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A pilot study comparing the performance of the microprocessor-controlled RHEO knee and the 3R80 hydraulic knee

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Abstract

This project studied the performance of two prosthetic systems utilising different technology. There is a lack of independent studies investigating the performance of microprocessor controlled knees, which supports manufacturers´ claims, that microprocessor controlled knees improve gait for trans-femoral amputees.

The intention of this project was to provide independent information about the performance of microprocessor controlled knee. That was done by comparing the mechanical knee joint, 3R80, and the microprocessor controlled RHEO knee.

Performance tests were implemented and data was collected for one trans-femoral amputee using the prostheses in imitated “real life situations”.

The results did show advantage and better performance of the microprocessor controlled knee for two out of three circumstances the knees were tested in. Also the feedback from the subject supports manufacturers´ claims that microprocessor controlled knees enhance smoother and more natural gait.

Acknowledgements

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1 Introduction and Background

In order to aid trans-femoral amputees to maintain the most active lifestyle possible the choice of prosthesis is a critical factor. Designing of knee joint prosthetics is in continuous progress and with technology advancements many above knee amputees are able to live active life and perform same things as intact individuals. The increased need for more advanced prostheses has led to a great market competition where prosthetics manufacturers worldwide strive for designing the best prostheses on the market.

The two most modern types of prosthetic knee joints are mechanical hydraulic knees and even more modern are microprocessor controlled knees. These types of prosthetic knees are widely used by trans-femoral amputees that are active in life and benefit from the use of the advanced technology offered by the knees. Both types of knees provide stance and swing phase control but achieve this through very different mechanisms (Lambrecht, 2008).

Prosthetics manufacturers claim that the technology utilised in microprocessor controlled knees will generally improve gait for trans-femoral amputees and solve numerous problems associated with amputee ambulation. Among the advantages that these devices should offer are gait efficiency, gait symmetry, user should be more confident walking and movements should be more natural including in situations like descending stairs and slopes. The microprocessor controlled knees are claimed to be safer and should decrease risk of falling and users can rely on that the prosthesis will not buckle when loaded (Össur, 2011; Otto Bock, 2012). To support manufacturers´ claims funded studies have been reported and results show that the studies were obviously part of marketing the devices. Depending on what manufacturer was behind the study the results did without exception illustrate significant improvements in favour of certain prosthesis manufactured by the funder of the study.

Nevertheless, feedback from users has generally been positive and questionnaires have shown that many prefer the microprocessor controlled knees and mention the decreased risk that the knee will buckle under their weight as their favourite feature. Certainly media attention and manufacturers promises can influence people's opinion and the new design and high technology that the microprocessor knees offer is appealing and that can be enough for people to believe that they experience enhanced performance (Kahle, 2008).

Microprocessor controlled knees are generally only prescribed to young and very active individuals. However, the prescription of these types of prostheses is often discouraged due to how considerable expensive they are and the lack of evidence indicating the advantages over other less expensive devices. The literature and comparative studies are contradictory (Johansson, 2005). Certain studies have shown significant improvements when using a microprocessor knee, such as decrease in energy expenditure, improved gait symmetry and enhanced balance but others studies contradict and show no significant difference (Segal, 2006). Previous studies have been focused on gait efficiency, thus assessed the metabolic cost associated with different knee designs. Generally the studies have failed to show significant decrease in metabolic cost when using microprocessor controlled knees compared with passive mechanical knee prosthesis.

The lack of objective studies is the main reason why the debate about if microprocessor controlled knees do in fact enhance better overall performance for amputees (Segal, 2006).

1.1 Passive mechanical knees

Mechanical knee prostheses are the most commonly used prosthetic knees and have been used with a great success. The major advantages of the mechanical knees is that they are low maintenance and relatively inexpensive. Numerous designs are available from different manufacturers. The technology used to control the level of resistance is where the main difference lies between the devices.

Mechanical locking knees are the simplest designs of mechanical knees and do not utilise any adjustments for the level of resistance, the knees are kept locked during walking and the user manually unlocks the joint to flex while sitting (Lambrecht, 2008).

Stance control knees are another type of mechanical knees. Weight activated stance control knees are one of the mostly used knee prostheses. When the knees are loaded, during walking, a braking mechanism secures that the knee will not buckle when weight bearing. However, a swing phase control is not provided by the knees, thus, when the knee is unloaded the shank can swing freely and that can cause problems for the user during the swing phase (Lambrecht, 2008).

Mechanical hydraulic knees are the most modern mechanical prosthetic knees. These knees are often single axis knees incorporated with a hydraulic cylinder that controls the knee´s level of resistance. These knees utilise stance and swing phase control and level of resistance can be adjusted (Lambrecht, 2008).

1.2 Microprocessor controlled knees

Microprocessor controlled knees are the most modern prosthetic knees and also the most expensive. These types of knees utilise a built in microprocessor. With the use of a microprocessor and sensors the knee monitors the user´s gait and reacts rapidly to changes in the environment and user´s behaviour. The knee provides different level of resistance according to the conditions. The knee has the ability to learn how the user walks and that facilitates the user to walk at variable speeds in different environments (OttoBock, 2010).

The aim of the microprocessor knees is to enhance natural gait for trans-femoral amputees by providing the appropriate amount of resistance throughout the gait cycle. Microprocessor knees detect changes in the environment, different phases of the gait cycle and changes in load and that way control stance and swing phase movements ((Herr, 2003) (Segal, 2006)).

Since the first microprocessor knee came commercially available in 1993, the Endolite's Intelligent Prosthesis, numbers of microprocessor controlled knees have been designed by different manufacturers and all feature different technologies. The C-Leg and the RHEO knee are both microprocessor controlled knees but differ in the way how the knees´ resistance is controlled. The C-Leg uses hydraulics to control the level of resistance but the RHEO knee uses magnetorheological fluid, a fluid that changes viscosity when exposed to an electromagnetic field.

1.3 Aim and Objectives

This study is a comparative study where the performance of the microprocessor controlled RHEO knee was tested and compared with the performance of the 3R80, passive mechanical knee. The aim of this study was to identify whether the RHEO knee has advantages over the 3R80.

Important parameters were considered, lower extremity joint moments and joint angles. The results from the study were used to evaluate these parameters and important characteristics within the subject's gait associated with the different type of prostheses were assessed.

The objectives with this research were to:

1. Identify whether there is a significant difference in the following parameters between the two tested prostheses :

    a. Hip angle pattern

    b. Knee angle pattern

    c. Ankle angle pattern

    d. Hip moment

    e. Knee moment

    f. Ankle moment

2 Literature review

Numerous studies have been performed to evaluate the performance of microprocessor knees and compare to the performance of passive mechanical knees. As previously mentioned comparative studies contradict each other and researchers do not agree whether the microprocessor knees perform better or similar compared to mechanical knees. The literature is focused on comparison on gait efficiency between the two different types of prostheses as well as biomechanical properties and gait pattern.

The following chapter will discuss the different knee prostheses designs and emphasise on different resistance control technology used. Studies that have been performed will be discussed, as well as their contradicting results.

The majority of the comparative studies assessing the performance of microprocessor knees and mechanical knees use the microprocessor controlled C-Leg and the mechanical Mauch SNS prosthetic knees for comparison. In fact, the C-Leg is the only adequately studied microprocessor knee. Thus, the literature is focused on results associated with the C-Leg and the Mauch SNS.

The C-Leg and the RHEO knee provide similar functions but achieve it through very different technology. Although this study is evaluating the performance and function of the RHEO knee, this literature review will include studies and results associated with the C-Leg.

The 3R80 and the Mauch SNS are both passive mechanical knees with hydraulic controlled resistance. These knees have similar mechanisms and therefore this literature review will include studies associated with the Mauch SNS.

2.1 RHEO Knee

Hugh Herr (Herr, 2003), the head of the Biomechatronics research group of the Massachusetts Institute of Technology Media Lab, is the patent holder of the RHEO knee. The knee is manufactured by the Icelandic prosthetics company Össur and came on the market in the year 2005. Herr is a double amputee and felt the need for improvements on previous designs of prosthetic knees and his aim was to design user-adaptive knee that would enhance performance when walking. The outcome was the magnetoRHEOlogical RHEO knee (Herr, 2003). The RHEO knee prosthesis is shown in figure 2-1 (Össur, 2011).

Figure 2-1 The RHEO knee (Össur, 2011)

The RHEO knee has an on board microprocessor that constantly samples knee position and applied load as the user walks and sends instructions to a magnetorheologic (MR) actuator that adjusts the level of resistance throughout the gait cycle. The microprocessor of the knee learns how the user´s gait is and therefore can rapidly respond to changes in environment, walking speed and load.

2.1.1 Magnetorheological resistance technique

The resistance technology used in the RHEO knee design is magnetorheological (MR) resistance technique. Inside the knee there is a MR fluid with small iron particles. When magnetic field is applied to the MR fluid the particles line up and form chains, with the chain formation the viscosity of the MR fluid increases and moment is produced at the knee. The magnetic field within the MR fluid is generated with electromagnetic and a magnetic circuit. The amount of resistance is controlled by varying the current applied to the electromagnet and thus controlling the magnetic field within the magnetic circuit and the level of resistance (Herr, 2003). The coronal section of the RHEO knee´s magnetic is shown in figure 2-2.

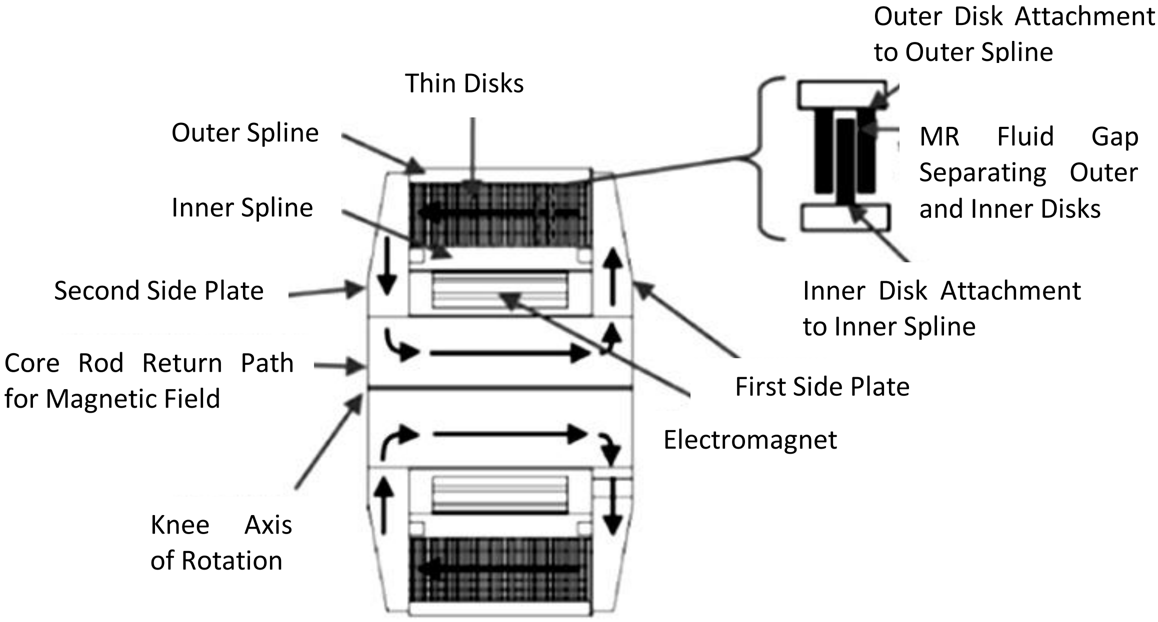


Figure 2-2 Coronal section of the RHEO knee´s magnetic circuit (Herr, 2003)

Figure 2-2 shows how thin disks separated by the MR fluid are aligned up between two sided plates. These disks are either attached to the inner or outer spline. As current is applied to an electro magnet, a magnetic field is created. This magnetic field passes through core rod or the axis of the knee, then outwards throughout the first side plate, next all the way through the sets of thin plates and after that inside to the core rod. The gap between two adjacent disks is only about 20 microns, thus, the fluid forms thin film between each set of inner and outer disks. The iron particles chains in the fluid that are created with the magnetic field make connection between these disks and generate resistance in movement of the knee, that is, the damping feature (Herr, 2003).

2.1.2 Resistance control

The knee resistance is controlled by position, moment and force sensors located on the knee that sample data at rate of 1000 times per second. Figure 2-3 shows the actuator (1), angle sensor (2), strain gage sensors (3), and electronic board and battery (4) (Herr, 2003).

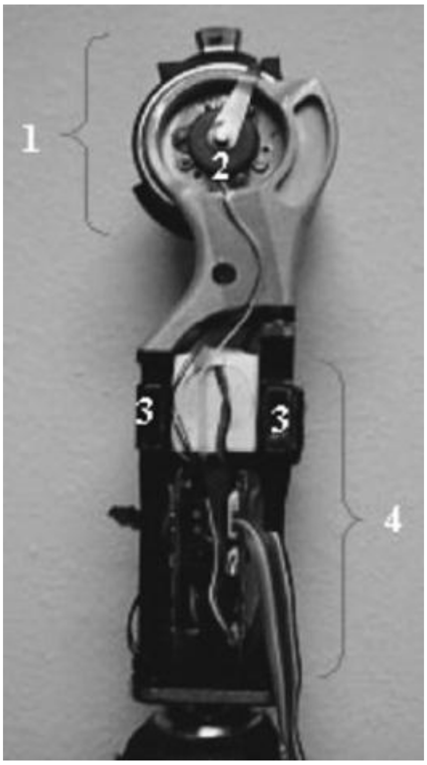


Figure 2-3 The arrangement of the RHEO knee device (Herr, 2003)

Angle sensor (2) measures flexion angle of the knee and from that angular velocity of the knee is estimated. The angular velocity indicates whether the knee is flexing or extending. Force sensors, four strain gages (3), two at the front and two at the rear of the knee, measure ground reaction force and knee moment. The measured ground reaction force indicates the position of the prosthetic foot, whether it is in contact with the ground or not. The measured knee moment indicates whether the knee is in flexion or extension. In early stance only the heel is loaded and strain gage sensors measure positive flexion moment that indicates that user´s load is behind the knee´s rotational axis and the knee is in risk of buckling. In distinction when toe is loaded during late stance negative extension moment is measured indicating that load line is in front of the knee´s rotational axis (Herr, 2003).

2.1.3 Five phases of the RHEO gait cycle

The on board microprocessor regulates the knee resistance depending on loading conditions that change throughout the gait cycle. The gait cycle is defined by the knee into five different phases shown in figure 2-4 (Herr, 2003).

