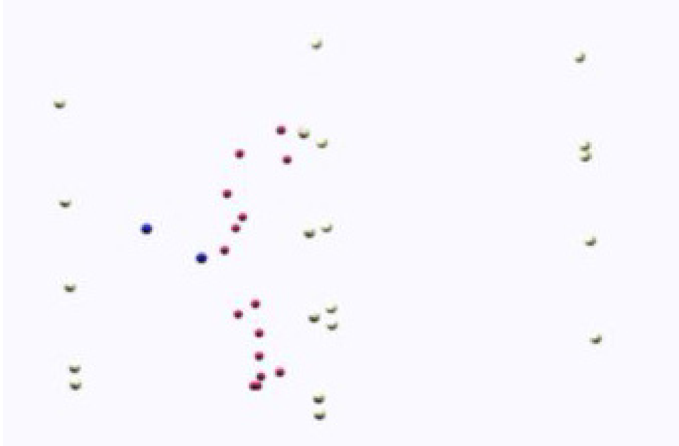


Figure 7-4 – The reconstructed markers in Figure 7-1 from the same viewpoint. This is all the Vicon will "see" since it only detects the infrared reflective markers. Colour has been added to aid presentation where red represents the segment markers, blue the wand markers, and yellow the calibration object markers.

The method used to capture the geometric relationship between the random markers and (virtual) anatomical landmarks involves using a pointer with two reflective markers attached along its length (this will be referred to as a wand). Calibration data is hence captured, with the Vicon, in which the tip of the wand is positioned on the landmarks. To improve the cameras' chance of capturing the wand, it was swivelled around using the landmark as the pivot point (see inset of Figure 7-1). Therefore, a separate calibration trial was created for each landmark that would be used to define the local axis systems. The global co-ordinates of the landmarks were then derived from the wand trajectories during analysis. The advantage of using a wand is that it allows accurate definition of anatomical landmarks, and it avoids the problem caused by lack of resolution, for example if anatomical landmarks are too close to each other for direct placement of markers. Since markers do not have to be placed on any specific location on the segment, awkward areas can also be avoided.

During the calibration trials, it was important for the person using the wand not to obscure the markers with their body. As a precaution the operator of the Vicon would check that all of the required markers (i.e. the wand and markers on the segment of interest) had indeed been captured before calibrating the next landmark. To do this the camera data had to be reconstructed so that the 3D positions of all reflections resembling a marker could be graphically presented. An example of this reconstruction is given in *Figure 7-4*, where the illustration of Figure 7-1 is shown from the Vicon's point of view (i.e. only IR reflections are visible). It should be noted that the calibration reference frame would ordinarily have been removed so that its markers (shown in yellow) would not add further to the confusion of segment and wand markers. By rotating the angle of view displayed by the Vicon and also zooming in and out, it was easier to try to spot if any markers were missing, a fiddly operation which required patience and practice.

7.1.2.1.3.2.c Virtual Landmarks for Prosthetic components

Since there were no palpable anatomical landmarks on prosthetic components, a standard practice was also required to define where these should be. This protocol essentially involved pointing at the imaginary positions where non-amputated anatomical landmarks would have been if amputation had not occurred. This was achieved by referencing the sound limb. For example, LM and MM for the shank can easily be mirrored from the sound ankle whilst the subject stands. This task is made easier if they are wearing a cosmetic foot covering which approximates to the surface of the ankle. The foot landmarks were of interest because it was decided that amputees should keep their shoes on whilst data is gathered as this is most likely to be how they will spend the majority of time walking. Therefore, the accuracy of locating the foot landmarks on the sound side was reduced which also had a knock on effect to the mirroring process. However, with practice, a standard technique was developed which was transferable between subjects and provided intra-subject repeatability. Similarly, although trans-tibial amputees possess the TT, HF, LE and ME landmarks, they are obscured by the socket, however with the aid of the sound side and the shape of the socket this was adequately dealt with.

Therefore an approach was used which was originally developed to reduce skin movement artefacts, however it also, fortuitously, provided the author with a means of addressing issues concerning the lack of palpable landmarks and limits on how close markers can be placed next to each other.

7.1.2.1.3.2.d Transformation of Local Reference Frames in Calibration Trials to the Experimental Trials

It was therefore possible, using the wand trajectories, to derive the global co-ordinates of each individual landmark within the calibration trials. These were subsequently used to calculate the local frames of reference during the experimental trials [section 7.2.1.3].

7.1.2.1.4 Timing Parameters

7.1.2.1.5 Speed detection

As described, it was thought that speed dependent variables were likely to be predominant candidates for control variables. Therefore, to provide an indication of what such a variable should be or rather give reference values for subsequent analysis, it was necessary to measure the speed of walking. Of course, the speed of the body segments varies harmonically throughout the gait cycle, where the head experiences the least fluctuation. However, it was thought better to represent the overall task by the average speed.

Two methods were thus used in order to measure the average speed of the subject and allow a comparative verification. The first was to measure the time it took the subject to walk over a short distance (6.08m as it turned out). Two infrared beams, placed at chest height, were thus used to trigger the start and stop of a timer; the subjects were given a sufficient walk up to ensure they had accelerated to a steady walking speed before the start and could maintain it before the watch was stopped. The second method involved using the trajectories of particular markers or anatomical landmarks or joint centres to measure the distance they had travelled over a particular temporal period. It was important, because of the periodicity of walking, that the temporal period was taken for a whole gait cycle e.g. as measured from initial contact (IC) to IC or toe off (TO) to TO. Otherwise, a particularly fast or slow interval might be taken rather than the overall average speed.

With the beam method, the speed was taken over several stride lengths whereas the other method only allowed for one stride-length. However, the second method enabled a better synchronisation of the timing points i.e. so that harmonic speed fluctuations were reduced by measuring between periodic gait events.

7.1.2.2 KINETIC DATA ACQUISITION

Rather than directly measuring the forces, moments, and powers accruing across joints, an **Inverse Dynamics** approach was used to calculate such variables. This technique relates the overall acceleration of the entire body mass, as measured by ground reaction forces, into its constituent parts pertaining to the kinematic properties of each segment. The details of this method of analysis will be described later, however, from an experimental point of view, there were certain prerequisites in order for such calculations to take place.

7.1.2.2.1 Kistler Force Plates

The foremost of these requisites was the necessity to measure the ground reaction forces. This was achieved using the Kistler Force plates, which were preinstalled, into the lab floor. Each force plate outputs a variety of signals in response to any force applied to the plate. These signals correspond to the:

- Three components of the applied force, which are oriented in convention with an orthogonal Cartesian system.
- Moments about predefined axes caused by the applied force

The signals are derived from Piezo-electric force transducer crystals, which are positioned at each corner of the force plates.

The Vicon system incorporates an analogue to digital converter (ADC), which can have up to 32 separate channels of analogue input. These channels are synchronised with the camera data. The first 20 channels are dedicated to the signals from the three force plates. All these channels had been calibrated so that the components of the ground reaction force were given in Newtons (N), and the moments in Nm.

7.1.2.2.2 Body Parametric Measurements

Using the signals derived from the force plates it is possible to calculate the resultant ground reaction force caused by the acceleration of the whole subject mass. However, in order to use inverse dynamics to calculate additional kinetic variables such as the moments and powers experienced across joints, certain parameters relating to the subjects must be measured. These include relative distances to the centres of gravity (COG), mass, density and most awkwardly the moments of inertia, of segments. The underlying mathematics relating to these parameters is described in section 7.2. The segments in question relate to each articulation of the lower limb [section 7.1.2.1.3], i.e. the foot, shank, and thigh.

Although techniques have long been available to measure and gather data on segment mass related parameters [48, 49, 63, 45, 97], some of these are lengthy procedures. A convenient method was required for the project, which did not place the subject in too much discomfort. The mass of segments is one of the most basic parameters that is required for the kinetic calculations, however this is not easy to directly measure given the inseparability of the segments; the method of reaction change is normally used for such direct measurement [63]. However, to simplify matters, it was decided instead to calculate the mass according to the ratio described by density. It was thought that the values for density and volume could be more easily obtained or estimated with reasonable accuracy than the segmental masses.

7.1.2.2.2.1 Segment Density Estimations

Segment	Density
Upper Arm	1.081
Forearm	1.122
Hand	1.144
Thigh	1.069
Shank	1.095
Foot	1.100
Head and Neck	1.111
Trunk	1.030

Table 7-C – Average Body Segment Densities (kg/l), compiled by Drillis et al [63]

The density of body segments is required in order to perform the various kinetic calculations. Such densities will be affected by the composition of the various segments in terms of bone, fat, muscle etc. There will also be inter-subject variations dependent on age, sex, and body build. Therefore, ideally each segment density for each subject would be measured and if it were possible its distribution within the segment would also be known. However, given the inseparability of segments with the exception of the prosthetic components, it would be time consuming, complex and consequently impractical to obtain such values for all segments on each individual.

Therefore, it was thought more appropriate to try to estimate the densities of natural segments using statistical techniques applied to data gathered from in depth investigations into these matters. Although fluctuations do occur between individuals, it was decided that they were insignificant as compared to differences in volume and mass between subjects. Therefore, it seemed feasible to use the data collected by various researchers for example as obtained from cadaver samples or by using the reaction change method on live subjects. The report by Drillis and Contini[63], compiles such results from several investigations to give an average density value for different body segments [Table 7-C]; these values were subsequently applied in the kinetic calculations.

7.1.2.2.2.2 Volume & Mass Measurements

Several methods were available for measuring the volume of various segments including using Archimedes' principle or by scanning the segments into a computer and producing 3D models of the segments. Although an attempt had been made to scan segments using the Polhemus scanner, which locates the 3D coordinates of a hand held pointer according to a magnetic field, its output format was too crude to be easily implemented into solid modelling such as "Autocad" and "3D Studio Max" by Kinetix. Therefore, the traditional approach of displacing a volume of water as devised by Archimedes was applied.

Although it is possible for the amputee to progressively displace water by succeeding proximal segments, it was decided to make a solid cast of each segment. This method provided for greater control and accuracy of what was being measured. It was also convenient because the cast was also needed to measure other segmental parameters. Plaster cast bandages were thus wrapped around the amputee's limb so that all the relevant anatomical segments were encased. The plaster cast was formed around the stump and thigh, reaching up as high as the greater trochanter. Once it had set, it was removed from the subject so that a mould of the whole limb was attained. This mould was cut in half through the transverse plane just above the knee joint centre so that the stump and thigh segments were separated. The distal end of the thigh segment was sealed/plugged with plaster bandages and the proximal end built up to form what represents the thigh region around the femoral head as distinct from the pelvis region.

In preparation for subsequent inertial moment measurements [section 7.1.2.2.2.3], two rods formed into cross hairs were inserted so that they aligned with the local x and z-axis of the segment [Figure 7-6]. The moulds were now ready to have plaster of Paris poured in them to create a solid cast of the stump and thigh segments. Note that although water could have simply been poured into the female mould and this volume of water used to estimate the volume of the segment it was preferable to produce a solid cast because of the error caused by spillage as the mould may not be watertight and besides a cast had to be made anyway. Any shrinkage of the cast was thought to be negligible especially if the original plaster bandages were not wrapped too tightly.

Therefore, the volumes of the solid plaster segments were measured according to Archimedes' principle. A large container was filled with water until it overflowed through a hole, which was cut near the top of the container. A tube was connected to the hole so that the water could be guided into measuring cylinders. Once the water stopped overflowing and was thus level with the bottom of the hole, the cast was gently lowered into the container until it was fully immersed. The displaced water was caught in measuring cylinders and the volume of each cast was measured.

The mass of the segment was then found by using the density estimations in combination with the measured volume.

7.1.2.2.2.3 Segment Moments of Inertia Measurements

Moment of inertia is "the rotational equivalent of mass in its mechanical effect, that is, the resistance to a change of state (a speeding up or slowing down) during rotation" quoting Rodgers & Cavanagh [168]. The mass moment of inertia (I) will depend upon how mass is distributed about an axis of rotation so that $\mathbf{I} = \sum_{i=1}^{n} m_i r_i^2$, see Figure 7-5 below.

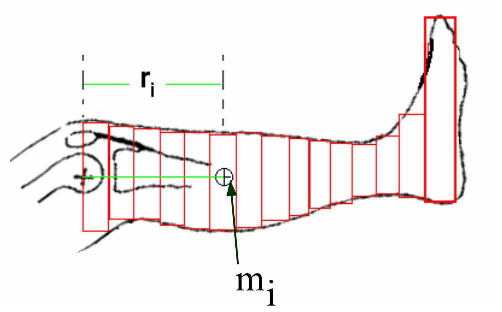


Figure 7-5; Example of the distribution of mass elements in the shank segment for rotation about the medio-lateral axis.

Numerous methods were available for measuring or estimating the mass moment of inertia [63, 49]. The method that was finally used was the compound pendulum method, which relied on measuring the period of a freely swinging body. With this method, it was possible to use the casts of the leg segments since objects of the same shape will have the same radius of gyration. Therefore, if the mass of the actual segment has already been estimated then it is a simple matter to convert the cast inertia to the actual inertia.

7.1.2.2.2.3.a Compound Pendulum Method

If an object of mass (m) is allowed to swing freely (with no friction) over a *small angular range* about an axis, then its *period of oscillation* (T) is related to its *inertial moment* about the *swing axis* (I_{swing}), and the *distance* (h) from the *swing axis* to the *centre of gravity* (COG) of the object so that

Equation 7 $$T = 2\pi\sqrt{\frac{I_{swing}}{mgh}}$$

Thereby this equation could be applied to calculate the swing period of the segmental casts that were made for volumetric measurements [section 7.1.2.2.2.2], and by using the parallax theorem to expand I_{swing} in terms of the radius of gyration (K) and h, the swing period could be written:

Equation 8 $$T_{cast} = 2\pi\sqrt{\frac{K_{cast}^2 + h_{cast}^2}{gh_{cast}}}$$

Since the angular velocity (ω) = $2\pi/T$, Equation 8 can be rearranged to yield the radius of gyration about the cast's centre of gravity (K_{cast}).

Equation 9 $$K_{cast}^2 = \frac{gh_{cast}}{\omega_{cast}^2} - h_{cast}^2$$

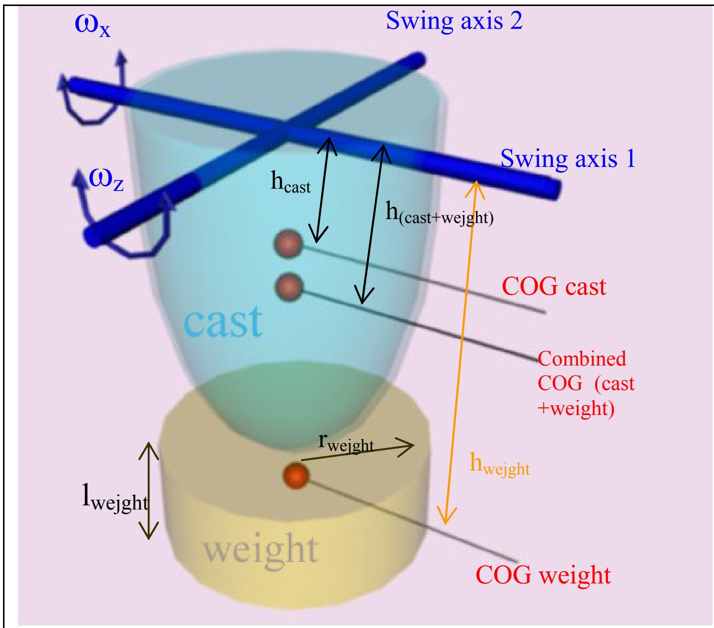


Figure 7-6: Compound pendulum set up. The swing period of the segment cast is first measured on its own, then the weight is attached, and the period measured again.

It can be seen from Equation 9 that it is necessary to measure h_{cast} and ω_{cast} to obtain the radius of gyration. There are several options available to locate the centre of gravity of the segment, and consequently the distance h_{cast}; these methods mostly rely on finding the balance point using moments. However, it was thought that by slightly altering conditions for the compound pendulum, it would be possible to solve for h_{cast} by generating two simultaneous equations. As a precaution, the technique of balancing moments was also used to measure the position of the segment COG, in order to cross validate the compound pendulum apparatus and estimate its accuracy.

The technique was to attach a weight of known dimensions to the cast and perform the pendulum experiment using these two combined objects [Figure 7-6]. Once again, it was possible to expand Equation 7, although this time I_{swing} was expressed as the combination of the separate inertial moments about the swing axis of the cast and the weight, i.e.:

$$I_{swing} = I_{(cast+weight)} + m_{(cast+weight)} h_{(cast+weight)}^2 = m_{cast}(K_{cast}^2 + h_{cast}^2) + m_{weight}(K_{weight}^2 + h_{weight}^2)$$

The distance from the swing axis to the COG of the combined object's ($h_{(cast+weight)}$) could also expressed in terms of the individual components h_{cast} and h_{weight} by equating the pivotal moments about the COG, so that

$h_{(cast+weight)} = \dfrac{m_{cast}h_{cast} + m_{weight}h_{weight}}{m_{(cast+weight)}}$. By substituting these elements into Equation 7 and manipulating, the swing period of the entity may be written in terms of, K_{cast}, K_{weight}, h_{cast} *and* h_{weight}:

Equation 10 $$T_{(cast+weight)} = 2\pi\sqrt{\frac{m_{cast}(K_{cast}^2 + h_{cast}^2) + m_{weight}(K_{weight}^2 + h_{weight}^2)}{g(m_{cast}h_{cast} + m_{weight}h_{weight})}}$$

Thus by once again introducing the angular velocity into the equation, substituting K^2_{cast} with Equation 9, and rearranging, it was possible to solve for h_{cast}:

Equation 11 $$h_{cast} = \frac{gm_{weight}h_{weight} - \omega^2_{(cast+weight)}m_{weight}\left(K^2_{weight} + h^2_{weight}\right)}{gm_{cast}\left(\dfrac{\omega^2_{(cast+weight)}}{\omega^2_{cast}} - 1\right)}$$

With the exception of K_{weight}, this equation was now in terms of measurable parameters; however, by carefully choosing a weight with known geometry, it was possible to apply standard predefined formulae to calculate its radius of gyration. The chosen weight was a steel cylindrical block with dimensions r_{weight} by l_{weight} [Figure 7-6]. The expression for the moments of inertia of such a cylindrical block rotating about an axis running through the transverse plane is $I = \frac{1}{4}mr^2 + \frac{1}{12}ml^2$, which allows K_{weight} to be easily derived.

The result of all this manipulation is that the only parameters that had to physically be measured to calculate h_{cast} and subsequently K^2_{cast} were:

- h_{weight} which was the distance from the axis of rotation to the COG of the weight (i.e. located half way up the height of the cylinder (l/2)).
- m_{weight} the mass of the attached weight.
- m_{cast} the mass of the cast.
- $\omega_{(cast+weight)}$ the angular velocity of the combined cast and weight entity as they freely swing (over a small swing angle)
- ω_{cast} the angular velocity of just the cast as it freely swings (over a small swing angle.

- r_{weight} the radius of the attached weight (assuming it's a cylindrical block)
- l_{weight} the height of the attached weight (assuming it's a cylindrical block)

All this manipulation to calculate h_{cast} therefore gave a meaningful way of verifying whether the apparatus was functioning properly, since as mentioned the distance h_{cast} could simply be confirmed by balancing the cast.

Ultimately, however, the object of this exercise was not to find I_{cast}, but rather the moment of inertia of the original segment ($I_{segment}$). However, because similar shaped objects have the same radius of gyration, then it was simply a matter of taking the product of the estimated segmental mass and K^2_{cast} to calculate $I_{segment}$. The mass was estimated from the measured volume section [section 7.1.2.2.2.2] and the density estimates [section 7.1.2.2.2.1] of the particular segment.

In practical terms, a variety of methods was devised to try to measure the swing periods and thus the angular velocities of both the cast on its own and the cast with the added weight. These are discussed more in the results [section 8.2.2.1.2 and 8.2.2.1.4]. It was fortunately possible to base the practical equipment on the rig that Stewart [194] had made to take similar measurements during her PhD.

7.1.2.2.2.3.b Torsional Pendulum Technique

The above technique was used to measure the inertial moments about the x, and z local segment axes (i.e. in the frontal and sagittal plane). However, it was more convenient to use a slightly different method using a different formula to measure inertia in the transverse plane, or the long axis. This involved the torsional pendulum technique in which the cast was suspended vertically using a trifilar suspension system (i.e. a harness attached by three strings). The strings were equidistant and the cast was suspended in the centre so that it was free to rotate about its y-axis. The formula for the inertia using this method is given below:

Equation 12

$$I_y = \frac{mgr^2}{{\omega_y}^2 l}$$

Where I_y is the moment of inertia about the long axis

r is the radius of the circle that the strings describe.

m is the total mass that is suspended by the strings (this includes the harness mass.

l is the length of the strings

It should be noted that because the harness rotates with the cast, its inertia would also be included in the above formulae. Therefore, this must be subtracted from I_y to yield the inertia of the cast. The apparatus inertia can be found by simply performing the experiment with only the harness attached to the strings.

Finally, because the formulae used in both pendulum methods require that only simple harmonic motion is present, the swing must therefore be kept small. In practice, this meant that the period would be slightly dependent on the size of the initial swing; therefore, the procedure was repeated many times to try to ascertain the variability of the measurements. An algorithm was thus written which analysed every combination of angular velocity (with and without a weight) to give the spread of h_{cast}. These were then consequentially used with the original ω_{cast} values to yield all possible moment of inertia values.

7.2 ANALYSING DATA

Once the experiments have been performed for a subject, as outlined in section 7.1.1 (pp. 195-202), and data acquired via the Vicon system, the data had to be processed in order to be useful and have meaning.

7.2.1 PREPARATION OF DATA FOR ANALYSIS

The Vicon cameras captured IR reflections at a predetermined rate; its software used the combination of these images to *reconstruct* only reflections that were most probably from spherical markers into a 3D mapping. Reflections that did not fit this spherical template were supposed to be ignored by the reconstruction algorithm. The cameras were left to sample at the standard rate of 50 Hz, which just lies on the borders specified by constraint 3; a higher rate was not possible because the cameras would become de-synchronised with the strobe and would cause a reduction in the camera imaging area. The reconstruction thus left a plethora of markers, which were indistinguishable from each other as they were monochrome, unlike the illustration in Figure 7-4.

Therefore, before this animated reconstruction of the visible markers could be used for kinematic analysis some pre-processing was necessary.

Segment	Label
Wand Markers	Distal Proximal
Left thigh markers	LT1 LT2 LT3 LT4
Left shank	L1 L2 L3 L4
Left foot (shoe)	LF1 LF2 LF3 LF4
Right thigh	RT1 RT2 RT3 RT4
Right shank	R1 R2 R3 R4
Right foot	RF1 RF2 RF3 RF4
Pelvis	LAS LPS RAS RPS
Cervical	C6

Table 7-D Marker Labels

The first step for preparing the data was to identify and label all of the segmental markers that appeared in each frame of the reconstructed trials (this included the calibration trials). A protocol was developed for labelling these markers so that subsequent algorithms could readily sort the data for analysis. In this protocol, the markers were classified according to which segment they were attached, and were labelled in the order given by Table 7-D. The Vicon software facilitated 3D viewing of reconstructed markers by allowing rotation and zooming of the field of view; using such manoeuvres and with practice, it was possible to identify whether markers were genuine or noise apparitions. Then the markers were clicked in the correct order and associated with the list of labels. Once labelling was finished, the processed video data was saved into C3D files that contain the capture and reconstruction parameters, the 3D trajectory data, and any analogue data that was unprocessed. It was also possible, before saving, to delete the trajectories of any markers that remained unlabelled thus avoiding noise from spurious data.

Unfortunately, the process was very time consuming and tedious and mistakes would permeate through subsequent analysis necessitating backtracking. Attempts were made at using the automatic labelling system, which relied on using relative spatial distances between sets of markers, however this was not successful, possibly because of too much relative movement between sets of markers. The use of elastic cuffs might have improved the chance of the automatic labeller however this method was avoided for the reasons given previously [section 7.1.2.1.3].

7.2.1.1 EXPORTING 3D CO-ORDINATES

The 3D trajectory data contained within the C3D files was extracted and converted into a TXT file using C3Dexporter; this also enabled the user to select the time range in which relevant data was thought to exist within a trial. A matrix was then produced containing in each column the X, Y, Z co-ordinates of each marker as ordered in Table 7-D. The rows represented the particular frame of data over the range that was manually selected.

"MATLAB" is analysis software, which was primarily designed for matrix manipulation [132]; it provided a high-level language and tools with which codes could be written. This software was thus primarily used to analyse and display the data gathered from the Vicon system. The basic structure of the codes for kinematic and kinetic analysis was available through a previous project within the department undertaken by van der Linden [205]; these were subsequently modified for the purposes of this project.

Therefore, MATLAB was used to upload the data and organise them according to segments so that various kinematical calculations could be performed. This included both the calibration trials, and the experimental trials.

7.2.1.2 PRE-PROCESSING THE TRAJECTORIES OF THE IR MARKERS

To improve the quality of the eventual kinematical analysis it was essential to pre-process the trajectory data at an early stage otherwise noise and errors would propagate through the analysis.

7.2.1.2.1 Average Anatomical Landmark Co-ordinates in Calibration Trials

As mentioned already [section 7.1.2] the end of a wand is used to locate the anatomical landmarks (including virtual ones) whilst the subject stands still during the calibration trials. To increase the chance of the cameras capturing the two markers on the wand, the wand was moved about the anatomical landmark, the endpoint always being in the same place. Therefore, as a first pre-processing step, the averages of the calculated endpoint co-ordinates, taken over the duration of the calibration trial (1 second), were used as the landmark co-ordinate. The use of averaging was intended to reduce the noise that may be present in the trajectory data.

7.2.1.2.2 Patching Gaps smaller than 4 frames

It was quite possible that during a trial the Vicon system would lose track of a marker that had been observed in other frames. Therefore, since the position of the anatomical landmarks during the experimental trials depends on having as much marker data as possible it was desirable to try to patch some of these gaps without distorting the data.

Therefore, given the relatively low frequency of trajectory data, it was thought reasonable to patch gaps of less than four frames (sampled @ 50Hz) using linear interpolation techniques. A linear relation was thought to hold as long as only a small window of data around the gap was used. Since this was a rough benchmark and depended on well-behaved samples of data, the results were also checked visually to ensure validity.

In addition to using linear interpolation for patching vanishing markers it was also possible to estimate where a missing marker should be if the other three segment markers were visible during that frame. This used a similar technique to that in section 7.1.2.1.3.2 where the position of landmarks was transformed from the calibration files to the experimental trials. However to avoid information replication the same reference set of segment markers were not used as in 7.1.2.1.3.2 but instead a complete set of relevant segment markers were found from another frame in the experimental trial or if that was not possible, from a different set of calibration trials. The transformation matrix could thus be found between the reference set of markers and the three visible markers. The algorithm used the method devised by Veldpaus [208, 190] to estimate the transformation matrices. This was useful because it not only gave the rotation and translation matrices but also an associated error depending on how closely the two sets of markers matched each other.

7.2.1.2.3 Filtering

A 4th order low-pass digital Butterworth filter was applied to all of the segment marker trajectories gathered in the experimental trials. A cut-off frequency of 7 Hz was chosen for the filter, which was generally accepted as containing most of the information content on gait dynamics so that only high frequency noise components were removed.

7.2.1.2.4 Patching larger gaps

Once filtering had been performed, larger gaps were plugged using a third order polynomial. This was fitted to a portion of the marker trajectory that had at least five frames before and after the gap; its coefficients were used to evaluate the polynomial at selected points, which corresponded to any vanished marker frames. A visual check is once again performed in case the algorithm has unreasonably attempted to plug in large gaps.

7.2.1.3 MAPPING ANATOMICAL LANDMARKS FROM CALIBRATION TRIALS TO EXPERIMENTAL TRIALS

The location of anatomical landmarks is only known during the calibration trials, however their virtual position during the experimental trials must also be located if kinematic analysis of the body segments is to proceed. Because each segment was assumed solid, it was possible to make such estimations based on deriving the rotation and translation tensors between the segments in the calibration trials and their corresponding segments in the experimental trial. Four markers per segment were more than sufficient to theoretically track the position and orientation of the segments by enabling the calculation of rotation and translation matrixes as suggested by Veldpaus [208, 190]. Although three markers would ordinarily suffice, this chosen technique allowed the median position of all four markers to be tracked and a least squares algorithm could be used to decide the most probable rotation given that noise and movement artefacts were likely to introduce discrepancies between the calibration set of markers and the experimental set.

These transformations (between each segment and experimental frame) were then used to map the co-ordinates of related landmarks as calculated in the calibration trials to their positions within the experimental trials. Eventually a full picture of each landmark trajectory of each segment was built up throughout each experimental trial. If any of the transformations had errors greater than 8mm associated with them, then a visual check of the trajectories for the segment landmarks was performed to ensure that the correct landmark had been transformed. Hence, this method allowed tracking of landmarks using only the more conveniently placed segment markers.

7.2.2 CALCULATING TASK (KINEMATIC) VARIABLES

Once the trajectories of all the required landmarks were known throughout the experimental trials, this data was analysed to yield the kinematic variables relating to the movement.

7.2.2.1 DEFINING THE LOCAL/ANATOMICAL FRAMES OF REFERENCE

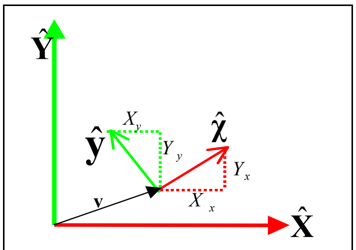


Figure 7-7: 2D view of local & global reference frames.

The first step was to define the local axis systems of each segment, as suggested by Cappozzo [section 7.1.2.1.3.1]; these would describe the orientation and position of the segments in each frame of the experimental trials. The origins of these local reference frames were calculated by simple vector manipulation of relevant landmark co-ordinates. Some care however had to be taken to construct the axis systems. If an axis lay on the intersection of two planes then the normal of the planes would be defined by using the cross products of two vectors that lay in that plane. For example, the y-axis of the foot lies on the intersection of the transverse and sagittal plane. The vector ($\mathbf{n}_{\mathbf{tran}}$) that is normal to the transverse plane of the foot can be found from the cross product of the vectors $\overrightarrow{O,FM}$ and $\overrightarrow{O,VM}$ (where O is the origin which is CA in this case). Similarly for the sagittal plane, its normal ($\mathbf{n}_{\mathbf{sag}}$) was found from the cross product of $\mathbf{n}_{\mathbf{tran}}$ and $\overrightarrow{O,SM}$. Therefore, a vector in the direction of the y-axis can be defined by the cross product of $\mathbf{n}_{\mathbf{tran}}$ and $\mathbf{n}_{\mathbf{sag}}$. This vector was divided by its modulus to give the y-axis unit vector ($\hat{\mathbf{y}}$). Since the remaining unit vectors are orthogonal to this one, cross products were once again used to define the remaining axis vectors, e.g. the z-axis was the cross product of $\hat{\mathbf{y}}$ and $\mathbf{n_{tran}}$, and the x-axis the cross product of $\hat{\mathbf{y}}$ and $\hat{\mathbf{z}}$. Similar techniques were also used to construct the local frames of reference for the other segments as illustrated in Figure 7-3 and Table 7-B. It should be noted for future reference that each of these unit vectors was described in terms of the global unit vectors, e.g. $\hat{\mathbf{x}} = X_x\hat{\mathbf{X}} + Y_x\hat{\mathbf{Y}} + Z_x\hat{\mathbf{Z}}$, where $\hat{\mathbf{X}}, \hat{\mathbf{Y}}$ and $\hat{\mathbf{Z}}$ are the global unit vectors and their coefficients (in non-bold type) are their magnitudes (see Figure 7-7 for a 2D view of concept). Such concepts will form the basis of the kinematical analysis.

7.2.2.2 ANGULAR VELOCITIES & ACCELERATIONS OF SEGMENTS

Thus far, vectors may be represented in terms of the global reference frame or the local reference frames. To aid with kinetic and kinematical analyses it was useful to switch between these two systems. Therefore, the orientation of the global reference frame to the local frames was described using the rotation matrix ($\mathcal{R}$) and its translation was described by the vector **v.**

The translation vector has effectively already been derived when the origins of the local frames were located. The rotation matrices however had to be calculated by taking the cosines of the angles between the local axes and the global. For example, the cosine between the local x-axis and the global X-axis was $\frac{X_x}{|\hat{\mathbf{x}}|}$ (see Figure 7-7). Now since $|\hat{\mathbf{x}}|$ is equivalent to one as it is a normalised unit vector, the cosine becomes X_x which would be the first element in the rotation matrix. Similarly the second and third elements would be Y_x and Z_x. Each row of the matrix thus represented one of the local unit vectors as derived above [section 7.2.2.1]. The rotation matrices thus took the following form: $\Re = \begin{bmatrix} X_x & Y_x & Z_x \\ X_y & Y_y & Z_y \\ X_z & Y_z & Z_z \end{bmatrix}$.

With such matrices, it was possible to express a vector ($\mathbf{r_l}$), which was written in the local system, in terms of the global reference system to give ($\mathbf{r_g}$):

Equation 13 $\quad \boldsymbol{r_g} = \mathcal{R}\boldsymbol{r_l}$

The reverse was also possible by applying the inverse of $\Re$ to $\mathbf{r_g}$ to give $\mathbf{r_l}$.

Therefore, if a point on a segment is expressed in the local system by a vector $\mathbf{r_l}$, then because the local frame of reference moves with respect to the body, this vector will be time invariant. However, because the body is moving relative to the global frame, the same point expressed in the global coordinate system will be time variant to give $\mathbf{r_g(t)}$.

This implies that the transformation matrix, of Equation 13, must be time variant in order to maintain the equality. Therefore, the velocity of such a point on the segment can be written in terms of its angular velocity (ω), which describes the local system relative to the global system, expressed as:

Equation 14 $$\frac{d\,\mathbf{r_g}}{dt} = \boldsymbol{\omega}_g \wedge \mathbf{r}_g = \Re(\boldsymbol{\omega}_l \wedge \mathbf{r}_l).$$

$$= (\boldsymbol{\omega}_y \mathbf{r}_z - \boldsymbol{\omega}_z \mathbf{r}_y)\hat{\mathbf{x}} - (\boldsymbol{\omega}_\mathbf{x} \mathbf{r_z} - \boldsymbol{\omega}_\mathbf{z} \mathbf{r_x})\hat{\mathbf{y}} + (\boldsymbol{\omega}_\mathbf{x} \mathbf{r_y} - \boldsymbol{\omega}_\mathbf{y} \mathbf{r_x})\hat{\mathbf{z}}$$

This can be written in matrix form as:

$$\frac{d\,\mathbf{r_g}}{dt} = \Re.\begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix}\begin{Bmatrix} r_x \\ r_y \\ r_z \end{Bmatrix}$$

The matrix contains the components of the angular velocity of the segment (relative to global frame) taken about the local axes.

However, Equation 14 can also be written as

$$\frac{d\,\mathbf{r_g}}{dt} = \mathbf{r_l}\frac{d\Re}{dt} \qquad \text{since } \Re\frac{d\,\mathbf{r}_l}{dt} = 0$$

Therefore, the matrix form of Equation 14 can be rearranged to give

Equation 15 $$\mathbf{R}^{-1}\frac{d\mathbf{R}}{dt} = \begin{bmatrix} 0 & -\omega_z & \omega_y \\ \omega_z & 0 & -\omega_x \\ -\omega_y & \omega_x & 0 \end{bmatrix}$$

This means that the local angular velocities can be directly calculated using the rotation matrices. Of course, the global angular velocities can be found by transforming the local ones using $\Re$.

The angular accelerations were found by differentiation of the angular velocities.

7.2.2.3 JOINT ANGLES / GROOD ANGLES

Although it is now possible to know the instantaneous orientation and position of each body segment using the local axis systems, it is also desirable to measure the relative movement between two adjoining segments. Since there exist 3 degrees of freedom for the relative rotations between two bodies, a variety of conventions have been defined to express these relative movements. For example, the Euler angles θ, ψ, and φ represent the rotation of a body as it moves about the Cartesian-axes of a fixed body. Unfortunately, because these angles have a dependency on each other as the individual rotations are performed, the order of the rotations is critical to truly give a full description of the relative orientation of the two bodies; if the rotation sequence is changed, the body will not necessarily end up in the same orientation and position[41]. As an alternative to the Euler angles, Grood and Suntay devised a joint coordinate system, which was more clinically relevant and avoided having to specify the order of the rotations since they were independent of each other [91]. With this protocol, a fixed axis was defined within the body of both segments; a reference axis that was orthogonal to the fixed was then also created with its direction chosen to have clinical relevance; this formed a base to measure joint angles along certain anatomical planes. Finally, a floating axis was also defined which is the cross product of both fixed axes from each segment. The axes systems could thus be selected so that:

- **Flexion-extension** rotation occurred about the fixed axis of the proximal body
- **External-internal** rotation occurred about the fixed axis of the distal body
- **Abduction-adduction** occurred about the floating axis.

[41] Three ordered sets of rotations $\mathfrak{R}_\psi$, $\mathfrak{R}_\theta$ **and** $\mathfrak{R}_\varphi$ are effectively performed on the rotating body. A vector which was expressed as $\mathbf{r_1}$ in the system before transformation may thus be expressed as $\mathbf{r_2}$ in the new system so that $\mathbf{r_2} = \mathfrak{R}_\psi\mathfrak{R}_\theta\, \mathfrak{R}_\varphi\, \mathbf{r_1}$ (note that it is still the same vector in both cases). Since the product of two matrices is not commutative ($\mathfrak{R}_\psi\mathfrak{R}_\theta \neq \mathfrak{R}_\theta\mathfrak{R}_\psi$), the order of the rotations must be preserved for $\mathbf{r_2}$ to remain the same (which it must do since it has not moved in the new system).

It was then possible to define a set of angles (α, β and γ) using the axes just described where:

- **Flex-ext angle** (α) is the angle between the floating axis and the reference axis of the proximal segment
- **Abduction-adduction angle** (β) is the angle between the two fixed axes
- **External-internal angle** (γ) is the angle between the floating axis and the reference axis of the distal segment

Fortunately, the local axes systems described by Cappozzo [Table 7-B, Figure 7-3] were valid candidates for the fixed and reference axes where clinical relevance could still be maintained. Take for example the knee joint, where:

- the proximal fixed axis was taken as the z-axis of the femur
- the proximal reference axis was taken as the x-axis of the femur
- the distal fixed axis was the y-axis of the shank
- and the distal reference axis was taken as the x-axis of the shank.

Then α, β, and γ thus described joint angles measured in the sagittal, frontal, and transverse planes respectively, which are clinically pertinent.

7.2.3 CALCULATING KINETIC VARIABLES

In conjunction with gathering camera data, force plates were used to enable the determination of ground reaction forces. Using the combination of this information and the kinematic data it was possible to calculate other kinetic variables related to the segments of interest.

7.2.3.1 GROUND REACTION FORCES

The calibrated signals derived from each force plate were exported into text files using an algorithm called VAD2TXT. According to the calibration process, the components of the applied force and moments on the plates were obtained. Using this data it was possible to reconstruct the resultant ground reaction force and its point of application, described as the centre of pressure (COP).

The COP coordinates relative to the force plate coordinate system can be calculated according to

$$\mathbf{COP_x} = \frac{\mathbf{M_z} + \mathbf{F_x} a_y}{\mathbf{F_y}}$$

and similarly

$$\mathbf{COP}_z = \frac{\mathbf{M}_x + \mathbf{F}_z a_y}{\mathbf{F_y}}$$

where, **M** and **F** are the moments and forces derived from the transducers, and a_y is the vertical distance from the force plate coordinate system to the plate surface. These values were translated into the global reference system by adding the relevant component of the vector between the force plate origin and the global origin.

7.2.3.2 INTER-SEGMENTAL FORCES

As discussed previously, it is possible to calculate inter-segmental forces by iteratively applying an inverse dynamic technique.

According to Newton's second law of motion, which states that the acceleration (**a**) of an object is directly proportional the resultant force acting on it and inversely proportional to its mass (*m)*, then the resultant force = $\sum \mathbf{F} = m\mathbf{a}$.

The acceleration of a segment can already be calculated as described in section 7.2.2.2, and its mass can be measured [section 7.1.2.2.2] therefore what remains is the components of the resultant force, in particular those that are applied to the distal and proximal ends of the segment. To calculate these components for each segment requires an inverse dynamic approach since for all segments except the foot there will be two unknowns but only one equation. Fortunately, the ground reaction force was equivalent to the distal force on the foot; this left sufficient information to calculate the proximal force through Newton's second law.

To calculate the forces acting on the adjoining segments then required the application of Newton's third law, which states that "if two bodies interact, the force exerted on body 1 by body 2 is equal and opposite to the force exerted on body 2 by body 1", i.e. $\mathbf{F_{12}} = -\mathbf{F_{21}}$. Therefore assuming such an interaction between adjoining segments, the proximal force of the distal segment is equal and opposite to the distal force of the proximal segment. Therefore, $\mathbf{F_d}$**(shank) = -**$\mathbf{F_p}$**(foot)** which means there will only be one unknown for the shank segment, i.e. the proximal force which can once again be calculated using the 2nd law. This process was then repeated for the thigh segment until all of the inter-segmental forces had been established.

7.2.3.3 INTER-SEGMENTAL MOMENTS

The resultant moment of a rigid body about some axis is equal to the derivative of the angular momentum (**L**), so that $\sum \mathbf{M} = \frac{d\mathbf{L}}{dt}$. However, it can be said that the angular momentum is the cross product of the moment of inertia about this axis (I) and the angular velocity (ω) of the body, i.e. **L**=I∧ω. The derivative of this will once again give the resultant moment, which has been expanded using Lagragian multipliers.

Equation 16 $$\sum \mathbf{M} = \frac{d\mathbf{L}}{dt} = \frac{d(I \wedge \boldsymbol{\omega})}{dt} = I \wedge \frac{d\boldsymbol{\omega}}{dt} + \frac{dI}{dt} \wedge \boldsymbol{\omega} = I \wedge \boldsymbol{\alpha} + \boldsymbol{\omega}(I\boldsymbol{\omega})$$

Where α is the angular acceleration of the segment given in the local axis system.

The components of the resultant moment on a segment are derived from the proximal and distal moments experienced at the joints as well as the moments caused by the proximal and distal inter-segmental forces, as calculated above. Therefore, Equation 16 becomes

Equation 17 $$I \wedge \boldsymbol{\alpha} + \boldsymbol{\omega}(I\boldsymbol{\omega}) = \mathbf{M_d} + \mathbf{M_p} + \mathbf{h_d} \wedge \mathbf{F_d} + \mathbf{h}_p \wedge \mathbf{F}_p$$

Where, **h** represents the moment arm caused by the proximal and distal forces on the centre [section 7.1.2.2.2.3].

