An Investigation into bench testing of Microprocessor Controlled Prosthetic Knee Joints

Mayank Seth BSc

This thesis is submitted in partial fulfillment of the requirements for the award of MSc in Bioengineering

16 / 08 / 2011

Bioengineering Unit

University of Strathclyde

Glasgow, UK

August 2011

COPYRIGHT

Signed:

Date:

ACKNOWLEDGEMENTS

I am deeply indebted to my guide Mr. Stephanos Solomonidis for his invaluable suggestions, patience and moral support throughout this endeavor.

I acknowledge the support and camaraderie of my friend Mr. Kaustav Ray who managed to help me even while having his own project to look for.

I am indeed grateful to the teachers and staff of the Bioengineering Department, University of Strathclyde for making this effort possible

Lastly i wish to record my gratitude to my parents who upheld and applauded all my proposals and time and again supported me with their wisdom and timely reminders.

ABSTRACT

Prosthetics has its share of technological advancements more than many other fields of rapidly evolving Bio-Medical Engineering. Microprocessor controlled prosthetic knee joints represent one such advancement. They can impart an almost natural physiological rhythm to an amputee's gait. However, these joints are still short of perfection, despite a lot of research having gone into their production.

The evaluation and testing of the artificial knees is usually carried out on live patients. This generally hinders research. It is time consuming and tiresome for the patients to constantly adjust their gait for different systems during the testing procedures. It is expensive too.

The aim of this project is to look into the functionality of the Microprocessor controlled prosthetic knee joints based on knowledge of human gait cycle and subsequently presenting specifications of a machine for testing different parameters on the Prosthetic Knee joints. Existence of a bench tester would reduce the loss of working time for patients. They need not constantly be present for evaluation and bear the load of testing each system. A literature survey indicated that there are no such testing systems of this kind readily available till date.

Contents

CHAPTER 1. INTRODUCTION

The science of replacing a lost limb with another, functionally active or inactive, device is termed Prosthetics. The history of Prosthetics goes back many, many centuries and there has been mention of artificial limbs even in ancient texts like the *Vedas*. This science has both attracted and baffled people. To make an amputee walk with a metal or a wood assembly is not an easy task; getting him used to it, so as to accept it as his own limb is even tougher. From the technical aspect prosthetics has always generated curiosity. It is this intense desire to know that has led to pioneering developments in the field. The past few decades, as in other fields of medicine, have shown a rapid increase in research in the lower limb prosthetic segment.

A lower limb prosthesis has majorly two functions: support the upper body in stance and provide a comfortable gait. Man is the only quadruped to have mastered the ability to stand and walk two legged! Both functions are equally important. However, it is the former that is of greater concern to the amputee. One relies insensibly on the knee joint, along with its muscle complex, for stability and balance in the erect posture, while standing and walking. Reflex correction of balance to keep center of gravity of the body steady by quadriceps, calf muscles and hamstrings keeps a person confidently standing. Fear of misbalancing and falling is very real and constant for fresh amputees. This is true most of all for an Above Knee (AK) amputee in whom the insecurity and instability increases tremendously. In the AK prosthesis the knee unit is the most vital part. In the absence of the innate knee joint the amputee has to be solely balanced by the mechanical knee unit. Thus, it is important for all the components within a prosthetic knee system to be structurally and mechanically sound, for the amputee to be able to develop confidence and security. Consequently, it is crucial to test the knee component of a newly designed AK prosthesis in all aspects prior to applying it on an amputee.

Bench testing is the first step in the evaluation and analysis of a knee joint. It is a good way of ensuring the correctness & soundness of the knee unit. In the past few decades quite a few tests have been carried out to check for the fatigue life of prosthetic knee joints.

Work done at University of California, Berkley, 1947 still stands as the basis of prosthetic testing. It also aids understanding of the gait cycle. However, the major concern at that time was the life of the prosthesis and majority of tests were conducted to test the fatigue of the prosthetic systems instead of their ability to provide the biomechanically correct knee motion. Nowadays there are knee systems which have the ability to replicate natural knee motion without conscious effort from the amputee. These are referred to as the Microprocessor Controlled (MPC) Prosthetic knee units. These units provide a variety of functions that rely on the data processing and commands of an on board microprocessor. These are a complicated and sophisticated genre of appliances. It is important for the researcher to test them for their stabilization features.

A lot of work has been done in the past on the fatigue of earlier designs of prosthetic knee units. The thoroughness of those designs led to the present highly efficient units. However, it was difficult to locate recent studies which enquire into pre-testing of MPC prosthetic knee units. It is possible that manufacturers might be having equipment for the bench testing of their specific design. But published data was hard to find. This point has been of particular motivation for the project.

1.1 OBJECTIVES

- To try to review as many bench tester designs made in the past. Not a lot of work has been done in this aspect, but any kind of past work should turn out to be helpful for the project
- To present a good understanding of the method of operation of the Microprocessor Controlled Prosthetic knee units
- Producing essential information about the human gait biomechanics, which will be helpful in setting some standards for testing a prosthetic joint
- To produce a possible idea for a bench tester which can be helpful for the testing of Microprocessor controlled prosthetic knee units

1.2 AIMS

The project necessarily deals mainly with understanding the basics of prosthetic knee joint testing and suggesting a plan for a bench tester which can be used to test properties specific to a Microprocessor controlled prosthetic knee joint. Presenting the manufacturing details of the plan however, is not a part of the project.

1.3 CHAPTER ORGANISATION

Chapter 2 will deal with a brief history of knee joint testing and the apparatuses which were made use of for this purpose. Also, it is worth mentioning some studies which were done on patients fitted with a Microprocessor Controlled knee unit.

Details of the microprocessor mechanism within the knee joint will be talked about in Chapter 3. The chapter mainly concerns itself with two Microprocessor controlled prosthetic knee units, the C-Leg® by Otto Bock and the Rheo® by Ossur. Both the systems present different mechanisms for providing variable damping.

The kinematics and kinetics of the human gait is covered in Chapter 4. This also talks about the normal gait and how the three major joints of the lower limb change their positions throughout the cycle. For testing the knee joint certain parameters need to be decided. The chapter will deal with the parameters and their importance.

Finally, Chapter 5 will talk about the proposed plan for the tester. Care will be taken to describe all the components of the design and how all the proposed parameters can be tested. It has not been possible to include a few things in the plan such as the controller circuit for the machine, the material considerations etc. these will talked about as future work for the machine.

CHAPTER 2. LITERATURE REVIEW

The following chapter concerns itself with a review of studies which relate to the project. The ideas presented have been of assistance in the work. Not a lot of data was available on Bench testing of prosthetic knee joints, but there have been certain studies dealing with fatigue testing of prosthesis. These have been included. Also, stress has been given on the studies concerning themselves with the functioning of microprocessor controlled knee joints.

2.1 TERRAIN IDENTIFICATION FOR PROSTHETIC KNEES BASED ON ELECTROMYOGRAPHIC SIGNAL FEATURES. *Jin et all 2006*

2.1.1 STUDY GOAL

The study aims to record and evaluate the EMG signals from a set of Hip muscles from a group of healthy individuals. The obtained EMG signals can then be used for terrain identification by a MPC knee joint that adapts to different conditions.

2.1.2 STUDY DESIGN

2.1.2.1 TEST SUBJECTS – 13 healthy individuals and 1 unilateral amputee

2.1.2.2 IMPLEMENTED IN – laboratory settings

2.1.3 METHOD

For the purpose of the study EMG signals were recorded from 8 different muscles adductor poilicuslongus (ADDL), tensor fascia laiae (TFL), rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris

(BF), semimembranosus (SM), and semitendinosus (ST) of the healthy individuals and of the normal side of the amputee.

2.1.4 RECORDING AND EVALUATION

The subjects were asked to walk 3 – 5 times on a walkway 8m for all of the patterns. The experimental setup consisted of the walkway, a force platform, a camera, two A/D convertors, and one PC computer.

The signals from the muscles were recorded when the patients were asked to walk up and down stairs, up and down ramps, and on level ground with different speeds.

The features of the signals were used to create a terrain identification model. However, the model was prepared from only the readings of the RF, VM, and SM for practical purposes. The identification process was then implemented in an intelligent prosthetic knee developed for laboratory tests.

2.1.5 RESULTS

With the obtained signals it was easy to assess the activity of different muscles. Also, the different waveforms of a single muscle under different walking patterns.

The MPC knee joint managed to present gait patterns which were very similar to the ones expected by the authors. They were able to conclude that the method used for terrain identification by the MPC was correct.

2.1.6 PERSONAL IMPRESSION

The experiment certainly presents a very good way of implementing terrain identification in the MPC knee joints. However, the whole process consisted mainly of individuals who had not experienced amputation. The data obtained thus, was more from normal individuals. But, the walking pattern of an amputee with or without a MPC knee may be very different from that of a normal individual. Therefore, in practical usage it will be more convenient to make use of data taken from the hip muscles of an amputee, so as to avoid any in congruency between the MPC knee and the normal knee of the patient.

Further, in an amputee hip muscles contributing the most are likely to have been severed. Therefore, while making the terrain model it will be beneficial to make use of data from the other muscles also, as a few of them are higher up and are spared during the trans femoral amputation.

2.2 EVALUATION OF FUNCTION, PERFORMANCE AND PREFERENCE AS TRANS FEMORAL AMPUTEE'S TRANSITION FROM MECHANICAL TO MICROPROCESSOR CONTROL OF THE PROSTHETIC KNEE. *Hafner et all 2007*

2.2.1 STUDY GOAL

An extensive examination and evaluation of unilateral trans femoral amputees shifting from the use of a mechanically controlled prosthetic Knee joint to a Microprocessor controlled (MPC) and guided prosthetic knee system namely the C-Leg®. It was also a part of the study to record the responses of the subjects with regard to their preference.

2.2.2 STUDY DESIGN

2.2.2.1 TEST SUBJECTS – 21 individuals around 18 years of age were taken for the test, having 2 to 3 grades of mobility (Otto bock mobility grade).

2.2.2.2 IMPLEMENTED IN – day to day life and laboratory settings

2.2.3 METHOD

The process was divided in four stages

1) In the first part of the study the individuals used the mechanical joints for 2 months. During this process the patient activity was recorded. The function and performance of the joint was noted. Also, the joint was assessed post the walking period.
2) The next stage consisted of the individuals walking on the MPC knee joint for the same time period. And a similar assessment was done.

3) The third stage required the individuals to adapt to the mechanical joint for a period of two weeks
4) In the final stage the subjects were able to choose between their knee joints for a period of 4 months, with subsequent recording of the functionality of the joints.

2.2.4 RECORDING AND EVALUATION

The subject's activities were recorded during walking in different patterns. The study required them to walk on level ground, up and down slope, up and down ramps and on uneven terrain. An innovative method for data recording was to make them walk while talking on phone with the examiner.

For the purpose of recording the step frequency the StepWatch2 step activity monitor was utilized. Base modality was performed by therapists with the help of Amputee Mobility Predictor (AMP).

The evaluation of the performance capabilities and satisfaction levels were done by the patient himself with the aid of a Prosthesis evaluation sheet.

2.2.5 RESULTS

The test results showed a slight increase in the step length with the C-leg as compared to the unaffected side. Walking down stairs and ramps took a vastly different time by the two knee systems. The required time and the evaluation index were significantly less with the C-Leg. The required time for passing through the obstacle course and the evaluation of split attention were better with the C-Leg. On the patient preference side, a majority of the subjects liked the C-Leg more, reflecting the improved capabilities and satisfaction levels attributed to it.

2.2.6 PERSONAL IMPRESSION

The study highlights the fact that the MPC knee joints may enhance the capabilities of the patients and sometimes even bring out a few capabilities. Also worth mentioning is the experience of the patient undergoing the change from a mechanical to a MPC knee joint.

The differences between the two systems were also well brought out.

An important aspect here is the short time period given to the patients for the use of the mechanical joint. This might have affected the results to a certain degree, with a longer test period the results could have been less divergent.

The authors do not specify which mechanical knee units they used. If certain hydraulic knee units were to be used which are pretty similar to the MPC units in function, a better evaluation could have been done.

2.3 KINEMATIC AND KINETIC COMPARISONS OF TRANS FEMORAL AMPUTEE GAIT USING C-Leg® AND MAUCH® SNS PROSTHETIC KNEES. *Segal et all 2006*

2.3.1 STUDY GOAL

The study aimed to make the use of biomechanical parameters to measure out the functional differences between two prosthetic Knee units - The Mauch® SNS and the C-Leg® prosthetic knees.

Figure 2.1Mauch SNS and C-Leg (picture adopted from mayoresearch.mayo.edu)

2.3.2 STUDY DESIGN

2.3.2.1 TEST SUBJECTS – Initially 12 unilateral amputees were selected. 8 of them managed to complete the study. Additionally 9 healthy individuals were also selected to form a control group which provided for the baseline measurements.

2.3.2.2 IMPLEMENTED IN – day to day life and laboratory settings

2.3.3 METHOD

All the chosen amputees were regular users of the Mauch® SNS for 1 year. From the group 2 amputees were enrolled randomly and were given 1 month period with the Mauch® SNS, to take baseline measurements. After this one of them was randomly chosen to use the C-Leg®. The subjects used the respective prosthesis for 3 months and at the end of this the lab measurements were done. A crossover was made and again the subjects were given 3 months to acclimatize with the prosthesis and later the data was collected.

2.3.4 RECORDING AND EVALUATION

The subjects were supposed to walk on level surfaces with a controlled walking speed and other speeds which were comfortable for them.

The data recording for evaluation was done with the help of 10 camera Vicon 612 with 120Hz recording frequency, 1 force measuring plate with a sensing rate of 600Hz, marker model Plug-in-gait by Vicon.

2.3.5 RESULTS

The results for the C-Leg® data were based on the Controlled walking speed only, as the individuals tended to walk faster while using the C-Leg®. Thus, by using selected parameters it was found that the step length with the C-Leg® was shorter and the difference between the prosthetic side and the healthy side was less, showing an increased symmetry of the limbs

For the C-Leg® the knee flexion momentum is less at the beginning of the stance phase in comparison to the control group, but greater than the Mauch® SNS. This has been evaluated to bring out a higher potential for stance phase flexion on level walking with the C-leg®.

2.3.6 PERSONAL IMPRESSION

The study reiterates the fact that a MPC knee joint has a superior control over the users' gait as compared to a mechanical joint. The stability and ease of ambulation provided by the MPC knees certainly surpasses that of the others. A majority of the test subjects at the end of the research preferred the C-leg® over the Mauch® SNS.

However, one point to be noted here is that the study utilized the data from the controlled walking speed (CWS) and the other faster or slower speeds were not taken into consideration. The results for that might differ surprisingly.

One thing the authors do mention in the end is that the much talked about reduction in joint muscle effort by the C-Leg® was not experienced. For this to happen, one has to be able to record a reduction in the knee movement in the sagittal and coronal plane. But, the knee movements for the joints were the same in this respect. This makes it hard to draw a conclusion about the required muscle force. Here again testing with higher speeds might bring out differences.

2.4 “What does the C-Leg® achieve?” – A GAIT ANALYSIS COMPARISON OF C-Leg®, 3R45, 3R80. *Kastner et all 1999*

2.4.1 STUDY GOAL

To establish the differences between the C-Leg® and two more mechanical knee joints with the help of data obtained from gait analysis

2.4.2 STUDY DESIGN

2.4.2.1 TEST SUBJECTS – 10 unilateral subjects were chosen for the study

2.4.2.2 IMPLEMENTED IN – laboratory settings

2.4.3 METHOD

The data gathering was done while the patients were walking on treadmill (3, 4 or 5KmpH). The analysis involved patients doing uphill and downhill walking on the treadmill. Swing phase behavior was investigated using kinematic data gathered. Also the heart rate was monitored during the test for investigating the load intensities

2.4.4 RECORDING AND EVALUATION

Kistler force plates were made use of for determining the position of the vector during loading of the prosthesis, 2D kinematic analysis, treadmill.

The patients were not given much time for familiarization with the prosthesis and were randomly given the prosthesis without prior information for the loading test. The treadmill test required them to get to know the system, but that was also for 10 minutes. Measurements were made in the following sequence: loading, swing phase control, treadmill (uphill/downhill), 1,000 m field test.

2.4.5 RESULTS

For the loading sequence significant differences could not be found. Load time, total and medium load were similar for all the systems. The authors also mention asymmetry between the prosthetic side and the normal side for the C-Leg®. The maximum flexion angle however for the C-Leg® was very minimum and comes very close to the one recorded for the normal side. The same applies to extension and flexion speed which also come very near to the normal side. The thigh angle for the C-Leg® is very less compared with the other joints. This shows a reduced activity by the patient. Within the uphill and downhill test no significant differences were found. However, in the 1000m field test all the subjects walked faster with the C-Leg®.

2.4.6 PERSONAL IMPRESSION

The study does establish some plus points with the use of the C-Leg®. Also, it is helpful in learning about the differences between C-Leg® and a few of the mechanical joints. The study however fails to produce big advantages of the C-Leg® over the other joints.

Due to lack of any acclimatization of the patients to the assemblies the C-Leg® shows asymmetry with the healthy side. Also, use of treadmill for the test is not a very good option. The data gathered cannot be taken for as data on a normal ground walking.

The 1000m field test was a good way of showing how comfortable the patients can get whilst using the C-Leg® and thus the increased speeds.

2.5 A CADAVER KNEE SIMULATOR TO EVALUATE RECTUS FEMORIS TRANSFER. *Anderson et all 2009*

2.5.1 STUDY GOAL

Spasticity of the Rectus femoris is a common observation of CP patients and is attributed to be the main cause of stiff knee gait. Concomitant RF transfer and hamstrings lengthening surgeries are often performed to improve the gait patterns of patients with these gait abnormalities. Clinical studies of the gait of the patients post transfer have not been able to demonstrate satisfactory results of any kind. For the same purpose cadaveric knee simulator has been developed to model the surgical transfer of rectus femoris.

2.5.2 STUDY DESIGN

2.5.2.1 DEVELOPMENT OF KNEE SIMULATOR

The simulator was encased within a rigid load frame. Hip and ankle gimbals were present within which was mounted the cadaver knee specimen for testing and several pistons and motors to replicate the contractile action of the various thigh muscles (Fig 2.2). The hip gimbal rotated in the frontal and sagittal planes and translated vertically. In stance the vertical translation of the gimbal made changes in the hip, knee and ankle flexion angles. While adhering to the natural anatomy of the lower limb, the frontal plane axis of the hip gimbal was displaced 4.4cm offset from the femur shaft to create the natural offset made by the femoral neck. The ankle gimbal had similar motion to the hip gimbal. The knee simulator thus had 6 degrees of freedom.

A piston (unsure if Pneumatic or Hydraulic) was mounted 34mm anterior and aligned with the axis of the femoral shaft. This modeled the combined extension effort of the three vasti muscles. The location corresponds to the origin of the vastus intermedus. A 4KN load was mounted to measure the applied load

A second piston mounted 34mm posterior and along the axis of the femoral shaft modeled the behavior of the hamstring muscles. This position comes between the origins of the biceps femoris and semimembranosus. A 2.2KN load cell was mounted for measuring the applied load.

The action of the RF muscle was modeled by a pneumatic piston, which was fixed in a yoke and that rotated in the sagittal plane independent of the Hip gimbal. The yoke was mounted 27mm superior to and 15mm anterior to the sagittal axis of rotation of the hip gimbal at a location corresponding to the origin of the RF muscle. A 2.2KN mounted load cell measured the applied load.

Figure 2.2 Knee simulator (adopted from Anderson et all 2009)

2.5.2.2 SWING PHASE SIMULATION

To model the swing phase a pendulum model of swing was created by mounting the thigh segment of the hinge knee (made earlier) on the hip gimbal and attaching a 44N weight to the end of the shank to act as foot. The motors at the hip gimbal were made to apply hip flexion and move the thigh from 10° to 30° of flexion. The body sensors were used to calculate the Range of motion and the total peak knee flexion angles.

2.5.2.3 STANCE PHASE SIMULATION

The major aim was to understand the force required to maintain knee extension for stance phase. This was carried out under 4 loading conditions: vasti only (V), vasti with hamstrings cocontraction (VH), vasti with hip and ankle torque (VT), and vasti with hamstrings cocontraction and torques (VHT) was measured. The data was compared with an analytical model.

Stance phase model was prepared by mounting the thigh and shank segments of the hinge knee to the hip and ankle gimbals. The vasti and the hamstring cables were connected to their respective pistons. The upper body weight of a normal male (174N) coming onto each limb was added to the sliding stage of the hip gimbal. The vasti actuator applied tension to extend the hinge knee from 50° to 10° at a rate of .75°/s. The amount of force required to complete this motion was recorded by the vasti actuator load cell.