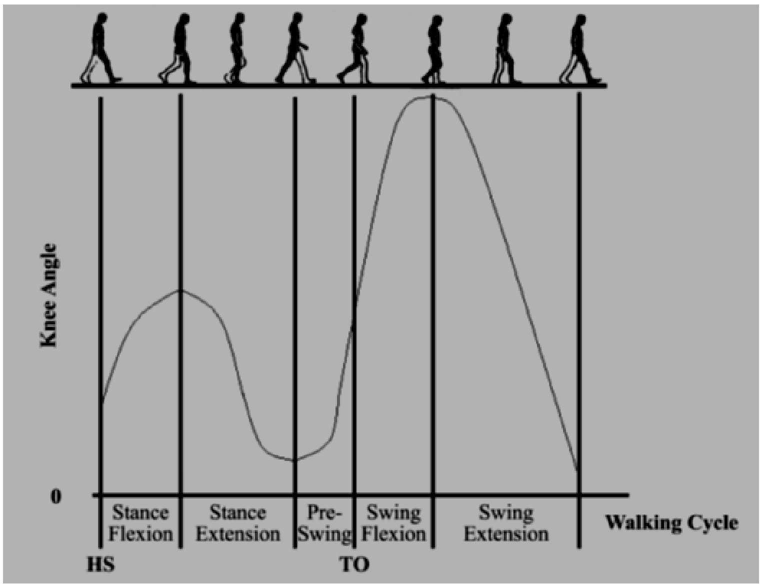


Figure 2-4 The five phases of the gait cycle (Herr, 2003)

1) Heel strike is the first phase of the gait cycle and occurs at the initial contact of the foot with the ground. At heel strike the knee is extended but immediately begins to flex to allow shock absorption and loading (Herr, 2003).

2) When knee stance flexion is reached the joint begins to go into extension again until maximum extension is reached (Herr, 2003).

3) Late stance is the period of double limb support. The body is carried forward over the stance limb and the knee starts to go from extension into flexion. In this phase the adjacent foot strikes the ground and both legs are supporting body weight (Herr, 2003).

4) In the initial swing phase the hip is in flexion and the knee flexes as well to allow the swinging limb to clear the ground as it moves forward (Herr, 2003).

5) During the swing phase the knee goes from maximum flexion into maximum extension. After maximum extension is reached the motion slows down as the heel strikes ground and another gait cycle begins (Herr, 2003).

2.1.4 Five states of the RHEO microcontroller

The microcontroller of the RHEO knee operates as a state machine. The five phases of the gait cycle (figure 2-4) correspond to different states in the microcontroller. Data from the angle, force and moment sensors are processed in the microcontroller and from these, system´s state is determined. As the user moves through the gait cycle, the controller cycles through the state machine (Herr, 2003).

Figure 2-5 shows conditions for phases in a typical gait cycle, these conditions have to be fulfilled so the controller can move from one state to another (Herr, 2003).

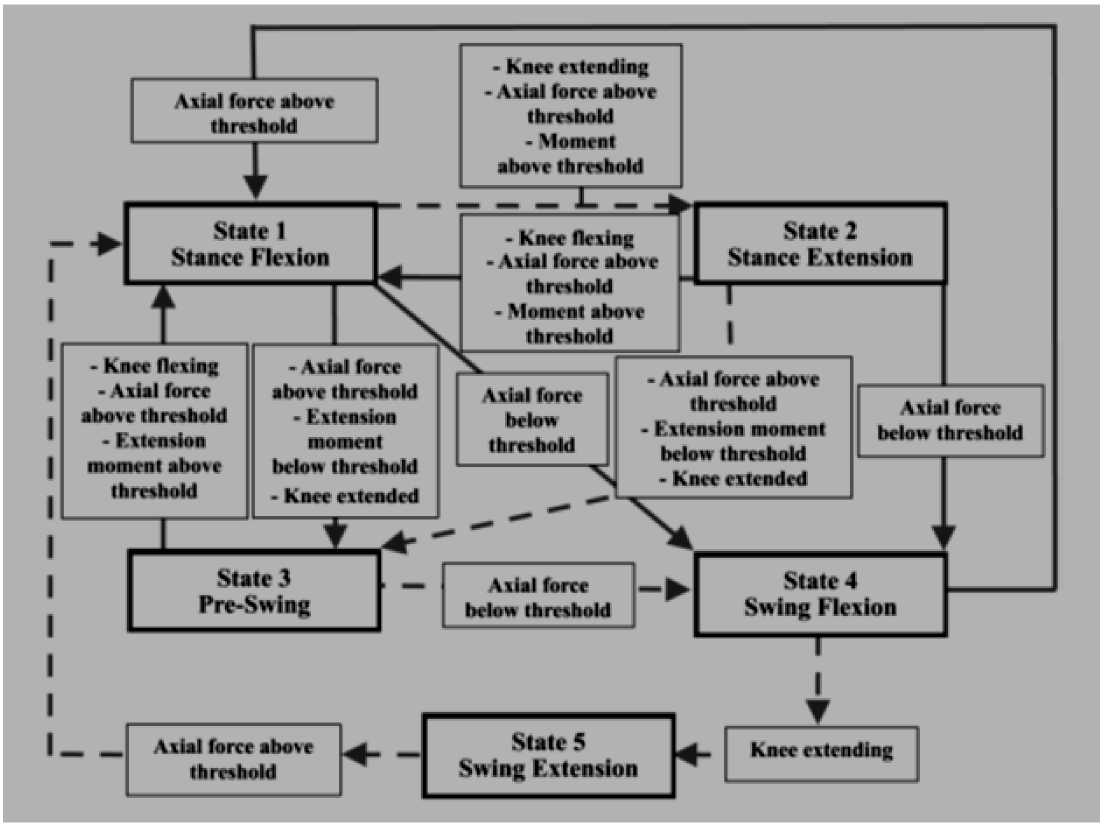


Figure 2-5 Conditions applied so state machine can shift from one state to the next one (Herr, 2003)

The current is controlled through each of the five states so the knee moment is relative to the square of the rotational velocity of the knee (Herr, 2003).

$$Torque = BV^2 \qquad (1)$$

Where V is the angular velocity of the knee and B is the active knee damping constant. The knee damping constant has five different values corresponding to the phases of the gait cycle and states of the microcontroller. The value of the damping constant for each of the five states is chosen to optimise the gait and specifically aim to acquire maximum flexion angle during the swing phase and allow early stance flexion to extension transition, which is important to allow shock absorption. The five states of the state machine are all controlled with different actions. States 1, 2 and 3 are under stance phase control and states 4 and 5 under swing phase control (Herr, 2003).

2.1.5 Dynamic learning matrix algorithm

The knee provides different level of resistance throughout the gait cycle as well as at different walking speeds. The Dynamic Learning Matrix Algorithm (DML) control calculates optimal values of resistance at any given time in the swing and stance phase. During the stance phase the knee will increase resistance and then reduce stance support in the swing phase. At loading response the knee will provide resistance to allow the user to slightly flex the knee and at heel strike the knee will work as a shock absorber. The knee also controls the rate of flexion and extension and provides extension resistance (Össur, 2011).

2.1.5.1 Stance phase control – States 1, 2 & 3

In state 1, relatively high level of knee resistance is applied, where appropriate value of the active knee damping constant (B, equation 1) prevents the knee from buckling. Resistance provided by the knee continues to be relatively high through state 2, to control the extension of the knee in stance phase. The amount of resistance provided by the knee in flexion and extension is in relation to the forces applied on the prosthesis in stance phase. With data provided by the knee sensors the level of stance resistance is automatically adjusted to user's weight (Herr, 2003).

In state 3, the phase of double limb support the current flow is shut off and the value of the active knee damping constant set equal to zero. Resistance provided by the knee at this state is a result of increased viscosity of the MR fluid, which increases the shear between the disks sets (figure 2-2) (Herr, 2003).

2.1.5.2 Swing phase control – State 4 & 5

Knee sensors measure the walking speed and the contact time with ground and together the information is used to determine the level of resistance in swing phase. In state 4 the level of resistance is controlled according to the walking speed and a reference flexion angle. In normal gait the peak flexion angle during early swing does generally not exceed 70°. To support normal gait the microcontroller has stored in memory an adaptive scheme, according to the maximum 70° reference flexion angle and ranges of walking speed. This way the microcontroller determines the appropriate amount of resistance according to user´s walking speed at any given time. The level of extension resistance, state 5, is also determined by these parameters, the walking speed and the reference angle (Herr, 2003).

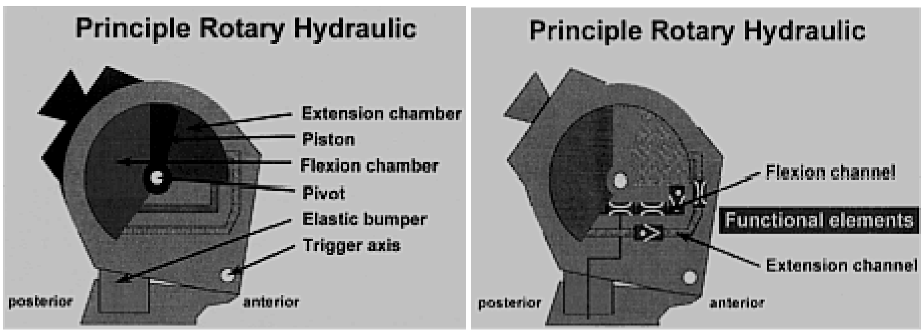
2.2 3R80

The 3R80 is a single axis mechanical knee joint manufactured by Otto Bock. Figure 2-6 shows the 3R80 knee prosthesis.

Figure 2-6 The 3R80 knee (Otto Bock, 2011)

The knee utilises a hydraulic resistance and does not involve microprocessor control.

A schematic overview of the 3R80 knee is shown in figure 2-7, (A) and (B).



(A) (B)

Figure 2-7 Schematic figure (A) and functional diagram (B) of the rotary hydraulic knee (Blumentritt, 1998)

Figure 2-7 (A) shows the hydraulic chamber which is filled with hydraulic fluids. The hydraulic chamber is divided into two chambers by the rotary axis of the piston vane. These chambers control extension and flexion movements of the knee joint as the rotary axis moves back and forward inside the chamber as the knee is flexed and extended. The two chambers are connected by extension and flexion channels containing one-way flow valves and adjustable resistance valves. Limiting the flow through these manually adjusted valves, controls the level of resistance to knee flexion and extension. Additionally to the rotary axis the knee joint utilises another trigger axis that enables the knee joint to rotate into about four degrees of stance knee flexion (Blumentritt, 1998).

Figure 2-7 (B) illustrates the control mechanism of the level of resistance, thus the resistance to movement of the hydraulic fluid inside the chamber and the channels. To initiate rotation about the trigger axis and close the stance flexion resistance valve the knee joint must be loaded properly, or sufficiently to compress the elastic bumper (Blumentritt, 1998).

The level of resistance during stance phase is determined by the position of the ground reaction forces. Figure 2-8 shows the impact from these forces. The position of the forces F1, F2 and F3 determines the movement of the knee joint, into extension or flexion (Blumentritt, 1998).

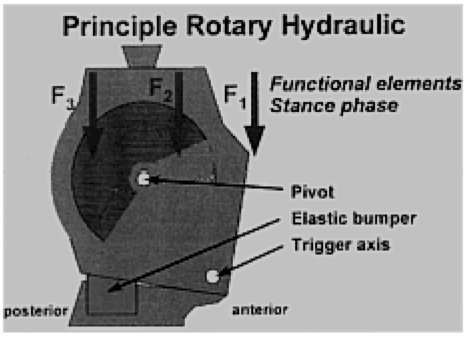


Figure 2-8 Principle of rotary hydraulic (Blumentritt, 1998)

The degree of stance control provided during walking is determined by ground reaction forces (F1, F2 and F3, see figure 2-8). F1 is the most anterior force. Knee extension moment is created by it and no stance control is required. The knee pivots about the anterior trigger axis caused by F2. The knee can pivot up to four degrees, in that way the stance control valve closes. The knee flexes around the rotary axis, when impacted by F3 and this motion is resisted by the hydraulic stance control element.

2.3 C-Leg

The C-Leg is a microprocessor knee joint manufactured by the prosthetics company Otto Bock. Figure 2-9 shows the C-Leg prosthetic knee (Otto Bock, 2012)

Figure 2-9 The C-Leg (Otto Bock, 2011)

The C-Leg utilises a hydraulic regulated resistance, where a hydraulic cylinder regulates the knee´s resistance under the control of an on-board microprocessor. The microprocessor analysis data acquired from sensors. The knee sensors monitor angular velocity, knee angle and the load on the prosthesis throughout the gait cycle. The microprocessor receives signals from its sensors to determine the type of motion of the user. The microprocessor then signals the hydraulic cylinder to act accordingly and adjust the knee´s resistance during extension and flexion. (Otto Bock, 2012).

The C-Leg is probably the most used microprocessor knee joint worldwide and according to Otto Bock, since it was introduced in 1999 over 40.000 above-knee amputees worldwide have been using the prosthesis.

2.4 Mauch SNS

The Mauch SNS, figure 2-10, is a mechanical hydraulic knee manufactured by Össur.

Figure 2-10 The Mauch SNS (Össur, 2011)

The Mauch SNS is one of the most common prosthetic knees designs used today. Like the 3R80 the Mauch SNS utilises a linear hydraulic cylinder inside the knee which regulates the knee´s resistance. The Mauch SNS prosthesis is like the C-Leg, very commonly used in studies, and that is reflected in a great amount of literature associated with the knee.

2.5 The gait of intact individuals

2.5.1 Ground reaction force

Gait cycle is the stage from where one foot touches the ground and until it touches the ground again. The cycle is divided into two phases, stance phase and swing phase. The stance phase relates to about 60 per cent of the gait cycle and is divided into 5 events seen in figure 2-11 (Vaughan, 1996).

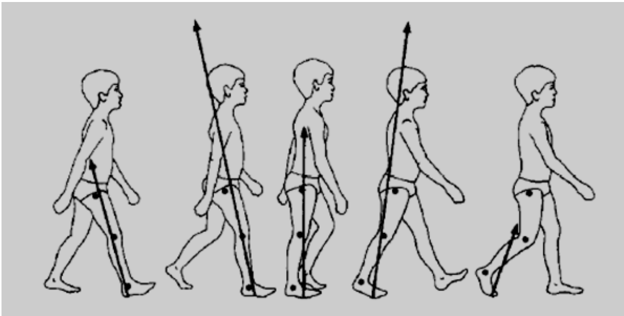


Figure 2-11 Stance phase of the gait cycle and the alignment of ground reaction force (Vaughan, 1996)

Heel strike occurs when the foot first contacts the ground. At heel strike the knee is almost fully extended but right after the heel strike it flexes slightly and the ground reaction force passes in front of the hip joint and behind the ankle and knee joints (Vaughan, 1996).

After heel strike the foot goes flat on the ground and ground reaction force exceeds body weight and translates in front of the hip, through the knee and in front of the ankle joint. The alignment of the force causes different joint moments. The knee moment at this instant is zero, however, the hip extension and ankle plantar flexion moment are great in magnitude (Vaughan, 1996).

At mid-stance the force decreases in magnitude and it translates through the hip and in front of the ankle and knee joints. At this instant the hip moment is zero but the ankle plantar flexion moment is relatively high and the knee moment is flexion (Vaughan, 1996).

After mid-stance the heel goes off the ground and again the reaction force exceeds body weight. At this instant the force passes behind the hip and knee joints but stays in front of the ankle joint. The alignment of the force causes flexion moment at the hip joint and extension moment at the knee joint (Vaughan, 1996).

At the end of the stance phase the toes loses contact with ground, the alignment of the ground reaction force is the same as at heel off but its' magnitude decreases (Vaughan, 1996).

2.5.2 Lower limb Joint motions and moments

Studying kinetics and kinematics of the lower limbs during walking can provide important information about individual´s gait. Motions and moments curves in sagittal plane at the hip, knee and ankle joints of a normal intact individual during walking are shown in figures 2-12, 2-13 and 2-14. In the top portion of the figures the five key gait events are shown, that is, the heel contact, foot flat, mid stance, heel off, and toe off. The upper curves on the figures show the pattern of normal joint motion and the bottom curve represent flexion and extension moments experienced by the joints (Murdoch, 1970).