Experimentally, the angular velocities, accelerations, and moments of inertia were measured about the centre of mass. Therefore, the only unknowns within this equation were the moments at the distal and proximal ends of the segment ($\mathbf{M_d}$ and $\mathbf{M_p}$),. However, using the force plate data, and assuming that no slippage occurs between the foot and force plates, it was possible to measure $\mathbf{M_d}$ for the foot; hence, the equation was solvable for the foot. Inverse dynamics was applied to calculate the moments for the remaining segments where, for example, the proximal moment of the shank was taken as the negative of the distal moment at the foot leaving only the distal moment as the unknown for the shank segment. Similarly, the moments for the thigh were calculated. These inter-segment moments often referred to as internal moments, were due to the effects of muscle and tissue.

7.2.3.4 INTER-SEGMENTAL POWERS

For this project, of greatest concern was the resultant effect of muscular control on the knee. Therefore, because muscles act to produce moments about a joint and thereby angular rotation, only the angular component of the inter-segmental power was calculated. The inter-segmental powers were calculated according to the dot product of the inter-segment moment (due to muscles) and the joint angular velocity. Note that it is important that these variables are all measured in the same axis system, therefore all the calculations are made in the global coordinate system where the joint angular velocity is taken to be the difference between the angular velocities of the distal and proximal segments, i.e.

Equation 18 $$P_{joint} = \mathbf{M}_{joint} \cdot (\boldsymbol{\omega}_d - \boldsymbol{\omega}_p)$$

7.2.3.5 TIMING PARAMETERS (STRIDE LENGTH / CADENCE)

Although a timing mechanism was used to assess the speed of the subject, for comparative reasons, it was also desired to measure speed using the trajectory data [section 7.1.2.1.5]. In addition, it was also useful to calculate cadence so that data may be represented in a different format between subjects (i.e. cadence is a speed normalisation with respect to stride length).

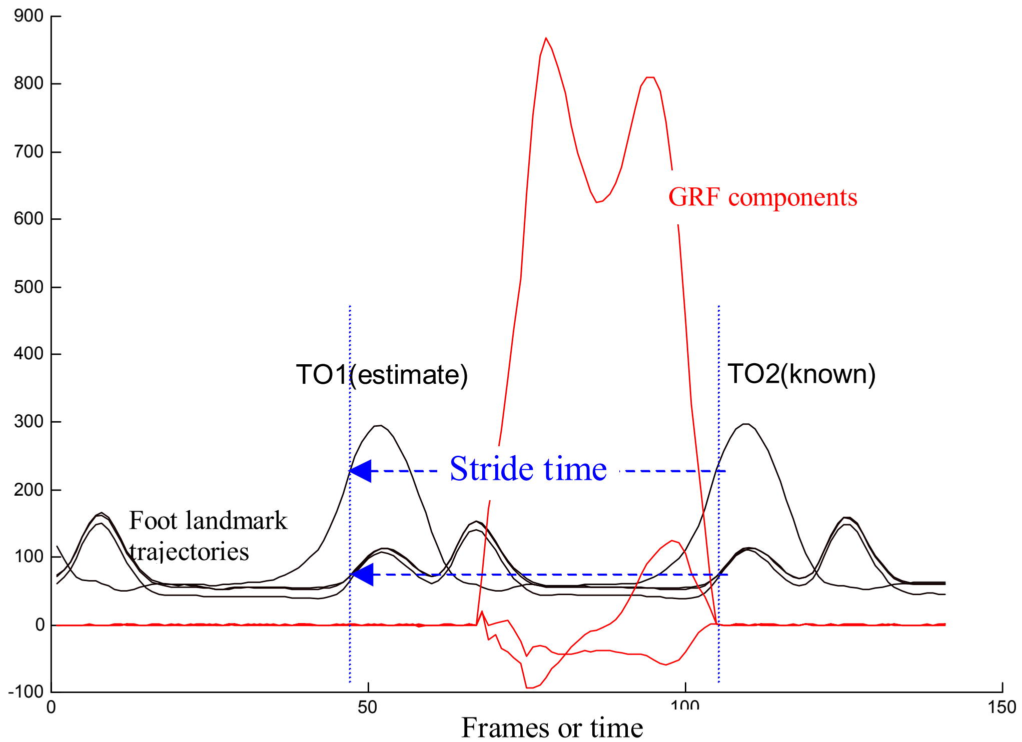


Figure 7-8: Estimation of missing timing event using foot trajectories in conjunction with force plate data. In this example, the first TO event is missed by the force plates.

The stride length was therefore taken as the distance covered between harmonic *ipsi-lateral* timing events, e.g. from toe off (TO) to toe off, or heel strike (HS) to heel strike. These timing events were established by primarily using the ground reaction force as an indication of when either heel strike occurred or toe off occurred, i.e. at the point of force application or removal from the relevant force plate. Unfortunately, because it was only possible for each leg to strike one force plate during the trial, the timing events (HS, and TO) of only one step could be directly identified from the ground reaction forces. Therefore, to estimate either TO from the previous step or HS for the next step, the trajectories of the foot and shank landmarks were superimposed onto the ground reaction force data. The timing of the event was then evident by eye from the cyclical trajectory patterns of the known event [Figure 7-8]. The temporal period between two adjacent equivalent events is thus the stride time.

Since Cadence = speed/stride length and the speed over this interval is stride length/stride time; Cadence = 1/stride time, a predictable result as frequency is the reciprocal of the period.

In practice, to measure the walking speed, the most proximal markers possible were used to calculate the stride length because they were felt to exhibit less localised variation during the timing events; these were the hip joint trajectories of both sides. The average of all possible measurements were thus made, i.e. stride times according to both TO-TO and HS-HS, as well as stride lengths measured for the sound and prosthetic side over both time intervals.

8 RESULTS / DISCUSSION

8.1 INTRODUCTION

Data has been gathered according to the philosophies described throughout the thesis. An examination of whether such information will be sufficient to control the prosthesis as prescribed in the aims of section 4.2.1, has been made.

8.1.1 SUBJECT EVALUATION

Preliminary testing of the equipment and experimental procedures were performed on non-pathological subjects. Subsequently, Bill Spence, the departmental prosthetist, was able to recruit 7 trans-tibial amputees who were thought to be suitable candidates for the trials, however due to subsequent illness with one subject, and another suffering stump problems, it was not possible to gather as much data as desired from some of them. In addition, some trans-femoral amputees were also asked to take part in the trials; of note was one subject who wished to try out the Ottobock C-leg. Although the work with the trans-femoral subjects was not used to provide data for the control system, it had been very beneficial to evaluate the other prosthetic systems and contributed to the thinking and development of the author's system as described in prior chapters.

After processing and analysis of the gathered data, approximately 750 trials were thought to be suitable for training of the neural networks, whilst the data from 250 were left for testing purposes. The majority of useful trials were predominantly obtained from five subjects, some of who were able to come in on four separate occasions to provide information on intra-subject variability. The data from the remaining two subjects was less reliable because it was considered too noisy due to problems with the Vicon system inconsistently capturing markers, and they were unfortunately unable to return for additional sessions.

A summary is now made of the trans-tibial subjects and the extent of the data they were able to provide. This includes assessing somatotypes and walking styles to try to understand how the generalisation capability of the neural networks may be influenced.

8.1.1.1 SUBJECT DETAILS

As explained during various sections [6.2.3, 6.2.3.3, 7.1.1.2] the gathered data would be used for both intent recognition and for reference profile purposes. Since the intent data should span the variability that is likely to be experienced from the end user and ideally (although not entirely necessary) should be comprehensive, it was realised that the reference profile would just be a subset of this data. Therefore, an overview is first made of the variability of the subjects. The following table [Table 8-A] provides details on the amputees who were thought suitable for this project:

	GENERAL			Morphology / (body build and shape)						
				Stature				SOMATYPE: AVE.=(4,4,4)		
				Mass	Height	Body mass Index.	Ponderal Index	Endo-morph	Ecto-morph	Meso-morph
Subject	Gender	Age	Perceived fitness	(kg)	(m)	kg./m^2	(stature)	Fat	Lean	Muscular
B	Male	75	4	85	1.80	26.2	435367	4	4	3
C	Male	29	6	55.3	1.73	18.5	237384	1	5	2
E	Male	39	8	73.5	1.78	23.2	333440	4	4	4
F	Male	35	7	87	1.77	27.8	458464	5	2	6
H	Male	48	3	84.7	1.82	25.7	429485	6	3	2
G	Male	60	4	84.6	1.73	28.3	449790			
I	Male	29	2	90	1.82	27.2	469129			
Average		45	5.6	77.1	1.78	24.3	382941	4.0	3.6	3.4
Std		18	2.1	13.3	0.03	3.6	90336	1.87	1.14	1.67
Reasonable range		17-90	1-10	50-110	1.5-1.9	13.8–48.9	186000 – 769000	1-7	1-7	1-7
Std (% ideal interval)		85	68	75	28	41	54	87	53	77

Table 8-A Details of the Trans-tibial amputees used during the trials. Ponderal index = 1000×(mass$^{3/2}$/height). The statistics that are shown were only inclusive of the subjects that provided usable data for the neural networks (i.e. B, C, E, F, and H.) Subjects in red were not used for statistical summary.

An attempt has been made by the author to measure the diversity that is within this sample population. It was wished to assess this diversity with respect to an idealised population for training of neural networks. The argument was as follows. Since in a normal distribution, 95% of the observations should lie within two standard deviations (std.) of the mean, this implies that 95% of the gathered data will extend 4 stds'. Therefore, it was of interest to know how the deviation in the sample compares to an idealistic population range. The idealistic interval was based on estimates of what a "reasonable range" should be, e.g. it was considered possible to find amputees with a mature gait, and adequate walking capabilities in the age range 17 – 90 years. Two options were available to derive the ideal interval. The simplest was that the reasonable range covers 95% of the applicable population so that the idealised extent is simply the difference, i.e. 73. The second is to argue that neural networks, for training purposes, should ideally be presented by a set of homogeneously spaced data that spans the reasonable range (i.e. examples are presented at discrete intervals throughout the range). The ideal interval would then be equivalent to four times the standard deviation of such a homogeneous sample. It should be noted that the second method yields a slightly larger idealised extent than the first; in this example, the second method gives 86 years[42] as opposed to 73. Whichever method is used, an indication of the diversity of the sample may thus be gained by taking the percentage of (sample interval/idealised interval). Such values have been calculated and presented in the grey coloured row of Table 8-A where the idealised interval assumes a homogeneous population; a value of one hundred % would imply that the sample population exhibits the same diversity as the idealised population. Although seven subjects provided data, the values in this row were based on the five amputees from whom useable data was obtained (i.e. not ruined by noisy artefacts due to problems with the Vicon data acquisition).

Such analysis shows that the mass of sample population exhibited a much larger relative diversity than the height. The height diversity had by far the lowest value and its effects were consequently reflected in the two measures of stature. The body mass index has a greater height dependency than the Ponderal-index with its emphasis on mass; hence, the body mass index gave a lower result. The distribution of height, in the sample population, may have been improved upon by the inclusion of a few more amputees with extreme builds i.e. short or tall since this population seems to cluster about the mid point of the reasonable range.

The distribution of the somatotypes (see section 2.2.2) was quite reasonable with the lowest value above 50% for the ectomorph composition and the diversities for the other types both high at over 75%. The ectomorph diversity was lower because of a concentration of the population around a value of four (it seemed difficult to enlist

[42] As a note of caution, when calculating the standard deviation of the homogeneous sample, the actual value will depend upon the spacing between samples. For example, a population that includes every year between 17 and 90 has a standard deviation of 21.52, whereas this will increase as the spacing increases (e.g. 2 years gives std. of 21.65 and 5 years gives 22.36). One can however estimate where the std. levels off as the spacing approaches zero.

the extremes of this type). This phenomenon may also be a reflection/consequence of the lesser diversity of height already described for this population.

The diversity in age and fitness was also thought to be good at 85 and 68% respectively. For practical reasons amputees with very low levels of fitness were not chosen which would automatically reduce the diversity, this should possibly have been reflected in the reasonable range, e.g. subjects below a level of three should not have been included.

In summary, although the sample population was small, it was possible to show that it incorporated a good diversity of subject parameters including mass, morphology, fitness and age. The main point of concern was the lack of height diversity, however the overall general diversity should yield a good variability in motor patterns with which to allow preliminary testing and training of neural networks.

It was thought that, if necessary, the variability of the overall data could further be increased in other ways, such as manipulating the biomechanical properties used in calculations (e.g. moments of inertia or joint centre positions), using gain or adding random noise to some data.

The requirements for selecting/modelling reference output profiles however are somewhat different [section 6.2.4]. It was recognised that if such patterns were to be dependent on a plethora of possible subjective variables, many more subjects would be required and beyond the scope of this project. Alternative strategies to address such an issue have however been discussed in section 6.2.4 such as the use of a gain switch to allow an element of tuning of a standard profile as required by each individual amputee. The standard profile could then be derived from one or two subjects whom are deemed to have the best walking gaits e.g. as judged by the prosthetist.

8.1.1.2 IMPRESSION OF SUBJECTS AND WALKING STYLES

Several criteria were set in order to try to define the task that the controller would attempt to imitate [section 4.2]. In particular, walking should be as autonomic as possible and the full scope of this task, as may be classified by some intangible attractor state, should be represented by the gathered data. Therefore, it seemed quite reasonable to give a general impression of the walking styles that were observed, given that such impressions are created using our very own highly evolved pattern recognition system.

Bill Spence, the departmental prosthetist, considered that all the subjects had quite reasonable walking styles for trans-tibial amputees; they were able to ambulate at satisfactory speeds although differences in ability were apparent. For example, although the oldest amputee seemed to be quite fit and agile for his age and even professed to ballroom dancing in his spare time, he was prone to tire quicker than the other amputees were. He required rest periods before commencing the later trials. Two other amputees had quite high levels of fitness due to their reasonably active jobs, one as a builder and the other as a postman. The remaining amputees, with the exception of one, seemed adequately fit and were able to effectively perform the task of walking for all the trials during the session.

It might have been desirable to include a few less fit amputees to extend the range of patterns, however this should not be pushed to extremes as they may start to produce far less autonomic patterns because of discomfort during walking; additionally Aim 4 states that this controller is intended for most reasonably fit amputees. Therefore, it was concluded that the selection of amputees, although small, would be sufficient to cover a reasonable spread of abilities, especially for demonstration purposes.

Regarding autonomicity of gait, the majority of amputees, once they were used to the instructions, appeared to walk very naturally. After a while, they took the task as routine and did not think about what they were doing to the extent of boredom. This was ideal in terms of gathering walking at a subconscious level. For one subject however, it was clear that during later trials he had become aware of aiming for the force plates. The amputee was discouraged from making such adjustments by looking straight ahead and being told that the force plates were not in use and that he should ignore them. After a couple of trials, this appeared to work.

8.1.2 DISTRIBUTION OF THE CONTROL VARIABLE (SPEED/CADENCE)

As explained in section 7.1.1.2, in order to collect the full range of data representing autonomic level walking, the subjects were asked to walk at self-determined speeds according to five speed instructions given by the author. The term session was used to describe each separate occasion that a subject came in and provided data consisting of several walking trials for each speed instruction (giving about 50 trials per session when possible). Table 8-B lists sessions in which useable data was gathered for each subject.

Subject	B	C	E	F	H
Sessions	Ba, Bb, Bd	Ca, Cb	Ea, Ed, Ee	Fb, Fc, Fe	Ha, Hc

Table 8-B: List of sessions in which useable data was gathered. A session consists of many walking trials spanning the subjects full walking speed range.

Speeds were monitored, first using a stopwatch with an IR beam triggering mechanism and secondly from the duration and distance travelled of the hip joint centres over two consecutive periodic events [section 7.2.3.5]. The second method uses an average of all the relevant hip joint trajectories gathered from the Vicon data.

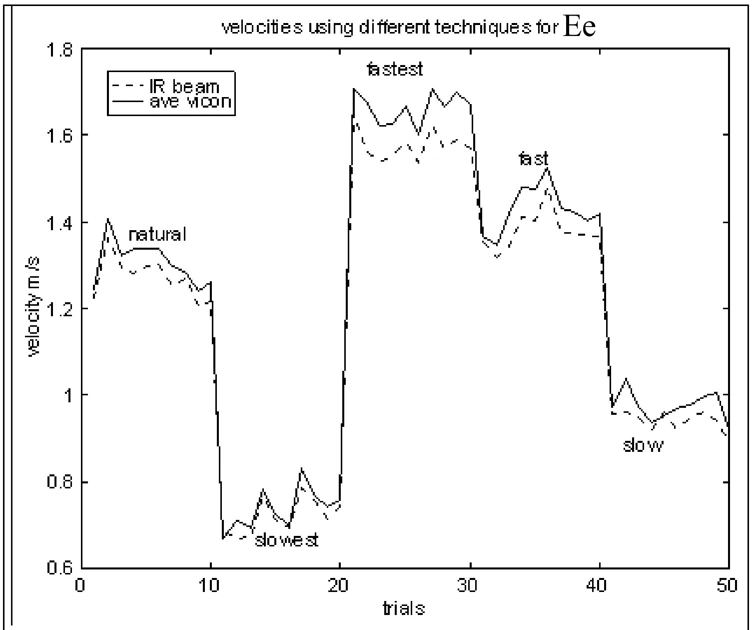


Figure 8-1: Walking speeds of trials taken during a session. The dashed line is the speeds as measured using the IR beam timing mechanism, and the solid line is the average speed of several calculations involving Vicon trajectory data [section 7.2.3.5].

An example of the speeds during one session, as calculated by both of these methods, is given in Figure 8-1; the trials are presented in the chronological order that they were gathered, i.e. group-by-group. It can be seen that the IR beam mechanism recorded slightly lower speeds than that calculated using the Vicon data; this was also generally observed for other sessions involving both this subject and others. A statistical summary of each session is presented in Table E-A of the appendix; this confirms such observation since only two sessions had a mean IR beam speed that was higher than the Vicon speed, and the mean beam speed was 0.02m/s less than the mean Vicon speed on average. This difference, although small, therefore appeared to be systematic and was likely to be a genuine phenomenon and not noise. The data in the table is graphically shown in Figure 8-2, where it can additionally be seen that although the beam speed is less in most sessions than the Vicon speed, the difference was negligible for sessions Fb, Fc, and Fe (i.e. all from the same subject F). This may be indicative of a subjective element to the speed differences; a purely systematic cause would have been evident in all cases and a random one would likely have influenced at least one session from this subject. A possible explanation was thought to be that subjects (with the exception of F), despite discouragements, slowed down before crossing the stop beam towards the end of runway. Therefore, the velocity averaged over this whole distance may be less than the velocity measured over one stride length. With such a possibility, it was considered better to use the speed values calculated from the Vicon data as these are directly linked to the analysis window. Furthermore, the precision of the beam method was of greater doubt because the variability of the distance between the two beams was up to about 10cm (or about 2 % of the distance).

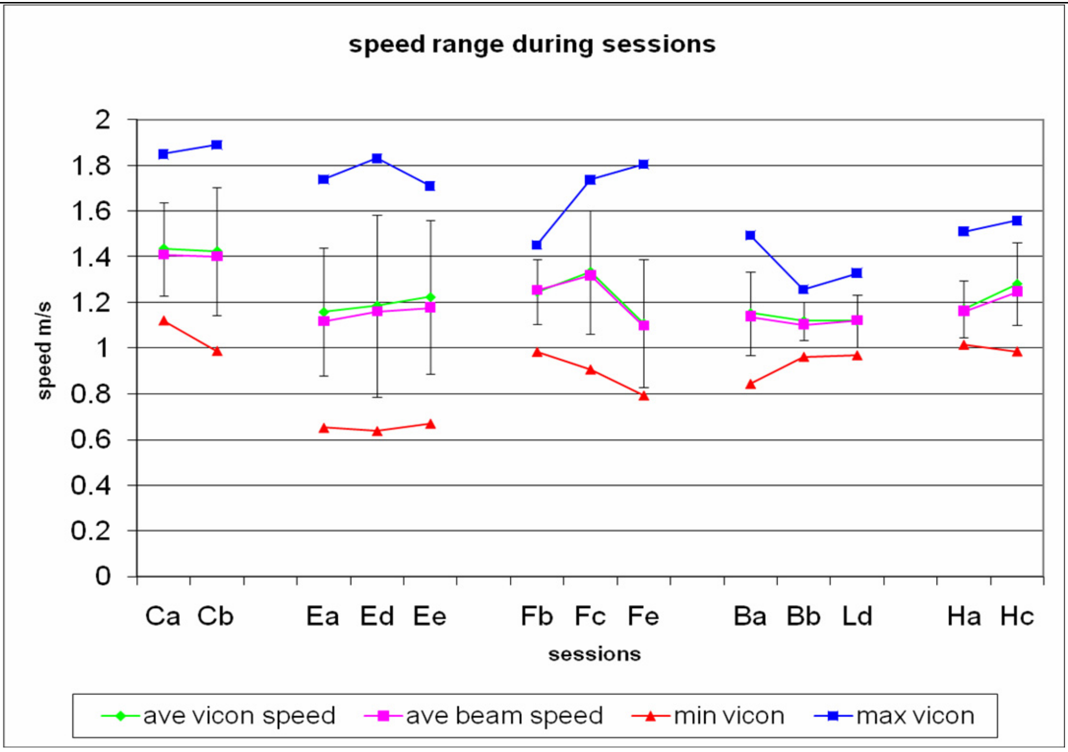


Figure 8-2; Summary of the range of walking speeds experienced throughout the sessions on different trans-tibial amputees. The mean velocity for all the trials during a session is calculated using both the IR beam mechanism and Vicon trajectories. The standard deviation of these samples is given as an error bar.

Returning to Figure 8-1, some further comment should be made regarding the intra-session speed distribution of trials. The effectiveness of the instructions is evident because - first, a good variety of speeds has been obtained where each speed group is quite distinctive - and second the distribution is evenly spread as dictated by the instructions. For example, faster than natural and slower than fastest gives a subjective way of insuring speed diversity. The first point can be seen from the average range (max speed – min speed) across all sessions, which was 0.74 m/s, or 60% of the overall mean speed. The response for each speed group was remarkably consistent, especially considering that subjects were told, “they did not have to concentrate and repeat exactly the same speed for a particular speed instruction” e.g. “natural speed”. This gave confidence that the subjects were performing the task in an autonomic fashion, i.e. at self-motivated speeds without too much thought. It seemed that once they were happy with the instructions they were inclined to repeat their performance possibly according to the “feeling” of previous trials with the same instructions. The feeling may thus set the tempo whilst the intricacies of the movement occur at the subconscious level in which basic walking patterns are utilised.

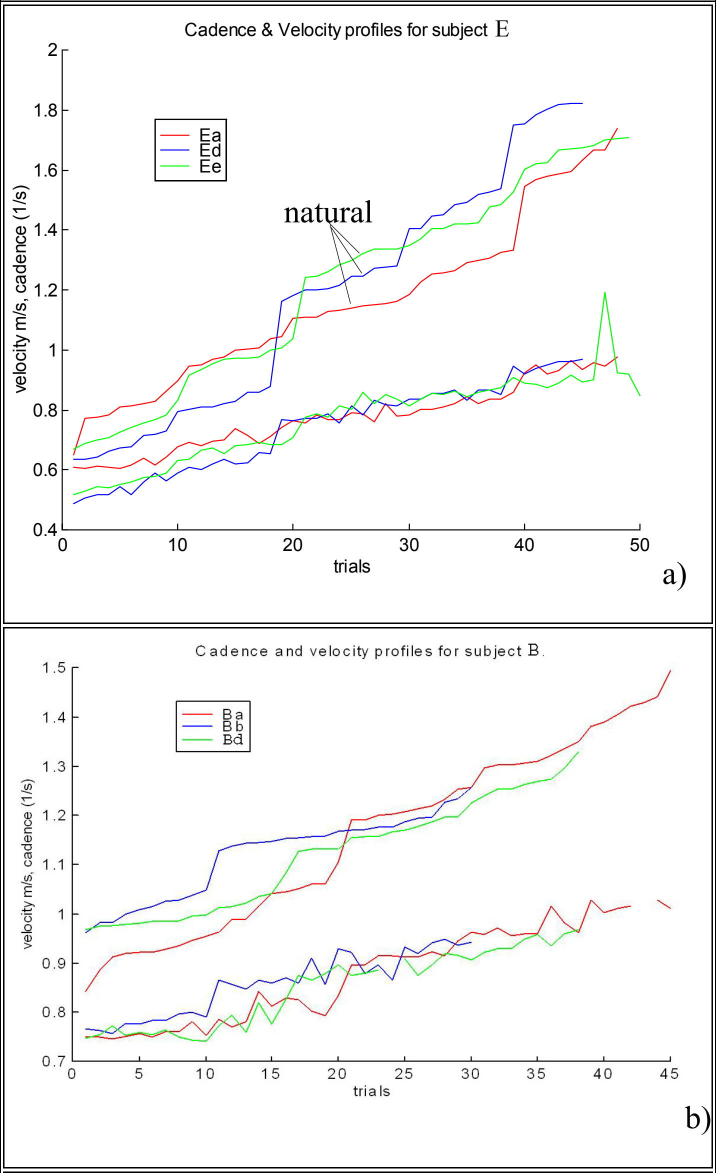


Figure 8-3 – Velocity and cadence profiles, the trials have been sorted according to velocity. Solid line = velocity (m/s), dotted = cadence (s^{-1}). Red = earliest session, Blue = subsequent session, Green = later session.

Figure 8-3 shows intra-subject plots of both speed and cadence. The trials have been ordered with respect to speed in this case. The set of plots (a) for the subject is typical for the majority of subjects in the tests. The distribution of speeds has good continuity when all session are used. The first session (red), appeared to be the smoothest session, with only small incremental jumps between speed groupings. The range was however smaller than for the other sessions but it did give a base level to build upon. A more diverse distribution was subsequently achieved in the second session by giving feedback through encouragement; this enhanced the fastest and slowest speed range but made the speed groupings far more distinctive. Interestingly, not only did the fastest and slowest speed grouping change between the different sessions but the average natural speed also changed. The interpretation of such instructions by the amputee from session to session may therefore be somewhat whimsical and depend on how the subject is feeling on a daily basis. It should be noted that the highest speed achieved by any of the amputees was 1.9m/s or 63% of the maximum theoretical value of 3m/s predicted by Alexander [2, or section 2.2.2.1.1]. This was considered quite reasonable for an amputee, although it gives scope for the inclusion of more athletic amputees who weren't available.

Plots b) are shown because they represent the smallest speed range of any subject, and were taken from the oldest subject. The speed groupings for this subject are not as distinguishable, and only three groupings were provided for the latter two sessions. Possibly, because of this lower speed range, his natural speed seemed more consistent at around 1.15-1.2 m/s. It was thought however, because of fitness issues, that this subject should provide a distinctive set of data, which can only be a good thing for the generalisation ability of the eventual ANN.

Cadence has also been included with these plots. The first observation is that the relative variation across trials is less for cadence than speed; speed appears to give a better discrimination from trial to trial than cadence. However, the combination of cadence and speed may have more merit for categorising the task.

Scatter plots for the sessions have thus been made showing the combinations of velocity and cadence particular to each trial [Figure 8-4]. The first chart gives an example of the intra-subject relationship, whereas the second gives the inter-subject relationship; for emphasis, a second order best-fit line has been superimposed onto each data set. Additionally, predictions based on Dean and Grieve models have also been included (see section 2.2.2.1.1 for review).

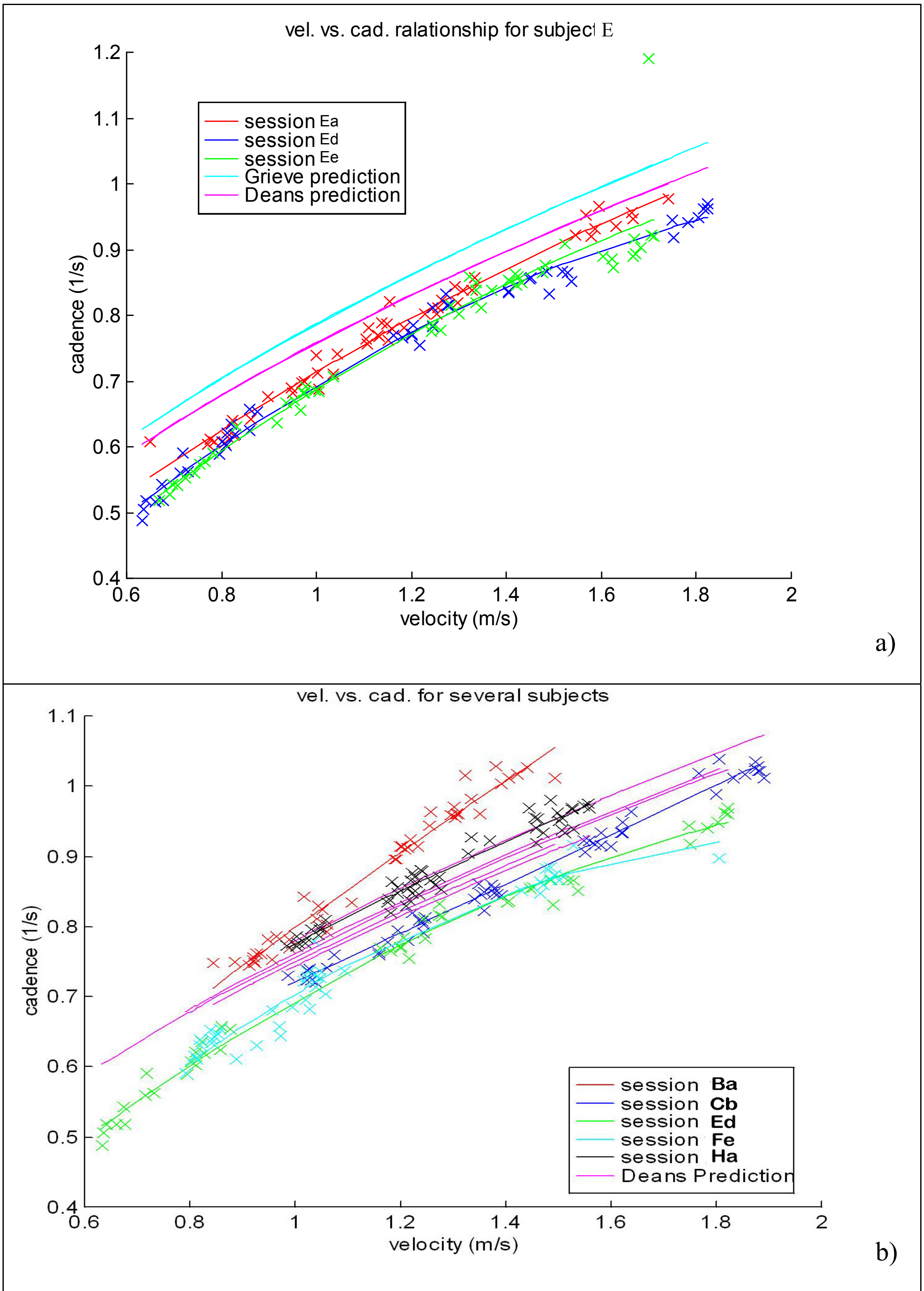


Figure 8-4; Plots showing the velocity/ speed combination for each trial of sessions for a) one subject, and b) several subjects. Predictions of the velocity / speed relationship are showing both Grieve's [87] and Dean's [56] model in the first figure. These predictions were developed from non-pathological subject data.

The first chart shows good repeatability of the scatter plots from session to session on the same subject, which was also observed for all the other subjects. Although both predictions utilise slighter higher cadence values for a particular speed, Dean's prediction, based on a collation of several studies, was closest to the experimental results. Grieve's subjects walked barefoot, which may explain the discrepancy, therefore for the second plots only Dean's prediction is included. The inter-subject plot (b) shows that apart from the oldest subject (red data points), the form of the predictions was closely followed. The slope of the best fit for this subject was steeper than for all the other subjects and shifted to a higher cadence range. This subject therefore appeared to rely on increasing cadence in order to walk at fast speeds rather than stride length. The majority of amputees opted for the alternative strategy, where they used a lower cadence (and hence longer stride lengths) to walk at particular speeds than the predictions on non-pathological subjects would have suggested. Only one subject's (black data points) plot agreed with the predictions. Interestingly he was less fit than the other amputees were, and thus similar to the oldest amputee where higher cadence seems to be a strategy.

These plots overall reveal a good variability in the methods/styles of walking across the sample population; there is diversity either side of the predicted models. The velocity/cadence plots shows discrepancies between predicted models, which may be a reflection of the pathology and further validates the use of trans-tibial amputees for the provision of reference data.

8.2 SYNOPSIS OF GATHERED DATA

The Vicon camera system was used to capture the trajectories of markers, from which kinematic and kinetic variables were calculated. This section therefore briefly outlines the success of this process in terms of whether the data can be considered a good reflection of the movement in terms of reliability, noise, and to an extent, accuracy[43].

[43] High accuracy, in the neural network training data, was of lesser importance since any real sensors that would be used in the eventual prosthesis would also be prone to error. Instead, a notion of the likely patterns that neural networks would encounter was more critical.

To examine the plethora of possible gait variables (kinematic and kinetic) a routine was written that allowed user-friendly access to 3D plots of the data from any session. This way, it was possible to view the speed or cadence dependence of the variable. For practical ease, the phase of the data was normalised with respect to the stride-length, i.e. toe off (TO) to TO for variables taken on the prosthetic side, and Heel strike (HS) to HS for the sound side. The algorithm also incorporated a number of additional viewing aids. For example, it was sometimes difficult to easily view the pure raw data because of noise and large error spikes, therefore routines to smooth the data as well as remove outliers were developed.

8.2.1 KINEMATIC DATA

8.2.1.1 TRAJECTORIES

The first step in processing data was to calculate the coordinates of anatomical landmarks during the calibration trials according to the average position pointed at by the wand [section 7.2.1.2.1]. It was realised, however, that spurious data was liable to arise where wand markers would disappear or their position erroneously captured for some frames during a trial. Therefore, to avoid such problems, the coordinates of the markers were visually checked and the distance between the two wand markers calculated. Thereby it was possible to select regions of data in which only reasonable results were apparent. Once all four anatomical landmarks had been calculated, their positions were plotted in three dimensions so that a visual check could be made to see all had gone well [Figure 8-5]. This avoided problems, for example, as caused by incorrect labelling of anatomical landmarks.

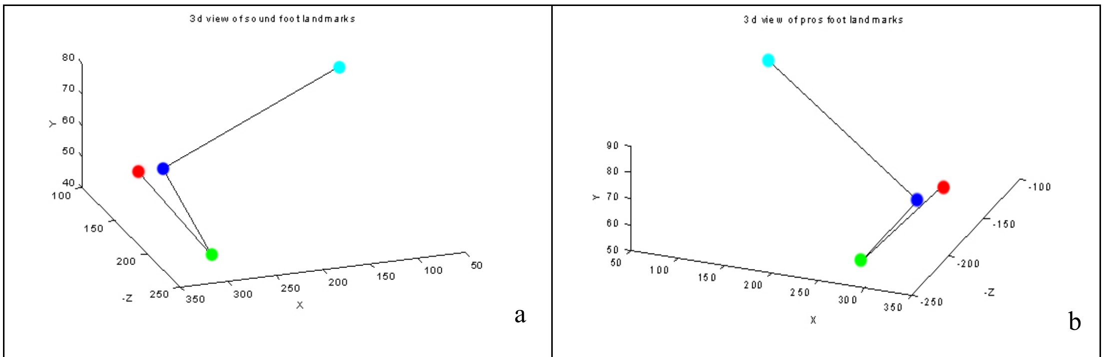


Figure 8-5 – Visual check to see if landmarks have correctly been located. The colours red, green, blue, and cyan represent landmarks FM, VM, SM, and CA respectively for the sound (a) and prosthetic (b) foot landmarks.

Once the position of the anatomical landmarks had been located within the calibration trials, they critically had to be transformed to give the local reference frames during the experimental trials. It was found that much care had to be undertaken with this process to ensure that mistakes had not occurred and that the anatomical landmark trajectories were as accurate as possible within the experimental trials. For example, if the error profiles (with respect to frames) re-occurred for each anatomical landmark within a particular segment, then it was likely that the error emanated from the trajectories of the markers in the experimental trial data as opposed to the calibration data. If a calibration trial was deemed to be in error however, then it was often possible to use the data from either the previous or next calibration trial since subjects were asked to stand as still as possible during the calibration process, they were in general quite able to oblige.

Once the trajectories of all the anatomical landmarks in the experimental trials had been calculated, it was possible to conduct further checks as to the precision and reliability of the kinematic data. A good indication was obtained by calculating the lengths of different segments throughout the experimental trials. This check was intuitively meaningful, as segmental lengths should remain relatively constant. For example, the shank length was initially measured according to the distance between the ankle and knee joint centres during the experimental trials. This was thought to be more meaningful as most other calculations depend on the location of the joint centres. This distance is shown in Figure 8-7 where three different trials were measured during a particular session.

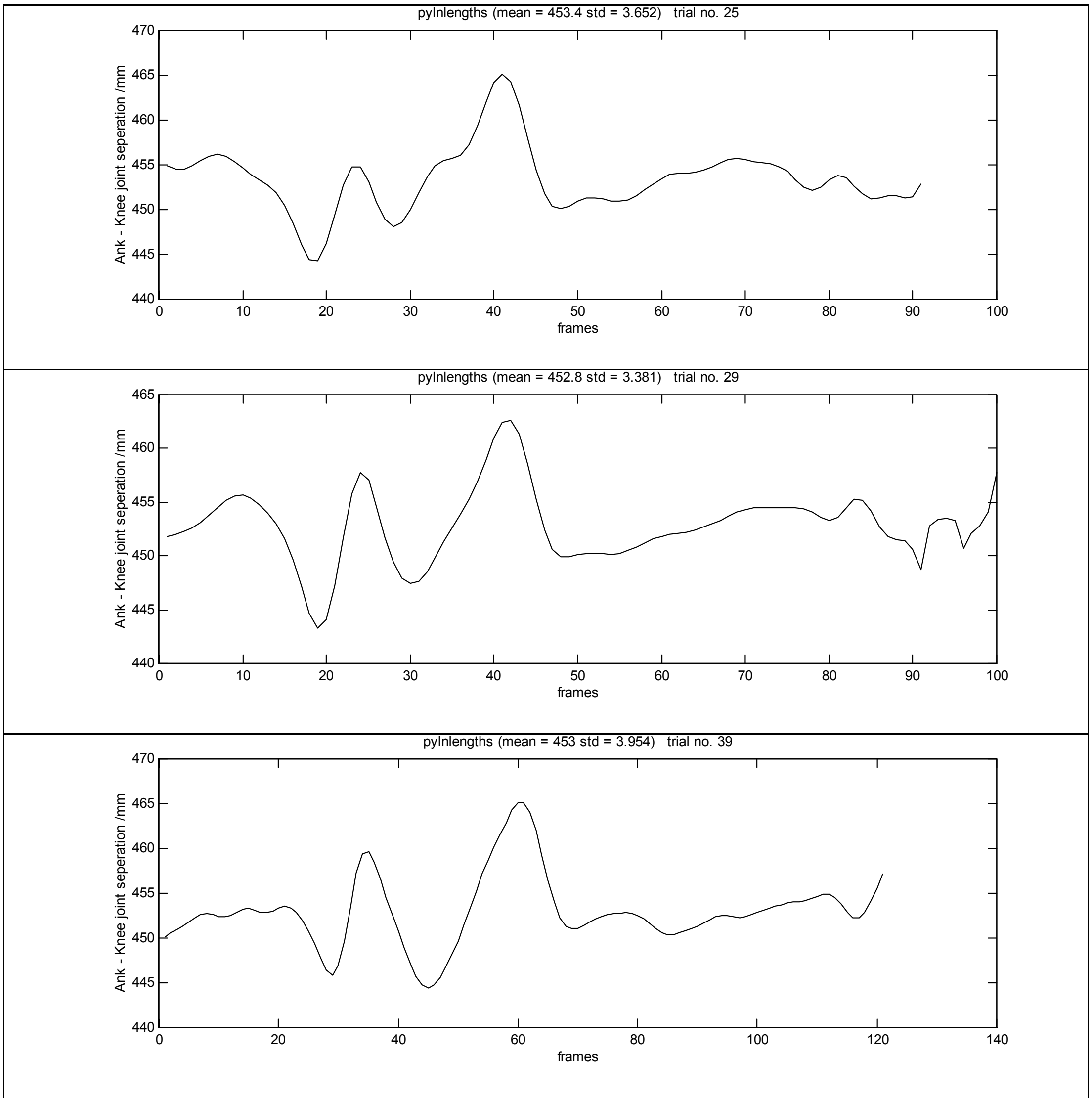


Figure 8-6 – Length of the prosthetic shank during three experimental trials as measured by the joint separation. Note how certain features are reproduced from trial to trial.

At first sight, the variation was thought to be due to noise, until comparisons with other trials revealed similar repeated features that could not be attributed to noise. In fact, the variations were simply due to the fact that the joint centres were calculated according to landmarks on two different segments (i.e. the shank for the ankle joint and the thigh for the knee joint), therefore a similar relative movement occurred between successive trials. This phenomenon however gave confidence in the reliability of the data.

To demonstrate that the variability was due to moving joint centres, the segment length was also calculated according to anatomical landmarks that were directly on the same segment [Figure 8-7]. It can now be seen that this length was very consistent throughout the trial (where the standard deviation was just 0.0002% of the mean), this also gave a basis to compare the repeatability/reliability of data from trial to trial and eventually session to session.

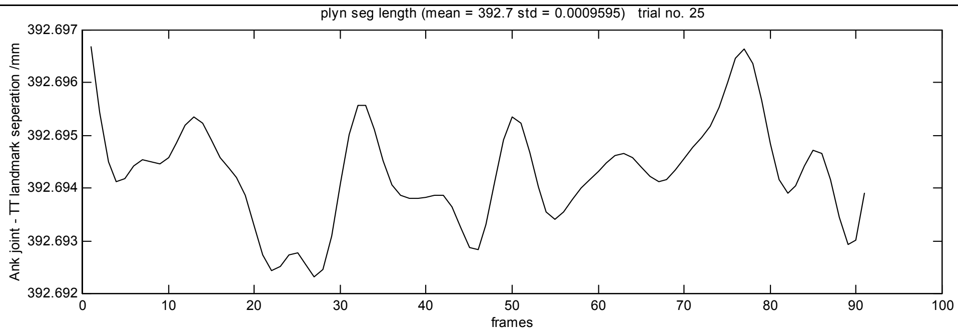


Figure 8-7 – Prosthetic shank length as measured from Ankle joint to tibial tuberosity. Amazingly, across all the exp. trials the mean of this distance remained the same, in this session, to a degree of 0.01mm.