For the VT and VHT hip and ankle torques were measured through an in vivo motion analysis study. 11 healthy males were taken for this study. The subjects were required to squat from a standing position by slowly flexing the hip, knee and the ankles. They were free to move the upper body parts. A motion capture system tracked the motion with the help of body markers. Ground reaction forces were recorded by force plates. Following this a biomechanical model calculated the full body torques on the hip and the ankle segments and applied this to the knee hinge model

2.5.4 RESULTS

During the swing phase simulation the initiation of hinged knee flexion coincided with the initiation of hip joint flexion. The reported total knee peak flexion occurred significantly earlier than the one in the simulated hinged knee models.

The knee extension forces to maintain the knee stance were not different from the ones produced in the analytical model, showing that the knee simulator did not apply artificial constraints that would lead to knee extensor force error

2.5.5 PERSONAL IMPRESSION

The above study does not deal directly with the concerned project at hand. However, my interest lies in the knee simulator developed by the researchers. It does give some ideas about the loading of the different lower limb segments. The use of pistons for modeling the muscles is an excellent method. This replication of the lower limb anatomy does help to produce near to normal forces on the knee.

The part that was a bit unclear was about the linkages of the different segments to each other as well as the gimbals. The use of gimbals is very unique idea unlike other knee simulating machines. But, it is hard to understand if the authors are making full use of the gimbals i.e. using full revolutions or just producing pendulum motions. These loopholes force one to make one’s own picture of the machine and its operation.

2.6 ABOVE KNEE TESTER. FUNDAMENTAL STUDIES OF HUMAN LOCOMOTION. *University of California, Berkeley, 1947*

2.6.1 STUDY GOAL

The study was involved with accelerated dynamic testing of Above knee (AK) prosthesis. For this purpose a testing machine was made.

My interest lies in the design of the machine.

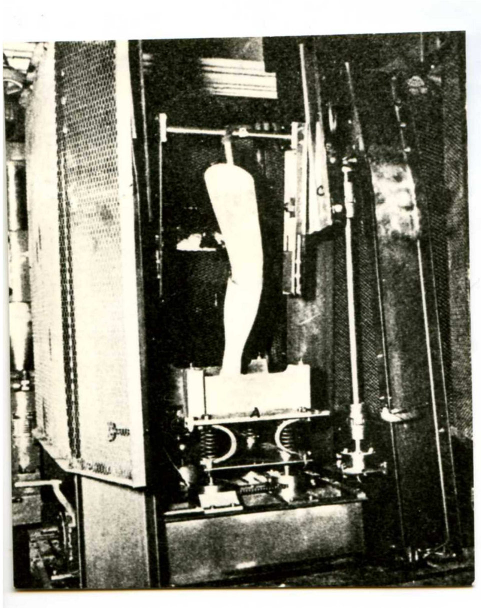
2.6.2 MACHINE DESIGN

The initial work by the organization consisted of developing a machine for testing of the pylon forces by producing cyclic loading conditions. Subsequently they came onto the testing of AK prosthesis.

The aim of the above knee tester was to model the walking pattern of a normal limb. To do this they made use of the elliptical pathway followed by the trochanter. This was found by doing studies of the trochanter with the toe fixed in a position.

The prosthesis to be tested is placed between a bar and a foot plate. The motion is produced at the proximal portion i.e. the bar, which is rotated to produce an elliptical path (Fig 2.3). The machine makes use of an epicyclic gear system (Fig 2.4) which is contained in a rotating arm. This helps in giving a high power density while being contained in a smaller area. The gears are driven by a motor which permits three levels of transmission speed 60, 90 and 120 cycles per minute.

Knee locking is also achievable in this design. The point at which the trochanter passes through the lower half of the ellipse the knee locks. This particular instance is the imitation of the midstance of a normal knee which also locks to provide for stability.



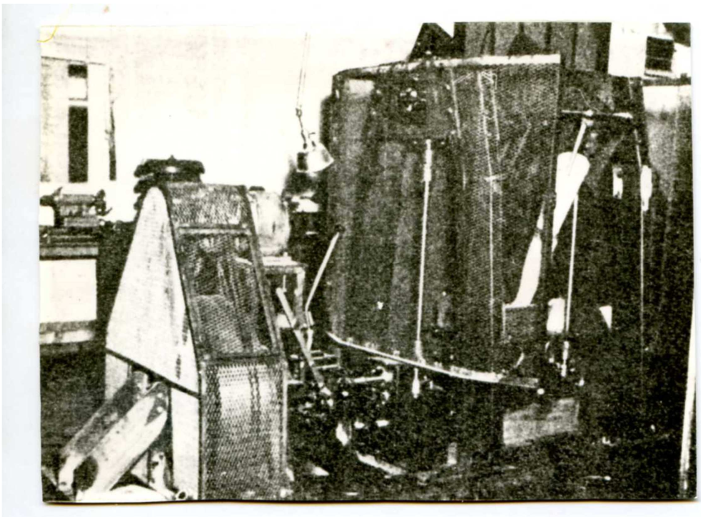


Figure 2.3 Above Knee Tester (adopted from University of California, Berkeley, 1947)

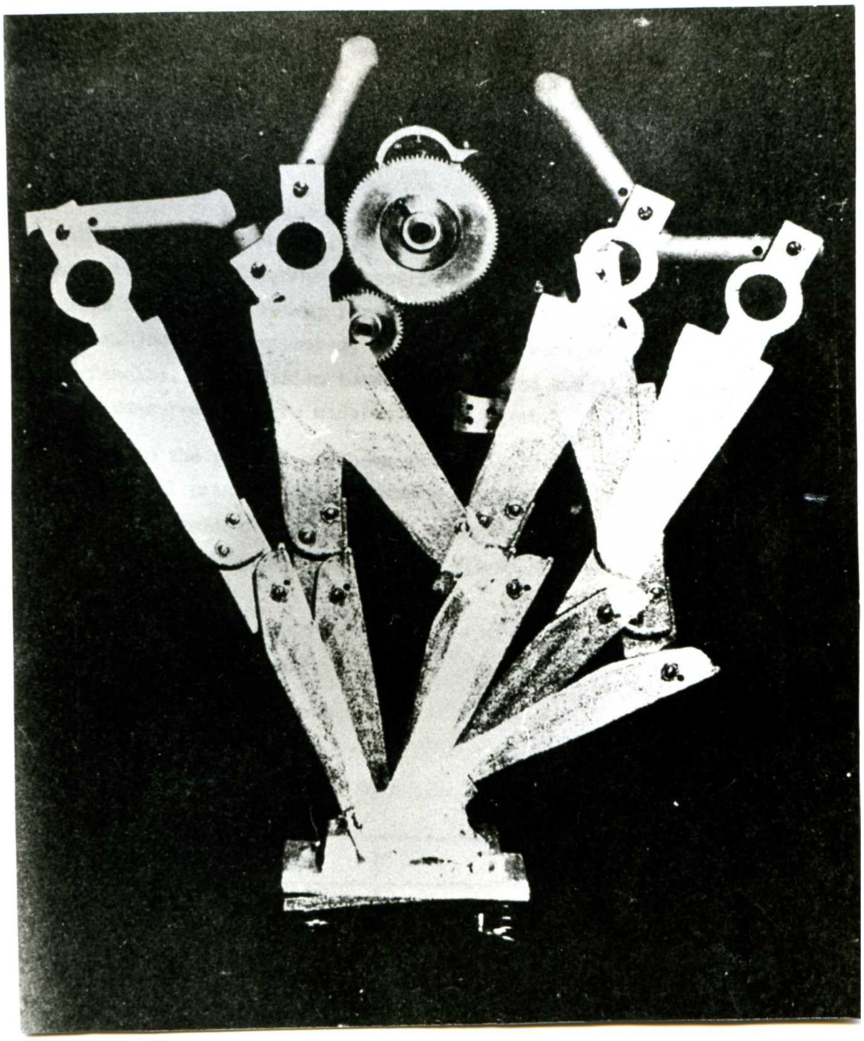


Figure 2.4 Position of prosthesis during various stages (adopted from University of California, Berkeley, 1947)

Through the available data the only applied load found was by the foot on the foot plate. Thus, it becomes essential to have an appropriate choice of springs for the load cycle to complete. Force measurement was done via strain gauges placed on the point of contact of the thigh to the machine. The design also allows for torsional loads to be applied along the axis of the test piece.

2.6.3 PERSONAL IMPRESSION

The data available for the design is very limited and it does leave a lot of doubts in the mind about quite a few things. It is hard to assess even from the available pictures the kind of linkages the machine and the prosthesis has. This is crucial in understanding the load application at the proximal aspect. From the present data one can only gather that the only load the authors talk about is the one by the foot on the plate.

It is possible that with the use of springs at the base of the design the machine could experience vibrations while the prosthesis is in its cycle. The data obtained could be far from being the actual one. In the present context use of hydraulics could solve this problem.

The use of the elliptical path of the trochanter by the authors as a means of guiding the whole cycle is a very innovative idea and one certainly worth appreciation.

2.7 “Lower Limb Modular Prosthesis” DEPARTMENT OF HEALTH AND SOCIAL SECURITY, SCIENTIFIC AND TECHNICAL BRANCH, ROEHAMPTON, LONDON 1973.

The following data has been obtained from “Fatigue testing of a Lower Limb” Phillips (1977). The original studies could not be located.

2.7.1 STUDY GOAL

The essential aim of the project was the dynamic testing of the Lower limb prosthesis by design and development of a testing machine. However, the available text cites a few shortcomings of the machine due to which the machine had to be constantly kept under observation. These will be discussed subsequently.

2.7.2 MACHINE DESIGN

Within the design the test prosthesis was meant to be kept between an upper cross segment and an ankle block for support. A foot piece was not part of the setup.

The rotation motion is produced from the base of the prosthesis i.e. the ankle. With the help of an upright lever assembly the prosthesis is moved back and forth. This motion is produced with the assistance of another lever which is attached at a position midway between the upright lever assembly and somewhere on the radius of a flywheel. The flywheel is in turn driven by an electric motor.

The load application is achieved with the help of a compressible spring lying proximal to the test prosthesis and parallel to the lever assembly (Fig 2.5). As the prosthesis is pushed into stance the spring is compressed by the action of a connecting roller which is deflected in accordance with a template. The compression of the spring is thus a direct measure of the force sustained by the prosthesis. As the prosthesis comes back to its initial position the template gets unlocked and goes back upwards, thus no axial loads. The tests were conducted at 60 cycles per minute.

2.7.3 PERSONAL IMPRESSION

As mentioned earlier the machine had some short comings. The calibration of the load spring had some problems and thus the measured forces would not be accurate. The rate of operation of the whole mechanism was considered slow and unlike the literature mentioned earlier did not have different operating speeds. Lastly, no fail-safe mechanism was incorporated within the machine.

Apart from the inherent disadvantages mentioned in the text there are other points which need to be mentioned. In my opinion the design is apparently not suitable for the testing of above knee prosthesis. With the addition of a mechanical knee joint the instability increases. The proposed method of producing motion might not be able to bring out the expected pathway of motion of the knee. However, the design might be good enough for static load testing of the prosthesis.

The absence of the foot adds to the failure of the design. With no foot one cannot expect to attain the normal gait. Also with the ankle block fixed proper imitation of the gait would be poor.

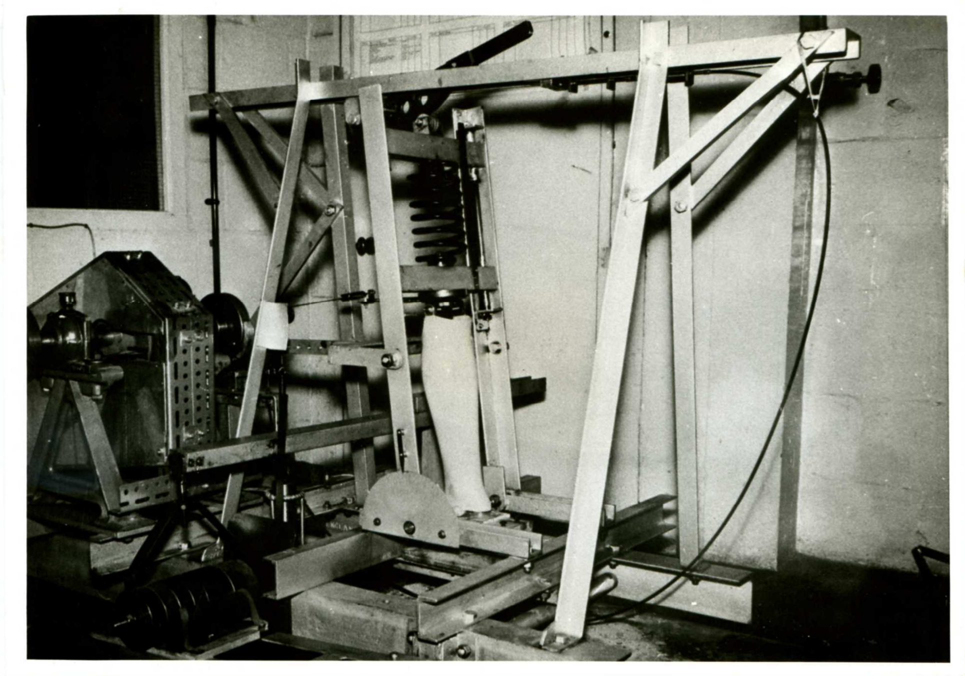


Figure 2.5 Prosthesis testing apparatus. Spring lying proximal to prosthesis (adopted from Dept. of health and social security 1973)

2.8 THE SPECIFICATION AND DESIGN OF A SYSTEM FOR THE MECHANICAL TESTING OF LOWER LIMB PROSTHESIS. *Phillips 1977*

2.8.1 STUDY GOAL

The particular chapter from the Thesis concerns itself with the design of a testing system for the evaluation of AK prosthetic systems.

2.8.2 STRUCTURAL DESIGN

Load producing mechanism: the author makes use of a hydraulic system for producing loads since the hydraulic actuators have higher velocity stiffness and leakages are low, therefore speed drops as load is applied, are not present.

2.8.2.1 FRAMEWORK: the different components of the testing machine are assembled and supported by a framework. The testing frequency is < 2Hz, therefore it is vital that the natural frequency of the framework be more than this to eliminate the possibility of resonance. This point is taken care by adding heavy members within the unit and giving more stiff joints.

The basic structure consists of four upright members (10cm X 10cm). These provide for a method to raise or lower the top support members as and when required for accommodating different sized prosthesis or even just the foot assembly.

This height-adjustable assembly supports the upper part of the prosthesis and aids rigidity of the test mechanism. Coupled with this are the four cross members which secure the top four of the upright members.

The lower load is taken up by the foot platform. This platform rolls on a heavy flat baseplate upon which the entire mechanism is fixed. It is taken care that all the pin-jointed parts are bearings selected to be suitable for the loading and speed operation used.

2.8.2.2 LOAD TRANSMITTING MEMBER: This is an essential feature of the testing machine and of extreme importance if the mechanism needs to function effectively and give reliable results.

The author makes use of a socket filled with a cold setting rubber insert between a polyester plug and the socket, at the center of the polyester is embedded a strong shaft fixture. The randomness of the rubber density is also of importance. Hard rubber at the loaded areas and soft rubber at the pressure relieved areas. The load bearing and relieving areas will correspond to the type of socket chosen for the experiment.

The position of the shaft needs to be carefully angled in relation to the foot plate. Thus it becomes vital to find the head of femur in relation to the limb structure. The bar needs to be held between this hip point and the center of the socket for proper load transmission. Failure to this will result in unwanted bending moments being applied during the tests, as a result of applied loads which are offset to the axis of the limb.

2.8.2.3 CONTROL SYSTEM: the testing machine needs to be operated and controlled by another system. The control system is the heart of the machine. Without careful maintenance and compensation of pre-selected stress levels and displacements, instability may arise rendering either the results meaningless or producing a situation where loads are excessive, causing premature breakage of the test limb.

The author makes use of a system developed by Dartec Ltd, Slourbridge, Worcestershire, England. The company could supply and assemble the hydraulic hardware.

2.8.3 OPERATING MECHANISM

The mechanism designed has three interdependent electro-hydraulic servo mechanisms producing controlled load and displacements in the vertical plane, anterior-posterior plane and in the anterior-posterior plane about a fixed point.

2.8.3.1 VERTICAL LOADING AND DISPLACEMENT: the vertical displacement of the foot plate is brought about by a motor. The motor is a linear actuator and is situated in the horizontal plane. With the help of levers it controls the vertical displacement of the plate. Throughout the vertical motion the plate remains in the horizontal plane. The footplate movement is frictionless as it is guided by four upright pillars which have a roller bearing interface. Pressure supply to and from the actuator is by flexible piping connected to an electro-servo valve situated on the main framework and as near as possible to the actuator.

2.8.3.2 ANTERIOR-POSTERIOR LOAD AND DISPLACEMENT: this motion is produced by the horizontal motion of the foot plate by the action of the actuator. This is in fact similar to the motion of a conveyer belt. The complete foot plate platform is displaced along flat guideways. Motion of the surface is restricted to the horizontal plane as its weight prevents it from displacing vertically. The foot plate runs on wheels and is prevented from displacing sideways by wheels acting against the inside of the guideway.

2.8.3.3 ANTERIOR-POSTERIOR MOMENT AND ROTATION OF SOCKET END: The motor is rotary and is held rigid by an upper cross member of the framework structure. The rotary shaft of the motor is coupled directly to a shaft perpendicular to the test piece fixture. The mid-point of this shaft represents the center of rotation of the applied anterior-posterior movement.

2.8.3.4 OPERATING CYCLE: To begin the cycle the foot plate is raised until the foot of the prosthesis is in contact with it. At heel strike a vertical load feedback sensor triggers the stance phase cycle to begin. The vertical actuator load is now balanced by both the anterior-posterior bending moment applied through the rotary actuator at the hip and the anterior-posterior shear force applied through the horizontal actuator.

The foot plate platform constantly displaces the prosthesis at a controlled rate by applying a shear force at the foot. At midstance the prosthesis is kept in extension by applying anterior-posterior bending moment. As the platform is moving towards the toe-off position, the controlled vertical loads fall back to zero. At toe-off the footplate is lowered and the test prosthesis is allowed to swing through and reach back to the starting position.

Once again the foot plate platform is raised to allow for heel strike to be initiated. Thus, the whole cycle is repeated.

2.8.4 PERSONAL IMPRESSION

The particular design remained only on papers and has never been constructed. Thus, it has never been physically tested and remains just a concept. However, the idea certainly presents a good method for a prosthesis testing. One point of particular interest is the appropriate use of hydraulics to control the rise and fall of the foot plate. With the help of which the prosthesis is able to swing through and complete the whole gait cycle. This is very important in the physical testing of prosthesis. In the other mentioned designs it is unclear whether they allow for the prosthesis to swing or just maintain it in constant stance.

The loading of the prosthesis is again a very important feature and has been discussed & concluded very well by the author. The described method for loading the prosthesis is very useful but, at times when it is required to change the load it will be difficult as the shaft is embedded within the polyester resin. This problem could be solved by the use of high density collapsible Poly urethane foam. The shaft can be added and removed anytime from the socket.

The author has very well described the operating mechanism of the design and how the foot plate and the foot plate platform bring about the motion of the prosthesis. But, it remains a bit unclear how he has fixated or linked the prosthesis with the whole framework proximally. The author does provide a sketch but with no write up it is difficult to understand the linkage. But, it is clear that the movement comes from the distal portion.

The design presented does produce a lot of inspiration. It certainly feels superior to the other machine design discussed on paper. But, without a physical machine it becomes hard to say if the design will produce the expected results.

2.9 A CLINICAL COMPARISON OF VARIABLE DAMPING AND MECHANICALLY PASSIVE PROSTHETIC KNEE DEVICES. *Johannson et all 2005*

2.9.1 STUDY GOAL

The study intends to compare the metabolic energy consumption and user gait with the use of a mechanical knee (Mauch® SNS) and two microprocessor controlled knee units (Rheo® from Ossur and C-Leg® from Otto Bock).

2.9.2 STUDY DESIGN

2.9.2.1 TEST SUBJECTS – 8 unilateral amputees participated in the study. All of the subjects had a K3 activity level of ambulation.

2.9.2.1 IMPLEMENTED IN - laboratory settings

2.9.3 METHOD

Each amputee had been given 10hrs for acclimatization to the knee he had not used. For the test the amputee's prosthesis was kept constant and only the knee assembly was replaced. Each amputee was to undergo three testing sessions. One of the sessions was performed using an indoor track and the metabolic oxygen consumption was recorded with one of the knees. The other two sessions required the amputee to walk within a gait laboratory and the kinematic and kinetic gait data was obtained for the other knees. For each session and amputee the knee joint used was randomized.

The subjects were asked to walk for a quarter of a mile with a portable breath by breath telemetric system .Before walking on the indoor track the amputee was asked to select a comfortable speed and walk alongside an electrical vehicle. For the sessions in the laboratory kinematic and kinetic analysis was made with the help of a motion analysis system. The kinematic data was obtained using positions of reflective markers placed on the amputee. The kinetic data was obtained by measuring the ground reaction forces with the help of two staggered force plates.