Figure 2-12 represents normal kinetics of the hip joint during walking.

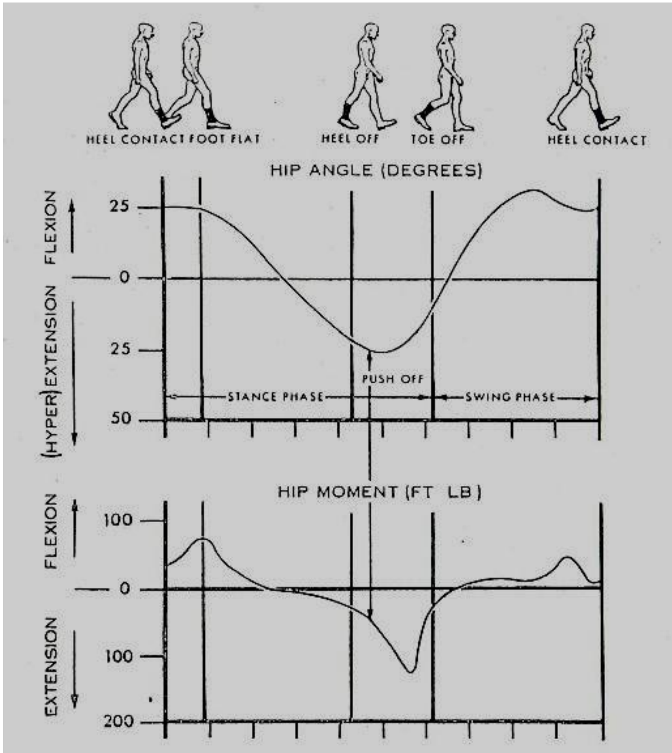


Figure 2-12 Hip motion and moment of intact individual during walking (Murdoch, 1970)

At initial contact of the heel with the ground the hip is in flexed position, about 25°. At the same instant the hip experiences flexion moment which peaks just before the foot goes flat on ground. For the rest of the stance phase the hip is in extended position, with peak extension occurring at heel off, with about 20° extension. At similar instant peak extension moment occurs (Murdoch, 1970).

Figure 2-13 represents normal pattern of the kinetics of the knee joint during walking.

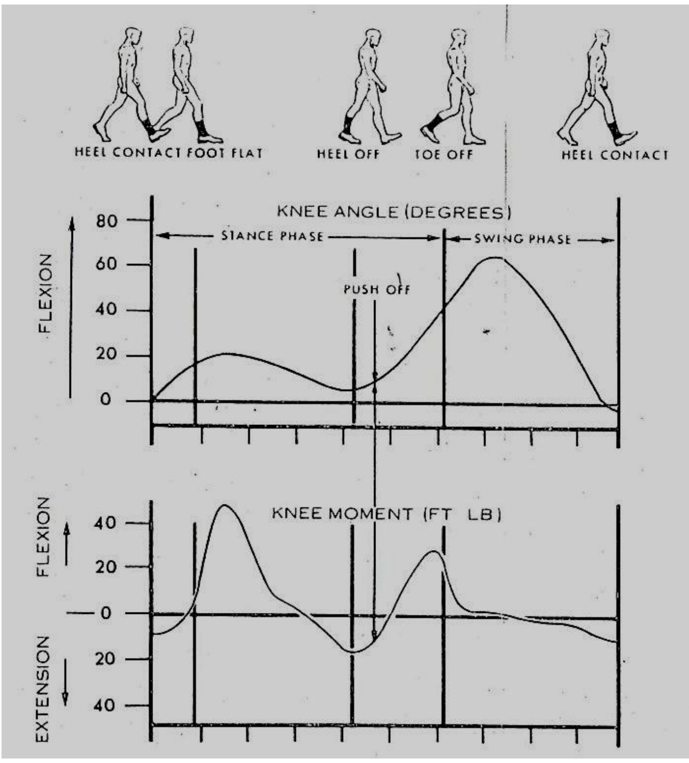


Figure 2-13 Knee motion and moment of intact individual during walking (Murdoch, 1970)

At the instance of heel contact, the knee flexes and peak knee flexion occurs at early stance, where the knee is in approximately 20° flexion. At the instance of heel contact, the knee joint experiences a flexion moment which reaches peak value at foot flat. After the flexion peak the flexion moment reduces rapidly and at mid stance, there is zero moment around the knee. Before the heel comes off ground the knee is in fully extended position and an extension moment about the knee peaks. When heel comes off the ground the knee joint starts to flex and another flexion moment peak can be seen on the curve before the moment goes to zero during the swing phase (Murdoch, 1970).

Figure 2-14 represents normal kinetics of the ankle joint during walking.

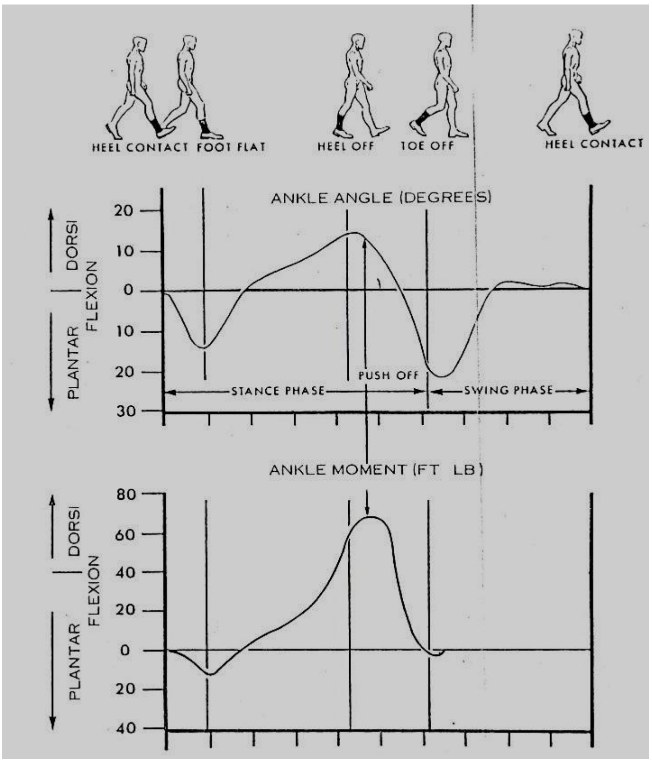


Figure 2-14 Ankle motion and moment of intact individual during walking (Murdoch, 1970)

Right after the initial contact of the heel, the ankle plantar flexes as the weight is transferred on the limb. At foot flat, the peak plantar flexion occurs, about 15°. At the same time a peak plantar flexion moment about the ankle is reached. After foot flat, the motion of the ankle changes to dorsiflexion, which peaks when heel comes off the ground with the ankle in 15° dorsiflexion. At the same time a dorsiflexion moment about the ankle reaches peak. After heel off the ankle motion goes from being dorsiflexion to plantar flexion and at toe off, the peak plantar flexion is reached, about 20°. At toe off, the ankle is 15° plantar flexed and during the swing phase, the position of the ankle joint is approximately 90° (Murdoch, 1970).

These gait characteristics are generally associated with intact individuals with relatively normal gait (Vaughan, 1996). For trans-femoral amputees gait characteristics are generally different from the characteristics in intact gait. The deviations in prosthetic gait are generally related to compensations for the difficulties in walking with prosthesis and the fact that trans-femoral amputees suffer from loss of active moment generation and somatosensory feedback and limb position awareness (Jaegers, 1995).

2.6 Above-knee amputee gait

2.6.1 Gait deviations

For individuals that have undergone trans-femoral amputation; a combination of training, appropriate socket fit and the optimal prostheses components is necessary to be able to establish good gait pattern (Lenhart, 2009).

Generally knee kinetics and kinematics are significantly different for trans-femoral amputees compared with intact individuals. Most trans-femoral amputees show little or no knee flexion throughout stance phase. This insufficient knee flexion can be related to the training that amputees go through after amputation. Patients are in some cases encouraged to stabilise the prosthetic knee joint by using hip extension to fully extend the joint during stance phase. There are several benefits from this training philosophy. If the knee is locked in extension the amputees are not depending on the ten to fifteen degrees of stance knee flexion that most prosthetic knee joints allow. By creating this stability with the hip, the muscles are active and amputees are more likely to be able to avoid falling if the knee begins to buckle due to disturbance or hindrances. Also by keeping the hip extension active, strengthens the hip extension musculature, which can decrease the demand on the intact limb and enhance more comfortable and better socket fit (Lenhart, 2009).

The altered knee kinematics during stance phase generally results in kinematics deviations in the hip joint. In normal gait, hip flexion angle stays similar through the beginning of the gait cycle. In a trans-femoral amputee gait the hip extension velocity is more constant from heal-strike until peak hip extension is reached. The lack of knee flexion during stance phase results in a constant hip extension velocity where the pelvis progresses over the fully extended prosthetic limb (Lenhart, 2009).

Generally trans-femoral amputees demonstrate a great decrease in ankle joint power generation. During terminal stance, where body weight moves forward on the supporting foot, the knee position changes quickly from being extension during mid-stance to being peak flexion during swing phase. Extension moment generated by the prosthetic knee joint controls the transition of the knee from extension to flexion (Lenhart, 2009).

Although the mechanical knee prostheses are generally used with good results the limiting factor of the mechanical mechanisms is that the level of resistance in swing phase is manually adjusted and does not change relatively to the walking speed. Having the rate of knee extension set as a constant means that rate of knee extension is only optimal at certain range of walking speeds. When walking speed changes it can cause gait deviations and changes in kinetics and kinematics of the gait. Other problems that have been associated with the mechanical knee mechanisms are that during the stance phase, when the prosthesis is loaded, there is increased risk of the knee buckling under the user´s weight (Lenhart, 2009).

Studies assessing the above-knee amputee kinetics have reported that compared with intact individuals trans-femoral amputees generally experience a decrease in joint moments and joint mechanical energy demands on the prosthetic limb but equivalent increase in mechanical energy demands and joint moments on the intact limb((Yang, 1991) (DiAngelo, 1989)).

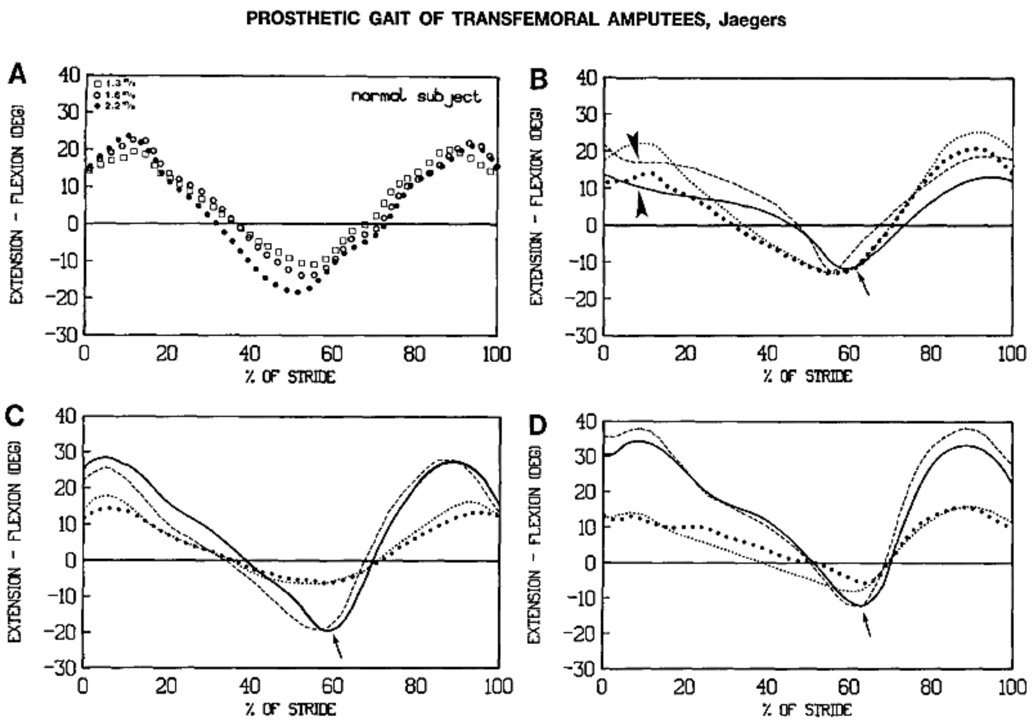
2.6.2 Above knee amputee gait studies

Numerous studies have studied gait characteristics of trans-femoral amputees but most studies assess kinematic parameters when subjects walk at self-selected comfortable walking speed. Murray et al (Murray, 1980), James and Oberg (James, 1973) and Zuniga et al (Zuniga, 1972) studied kinematic parameters of amputee ambulation while walking at self-selected walking speed and all reported that gait asymmetry is a problem associated with trans-femoral amputees. Longer stance phase on the intact side and longer swing phase on the prosthetic side was observed. A decrease in joint moments, mechanical energy demands and longer swing phase on the prosthetic side indicates that trans-femoral amputees are associated with limited loading on the prosthetic side limb when walking and compensations result in increased load on the intact side limb (Murray, 1980).

2.6.2.1 Hip, knee and ankle joint movements and moments

Jaegers et al (1995) studied gait characteristics of trans-femoral amputees while walking at comfortable and rapid walking speeds. Eleven trans-femoral amputees participated in the study and two intact individuals were studied and used as controls. The subjects had different stump lengths and were categorised to three different groups according to their stump length.

Generally the subjects showed a slightly different pattern of knee and hip flexion and extension compared with the controls. Figures 2-15 and 2-16 represent the results of the hip and knee motion patterns obtained for the intact individuals and the three groups of the amputee subjects, depending on their stump length and different walking speeds. Figure 2-15 shows the hip motion of the intact individuals (A) and the amputee subjects (B, C, and D).



2-15 The patterns of hip flexion-extension of three amputees with a long (B), medium (C), and short (D) stump length and a normal subjet (A) at the v_{conf} and v_{rapid} (IS, Intact leg; AS, prosthesis leg). •••, IS comf; ——, AS comf; ˙˙˙˙˙, IS rapid; ····, AS rapid (Jaegers, 1995)

Results showed that hip flexion and extension of the intact limb of the amputees were not significantly different from the intact individuals. Some subjects showed significantly larger hip flexion of the prosthetic limb compared to the intact individuals and generally the subjects showed significantly larger hip extension on the prosthetic side than the intact side limb. The intact individuals showed peak hip extension right after heel strike of the other foot, but for the amputee subjects the maximum hip extension on the affected side occurred later When curves on figure 2-15 are compared it can be seen that for all amputee subjects, graphs B, C and D there is a sharp transition from hip extension to hip flexion in late stance for the prosthetic leg compared with more smooth curve for the intact individuals (A).

Authors discuss that right before toe off the pelvis rotates slightly backwards causing an increased hip extension on the prosthetic side and right after that a rapid transition from hip extension to flexion occurs at the amputated thigh. Authors conclude that when peak hip extension occurs and the limb goes into swing phase the angular velocity has increased and with shorter stump length the angular velocity of the hip will increase (Jaegers, 1995).

