The effect a cosmetic cover has on the swing phase of a transfemoral prosthesis during level walking: A pilot study

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Abstract

Cosmetic covering of the prosthesis is one of the stages in getting the patient to be confidently rehabilitated back into society. The cosmetic covers are made from polyurethane (PU) foam, a versatile polymer material. However, PU foam does not come without its problems. It is found that when the amputee walks it causes interference with the prostheses components and hence causes changes to the amputee's gait.

In research studies, cosmetic covers of the lower limb prostheses are often briefly mentioned but not fully explored or even omitted. It is hypothesised that the cosmetic cover does cause a certain effect on the amputee gait especially during the swing phase. However, this conclusion is largely drawn from anecdotal information from prosthetists and amputees and therefore has not been scientifically verified.

In this thesis, this hypothesis was tested by recruiting six transfemoral amputee subjects. Four scenarios were investigated: Bare prosthesis (no cosmetic cover), subject's own cosmetic cover with stockings, new cosmetic cover (no stockings) and new cosmetic cover with stockings. The VICON system was used to motion capture the markers attached to the subject's lower limb on each walk trials of each scenario. The parameters investigated were: knee flexion range, knee angle during one gait cycle, step length, stride length and velocity. Successful walk trials were then loaded into a MATLAB program for further analysis and the results of each scenario were compared within each subject.

The results compared showed differences in the parameters of each scenario when compared with each other. This however, varied from subject to subject and therefore further research is needed to provide conclusive results.

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Contents

List of Figures

List of Tables

List of Abbreviations

LASIS – Left anterior superior iliac spine marker.

RASIS – Right anterior superior iliac spine marker.

LPSIS – Left posterior superior iliac spine marker.

RPSIS – Right posterior superior iliac spine marker.

LLEF – Left lateral epicondyle marker.

LMEF – Left medial epicondyle marker.

LITB – Left tibial tuberosity marker.

RLEF – Right lateral epicondyle marker.

RMEF – Right medial epicondyle marker.

RITB – Right tibial tuberosity marker.

LLMAL – Left lateral malleolus marker.

LMMAL – Left medial malleolus marker.

RLMAL – Right lateral malleolus marker.

RMMAL – Right medial malleolus marker.

LCAL – Left hindfoot marker.

LLMET– Left lateral metatarsal marker placed at head of fifth metatarsal.

LLMET – Left medial metatarsal marker places at head of first metatarsal.

RCAL – Right hindfoot marker.

RLMET – Right lateral metatarsal marker placed at head of fifth metatarsal.

RLMET – Right medial metatarsal marker places at head of first metatarsal.

1. Introduction

Advances in technology have greatly influenced the development and manufacturing of artificial limbs such as the components of transfemoral prosthesis. Thus the materials of prostheses have become more durable and most importantly modelled alike to a limb. Technology such as the use of motion capture has aided in the understanding of gait and as a result, there are prostheses that have been developed to mimic closely a normal gait for amputees.

Technology such as advancements in or the use of microprocessors has been an integral of the development of the prosthesis – constantly upgrading the design of the prosthesis' components each time to be more effective in providing normal gait-like performance to the user. However rehabilitating back into society, has not been a straightforward solution – it involves physical (learning to use the prosthesis) and mental (acceptance and confidence in using the prosthesis) rehabilitation, which are closely associated **(Furst et al., 1983; Murdoch, 1988)**.

Successful rehabilitation involves: a socket that can comfortably house a residual limb, a prosthetic limb that is stable when walking, can withstand weight during stance, give toe clearance during swing and also have an appearance that is acceptable **(Lusardi et al., 2000)**.

Walking is a locomotive bipedal activity to get from one location to another. It is an activity that we undertake without constant thought. On the contrary, such a simplistic movement relies on the joint unison of muscle activity and the nervous system to move the body with the least energy expenditure to do so.

The lower limbs require fine-tuned muscle activity when walking to prevent buckling and falling when walking. However, this intricate interaction of the hip and knee flexors and extensors muscles becomes disrupted in the case of an amputation of the lower limb. Alternative gait pattern is adopted by the amputee as a result of the lost of knee flexors and extensors muscles and the increase output activities of the hip flexor and extensors muscles to compensate this. Amputee gait is not energy efficient compared to normal gait therefore research for the past decade have been aspiring to designing a knee joint mechanism that can mimic a normal gait as closely as possible. This has lead to designs of knee joint units from being just a hinge joint knee mechanism to that of being controlled by a hydraulic mechanism or by microprocessors (e.g. C-Leg and Rheo).

Research studies have emphasised the importance of functionality of the prosthesis and little on making the acceptable to the user. A popular method of making the prosthesis acceptable to the user is by the use of a cosmetic cover placed over the prosthesis to mirror the look of the contralateral limb. Anecdotal information from Prosthetists **(Solomonidis et al., 2012)** have voiced a cosmetic cover has an effect to some extent on the prosthesis however, often in literature there is little mentioning of cosmetic covers and or the effect they have on prosthesis' performance.

To the author's knowledge, the research studies read and mention throughout this thesis, the journal article authors have not specified whether a cosmetic cover was worn or not and therefore is difficult to deduce the conclusions presented by these studies whether wearing a cosmetic cover or not would have an influence in the results.

1.1 Aims and objectives

This study aims to demonstrate the influence of polyurethane (PU) cosmetic cover has on the gait performance of the prosthesis during level walking.

The purpose of this investigation is to validate the above hypothesis and therefore illustrate the importance of including the cosmetic cover when performing evaluation or gait performance studies of a prosthesis. For example, in studies that simulates the "everyday activities" of an amputee or for studies evaluating the prostheses in everyday use. Clinically, this investigation can aid prosthetists who program the microprocessor of the intelligence knee units to factor in if the amputee would be wearing a cosmetic cover or not to adjust the microprocessor programming accordingly to provide the amputee with a smooth and energy sufficient gait.

1.2 Layout of thesis

Chapter 1.3 presents a review of the history of amputation from ancient to recent times to show the progress medical and technology has input to the prosthetic industry.

Chapter 2 presents a review of transfemoral amputation and prosthesis, introducing the level, cause of amputation and the components available and used by transfemoral amputee. This chapter includes a section on cosmetic covers – the purpose for them and how it is manufactured.

Chapter 3 is the methodology procedure used in this study and the results are given in Chapter 4, were the results are compared within each subject as each subject used a different prosthesis and therefore the results cannot be compared between subjects.

Chapter 5 discusses the results presented in chapter 4 and chapter 6 concludes with an overall summary of the findings from chapter 5 and suggested future work that can be done.

1.3 History of amputation and prosthesis

Many of amputations from skeletal remains have been documented from all over the world. Records of amputations and artificial replacements have been documented as far back as around 50BC on figurines or on vases.

During the Middle Ages iron prostheses were made mainly for knights who had lost their limbs in battles, acting as aesthetic replacements. Surgical techniques during this time involved no anaesthesia or control of bleeding and as a result the procedure was done swiftly and usually by the guillotine fashion. Hippocrates was the first to describe the use of ligature techniques, which in the Dark Ages was lost and instead replaced with the crude methods of boiling oil or by the method of crushing the stump, was utilised instead, which unfortunately left the patient with a non-ideal residual limb.

In around 14th – 15th century of wound treatment methods advanced with the widely popular use of the method of controlling bleeding primarily done via cauterising the wound. It was not until Ambroise Paré who developed and re-introduced further the use of ligatures around 1560. Advances such as re-introduction use of tourniquet increased the use of ligature technique during the time when cauterising was used, and when speed which was seen as the upmost important factor, as a majority of all amputation procedures occured during the times of war and when anaesthesia was not invented.

In 1679, James Young of Plymouth was credited as the being the first in introducing the true flap amputations – technique to cover the protruding bone after amputation, which by 1837, Robert Liston developed this idea further, introducing a flap amputation that is still performed nowadays.

In 1718, J.L. Petit designed the most effective tourniquet were it was tied onto the abdomen to avoid slipping and pressured onto the major artery of the limb that needed to be amputated, which this provided good haemostasis **(Sellegren, 1982)**.

With the increase of innovative surgical procedures came the development of sophisticated surgical instruments and most importantly, the invention and development use of anaesthesia in the late 18th century, which meant that procedures could be performed painless for the patient and speed was not a problematic issue **(Kirkup, 2010)**. New techniques of amputations were introduced as a result of the American Civil War, were ether or anaesthesia was used. Apart from the improvements in surgical methods, with the use of anaesthesia and accurate haemostasis it contributed to improving primary healing and as a result also the improvement of the residual limb, leading to prostheses that were not only designed solely for aesthetic reasons but could be actually functionally walking prostheses.

It has been documented that the first above-knee prosthesis with an articulating knee joint was designed by Paré himself (figure 1).

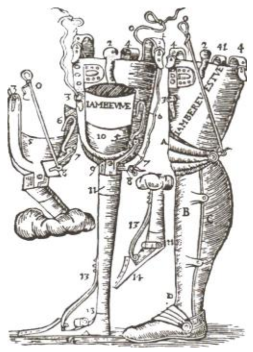

Figure 1: Above-knee prosthesis designed by Paré **(van de Veen, 2000)**.

It was not till around the 1800s were James Potts, designed an above-knee prosthesis that was lighter (figure 2), as the components were made of wood than Paré’s prosthesis which was made primarily of metal. It was then introduced to America in 1839 modification were made – a rubber plate was added to the ankle and a rubber sole added – this became widely known as the “American Leg”.

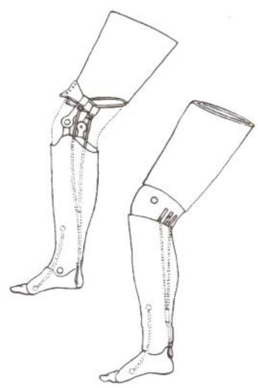

Figure 2: Above-knee prosthesis designed by Potts **(van de Veen, 2000)**.

In 1857, Gritt introduced a surgical procedure, which provided an end weight bearing above-knee residual limb, which in 1870 was slightly altered by Stokes and is known as the Gritt-Stokes procedure. With the, outbreak of the First and Second World Wars, brought around the refinement of amputation procedures that resulted in the end result of the patient having an weight end-bearing residual limb. From 1950s, and onwards, developments into prostheses and amputation have been rapid. Amputations have been emphasised to be more weight end-bearing. Prosthesis with the aid of new material development, have designs to have prostheses being made of lightweight material yet still be functional for the user **(Sellegren, 1982)**.

In recent times, over the last decade of the twentieth century has seen no major leap in the area of amputation surgery. Improvements in the area of diabetic area, vascular surgery, protective footwear and the invention and use of antibiotics have allowed gradual improvements into this area **(Cochrane et al., 2001)**. Emphasis have been primarily made along the idea of making the prosthesis to have the functionality of simulating the same performance as normal gait as well as keeping the material used to be robust yet still maintain being lightweight **(Sellegren, 1982)**.

2. Literature review

2.1 Transfemoral amputation and prosthesis

2.1.1 Level of amputation

Amputation levels of the lower limb depend on how far the damage is on the limb. Figure 3 illustrates the typical levels of lower limb amputation that can be performed.

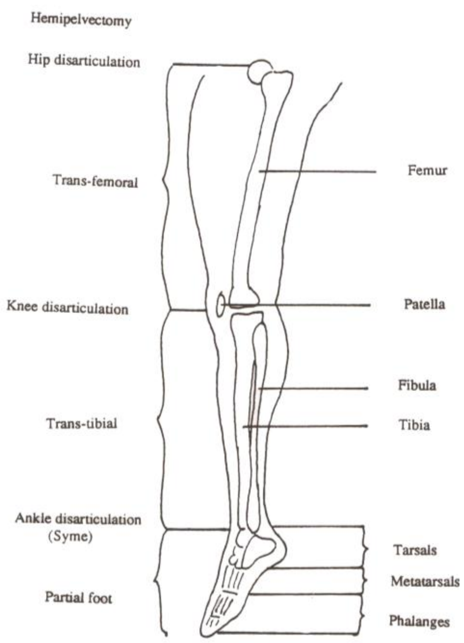

Figure 3: Level of lower limb amputations that can be performed **(Stewart, 1995).**

Symes's amputation involves partial amputation of the foot leaving the heel intact. This type of amputation has the advantage of preserving the heel for weight bearing purposes; however it has the drawback of being not aesthetic. Hemipelvectomy is rarely performed – approximately around 6 per year compared to transtibial and transfemoral amputations which are more commonly performed. For transfemoral amputation, the shape of the residual limb can be long, medium or short depending on the viable factors of the muscles as well as the amount of muscle coverage remaining for cushioning the bony end of the residual limb **(Cochrane et al., 2001)**.

2.1.2 Causes of amputation

Amputation is an irreversible procedure that can present physical and psychosocial obstacles **(Gallagher et al., 2000)**, hence, is a last resort method if there is no viable blood circulation at the extremities as the limb cannot be salvaged. The cause of amputation for each case differs – from as a result of a traumatic incident (i.e. car accident) to non- traumatic reasons such tumours.

On the whole, Peripheral Vascular Disease (PVD) is the major cause of all amputations, and approximately 80% of PVD patients are also diabetic patients **(Global Lower Extremity Amputation Study Group, 2000)**. Other causes of amputation, such as trauma, tumours and infection only constitute a small percentage compared to PVD caused amputations due to improvements in medical procedures and stricter guidelines in health and hygiene in treating diseases that would have otherwise resort in amputation. Diabetes remains a major factor in the cause of amputation, were diabetic patients are nine times more likely to have a major amputation compared to non-diabetic patients.

Study by Moxey et al., reported from the five years that was investigated that the major amputation rate in England was 5.1 per 100,000 population with minor amputation rate was 6.3 per 100,000 population **(Moxey et al., 2010)** were the ratio of below and above knee amputation (BK: AK) is approximately 2.5:1 **(Unwin, 2000)**. Approximately 80% of lower limb amputees are over 60 years of age **(Moxey et al., 2010; Geertzen et al., 2001)** and the majority of amputees in years 2001/02 were males **(Pitkin, 2009)**. Factors that can also influence the incidence of amputation is dependent on the country, such as a high incidence in Northern America and Northern Europe compared to low incidence in Spain, Taiwan and Japan and therefore proper medical care can be given to the patient **(Unwin, 2000).**

2.2 Components of a transfemoral prosthesis

Prostheses have gradually improved in design, material and performance; they have evolved from being functional but not aesthetic to having a gait-like performance. As a result, advancement in material manufacturing technology, prostheses are made widely from various materials such as alloys of aluminium or titanium, thermoplastics or even carbon fibre which have made prostheses far more durable and most importantly lightweight without highly compromising their performance (unlike their predecessors).

To fit a prosthesis securely onto a residual limb, requires an assortment of components to ensure that residual limb fits perfectly in the prosthesis whilst the user is moving. These components are described in turn in sections 2.2.1 to 2.2.4.

2.2.1 Socket

A socket is used as an interface between the residual limb and the prosthesis. It serves the important function of ensuring the residual limb is properly and securely locked or fitted to the prosthesis. Socket designs are unique for each amputee to provide an intimate fit so the amputee **(Lusardi et al., 2000)** can comfortably load a proportion of their bodyweight onto it without becoming uncomfortable over time. They are usually manufactured from thermoplastics, thermosetting resin or can be produced from flexible thermoplastics that have the advantage of accommodating any changes in residual limb volumes as well as during muscle contraction. **(Lusardi et al., 2000)**

Sockets that have been successful through the decades since they have been implemented are either the ischial containment or the quadrilateral (quad) socket. Both sockets have both positive and negative attributes associated with them and therefore, which is used depends on the daily activities of the amputee as well as other issues that need to be factored in.

The quadrilateral socket (figure 4) is named as such as it is narrow anterio-posteriorly and wide medio-laterally. It was introduced during the 1950s, and was the primarily socket choice till the 1980s. The ischial containment socket (figure 5) instead has the dimensions that are opposite of the quadrilateral socket – begin wider anterioposteriorly and narrow mediolaterally , and was introduced later on in the 1980s **(Lusardi et al.,2000; Schuch et al., 1999)** as a design that could stabilise the residual limb and prevent the rotation of the residual limb **(Schuch et al., 1999)**.

The quadilateral has a narrow antero-posterior dimension as the ischium tuberosity and ramus – where the primarily weight loading and where the primary-weight bearing surface of the gluteal and ischial muscles sits on the posterior wall of the socket. The anterior is shaped to produce a pressure on the Scarpa's triangle to prevent ischium movement **(Lusardi et al., 2000)**. The lateral wall is contoured for the femur to be in an adduction position as well as to help stabilise the pelvis. The medial wall is also contoured to act as a counterpressure to the pressure on the lateral wall. The contoured walls on the quadrilateral socket aid the functional muscles to be at the optimum position to provide maximum power during the gait cycle when the muscle is activated. Stability forces are mainly applied on the anterio-posteriorly dimensions of the quad socket to prevent the femur from laterally shifting. Stabilising pressures are mostly applied onto the bone instead of being transferred onto the muscles **(Bowker et al., 2002)**.

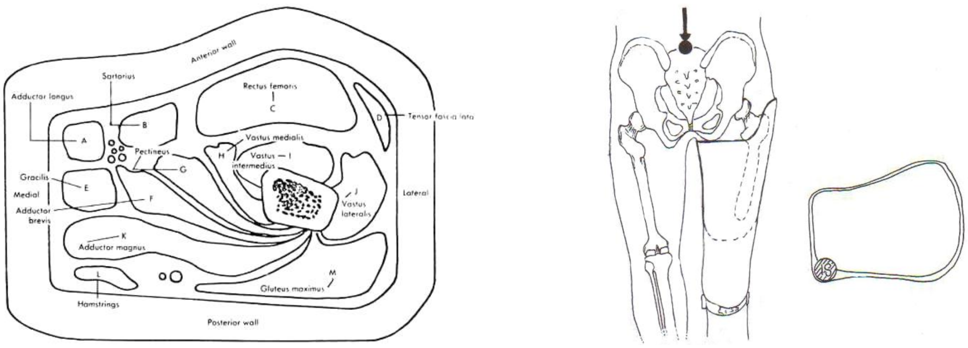

Figure 4: Transverse section of the quad socket and upper thigh (left) **(Bowker et al.,2002)**, posterior view when worn and dimension of the quad socket, IS = Ischium (right) **(Lusardi et al., 2000)**.

The Ischial Containment Socket differs by having the the anterio-posterior dimension wider, provides more room for the muscles to optimally function. The ischium tuberosity and ramus are instead contained in the ischial containment socket which creates a “bony lock”. The socket is contoured to prevent residual limb rotation and for initial femoral adduction Forces are instead distributed down the femur **(Lusardi et al., 2000)** and the weight-bearing occurs on the medial side of the ischium and ramus were countersupport for this occurs along the medio-posterior wall **(Bowker et al., 2002)**.

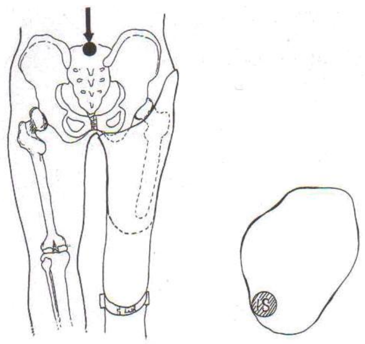

Figure 5: Posterior view when worn and the dimensions of the ischial containment socket. IS = Ischial**. (Lusardi et al., 2000).**

2.2.2 Suspension systems

Suspension systems are used to hold the socket to the residual limb. There are various types of suspension systems used such as the Silesian belt that is strapped around the waist and is flexible. It has the added advantage of being simple to use and therefore can be prescribed to geriatric amputees. Pelvic bands (figure 6 right) are made from leather **(Lusardi et al., 2000)** also have similar qualities and advantages as the Silesian belt except is strapped around the pelvis instead. Silesian belt suspensions are made from either leather or a lightweight webbing material. It is attached on the lateral side of the socket then around the waist were it is the fastened by a buckle or hoop on the anterior side of the socket. They do not prevent rotation of the residual limb in the socket and therefore are usually used as a backup suspension support **(Bowker et al., 2002, Lusardi et al., 2000)**.

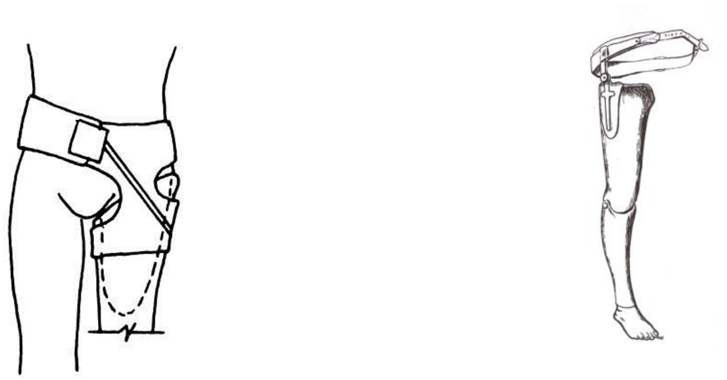

Figure 6: TES belt (left) **(Bowker et al., 2002).** Pelvic belt (right) **(Lusardi et al., 2000)**.

TES (Total Elastic Suspension) (figure 6 left), is also in the same group as the Silesian belts and Pelvic bands in that it is strapped and fastened with Velcro around the waist as well as around approximately eight inches of the prosthesis **(Bowker et al., 2002, Lusardi et al., 2000)**. The TES has the added advantage of being easily donned, comfortable **(Bowker et al., 2002)**, provides control, such as reducing the possibility of the prosthesis rotating on the socket and can be. The disadvantage of these types of suspension systems is the hygienic aspect of using these suspension systems if not cleaned regularly. For example, for the TES, it is usually manufactured using elastic neo-prene and lined with nylon material, and therefore can retain heat and is not durable over time **(Bowker et al., 2002, Lusardi et al., 2000)**.

Suction liners are more popular as they provide a more intimate fit (i.e. ICEROSS) (figure 7). There are different types of suction liners, for example suction socket with valve or locking mechanism. Suction sockets are usually favoured as they provide a better range of motion **(Kapp, 1999)**, better skin to socket contact, prosthetic control and as well as better proprioception for the amputee **(Kapp, 1999; Lusardi et al., 2000)**. Suction liners are mainly produced from: silicone, urethane or elastomer. The liner has the advantage of reducing shear and friction forces on the residual limb however they are not durable **(Lusardi et al., 2000)**.

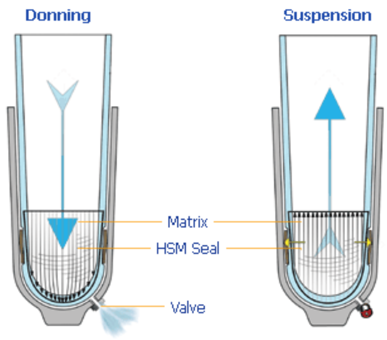

Figure 7: Application of a suction socket. **(Össur, 2012a)**

2.2.3 Knee mechanisms

The type of knee mechanism prescribed again, depends on the amputee as well as other factors such as their daily activities. It primarily has two functions: to simulate a normal gait – can flex and extend smoothly and provide stability in stance and during swing when appropriate and to be able to maintain stability when the body weight forwardly progresses during stance **(Lusardi et al., 2000)**. The rate of shin advancement in swing is dependent on the knee unit (i.e the built-in resistance) **(Lusardi et al., 2000)**. Nolan et al. study have shown that the amount of knee flexion that can be produced depends on the prosthesis used **(Nolan et al., 2000)**.

Single axis knee joint (figure 8 left), which is akin to a single hinge mechanism **(Lusardi et al., 2000)** is a basic knee mechanism such as are prescribed to those that have a loss of physical energy or knee joint movement and therefore are commonly prescribed to certain geriatric amputees as they provide control on either swing or stance phase of gait **(van de Veen, 2000).** They can have additional components such as manual (locking pin) or semi-automatic lock mechanisms that provide further stability for example, automatically locking when the knee is in full extension during stance however this can compromise the ability to have total toe clearance in swing **(Lusardi et al., 2000)**.

Figure 8: Single axis manual lock knee (left) (Lusardi & Nielsen 2000)**.** Four bar linkage polycentric knee depicting the scribed path of the centre of rotation. (right) **(Bowker et al., 2002)**.

Multiaxial knee joints such as polycentric knee mechanisms (figure 8 right) are those that do not have a fixed point of rotation. Instead the upper and lower part of the prosthesis are constantly in contact with one another and hence the centre of rotation is constantly in movement and is not fixed – acting similarly to the anatomic knee joint **(Lusardi et al., 2000, Veen 2000)**. This mechanism provides stability during stance and flexes the knee thereby shortening the shin providing toe clearance during swing. Unfortunately they are not durable and are prescribed to those that have a short residual limb or have weak hip extensors. **(Lusardi et al., 2000).**

Technology plays a prominent factor in the advanced development of knee mechanisms, for example, the popular hydraulic mechanism **(van de Veen, 2000)** (figure 9 left). The hydraulic mechanism was designed during World War II by Henschke Mauch who then adapted the design that was originally for rocket guidance systems for amputated veterans **(Lusardi et al., 2000)** . It utilises the viscosity of the fluids contained inside (usually oil or can be air (figure 9 right)) in the piston. When walking, the flow of fluid or air is ejected into the valves and chambers in the unit that provides frictional resistance which results in increasing the compression speed, which allows it to simulate a normal swing phase. If the speed of walking increases the unit can accommodate this change by increasing prosthetic shin extension. **(Lusardi et al., 2000)** Advantages of the hydraulic system are that: it can emulate the quadriceps activities at early swing and the hamstring activities during late swing (Sjödahl et al., 2002). It can also offer a wide range of cadence speeds **(Lusardi et al., 2000; van de Veen, 2000)** for the user that can be manually adjusted depending on the activity the user is involved with. However, they are heavier, require extra maintenance and in cold weather for fluid hydraulic systems, are late to respond to changes due to temperature affecting the viscosity of the fluid. On the market, hydraulic units can control both phases of gait (known as SNS – Swing and Stance control) or one of the phases. **(Lusardi et al., 2000)**

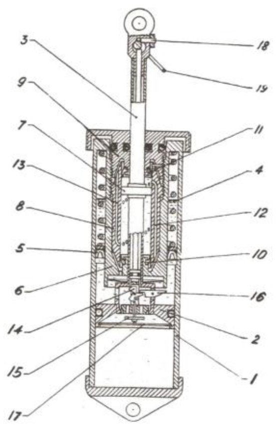

Figure 9: Cross-section of Mauch HYDRAULIK model of a hydraulic knee unit depicting the various components (i.e. piston rod, swing-control piston, fluid channels etc.) (left) (Mauch 1968; van de Veen 2000). Model of Endolite ESK+pspc pneumatic knee unit (right) **(Endolite, 2012a).**

Microprocessor controlled knee joint unit mechanisms (figure 10 left) was first developed in Japan during the early 1980s, that allows the amputee to an imitated normal gait **(Lusardi et al., 2000)** and is also a preferred mechanism, such as the Ottobock C-Leg. They are knee joint mechanism that can control either stance, swing or both phases **(Lee et al., 2009)**. The microprocessor controls the hydraulic mechanism, (which in turn is pre-programmed by the prosthetist beforehand), the stance and swing control and the hydraulic resistance valves **(Lusardi et al., 2000)**. The microprocessor technology functions by evaluating the data collected real-time from sensors on the shin (e.g. force) and strain gauges which then adapts the hydraulic dampers via software algorithms and adjust the knee accordingly to the stance and swing phase of the gait cycle the amputee is in. Studies into these mechanism units have shown to significantly reduce energy expenditure during gait **(Lee et al., 2009)**.

Figure 10: Endolite Orion model of microprocessor controlled knee unit (left) (Endolite 2012b) Model of Össur Rheo knee unit. (right) **(Össur, 2012b).**

The latest technological development of knee mechanism is the magnetorheological (MR), e.g. Ossür's Rheo knee (figure 10 right). The MR fluid is a mixture of oil and anti-coagulant coated iron particles. When an electromagnetic field is applied, it causes the iron particles to align themselves to the magnetic flux lines and therefore causes the viscosity of the fluid to change. The viscosity varies depending on the magnetic field is applied **(Poynor, 2001)** and therefore the knee can be controlled and attenuated depending on what stage of the gait cycle the user is in.

2.2.4 Prosthetic foot

The type of prosthetic foot can also have an effect on the user's gait such as from Lee et al. study found the amount of knee flexion and extension a prosthesis can achieve during early stance depended on the prosthetic foot selected **(Lee et al., 2009; Wang et al., 2011)** hence the ground reaction forces can influence the knee mechanism. Ideally the prosthetic foot should function as what the anatomical foot does. The anatomical foot functions as a "rocker" were the ground reaction forces are gradually shifted forward from the heel (after heel strike) to the front of the foot for toe off. One of the functions of the prosthetic foot is to emulate this as much as possible. Other functional requirements of the prosthetic foot are: to be able to take the bodyweight load and be stable at stance phase of the gait cycle **(Lusardi et al., 2000)**, shock absorption at heelstrike and compensation when walking on uneven grounds. Prosthetic feet are grouped into either non-articulating or articulating types.

Non-articulating foot types such as the SACH (figure 11 left), Dynamic and Seattle litefoot are those designed to offer very little movement around the ankle and therefore are stable and durable, thus are generally prescribed to older geriatric patients. The Solid Ankle, Cushion Heel (SACH) is a simplistic prosthetic foot design. The heel wedge can be customised to have various densities for various walking grounds.

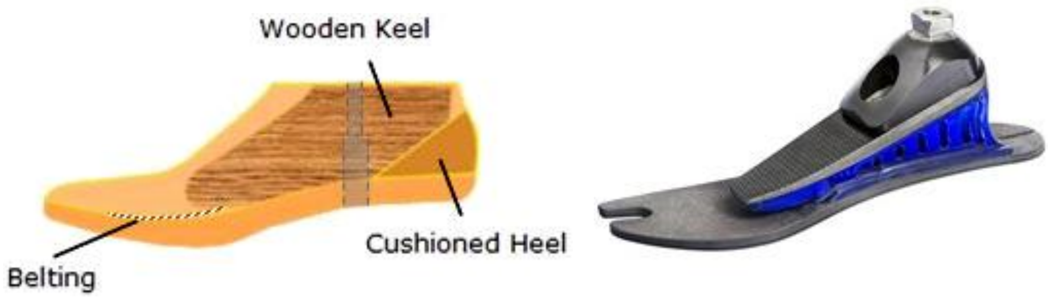

Figure 11: Depicts a sagittal cross section of the SACH foot (left) (Leimkuehler 2012) Össur Flex-Foot® Axia model of a multiaxial foot (right) **(Össur, 2012c).**

Articulating foot types (e.g. multiaxial foot) (figure 11 right), are able to function on one or more than one axis depending on their design. Plantaflexion of articulating feet are controlled by the stiffness of the rubber bumpers in the foot. They have the advantage of shock absorption and increased stability on uneven grounds however they have the disadvantage of being heavy and hence are not ideal for geriatric or less active patients.

Figure 12: Examples of dynamic response feet: Ottobock 1E58 Axtion® DP (left) **(Ottobock, 2012a)** Össur Cheetah® model (right) **(Össur, 2012a).**

Recent designs into prosthetic feet have been geared towards producing more energy efficient or storing. This has led to the design of dynamic response foot, a unique design at the ankle, were the material is bent at the position equivalent to the anatomical ankle. The theory behind dynamic response foot is that as it is bent, it acts as a spring when deflected during initial contact with the ground and at heel off or toe off during swing **Lusardi et al., 2000)**, the energy from this is absorbed in the spring therefore making it energy storing/saving. The above types of dynamic response feet depicted in figure 12 are largely prescribed to active patients such as competing athletes or highly active amputees for sporting events **(Stewart, 1995).**

2.3 Cosmetic covers

A Cosmetic or cosmesis cover is a device that is fitted over the prosthetic leg. The cosmesis, if the user chooses to wear one, serves two important purposes: the first, which is often overlooked is psychological purposes and the second - for protection.

2.3.1 Psychological reasons

For many amputees, having a functional prosthesis is important, but a cosmetic appearance is also equally important. Hence, the golden requirements of a cosmesis are for it to be: durable, lightweight and life-like as possible. The condition of making it like-life requires not only for it to have the same skin colour but also shaped as the other leg **(Bowker et al., 2002)**.

Life-likeness of the cosmesis is very important for an amputee to be rehabilitated into the society; a method of presenting normality is by concealment of visible disability **(Kaiser et al., 1985)**; however this is rarely done to these requirements **(Murdoch, 1988,).** Rehabilitation requires two factors to be fulfilled: Functionality of the body to be restored and restoration of the “body image” that is acceptable for the person and for society.

The relationship between body image of amputees and psychosocial adjustment has been researched **(Furst et al., 1983; Murdoch, 1988; Murray et al., 2002; Smith et al., 2004; Murray, 2005; Murray, 2009)**. These are at least three types of “body image”. First is the “image of oneself”, the image of oneself in the mirror. The main characteristic of this “on-face” image is that it is static. The second is the image that other people have of them and therefore its primary characteristic is that of being dynamic. For example a spectator is more aware of gait deviations such as someone limping. The third image, which is seen as the most crucial of the three, is the inner image a person has of themselves and is normally comprised of characteristics that they would ideally desire and therefore is neither a dynamic nor a static image **(Smith et al., 2004)**.

This third image has one important characteristic in that it is the image of oneself’s body image – a complete, whole self image were the limbs are intact. For a person, to having a blemish, emotional stress to using a prosthesis or any kind of impairment can destroy this image or illusion of able-bodied-ness **(Smith et al., 2004; Murray 2005)**.

Apart from functional rehabilitation for the amputee to learn to use the prosthesis, psychological rehabilitation is also crucial; to rebuild the broken self images so as to rebuild their confidence, reaccepting themselves and also reaccepting themselves in the face of society **(Murdoch, 1988)**. Part of the psychological rehabilitation solution is for the prosthesis to be aesthetically same as the contralateral leg.

Amputation is akin to a diverse disability therefore people react to such an event differently, in most cases the amputees feel victimised by others, as a result inflict onto themselves self-humiliation and or a sense of vulnerability **(Murray, 2005).** The first stage is initially of sadness, which then after a certain amount of time gives way to physical rehabilitation improvement, which in turn boosts confidence, although in some cases temporary setbacks can occur. Hence, there is a relationship between physical and psychological rehabilitation **(Furst et al., 1983)**.

It is highly seen that prosthetics are used for the presentation of oneself to others (also known as 'social interaction') were there is a high public self-consciousness and therefore associated with the amputee's sense of self identity **(Murray, 2005)**. Amputation does not only mean the physical loss of a limb but also a great disturbance to the perception of their body image **(Desmond et al., 2002)**. It has been suggested that when amputees exhibit emotions of self-denial or inability to use the prosthesis it causes a downward spiral of disappointment in the functionality and aesthetic appearance of the prosthesis and hence psychological rehabilitation is needed to remove such negative emotions and reconstitute this traumatic incidence. If not reconceptualised, it can lead to an increase in social dependence and so, a broken self image requires mending for proper integration back into society.

Cosmetic covers are individually tailored to look realistically like a limb (characteristics such as skin colour, or even veins) and therefore these covers are seen for amputees to be of personal great value – A study by Millstein et al., claimed that for a prosthesis to be accepted for use, it had to first have a "*pleasing appearance*" thereby contributing to the amputee's whole self image **(Millstein et al. 1986; Murray, 2009)**.

Anecdotal records have proposed that there is relationship between having a negative self image of oneself and not coping psychologically with limb loss. It has been found in many studies, the idea of self image is varied amongst the two genders. Males lean towards the importance of functionality while females in the studies stress the importance of cosmetic covers as it was seen to be linked with femininity e.g. the ability to continue to wear dresses and skirts, and therefore having a cosmetic cover that look like the contralateral leg is important. Emphasis has been made to stress the important point that a "successfully" fitted prosthesis lies in the psychology of the wearer rather than solely the physical aspect **(Murray et al., 2002)**.

2.3.2 Physical reasons

Apart from providing a realistic mirror image of the contralateral leg, it also has the functional purpose of protecting the prosthesis from external damage, preventing foreign particles and water from interfering with the knee joint mechanism.

2.3.3 Types of foam covers

Above knee cosmetic prosthesis covers are largely made from a material that must comprise of the following characteristics **(Krouskop et al., 1974; Radcliffe, 1974)**:

- Must be made from a material that when stained can be easily cleaned.
- Material must be durable.
- The material has to be compatible with the process of aesthetic pigmentation to match the amputee's skin tone.
- Have no strong odours.
- Be able to maintain its flexible nature after flexing repeatedly.

Foam covers are usually made from thermoplastic polymers such as Polyethylene, Polyproprene for transtibial amputees, Polyurethane (PU) **(Jensen et al., 2000)** and Plastazote® foam made from cross-linking polyethylene that is impregnated with nitrogen gas **(Paulamar, 2007).**

For lower limb prostheses, foam is generally the chosen material for cosmetic covers – most commonly for above-knee, made from polyurethane as it satisfies the characteristics listed above.

2.3.4 How it is made

Cosmetic covers are typically made from foam and these are pre-made from the manufacturer. The first scientific documentation of manufacturing PU foam was in 1947. Over the years, synthesising and therefore the properties have evolved. The mechanical properties of PU foam such as the Young's Modulus is around 3.5 MPa **(Rampf et al., 2011)**.

PU foam is synthesised from reacting isocyanate (a compound with a –N=C=O group) with a polyol FOAM (poly-alcohol group) **(Kaushiva, 1999)** while at the same time the isocyanate also reacts with water from the atmosphere to produce polyuria, heat but most importantly carbon dioxide that causes the liquid mixture to expand giving the material a foam consistency.

A wide range of addition chemicals can be added to modify further the finally properties. Catalysts may be added to speed the whole reaction process up such as tertiary amines. Filler agents can be added to strengthen or add weight to the foam. Surfactants are often utilised as they are able to for example, make foam cells, reduce surface tension, act as an emulsifier, stabilise the foam when rising and hence reduce the effect of deforming. Cross-linking agents such as diethanolamine (DEOA), or that of short-chained molecules that contain an amine or hydroxyl group can also be added to increase stabilisation of the foam. Other additives such as antistatic agents, dye colours or bacteriostats can be added to the foam to have additional qualities.

The steps involved in making PU foam are: bubble initiation, bubble growth bubble packaging and cell opening. Bubble initiation involves the need for gas to be produced, it can be bled into the liquid mixture or produced from the mixture itself, as a by-product or as a side reaction as a result of additional agents added in. Bubble growth is the lengthening stage when gas is diffused into the bubbles and thus increasing the size of the bubbles. Bubble packing is maintaining the perfect shape consistency of the bubble from being distorted or interfering with one another. Lastly, is cell opening, the crucial timing when the bubbles rupture and the structure is thus supported by the "honeycomb-like" morphology **(Rampf et al., 2011)**.

A block of PU foam can then be shaped in a generic limb shape using methods such as inversion cast process **(Krouskop et al., 1974)**. A prosthetist assembles the entire limb from socket to the artificial foot. The foam cover is then attached, enveloping the socket, the knee joint mechanism and pylon **(Lusardi et al., 2000)** as shown in figure 13.

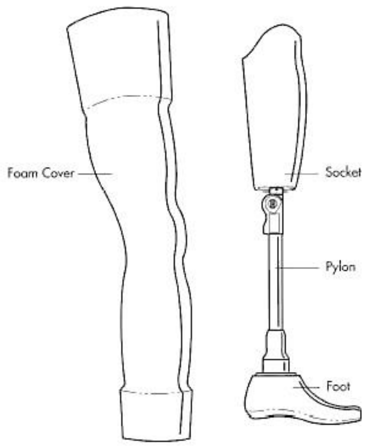

Figure 13: Illustration of how the cosmetic cover attached to prosthetic limb. **(Cengage, 2002).**

The cover can be further aesthetically customised and further reshaping is completed manually by the prosthetic technicians in the limbing fitting centre using the dimensions of the amputee's sound limb. The colour can be further pigmented with a range of skin colours or by using stockings (these are mainly for transtibial prostheses), all which that can be acquired from the manufacturer **(Ottobock, 2012b)**.