2.9.4 RESULTS

The authors have hypothesized that the amputees will benefit by using variable damping prosthetic knees. They have tried to prove this by the test procedure mentioned above. The metabolic cost reduces by 5% on using the Rheo® as compared with the Mauch® SNS. The metabolic cost is also less by 3% for the Rheo® against the C-Leg®. The authors observed several biomechanical advantages of the variable damping mechanisms over the mechanical. An increase in the smoothness of the gait shown by a lower jerk RMS, a decrease in hip work production in stance and swing, a lower peak hip flexion moment at terminal stance and a reduction in peak hip power generated at toe-off.

In comparison with the C-leg® the Rheo® knee offers a better foot and ground interaction and improved swing phase hip biomechanics. Also, the Rheo gives a better heel compression in stance and a reduction in peak hip extension torque during terminal swing.

2.9.5 PERSONAL IMPRESSION

Through the paper the authors have tried to prove their hypothesis that microprocessor controlled knee have quite a few advantages over mechanically controlled knee units. However, this is a known fact now in the field of prosthetics. But, the other very important point to be considered about the paper is the comparison of the two variable damped knee units. There not a lot of studies which have tried to do this work. Through evaluation the authors managed to prove the benefits of the Rheo over the C-Leg®. However, the test methodology was more directed towards the comparison of non-microprocessor with microprocessor. But, for testing two variable damping units one needs to take into consideration many other parameters. In which case the results could be different.

CHAPTER 3. MICROPORCESSOR CONTROLLED PROSTHETIC KNEE UNITS

3.1 INTRODUCTION

Walking is an activity which is common to all humans. Its necessity cannot be debated. Be it moving from one place to another or just for the purpose of exercise, it is hard living without being able to walk. For a smooth gait cycle the anatomical knee joint needs to be in a healthy state. Any deviation from the normal causes changes which are visible to the naked eye. Therefore one can surely understand the psychological state of a trans femoral amputee. Walking becomes a challenge for him. So it is vital that the amputee undergo a good rehabilitation program to bring him back to his feet and give him his confidence back.

One good step towards the amputee's rehabilitation is the prescription of an appropriate prosthesis. With the right prescription a smooth rehabilitation can be anticipated. The mechanical knee unit here plays an important role. As mentioned earlier the knee joint is a vital part of the lower limb, same would apply here. Presently in the market 100's of knee units are available for the Prosthetist to choose from. Careful selection will make a lot of difference in the gait of the amputee.

Microprocessor controlled (MPC) knee units form one of the categories. As the name suggests the knee utilizes the help of computer panel to control the stance and swing phases of the gait cycle. The technology was first introduced to the world in the early 90's when the Intelligent prosthesis was launched by Blatchford, United Kingdom. Subsequent years saw many changes within this particular knee unit. Also, many other companies came out with their own more sophisticated versions. The basis however remained same, appropriate control of swing and stance for a smooth and comfortable gait cycle. What is important here is how the knee units mange to control the phases of gait. One of the knee units for example makes use of sensors to detect different phases of the gait cycle. Depending on the phase the sensor has detected, the microprocessor controls the knee flexion. Therefore, by giving appropriate resistance to flexion and extension the knee unit can even control the speed of walking. The features of the knees are not limited to just alter walking speeds. But by using the same idea the microprocessor can provide for a better stability, safety and improve the function. One example of this would be the prevention of stumble or fall, the sensor is designed to recognize stumble and in return it sends the signal in reply to which the knee unit stiffens, to prevent the fall. Other advantageous features of the knee units are improved ability to navigate stairs, slopes and adjustability to uneven terrain.

Most of the MPC knee units can provide for control in both the phases of gait. They usually make use of hydraulic or pneumatic cylinders for replicating the function of the anatomical knee joint. These in turn are heavier as compared to the other mechanisms. Thus, making the whole knee unit a lot heavier comparatively. Hence, the prescription of the knee unit is very important. Mostly the units are best suitable for high activity patients. However, some designs are in the markets which are seen suitable for very low activity patients or even for geriatrics such as the C-Leg® Compact®.

All in all it is an understood fact that the MPC knee units are supposed to have the ability to mimic the anatomical knee motion. This is a very difficult task in itself. However, as mentioned in above paragraphs the MPC knee units employ both electrical and mechanical control in fulfilling this claim. The intention of this project is to design means for testing the claims made by the manufacturers about the MPC knee units. For the purpose of this project concentration has been given to two MPC knee units the C-leg® manufactured by Otto Bock Healthcare and Rheo® Knee manufactured by Ossur.

3.2 C-Leg®

Otto Bock Orthopadische lndustrie Besitz- und Verwaltungs-Kommanditgesellschaft lndustriestrasse 1993. System for controlling artificial knee joint in an above knee prosthesis. Euro Pat No 0 549 855 A2

3.2.1 INTRODUCTION

Within the family of Microprocessor controlled Prosthetic knee joints the C-Leg has been considered as a pioneering invention. Functionally the knee makes use of preprogramed microprocessor which interprets the common gait patterns received from strain and knee angle sensors on the prosthesis. At various instances during the gait cycle the microprocessor activates a motor which controls the opening and closing of a valve assembly within the damper. The valve assembly is capable of damping the knee joint in each of flexion and extension. With the good control over knee flexion throughout the cycle the user tends to have a comfortable gait. Activities such as stair and ramp ascending are also smoothened with the joint.

3.2.2 TECHNICALITIES

As mentioned earlier the Knee joint makes use of an on-board computer controlled system for damping or resisting the amount of rotation of the knee joint. The joint forms a connection between the upper and the lower segment of the prosthesis and plays an important role. The system comprises of the following functions:

- A linear hydraulic damper which can separately and variable damp or resist the flexion extension moments about the knee joint.
- An electronic sensing element which takes signals about the knee angle and the strains, both of which are indicative of the angle of segments and the relative position of the center of Gravity of the body with regard to the load.

The emitted signals provide for:

- Means to activate the motor which subsequently provides control of the valve assemble to damp or resist the knee rotation in either flexion or extension depending on the requirement.

- A constant knowledge of the position of the prosthesis in the gait cycle is again and again updated. To do this the microprocessor is fed with a set of threshold values. For damping to happen the received signals are constantly correlated with the stored set of values to give the microprocessor knowledge of the position of the prosthesis.

3.2.3 LINEAR HYDRAULIC DAMPER

The knee joint has the capability to provide for variable and separate damping of knee joint rotation moment in each of flexion and extension. There could be three phases:

- The resistance may be substantially complete, in which case the knee joint is substantially prevented from rotating in either flexion or extension
- The resistance may be partial, in which case the rate of rotation of the knee joint is restricted in one or both of flexion and extension.
- There might not be any resistance, the knee joint is hence, free to rotate in one or both of flexion and extension.

The hydraulic damper enables such bi-directional damping and it comprises of:

- A hollow cylinder filled with hydraulic fluid and a cylindrical piston adapted to move vertically within the cylinder chamber.
- Piston has axial rods extending from the sealed openings of the cylinder

- A first aperture and check valve associated with the first end wall of piston and enables fluid to enter the piston chamber from the first end of the cylinder chamber
- A second aperture and check valve associated with the second end wall of piston and enables fluid to enter the piston chamber from the second end of the cylinder chamber.
- A first pair of diametrically opposite ports extend through side of the first end of the piston wall
- A second pair of diametrically opposed ports extend through the side of the second end of the piston wall
- A vale extends into the cylinder and piston chambers and is adapted to increase or decrease the area of the first end ports available for fluid flow and subsequently increase or decrease the area of the second port.

3.2.4 PHYSICAL OPERATION

As said earlier the knee joint is attached between the upper and lower segments of the prosthesis, the damper unit here is the attachment point. The upper push rod of the piston is pivotally connected to the upper leg segment of the prosthesis and the cylinder is connected pivotally to the lower end of the segment. Thus, in flexion the piston will move downward with the body load and during extension the piston will be pulled upwards by body action.

3.2.4.1 FLUID FLOW

- If the valve is positioned to enable flexion (Fig 3.1) and if the piston is pushed downward, thereby pressurizing the fluid in the lower end of the cylinder chamber to flow upward through the lower end check valve and extension port into the piston chamber. The fluid will leave the piston chamber from the upper flexion ports
- If the valve is positioned to enable extension (Figure 3.2) and if the piston is pulled upwards, thereby pressurizing fluid in the upper end of the cylinder chamber to flow downward through the upper check valve and flexion port into the piston chamber. The fluid will not leave the piston chamber via the flexion port due to insignificant pressure difference between the upper piston chamber and cylinder chamber.

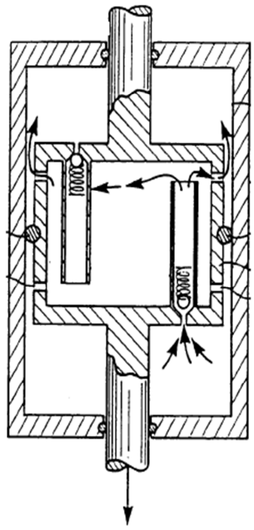


Figure 3.1 Downward motion of piston by action of fluid (adopted from Otto bock, Euro Pat No 0 549 855 A2)

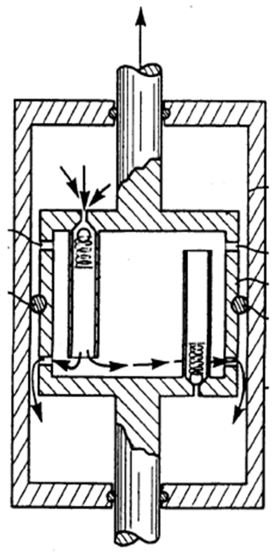


Figure 3.2 Upward motion of piston (adopted from Otto bock, Euro Pat No 0 549 855 A2)

3.2.4.2 MOTION REPETITION AND MEMORY STORAGE

For the amputee to move around the prosthesis has to perform repetitive motion cycles. This is one of the features of the C-Leg®. During normal walking on a level surface, the patterns of knee angle and the lower leg strains do not change significantly. By monitoring these two sets of signals from the sensors the computer software can identify which stage of the gait cycle the knee is experiencing and thus make the necessary changes in flexion and extension. With any kind of deviation from this pattern such as a sudden stubbing of the prosthesis , the knee joint software can swiftly detect this change and provide corrective measures. Figure 3.3 shows the basic Physical Operation of the Joint. The above mentioned actions can be performed in the following manner:

- Within the computer memory are stored certain threshold values of knee angles and lower leg strains, which are indicative of knee angle in stance phase, the position of the center of gravity in relation to the weight on the ankle and the foot, also the swing phase bending.
- The sensors embedded within the joint continuously keep track of the knee angle and the lower leg strains of the prosthesis and keep producing the corresponding electronic signals
- Subsequently the sensors compare the signals with the recorded threshold values. Once, the signals correlate, they produce actuating means for the alteration of the rate of rotation of the knee either to initiate flexion during the beginning of stance phase,

to lock the knee in extension during md-stance phase or to free the joint as it nears towards the swing phase, thereby subsequently providing natural movement and repeating the above process continuously.

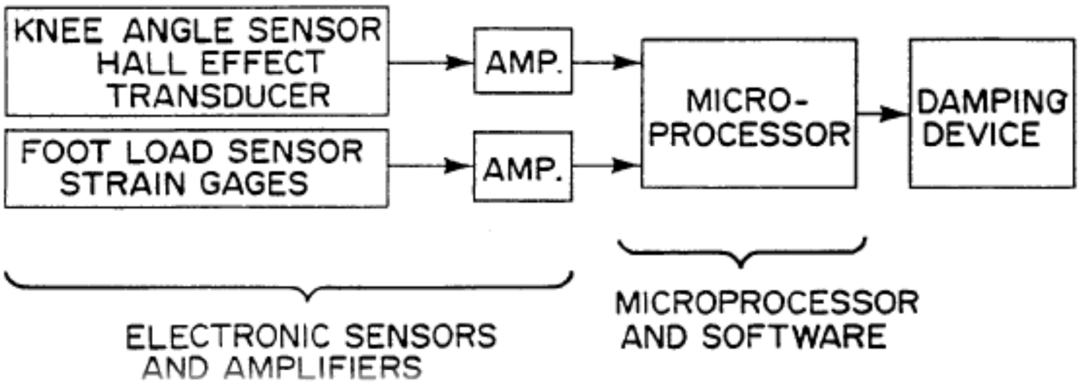


Figure 3.3 Basic operation principle (adopted from Otto bock, Euro Pat No 0 549 855 A2)

By combining the sensing elements within the joint, the hydraulic damper having means to simultaneously and separately control the flexion and extension and software which works on repetitive profiles the knee joint can closely control and predict the responses. The result is the user gaining confidence while walking and the knee joint adapting to the particular gait type and giving a more longer & rhythmic gait. Furthermore, the software can be fine-tuned to the needs of the user or altered to modify the operation of the prosthesis. In addition to level walking the system is also adaptable to stair descent and sitting down

3.3 RHEO® KNEE

Herr H & WIlkenfeld A 2003. User adaptive control of a Magnetorheological prosthetic knee. Industrial Robot: An International Journal 30:42-55

3.3.1 INTRODUCTION

Developed at the Massachusetts Institute of Technology the Rheo® Knee presents a very new concept to the field of Prosthetics. The basic function of any MPC prosthetic knee unit is to provide for variable damping during stance and swing phases of the gait cycle. The Rheo® Knee achieves this via the use of an on board microprocessor which makes use of Artificial Intelligence. The resistive torque or the damping of the knee is produced through the control of magnetic field within a Magnetorheological fluid (explained further).

3.3.2 COMPONENTS

The function of the knee unit is divided among the following segments:

3.3.2.1 ACTUATOR

As mentioned above the Knee unit makes use of Magnetorheological (MR) fluid as the braking mechanism. MR fluid basically consists of small iron particles (1 micron) suspended in oil that forms torque producing chains in response to an applied magnetic field. For production of magnetic field the system is comprised of an electromagnet and a magnet circuit. By varying the current in the electromagnet the magnetic field was controlled and the hence the damping.

Figure 3.4 shows path of the magnetic field within the knee. When the current is applied to the electromagnet, a magnetic field is generated which follows path around the rotary axis of the knee. The field then moves radially outwards through the first side plate, laterally through an interspersed set of inner and outer metal disks and then radially inwards through a second plate. As is seen in the figure 3.4.each inner disk is coupled to an inner spline and outer disk to an outer spline. Injected between each inner and outer disk is the MR fluid. On passing magnetic field MR chains develop and connect each lower disk surface to an upper disk surface. These chains hence enhance the necessary torque required for flexion.

3.3.2.2 SENSORS

For controlling the damping of the knee the system uses only local mechanical sensing of knee position, force and torque. All the sensors were placed relatively near to the knee axis. The angle measured by the knee angle sensor is differentiated in analog circuitry for estimation of the knee velocity. This is necessary to estimate whether the knee is flexing or extending. The axial force sensors measured the forces on the knee coming from the ground in the longitudinal axis of the knee. This is beneficial in knowing if the foot is on or off the ground. The axial force sensors also measure the torque of the knee. During the early phase of stance when only the heel is being loaded the sensor shows a positive flexion moment indicating that the load is passing posterior to the knee and the knee is at a risk of buckling. Whereas, during late stance when the toe is being loaded the sensor presents a positive extension moment indicating that the load is anterior to the knee and there is no risk of buckling. Figure 3.5 shows the various positions of the sensors

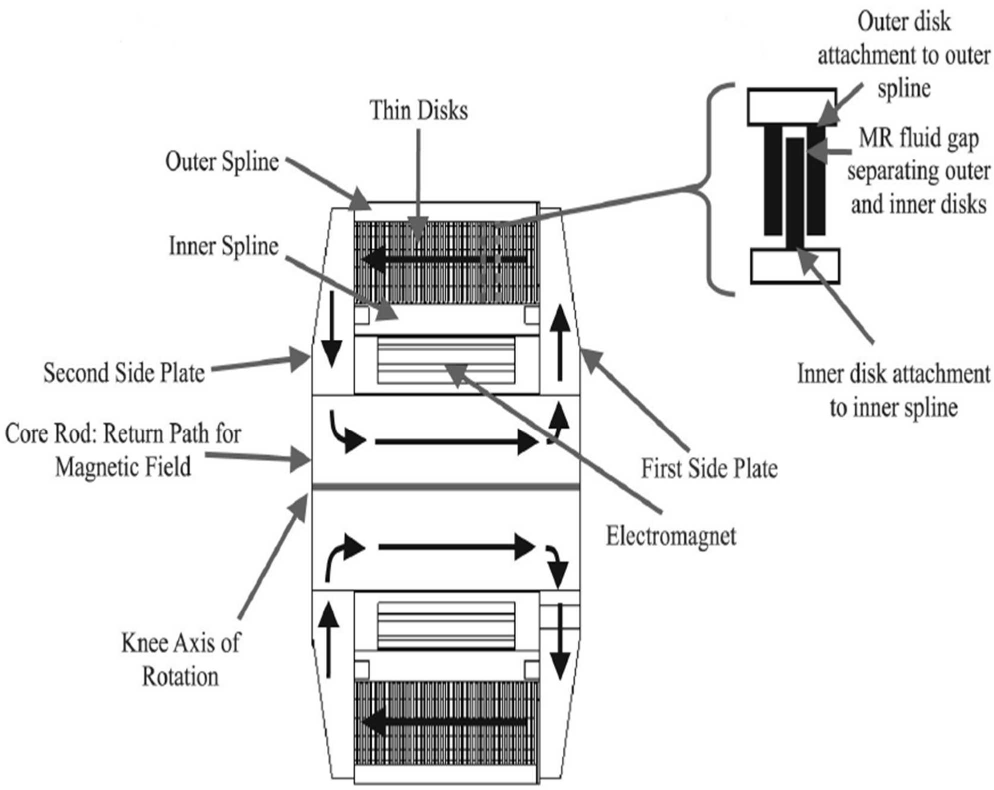


Figure 3.4 Operation of the actuator (adopted from Herr & WIlkenfeld 2003)

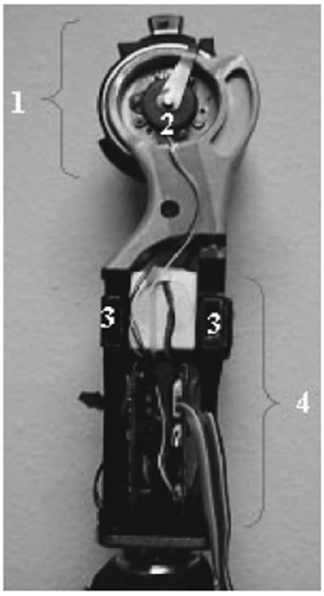


Figure 3.5 Position of various components (adopted from Herr & WIlkenfeld 2003)

From the figure:

1) *Magnetorheological Brake*
2) *Potentiometer angle sensor*
3) *Force Strain Gauge Sensors*
4) *Battery and electronic Board*

3.3.2.3 ELECTRONIC BOARD AND BATTERY

The system makes use of a 6812 Motorola microprocessor for computation of all the sensor data. The power is generated by four rechargeable Lithium ion batteries.

3.3.3 PHYSICAL OPERATION

The knee joint makes use of the different phases of the biological gait cycle. For the system the gait has been divided into 5 phases:

1) Starts with the heel strike, after which the knee immediately begins to flex. The flexion here allows for shock absorption upon impact.
2) Once the maximum flexion is reached the knee begins to move towards maximum extension in stance.
3) During the late stance the knee of the stance leg again begins to flex in preparation for leaving the ground. At the same time the adjacent foot strikes the ground and the stage is referred as the Double support mode.
4) As the knee keeps flexing, the hip also goes into flexion and the foot leaves the ground.
5) Once reaching a maximum angle of flexion in swing the knee again begins extend. Once, full extension is achieved the foot again makes a heel strike and the whole process gets repeated.

The phases control the functioning of the knee mechanism. Each of the mentioned phases corresponds to a state of the knee. As talked earlier, the on-board sensors were used to determine the state of the knee by measuring knee angle, velocity and the torque. Based on this acquired data the controller cycles through the different phases and gives the respective signal.

Within each of the stage the electric current of the electromagnet is controlled such that the knee resistive torque is proportional to the square of the knee rotational velocity,

$$Torque = B(V^2)$$

Where 'V' is the angular velocity determined from the differential signal of the angle sensor. 'B' is the active knee damping constant. For the 5 different phases or stages of the knee 5 different values of damping constant are used. The damping of the knee is only from cycle to cycle or from stage to stage and never during a particular stage. The objective here is to determine knee damping value for each stage which in turn would provide an improved gait.

The major aim of the controller here is to give a biologically realistic maximum angle of flexion during swing and stance flexion & extension critical for effective shock absorption .By following the mentioned principles the user-adaptive knee system is able to provide for a trans femoral gait which should prove to be improved in terms of biological realism and symmetry between affected and unaffected sides.