In general, segment lengths appeared to vary by up to 2cm between sessions. This variability may be a cause for concern in some biomechanical studies; however, because of the nature of this project where it is desirable to experience as much likely variation as possible this difference is a positive benefit. Below [Table 8-C] the joint-joint distances have been given along with a prediction based on segment length to height ratios given by Contini [49]. It can be seen that there are differences on an intra-subject basis from session to session. It was thought that these could be attributed to the difficulty in locating the anatomical landmarks around the knee joint because of the prosthesis; it was also often difficult to position the hip markers so that they were readily visible and did not get knocked off, therefore some compromise was often made as to the precision of these markers.

Session	Shank (ank-knee joint distance/m)		Thigh (knee-hip joint distance /m)	
	Measured	Predicted (0.246 x height)	Measured	Predicted (0.245 x height)
Ed Ea Ea & Ed Ee	0.412 ± 0.0015 0.431 ± 0.0020 0.422 ± 0.0094 0.453 ± 0.0008	0.438	0.448 ± 0.0001 0.476 ± 0.0162 0.462 ± 0.0178 $0.419 \pm 2e^{-6}$	0.436
Fe Fc Fb	0.4751 ± 0.0012 0.4363 ± 0.0015 0.4517 ± 0.0021	0.435	0.4273±0.00001 0.4345± 0.00068 0.4447± 0.00003	0.434
Ca Cb	0.392 ± 0.0027 0.396 ± 0.0004	0.426	0.449 ± 0.0001 $0.425 \pm 2e^{-6}$	0.424
Bf Bc new foot Be Bb Ba	0.430 0.435 0.433 ± 0.0028 0.404 ± 0.0008 0.359 ± 0.0039	0.443	0.456 0.452 0.453 ± 0.0018 $0.493 \pm 4e^{-6}$ $0.521 \pm 5e^{-6}$	0.441
Ha Hc	0.460 ± 0.0013 0.455	0.446	0.451 0.438	0.445

Table 8-C – Segment lengths calculated from the joint trajectories and averaged; predictions based on Contini's work are also given. Note that sessions are grouped with respect to subjects (see Table 8-B). Session's Be, Bf, and Bc were not listed in Table 8-B because the data was too poor for gait analysis, although it was sufficient for these measurements.

8.2.1.2 JOINT ANGLES

The following summarises the joint angles that were calculated from the trans-tibial data. The angles that are shown were derived according to the Grood and Suntay method [91]. For visual purposes, data was normalised with respect to the stride length as measured from either heel-strike (HS) to HS or toe off (TO) to TO depending on whether the sound or prosthetic limb was under scrutiny. Normalisation had the advantage of enabling the data to be plotted in 3D so that the velocity axis was also visible, and ensemble averages to be made of the angular profiles within a session.

8.2.1.2.1 Ankle

To allow comparison with non-pathological subjects and verify the processed data, profiles of ankle joint angles for the sound limb were plotted. The data in this case was normalised with respect to HS, and all of the trials within a session were plotted in 3D [Figure 8-8a]. The ensemble averages and standard deviations of the profiles are shown in Figure 8-8b), and c).

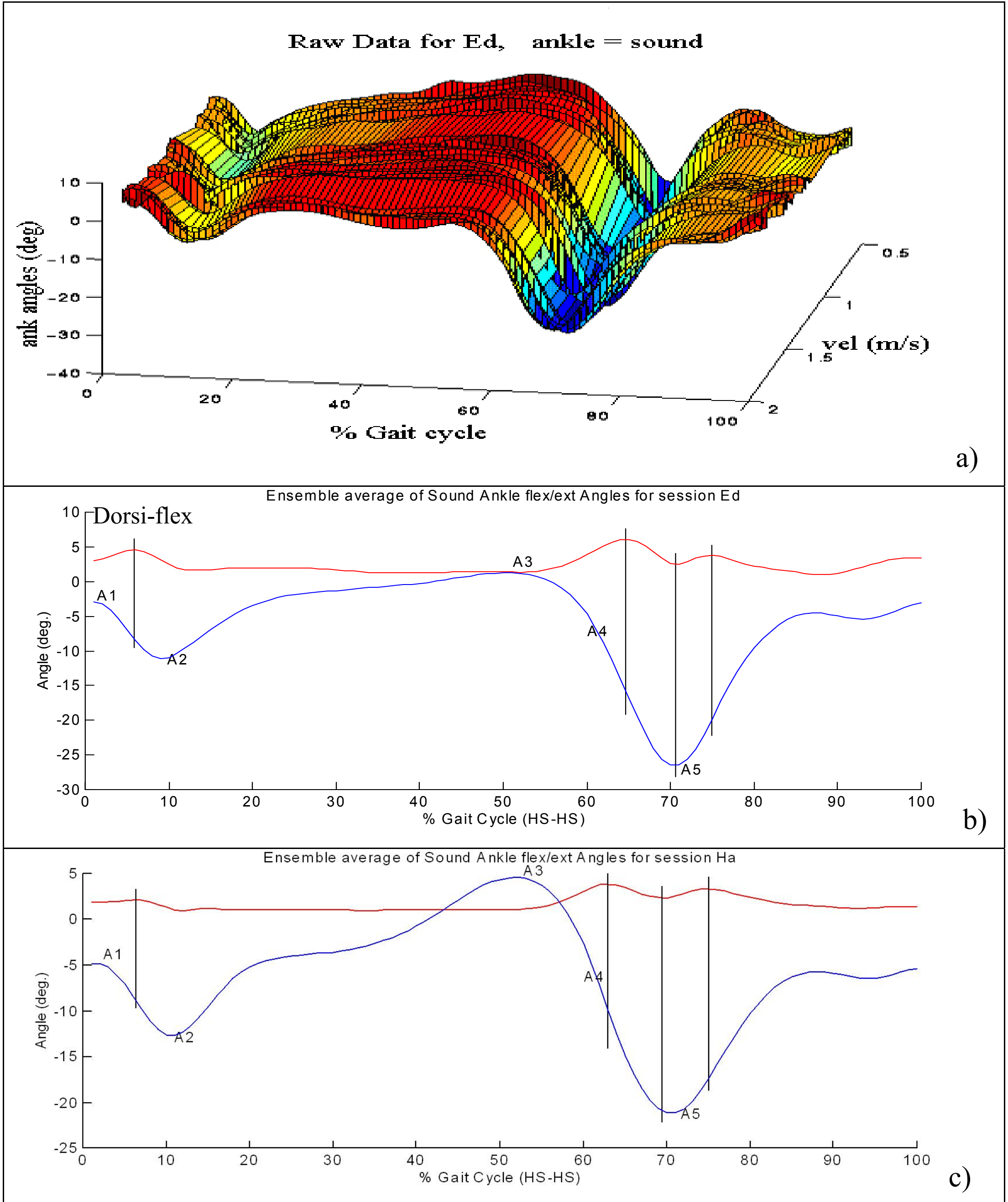


Figure 8-8; Sound ankle angles – a) Trials of session Ed plotted with respect to velocity, b) ensemble average (blue) of session Ed and ensemble standard deviation (red), c) ensemble average for another subject. The labels refer to the gait parameters suggested by Benedetti [16].

Of first note is that the general appearance of the profiles is similar to that obtained by other researchers; see for example those used by Benedetti to illustrate his convention for the gait parameters [Table A-A]. In this respect, all of the ankle-joint angle parameters are discernible and accordingly labelled in the diagrams. The A1 and A2 peaks closely resemble the values given by Benedetti [16], although there seems a slight shift of 5^0 toward the plantar flexion direction of the A3 and A5 peak; their overall excursion however is similar to Benedetti's at 35^0.

A first indication of the organisation of the data with respect to speed may be gained from the standard deviation profile. It can be seen that both profiles (red lines) depicted in diagrams b) and c) [Figure 8-8] appear to be similar despite having been collected from different subjects. There are discernible maxima and minima, which are marked by dotted lines on the figures. The maxima appear to correspond to regions of high angular change whereas the most notable minima occur around the A5 peak. One interpretation of this phenomenon, where random fluctuations may be ruled out (given the inter-subject repeatability), is that such flexion peaks may be said to occur relatively consistently as timed on the % gait cycle scale regardless of walking speed. If the peaks were to occur slightly adrift of each other this would produce large standard deviations as one maximum may be compared with a non peak of much lower value. Therefore, to maintain this consistency of peak timing irrespective of walking speed, some controlling force must counteract the increasing kinetic energy experienced with increased speed. It may however also be said that as speed is reduced, the peaks are flattened although maintaining their temporal position on this scale; this is reflected by the greatest standard deviation peaks occurring when the slope of the angular profile is at its greatest. It will however be shown later that this dependency of the amplitudes is very small when compared to the changes that occur with kinetic variables for example.

The ankle joint angles on the prosthetic side; as expected, has more discrepancies with the non-pathological subject were apparent than with the intact side; and the type of prosthetic foot had some influence on these profiles. To illustrate, the ensemble averages of data from several sessions have been calculated and superimposed upon each other [Figure 8-9].

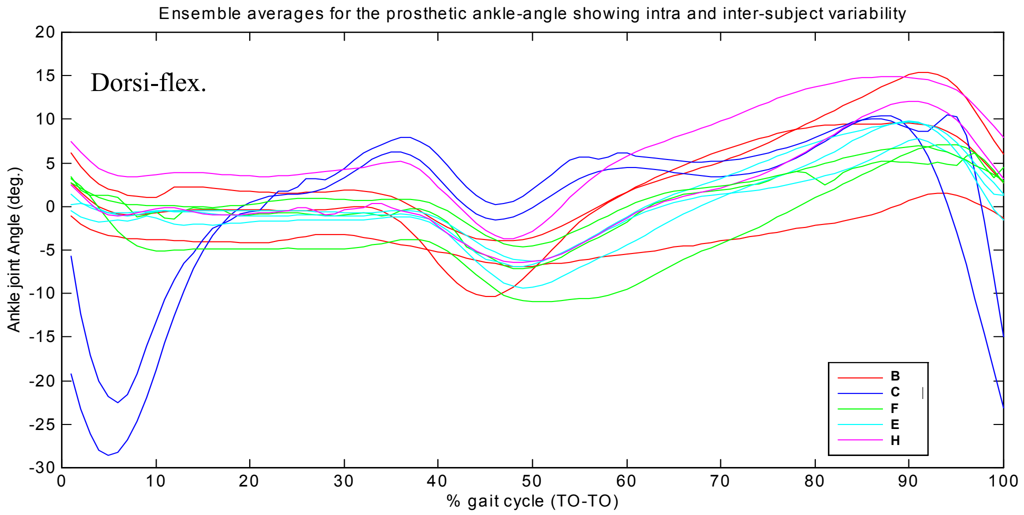


Figure 8-9; Ensemble averages of the prosthetic ankle joint angles derived from the trials taken during several sessions and from several subjects. Each colour represents the sessions from a particular subject.

At first sight, most profiles appear to be spread over a wide band; a large portion of this however may be attributed to an inexact definition of the angles. This is because the angles were deliberately not calibrated, for example by setting zero degrees to occur when standing up straight. Instead, the angles were only dependent on the definition of local reference systems through anatomical landmarks. Since the pinpointing of anatomical landmarks for prosthetic components relied on mirroring the sound limb [section 7.1.2.1.3.2.c], some discrepancy was likely and a likely source of such angular drift. However, this policy helped to increase the variability of data that may be presented to the artificial neural networks as may well be expected in a practical system. An impression of the drift can be seen for some of the subjects where the profile remains relatively flat before 40% of the gait cycle; the extent of the drift is just less than 10 degrees.

The drift effect aside, the subject C also exhibited a very different general profile to the other subjects. A significant plantar flexion angle just after TO can be observed for this subject which is not evident for the other subjects. In fact, this plantar flexion would be expected in non-pathological gait (A5 gait parameter), whereas according to the general literature on trans-tibial gait it is much reduced. The occurrence must be generated by the springiness of the prosthetic foot given that the plantar flexion occurs when no external forces are present on the foot; its unusual dynamics may possibly be attributed to the subject's relatively light mass. Whatever the cause, it is certainly of advantage to present to the neural networks.

8.2.1.2.2 Knee

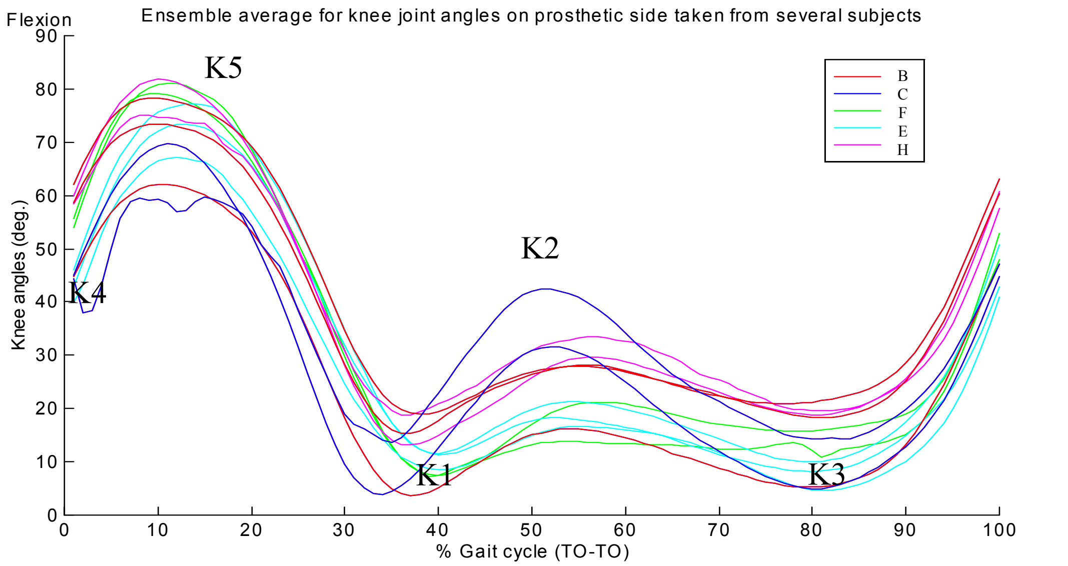


Figure 8-10; Ensemble averages of the prosthetic knee joint angles derived from the trials taken during several sessions and from several subjects. Each colour represents the sessions from a particular subject.

The ensemble averages of the knee angles for the prosthetic side are shown in Figure 8-10; several sessions from each subject are illustrated and once again colour co-ordinated. Although all of the patterns are very similar in shape, there is a considerable difference of up to 20 degrees between the highest and lowest. These differences could be a true reflection of the variability in speed and subjective parameters, but may also be caused by inaccuracies in the definition of anatomical reference systems. The patterns from the same subjects however appear to match each other much more closely with less drift, therefore the problem of definition of anatomical systems possibly relates more to its general applicability to all trans-tibial amputees, rather than random inaccuracies with palpation from session to session. Although the process of calibrating such angles may reduce such variability, as mentioned before, it would be detrimental to training the neural networks that are likely to experience such errors with sensors.

Subject/ session		Mean Excursion (K1-K5)	Mean Excursion (K1-K2)
B	Ba	58	13
	Bb	58	12
	Bd	59	9
C	Ca	56	29
	Cb	66	28
E	Ea	65	10
	Ed	63	12
	Ee	68	11
F	Fb	72	7
	Fc	74	14
	Fe	76	12
H	Ha	69	17
	Hb	59	15

Table 8-D; Excursions between gait parameters of the ensemble average prosthetic knee angle.

Subject C once again, as with the prosthetic ankle-angles, had the most notable angular profile, where the K2 peak just after heel-strike was more accentuated than for other subjects. This can be seen in Table 8-D, which summarises the excursions made by the K2 and K5 gait parameters. The K2 excursion as measured from the K1 extension peak was almost 30 degrees, which was as high as non-pathological subjects may achieve whilst walking fast. In contrast, the same peak for subject F was only just apparent during session Fb with an excursion of less than 10 degrees; this is much lower than the mean speed value of 20 degrees for non-pathological subjects, although comparable with their lowest speed results. The remaining subjects also had K2 excursions less than 20 degrees but greater than 10. This general observation agrees with the literature, which usually shows how this peak is the only marked difference between non-pathological subjects and trans-tibial amputees (see Sanderson and Martin [175] for examples.)

The K5 peak seemed more reliable for all subjects where excursions of about 60 degrees and over were generally observed; this is in line with Winter's results on non-pathological subjects [222], although slightly less than Benedetti's mean value of 67 degrees [16]. Subject B consistently had the lowest value of around 59 degrees, whereas subject F who displayed the smallest K2 excursion in contrast produced the largest K5 excursions reaching 76 degrees.

Overall, these knee angular patterns for the trans-tibial amputees are similar to non-pathological subjects and demonstrate their ability to still control the knee joint. The overall extent of variations between subjects and within subjects was pleasing with regard to neural network training, where observations described in the literature were not only captured but also exceeded upon with similar K2 peak examples to non-pathological subjects.

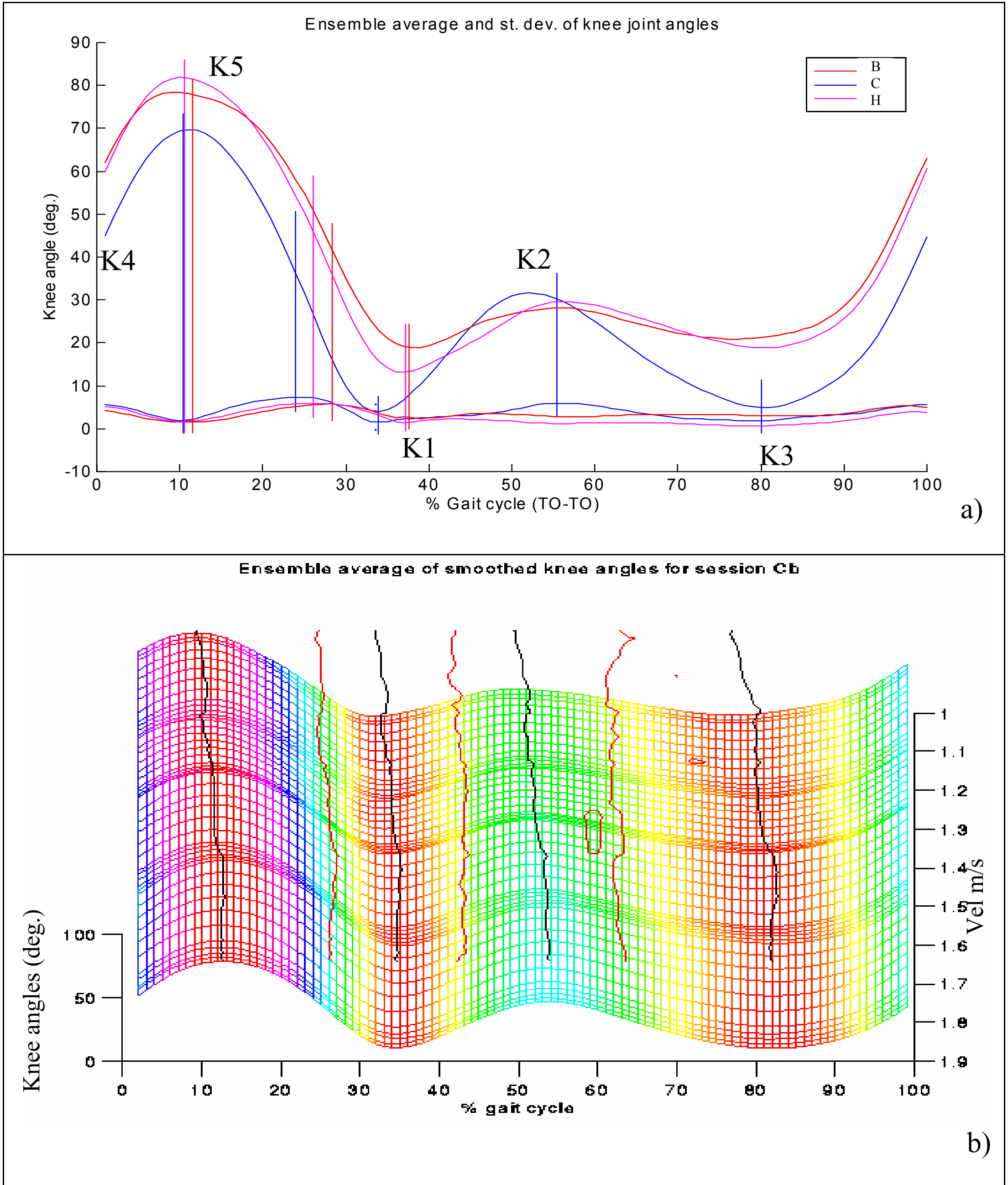


Figure 8-11; Plot a) shows the ensemble knee angle averages and standard deviations for three of the trans-tibial amputees. Diagram b) is a 3D plot viewed from an elevation of 70^0, of the knee angle profiles taken during session Cb. In addition, the zero gradient (black lines) and highest gradients (red lines) have been superimposed onto the diagram to show where the peaks lie as velocity changes.

Because of the relatively high degree of curvature of the peaks K1, K2, and K5, the standard deviation profile elicited some temporal information about the distribution of the gait parameters. In Figure 8-11a, example sessions have been selected, from different subjects, in which the data exhibited the least noise artefact, i.e. so that the standard deviation reflected changes due to walking speed. Notable peaks in the standard deviation profiles have thus been highlighted with a vertical line, which also bisects the mean profile. The standard deviation maxima happen when the mean profile experiences regions of greatest angular change whereas the relative minima in deviations coincide with peak angles (in particular K5 and K1). This was thought to be indicative of consistent timing of these peak values relative to the gait cycle. This postulate was reaffirmed by plotting the position of the zero gradient and max gradient onto a 3D plot of the data [Figure 8-11b], which shows the knee angular profiles taken during session CB as viewed from a high elevation in order to show the speed dependence. The contours are almost vertical and demonstrate the speed invariant characteristics of the ankle parameters on the normalised temporal scale.

8.2.1.2.3 Hip

Subject	Session	Max Hip angle Excursions
B	Ba	44
	Bb	39
	Bd	46
C	Ca	40
	Cb	46
E	Ea	46
	Ed	47
	Ee	47
F	Fb	43
	Fc	49
	Fe	46
H	Ha	40
	Hb	44

Table 8-E; Hip excursion angles as calculated from the hip joint angles shown in Figure 8-12. The value is taken as the difference between the minimum and maximum value.

The ensemble averages of the hip joint angles are shown in Figure 8-12 for the trans-tibial subjects. All of the profiles display a similar pattern, which is consistent with the general shape produced by non-pathological subjects. Because once again there appears to be drifting between the profiles, the maximum excursion has been calculated [Table 8-E]. The literature on non-pathological subjects shows an excursion of about 30 degrees to 40 degrees (see section 2.2.2.1 for review). These trans-tibial amputees seem to have much more active hip excursions ranging between 40 and 50 degrees, which is in agreement with Sanderson and Martin's work on trans-tibial amputees [175].

Profiles appear quite consistent on an intra-subject basis so that the drift associated with the inter-subject data may once again indicate inconsistencies with identifying anatomical landmarks between trans-tibial subjects.

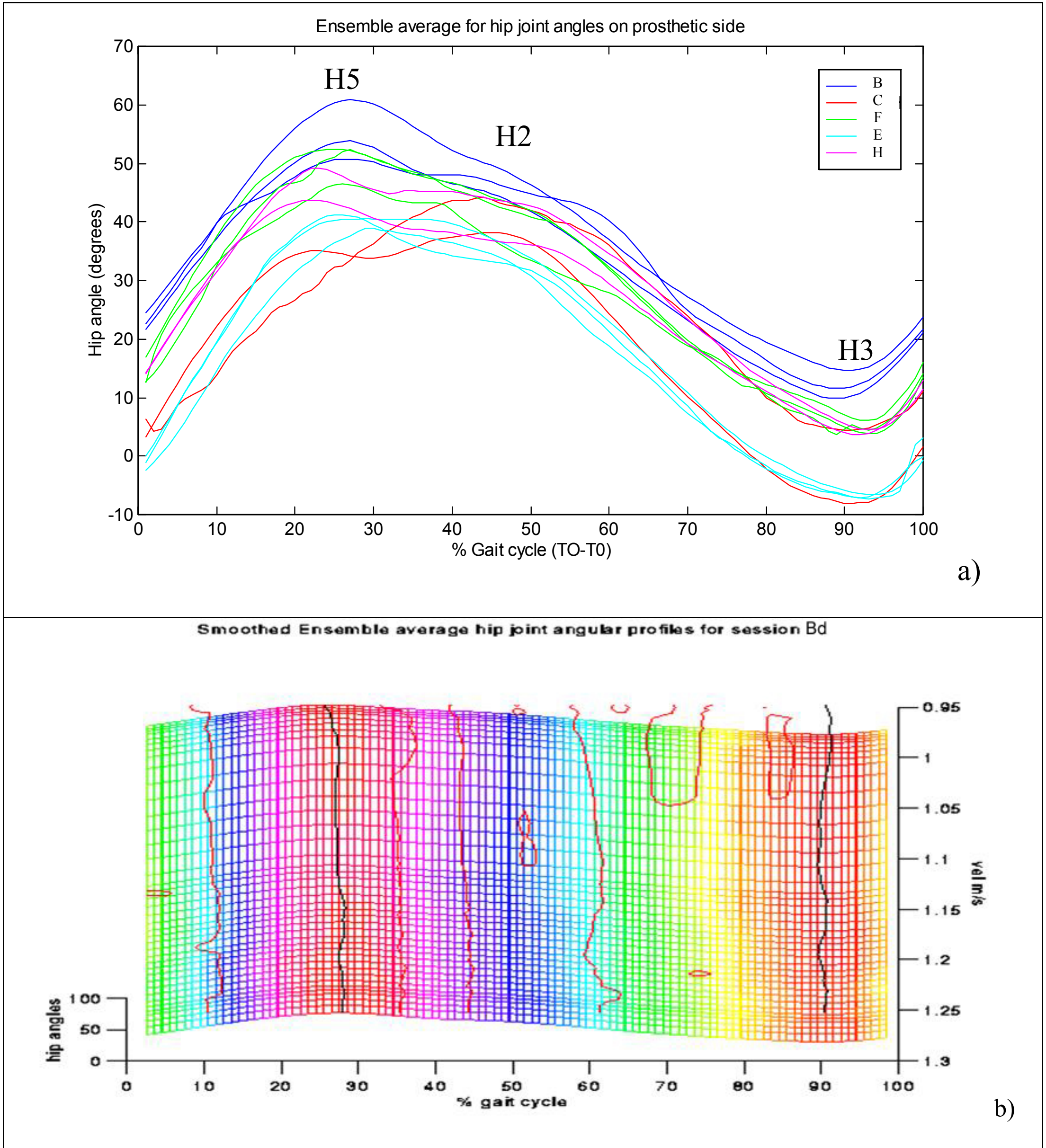


Figure 8-12; Plot a) shows the Ensemble averages of the hip joint angles taken for several sessions and from several trans-tibial amputees. The labels are according to Benedetti's convention [16]. The second diagram b) is the hip angle data from session Bd viewed from an elevation of 80 degrees with respect to the angle axis. The zero gradient contour lines have been superimposed (black lines) as well as the highest gradient contour lines (red lines).

Because of the much gentler peaks in the hip angles, the standard deviation patterns reveal much less about the temporal content of the data and merely show an overall standard deviation that would be expected between the angular profiles. However, by plotting, for each session, the contour lines of the peak values (e.g. H5 and H3) with respect to speed and gait cycle [Figure 8-12b], it could be seen that as speed varied, these lines remained fairly vertical, i.e. remained at the same percentage of gait cycle irrespective of speed.

To summarise, in general all of the joint angular profiles were distinguishable and were consistent with the characteristics of non-pathological patterns. The greatest divergence from non-pathological patterns occurred with the prosthetic ankle angles, where for most subjects, no plantar-flexion was observed just after toe off. Subject C was the exception to the rule, and could provide some plantar-flexion, in general all of his profiles were closer to the non-pathological subject, and may be a reflection that he is a congenital amputee who has used prostheses since the age of one.

In general, there was also some up and down drifting of profiles in what appeared to be a "zero" offset problem, especially between subjects. This was however thought to be beneficial for training of neural networks, and if it was too much a problem, it could always be corrected later. Ignoring the difference caused by offset, although regular features within profiles were apparent, profiles did not perfectly match each other. This gave optimism that the data had good variation on both an intra and inter-subject level.

Although an in-depth analysis of the effects of speed was not made, as it was deemed unnecessary at this time, certain observations could be made. In particular, various features of the joint angulations during the gait cycle seemed to reoccur consistently at the same point on the normalised time scale. Although this did not hold one hundred percent, it was viewed as an attempt by the central nervous system to constrain the limb movement with speed.

8.2.1.3 JOINT ANGULAR VELOCITIES

The joint angular velocities were taken as the difference between the angular velocities of adjoining segments. Since they depend on the differentials of the joint angulations, they were independent of the offsets experienced with the angles. Only the sagittal angular velocities are presented for the sake of brevity since this was the most significant component. The ensemble-averaged profiles about the ankle, knee, and hip joint for each session are presented below [Figure 8-13, Figure 8-14, and Figure 8-15].

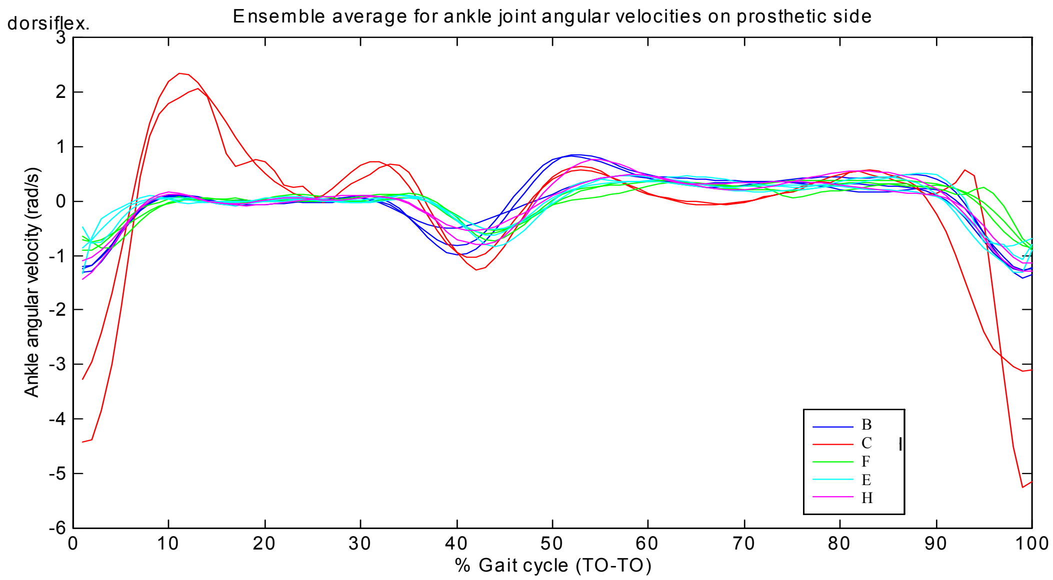


Figure 8-13: Angular velocities about the flexion/extension axis of the prosthetic ankle joint. Sessions are colour coded according to subject.

Apart from one subject (C), it can be seen that the angular velocities are remarkably consistent, even between subjects. Subject C in contrast to the other amputees produced a dorsiflexion peak just after toe off which is also evident in non-pathological gait however rare in amputee gait. It is unexpected because after toe-off there are no ground reaction forces to cause such a movement, therefore it is likely to be caused by a function of the elastic mechanism of the prosthetic foot.

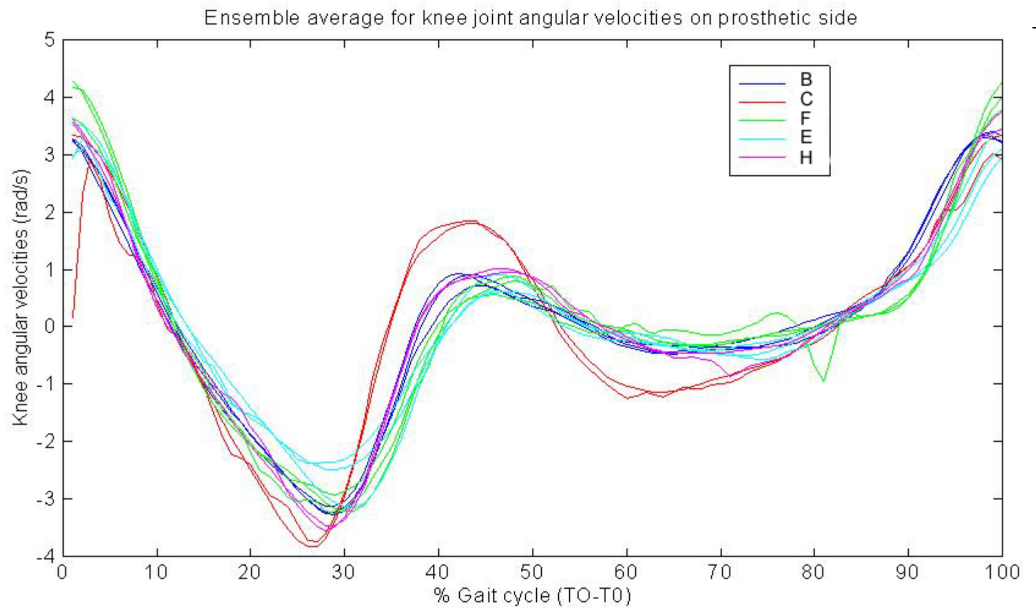


Figure 8-14: Angular velocities about the flexion/extension axis of the knee joint on the prosthetic side. Sessions are colour coded according to subject.

The angular velocities about the flex/ext axis of the knee joint are also very consistent including on an inter-subject level. Subject C once again had the most pronounced pattern in which the flexion peak just after heel-strike (37% gait cycle) was greatest (and closer to non-pathological values of up to 2.5 rads/s).

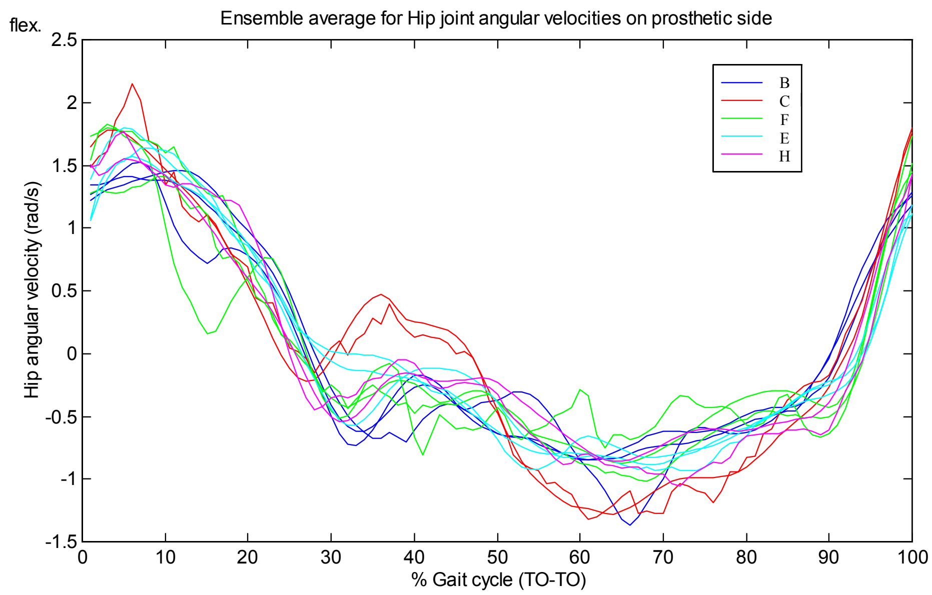


Figure 8-15: Angular velocities about the flexion/extension axis of the hip joint on the prosthetic side. Sessions are colour coded according to subject.

A general bowl shaped pattern in the hip joint angular velocities is observable between all subjects although there seems to be a much more erratic high frequency component imposed than can be observed for the other joints. It is unclear whether this is due to slight adjustments throughout gait by the subjects or some form of noise (e.g. due to an increase in unwanted thigh marker movement because of thicker underlying tissue). Once again, subject C stands out a little more in particular after 30% gait cycle where there is an unusually high flexing velocity.

8.2.2 KINETIC DATA

The kinematic data exhibited consistency, in particular with the temporal occurrence of various gait parameters on the normalised gait cycle scale. The same could not be assumed for the kinetic variables and a visual check was undertaken using three-dimensional plots that included the velocity dependency. However, for these plots to be interpretable, removal of noise and artefacts was performed by a process of smoothing the surface. Two methods were available for this: the removal of outliers, and an averaging of related points. For best results, it was desired to perform such calculations on a localised area rather than taking the statistics of the whole surface so that the temporal information and velocity dependence was preserved. Therefore, the surface was scrutinised using an averaging window that was iteratively passed over the whole surface in increments, which allowed fine overlap. The average value of this area was thus plotted in the centre of the window. The size of the window depended on the nature of the surface, e.g. a highly changeable profile would require a small gait cycle width, whereas the length of the window was determined by the number of trials taken and the extent of the speed range. In general, window dimensions of around 2-4% of gait cycle by up to 10 trials were used. Similarly, outliers of a particular window area could be determined and removed, although the window was generally made larger.

8.2.2.1 SEGMENT MASS PROPERTIES

8.2.2.1.1 Volume\Mass

As explained previously, in order to ascertain segment properties, casts were formed of the thigh (prosthetic side), and the stump of each trans-tibial subject. It was then possible to calculate the mass of the segment by multiplying the volume of the cast by the density of the segment; these results are presented in Table 8-F. In addition, for the thigh segments it was possible to make comparisons with the predictions that were based on body proportions [49]. It should be noted that an average coefficient of 20.9% was selected from several studies presented by Contini. The standard deviation of these coefficients was 2.8% of total body mass, which is given as the error in the prediction.

Segment	Volume (l)	Mass (kg) = Density*vol.	Predicted Mass (kg) Thigh =10.45 % BM Shank = 4.6% BM	Adjusted mass (kg)
E Stump	0.68	0.74	3.381	
E Thigh	4.88	5.22	7.68 ± 0.98	7.22
F Stump	0.822	0.90	4.002	
F Thigh	7.320	7.83	9.09 ± 1.21	8.15
C Stump	0.455	0.50	2.542	
C Thigh	3.62	3.87	5.77 ± 0.77	6.14
B Stump	1.32	1.45	3.91	
B Thigh	6.131	6.55	8.67 ± 1.18	8.91
H Stump	1.7	1.86	3.896	
H Thigh	6.9	7.35	8.85 ± 1.18	8.91
I Stump				
I Thigh	7.9	8.41		

Table 8-F – Masses of anatomical segments as determined from: a) casts, b) thigh predictions based on body proportions according to Contini, c) adjusted cast volume using cylindrical extension at proximal end to allow for shortfall in casting procedure. Note that the predicted shank mass applies to the intact shank, the mass of the prosthesis is not included in the stump masses hence much lower values. (A standard BK socket with pylon is assumed to weigh 1.04kg in much of the analysis that still leaves a substantially lighter shank for trans-tibial amputees).

Despite the deviation present in the thigh predictions, it could be seen that the masses calculated from the thigh casts for all subjects except subject F appeared to be lighter than the predictions. Two main possible explanations immediately came to hand, the first that trans-tibial amputees may indeed have smaller thighs than non pathological subjects, possibly due to muscle atrophy or that the thigh casts were systematically smaller than those used to produce the prediction models. In addition, because of the low number of

Subject	Cast thigh length (m)	Knee-Hip distance (m)	Predicted length (m) = 24.5% height
E	0.37	0.45	0.436
F	0.42	0.43	0.434
C	0.34	0.425	0.424
B	0.36	0.45	0.441
H	0.40	0.451	0.445
I	0.39		

Table 8-G: Length of the thigh segment as measured from: a) the thigh cast, b) knee joint centre to hip joint centre obtained from the kinematic data, c) predictions based on body proportions.

subjects it was also possible to get such statistical anomalies where the available subjects do not quite follow predictions.

The first suspicion was to check the size of the cast since it was realised that for some of the subjects, it had not been practical to make casts all the way up to the hip joint. Therefore, the lengths of the thigh casts were compared to the lengths calculated from both the Vicon data, and predictions made on body proportions [Table 8-G]. The lengths of the shank and thigh segments were thus calculated for every trial and the average of this after outliers removed used as the joint-to-joint distances. This method had an unexpected benefit in ensuring intra-subject reliability of data between trials and indeed between sessions, where discrepancies are immediately obvious. For example, during one session the length of the thigh appeared to dramatically change for later trials. The sudden cut off is indicative of a mark of relocation on the subject.

Reassuringly, the joint distances calculated from the Vicon data agreed with predictions, however there was a significant discrepancy with the casts for most subjects (except F) where they were up to 8cm short.

Therefore, since the shortfall could only come from the hip end of the cast, some allowance for this was made in the calculations by mathematically adding the missing chunk of thigh to the cast. A cylindrical shape was assumed for the missing piece with a radius equivalent to the cast radius at the proximal end and a height equal to the shortfall in segment length (i.e. cast length subtracted from knee-hip distance). Therefore, the masses of the thigh segments were recalculated using this additional volume (see the adjusted mass column [Table 8-F]). The results now appear to be closer to the predicted values and therefore it cannot be supposed that muscle atrophy at the thigh is symptomatic of all the trans-tibial subjects.

In conclusion, once adjustments had been made for the thigh casts, these results were thought to be acceptable for further calculations.