The results for the hip motions indicate that the amputee subjects were stabilising the prosthesis during stance phase by using their hip extension to fully extend the prosthetic limb. As a result they show exaggerated hip extension movement on the prosthetic side and wider range of motion than shown by the intact subjects. The increased hip flexion associated with the prosthetic limb further indicates that the subjects needed exaggerated hip movements to control and support stabilisation of the prosthetic limb during walking. Authors conclude that the delayed peak hip extension on the prosthetic side could be an indication of that the subjects were trying to increase the step length of the intact limb to compensate for the prosthetic limb (Jaegers, 1995).

Figure 2-16 represents the results for the knee motion of the intact individuals (A) and the amputee subjects (B, C, and D).

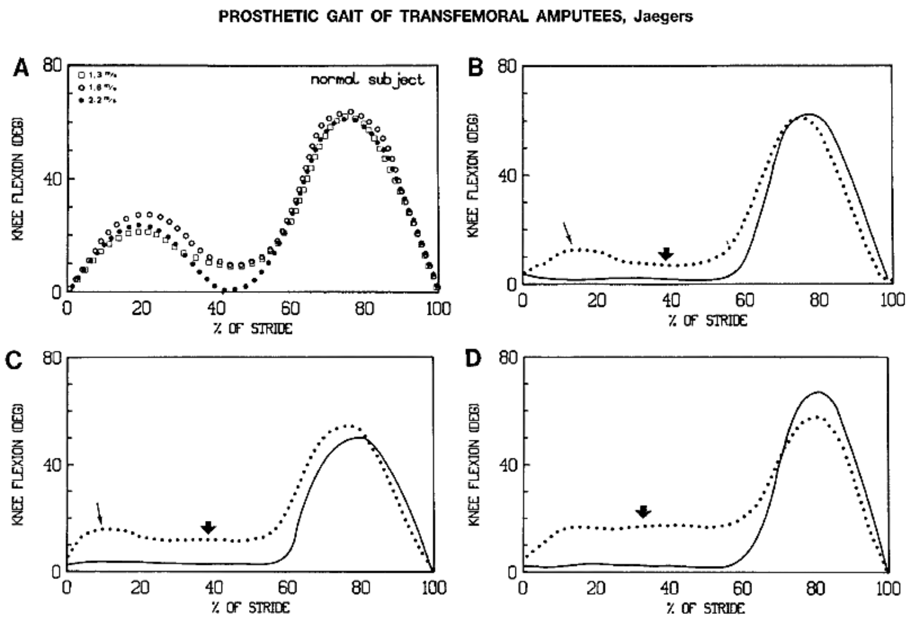


Figure 2-16 The patterns of knee flexion-extension of three amputees with a long (B), medium (C), and short (D) stump length and a normal subjet (A) at the v_{conf} . •••; ——, prosthetic leg (Jaegers, 1995)

Results for the intact individuals showed, as was expected, knee flexion at the instance of heel strike and after the heel strike they flexed the knee further. For the intact individuals the knee flexion increased with increased walking speed. Results for the amputee subjects showed that for the intact limb their heel strike was normal. However, with slightly

decreased knee flexion at heel strike compared with the intact individuals. Additionally the knee stayed slightly flexed throughout the stance phase which is also unlike what was seen for the intact subjects. Authors discuss that the decreased knee flexion on the intact limb could be explained by lower walking speed of the amputee subjects than the intact ones (Jaegers, 1995). The fact that the intact limb was flexed throughout the stance phase can be a result of inadequate load on the prosthetic limb, as walking the amputee sort of throws his weight on the intact limb and as a result the knee is flexed to allow cushioning and shock absorption.
For the prosthetic limb there was a significantly decreased knee flexion during early stance. Some of the amputee subjects had also decreased knee flexion for both limbs during the swing phase, however, other amputee subjects were associated with greater knee flexion in terminal swing compared with the intact subjects (Jaegers, 1995). As discussed, amputee subjects showed exaggerated hip motion on the prosthetic side, which could be a result of the lack of the prosthetic knee flexion during early stance.

Authors discuss that the decreased peak knee flexion during swing phase that was associated with the prosthetic knee could indicate that the level of resistance provided by the prosthetic knee in swing restricted the subjects to normally flex the knee joint (Jaegers, 1995). The subjects used passive mechanical knee joint in the study and as mentioned earlier, a problem associated with the mechanical knee mechanisms is that the level of resistance is manually adjusted. If the walking speed was not the optimal speed according to the resistance adjustments that can result in altered knee motion.

Seroussi et al (1996) and Blumentritt et al (1998) also studied the differences in joint movements and moments of trans-femoral amputee gait and normal gait and also the differences between the prosthetic and the intact limbs. Throughout both studies mechanical prosthetic knee joints were used, the Mauch SNS and the 3R80. Subjects were given weeks to months to acclimatise to the prostheses.

Seroussi et al (1996) reported similar motion of the joints for the intact subjects and the intact limb on the amputees. Conversely, comparison on the prosthetic limbs on the amputee subjects and the intact individuals revealed significant difference. Blumentritt et al (1998) reported great variations between the subjects.

Results from Seroussi et al (1996) showed that through all phases of the gait cycle, prosthetic ankle motion was reduced, especially at push-off where the plantar flexion range of motion for the intact subjects was 28° compared to 7° for the prosthetic ankles. The significant decrease in plantar flexion at push-off leads to decreased energy generation by the prosthetic ankle and authors concluded that will likely cause other compensatory movements (Seroussi, 1996).

Figure 2-17 represents the results from Blumentritt et al (1998) for prosthetic limb and intact limb knee motions. The variations between the seven subjects can be well seen.

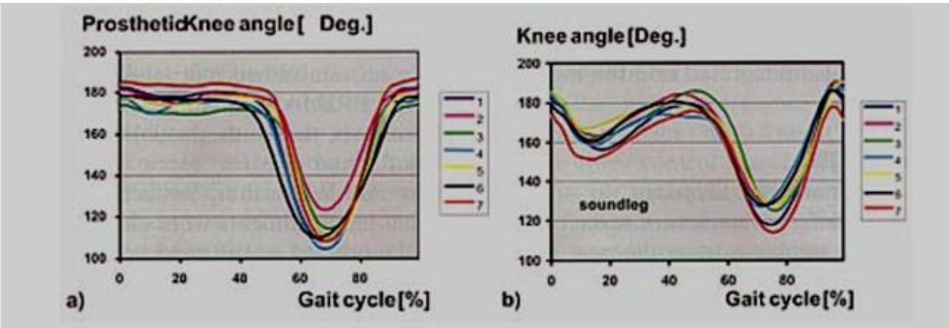


Figure 2-17 Knee motions for the prosthetic limbs (a) and the intact limbs (b) of seven subjects (Blumentritt, 1998)

Agreeing with the reports by Jaegers et al (1995) the results from Blumentritt et al (1998) (figure 2-17) and Seroussi et al (1996) showed that at heel strike and during early stance the prosthetic knee was locked in extension and did not flex as in normal gait. Reports from Seroussi et al (1996) agree to reports by Jaegers et al (1995), that throughout late stance and the swing phase the prosthetic knee motion was close to normal knee motion apart from slightly delayed flexion peak in the swing phase. As seen on figure 2-17 results from Blumentritt et al (1998) do not show delayed peak flexion for the prosthetic knee during swing. However, their results generally showed slightly greater peak knee flexion for the prosthetic knee during mid-swing phase when compared to the intact knee.

Seroussi et al (1996) compared joint moments of the limbs of the amputee subjects to the intact subjects. Blumentritt et al (1998) studied the differences in the joint moments among the prosthetic and the intact limbs. Figure 2-18 represents the results from Blumentritt et al (1998) for the hip (a), prosthetic knee (b), the ankle (c) and the intact knee (d) joint moments.

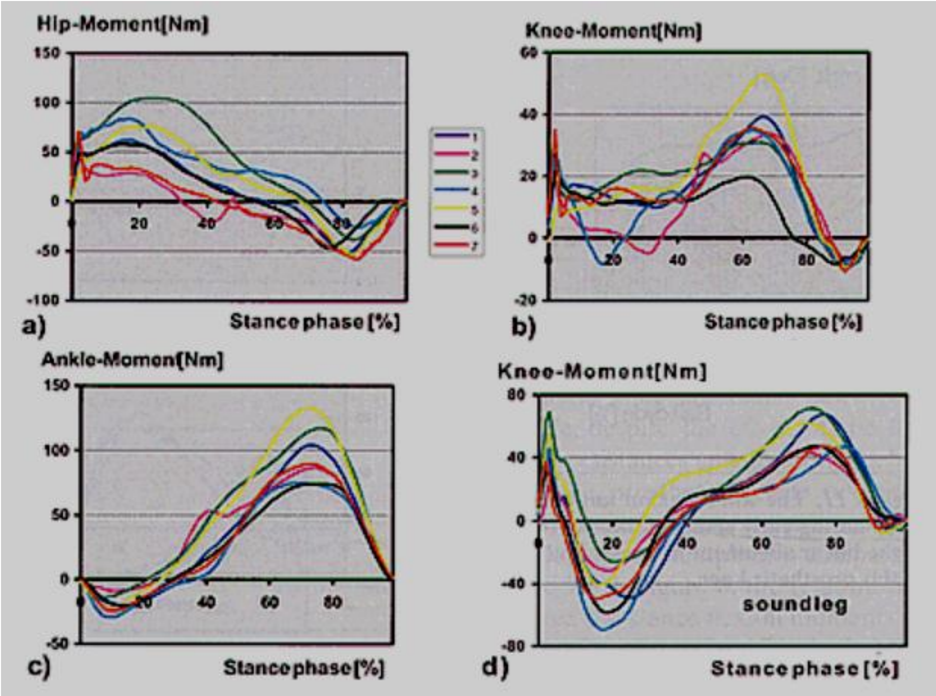


Figure 2-18 Hip moments (a), knee moments for the prosthetic limb (b), ankle moments (c) and intact limb knee moments (c) of seven subjects (Blumentritt, 1998)

Results from Blumentritt et al (1998) (figure 2-18) and Seroussi et al (1996) both showed ankle plantar flexion moment about the prosthetic ankle during early stance which changed to dorsiflexion moment about mid stance and lasted throughout the stance phase. Seroussi et al (1996) reported that the prosthetic ankles were associated with smaller plantar flexion moment when compared with intact ankle joints

As seen on figure 2-18, results reported by Blumentritt et al (1998), the prosthetic knee moments showed great inter individual variations among the subjects. Results showed that some subjects had flexion moment about the prosthetic knee during early stance, with peaks occurring at different instances. Other subjects had no or very small flexion moment about the prosthetic knee during weight bearing phase, similar results were reported by Seroussi et al (1996). Seroussi et al (1996) reported an absences or very small knee extension moment in early to mid-stance and Blumentritt et al (1998) reported varying prosthetic knee extension moment among the subjects, with very different peak values and different trajectories. As seen on figure 2-18 the intact knees showed various results among the subjects. The moment patterns were similar, however different peak values occurring at different times during the gait cycle. Authors discuss that the varying results are likely due to different ways and needs of the subjects to compensate for the prosthetic knee joint and also the influence from the anatomical muscles that are active in the intact limb and the biological knee joint. Furthermore, the authors discuss that during walking, the movements of the prosthetic knee joint is more predictable than movements of the biological knee as the mechanical device reacts mainly to moments about the knee (Blumentritt, 1998).

Similarly to the prosthetic knee moments, the hip moments results reported by Blumentritt et al (1998) varied greatly among the subjects (figure 2-18 (a)). The peak hip extension moment during the stance phase varied between 25 and 105 Nm and the peak hip flexion moment during pre-swing varied between 27 and 59 Nm. Seroussi et al (Seroussi, 1996) reported for the hip moments that the intact sides on the amputee subjects were associated with greater hip extension moment in early stance compared with the prosthetic side and the intact subjects. In late stance the peak hip flexion moment was found to be greater for the prosthetic limb compared with the intact limb and the intact subjects (Seroussi, 1996). The increased hip flexion moment in late stance associated with the prosthetic limb could indicate that the subjects were slowing down the extended hip to be able to progress the body over the prosthetic limb.

Results showed increased hip extension and flexion moment associated with the amputee subjects and authors discuss that the exaggerated hip movements are likely results of the hip compensating for the decreased push-off of the prosthetic ankle joint (Seroussi, 1996)

Due to the lack of knee flexion of the prosthetic knee, authors discuss that a lack of hip flexion in the early stance will exist. Thus, more eccentric hip flexion work is required in prosthetic limb compared to normal limb. Consequently, the centre of mass of the subject´s has the tendency to be placed posteriorly to assist in hip flexion moment generation to control the extension of the hip during early to mid-stance (Seroussi, 1996).

As discussed, there has been a continuous progress in knee joint prosthesis design. The design of the microprocessor knees aims to minimise the gait deviations discussed and enhance normal biomechanics as possible. Studies discussed earlier conclude that trans-femoral amputees are associated with altered gait biomechanics, where the hip is compensating for both lack of power generation by the ankle at push-off and absence of flexion in the prosthetic knee during stance phase. Manufactures' of microprocessor-controlled knees claim that these gait deviations can be minimised with the use of microprocessor controlled knee joints. The devices allow users´ to walk with normally flexed knee without the knee buckling during weigh bearing and the user should experience improved balance. Additionally microprocessor controlled ankle prostheses are claimed to solve the problem associated with the lack of power generation of prosthetic ankles. Further studies where microprocessor controlled knees were studied will be discussed later.

2.7 Above-knee amputee stair descending pattern

Stair descending can be a challenge or even a hindrance for above-knee amputees. Generally trans-femoral amputees descend stairs with one step at a time pattern but the advances in prosthetic knee technology have enabled stronger individuals to descend stairs more naturally, using step-over-step pattern. Microprocessor controlled knees are claimed to offer a safer and more natural way to descend stairs. The manufacturer of the RHEO knee and the C-Leg claim that when users´ descend stairs, the knee joint adjusts the level of resistance so walking with step-over-step pattern will not cause the knee to buckle.

Kahle et al (2008) investigated manufacturers´ claim and studied whether subjects were able to descend stairs with step-over-step pattern when using a microprocessor-controlled knee. Nineteen trans-femoral amputees participated in the study and were given 90 days prior to the study to acclimatise to the microprocessor controlled C-Leg. Results revealed that twelve out of nineteen subjects were able to descend stairs with step-over-step pattern using the C-Leg. However, study's results do not indicate whether step-over-step is the pattern that subjects would choose to descend stairs and if it is the most efficient and safest way Kahle et al (2008).

Bellman et al (2010) also studied how microprocessor-controlled knees performed during stair and slope walking. Results reported by Bellman et al will be discussed in chapter 2.10.3.

2.8 Normal and above-knee amputee gait during slope walking

Kinematics and kinetics of normal gait during level walking has been extensively studied but in the literature there are not many studies that have assessed gait during slope walking. Kuster et al (1995) studied the differences in biomechanics during level and downhill walking. Twelve able bodied subjects participated in the study and walked both on level ground and slope with -19° downhill gradient.

Results showed significant difference in knee joint kinematics during the stance phase and early-to-mid swing when level and downhill walking was compared. When walking down the slope subjects showed greater knee flexion and during mid-stance the difference was more than 20°. Authors concluded that results indicated that when walking down the slope the ankle joint compensated for the gradient at push off and during swing phase. Comparison on ground reaction forces for level and downhill walking showed no significant difference although force peaks were significantly different (Kuster, 1995).