CHAPTER 4. REVIEW OF THE KINEMATIC AND KINETIC PARAMETERS OF NORMAL SUBJECTS AND AMPUTEES DURING GAIT

4.1 INTRODUCTION

Testing and evaluation is an essential part of commercial production. The manufacturer and vendor often aim to give attractive features that their product fails to deliver. This applies to a wide range of products, from household appliances to industrial machines to healthcare equipment. Therefore, for each promised feature a testing and evaluation tool needs to be evolved. Quality assurance is required for the security of the producer as well as the consumer. The limb prosthesis forms one such product for which strict quality control is crucial. It is vital that all the components of AK prosthesis be tested in all aspects before they can be made available to the clinical centers for fitment. Failure to do so can place the patient/amputee in an accident prone state.

The knee unit fitted within AK prosthesis forms its most vital part. Its evaluation needs to be done separately with a system that can take care of all its characteristics. Historically the knee units were placed under cyclic loading to check their development of fatigue & test their longevity. Nowadays the concern is not just for the life of the knee units, but also on their functionality. The designs offered in the market today claim a variety of functions. It is important for authorities to test these claims.

A microprocessor controlled prosthetic knee unit takes help of liquid and gas pistons to control the stance and the swing phases of gait cycle of an amputee. Along with the stance and swing control, the MPC knee units are capable of providing many other features which help the amputee in attaining a comfortable and smooth gait. Apart from this, these knee units provide features for safety. As a result of all these advantages to the amputee, they get largely dependent on the effective functioning of the MPC knee units. Walking becomes a very unconscious effort for amputees using these knee units. This is a great achievement from the point of view of both the clinician and the amputee. To maintain it like that, it is of extreme importance that the MPC knee units work in the projected manner. Testing and evaluation of the MPC knee units is thus a very important task, not just to check them for their level of safety and security, but also to know if they reach the claims made by their manufacturers.

4.2 GAIT ANALYSIS FOR HEALTHY INDIVIDUAL

The MPC knee unit is expected to replicate the movements of a real anatomical knee joint during the gait cycle. Therefore, it becomes necessary to first understand the positions that lower limb joints take in the gait cycle of a normal, healthy individual. This can be used as a basis for comparative analysis of the MPC knee unit to further support the claim that it does indeed behave like an anatomical knee joint. This data also helps in design of the bench tester machine.

Gait is the process of locomotion achieved through the movement of limbs. When performed in repetition it forms the Gait Cycle. The gait cycle is composed of two phases: *Stance phase and swing phase.* In simple words it can be said that the stance is the phase wherein both the feet are in contact with the floor either partially or completely and in swing one of the feet is lifted off the ground to move forward.

4.2.1 ANALYSIS OF THE MOTION IN SAGITTAL PLANE DURING LOCOMOTION (Peizer & Wright 1970), (Whittle 1991)

4.2.1.1 STANCE PHASE

4.2.1.1.1 Heel Strike

The beginning of the Stance phase is characterized by the heel of the foot striking the floor. The ground reaction moves anterior to the hip causing flexion. At knee the ground reaction is again in front causing an extension moment. At the ankle the vector lies posteriorly causing a plantar flexion moment.

Hip – The hip is flexed to 25°. Further flexion is prevented by the action of the Gluteus maximus and hamstrings

Knee – The knee should be most stable at this time by having full extension. Later on this extension moment is overcome by the action of the Hamstrings and it begins to flex.

Ankle – the ankle presents itself in a neutral position and then later begins to plantar flex. This is controlled by the action of pretibial muscles.

4.2.1.1.2 Foot Flat

Shortly after heel strike the foot lands completely on the ground and assumes a flat position. The ground reaction vector is still in front of the hip causing a flexion moment. At the knee the vector moves posteriorly causing a flexion moment. At the ankle the vector lies just posterior to it causing a plantar flexion moment.

Hip – the hip lies in 22° of flexion but begins to extend by the action of the Gluteus maximus and the hamstrings

Knee – the knee starts off with a 15° flexion angle and goes on till 20° of flexion is reached. It then begins to extend. The quadriceps is active in controlling the angle of flexion.

Ankle – the ankle lies in 15° of plantar flexion.

4.2.1.1.3 Mid Stance

This stage refers to the mid-point of the stance phase. The lower limb almost behaves as a pillar during this phase. The vector almost passes through the hip joint giving negligible moment. At the knee the vector passes slightly posterior causing a flexion moment. At the ankle the vector lies anteriorly and causes a dorsiflexion moment.

Hip – the hip is in 10° of flexion and begins to move towards 3° extension as the ground reaction shift posterior to the hip just after mid stance

Knee – the knee reaches 10°of flexion and it continues to extend. The quadriceps action is no longer active and the soleus is active in controlling flexion

Ankle – The ankle has 7° of dorsiflexion and continues to do so.

4.2.1.1.4 Heel Off

within this phase the ground reaction has a posterior position at the hip and gives an extension moment. At the knee it lies anterior giving an extension moment. The vector is still anterior to the ankle causing further dorsiflexion.

Hip – the hip reaches a maximum of 25° of extension just after heel off and thereafter starts flexing. The iliacus and the psoas major are active in controlling extension and initialing flexion.

Knee – The knee remains in 5° of flexion. The gastrocnemius may be active in preventing further extension

Ankle – the ankle reaches 15° of dorsiflexion after which it starts to plantar flex due to a powerful contraction of the calf muscles which counteracts the dorsiflexion moment and assists in propelling the body forward.

4.2.1.1.5 Toe off

The final phase in stance phase wherein the ground reaction has lost most of its significance as the majority of weight as borne by the other limb.

Hip – Lies in 16° of extension and continues to flex due to the plantar flexion of the foot and the action of the rectus femoris.

Knee – The knee is in about 40° of flexion and continues to flex

Ankle – With the action of the calf muscles the ankle reaches plantar flexion of 17°. This muscle group becomes inactive directly after toe off.

4.2.1.2 SWING PHASE

4.2.1.2.1 Acceleration

The first stage in the Swing Phase is characterized by lower limb accelerating forward.

Hip – The hip is in 10° of extension and moves towards 10° of flexion as the hip flexors accelerate the limb forward.

Knee – The knee is in 60° of flexion and continues to do so.

Ankle – The ankle lies in 12° of plantar flexion directly after toe off. It then begins to dorsiflex.

4.2.1.2.2 Mid Swing

Hip – The hip is flexed to about 20° and continues to flex

Knee – the knee reaches about 50° of flexion and then begins to extend

Ankle – the ankle reaches towards a neutral position and is kept there by the action of the pretibial muscles.

4.2.1.2.3 Deceleration

The final stage for swing is when the lower limb tends to lose its pace and gets ready for the heel strike for the next cycle.

Hip – the hip moves towards 25° of flexion and is restrained by gluteus maximus and the hamstrings

Knee – The knee is in 15° of flexion and is moving into full extension and thereafter is restrained by the action of hamstrings

Ankle – the ankle is still held in the neutral position by pretibial muscles.

The importance of understanding the natural gait cycle in regard to this project is for knowing whether the motion provided by the MPC knee units is appropriate or not. It is understood that the MPC knee units might not be giving the exact flexion angles. What is important is that, on rotation of the (voluntary) hip joint the (prosthetic) knee unit follows suit and completes the stages of gait.

4.2.2KINEMATIC AND KINETIC ANALYSIS OF GAIT

The preceding section concentrates on the different positions of the three joints during the gait cycle. This is an important consideration for the project as this helps in designing the motion for the lower limb component of the machine. For evaluation of the other parameters it would be helpful to have a data set dealing with the kinetic and kinematic analysis of gait.

A study conducted by S Zahedi et all, 1987 concerns itself with the kinematic and kinetic analysis of a healthy individual, a BK amputee and AK amputee. They have also looked at the changes brought about by different alignment of the prosthesis. This study deals with the comparison of gait data of the healthy subject with that of an amputee. This would provide for a picture of what the results can be. Also, it will be easier to comprehend the motion of the knee with the MPC knee unit. The authors have made use of a single axis knee unit for the purpose of the study.

4.2.2.1 VERTICAL LOAD, A-P SHEAR FORCE

In a prosthetics study it becomes necessary to calculate or assess the vertical and the horizontal components of forces acting either on a foot plate or to hypothesize on the ground. There is a marked difference between the forces by a normal subject and the one by an amputee. From a MPC knee unit it is expected that there would not be excessive loading and it is able to follow the functional pattern of the normal subject. Further, when designing a machine for testing any knee unit it is vital to make sure that there are no unwanted loads on it, which could damage the equipment.

Figure 4.1 shows typical variation of loads on a normal subject and also the A-P shear forces. Figure 4.2 shows the variation of load and shear forces in the pylon transducer frame of reference on an amputee used for the study. The thickness of the envelope represents the variations of the tests.

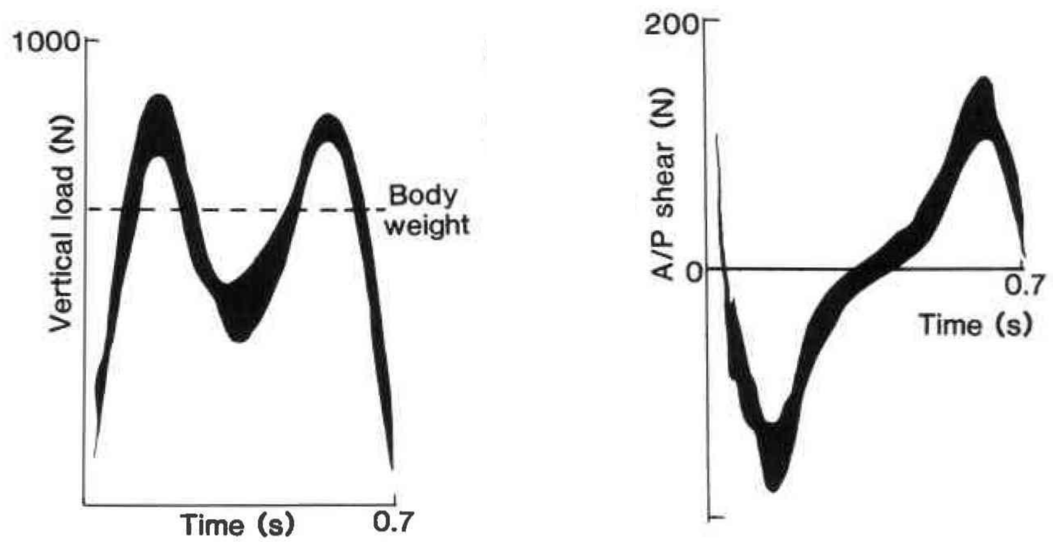


Figure 4.1 Vertical load and AP shear force of normal subject (adopted from Zahedi et all 1987)

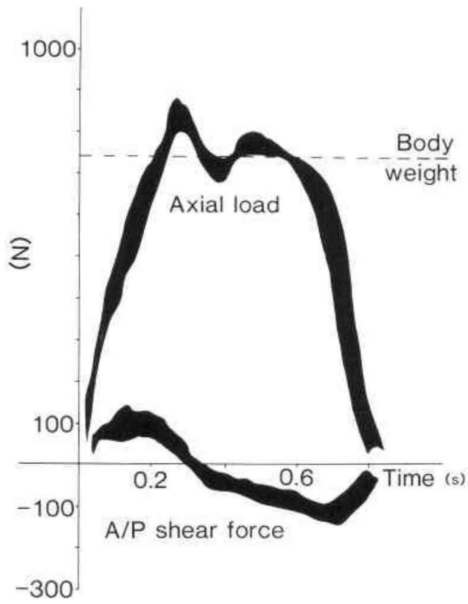


Figure 4.2 Vertical force and AP shear force transducer frame of reference (adopted from Zahedi et all 1987)

4.2.2.2 A-P MOMENTS AT THE ANKLE, KNEE AND HIP

Within the human body flexion and extension movement is brought about by moments generated by the forceful action of muscles. Similarly in any mechanical system the moments are generated by the direction of applied forces. When applied in a correct manner they can bring about the right movement, but improper control could lead to jamming of the system or excessive movement in a certain direction.

In the varied stages of the stance phase the knee goes in unequal amounts of flexion and extension. The direction of forces however is not constant and in some phases the forces lie behind it (midstance) and can cause excessive flexion, which would lead to unwanted buckling. Within a healthy individual this is controlled by the action of the muscles and the position of the anatomical knee joint. However, in a trans femoral amputee with the loss of the knee joint the quadriceps lose their advantage (long lever arm due to patella) and the amputee has to put in an extra effort to keep the knee in extension or maintain a certain degree of flexion. A partial solution to this problem is the posterior alignment of the mechanical knee unit. This just helps in giving a bit of stability to the amputee. However, flexion is hampered. But, with the use of a MPC knee unit this problem could be solved. The MPC knees are equipped to understand the different phases of the gait cycle. Memory storage feature allows them to understand the phase of maximum moment. Also they can adjust to the changing direction of vertical and shear forces. Thus, they should be able to provide for the right amount of flexion without causing any buckling and without any extra effort by the amputee.

Figure 4.3 shows the moments experienced by the three joints during the different stages of the gait cycle. The data corresponds to 14 steps of a normal subject.

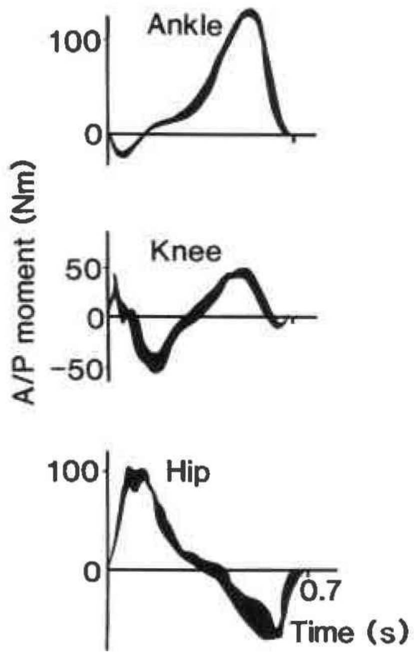


Figure 4.3 Moments at three joints of lower limb of normal subject (adopted from Zahedi et all 1987)

4.2.2.3 KNEE JOINT ANGLE VARIATION WITH TIME

The following data deals with the motion of the knee joint during normal walking of a healthy individual, a BK amputee and an AK amputee using a single axis knee unit. Figure 4.4 shows the variation of the knee movement in all three subjects.

For the healthy subject the figure shows data from one Heel Strike (HS) to another HS. As can be seen the knee keeps on flexing just after HS and continues to do so until midstance. After which it again rises and goes to a maximum flexion at the time of mid-swing. Thereafter, it again goes into extension for the next HS.

For the BK amputee the data set is from one Toe-off (TO) to another TO. Due to the presence of an anatomical knee joint the BK amputee is capable of flexing in a similar manner as that of a healthy individual. This is evident within figure 4.4.

For the AK amputee the data set is again from one TO to another TO. The amputee is able to provide for flexion during the swing phase for a proper ground clearance. But, during stance phase the flat portion of the figure reveals the incapability to flex. The subject is hence walking with a stiff knee gait. This results from the fear of falling while the normal limb of the patient is in swing and the prosthetic limb is on the ground supporting the whole body. However, with a MPC knee unit after a good practice the knee should be able to show some improvement in the flexion in stance. And over time the curve might even resemble that of a normal subject.

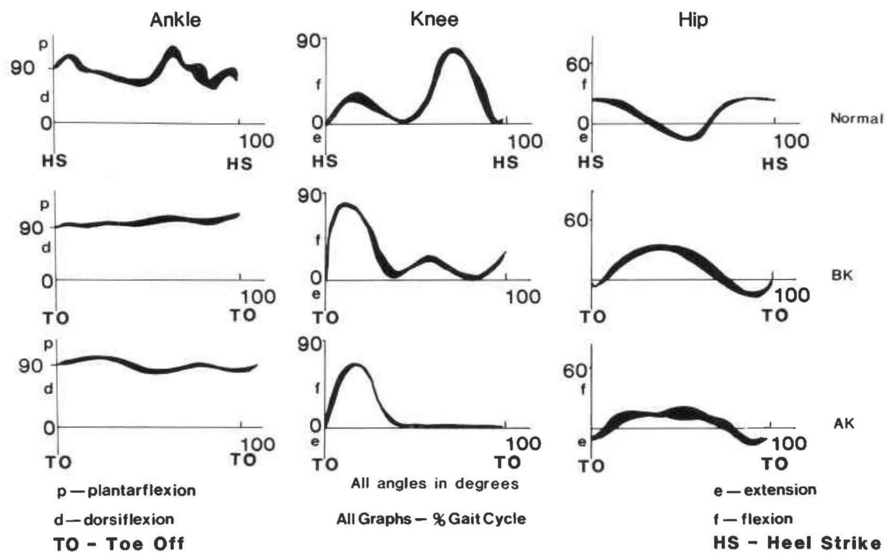


Figure4.4 Joint activity of normal subject, BK and AK amputee (adopted from Zahedi et all 1987)

4.3 PROPOSED TESTING PARAMETERS

It is necessary to test the MPC knee units on certain grounds. A smooth and comfortable gait cycle requires for a set of features to be carried out properly. Dependence of different movements upon one another is also seen. Thus, correct execution of knee movements and their timing is of utmost importance. The following are a few of the parameters that are to be tested within the MPC knee units:

4.3.1 FLEXION IN STANCE

As seen earlier a healthy gait requires for the knee to have the ability to provide for the right amount of flexion at each phase. A stiff knee gait is characterized by insufficient flexion, majorly during the stance phase. Though very stable, the gait gives a lot of discomfort to the stump of the amputee and over the years can completely damage the distal of the stump.

Excessive knee flexion is probably more fatal. Without a very good muscle control the knee can be very naughty and make the amputee more accident prone.

The present knee joints in the prosthetic market have a variable flexion resistance control, which needs to be adjusted manually. However, in a MPC knee unit it is expected that the knee adjusts to various walking speeds and gives just the right amount of flexion.

4.3.2 KNEE LOCKING AND UNLOCKING

Another very important characteristic of the gait cycle is the locking and unlocking of the knee joint at the right instances. But, this does not necessarily mean the locking of the knee without providing for the required flexion angle. For instance at midstance the knee joint has 10° of flexion, but for that phase the knee has to be locked for a fraction of a second to act as a pillar for taking the load of the whole body.

The unlocking of the knee joint is an equally important aspect of the cycle. An example will be the phase just after heel strike, wherein the knee has to have 5° of flexion before it can enter foot flat. At heel strike the knee is in a locked state to provide stability. But, the next instance it needs to be unlocked. While using a mechanical knee unit the amputee has to take care of the locking and the unlocking which comes subsequently with time and practice. In a MPC knee unit it is expected that the knee itself will perform the locking and the unlocking at the correct interval.

4.3.3 A-P KNEE MOMENTS

Flexion and extension moments dictate the movement of a knee joint completely. During stance the expectation from the MPC knee unit is to control the moments in a manner so as to provide for the right amount of flexion and extension without any strenuous effort by the user. With a non MPC knee unit this is however not possible and the patient relies on himself and the posterior alignment of the knee unit. For a MPC knee unit the alignment is a bit anterior so as to keep the knee in an advantageous position in regard to flexion.

During the swing phase the MPC knee unit should be having the ability to provide a flexion moment at the knee for the whole limb to clear the ground. Failure to do so will result in the amputee dragging the limb on the ground. Subsequently the amputee will end up vaulting for ground clearance.

Thus, it becomes necessary to keep a check on the knee moments of a MPC knee unit. While testing this would be usually visible to the naked eye in the gait cycle of the tester. But for a more close examination strain gauges can be made use of.

4.3.4 KNEE STUMBLING PREVENTION

This is one of the special features that the manufacturers of MPC knee units claim to provide in their designs and a very big concern for AK amputees while using prosthesis. This makes them very conscious of their walking environment and in a constant tensed state of mind. The resulting gait tends to have many deviations. Any form of mechanical knee unit cannot provide for this feature and the amputees just need to get used to walking with the eyes on the road and the mind on controlling the knee motion. However, with the MPC knee units it is anticipated that the sensors can pick up the signal of the foot hitting an object and instantly the microprocessor would lock the knee to prevent the fall.

4.3.5 SLOPE ASCENDING/DESCENDING

The state of an individual changes drastically while climbing up or down a slope or hill. This is even truer for an AK amputee. Walking uphill or downhill with an external knee joint is not a very comfortable task. However, with the use of a MPC knee unit it is expected that the user will not need to worry about the control of the joint. Special programs within the on board microprocessor of the knee joint enable the amputee to walk more naturally on a slope or hill. Walking downhill is however the more dangerous of the two. With the MPC knee unit it is expected that the amputee will not need to make a conscious effort to control the flexion of the knee. The microprocessor keeps it in extension, as a healthy individual will do while coming down a steep slope.

But it is very essential to evaluate this function of the knee unit. If the knee fails to perform the claimed task, it can be very fatal for the amputee. Thus, it is an intention to make provisions for checking this property of the knee joint.