8.2.2.1.2 Moments of Inertia

Two main variables affected the calculation of moment of inertia values: the angular velocity of the object as it swung on its own, and the distance of its COG from the swing axis. As an additional check of the apparatus, the COG was derived in two ways: 1) by direct measurement where the cast was balanced on a knife-edge (h^{meas}_{cast}), and 2) by using the compound pendulum method but with an additional standard object/weight to give a simultaneous calculation and yield h^{swing}_{cast} (see Figure 7-6). For the second method, several readings of both the angular velocity of the cast on its own (ω_{cast}) and the angular velocity of the combined cast and weight ($\omega_{(cast+weight)}$) were made by measuring the respective swing periods. Therefore, it was possible to calculate a value of h^{swing}_{cast} for every combination of ω_{cast} and $\omega_{(cast+weight)}$ that had been gathered. An example is given in the table below in which a matrix has been produced of h^{swing}_{cast} from each combination of angular velocities [Table 8-H]. In this example, the values are in good agreement with the measured value, h^{meas}_{cast} of 0.068m. It should however be noticed, from the standard deviations (which are written as a percentage of the mean), that h^{swing}_{cast} appears to be far more sensitive to the changes in $\omega_{(cast+weight)}$ than ω_{cast} where the deviation is around 8% as opposed to 2.5%. This difference cannot be attributed to the deviation within the angular velocities, as they are reasonably similar with 0.5% and 0.7%. Therefore, the question remained what would be the most applicable value to use for the centre of gravity. It was decided that since the values derived from the compound method were more directly related to the swing axis than measurements using the balance technique that these values would be used, although some attempt was made to remove outliers/unlikely values. Outliers were removed by using h^{meas}_{cast} as a guide, any h^{swing}_{cast} values that differed by more than 30% of h^{meas}_{cast}, (or 0.01m which ever was greater) were eliminated. Therefore, only these good values of h^{swing}_{cast} were used to calculate a matrix showing the radius of gyration (K) for every combination of the good h^{swing}_{cast} and ω_{cast}. This meant ultimately that only those values of ω_{cast} and h^{swing}_{cast} (or in effect $\omega_{(cast+weight)}$), which produced reasonable results, had been used to assess the moment of inertia.

															mean	std %
		$\omega_{(cast+weight)}$ rad/s	6.4	6.373	6.338	6.368	6.384	6.338	6.367	6.417	6.36	6.442	6.433	6.404	6.385	0.538
	ω_{cast} rad/s															
	7.899		0.068	0.064	0.058	0.063	0.065	0.058	0.063	0.071	0.062	0.075	0.073	0.068	0.066	8.235
	7.995		0.065	0.061	0.056	0.060	0.062	0.056	0.060	0.067	0.059	0.071	0.070	0.065	0.063	8.099
	8.027		0.064	0.060	0.055	0.059	0.062	0.055	0.059	0.066	0.058	0.070	0.069	0.065	0.062	8.056
	8.02		0.064	0.060	0.055	0.059	0.062	0.055	0.059	0.067	0.058	0.070	0.069	0.065	0.062	8.065
	8.053		0.063	0.059	0.055	0.059	0.061	0.055	0.059	0.066	0.058	0.069	0.068	0.064	0.061	8.024
	8.105		0.062	0.058	0.053	0.057	0.060	0.053	0.057	0.064	0.056	0.068	0.067	0.062	0.060	7.961
	8.057		0.063	0.059	0.054	0.059	0.061	0.054	0.058	0.066	0.057	0.069	0.068	0.064	0.061	8.019
	8.019		0.064	0.060	0.055	0.059	0.062	0.055	0.059	0.067	0.058	0.070	0.069	0.065	0.062	8.067
	7.965		0.066	0.062	0.057	0.061	0.063	0.057	0.061	0.068	0.060	0.072	0.071	0.066	0.064	8.140
	7.984		0.065	0.061	0.056	0.060	0.063	0.056	0.060	0.068	0.059	0.072	0.070	0.066	0.063	8.114
mean	8.012		0.064	0.061	0.055	0.060	0.062	0.055	0.060	0.067	0.059	0.071	0.069	0.065		
Std %	0.707		2.536	2.478	2.403	2.466	2.500	2.403	2.464	2.575	2.450	2.631	2.612	2.545		

Table 8-H: Matrix with resultant h^{swing}_{cast} (in grey), as calculated from the compound pendulum method using ω_{cast} and $\omega_{(cast+weight)}$ (see section 7.1.2.2.2.3). In this example, a BK stump cast was used, where h^{meas}_{cast} was 0.068m from the balance techique. The mean and standard deviations shaded blue are calculated for values in rows, and the green for columns. This way, it was possible to see which parameter that h^{swing}_{cast} was most sensitive to.

Although the method of analysing the swing period data was thought to be sound, it was decided that, with some of the casts, the value of h^{swing}_{cast} was still far too erratic where some values would differ by more than 10cm from not only each other but also the measured value. It was concluded that much of this variation had to be attributed to the method used to measure the swing period, which was fraught with practical problems, difficult to reliably repeat and prone to error. Therefore, alternative techniques were devised with which at least comparisons could be made and hopefully accuracy and reliability improved upon.

8.2.2.1.3 New techniques to measure the compound pendulum swing period.

Because of the uncertainty in the inertial measurements using the stopwatch technique to measure the swing period, other techniques were considered. The first was the use of a small accelerometer that was to be mounted on the swing cast. This relies on the fact that for a body of mass (m) that swings about an axis a distance (h) away from the centre of gravity (COG), the resultant moment is equal to the product of the inertial moment about the axis (I) and the angular acceleration (α). However this moment is provided by the force of gravity so that $I\alpha = -mgh\sin\theta$ where θ is the angle between the vertical and the line from the pivot to the COG. Therefore the angular acceleration would be harmonic as it is proportional to sin θ and allow the swing period to be measured. Unfortunately, the only accelerometer that was readily available was too insensitive as it was designed for much greater accelerations, and its small signal was thus swamped by noise.

The next method that was tested was the use of a digital video camera to try to record the harmonic motion of the swinging object. Since the whole object moved as one, it was convenient and more accurate to focus the camera on a point on the object whose position could easily be identified with respect to an external fixed marker. To improve accuracy the feature on the object was chosen in a location that would give the most visible movement [see Figure 8-16 for example].

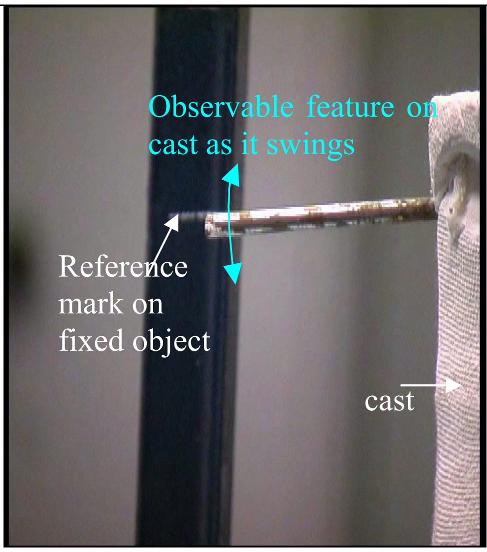


Figure 8-16 - Video Camera view of swinging cast.

Consequently, it was a simple matter of measuring the period by taking a note of two video frames in which the observed feature appeared in the same relative place. Generally, these two frames were chosen to span 10 cycles to help with accuracy. The method was shown to be consistent over repeated trials although the video sampling rate of 50Hz proved to be the main limiting factor possibly giving a false impression of the repeatability of data.

Therefore, another method was devised which was finally chosen. This utilised the Vicon camera system to record the motion of the swinging object. This provided an increased sampling rate of 120Hz and enabled further processing of the data. Since all that was required was some measure of the harmonic motion relative to the global (lab) reference frame, the distance from the lab origin to a marker on the object sufficed for this. In practice, this was simply the modulus of the co-ordinates of reflective markers placed on the object. Thereby, the Vicon system was used to collect several repeat trials of data each spanning at least ten swing cycles.

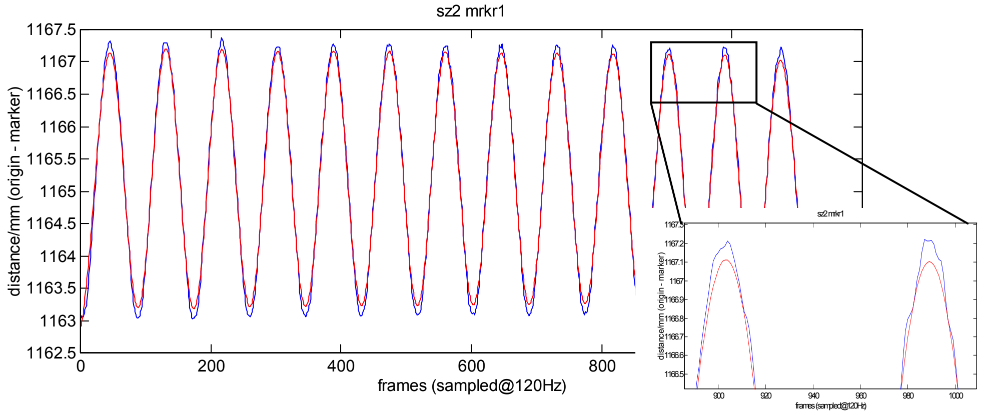


Figure 8-17 – Distance from origin of marker placed on swinging cast. Blue = raw data, red = filtered data.

Two techniques were used to calculate the period from the gathered data. The first was simply by directly measuring some interval such as peak-to-peak (usually including 10 peaks), the second involved performing a spectral analysis of the data with a Fast Fourier Transform algorithm (FFT). In the first method, it was necessary to filter out some of the high frequency noise created by the data acquisition process; otherwise, it was not easy to establish exactly where a peak might lie (see zoomed portion of Figure 8-17). A fourth order Butterworth filter with a cut off frequency of at most 3Hz was found to be sufficient and produced good results. This method was primarily used as an initial check and confirmation of the results, however it was found to be more convenient, reliable, and accurate to automate the process by utilising FFT. Matlab was used to calculate a power spectrum-density estimate, using a function called “psd.” It was speculated that the fundamental swing frequency of the cast would dominate all other frequencies and thus have the largest power magnitude, thereby making automatic detection possible. The function utilised the FFT to be able to calculate the power spectrum; therefore, in practice much care had to be taken in order to set up the parameters to give a sufficiently resolved and precise frequency. It was discovered that although most samples of data contained about 1000 data points, this was not sufficient alone to give the same accuracy as found by measuring peak to peak. Therefore, it was necessary to zero-pad the data to length 4096 (which is a power of 2 and helps computation). A Hanning window of length 512 was also used to try to reduce leakage. The spectrum of Figure 8-17 has thus been calculated and is shown [Figure 8-18]. Although the main frequency was clearly discernable, it was found to be more accurate to use a cubic spline fit of the spectrum in order to smooth and interpolate the data to a finer frequency resolution (see zoomed portion of Figure 8-18).

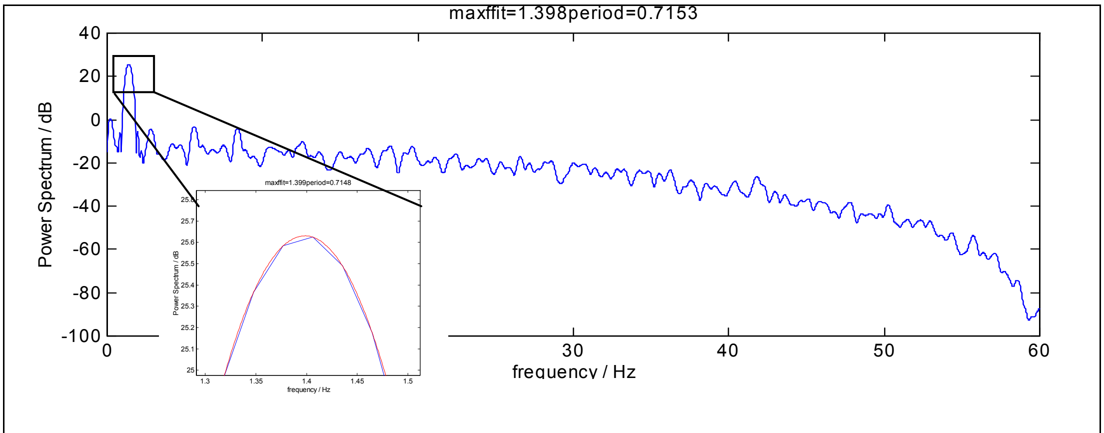
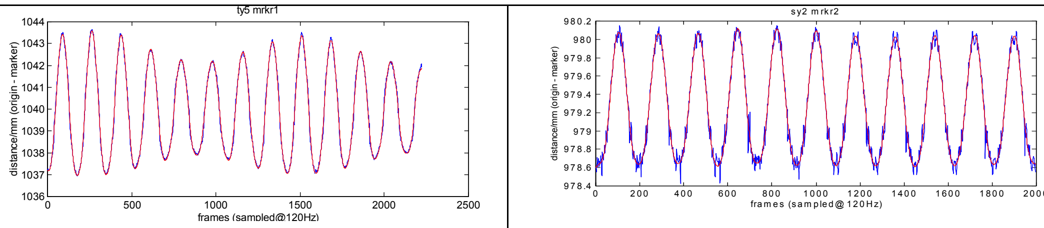


Figure 8-18 – Chart showing frequency content of the marker distance data in Figure 8-17. Blue line shows the direct result of the FFT whereas the red line is a cubic spline interpolation. Note that the fundamental frequency as highlighted in the zoomed section is much greater than the other frequencies. The spline fit enables the maximum frequency to be resolved with greater precision.

It should be noted that this method was also applicable to measure the swing period of the torsion pendulum [7.1.2.2.2.3.b]. It fortuitously also provided an unexpected benefit with such data, where it was possible to judge the quality of individual torsion swings. Great care had to be taken when initiating these rotations in order to minimise unwanted translation forces. Therefore, with the Vicon data, it was possible to observe such occurrences as they often created additional harmonics that seemed to disrupt the desired fundamental harmonic and give erroneous results. A case in point [Figure 8-19] is the data collected for the torsion harness being swung on its own (i.e. so its moment of inertia can be accounted for in subsequent measurements of objects). In this figure, the first example shows the effect of perturbation from translation forces and the second where such influences were minimal. These yielded values for the moments of inertia of 0.0062435 and 0.0063937 kgm^2 respectively, a difference of 2.3%.

The compound pendulum method did not suffer as severely from such perturbations since a string was used to pull the cast to its initial position and then released thereby minimising unwanted influences such as hand movement.



a) Perturbed motion; freq. = 0.677 Hz

b) Non-perturbed motion; freq = 0.669 Hz

Figure 8-19 – Distance from origin results for torsional pendulum harness being swung on its own. The observation in a) shows the effect of translation forces perturbing the rotational motion. It can be seen that the fundamental frequency is different to b) where no additional harmonics were induced.

In addition, the Vicon system was also used to measure the gap between the weight's COG and the cast swing axis. This was achieved by placing two markers on opposite sides of the weight, at a height that was in line with the COG (i.e. half way up the side of the cylindrical weight). Similarly, two markers were threaded at both ends of the rod that protruded from the cast. The halfway points of both of these sets of markers were then used to define the distance to the weight's COG.

8.2.2.1.4 Comparison of inertial results and validity

The distance of the COG from the swing axis is analysed below for a variety of objects since it is a critical parameter for determining the moment of inertia about the COG of an object. Table 8-I gives a comparison of the COG as measured using the balance technique and as calculated from the compound pendulum technique. It should be noted that the compound value is taken as the average of the COGs' calculated from each combination of at least three ω_{cast} measurements and at least three $\omega_{(cast+weight)}$ readings.

In addition, for the thigh segments, adjustments have had to be made in order to mathematically include the missing proximal portion, which was often unable to be cast. The same cylinder that was used to form the adjusted thigh volumes [Table 8-F] was thus incorporated to make up the shortfall. It should be noted that when calculating the adjusted COG, both the masses of thigh and cylinder had to be converted into equivalent forms. An anatomical composition was used rather than plaster for both the cast and the added cylinder. Predicted COGs' for the thigh segments have also been given. These are based on the location of the centre of mass from the proximal joint being 42.7% of the thigh length [48], and the thigh length as predicted in Table 8-G. In general, it can be seen that the calculated COG values came within reasonable agreement of the measured ones, often differing by less than 1cm. Considering the difficulty in making such precise measurements this was thought to be very satisfactory and a good vindication of the compound pendulum technique. Dubious values have been highlighted.

Subject	Leg Component	Method	Distance to COG from Swing axis along y axis (m) From x axis	From z axis	Prediction
Standard	Trans-tibial Socket & Pylon	Balance Compound	0.1775 0.2173 ± 0.0069	0.1865 0.1893 ± 0.0024	
Standard	SACH Foot	Balance* Compound	0.07 0.0910 ± 0.0050	0.07 0.0605 ± 0.0018	
E	Shank	Balance Compound	0.1337 0.1281 ± 0.0007	0.1262 0.1276± 0.0002	0.177
E	Thigh	Balance Compound	0.250 0.264 ± 0.0062	0.247 0.26 ± 0.0055	0.250
F	Shank	Balance Compound	0.136 0.139 ± 0.0009	0.132 0.138 ± 0.0013	0.176
F	Thigh	Balance Compound	0.237 0.289 ± 0.0010	0.237 0.241 ± 0.0001	0.248
C	Shank	Balance Compound	0.1339 0.1377 ± 0.0011	0.1391 0.1396 ± 0.0004	0.172
C	Thigh	Balance Compound**	0.245 0.229 ± 0.0021	0.239 0.244 ± 0.0011	0.242
B	Shank	Balance Compound	0.1189 0.111± 0.0024	0.1189 0.107 ± 0.0005	0.179
B	Thigh	Balance Compound**	0.247 0.254 ± 0.0006	0.233 0.0.2452 ± 0.0015	0.253
H	Shank	Balance Compound	0.1063 0.1029	0.1072 0.1014	0.180
H	Thigh	Balance Compound	0.2003 0.2526	0.1965 0.2437	0.255

Table 8-I – Distance to the COG of various leg components from the swing axis as measured by the balance technique and as calculated from the compound pendulum method. Note that for these trans-tibial subjects the term shank represents the stump, socket, and pylon.
** Awkward hard to balance shape therefore difficult to measure COG.*
***Thigh cast swung about hip joint rather than knee joint.*

If discrepancies did however occur, it was usually obvious whether the source of error was from the balance technique or the compound. For instance, there was always a small degree of uncertainty with the directly measured values. This uncertainty was greatly increased when awkward shapes were used that were difficult to balance. The main cause for error with the calculated values of COG were thought to arise due to the standard weight being poorly positioned below the cast; it was sometimes tricky to attach so that it lay in the correct orientation along the y axis. In future, it may be preferable to use a spherical weight in which the moment of inertia is constant for all orientations. Therefore, in situations of doubt, a deduction could usually be made of which was the most reasonable value to use for calculating the moment of inertia; generally the calculated COG values were preferred if they were reasonably close to the measured values, otherwise the measured ones were used. As a further indication, the COGs obtained for the x and y-axis should yield similar distances. This knowledge helped to identify dubious values, e.g. see the calculated x-axis value for the BK socket and pylon, or x-axis on subject F's thigh. In both cases, they clearly stand out.

The small deviation in the calculated COGs demonstrates the high repeatability of the frequency measures. This gave great confidence in the validity of using Vicon for such measurements.

For a final confirmation, the COG for the thigh segments was also calculated according to models made by other researchers. It should also be noted that for most of the thigh measurements it was more practical to locate the swing axis through the knee joint. Therefore, care had to be taken when comparisons were being made with other researchers since most results that were published e.g. Drillis and Contini [63] measured the distance to the COG from the proximal joint. The use of the measurement from the knee joint however gave an added benefit since it was thought that the knee joint centre could be located more accurately by palpation than the hip joint centre, thereby the co-ordinates of the COG could be calculated more accurately using this distance from the knee joint. Therefore, all thigh COG distances have been given according to the distal joint, where predictions have been converted. Therefore, it can be seen that the predicted COGs' closely match the results that have been obtained from the casts.

Predictions have not been made for the shank inertial properties of trans-tibial amputees since these drastically differ from non-pathological subjects. This is immediately evident with the distances to COG e.g. with the prosthetic shank COG (i.e. the stump + socket + pylon). According to estimates based on body proportions such as Contini [48, 49], the average location of mass centre from the knee joint for the shank is 40.4% of the segment length. Therefore, taking a segment length of 0.453m, as calculated for a subject in section 8.2.1.1, [Figure 8-6, pp. 254], as a case in point, this would give a distance of 18.3m to the COG. Now comparing all the COG values in Table 8-I, the mean is only 0.13m. The COG for BK subjects is clearly closer to the knee joint than might be expected by such models that are normally based on non-pathological data[44]. This would be expected given that the pylon portion of a prosthesis is generally constructed to be much lighter than its anatomical counterpart. This situation is also likely to be experienced with trans-femoral amputees, an important consideration for control purposes [see section 6.2.4.1]. Such differences also justify the effort made in directly measuring moments of inertia rather than relying on predictions.

The moment of inertia values have thus been calculated using every combination of COG and ω_{cast} that was available [Table 8-J]. The mean and standard deviations of the results, using the calculated COGs' from the compound pendulum method are displayed in the rows labelled "compound", and those which were directly measured in the rows labelled "balance". The frequency data has been obtained using the Vicon system unless otherwise specified within the "Object" column. It should be noted that the distance to the COG from the y-axis was assumed zero in the torsion pendulum measurements since for the objects the axis of rotation was generally aligned to go approximately through the COG.

The first few objects that are shown in the table are standard objects whilst the remaining shows the moments of inertia of different body segments. The Y-harness was used to hold objects in place during the torsion pendulum measurements. Its inertia had to be known so that it could be subtracted from any subsequent y-axis

[44] Although Contini has released a report on Pathological data [48], the findings for BK subjects were not conclusive, and no attempt had been made at any specific model.

inertial measurements. In addition, because it was not practical to measure the moments of inertia of the prosthetic components for every subject, sample components were measured which would enable an estimate. Of particular note was the measurement for the BK prosthetic shank (socket and pylon), where the radius of gyration of the sample was used to estimate specific inertias of individuals. The assumption for this was that the shape of sockets and thus the radius of gyration of sockets would be similar between subjects. Differences in mass e.g. because of thicker socket walls could thus be incorporated in the moment of inertia estimation.

It should be noted that for the overall shank estimation (socket, pylon and stump), the COG of both the stump and the prosthesis had to be known. If it had not been possible to measure the prostheses COG using the balance technique, then an estimate based on the standard sample was used.

Since the inertias of the prosthetic feet were so small, it was also thought acceptable to use a sample SACH foot for all subject calculations.

Subject	Component	COG acquisition	Moments of inertia about respective axis /kgm^2 x-axis	y-axis	z-axis	Predicted
N/A	Y-harness*	-	-	0.0063936	-	-
N/A	Y-harness on another day	-	-	0.00642 ±0.00001	-	-
Standard	Trans-tibial socket and pylon	Balance Compound	0.0216 ± 0.0001 0.0175 ± 0.0010	0.0015 ± 0.000008	0.0184 ± 0.00029 0.0182 ± 0.00034	-
Standard	SACH Foot	Balance Compound	0.0017 0.0011	0.0015	0.0019 0.0019	-
E	Shank	Balance Compound	0.0390 ± 0.0001 0.0407 ± 0.0002	0.0033 ± 0.00005	0.0425 ± 0.00001 0.0422 ± 0.0001	0.0452
E	Thigh	Balance Compound	0.1199 ± 0.0003 0.0923 ± 0.0123	0.0255 ± 0.00009	0.1178 ± 0.0006 0.0971 ± 0.0107	0.0913
E	Thigh	Balance Compound (digital camera)	0.1221 ± 0** 0.0710 ± 0.0102		0.1188 ± 0.0017 0.0576 ± 0.0181	0.0913
F	Shank inc. light prost.	Balance Compound	0.0327 ± 0.0001 0.0316 ± 0.0003	0.0032 ±0.00007	0.0348 ± 0.00001 0.0327 ± 0.0004	0.0529
F	Thigh	Balance Compound	0.1097 ± 0.0005 0.0243. ± .0169	0.0295 ± 0.00007	0.0993 ± 0.0001 0.0921± 0.0002	0.1069
C	Shank	Balance Compound	0.0290 ± 0.0001 0.0281 ± 0.0003	0.0025 ± 0.00001	0.0281 0.0279 ± 0.0001	0.0321
C ***	Thigh	Balance Compound	0.0436 ± 0.0001 0.0442 ± 0.0002	0.0118 ± 0.0001	0.0428 ± 0.0002 0.0423 ± 0.0002	0.0648
C Shank	(IR beam timer)	Balance Compound	0.0199 ± 0.0009		0.0166 ± 0.026	0.0321
C Thigh	(IR beam timer)	Balance Compound	0.0354 ± 0.0001 0.0387	0.0118 ± 0.0001	0.0332 ± 0.0002 0.0366 ± 0.0002	0.0648
B Thigh		Balance Compound	0.0818 ± 0.0002 0.0813 ± 0.0002	0.0334	0.0820 ± 0.0004 0.0835 ± 0.0003	0.1054
B Shank		Balance Compound	0.0398 0.0395 ± 0.0005	0.0044	0.0408 0.0413 ± 0.0001	0.0522
H Thigh		Balance Compound	0.1125		0.1125	0.1094
H Shank		Balance Compound				0.0541
I Thigh						
I Shank						

Table 8-J – These moment of inertia figures were derived by analysing all possible combinations of object swing frequency and object centre of gravity. Thereby, the standard deviations indicate the repeatable of the results. Two values are given for the moment of inertia depending on whether the COG is calculated from the compound pendulum method or measured from the balance technique. The predicted results are calculated from the relative ratio of the radius of gyration to segment length, which is 0.250. The predicted segment masses and lengths are taken from Table 8-F and Table 8-G. The term shank represents the stump, socket, and pylon
** Only one acceptable trial, rest had too much perturbation*
*** All trials had same frequency*
**** Lightest amputee therefore figures not surprising,*

Once again, there is good agreement with the comparable inertia values. Only two questionable ones have been highlighted, and these were mostly due to COG discrepancies. The mean shank values were 0.0285 (14.0%), 0.0031 (28.2%), and 0.0296 (14.4%) in the x, y, and z-axes respectively; numbers in brackets show the S.D as a percentage of the mean. Similarly, for the thigh, the mean was 0.0752 (47.5%), 0.0247 (37.3%), and 0.0832 (34.3%). The thigh values seem to show larger deviations than the stump, where for the smallest amputee there was a very low moment of inertia value of about 0.04 for the x and z-axes.

Predictions were also made for the thigh moments of inertia; these are based on a ratio of 0.25 for the radius of gyration to segment length. The segment mass and length can be estimated as before so that the inertia can be derived (see Table 8-J). Although the predictions are not precisely the same as the measurements, they are all certainly in the same ballpark and indicate that the compound pendulum method was producing convincing results. Deviations from the predictions are therefore likely to be indicative of actual subjective differences and relate more to variations that may be expected in a population. For example, the smallest subject mentioned above had a thigh inertia that was significantly smaller than the predicted value; however, given his diminutive morphology this would not be too surprising.

In summary, the method of directly measuring mass properties proved to be most beneficial for the prosthetic shank and foot where predictive methods differed from measured results. Although the predicted thigh results were probably more valid, measuring the thigh inertia was beneficial for subjects who suffered from muscle atrophy or whose general soma-type deviated from the model.

8.2.2.2 GROUND REACTION FORCES

Below, the ensemble averages of the ground reaction force components have been plotted so that intra-and inter-subject inspection of the general profiles can be made [Figure 8-20]. The first diagram a) demonstrates remarkable consistency of the vertical component of the ground reaction force on an intra-subject level. There is more variability with the other two components [Figure 8-20 c) and d)]. The variation between subjects is much more apparent, as might be expected, however in the first diagram the effect of gravity on body mass gives a bias to these profiles therefore normalised plots with respect to body weight are also included [Figure 8-20 b, c, and d].

It can be seen that the patterns produced by subject C are far more prominent as a proportion of body mass than the other subjects are, with larger peaks and troughs. In fact, for this subject, the amplitudes of the gait parameters F1-F9 are comparable to non-pathological subjects and may help explain how this subject's kinematic profiles resemble non-pathological subjects. In particular, the vertical and fore/aft forces are significantly larger and may be a result of greater movement of the body's centre of mass. It is suspected that this subject, in particular, owes much of this high level of "performance" to being a congenital amputee who has learnt to use prostheses since childhood. In this context is also interesting to note that subject B had the second most dynamic pattern despite being one of the less fit amputees [section 8.1.1.1, pp. 240]. It may therefore be no coincidence that as the oldest amputee, this subject also has had long experience with a prosthesis having had an amputation about 50 years ago. The other amputees were traumatic amputees with less experience. From a temporal viewpoint, this subject is also slightly different from the other subjects where the onset of F1 (with the exception of B, a long-term amputee) seems to occur earlier and F3 later.

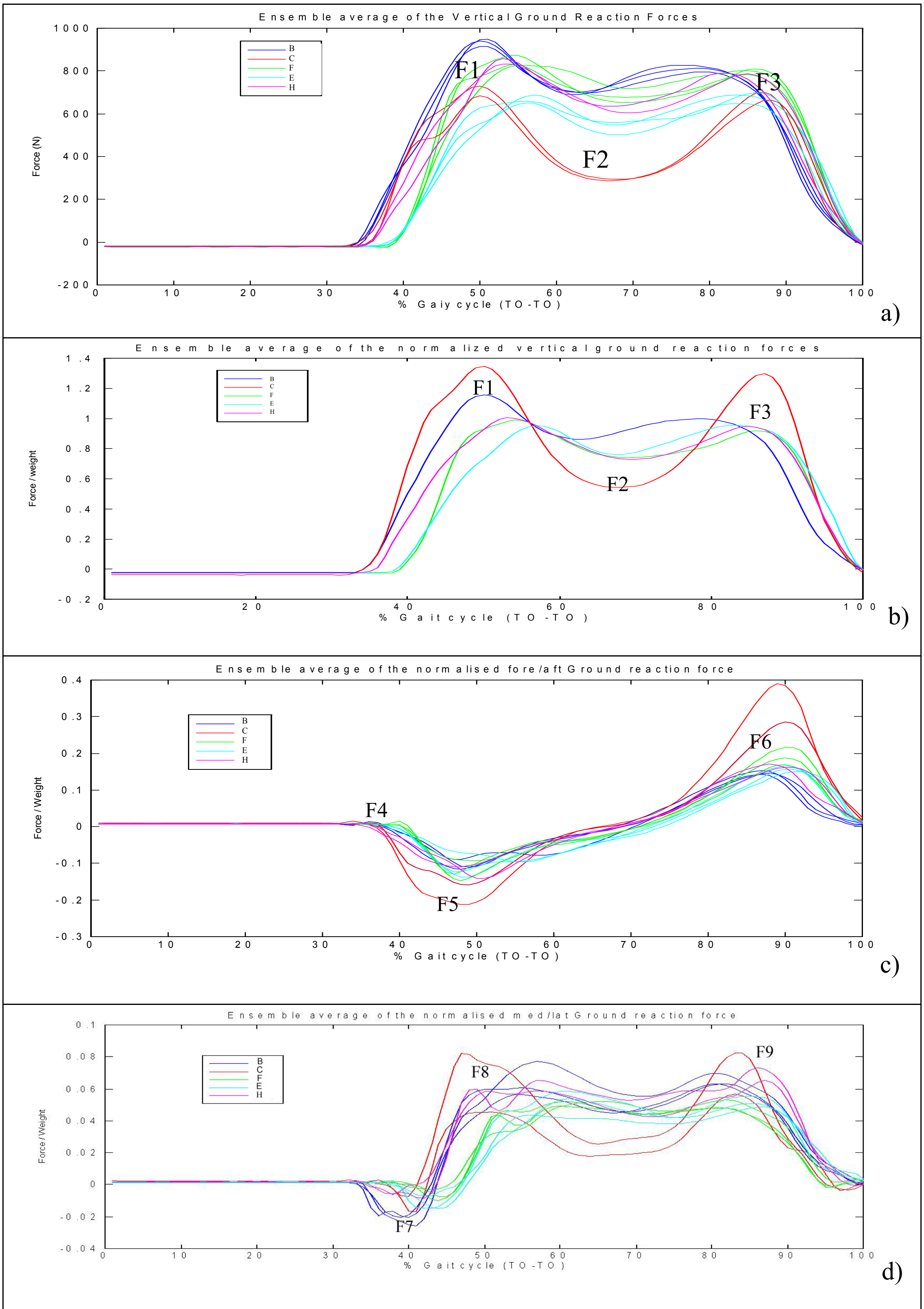


Figure 8-20; Ensemble average plots of the ground reaction force components experienced by the trans-tibial amputees. Figure a) gives the vertical component measured in Newton and includes several sessions from each subject. The remaining figures have been normalised with respect to the body weight (M) and for clarity, only one session from each subject has been included. The vertical component, fore-aft, and medial-lateral components have thus been shown in diagram b), c) and d) respectively.

Session Cb from subject C is used to display the effects of speed on the ground reaction force components [Figure 8-21]. The vertical and fore/aft components appear to be very sensitive to speed, whereas the mediolateral force was more chaotic especially with other subjects. Diagram b) shows that the F1 parameter drifts to the right with speed whereas F3 remains relatively consistent. In contrast, for subject B, the F3 parameter drifted towards the right as speed increased, whereas for subject F all three parameters diverged from each other. Therefore, there are variations within these patterns from subject to subject, which makes it difficult to use this normalised temporal scale to indicate the exact phase of gait parameters.

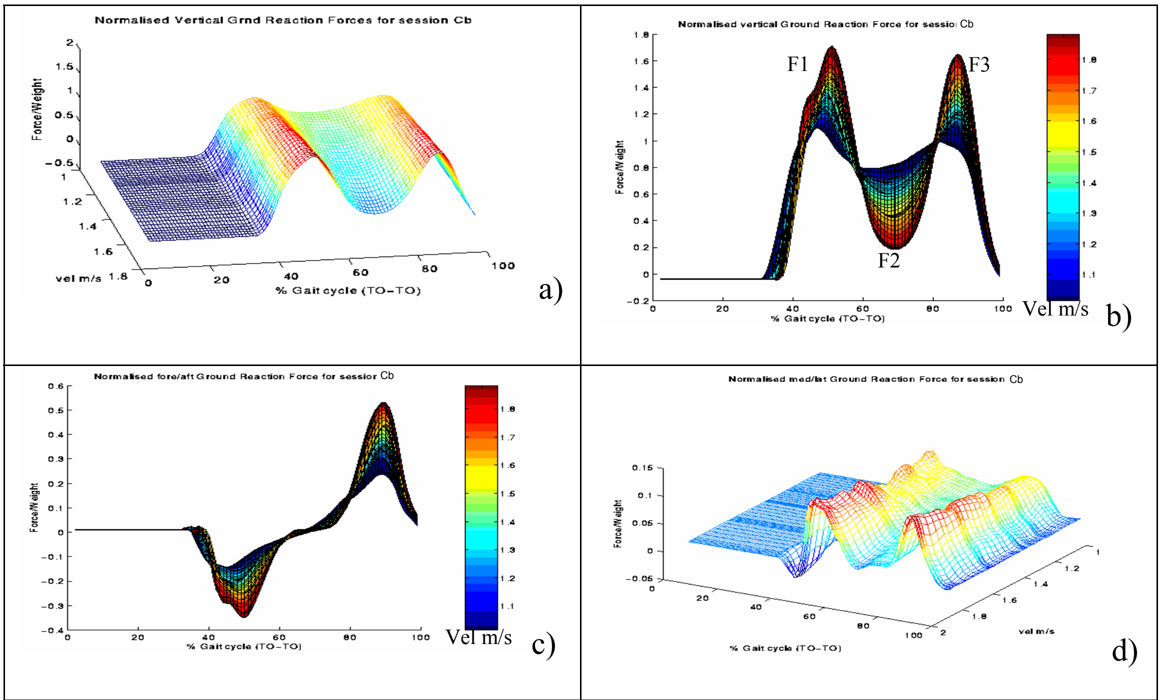


Figure 8-21; Plots of ground reaction force components for session Cb. Plot a) vertical component, b) vertical component with profiles colour coordinated with respect to speed, c) fore/aft component colour coordinated with respect to speed, and d) mediolateral component.

8.2.2.3 INTER-SEGMENT/MUSCLE MOMENTS

The morphology of inter-segment moments, as a surface on the gait cycle/velocity plane, was of particular interest in order to show precursors to control strategies. In particular, when should particular "*actions*" be performed?

8.2.2.3.1 Ankle

The ensemble-averaged moments about the prosthetic ankle joint are shown in order to demonstrate the intra and inter-subject variability of the profiles [Figure 8-22]. It can be seen that apart from one session with subject B, all profiles are contained within an envelope that resembles the non-pathological. The subject variability is surprisingly low considering the use of different prosthetic feet (including on an intra-subject level for some sessions).

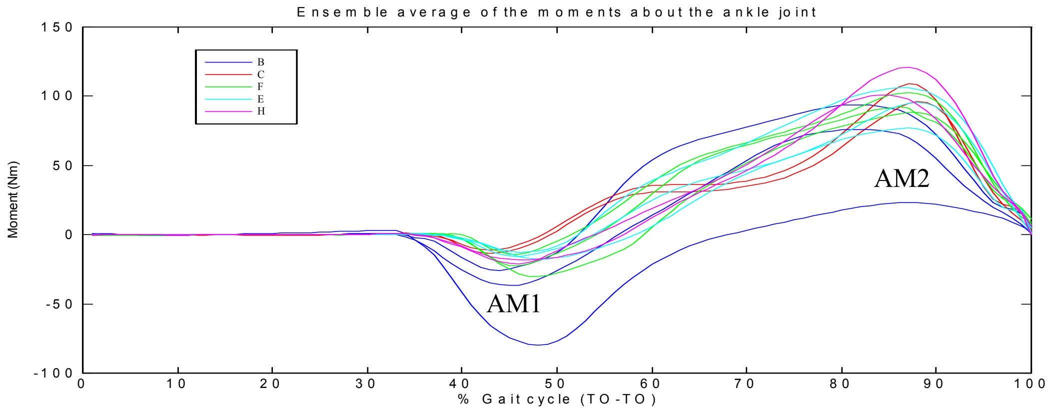


Figure 8-22: Ensemble average of the inter-segmental moments about the flexion axis of the prosthetic ankle joint. Sessions are colour coordinated according to the amputee.

To view the general variation of these profiles with speed, session Cb has been plotted in three dimensions [Figure 8-23]. Clearly, the AM2 peak increases with speed, however the AM1 peak remains relatively constant. The AM2 peak in addition appears to be invariant in the temporal axis, whereas the AM1 peak follows the heel strike outline, which occurs earlier on this scale for lower speeds. The AM2 speed dependence for other subjects is however far less sensitive so that changes are not as vivid and for some subjects e.g. B barely noticeable.

Comparing the values of these amplitudes with non-pathological subjects, only subject C (with a max. value of 2.5Nm/kg) exceeds the fast speed values presented by Winter [222] of 1.8 Nm/kg for AM2 and –0.1 for AM1. In contrast, subject B does not even achieve Winter's low speed value of 1.5 Nm/kg.

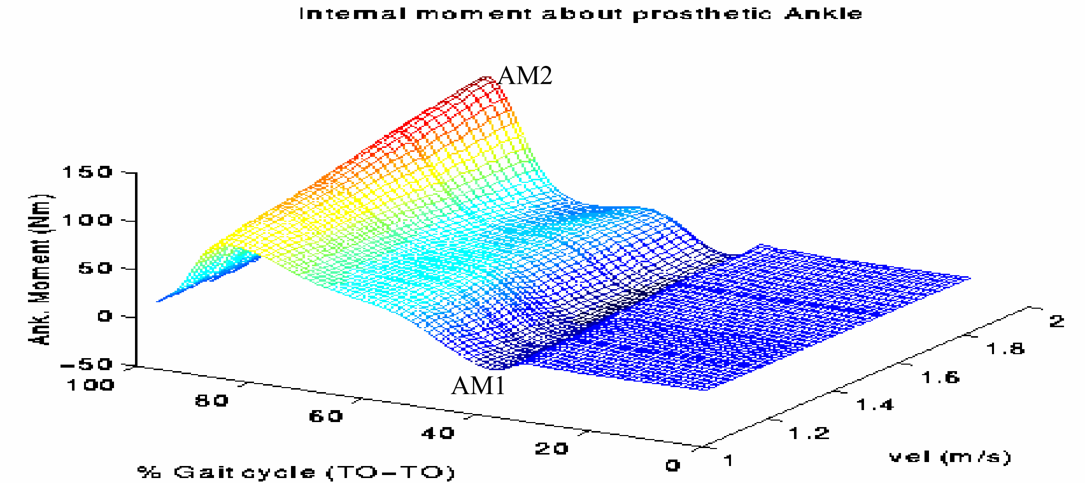


Figure 8-23; 3D plot showing velocity relationship of the moments about the ankle of the prosthetic foot for session Cb.

8.2.2.3.2 Knee

The general moment profiles about the knee flexion/extension axis as caused by muscles are shown in Figure 8-24 for the sample of trans-tibial amputees.

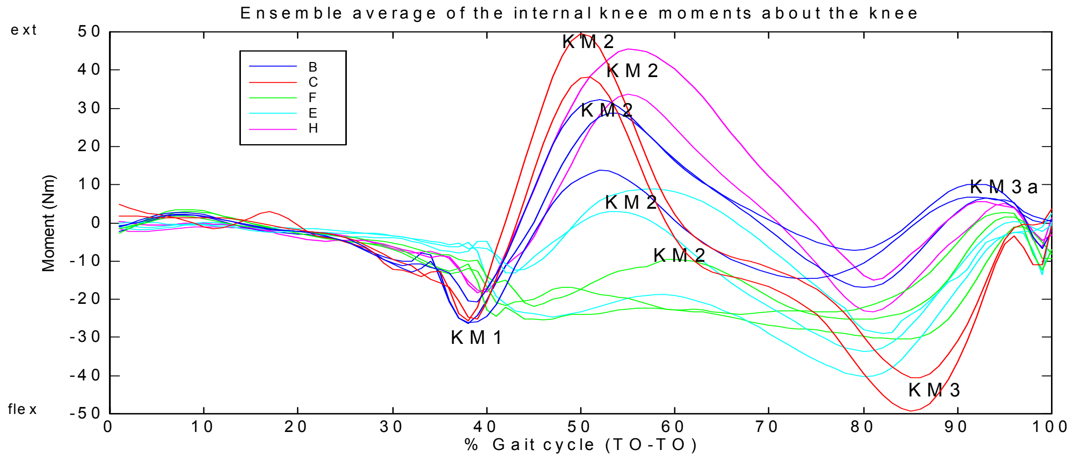


Figure 8-24; Ensemble average of the inter-segmental flex/ext. moments about the knee joint on the prosthetic side. Sessions are colour coordinated according to the amputee. Labels are according to Benedetti's convention [16] apart from KM3a, which has been additionally inserted.

Unlike the moment patterns about the prosthetic ankle, which must be governed passively, there appears to be much greater variability between individuals of the moments generated by muscles spanning the knee joint. This inter-subject variability of the ensemble knee moments is evident not only with amplitude but also with respect to the phase of the gait cycle. For example, the KM2 gait parameter occurs earliest in the gait cycle for subject C, then followed by B, H, E and F in succession.

In fact, for subject E and F, this parameter is so late and flattened in curvature that its definition is questionable.