Authors further concluded that when walking down slope, the required muscle power in hip, knee and ankle joints is greater through the whole gait cycle excluding the push-off phase. Knee joints moments and muscle power showed the most significant difference when walking down the slope as the required muscle power was greater and moments increased throughout the stance. Hip moments and muscle power did not increase as much compared to the knee joint but authors concluded that during downhill walking adjustments in the gait at heel strike do mainly occur in the hip joint. Despite increase in muscle power in knee, hip and ankle joints, overall downhill walking is less energy demanding compared to level walking (Kuster, 1995).

Vrieling et al (2008) studied uphill and downhill walking gait for above knee amputees. Results showed that when subjects were walking uphill and downhill they generally did not increase prosthetic knee flexion. During late stance the subjects showed reduced hip extension when compared with intact individuals, which authors concluded was related to a smaller step length on the prosthetic side. All subjects in this study were fitted with mechanical knee joints except one, which was fitted with a microprocessor controlled knee. The subject with the microprocessor knee did not show increased knee flexion during the stance phase as was expected when compared with the remaining subjects, however, in late stance he showed increased knee flexion when walking on level ground, uphill and downhill (Vrieling, 2008).

While walking downhill the subjects were associated with smaller hip flexion on the prosthetic side during the swing phase. Authors discussed that it could indicate a smaller step length on the prosthetic side and that way the subjects make sure they are able to place the prosthetic foot safely on the declining surface (Vrieling, 2008).

In uphill walking, results showed that the intact limb was associated with smaller peak hip extension, increased knee flexion and ankle plantar flexion at toe-off. Authors discuss that the increased knee flexion and the plantar flexion of the ankle could be an adjustment to ease the loading of the prosthetic limb, as it is more difficult to load the limb during uphill walking as the height of the slope has to be overcome (Vrieling, 2008).

Authors concluded that knee flexion throughout the gait cycle is the parameter that should be focused on in prosthetic design. Without compromising the stability of prosthetic knees the flexion properties need to be improved (Vrieling, 2008).

2.9 Gait efficiency and energy expenditure

Trans-femoral amputees use significantly more energy when walking. Increase in energy expenditure can be from 22 to 88% compared with intact individuals. Mass asymmetry of the limbs and the fact that amputees lack the sensorimotor control of the prosthetic limb are prospective reasons for this significant increase. Abnormal gait pattern and factors like abnormal trunk movements and energy transfer are common problems associated with amputees and do increase energy expenditure (Chin, 2003).

Chin et al (2003) performed a comparative study on energy expenditure for trans-femoral amputees and intact individuals. The study involved comparison on eight trans-femoral subjects and same number of intact individuals. Results showed averaged of 24% increased metabolic cost for the amputees (Chin, 2003).

Numerous studies have been performed to compare energy expenditure when using microprocessor-controlled knees and mechanical knees. Results have varied, some studies report significant difference, decreased metabolic cost for the microprocessor knees and other have reported no or minor differences.

The first commercially available microprocessor knee, the intelligent knee prosthesis (IP) has been studied with varying reported results. Buckley et al (1997) and Taylor et al (Taylor, 1996) reported slightly decreased metabolic cost when using the IP compared with mechanical prosthesis.

Kirker et al (1996) also studied the IP and implemented a survey to study subjects' preferences. Six trans-femoral subjects, all used to wearing the IP participated in the study. Authors studied energy expenditure, thus investigated manufacturers' claims that microprocessor knees enhance gait efficiency. The metabolic cost associated with the prostheses was compared with a pneumatic prosthetic knee. Treadmill was used where subjects walked at variable walking speeds (Kirker, 1996).

Results showed no significant difference in energy expenditure on any of the walking speeds. Results from a questionnaire, where fourteen trans-femoral amputees participated, showed that all subjects preferred the IP over the mechanical knee joint. Their preference was based on that they felt it took less effort walking using the IP, especially at faster walking speeds (Kirker, 1996).

The authors criticise the method used in the study to measure energy expenditure. More accurate methods exist but are more time consuming and therefore difficult to implement when time with subjects is limited. The method used in this study to evaluate energy expenditure was Physiological Cost Index, which is based on pulse rate and walking speed (Kirker, 1996).

Datta et al (2005) also performed a comparative study on oxygen consumption. A microprocessor knee and pneumatic knee were tested and compared. Results were similar to results reported by Kirker et al (1996). No significant difference was found in oxygen consumption at normal walking speed, however, when walking at slow speed the microprocessor knee was associated with significant decrease in oxygen consumption (Datta, 2005).

The method Kirker et al (1996) and Datta et al (2005) used to measure oxygen consumption was similar. Both studies used treadmill for their testing, which can cause errors. In both studies the IP was compared to pneumatic knee prosthesis, thus the comparison should be relevant and both studies had relatively few participants. The fact that Datta et al (2005) reported significant difference in oxygen consumption only at slow walking speed does not indicate that the microprocessor knees enhance overall gait efficiency. It is questionable if it is a major advantage for relatively young and fit individuals to be able to walk at slow walking speed and save energy to some extent.

Schmalz et al (2002) studied oxygen consumption rate when using the C-Leg and a mechanical hydraulic knee. Reported results tie up with results reported by Datta et al, slightly decreased metabolic cost when walking at slow speed, about 6% decrease for the C-Leg, but no significant difference was observed at faster walking speeds. Schmalz et al

(2002) discuss that in this study subjects were not allowed to acclimatise to the C-Leg prior to the study. That will have affected results where subjects were unfamiliar with the prosthesis.

Johansson et al (2005) discuss the fact that no significant difference was observed between the C-Leg and the mechanical knee at greater walking speed. Johansson et al claim that is associated with the adjustments on the mechanical knee joint, where the swing phase damping was optimised for walking at greater speed.

Reports indicate that metabolic cost does generally not decrease significantly when using a microprocessor knee, although manufacturers claim otherwise. Reported results indicate that energy expenditure focused studies are not the best way to investigate the performance of microprocessor controlled knees. Metabolic studies have failed to show any significant difference between the knee devices other studies did further comparison where gait biomechanics were considered and assessed.

2.10 Comparative studies on prosthetic knees

The following studies investigated the performance of microprocessor controlled knees. Metabolic cost was evaluated and kinetics and kinematics associated with different knee prostheses were investigated. The reports were published in the same time order as they appear here. Due to lack of independent studies in the literature assessing microprocessor knees, particularly the RHEO knee, both independent and dependant studies are discussed in this section.

2.10.1 RHEO knee

As mentioned, numerous dependant studies have been reported where the performance of microprocessor knee prostheses has been investigated. Generally the studies have been implemented by researchers that are in some way linked to the manufacturers or the design of the devices and obvious competing interests exist.

Johansson et al (2005) performed a comparative study on microprocessor controlled knees and mechanical knee. Among the authors of the study is Hugh Herr, the developer of the RHEO knee. The study compared two microprocessor knees, the RHEO knee and the C-Leg and the mechanical knee Mauch SNS.

Tests were implemented to compare metabolic cost across the prostheses and evaluate if there is a significant difference in gait biomechanics when walking at self-selected comfortable speed and using different types of prostheses. The RHEO knee and the C-Leg were compared and they were then compared to the mechanical knee joint, Mauch SNS (Johansson, 2005).

Eight trans-femoral amputees took part in the study and they were given ten hours to adapt to the different knee prostheses. To measure oxygen uptake, subjects walked at self-selected walking speed over an indoor track. To analyse the difference in kinetics and kinematics between the devices a motion analysing system and force plates were used to acquire data for nine walking trials. Electromyography (EMG) electrodes and accelerometers were used to monitor EMG activity and patterns of motions. Same socket, prosthetic foot and shoe were used throughout the study to ensure compliance. Results were evaluated to test for significant difference among the prosthesis, with statistical significance set at 5% (Johansson, 2005).

Results from oxygen consumption evaluation showed a difference in oxygen consumption across the three knees during walking at self-selected speed. The RHEO knee was associated with an average of 5% lower rate of oxygen consumption compared with the Mauch SNS and 3% lower compared with the C-Leg. Comparison of the oxygen consumption between the Mauch SNS and the C-Leg did not show significant difference (Johansson, 2005).

Authors do not mention the adjustment settings on the Mauch SNS prosthesis, for what walking speed the swing phase damping was optimised. The results were not consistent with results reported by Schmalz et al (2002), which reported no significant difference in oxygen consumption across these knees. Schmalz et al (2002) discussed that the results reported by Johansson et al (2005) indicate that the Mauch SNS was optimised for slower walking speed (Schmalz, 2002).

Results from the gait analysis revealed that walking speed was similar across the knees but for the RHEO knee step time was significantly longer (Johansson, 2005). The fact that increased step time was associated with the RHEO knee indicates that subjects were loading the prosthesis more naturally compared with other tested prostheses and will result in more natural gait and better balance over the prosthesis.

Comparison of hip biomechanics across the prostheses showed that during the stance phase the Mauch SNS was associated with greater negative hip work compared to the microprocessor controlled knees. The authors' claim that the greater negative hip work associated with the Mauch SNS indicates exaggerated hip control when using the Mauch SNS compared with the microprocessor knees (Johansson, 2005). During late swing the RHEO knee showed a lower peak hip extension moment compared to the two hydraulic prostheses. The authors do not mention how great these biomechanics differences were, but importantly none of these parameters were significantly different across the knees.

Comparison of knee biomechanics revealed that during terminal swing the C-Leg was associated with significantly lower angular velocity and greater peak knee extension angle compared to both the RHEO knee and the Mauch SNS. At toe-off the Mauch SNS was associated with significantly greater peak knee extension moment and maximum knee power absorption compared to the microprocessor controlled knees. Conversely, the RHEO knee was associated with significantly lower peak knee flexion moment in terminal swing compared to the hydraulic knees (Johansson, 2005).

When ankle biomechanics associated with the knees were compared, results showed differences in ankle dorsiflexion and plantar flexion angles across the knees. During early stance the RHEO knee and the Mauch SNS were associated with considerably greater foot compression and peak ankle plantar flexion angles. During mid-to terminal stance they showed significantly lower peak ankle dorsiflexion angles (Johansson, 2005).

The fact that greater plantar flexion was found to be associated with the RHEO knee and the Mauch SNS indicates that subjects were able to generate some amount of energy at toe-off. That could result in less compensatory movements such as increased hip extension. However, as previously mentioned the Mauch SNS was found to be associated with exaggerated hip work.

Activity of gluteus maximus and gluteus medius muscles were investigated using the EMG electrodes and accelerometers. The RMS value of the EMG recordings showed that the RHEO knee was associated with lower level of muscular activity compared to the C-Leg and the Mauch SNS. Comparison of the data from the accelerometer across the three prostheses showed that the Mauch SNS was associated with significantly higher RMS values of jerk about toe-off compared with the RHEO knee and the C-Leg. Authors claim that this difference indicates that the microprocessor knees enhance smoother transition from swing-to stance phase (Johansson, 2005).

Authors discuss the differences in the alignment of the RHEO knee and the Mauch SNS. The RHEO knee is generally aligned so the ground reaction force lies posterior to the knee joint axis, as in intact limb, and early stance stability and prevention of knee buckling is achieved by increasing the level of resistance. The Mauch SNS is generally aligned differently. The alignment is generally set so the ground reaction force lies anterior to the knee joint axis when the user is standing in a quiet position. This alignment prevents the knee to flex at heel strike and buckle when the prosthesis is loaded. The anterior alignment of the Mauch SNS enhances stability during early stance but during pre-swing this alignment slows down rapid knee flexion compared to the posterior alignment (Johansson, 2005).

Authors claim that the results from the study indicate that the microprocessor knees have clear advantages over the mechanical knee. Additionally they claim that the fact that the RHEO knee was associated with decreased metabolic cost compared with both hydraulic knees, indicates that the combination of the magnetorheological and the microprocessor technology has advantages over the hydraulic technology. Authors discuss that there are certain design factors that are likely to be key features to the success of the RHEO knee. The prostheses have different peak extension angles. When full extension of the RHEO knee is reached the joint is in zero flexion. For the C-Leg it is different, when the joint is fully extended the C-Leg assumes a slightly flexed knee. Authors claim that the difference in peak extension angle between the prostheses explains the fact that the RHEO knee was associated with greater heel compression, thus energy storage (Johansson, 2005).

As mentioned earlier, the developer of the RHEO knee is among authors of this study. Although the implementation of the study was generally good and the authors very experienced researchers there is no doubt that throughout the study the results from the study are interpreted in favour of the RHEO knee. The study fails to show statistical difference in most of the parameters investigated. However, authors translate results into improvements in function, smoother gait and increased energy storage, all associated with the RHEO knee.

Subjects were given ten hours to acclimatise to the prostheses, both the knee joint and the prosthetic foot. All subjects were long term microprocessor knee users´ except one subject that used the Mauch SNS prior to the study. All subjects used the same prosthetic foot throughout the study, and perhaps this foot was not the optimal choice of foot prosthesis to use either with the Mauch SNS or the C-Leg. Authors conclude that the RHEO knee offers improvements over the Mauch SNS mechanical knee. They conclude that from studying eight subjects and averaged data from nine trials in a gait laboratory where subjects only walked in a straight line. The study does not include how the RHEO knee would perform in real-life situations such as walking at variable walking speeds, walking up and down stairs and slopes. Additionally, the method used to investigate biomechanics in this study was using the Plug-In gait from the Vicon system. That can cause errors and will be discussed more detail in chapter 3.3.2.

2.10.2 C-Leg and Mauch SNS study

Following the reports by Johansson et al (Johansson, 2005), Segal et al (2006) studied the gait biomechanics of trans-femoral amputees when walking with the C-Leg and the Mauch SNS. Authors claimed that this was an independent study and no competition interests existed.

Eight trans-femoral subjects participated in the study and all subjects were used to using the Mauch SNS prior to the study. In this study subjects were given three months to adapt to the C-Leg. Nine control subjects were tested for comparison. Gait biomechanics were compared across the knees when subjects walked on controlled walking speed and each subject walked ten trials. Results were evaluated to test for significant difference among the prostheses, with statistical significance set at 5% (Segal, 2006).

Each subject had three testing sessions in the gait lab. First session involved collecting baseline data. Then subjects had three months to acclimatise to either of the prosthesis, Mauch SNS or the C-Leg, and participate in a test session. Again subjects were given other three months to acclimatise to the other prosthesis and then participate in a test session. When data from baseline study and Mauch SNS study were compared no significant difference was observed. Therefore further comparison was done, where results for the Mauch SNS were compared to the C-Leg results (Segal, 2006).

Comparison on kinematics across the knees showed that in stance phase the peak knee flexion angle did not differ between the prostheses. However, during swing phase the Mauch SNS was found to be associated with significantly greater peak knee flexion angle compared to the C-Leg and the intact controls (Segal, 2006).

Both prostheses were associated with significantly decreased knee moments compared with the controls and when ground reaction forces were studied results showed that both prosthetic knees were associated with decreased ground reaction force on the prosthetic limb. When using the Mauch SNS the ground reaction force was lower when compared with the C-Leg but authors discuss that can be a result of smaller step length when subjects were using the C-Leg (Segal, 2006).

When sagittal and coronal plane knee moments were studied no significant difference was observed between the prostheses. Authors discuss that to further investigate the knee moments a study involving variable walking speeds and not only straight line walking is needed (Segal, 2006).