2.3.4.1 Advantages

As a versatile material, it has many positive attributes; most importantly its characteristics fulfil the requirements stated in 2.3.3 section.

PU foam is a flexible material and therefore a useful material for covering over prosthetics, the flexibility allows the knee joint mechanisms to operate and the material maintains its integrity (up to a point) during repeated flexion.

As covers are made into foam, it is therefore lightweight which is beneficial, as it does not contribute substantial weight to the amputee when walking.

PU is versatile in addition to its morphological characteristics it can include additional agents to its composition, such as anti-static agents or bacteriostats and can be customised. Therefore it is a material were it is possible add skin-like colour sprayed on it to make a personalised cosmetic cover for the amputee **(Rampf et al., 2011)**.

As it is used widely in all industries it is relatively inexpensive to manufacture and therefore PU cosmetic covers are an acceptable device for NHS provision in the UK **(Cairns, 2012)**.

2.3.4.2 Disadvantages

Despite the advantages of PU provided in section 2.3.4.1, there are numerous problems related to the use of these cosmetic covers. After a certain amount of time – usually around three to six months depending on the amputee’s activities (can be shorter or longer), the cosmetic covers have a tendency to rupture. This is due to repetitive flexion and extension of the limb which fatigues the cosmetic cover and results in material failure. The rupture most commonly occurs, around the anterior area of the knee joint - a prominent location that is visible to the amputee and therefore has a negative impact on the cosmetic appearance of the cover, and also on the amputee’s self image. In addition, friction between the soft foam and the knee joint mechanism causes the interior surface of the cover to break down **(Mauch, 1974; Radcliffe, 1974)** . This causes the additional problem of small foam particles to become wedged in the knee joint mechanism **(Solomonidis et al., 2012)** causing damage to the costly components and thereby requiring regular limb maintenance to clean out the foam debris from the joint mechanism.

The cosmetic covers require to be regularly replaced every 3 months for a transfemoral amputee **(Cairns, 2012)** The cost to the NHS bi-annual cover replacement includes the cost of technical and clinical time necessary to reshape and fit a new foam cosmetic cover to the prosthesis and is estimated to cost approximately £500 per patient per year **(Datta et al., 1999; Nair et al., 2008)**. Whilst the raw material and mass production of manufacturing the cosmetic cover are reasonable for the NHS, the additional costs associated with the cosmetic cover over the lifetime of an amputee are excessive.

Furthermore, it has also been reported anecdotally by prosthetists that the cover affects the function of the knee joint mechanism and thereby affecting the gait of the transfemoral amputee especially during the swing phase on the gait cycle of the prosthesis **(Solomonidis et al., 2012).**

The method of which the cosmetic cover is attached to the thigh and ankle of the prosthesis results in the cosmetic cover, to be unstrained when the knee is in full extension in gait. In the case of when the knee is flexed, the cover is stretched, thereby being under tension at the anterior side of the knee, this influences the amount of knee flexion that can be produced, which can affect the swing phase of gait. On the posterior side, the foam cosmetic cover is compressed during knee flexion and can be wedged in available spaces around the knee unit, which can affect the knee mechanism.

The impact that the cosmetic cover has on the knee joint mechanism functionality has received very little attention in literature; to the author's knowledge only four sources have stated the limitations of cosmetic covers **(Mauch, 1974; Radcliffe, 1974; Solomonidis, 1980; Smith et al., 2004)** of which two have stated the limitations of PU made cosmetic covers **(Solomonidis 1980; Smith et al., 2004).** To date, to the author's knowledge, there have been no studies that have quantified the effect of PU cosmetic cover on the knee mechanism and therefore how it can influence the gait of an amputee. This is a current gap in prosthetic knowledge. It is an area of research that has important clinical considerations; investigating into the impact it has on the knee joint mechanism can provide information to assist clinicians in setting the prosthesis alignment to compensate for when the cosmetic cover is fitted on. The research could also have an influence on how cosmetic covers are designed; evaluating alternative cover materials in comparison to the performance of PU cosmetic covers to identify possible new products that could improve the knee joint mechanism with a fitted cosmetic cover.

2.4 The Gait cycle

Locomotion is the process which we use to move ourselves from one location to another and relies on the combination activities of the physiological systems of the body to produce efficient walking. From research studies, every walk is unique to each person, however from many observations made from observing common observations can be seen in one gait cycle of walking.

Gait is known as the pattern of movements of the limb and torso of the body during locomotion on solid terrain **(Rose et al., 1994)**. Studies have shown that the muscles contribute to the acceleration of the body forward during gait **(Liu et al., 2006)**. Apart from involving the movements of the body, and contraction of the muscles, it also involves the transfer of body weight from one limb to the contralateral limb and is classified as being symmetrical **(Curtze et al., 2011).**

A gait cycle is defined as *"an interval of time which one sequence of regularly recurring succession of events is completed."* In this scenario when walking, the cycle begins with one foot and ends with the same foot **(Rose et al., 1994)**.

2.4.1 Normal gait cycle

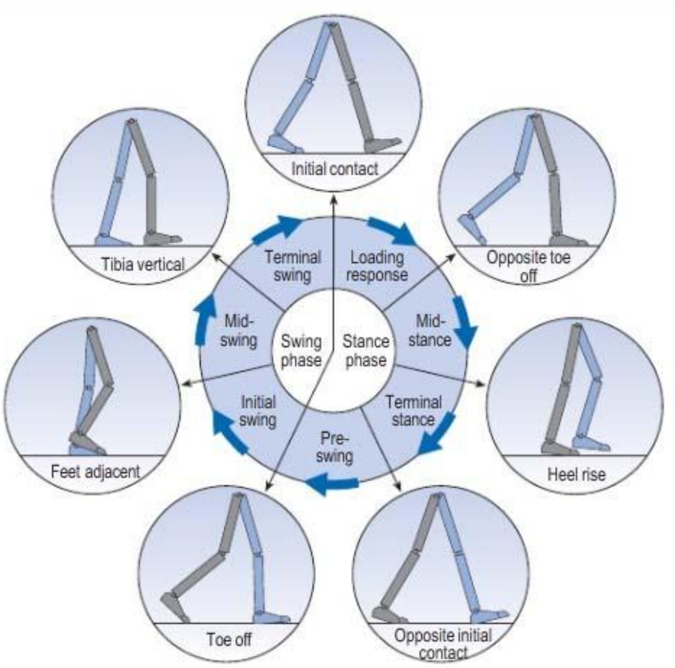

Figure 14: Illustration of the gait cycle. **(Whittle, 2007)**.

The gait cycle, depicted in figure 14 is sectioned into two phases: beginning with the stance phase and finishing with swing phase of the same limb. During gait, the centre of mass undergoes cyclic accelerations and decelerations to propel the body forward **(Liu et al., 2006)**. For a normal gait, it also requires stability from anti-gravity support for the bodyweight during stance, mobility for smooth body segment movement and fine motor control of the body segment when the bodyweight is transferred from one limb to the other **(Rose et al., 1994)**.

The stance phase occupies 60 per cent of the gait cycle **(Rose et al., 1994)** and is when the foot is in contact with the ground. The stance phase itself is sub-divided into separate well defined events.

During stance the hip, knee and ankle perform together to produce an extension moment to prevent the limb from collapsing **(Nolan et al., 2000)**. It begins with heel strike – the initial contact of the heel of the leading foot contacting the ground. During this time, the knee flexes to absorb shock **(Jaegers et al., 1996)** , aid in weight acceptance during the late stage of single limb support when the foot contacts the ground and to counteract the activities of the quadriceps **(Rose et al., 1994; Sjödahl et al., 2002)**. The ground reaction forced during stage is greater than that of the body's weight thereby causing the body to accelerate forward **(Liu et al., 2006)** and centre of mass to accelerate smoothly downwards **(Rose et al., 1994)**. The foot rolls over on the heel **(Curtze et al., 2011)** and load is transferred from the heel forward towards the toes.

Midstance is when the leading foot is rocked forwards primarily by the dorsiflexor muscles **(Anderson et al., 2003)** and hamstrings generate the bulk of acceleration from the heel and into foot flat on the ground by contractions instead from the gluteus maximus and vasti causes the deceleration of the centre of mass **(Anderson et al., 2003; Liu et al., 2006)** whilst the contralateral leg is on toe off and the bodyweight is pivoted on a "rocker" - the ankle, for the leading limb to begin to swing **(Rose et al.,1994).** Research by Anderson et al., have shown that the hip abductor muscles were powerful enough during this stage to support the body when the ground reaction force is lower than the body weight, causing the centre of mass to decelerate **(Anderson et al., 2003)**.

Weight is then shifted from being double limb support of the two limbs at initial contact, known as loading response to being single limb (primarily load is exerted onto the leading limb), thereby letting the contralateral leg free from load to swing forward. The gluteus medius and maximus were found in several studies to contribute significantly during single limb support **(Liu et al., 2006)**. Midstance ends when heel rise occurs on the leading foot, where the terminal stance begins.

In terminal stance, load is transferred anteriorly to the toes and heel-off for pre-swing occurs. At heel-off, it was found in studies that the plantaflexor muscles produced the second largest ground reaction force for accelerating **(Anderson et al., 2003)** the limb for swing. The ground reaction forces exceeds that of the body weight accelerating the centre of mass **(Liu et al., 2006)**. The knee extends, allowing the trunk to smoothly glide forward **(Rose et al.,. 1994)**.

Pre-swing occurs were the knee flexes rapidly to shorten the limb for clearance and the quadriceps muscles contract to counteract this flexion and limit it to be at the maximum of sixty-five degrees **(Sjödahl et al., 2002)**. The bodyweight is then transferred onto the contralateral limb to allow the leading limb to swing **(Rose et al., 1994)**. Additionally, the contralateral foot begins with heelstrike returning back to double limb support in preparation of swing phase **(Whittle, 2007)**.

Swing phase begins with the initial swing of the leading limb, from toe off of the leading limb by the ankle dorisflexors to allow swing **(Rose et al., 1994)** and foot flat of the contralateral limb.

This is then followed by mid-swing were the other limb is on stance phase and the quadriceps of the leading limb stops contracting to give enough clearance to the limb to swing forward, at the same time the hamstring muscles contract to reduce the speed of the swinging limb for ground contact of the foot **(Sjödahl et al., 2002)**. The knee remains in extension and the ankle maintains dorsiflexion contraction to allow foot clearance on the ground.

Lastly, is the terminal swing, the knee remains in extension the leg is decelerated by the hamstrings and gluteus maximus for the swinging limb to be positioned well for a stable stance. This phase is indicated by tibia vertical of the leading limb – this is when the tibial bone is perpendicular to the floor. The quadriceps and dorsiflexor muscles are then activated for stance to occur again **(Rose et al., 1994)**. It is also the moment when the leading limb is about to heel strike to repeat the gait cycle **(Whittle, 2007)**.

2.4.2 Gait cycle of an amputee with prosthetic limb

In bipedal locomotion studies, the most optimal method of walking is by having a symmetrical gait, this yields a smooth gradual deflection of the centre of gravity in a sinusoidal pathway that uses as little energy as possible in producing acceleration and deceleration forces to propel the body forward **(Murray et al., 1980; Schaarschmidt et al., 2012)**. Disruption to normal gait such as disease or trauma causes limitations and introduces the utilisation of compensatory methods in walking that depends on the strength of the residual limb, joint mobility, motor control and sensory capabilities **(Rose et al., 1994)**.

Instead, due to the loss of a limb, there is an increased need of the patient to posturally adapt and balance forces such as the sound limb compensates this functional loss by the means of utilising an alternative gait.

Transfemoral gait differs to normal gait as the muscles used to co-ordinate a smooth gait are absent or weakened and the knee joint mechanism can only mimic an anatomical knee joint to a certain extent that there are notable differences **(Sjödahl et al., 2002)**.

For an amputee energy is conserved in a particular way that the pathway seen is not sinusoidal but irregular, which inadvertently results in higher energy expenditure compared to normal locomotion. It was concluded by a study conducted by Jaegars et al., that the hip stabilising muscles become increasing insufficient with the increasing levels of amputation resulting in an unstabilised pelvis during gait **(Jaegers et al., 1995)**. Tura et al. study established that the amount of asymmetrical gait seen was related to the length of the residual limb **(Tura et al., 2010)**.

The deceleration and acceleration forces differ highly when compared with normal such as there is increased movement from the trunk, pelvis and thigh due to the prosthetic limb **(Jaegers et al., 1996),** therefore requiring high amount of energy from the amputee to counteract or make these forces for motion resulting in asymmetrical limb movement during walking **(Curtze et al., 2011)**.

When hip abductor muscles are utilised, they undergo lengthening contraction during stance phase there is an inability to flex the knee. During stance phase, control of the pelvic descend onto the contralateral limb to the limb in swing phase is needed. If the amputee allows the pelvis to undergo normal descent onto the sound limb it would result in discomfort as the tissues below the ischium and ramus are compressed **(Murray et al., 1980)**. At loading response, to counteract the ground reaction force in absent of knee extensors require the activation of the hip extensors and crucially the prosthetic alignment for stability during when weight is on a single limb and prevent buckling during mid to late stance **(Lusardi & Nielsen 2000)**.

In midstance, transfemoral amputee exhibit lateral leaning, a strategy used for the pelvis to remain level when the sound limb is swinging forward (Lusardi & Nielsen 2000)**.** As a result, amputee would use alternative ways to maintain comfort during walking such as increase stride length to increase lateral stability **(Murray et al., 1980)**.

During swing, the pelvis on the amputated side rotates forward to allow the prosthesis to advance forward **(Lusardi et al., 2000)**.

For transfemoral gait, there is greater plantaflexion angle during toe off on the sound limb to allow the sound limb to be stable to allow clearance in swing on the prosthetic limb and control of the rate of flexion during toe off. Transfemoral amputee also show large hip extensor moments and shock absorption during stance phase **(Nolan et al., 2000)**.

This causes the transition from stance to swing uneven with the stance being elongated on the sound limb and on the prosthesis side, the swing phase is longer **(Murray et al., 1980)**. For an amputee there is a shorter stance period and a longer swing phase on the prosthetic limb compared to when on the sound limb **(Nolan et al., 2000; Sjödahl et al., 2002)**.

Due to amputation, muscles not utilised at a particular stage of gait are activated, such as a study conducted by Winter et al., reported that there was significant activity of the hip extensors during heel strike and mid-stance. This was presumed to be a compensation mechanism of the body as a result of diminished activity of the absent plantaflexors that should be activated during the two stages **(Sjödahl et al., 2002)**.

The extensor knee moment that usually occurs during absorption in stance is increased for the sound limb. During the stance phase the sound limb's main requirement is for anti-buckling during loading. At terminal phase knee flexion is required for pre-swing to occur as well as during swing phase for ground clearance **(Murray et al., 1980)**.

Sjödahl et al. study., found that the duration of the stance phase is prolonged with the decreasing length of the residual limb **(Sjödahl et al., 2002)**.

In normal individuals, walking velocity is fairly uniform however for amputees, during stance is hastened as the torso is inclined forward for the line of gravity to be anterior of the prosthetic knee axis to reduce to possibility of a flexion moment. When the sound limb is on stance phase, it is lengthened to provide time and enough toe clearance for the prosthesis to go into swing phase **(Murray et al., 1980)**.

In amputee gait, the walking speed is longer and is forty-three per cent slower than of normal healthy subjects **(Lusardi et al., 2000)** with abnormal step lengths compared to normal gait. For amputee, gait symmetry is not merely restored with the sound limb adapting to the prosthesis but by functionality compensation.

By reducing functionality of the prosthesis, it results in shorter step time and as a result of a loss of a sound limb there are functional deficits such as missing knee extension and ankle push off during gait.

Compensatory methods by the sound limb are used to adjust with the prosthesis to enable forward movement when walking whilst ensuring that load bearing on the prosthetic limb is keep as minimal as possible.

For an amputee to be able to successfully use a prosthesis successfully requires a length period of training to achieve an optimal and acceptable gait **(Tura et al., 2010)**. The adaptations by the amputee on the sound limb leads to increase forces on the joint that can lead to problems such as joint pain and degeneration **(Nolan et al., 2000)**, and in most cases the gait exhibits asymmetry and deviation.

2.4.3. Types of transfemoral gait deviations

There are approximately fifteen deviations of which eleven are seen very commonly by prosthetists **(Lusardi et al., 2000)**. Observing and correcting (if possible) gait deviations are important to ensure that: the amputee is using the prosthesis correctly, ensuring that the prosthetic alignment is accurate and the correct components are prescribed. Detecting gait deviations is important as they can lead to ailments if not identified such as: back pain, osteoarthritis and the increase risk of falling **(Tura et al., 2010)** and cause increase in energy expenditure for the amputee **(Tura et al., 2012)**.

Lateral bending of the trunk (figure 15) the most common seen deviation during mid to late stance **(Lusardi et al.,2000)**, seen from the rear or front is characterised by the excessive bending of the trunk laterally to the midline or commonly towards the prosthetic side **(Gait Analysis; 1966, Bowker et al., 2002,)**. There are numerous causes for this deviation. For examples: the prosthesis is too short for the amputee, problems with the socket design – the lateral wall is shaped as such that the muscles cannot optimally contract to aid the femur or the medial wall is made too high that the amputee has to lean towards the lateral side to avoid discomfort when walking, the socket is aligned in an abducted position **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002)**causing the hip abductor muscles to be not at optimal position to stabilise the pelvis during gait, amputee has weak hip abductors **(Sjödahl et al., 2002)**, due to a sensitive residual limb or due to amputee originally having a abducted gait **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**

Figure 15: Lateral trunk bending. **(Bowker et al., 2002)**.

Abducted gait (figure 16) is when the amputee walks with a wide walking base during swing phase and due to the foot lying laterally out at initial contact **(Lusardi et al., 2000)** during which the prosthesis is pointed away from the away from the midline with little or no knee moment seen. This could arise due to: the prosthesis being too long, the socket being made in an abducted position, medial brim of the socket made too high causing discomforting resulting in the amputee widening their walking base to relieve pressure **(Radcliffe, 1969; Lusardi et al., 2000; Bowker et al., 2002)**, weak hip abductors or as a result of inadequate lateral brim not supporting the femur during gait **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**.

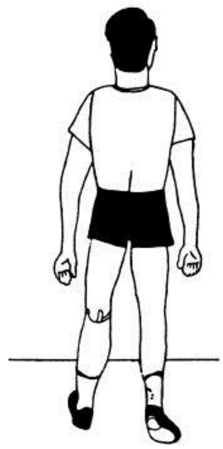

Figure 16: Abducted gait. **(Bowker et al., 2002)**.

Circumduction (figure 17 left) is the swinging of the prosthesis laterally around to clear the ground whilst the knee remains in extension **(Lusardi et al., 2000)** during swing phase. This could be due to: a long prosthesis, too much stability or excess friction resulting in difficulty in flexing the knee joint during swing phase, the socket made too small that the ischial tuberosity is sitting above its correct position or due to a lack in confidence in the use of the prosthesis **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**.

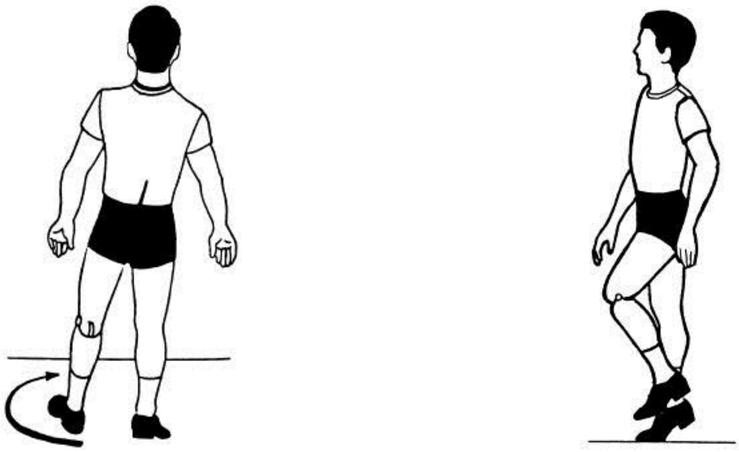

Figure 17: Circumduction (left). Vaulting (right). **(Bowker et al., 2002)**.

Vaulting (figure 17 right) is characterised by the rise onto the toes on the sound limb during swing phase **(Lusardi et al., 2000; Bowker et al., 2002)** in allowing minimal flexion or to allow enough toe clearance **(Sjödahl et al., 2002)** for the prosthesis during swing. This can arise from: having an excessively long prosthesis, lack of an appropriate socket suspension **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**, excessive plantaflexion of the foot or from excessive amount of extension that the knee joint cannot be flexed properly **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002).**

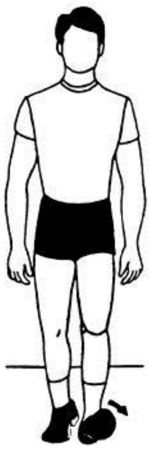

Figure 18: Foot rotation at heelstrike. **(Bowker et al., 2002)**.

Foot rotation (figure 18) at heel strike is characterised by when the foot rotates laterally during heel strike. This deviation can arise due to: the prosthetic foot's plantaflexion bumper carrying excessive resistance, socket is loose, from the amputee extending the stump forcefully during heelstrike or having poor muscle control of the residual limb **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**.

Uneven arm swing – characterised by diminished or missing arm swing when walking **(Lusardi et al.,2000)**. The arm on the prosthesis' side held too close to the trunk (i.e., as if the patient is trying to hold on the prosthesis on) **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002).** The cause for this either due to a painful residual limb **(Lusardi et al.,2000)** or an ill-fitted socket causing discomfort or the patient has yet to develop a good balance on the prosthesis **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**.

Uneven timing is classified as steps taken that are of uneven duration, for example a short stance phase on the prosthesis side – a known transfemoral gait deviation. This can be from: ill-fitted socket lack of extension from the prosthesis, weak residual limb or stemmed from a lack of confidence in prosthetic use **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002).**

Uneven heel rise (figure 19) – when the heel is raised rapidly during when the knee flexes in early swing and the foot rises rapidly from the ground causing an extension delay of the shin (Lusardi & Nielsen 2000). This is caused by an inadequate extension aid from the prosthesis or excessive force from the amputee is used to flex the knee joint **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002)**.

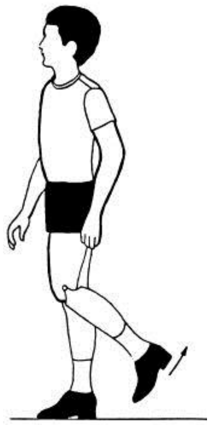

Figure 19: Uneven heel rise. **(Bowker et al., 2002).**

Terminal swing impact (figure 20 left) is characterised by rapidly allowing the knee joint to be in maximum extension during swing and is often accompanied with an audible impact during swing before heel strike. Terminal swing impact can be as a result of: excessive extension aid, insufficient friction, worn out or missing extension bumper or intentionally caused by the amputee causing this deviation to produce an audio cue **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002)**.

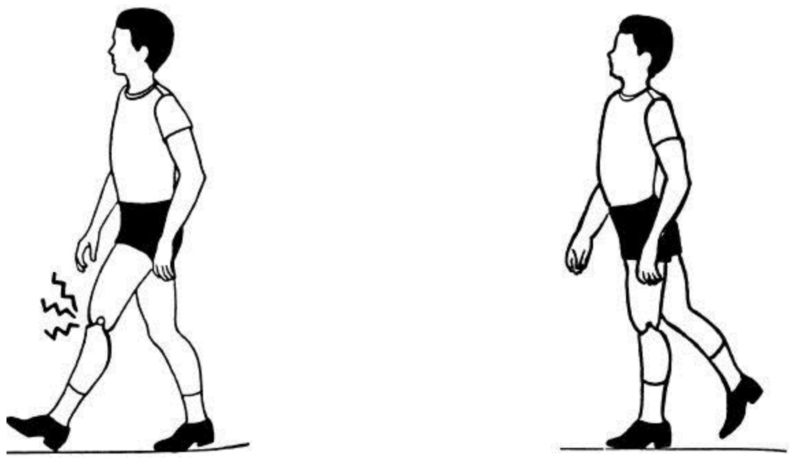

Figure 20: Terminal swing impact (left). Exaggerated lordosis (right). **(Bowker et al., 2002).**

Exaggerated lordosis (figure 20 right) normally occurs during late to early swing **(Lusardi et al., 2000)** and is classified as when the trunk is places more posteriorly during stance of the prosthetic limb. Exaggerated lordosis or excessive trunk extension can occur from: an improper posterior socket brim causing the pelvis to rotate forward from insufficient flexion built in the socket, from the amputee having weak hip extensors causing the pelvis to shift forward and the trunk and to counteract this posteriorly bend the trunk backwards , weak abdominal muscles, constraint of the hip flexor muscles, the TKA (trochanter knee ankle) line – the mechanical knee joint being placed too anteriorly **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002).**

Swing phase lateral or medial whips (figure 21) can be observed: were medial whips is the heel when walking moves around medially and lateral is the vice versa of medial whips during flexion in early swing. Lateral and medial whips occur as a result of: the external rotation (for medial whips) or internal rotation (for lateral whips) of the knee being excessive, tight socket or from an ill-contoured socket causing rotation of the residual limb **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002).**

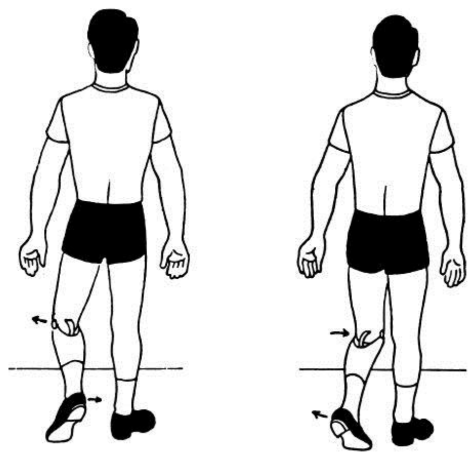

Figure 21: Swing phase lateral (left) and medial (right) whips**. (Bowker et al., 2002).**

Foot slap (figure 22) is categorised by the rapid plantaflexion of the foot to the ground producing a “slap” noise during initial contact to loading response of stance phase. This can arise from: amputee applying excessive force on to the prosthesis to extend as quickly as possible or from a weak resistance plantaflexion bumper in the prosthetic foot **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002)**.

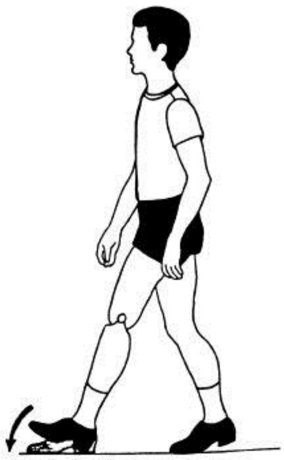

Figure 22: Foot slap. **(Bowker et al., 2002)**.

Drop off at late stance were the trunk moves downwards when the body moves over the prosthesis. This can be due to the prosthetic foot that has an inadequate dorsiflexion resistance or from the socket being placed too anteriorly with respect to the foot **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Lusardi et al., 2000; Bowker et al., 2002)**.

Long prosthetic step is characterised by the prosthetic limb when taking a step, the step duration is longer compared to the sound limb and can be a result of insufficient integrated socket flexion **(Illinois: Prosthetic-Orthotic Education Northwestern University Medical School, 1966; Bowker et al., 2002)**.

2.4.4. Gait analysis

Gait analysis is known as a measurement, description and assessment of the gait cycle – a method that is essential for all gait analysis methods (Sutherland, 2002) in evaluating the walking patterns of a subject **(Davis III et al., 1991)**. Methods of acquiring measurement data has not deviated far since Braune and Fischer in 1895, but instead technology instead have improved the efficiently of collecting data **(Rose et al., 1994).**

Gait analysis involves the tracking systems of: kinematics, kinetics, video recording and clinical observation analysis as well as energy expenditure and the use of electromyographic data. One or more of these techniques can be used to evaluate a person's gait – giving different information of their gait. Data acquired from one or more of these methods are then processed into a format that is descriptive – a format that can be easily interpreted by the clinicians and patients.

Kinematic describes the spatial movements of the body during locomotion and is useful in studying the parameters of: angular displacements, velocities and accelerations. Kinetic on the other hand involves the forces that drive locomotion such as ground reaction forces, joint movements and forces and require the use of devices for instance force plates **(Davis III et al., 1991; Sutherland, 2002)**.

Electromyographic data is data recorded from active muscles contracting by inserting surface electrodes or fine wires into the muscles of interest. Electromyographic data provides information on energy consumption, carbon dioxide or oxygen consumption.

Gait analysis is highly useful in areas such as in sports and for evaluating gaits of neuromuscular disorders, for example cerebral palsy. With the use of gait analysis, it aids in being utilised in preoperative assessment and postoperative evaluation providing the surgeons with information from the patient's gait that would influence operative planning are carried out or if orthotic intervention can be used instead. Hence, gait analysis is one of many methods used to save time that would otherwise be used elsewhere in a patient's treatment. Apart from assessing diseases it can be used as a health diagnostic tool – used to differentiate and classify pathological gait deviations such as those seen in cerebral palsy patient and with idiopathic toe walking. In the prosthetic and orthotic industry it is used primarily to evaluate performance, design and alignment of prostheses and orthoses. In sports, it is used in the area of rehabilitation to assess and monitor injuries as well as to monitor possible methods in improving athletes' performances **(Davis III et al., 1991)**.

Gait analysis has provided a method for clinicians to assess, monitor and document details of normal and abnormal gait more precisely than previously before. Recent studies emphasise on the use of gait analysis in evaluating the functional performances of different types of knee joint mechanisms that have been developed.

There are 17 parameters identified that are frequently used in transfemoral and transtibial research, for example spatio-temporal parameters such as step or stride length. The use of knee flexion angles and moments produced during gait are frequently looked at to determine whether there is significant change in gait as a result of the mechanism used **(Sagawa Jr. et al., 2011)**.

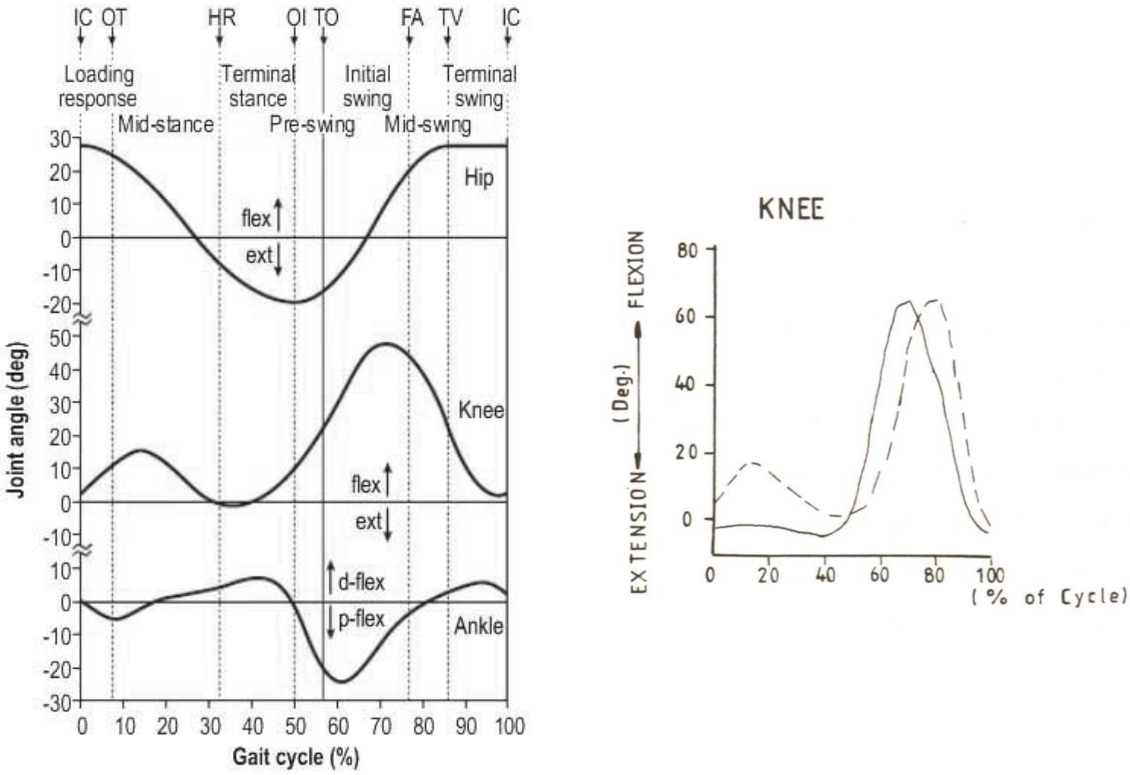

Figure 23: Sagittal plane of joint angles (degrees) during one normal gait cycle. IC = Initial contact. OT = Opposite toe-off. HR = Heel rise; OI = Opposite initial contact; TO = Toe off; FA = Feet adjacent; TV = Tibia vertical. (Left) **(Whittle, 2007)**. Sagittal plane of above-knee amputee joint angle. Line = Prosthetic limb; Dash line = Sound limb. (Right). **(Dabestani, 1984).**

Maximum knee flexion occurs during the swing phase and for an amputee is greater than 60° **(Rose et al., 1994)** (see figure 23 right). It is a parameter frequently looked at as an indication of any present gait deviations **(Global Lower Extremity Amputation Study Group, 2000)** seen or in knee mechanism evaluation studies, as the flexion angle can be influenced by the type of mechanism used **(Nolan et al., 2000)**. An evaluation study of the pneumatic swing-controlled knee mechanisms concluded that the maximum knee flexion angle achieved was in the range between 64 and 68° **(Furse et al., 2011)**. This is a result of the quadricep muscles maintaining contraction whilst counteracting force produced by the hip flexors muscles are lacking or absent to reduce this maximum flexion angle. Other studies have shown, were the C-Leg and Mauch SNS mechanism have shown that this maximum knee flexion angle can be reduced to 55.2 ± 7° almost that of the sound limb (approximately 52°) (figure 23 left).

In a transfemoral amputee, the knee moments range from values that are relatively small to being completely absent **(Sagawa Jr. et al., 2011)** and again, is influenced by the knee mechanism used. For an able-bodied subject, the knee moment produced was found to be 0.47 ± 0.1 Nm/kg compared this to the knee moment values found in studies have shown that the knee unit produces significantly less moment compared to that of an able-bodied subject, C-Leg (0.14 ± 0.05 Nm/kg) **(Sjödahl et al., 2002)** and for a mechanical knee unit (0.06 ± 0.07 Nm/kg) **(Sagawa Jr. et al., 2011)**.

Step and stride length are parameters frequently used to evaluate prosthetic foot performances as the type of foot worn, can affect the gait performance. Step length is the distance between two successive points of one foot and the other. Stride length, is the distance between two consecutive points of the same foot. For a transfemoral amputee, the step length is approximately 0.74m and the stride length to 1.33m **(Sagawa Jr. et al., 2011)**.

2.4.5. Motion analysis systems

Motion analysis systems over the years have developed to carefully capture movement and after processing, produce data that can be easily interpreted. Motion analysis systems such as VICON® are used widely in various industries such as in film, visual effect, gaming, animation, biomechanics and in sports **(Han et al., 2008)**.

Motion analysis systems such as optical motion analysis require various components to be able to record motion. The subject is required to wear skin tight clothing **(Wang et al., 2011)** were on specific bony landmarks of the body are placed retroreflective markers that the specialised calibrated cameras that are placed strategically around the area where the subject would be can be detected. The cameras are fixed in the area and emit a strobe of light which the markers reflect back when insight of the cameras. The ray of light is detected and gives the precise 3-dimensional coordinates of where the marker is. This data is then processed by a programme and generate an image – a computer generated human model **(Vicon, 2012).**

Other motion analysis systems used to collect force data (e.g. ground reaction forces) require the use of force plates (figure 24). These are placed four transducers on each of the corners of the plate, when the subject walks on it, thereby applying force to it, it generates an electrical output that is proportional to the force applied onto it producing ground reaction forces that is directly proportional to the forces produced by the subject **(Wang et al., 2011).**

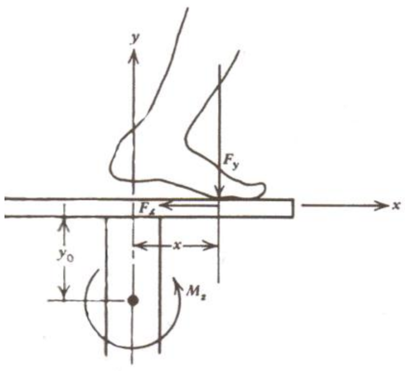

Figure 24: Forces and moments produced when forceplate is stepped upon. **(Winter, 2009)**.

Motion capturing can also be performed by the use of digital video cameras (typically at least two cameras are utilised) that can clearly capture the view of the subject. Filming captures consecutive images at a specific sample rate, however a disadvantage of using such a system is that to obtain more precise data requires increasing the sample rate and as a result would increase the amount of data needed to be processed **(Rose et al., 1994)**. The subject is then recorded and saved for further analysis, such as fine detail or slow motion analysis.

2.4.5.1. What it is used for?

Optical motion analysis systems are used widely from in research into human or animal locomotion, medical applications, producing virtual reality to sports rehabilitation **(Einsmann et al., 2005; Wang et al., 2011).** Since the 1980s, motion analysis systems have given clinicians the opportunity to analyse and provide patients with abnormal gait, a well-tailored treatment plan or orthotic device to use. It can be used postoperatively to verify if treatment have been successful or further robust treatment is required. It is also used in orthopaedic industry as a tool in evaluating the effectiveness of, for example, total knee arthroplasty.

After the 1980s, motion analysis systems have been widely used in the analysis of prosthetics, most frequently in research studies in the areas of evaluating the performance of knee mechanisms, prosthetic limbs, prosthetic components, gait deviations and energy expenditure. They are also used widely in a clinical aspect, aiding prosthetists and clinicians in evaluating prosthetic limb alignment and gait deviation correction before it becomes habitual **(Gage et al., 1995)**.

In sport rehabilitation monitoring of movements is heavily emphasised in the field of medicine and sports. In sports measuring exercise is done qualitatively and measuring such parameters of movement speed, angle of joint rotation are not frequently measured accurately. Therefore, motion analysis systems are used to measure effectively the mentioned parameters as well as used as an evaluation tool in assisting whether innovative methods are effective, improve an athletes' performance and also in injury prevention (e.g. identifying and correcting postures used) **(Mirabella et al., 2011).**

Overall, motion analysis systems provide researchers the ability to evaluate the performances of innovate designs or mechanisms that have become available on the market and to clinicians an opportunity to provide continuous monitoring of their patient's progress **(Tao et al., 2012)**.

2.4.5.2. Advantages of motion analysis systems

Over the decades, motion analysis systems have become a major useful tool in many industries for being able to provide an assortment of information in a matter of minutes. By using such methods, kinematic and kinetic data can be found and combined with other methods such as EMG can provide further information into the mechanics of gait.

Advantages of using motion analysis systems is that copious amount of data can be recorded in real time, saved, processed and the added advantage of the stage of evaluating data can occur on a later date. It also provides the subject of full freedom of movement not constraint by cables **(Wang et al., 2011)** thereby also not constraining what movements can or cannot be used for motion analysis. Different parameters can be captured simultaneously therefore saving time and after processing, the data can be presented in a format that can be easily understandable. High numbers of cameras are used to enable the ability to properly verify and detect errors **(Sutherland, 2002)** whilst the subject is still in the laboratory and therefore experiments can be repeated.

2.4.5.3. Disadvantages of motion analysis systems

Even though motion analysis systems have become a major benefit in various industries, they also have major flaws and drawbacks associated with them **(Gamarnik et al., 2009)**.

The cost of purchasing and maintenance of such a system can be expensive as to collect relevant good data requires a high number of cameras and dedicated computer station **(Sutherland, 2002)** with a high core processing unit and numerous apparatus to link the system to the computer to save data. Housing such a system requires a large laboratory to allow enough area for the subject to freely move around to collect enough data to be processed **(Einsmann et al., 2005)**.