	Session	KM1	KM2	KM3	KM3a
B	Ba	-21	32	-7	10
	Bb	-26	28	-17	7
	Bd	-26	14	-15	7
C	Ca	-26	49	-41	-1
	Cb	-25	38	-50	-4
E	Ea	-13	3	-34	0
	Ed	-13	9	-27	-2
	Ee	-24	-19	-40	-3
F	Fb	-22	-17	-23	2
	Fc	-22	-17	-24	3
	Fe	-22	-10	-25	2
H	Ha	-18	34	-23	5
	Hc	-18	45	-15	6

Table 8-K: Gait parameters measured in Nm for the average inter-segmental knee moment about the sagittal axis. Unusual values are highlighted.

The session averages of the knee moment gait parameters are listed in Table 8-K. KM1 appears to be very consistent on an intra-subject basis apart from E. The values are also comparable with non-pathological values. KM2 was much more variable on both an intra and inter-subject basis. Subjects E and F had the least prominent KM2 profile, which for one E session and all F sessions was unusually a flexion moment. Only subject C produced KM2 values that were as high as the non-pathological subjects presented by Winter [222]. The KM3 parameter seemed in general to be less than for non-pathological subjects where KM3 is roughly level with KM1. KM3a although not labelled by Benedetti, was observable for most of the trans-tibial amputees. Benedetti may have omitted to label it because of its inconsistency; for example with these subjects, KM3a is unpredictable as to whether it acts as a flexion or extension moment.

To show the speed variation of profiles within a session, a full session from each subject is presented in Figure 8-25.

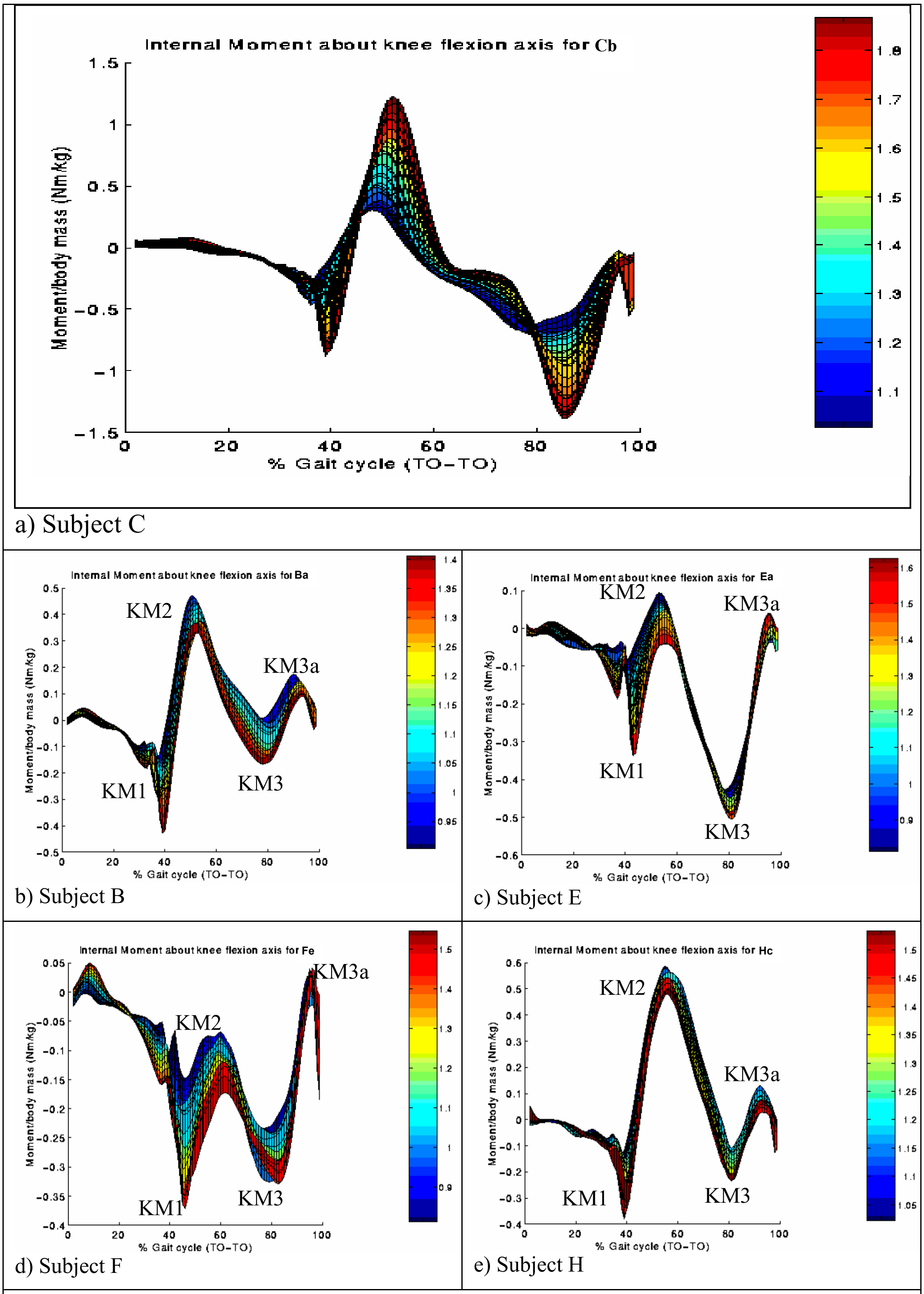


Figure 8-25: Plots of the inter-segmental knee moment about the flexion axis. Each trial within a session is colour coordinated with respect to speed so that red is fastest and blue slowest (see colour bar). These examples have been normalised with respect to body mass.

Of first note is that, as might be expected, all subjects display some variance with respect to speed. The exact nature of this however differs uniquely for subject C compared to the other amputees. For subject C at KM2, the fastest speeds (red) have the greatest moment value whereas for the other subjects the opposite is true. Therefore, it appears that subject C once again has the closest profile to non-pathological subjects, who according to Winter's data also increase the KM2 peak with speed [222]. One anomaly with Winter's data however was that for the KM3 minima, the fastest speed had the smallest value, whereas slow was also smaller than natural. With all of the amputees in the project, the KM3 peak became larger with speed.

The KM2 peak also displays the greatest temporal variability where it changed phase by up to 5% of the gait cycle between fast and slow speeds. Interestingly the direction of this change was opposite for subject C compared to the other subjects, where for fast speeds this apex occurred later than for slow speeds.

8.2.2.3.3 Hip

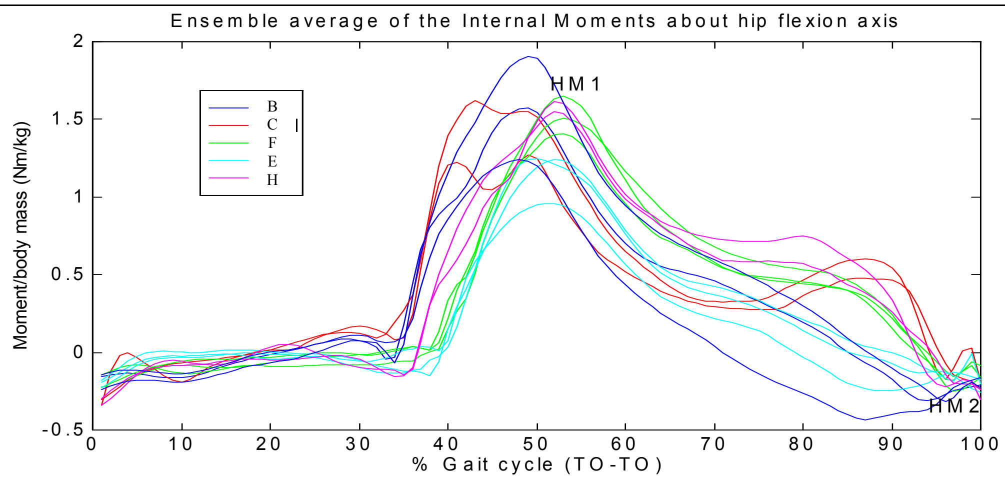


Figure 8-26: Ensemble average values have been taken for the trials within each session. The inter-segment moments about the hip flexion axis are thus colour coordinated according to amputee. Benedetti's label convention is used [16].

The average inter-segmental moments about the hip flexion axis for each session are shown in Figure 8-26. From this figure, it can be seen that there is good intra-subject consistency, whilst much variability is apparent between subjects. A basic shape is apparent however, which is similar to the profile produced by non-pathological subjects. Interestingly, the moment values at the HM1 gait parameter appear to be strikingly larger than even the equivalent fast speed non-pathological parameter. The increased activity may be compensatory for the lack of activity with the prosthetic foot.

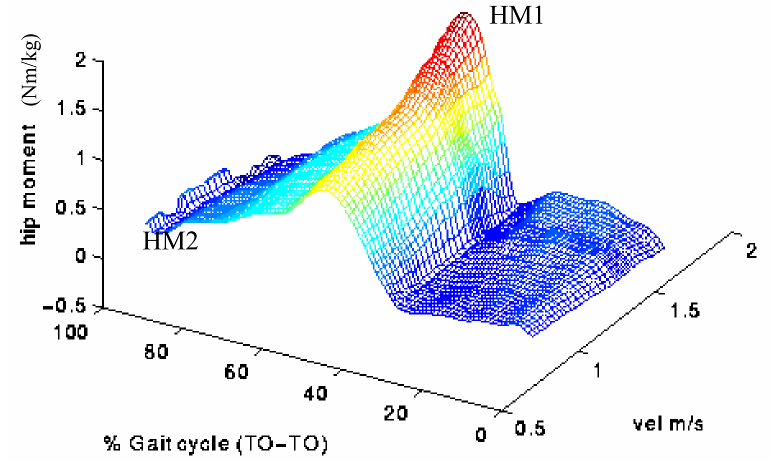


Figure 8-27; Inter-segmental hip moments about the flexion axis for session Ee (subject E). These values are normalised with respect to body mass

The speed dependence of the profiles was similar for all the subjects; an example is shown in Figure 8-27. It can be seen that the HM1gait parameter is most sensitive to speed. There was no clear trend for the temporal variance of the gait parameters. For example, the HM1 apex, for some subjects, followed the temporal path of the heel strike event, which as speed increases occurs later in the gait cycle. The converse was however observed with other amputees.

8.2.2.4 INTER-SEGMENT JOINT POWERS

The calculated inter-segment power described the rate at which work was done by muscles and tissue on adjoining segments, thereby influencing each segment's rotational kinetic energy. Since two segments are coupled at the joint, the power was calculated from the product of the inter-segment moment and the "relative" angular velocity between the two segments. This meant that the power reflected the relative rate that kinetic energy was generated or absorbed between the segments, which is often interpreted as concentric or eccentric muscular contraction respectively [section 2.2.2.2.3], (zero power reflects conditions of isometric contraction where force/moments only are transmitted between segments). Therefore, it was hoped that this variable would give a good indication of the timing of control events useful for a knee controller. Since power is the product between the moment and angular velocity, it was also thought a useful way of condensing more kinetic information than just using the inter-segment moments.

Therefore, particular attention is given in this section to the knee power; the general features of the knee power profiles are visible in Figure 8-28, which is a normalised plot with respect to body mass of each session.

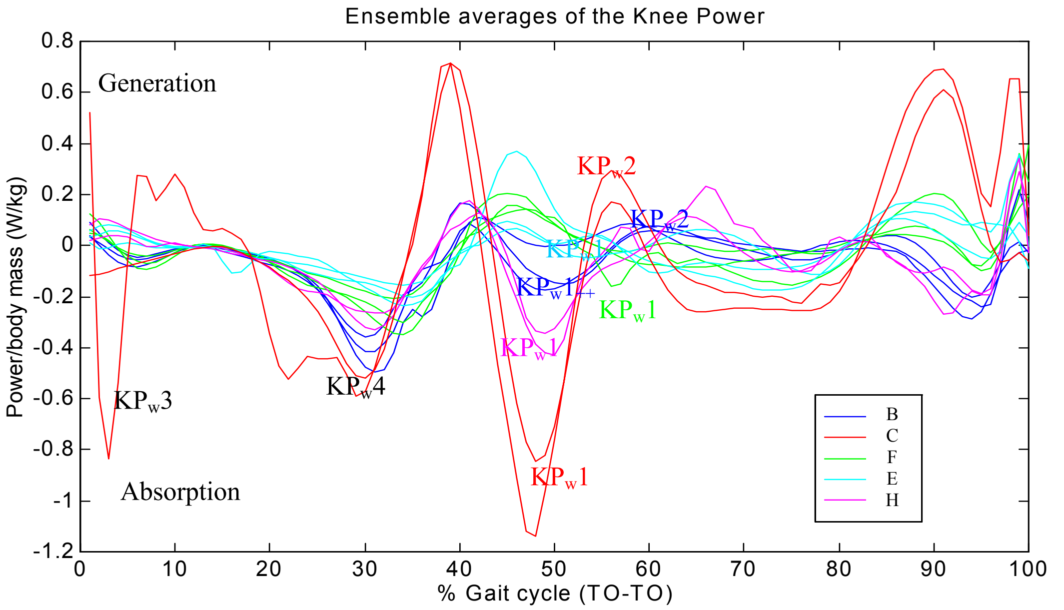


Figure 8-28: Ensemble average taken for each session of the inter-segmental/muscle Power at the knee joint. The power has been normalised with respect to body mass. Each session is colour coded according to amputee, and the same parameters that Winter [222] chose have been used hence the labels (KP_w).

Although, intra-subject consistency was apparent, the patterns are much more erratic than the other variables that have been reviewed. On an intra-subject level, the variability manifests in the temporal direction, where various gait features, in particular, KP_w1 and KP_w2 are easily confused between different subjects. In addition, the amplitudes of different parameters range between what can be observed with non-pathological subjects to almost non-existent, for example KP_w2. Such anomalies may be the reason for Winter not labelling more features that may be distinguishable.

The power profiles also exhibited a speed dependent change. Examples from two subjects are given in Figure 8-29, where each trial of a session has been plotted in three dimensions. It is immediately obvious that as speed reduces, the amplitudes are considerably reduced almost to the point of losing some features.

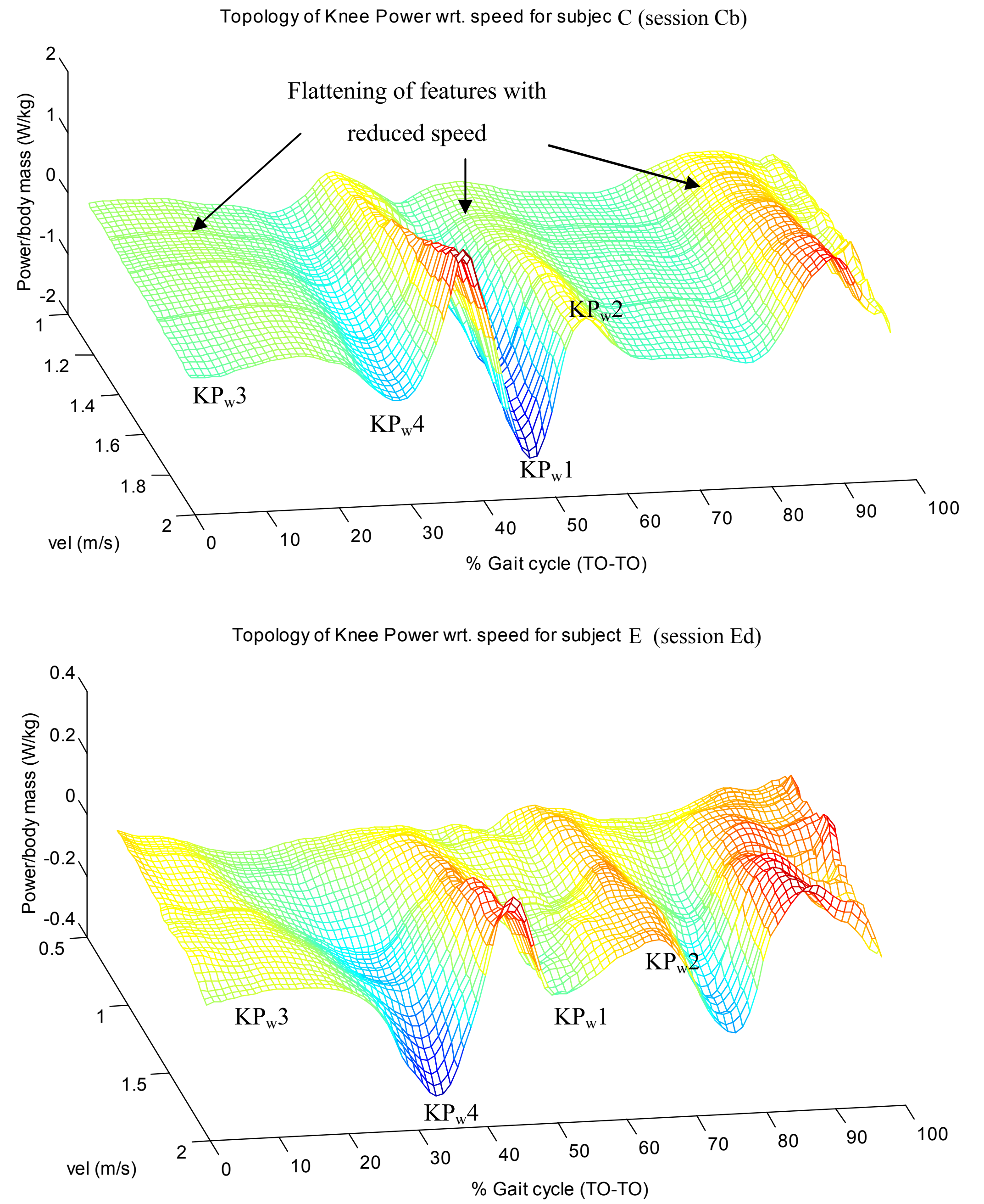
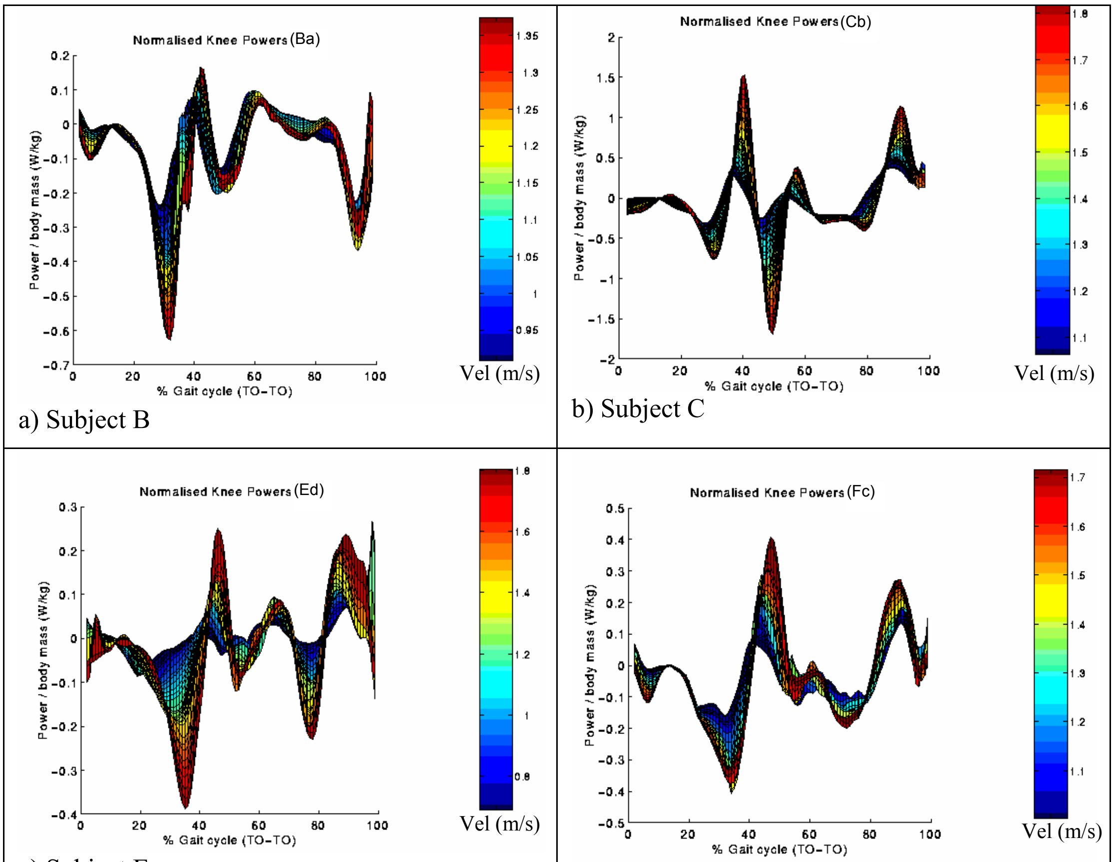


Figure 8-29; Surfaces generated from trans-tibial amputees' knee power (for the prosthetic side) with respect to speed and % gait cycle. Two contrasting subjects have been chosen (C and E). The data has been smoothed by iteratively passing over an averaging window (8 trials by 2 % gait cycle).

The chosen subjects exemplify the temporal differences that occur around KP_w2 during mid-stance. It can be seen that for both of the subjects the KP_w3 minima even at high speed are not great, especially considering that in non-pathological subjects these are as large as the KP_w4 minima. The following figure however gives a better quantitative description of how these features vary with speed [Figure 8-30].



a) Subject B
b) Subject C
c) Subject E
d) Subject F

Figure 8-30: Example sessions from different subjects showing the variation of knee power with speed. The colour bar indicates the speed.

8.3 INTENT RECOGNISER

8.3.1 CHOOSING INTENT STATES FROM KNEE POWER PROFILE

The culmination of this project will be to establish and recognise when control phenomena should occur relating to a particular movement task. This problem was reduced to finding characteristic features of the knee power experienced during the task. It has just been shown that the reference group of trans-tibial amputees utilise a varied power profile whilst walking from session to session and even more from subject to subject. Clearly, there will be even more variability for the individual trials so that great care had to be taken when selecting topological features that was consistent from trial-to-trial, session-to-session, and subject-to-subject.

After much scrutiny of the profiles, it was realised that some additional aids were required to reliably distinguish power events especially in order to ensure consistency from subject to subject. The power events that were initially selected were those that were considered more robust. The method of selecting power events, from every trial and every session, was largely visual because of the complexity of automating such a process. However, to assist this, various steps were taken which were incorporated into algorithms written for Matlab [132]:

The first step was to plot the ensemble averaged knee power of the current session to use as a visual reference of smoothed data. In addition, the ensemble average profiles of the moment and angular velocities about the flexion axis were superimposed onto the power profile. These were used as they represent the most significant contribution to the power. It should be noted that gain factors were applied to the moments and angular velocities in order to reduce their profiles to similar levels of amplitude to the knee power. These ensemble average patterns for each session are shown in Appendix F and the topological features that were thought to be feasible intent phases were labelled [Figure F-1].

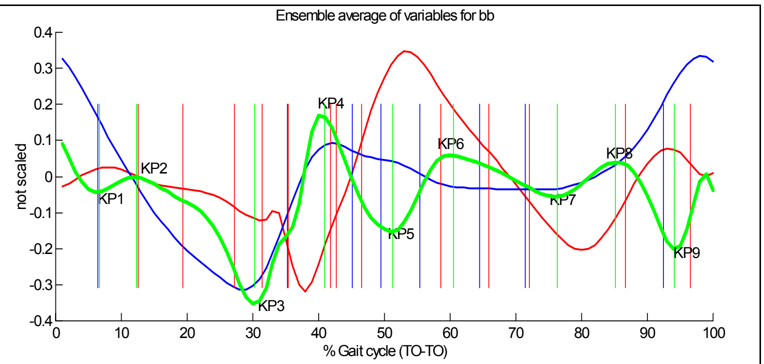


Figure 8-31; Ensemble averages of the knee power (green), knee moment (red), and knee angular velocities (blue) superimposed in order to show the features that were used to distinguish intent states (these states have the prefix KP not to be confused with Winter's gait parameters KP_w).

An example, from Appendix F, is given in which all of these features are readily visible [Figure 8-31]. The session (Bb) was from subject "B" who was the oldest subject and who, according to the previous section, appeared to have gait patterns closer to non-pathological subjects maybe as consequence of longer experience as an amputee. The intent states labelled KP1 – KP9 are marked according to peaks or troughs within the knee power profile. Although selection methods that are more complex are possible such as using maximal gradients, the process was felt adequate in order to see whether the overall philosophy was reasonable.

Therefore, troughs and peaks were selected that had some repeatability between sessions. Closer inspection of these ensemble patterns shows that certain topological features were however more readily identifiable between subjects than others were; the quality of each of these features was thus examined for all the sessions and tabulated [Table 8-A].

Assesment of average Knee Power Topology

phases	KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
expected shape	trough	peak	trough	peak	trough	peak	trough	peak	trough
Sessions: Ba	good	good	good	good	good	good	small trough	small peak	good
Bb	good	good	good	good	good	good	good	good	good
Bd	good	good	good	good	small trough	good	good	good	good
Ca	noisy	bit flat	bit noisy	good	good	good	good but non sharp	good	good
Cb	off edge	good	good	good	good	good	flat	good	good
Fb	good	good	good	right shift	ambiguous	bad noisy and flat	noisy	noisy	ok
Fc	good	good	good	ok right shift	ambiguous right shift	ok but right shift	good	good	good
Fe	good	bit flat	good	ok right shift	blurs into KP7	non existent	blurred	good	good
Ea	small trough	ok	good	ok right shift	small trough	ambiguous confusion with KP5	flat	good	flat
Ed	no trough	ok	good	ok right shift	good	good but right shift	good	good	good
Ee	no trough	bit flat	good	ok right shift	blurs into KP7	non existent	ok blured	good	good
Ha	no trough	bit flat	good	good	good	good but right shift	good	ambiguous	ok
Hc	limited trough	bit flat	good	good	good	noisy right shift	good	good	good

Table 8-L: Comments are made for each intent state regarding the topological quality of the knee power as observed from the ensemble average of each session [Figure F-1]. Note that shifts to the right of a particular state are with respect to previous good sessions.

Of note was that although the KP1 trough was visible in most sessions, with some sessions the average profile had the trough flattened and pushed out to the left of the window. When viewed on a trial-by-trial basis, the profiles taken at slower speeds generally formed this much less distinctive appearance; the trough was however distinguishable on most occasions and it seems the averaging process has removed it. The second area of greatest concern was for the topology surrounding KP6. It can be seen from the table that this peak was ambiguous or even not evident for sessions “Fe”, “Ea” and “Ee”. This in turn caused problems with the definition of the surrounding phases KP5 and KP7. For the sessions that had a distinctive KP6, it seemed that this feature coincided with the maximal knee moment gradient. Therefore, when the phases were being pinpointed on a trial-by-trial basis, the moment pattern significantly helped to identify the correct peak. This was particular the case with session “Fe” where noise or other aberrations seemed to cause ripples to propagate through the profile and obscure the main features. This may have therefore caused a flattened ensemble average profile as various fluctuations cancelled each other out.

Therefore, once generic features had been determined from the ensemble averages, the next stage was to pinpoint each identifiable event within every trial. Therefore, the power profiles of each trial were plotted in succession and the PC mouse used to "point and left click" the positions of each of the nine intent states. To aid the process, contour lines corresponding to the zero gradient and max gradient values of each variable were superimposed as well as the individual moment and angular profiles. The lines represented the contours taken over a small speed range about the current trial speed. Finally, a filtered version of the current power profile was also added to smooth out undulations in the peak to allow a more consistent positioning.

If after all this, noise or ambiguous topologies prevented the identification of a phase then a right click on the mouse signified this. To help increase the amount of data available for testing, the loci of right click events were estimated according to non-ambiguous events. Note that such data was not used to train the neural networks.

In order to estimate the temporal position of ambiguous events, it was desirable not to assume any positioning of events in terms of gait cycle percentage since this normalises the gait cycle according to some non-control event (usually heel strike or toe off as in the case of this study). Instead, all known events from previous trials were used to estimate the normalised periods between all possible events. For example, if a temporal estimate of KP1 is required: the period KP1-KP2 can be normalised with respect to KP1-KP3, or KP2-KP3 etc. and similarly KP1-KP3 can also be normalised until a matrix is built with all possible permutations. One can then find which normalised coefficient is the most consistent across all trials. If the timing of the other phases was known, then these events along with the coefficient were algebraically manipulated and solved to give the timing of the missing phase. In circumstances where the timing of an event related to the normalised coefficient was also unknown, then a different coefficient was used which did not cause conflict when being solved. Finally, as an additional selection pressure, more weight was given to trials that had a similar speed to the trial with the ambiguous event.

8.3.2 TRAINING NETWORKS TO RECOGNISE INTENT STATES

Where possible, the temporal position (using frames) of the knee power intent events has been located within every trial of every session. The objective, as described in section 6.2.3, was to train individual artificial neural networks (ANN) to be able to recognise these states according to a "yet to be determined" set of inputs (or task variables using previous nomenclature). Neural networks would be used to effectively trawl through the data in search of unique solutions representative of each power event.

Artificial Neural Networks are constructed by connecting neurones (or nodes). Each input connection (indexed by the letter i) to a neurone (index j) can multiply the strength of the input signal (p_i) by a weight (w_{ji}). Usually, the sum of each of these weighted inputs to a neurone is taken and passed through a transfer function **F**. The output (a_j) from the neurone can thus be calculated as $a_j = \mathrm{F}\left(\sum_i w_{ji} \times p_i\right)$. The input to a neurone can be some external signal or even the output of another neurone, so that connections can be developed to form a network. By adjusting these basic elements it is possible to design many different types of artificial neural networks; a commonly used classification scheme uses five properties to describe them:

- **Topology** refers to the set of connections between inputs, outputs, and neurones. All topologies will be subsets of the fully connected topology.
- **Architecture** refers to the overall operation of the network and can take on four different general forms although more complex formations are possible:
    - *Feed-forward:* where there is a straightforward mapping from inputs to outputs, the inputs to a layer of neurones derive from previous layers.
    - *External feedback:* where the output is fed back to the input so that the previous output provides the new input. Only one external stimulus is required at the start to initiate a particular response. If additional stimuli are applied the network will effectively be reset and respond to these without knowledge of prior outputs. This property enables networks to generate temporal patterns as well as spatial.

- *Internal feedback:* where the input to the network uses both an external stimulus and the previous output.
- *Unsupervised:* which have the ability to self organise in order to recognise frequently presented input vectors. The network is effectively sculpted according to rules that generally monitor the frequency or spatial distribution of inputs.

- **Neurone model** refers to the transfer functions utilised by the neurones. Most networks generally sum all of the weighted inputs before applying the transformation. In addition, a bias (b_j) may also be applied to the sum of the weighted inputs so that the transfer function is shifted left or right depending on the bias value. Many different transfer functions exist but commonly used ones include step, linear, log-sigmoid, tan-sigmoid, radial basis functions etc.

- **Training algorithms** are used to adapt network parameters such as biases and weights in order to reduce the output error produced in response to a particular set of input data. With supervised networks, the error is calculated according to a pre-established set of target outputs. Different training algorithms have been developed to efficiently iterate through the different network parameters (dependent on topology, architecture and neurone model), in order to find where the error minima may lie. For example, the most commonly used algorithm is back-propagation and its variants for multiple layer networks using non-linear differential transfer functions. It is a procedure for efficiently calculating the derivatives and using these to adapt weights or parameters of the non-linear system. Generally, the fastest back-propagation algorithm for estimating functions is the Levenberg Marquardt optimisation, whereas for pattern recognition problems the resilient back propagation algorithm is often found to be quicker [200].

- **Operation Schedule** controls which neurones respond and when. In the multi-layered feed-forward the input layer will respond before hidden layers as the outputs are fed into them; the neurones within a layer will therefore be synchronised to produce their outputs at the same time. In networks with feedback, the set of neurones responding at a particular time may be chosen randomly. A synchronous feedback network in this case would have all neurones responding at the same time, whereas an asynchronous feedback network would have one neurone responding at a time.

There is thus a myriad of options to design artificial neural networks; a strategy was therefore required to sort through these permutations and find networks that yield the desired response for this project.

8.3.2.1 SELECTING / DESIGNING NEURAL NETWORKS:

To constrain the network design, information regarding the nature of the problem was first considered.

Constraining the Neurone model:

A linear relationship between possible inputs and outputs could not be assumed. It was unlikely that a particular knee power state mapped in the input space could be linearly separable from all other states. Therefore, despite an increased computational cost, it was thought better to start with a more generalist approach, in which non-linear regions are defined (see Figure 6-2 for a visual analogy). In practice, this can be achieved by using non-linear differentiable transfer functions for the input layer at least. A two-layered network for example, with the first a non-linear tan-sigmoid layer and a linear output layer, was capable of approximating any function, the accuracy being dependent on the number of neurones.

Constraining Output:

For this project, the output did not need an infinite range - but could be constrained between one and zero to determine whether a particular intent state was present or not. Therefore, a log-sigmoid transfer function was chosen for the output layer as it ranged between zero and one. This was considered preferable to a step function because it gave the networks the ability to indicate the probability of an intent trigger state, which could be useful later for decision making when inputs are ambiguous. Only one neurone was therefore required for the output layer since a single number between zero and one could describe the target.

Constraining Operation Schedule:

Recurrent networks were also ruled out as an initial start point because although gait may be viewed as an oscillation, the detection of power states was meant to act as a trigger point that could occur anytime. In this situation, the output is viewed as a single temporal point.

Therefore, experimentations began with two layered feed-forward networks. The log-sigmoid transfer function was used for the output layer and either the tan-sigmoid or the sigmoid function was used for the input layer. A hidden layer was omitted.

Constraining training algorithm and training time: With the architecture described thus far, back propagation could be used to adjust the weights and biases in order to minimise the mean square error (mse) of the training set. It can do this by estimating an instantaneous gradient and following the path of steepest descent and minimize the mean square error of the training set. This algorithm according to popular wisdom generalises better than alternatives such as radial basis function networks although training time is longer, for example, Musavi et al. examine such issues [145]. To try to improve training times, a variant of back propagation known as the Levenberg-Marquardt optimisation was finally used for training of networks as it was considered one of the fastest although at the expense of memory requirements. The compromise with memory requirements meant that a sufficiently powerful computer was necessary for the larger network topologies. If this method proves problematic, then an alternative algorithm was available. The Resilience back propagation algorithm was also considered because of its effectiveness with pattern recognition problems. In this context, the nature of the problem may be interpreted as the recognition of particular patterns of inputs synonymous with the motor intent state. In this regard, this highlights the advantage of using two layer feed-forward networks, where because of their popularity a variety of learning algorithms was available. In particular, because of the extensive use of Matlab [132] for data preparation and analysis during this project, its use was naturally extended to implementing neural networks. Towards this, Matlab even provided a Neural Network toolbox that included useful functions such as the Levenberg-Marquardt optimisation technique. It was still necessary however to write code to organise the data, train the neural networks and test them.

The time taken for a network to be trained towards an acceptable error, is also dependent on many other factors including size of training data, size of network, computer processing power, and training parameters such as the error goal, and learning rate (which is a gain on the adjustment of weights and biases according to errors). Whatever combination is used, a practical limit to the maximum amount of training time for a network will exist. For example, when the training of neural networks was first attempted, the available PC was very slow for the task, and to train any network with over 20 input neurones would necessitate overnight processing. Although things have improved one cannot go mad with the network size for example. A compromise is necessary between the number of networks that can be designed and the size of them.

Constraining Inputs: Finally, given that 48 different gait variables[45] were available from the analysis of the Vicon data this presented a vast choice to sort through and find the best combination. There were some limiting requirements however; first there needed to be sufficient information to coordinate with the natural control system. Returning to the discussions of section 4.3 the output of the natural system $\tau_{\mathbf{stump}}$ should be incorporated into the artificial controller's target task variable. This is most easily derived from the movement of the stump e.g. as described by hip joint variables or socket variables. In addition, there needs to be feedback as to the outcome produced by the system. This can be any variable relating to the movement of the shank, and can even include events at the prosthetic foot such as the ankle force components which are highly correlated with force components at the knee joint. Therefore, this basic requirement means that inputs cannot be composed of

[45] The following variables were available for input to the neural networks: The angles, angular velocities, and angular accelerations between segments across the ankle, knee and hip joint as measured in the flexion/extension, abduction/abduction and internal/external rotation planes. In addition, kinetic variables including the three orthogonal force components at each joint as well as the original ground reaction forces.

variables that are purely distal to the knee or purely proximal; there must be some mixture.

Although not a primary concern, in terms of future practicality regarding implementation of sensors, it was felt that networks that required fewer inputs were preferable. Therefore, in general, it was desirable to keep the number of inputs below 10, although inputs that could be incorporated as a package would not add to the tally. For example, it is conceivable that the three force components may be detected by a single sensor package; although the network will see three input variables, for practical purposes only one sensor unit is needed.

Data preparation: Redundant inputs will slow down training and if there is a lack of training data could give poor generalisation. However, as long the source of noise for each input is independent they can be advantageous for improving noise robustness.

Principal Component Analysis can be used to help reduce dimensionality by reducing the number of highly correlated input vectors, i.e. removing redundant information. A hidden layer that is smaller than the input layer can also act like principal component analysis by funnelling the inputs and weighting the most important.

Training algorithms can also be helped by **normalising** the input data and target data so that they have means of zero and standard deviation of one. This puts the magnitudes of weights affected by each input on an equal footing. For example, without normalisation, an input ranging from 0-5000 would need much larger weights to yield noticeable effects than inputs ranging between zero and 0.05.

Improving Generalisation: It was recognised that generalisation is improved by keeping networks “simple” so that one with fewer connections and smaller weights is less likely to over fit data. The compromise is that networks, which are too small, cannot accurately approximate the desired input to output mapping. One may therefore use a variety of techniques to try to find the optimal solution which is likely to be the smallest network that will fit the training data. In general terms this may be achieved by either **pruning**, **growing, weight sharing, early stopping, regularization** of networks or a combination.

With the **pruning** technique, one can start with a highly connected neural network and try to remove/prune weights when they approach zero. One problem of starting with complex networks is that the shape of the error surface is likely to contain more minima in which the training might be stuck before finding the optimal.

The **growing** technique contrasts by starting with a minimal network topology and expanding upon it by adding neurones e.g. using the Fahlman's cascade correlation technique [217]. One advantage is that the smaller networks will be much faster to train although there is no limit as to how large the network may be increased. If information gathered from training the smaller networks is used to initiate the larger networks then training times for the larger networks can be significantly reduced.

Early stopping: With this technique, the data is also divided into validation sets as well as training and test sets. Although the validation data is not used to adjust weights it is used to decide if training should be brought to an early halt and thereby reduce over fitting. Over fitting will tend to occur when the error for the validation set begins to rise, and is therefore monitored with this technique.

Regularization is applied in order to bias the learning algorithm away from over-fitting. A term is added to the error reduction so that there is a trade off between smoothing and error reduction (i.e. so that small changes in input do not cause large changes in output). Care must be taken regarding it's applicability for this project because a sharp and crisp output is desired so that there is a sharp transition between intent states and non-intent states. One feature of the Bayesian regularisation algorithm is that it provides a measure of how many network parameters (weights and biases) are being effectively used by the network. It can therefore act to almost prune a network. For this algorithm to work best, the inputs and outputs should be normalised as explained earlier with data preparation.

Ultimately, the search must include random attempts in which new connections are examined; it is akin to avoiding being caught in a local minimum. This problem is two fold, in that a choice must be made of not only the topology of the network in terms of number of neurones and layers but also the choice of input variables.

Input/ Target Data Organisation:

Whatever architecture was finally examined, it had to be presented by a training set of input variables. If the input variables form a vector size R, then each sample of training data (1 … Q) could be presented in batch form as a matrix **P** size R by Q. Similarly for network training and testing, a set of corresponding target outputs defined by matrix **T** sized according to the number of output neurones (1 in this case) by Q was required. The samples that represented a power event were assigned a target value of one (i.e. maximum probability of this frame being the particular event) and non events a value of zero. These events were pre-selected by eye from the plot of the knee power patterns as discussed above [section 8.3.1]. Any trials where the events/intent states could not be determined were not used for training. It was initially assumed that all non-event frames within a trial had a target value of zero. However, because of the imprecise nature of defining events, there was some ambiguity with the target value of frames directly adjacent to the event frame; there was some probability of such frames to be the actual event rather than the one selected by eye from the knee power. Although some Gaussian type function centred about the selected event could have been used to estimate the probability of the target output, it was thought better to avoid presenting unknown information by simply omitting adjacent frames from the training set of data.

Given this formulation for the input and target matrices, the available data was split into training and test sets, and if "Early stopping" was used, into validation sets. This split could happen on several levels according to subjects, sessions, trials, and even frames of data. It was decided to have the split based primarily on a trial-by-trial basis, so that only every other trial within a session would be used for training. This enabled within session testing of the output data. It was then a design choice as to which sessions were to be used for training, although there were some restrictions applied. For example, it was desirable to have test data from subjects who had not been used for training purposes at all. In practice, a limit of data from three different subjects was imposed to give sufficient uncharted test data.

8.3.3 TESTING OF NETWORKS

The purpose of testing is both to see the capability of a particular network, and to gather information to enable improvements with subsequent network designs. Tests should comprise data that the network has never experienced, which for generalisation purposes should include data from different sessions and subjects that were not used for training purposes. In addition, trials within a training session that were not selected for training purposes were also examined to determine how well the network learnt the training set of data. If good results are not found for these within trials then that network is unlikely to yield adequate solutions.

It was anticipated that the whole strategy of designing/finding a useful network to detect each intent state would involve a considerable amount of iteration – and generate considerable data to examine. It was therefore important to quickly assess the performance and capability of each network that was designed so that some clue with regard to changing the design parameters (e.g. input variables, number of first layer neurones, training algorithm etc.) for the next iteration could be gleaned. Therefore, the analysis of the test results was automated. This meant for each network that was designed, the simulated outputs from the test set of data had to be examined for performance as follows. In terms of error, three issues were addressed: 1) consistency of results 2) resolution and 3) accuracy.

8.3.3.1 ANALYSING NETWORK PERFORMANCE

As an addendum, error can occur in two dimensions, in the temporal and in the output amplitude field. For example, if a high-level output, with a value of 0.99, occurred one frame before the designated intent frame, then one might view this mainly as a temporal error. This is in contrast to the conventional error reading (for example used to train the networks), which would judge the intent frame to be completely wrong, and the high value frame to have an error of 0.99. In addition, as long as one can reliably distinguish high level and low-level amplitude values then amplitude errors are not so critical. For example, a low level threshold may be used to distinguish all non-intent states as long as it is clear that the networks are not producing high values of this order. Therefore, the temporal distributions of high and low output levels are of greatest interest.