4.3.6 CADENCE RESPONSE

The cadence of an amputee is a very good determinant of the flexion of the knee during gait. The amount of flexion will decide the number of steps taken, which will further decide the speed of the amputee. With the use of a MPC knee unit it is anticipated that the patient will have a more comfortable knee flexion without any fear of falling as the knee itself initiates flexion. With a good flexion control one could expect to find good speed and even variable speed. Hence, it becomes necessary to check for the cadence response of a knee joint.

4.3.7 TIMING OF ALL KEY MOVEMENTS

Finally to get a satisfactory gait it is required that all the above mentioned processes happen at the right time and the right phase of the gait cycle. Failure to do so could prove to be fatal as the amputee relies completely on the functions of the MPC knee unit while walking.

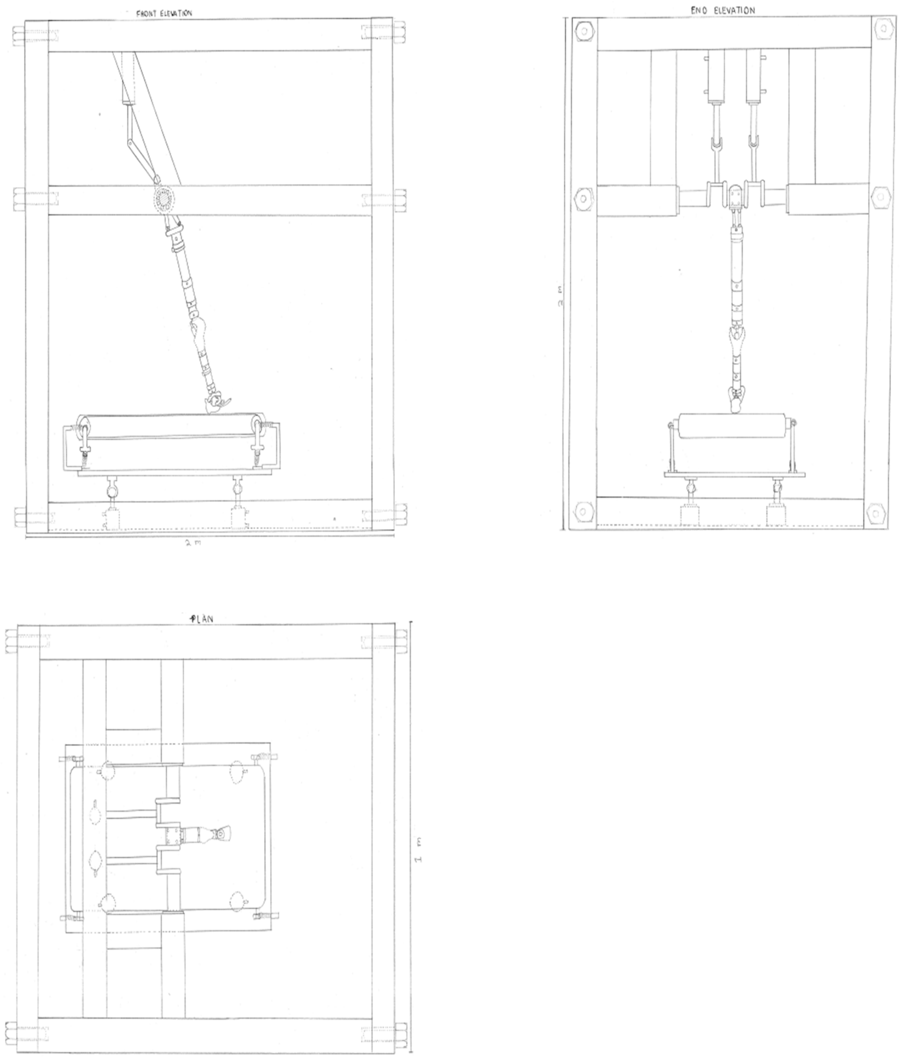
CHAPTER 5. GENERAL SPECIFICATIONS OF A DESIGN FOR A BENCH TESTER FOR MPC PROSTHETIC KNEES

5.1 INTRODUCTION

A bench tester or test bench is a device which aims at creating an environment for checking and evaluating the correctness and/or soundness of a device. The aim here lies in planning and devising a bench tester which can evaluate the functional qualities of MPC knee units. As mentioned earlier, there has been very little work published in this aspect. Hence, it has been difficult to compare the proposed plans/ideas with any other concepts.

The MPC knee units are capable of functions similar to the anatomical knee joint. This is the major reason for the popularity. The anatomical knee joint has a complex structure as do MPC knee unit designs. It is important that the MPC's be tested and evaluated according to their claimed functions.

To create the right environment for testing knee units it is required to produce conditions similar to the anatomical limb. For functional testing it is essential to devise forces similar to those of the human body acting on the knee joint. Simulation of the right hip angles is also very important to get the right movement from the knee unit. It has been my endeavor to incorporate the inherent properties of human locomotion within the design of the test bench. In the following sections the components of the design will be presented along with the functioning of the system. Figure 5.1 shows the plan for the machine.



For Figure 5.1Machine view in 1st Angle - Front elevation, End elevation & Plan

5.2 STRUCTURAL COMPONENTS

5.2.1 FRAMEWORK

The exoskeleton of the machine needs to be a sturdy construction. It is this member of the system which carries the load of all the other members. It is required that while the machine operates, the framework should not vibrate or produce unwanted movements. For this purpose it is essential to do a vibrational analysis before construction of the machine. The resonant frequency of the framework has to be more than the working frequency of the machine.

For the purpose of this design 4 cross members will be used. These will support the upper portion of the frame. In the upper portion a horizontal 'I' beam will be present, in the center of which can be attached the proximal most part of the machine (load producing actuators). Coming down approximately one third from the top there will be two horizontal beams (opposite to each other) while looking in Front Elevation (1^{st} angle) (Fig 5.2). While looking from the End Elevation (Fig 5.3) two L shaped cross members will be coming down from the top portion of the frame and be connected to the two horizontal beams. Between these members will be connected the shaft. At the base a plate will need to be fused with the framework for linkage of hydraulic actuators.

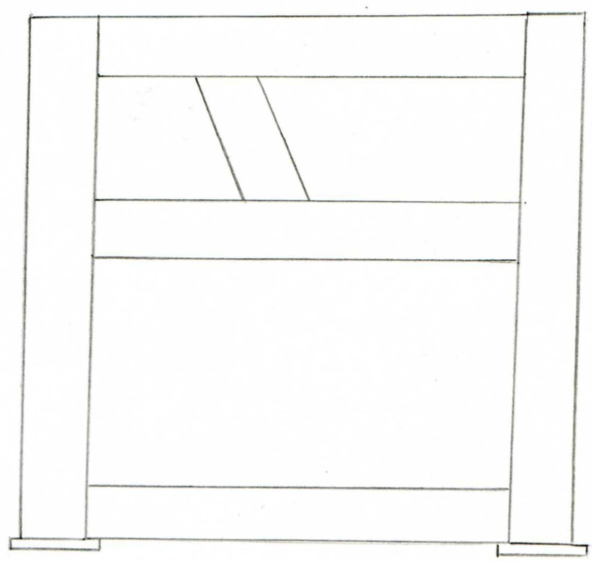


Figure 5.2 Framework in Front Elevation (1st angle)

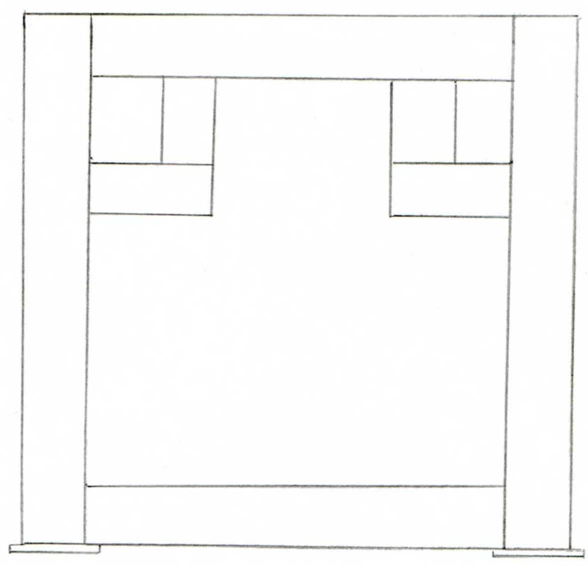


Figure 5.3 Framework in End Elevation (1st angle)

5.2.2 LOAD PRODUCING SEGMENT

To provide conditions similar to the anatomical lower limb a load simulating that of an amputee is to be put on the MPC knee. This section will deal with the load actuator member of the system. A further section will be dealing with the loads and forces experienced by the knee unit during the process.

It has been time and again proved that Hydraulic systems are the best method for moving an object by the right application of load. Pneumatic systems are handy, but the advantages of the hydraulic systems surpass them. A special mention needs to be made about the amount of force that can be delivered under control. This is possible by the use of fluids. Liquids being incompressible, the force rapidly gets transmitted equally in all directions.

Hydraulic actuators are majorly available in two types: Linear and rotary (Parr 1998). For the purpose of this project a linear actuator has been made use of. Linear, as the name implies, is used to apply force or move an object in a straight line. The basic linear actuator is shown in the figure 5.4. It basically consists of a cylinder within which is a piston of radius R, moving within a bore. The piston is connected to a rod of radius r, which drives the load. For moving the piston up or down the fluid needs to be pushed in from either one of the valves.

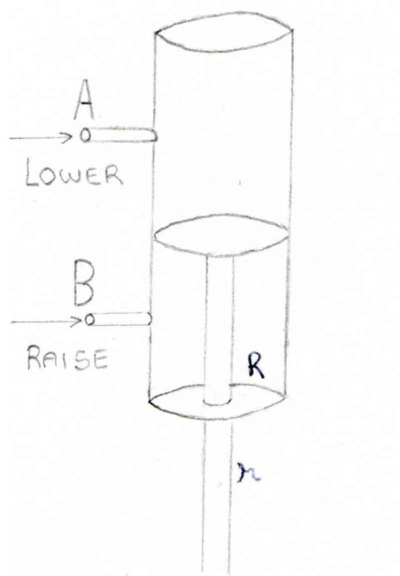


Figure 5.4 Linear actuator

5.2.2.1 ACTUATOR COMPONENTS AND WORKING

The machine design utilizes 2 actuators. Both of these are required to give a load of 400 N each. The figure 5.5 shows the circuit being used for each of the actuators for loading the machine. The system requires a liquid to operate it. Fluid is drawn from the reservoir by a pump which produces a fluid flow of a very high pressure. At such high pressures the fluid cannot be allowed to go into dead end space. Thus, there needs to be a form of pressure regulation. Cylinder movement is controlled by a three position valve. To take the piston down for producing the load the port A is connected to the pressure line and port B is connected to the tank. For reversing the motion of cylinder port B is connected to pressure line and port A is connected to tank.

The speed control is achieved by the regulation of the volume of fluid being sent to the cylinder. The precise control of speed is a big advantage of Hydraulic systems. The travel limits of the cylinder will be determined by the cylinder stroke. In subsequent sections it will be clarified how the length of the piston rod will control the motion of the whole assembly.

The pump needs to be turned on by an external power supply, most commonly an AC induction motor. The fluid within the reservoir needs to be kept clean; hence a filter is added to keep it clean as it passes from the reservoir to the cylinder.

5.2.2.2 FRAMEWORK LINKAGE

The hydraulic actuators are the proximal most part of the machine system. For the system 2 actuators are being utilized. They will be bolt jointed with the framework at its proximal most part. The actuators will be linked at the center of a beam. A solid attachment gives them the advantage of stabilizing the lower segments and also providing the proper load required.

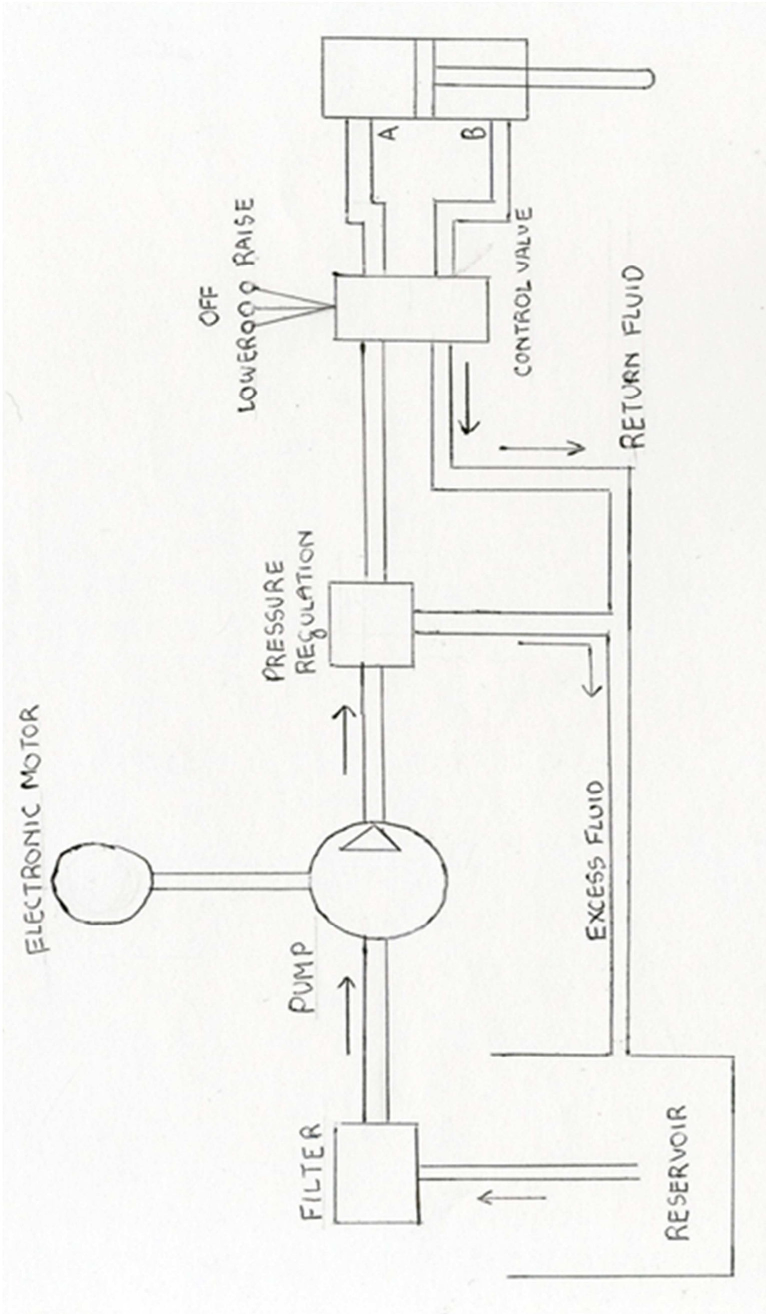


Figure 5.5 Flow of fluid in load producing actuator

5.2.3 MOTION PRODUCING SEGMENT

The movement of the lower limb comes mainly because of the applied force by the actuator. But the force by the actuator is not transmitted directly to the lower limb assembly. The linear force is transmitted into a rotation, which further brings about swinging of the whole limb assembly. This is carried out via the help of a crankshaft assembly.

5.2.3.1 CRANKSHAFT

"*The Crankshaft, sometimes casually abbreviated to Crank, is the part of an engine which translates reciprocating linear piston motion into rotation",* Wikipidea.org.

Here, the crankshaft is not used for a whole cycle as it would be within an engine. The endeavor here is to make use of only partial movement of the crankshaft. The hip joint of a healthy person goes into a maximum of 25° of flexion and 25°of extension in normal speed walking. Hence, a total of 50° is covered in locomotion. The intention thus, is to just bring about only 50° of rotation of the crankshaft assembly.

It was initially planned to make use of 'stops' to limit the motion of the crankshaft. But 'stops' have a quite a few inherent disadvantages -- wearing out being one of them. Hence, it was decided to use the piston rod as a means of controlling the rotation. With the right length of the piston rod the desired amount of rotation of the cranks can be achieved.

For calculating of the right length of piston rod it is first vital to understand that the connecting rod (fig 5.6) needs to be kept at angle of 50° to the horizontal. The piston rod length needs to be equal to the vertical distance of the connecting rod. Beyond the connecting rod are the cranks connected to each other via a crank pin. The crank pin is also the point of contact of the connecting rod. The cranks need to be aligned at a certain angle for the shaft to rotate.

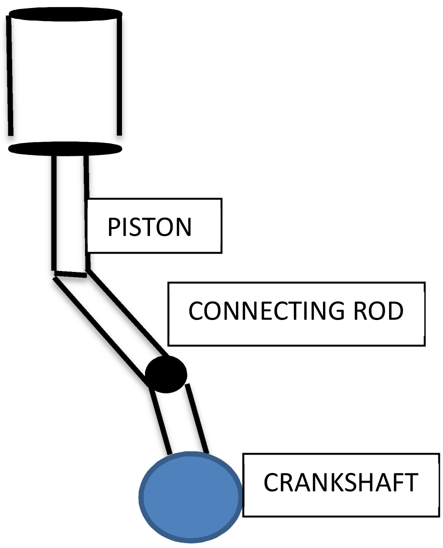


Figure 5.6 Position of the connecting rod in respect to the piston rod

So in a sense the shaft rotates when the piston rod moves down because of the vertical load. The piston compresses the connecting rod, eventually making it parallel to horizontal. With a steady compression of the connecting rod, the crank also starts to rotate and covers a whole of 50° in clockwise direction. In the second half the piston rod moves up and the crank returns to its initial position.

5.2.3.2 LINKAGE TO FRAMEWORK

The crankshaft will need to be connected between the two L shaped vertical members of the Framework. This position will be a bit offset as compared to the hydraulic actuators as the connecting rod is aligned at 50° to the horizontal. To incorporate the rotation of the shaft it will be connected via ball bearings to the Framework.

5.2.4 LOWER LIMB ASSEMBLY

The lower limb assembly onto which the MPC knee unit will be attached is made of a number of components. Some of these components remain specific for each MPC knee unit. For the sake of this chapter the prosthetic components required for fitting a C-Leg® have been listed.

5.2.4.1 HIP JOINT

The commercially available hip joints for Hip Disarticulation patients have a plate for attachment to the socket. But, in the case of the machine design, this plate can be taken out and the joint will need to be attached with a clamp, which will be fit around the shaft

The requirement of the hip joint in the present scenario is not to provide for motion at the hip region. This will be enacted by the shaft. The hip joint will always be immovable during the motion of the assembly.

However, it serves the purpose of keeping the leg in a fixed angle and then locking it. This assists in the operation of the machine. It is decided to keep the leg in 25° of flexion so as to start of the operation with a Heel Strike. The presence of the hip joint hence, is beneficial and justified. Also while testing for ascending or descending a ramp the assembly is meant to be kept in further 10° of flexion i.e. 35°.

Figure5.7 Prosthetic Hip Joint (adopted from ottobock.com)

5.2.4.2 THIGH PYLON

The gap between the hip region and the knee joint is filled by a pylon. It represents the thigh region. Initially it was thought of to use some form of loading at the high region and also to make use of pneumatic actuators to replicate the muscle action. But, it just complicated matters and the goal shifted towards the muscle action rather than the evaluation of the MPC knee unit. Therefore, it was decided to keep the thigh region as simple as possible and as near to normal amputee conditions. Hence, a thigh pylon is made use of in the design.

A thigh pylon is usually made out of Stainless Steel, Aluminum or Titanium. The weight of a thigh pylon is pretty much negligible. But, it is an excellent way of transmitting the forces further below to the knee joint. The figure 5.8 shows a basic thigh pylon. At its lower end it is inserted into a tube clamp. The clamp allows its attachment to a pyramid. At its upper end it goes into a tube clamp adaptor.

Figure 5.8 Thigh pylon with adaptor proximally and tube clamp distally (adopted from ottobock.com)

5.2.4.3 STRAIN GAUGING

It is vital to keep a constant check of the forces and the moments at regions above and below the MPC knee unit. For this purpose tubular transducers are needed. Berme et all, 1975 at the Bioengineering Dept. of University of Strathclyde, developed a shorter Pylon transducer to measure forces and moments during the amputee gait. The transducer can be fitted at any desired location on the pylon. With the presence of end adaptors it can be clamped with the pylon (thigh or shank).

5.2.4.3.1 Details of the structure

The basic structure of the transducer consists of a tubular structure with flanges on which were fitted the adaptors for mounting on the prosthesis. The tubular structure and the flanges were machined as one component. 6 strain gauges forming full bridges were attached using a heat curing adhesive. The gauges were placed at two levels as it was not possible to accommodate them together. The bending moment and the axial loads were grouped together as were those measuring axial loads and torque.

Bending moment and shear were each measured in the two orthogonal planes (A/P and M/L). Axial load was obtained using four 90° rosettes. Torque and shear were measured using 90° rosettes with gauges aligned at 45° to the principal axis of the transducer. However, the positioning and the wiring of the gauges differed between torque and shear measurement. This is illustrated in the figure 5.9

Figure 5.10 shows the short pylon transducer. The adaptors used in this case are different from the ones utilized in the lower limb assembly. This however, can be solved by just changing the type of adaptor. At both the upper and lower ends of the transducer it will need to be fit with pyramids, which can be inserted into the pyramid adaptors.