Complex software and hardware components are designed to produce data that is accurate and reliable, which till now remains to be a challenge and progressive research is currently underway to produce a programme to produce accurate results **(Sutherland, 2002)**.

Before collecting data can begin, the cameras and force plates requires extensive amount of time to be used to calibrate beforehand as well as in the post processing stage. Artefacts can be recorded, for example, if the subject is wearing reflective clothing and is not masked properly the camera can misinterpret it as a marker producing incorrect data.

Most importantly, an ongoing problem with optical motion analysis systems that is still under study is the problems associated with the use of the marker system and how to overcome them.

Markers

A problem using markers is that for the cameras to successfully detect them, on the marker's surface is coated with reflective material, if this material is scratched, grease or oil secreted naturally from the hands can diminish the retroreflectiveness of the markers and cause the cameras to fail to detect the markers. Movements from the skin or the underlying soft tissues can cause the marker to be misplaced out of the bony landmark that the marker intended to represent **(Rose & et al., 1994)**.

Another problem associated with markers that are currently being addressed is the problem of occlusion – the deficiency of data captured as a result of hidden markers. Occlusion can occur as a result of loose clothing masking over the marker or because of the markers placed in close proximity to one another especially during movement and as a result the cameras interprets it has one marker thereby producing false data **(Wang et al., 2011)**.

3. Methodology

3.1 Participant recruitment

Ethics approval to conduct the study and recruit the subjects was proposed to the University Ethics Committee and was received on 28th March 2012. Recruitment letter packages (see Appendix A) were sent out to members of the Murray Foundation – a registered amputee charity in Scotland. Only amputees who fulfilled certain criteria were contacted to volunteer in this study: those aged between 18 and 70 with a transfemoral amputation and who lived within reasonable travelling distance from the university. Consequently a total of 11 amputees were contacted.

Six of the eight amputees who replied with interest to participate in this project were selected. Selection was based upon their reported activity level (activity level K3 or above) given when a follow up telephone call with a Prosthetist as the subjects needed to be able to repeatedly walk the length of the gait laboratory for a duration of the test day. The demographic participant data are provided in Appendix B.

3.2 Experimental procedures

3.2.1 Participant preparation

Measurements of the subject's prosthesis side were documented before testing day. The new cosmetic covers were purchased (non-shaped) and then shape by hand by the Prosthetic technicians to the subject's measurements that were previously noted.

At the beginning of each test day, the subject, prior to participating in the experimental walk trials were assessed by a Prosthetist from the National Centre for Prosthetics and Orthotics. The assessment involved general checks and to ensure the prosthesis was in a condition to be able to participate in this study and the new cosmetic cover was shaped to the correct dimensions to fit over the prosthesis.

They were asked to change into shorts and then measurements of the subject's residual limb, height and weight were taken (information needed to be fed into the MATLAB program for data processing in section 3.2.4).

Retroreflective markers (13 markers) were placed on certain bony landmarks of the lower limb of the subject using double-sided tape and five cluster markers were strapped onto the body (the pelvis, right and left lower limb and right and left foot) (see figure 25, 26 and 27). As the prosthetic foot does not have any bony landmarks equivalent to the sound limb ankles, the markers for the L or R MMAL and L or R MMAL, the Prosthetist chose an area that closely resembled the markers on the ankle on the sound limb (see figure 27). For certain lower limb landmarks, marker placements were not practical as they are on places were the markers can easily become detached during walking.

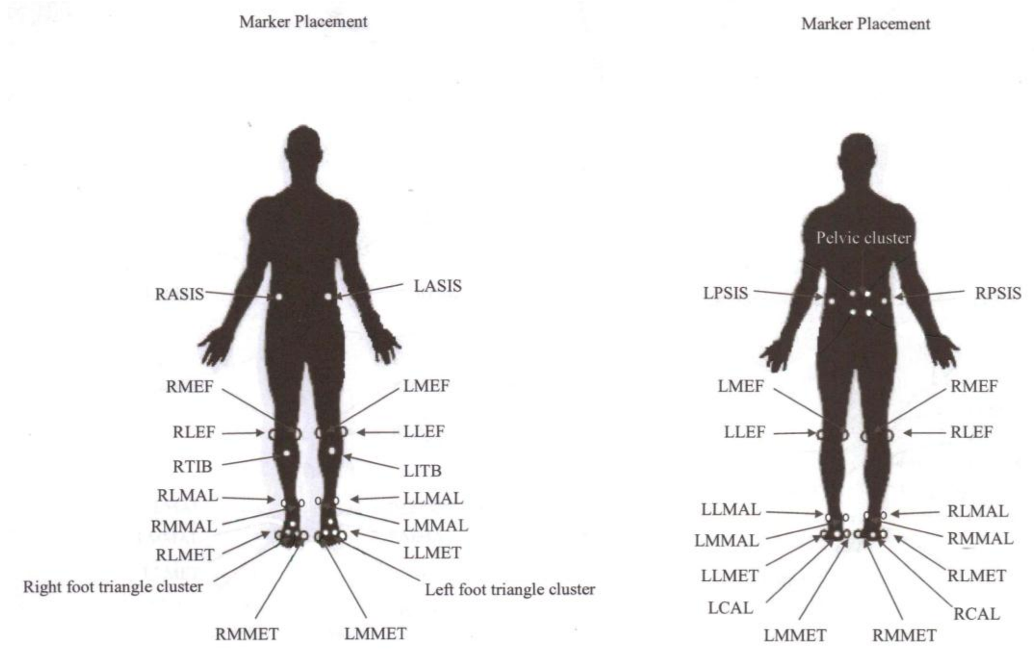

Figure 25: Illustration of marker placement when looking at the anterior side (left) and posterior side (right) of the subject **(Crimin, 2012)**.

Example places are (LASIS, PASIS, LPSIS, RPSIS and certain landmarks on the prosthesis (MEF, LEF and ITB)). These places required the use of the 'wand' to be marked.

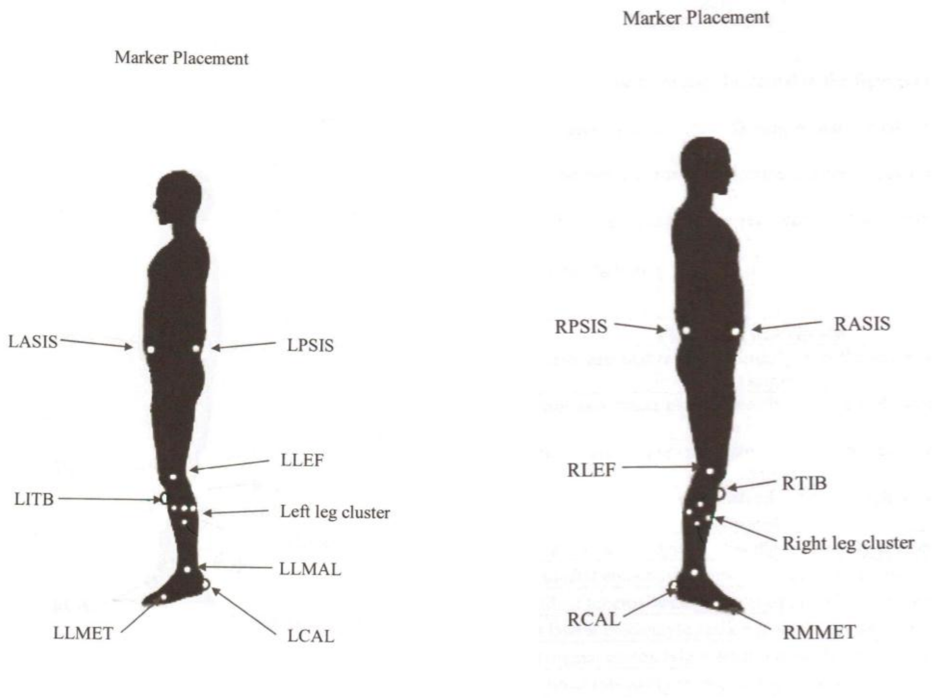

Figure 26: Illustration of marker placement when looking at the left side (left) and right side (right) of the subject **(Crimin, 2012)**.

Figure 27: Photograph of the ankle markers (for the subject prosthetic side is on the right) RLMAL and RMMAL is placed similarly to that on the sound limb side.

3.2.2. Laboratory preparation

Gait analysis was conducted using the VICON system (Vicon Nexus 1.7.3 version, VICON, UK). Before any experimental procedures were carried out, the VICON system required calibration. All existing reflective markers that were being captured by the cameras were masked to prevent them from interacting with the capturing of the markers on the subject during the trial walking.

The cameras were then calibrated to ensure that they all used the same co-ordinate system when a marker was seen in space during the walking trials (see figure 28). The origin of the room was set, this is to correctly orientate the room and the cameras that the VICON system was seeing to ensure the laboratory layout shown by the VICON system is identical to the laboratory the trial walking were being held (see figure 29). The force plates were 'zeroed' before each walk was captured.

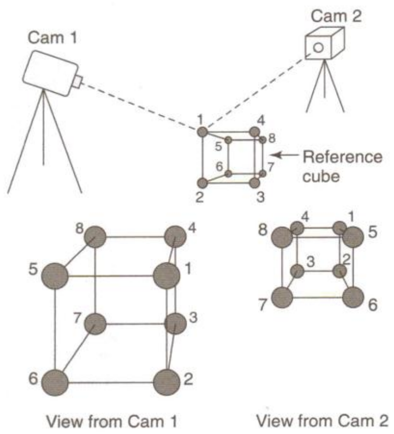

Figure 28: Illustration of the need to calibrate the camera beforehand to confirm all marker references are identical for each camera. **(Mirabella et al., 2011)**

3.2.3 Marker calibration

Once fitted with markers, a static calibration was captured for at least five seconds before each condition trial walk began. This consisted of the subject fitted with the markers to stand still and relaxed within the camera capture (figure 30).

For the markers that require the use of the wand, the Prosthetist identified one landmark with the end of the wand and once all the cluster markers and the two wand markers were clearly visibly on the system, a five second static wand calibration capture of that one marker was taken (see figure 31). This was then repeated for each body landmark that was represented by a marker.

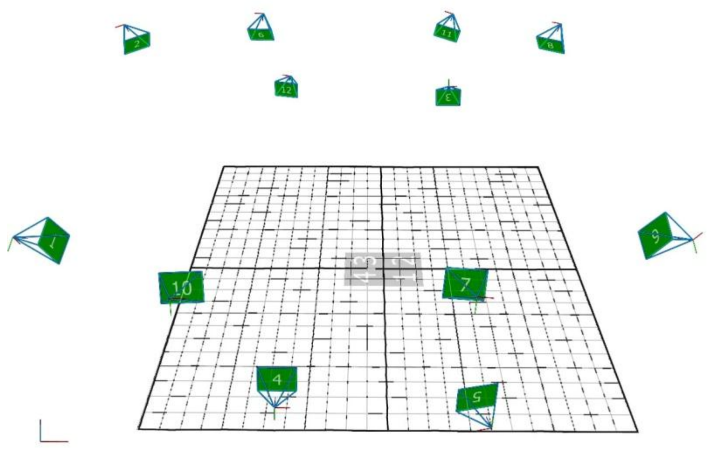

Figure 29: Layout of the cameras (shown as green boxes) and force plates (the grey boxes) of the gait laboratory.

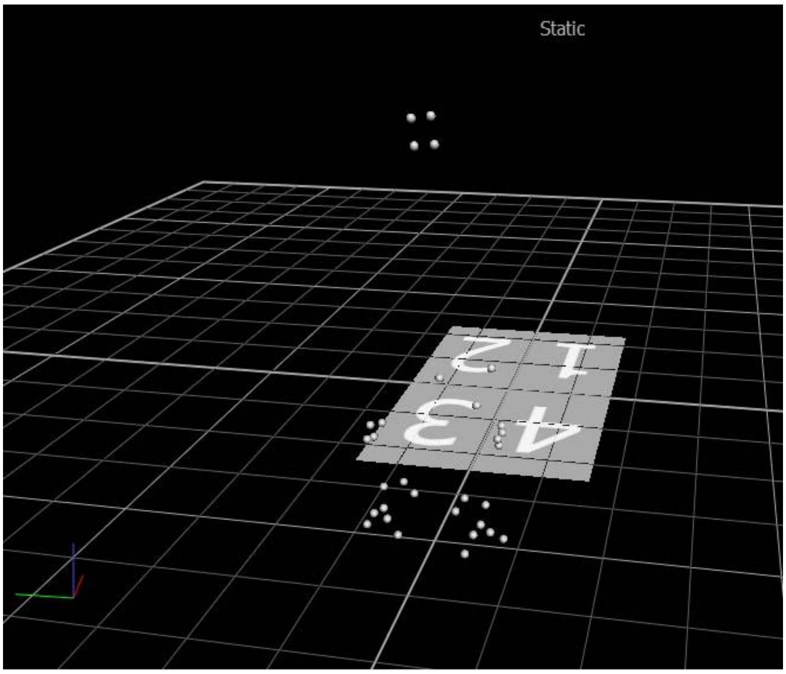

Figure 30: Illustration of what is seen when capturing a five second static calibration.

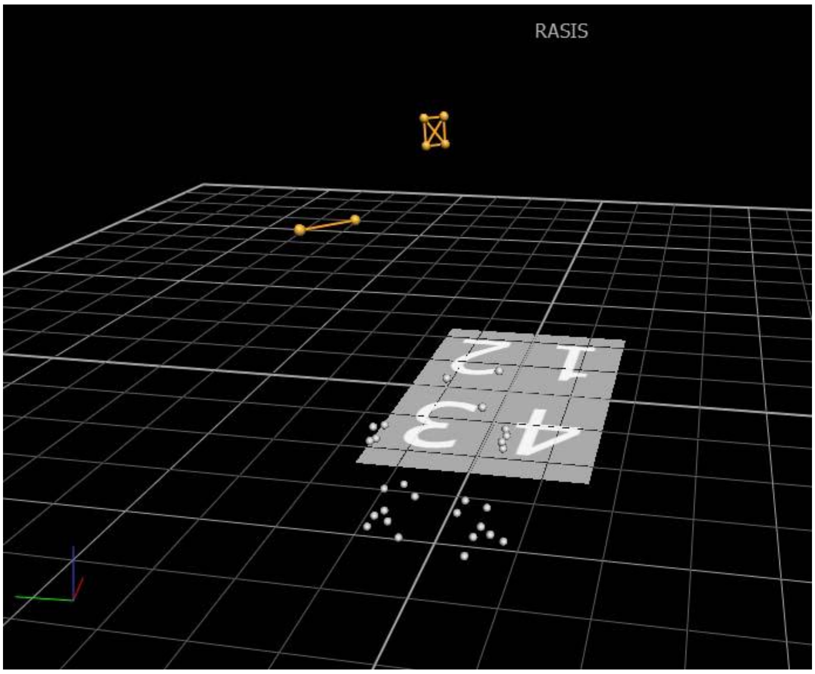

Figure 31: Illustration of what is seen when capturing a static calibration of the marker RASIS using the wand (the wand is represented by the two markers connected together and the four markers connected together are the pelvis cluster markers in orange).

3.2.4. Trial walk procedures

With the static and static wand calibration completed, the walking trials were ready to be recorded. Four walking scenarios were considered for this study. The procedures noted in section 3.2.2 were repeated for each of the four scenarios:

- Subject’s own cosmetic cover with two stockings.
- Bare prosthesis only.
- Subject with new cosmetic cover.
- Subject with new cosmetic cover with two stockings.

As subjects 1, 2, 4, 5 and 6 wore a cosmetic cover, trials walks for the subjects own cosmetic cover with two stockings was conducted first. For bare prosthesis scenario, the subject’s own cosmetic cover was removed by the Prosthetist, who then placed the individual markers onto the body, strapped the lower leg markers clusters on the prosthetic side with foam and Velcro and placed the feet marker clusters on (see figure 32).

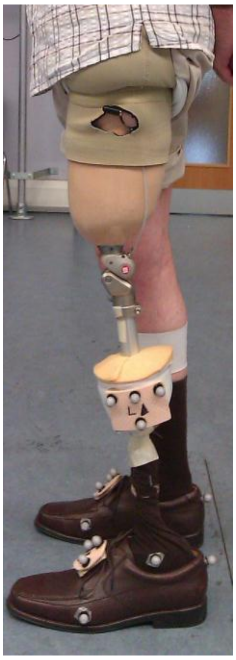

Figure 32: Photograph of how the leg marker cluster was attached to the prosthesis by the Prosthetist.

For the new cosmetic cover scenario the markers on the bare prosthesis was removed and the Prosthetist attached the new cosmetic cover onto the bare prosthesis which was then secured at the top (on the socket) by Velcro.

In all scenarios involving cosmetic covers, the cover was ‘rolled down’ by the Prosthetist before the static wand calibrations was conducted.

When walking the subject was asked to hit the force plate ‘cleanly’ (i.e. step within the force plate as much as possible avoiding stepping on the boundaries of the force plate) and walk at their own self-selected gait speed.

One walk was classified as walking one length of the gait laboratory with each foot hitting a force plate once. Each walk was captured and at least ten ‘good’ walks were recorded (i.e. clean strikes on the force plates, markers still attached to the subject, no noticeable deliberate adjustment of subject’s gait speed or step length to strike the force plate cleanly and the VICON system program indicating that force plate and marker trajectory data were recorded).

Subject 3 is not a cosmetic cover wearer and therefore only three of the scenarios were conducted for this subject. For when conducting the trial walks for the new cover scenario on subjects 3 and 6 due to the knee joint unit being larger than the internal space originally carved out inside for the prosthesis. It required the Prosthetist to cut a slit at the back in the new cosmetic cover to accommodate the knee joint unit and was sealed with double-sided tape (see figure 33).

Figure 33: Photograph example of the slit made to the back of the new cosmetic cover.

3.2.5. Post-processing

The static calibrations and 'good' walk trials were then reconstructed so visible 3D co-ordinates of the markers could be checked. The trial walks that showed no force plate data recorded, missing markers and deliberate alterations of speed and step length were discarded from further processing.

The markers in each captured static calibrations were manually labelled according to section 3.2.1 figures 25 and 26. For the static wand calibrations, only the cluster markers of the pelvis, left and right lower leg and feet were labelled. For the trial walks, the cluster markers were labelled between the time frames of before each limb underwent one gait cycle on the force plate. This manual labelling was required for the MATLAB program (MATLAB version R2012a 7.14.0.739 Natick, Massachusetts: The MathWorks Inc., 2012) detailed below.

Once labelled, the trial walks and calibrations were exported out in .csv format (exported data included force plate and labelled marker trajectories). After exportation, each static calibration and trial walk was checked to ensure it was exported correctly and any missing labelling or gaps in trajectories were amended on the VICON and exported out again before being input into the MATLAB program.

The MATLAB program required manual input of the subject's measurements previously noted down by the Prosthetist during testing day. Once these measurements and the name of one walk were input into the program, hip, knee and ankle flexion angle graphs were produced for that one walk that was input. This was to verify that no irregularity were seen (e.g. noise) and any walks that showed irregularity were discarded from further analysis.

The MATLAB program (see Appendix C) also normalised the data of each walk trial to one gait cycle. This normalised data was then averaged and produced graphs (see Appendix E).

Step length was calculated by looking at the reconstructed markers on VICON of the walks that were input into the MATLAB program. The time frame of when heel strike occurred on the force plate was noted and the time frame of the other foot was noted. Then the LCAL and RCAL markers was labelled and exported out to .csv file. The X co-ordinates of LCAL and RCAL were used to calculate the step length. The same procedure used for step length was also used to calculate the stride length. Velocity was calculated using the equation:

$$Velocity = \frac{Stride}{Time}$$

Time is taken as the duration from heel strike of the foot on the force plate to the next consecutive heel strike of the same foot.

As each subject used a different type of knee joint mechanism in the trial walks, ANOVA and a multi-comparison test on the normalised data was performed to compare the scenarios against each other of one subject, using MATLAB (see Appendix D).

4. Results

The results presented in this chapter are on a subject to subject basis as each subject uses different types of knee joint unit (see Appendix E). As a result, each same scenario of each subject cannot be compared as each knee joint unit influences the outcomes differently; therefore the results are grouped into each subject and compared internally.

Five gait parameters are reported in this chapter: knee flexion angle range, step length, stride length and velocity. The mean and one standard deviation are presented on tables in this chapter and were calculated from trial walks of each scenario. Differences in a selected parameter of each walk scenario was tested using the multiple comparison ANOVA described in section 3.2.4 at a significance level, P=0.05. For multiple comparison ANOVA test results see Appendix D.

4.1. Subject 1

For bare prosthesis and new cosmetic cover with stockings, nine walk trials were used, 10 walk trials for subject's own cosmetic cover was used and 11 walk trials from when the subject and the new cosmetic cover were used for further analysis.

From Appendix E figure 34 and 35 the average knee angle produced for each scenario is consistent, were the maximum flexion angle was above 50°.

Knee flexion angle range for all 4 scenarios for subject 1.		
	Mean (degrees)	**Standard deviation (degrees)**
Subject's own cosmetic cover + 2 stockings	58.17	0.97
Bare prosthesis	56.41	2.88
New cosmetic cover	55.97	2.21
New cosmetic cover + 2 stockings	53.31	2.63

Table 3: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle for subject 1.

From the multi-comparison ANOVA test, it was found there were significant differences between the flexion angle ranges of at least one of the scenarios used ($P < 0.05$). Flexion angle range (table 3) found in bare prosthesis scenario was significantly different from when the subject was wearing the new cosmetic cover with two stockings ($P<0.05$). There are no statistical differences between the scenarios of bare prosthesis, subject's own cosmetic cover and when the new cosmetic cover was worn. However, when comparing the scenarios of when the subject's own cosmetic cover was worn and the new cosmetic cover with two stockings there is a considerable difference.

Step length for all 4 scenarios for subject 1.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	0.69	0.03
Bare prosthesis	0.72	0.04
New cosmetic cover	0.68	0.09
New cosmetic cover + 2 stockings	0.71	0.02

Table 4: Mean and standard deviation of step length calculated for each scenario for subject 1.

There are no statistic differences between the step length found from each scenario when compared with each other ($P=0.487$) (table 4). For stride length, there was only a considerable difference between when the new cover was worn and when the new cover was worn with two stocking (table 5).

Stride length for all 4 scenarios for subject 1.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	1.21	0.04
Bare prosthesis	1.22	0.06
New cosmetic cover	1.19	0.12
New cosmetic cover + 2 stockings	1.25	0.03

Table 5: Mean and standard deviation of stride length calculated for each scenario for subject 1.

Velocity for all 4 scenarios for subject 1.		
	Mean (m/s)	**Standard deviation (m/s)**
Subject's own cosmetic cover + 2 stockings	1.1	0.02
Bare prosthesis	1.0	0.05
New cosmetic cover	1.0	0.05
New cosmetic cover + 2 stockings	1.20	0.03

Table 6: Mean and standard deviation of velocity calculated for each scenario for subject 1.

Statistical differences however was found in the velocity between bare prosthesis, when subject's own cosmetic cover was worn and when the new cover with two stockings was worn (table 6). No significant differences were found between when the new cosmetic cover was worn when compared to bare prosthesis and subject's own cosmetic cover. Instead, when the new cosmetic cover was compared to when stockings was worn with the new cover there was a difference found in the velocity.

4.2. Subject 2

For subject's own cosmetic cover only eight walk trials were used, bare prosthesis 11 walk trials were used, new cosmetic cover and new cosmetic cover with stockings 13 walk trials were used for further analysis.

From Appendix E figure 36 and 37, the maximum knee angle achieved for each scenario did not go beyond 30° except for new cosmetic cover with stockings were the maximum knee angle was below 10°.

Knee flexion angle range for all 4 scenarios for subject 2.		
	Mean (degrees)	**Standard deviation (degrees)**
Subject's own cosmetic cover + 2 stockings	25.01	0.83
Bare prosthesis	25.50	2.48
New cosmetic cover	19.47	3.35
New cosmetic cover + 2 stockings	11.45	1.54

Table 7: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle for subject 2.

Statistical differences was found between each scenario for subject 2 (P<0.05) when comparing the range of flexion angle (table 7). However no significant differences was found when comparing bare prosthesis to when the subject's own cover was worn (P>0.05) (table 8).

Step length for all 4 scenarios for subject 2.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	0.56	0.05
Bare prosthesis	0.54	0.06
New cosmetic cover	0.61	0.18
New cosmetic cover + 2 stockings	0.55	0.03

Table 8: Mean and standard deviation of step length calculated for each scenario for subject 2.

No significant differences was found for step or stride length when each scenario was compared with one another (P=0.511 and P=0.351 respectively) (table 9).

Stride length for all 4 scenarios for subject 2.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	1.22	0.06
Bare prosthesis	1.21	0.04

New cosmetic cover	1.19	0.12
New cosmetic cover + 2 stockings	1.25	0.03

Table 9: Mean and standard deviation of stride length calculated for each scenario for subject 2.

Velocity for all 4 scenarios for subject 2.		
	Mean (m/s)	**Standard deviation (m/s)**
Subject's own cosmetic cover + 2 stockings	0.67	0.13
Bare prosthesis	0.69	0.06
New cosmetic cover	0.65	0.10
New cosmetic cover + 2 stockings	0.85	0.20

Table 10: Mean and standard deviation of velocity calculated for each scenario for subject 2.

When comparing velocities, considerable differences was only found when comparing when the new cover with two stockings with when the subject's own cosmetic cover was worn ($P<0.05$) (table 10).

4.3. Subject 3

For subject 3, only nine walk trials were used for bare prosthesis, for new cosmetic cover and when the cover included stockings only 13 walk trials were used for further analysis.

Appendix E figure 38 and 39, the average knee angle produced for each scenario is shown to have a consistent trend pattern and maximum knee flexion angle below 50°.

Knee flexion angle range for all 4 scenarios for subject 3.		
	Mean (degrees)	**Standard deviation (degrees)**
Bare prosthesis	31.11	1.20
New cosmetic cover	31.28	1.37
New cosmetic cover + 2 stockings	24.11	1.65

Table 11: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle for subject 3.

Statistical differences was found when comparing the flexion angle range of bare prosthesis with when new cosmetic cover was worn with two stockings and also when the new cosmetic cover was compared to when two stockings was worn with the new cosmetic cover ($P<0.05$). However no differences were found when comparing the flexion angle range of the bare prosthesis and when the new cover was worn (table 11).

Step length for all 4 scenarios for subject 3.		
	Mean (m)	**Standard deviation (m)**
Bare prosthesis	0.71	0.06
New cosmetic cover	0.76	0.03
New cosmetic cover + 2 stockings	0.70	0.03

Table 12: Mean and standard deviation of step length calculated for each scenario for subject 3.

Stride length for all 4 scenarios for subject 3.		
	Mean (m)	**Standard deviation (m)**
Bare prosthesis	1.40	0.04
New cosmetic cover	1.40	0.03
New cosmetic cover + 2 stockings	1.40	0.09

Table 13: Mean and standard deviation of stride length calculated for each scenario for subject 3.

Velocity for all 4 scenarios for subject 3.		
	Mean (m/s)	**Standard deviation (m/s)**
Bare prosthesis	1.40	0.10
New cosmetic cover	1.34	0.06
New cosmetic cover + 2 stockings	1.40	0.15

Table 14: Mean and standard deviation of velocity calculated for each scenario for subject 3.

No significant differences were seen when comparing the step length (table 12), stride length (table 13), and velocity (table 14) of each of the scenarios for subject 3 (P=0.074, P=0.742 and P=0.783 respectively).

4.4. Subject 4

12 walk trials were used for the bare prosthesis scenario, 11 walk trials were used for subject's own cosmetic cover, seven walk trials were used for new cosmetic cover and

eight were used for the new cosmetic cover with stockings for analysis in the MATLAB program.

Appendix E figure 40 and 41 shows all average knee angle graphs show a consistent trend except bare prosthesis where there is a slight noticeable flexion at the beginning of the gait cycle compared to the other graphs.

Knee flexion angle range for all 4 scenarios for subject 4.		
	Mean (degrees)	**Standard deviation (degrees)**
Subject's own cosmetic cover + 2 stockings	27.81	2.33
Bare prosthesis	51.17	3.71
New cosmetic cover	42.43	1.69
New cosmetic cover + 2 stockings	44.42	1.76

Table 15: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle subject for 4.

There was significant differences between the flexion angle range of at least two of the scenarios. Bare prosthesis was significantly different from when the subject's cosmetic cover was worn, when the new cover was worn and when the new cover included stockings ($P<0.05$). However when comparing the scenario of when the new cosmetic cover was worn against when the new cosmetic worn with stockings there was no significant difference (table 15).

Step length for all 4 scenarios for subject 4.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	0.59	0.02
Bare prosthesis	0.60	0.02
New cosmetic cover	0.67	0.03
New cosmetic cover + 2 stockings	0.57	0.04

Table 16: Mean and standard deviation of step length calculated for each scenario for subject 4.

For step length, bare prosthesis was significantly different with when the new cosmetic cover was worn ($P<0.05$). When the subject wore their own cosmetic cover, it was only considerably different from when the new cosmetic cover was worn ($P<0.05$). On the other hand, when the new cosmetic cover was worn the step length was considerably different to the other three scenarios' step length ($P<0.05$) (table 16).

Stride length for all 4 scenarios for subject 4.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	1.24	0.03
Bare prosthesis	1.24	0.03
New cosmetic cover	1.30	0.01
New cosmetic cover + 2 stockings	1.26	0.01

Table 17: Mean and standard deviation of stride length calculated for each scenario for subject 4.

For stride length, statistical difference was only seen when comparing the scenarios of when the new cosmetic cover was worn, bare prosthesis and when the subject worn their own cosmetic cover (P<0.05). However, when comparing bare prosthesis with the stride length found when the subject was wearing their own cosmetic cover, there was no difference seen or when the new cosmetic cover worn with stockings was compared with the other three scenarios' stride length subject 4 (table 17).

Velocity for all 4 scenarios for subject 4.		
	Mean (m/s)	**Standard deviation (m/s)**
Subject's own cosmetic cover + 2 stockings	0.99	0.09
Bare prosthesis	1.02	0.10
New cosmetic cover	0.91	0.08
New cosmetic cover + 2 stockings	0.91	0.05

Table 18: Mean and standard deviation of velocity calculated for each scenario for subject 4.

No significant difference in velocities was found when comparing the scenarios with each other (table 18).

4.5. Subject 5

14 good walk trials were use for bare prosthesis, 15 walk trials were used for the subject's own cosmetic cover scenario, 10 walk trials were used for the new cosmetic cover and 13 walk trails were used for the new cosmetic cover with stockings for further analysis.

Appendix E figure 42 and 43 shows each scenario's average knee angle graph show similar trend pattern but show the different maximum knee angle each scenario had achieved.

Knee flexion angle range for all 4 scenarios for subject 5.		
	Mean (degrees)	**Standard deviation (degrees)**
Subject's own cosmetic cover + 2 stockings	40.25	1.71
Bare prosthesis	54.42	2.07
New cosmetic cover	37.79	1.34
New cosmetic cover + 2 stockings	44.46	3.32

Table 19: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle for subject 5.

Comparing the flexion angle range of the bare prosthesis showed a significant difference to the other three scenarios (P<0.05). However from the multi-comparison ANOVA test, comparing the flexion angle range of when the subject's own cosmetic cover was worn, it was only significantly different to bare prosthesis and when the new cosmetic cover with stockings was worn. Another difference was seen when comparing the flexion angle range of the new cosmetic cover with bare prosthesis and new cosmetic cover with stockings (P<0.05) (table 19).

Step length for all 4 scenarios for subject 5.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	0.60	0.02
Bare prosthesis	0.59	0.02
New cosmetic cover	0.60	0.04
New cosmetic cover + 2 stockings	0.61	0.02

Table 20: Mean and standard deviation of step length calculated for each scenario subject 5.

Stride length for all 4 scenarios for subject 5.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	1.30	0.01
Bare prosthesis	1.28	0.02
New cosmetic cover	1.29	0.05
New cosmetic cover + 2 stockings	1.31	0.02

Table 21: Mean and standard deviation of stride length calculated for each scenario for subject 5.

No differences were seen when step length (table 20), stride length (table 21) or velocities (table 22) of each four scenarios were compared (P=0.11, P=0.10 and P=0.095 respectively).

Velocity for all 4 scenarios for subject 5.		
	Mean (m/s)	**Standard deviation (m/s)**
Subject's own cosmetic cover + 2 stockings	1.06	0.04
Bare prosthesis	1.03	0.03
New cosmetic cover	1.02	0.05
New cosmetic cover + 2 stockings	1.04	0.03

Table 22: Mean and standard deviation of velocity calculated for each scenario for subject 5.

4.6. Subject 6

10 good walk trials from bare prosthesis, 17 walk trials from the subject's own cosmetic cover scenario, nine walk trials from new cosmetic cover and 12 walk trials from new cosmetic cover with stockings were used for analysis.

Appendix E figure 44 and 45 illustrates the averaged knee angle from each scenario. The graphs follow similar trend but do not have similar maximum knee flexion angle.

Knee flexion angle range for all 4 scenarios for subject 6.		
	Mean (degrees)	**Standard deviation (degrees)**
Subject's own cosmetic cover + 2 stockings	34.09	2.08
Bare prosthesis	41.67	1.94

New cosmetic cover	38.89	3.46
New cosmetic cover + 2 stockings	43.84	2.18

Table 23: Mean and standard deviation of flexion angle range taken from during pre-swing and swing (from approximately 34% and onwards) of one gait cycle for subject 6.

Bare prosthesis flexion angle range was statistically different from the flexion angle range found when the new cosmetic cover was worn with stockings ($P<0.05$). When the subject wore their own cosmetic cover, significant differences were seen when compared to the other three scenarios. When the new cosmetic cover was worn with stockings, the flexion angle range was shown to have a notable difference when compared with when the subject wore their own cosmetic cover and when the new cosmetic cover was worn ($P<0.05$). No significant difference was found when the flexion angle range was compared to when the new cosmetic cover was worn with stockings (table 23).

Step length for all 4 scenarios for subject 6.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	0.66	0.04
Bare prosthesis	0.68	0.03
New cosmetic cover	0.68	0.02
New cosmetic cover + 2 stockings	0.72	0.02

Table 24: Mean and standard deviation of step length calculated for each scenario for subject 6.

Step length for some of the scenarios were found to be significant different from each other. Bare prosthesis was different from when the new cover was worn with stockings but not when the new cover was solely worn or when the subject's own cosmetic cover was worn. Subject's own cosmetic cover was only statistically different from new cosmetic cover ($P<0.05$). Step length of when the new cosmetic cover was worn was only significantly different from when it was also worn with new stockings. Whilst, when wearing the new cosmetic cover with stockings, the step length found was shown to have a significant difference to the other three scenarios investigated (table 24).

Stride length for all 4 scenarios for subject 6.		
	Mean (m)	**Standard deviation (m)**
Subject's own cosmetic cover + 2 stockings	1.46	0.03
Bare prosthesis	1.44	0.03
New cosmetic cover	1.45	0.03
New cosmetic cover + 2 stockings	1.47	0.03

Table 25: Mean and standard deviation of stride length calculated for each scenario for subject 6.

Stride length of bare prosthesis and when new cosmetic cover was worn with stockings, when compared together was showed to have considerable differences from each other. However, when the subject's own cosmetic cover was worn or when the new cosmetic cover was worn, when compared to the other three scenarios no significant differences were found (table 25).

Velocity for all 4 scenarios for subject 6.		
	Mean (m/s)	**Standard deviation (m/s)**
Subject's own cosmetic cover + 2 stockings	1.06	0.04
Bare prosthesis	1.03	0.03
New cosmetic cover	1.02	0.05
New cosmetic cover + 2 stockings	1.04	0.03

Table 26: Mean and standard deviation of velocity calculated for each scenario for subject 6.

Bare prosthesis and when the subject's own cosmetic cover was worn showed differences in velocities when compared to the other three scenarios (P<0.05). Velocity of when the new cosmetic cover was worn showed only to be significantly different from bare prosthesis and when the subject's own cosmetic cover was worn. The velocity of the new cover with two stockings was not significantly different from when only wearing the new cosmetic cover but was different when compared to bare prosthesis and when subject's own cosmetic cover was worn (table 26).

5. Discussion

This chapter is presented in the same format as chapter 4 and will explain further the results and graphs presented in chapter 4 and ends with a summary of findings of each subject.

5.1. Subject 1

Appendix E figure 34 and 35 illustrates, especially the bare prosthesis that the maximum knee flexion angle that can be achieved is 60° **(Hale, 1991).** Interestingly the subject's own cosmetic cover was able to achieve a maximum knee flexion greater than 60° compared to bare prosthesis. For the new cosmetic cover with stocking graph, it showed slight extension by the subject during the beginning 10% of the gait cycle to prevent buckling, which is not uncommon to see.

Comparing the mean range of knee flexion angle of all four scenarios, it can be seen that there is a significant reduction of the range of flexion that the subject could achieve when the new cosmetic cover with stocking was worn (53.31 ± 2.63°) compared to when the subject's cosmetic cover was worn (58.17 ± 0.97°). This indicates that the new cover including stockings is restrictive, and for subject 1, the maximum knee flexion was found to be similar to that of the sound limb (52°) **(Rose et al., 1994).** It is not surprising to see there is no significant difference for subject 1 between the knee flexion ranges of bare prosthesis and when the subject is wearing own cosmetic cover (56.41 ± 2.88° and 58.17 ± 0.97° respectively) as the condition of the cosmetic cover showed typical signs of extensive use (e.g. thin wall thickness, ripped at the anterior side of the knee joint mechanism) and therefore would have no restrictions and could freely flex and achieve similar flexion angle range as a bare prosthesis (see figure 34).

Differences in step length were expected to be observed between each scenario for example, if the new cosmetic cover showed restriction of flexion throughout swing phase then the step length should be shorter. However, when the step length of each of the scenarios was compared to the step length found in bare prosthesis, no significant changes between all scenarios were seen. This can be due to various reasons such as misplaced markers or changes in gait when attempting to strike force plates commonly as (the patient was instructed to 'cleanly strike' each force plate before the walk trials commenced) and as a result the subject could deliberately alter their step length to accommodate this.

For stride length, the only difference seen was the reduction of stride length when the new cosmetic cover included the stockings compared to when only the new cosmetic cover was worn (1.37 ± 0.04m and 1.4 ± 0.03m respectively). It is expected that since there is a reduction of flexion angle range between the subject's own cover and the new cosmetic cover it would cause the stride length to decrease, however when comparing the stride length of the subject's own cosmetic cover and the new cosmetic cover no differences was found in this case. Instead, there was a significant increase of stride length between the new cosmetic cover and when the new cover was worn with stockings (1.37 ± 0.04m and 1.42 ± 0.3m respectively), figure 35 shows a decrease in flexion angle between new cosmetic cover and when stockings was included. As the speed was self-selected, the step and stride length can be deliberately altered by the subject.

The velocity, when the subject had a bare prosthesis on, showed the slowest than the other three scenarios examined (1.08 ± 0.05m/s). Unexpectedly when the subject wore the new cosmetic cover with stockings it was shown the velocity increased (1.20 ± 0.03m/s) when compared to the other three conditions. When the subject wore the new cosmetic cover the velocity (1.10 ± 0.05m/s) was found to be surprisingly not different from the velocity when the subject wore their own cosmetic cover or the bare prosthesis. Conversely, when the new cosmetic cover included stockings the velocity significantly increased (1.20 ± 0.03m/s) compared to subject's own cosmetic cover and bare prosthesis (1.14 ± 0.02 and 1.10 ± 0.05m/s respectively).

5.2. Subject 2

Appendix E figure 36 and 37, the gait cycle begins with a 5° extension at the beginning of each cycle this can be due to the two quad walking stick aids used by the subject during each walk trials of each scenario. By relying on the walking aids this can reduce the flexion that is usually seen in early stance as a shock absorber after heel strike. New cosmetic cover with stockings shows a 10° extension and could be as a combined effect of using walking aids and the new cosmetic cover being restrictive.