Ideally, the simulated output of the neural networks should give values of one or zero in response to the universal set of input data depending on whether the target is an intent state or non-intent. It was however recognised that real data will contain some uncertainties, both with the pre-identification of intent states and the simulated output using test data. As discussed in the training section [section 8.3.2], frames of data that were immediately adjacent to target frames were not used for training; this was in a bid to avoid presenting erroneous target values to the network because of a certain amount of leeway in the temporal accuracy of the identification of power phases. With test data however, it was important to simulate these periods where a Gaussian type response may be expected that is centred about the intent state and diminishes practically to zero within a few frames either side of the pre-determined intent state (in practice this was taken to be five frames). This formed regions throughout the data where an intent state might be expected. With such points in mind, measures of network performance were attempted.

1) **Consistency:** If the network output, in response to input from data outside the intent Gaussian probability envelope was non-zero (anything above a 0.2 threshold), then this was deemed unacceptable. Code was written to measure the frequency of such high values in non-intent regions. Conversely, if all output values were low (anything below 0.8 threshold) during periods encompassed by the Gaussian envelope, then this was also undesirable. The proportion of such circumstances taken over all such periods was used as a measure of the high output consistency. Two separate consistency measures were therefore made, one for the zero valued target range (non-intent consistency) and the other for the high valued target range (intent consistency) - details are given below:

> **Non-intent consistency:** A non-intent state, in theory, should reliably occur at all times except during the intent regions. Therefore, these could be presented to the network at any time so that it seemed reasonable to simply measure the proportion of erroneous high output values out of all non-intent test data. It did not seem necessary to have a trial by trial analysis, although a session by session analysis could give information as to how well networks learnt for different subjects. Therefore, values above the 0.2 thresholds were considered erroneous and totted up. A good network should therefore not

produce any high values in these regions and give a zero percentage non-intent consistency result.

Intent consistency: It was not a good measure to simply take the proportion of high values out of all of the frames designated by intent regions because only one high value was needed within each region for the network to be successful. Therefore, high intent state consistency had to be measured from intent region to intent region (which should be one region per walking trial). A count was thus made of the intent regions with at least one high output value (taken as anything above 0.8) within the region. A good network should therefore yield a 100% high consistency score. The quality of the high value response within regions was assessed by separate measures.

2) Resolution/Distribution: Before the accuracy of the high outputs could be determined, it was necessary to know the distribution of the high-level output values within each intent region. If only 1 high valued output frame exists within an intent region, then the resolution is optimum and the accuracy can be measured in temporal terms by the number of frames from the high level target. Otherwise, the validity of the data depends upon the sparseness of the high level values. For example, a sparse distribution of high values within the intent region will imply a lack of reliability and jitteriness, whereas several adjacent frames with high outputs would indicate that the networks have learnt to generalise the phase into a narrow time bandwidth. The latter solution, although less precise, may be of benefit for some circuitry where a band of several frames would be more likely to be picked up and give a robust triggering event than a sharp single high value frame. The band should however not be so large as to cause confusion with other phases. Therefore, code was written which examined each high target region and measured:

- the temporal **width** between the first high valued output and the last
- and the **sparseness** of the high values within the width, i.e. the proportion of high values per width size.

For example, a high target region with three high level outputs all next to each other will have a width of 3, and sparseness of 100%, whereas if the middle frame has a low output value then the sparseness becomes 67%. Ideally, a sparseness of 100% for each intent region was sought with a small width of up to three frames.

3) Accuracy: Finally, to get a complete picture of each network output, it was desirable to know how close, in temporal terms (frames), the high level values (within the intent regions) were to the original target event. The ideal result would be that a high output would correspond to the particular frame relating to the power event. Otherwise, the accuracy was determined as the distance in frames of the high output from the target intent state. However, if the width of the response was greater than one, then several measures were useful to assess the temporal accuracy. The first was simply to take the mean temporal position of all the high responses to estimate the locus of the response and hence the distance from the target. Alternatively, because it is possible to have positive and negative edge detectors, the distance from the target of the first and last high value was also given.

Automation of Network Analysis; Therefore, for each session of test data that was presented to a particular network structure, Matlab code was written which would automatically compare the simulated outputs with the target values for the different intent states. To facilitate easy examination, the above performance measures taken were directly written to an Excel spread sheet using Matlab. Since each of the nine intent states was recognised by a different network with different weights (although using the same structures), a separate worksheet of results was produced for each phase. Finally, a summary worksheet was also written automatically, which allowed networks that performed well in detecting their respective intent phase to be singled out quickly and easily. The summary result for the networks was formulated from the sum of the intent and (100% - non intent) consistency measures made for each test session and each phase. Since the networks should perform well for both intent and non-intent regions, high scores of up to 200% indicated networks that were working very well over all of the test data, and were worthy of closer inspection e.g. using the remaining performance measures and ultimately graphical checking of data. To allow quick assessment, the consistency summary result was categorised so that any network with a score above 190% was considered “very good”, a score below 160 % “bad” and in between as “good”. In this manner, it was possible to count, for each network, the number of sessions that produced “very good” and "good" consistency results. Ideally, if a network has generalised well, every test session from all subjects should give "very good" results, and it was a matter of selecting networks with such high "very good" counts.

Improving Networks: The above performance measures provide information as to the effectiveness of a network structure with each intent phase. This was useful to see whether to persevere with such a structure regarding input selection, first layer size, and indeed particular training algorithms. In addition, it was possible to see which subjects had gait patterns that were more representative of other subjects and which had uniquely distinctive patterns. In this sense it was a good measure of the extent of generalisation where for example, if only test trials from a session that was also used for training purposes yielded good results then very poor generalisation was implied and the data had been over fitted. Therefore, it may be worth trying to increase generalisation by reducing the number of neurones in the first layer or the number of input variables, or by using early stopping, Bayesian regularisation, using principal component analysis etc. [section 8.3.2]. Alternatively, more training sessions may be used for adapting the network weights, although the restriction of using no more than three different subjects was given regard.

If test trials from a training session gave a poor performance, then the problem was under fit or indeed there was no solution readily extractable using that particular set of input variables.

Despite some clues from the performance measures of each network, there was no easy way to identify which inputs had been most useful to the neural networks for solving their particular task. This included the use of Principal Component Analysis (PCA), which, although is often used to pre-process data, still requires the same number of original input variables. For example, principal component analysis allowed one to organise the data so that the bulk of the informational content of the data is condensed into the first few principal components. One can measure how much of the variance is explained by each component so that it was possible to reduce the data by removing what is considered redundant information e.g. any components which explained less than 10% of the data. With this method of reduction however, all of the original input variables are still required before being transformed into the principal components and applied to the network.

Although this technique could not be used to really show which input variables were directly of greatest use for the networks, it was beneficial to help visualise the data. One wants to determine how uniquely the target states can be defined in some input space; a space was ideally required in which the intent states could easily be extracted. It was realised that given the multidimensionality of the input space it would be very difficult to visualise the problem for dimensions in excess of three. Therefore, as a basic guide, it was thought that 3D plots of the first three principal components of the data would be of use. In this way, the bulk of the information should be present within the diagram. By using colour codes to discriminate intent states from non-intent states, a visual idea of how well intent states cluster was developed and whether the data was trainable. See for example the first three components derived from the PCA performed on all 48 possible input variables for session Ba [Figure 8-32]. Even with such a large number of variables, this method is useful because the first three components explained almost 55% of the data. It can be seen how the data is spread out mostly in the PC1/PC2 plane, although phase KP5 protrudes more in the PC3 direction. Close inspection by rotation and zooming of the figure showed that most of the intent states occupy distinctive regions although of varying sizes. It is interesting to note that this distinction for KP5 is only visible when viewed in the PC3/PC1 plane; it highlights the importance of using a sufficient number of inputs for the problem. In this sense, more inputs are also needed because it could also be seen that some non-intent states do transgress into the intent state regions (although in this the picture is made slightly worse because directly adjacent frames to the intent state have not been omitted or colour coded). This scenario reflects the compromise required in using enough inputs to get sufficient partitioning of intent states without having too many so that the network is not generalised enough for new data sets.

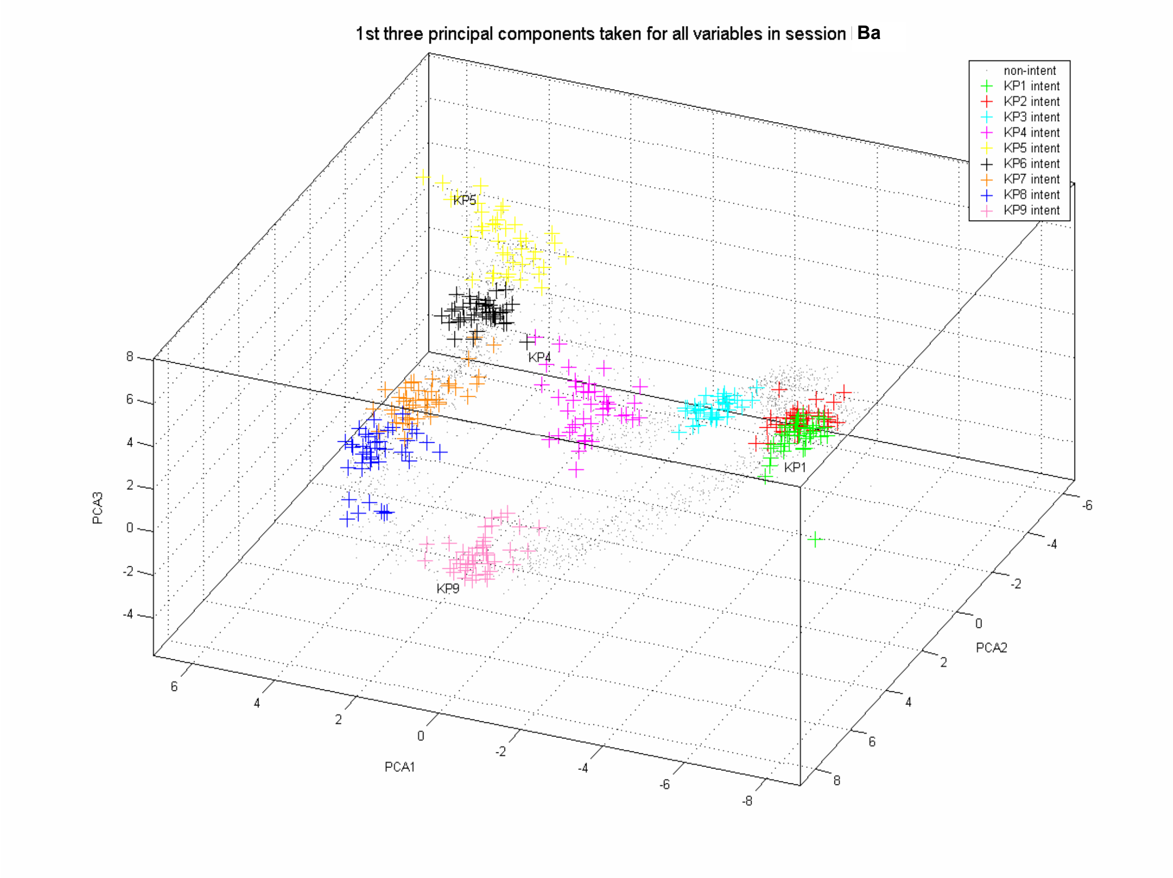


Figure 8-32 Most intent states can be seen to occupy distinctive regions within the principal component space. However, the orange KP7 region seems to be encroached by the blue KP8 states. This indicates that more inputs are required than three. These three Principal components explain about 50% of the variance in data. This is not bad considering there were originally 48 inputs. The green markers show states that are up to four frames either side of the intent state. Even these frames represent quite a narrow time window; it can be seen how quickly they disperse away from the target states.

It was therefore a considerable problem selecting the appropriate set of input variables for dimensions greater than three because one couldn't judge which inputs help the network more than others and which did not. This meant that a slow process of trial and error had to be undertaken in order to find the best combination of inputs.

Since it was not easy to automate some process for choosing the effectiveness of inputs, an educated guessing approach was employed. To increase the chances of success, input variables that were well behaved were first employed. This was in terms of choosing variables that were not too noisy or had too high a frequency content in which there are too many roots within a gait cycle that cannot easily be discriminated according to amplitude. Finally, some regard had to be given to the fundamentals governing the nature of this project, in that two control systems (artificial and natural) are involved, therefore feed-forward information will primarily be supplied via the stump motion, whereas some feedback must be provided to the movement of the artificial prosthetic component below the stump.

8.3.3.2 SUMMARY OF RESPONSES FROM NETWORKS

Example Network Performance: To illustrate the process of testing networks, some of the results from a well-behaved network are presented [Figure 8-33]. This example shows the simulated output of the network produced in response to input test data from session "Cb" (corresponding to a subject whose data was not used for training). Since this network was designed to detect phase KP2, the corresponding high target values for this phase have been superimposed as determined previously from analysis of the power patterns in this session. To illustrate the uncertainty of this target value, Gaussian type envelopes (in purple) have also been added that are centred about the intent state of each trial. These envelopes exemplify the probability of particular frames being an intent state, thereby allowing intent regions to be formed. Even though only one session has been shown, it can be seen from the top figure how there was too much data to enable a full graphical examination of the results from every net as tested with all available sessions. In fact, over 74 different network structures were trained to recognise each of the nine intent states, giving over 666 individual networks to test.

The zoomed portion in the figure enables one to see how effective this example net was for at least the couple of trials within the window. Of first consideration was that no high output values (in green) were present within the non-intent regions. This was to reduce the chances of inadvertent intent signals occurring at incorrect times. Secondly, it can be seen that there are easily discernible high output values within each intent region which occupy a solid region about three frames wide just before the intent state.

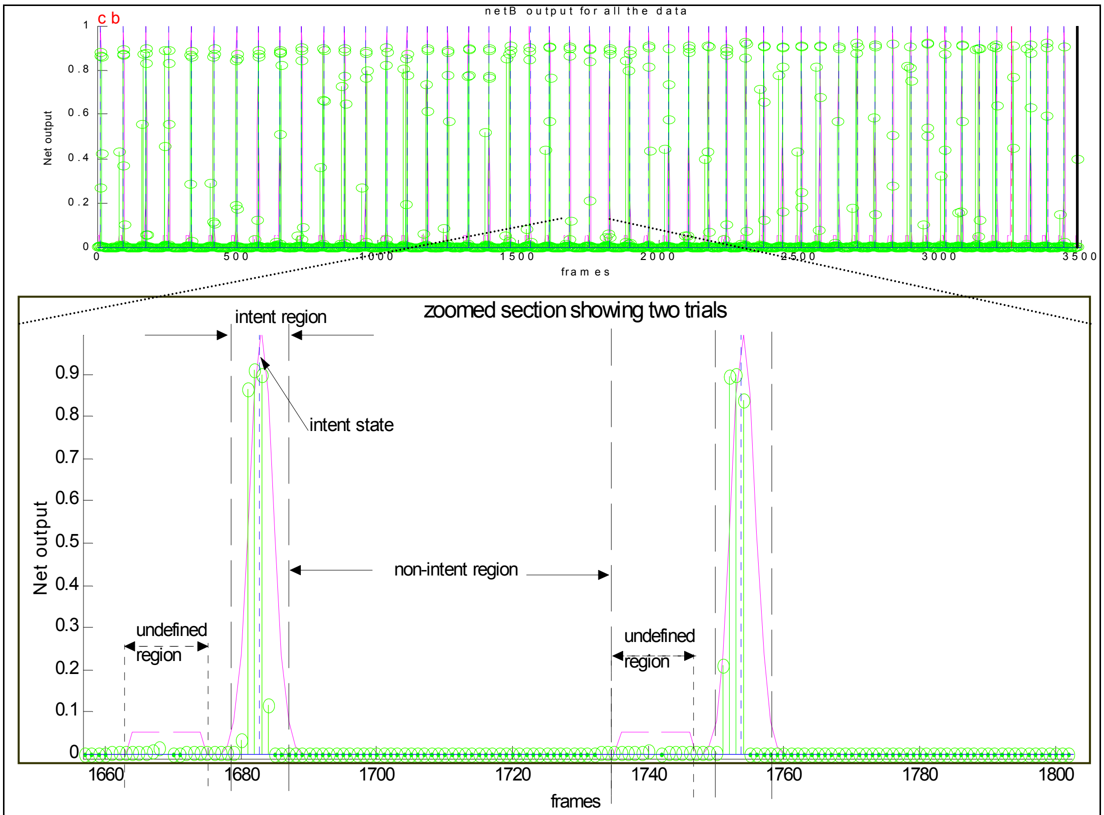


Figure 8-33; Simulated outputs (green) from a KP2 network (B4zd15) in response to trials of data from session Cb; the zoomed view focuses on two trials for more clarity. Each trial has been split into intent and non-intent regions where the probability of intent is designated by the purple line whereby Gaussian envelopes mark out the intent regions. In this example, high valued outputs are produced with a bandwidth of about 3 frames just before the target intent state; zero values are produced at the other times indicating that this network is performing quite well.

However, rather than graphically examining all this data, a fuller picture of the response of the networks to test data was given by the performance values (as discussed previously [section 8.3.3.1]. As an example, the tabulated results from the network shown above are presented [Table 8-M]. In this table, all of the test sessions have been included and not just "Cb." However looking at "Cb" first, both the consistency results (0% non-intent and 100% intent) confirm that the results from the above zoomed portion were not just localised but occurred across all trials. There was an average high output width of 2.74 frames and an average sparseness of 100% indicating, as in the diagram, that all frames within the width had a high value. The accuracy measure, in which the first value is at 5.41 and the last at 6.28, shows that on average, the high values occurred just on and before the intent state (located at a value of 6 frames within the intent region that is 11 frames wide). Such a response in practice may utilise negative edge detection circuits as a trigger point for the state. The overall performance of this particular network as tested with session "Cb" was summarised with a perfect score of 200%, signifying that all high output values occurred only within the intent regions of the test data.

		Consistency Non-intent			Consistency Intent Region			Distribution		Accuracy		
test sessions	summary 200=good	no. high hits	size region/no. trials	pcent hits (0 = good)	no. high hits	size region/no. trials	hits (100 good)	width (frames)	sparsity (100=solid)	mean pos	first high	last high
Ba	192.7	0.0	1944	0.0	38.0	41.0	92.7	1.3	92.7	4.9	4.7	5.1
Bb	200.0	0.0	1441	0.0	30.0	30.0	100.0	3.6	100.0	6.4	5.1	7.6
Bd	200.0	0.0	1066	0.0	22.0	22.0	100.0	3.5	100.0	6.1	4.9	7.4
Ca	195.5	23.0	1542	1.5	32.0	33.0	97.0	3.6	91.0	6.5	5.1	7.7
Cb	200.0	0.0	2393	0.0	50.0	50.0	100.0	2.7	100.0	5.4	4.5	6.3
Fb	199.4	6.0	945	0.6	19.0	19.0	100.0	4.8	98.7	6.6	4.6	8.5
Fc	199.6	5.0	1246	0.4	23.0	23.0	100.0	4.3	99.3	5.9	4.2	7.5
Fe	200.0	1.0	2347	0.0	40.0	40.0	100.0	4.4	99.5	6.4	4.7	8.1
Ea	199.9	2.0	2596	0.1	47.0	47.0	100.0	5.0	100.0	6.5	4.5	8.6
Ed	199.8	2.0	1281	0.2	22.0	22.0	100.0	3.9	94.6	5.9	4.5	7.4
Ee	197.2	14.0	2559	0.5	44.0	45.0	97.8	5.0	95.2	5.2	3.3	7.3
Ha	200.0	0.0	1058	0.0	21.0	21.0	100.0	5.0	100.0	6.4	4.3	8.4
Hc	195.2	18.0	2533	0.7	47.0	49.0	95.9	4.0	93.8	6.0	4.4	7.5

Table 8-M: KP2 phase Excel worksheet showing the performance measures of network B4zd15 that was trained with the Levenberg Marquardt algorithm ,and used logsig transfer functions. The accuracy measure assumes that the width of the intent region is 11 frames so that the central position with a target value of one would be at position 6. Note that the sessions highlighted in grey represent those that provided training data. The summary column reflects the overall consistency for the session and was taken as the sum of (100 – non-intent hits) and intent hits.

Similarly, the response to the other test sessions was also impressive where the lowest summary measure was only 192% for session “Ba”. It was however surprising that this particular session had the worst reading because a portion of data from this subject had been used for training (session Bd). It would have been expected that testing with intra-subject data would have produced better results because this data would be more likely to be similar to the training data than inter-subject data. Since the differences were however so small, it was thought that noisy data sets could easily account for them.

Summarised Network Results: The consistency performance summary therefore allowed networks that were likely to be successful to be quickly spotted. As described earlier, the performance measures which gave “V Good” and “Good” results were totted up so that such networks could easily be spotted. An Excel file containing the full summary worksheet is included showing the summarised consistency values that were made for each test session [Table G-C]. For visual clarity, only the number of test sessions that produced “V Good” and “Good” consistency measures is tabulated below [Table 8-N]; when the tally of both of these categories reached specified thresholds, a comment reflecting the overall network performance was also made in the yellow rows. Networks that resulted in an overall performance comment of “GREAT” were primarily sought, i.e. in which at least 12 test sessions produced “V Good” or “Good” consistency scores.

net name		KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
1ba5	Good Count	0	4	0	1	0	0	2	0	1
	V Good count	0	2	1	1	0	1	3	1	7
	summary									OK
1ba10	Good Count	4	6	0	0	0	0	0	0	2
	V Good count	8	7	0	0	0	0	0	0	0
	summary	V GOOD	GREAT							
1ba5tan	Good Count	0	0	0	0	0	0	0	0	0
	V Good count	0	0	0	0	0	0	0	0	0
	summary									
1cb5	Good Count	0	0	0	6	2	6	2	0	2
	V Good count	0	0	0	5	6	0	2	0	6
	summary				V GOOD	OK				OK
1cb10	Good Count	0	4	0	5	0	4	3	0	2
	V Good count	0	8	0	6	0	0	0	2	8
	summary		V GOOD		V GOOD					GOOD
1ed5	Good Count	2	7	6	2	2	2	2	2	2
	V Good count	4	5	3	4	4	4	4	4	4
	summary		V GOOD	GOOD						
1ed10	Good Count	0	0	0	0	0	0	0	0	0
	V Good count	0	0	0	0	0	0	0	0	0
	summary									
1fe40	Good Count	0	0	0	0	0	0	0	0	0
	V Good count	0	0	0	0	0	0	0	0	0
	summary									
1hc10	Good Count	0	0	0	0	0	0	0	0	0
	V Good count	0	0	0	0	0	0	0	0	0
	summary									
1cb20	Good Count	0	0	0	0	0	0	0	0	0
	V Good count	0	0	0	0	0	0	0	0	0
	summary									
1lc10	Good Count	2	4	0	0	0	0	0	0	1
	V Good count	7	5	0	0	0	0	0	0	1
	summary	GOOD	GOOD							
1ba10tan	Good Count	2	0	0	1	0	1	0	2	1
	V Good count	1	1	1	1	0	0	0	0	6
	summary									OK
4ba25tan	Good Count	0	1	4	1	0	2	0	0	0
	V Good count	0	0	1	0	0	0	0	0	0
	summary									
3za15	Good Count	6	7	1	1	0	0	0	0	3
	V Good count	6	4	0	0	0	0	0	0	0
	summary	V GOOD	V GOOD							
3za18tan	Good Count	2	0	0	0	0	0	0	4	10
	V Good count	0	0	0	0	0	0	0	0	0
	summary									GOOD
3za12tan	Good Count	1	4	3	8	5	0	0	2	9
	V Good count	0	2	1	2	0	0	0	0	0
	summary				GOOD					GOOD
4zd15tan	Good Count	2	3	3	1	3	0	0	0	4
	V Good count	9	5	9	0	0	0	0	0	7
	summary	V GOOD	OK	V GOOD						V GOOD
5ze15tan	Good Count	3	4	3	6	0	1	4	1	2
	V Good count	3	3	1	6	2	1	0	0	5
	summary		OK		V GOOD					OK
5ze10tan	Good Count	0	2	7	7	4	1	0	1	2
	V Good count	0	0	2	5	2	0	0	1	4
	summary			GOOD	V GOOD					
4zd15	Good Count	3	0	5	0	5	0	6	0	0
	V Good count	8	13	6	0	0	0	0	0	1
	summary	V GOOD	GREAT	V GOOD						
5ze10	Good Count	3	1	2	6	3	3	0	4	5
	V Good count	2	3	1	4	1	1	0	2	7
	summary				GOOD					V GOOD
5ze15tan	Good Count	3	4	3	6	0	1	4	1	2
	V Good count	3	3	1	6	2	1	0	0	5
	summary		OK		V GOOD					OK
6zf15	Good Count	6	6	3	4	5	2	0	4	7
	V Good count	1	3	8	9	3	0	0	3	6
	summary	OK	GOOD	V GOOD	GREAT	OK			OK	GREAT
7zg80tan	Good Count	1	0	0	3	0	0	2	1	1
	V Good count	0	0	0	0	0	0	0	2	12
	summary									GREAT
7zg30tan	Good Count	1	0	2	0	0	0	2	1	3
	V Good count	1	0	1	0	0	0	0	3	8
	summary									V GOOD
6zh25tan	Good Count	2	6	0	2	5	7	8	2	1
	V Good count	0	4	0	5	0	0	0	6	7
	summary		GOOD		OK		OK	OK	OK	OK
6zh15tan	Good Count	0	3	1	4	3	3	7	4	2
	V Good count	0	3	7	7	0	0	2	4	9
	summary			OK	V GOOD			GOOD	OK	V GOOD
8zh15tan	Good Count	5	1	0	5	3	7	5	0	2
	V Good count	0	0	0	4	0	3	0	0	5
	summary				GOOD		GOOD			OK
8zh25tan	Good Count	4	2	0	6	3	5	0	5	2
	V Good count	4	10	0	5	1	0	0	2	10
	summary	OK	V GOOD		V GOOD				OK	V GOOD
8zf20tanx	Good Count	0	4	8	2	6	3	3	0	1
	V Good count	0	3	3	3	0	0	0	0	12
	summary		OK	V GOOD						GREAT
9zj35tan	Good Count	0	2	0	0	3	0	0	0	0
	V Good count	0	2	0	0	2	0	0	0	0
	summary									
9zk15tan	Good Count	3	0	0	4	3	0	2	3	4
	V Good count	5	0	0	5	0	0	0	1	8
	summary	OK			GOOD					V GOOD
8zf20tanx	Good Count	5	4	8	2	6	3	3	0	1
	V Good count	4	3	3	3	0	0	0	0	12
	summary	GOOD	OK	V GOOD						GREAT
11zl17tan	Good Count	4	0	4	3	2	3	0	0	0
	V Good count	1	0	3	5	2	0	0	0	0
	summary			OK	OK					
11zd17tan	Good Count	2	5	9	4	3	3	1	1	5
	V Good count	0	7	1	3	1	0	0	1	4
	summary		V GOOD	GOOD	OK					GOOD
11zd20tan	Good Count	0	4	2	0	2	2	0	0	5
	V Good count	0	5	5	0	0	0	0	0	7
	summary		GOOD	OK						V GOOD

Cont….

net name		KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
11zf15tan	Good Count	0	6	3	7	4	0	0	7	1
	V Good count	0	3	4	1	4	0	0	5	0
	summary		GOOD	OK	OK	OK			V GOOD	
11zd8	Good Count	3	0	6	5	0	2	1	2	3
	V Good count	1	0	4	4	0	0	1	1	7
	summary			GOOD	GOOD					GOOD
3zm15	Good Count	3	0	0	5	3	0	0	0	1
	V Good count	2	0	0	7	1	0	0	0	4
	summary				V GOOD					
12zd15	Good Count	0	4	7	0	0	2	0	6	2
	V Good count	0	2	3	0	0	0	0	0	2
	summary			GOOD						
12zd12	Good Count	5	6	2	5	0	4	0	7	3
	V Good count	4	2	3	6	0	0	0	0	6
	summary	GOOD	OK		V GOOD				OK	GOOD
13zd15	Good Count	5	4	6	0	0	1	1	7	1
	V Good count	1	1	2	0	0	0	0	2	10
	summary			OK					GOOD	V GOOD
14zm15	Good Count	0	6	7	0	0	0	0	1	4
	V Good count	0	1	4	0	0	0	0	0	7
	summary		OK	V GOOD						V GOOD
14zd10	Good Count	2	6	1	6	1	1	4	0	3
	V Good count	0	1	1	1	0	0	0	0	6
	summary		OK		OK					GOOD
14zd25	Good Count	1	3	0	3	3	1	2	1	6
	V Good count	0	0	0	0	0	0	0	0	4
	summary									GOOD
16za15	Good Count	2	4	4	5	4	4	0	4	0
	V Good count	1	5	1	2	0	0	0	1	0
	summary		GOOD		OK					
16za10	Good Count	2	0	0	6	2	4	0	3	4
	V Good count	1	0	0	1	0	4	0	0	4
	summary				OK		OK			OK
17zm15	Good Count	5	6	2	3	2	5	2	0	0
	V Good count	4	3	0	6	1	0	0	0	0
	summary	GOOD	GOOD		GOOD					
17zm12	Good Count	0	6	4	4	0	0	1	0	4
	V Good count	0	1	0	5	0	0	0	0	4
	summary		OK		GOOD					OK
18zd15	Good Count	4	4	3	0	1	1	4	3	7
	V Good count	2	1	0	0	0	0	0	0	3
	summary									GOOD
19zn20	Good Count	8	0	2	5	4	1	0	1	6
	V Good count	4	0	3	8	0	0	0	2	6
	summary	V GOOD			GREAT					V GOOD
19zn25	Good Count	4	5	0	5	6	0	4	3	2
	V Good count	3	1	0	6	5	0	0	2	8
	summary	OK			V GOOD	V GOOD				GOOD
6zn15	Good Count	4	0	6	2	0	3	5	2	0
	V Good count	4	0	0	8	0	2	0	0	0
	summary	OK			GOOD					
6zm20tan	Good Count	0	0	5	3	0	3	0	2	4
	V Good count	0	0	0	10	0	0	0	2	9
	summary				GREAT					GREAT

Single training session		KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
6ba15	Good Count	0	3	0	8	0	0	0	5	4
	V Good count	0	5	0	4	0	0	0	1	7
	summary		OK		V GOOD					V GOOD
6bb15	Good Count	0	0	2	1	2	0	0	0	2
	V Good count	0	0	4	0	2	0	0	0	1
	summary									
6ld15	Good Count	0	1	1	3	1	0	0	0	5
	V Good count	0	0	0	1	0	0	0	0	3
	summary									OK
6ca15	Good Count	0	0	0	11	3	2	0	1	2
	V Good count	0	0	0	1	0	0	0	0	10
	summary				V GOOD					V GOOD
6cb15	Good Count	0	4	3	11	0	0	0	0	3
	V Good count	0	0	0	1	0	0	0	0	5
	summary				V GOOD					OK
6ea15	Good Count	3	4	0	5	1	1	0	3	1
	V Good count	0	4	0	8	0	1	0	8	1
	summary		OK		GREAT				V GOOD	
6ed15	Good Count	4	0	3	4	6	5	0	1	0
	V Good count	2	0	1	2	0	0	0	0	1
	summary									
6ee15	Good Count	2	7	0	4	3	0	5	2	4
	V Good count	1	0	0	7	0	0	3	4	5
	summary		OK		V GOOD			OK		GOOD
6fb15	Good Count	3	2	1	0	0	1	0	5	6
	V Good count	0	8	0	0	0	0	0	3	2
	summary		GOOD						OK	OK
6fc15	Good Count	1	4	3	0	1	0	0	0	0
	V Good count	0	5	2	0	0	0	0	0	0
	summary		GOOD							
6fe15	Good Count	4	4	3	0	0	0	3	4	0
	V Good count	0	3	0	0	0	0	0	1	0
	summary		OK							
6ha15	Good Count	0	1	0	4	9	2	0	0	2
	V Good count	0	0	0	4	4	0	0	0	7
	summary				OK	GREAT				GOOD
6hc15	Good Count	0	2	3	4	0	0	0	4	6
	V Good count	0	1	1	8	0	0	0	3	4
	summary				V GOOD				OK	GOOD
6zm15	Good Count	3	3	5	4	6	3	3	5	3
	V Good count	3	1	1	8	5	0	0	3	9
	summary				V GOOD	V GOOD			OK	V GOOD
6zm20	Good Count	4	0	2	3	0	4	0	4	0
	V Good count	2	0	0	9	0	1	0	1	0
	summary				V GOOD					
6zm12	Good Count	3	2	0	3	6	0	0	3	1
	V Good count	4	1	0	9	6	0	0	2	12
	summary	OK			V GOOD	V GOOD				GREAT

Table 8-N; Tabulation of each network performance with respect to the nine intent states. The number of test sessions that had consistency scores over190% contributed to the "V Good" count and the number over 160% to the "Good" count. If the sum of the "V Good" and "Good" counts exceeds 12 sessions then the comment "GREAT" is written to summarise the overall performance of the network; if more than 10 then "V GOOD", more than 8 gives "GOOD" and more than 6 gives "OK".

The naming convention for the networks (e.g. A1ba10tan) that were designed was:

- First, a letter prefix representing the intent state that the network has been trained to recognise, where "A" represents KP1, B – KP2, … and I gives KP9.
- Second, an input group number coding for particular input sets [Table G-A].
- Third, the training session group name, which took its name from the session code when only one session was used, otherwise another two-letter prefix was made, usually starting with "z" see Table G-C for codes.
- Fourth, the number of neurones in the first layer.
- Fifth, the function used by the neurones of the first layer, where "tan" signifies a tan sigmoid function, and absence of a label - the log sigmoid function.

Table 8-N therefore shows the performance summary of all the networks that were trained and tested and below is a description of some of the strategies that were attempted to iterate through all these networks.

Effect of training sessions: After trying a few random network structures to get a feel for the problem, it was desired to see what effect different training sessions had on network performance. Thereby, to enable fair comparisons, the network structure type was made invariant in which its input group, number of first layer neurones, first layer transfer function, and training algorithm were all kept constant. Each available session of data was then used independently to train separate sets of networks to recognise each intent state. The chosen network structure for this examination used input group 6, a log sigmoid function for the first layer, which consisted of 15 neurones. This structure was selected because previous random attempts had already yielded a few reasonable results with this type of structure. The number of first layer neurones had been confined to 15 because it had already been observed from the few network examples that first layers of 10-20 seemed to give a compromise between over fitting and generalisation. The results of these networks trained with individual sessions are tabulated at the bottom of Table 8-N, and graphically shown in Figure 8-34. The figure shows for each intent state and each

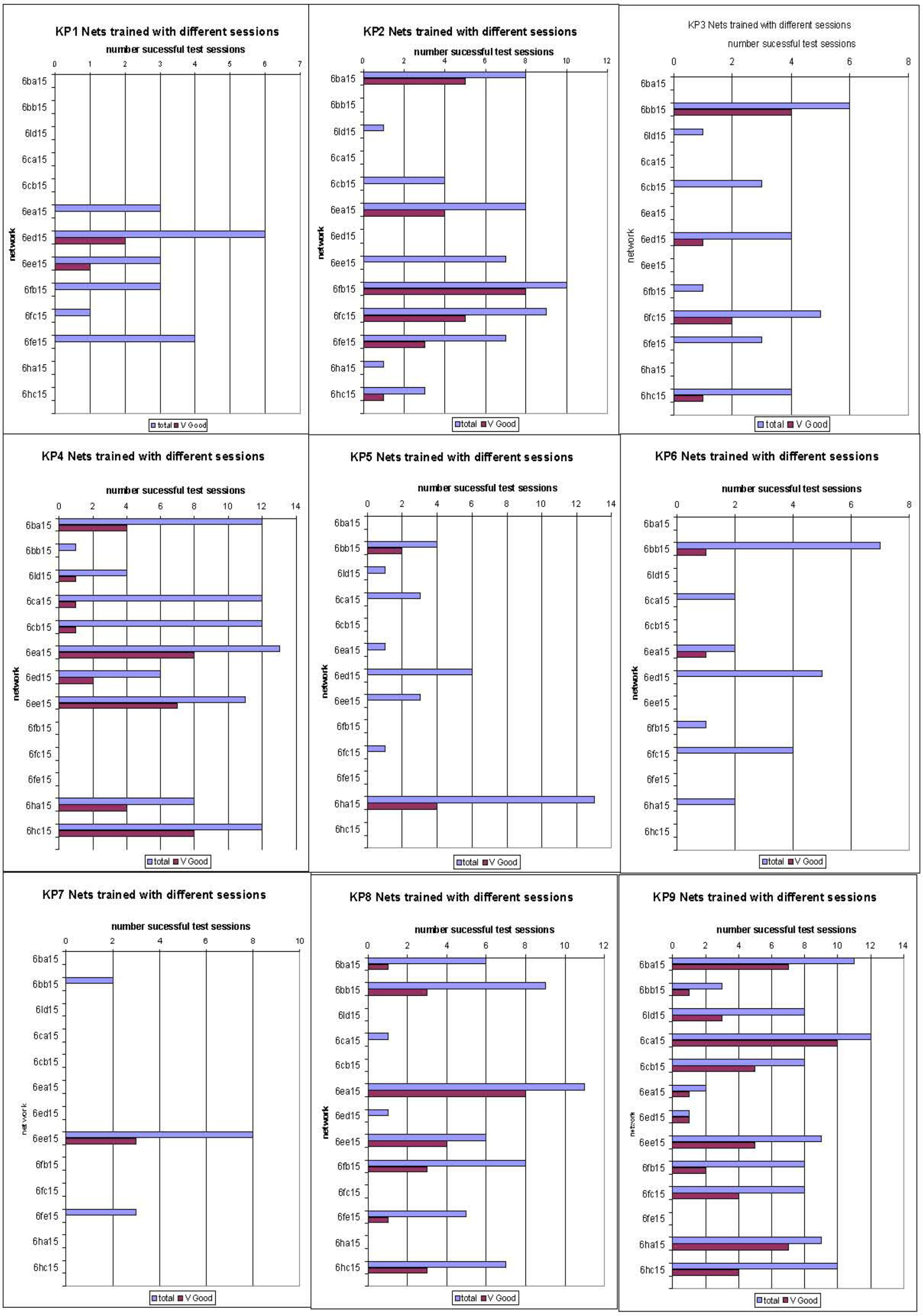


Figure 8-34; Charts for each intent state comparing performance of several networks with the same structure but trained with different individual sessions. The purple bars give the total number of test sessions that produced "V Good" or "Good" consistency results; the mauve bar gives the number of test sessions that only produced "V Good" results.

training session, how many test sessions yielded "V Good" or "Good" performance values, and shows what proportion of this total value was "V Good."

From a glance at high valued bars, it appeared that this particular network structure was not uniformly effective for all intent states; for instance, phases KP4 and 9 appeared to be most readily detectable by a greater number of networks whereas KP5 and 8 yielded fewer good results. For these intent phases where most networks gave good results, it was of interest to note which training sessions resulted in a poor performance. Training sessions "Fb", "Fc" and "Fe" from subject "F" seemed to give an overall bad performance although the worst session was "Ed". One interpretation of the poor results caused by using sessions from subject "F" for training is that his data for these phases were unique as compared to data from other sessions, hence the network was not able to generalise; the subject may be performing the particular phase in an unusual manner. Therefore, to avoid training networks that learn features unique to this subject, it was thought wise not to include sessions from this subject for training. The hope would be that training with data from a subject with a more general pattern should be less focused and still be able to pick up subject F's intent. If this does not occur then one may be forced to use multiple training sessions that include a session from this subject.

The phases that yielded bad performance values, irrespective of the training sessions, were KP1, KP3, KP6, and KP7. The reasons for this could also be in a similar vein to that just discussed in which individual sessions are too unique, however other causes were also considered. For example, because the performance of **all** the networks for phase KP7 was poor it suggested that the problem was intractable at least by this particular network structure. For these phases, little can thus be gleaned about differences between individual training sessions and other network structures should possibly be examined.

Good training sessions:

Therefore, considering again the more successful phases, collation of the best networks within these phases provided some clues as to which sessions were more generalist. Training with session "Ea" gave the best performance for both KP4 and KP8, whereas session "Ba" provided the second best results for KP4 and KP9, and training session "Hc" yielded third best for KP4, 8 and 9. Although session "Ca" gave the best results for KP9, it was less successful for the other phases. Therefore, it was anticipated that training sessions "Ea", "Ba", and "Hc" would be good candidates for a group where they should provide more generalised data for at least these three phases. In fact, this was initially partially borne out by previous results of networks using group "zf" for training, which consisted of sessions "Ba", "Ea", and "Ha". To test this idea, network 6zm15 was trained, using sessions "Ba", "Ea", and "Hc". This yielded results that were about as good as the best result with a single training session [bottom of Table 8-N]. It should however be remembered that this network structure was not necessarily the optimum and as a check the number of neurones in the first layer was adjusted. The number was first increased to 20, which gave a poorer performance especially for KP9 indicating that over fitting was occurring. Reduction to 12 neurones gave the most promising results where KP1, KP5, and KP9 achieved their highest performance with respect to all of the single session networks.

Choosing Network Structure:

As can be seen, the choice of training sessions affects the success of the network but in addition, network structure was also crucial even between phases. The set of inputs of a structure was most difficult to determine, especially considering the numerous permutations that are possible with the 48 candidate input variables. It was further complicated since, for each set of inputs, there was likely to be a different optimum number of first layer neurones. However, given time and computational constraints, it was not possible to attempt every input combination and optimise the net for each input set, not to mention trying different groups of training sessions.

Therefore, in general, less effort was given for optimisation, especially if a particular set of inputs produced ineffective results across all test sessions; optimisation would only be performed when likely networks had been anticipated. Therefore, most new input attempts were trained using the sessions mentioned above, and 15 neurones in the first layer as this was often close to an optimum number with some of the earlier network attempts. For networks with larger sets of inputs, the number of first layer neurones was however increased although ultimately this was of less concern as networks could be optimised once some idea was ascertained as to which structure was most likely to be successful.

As mentioned all of the results are summarised in Table 8-N. After iterating through about 10 different random variable groups, the performance scores were scanned for networks that appeared to be having some success and their input groups noted. It came to attention that two different network structures seemed to be producing promising results; one structure used input group 4 and the other group 6. The variables from these groups formed the basis of the next generation of network designs where a variety of input groups was tried [Table G-A]. In this table, the variables of the two most promising groups were colour coded to make it visually clear how subsequent groups related to this group. For example, group 11 is an amalgamation of group 4 and 6, and group 12 additionally includes the thigh internal/external angular velocity; group 15 is a variation on 6 in which the force about the abduction/adduction plane of the ankle joint is also included etc.