5.2.4.3.2 Transducer mounting

Two transducers will be needed for the design. One of the transducer needs to be placed mid-way between the knee joint and the shaft. Another one would need to be placed just above the foot. The transducer will need to be aligned with the thigh and shank pylon axis.

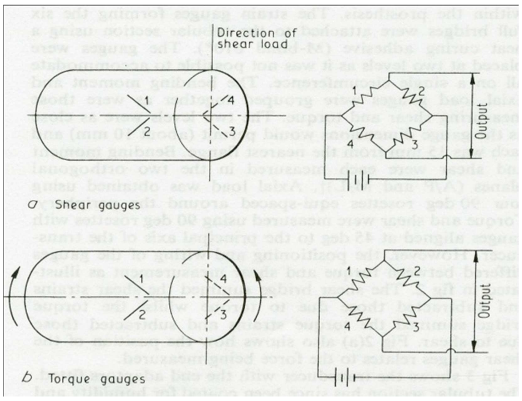


Figure 5.9 Positioning of the strain gauges on the pylon transducer (adopted from Berme et all 1975)

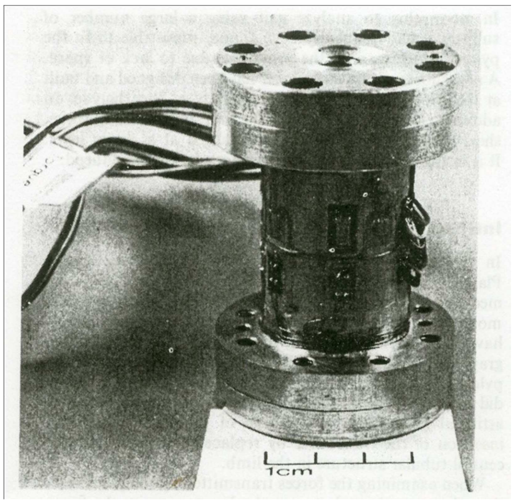


Figure 5.10 Pylon transducer with removable adaptors proximally and distally (adopted from Berme et all 1975)

5.2.4.4 KNEE JOINT ADAPTOR

The link of the MPC knee joint with the Thigh pylon tube is via the tube clamp adaptor in this case it is a double ended adaptor (fig 5.11). The adaptor type can be a Pyramid or even a threaded one. Preferably a pyramid adaptor is used. It gives better grip of the knee joint.

The knee adaptor forms a very important member of the prosthetic system. Its importance lies not just as a linkage, but also in the fact that it is the member that brings about the alignment of the trans-femoral socket. Without the right alignment of a socket one cannot expect to get the right results from the knee joint. A pyramid adaptor provides rotational alignment to the socket i.e. flexion-extension, adduction-abduction. Another type of adaptor manufactured by Blatchford, provides linear adjustment of the socket in relation to the knee joint.

Figure 5.11 Double ended adaptor (adopted from www.ottobock.com)

5.2.4.5 SHANK PYLON

As a thigh pylon formed the link between the hip region and the knee joint, a shank pylon is the link between the knee joint and the prosthetic foot or prosthetic ankle. It is again made out of Stainless Steel, Aluminum or Titanium. It is a very light component of the whole assembly. But, it carries a major load of the amputee's body.

A very interesting observation made by early researchers was that the shank pylon of lower limb amputees tends to crack with use at its distal portion. This was due to the fact that the major part of the load came on the distal portion during final stages of the stance phase. To prevent this, a special feature was added to the pylons. Manufacturers started adding an extra few layers just for the distal portion of the pylon. If one would to view a side section of the pylon, it would be found that lower distal segment is thicker internally than the rest of the pylon.

Figure 5.12 shows a general pylon to be used with the C-Leg® knee joint. The pylon is inserted at its distal portion into a tube clamp and proximally into a tube adapter.

Figure 5.12 Shank pylon (adopted from www.ottobock.com)

5.2.4.6 PROSTHETIC FOOT

For evaluating the function of a MPC knee unit by replicating lower limb movement it is vital to install a prosthetic foot which is the distal most segment of the lower limb assembly. All the load of the upper body is transmitted to the ground via it and also the ground reaction force comes through it. Hence, it is required that the foot be sturdy and also resilient enough to provide for a very smooth stride. The anatomical foot is a very complex structure and it is difficult to replicate its functions. But, there have been quite a few advancements in prosthetic feet. Manufacturers present a variety of feet suitable for each activity level. Also there are knee joint specific feet. Figure 5.13 shows the prosthetic foot chosen for the design.

Figure 5.13 C-Walk Prosthetic foot (picture adopted from www.ottobock.com)

5.2.4.7 LINKAGE TO SHAFT

The limb assembly will be required to be connected to the shaft of the Crankshaft mechanism for its movement. This linkage needs to be via a clamped connection, so that the assembly moves with the shaft. Figure 5.14 shows the clamping of the assembly to the shaft. Also at its distal portion the clamp will need to accommodate the pyramid head of the hip joint. Thus, at the base of clamp there needs to be a screw able plate on the lower portion of which a pyramid adaptor can be fused.

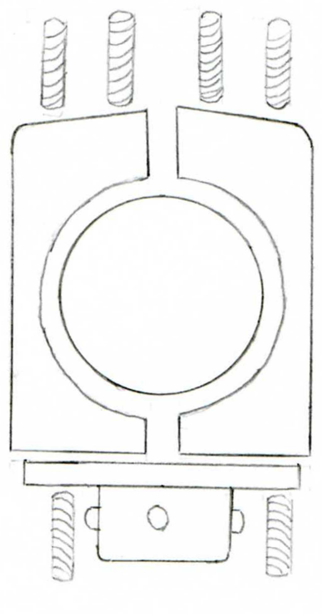


Figure 5.14 Clamp for shaft linkage, with adaptor at distal end

5.2.4.8 MACHINE BASE

5.2.4.8.1 Considerations

- It is understood that with the movement of the lower limb the whole machine is not expected to move. Therefore, for simulation of a good walking cycle it is necessary to make use of a conveyer belt or a treadmill type of base.
- Also the base needs to be flexible enough to compensate for the shortening and lengthening of the lower limb while moving in stance.
- For swing phase there needs to be a mechanism which will make the base drop down, to let the leg clear the ground.
- The shear forces need to be taken into consideration and some form of compensation has to be given for them.

5.2.4.8.2 Components

5.2.4.8.2.1 Belt Drive Platform

For a smooth roll off of the prosthetic foot a rotating belt or a conveyer belt is required. This can be similar to any treadmill, commonly seen in health clubs. With just a hard surface there are chances the system might just jam and cause damage to the MPC knee unit. With the presence of a belt drive and a platform underneath it, it is expected that the knee joint shall complete all the stages of the gait cycle. The belt will need to be connected to an AC motor supply for it to run.

Figure 5.1 shows the belt moving on a platform having roller ends for the belt rotation. The platform lies just underneath the belt and the belt is flexible enough to allow the foot to have a contact with the platform.

5.2.4.8.2.2 Brackets

The belt platform is held upright with the help of brackets attached to the roller ends of the platform. For providing the upward and downward motion of the platform, which will further compensate for the leg shortening and lengthening (similar to the vertical motion of the Centre of Gravity) the brackets will be stabilized on 4 springs. Also for compensating for shear forces (AP), 4 springs will be attached horizontally to the four ends of the bracket. During the start of the operation the vertical springs will already be in compression by the load of the belt platform and subsequently by the force transmitted by the prosthesis. Hence, during the cycle the sprigs will not be allowed to expand completely. This will help in keeping them stable. Also at the base they will be attached to spring keepers.

5.2.4.8.2.3 Base Hydraulics

For a complete gait analysis the lower limb will need to perform both stance and swing. For simulating the swing phase either the limb will need to move (normal movement) or the ground will need to drop down. For this project it has been decided to include a mechanism by which the platform and the bracket can be made to come down during swing, so as to allow the lower limb to swing through. For the same purpose it will be convenient to make use of a hydraulic system, which just provides enough power to lift the whole platform up and bring it down.

The design makes use of 4 hydraulic actuators to bring the vertical motion of the platform. The cylinders will be connected to a horizontal plate, which will be fused to the Framework. The piston rod will in turn be connected to another plate upon which the springs are localized. The end of each piston will need to be shaped as a spherical rod end. This is to facilitate positioning of the belt platform at an angle, which will be helpful when testing for the MPC knee joint properties in ascending or descending a ramp. The length of the pistons has to be just enough to make contact of the belt with the prosthetic foot and enough space is left for the foot to clear the belt when in swing

5.2 FORCE ANALYSIS OF STANCE PHASE

The main aim for doing a force analysis is to understand whether or not the knee joint is getting the right amount of force. This is essential for two reasons. Firstly, it is important that the properties of the knee joint be evaluated under the most natural conditions as much as possible. Secondly, overloading the knee joint can be fatal and the load must not exceed the manufacturers given limit.

As mentioned in the above section the hydraulic actuators will be providing a load of 400N each to the crankshaft for its rotation. This load combined comes onto the center of the shaft at the point of clamping of the shaft with the lower limb assembly. So in a sense the combined load from the hydraulic cylinders comes onto the hip region of the lower limb assembly.

Figure 5.15 shows the vertical load from the cylinders. The connecting rod lies at 50° to the horizontal and 40° to the vertical.

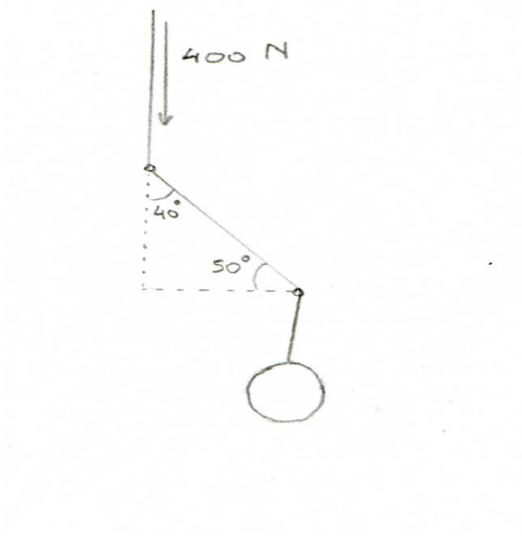


Figure 5.15 Stick diagram of piston rod, connecting rod and crankshaft.

Hence, the resultant force coming onto the Crankpin, F_{pin} is given by:

$$F_{pin} = \frac{400}{\cos 40}$$

$$= 522.1 \text{ N} \qquad (5.1)$$

Equation (5.1) gives the force by one hydraulic actuator. Hence, the combined force on the shaft..

$$F_{shaft} = 2 \times F_{pin}$$

$$= 2 \times 522.16$$

$$= 1044.32 \text{ N} \tag{5.2}$$

Equation (5.2) gives the total force coming onto the centre of the shaft. It can be approximated to 1044 N. At this point is also the attachment of the hip unit. Hence, from here on force analysis can be done according to the different stages of stance.

5.2.1 HEEL STRIKE

At the begening of the gait cycle the hip will remain in an intial flexion of 25° to simulate heel strike. So effectively at the instant of heel strike there is no force acting on the joints. It is only when the limb crosses heel strike and starts to move into the next stage that the force starts to increase. So this analysis will be at a stage just after heel strike.

At this point the hip still remains in 25° of flexion, but the knee starts to move from extension into flexion of approxiamtely 5 to 7°. The ankle moves into just about 5 ° of plantar flexion.

From the figure 5.16 the vertical and shear forces in Anterior – Posterior (AP) plane at the hip can be given as…

$$F_V = 1044 \cos 25$$

$$= 946.2 \text{ N} \tag{5.3}$$

$$F_S = 1044 \sin 25$$

$$= 441.2 \text{ N} \tag{5.4}$$

The direction of forces can be seen in the figure 5.17.The force direction at each joint will be opposite to the one preceeding it. The forces will remain constant within one frame of reference.

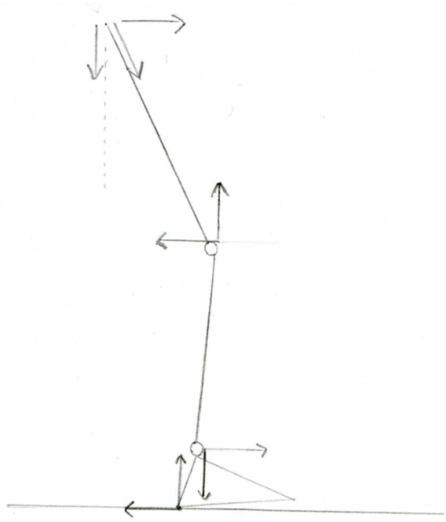


Figure 5.16 Force directions during heel strike

5.2.2 FOOT FLAT

The hip moves into 22° of flexion and the knee lies in 15° of flexion. The ankle remains in 10° of plantarflexion.

From the figure 5.17 the vertical and shear forces in AP can be given as..

$$F_V = 1044\ \cos 22$$

$$= 967.98$$

$$= 968 \text{ N approx.} \tag{5.5}$$

$$F_S = 1044 \ \sin 22$$

$$= 391.09 \text{ N} \tag{5.6}$$

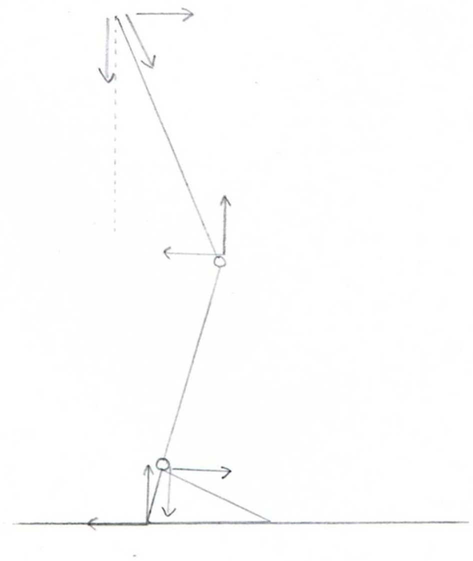


Figure 5.17 Force directions during foot flat

5.2.3 MIDSTANCE

During this stage the hip moves into 3° of extension from 10° of flexion at this stage. From the figure 5.18 the vertical forces and shear forces in AP can be given as….

$$F_V = 1044 \ \cos 3$$

$$= 1042.6 \text{ N} \tag{5.7}$$

$$F_S = 1044 \ \sin 3$$

$$= 54.64 \text{ N} \tag{5.8}$$

Figure 5.18 Force directions during midstance

5.2.4 HEEL OFF

The hip reaches its maximum in extension i.e. 25° and thereafter begins to flex for the next phase. From the figure 5.19 the vertical and shear forces in Ap can be given as...

$$F_V = 1044 \ \cos 25$$

$$= 946.18 \text{ N} \tag{5.9}$$

$$F_S = 1044 \ \sin 25$$

$$= 441.2 \text{ N} \tag{5.10}$$

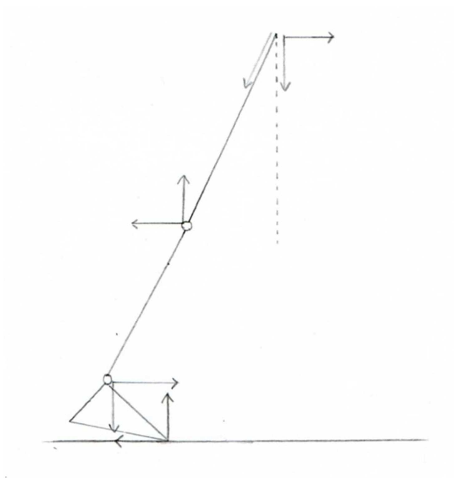


Figure 5.19 Force directions during heel off

5.2.5 TOE OFF

The final stage of stance phase. The hip lies in 16° of extension and starts moving towards flexion. From the figure 5.20 the vertical and shear forces in AP can be given as....

$$F_V = 1044 \ \cos 16$$

$$= 1003.56 \text{ N} \tag{5.11}$$

$$F_S = 1044 \ \sin 16$$

$$= 287.77 \text{ N} \tag{5.12}$$

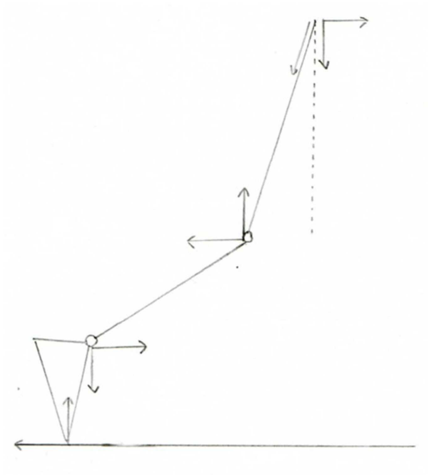


Figure 5.20 Force directions during toe off

5.2.6 BRIEF ANALYSIS

From equation (5.7) it is clear that the maximum vertical force comes in the Midstance phase. The maximum shear force comes just after Heel strike and during Heel off as shown by equations (5.4) and (5.10).

This is an important conisderation, as the maximum force in vertical and shear will need to be taken into consideration while making the choice of springs for the Belt Platform. The springs should be able to take the force and provide the necessary compression or elongation. Also the resultant force at the shaft is 1044 N, which lies just below the maximum limit (1060 N) set for a 80kg man by the Phyladelphia Standards committee.

5.4 OPERATING CYCLE

5.4.1 GENERAL OPERATION

The process starts of with the hydraulic actuators at the base of the desgn rising to bring the belt platform in contact with the heel of the foot. Subsequently the actuator applies a downward load on the piston and the belt platform starts to rotate at the same time. This is brought about by the motor of the hydraulics and the belt drive motor. Following this, the connecting rod starts to get compressed under the action of the load. With the compression of the connecting rod, the crank is pushed to rotate in a clock wise manner, the crank in turn rotates the shaft. The shaft finally makes the lower limb assembly move from a stage wherein it is in 25° flexion initially and subsequently it moves into the other stages of gait cycle.

As the piston compresses the connecting rod completely and brings it to an angle of 0° to the horizontal, the valve of the hydraulic cylinders changes the direction of flow of the liquid fluid. The liquid flows into the other chamber of the cylinder and makes the piston move in the upper direction. Thereby, pulling the connecting rod back up steadily. This motion also makes the shaft return to its starting position. As the shaft reaches an angle of 16° in extension, the base of the machine drops down by the action of the hydarulic units. This is to allow the knee joint to flex and clear the ground for the swing phase.

When the piston has gone completely up and the connecting rod is back to its original position, the shaft has also completed its complete cycle and the limb is again back to 25° of flexion for making the heel contact. The base again rises and the limb is ready for the next cycle.

5.4.2 STAGE DIVISION

The operating cycle of the design can be divided into 2 stages. This is dependent on the direction of rotation fo the shaft. The first stage of the cycle is when the shaft rotates clock-wise and the lower limb completes the first 4 phases of stance phase. The second stage of the cycle comes when the shaft is to return to its intial position and the lower limb completes the other phases of stance and swing. Hence, in a way the two stages of the operating cycle can further be classified into 4 phases each covering the whole of the gait cycle.

The following sections will deal with the motion of the lower limb during the operation of the machine. Also, the lenghtening and the shortening of the lower limb as a result of stance flexion will be discussed.

5.4.2.1 STAGE I

5.4.2.1.1 Phase 1

With the help of the hip joint the lower limb is kept in an initial flexion position of 25°. As the actuators have not yet started to apply the load . The hip lies in a flexion angle. The knee lies in complete extension, but starts to move towards flexion later on. Figure 5.21 shows the relative position of the joints att his time

The original length of the whole lower limb assembly is 1.086m. At the present stage the effective vertical length of the limb is given by the following equations:

$$L_1 = 1.086 \cos 25$$

$$= .984 \text{ m} \tag{5.13}$$

This stage refers to the Heel Strike Stage of the Stance phase of the human gait cycle. The loading of the prosthesis begins immediately after this stage. The springs of the base are kept in slight compression at this stage.