Notably, the flexion angle range was seen between at least three difference scenarios. Bare prosthesis was shown to have the highest flexion angle range (25.5 ± 2.48°) compared to when the new cosmetic cover was worn (19.45 ± 3.35°) and when the new cosmetic cover included stockings (11.45 ± 1.54°). Again, there was no difference between bare prosthesis and the subject's own cosmetic cover. The reduced flexion angle range for the new cosmetic cover and when stockings were included (19.45 ± 3.35° and 11.45 ± 1.54° respectively) was hypothesised as a new cover restricts flexion and therefore angle range. The bare prosthesis and subject's own cosmetic cover showed a low flexion angle range compared to the other subjects. This can be attributed to the subject requiring for the walk trials, two quad walking stick aids as the subject was not confident relying only on the 3R49 knee joint unit without the swing-phase stabiliser on. The new cosmetic cover with stockings flexion angle range (11.45 ± 1.54°) was significantly reduced that the prosthesis side during swing phase seemed to remain in extension.

Step and stride length remained fairly constant this can be as a result of the use of the two walking aids. Velocity on the other hand from the statistical analysis showed differences. New cosmetic cover with stockings was shown to be faster compared to when the subject's own cosmetic cover was worn. It would have been expected the opposite to occur as the old cosmetic cover would be worn out therefore easier to flex to move faster.

5.3. Subject 3

Appendix E Figure 38 and 39, the graphs show a constant trend pattern, however the gait cycle begins with 20° extension, which most probably is due to markers being misplaced (usually the LASIS, RASIS, RPSIS and LPSIS as they are difficult to find) and therefore would cause the referencing of the hip markers to the knee clusters to be inaccurate and therefore the inaccurate flexion angles to be calculated by the MATLAB program. As a result the maximum knee angles shown on figure 35 are inaccurate.

Unlike the previous subjects, subject 3 there are no significant differences of the flexion angle range between the bare prosthesis and when the new cosmetic cover is worn (31.11 ± 1.2° and 31.28 ± 1.37° respectively). A reason for this could be due to as the knee joint unit is a C-leg (see Appendix B) or as a result of the slit at the back of the new cosmetic cover (as shown in figure 38), therefore the flexion angle range for the new cosmetic cover is similar to that found for the bare prosthesis. It was noted by subject 3 that during the walk trials for new cosmetic cover with stockings, a difference was felt with the new cosmetic cover on as the prosthesis now felt "*much heavier"* and felt their "*stride was shorter*" **(Subject 3, 2012)**. This is reflected on the flexion angle range when the new cosmetic cover was worn with stockings (24.11 ± 1.65°) was significantly reduced when compared to that of bare prosthesis and when the new cosmetic cover was worn. The subject commented on the difficulty of flexing the knee joint as a result of the new cosmetic cover restricting such as "*cannot bend, pulls on the hip."* and is "*much stiffer*." **(Subject 3, 2012).** However, statistical analysis on the knee flexion angle range and the knee angle (figure 38 and 39) showed no significant variance in the knee flexion achieved. Step and stride length revealed there was no significant difference of the step or stride length as reported by the subject (bare prosthesis step length and new cosmetic cover 0.71 ± 0.06m and 0.7 ± 0.03m respectively). Statistical analysis also showed there was no difference in velocities between all three scenarios. The constant step length, stride length and velocity throughout the three conditions can be possibly attributed to the microprocessor of the C-Leg to prevent falling; however the subject felt "*needed to work harder even with the C-Leg microprocessor aid."* Apart from this, subject 3 also commented that to walk, required "*to use the normal leg to drag the prosthetic leg"* **(Subject 3, 2012).**

5.4. Subject 4

Appendix E figure 40 and 41 subject's own cosmetic cover, new cosmetic cover and new cosmetic cover with stockings also showed the same marker misplacement as that seen on subject 3 and figure 35.

Subject 4 showed significant differences between the flexion angle range of bare prosthesis and with the subject's own cosmetic cover (51.18 ± 3.72° and 27.81 ± 2.33° respectively). A possible reason for this could be due to the condition of the subject's own cover: As noted by the Prosthetist that the cover looked to be (in comparison with the other subjects tested) in a "*hardly used*" condition **(Murray, 2012)**. Hence, the subject's own cosmetic cover would act as a new cosmetic cover and cause restriction in the amount of flexion available. The flexion angle range however, for when the new cosmetic cover was worn showed considerable larger range of flexion (42.43 ± 1.7°) compared to the range found when the subject's own cosmetic cover was worn. No differences was found between the new cosmetic cover and when with stockings (44.42 ± 1.76°).

Step length for subject 4 was shown to have a significant difference as hypothesised. When the new cosmetic cover was worn, the step length (0.67 ± 0.03m) was greater than the subject's own cosmetic cover (0.59 ± 0.02m) and bare prosthesis (0.60 ± 0.02m). As the new cosmetic cover was shown to be restrictive, it reduces the flexion angle that is possible during swing phase therefore the step length would increase. When the new cosmetic cover was worn with stocking the step length reduced (0.57 ± 0.04m) and similar length as when the subject's own cosmetic cover.

The stride length was also seen to be significantly altered. When the new cosmetic cover was worn again, it showed to have the longest stride when compared to the other scenarios (1.30 ± 0.01m). Even though the flexion angle range of when the subject's own cosmetic cover was reduced, the stride length was similar to that of bare prosthesis (1.24 ± 0.03m and 1.24 ± 0.03m respectively)

For subject 4, no velocity changes were seen when comparing the scenarios against each other (see table 18).

5.5. Subject 5

Subject 5, appendix E figure 42 and 43, the new cosmetic cover showed a 5° extension at the beginning of the gait cycle compared to the graphs of the other scenarios. Bare prosthesis showed the maximum knee flexion angle as expected and the scenarios involving the new cosmetic cover showed to have the similar flexion angle however, statistically the flexion angle range of new cosmetic cover and new cosmetic cover with stockings was shown to be significantly different.

Bare prosthesis showed the highest flexion range (54.4 ± 2.07°) as there is nothing to inhibit the amount of flexion that it can achieve. New cosmetic cover showed the least flexion angle possible (37.79 ± 1.34°) as expected but was not significantly different to when the subject's own cosmetic cover was worn (40.25 ± 1.71°). When stockings were added to the new cosmetic cover however the range of flexion angle increased instead (44.46 ± 3.32°).

Step length, stride length and velocity however did not show any significant differences for this subject.

5.6. Subject 6

Subject 6 Appendix E figure 44 and 45, the bare prosthesis, subject's own cosmetic cover scenario and new cosmetic cover graphs shows the same problem as figure 35 and 36.

Subject 6 also commented that the new foam felt "*stiffer and the muscles had to work harder.*" **(Subject 6, 2012)** however the new cosmetic cover with stocking showed the highest flexion angle range (43.84 ± 2.18°) compared to the other scenarios. The subject's own cosmetic cover had the lowest flexion angle range (34.09 ± 2.09°) when also compared to the other scenarios.

The step length of the new cosmetic cover with stockings also had the longest step length (0.72 ± 0.02m) when compared to the other scenarios. Step length of the subject's own cosmetic cover and new cosmetic cover was shown to be similar (0.66 ± 0.04m and 0.68 ± 0.02m). Stride length also reflected this with the new cosmetic cover with stockings having a stride of 1.48 ± 0.02m compared to bare prosthesis 1.44 ± 0.03m, which shows the new cosmetic cover has an effect in causing the stride length to increase.

The velocity of when subject is wearing their own cosmetic cover (1.33 ± 0.04m/s) was significantly different from the other scenarios; this can be attributed to as subject 6 is a cosmetic cover wearer, and therefore is comfortable with that. Bare prosthesis is significantly lower at 1.2 ± 0.02m/s can be because the subject is not comfortable with not wearing a cover and would therefore walk slower. New cosmetic cover and with stockings (1.26 ± 0.05 m/s and 1.25 ± 0.03m/s respectively) showed no differences but when both are individually compared with when the subject's own cosmetic cover is worn, both are significantly lower than the velocity of the subject's own cosmetic cover. This can be due to the subject not adjusted to the new cosmetic cover and therefore walks slower.

5.7. Summary of discussion

This sub-section summaries the findings for each subject.

For subject 1 the bare prosthesis had similar range of knee flexion angle as when the subject's own cosmetic cover was worn. New cosmetic cover was shown to have the least knee flexion range available. No step length changes were observed when comparing each scenario. Reduction of stride length was seen when comparing new cosmetic cover with subject's own cosmetic cover. Increase of stride length when comparing new cosmetic cover with when stockings were included. Bare prosthesis had the slowest velocity and when new cosmetic cover with stockings had the fastest velocity.

For subject 2, bare prosthesis knee flexion angle range had the highest knee flexion angle range. Reduction of flexion angle range was seen between new cosmetic cover and when stockings were also worn.

For subject 3, reduction of knee flexion angle range was seen for when new cosmetic cover was worn with stockings. No significant differences of step, stride length and velocity were seen.

For subject 4, there were differences between the knee flexion angle range of bare prosthesis and subject's own cosmetic cover. No differences found between new cosmetic cover and when stockings was included. Step length of when new cosmetic cover was worn was larger than subject's own cosmetic cover and bare prosthesis. Stride length of new cosmetic cover was largest compared to the other scenarios. No velocity changes were seen between each scenario.

Subject 5, bare prosthesis showed the largest knee flexion angle range compared to new cosmetic cover and subject's own cosmetic cover, which showed the least knee flexion angle range. No differences in step, stride length and velocity were seen.

For subject 6, the subject's own cosmetic cover showed the least knee flexion range and then new cosmetic cover with stockings showed the highest knee flexion angle range. For step and stride length, the new cosmetic cover with stockings showed the largest step and stride length.

The findings for this study have shown to be varied on a subject to subject basis. However the step and stride length have shown to be consistent with those reported in literature **(Jaegers et al., 1995; Sagawa Jr. et al., 2011).** On the other hand, there could be errors in calculating stride length as the time for one stride to occur was calculated by the heel strike of the foot on the force plate and the next consecutive heel strike of the same foot. Here, inaccuracy can arise as the consecutive heel strike does not occur on a force plate therefore is difficult to see the time when that heel strike occurs. Subjects 1 and 3 both used an intelligence prosthesis (Endolite smart IP and C-Leg model respectively) were in both cases the step and stride length for each scenario, when compared with each other showed no differences. This could be as a result of the microprocessor knee unit aid in maintaining consistent length to avoid having the amputee walk with an incorrect gait.

For some subjects a significant change was seen when the new cosmetic cover was worn such as the maximum knee flexion angle however they also shown some discrepancy as a result of marker misplacement or marker labelling errors.

6. Conclusion and future work

The aim of this study was prove the hypothesis that a cosmetic cover has an effect on the gait performance of the prosthesis by analysing the gait parameters, such as knee flexion angle range, step length, stride length and velocity to determine if the hypothesis is true by using motion capturing system.

The results derived from the subjects remain inconclusive, for some subjects the new cosmetic cover does lower the knee flexion angle during swing phase however in other subjects, it causes the knee flexion angle range to increase. The step length and stride length as well as the velocity are seen in some subjects to be influenced by the restrictiveness caused by the new cosmetic cover however; this is not consistently seen in other subjects. Subject's use of microprocessor controlled prosthesis have should that the cosmetic cover does not cause significant restrictiveness. This can be due to numerous reasons, such as one subject required a walking aid and there did not fully flex the prosthesis as would be done when in a non-laboratory environment.

Inclusive results can be due to incorrect marker placements especially on multi-axis feet as they do not have any landmarks that closely resemble the bony landmarks on the anatomical foot. This would result in inaccurate calculations of the knee angles as the relationship points between places (e.g. the relationship distances between the LLEF and LLMAL marker) would not be at the correct distance and therefore can result in calculating knee angles of a gait cycle to show the subject is walking with an extension when they are not. The subjects also used different prosthesis and therefore results of one scenario could not be compared with the results from another subject of the same scenario. Or errors could have occurred during data processing for example, subject 4 new cosmetic cover scenario, only seven walk trials were analysed due to some of the walk trials did not have force plate data recorded or the marker trajectories showed considerable amount of markers gaps during the trial walks and therefore could not be used to be further analysed by the MATLAB program. As a result, new cosmetic cover is not truly represented for subject 4. The parameters, stride and velocity can also be miscalculated as the first heel strike occurs on the force plate, the second heel strike of that same foot however occurs on the ground and not on another force plate. Thus, it is difficult to work out when the second heel strike occurs as there is no force vector to indicate this moment.

Two subjects noticed and commented on when the new cosmetic cover was worn, how it was affecting the way they were walking (e.g. making it harder to walk or need to use more energy to move the prosthesis). This shows that not only has it been noticed by Prosthetists but also by the amputees and therefore is an area that should be explored as to see the extent of influence the cosmetic cover can have. This can be useful as it can lead to research into the use of other suitable materials to be used as cosmetic covers.

Overall, some findings by comparing the results of each scenario in each subject of this study have shown that the knee flexion angle range during swing phase of gait can be reduced by the use of a new cosmetic cover however to prove this requires keeping certain factors constant (i.e. marker placements) to provide more accurate results. For more statistically significant result, more time is needed to acquire enough data to also run statistical t-tests to examine the significant difference of each scenario.

For future works in this study, apart from investigating into the parameters looked at in this study, examining into the internal shear and friction forces between the prosthesis and internally of the cosmetic cover without the influence of the subject should also be considered. This control experiment can be conducted by having a machine simulate a gait cycle of the prosthesis in all four scenarios. More gait parameters (such as knee moments and knee joint power) could be considered to be looked at to provide a further more conclusive result. Other factors, for example, prosthesis type should be kept identical so results can be compared via scenario and not by, as in this study by a subject to subject basis.

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Appendix A

Invitation to Participate

Name of department: Department of Design, Manufacture and Engineering Management

Title of the study: Assessment of cosmetic cover influence on the functional performance of above knee prostheses

4/4/12

Sir/Madam,

You have been contacted as a member of the Murray Foundation to ask for your participation in a research investigation conducted by the University of Strathclyde, Glasgow. The study is investigating cosmetic covers for artificial limbs and their influence on the performance of the limb. Your participation in this investigation would be very much appreciated and may help improve the design of current cosmetic covers.

Please take the time to read the enclosed Participant Information Sheet and decide if you would like to participate.

Please note that you have been contacted using the most up to date information that the Murray Foundation has. If you believe there to be an error in the information, please contact the Murray Foundation directly.

Thank you for your time and attention.

Sincerely,

Dr Nicola Cairns

Participant Information Sheet

Name of department: Department of Design, Manufacture and Engineering Management

Title of the study: Assessment of cosmetic cover influence on the functional performance of trans-femoral prostheses

Introduction

My name is Nicola Cairns and I am a Research Fellow in the Department of Design, Manufacture and Engineering Management at the University of Strathclyde, Glasgow. My contact details can be found at the end of this form.

What is the purpose of this investigation?

The purpose of this investigation is to determine if the cosmetic cover (cosmesis) that is worn around the artificial limb (prosthesis) influences the way the mechanical components of the limb operate. The overall aim of the research is to improve the foams used to make cosmeses. Your participation in the study may help this process.

Why have you been invited to take part?

You have been invited to take part because you are an above knee amputee who uses a prosthesis with a cosmesis fitted over it.

Do you have to take part?

Taking part in this investigation is voluntary. If you decide not to participate, it will not affect you in any way. You also have the right to withdraw your participation at any time. If you decide to withdraw, any information you have already provided will be destroyed.

What will you do in the project?

Should you decide to participate, you will be asked to visit the Bioengineering Department, University of Strathclyde, 103 Rottenrow East, Glasgow, for one day. You will be reimbursed for travel expenses to and from the university.

When at the Bioengineering Department removable markers will be placed on your cosmesis. The markers are small and lightweight (like small ping pong balls) and are easy to remove. The cosmesis will not be damaged. You will then be asked to walk a 10 metre length in a straight line at your normal speed. The walking test is indoors and on a flat surface. During the walking test the position of the markers will be recorded using video. The video cameras will be focused on the markers on your leg. You will NOT be identifiable from the recording.

You will be asked to repeat the straight line walking several times to make sure the information has been recorded. Rest breaks will be given in between the walking tests where you can sit down.

Next you will be asked to remove your prosthesis. There are toilets and changing rooms available for this. A clinically registered prosthetist will then remove the cosmesis from your limb. The cosmesis will not be damaged by this process. The small markers will be placed on the prosthesis. You will then be asked to put the limb back on and repeat the walking test.

Finally, a new cosmesis will be fitted to the prosthesis. The markers will be placed on the foam and you will be asked to repeat the walking test again.
At the end the original cosmesis will be replaced on your prosthesis and adjusted to your satisfaction.

What are the potential risks to you in taking part?
When taking part in the walking test there is no more risk to falling than you would normally experience at home or walking outside. You will only be asked to walk a short distance on a level surface indoors and rest breaks will be provided regularly where you can sit down.

What happens to the information in the project?
The information you provide is anonymous and will be treated in the strictest confidence. The electronic information collected in the walking test will be stored securely at the University of Strathclyde. Your personal details will not be kept. The information will be used by researchers at the University of Strathclyde to help determine possible areas of improvement in the design of cosmeses. The information will not be used for any other purpose or in any other investigation.

The University of Strathclyde is registered with the Information Commissioner's Office who implements the Data Protection Act 1998. All personal data on participants will be processed in accordance with the provisions of the Data Protection Act 1998.

Thank you for reading this information – please contact the researcher if you are unsure about what is written here.

What happens next?
If you are willing to participate you need to sign the attached consent form including your address and contact telephone number and return in the prepaid envelope. The researcher will then contact you to discuss the project further and, if you are able to participate in the study, arrange your visit to the university. Thank you very much for your help in this investigation.

If you do not want to participate, simply dispose of this information letter. Thank you for your time and attention.

[This investigation has received ethical approval from the University of Strathclyde Ethics Committee.]

If you have any questions/concerns, during or after the investigation, or wish to contact an independent person to whom any questions may be directed or further information may be sought from, please contact:

Secretary to the University Ethics Committee
Research & Knowledge Exchange Services
University of Strathclyde
Graham Hills Building
50 George Street
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Consent Form

Name of department: Department of Design, Manufacture and Engineering Management

Title of study: Assessment of cosmetic cover influence on the functional performance of above knee prostheses

- I confirm that I have read and understood the information sheet for the above project and the researcher has answered any queries to my satisfaction.
- I understand that my participation is voluntary and that I am free to withdraw from the project at any time, without having to give a reason and without any consequences.
- I understand that I can withdraw my data from the study at any time.
- I understand that my participation involves visiting the University of Strathclyde for one day and doing a series of short walking tests in a straight line and on a level surface.
- I understand that video recording equipment will be used in the testing, but that I will not be identifiable in the video
- I understand that I will be reimbursed for any travel expenses to and from the university on the day of the test.
- I understand that any information recorded in the investigation will remain confidential and no information that identifies me will be made publicly available.
- I consent to being a participant in the project.

(PRINT NAME)	Hereby agree to take part in the above project
Signature of Participant:	Date

Please return this consent form in the prepaid envelope.

If you would like to be informed of the outcome of the investigation or any publications concerning the outcomes, please tick the box [ ].

Please provide your address and contact telephone number so that the researcher can contact you about the project. Your details will be stored separately and will NOT be used to identify you.

Address and telephone number:

Appendix B

Subject	Gender	Age	Side of Amputation	Year of Amputation	Cause of Amputation	Activity Level	Use of Assistive Device	Cosmetic Cover wearer
1	M	54	Left	1991	Trauma	K2	Walking Stick - for uneven terrain	Yes
2	M	67	Left	2009	PVD	K2 -3	Walking stick	Yes
3	F	37	Right	1993	Tumour	K1-2	No	No
4	M	52	Right	1996	Trauma (RTA)	K2	Walking Stick	Yes
5	M	56	Right	1995	Trauma	K2-3	Walking stick (Occasionally)	Yes (skinergy to knee)
6	M	52	Right	1979	Trauma	K2	Walking Stick	Yes

Table 1: General information on the subjects.

Subject	Suspension type	Knee unit	Foot unit
1	Si liner + one-way valve	Endolite Smart IP	Multi-flex
2	TES belt	3R49 (Stabilised)	OB dynamic
3	Si liner + one-way valve (vacuum)	C-Leg	OB Dynamic plus
4	Liner + pin	3R80 (Hydraulic)	OB Dynamic
5	Suction	3R60 (Stance flex)	Blatchford Multiflex
6	Suction	3R80 (Hydraulic)	Flex foot

Table 2: Information on the various types of prosthetic components each subject used for the trial walks.

Appendix C

MATLAB programs (written by Crimin A.) were used to extract the knee flexion angle produced during one gait cycle graphs.

```
clearvars -except static normalised_data
%===========================anthropomrtrics=================================
gender='male';
data_sheet.trunk_length=0;%mm
data_sheet.marker_diameter=14;%mm
data_sheet.height=0;%mm
%for amputee only
data_sheet.waist=0;%mm
data_sheet.hip=0;%mm
data_sheet.body_mass=0;%kg
data_sheet.foot_length=0;%mm
%mm from tibial tuberosity marker to condyle plateu
data_sheet.condyle_plateu_L=0;
data_sheet.condyle_plateu_R=0;
%STUMP and PROSTHETIC PARAMTERS
%specify whether above (AK) or below (BK) knee amputation
data_sheet.level='N/A';
%dimensions in mm or kg
data_sheet.proximal_end_circumference=0;
data_sheet.distal_end_circumference=0;
data_sheet.stump_length=0;
%leg mass for AK
data_sheet.socket_wall_thickness=0;
data_sheet.socket_mass=0;
%leg mass for BK
data_sheet.leg_mass=0;
%number of leg ossolations in ten seconds
data_sheet.time_leg_coronal=0;
data_sheet.time_leg_transverse=0;
data_sheet.time_leg_sagittal=0;
%foot mass
data_sheet.foot_mass=0;
%number of foot ossolations in ten seconds
data_sheet.time_foot_coronal=0;
data_sheet.time_foot_transverse=0;
data_sheet.time_foot_sagittal=0;
%centre of mass in mm measured from proximal end
data_sheet.leg_centre_of_mass=0;
data_sheet.foot_centre_of_mass=0;

%======================data uploading and processing ====================
if exist('static','var')==0
static=static_upload1('static.CSV',1-amputee/0-non amputee);
end
NAME='name the sheet to upload';
dynamic=dynamic_upload1(strcat(NAME,'.CSV'));
```

```
%adjusting total body mass used for amputee when determining anthropometric
measures
if data_sheet.waist~=0 && data_sheet.hip~=0
[ data_sheet.body_mass ] =
body_weight(data_sheet.height,data_sheet.waist,data_sheet.hip,gender);
end
%gait analysis function
[joint_angles,kinematics,moments,force,powers,prosthetic_left,prosthetic_right,
dynamic,segment_mass]=...
    gait_analysis(static,dynamic,data_sheet,gender,100);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%momentum
clear('momentum')

%total mass (m)
momentum.m=data_sheet.body_mass;

%com positions
momentum.R_foot_L=0.001*dynamic.anatomical_centres.com_foot_L;
momentum.R_foot_R=0.001*dynamic.anatomical_centres.com_foot_R;
momentum.R_leg_L=0.001*dynamic.anatomical_centres.com_leg_L;
momentum.R_leg_R=0.001*dynamic.anatomical_centres.com_leg_R;
momentum.R_thigh_L=0.001*dynamic.anatomical_centres.com_thigh_L;
momentum.R_thigh_R=0.001*dynamic.anatomical_centres.com_thigh_R;
momentum.R_trunk=0.001*dynamic.anatomical_centres.com_trunk;


%com of system
momentum.com=(segment_mass.foot_mass_L*momentum.R_foot_L+segment_mass.foot_mass
_R*momentum.R_foot_R+...

segment_mass.leg_mass_L*momentum.R_leg_L+segment_mass.leg_mass_R*momentum.R_leg
_R+...

segment_mass.thigh_mass_L*momentum.R_thigh_L+segment_mass.thigh_mass_R*momentum
.R_thigh_R+...
    segment_mass.trunk_mass*momentum.R_trunk)/momentum.m;


%momentum of segments around body com
for n=1:length(kinematics.v_trunk_com)

momentum.foot_L(n,:)=segment_mass.foot_mass_L*cross(momentum.R_foot_L(n,:)-
momentum.com(n,:),...
    kinematics.v_foot_L(n,:)-kinematics.v_trunk_com(n,:));
momentum.foot_R(n,:)=segment_mass.foot_mass_R*cross(momentum.R_foot_R(n,:)-
momentum.com(n,:),...
    kinematics.v_foot_R(n,:)-kinematics.v_trunk_com(n,:));
momentum.leg_L(n,:)=segment_mass.leg_mass_L*cross(momentum.R_leg_L(n,:)-
momentum.com(n,:),...
    kinematics.v_leg_L(n,:)-kinematics.v_trunk_com(n,:));
momentum.leg_R(n,:)=segment_mass.leg_mass_R*cross(momentum.R_leg_R(n,:)-
momentum.com(n,:),...
    kinematics.v_leg_R(n,:)-kinematics.v_trunk_com(n,:));
```

```
momentum.thigh_L(n,:)=segment_mass.thigh_mass_L*cross(momentum.R_thigh_L(n,:)-momentum.com(n,:),...
    kinematics.v_thigh_L(n,:)-kinematics.v_trunk_com(n,:));
momentum.thigh_R(n,:)=segment_mass.thigh_mass_R*cross(momentum.R_thigh_R(n,:)-momentum.com(n,:),...
    kinematics.v_thigh_R(n,:)-kinematics.v_trunk_com(n,:));
%momentum of the body around the com
momentum.com_momentum(n,:)=momentum.foot_L(n,:)+momentum.foot_R(n,:)+...
    momentum.leg_L(n,:)+momentum.leg_R(n,:)+...
    +momentum.thigh_L(n,:)+momentum.thigh_R(n,:);

momentum.AC_L(n,:)=momentum.com_momentum(n,:)+...
    cross(momentum.com(n,:)-momentum.R_foot_L(n,:),momentum.m*(kinematics.v_trunk_com(n,:)-kinematics.v_foot_L(n,:)));

momentum.AC_R(n,:)=momentum.com_momentum(n,:)+...
    cross(momentum.com(n,:)-momentum.R_foot_R(n,:),momentum.m*(kinematics.v_trunk_com(n,:)-kinematics.v_foot_R(n,:)));
end

momentum.ankle=momentum.AC_R+momentum.AC_L;


%ensuring momentum is always expressed in a positive sense which means on
%the x and z directiion is affected

    if max(kinematics.v_AC_L(:,1))< 0.5
        momentum.com_momentum(:,3)=-momentum.com_momentum(:,3);
        momentum.AC_L(:,3)=-momentum.AC_L(:,3);
        momentum.AC_R(:,3)=-momentum.AC_R(:,3);
        momentum.ankle(:,3)=-momentum.ankle(:,3);
        momentum.com_momentum(:,1)=-momentum.com_momentum(:,1);
        momentum.AC_L(:,1)=-momentum.AC_L(:,1);
        momentum.AC_R(:,1)=-momentum.AC_R(:,1);
        momentum.ankle(:,1)=-momentum.ankle(:,1);
    end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%energy

clear('conservative_energies')
%mass of HAT
conservative_energies.m_HAT=data_sheet.body_mass-(segment_mass.thigh_mass_R+...

segment_mass.leg_mass_R+segment_mass.foot_mass_R+segment_mass.thigh_mass_L+...
    segment_mass.leg_mass_L+segment_mass.foot_mass_L);

%conservative energies

for n=1:length(kinematics.a_trunk_com)

%for the trunk
conservative_energies.KE_trunk(n,1)=0.5*conservative_energies.m_HAT*kinematics.v_trunk_com(n,1)^2;
```

```
%for left thigh
conservative_energies.KE_thigh_L(n,1)=0.5*segment_mass.thigh_mass_L*norm(kinema
tics.v_thigh_L(n,:))^2;

%for left leg
conservative_energies.KE_leg_L(n,1)=0.5*segment_mass.leg_mass_L*norm(kinematics
.v_leg_L(n,:))^2;

%for left foot
conservative_energies.KE_foot_L(n,1)=0.5*segment_mass.foot_mass_L*norm(kinemati
cs.v_foot_L(n,:))^2;

%for right thigh
conservative_energies.KE_thigh_R(n,1)=0.5*segment_mass.thigh_mass_R*norm(kinema
tics.v_thigh_R(n,:))^2;

%for right leg
conservative_energies.KE_leg_R(n,1)=0.5*segment_mass.leg_mass_R*norm(kinematics
.v_leg_R(n,:))^2;

%for right foot
conservative_energies.KE_foot_R(n,1)=0.5*segment_mass.foot_mass_R*norm(kinemati
cs.v_foot_R(n,:))^2;

%total kinetic energy
conservative_energies.KE(n,:)=conservative_energies.KE_trunk(n,1)+...

conservative_energies.KE_thigh_L(n,1)+conservative_energies.KE_thigh_R(n,1)+...
    conservative_energies.KE_leg_L(n,1)+conservative_energies.KE_leg_R(n,1)+...
    conservative_energies.KE_foot_L(n,1)+conservative_energies.KE_foot_R(n,1);

end

conservative_energies.g=9.81;

%com positions
conservative_energies.R_foot_L=0.001*dynamic.anatomical_centres.com_foot_L;
conservative_energies.R_foot_R=0.001*dynamic.anatomical_centres.com_foot_R;
conservative_energies.R_leg_L=0.001*dynamic.anatomical_centres.com_leg_L;
conservative_energies.R_leg_R=0.001*dynamic.anatomical_centres.com_leg_R;
conservative_energies.R_thigh_L=0.001*dynamic.anatomical_centres.com_thigh_L;
conservative_energies.R_thigh_R=0.001*dynamic.anatomical_centres.com_thigh_R;
conservative_energies.R_trunk=0.001*dynamic.anatomical_centres.com_trunk;

for n=1:length(kinematics.a_trunk_com)

%for the trunk
conservative_energies.PE_trunk(n,:)=conservative_energies.m_HAT*...
    conservative_energies.g*conservative_energies.R_trunk(n,2);

%for left thigh
conservative_energies.PE_thigh_L(n,:)=segment_mass.thigh_mass_L*...
    conservative_energies.g*conservative_energies.R_thigh_L(n,2);
```

```
%for left leg
conservative_energies.PE_leg_L(n,:)=segment_mass.leg_mass_L*...
    conservative_energies.g*conservative_energies.R_leg_L(n,2);

%for left foot
conservative_energies.PE_foot_L(n,:)=segment_mass.foot_mass_L*...
    conservative_energies.g*conservative_energies.R_foot_L(n,2);

%for right thigh
conservative_energies.PE_thigh_R(n,:)=segment_mass.thigh_mass_R*...
    conservative_energies.g*conservative_energies.R_thigh_R(n,2);

%for right leg
conservative_energies.PE_leg_R(n,:)=segment_mass.leg_mass_R*...
    conservative_energies.g*conservative_energies.R_leg_R(n,2);

%for right foot
conservative_energies.PE_foot_R(n,:)=segment_mass.foot_mass_R*...
    conservative_energies.g*conservative_energies.R_foot_R(n,2);

conservative_energies.PE(n,:)=conservative_energies.PE_trunk(n,1)+...

conservative_energies.PE_thigh_L(n,1)+conservative_energies.PE_thigh_R(n,1)+...
    conservative_energies.PE_leg_L(n,1)+conservative_energies.PE_leg_R(n,1)+...
    conservative_energies.PE_foot_L(n,1)+conservative_energies.PE_foot_R(n,1);
end


%plot(dynamic.time,conservative_energies.PE,'r')
%hold on
%plot(dynamic.time,conservative_energies.KE,'b')

%single pendulum model

%total mass (m)
conservative_energies.m=data_sheet.body_mass;

%com of system
conservative_energies.com=(segment_mass.foot_mass_L*conservative_energies.R_foo
t_L+...
    segment_mass.foot_mass_R*conservative_energies.R_foot_R+...
    segment_mass.leg_mass_L*conservative_energies.R_leg_L+...
    segment_mass.leg_mass_R*conservative_energies.R_leg_R+...
    segment_mass.thigh_mass_L*conservative_energies.R_thigh_L+...
    segment_mass.thigh_mass_R*conservative_energies.R_thigh_R+...

conservative_energies.m_HAT*conservative_energies.R_trunk)/conservative_energie
s.m;


conservative_energies.V(:,1)=gradient(conservative_energies.com(:,1),0.01);
conservative_energies.V(:,2)=gradient(conservative_energies.com(:,2),0.01);
conservative_energies.V(:,3)=gradient(conservative_energies.com(:,3),0.01);
```

```
conservative_energies.a(:,1)=gradient(conservative_energies.V(:,1),0.01);
conservative_energies.a(:,2)=gradient(conservative_energies.V(:,2),0.01);
conservative_energies.a(:,3)=gradient(conservative_energies.V(:,3),0.01);

for n=1:length(conservative_energies.V)

conservative_energies.V_magnitude(n,1)=norm(conservative_energies.V(n,:));
conservative_energies.a_magnitude(n,1)=norm(conservative_energies.a(n,:));

conservative_energies.pendulum_KE(n,:)=...
    0.5*conservative_energies.m*norm(conservative_energies.V(n,:))^2;
conservative_energies.pendulum_PE(n,:)=...
    conservative_energies.m*conservative_energies.g*...
    conservative_energies.com(n,2);

end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%



%include more folders if required
folders={'joint_angles','moments.local','powers'};

%===========================================================================
%===========================================================================

%creating a count folder incase one does not exist
if exist('normalised_data','var')==0
    normalised_data.count3=0;
end
%checking the column to insert data, storing the number of passes
m=normalised_data.count3(1,1);
normalised_data.count3(1,1)=m+1;


%locating a whole period during gait cycle for left and right leg
if isfield(dynamic,'GRF_L') ==1
    GRF_L=dynamic.GRF_L(:,2);
    GRF_L(GRF_L<0.05,:)=[];
    normalised_data.GRF.GRF_L(:,normalised_data.count3)=...
        interpft(GRF_L,50);
    normalised_data.GRF.GRF_L(:,sum(normalised_data.GRF.GRF_L,1)==0)=[];
[ normalised_data.count_L,normalised_data.count1_L ] =...
    gait_period1(kinematics.v_LCAL,dynamic.GRF_L);
%momentum data
    normalised_data.coronal.H_AC_L(:,normalised_data.count3)=...

interpft(momentum.AC_L(normalised_data.count_L:normalised_data.count1_L,1),50);
    normalised_data.transverse.H_AC_L(:,normalised_data.count3)=...

interpft(momentum.AC_L(normalised_data.count_L:normalised_data.count1_L,2),50);
    normalised_data.sagittal.H_AC_L(:,normalised_data.count3)=...

interpft(momentum.AC_L(normalised_data.count_L:normalised_data.count1_L,3),50);
```

```
    normalised_data.coronal.H_ankle_L(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_L:normalised_data.count1_L,1),50)
;
    normalised_data.transverse.H_ankle_L(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_L:normalised_data.count1_L,2),50)
;
    normalised_data.sagittal.H_ankle_L(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_L:normalised_data.count1_L,3),50)
;

    normalised_data.coronal.H_com_momentum_L(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_L:normalised_data.count1_L
,1),50);
    normalised_data.transverse.H_com_momentum_L(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_L:normalised_data.count1_L
,2),50);
    normalised_data.sagittal.H_com_momentum_L(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_L:normalised_data.count1_L
,3),50);

%conservative energies data
     normalised_data.energies.pendulum_KE_L(:,normalised_data.count3)=...

interpft(conservative_energies.pendulum_KE(normalised_data.count_L:normalised_d
ata.count1_L,1),50);

     normalised_data.energies.pendulum_PE_L(:,normalised_data.count3)=...

interpft(conservative_energies.pendulum_PE(normalised_data.count_L:normalised_d
ata.count1_L,1),50);

end



if isfield(dynamic,'GRF_R') ==1
    GRF_R=dynamic.GRF_R(:,2);
    GRF_R(GRF_R<0.05,:)=[];
    normalised_data.GRF.GRF_R(:,normalised_data.count3)=...
        interpft(GRF_R,50);
    normalised_data.GRF.GRF_R(:,sum(normalised_data.GRF.GRF_R,1)==0)=[];
[ normalised_data.count_R,normalised_data.count1_R ] =...
    gait_period1(kinematics.V_RCAL,dynamic.GRF_R);
%momentum data
    normalised_data.coronal.H_AC_R(:,normalised_data.count3)=...

interpft(momentum.AC_R(normalised_data.count_R:normalised_data.count1_R,1),50);
    normalised_data.transverse.H_AC_R(:,normalised_data.count3)=...
```

```
interpft(momentum.AC_R(normalised_data.count_R:normalised_data.count1_R,2),50);
    normalised_data.sagittal.H_AC_R(:,normalised_data.count3)=...

interpft(momentum.AC_R(normalised_data.count_R:normalised_data.count1_R,3),50);

    normalised_data.coronal.H_ankle_R(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_R:normalised_data.count1_R,1),50)
;
    normalised_data.transverse.H_ankle_R(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_R:normalised_data.count1_R,2),50)
;
    normalised_data.sagittal.H_ankle_R(:,normalised_data.count3)=...

interpft(momentum.ankle(normalised_data.count_R:normalised_data.count1_R,3),50)
;

    normalised_data.coronal.H_com_momentum_R(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_R:normalised_data.count1_R
,1),50);
    normalised_data.transverse.H_com_momentum_R(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_R:normalised_data.count1_R
,2),50);
    normalised_data.sagittal.H_com_momentum_R(:,normalised_data.count3)=...

interpft(momentum.com_momentum(normalised_data.count_R:normalised_data.count1_R
,3),50);

    %conservative energies data
     normalised_data.energies.pendulum_KE_R(:,normalised_data.count3)=...

interpft(conservative_energies.pendulum_KE(normalised_data.count_R:normalised_d
ata.count1_R,1),50);

     normalised_data.energies.pendulum_PE_R(:,normalised_data.count3)=...

interpft(conservative_energies.pendulum_PE(normalised_data.count_R:normalised_d
ata.count1_R,1),50);

end

clear 'GRF_L' 'GRF_R'

if length(fieldnames(joint_angles))==length(fieldnames(moments.local))
    angle=fieldnames(joint_angles);
    moment=fieldnames(moments.local);
    power=fieldnames(powers);
    varibles=horzcat(angle,moment,power);
elseif length(fieldnames(joint_angles))~=length(fieldnames(moments.local))
    if isfield(normalised_data,'count_L')==1 &&
isfield(normalised_data,'count_R')==0
        count=0;
```

```
        for n=1:length(folders)
            name=fieldnames(eval(folders{1,n}));
            for r=1:length(name)
            if sum(strfind(name{r,:},'L'))~=0
                count=count+1;
                vars{count,n}=name{r,:};
            end
            end
            count=0;
        end
    elseif isfield(normalised_data,'count_L')==0 &&
isfield(normalised_data,'count_R')==1
        count=0;
        for n=1:length(folders)
            name=fieldnames(eval(folders{1,n}));
            for r=1:length(name)
            if sum(strfind(name{r,:},'R'))~=0
                count=count+1;
                vars{count,n}=name{r,:};
            end
            end
            count=0;
        end
    end
    varibles=vars;
end

clear 'name' 'vars' 'count' 'r'

for f=1:length(folders)
for m=1:length(varibles)

    name1{1,1}=strcat(folders{1,f}, '.');
data=eval(strcat(name1{1,1}, varibles{m,f}));

%selecting the data to be normalised
if norm(strfind(varibles{m,f},'_L'))~=0
    select_x=data(normalised_data.count_L:normalised_data.count1_L,1);
    select_y=data(normalised_data.count_L:normalised_data.count1_L,2);
    select_z=data(normalised_data.count_L:normalised_data.count1_L,3);
elseif norm(strfind(varibles{m,f},'_R'))~=0
    select_x=data(normalised_data.count_R:normalised_data.count1_R,1);
    select_y=data(normalised_data.count_R:normalised_data.count1_R,2);
    select_z=data(normalised_data.count_R:normalised_data.count1_R,3);
end


% normalised_data is the assosiated values with normalised points

check=fieldnames(normalised_data);
if sum(strcmp(check,'coronal'))==1
    check1=fieldnames(normalised_data.coronal);
    if sum(strcmp(check1,varibles{m,f}))==1
    name=strcat('normalised_data.coronal.', varibles{m,f});
    ndata=normalised_data.coronal.(varibles{m,f});
```

```
    ndata(:,normalised_data.count3)=interpft(select_x,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
    if sum(strcmp(check1,varibles{m,f}))==0
    name=strcat('normalised_data.coronal.', varibles{m,f});
    ndata=interpft(select_x,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
end

if sum(strcmp(check,'coronal'))==0
    name=strcat('normalised_data.coronal.', varibles{m,f});
    ndata=interpft(select_x,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
end



check=fieldnames(normalised_data);
if sum(strcmp(check,'transverse'))==1
    check1=fieldnames(normalised_data.transverse);
    if sum(strcmp(check1,varibles{m,f}))==1
    name=strcat('normalised_data.transverse.', varibles{m,f});
    ndata=normalised_data.transverse.(varibles{m,f});
    ndata(:,normalised_data.count3)=interpft(select_y,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
    if sum(strcmp(check1,varibles{m,f}))==0
    name=strcat('normalised_data.transverse.', varibles{m,f});
    ndata=interpft(select_y,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
end

if sum(strcmp(check,'transverse'))==0
    name=strcat('normalised_data.transverse.', varibles{m,f});
    ndata=interpft(select_y,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
end



check=fieldnames(normalised_data);
```

```
if sum(strcmp(check,'sagittal'))==1
    check1=fieldnames(normalised_data.sagittal);
    if sum(strcmp(check1,varibles{m,f}))==1
    name=strcat('normalised_data.sagittal.', varibles{m,f});
    ndata=normalised_data.sagittal.(varibles{m,f});
    ndata(:,normalised_data.count3)=interpft(select_z,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
    if sum(strcmp(check1,varibles{m,f}))==0
    name=strcat('normalised_data.sagittal.', varibles{m,f});
    ndata=interpft(select_z,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
    end
    %saving to excel spread sheet
%xlswrite(strcat('sagittal',
varibles{m,f}),normalised_data.coronal.(varibles{m,f}));
end

if sum(strcmp(check,'sagittal'))==0
    name=strcat('normalised_data.sagittal.', varibles{m,f});
    ndata=interpft(select_z,50);
    ndata(:,sum(ndata,1)==0)=[];
    eval([name '= ndata;']);
    clear 'ndata'
end

end
end


if isfield(normalised_data,'count_R')==1
normalised_data=rmfield(normalised_data,'count_R');
normalised_data=rmfield(normalised_data,'count1_R');
end
if isfield(normalised_data,'count_L')==1
normalised_data=rmfield(normalised_data,'count_L');
normalised_data=rmfield(normalised_data,'count1_L');
end

if sum(strcmp('H_AC_L',fieldnames(normalised_data.sagittal)))==1
%clearing up momentum and conservative energies
normalised_data.sagittal.H_AC_L(:,sum(normalised_data.sagittal.H_AC_L,1)==0)=[]
;
normalised_data.sagittal.H_ankle_L(:,sum(normalised_data.sagittal.H_ankle_L,1)=
=0)=[];
normalised_data.sagittal.H_com_momentum_L(:,sum(normalised_data.sagittal.H_com_
momentum_L,1)==0)=[];
end

if sum(strcmp('H_AC_R',fieldnames(normalised_data.sagittal)))==1
normalised_data.sagittal.H_AC_R(:,sum(normalised_data.sagittal.H_AC_R,1)==0)=[]
```

```
;
normalised_data.sagittal.H_ankle_R(:,sum(normalised_data.sagittal.H_ankle_R,1)=
=0)=[];
normalised_data.sagittal.H_com_momentum_R(:,sum(normalised_data.sagittal.H_com_
momentum_R,1)==0)=[];
end

if sum(strcmp('pendulum_KE_L',fieldnames(normalised_data.energies)))==1
normalised_data.energies.pendulum_KE_L(:,sum(normalised_data.energies.pendulum_
KE_L,1)==0)=[];
normalised_data.energies.pendulum_PE_L(:,sum(normalised_data.energies.pendulum_
PE_L,1)==0)=[];
end


if sum(strcmp('pendulum_KE_R',fieldnames(normalised_data.energies)))==1
normalised_data.energies.pendulum_KE_R(:,sum(normalised_data.energies.pendulum_
KE_R,1)==0)=[];
normalised_data.energies.pendulum_PE_R(:,sum(normalised_data.energies.pendulum_
PE_R,1)==0)=[];
end

save(NAME)

clear 'select_x' 'select_y' 'select_z' 'average' 'deviation'
'lower_confidence'...
    'm' 'n' 'percentage' 'time' 'upper_confidence' 'angle' 'coronal' 'data'
'name1'...
    'f' 'moment' 'name' 'sagittal' 'transverse' 'check' 'check1' 'folders'
'varibles'...
    'GRF_L' 'GRF_R' 'power'

if sum(strcmp('hip_flex_angle_L',fieldnames(normalised_data.sagittal)))==1
set(figure, 'Position', [0 0 4000 4000])
subplot(1,3,1)
plot(normalised_data.sagittal.hip_flex_angle_L)
subplot(1,3,2)
plot(normalised_data.sagittal.knee_flex_angle_L)
subplot(1,3,3)
plot(normalised_data.sagittal.ankle_flex_angle_L)
title('left')
end
if sum(strcmp('hip_flex_angle_R',fieldnames(normalised_data.sagittal)))==1
set(figure, 'Position', [0 0 4000 4000])
subplot(1,3,1)
plot(normalised_data.sagittal.hip_flex_angle_R)
subplot(1,3,2)
plot(normalised_data.sagittal.knee_flex_angle_R)
subplot(1,3,3)
plot(normalised_data.sagittal.ankle_flex_angle_R)
title('right')
end
```

```
%load('normalised_data')
%folders={'normalised_data.sagittal'};
font_size=20;
%folders={'normalised_data.coronal','normalised_data.transverse',...
    %'normalised_data.sagittal','normalised_data.GRF'};

    if exist('fighandle1','var')==0
    fighandle1=figure;
    elseif exist('fighandle1','var')==1
        figure(fighandle1)
        hold on
    end

    varibles={'normalised_data.sagittal.hip_flex_angle_L';...
        'normalised_data.sagittal.knee_flex_angle_L';...
        'normalised_data.sagittal.ankle_flex_angle_L';...
        'normalised_data.sagittal.M_HC_L';...
        'normalised_data.sagittal.M_KC_L';...
        'normalised_data.sagittal.M_AC_L';...
        'normalised_data.sagittal.hip_flex_angle_R';...
        'normalised_data.sagittal.knee_flex_angle_R';...
        'normalised_data.sagittal.ankle_flex_angle_R';...
        'normalised_data.sagittal.M_HC_R';...
        'normalised_data.sagittal.M_KC_R';...
        'normalised_data.sagittal.M_AC_R'};
    count=0;


for m=1:length(varibles)


    %selecting data out of each folder
    data=eval(varibles{m,1});

    average=mean(data,2);


    for n=1:50
    [h(n,:),p(n,:),ci(n,:)] = ttest(data(n,:),0);
    end

    upper_confidence=ci(:,2);
    lower_confidence=ci(:,1);
    if count>5 && count <7
        count=0;
        if exist('fighandle2','var')==0
        fighandle2=figure;
        elseif exist('fighandle2','var')==1
        figure(fighandle2)
        hold on
        end
    end
    count=count+1;
    subplot(2,3,count)
```

```
    hold on
    set(gcf, 'color', 'white');
    %upper
    plot(1:2:100,upper_confidence,'+','LineWidth',2)
    %average
    plot(1:2:100,average,'LineWidth',2)
    %lower
    plot(1:2:100,lower_confidence,'--','LineWidth',2)
    if count==3
    AX=legend('95% upper confidence','average','95% lower confidence',...
        'Location','NorthEastOutside');
    LEG = findobj(AX,'type','text');
    set(LEG,'FontSize',10);
    end
    %removing underscores
    if strcmp('normalised_data.sagittal.M_AC_L',varibles{m,1})~=0
    header='Left Ankle Moment';
    ylabel('<-plantarflexion [moment Nm] dorsiflexion->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.M_AC_R',varibles{m,1})~=0
    header='Right Ankle Moment';
    ylabel('<-plantarflexion [moment Nm] dorsiflexion->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.M_KC_L',varibles{m,1})~=0
    header='Left Knee Moment';
    ylabel('<-flexion [moment Nm] extension->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.M_KC_R',varibles{m,1})~=0
    header='Right Knee Moment';
    ylabel('<-flexion [moment Nm] extension->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.M_HC_L',varibles{m,1})~=0
    header='Left Hip Moment';
    ylabel('<-extension [moment Nm] flexion->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.M_HC_R',varibles{m,1})~=0
    header='Right Hip Moment';
    ylabel('<-extension [moment Nm] flexion->','FontSize',font_size)
    elseif
strcmp('normalised_data.sagittal.ankle_flex_angle_L',varibles{m,1})~=0
    header='Left sagittal Ankle Angle';
    ylabel('<-plantarflexion [angle degrees] dorsiflexion-
>','FontSize',font_size)
    elseif
strcmp('normalised_data.sagittal.ankle_flex_angle_R',varibles{m,1})~=0
    header='Right Sagittal Ankle Angle';
    ylabel('<-plantarflexion [angle degrees] dorsiflexion-
>','FontSize',font_size)
    elseif
strcmp('normalised_data.sagittal.knee_flex_angle_L',varibles{m,1})~=0
    header='Left Sagittal Knee Angle';
    ylabel('<-extension [angle degrees] flexion->','FontSize',font_size)
    elseif
strcmp('normalised_data.sagittal.knee_flex_angle_R',varibles{m,1})~=0
    header='Right Sagittal Knee Angle';
    ylabel('<-extension [angle degrees] flexion->','FontSize',font_size)
    elseif strcmp('normalised_data.sagittal.hip_flex_angle_L',varibles{m,1})~=0
    header='Left Sagittal Hip Angle';
    ylabel('<-extension [angle degrees] flexion->','FontSize',font_size)
```

```
    elseif strcmp('normalised_data.sagittal.hip_flex_angle_R',varibles{m,1})~=0
    header='Right Sagittal Hip Angle';
    ylabel('<-extension [angle degrees] flexion->','FontSize',font_size)
    end

    xlabel('Percentage Gait Cycle %','FontSize',font_size)
    title(header,'FontSize',font_size)

    %clear('data')
    %close all

end


figure(fighandle1);
%This sets the units of the current  figure (gcf = get current  figure) on
paper  to centimeters.
xSize = 2*29.7; ySize = 2*21;

%These are my size variables, width of 30 and a height of 21
xLeft = (29.7-xSize)/2; yTop = (21-ySize)/2;

%Additional coordinates to center  the figure on A4-paper
set(gcf,'PaperPosition',[xLeft yTop xSize ySize])
saveas(gcf,'left','fig');
saveas(gcf,'left','jpg');

figure(fighandle2);
%This sets the units of the current  figure (gcf = get current  figure) on
paper  to centimeters.
xSize = 2*29.7; ySize = 2*21;

%These are my size variables, width of 30 and a height of 21
xLeft = (29.7-xSize)/2; yTop = (21-ySize)/2;

%Additional coordinates to center  the figure on A4-paper
set(gcf,'PaperPosition',[xLeft yTop xSize ySize])
saveas(gcf,'right','fig');
saveas(gcf,'right','jpg');
%close all
%save('normalised_data','normalised_data');

clear 'average' 'ci' 'f' 'h' 'lower_confidence' 'm' 'n' 'p' 'str'...
    'title_str' 'title_str1' 'upper_confidence' 'varibles' 'folders'...
    'font_size' 'header' 'xLeft' 'xSize' 'ySize' 'yTop'
```