Finally, when sufficient iterations had been performed and the possibility of significant leaps in performance reduced, the networks were examined and compared. The best set of networks for each phase was then collated and presented [Table 8-O]. Performances classed as "GREAT" were achieved for three of the phases, KP2, KP4, and KP9. It was interesting to note that for KP9, numerous network structures were able to achieve good performance whereas in contrast for KP6 and KP7 only one or two networks were able to partially solve the problem. The responses of KP1, KP5, and KP8 were also not ideal in that the number of test sessions with "V Good" results was below 10, although nearly all sessions produced at least a "Good" result. For these mediocre phases, as it were, the possibility of further optimisation was strong for example by examining which test sessions produced poor results or by trying to use another combination of input variables. More consideration was required however for the worst two phases, as the response of the networks for these phases seemed to be more endemic, and therefore possibly due to irresolvable data.

net name		KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
	good phase	KP1								
1ba10	Good Count	4	6	0	0	0	0	0	0	2
	V Good count	8	7	0	0	0	0	0	0	0
	summary	V GOOD	GREAT							
3za15	Good Count	6	7	1	1	0	0	0	0	3
	V Good count	6	4	0	0	0	0	0	0	0
	summary	V GOOD	V GOOD							
4zd15tan	Good Count	2	3	3	1	3	0	0	0	4
	V Good count	9	5	9	0	0	0	0	0	7
	summary	V GOOD	OK	V GOOD						V GOOD
4zd15	Good Count	3	0	5	0	5	0	6	0	0
	V Good count	8	13	6	0	0	0	0	0	1
	summary	V GOOD	GREAT	V GOOD						
19zn20	Good Count	8	0	2	5	4	1	0	1	6
	V Good count	4	0	3	8	0	0	0	2	6
	summary	V GOOD			GREAT					V GOOD
1lc10	Good Count	2	4	0	0	0	0	0	0	1
	V Good count	7	5	0	0	0	0	0	0	1
	summary	GOOD	GOOD							
	good phase		KP2							
1ba10	Good Count	4	6	0	0	0	0	0	0	2
	V Good count	8	7	0	0	0	0	0	0	0
	summary	V GOOD	GREAT							
1cb10	Good Count	0	4	0	5	0	4	3	0	2
	V Good count	0	8	0	6	0	0	0	2	8
	summary		V GOOD		V GOOD					GOOD
4zd15	Good Count	3	0	5	0	5	0	6	0	0
	V Good count	8	13	6	0	0	0	0	0	1
	summary	V GOOD	GREAT	V GOOD						
8zh25tan	Good Count	4	2	0	6	3	5	0	5	2
	V Good count	4	10	0	5	1	0	0	2	10
	summary	OK	V GOOD		V GOOD				OK	V GOOD
11zd17tan	Good Count	2	5	9	4	3	3	1	1	5
	V Good count	0	7	1	3	1	0	0	1	4
	summary		V GOOD	GOOD	OK					GOOD
1ed5	Good Count	2	7	6	2	2	2	2	2	2
	V Good count	4	5	3	4	4	4	4	4	4
	summary		V GOOD	GOOD						
	good phase			KP3						
4zd15tan	Good Count	2	3	3	1	3	0	0	0	4
	V Good count	9	5	9	0	0	0	0	0	7
	summary	V GOOD	OK	V GOOD						V GOOD
4zd15	Good Count	3	0	5	0	5	0	6	0	0
	V Good count	8	13	6	0	0	0	0	0	1
	summary	V GOOD	GREAT	V GOOD						
6zf15	Good Count	6	6	3	4	5	2	0	4	7
	V Good count	1	3	8	9	3	0	0	3	6
	summary	OK	GOOD	V GOOD	GREAT	OK			OK	GREAT
6zf20tan	Good Count	0	4	3	3	5	1	0	0	3
	V Good count	0	3	6	10	4	0	0	0	10
	summary		OK	GOOD	GREAT	GOOD				GREAT
8zf20tanx	Good Count	0	4	8	2	6	3	3	0	1
	V Good count	0	3	3	3	0	0	0	0	12
	summary		OK	V GOOD						GREAT
14zm15	Good Count	0	6	7	0	0	0	0	1	4
	V Good count	0	1	4	0	0	0	0	0	7
	summary		OK	V GOOD						V GOOD
	good phase				KP4					
1cb5	Good Count	0	0	0	6	2	6	2	0	2
	V Good count	0	0	0	5	6	0	2	0	6
	summary				V GOOD	OK				OK
1cb10	Good Count	0	4	0	5	0	4	3	0	2
	V Good count	0	8	0	6	0	0	0	2	8
	summary		V GOOD		V GOOD					GOOD
5ze15tan	Good Count	3	4	3	6	0	1	4	1	2
	V Good count	3	3	1	6	2	1	0	0	5
	summary		OK		V GOOD					OK
5ze15tan	Good Count	3	4	3	6	0	1	4	1	2
	V Good count	3	3	1	6	2	1	0	0	5
	summary		OK		V GOOD					OK

Cont…

6zf15	Good Count	6	6	3	4	5	2	0	4	7
	V Good count	1	3	8	9	3	0	0	3	6
	summary	OK	GOOD	V GOOD	GREAT	OK			OK	GREAT
19zn20	Good Count	8	0	2	5	4	1	0	1	6
	V Good count	4	0	3	8	0	0	0	2	6
	summary	V GOOD			GREAT					V GOOD
6zm20tan	Good Count	0	0	5	3	0	3	0	2	4
	V Good count	0	0	0	10	0	0	0	2	9
	summary				GREAT					GREAT
6cb15	Good Count	0	4	3	11	0	0	0	0	3
	V Good count	0	0	0	1	0	0	0	0	5
	summary				V GOOD					OK
6ea15	Good Count	3	4	0	5	1	1	0	3	1
	V Good count	0	4	0	8	0	1	0	8	1
	summary		OK		GREAT				V GOOD	
	good phase					KP5				
19zn25	Good Count	4	5	0	5	6	0	4	3	2
	V Good count	3	1	0	6	5	0	0	2	8
	summary	OK			V GOOD	V GOOD				GOOD
6ha15	Good Count	0	1	0	4	9	2	0	0	2
	V Good count	0	0	0	4	4	0	0	0	7
	summary				OK	GREAT				GOOD
6zm12	Good Count	3	2	0	3	6	0	0	3	1
	V Good count	4	1	0	9	6	0	0	2	12
	summary	OK			V GOOD	V GOOD				GREAT
	good phase						KP6			
8zh15tan	Good Count	5	1	0	5	3	7	5	0	2
	V Good count	0	0	0	4	0	3	0	0	5
	summary				GOOD		GOOD			OK
	good phase							KP7		
6zh25tan	Good Count	2	6	0	2	5	7	8	2	1
	V Good count	0	4	0	5	0	0	0	6	7
	summary		GOOD		OK		OK	OK	OK	OK
6zh15tan	Good Count	0	3	1	4	3	3	7	4	2
	V Good count	0	3	7	7	0	0	2	4	9
	summary			OK	V GOOD			GOOD	OK	V GOOD
	good phase								KP8	
11zf15tan	Good Count	0	6	3	7	4	0	0	7	1
	V Good count	0	3	4	1	4	0	0	5	0
	summary		GOOD	OK	OK	OK			V GOOD	
6ea15	Good Count	3	4	0	5	1	1	0	3	1
	V Good count	0	4	0	8	0	1	0	8	1
	summary		OK		GREAT				V GOOD	
	good phase									KP9
6zf20tan	Good Count	0	4	3	3	5	1	0	0	3
	V Good count	0	3	6	10	4	0	0	0	10
	summary		OK	GOOD	GREAT	GOOD				GREAT
6zf15	Good Count	6	6	3	4	5	2	0	4	7
	V Good count	1	3	8	9	3	0	0	3	6
	summary	OK	GOOD	V GOOD	GREAT	OK			OK	GREAT
7zg80tan	Good Count	1	0	0	3	0	0	2	1	1
	V Good count	0	0	0	0	0	0	0	2	12
	summary									GREAT
8zf20tanx	Good Count	0	4	8	2	6	3	3	0	1
	V Good count	0	3	3	3	0	0	0	0	12
	summary		OK	V GOOD						GREAT
6zm20tan	Good Count	0	0	5	3	0	3	0	2	4
	V Good count	0	0	0	10	0	0	0	2	9
	summary				GREAT					GREAT
6zm12	Good Count	3	2	0	3	6	0	0	3	1
	V Good count	4	1	0	9	6	0	0	2	12
	summary	OK			V GOOD	V GOOD				GREAT

phases	KP1	KP2	KP3	KP4	KP5	KP6	KP7	KP8	KP9
best net	4zd15tan	4zd15	4zd15tan	6zm20tan	6zm12	8zh15tan	6zh15tan	6ea15	6zm12
Good Count	2	0	3	3	6	7	7	3	1
V Good count	9	13	9	10	6	3	2	8	12
summary	V GOOD	GREAT	V GOOD	GREAT	V GOOD	GOOD	GOOD	V GOOD	GREAT

Table 8-O: Table showing, for each intent state, the networks that performed well. The performances related to the other phases are also included in order to show whether any particular structure also performed well across the board and not just for the selected phase. The very best network of each phase is highlighted in purple and then summarized at the bottom of the table.

Almost all results from network structures for KP6 and KP7 were poor. This was therefore considered symptomatic of the choice in the knee power topologies defining these phases [section 8.3.1] rather than just a poor network structure. In essence, a supervised network can only learn a pattern if it is presented with regular examples which are consistently defined; if targets are difficult to classify, one cannot expect good results. In fact, from the method of choosing intent states, one would not expect the quality of all the target states to be homogeneous and indeed, during this process certain topological characteristics did seem more ambiguous than others did. These ambiguities was often pronounced on a trial-by- trial basis, and indeed as an attempt to reduce this problem, trials with such uncertainty were deliberately excluded for training purposes. For example, on the intra-subject level, slow speed trials created the greatest concern where quite often, as speed decreased, many of the more sharply defined features would level off making it difficult to resolve the maxima or minima. The problem was slightly different on an inter-subject level where only certain periods of the gait cycle caused problems. Of particular note was the period between KP5 and KP8 where two different forms of the knee power patterns seemed to be possible, although at the time it was dismissed. For some subjects the blurring of some of the topological features, within this period, into others occasionally occurred. For example, with subject “F,” the KP6 peak often disappears leaving uncertainty not only with this state but also with the surrounding states [Figure F-1]. It was interesting to observe in hindsight that sessions with this occurrence had a shift of the KP4 peak to the right (e.g. from about 40% gait cycle to 45%), and with this shift appeared some confusion with regards to the KP5 and KP6 states. These sessions were “Fb”, “Fc”, “Fe”, “Ea”, and “Ed”, i.e. with the exception of session “Ed” all the sessions from subjects “F” and “E”. In correlation with this power pattern, there appeared to be very low relative moment amplitudes throughout the KP4 – KP8 period as compared to other sessions. Only session “Ed” for subject “E” had a knee moment that was pronounced, and in apparent consequence a small KP6 peak also became visible. Thus, it is conceivable that two styles of control patterns exist during the period KP4 – KP7 (which will be referred to as the old style e.g. like session “Ba”, or new style such as session “Ea”).

Two solutions occurred to the author both of which involve identifying new intent states. The first was more in keeping with the approach that has been undertaken in which these two styles are simply treated as an additional pattern to identify and will therefore represent new states for networks to specifically be trained to recognise. If, then test data representing the old style is presented, only certain networks should respond, or conversely if the new style gait is presented only networks that are trained to recognise this should respond.

The alternative method concedes that KP5, KP6, and possibly KP7 are not robust topological patterns among all subjects. Therefore, these features would not be used as intent states. This, however may produce problems concerning the temporal resolution of the controller, where too great a leap from one state to the next is required. The implication being that long feed-forward patterns are needed for control and thereby a greater chance of control error exists. The issue may however be addressed by simply splitting the time between the closest two robust states e.g. KP4 – KP9 so that additional states are set up relative to these two. This will not contravene the original philosophy because the phases are still with respect to controlling phenomena that are determined on a trial-by-trial basis.

In general, the repercussions of inaccurate target states may not only influence training of networks but also the testing process. Results can sometimes be deemed erroneous because the test target has been marked incorrectly so that a network may actually be performing perfectly well but is being judged as erroneous. The situation may arise for noisy test trials, maybe with additional waveforms induced through pre-processing of the Vicon data. However, considering the circumstance described above with the possibility of two walking patterns, if a network has been trained for one style then correct results would only be expected for test data with that style. If test data with the different style is able to produce a high value, the performance may be deemed poor simply because the target is incorrectly offset. The network may have correctly identified the state where the author was not able to. Therefore, the results should not immediately be interpreted too harshly for these phases with two possible styles.

At this stage, the network structures that produced the best results were considered [bottom of Table 8-O] and an attempt at optimising the structures performed. Taking the best KP8 network as an example (H6ea15), the process of optimisation may be demonstrated (see Table 8-P for results). The first examination was to see how the number of first layer neurones affected performance. Therefore whilst keeping the training algorithms and first layer functions, 10 neurones then 20 were attempted. Such steps however seemed too large as very poor results were attained. Eventually 17 neurones in the first layer were shown to produce the best results. Using this number of neurones, the neurone function was then changed to a tan-sigmoid function that gave the same performance according to the "good" and "V good" counts; however, scrutiny of the figures shows higher scores with the tan-sigmoid function. The only failure for this phase was for session "Ee," and that was most likely because the data was corrupted near the end of the trials because reflective markers were not being seen by the Vicon cameras.

# 1st layer neurons training algorithm 1st layer function test session	15 Levenberg logsig summary 200=good		10 Levenberg logsig summary 200=good		20 Levenberg logsig summary 200=good		17 Levenberg logsig summary 200=good		17 Levenberg tansig summary 200=good		13 Levenberg logsig summary 200=good		16 Quasi Newton logsig summary 200=good		18 Bayesian Regularisa logsig summary 200=good		17 Bayesian Regularisa logsig summary 200=good	
Ba	158	bad	100	bad	100	bad	196	V Good	189	Good	190	V Good	164	Good	99	bad	100	bad
Bb	199	V Good	100	bad	100	bad	195	V Good	199	V Good	199	V Good	98	bad	180	Good	100	bad
Bd	199	V Good	100	bad	100	bad	191	V Good	195	V Good	198	V Good	97	bad	183	Good	100	bad
Ca	198	V Good	100	bad	108	bad	196	V Good	194	V Good	196	V Good	161	Good	195	V Good	129	bad
Cb	198	V Good	100	bad	100	bad	196	V Good	197	V Good	197	V Good	178	Good	199	V Good	145	bad
Fb	183	Good	100	bad	100	bad	189	Good	189	Good	177	Good	105	bad	188	Good	100	bad
Fc	195	V Good	100	bad	100	bad	195	V Good	195	V Good	184	Good	100	bad	193	V Good	100	bad
Fe	194	V Good	100	bad	100	bad	195	V Good	199	V Good	189	Good	100	bad	195	V Good	100	bad
Ea	162	Good	100	bad	100	bad	179	Good	167	Good	171	Good	100	bad	154	bad	100	bad
Ed	196	V Good	100	bad	100	bad	196	V Good	197	V Good	193	V Good	100	bad	196	V Good	100	bad
Ee	109	bad	100	bad	100	bad	104	bad	109	bad	98	bad	100	bad	119	bad	100	bad
Ha	190	V Good	100	bad	100	bad	186	Good	200	V Good	200	V Good	100	bad	171	Good	100	bad
Hc	175	Good	100	bad	104	bad	173	Good	188	Good	186	Good	99	bad	168	Good	100	bad
bad count		2		13		13		1		1		1		10		3		13
good count		3		0		0		4		4		5		3		5		0
V good count		8		0		0		8		8		7		0		5		0

Table 8-P; Table showing optimisation of the KP8 network structure (trained with session Ea, and group 6 input variables). Three parameters are changed to see if improvements can be attained: number of first layer neurones, training algorithm and first layer neurone function. The two highlighted networks were the optimum for this set.

After trying a couple of different training algorithms, the Levenberg Marquardt optimisation proved to be the best for this problem.

Thereby, to conclude it has been shown how artificial neural networks can be used to recognise controlling phases and act as trigger points within a task. Although results weren't perfect, the faults were of greater benefit in order to help understand what was generically required from the procedure. Nevertheless, many networks proved successful and robust across most test sessions.

9 CONCLUSION / FUTURE WORK

A strategy was developed to design an artificial controller of powered knee mechanisms for trans-femoral amputees. This was formulated through a series of logical steps, starting with a consideration of what is meant by the term "movement" (i.e. what does it imply, what is it), and second how might it be accomplished? Accordingly, a review of the literature on gait analysis was first given, and then a review collating the studies that examined how the natural system may achieve these behaviours, while acknowledging the issues that encompass the problem.

The reviews formed a foundation for the project, and allowed a set of specific *aims* for the artificial controller to be drawn up; subsequently *constraints* based on the aims were imposed to focus the design process and define the objectives of the controller. In essence, the controller was to assist with autonomic movement tasks that would ordinarily be performed subconsciously; in a similar vein, the control system was also required not to demand high levels of concentration.

To formulate a strategy to achieve the aims required some consideration of how an artificial controller may interact with the natural control system where coordination was required between these two separate systems. As a first step, a schematic diagram was drawn up to illustrate the possible routes of information flow and consequential interactions. With this perspective it was possible to review control systems designed by other researchers and identify problem areas as well as glean further inspiration. In particular, the consequence of using feed-forward or feedback mechanisms was highlighted. For example, systems largely based on feedback signals (such as electromyography (EMG) controllers) were thought to be problematic in terms of coordinating the natural and artificial controllers, where delays formed a major concern requiring high concentration and contravening many of the aims of the project.

Instead, a design using feed-forward control techniques was formulated. In particular, control patterns were sought which could be implemented for feed-forward control. These patterns were derived from *trans-tibial* amputees because it was argued that they already had the capability of controlling the knee whilst having adapted to using a prosthetic device for the shank and foot. They afforded the closest available model for what was desired for a trans-femoral amputee that was to use an active knee mechanism in combination with a conventional foot and shank prosthesis.

Much thought was given to the method of identifying feed-forward patterns. Firstly, the patterns had to be representative of the movement task (which itself had to be defined), and secondly they had to be synchronised with the task that was being performed by the amputee. Synchronisation entailed that the *appropriate* feed-forward pattern for the task had to be activated *precisely* on time; this dual role was dealt with by the "*intent recogniser*". It was argued that the timing of *control events* could be ascertained from the *topological features of knee power* patterns gathered from trans-tibial amputees. This was a fundamentally different way of defining phases within a task. Unlike the conventional method of dividing gait patterns into a percentage of the gait cycle (typically between two repeating cyclic *outcome* events e.g. heel-strike), the use of power topology made no assumption that all such events would remain temporally fixed within this gait cycle scale. Instead, each topological feature that was identified within the task became a *control phase* (or *intent state*) in its own right; there was no attempt to pinpoint their occurrence to particular percentages of the gait cycle.

To recognise these control phases, the use of artificial neural networks was proposed, which enabled mappings of the knee power events with a selection of outcome variables, the best selection eventually forming an array of input sensors. However, rather than only using a single network to produce all mappings, separate neural networks were dedicated to recognise each of the events that were identified within each particular task. It was desirable to identify as many such power events as possible so that the duration between each control phase would be shortened and minimise the chance of errors accumulating as the feed-forward patterns proceeded. The detection of a new control phase thereby acted to reset and trigger new feed-forward patterns. Since each of these intent states would be recognised by a separate network (including those of other tasks), it was possible to conceptualise a simple parallel structure that would allow new tasks to be incorporated in a modular fashion.

Walking was chosen as the specific gait to analyse for demonstration purposes, and therefore care was taken to gather the full range of this pattern as it varied with speed. Sample data had to be gathered, which was sufficiently diverse for neural networks to generalise and enable robustness against subject differences in gait while being precise enough not to cause timing errors. Most of the data was provided by five trans-tibial unilateral amputees, where up to three sessions of useful data was gathered; each session including about 50 separate walking trials. Though the use of more amputees was always desirable to increase the variety of data, the amputees that were available were shown to be heterogeneous in stature and build, with only the height diversity being singled out as a main area for improvement; the population was considered satisfactory for demonstration purposes. Each amputee was able to provide data spanning their full range of walking speed, i.e. from very slow to very fast although it was important that they were still walking and did not begin to jog for example. Since so much data was provided by the order of 150 trials per amputee, covering their full speed range, it made possible three-dimensional analysis of the data to see how each gait variable developed with speed.

The Vicon camera data acquisition system was used to record trajectories (in three-dimensions) of markers that were strategically placed on the trans-tibial subjects, and force plates were used to measure ground reaction forces. This method was chosen to gather data because through subsequent analysis it was possible to calculate a multitude of kinematic and kinetic gait variables that could be used in many combinations to train different network structures. Much effort was also made to measure the various body parameters such as the moments of inertia of the lower limb segments to enable the kinetic analysis.

Pre-processing of marker trajectories was performed to reduce noise and fill in gaps in the data that the Vicon system did not capture. Various checks were performed to show the accuracy of the trajectories. For example, the distances between joint centres were calculated from the marker trajectories and compared to actual measurements. It was found that these distances were very consistent throughout a session, but slight discrepancies would occur between sessions that were thought to be due to inaccuracies in the determination of local segment reference frames. However, since these differences were only up to about 2cm, and given that variability was desirable to allow the networks to generalise, the uncertainty was not of great concern.

The subsequent kinematic analysis of this data provided patterns with obvious similarities to non-pathological subjects (where the two longest term amputees in particular were most comparable). Although the inter-segment angle patterns suffered some offset drift, thought to be due to inaccuracies in defining the local reference systems, they were considered suitable for neural network training. The inter-segment angular velocities, being derivatives, did not suffer the same from drift, and hence were closer matches to each other. The most discernible gait patterns were produced by a congenital amputee whose pattern was more pronounced and closer to non-pathological patterns. It seemed that his active style either facilitated, or was facilitated by, plantar flexion of the prosthetic foot just after toe-off, which was absent for the other subjects.

This distinction was also reflected in the kinetic variables, where for example the vertical component of his ground reaction force as normalised with respect to body mass was the most pronounced and closest to non-pathological levels. Overall, a good variation of gait patterns was provided by the amputees, ranging from the impressive congenital amputee’s patterns to those produced by less fit amputees.

Despite a prosthetic foot, the ankle moment patterns resembled non-pathological patterns albeit with slightly smaller amplitudes. The inter-segment knee moments exhibited a much greater range of variation than the passively controlled ankle moments, where in general, peak values were less than those produced by non-pathological subjects. The hip moment patterns also had a similar topology to non-pathological subjects, although peak values appeared to be strikingly larger which might reflect increased activity to cope with deficits in ankle function.

In general, the trans-tibial amputees’ inter-segment power patterns for the knee during walking had greater variability than the other variables. It was of note that this variability not only occurred with respect to amplitude but also with respect to the percentage of the gait cycle (this was a vindication of not using percentage of gait cycle to indicate control events). Despite the variability, features that were in common with non-pathological subjects were still recognisable, including the gait parameters that other researchers had labelled themselves. After much scrutiny of the power topologies, both using intra-subject comparisons related to speed changes, and inter-subject comparisons it became possible to establish a set of power events that would be used for intent recognition.

Initially the most robust events were established, and because of the variability from trial to trial and subject to subject, it became a useful guide to also superimpose the ensemble averagef knee moment patterns and the knee angular velocity patterns. Thereby, nine intent states (KP1 – KP9) were established for the task of walking, which were generally related to peaks and troughs.

After much patience, each trial of data was scrutinised and any of the nine intent states that were clearly visible were tagged in preparation for neural network training and testing. A separate neural network was required to recognise each one of the nine intent states, whereby their outputs represented the probability of the relevant intent state occurring. The process of establishing the most suitable neural networks for each state involved iterating through a variety of network structures, which included not only changing the number of neurones within the network but also the input variables, node functions, etc. The direction and step size of the iterations were influenced by their success using test data. Towards this, it was necessary to produce an algorithm to analyse the success of each network. Because 666 networks were generated and each one was tested with large quantities of data, the results were summarised according to 11 different performance values examining consistency, resolution/distribution, and accuracy. These performance values were then combined to give a qualitative indication of the networks with overall performances of “Great”, “V. Good”, “Good”, and “Ok” where high consistency scores were given the most weighting. Although these classes were somewhat arbitrary, they enabled a quick way of highlighting the most promising networks. For example, an overall performance of “Great” meant that the response of a particular network to each one of the test sessions was very consistent so that intent states were being identified only when they should occur.

As an initial observation, the success of the networks to test data seemed to depend on which trans-tibial amputee had provided the training data. It appeared the walking styles of some amputees were more generically comparable to other subjects while the less successful styles might have included peculiarities to the individual, which the networks focused upon.

It was also found that it was easiest to design networks to recognise intent states KP2, KP4, and KP9 while KP1, KP5, and KP8 required more tuning of the networks before good results could be achieved. However, it was noticeably more difficult to find networks that had sufficiently generalised to always recognise phases KP6 and KP7. It was thought that the problem for these states was related more to their definition according to power topology than to merely being able to iterate adequate network designs. For example, the minima used to define KP7 were sometimes vague and difficult to pinpoint because the topology was rather flat for some subjects, especially at slow speeds. In essence, the choice of this topological feature was not ideal for defining a state.

Such a situation also brought to attention that the analysis of network accuracy was first dependent on how accurately the data had been tagged. For example, it was possible to train networks with data from subjects with definitive KP6 states, and yield positive responses when tested with new subject test data, i.e. these networks would recognise a single state during all gait cycles. However, the algorithm to analyse the network response sometimes judged the results erroneous because the timing of the state was shifted from the manually tagged frames. It seemed the error reflected more a difference in opinion than a true error where the definition of the KP6 state for one amputee differed from another one. It was conceivable that during such ambiguous phases that more than one gait style existed between subjects.

Two solutions to ambiguous states were considered. The simplest was to ignore such contentious states so that networks only recognised robust states. This does of course imply that the duration between states would be longer, which may increase the possibility of cumulative error caused by using feed-forward patterns. If such duration was to become excessive then intervening states may have to be established which occur at normalised intervals between two adjacent robust states.

The more complex solution recognises that more than one gait pattern may exist. Fortunately, because of the modularity afforded by using a network for each intent state, it was thought the strategy could be extended to also include such dichotomous states. A separate network could be designed for each of these states so that only the one representing the appropriate style of the test subject would be activated. Ultimately, the more networks running in parallel the more information will be available to the system for decision-making.

Considering now the choice of input variables, although there were insufficient resources to design networks using every permutation of the 47 possible input variables, the most promising group of variables was established using both an iterative process and educated guessing. It was found that the best set of input arrays fell into three groups (4, 6, and 8) dependent on the state being recognised. These groups however had variables in common including hip flexion/extension angles, prosthetic shank angular velocity, thigh internal/external angular velocity, and the vertical ground reaction force. It was however recognised that several intent states performed almost as well with other combinations of input variables so that there appeared plenty of room for practical considerations in the choice of sensory inputs.

Once a set of input variables had been established, it was shown that the network could be further improved by further optimisation of the networks through adjustment of the number of first layer neurones, neurone functions, and training algorithms. For example, for intent state KP8: 17 input neurones with the tan-sigmoid transfer function, and the Levenberg Marquardt optimisation rule proved to give the best combination.

Overall, it can be said that a detailed strategy was developed and demonstrated for designing the control system. The strategy has incorporated and considered many of the issues that were realised when devising the aims. Areas of concern with the methodology have been highlighted and suggestions made as to how they could have been overcome. It appears that the method certainly looks promising towards recognising intent states with sufficient accuracy and reliability.

9.1 FUTURE WORK

It is clear that there is much further work required before a fully commercially available prosthesis based on these control principles can be realised. However, regarding this project, it can be seen how the strategy can be continued whereby improved network structures are found (e.g. with increasing computational power more network permutations may be examined), and the system expanded to include more states involving other tasks.

It would be beneficial if much of this process can be further automated, for example with the identification of topological features within tasks or quicker means of gathering data for the neural networks.

There is also more work to be done on providing definitive feed-forward profiles to span between intent states. Although it has been suggested that these patterns may be taken from exemplar trans-tibial amputee data, this area has not been pursued further within this project because ultimately the profiles must be transformed to suit the eventual dynamics of whatever mechanism is employed. It would also be much simpler to test the effects of minor adjustments in the profiles once a prototype is constructed. It was thought that the recognition of intent states had an overriding priority as once this is achieved then one can always make modifications of the feed-forward profiles. It is even possible to provide simple attenuation of patterns to suit the amputee. It must however be conceded that more work can always be done in modelling control variable (e.g. speed) related effects of feed-forward patterns, although as discussed other researchers are involved in this area.

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Appendix A STRATEGIES FOR NORMALISING GAIT DATA

Dimensionless numbers related to gait mechanics

Quantity	Symbol	Dimension	Dimensionless number
Mass	m	M	$\hat{m} = \frac{m}{m_0}$
Length, distance	l, x	L	$\hat{l} = \frac{l}{l_0}$
Time	t	T	$\hat{t} = \frac{t}{\sqrt{l_0/g}}$
Frequency	f	T^{-1}	$\hat{f} = \frac{f}{\sqrt{g/l_0}}$
Speed, velocity	$v = \dot{x}$	$L\ T^{-1}$	$\hat{v} = \frac{v}{\sqrt{g l_0}}$
Acceleration	$a = \ddot{x}$	$L\ T^{-2}$	$\hat{a} = \frac{a}{g}$
Force	F	$M\ L\ T^{-2}$	$\hat{F} = \frac{F}{m_0 g}$
Moment	M	$M\ L^2\ T^{-2}$	$\hat{M} = \frac{M}{m_0 g l_0}$
Work, energy	W, E	$M\ L^2\ T^{-2}$	$\hat{W} = \frac{W}{m_0 g l_0}$
Power	P	$M\ L^2 T^{-3}$	$\hat{P} = \frac{P}{m_0 g^{1/2} l_0^{3/2}}$
Angle	ϕ	(is already dimensionless)	
Angular velocity	$\omega = \dot{\phi}$	T^{-1}	$\hat{\omega} = \frac{\omega}{\sqrt{g/l_0}}$
Angular acceleration	$\alpha = \ddot{\phi}$	T^{-2}	$\hat{\alpha} = \frac{\alpha}{g/l_0}$
Moment of inertia	I	$M\ L^2$	$\hat{I} = \frac{I}{m_0 l_0^2}$

m_0, body mass; l_0, leg length (from greater trochanter to floor); g, acceleration of gravity ($= 9.81\ m/s^2$ on earth).

Table A-A - List of normalisation parameters suggested by Hof [99]

Appendix

Summary of scaling strategies that could be used to reduce inter-subject variation in gait data outcomes[a]

	Scaling strategy							
	I (none)	II (ad hoc)	III (dimensionless numbers[b])	IV_0 (hydrodynamic[d])	$IV_{1/2}$	IV_1 (Geometric[e]/ Kinematic[d]/ Geometric[c])	$IV_{3/2}$ (Dynamic[e]/ Mechanical[b]/ Elastic[e])	IV_2 (Static stress)
Diameter [$Do \propto Lo^{F}$]	–	–	–	Lo^0	$Lo^{1/2}$	Lo^1	$Lo^{3/2}$	Lo^2
Time [$To \propto Lo^2\ Do^{-1}$]	–	–	–	Lo^2	$Lo^{3/2}$	Lo^1	$Lo^{1/2}$	Lo^0
Stride time [To]	1.0	$Lo^{1/2}$	$Lo^{1/2}$	Lo^2	$Lo^{3/2}$	Lo^1	$Lo^{1/2}$	Lo^0
Stride length [Lo]	1.0	Lo^1	Lo^1	Lo^1	Lo^1	Lo^1	Lo^1	Lo^1
Progression velocity [$Lo\ To^{-1}$]	1.0	$Lo^{1/2}$	$Lo^{1/2}$	Lo^{-1}	$Lo^{-1/2}$	Lo^0	$Lo^{1/2}$	Lo^1
Angle (sagittal, frontal) [$Lo\ Do^{-1}$]	1.0	1.0	1.0	Lo^1	$Lo^{1/2}$	Lo^0	$Lo^{-1/2}$	Lo^{-1}
Angle (transverse) [$Do\ Do^{-1}$]	1.0	1.0	1.0	Lo^0	Lo^0	Lo^0	Lo^0	Lo^0
Ground reaction force [$Mo\ Lo\ To^{-2}$]	1.0	Mo	Mo	$Mo\ Lo^{-3}$	$Mo\ Lo^{-2}$	$Mo\ Lo^{-1}$	$Mo\ Lo^0$	$Mo\ Lo^1$
Ground reaction moment [$Mo\ Do^2\ To^{-2}$]	1.0	Mo	$Mo\ Lo^1$	$Mo\ Lo^{-4}$	$Mo\ Lo^{-2}$	$Mo\ Lo^0$	$Mo\ Lo^2$	$Mo\ Lo^4$
Joint force [$Mo\ Lo\ To^{-2}$]	1.0	Mo	Mo	$Mo\ Lo^{-3}$	$Mo\ Lo^{-2}$	$Mo\ Lo^{-1}$	$Mo\ Lo^0$	$Mo\ Lo^1$
Joint moment [$Mo\ Lo\ Do\ To^{-2}$]	1.0	Mo	$Mo\ Lo^1$	$Mo\ Lo^{-3}$	$Mo\ Lo^{-3/2}$	$Mo\ Lo^0$	$Mo\ Lo^{3/2}$	$Mo\ Lo^3$
Joint power [$Mo\ Lo^2\ To^{-3}$]	1.0	Mo	$Mo\ Lo^{1/2}$	$Mo\ Lo^{-4}$	$Mo\ Lo^{-5/2}$	$Mo\ Lo^{-1}$	$Mo\ Lo^{1/2}$	$Mo\ Lo^2$

[a] Mo = total body mass, Lo = leg length (hip to ankle joint centre).
[b] Hof [7].
[c] Duncan [10].
[d] Günther [11].
[e] McMahon [12–14]. Lo^0 is used, instead of 1, to convey the theoretical underpinnings of the connected scaling strategies.

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Table A-B: From Pierrynowski [161], Shows the 8 normalization strategies that they evaluated.

Inter-subject variability (mean, 95% confidence interval) of pooled gait data after modification using various scaling strategies[a]

	Scaling strategy							
	I (none)	II (ad hoc)	III (dimensionless numbers)	IV_0 (hydrodynamic)	$IV_{1/2}$	IV_1 (geometric/ kinematic/ geometric)	$IV_{3/2}$ (dynamic/ mechanical/ elastic)	IV_2 (static stress)
Inter-subject variations of the mass and length scaled pooled gait data outcomes	108.2 107.5–108.9	***47.7*** *47.4–48.0*	***47.7*** *47.4–48.0*	91.0 90.4–91.6	69.5 69.0–69.9	54.7 54.4–55.1	47.8 47.5–48.1	50.3 50.0–50.6
Reduction in the inter-subject variations of the mass and length scaled pooled gait data outcomes	1.00	0.44	0.44	0.84	0.64	0.51	0.44	0.46

[a] Within each row italicized bold faced entries indicate the lowest variability score and the shaded cells indicate the set of statistically equivalent lowest variation scores.

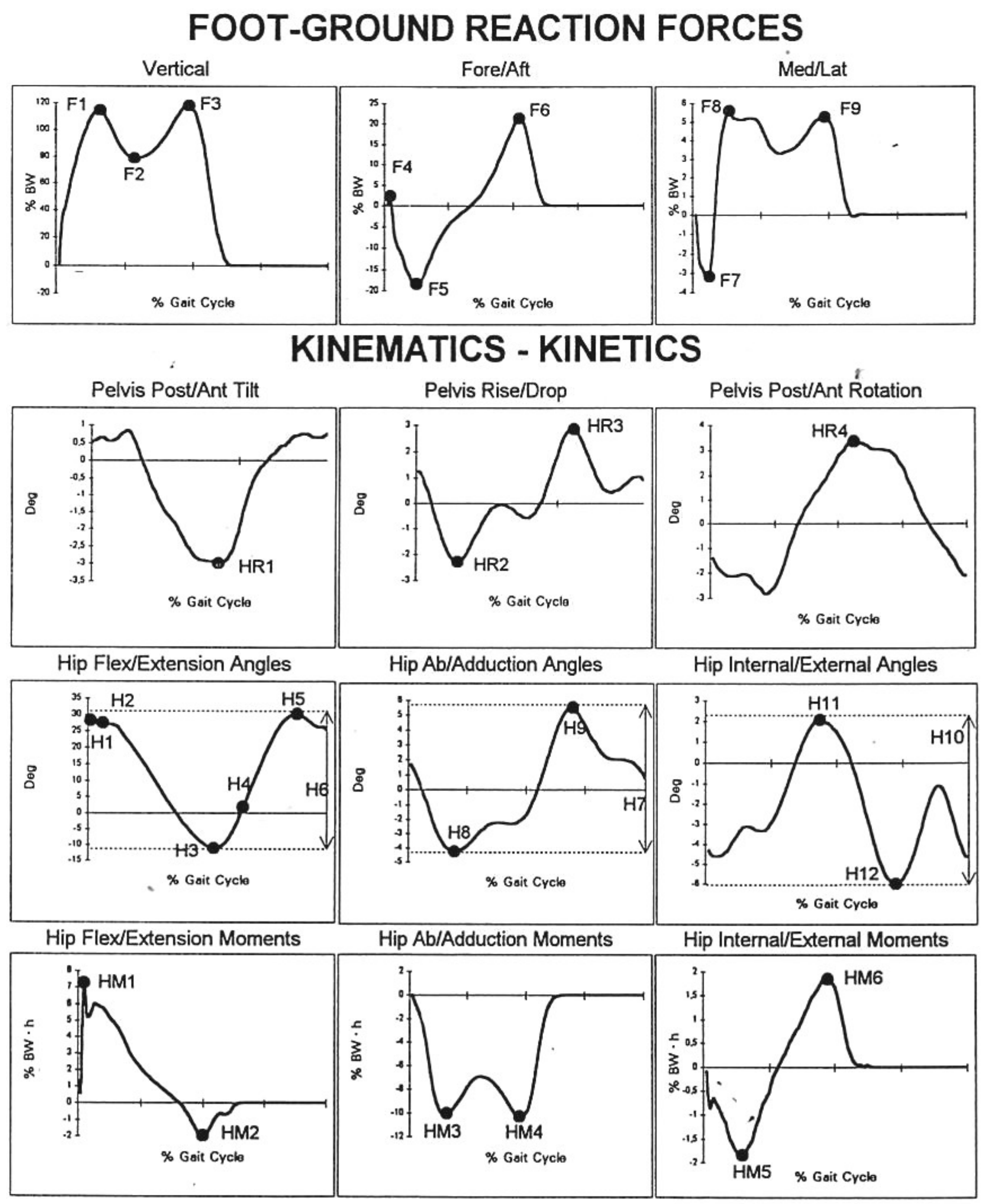
Table A-C: Comparison of scaling strategies inTable A-B, from Pierrynowski [161].

Results from the forward stepwise regression model for the unnormalized, body mass, and body weight times height normalized data sets. The adjusted r^2 represents the proportion of variance of the moment explained by height, weight, or speed. For example, for the unnormalized hip flexion moment, weight entered the equation first explaining 35% of the variance in this moment (adj $r^2 = 0.35$) followed by speed, which explained an additional 11% of the variance (Δadj $r^2 = 0.11$). Therefore, together these two variables accounted for 46% of the variance in the hip flexion moment (adj $r^2 = 0.46$)

Peak joint moment	Unnormalized data				Body mass normalized data				Body weight times height normalized data			
	Variable in	adj r^2	Δadj r^2	p-value	Variable in	adj r^2	Δadj r^2	p-value	Variable in	adj r^2	Δadj r^2	p-value
Hip flexion	Wt.	0.35	0.35	<0.001	Speed	0.18	0.18	<0.001	Speed	0.18	0.18	<0.001
	Speed	0.46	0.11	<0.001								
Hip extension	Wt.	0.18	0.18	<0.001	Speed	0.10	0.10	<0.001	Speed	0.10	0.10	<0.001
	Speed	0.25	0.07	<0.001	Wt. (−)	0.12	0.02	0.041	Wt. (−)	0.16	0.06	0.001
	Ht.	0.28	0.03	0.011	Ht.	0.15	0.03	0.016				
Hip abduction	Wt.	0.16	0.16	<0.001	None	None	—	—	None	None	—	—
Hip adduction	Wt.	0.60	0.60	<0.001	Speed	0.02	0.02	0.039	Ht. (−)	0.13	0.13	<0.001
	Ht. (−)	0.61	0.01	0.019					Speed	0.15	0.02	0.031
Hip internal rotation	Wt.	0.31	0.31	<0.001	Speed	0.03	0.03	0.015	Speed	0.03	0.03	0.026
	Speed	0.34	0.03	0.013								
Hip external rotation	Wt.	0.15	0.15	<0.001	Speed	0.07	0.07	<0.001	Speed	0.07	0.07	0.001
	Speed	0.20	0.05	0.001					Wt. (−)	0.09	0.02	0.049
Knee flexion	Speed	0.08	0.08	<0.001	Speed	0.11	0.11	<0.001	Speed	0.11	0.11	<0.001
	Ht.	0.15	0.07	<0.001					Wt. (−)	0.13	0.02	0.026
Knee extension	Wt.	0.47	0.47	<0.001	Speed	0.09	0.09	<0.001	Speed	0.09	0.09	<0.001
	Ht.	0.51	0.04	<0.001	Ht.	0.14	0.05	0.002				
	Speed	0.54	0.03	0.004								
Knee adduction	Wt.	0.45	0.45	<0.001	Ht.	0.05	0.05	0.003	None	None	—	—
	Ht.	0.48	0.03	0.002								
Ankle dorsiflexion	Wt.	0.82	0.82	<0.001	Ht.	0.22	0.22	<0.001	Speed	0.08	0.08	<0.001
	Ht.	0.85	0.03	<0.001	Speed	0.29	0.07	<0.001	Wt. (−)	0.1	0.02	0.037
	Speed	0.86	0.01	0.001								

*Table A-D: From Moisio et al. [140], Comparison of scaling moments to mass, and body weight*height.*

Appendix B GAIT ANALYSIS PARAMETERS



An example of parameter selection on GRFs, pelvic rotations, joint angles and moments on one individual's wavefo

Cont...