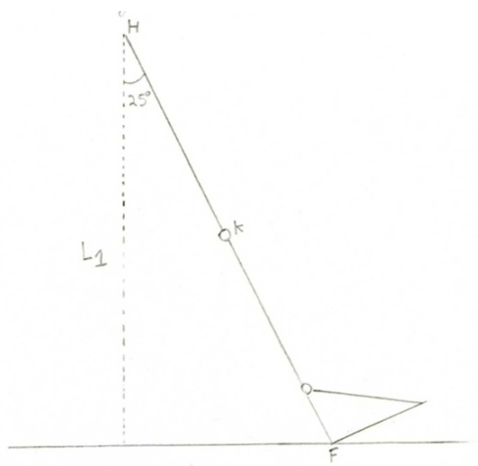


Figure 5.21 Vertical length of prosthesis at beginning of cycle

5.4.2.1.2 Phase 2

The second phase is when the actuators have actually started loading the lower limb assembly. The piston moves down and the crank rotates the shaft. This reduces the hip flexion angle and the knee flexion angle increases. The foot of the limb lies flat on the ground or the belt platform. The belt starts rotating and thus assists in the motion of the lower limb assembly. The effective length of the limb at this point is given by the following equations:

L_1 , Vertical distance from Hip to Knee

L_2, Vertical distance from Knee to foot

$$L_1 = HK \cos 22$$

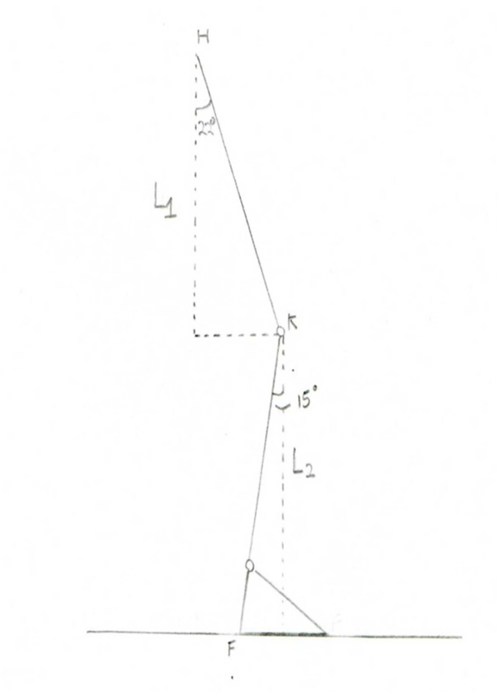
$$= .668 \cos 22$$

$$= .619 \text{ m}$$

$$L_2 = KF \cos 15$$

$$= .418 \cos 15$$

$$= .404 \text{ m}$$



$$T = L_1 + L_2 = .619 + .404$$
$$= 1.023\text{m} \qquad (5.14)$$

At this point the Prosthesis lies in the Foot Flat stage of the stance phase.

Figure 5.22 Vertical lengths during foot flat

5.4.2.1.3 Phase 3

The shaft reaches a rotation of about 10° in flexion. This brings the prosthesis at almost the center point. The knee also lies in flexion of about 10°. The prosthesis is reaching the stage of midstance. In an individual this stage requires for the leg to be able to take the complete load of the body. The maximum load comes around this point. The effective vertical position of the limb can be given by the following equations:

L_1 , Vertical distance from Hip to Knee

L_2, Vertical distance from Knee to foot

$$L_1 = HK\ \cos 10$$
$$= .668\ \cos 10$$
$$= .658 \text{ m}$$
$$L_2 = KF\ \cos 10$$
$$= .418\ \cos 10$$
$$= .412\text{m}$$
$$T = L_1 + L_2$$
$$= .658 + .412$$
$$= 1.07 \text{ m} \tag{5.15}$$

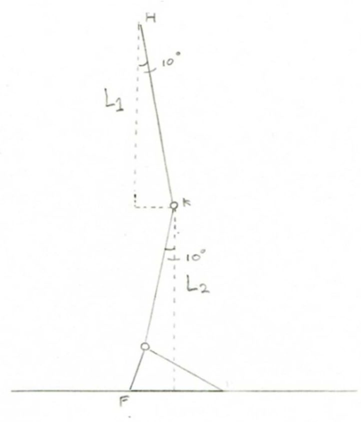


Figure 5.23 vertical lengths during midstance

5.4.2.1.4 Phase 4

The final part of the first stage is reached when the piston completely comes down its length. The shaft has had a complete rotation of 50°. It is now at an angle of 25° in extension. The knee joint is in about 5° of flexion. The foot is no more flat on the ground and the heel is not in contact (figure 5.25). This is the phase of Heel Off. The following equations give the effective vertical length of the prosthesis:

L_1 , Vertical distance from Hip to Knee

L_2, Vertical distance from Knee to foot

L_3, Vertical distance of heel to floor

$$L_1 = HK \cos 25$$
$$= .668 \cos 25$$
$$= .605 \text{ m}$$

$$L_2 = KF \cos(25 + 5)$$
$$= .418 \cos 30$$
$$= .362\text{m}$$

$$L_3 = FO \sin 15$$
$$= .24 \sin 15$$
$$= .062 \text{ m}$$

$$T = L_1 + L_2 + L_3$$
$$= .605 + .362 + .062$$
$$= 1.029 \text{ m} \tag{5.16}$$

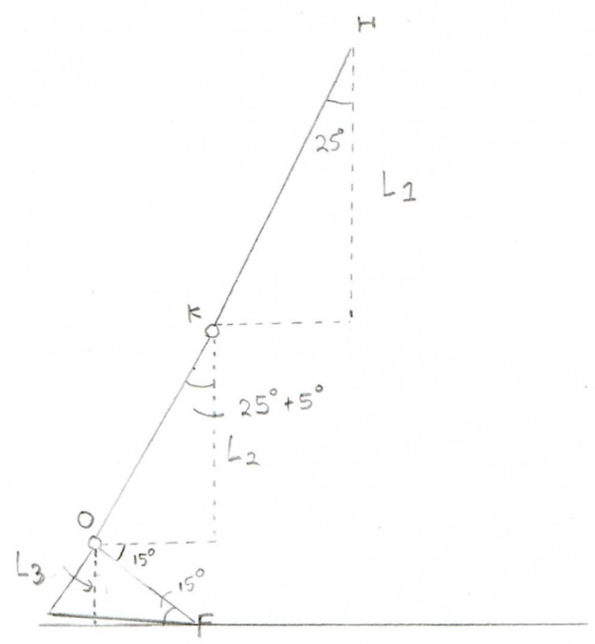


Figure 5.24 Vertical lengths during toe off

5.4.2.2 STAGE II

5.4.2.2.1 Phase 1

As soon as the pistons completely go down their length the direction of fluid flow within the chambers of the cylinders is changed and the piston now moves up. The shaft now rotates anti-clockwise and the prosthesis changes its direction of movement along with it. This is very similar to the motion of a normal healthy individual, wherein after the phase of heel off the limb advances forward to prepare for swing.

With the change of direction of the shaft the prosthesis enters the phase of Toe Off. The hip is still in extension. But the knee moves into a big flexion angle. With the flexion of the knee the foot is also moved into plantar flexion. The following equations give the effective vertical length of the prosthesis at this stage:

L_1 , Vertical distance from Hip to Knee

L_2, Vertical distance from Knee to ankle

L_3, Vertical distance of heel to floor

$$
\begin{aligned}
L_1 &= HK \cos 16 \\
&= .668 \cos 16 \\
&= .642 \text{ m}
\end{aligned}
$$

$$
\begin{aligned}
L_2 &= KA \cos(16+40) \\
&= .347 \cos 56 \\
&= .194 \text{ m}
\end{aligned}
$$

$$
\begin{aligned}
L_3 &= FO \cos 17 \\
&= .24 \cos 17 \\
&= .23 \text{ m}
\end{aligned}
$$

$$
\begin{aligned}
T &= L_1 + L_2 + L_3 \\
&= .642 + .194 + .23 \\
&= 1.066 \text{ m} \qquad (5.17)
\end{aligned}
$$

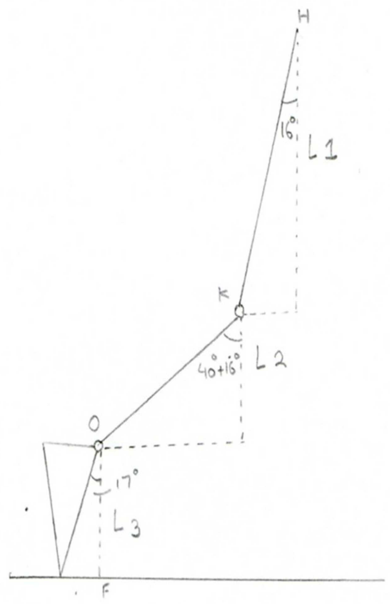


Figure 5.25 Vertical lengths during toe off

5.4.2.2.2 Phase 2

As soon as the prosthesis completes the different stages of the stance phase and starts to enter into swing, the base belt platform drops down by the action of the hydraulic actuators. The prosthesis can now perform the swing phase. This phase pertains to the first segment of the swing phase.

5.4.2.2.3 Phase 3

The prosthesis is almost at the center point. The hip is in about 20° of flexion. The knee joint should be sufficiently flexed for this phase even with the dropping down of the Belt platform. This phase relates to the Mid Swing stage of the swing phase.

5.4.2.2.4 Phase 4

The final phase is reached when the prosthesis is again at an angle of 25° to the vertical and the rest of the joints are coming towards an extension position. The prosthesis is preparing for another heel strike.

As soon as the limb finishes this stage, the belt platform rises and makes contact with the heel of the foot. The process is again repeated.

Figure 5.26 to 5.33 give the movement of the prosthesis during the different phases of the machine.