Appendix D

Below are the ANOVA and multi-comparison tests conducted in MATLAB. They are presented in the format of the ANOVA results then the multi-comparison test results.

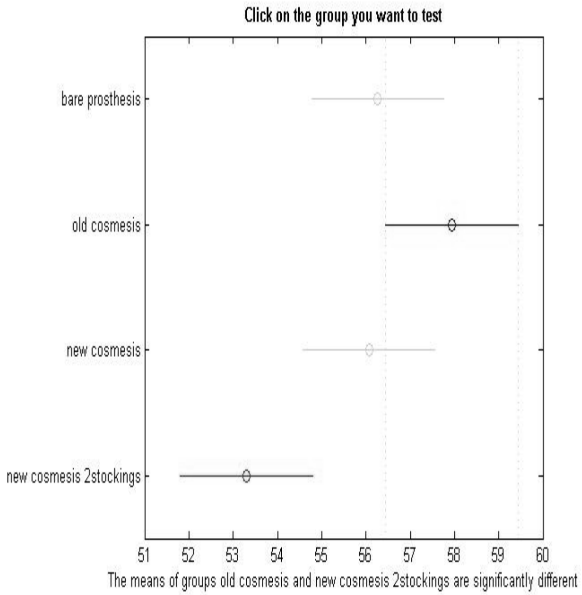
Old cosmesis refers to the subject's own cosmetic cover. New cosmesis refers to the new cosmetic cover. New cosmesis + 2 stockings refers to new cosmetic cover with two stockings.

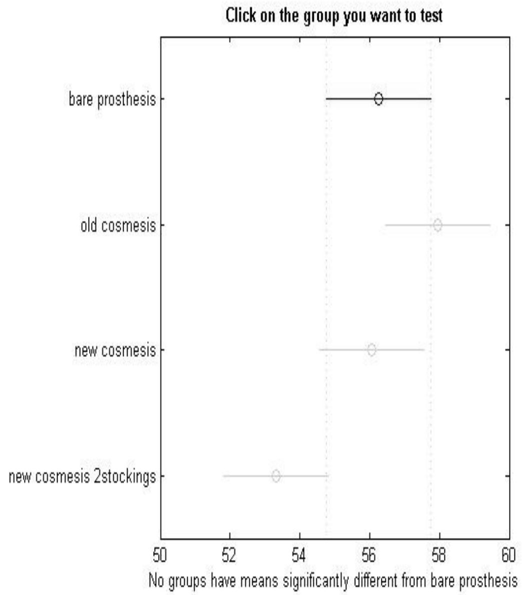
ANOVA and multi-comparison test on Subject 1, knee flexion range, step length, stride length and velocity.

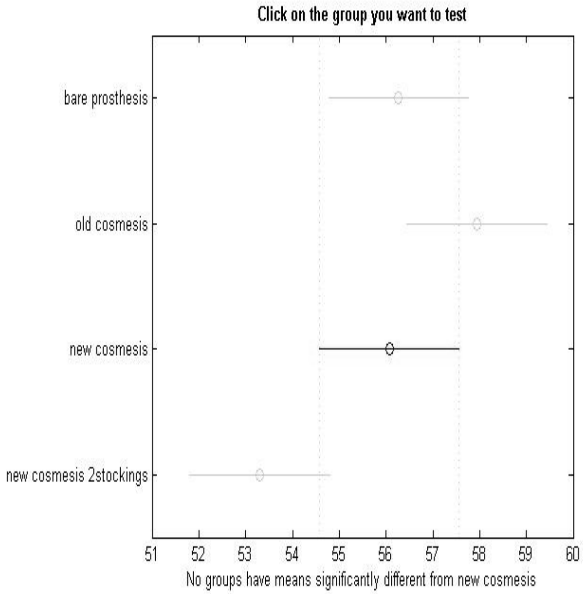
ANOVA Table

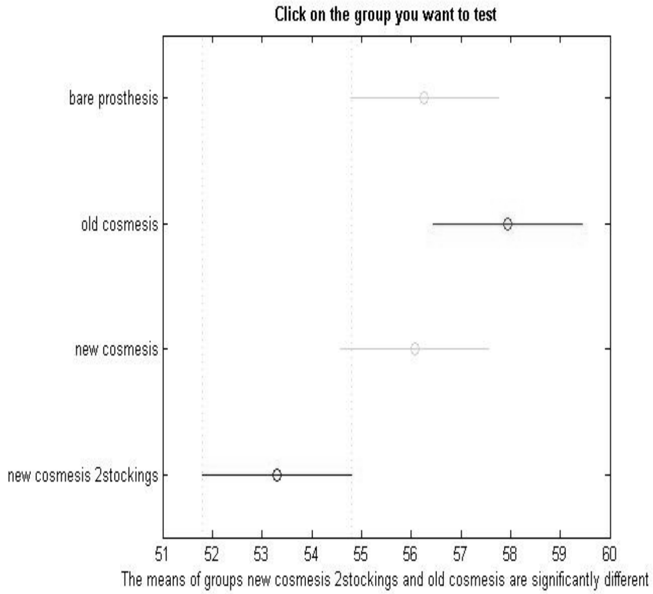
Source	SS	df	MS	F	Prob>F
Groups	99.788	3	33.2626	6.06	0.0022
Error	175.613	32	5.4879		
Total	275.401	35			

ANOVA analysis results on knee flexion angle range.



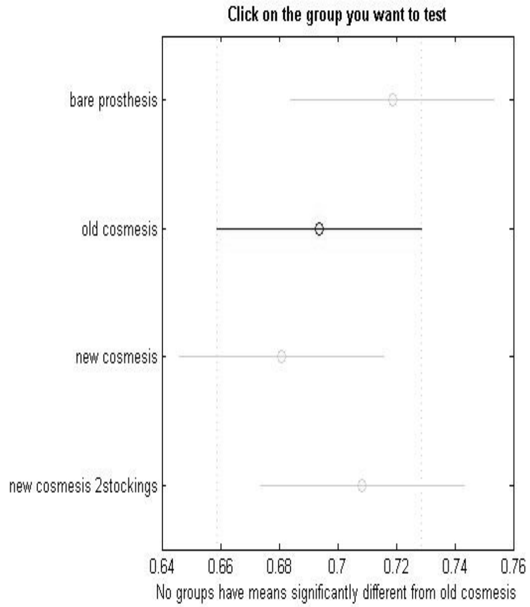


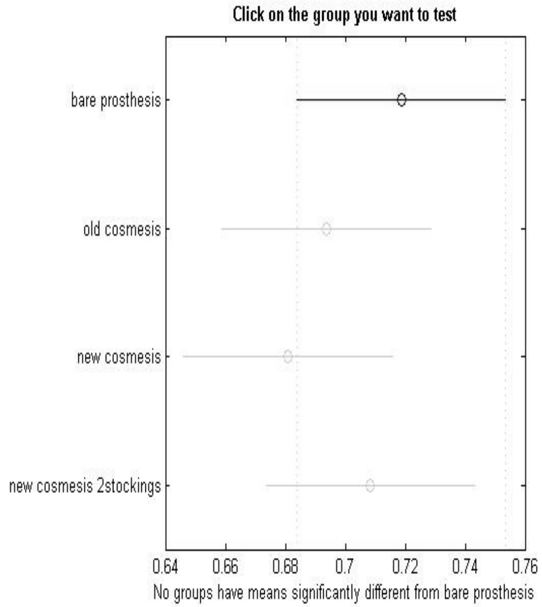

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
51
52
53
54
55
56
57
58
59
60
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
51
52
53
54
55
56
57
58
59
60
The means of groups new cosmesis 2stockings and old cosmesis are significantly different

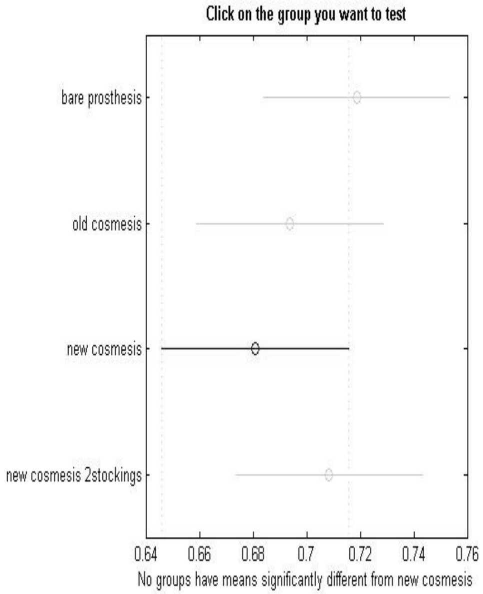
ANOVA Table

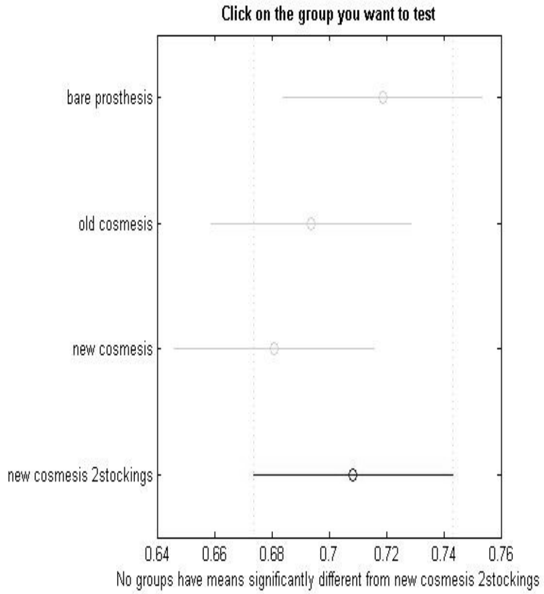
Source	SS	df	MS	F	Prob>F
Groups	0.00742	3	0.00247	0.83	0.487
Error	0.09528	32	0.00298		
Total	0.1027	35			





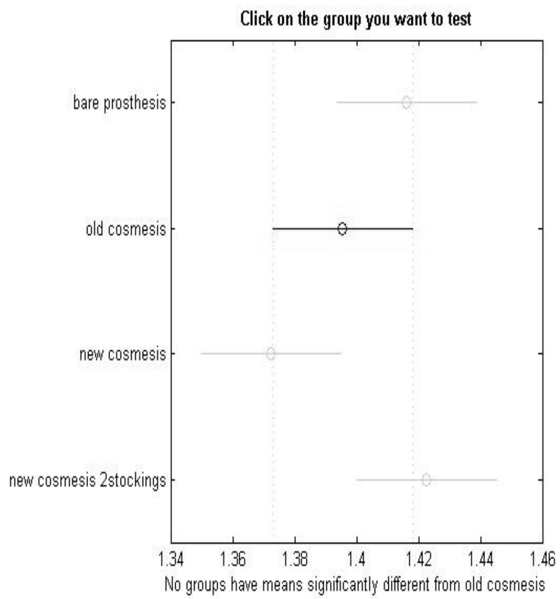
ANOVA test results of step length of subject 1.

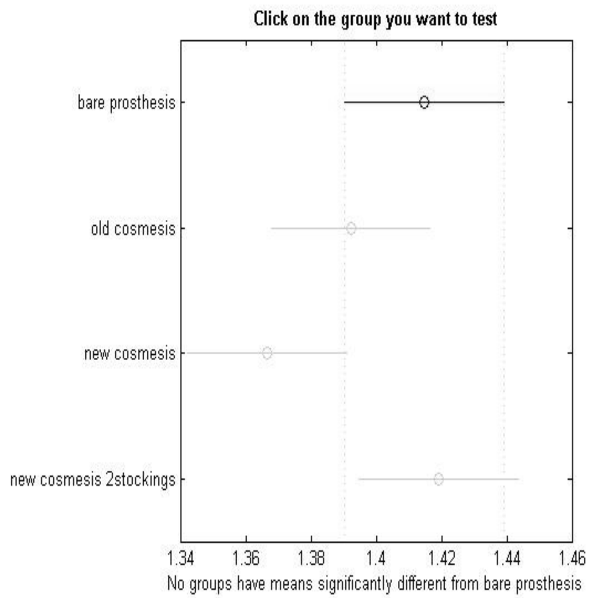

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.64
0.66
0.68
0.7
0.72
0.74
0.76
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.64
0.66
0.68
0.7
0.72
0.74
0.76
No groups have means significantly different from new cosmesis 2stockings

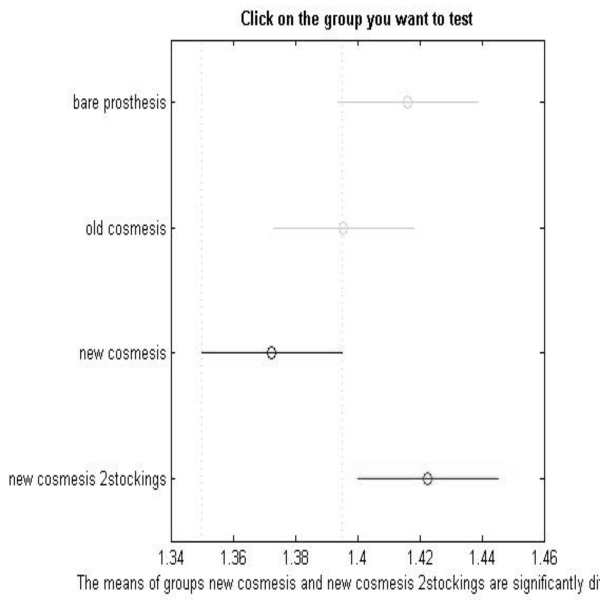
ANOVA Table

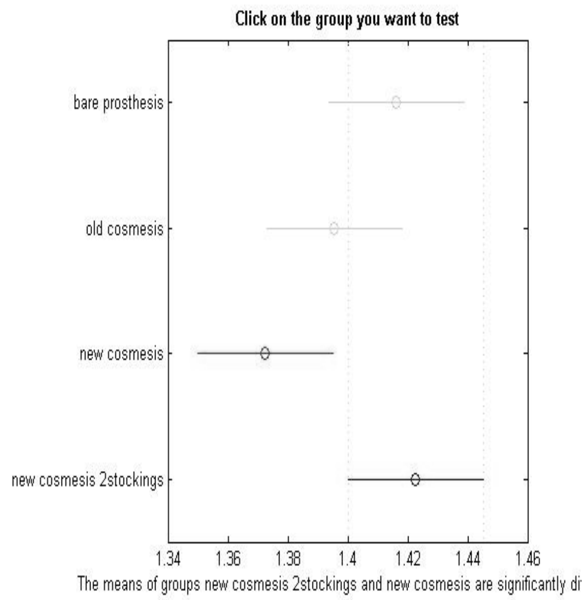
Source	SS	df	MS	F	Prob>F
Groups	0.01396	3	0.00465	3.64	0.0246
Error	0.03574	28	0.00128		
Total	0.0497	31			





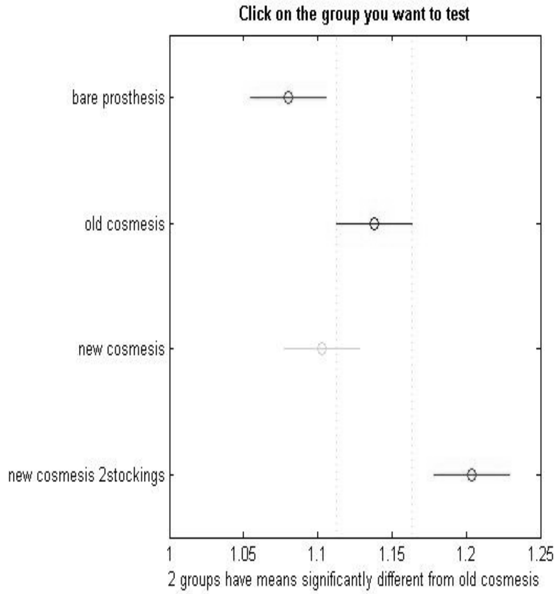
ANOVA test results of stride length of subject 1.

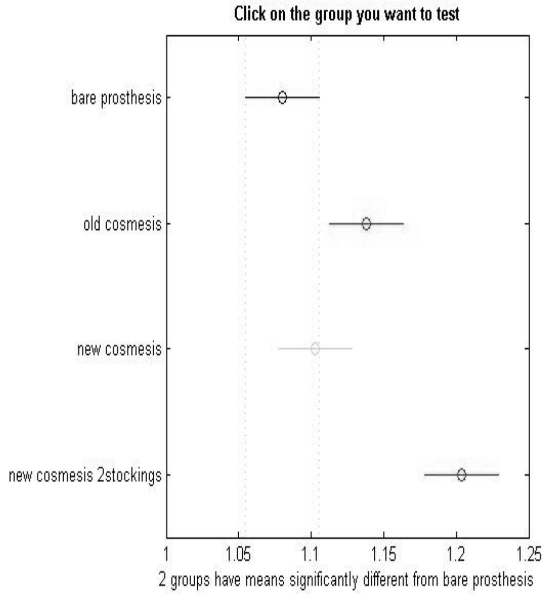

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.34
1.36
1.38
1.4
1.42
1.44
1.46
The means of groups new cosmesis and new cosmesis 2stockings are significantly di


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.34
1.36
1.38
1.4
1.42
1.44
1.46
The means of groups new cosmesis 2stockings and new cosmesis are significantly di

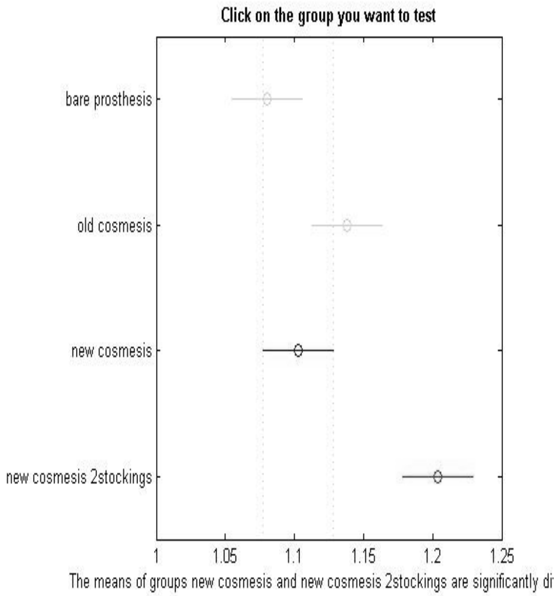
ANOVA Table

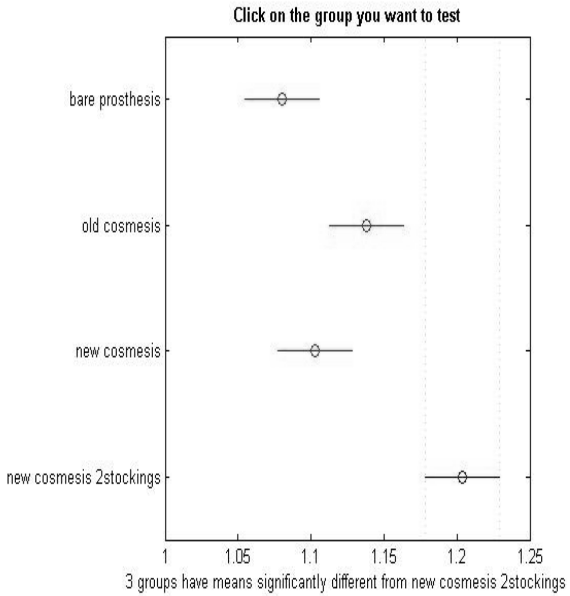
Source	SS	df	MS	F	Prob>F
Groups	0.07842	3	0.02614	16.34	1.2827e-06
Error	0.05119	32	0.0016		
Total	0.12961	35			





ANOVA tests results of velocity of subject 1.

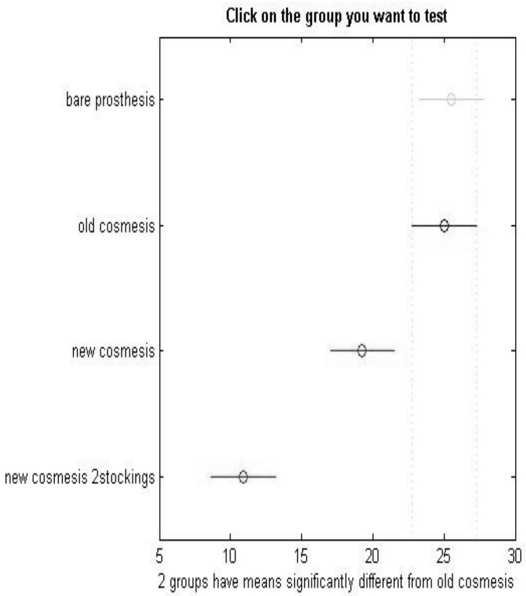

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1
1.05
1.1
1.15
1.2
1.25
The means of groups new cosmesis and new cosmesis 2stockings are significantly di

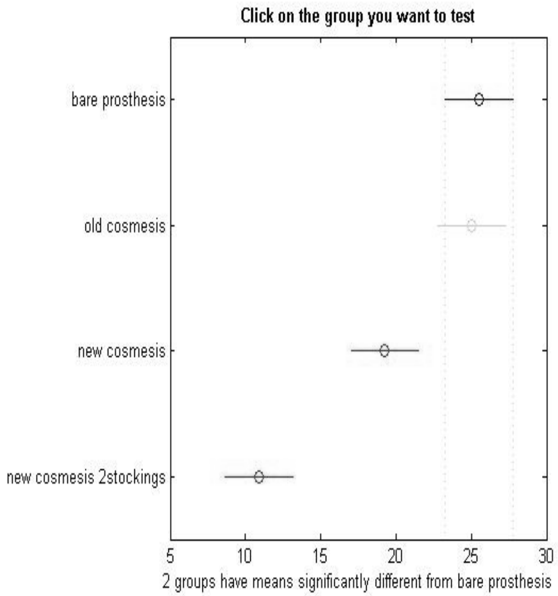

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1
1.05
1.1
1.15
1.2
1.25
3 groups have means significantly different from new cosmesis 2stockings

ANOVA and multi-comparison test on Subject 2, knee flexion range, step length, stride length and velocity.

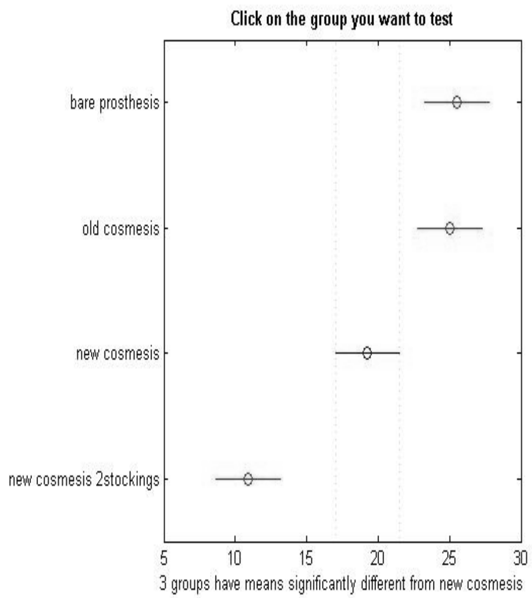
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	972.94	3	324.313	34.81	6.64615e-09
Error	223.6	24	9.316		
Total	1196.53	27			

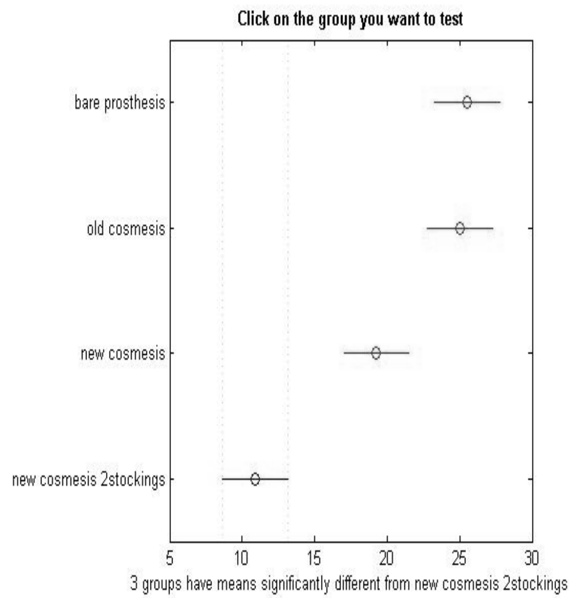




ANOVA analysis results on knee flexion angle range.



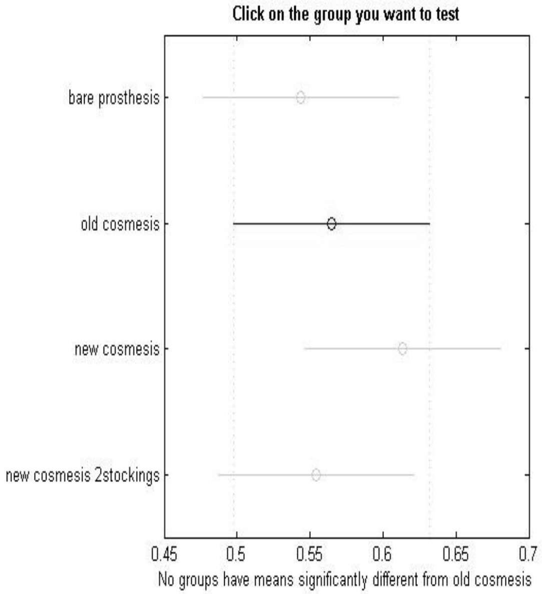
Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
5
10
15
20
25
30
3 groups have means significantly different from new cosmesis

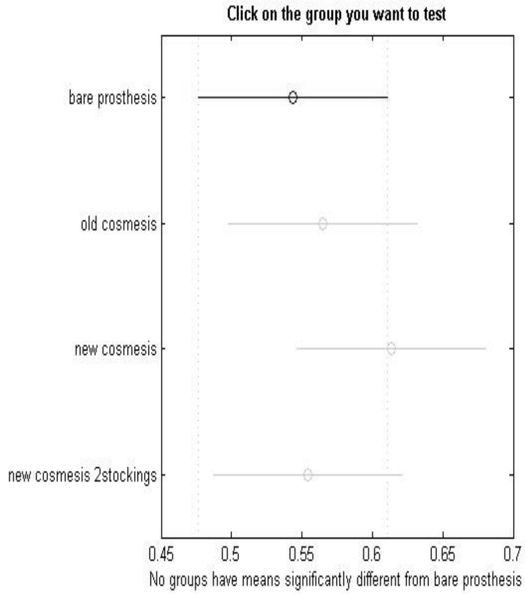


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
5
10
15
20
25
30
3 groups have means significantly different from new cosmesis 2stockings

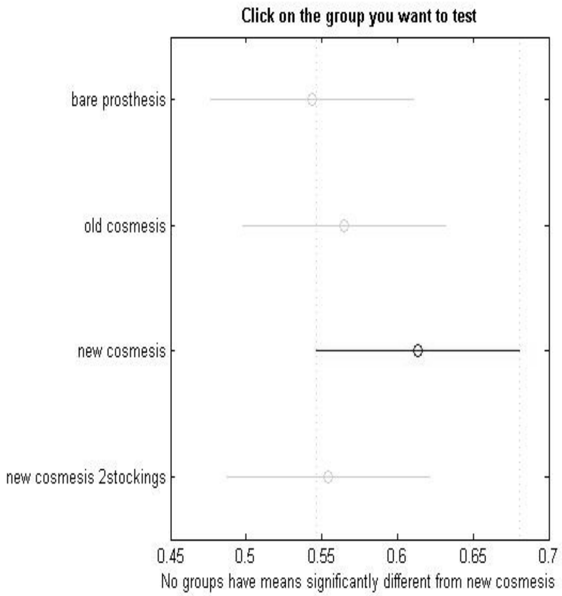
ANOVA Table

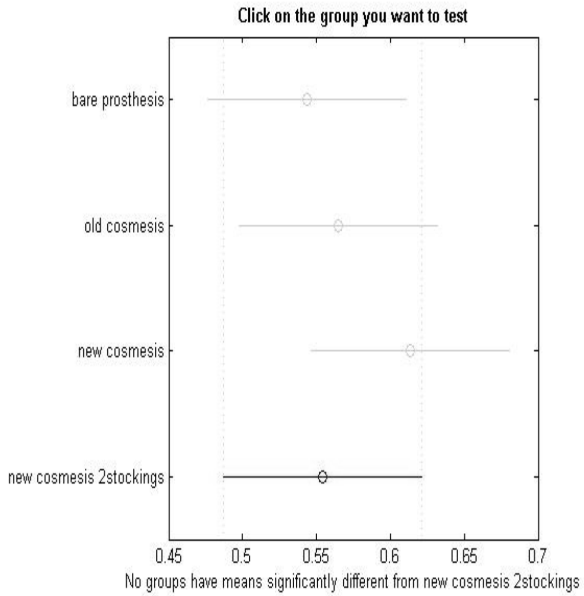
Source	SS	df	MS	F	Prob>F
Groups	0.02296	3	0.00765	0.79	0.5109
Error	0.27204	28	0.00972		
Total	0.295	31			





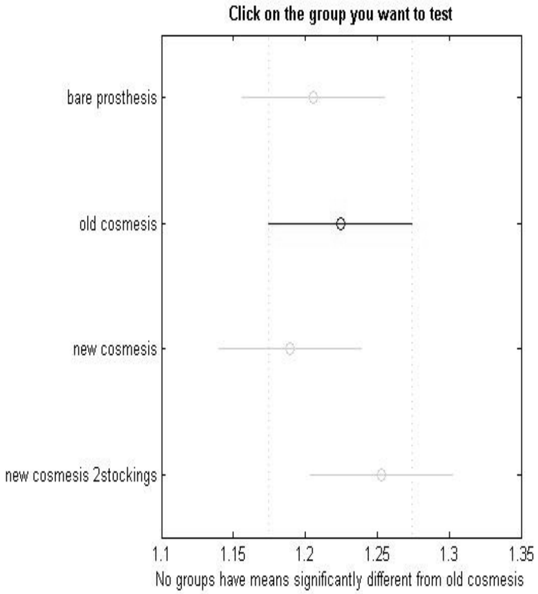
ANOVA test results of step length of subject 2.