When sagittal plane hip powers were studied, results did not show significant difference between the prostheses and thus contradict previously reported results by Johansson et al (Johansson, 2005). Authors discuss that difference in results can be a result of insufficient acclimation time in Johansson et al study (Segal, 2006). When stance phase knee flexion was studied results showed that both prostheses were associated with decreased stance flexion compared to the controls (Segal, 2006).

Authors conclude that the study only demonstrated minor differences between the gait biomechanics. Although results did not show clear advantage and better performance of the C-Leg, the feedback from subjects was very positive. Seven out of eight subjects chose to continue using the C-Leg (Segal, 2006).

All subjects in this study were long term Mauch SNS users but were given three months to acclimatise to the C-Leg. The fact that subjects were not loading the prostheses fully and as a result a decrease in ground reaction force on the prosthetic side was observed could be associated with lack of confidence when subjects were using the C-Leg. Despite having three months to acclimatise, that can be insufficient time to acquire full confidence in walking and learn how the prosthesis functions and trust it to bear full weight without buckling.

The method used in this study has advantages, where baseline measurements were done before the prostheses were compared. Also in this study three test sessions were conducted, therefore subjects' fatigue should not have affected the results.

In this study the prosthetic foot type did vary across the subjects. Subjects were prescribed foot prosthesis that would fit for the knee prosthesis and their individual needs. That has both certain advantages and disadvantages. Using the recommended prosthetic foot type for each knee joint lead to a fair comparison across the knees, however, changing parameters makes it more difficult to evaluate the impact that is due to the different knee technology.

Subjects chose different self-selected walking speed related to which prosthesis they were using, thus the authors investigated gait biomechanics during controlled walking speed. That makes the comparison across the prostheses more relevant, as changes in gait biomechanics observed were not induced by changes due to variations in walking speed. However, when the walking speed is controlled authors cannot be sure that this particular walking speed was the optimal speed for subjects and the fact that subjects chose faster walking speed when using the C-Leg can indicate that the most comfortable speed for the C-Leg is not the same as for the Mauch SNS.

2.10.3 Comparative study on various prostheses

Following the Johansson et al report, Bellman et al (2010) studied the functional differences between four prosthetic knee joints, the C-Leg and three different types of microprocessor controlled knees, the RHEO knee, the Energy knee and the Adaptive 2. Authors of the study work for the manufacturer of the C-Leg, Otto Bock, thus the study was not independent.

Nine unilateral, trans-femoral amputee subjects participated in the study. All subjects were experienced with the C-Leg prosthesis and used it as their daily prosthesis prior to the study. Subjects were given two hours to familiarise with the unknown prosthesis, authors claim that these two hours were sufficient time, where all subjects were experienced in walking with various prostheses.

Level walking at various speeds was studied to investigate swing phase feature and metabolic cost associated with the different prostheses. Stairs and 10° ramps descending and stumble and fall recovery were also studied. The study was conducted in two phases, first all prostheses were tested except the RHEO knee. The RHEO knee was tested a year later. When the RHEO knee was tested metabolic energy consumption for the C-Leg was repeated (Bellman, 2010).

Same socket and prosthetic foot was used throughout all testing and the different prostheses were tested in random order. Motion analysis system and force plates were used to record kinetic and kinematic data for both amputated and intact sides. Energy consumption was measured while subjects were walking on a treadmill. Measurements were done for comfortable self-selected speed and for slow and fast walking speeds. Results were compared with statistical significance set at 5% (Bellman, 2010).

When metabolic energy consumptions associated with the prostheses were compared, results showed no difference between the C-Leg and the Energy knee. Comparison on the C-Leg and the RHEO knee revealed that generally the C-Leg was associated with slightly lower energy consumption; however the difference was only significant at self-selected walking speed (Bellman, 2010). These results contradict reported results by Johansson et al (2010).

The mean peak knee flexion angles were determined across the knees during level walking. With increased walking speed all prostheses showed increased peak knee flexion angle. Comparison across the knees showed that when using the C-Leg the maximum flexion angle did not increase as much as for the other prostheses. Authors discuss the two most important features of microprocessor-controlled swing phase during level walking. The device has to provide appropriate resistance in late swing phase. To enhance natural gait pattern the maximum knee flexion angle should range between 60° to 65°. Authors claim that the C-Leg prosthesis's design is the most suitable (Bellman, 2010).

When subjects descended stairs and ramps they had the option to use handrail. For the subjects that walked down stairs and ramp with step-over-step pattern the performance of the knee joint was evaluated. Both intact side and prosthetic side were examined. Stance phase resistances provided during single limb stance on the prosthetic side was examined and on the intact side the ground reaction forces were investigated (Bellman, 2010)

When the prosthetic side was studied results showed that the maximum knee flexion moments were associated with the C-Leg when descending stairs and the ramp. When walking down the steps the RHEO knee and the other two joints except the C-Leg were slightly flexed prior to stair contact (Bellman, 2010).

When descending stairs, prosthesis must provide controlled flexion and at the instance when stepping onto a stair the knee joint must be fully extended so user can safely place the prosthetic foot on the stair. The RHEO knee was still extending at the instance before stair contact, authors claim that can cause difficulties to the user when it comes to placing the prosthetic foot on the step. Also authors discuss that the RHEO knee was associated with exaggerated hip movements on the prosthetic side. The movement was needed to encourage knee extension in terminal swing to prevent the knee from buckling when loading for next step (Bellman, 2010).

Authors discuss the method used to control the level of resistance in the RHEO knee. They claim that the fact that the level of resistance depends on the axial load on the prosthesis can cause the prosthesis to collapse if the users steps gently with the knee flexed. The inadequate load on the prosthesis will result in improper resistance and the user is in risk of falling (Bellman, 2010).

Examination on the intact side revealed that when descending the stairs the C-Leg was associated with the lowest recorded maximum ground reaction force. When using the RHEO knee the ground reaction force increased on the intact side but the difference across all knees was not significant (Bellman, 2010).

Knee angles and external knee and hip moments were used to evaluate the performance of the prostheses regarding fall prevention. To investigate how subjects used the knee joints to avoid falling, video recordings that showed compensatory movements were used. Results showed that when using the C-Leg and the Energy knee, movements like stopping and side stepping were achieved most effectively. The RHEO knee was associated with increased compensatory movements to avoid falling (Bellman, 2010).

To investigate how subjects dealt with stumbles wearing the different prostheses the authors studied how subjects dealt with clearing the toe off the prosthetic foot and compensatory movements to avoid falling. Results showed that if subjects experienced an interruption during swing extension, particularly between 10°-35° knee extension, subjects wearing the C-Leg managed to keep walking but wearing the RHEO knee subjects showed increased compensatory responses when trying not to fall. Also the RHEO knee did not show as strong knee extension after the user stumbled as the other prostheses. Additionally, in a flexed position and under load bearing, subjects showed strong compensatory movements to avoid falling and the knees buckled if the flexion angle exceeded 30° (Bellman, 2010).

When subjects had to stop or sidestep due to interruption they have to depend on the prosthetic knee joint´s resistance. If the knee joint fails to provide the adequate resistance there is a high risk that the prosthesis will buckle under the user´s weight. Authors claim that results from the study showed that when subjects were using the RHEO knee the flexion resistance was unreliable and the prosthesis had to be stabilised by using the extension muscles of the residual limb (Bellman, 2010).

As previously mentioned inadequate stance phase knee flexion is a common gait deviation among trans-femoral amputees. To allow stance phase knee flexion the prosthetic knee joint has to be able to go from low extension resistance to high flexion resistance prior to the initial contact of the prosthetic foot. Authors claim that the control of level of resistance offered by the C-Leg has advantages over the RHEO knee. They claim that the C-Leg is able to provide at the same time low extension resistance and high flexion resistance. Authors claim that with the RHEO knee a high level of resistance affects both the knee flexion and the extension (Bellman, 2010).

Authors discuss the limitations of the study. The facts that all subjects were used to the C-Leg and that it is impossible to blind the participants. The subjects in this study were identical, all relatively young healthy trans-femoral amputees. Further investigations have to be done to study typical elderly subject and evaluate if they can perform better with the use of the technology offered by the microprocessor controlled knees (Bellman, 2010).

Authors conclude that the fact that no significant difference was found in metabolic energy consumption across the knees shows that when functional difference between microprocessor controlled knees and mechanical knees is investigated, energy expenditure should not be the parameter that is focused on (Bellman, 2010).

It is obvious that this study is not an independent study. Similarly to Johansson et al report (Johansson, 2005), results tend to be interpreted in favour of the knee that authors are linked to, and this case the C-Leg. The study fails similarly to Johansson et al to show statistical difference in most of the parameters investigated. However, authors translate results into improvements in function, increased safety associated with stair descending and decreased risk of falling, all when using the C-Leg.

All subjects were long term C-Leg users and were given two hours prior to the study to acclimatise to the other prostheses tested. Two hours to get used to three different prostheses is most likely insufficient time. Especially, as this study was challenging, investigating stair descend and stumble and falls. Authors concluded that using the C-Leg will decrease risk of falling and subjects were more likely to be able to avoid fall and continue walking when using the C-Leg. It is predictable that amputees will perform better in difficult situations using a prosthesis that they are used to. When the subjects experienced a disturbance and had to deal with it and try as they could to avoid falling it makes a great difference to wear a prosthesis that they are fully familiar with, know its' boundaries and trust that it will not buckle under their weight.

2.10.4 Balance and gait

Kaufman et al (2007) studied the gait and balance of fifteen trans-femoral amputees when walking on the C-Leg and the Mauch SNS, all subjects were used to the Mauch SNS prosthesis. The aim of the study was to evaluate if the microprocessor controlled C-Leg enhances balance and more normal gait.

Gait analysis using motion capture system and force plates was performed and the peak knee extension moment during stance was examined. To study and compare the balance on the prostheses a test was conducted where sensory components of balance were assessed, the eyes, muscles and joints, and vestibular organs. These sources sense changes in the environment and a movable force platform that could rotate was used with movable visual surround to make changes in subjects´ environment. Subjects had to make postural adjustments to maintain balance and stability and the position of centre of mass was used to determine if subject was in balance or not. Results showed that wearing the C-Leg the balance was significantly better (Kaufman, 2007).

Results showed that when subjects were wearing the Mauch SNS, they maintained the prosthesis stable by controlling their walking pattern the way that the ground reaction force was maintained in front of the knee, knee was kept hyperextended and therefore unlikely to buckle. When subjects were wearing the C-Leg, the gait was more natural. In loading response ground reaction force was behind the knee joint like in normal gate and the knee slightly flexed (Kaufman, 2007).

Authors conclude that results from this study show that there is a significant difference in gait when wearing a microprocessor knee or a mechanical one and that the use of a microprocessor-controlled knee improves gait efficiency. Authors agree with results reported by Johansson et al (2005) but contradict results reported by Segal et al (2006). Authors discuss that reason for the inconsistency in results can be explained by insufficient time given to subjects to acclimatise to the C-Leg in Segal et al study, authors claim that ten hours is insufficient time to allow subjects to get used to the prosthesis. In this study the subjects were given four and a half months to acclimatise, which should be appropriate time referring to that study reported that 3.5 months is an average time required to get used to a microprocessor knee (Hafner, 2007; Kaufman, 2007). Another explanation for the contradictory could be the alignment of the microprocessor prosthesis was not optimised during Johansson et al testing and could have affected the results.

Authors discuss that results from study show significant difference between the two types of knees but do not indicate or identify the characteristics of patients, which can benefit from using of microprocessor controlled knee Kaufman et al (2007). The study conducted by Kaufman et al was funded by Otto Bock Health Care, Inc. which is the manufacturer of the C-Leg.

2.11 Discussion

This literature review indicates that overall there is a major lack of independent studies that assess the performance of the RHEO knee. Reported studies have some shown that the microprocessor knees are associated with lower energy expenditure and some reported positive impact on subjects´ gait.

Nevertheless, many of previous studies have limitations that affected reported results. Generally they involved few subjects and the studied group was relatively homogenous across all studies, relatively young, active and healthy individuals. Possible moderately active and relatively older amputees could benefit from these devices and help to improve their level of activity. In all independent studies the subjects were unfamiliar with the microprocessor knees and long term users´ of mechanical knees. Being a long term user of a mechanical knee could have an impact on the performance when first using a microprocessor controlled knee. Breaking the habit of keeping the prosthetic knee locked in extension throughout the stance phase can be difficult. To benefit the most from the microprocessor technology the user has to be confident in loading the prosthesis and use the prosthetic limb, which takes time and practice.

Major part of the studies focused on energy expenditure but kinematics and kinetics were not assessed in any details. Individuals that suffer from muscle weakness, great gait deviations and do generally struggle with walking could benefit from a device that would enhance gait efficiency. However, the criteria to be a candidate for microprocessor controlled knee are that the amputee is very active and has the ability to use the prosthesis fully. Some of the previous studies have shown trend in slightly decreased energy expenditure, but how much a relatively fit and active individual benefit from that would is a question. The prosthesis should be prescribed to amputees that will gain the most benefit from the technology, but what most amputee individuals value the most is enhanced safety and confidence.

Most previous studies have not investigated how the microprocessor knees perform in “real life” situations. Many studies have implemented tests where treadmills are used but only few have addressed the performance of the prostheses in situations like walking down ramps and stairs and deal with other obstacles that trans-femoral amputees experience daily.

A study that investigates the performance of the RHEO knee in daily life situations and focuses on other more relevant parameters than metabolic cost is needed. That emphasises the importance of this study.

3 Project Methodology

3.1 Introduction

This study assessed kinetic and kinematic parameters of the hip, knee and ankle joints of the intact and prosthetic limb of one trans-femoral subject. Tests were implemented and the results used to evaluate how the two tested devices, the RHEO knee and the 3R80, performed during activities that amputee might undertake in the course of their daily lives, such as walking up and down slopes and stairs and also biomechanics will be studied during level walking.

The study involved testing using a gait laboratory and the resulting data were used to evaluate the performance of the two different knee prostheses designs and compare them. Both limbs of the subject were evaluated, as the body works as a whole, and to evaluate the general impact of different prosthetic knee technologies on kinetics and kinematics parameters both sides have to be considerate.

Ethical approval for the study was sought and obtained from The West of Scotland Research Ethics Service for Evaluation of microprocessor controlled prosthetic knee mechanisms. REC reference: 11/WS/0109 and protocol number: EP/F50036X/1.

3.2 Literature Review

For the literature review, journals were searched to identify relevant publications to include in the literature review chapter. Relevant citations discovered from journal searches, such as citations to book sections or other journals, were also tracked and studied. All sources included in the literature review were read, evaluated and included into the final review if they were considered relevant.

3.3 Gait analysis

Gait analysis is widely used as a tool to study how individuals walk and to determine important parameters such as forces on joints and joint kinematics. Gait analysis is commonly used as a tool to evaluate and compare different prostheses components. Gait analysis has been used in studies that have investigated the performance of microprocessor controlled knees. Gait biomechanics have been compared where subjects walk using different prostheses, microprocessor knee and mechanical knee and from data the performance of the knees has been determined.

3.3.1 Motion and force analysis systems

Motion and force analysis systems are useful tools to investigate individual´s locomotion and identify walking abnormalities. With the use of modern motion and force analysing systems, three dimensional motion data can be acquired and kinematics and kinetics investigated. Where this study assessed kinetic and kinematic differences between the RHEO knee and the 3R80 a motion and force analysis system was used to acquire data. Typical motion and force analysis system comprises of motion capture system and force plates. The tests implemented in this study were carried out using the Vicon motion capture system and KISTLER force plates.