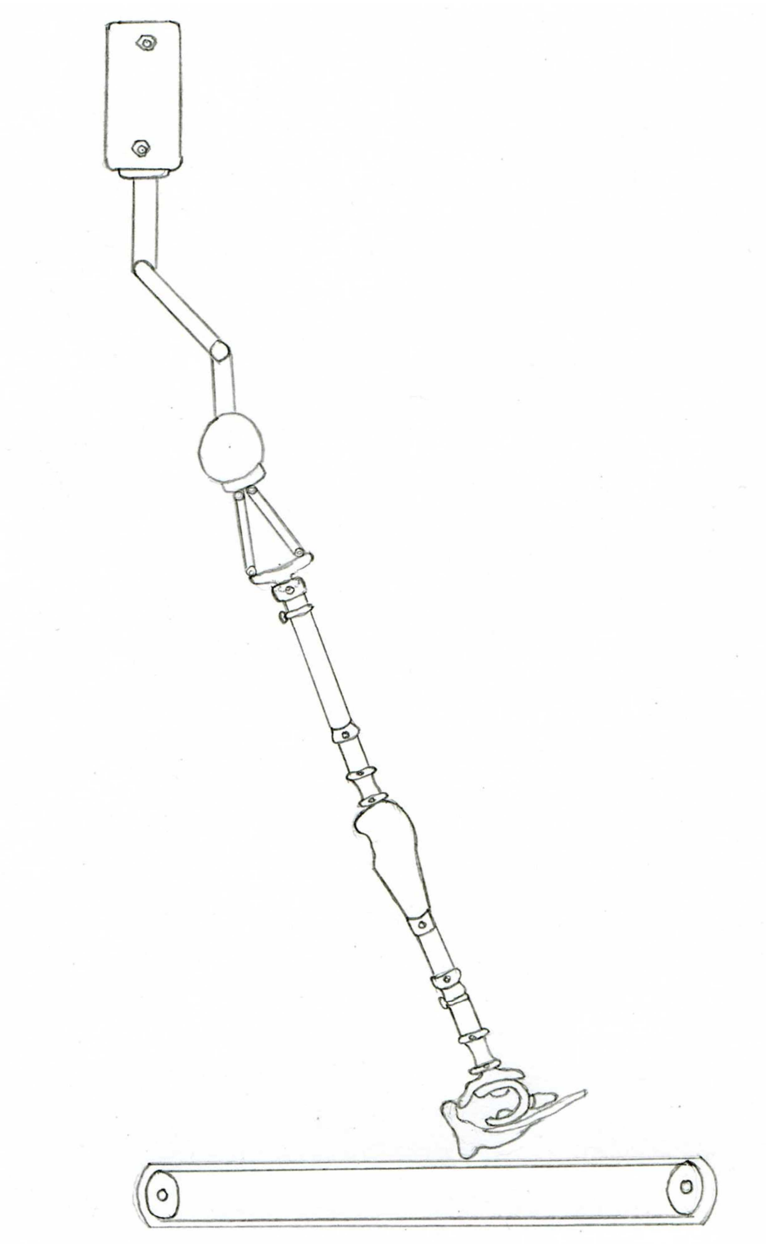


Figure 5.26

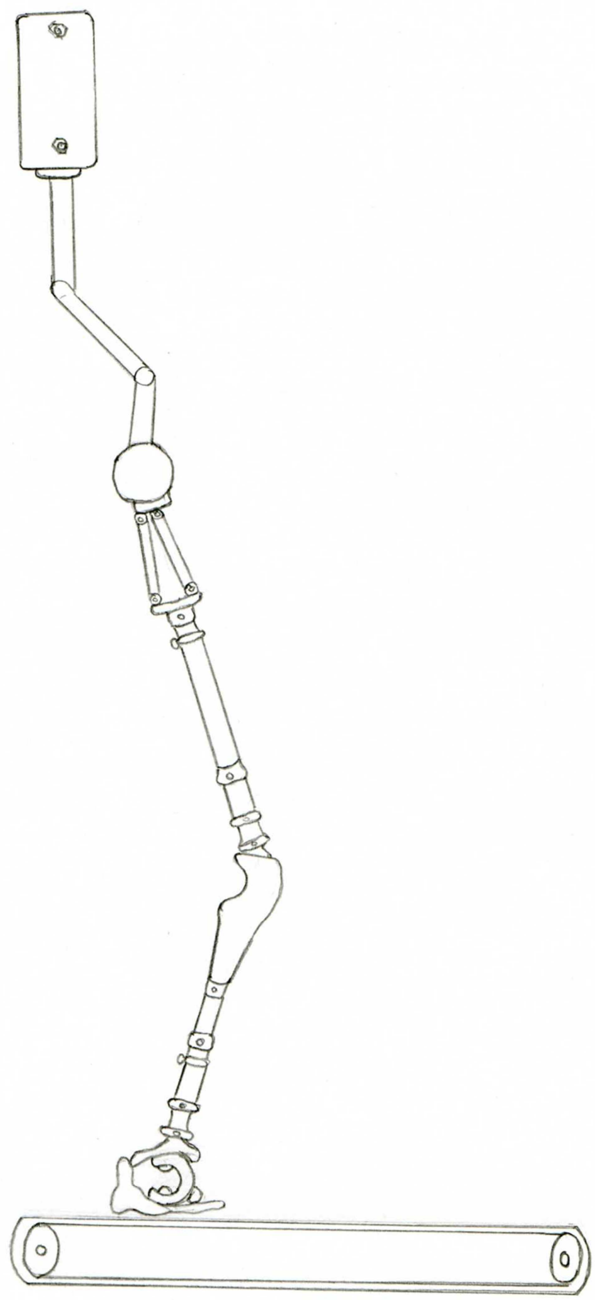


Figure 5.27

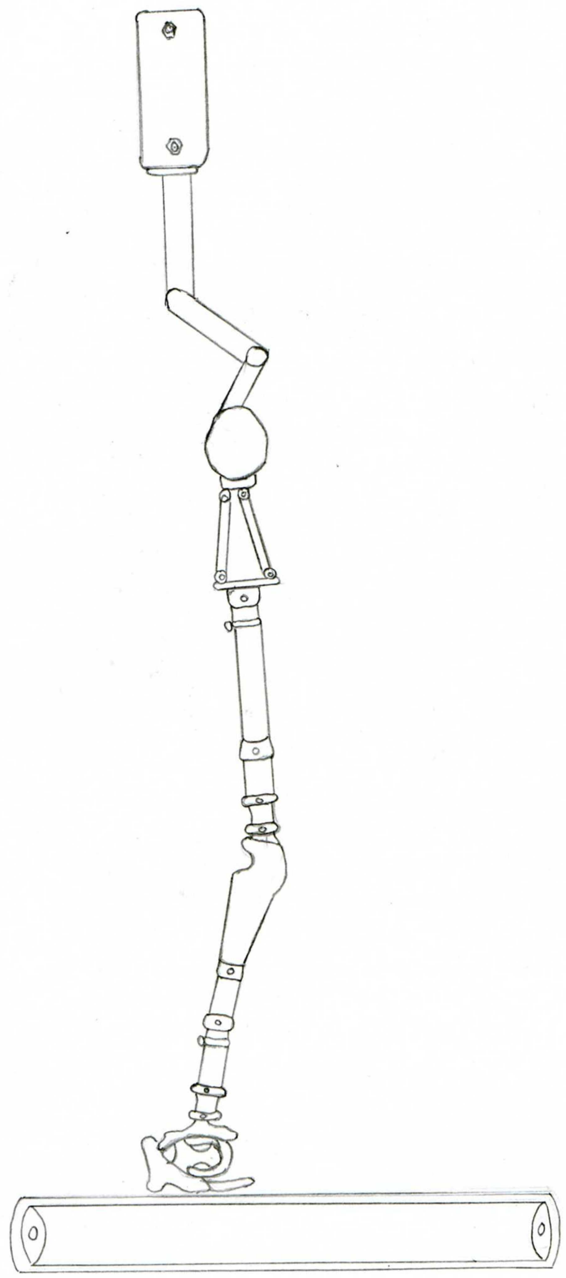


Figure 5.28

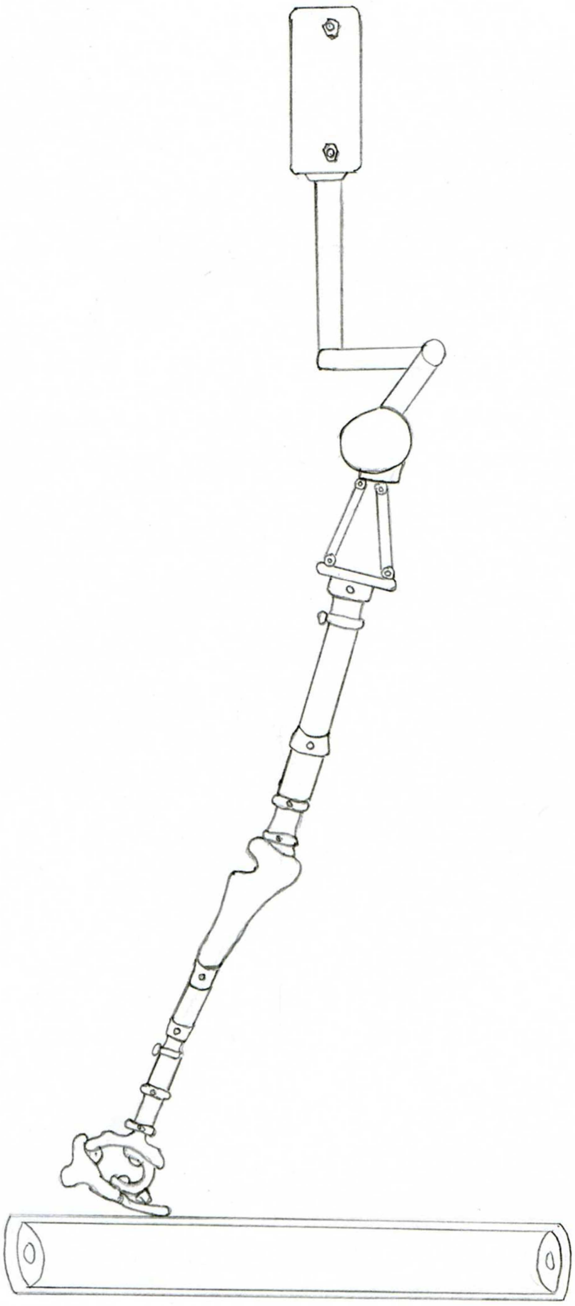


Figure 5.29

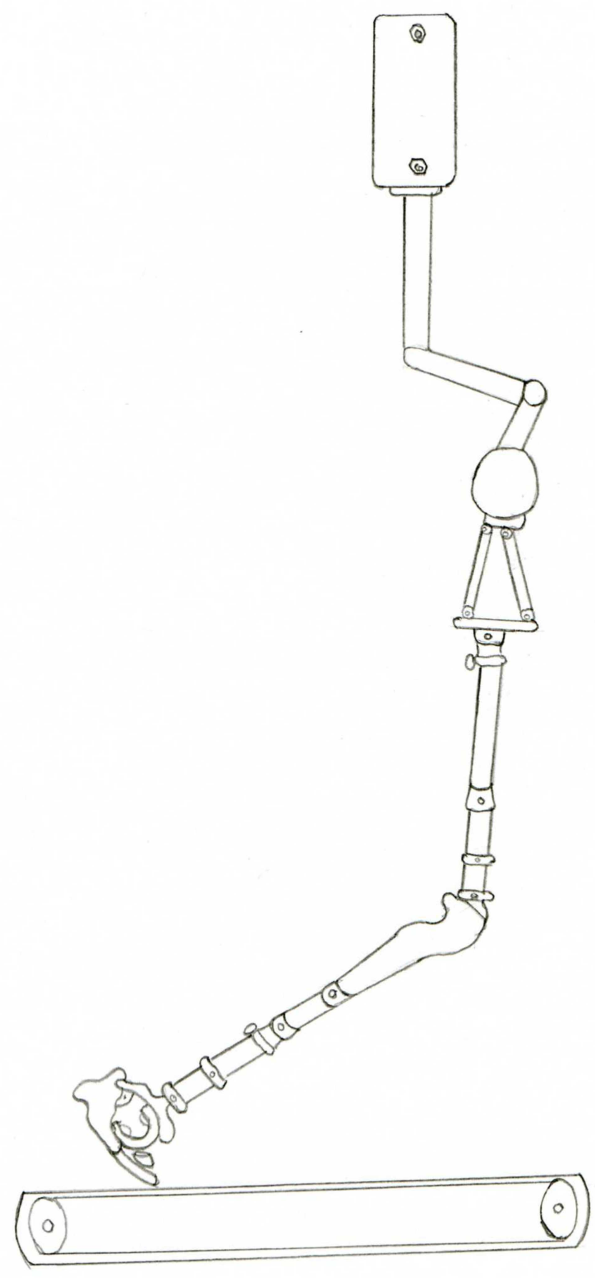


Figure 5.30

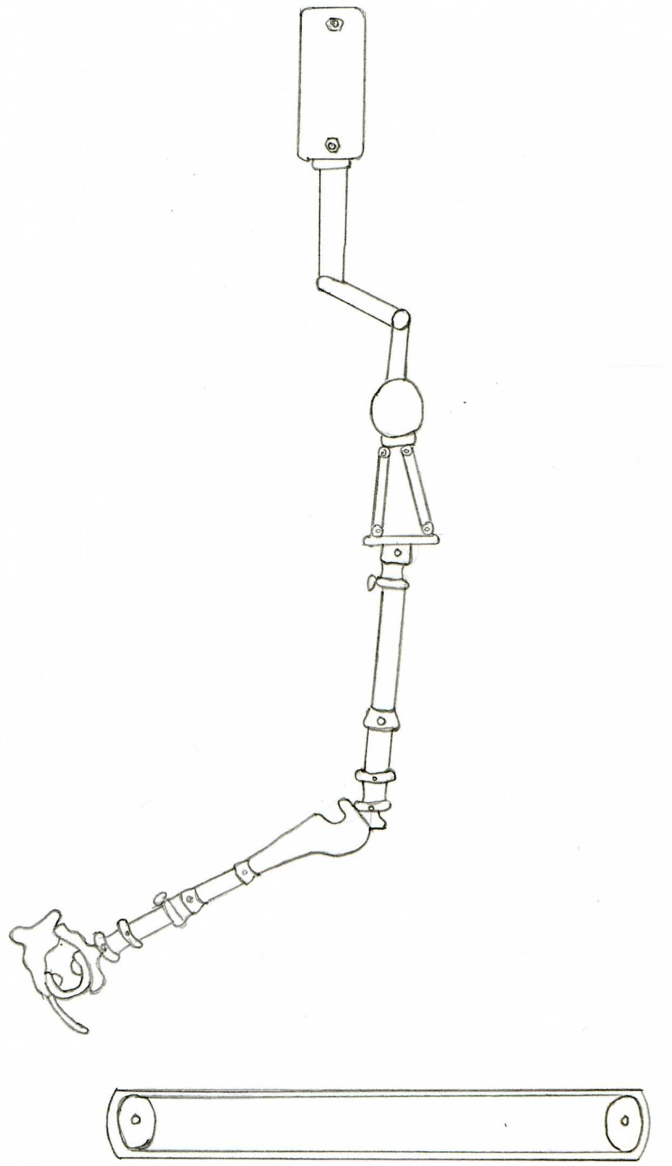


Figure 5.31

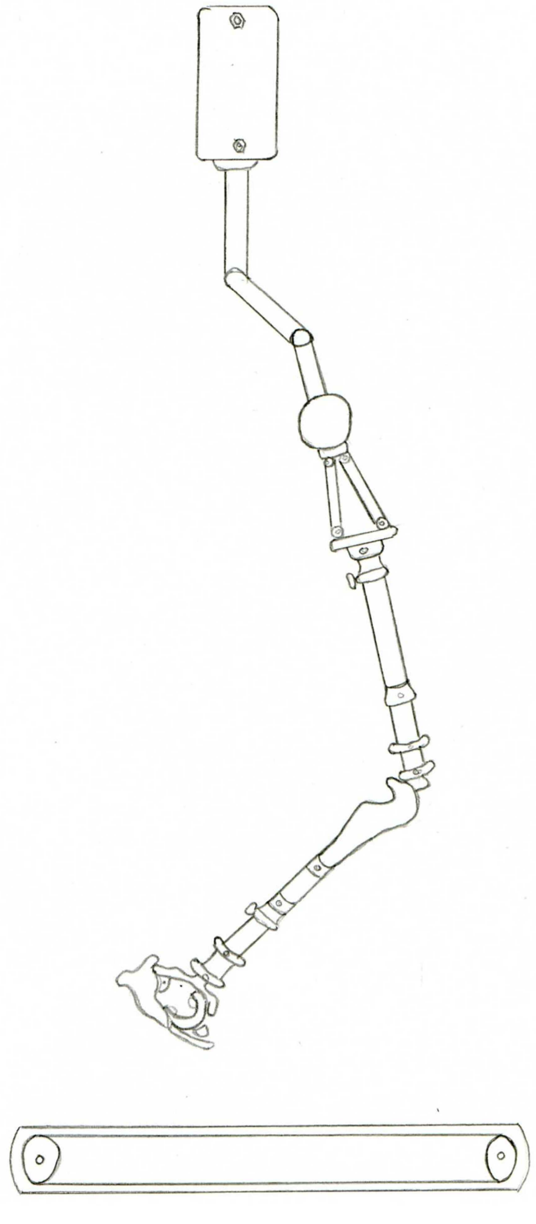


Figure 5.32

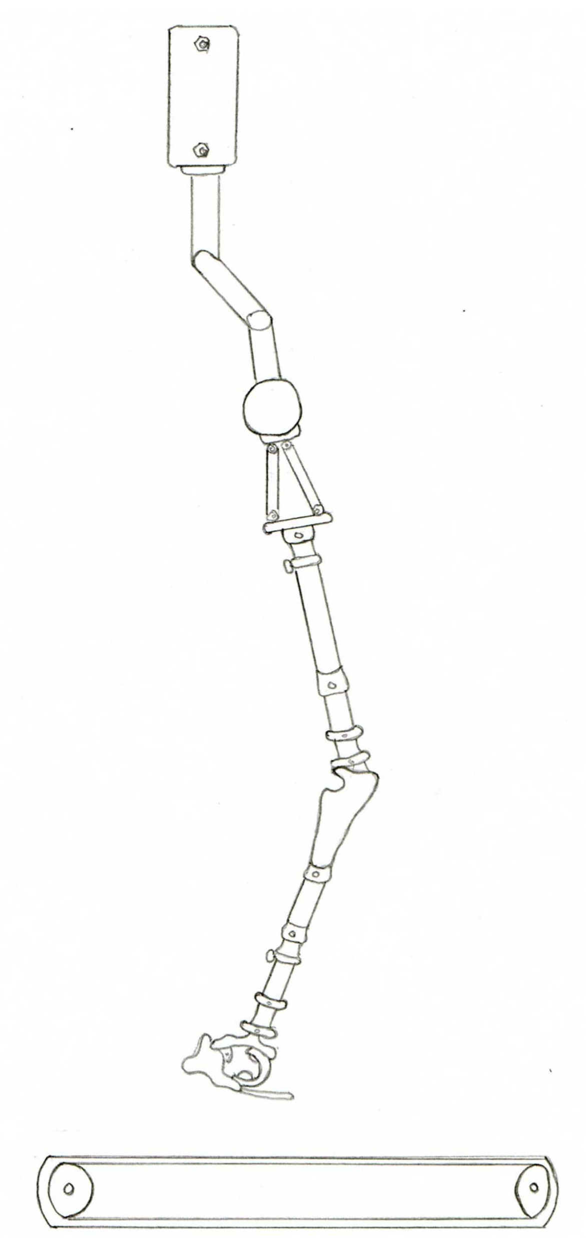


Figure 5.33

5.5 KNEE JOINT TESTING

The major idea behind designing a Bench tester device is to simulate conditions which are similar to an amputee walking with a MPC knee unit. By providing the right kind of environment for the MPC knee unit we can expect to get the right results.

In the previous chapter some testing parameters were proposed and their importance highlighted. This section concerns itself with the methodology of testing the proposed parameters.

5.5.1 FLEXION IN STANCE

The knee flexion can very well be observed by just closely observing the movement of the prosthesis. Any deviation from the normal shall be instantly visible even to the naked eye. But for a thorough analysis of the knee flexion in stance one could make use of an electronic goniometer. The goniometer can be used to plot the Knee angle Versus Time.

Another very important function of a Bench tester in this aspect is the change in flexion angle with the variation of speed. By varying the amount of fluid entering the valves of the Hydraulic actuator, the working speed of the machine can be varied. This shall be similar to normal walking situations, wherein the amputee keeps changing the speed of walking. A MPC knee unit however should be adaptable to that. With a faster walking speed the flexion angle will be less and vice versa for slower walking speed.

The motion of the MPC knee unit is basically a consequence of the movement at the hip region. With the right hip flexion, it is expected that the knee unit shall behave in the right manner. Hence, it has been taken care in the plan to provide simulations of the normal hip movement.

5.5.2 KNEE LOCKING AND UNLOCKING

Another very important feature of a MPC knee unit is the locking and the unlocking of the knee unit at all periods of the gait cycle. This feature however, will be very easy to identify and can be easily seen during the operation of the Bench Tester. Any deviation from the normal pattern of locking and unlocking can be comfortably caught. But, for evaluating the locking and unlocking of the knee unit the normal body weight must be supplied at the knee. In this case it is provided by the two Hydraulic actuators. Under this load the knee should be working efficiently.

5.5.3 AP KNEE MOMENTS

For noting the AP moments the tester makes use of two transducers placed in regions above and below the knee unit. From previous studies the AP moment's graph of a normal subject walking have been obtained. The moments produced by the MPC knee unit should be comparable.

The transducers also provide for a data of the forces experienced by the prosthesis. This helps in keeping a check on the loading of the prosthesis. Depending on the data being updated by the transducers the load can be reduced or increased.

5.5.4 KNEE STUMBLING PREVENTION

One of the major characteristics features of the MPC knee units is their ability to provide for good safety to the user. Stumbling is very big fear with the amputees. It is a normal reaction of a normal person is to bring his knee joint into extension during a sudden stumble. This phenomenon helps in keeping the body stable at that fixed position and preventing the fall. With a trans femoral amputee though, the absence of the knee joint makes it very difficult to lock the knee suddenly in extension. With MPC knee units this is possible. A MPC knee unit on having a stumble suddenly locks itself in the position, to prevent the amputee from falling.

To check for this ability a Bench tester can very well be used. Within the plan it is explained that the motion of the prosthesis comes from the actuators at the proximal portion. But, the belt platform also assists in the movement of the prosthesis. It is the same process of a person walking on a treadmill. To simulate stumble one simply needs to stop the belt first and subsequently the machine. The sudden stopping of the belt will be taken by the microprocessor as a stumble and it will lock itself into position. The machine needs to be simultaneously shut off, to prevent any damage that can happen to the prosthesis by the dragging of the foot on the stationary belt.

5.5.5 RAMP ASCENDING/DESCENDING

To check for the knee abilities for going up and down ramps provisions have been made within the design.

In ascending it is required for the prosthetic hip joint to be moved from a flexion angle of 25° to 35°. The belt platform will need to be lifted by the same difference I.e. 10°. For this purpose only one actuator will need to be taking the platform up. While the other one will only be providing for support. Similarly, in descending the hip joint will be needed to be kept in 15° and the platform raised from the posterior end to give the simulation of going down a ramp.

5.5.6 CADENCE RESPONSE

The cadence can be calculated with the help of the electronic goniometer. An electronic goniometer apart from giving information about the flexion angle also tells about the angular velocity and acceleration. With the help of these values, the speed of the prosthesis can be calculated and eventually the cadence.

5.5.7 TIMING OF ALL KEY MOVEMENTS

The timing of the MPC knee joint is a very important factor. It can be easily known by reviewing the graphs of flexion obtained by the knee unit during the gait cycle.

CHAPTER 6. FUTURE WORK AND CONCLUSION

6.1 FUTURE WORK

The present project has dealt with the plan & design of a bench tester. There is still a lot of work needed before the machine can be actually constructed.

6.1.1 SYSTEM CONTROLLER

The design relies to a great extent on timing. Improper timing can be fatal. Jamming of the prosthesis is a major. Hence, controller circuit is to be perfected before manufacturing. Once, the circuit has been worked it will need to be tried and tested with a real machine.

6.1.1.1 PRINCIPLE

The controller will initiate the operation by bringin belt platform up to contact heel of prosthesis. The working cycle:

6.1.1.1.1 Stage I

- Controller starts motor of load producing hydraulic actuators. Then opens valve for pushing piston
- As the hydraulics start, the belt also moves.
- The machine can now run without controller circuit to the end of the stage.

- Simultaneously a sensor gathers data from strain gauges for the computer.

6.1.1.1.2 Stage II

- As load producing actuators cover the length of the piston, controller circuit changes valve position to direct move of piston up. To be correctly timed, for a proper gait cycle.
- As motion of the piston changes, prosthesis moves anti-clockwise to 16° extension. Electronic goniometer is placed at the level of the shaft
- Stance phase gets completed, prosthesis moves into swing. Hydraulic actuators at the base are brought down to bring base assembly down to give enough space to the prosthetic foot to clear the belt.
- Swing part of gait complete, shaft comes to original position and controller reverses the motion of the load producing actuators and brings up the belt platform. Now the machine will make the prosthesis go into another cycle.

The circuit is to be customized for gathering the data about the MPC knee performance

6.1.2 MATERIAL CONSIDERATIONS

Some parts to withstand extreme load; others to be light enough. Different segments of a machine have their own criteria for material selection. Essential considerations:

- The framework has to take the load without causing vibrations
- The load producing actuators should have the required capacity
- The crankshaft has to be steady and take load of required rotation
- The belt of the platform should be sturdy enough to give the ground reaction
- Vertical springs in the base to be strong to take load from the weight of the platform + proximal portion of the machine; also flexible to be compressed in the stance phase, to counter lengthening /shortening of the prosthesis
- The base hydraulic actuators should have enough power to take the platform up and bear the subsequent load.

The weight of the prosthesis itself is negligible. It is the vertical loading that is of major concern, weight bearing ability of the different components needs to be taken into account.

6.1.3 CRANKSHAFT DESIGN

The crankshaft is to take the load and convert it into rotation. For that it will need to be sturdy enough, as well as light enough to be rotated by the load.

6.1.4 LOADING ACTUATOR SHEAR FORCE (AP) CONSIDERATIONS

During the vertical loading by the proximal actuators a force component in the horizontal direction is also created. It is significant enough to cause damage to the cylinders / cause their jamming. This will subsequently affect the functioning of the machine.

Compensation is given within the cylinders to accommodate the shear force component by adding 2 sets bearings on either side of the piston

6.1.5 FORCE ANALYSIS

Force and stress analysis for suitability for the machine. An analysis of the force on the knee unit plus segmental loading.

6.1.6 VIBRATIONAL ANALYSIS

For framework structure the resonant frequency of framework has to be more than the working frequency of the machine -- also base of the machine.

6.1.7 SYSTEM COOLING

Prosthetic knee joint has be tested for a number of hours, machines tend to get heated up, presence of a cooling mechanism is vital, to prevent damage to system.

6.2 CONCLUSION

A bench tester for testing MPC prosthetic knee units is a novel idea. It has been difficult to find literature pertaining to the topic. MPC prosthetic knee units have been in the prosthetic market for quite some time now. It was very surprising to not find studies which were dealing with testing of prosthetic knee units based on their functional capability. Earlier decades saw a lot of fatigue testing of the prosthetic knee units, but hardly any testing specifically taking the human gait into account. Therefore, the idea presented in the project is very fresh and nothing close to similar could be found. The project is aimed not only at giving a concept for the testing, but also in understanding the basic goals of testing a prosthetic knee unit.

MPC prosthetic knee units represent a novel concept in knee joint damping. The aim is to free the patient of worries while participating in day to day life. An anatomical knee joint is hard to duplicate, but the MPC knee units make an attempt to do so. Thus, for their testing it is important to simulate natural conditions as much as possible. The attempt within the project has been to present a plan of a machine which is suitable enough to test the MPC knee units. For providing natural conditions to the knee a lot of factors have been taken into consideration such as flexion at the hip region, leg length shortening etc. By providing the right environment to the knee it is expected that an amputee would achieve a confident gait.

Bench testing is not uncommon to other fields of science and is considered as essential in bringing a product into the market. Same is true for Prosthetics and any device needs to be tested on bench first before it can be put on a patient. The risk to the patient is greatly reduced by following this protocol. Also, a lot of features that cannot be tested on human beings can be examined by a machine, thus making the system further foolproof. This should be able to fulfill the primary aim i.e. amputees' wellbeing and satisfaction.

References

1. Anderson M C, Brown N A T, Bachus K N & McWilliams B A 2009. A cadaver KNEE SIMULATOR to evaluate Rectus femoris transfer. Gait & Posture 30:87-92

2. Berme N, Lawes P, Solomonidis S & Paul J P 1975. A shorter pylon transducer for measurement of prosthetic forces and moments during amputee gait. Engineering in Medicine 4:6-8

3. Crankshaft < en.wikipedia.org > accessed on 17 / 07/ 2011

4. Dept of Health and Social Security 1973. Lower Limb Modular Prosthesis, Her Majesty's' Stationary Office 1973. Bowden M, 1974 Personnel Communication, Dept of Health and Social Security, Scientific and Technical Branch, Roehampton, London *(as quoted by Phillips J, Fatigue testing of Lower Limb Prostheses 1977. Thesis, University of Strathclyde)*

5. Hafner B J, Willingham L L, Buell N C, Allyn K J & Smith D G 2007. Evaluation of function, performance and preference as trans femoral amputee's transition from mechanical to microprocessor control of the prosthetic knee. Archives of Phys Med and Rehabil 88:207-217

6. Herr H & WIlkenfeld A 2003. User adaptive control of a Magnetorheological prosthetic knee. Industrial Robot: An International Journal 30:42-55

7. Human gait < en.wikipedia.org> accessed on 11 / 07 / 2011

8. Inman T V, Ralston J H & Todd F 1981. Human Walking, ed J C Lieberman pp 1-22, Williams & Wilkins, Baltimore/London

9. Jin D, Zhang R, Wang R & Zhang J 2006. Terrain Identification for Prosthetic Knees Based on Electromyographic Signal Features. Tsinghua Science & Tech 11:74-79

10. Johannson J L, Sherill D M, Riley P O, Bonato P & Herr H 2005. A Clinical Comparison of Variable Damping and Mechanically Passive Prosthetic Knee Devices. Am J Phys Med Rehabil 84:563-575

11. Kastner J, Nimmerwall R & Wagner P 1999._“What does the C-Leg® achieve?” – A Gait Analysis Comparison of C-Leg®, 3R45, 3R80 . Medizinisch Orthopadische Technik 119:131-137

12. Mayo Clinic < mayoresearch.mayo.edu > accessed on 25 / 06 / 2011

13. McKenzie D S 1977. Standards for lower limb prostheses at ‘International Society of Prosthetics and orthotics’

14. University of California, Berkeley, Prosthetic devices research project submitted to the National Research Council (Committee on Artificial Limbs) 1947: Above Knee tester, “Fundamental Studies of Human Locomotion and other Information relating to the Design of Artificial Limbs”, 2:12.3-12.5

15. Ossur< www.ossur.com > accessed on 20 / 06 / 2011

16. Otto Bock Orthopadische lndustrie Besitz- und Verwaltungs-Kommanditgesellschaft lndustriestrasse 1993. System for controlling artificial knee joint in an above knee prosthesis. Euro Pat No 0 549 855 A2

17. Otto bock <www.ottobock.com> accessed on 13 / 07 / 2011

18. Parr A 1998. Hydraulics and Pneumatics A technician's and engineer's guide, Butterworth-Heinemann Ltd, Oxford

19. Peizer E & Wright D W 1970. Human locomotion in 'Prosthetic and Orthotic Practice', ed G Murdoch pp 15-35, Edward Arnold (publishers) ltd, London.

20. Philips J 1977, Fatigue Testing of Lower Limb Prostheses, MSc Thesis, University of Strathclyde, Glasgow

21. Segal A D, Orenduff M S, Klute G S, McDowell M L, Pecararo J A, Shofer J, Czerniecki J M 2006. Kinematic and kinetic comparisons of trans femoral amputee gait using C-Leg® and Mauch SNS prosthetic knees. J Rehabil Research Deve 43:857-870

22. Whittle M W 1991. Normal Gait in 'Gait Analysis an Introduction', ed M W Whittle pp 48-88, Butterworth-Heinemann Ltd, Oxford.

23. Zahedi M S, Spence W D, Solomonidis S E & Paul J P 1987. Repeatability of kinetic and kinematic measurements in gait studies of the lower limb amputee. Prosthetics and Orthotics International 11:55-64