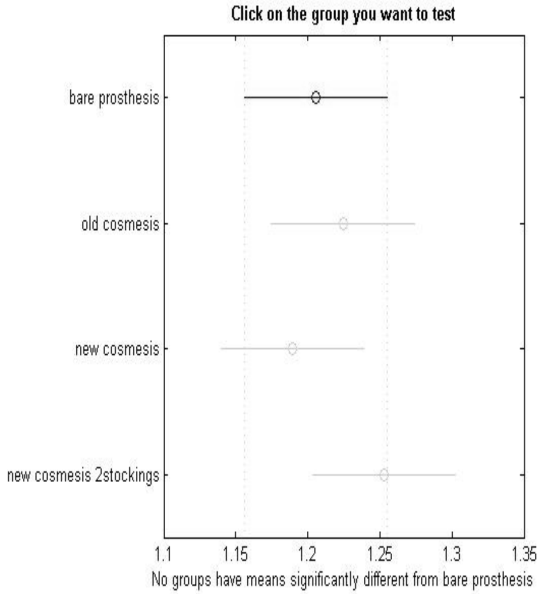

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.45
0.5
0.55
0.6
0.65
0.7
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.45
0.5
0.55
0.6
0.65
0.7
No groups have means significantly different from new cosmesis 2stockings

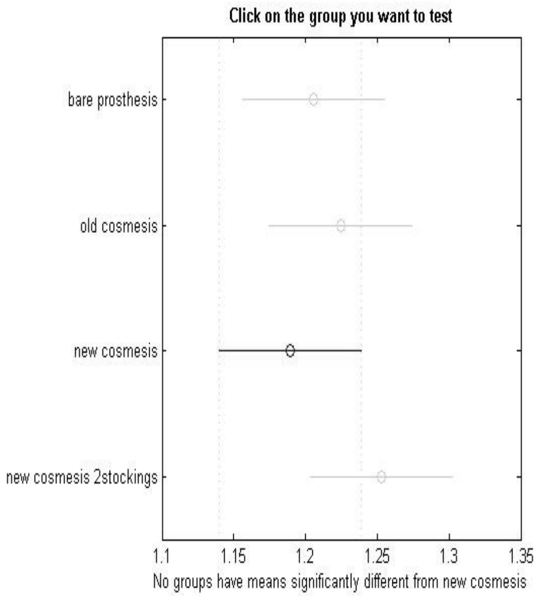
ANOVA Table

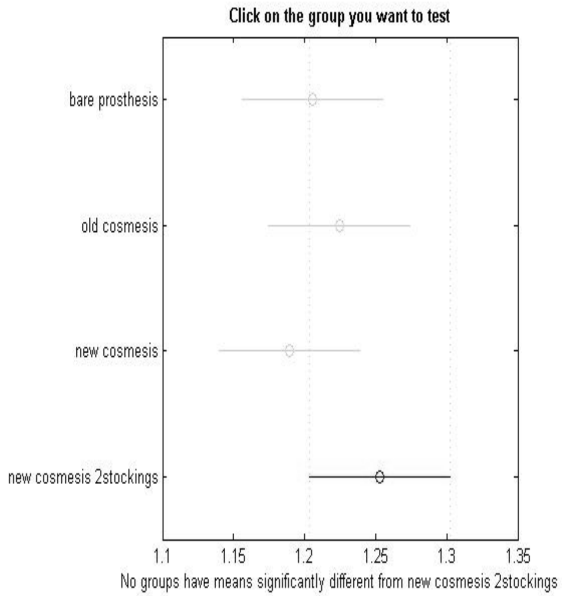
Source	SS	df	MS	F	Prob>F
Groups	0.01796	3	0.00599	1.14	0.3513
Error	0.14745	28	0.00527		
Total	0.16541	31			



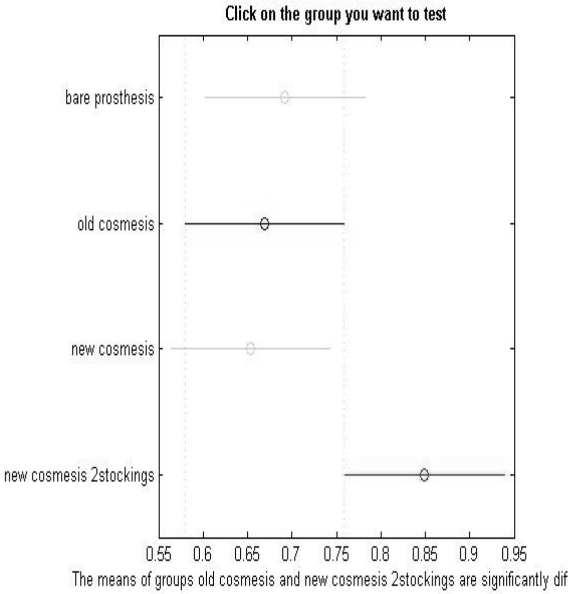


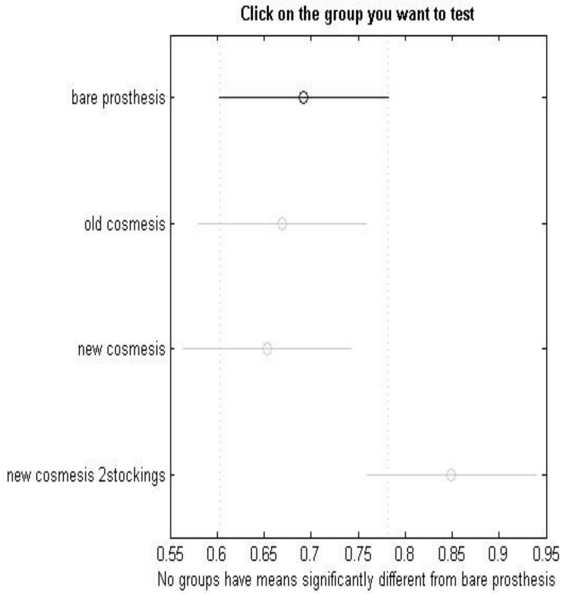
ANOVA test results of stride length of subject 2.


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.1
1.15
1.2
1.25
1.3
1.35
No groups have means significantly different from new cosmesis

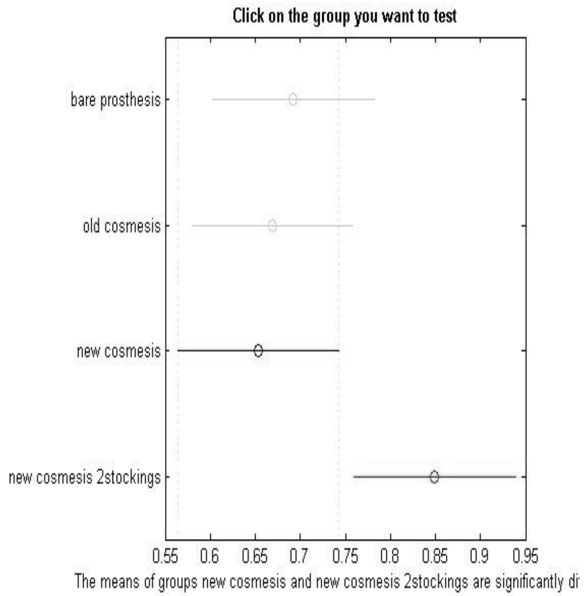

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.1
1.15
1.2
1.25
1.3
1.35
No groups have means significantly different from new cosmesis 2stockings

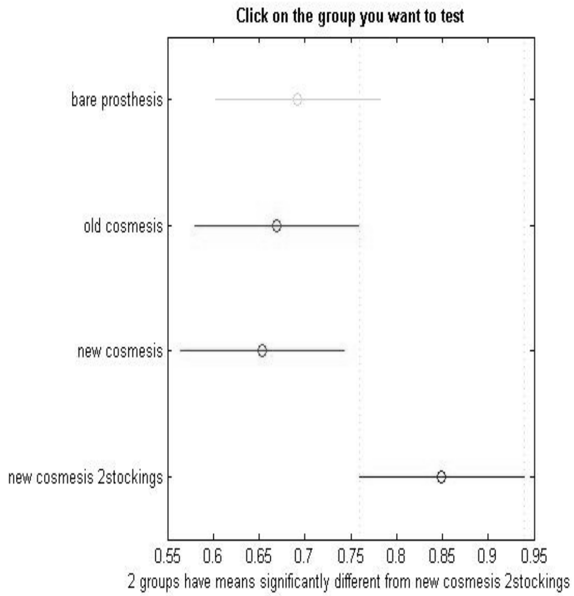
Source	SS	df	MS	F	Prob>F
Groups	0.19549	3	0.06516	3.78	0.0215
Error	0.48276	28	0.01724		
Total	0.67825	31			





ANOVA tests results of velocity of subject 2.


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.55 0.6 0.65 0.7 0.75 0.8 0.85 0.9 0.95
The means of groups new cosmesis and new cosmesis 2stockings are significantly di

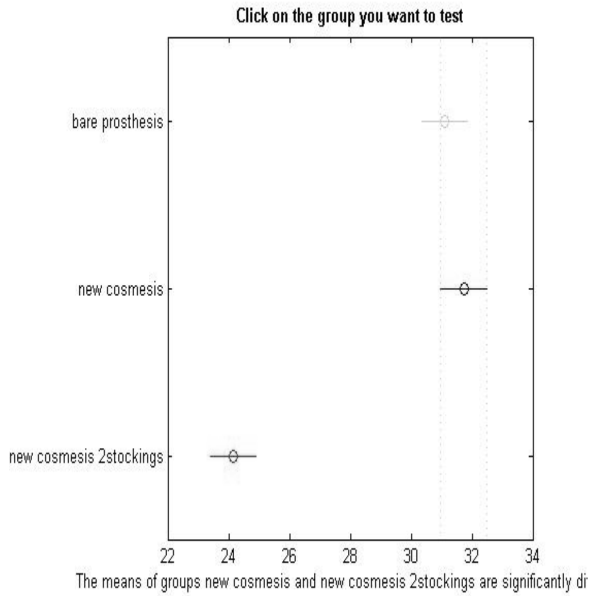

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.55 0.6 0.65 0.7 0.75 0.8 0.85 0.9 0.95
2 groups have means significantly different from new cosmesis 2stockings

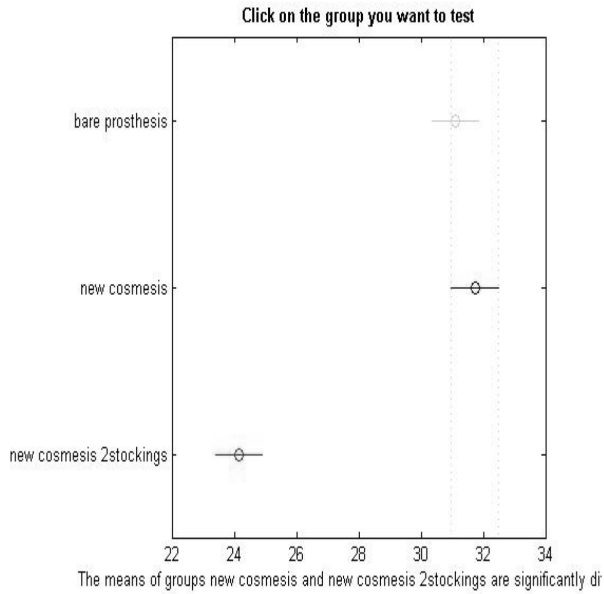
ANOVA and multi-comparison test on Subject 3, knee flexion range, step length, stride length and velocity.

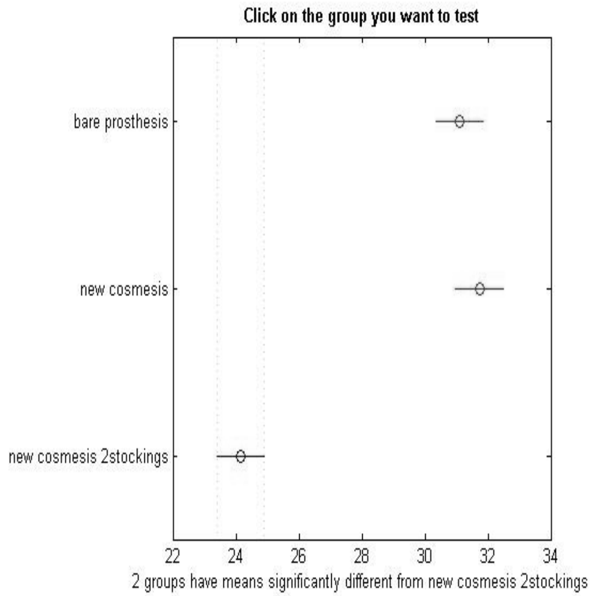
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	353.913	2	176.956	95.93	5.38349e-13
Error	49.804	27	1.845		
Total	403.717	29			

ANOVA analysis results on knee flexion angle range.

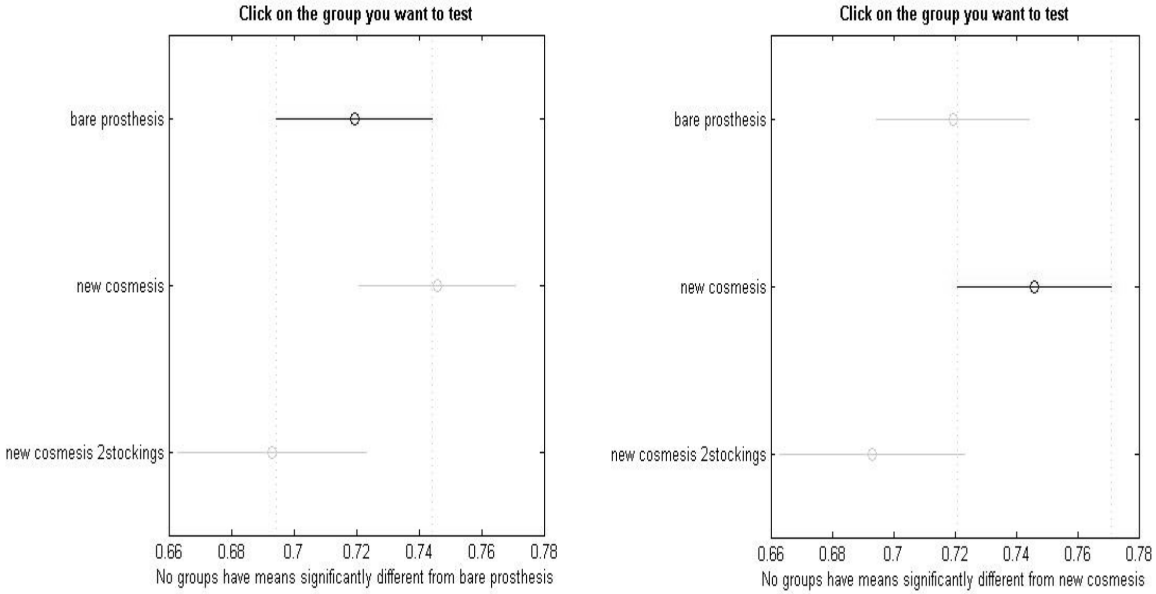




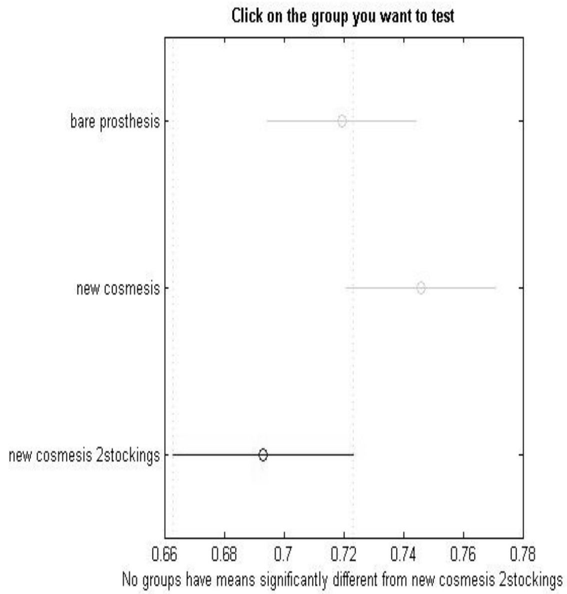

Click on the group you want to test
bare prosthesis
new cosmesis
new cosmesis 2stockings
22
24
26
28
30
32
34
2 groups have means significantly different from new cosmesis 2stockings

ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.01162	2	0.00581	2.9	0.0744
Error	0.04808	24	0.002		
Total	0.05971	26			

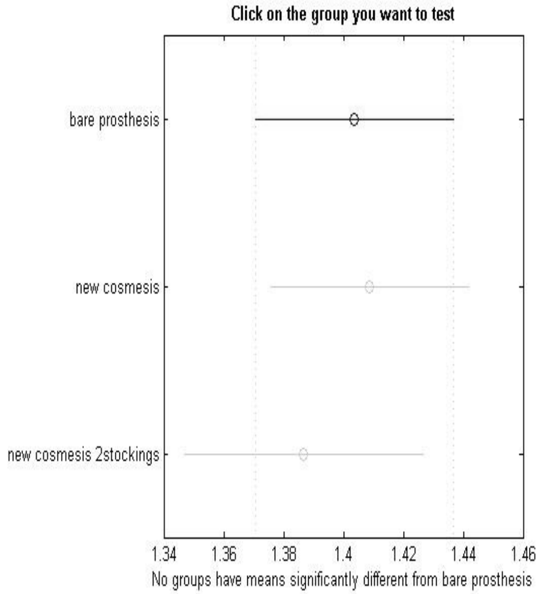


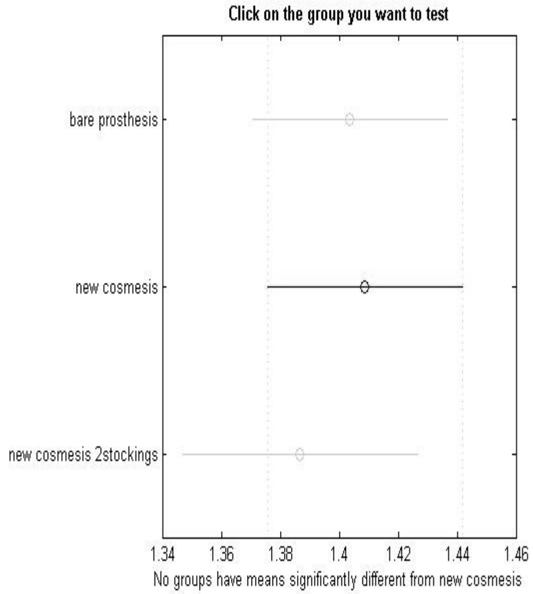
ANOVA test results of step length of subject 3.


Click on the group you want to test
bare prosthesis
new cosmesis
new cosmesis 2stockings
0.66
0.68
0.7
0.72
0.74
0.76
0.78
No groups have means significantly different from new cosmesis 2stockings

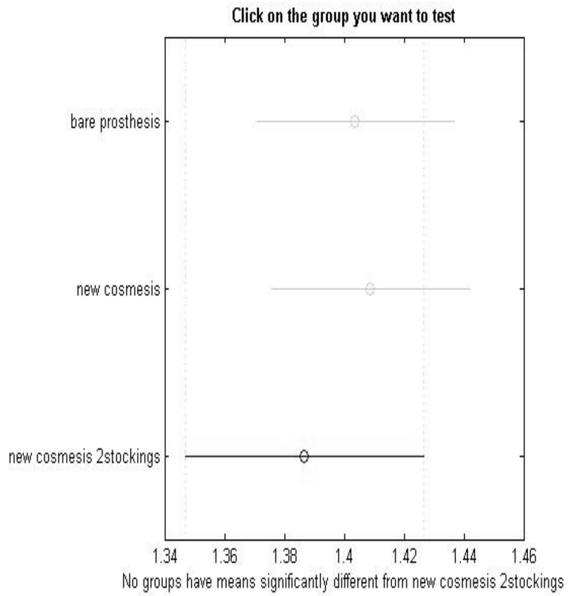
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.0021	2	0.00105	0.3	0.7428
Error	0.08385	24	0.00349		
Total	0.08596	26			



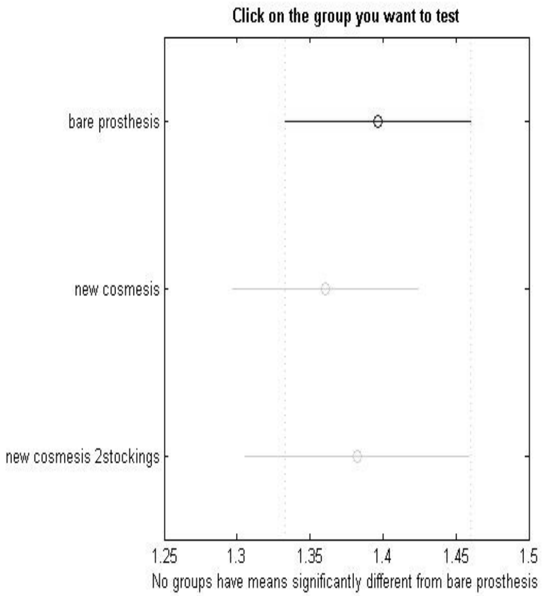


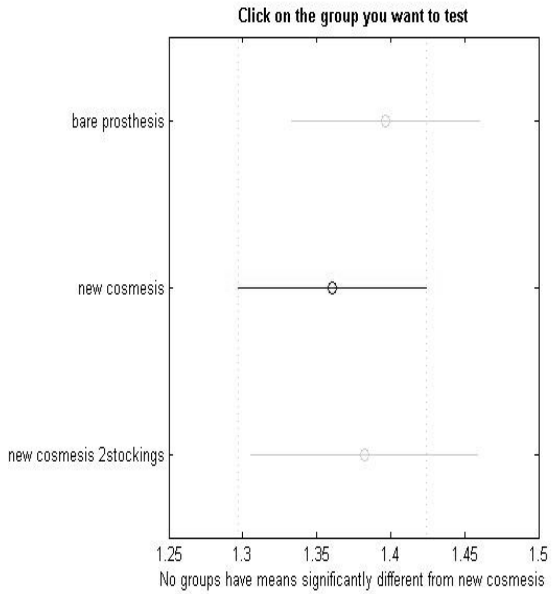
ANOVA test results of stride length of subject 3.


Click on the group you want to test
bare prosthesis
new cosmesis
new cosmesis 2stockings
1.34
1.36
1.38
1.4
1.42
1.44
1.46
No groups have means significantly different from new cosmesis 2stockings

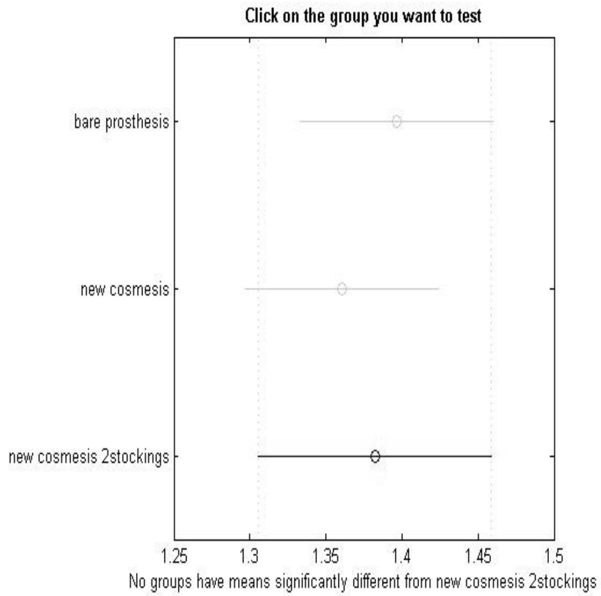
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.00642	2	0.00321	0.25	0.7828
Error	0.31158	24	0.01298		
Total	0.318	26			





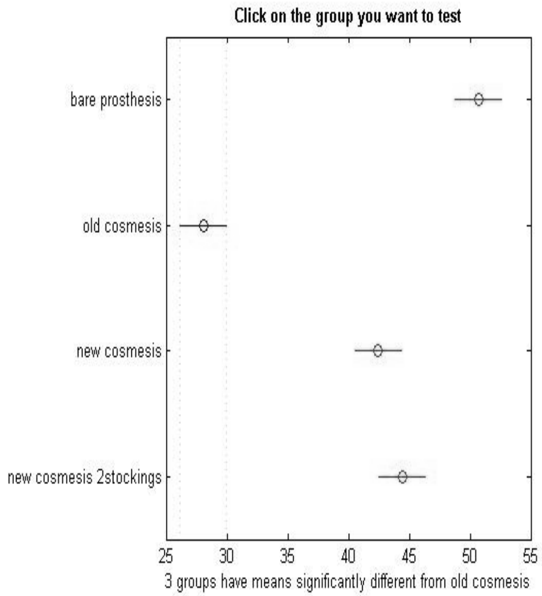
ANOVA tests results of velocity of subject 3.

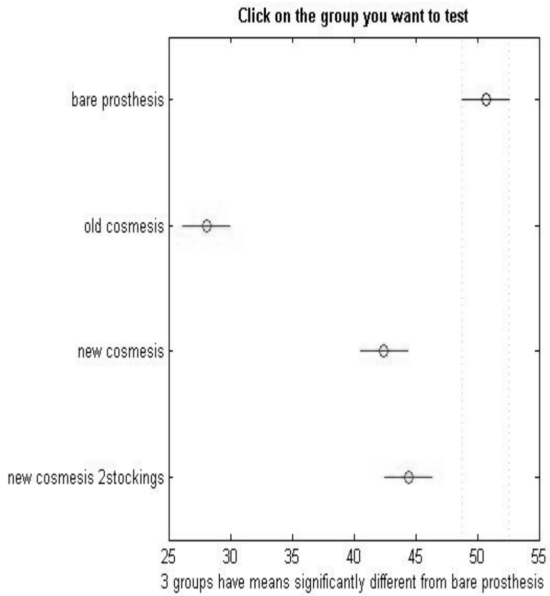

Click on the group you want to test
bare prosthesis
new cosmesis
new cosmesis 2stockings
1.25
1.3
1.35
1.4
1.45
1.5
No groups have means significantly different from new cosmesis 2stockings

ANOVA and multi-comparison test on Subject 4, knee flexion range, step length, stride length and velocity.

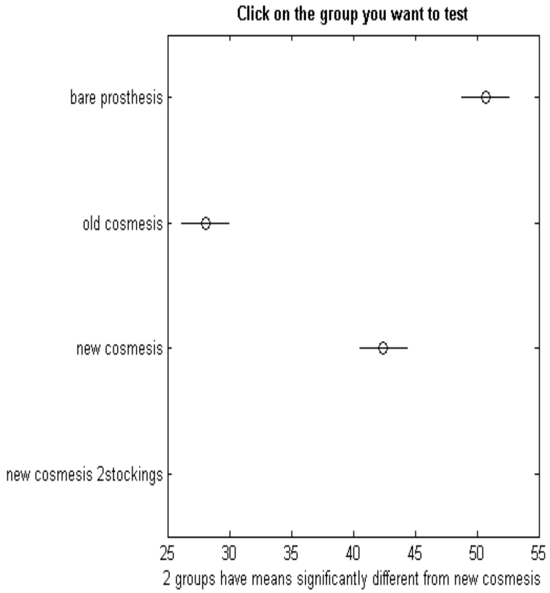
ANOVA Table

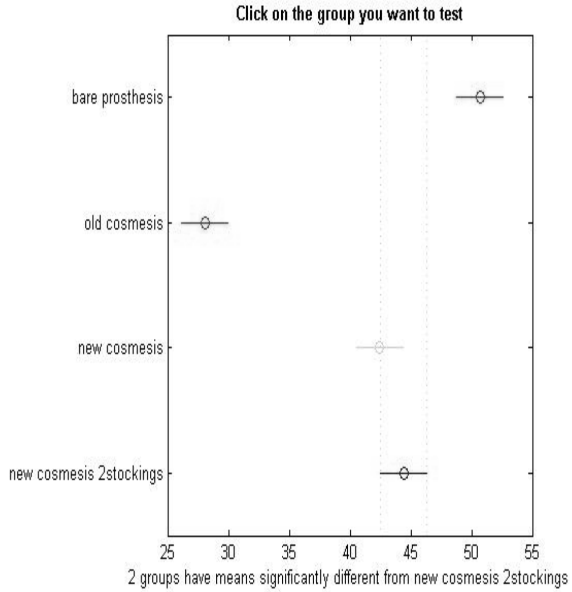
Source	SS	df	MS	F	Prob>F
Groups	2188.62	3	729.542	93.66	1.0444e-14
Error	218.1	28	7.789		
Total	2406.72	31			





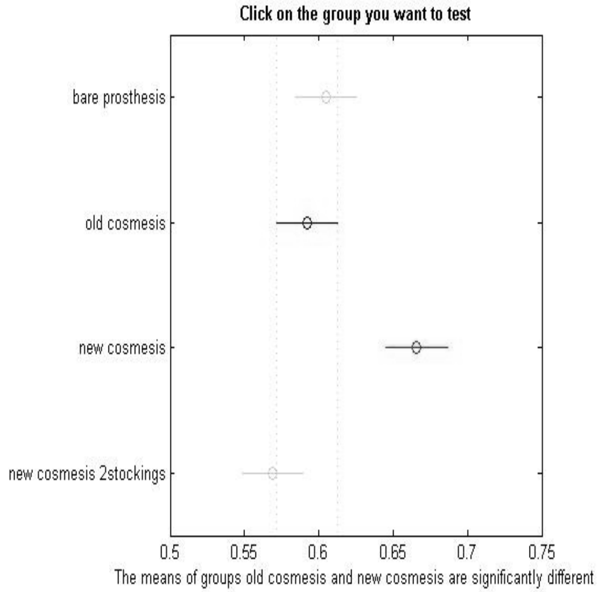
ANOVA analysis results on knee flexion angle range.

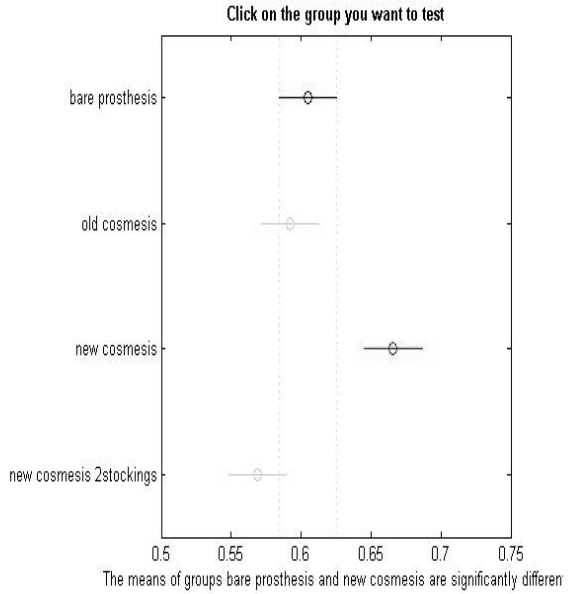

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
25 30 35 40 45 50 55
2 groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
25 30 35 40 45 50 55
2 groups have means significantly different from new cosmesis 2stockings

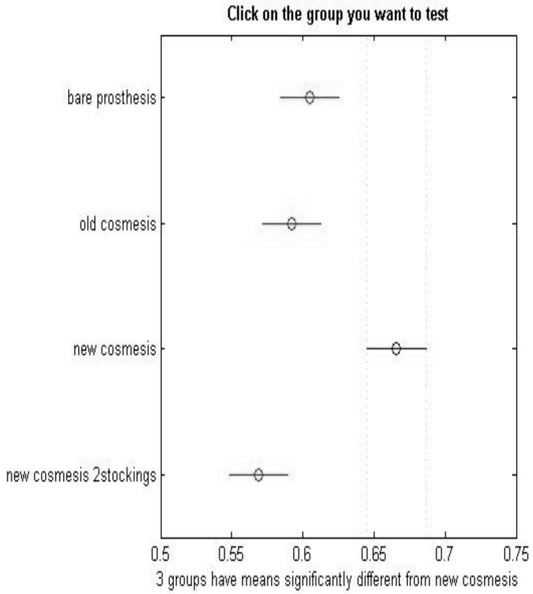
ANOVA Table

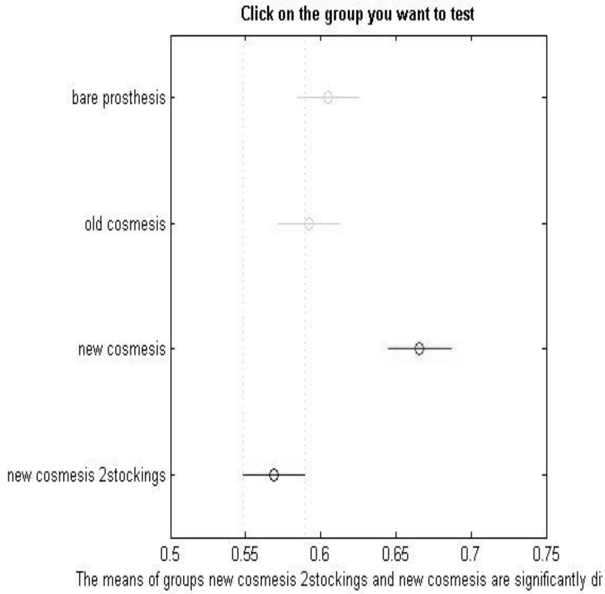
Source	SS	df	MS	F	Prob>F
Groups	0.036	3	0.012	15.37	8.61194e-06
Error	0.01874	24	0.00078		
Total	0.05474	27			





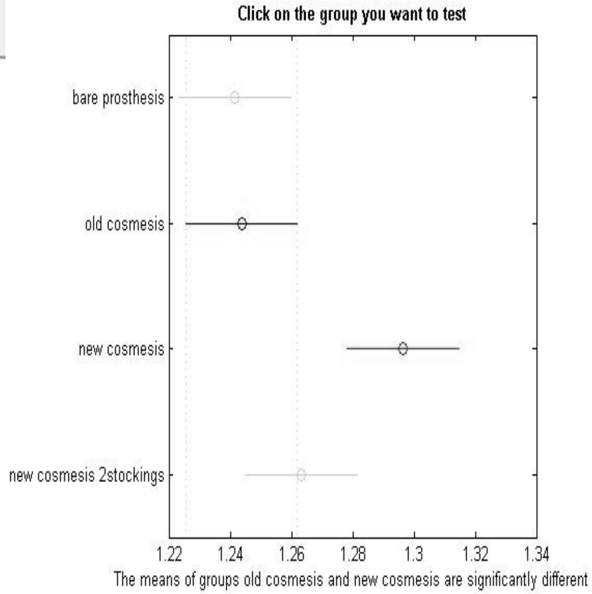
ANOVA test results of step length of subject 4.

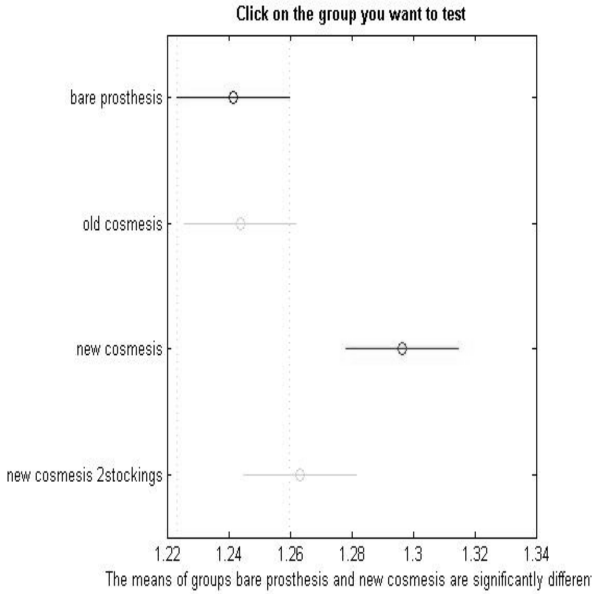

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.5
0.55
0.6
0.65
0.7
0.75
3 groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.5
0.55
0.6
0.65
0.7
0.75
The means of groups new cosmesis 2stockings and new cosmesis are significantly di

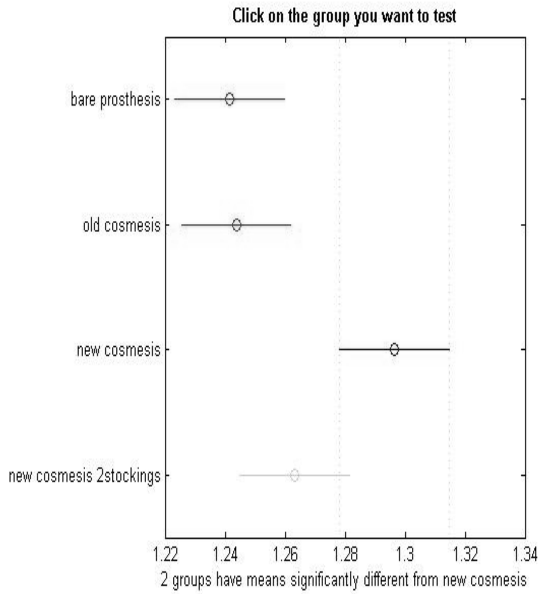
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.01355	3	0.00452	7.36	0.0012
Error	0.01472	24	0.00061		
Total	0.02827	27			

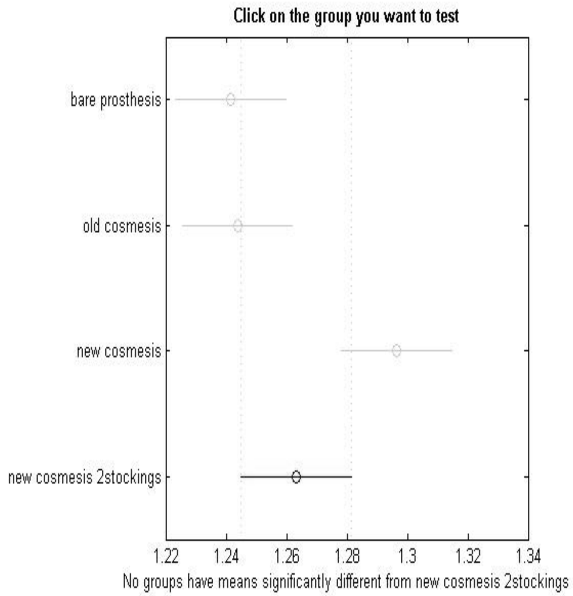




ANOVA test results of stride length of subject 4.