3.3.1.1 The Vicon system

General gait laboratory comprising the Vicon motion capture system includes a room that has a defined volume that is surrounded by number of high-resolution cameras that are used to capture data. The cameras in the Vicon system are infrared cameras; around the lens of each camera are rings of Led strobe lights. To capture data for individual´s gait or other activities a set of reflective markers is attached to subject´s body in relevant positions (Vicon, 2012).

Inaccurate location of joint centres, skin movements and markers used in the gait laboratory are parameters that can cause errors and inaccuracy. To determine accurately the location of the joint centres can be difficult, especially the hip joint located deep in the body. The positioning of the markers placed on subjects segments and skin movements can also cause inaccuracy (Zahedi, 1987).

Before starting the actual gait analysing study the system has to be calibrated. The system needs both static and dynamic calibration. To dynamic calibrate the system an individual walks around the workspace holding a calibration wand. The cameras detect the markers situated on the wand and collect data. From data the system verifies the relative positions and orientations of the cameras and prevents that cameras will fail to detect data. When dynamic calibration is completed the calibration wand is placed in the centre of the working space to static calibrate the system. With static calibration the system determines the origin and the orientation of the three-dimensional coordinate system, direction of the axis. Data from cameras during measurements and information from calibration are combined to reconstruct three dimensional motion data (Tebbut, 2002).

During measurements, as the subject walk and moves within the volume surrounded by the cameras, a video signal is created by reflection of the markers. A hardware component, called Datastation, controls the cameras and collects all data, both from cameras and force plates. While the system is running and data is being captured the Datastation generates two dimensional coordinates of each marker placed on subject´s segments. Workstation is the software that controls the Datastation, and recorded data is forwarded into a computer comprising the Workstation software (Tebbut, 2002).

The Workstation software then processes the raw data into two dimensional data from the Datastation. It calculates the three dimensional coordinates of each marker using data from cameras, calibration parameters and reconstruction parameters that are set for each study. The Workstation calculates the trajectories of the markers, and in the software the three dimensional digital motion can be seen. The trajectory displayed on the screen is the reflective markers, seen as white dots moving. To analyse the trajectory and data obtained the markers are labelled. Each marker used on a subject is labelled, and therefore identified (Tebbut, 2002).

When data has been reconstructed into three dimensions, it can be downloaded into another program for further gait analysis. The Plug-in Gait program is a part of the Vicon systems, and it is a conventional gait model that can be used for gait analysis (Tebbut, 2002).

3.3.1.2 Kistler force plates

Kistler force plates are a force platform that measures ground reaction forces acting on the body during walking. When the foot touches the plate the direction and the magnitude of the force is recorded throughout the stance phase. The force plates are synchronised with the Vicon motion capture system and ground reaction force measurements are done at the same time as capturing data with the cameras.

3.3.2 Motion analysis system data processing

To analyse data acquired with the Vicon system a simple software solutions can be used, where biomechanical analysis are performed by using customisable models. The conventional gait model, the so called plug-in-gait, from the Vicon system is a full-body-model that can be used to easily implement biomechanical models. Using the plug-in-gait, angles, moments and vectors can be calculated in simple operation, where all complex mathematics is done by the software. For researchers that do not have any engineering or mathematical background this software can be very useful, as designing a biomechanical model can be complicated (Vicon, 2012).

Using the Plug-In-Gait software can cause multiple errors in calculated results, where the software makes assumptions that are not relevant to individual subjects´. Additionally the Plug-In-Gait marker set may not be optimal for all subjects. The biomechanical model is a general model that is used to model all kinds of movement. The software does not distinct between specific characteristics of subjects, such as if subjects that are being tested are intact individuals, uni- or bilateral amputees and so forth. The marker set generally has thigh markers which is not optimal when the subject is an amputee and has a socket covering the thigh that moves and causes errors.

Therefore this conventional gait model was not used for this study. The biomechanical model used is an algorithm written in MATLAB by Anthony Crimin, a research student at the University of Strathclyde. The MATLAB algorithm will be discussed further in chapter 3.4.7.1.

3.4 Implementation of the study

3.4.1 Components

To assure that differences observed during the study were caused by the different knee prosthetic designs the subject used the same prosthetic foot, socket and shoe throughout the study. Prior to testing the subject was given time to acclimatise to the prostheses and surroundings. The tests were conducted with the RHEO knee and the 3R80 knee in combination with the Echelon prosthetic foot. Throughout the study the subject was wearing Lycra shorts and a t-shirt. That way errors caused by clothing movement, thus marker movements was minimised.

3.4.2 Subject

One subject with unilateral trans-femoral amputation gave his consent to participate in this study. The participant is 53 years old male, weighs 86 kg and underwent amputation on left limb about 21 years ago, due to trauma. The subject is used to prosthesis ambulation and can ambulate at least at a K3 level. That means the subject has the ability to traverse most barriers found in real life environment, such as stairs and inclines. The subject routinely uses the 3R80 knee joint and ischilateral socket. The subject has walked with various prosthetic knees, as he is used to participating in studies and therefore able to acclimatise to different prosthesis within a relatively short time. The subject has no other musculoskeletal problems or other disorders.

3.4.3 Test situation

The tests were conducted in the gait laboratory at the University of Strathclyde. The subject was asked to commit to two test sessions, one for the 3R80 prosthesis and another one for the RHEO knee. The conditions in the tests simulated three real-life situations, where subject was asked to walk on level ground, up and down slope (7° incline) and up and down stairs. Data was recorded for both the prosthetic side and the contralateral side for all situations. In order to minimise the effects of fatigue the subject was given sufficient time to recover as needed between trials.

3.4.4 Prosthetic alignment

Prior to the test sessions a certified licensed prosthetist fitted the subject with the prostheses and handled alignments and adjustments needed on the knees and the prosthetic foot. The prosthetist was presented throughout the test sessions in the gait laboratory.

A clinical specialist prosthetist at Össur UK brought the Össur knee that was used for the study. He presented background information and demonstrated the RHEO KNEE. The subject was fitted with the knee at the National Centre for Prosthetics and Orthotics and an initial dynamic assessment was performed. The following initial dynamic assessment for the RHEO knee is recommended by Össur and is an important factor in establishing trust on the prosthesis for a new user. The evaluation was performed in parallel bars, ramp and stairs and is classified into six major steps according to the manufacturer procedure:

1. *"Evaluate the user's level of voluntary control by having the user walk with the RHEO KNEE power off. The user should have the ability to maintain knee stability with power off.*

2. *Use hip extension to maintain stability into stance.*

3. *Have the user experience the response, the roll-over, of the Flex-Foot.*

4. *Load the heel and allow the knee to flex, thus experience the resistance provided by the knee.*

a. *Control the rate of the knee flexion with hip extension.*

b. *Perform rapid loading of the prosthesis with use of the parallel bars.*

5. *Train the user to maintain normal step length on the sound side.*

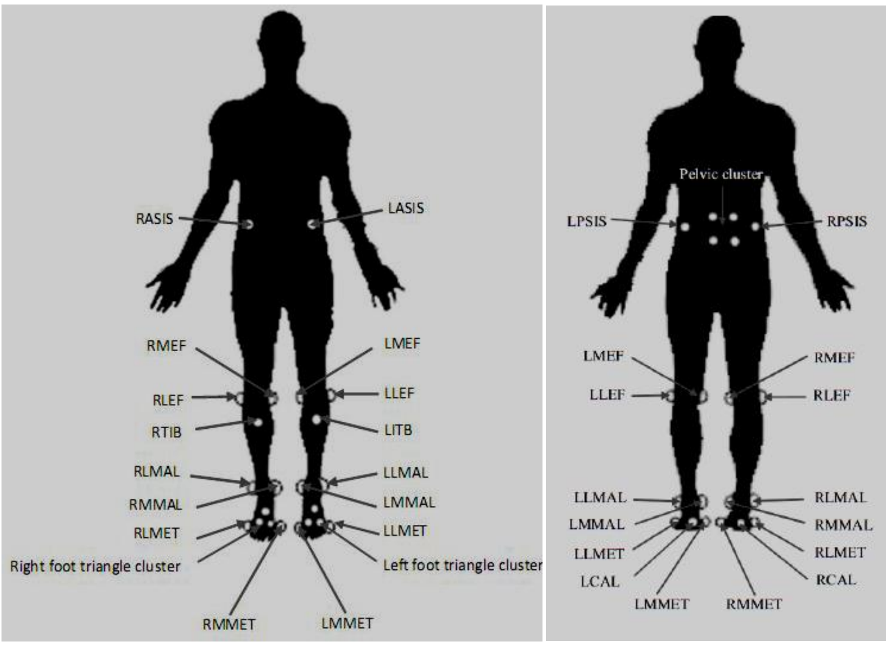
6. *Sit-to-stand. Have the user sit down using the stance resistance of the RHEO KNEE."* (Össur, 2011).

The assessment involved these steps, apart from step three where in this study the Echelon foot was used in combination with the RHEO knee. During the assessment the subject had few hours to adapt to the knee as simultaneously the knee collected information about its gait. The clinical prosthetist from Össur UK recommended that the subject would wear the knee for a week to allow it to adapt properly but due to circumstances that was not possible.

3.4.5 Marker set and data collection

3.4.5.1 Kinematic data

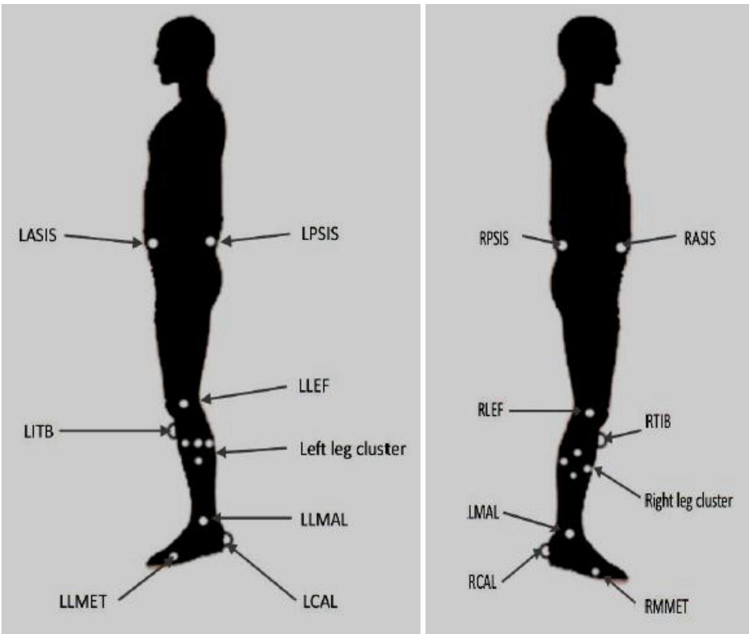
Kinematic data was collected by using the Vicon MX system, where twelve infrared cameras captured the trajectory of retro reflective markers placed on subject´s segments. A lower limb marker set was used, which is designed for gait analysis on amputee subjects in mind. This marker set determines the knee joint centre from the tibial frame of reference, which is advantageous when the participant has a socket covering the thigh which can cause errors due to movements (Crimin, 2012). Figures 3-1 and 3-2 show the positioning of the reflective 14 mm spheres markers placed on the subject.



A B

Figure 3-1 Frontal (A) and posterior (B) view of the marker set (Crimin, 2012)

Figure 3-1 shows the frontal and posterior view of the marker set. Table 1 provides further descriptions of each marker and the abbreviations presented in figures 3-1 and 3-2.



A B

Figure 3-2 Left sagittal (A) and right sagittal (B) view of marker set (Crimin, 2012)

In figure 3-2, the sagittal right and sagittal left view of the marker set is presented. In order to distinct between right and left leg, there is a difference in the alignment between the left and right leg clusters. This aids further analysis of the data generated in the test. Table 1, represents anatomical description of the marker set positioning.

Table 1 Anatomical description the marker set positioning (Crimin, 2012)

Marker	Placement and description
LASIS	Left anterior asis marker placed directly over the left anterior superior iliac spines
RASIS	Right anterior asis marker placed directly over the left anterior superior iliac spines
LPSIS	Left posterior asis marker placed directly over the left anterior superior iliac spines
RPSIS	Right posterior asis marker placed directly over the left anterior superior iliac spines
LLEF	Left lateral epicondyle marker placed directly over epicondyle
LMEF	Left medial epicondyle marker placed directly over epicondyle
LTIB	Left tibial tuberosity marker placed directly over tuberosity
RLEF	Right lateral epicondyle marker placed directly over epicondyle
RMEF	Right medial epicondyle marker placed directly over epicondyle
RTIB	Right tibial tuberosity marker placed directly over tuberosity
LLMAL	Left lateral malleolus marker placed directly over malleolus
LMMAL	Left medial malleolus marker placed directly over malleolus
RLMAL	Right lateral malleolus marker placed directly over malleolus
RMMAL	Right medial malleolus marker placed directly over malleolus
LCAL	Left hind foot marker placed directly over left calcaneus
LLMEL	Left lateral metatarsal marker placed at head of the fifth metatarsal
LMMET	Left medial metatarsal marker placed at head of the first metatarsal
RCAL	Right hind foot marker placed directly over left calcaneus
RLMET	Right lateral metatarsal marker placed at head of the fifth metatarsal
RMMET	Right medial metatarsal marker placed at head of the first metatarsal

As seen in Table 1, there are total of twenty markers in the marker set, ten for each leg and there is the same setup for both the prosthetic and the intact leg.

3.4.5.2 Kinetic data

Kinetic data was collected using Kistler force plates. During level walking the ground reaction force acting on the body was measured with four Kistler force plates embedded in the floor. To measure the ground reaction force while subject was walking up and down the slope a force plate was embedded in the ramp. Simultaneously as the kinetic and the kinematic data was collected videos of the lower limbs were taken with fixed cameras. Figure 3-3 shows the layout of the motion analysis system in the gait laboratory.

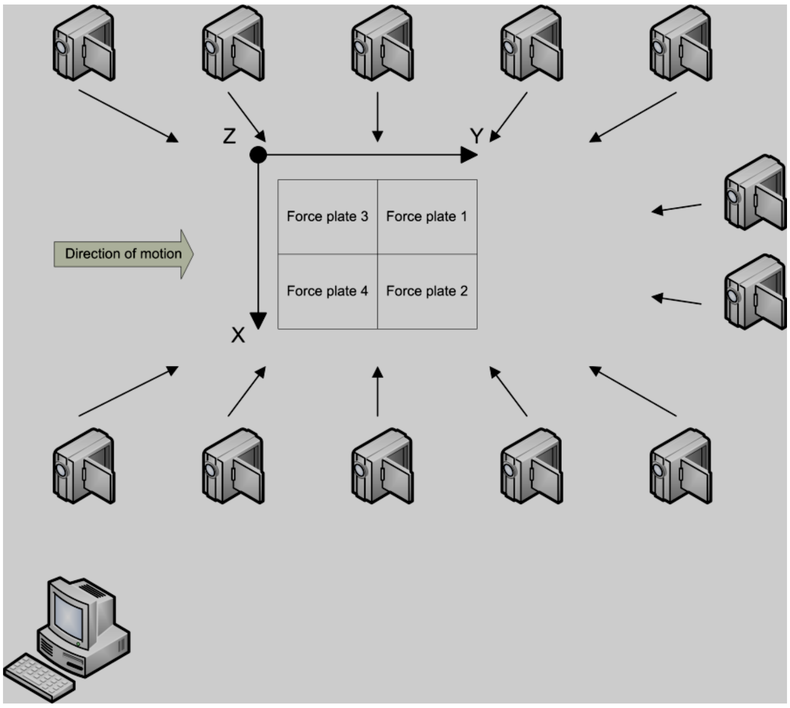


Figure 3-3 Layout of the motion analysis system in the gait laboratory

The number of trials needed depended on how many trials were needed to get quality data where the subject fully stepped on the force plates. Figures 3-4 and 3-5, show still images from the video recordings. In figure 3-4 the force plates embedded in the floor can be seen and how the subject fully stepped on two of the force plates.