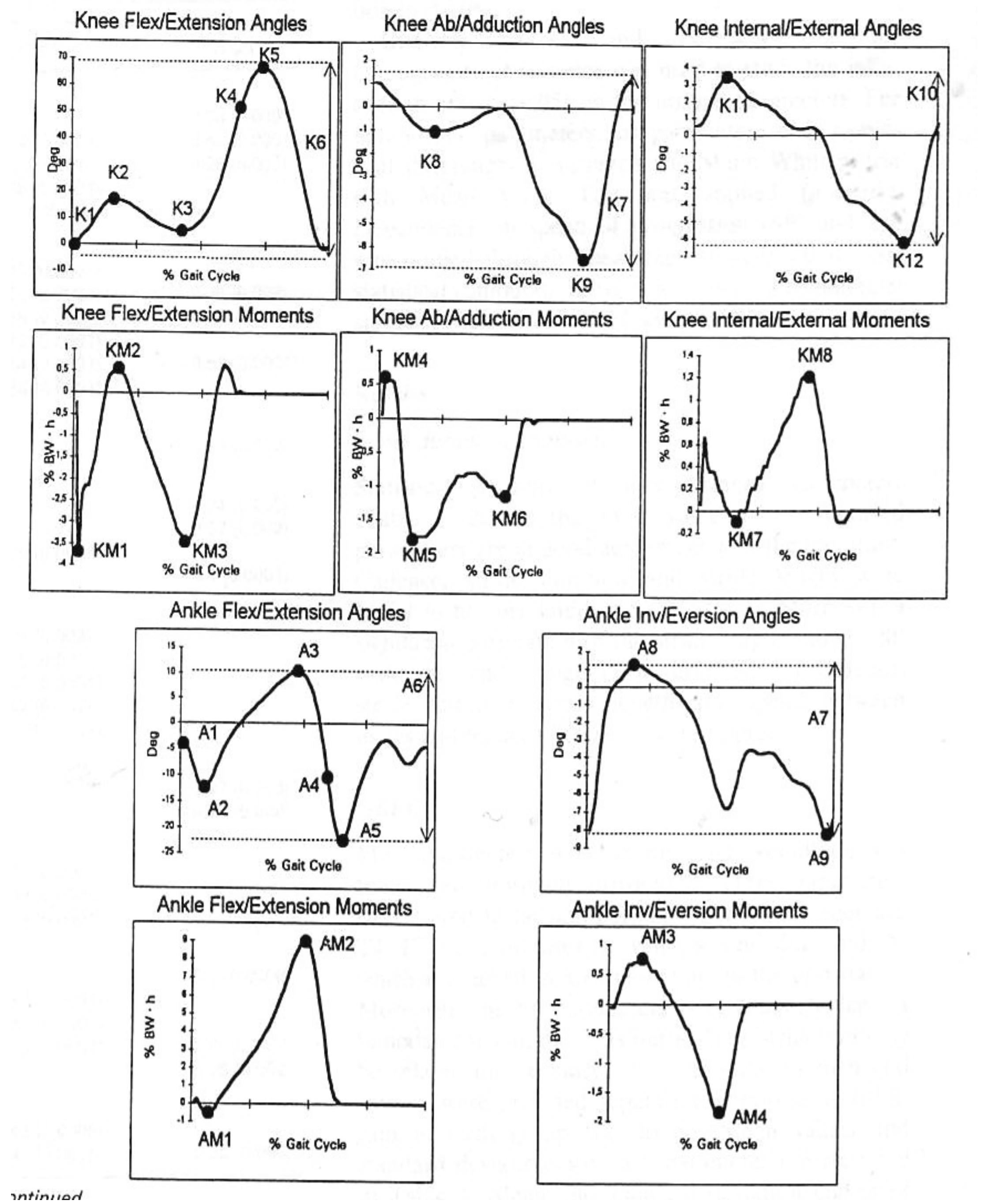


Figure B-1: Taken from Benedetti [16]. Suggested labelling convention for gait parameters.

Ground Reaction Forces (% body weight)	
F1	Max. vert. Force loading response
F2	Min. vert. Force mid-stance
F3	Max. vert. Force terminal stance
F4	Max. fore-aft force loading response
F5	Min. fore-aft force mid stance
F6	Max fore-aft force terminal stance
F7	Min. med-lat force loading response
F8	Max. med-lat force mid stance
F9	Max. med-lat force terminal stance

Table B-A: Prefixes for Ground Reaction Force parameters

Hip angles Parameters (deg.)		*Knee angles Parameters (deg.)*		*Ankle angles Parameters (deg.)*	
H1	Flex at HS	K1	Flexion at HS	A1	Flexion at HS
H2	Max. flex. at loading response	K2	Max. flex. at loading response	A2	Max. plant. flex. at loading response
H3	Max. ext. in stance phase	K3	Max. ext. in stance phase	A3	Max. dorsiflex. in stance phase
H4	Flexion at TO	K4	Flexion at TO	A4	Flexion at TO
H5	Max. flex in swing phase	K5	Max. flex. in swing phase	A5	Max. dorsiflex. in swing phase
H6	Total sagittal plane excursion	K6	Total. sagittal plane excursion	A6	Total. sagittal plane excursion
H7	Total coronal plane excursion	K7	Total. coronal plane excursion	A7	Total. coronal plane excursion
H8	Max. add. in stance phase	K8	Max. add. in stance phase	A8	Max. eversion in stance phase
H9	Max. abd. in swing phase	K9	Max. abd. in swing phase	A9	Max. inversion in swing phase
H10	Total transverse plane excursion	K10	Total transverse plane excursion		
H11	Max. int. rot. in stance phase	K11	Max. int. rot. in stance phase		
H12	Max. ext. rot. in swing phase	K12	Max. ext. rot. in swing phase		

Table B-B: Prefixes for angle parameters about the Joint Axes.

Hip Moment Parameters (% body weight * height)	Knee Moment Parameters (% body weight * height)	Ankle Moment Parameters (% body weight * height)
HM1 Max flex. moment	KM1 1st max. ext. moment	AM1 Max. plantar flex moment
HM2 Max. ext. moment	KM2 Max. flex. moment	AM2 Max. dorsiflex. moment
HM3 1st max. add. Moment	KM3 2nd max. ext. moment	AM3 Max. eversion moment
HM4 2st max. add. Moment	KM4 Max. abd. moment	AM4 Max. inversion moment
HM5 Max. ext, rot. moment	KM5 1st max. add. moment	
HM6 Max. int. rot. moment	KM6 2nd max. add. moment	
	KM7 Max. ext. rot. moment	
	KM8 Max. int. rot. moment	

Table B-C: Prefixes for Inter-segment Moment Parameters generated by External forces

Appendix C FEEDBACK CONTROL STRATEGIES FOR R

Two-Point Feedback Control (On-Off control):

This would be the simplest response that feedback controllers may utilise; it is commonly used in thermostats for example. In such controllers, two states trigger the controller to either switch on or switch off an actuator. In this example using a thermostat, a heater may be switched on when the controlled variable (e.g. temperature) falls below the first set point and it is switched off when the temperature rises above the second set point. This may be represented in terms of an error where the two states depend on the sign of the error as compared to the two threshold values. If the difference between the room temperature and the "off" set point is positive then the heater is switched off and if the difference between the room temperature and the "on" set point is positive the heater is switched on. This system therefore relies on external influences to provide potential energy to push the system back in one particular direction, i.e. cool the room down to the natural environmental temperature; it essentially uses a self-seeking rest state. The problem with this is that if on a particularly hot day the rest state is higher than the off set point then the room will never cool down. An air conditioner will additionally be required to force the temperature down (see the next method). This type of feedback control usually ends up cycling between on and off states so that the variable that is being controlled is constantly fluctuating between the cut-on setting and the cut-off setting at a certain frequency depending on how close the two limits are together. This is often used for slow applications where such movements between the two limits are acceptable. For the situation in this project, gravity may be used to provide the potential energy so that a one way motor that only outputs one value of torque would oppose the force of gravity. This however, would probably not yield very satisfactory results as there are other forces coming into effect such as ground reaction forces.

Three-Position Feedback Control:

Unlike two-point control, this method does not rely on using an external potential energy source to push the system back in a particular direction. Instead, the controller can itself push the system in the two directions. It will still suffer from oscillating between the set limits but will be more able to respond to supplementary influences. An example may be with air conditioners where both a heater and a cooler can be employed to force the room temperature in the required direction.

Proportional Feedback Control: As the name suggests, the output of this system is directly proportional to the error so that **R** is just some constant that acts as a simple gain on the error between the target task variable and the actual task variable. Therefore, in this system the larger the error the greater will be the restoring response. This method should help to reduce overshooting of the output since as the output approaches the target, the error will be small, and hence the response will be small. The sign of the error will also allow a bi-directional response, which should allow the system to home in on the set point despite overshoots. This has the advantage of being a very simple method of feedback control although suffering fundamentally from what is known as the **steady state error** problem**.** As mentioned because the restoring force becomes smaller as the detected error becomes smaller there will eventually be a situation in which the correcting force drops almost to zero, in which case it may no longer be sufficiently strong to overcome any opposing forces such as friction or gravity. This results in a new steady state that is not the same as the target state and the difference leaves a dead zone. Increasing the gain will reduce and eventually overcome this problem, however at the cost of increased instability by being more prone to overshooting the target. Some controllers can also overcome this problem by using a **bias** for situations in which the steady state error is caused by a predictable source i.e. that maintains a constant resistive force. The bias therefore will effectively compensate for the effects of the residual force so that it works almost like another source input to the plant.

Integral Control:

This method was developed in order to overcome any unexpected steady state problems that may be occurring during the course of the control. The output from the controller is not only dependent on the present error but also on past errors. By integrating all these errors over time and multiplying the result by a gain constant it is possible to have feedback control which eventually overcomes the problem of small steady state errors because the effect is accumulated over time so that eventually the response of the controller is sufficiently large to overcome the dead zone. This method however can suffer from overshooting. For example if the dead zone is caused by friction then because of the non-linear nature of friction[46], as soon as the restoring force has built up to overcome it, the initial friction suddenly drops. Once the body is moving, the plant is liable to overshoot its target because the initial force will now be too much. Herein lies a fundamental problem in that with integral control only friction is relied upon for braking before the plant gets to its target point. If this is not sufficient, active braking can only be applied when the error is in the opposite direction, i.e. when overshoot has occurred. Therefore, oscillations are also likely to occur. In this situation the transfer function **R** might be:

$$\mathbf{R}(\varepsilon) = \frac{1}{T_I}\int(\varepsilon)\mathrm{dt}$$

where

ε = the input to the transfer function, namely the error (Ψ-Ψ^T)

and $\frac{1}{T_I}$ = integral gain constant

This method does also tend to respond quite slowly since it takes time for the integration to accumulate the errors.

---

[46] Non-linear friction occurs when more force is required to overcome friction for a body at rest than when it's moving.

One other improvement that can be made is instead of integrating errors for all time since the beginning it is better to forget the errors from the distant past. Only prior errors over a short period before the present should be integrated, otherwise, it may be responding to irrelevant errors.

Derivative Control:

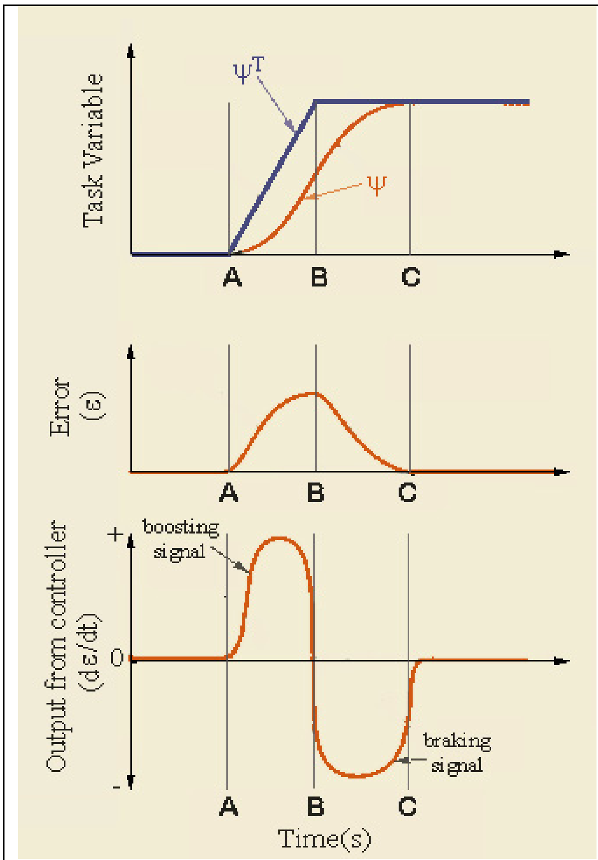


Figure C-1 - Example of how a system under Derivative control may perform

The main problem with integral control was its tendency to overshoot and its slowness to respond; hence, to resolve these problems derivative control was developed. As the name suggests the transfer function differentiates the error signal and applies a gain to it. The response tends to a bi-phasic output from the controller e.g. accelerating and then braking (see the illustrated example in Figure C-1). In this example, the set point is changed from one value to another over a short time interval (A-B). Initially there is no difference between the plant output and the target task variable/ set point, however as soon as the target starts to change there will be some error because it takes time for the controller and the plant to respond. The error begins to increase until a peak is reached around time B (note the timing of this error peak will depend on the plant dynamics and the gain of the feedback loop). Since the output of the controller is based on the gradient of the error slope, a peak restoring force (which pushes the system in the opposite direction of the error) is reached before the maximum error when the gradient becomes zero. After this peak, the force is rapidly reduced and then made to act in the other direction. This method has the advantage that it is therefore quick to respond and because an opposing force is also applied, it helps to prevent the system from overshooting. Derivative control does not however influence the accuracy of the system. It simply affects the response time so that this method is usually not used on its own.

For derivative feedback, the transfer function may be described by:

$$\mathbf{R}(\varepsilon) = K_D \frac{d\varepsilon}{dt}$$

where

K_D = derivative gain

Combined Proportional, Integral and Derivative Control.(PID):

As described, methods 3), 4) and 5) all have their advantages and disadvantages. Therefore with PID these methods are combined in varying proportions (according to the gain constants) in order to produce a more satisfactory response for the system. The final output from the feedback controller can be represented by:

$$\mathbf{R}(\varepsilon) = K_p\varepsilon + \frac{1}{T_I}\int \varepsilon \, dt + K_D \frac{d\varepsilon}{dt}$$

where K_P = proportional gain

$\frac{1}{T_I}$ = integral gain

K_D = derivative gain

ε = error between target task variable and resultant plant task variable

There are various iterative type methods, which are often employed to tune such PID controllers by adjusting the gain constants. The intention is to reduce instability and steady state errors. It should be noted that instabilities are not only caused by overshooting but also by phase lags which if sufficiently large can lead to positive feedback in which the controller output is in phase with the motion e.g. if the response is 180^0 out of phase with the target. In this situation, the controller will push the system further off course if the gain is at least one. Therefore, for a phase lag of 180^0 the gain must be less than 1, which means that the system's frequency response must be assessed since the phase lag will change according to the frequency. This would be very awkward to test for this project unless a good model can be produced of the entire process including the plant dynamics.

Phase Dependent Feed-back (PDF) Control:

As a final strategy, the combination of feedback methods may be varied according to the specific movement pattern that Ψ^T may be representing. This may allow for modulation of responses according, for instance, to the phase of gait (if walking has been broken down to several tasks according to phase). Therefore **R** is phase dependent and hence possibly time dependent. This more closely resembles the natural system where phase modulation of reflex arcs is believed to occur. In this situation, the coefficients for PID would be changed depending on the phase.

Appendix D ETHICAL APPROVAL FOR EXPERIMENTS

SGH

Application Ref________________

SOUTHERN GENERAL HOSPITAL NHS TRUST

ETHICS COMMITTEE

APPLICATION FOR APPROVAL OF A CLINICAL RESEARCH PROJECT

To: The Secretary
Medical Ethics Committee

Notes: i) All applications should be on this form

ii) This form should be typed

iii) "See protocol" is not an acceptable answer to any question. A summary sheet must be provided and other questions answered in full. One copy of any protocol should be provided for reference when necessary.

iv) Separate patient consent forms and information sheets should be provided with the application. In some cases these may not be applicable or necessary in which case a reason for not providing them should be given.

1. Title of Project:

Design of an Adaptable controller for the knee mechanism in a trans-femoral prosthesis.

2. Date of Submission:

June 1997

1

3. **Name, Personal Qualifications, Status of Principal Investigator(s):**

S.E. Solomonidis BSc, ARCST, CEng, M.I.Mech.E.
Senior lecturer, Bioengineering Unit, University of Strathclyde.
Bernard A. Conway BSc, PhD
Lecturer in Applied Physiology, Bioengineering Unit, University of Strathclyde
S.W. McCreath MBChB FRCS(Glasg), department of Orthopaedics, Prosthetics, Orthotics - Accident &Emergency, Southern General Hospital.
A Weir Bsc (Hons) MBChB FRCS Edinburgh MRCGP Dip Med. Rehab
Rehabilitation Unit at the Southern General Hospital
W.D. Spence MSc
Prosthetist/Orthotist, Bioengineering Unit, University of Strathclyde, 20 years experience

4. **Other Personnel Involved:**

B Altan BSc (Physics), PhD student, Bioengineering Unit
University of Strathclyde.

5. **Department(s):**

WESTMARC - Belvidere and SGH
Bioengineering Unit, University of Strathclyde

6. **Date discussed and approved by Department or Division:**

Jan - April 1997

7(a) **Has the proposed research been submitted to any other Ethics Committee?**

Yes ☐ No ☑

If "Yes", give details:

2

(b) **Has the proposed research been approved by any other Ethics Committee?**

Yes ☐ **No** ☑

If "Yes", give details:

8. **Has any similar research been carried out in any other Centres?**
Give details:

Not to our knowledge

9. **Describe the purpose, medical and scientific value of the investigation:**

Trans-femoral (AK) amputees prostheses suffer several draw backs due to the general lack of adaptability in most conventional prostheses. They generally are not easily able to vary their speed of walking away from the normal, and their style of gait is not as good as it might be for example during stance where most prostheses simply lock. It is therefore the aim of this project to develop a controller, for a variable friction knee mechanism, which will enable more adaptive movements to be performed by the amputee. To do this will require data on the gait of trans-tibial amputees (BK). This data should show how the knee should be behaving during various situations for a leg with a prosthetic foot. It should then be possible to develop a control scheme which will give the trans-femoral amputee similar gait characteristics resulting in an improved gait at a range of speeds. The BK data may also be useful for other investigators since a lot of the data will describe many aspects of gait.

10. <u>**Number of patients involved and duration of project:**</u>

No. of patients (this Hospital):	20 BKs, 6 AKs, 10 non amputees
Duration of Project:	2 years

11. Methods - Non-invasive:

Data will be collected on the gait of trans-tibial amputees who have had an amputation for non-vascular reasons and have no known pulmonary, cardiac or other condition limiting their test performance. Non amputees braced to try and reproduce BK motion may also be initially used to set up the experiments. The experiments will be performed in the gait lab of the bioengineering department, Strathclyde University. A commercial video system (VICON) will record the 3D co-ordinates of body markers. Concealed force plates in the path of the amputee will record the ground reaction force. Transducers in the pylon of the prosthesis may also be used to record forces in the leg. The amputee will be asked to walk a distance of about 7m, repeating this at least 5 times for that speed. He/she will then be asked to repeat this for up to 5 other speeds whilst data is collected. Eventually once a control system has been designed and checked the system will have to be tested on a selection of fit trans-femoral amputees. To this a walk way with hand railings on either side will be employed for safety reasons.

12. Methods - Invasive (Indicate where these are over and above the normal treatment of the patient, e.g. venepuncture, endoscopy etc.):

None

13. List any drugs or non-standard products which are to be given for experimental purposes. Indicate whether or not a product licence has been obtained for the purpose for which the preparation is to be used. If a Clinical Trials Exemption Certificate has been obtained from the Committee of Safety of Medicine, please indicate:

None

a) Product Licence Obtained: Yes ☐ No ☐

b) Clinical Trials Exemption Certificate: Yes ☐ No ☐

14. If the project involves the administration of radioactive materials to human subjects, please indicate the material and the name of the certificate holder:

15. List any hazards to the patients:

No hazards are envisaged. There will be at least 2 members of the team present during the tests to ensure safety.

Tape used to secure markers may cause slight skin irritation. This should be no more than the irritation caused by a plaster bandage.

16. Is the work being funded from any source?

a) A grant distributing body, e.g. Research Support Group

Yes ☑ No ☐

State the Body:

University of Strathclyde, General Research Source

b) A Drug or Scientific Material manufacturers

Yes ☐ No ☑

5

c) State the fee provided and to whom it is paid:

17. Are the products used being provided by the manufacturer?

Yes ☐ No ☑

18. If any of the following departments are providing resources, has the Departmental Head been asked and do they agree to the use of their facilities, resources and expertise?

Biochemistry		Pathology	
Microbiology		Haematology	
Diagnostic Radiology		Pharmacy	
Medical Records		Secretarial Services	
Neurophysiology		Others (Specify)	

19 (a) Is any non-standard product or unusual use being made of a product or an investigation?

Yes ☐ No ☑

(b) Is the investigator or the hospital authority indemnified in the event of an accident?

Yes ☑ No ☐

20. Please state who will have access to the data and what steps will be taken to keep the data confidential:

Only the investigators named will have access to the data.
The subjects will be identified by numbers in any publication.

21. In research involving the extraction of data from records, please indicate whether or not patients will be identified and associated with any specific information obtained:

None

22. Where the research is epidemiological, please state what steps are being taken to inform patients of their rights not to participate or if it is thought that this is not necessary, state why. This could include research by questionnaire or by extracting information from records:

N/A

23. <u>Consent</u>

(a) Is a Patient Consent Form enclosed and a <u>separate</u> Patient Information Sheet?

Yes ☑ No ☐

If "No", please state why:

(b) If "Yes", is the investigator satisfied that the sheet contains all the relevant information that enables the patient to properly consent:

Yes ☑ No ☐

24. Layman's Summary (About 100 words):

Data on gait will be collected using below knee amputees. This data will be used to develop a control system for an above knee prosthesis so that the amputee will be able to walk as naturally as possible at any chosen speed. The experiments will involve the BK amputee walking a short distance at 5 different speeds. This may be repeated up to 5 times for each speed to give a good statistical sample. Although the below knee amputee may not benefit directly from these experiments such data may also be used by other researchers interested in BK gait. Such data would be invaluable to decide how an above knee amputee might improve his gait. A below knee amputee is needed for this experiment because they still have control of their knee without direct control of their foot. This project therefore aims to replicate such control for an above knee amputee. Once a control system has been developed AK amputees may also be used to test the system.

SGH

SOUTHERN GENERAL HOSPITAL NHS TRUST

CONSENT FORM

PATIENT NAME..DATE OF BIRTH......................

To be completed by the Patient

	Please Tick Yes	No
. Have you read the Patient Information?	☐	☐
. Have you had an opportunity to ask questions and discuss this study?	☐	☐
. Have you received satisfactory answers to all your questions?	☐	☐
. Have you received enough information about the study?	☐	☐

Who have you spoken to? Dr/Mr/Ms________________________________

Do you understand that you are free to withdraw from the study -

- at any time
- without having to give a reason
- and without affecting your future medical care? ☐ ☐
- Do you agree to take part in this study? ☐ ☐

Do you have any reason to believe you are or may be pregnant?

YES, I may be pregnant ☐

NO, I am not pregnant ☐

Signed.. Date.........................

Name in Block Letters..

Signature of Witness.. Date.........................

Name in Block Letters..

I am familiar with the Declaration of Helsinki and I am satisfied that the work fits the criteria embodied within that declaration.

Signed: A Solomonidis

Date: 13th June 1997

9

Information for Participants

Purpose of Study:

To develop an adaptable controller for a knee mechanism to be used by an above-knee amputee. Conventional above knee prostheses have many draw backs due to the passive nature of their response to the requirements of the amputee. For example using these prostheses an amputee is unable to easily vary his/her speed from the normal setting under various circumstances. Therefore it is the intention of this project to develop a more active control scheme which can cater for a larger range of requirements. To do this, data on below knee amputee gait will be collected so that a control strategy may be formulated for the above knee amputee. Although this study is not intended to directly benefit the below knee amputee the data obtained will be of great interest to other researchers who have a more direct concern with their rehabilitation.

Experiments:

Small markers will be attached onto you using tape, this will enable data collection. You will be asked to walk at a particular speed, a distance of about 7m whilst data are collected. This data collection will be repeated about 5 times for that speed. You will then be asked to repeat the procedure for another 4 speed settings so that data are obtained from your slowest gait to your fastest gait. It may also be necessary to have a force transducer on the pylon of the prosthesis. If this is the case a new prosthesis will be made for you. A test session will only require about half a day to perform.

As a volunteer you are free to demand that the experiment be stopped, and that you can withdraw from the study at any time.

SGH
Southern General Hospital
NHS Trust

1345 Govan Road Glasgow G51 4TF
Tel 0141-201 1100 Fax 0141-201 2998

Our Ref: LETJUNEFMCG.LC
Enquiries to: Mr Frank McGuire - Secretary
0141-201-1150

27 June 1997

Mr S E Solomonidis,
Bioengineering Unit,
University of Strathclyde,
Wolfson Centre,
106 Rottenrow,
Glasgow, G4

Dear Mr Solomonidis,

DESIGN OF AN ADAPTABLE CONTROLLER FOR THE KNEE MECHANISM IN A TRANS-FEMORAL PROSTHESIS. EC/97/S/59

Further to your recent application for approval of the above study, I am pleased to advise that full ethical approval has been granted.

On behalf of the Committee may I take this opportunity to wish you every success with your trial and request that on the conclusion of this study, a full report is forwarded to the Ethics Committee.

Yours sincerely,

Frank McGuire

FRANK McGUIRE
SECRETARY - ETHICS COMMITTEE

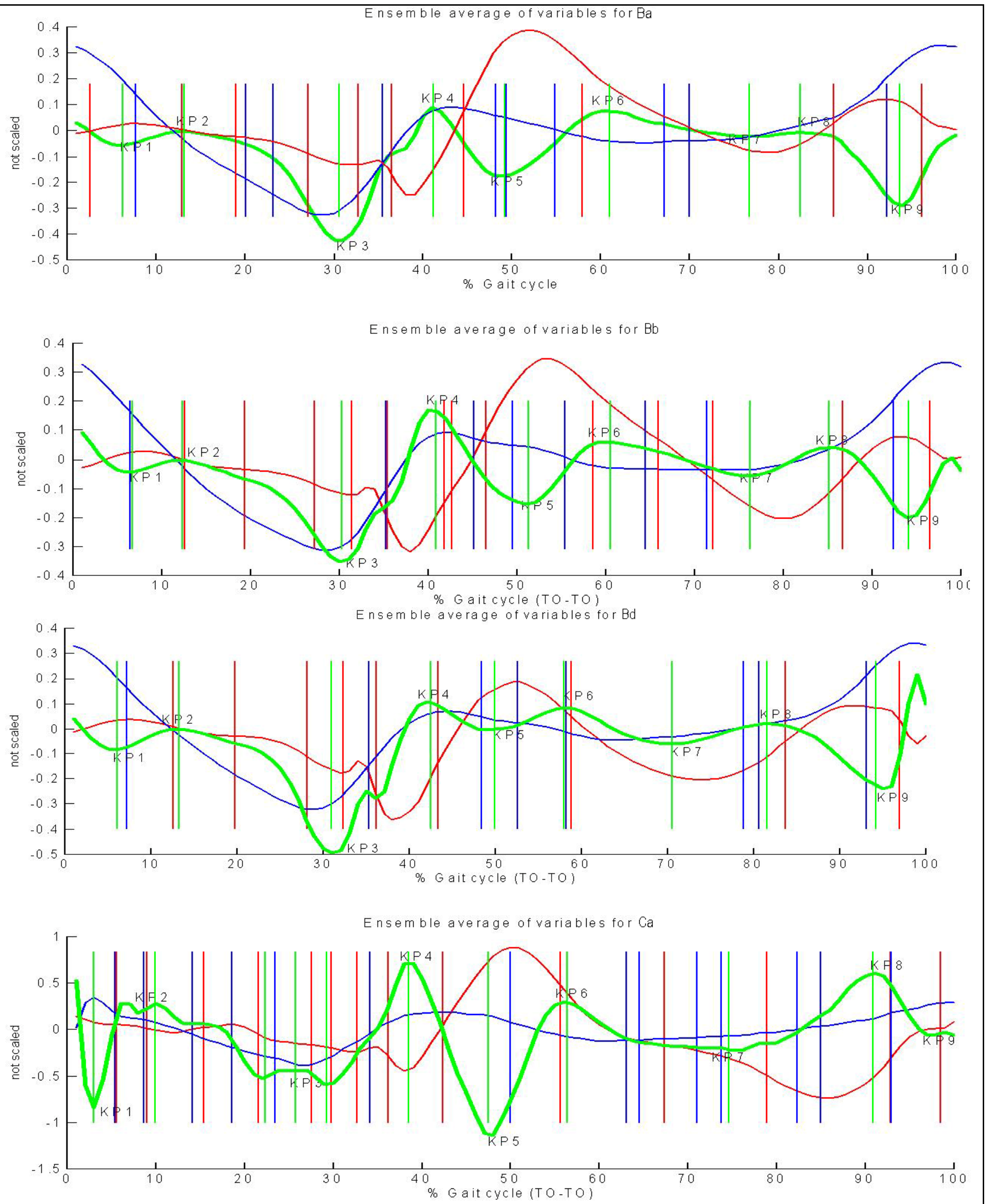
SGH 401 *(Incorporating Cowglen Hospital and the West of Scotland Mobility and Rehabilitation Centre)*
0368349 Continuing the Tradition of Developing NHS Services for the Community

Appendix E CONTROL VARIABLE STATISTICS

Session	mean Vicon speed	beam speed	speed diff	STD VICON	std beam	min Vicon	min beam	max Vicon	max beam	mean cadence	std cadence	min cadence	max cadence
Ca	1.435	1.41	0.025	0.204	0.197	1.12	1.14	1.85	1.78	0.887	0.084	0.765	1.063
Cb	1.424	1.404	0.02	0.281	0.273	0.987	0.997	1.89	1.852	0.87	0.1	0.721	1.038
Ea	1.158	1.118	0.04	0.28	0.252	0.651	0.662	1.74	1.583	0.773	0.111	0.604	0.978
Ed	1.186	1.161	0.025	0.399	0.385	0.637	0.622	1.831	1.781	0.748	0.148	0.492	0.974
Ee	1.223	1.178	0.045	0.336	0.314	0.669	0.667	1.71	1.648	0.768	0.141	0.517	1.191
Fb	1.246	1.255	0.009	0.143	0.138	0.983	1.014	1.452	1.445	0.83	0.05	0.723	0.927
Fc	1.334	1.319	0.015	0.27	0.23	0.906	0.95	1.738	1.628	0.82	0.079	0.649	0.925
Fe	1.108	1.099	0.009	0.279	0.239	0.792	0.823	1.806	1.62	0.733	0.101	0.591	0.916
Ba	1.153	1.139	0.014	0.183	0.178	0.843	0.876	1.494	1.427	0.879	0.098	0.746	1.029
Bb	1.119	1.104	0.015	0.085	0.089	0.961	0.962	1.258	1.233	0.857	0.063	0.757	0.949
Bd	1.147	1.145	0.002	0.114	0.117	0.971	0.965	1.329	1.33	0.867	0.077	0.74	0.98
Ha	1.172	1.162	0.01	0.124	0.13	1.014	0.998	1.511	1.52	0.826	0.043	0.772	0.926
Hc	1.281	1.248	0.033	0.181	0.177	0.985	0.983	1.559	1.504	0.878	0.065	0.771	0.98
mean	1.227	1.208	0.019	0.221	0.209	0.886	0.897	1.628	1.565	0.824	0.089	0.68	0.989

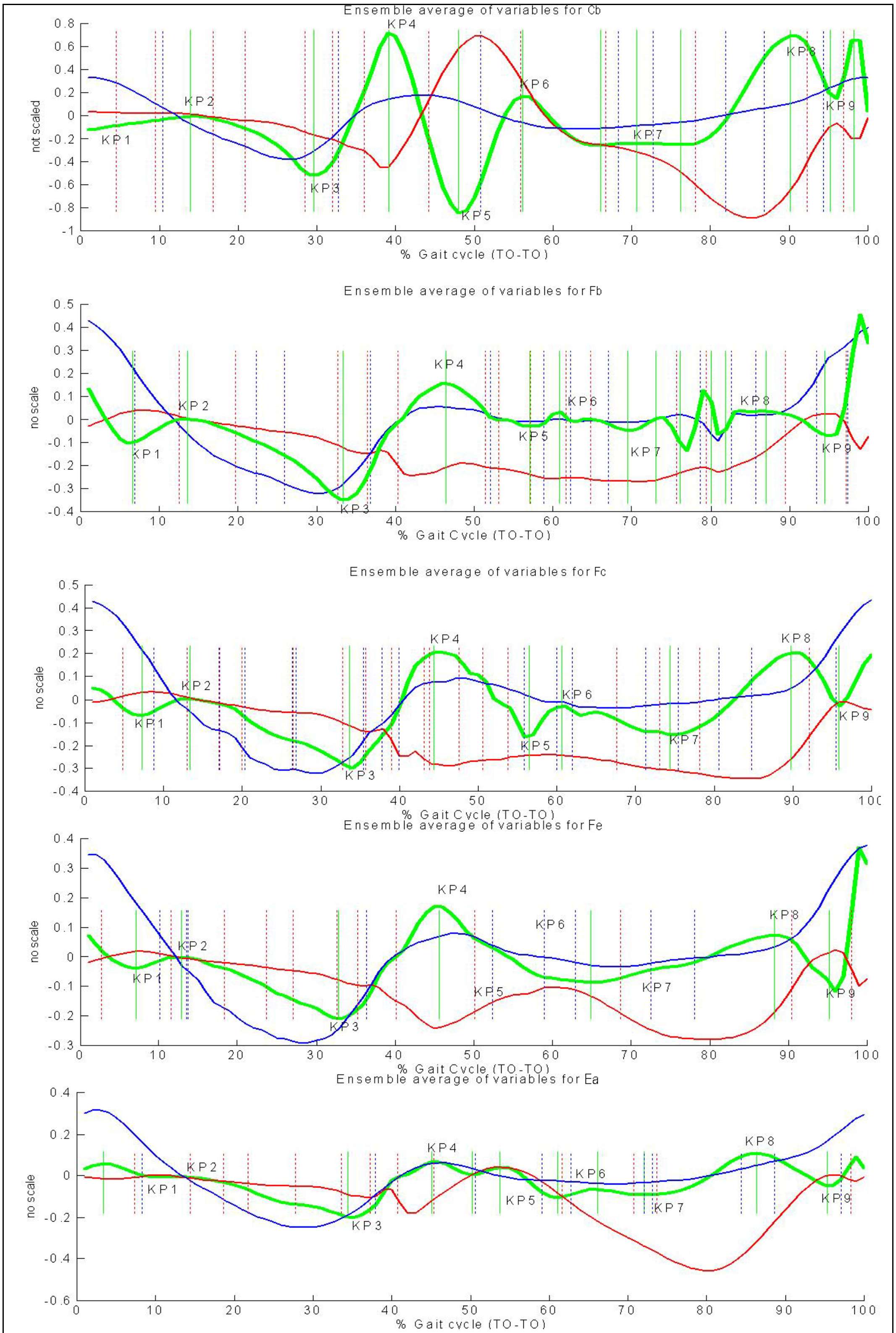
Table E-A – Statistics for all the trials during a session. The suffix "Vicon" refers to speeds that were calculated from Vicon trajectories and "beam" refers to those measured using the IR beam timing mechanism [section 7.2.3.5].

Appendix F INTENT STATES



Cont...

Appendix



Cont...

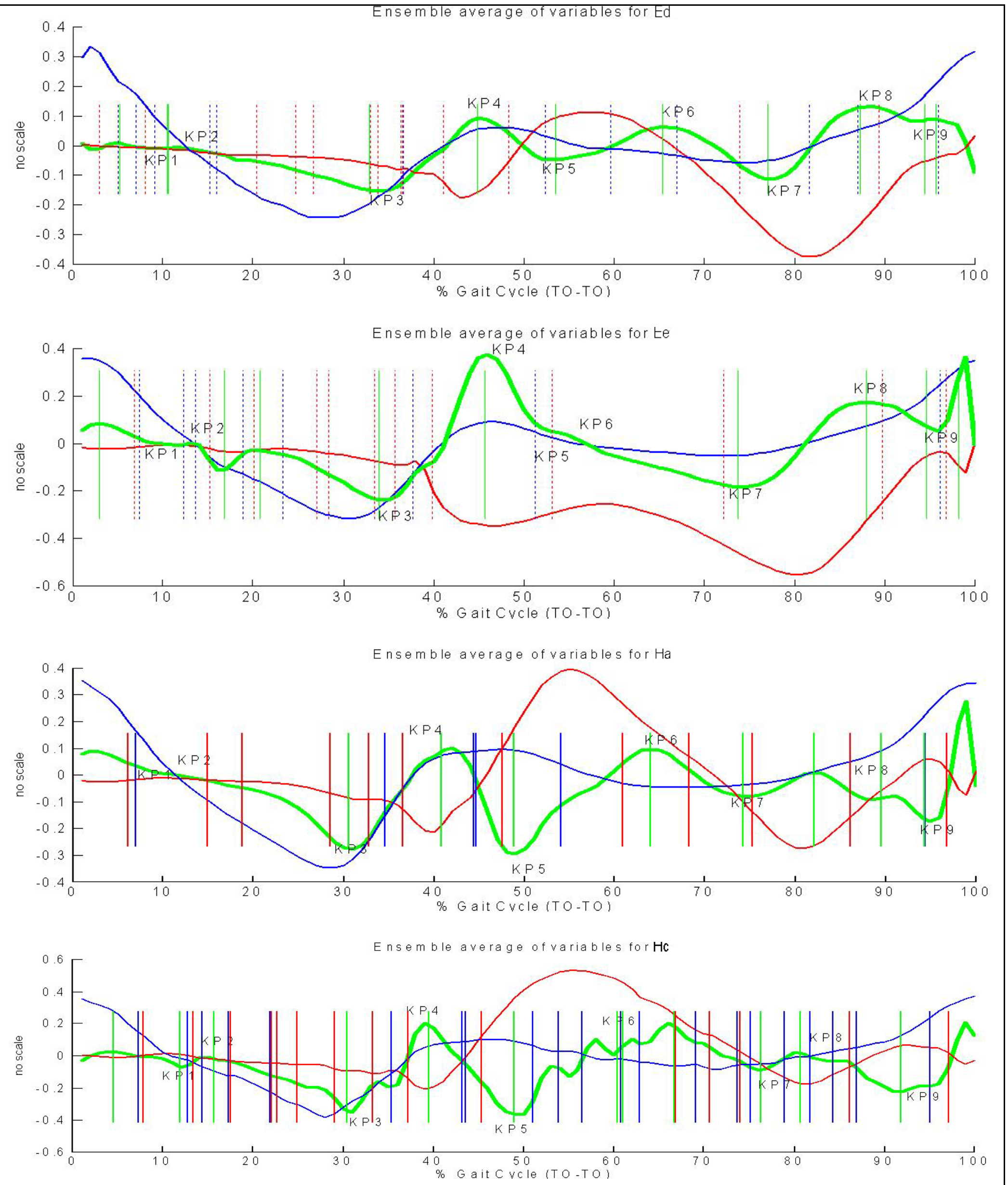


Figure F-1; Ensemble averages of the knee - power (green), moment (red), angular velocity (blue) superimposed upon each other. The data for each session is shown and the topological features that were thought to be viable intent states have been labelled. Note that these should ideally be consistent from session to session.

Appendix G PERFORMANCE SUMMARY OF NETWORKS

Input Variables	Input Variable Groupings 1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19
Ank fl/ex angles									X					X		X	X	X	
Ank ab/ad angles													X						
Ank in/ex angles																			
Kne fl/ex angles	X	X		X	X				X	X	X	X	X						X
Kne ab/ad angles																			
Kne in/ex angles																			
Hip fl/ex angles	X	X	X	X	X	X		X	X	X	X	X	X		X	X	X	X	X
Hip ab/ad angles																X	X	X	
Hip in/ex angles				X	X						X	X	X	X				X	X
Ank fl/ex ang vel													X						
Ank ab/ad ang vel																			X
Ank in/ex ang vel				X	X						X	X	X	X					
Kne fl/ex ang vel		X	X			X	X	X			X		X		X	X	X	X	X
Kne ab/ad ang vel				X	X														
Kne in/ex ang vel				X	X														
Hip fl/ex ang vel	X	X	X			X		X		X	X	X			X	X	X	X	X
Hip ab/ad ang vel									X					X					X
Hip in/ex ang vel					X				X	X						X	X	X	
Foot fl/ex ang vel														X					
Foot ab/ad ang vel																			
Foot in/ex ang vel																			
Pyln fl/ex ang vel	X	X	X	X	X	X				X	X	X	X		X	X	X	X	X
Pyln ab/ad ang vel																			
Pyln in/ex ang vel								X											
Thigh fl/ex ang vel							X						X	X				X	
Thigh ab/ad ang vel								X				X							
Thigh in/ex ang vel				X	X	X			X		X	X	X		X	X	X		X
Foot fl/ex ang acc																		X	
Foot ab/ad ang acc																			
Foot in/ex ang acc																			X
Pyln fl/ex ang acc					X		X												
Pyln ab/ad ang acc																	X	X	
Pyln in/ex ang acc																			X
Thigh fl/ex ang acc					X														X
Thigh ab/ad ang acc							X												X
Thigh in/ex ang acc							X	X						X					X
Ank fl/ex force							X	X		X									X
Ank ab/ad force			X					X		X					X		X		X
Ank in/ex force		X						X		X									
Kne fl/ex force						X					X								
Kne ab/ad force							X							X				X	
Kne in/ex force							X							X					
Hip fl/ex force																			
Hip ab/ad force																			
Hip in/ex force																			
X Gnd React force									X										
Y Gnd React force				X	X	X			X		X	X							X
Z Gnd React force					X				X					X					

Table G-A; Table showing the different groupings of the variables that were used as inputs to the networks. Note that the variables that were used for group 4 and 6 have been coloured blue and green because these were noted to be promising selections; orange was used for variables common to both.

	Codes for training session groups												
	Ic	za	zd	ze	zf	zg	zh	zi	zj	zk	zl	zm	zn
Training Sessions	Bd	Cb	Bd	Cb	Ba	Ee	Bb	Ba	Bd	Bd	Ba	Ba	Ba
	Cb	Fe	Fc	Hc	Ea	Fe	Ed	Bd	Cb	Fe	Ed	Ea	Bd
	Ed		Ed		Ha	Hc	Ha	Cb	Fe	Ha		Hc	Ea
													Hc

Table G-B; List of prefixes used to described different groups of sessions used to train the network

Table G-C: This table is on the Excel spreadsheet file named "Baran Thesis summarised performance of all network structures.xls", which is available on the CD insert. The table gives the summarised performance of all network structures. For each structural type, separate networks were trained and tested to detect each of the nine intent states (KP1 – KP9). Values above 190% were categorised as "V Good", above 160% "Good" and below as "bad". Tallies were made of the number of test sessions that produced "V good" and "Good" results from each individual network.