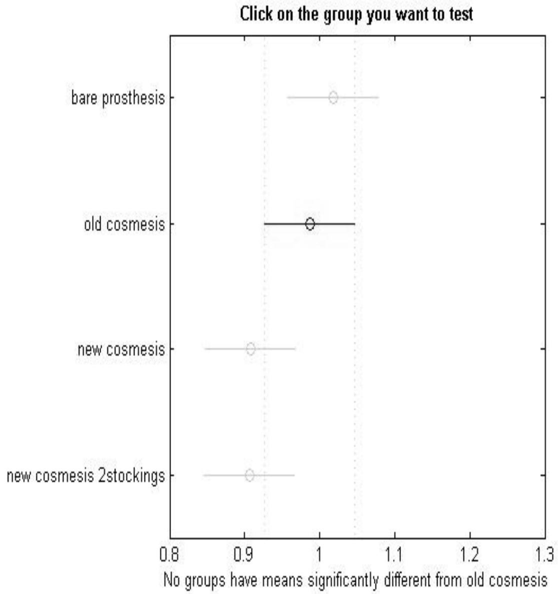
Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.22
1.24
1.26
1.28
1.3
1.32
1.34
2 groups have means significantly different from new cosmesis

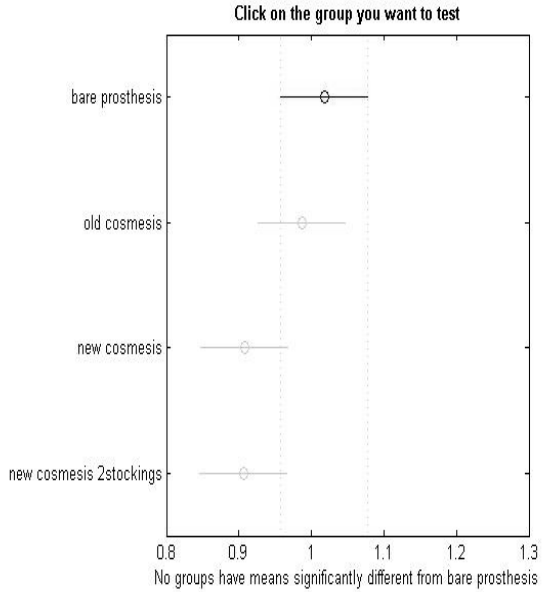


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.22
1.24
1.26
1.28
1.3
1.32
1.34
No groups have means significantly different from new cosmesis 2stockings

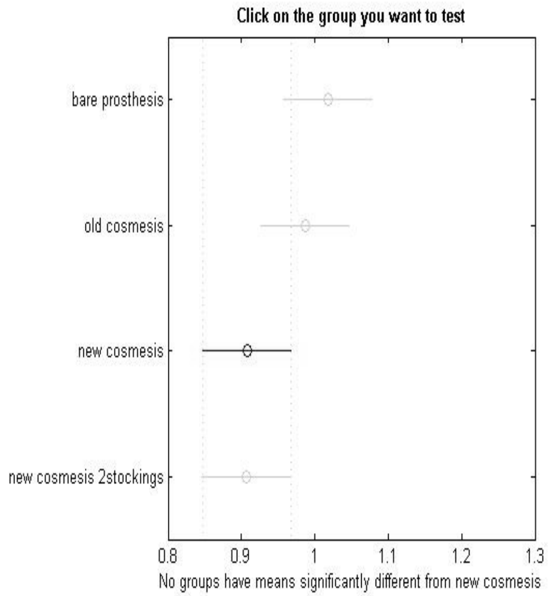
ANOVA Table

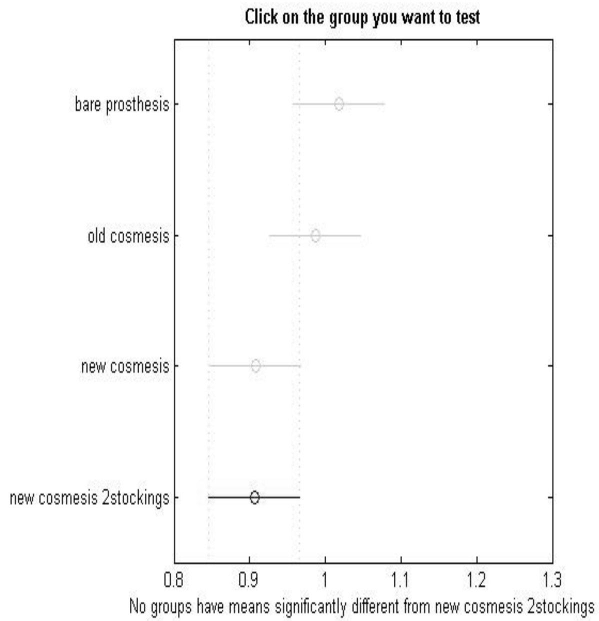
Source	SS	df	MS	F	Prob>F
Groups	0.06744	3	0.02248	3.4	0.0341
Error	0.15875	24	0.00661		
Total	0.22619	27			





ANOVA tests results of velocity of subject 4.

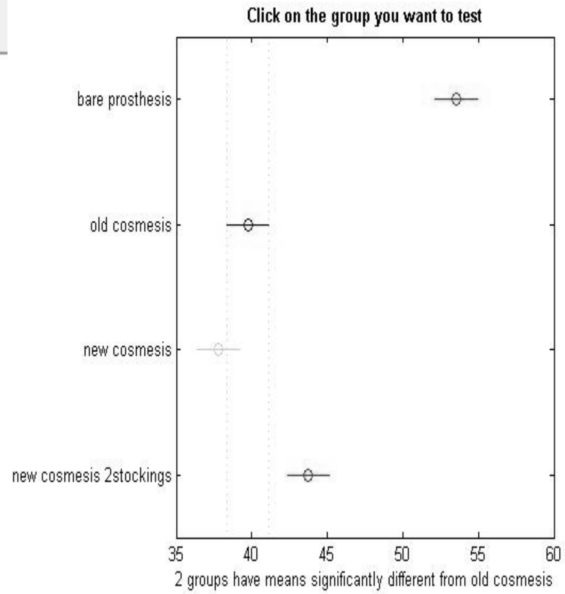

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.8
0.9
1
1.1
1.2
1.3
No groups have means significantly different from new cosmesis

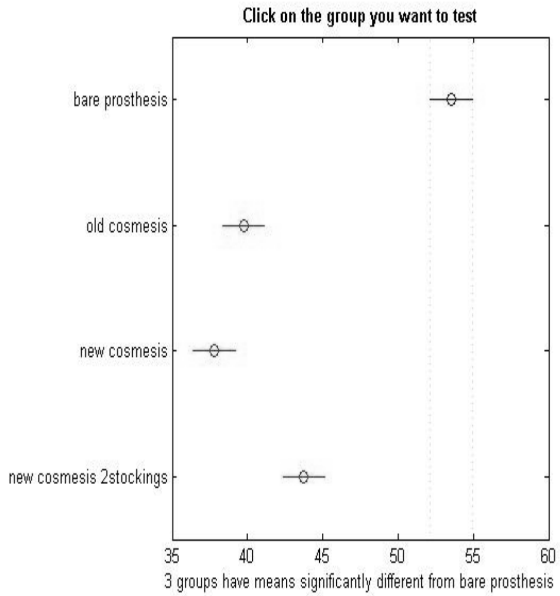

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.8
0.9
1
1.1
1.2
1.3
No groups have means significantly different from new cosmesis 2stockings

ANOVA and multi-comparison test on Subject 5, knee flexion range, step length, stride length and velocity.

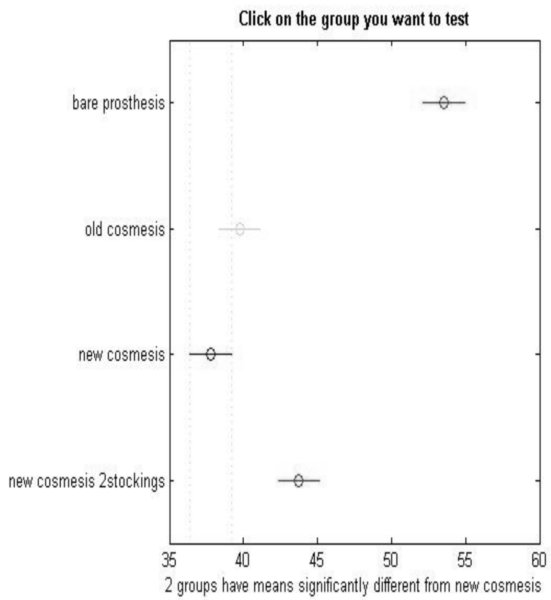
ANOVA Table

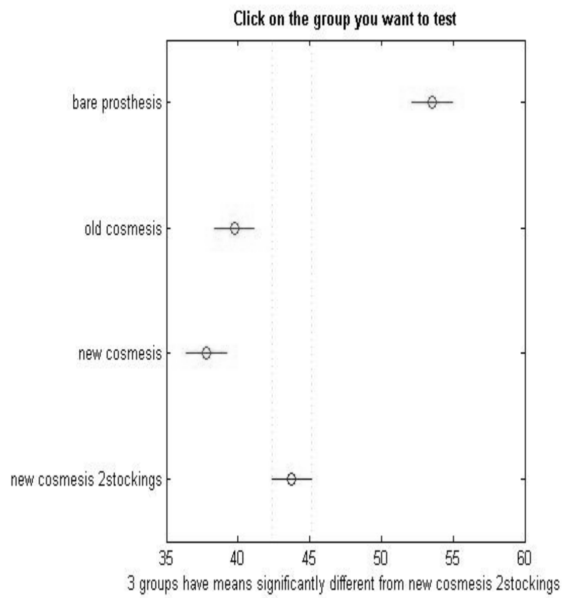
Source	SS	df	MS	F	Prob>F
Groups	1325.89	3	441.964	93.03	6.90093e-16
Error	152.03	32	4.751		
Total	1477.93	35			





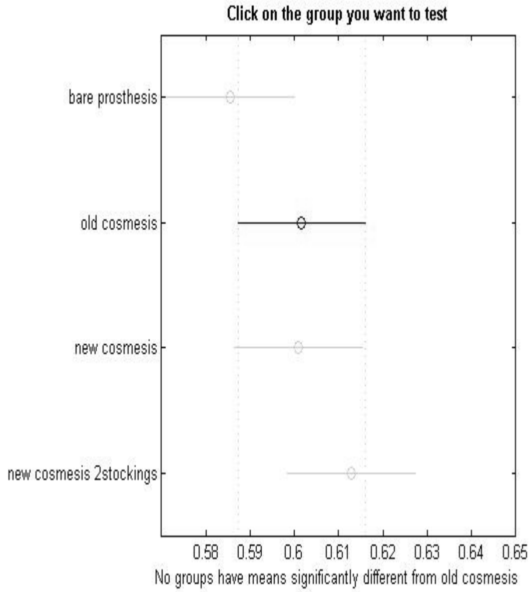
ANOVA analysis results on knee flexion angle range.

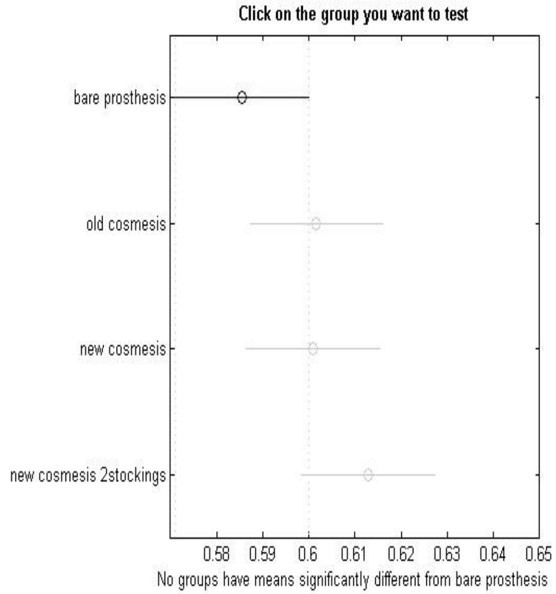

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
35
40
45
50
55
60
2 groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
35
40
45
50
55
60
3 groups have means significantly different from new cosmesis 2stockings

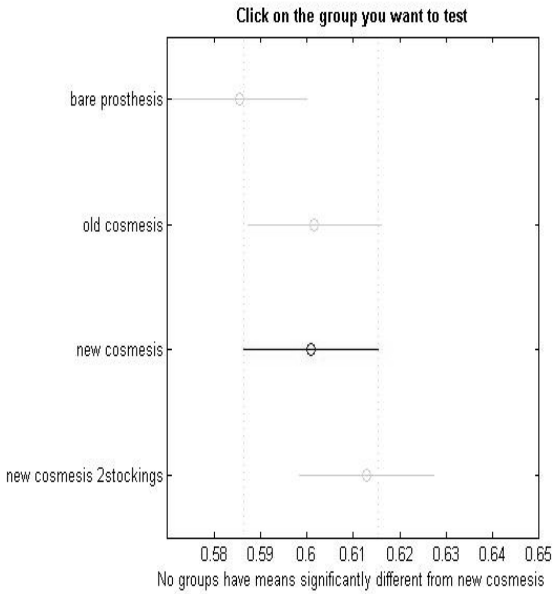
ANOVA Table

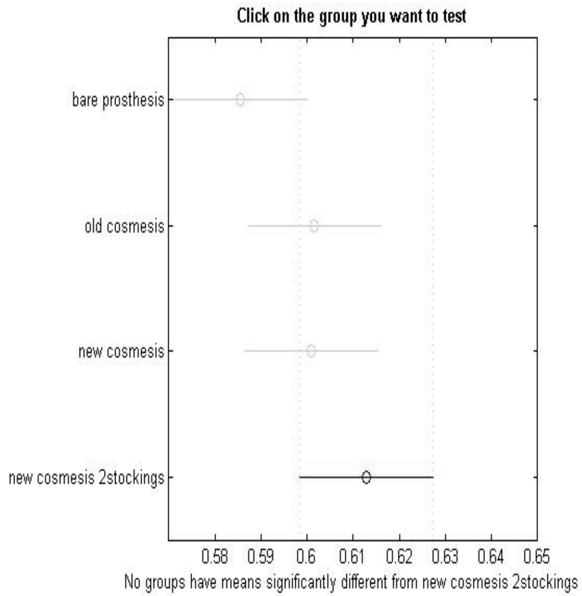
Source	SS	df	MS	F	Prob>F
Groups	0.00377	3	0.00126	2.19	0.1056
Error	0.02059	36	0.00057		
Total	0.02436	39			





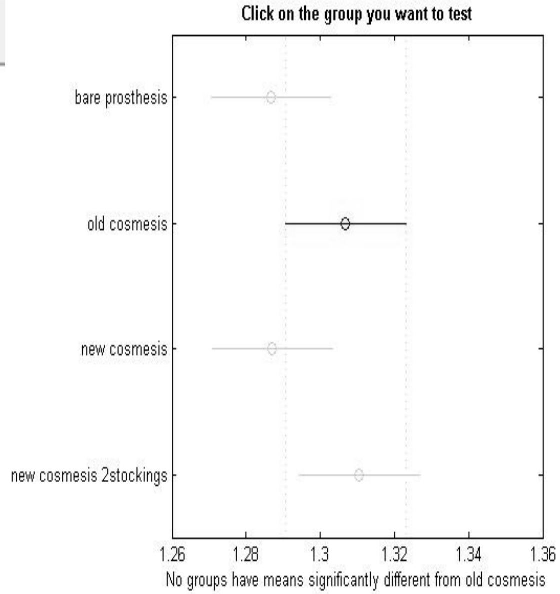
ANOVA test results of step length of subject 5.

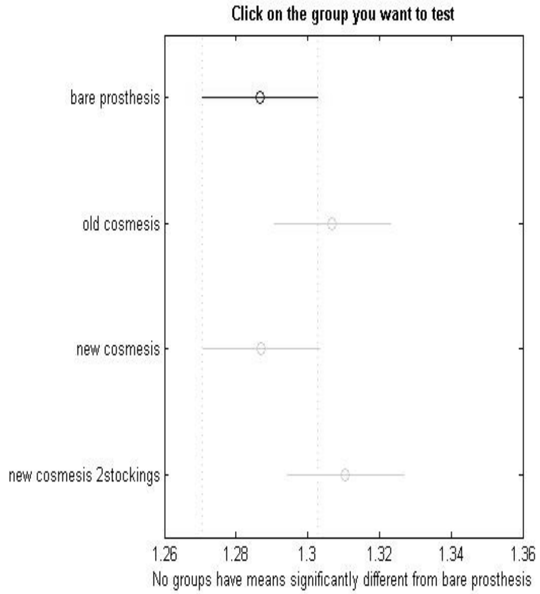

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.58 0.59 0.6 0.61 0.62 0.63 0.64 0.65
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.58 0.59 0.6 0.61 0.62 0.63 0.64 0.65
No groups have means significantly different from new cosmesis 2stockings

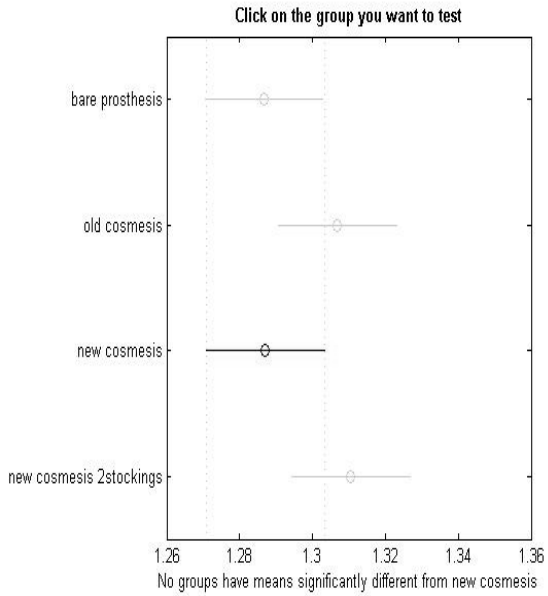
ANOVA Table

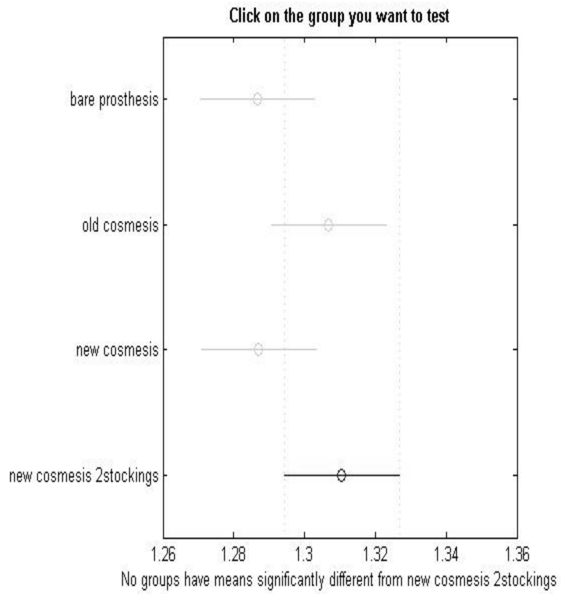
Source	SS	df	MS	F	Prob>F
Groups	0.00481	3	0.0016	2.23	0.1018
Error	0.02594	36	0.00072		
Total	0.03075	39			





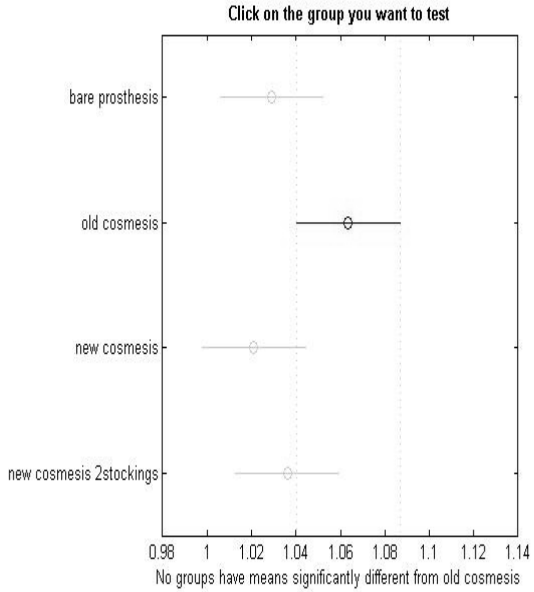
ANOVA test results of stride length of subject 5.

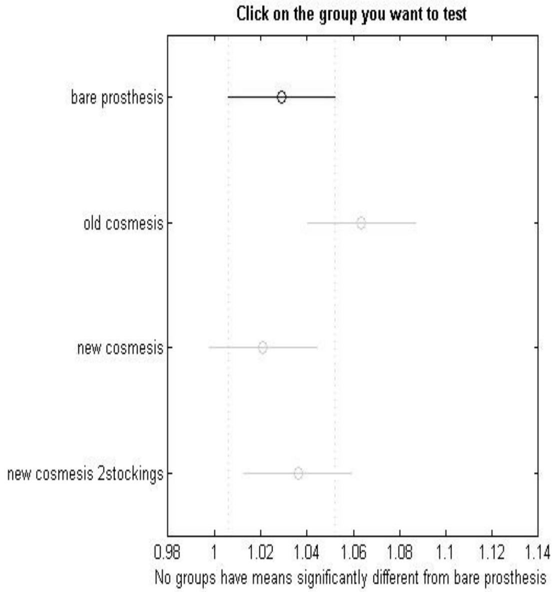

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.26
1.28
1.3
1.32
1.34
1.36
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.26
1.28
1.3
1.32
1.34
1.36
No groups have means significantly different from new cosmesis 2stockings

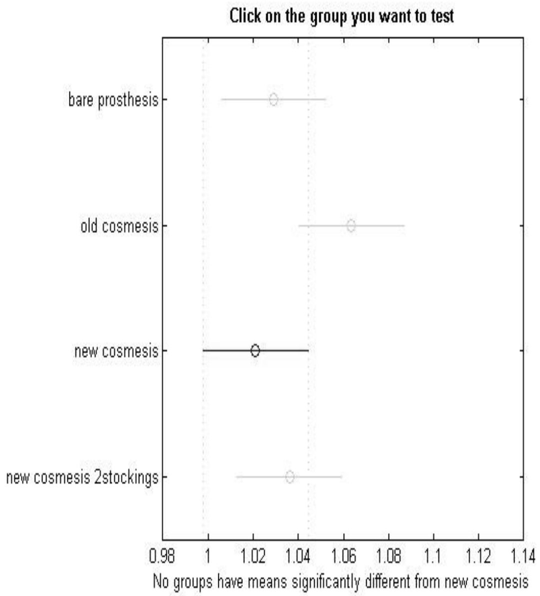
ANOVA Table

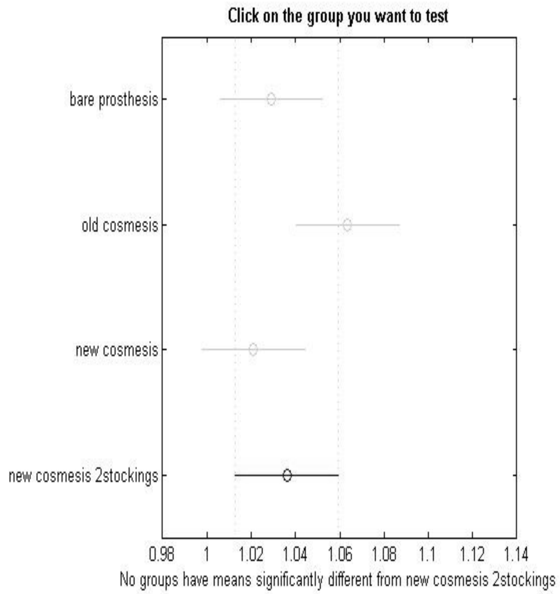
Source	SS	df	MS	F	Prob>F
Groups	0.01023	3	0.00341	2.29	0.095
Error	0.05361	36	0.00149		
Total	0.06384	39			





ANOVA tests results of velocity of subject 5.

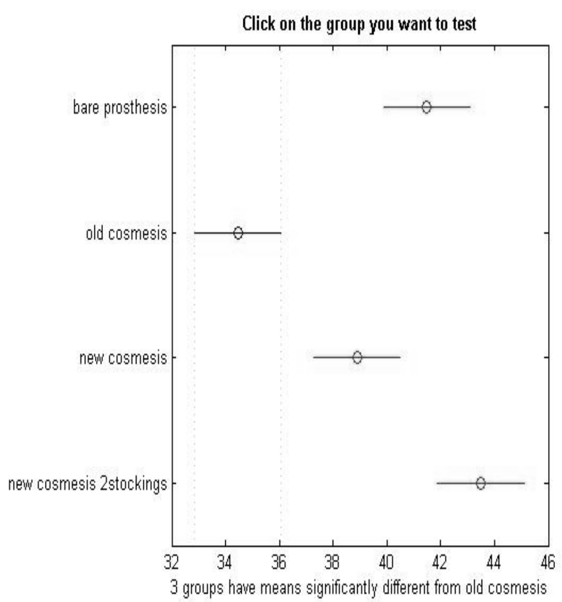

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.98
1
1.02
1.04
1.06
1.08
1.1
1.12
1.14
No groups have means significantly different from new cosmesis

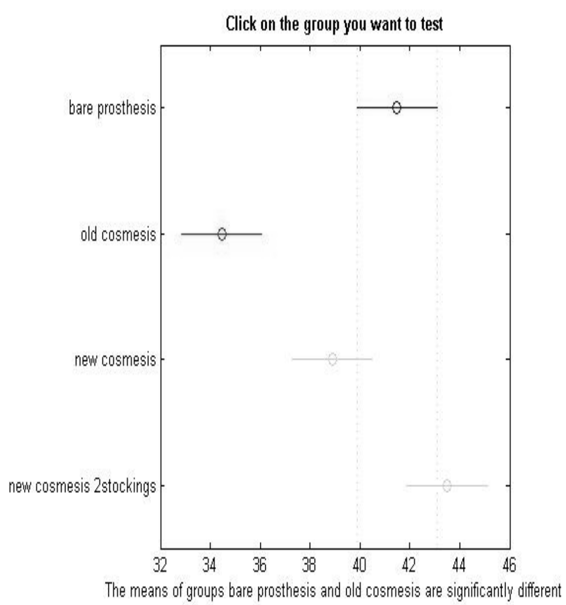

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.98
1
1.02
1.04
1.06
1.08
1.1
1.12
1.14
No groups have means significantly different from new cosmesis 2stockings

ANOVA and multi-comparison test on Subject 6, knee flexion range, step length, stride length and velocity.

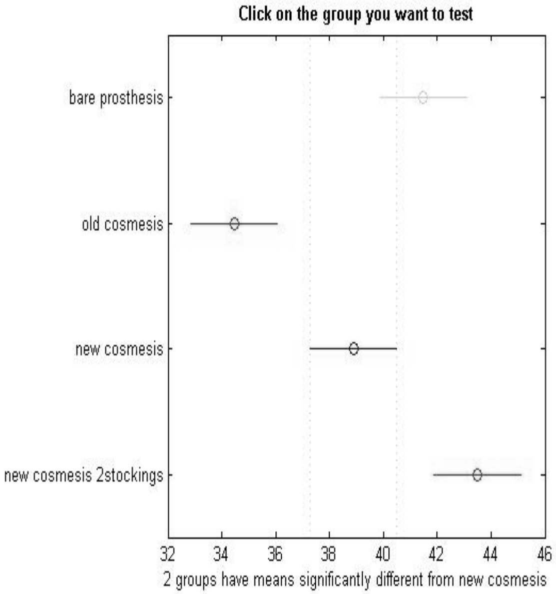
ANOVA Table

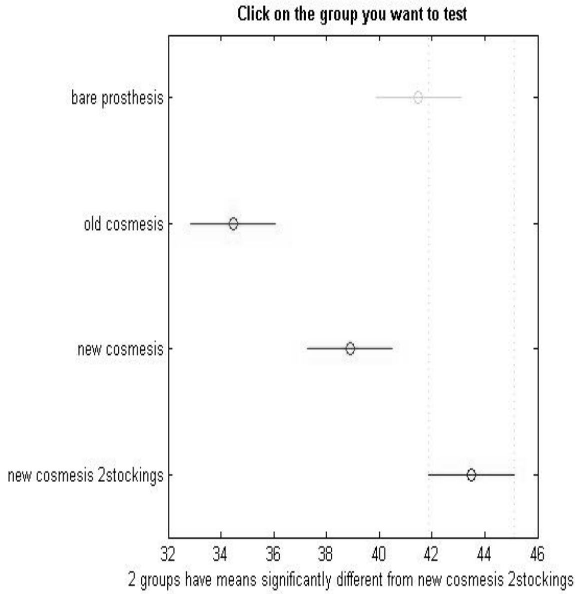
Source	SS	df	MS	F	Prob>F
Groups	409.986	3	136.662	21.67	7.5445e-08
Error	201.833	32	6.307		
Total	611.819	35			





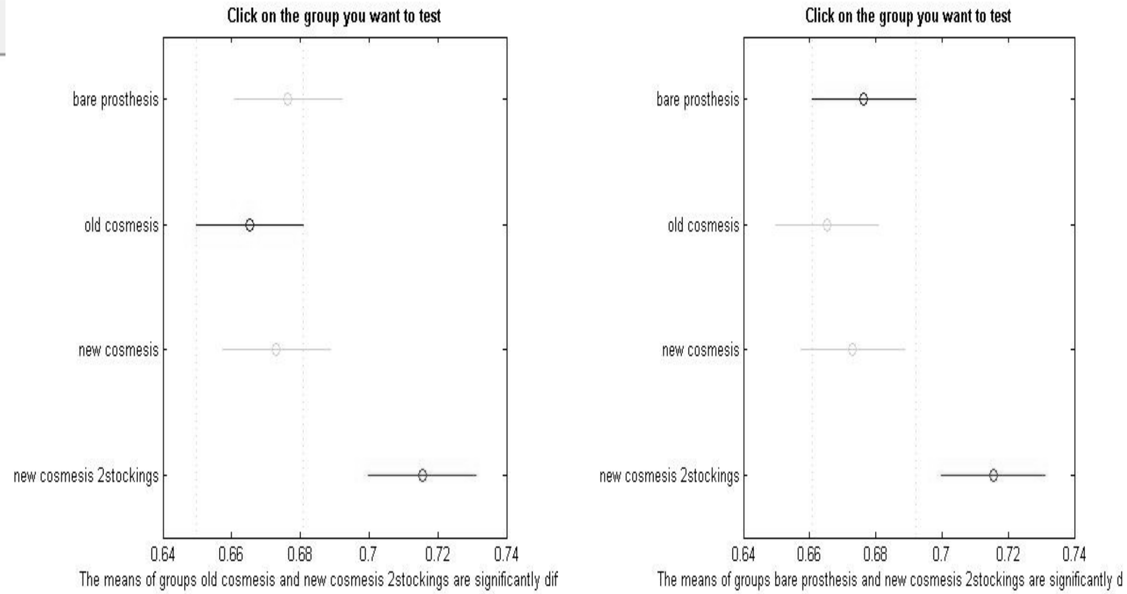
ANOVA analysis results on knee flexion angle range.


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
32
34
36
38
40
42
44
46
2 groups have means significantly different from new cosmesis

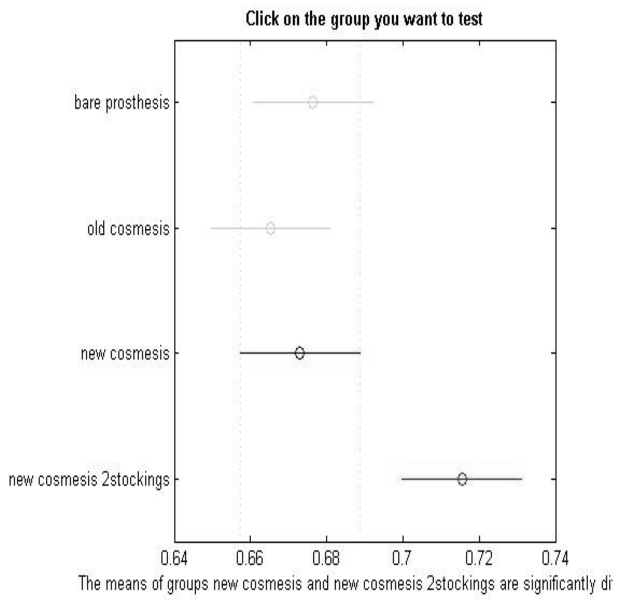

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
32
34
36
38
40
42
44
46
2 groups have means significantly different from new cosmesis 2stockings

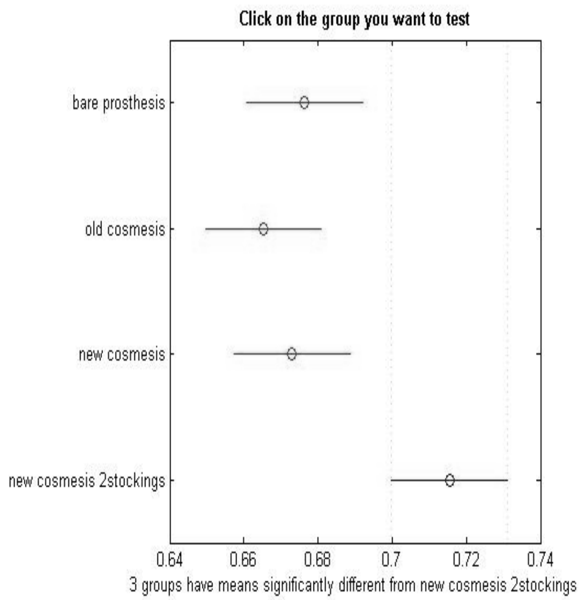
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.01658	3	0.00553	7.38	0.0005
Error	0.02994	40	0.00075		
Total	0.04652	43			



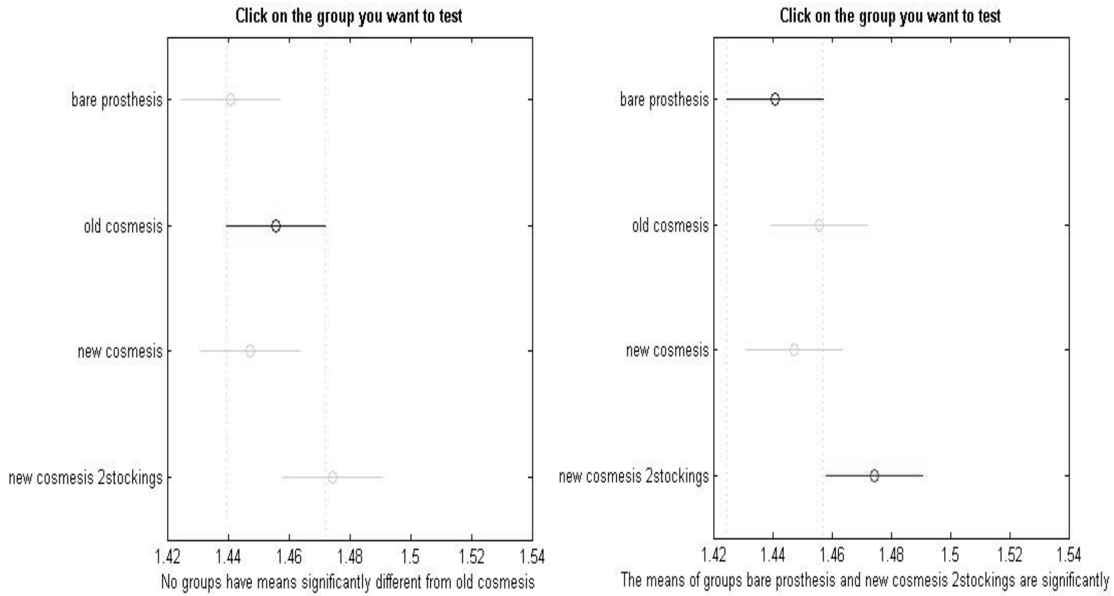
ANOVA test results of step length of subject 6.


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.64
0.66
0.68
0.7
0.72
0.74
The means of groups new cosmesis and new cosmesis 2stockings are significantly di

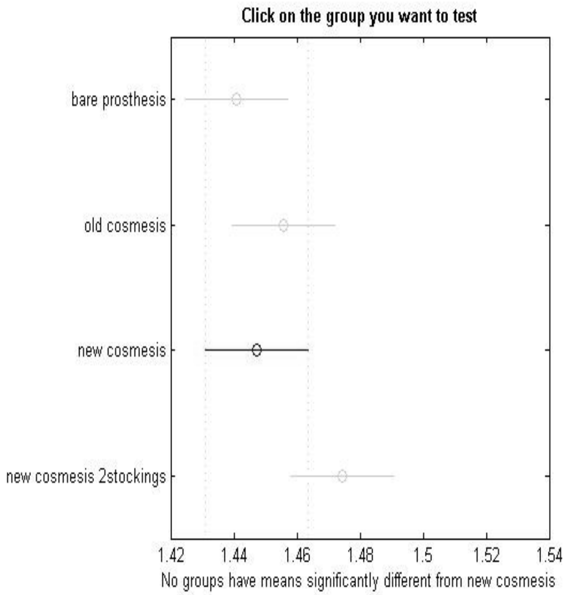

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
0.64
0.66
0.68
0.7
0.72
0.74
3 groups have means significantly different from new cosmesis 2stockings

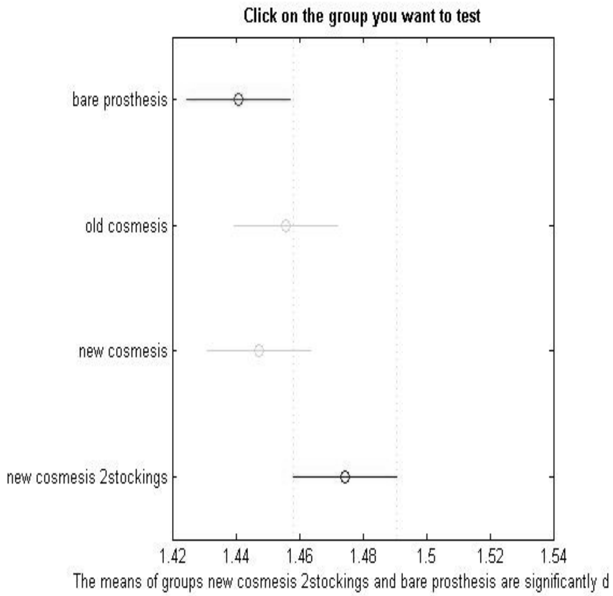
ANOVA Table

Source	SS	df	MS	F	Prob>F
Groups	0.00692	3	0.00231	2.84	0.05
Error	0.03252	40	0.00081		
Total	0.03944	43			



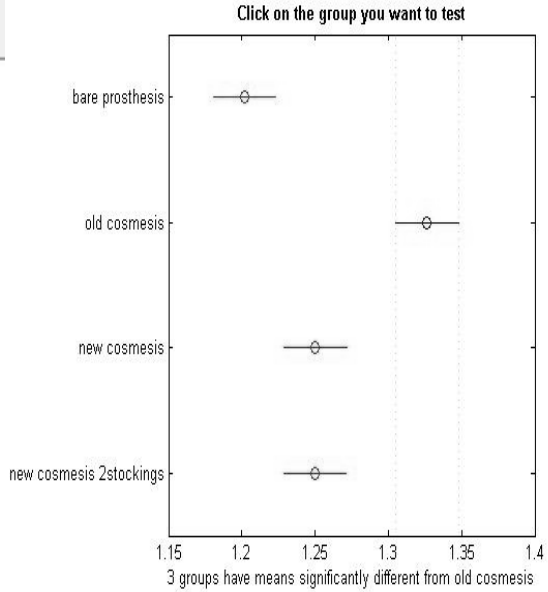
ANOVA test results of stride length of subject 6.

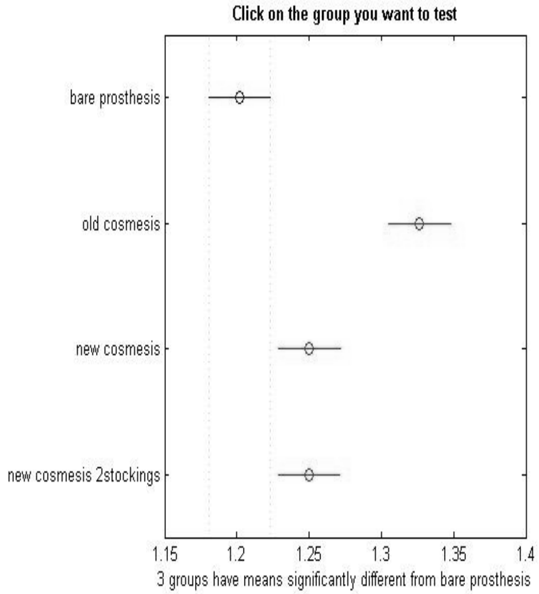

Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.42 1.44 1.46 1.48 1.5 1.52 1.54
No groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.42 1.44 1.46 1.48 1.5 1.52 1.54
The means of groups new cosmesis 2stockings and bare prosthesis are significantly d

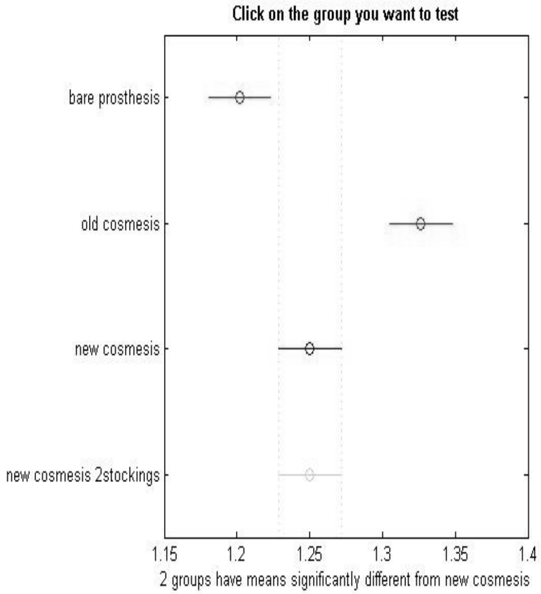
ANOVA Table

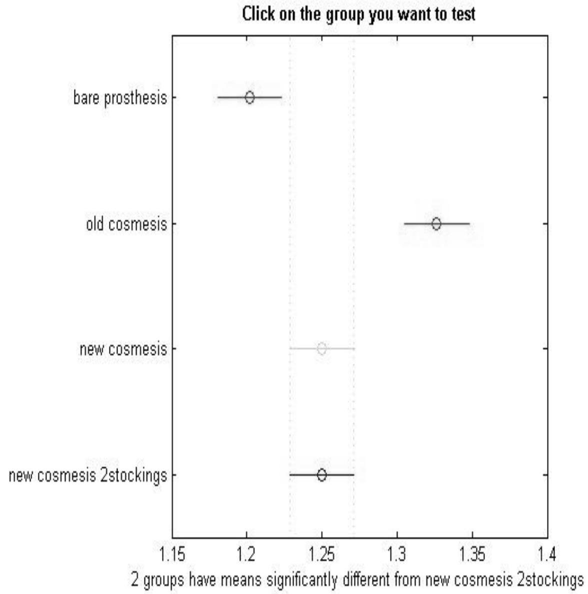
Source	SS	df	MS	F	Prob>F
Groups	0.08755	3	0.02918	20.76	2.84857e-08
Error	0.05622	40	0.00141		
Total	0.14377	43			





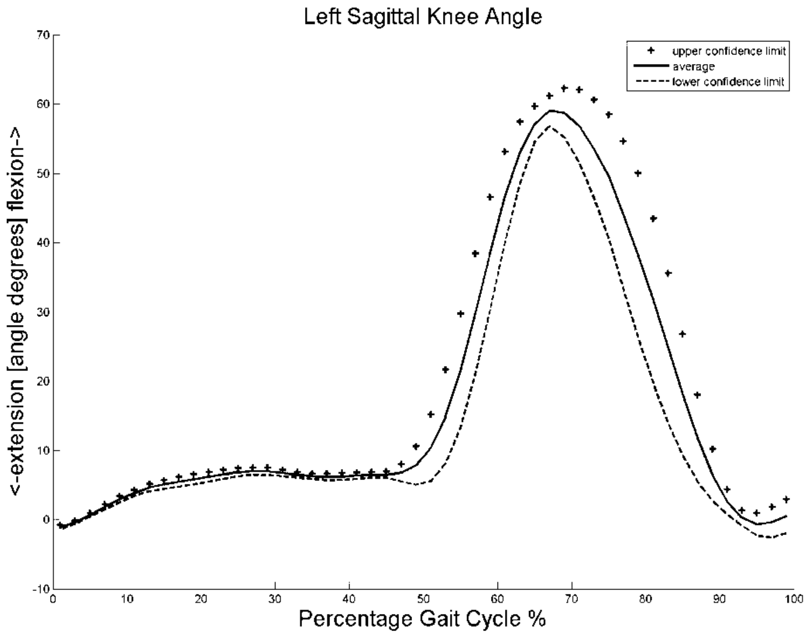
ANOVA tests results of velocity of subject 6.


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.15
1.2
1.25
1.3
1.35
1.4
2 groups have means significantly different from new cosmesis


Click on the group you want to test
bare prosthesis
old cosmesis
new cosmesis
new cosmesis 2stockings
1.15
1.2
1.25
1.3
1.35
1.4
2 groups have means significantly different from new cosmesis 2stockings

Appendix E

Sagittal plane knee flexion angles of the subject's prosthetic limb.



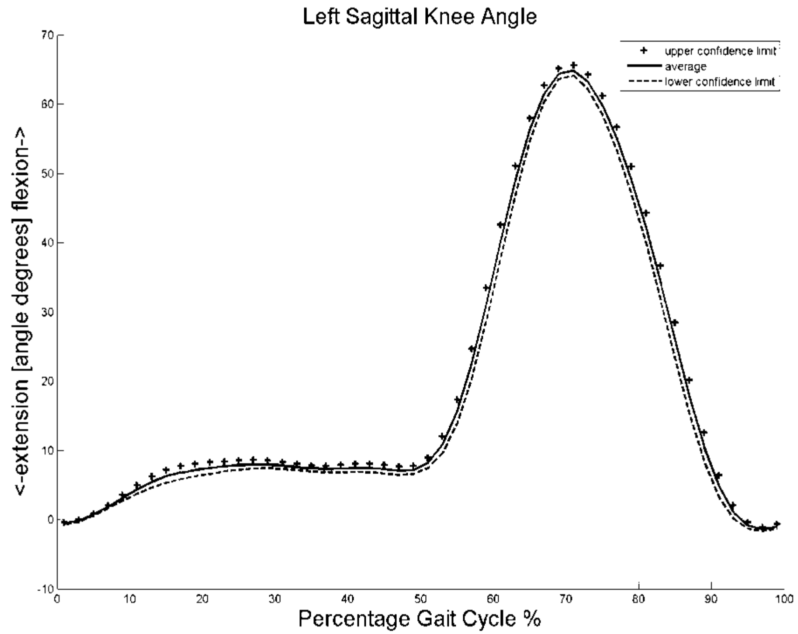


Figure 34: Subject 1 bare prosthesis (left), subject's own cosmetic cover (right), knee flexion angle in one gait cycle.

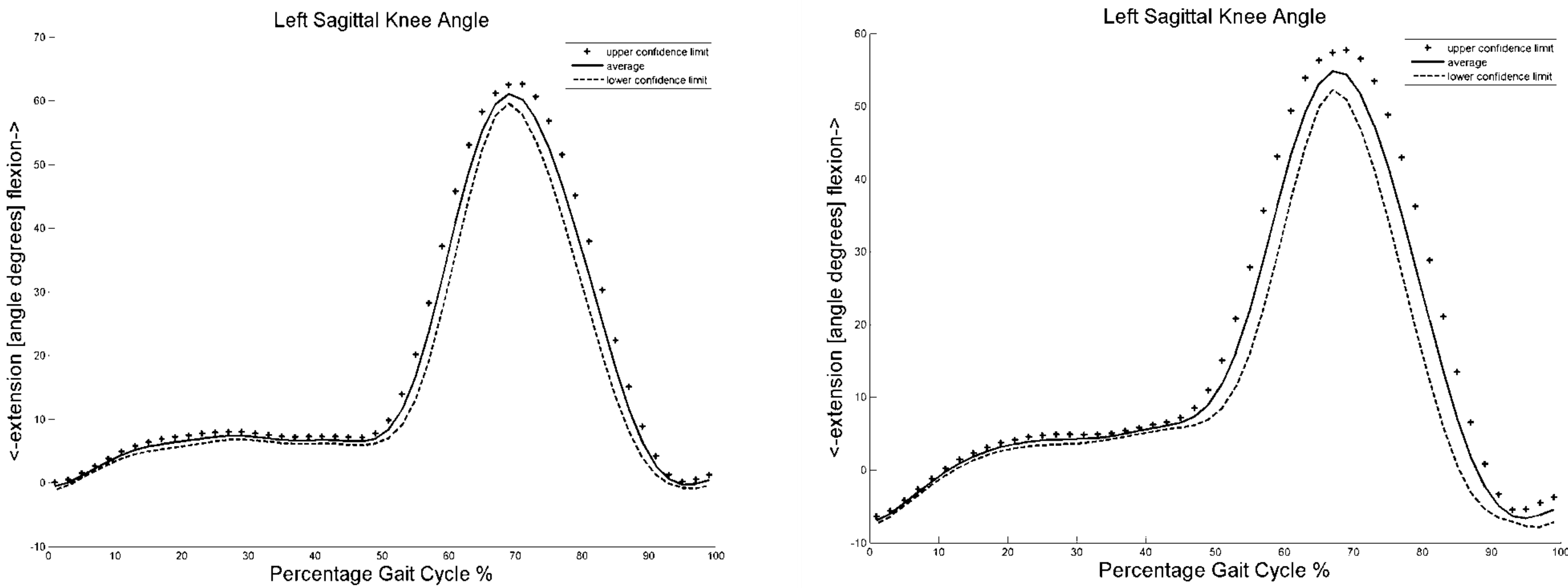


Figure 35: Subject 1 new cover (left) and new cover with two stockings (right) knee flexion angle in one gait cycle.

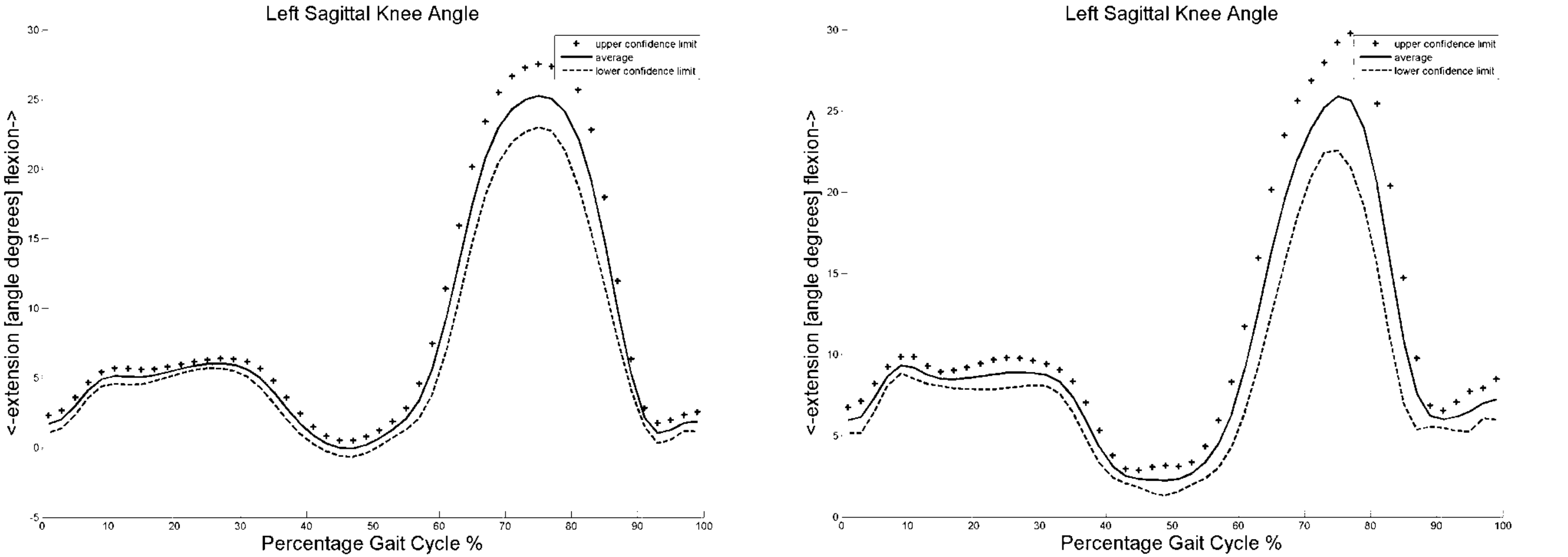
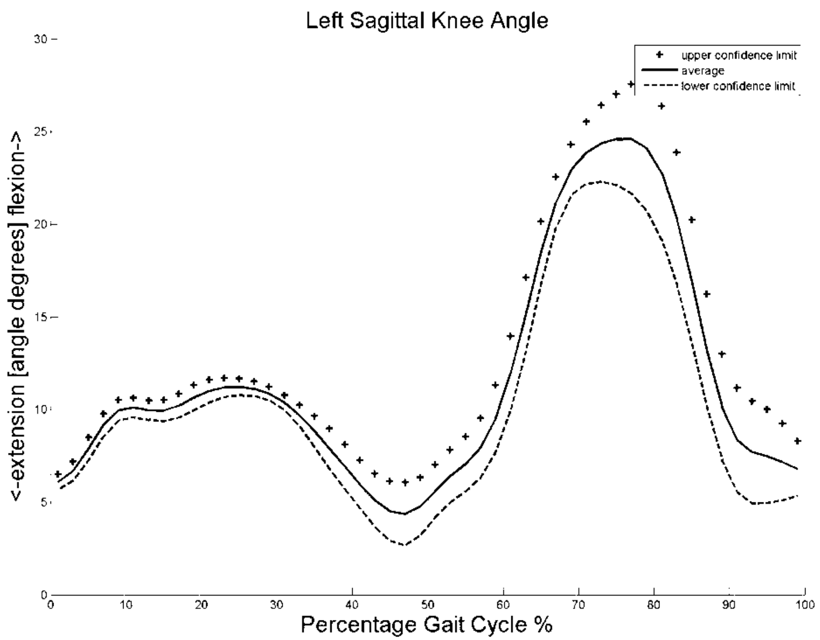


Figure 36: Subject 2 bare prosthesis (left), subject's own cosmetic cover (right), knee flexion angle in one gait cycle.



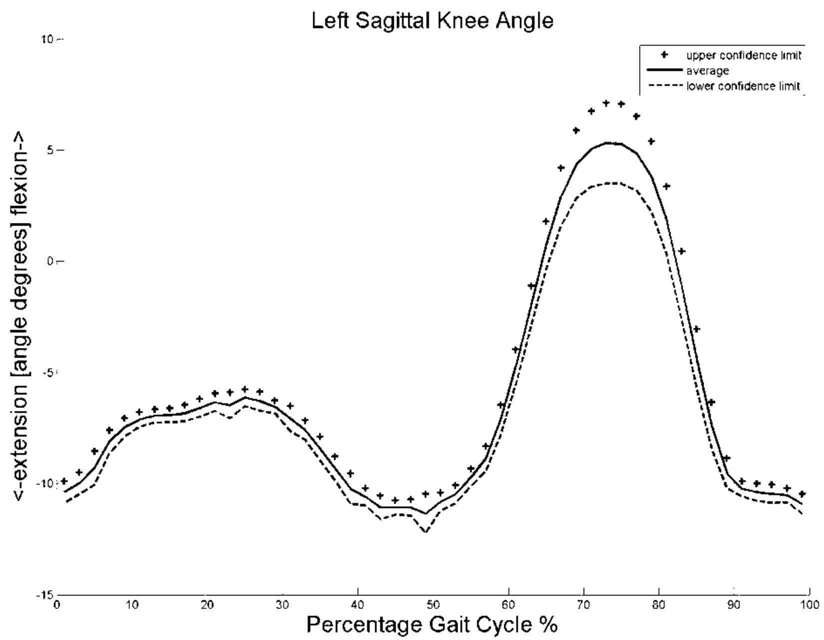


Figure 37: Subject 2 new cosmetic cover (left) and new cover with two stockings (right) knee flexion angle in one gait cycle.

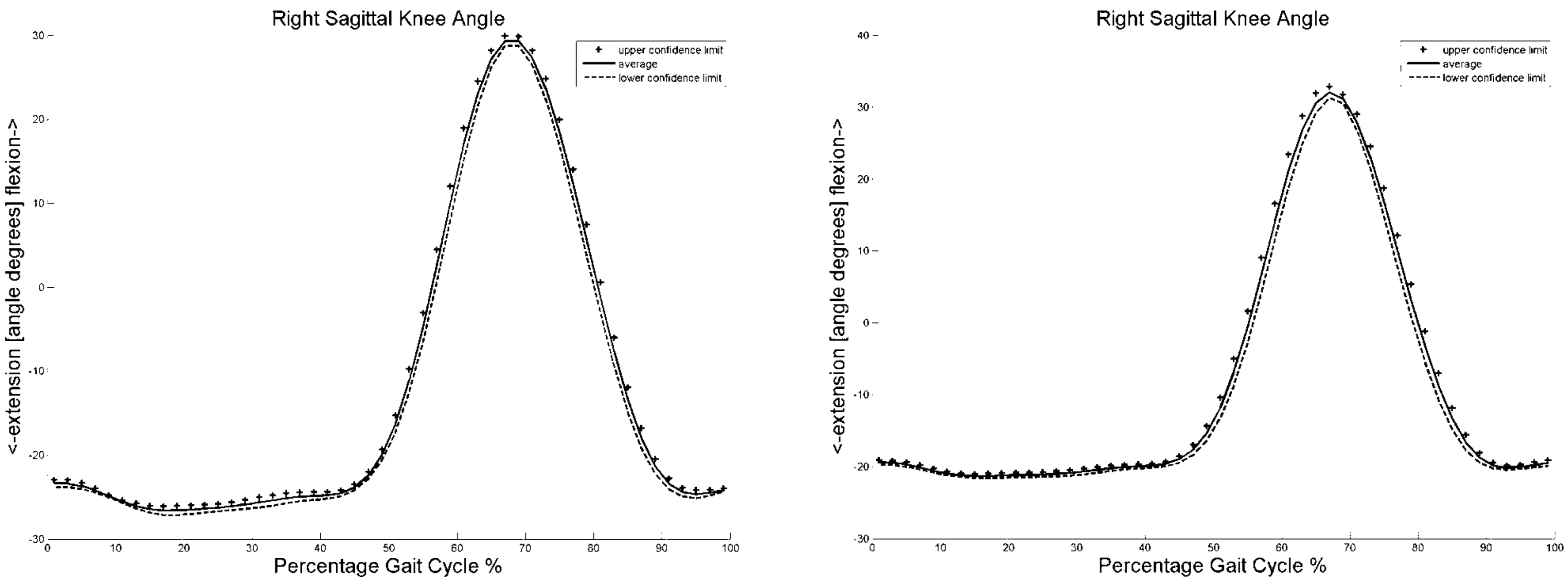


Figure 38: Subject 3 bare prosthesis (left), new cosmetic cover (right) knee flexion angle in one gait cycle.

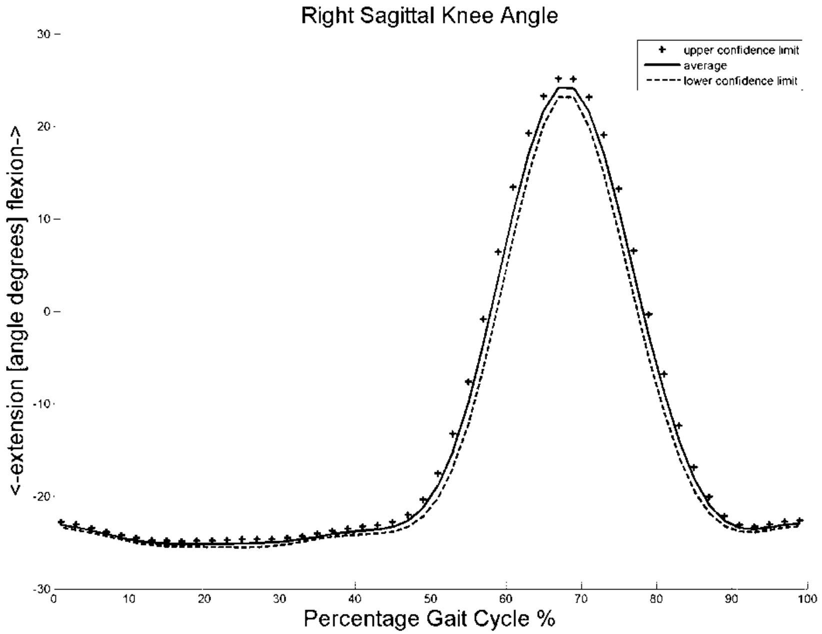
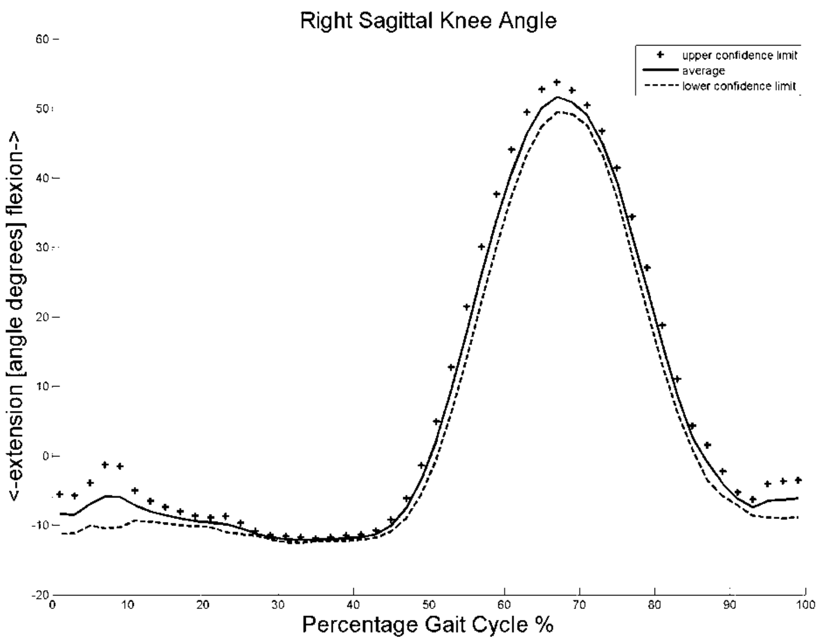


Figure 39: Subject 3 new cover with two stockings (left) knee flexion angle in one gait cycle.



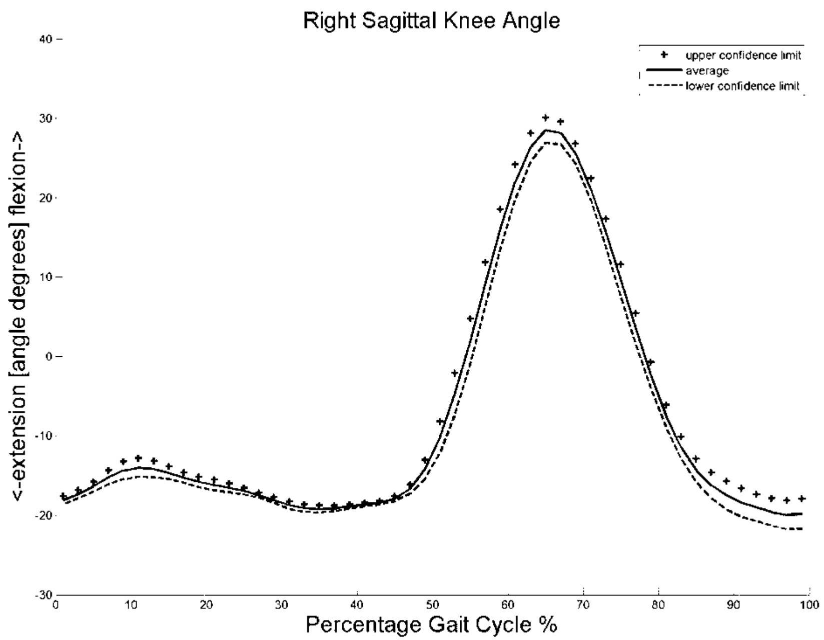


Figure 40: Subject 4 bare prosthesis (left), subject’s own cosmetic cover (right), knee flexion angle in one gait cycle.

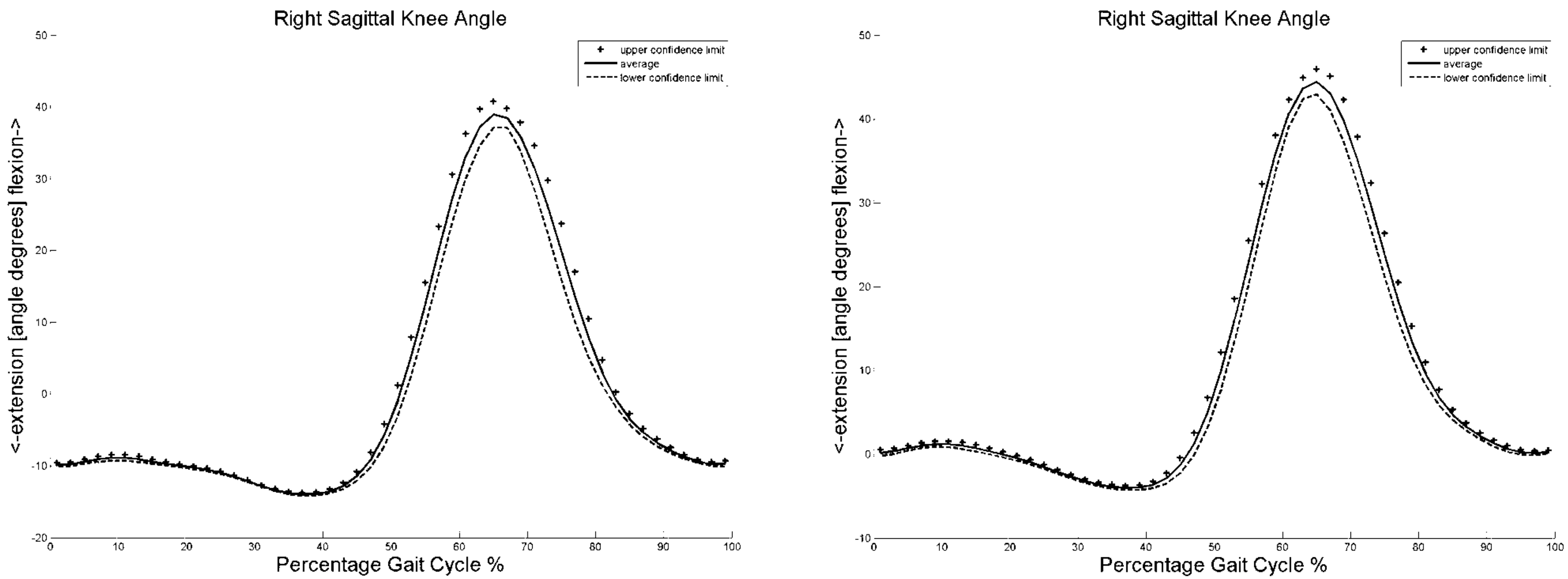


Figure 41: Subject 4 new cosmetic cover (left) and new cover with two stockings (right) knee flexion angle in one gait cycle.

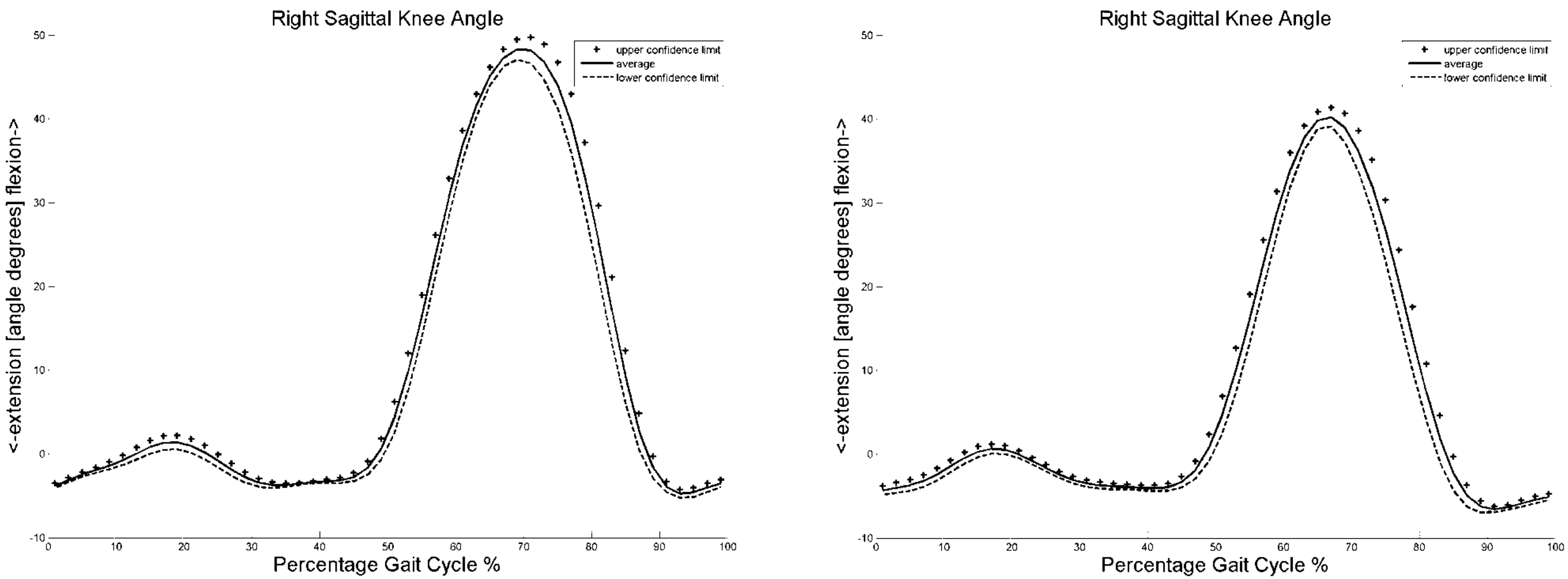


Figure 42: Subject 5 bare prosthesis (left), subject's own cosmetic cover (right), knee flexion angle in one gait cycle.

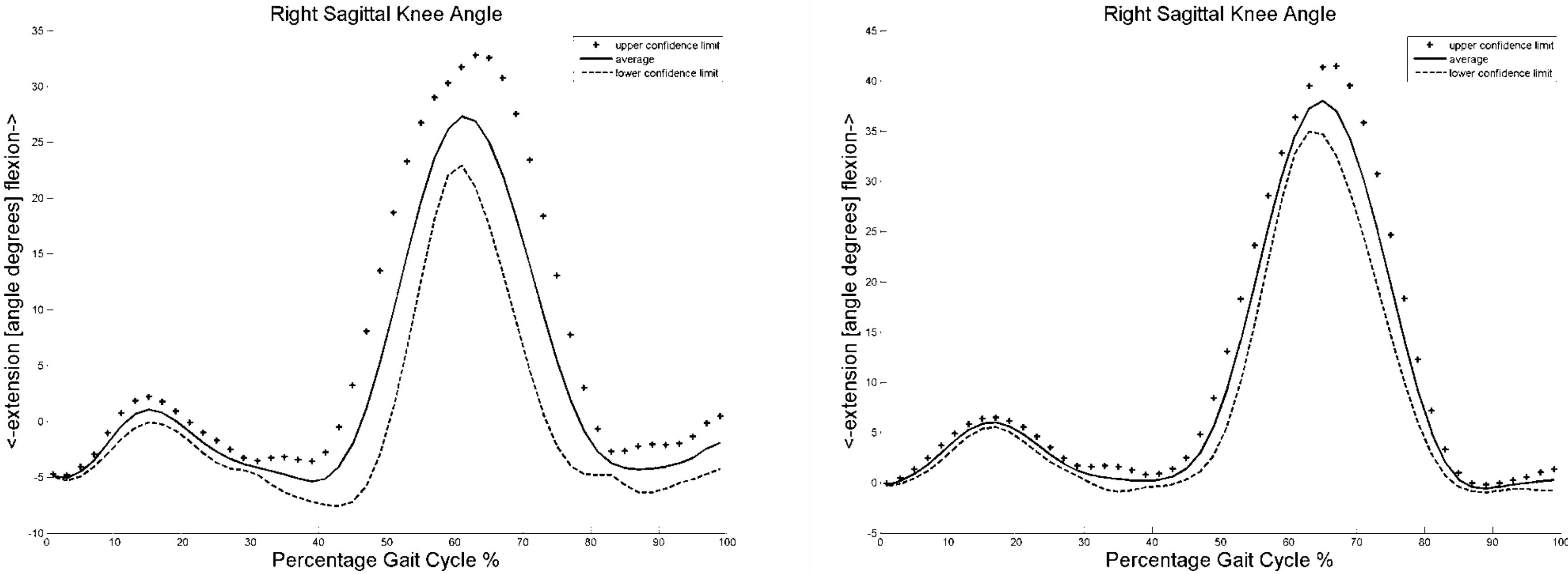


Figure 43: Subject 5 new cosmetic cover (left) and new cover with two stockings (right) knee flexion angle in one gait cycle.

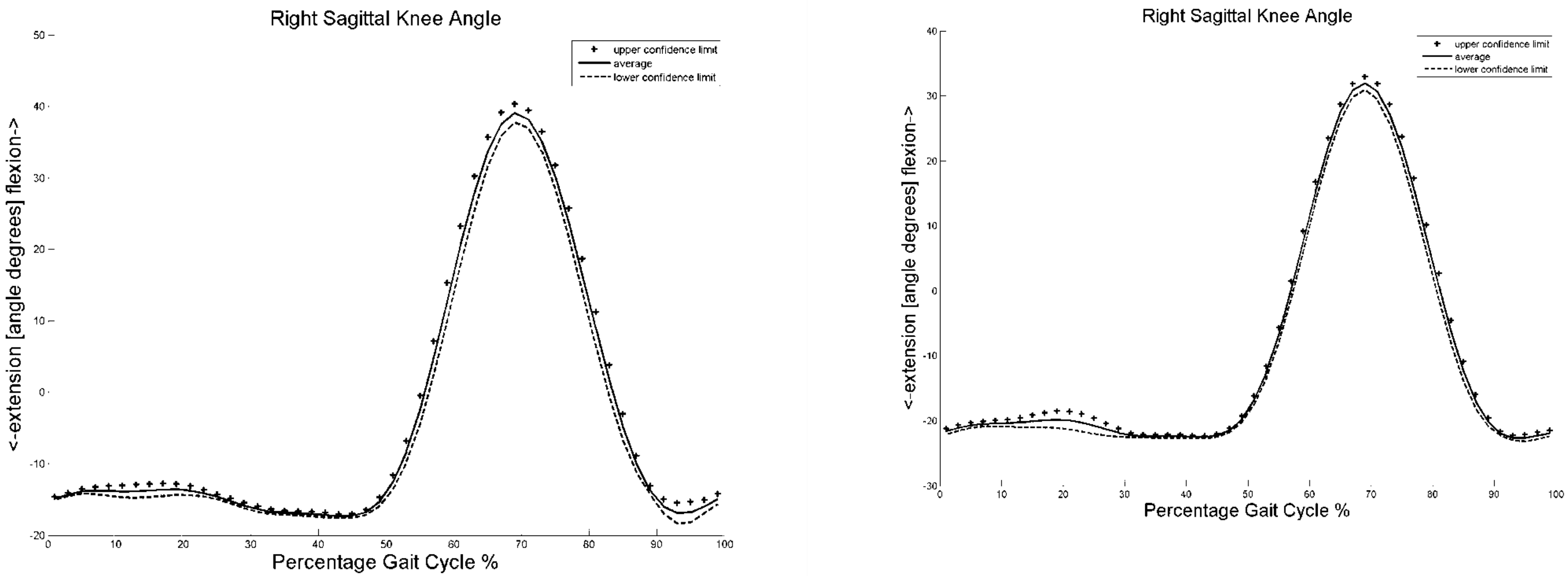


Figure 44: Subject 6 bare prosthesis (left), subject's own cosmetic cover (right), knee flexion angle in one gait cycle.

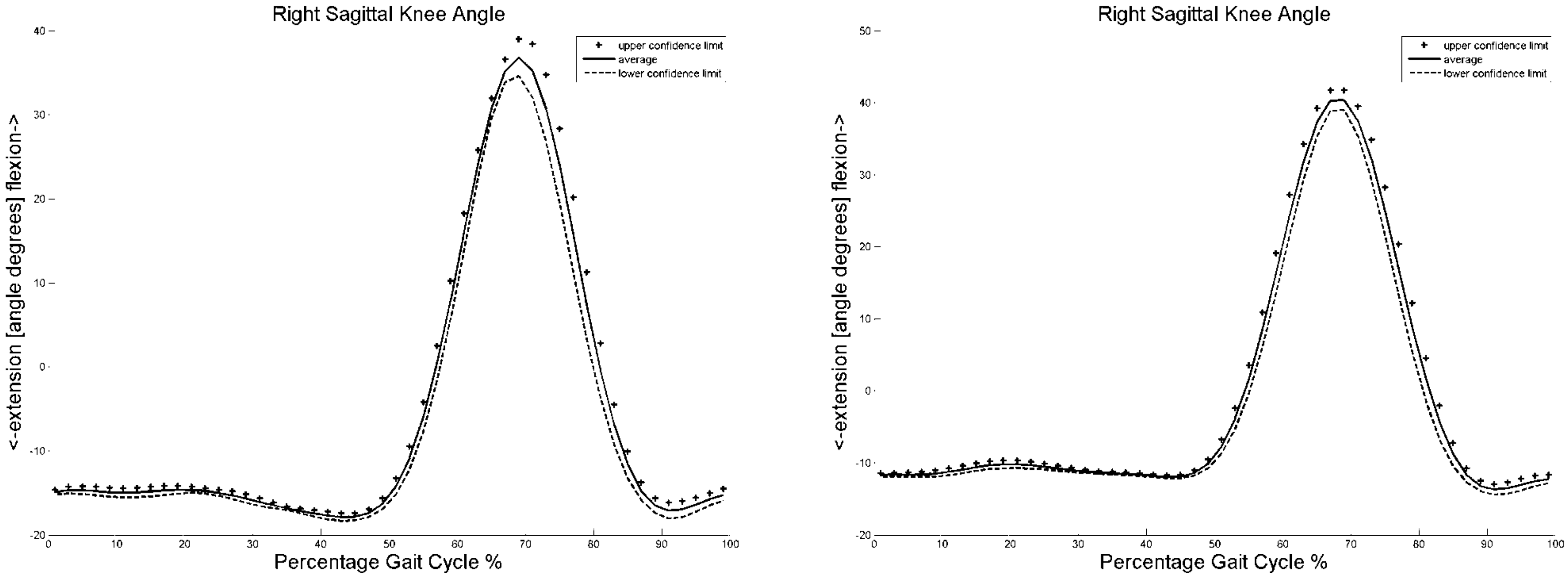


Figure 45: Subject 6 new cosmetic cover (left) and new cover with two stockings (right) knee flexion angle in one gait cycle.