Figure 3-4 Subject during level walking

Figure 3-4 shows images from the recordings during up and down slope walking. The single force plate embedded in the ramp can be seen and the subject fully steps on the plate with the foot.

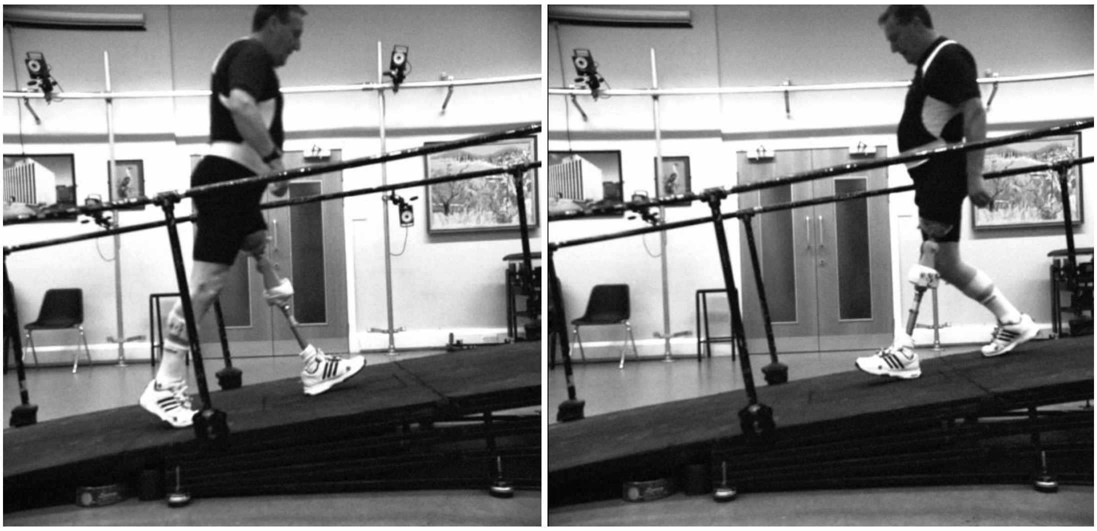


Figure 3-5 Subject during up and downhill walking

3.4.6 Study procedure

Figure 3-6 shows the structure of arrangements from prosthetic knee fitting to data capturing.

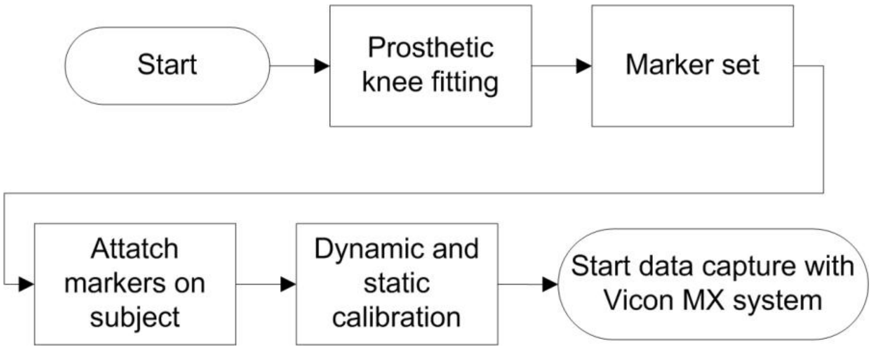


Figure 3-6 Block diagram shows structure of arrangements before data recording starts

Before starting the data collection the Vicon system had to be dynamically calibrated, details on how calibration is performed can be found in chapter 3.3.1.1. When the subject had successfully been fitted with the prosthesis he was prepared for the test by placing the marker set on. For the static calibration the calibration the subject was placed in the volume where all calibration markers were detected by the cameras. Each calibration marker was defined according to anatomical position and from the position of the markers the local hip, knee and ankle frame of references were created. The local coordinate system was used to describe the position within the segments during movement.

3.4.7 Data processing and analysing

Figure 3-7 shows the workflow from capturing data to data output.

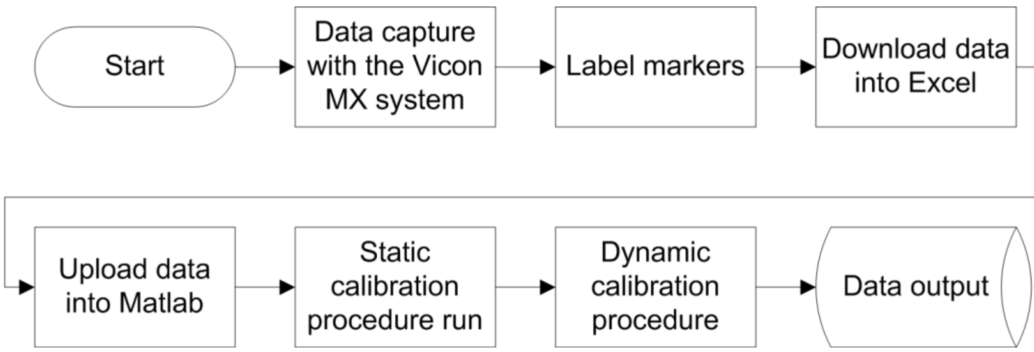


Figure 3-7 Structure of data processing

When data had been successfully captured parameters such as weight of the prosthetic knee and moment of inertia were investigated to use for calculations. The markers were then labelled as described in chapter 3.3.3.1 As discussed in chapter 3.3.2 the biomechanical model used for data analysing is a MATLAB algorithm. The algorithm was used to calculate three-dimensional kinetics and kinematics.

3.4.7.1 MATLAB algorithm

The local and global coordinate systems had to be defined. The local coordinate systems were determined for each segment and used to describe the position within the segments during movement. Once the local reference system had been determined the segment angles could be found. The Global coordinate system that was used was a right hand system where X direction is perpendicular to the frontal plane of the segment, Y direction is perpendicular to the horizontal segment plane and the Z direction is perpendicular to the sagittal plane (Crimin, 2012).

The MATLAB algorithm was used to calculate hip, knee and ankle joint intersegment forces and moments using the measured external ground reaction force acting on the joints and the segment trajectories (Crimin, 2012). Figure 3-8 shows the flowchart for the inverse dynamics methods that were used to calculate the intersegment moments during walking.

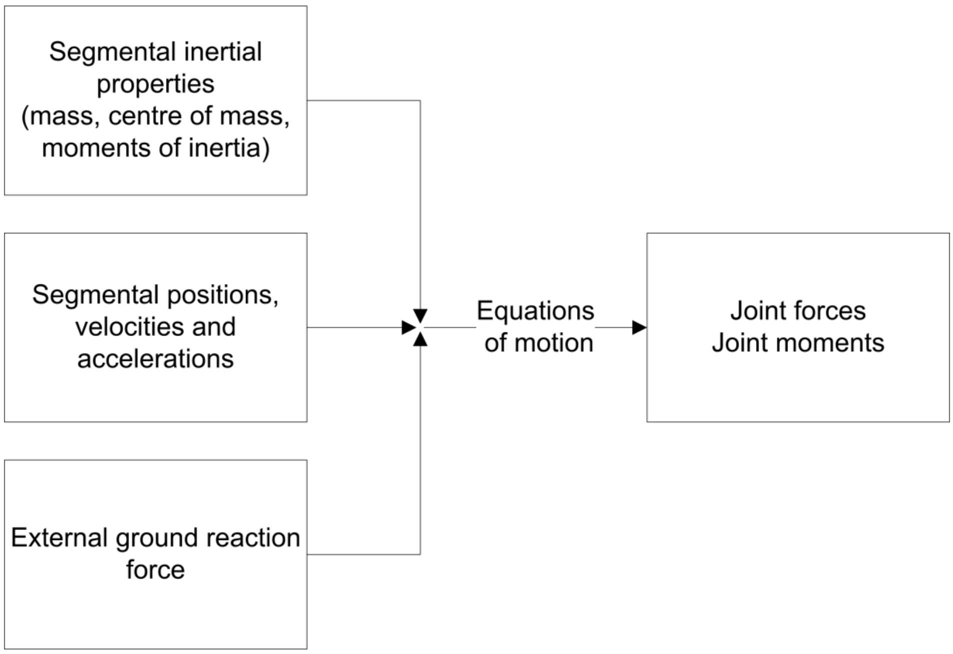


Figure 3-8 Inverse dynamics method

Using the inverse dynamics method to evaluate intersegment forces and moments the segmental displacement data obtained were differentiated twice to obtain acceleration data. After inertial characteristics of the segment were estimated, joint moments could be calculated since they were the only unknowns in equations of motion (Robertson, 2004).

3.5 Limitations of the study´s implementation

The main limitation of the study´s implementation is the fact that the subject got limited time to get used to the RHEO knee prosthesis and could have resulted in a worsen performance. Ideally the subject should have been given up to months to acclimatise to a new prosthesis but because of the circumstances that was not possible. The RHEO knee was a loan from the manufacturer and was returned back after the study. The socket used throughout the study was a rigid socket made ofa brittle plastic material and not intended for use outside the laboratory environment.

The subject is a long term non-microprocessor knee user and that could have affected his performance when walking with the RHEO knee. Additionally, the Echelon prosthetic foot that was used throughout the study was also a foot that the subject was not used to.

As discussed in chapter 3.4.1 the same prosthetic foot, socket and shoe were used throughout the study. However, to be able to use the same socket and the same alignment across the knees the RHEO knee joint had to be positioned higher up due to the different design of the prostheses. The higher position of the knee joint makes the knee more stable but it was decided that changing one variable, the height, would be more preferable than changing two variables, the socket and the alignment.

One subject participated in the present study. Due to limited time and the size of this project, detailed gait analysis on one subject using two different prostheses was considered relevant. Therefore the results from the study were focused on the impacts that the different prosthetic knee designs had on this particular subject.

4 Tests Results

The joint angle measures were considered as the angular position between two segments of the body, one segment relative to another in the sagittal plane. The hip joint angles were measured as the angles between the thigh and the pelvis and the knee joint angles the angle between the shank and the thigh. The hip and knee joint angular positions were assumed positive in flexion and negative in extension. The ankle angles were measured as the angles between the foot and the shank segments, positive in dorsiflexion and negative in plantar flexion. It should be noted that comparison on results from the present study and other studies is relevant; however, authors choose different frame of reference so comparison on differently scaled curves can show similar range of motion obtained using different reference system.

The joint moments are all in the sagittal plane, with positive direction indicating flexor hip moment and extensor knee moment and negative direction indicating extensor hip moment and flexor knee moment. The positive and negative directions indicate dorsiflexion and plantar flexion moments, respectively. The subject´s prosthetic limb is on the left side and the intact limb is on the right side, thus, data referring to left side are referring to the prosthetic limb

Data from every gait cycle were normalised from initial heel contact to the next heel contact of the same foot. Average data from all trials and 95% upper and lower confidence intervals are displayed. For the results and discussion the average data from trials was used.

In the following sections of this chapter, the results from studies on the RHEO knee and the 3R80 are discusses coherently. First each knee is considered with respect to the intact limb and distinction is made for each element. This is then followed by comparison of the performance between the two knees

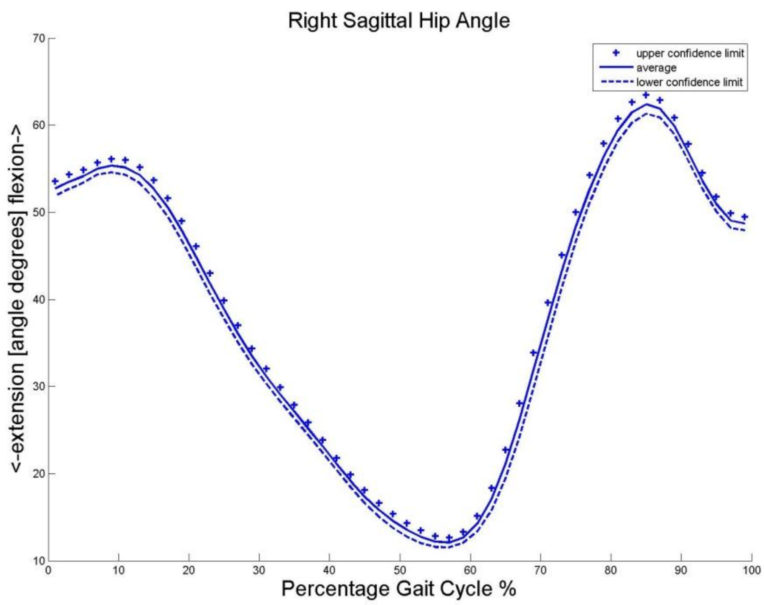
The intention was to investigate the performance of the knees during stair descending. It was assessed in the gait laboratory, however, the results obtained were not considered usable. The number of steps was not sufficient to obtain quality data and it was not possible to make any changes on the environment in the gait laboratory to get more number of steps. The subject used step-over-step pattern when descending the stairs but in real life he would not choose that pattern. The author of the present study decided in consultation with the project´s supervisor to exclude the assessment on stair descending from the project.

4.1 Level walking

4.1.1 Joint angles – Hip – 3R80 & RHEO

Figure 4-1 represents right (A) and left (B) sagittal hip angles for the 3R80.

(A)



(B)

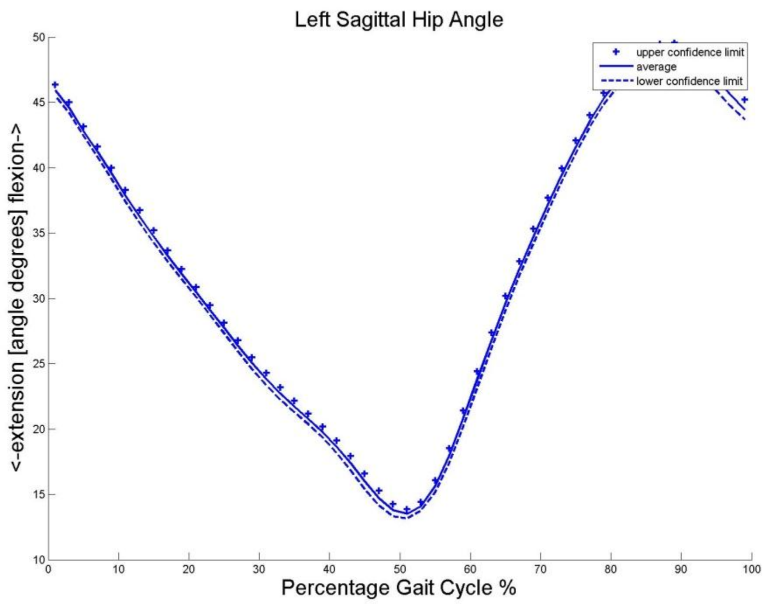
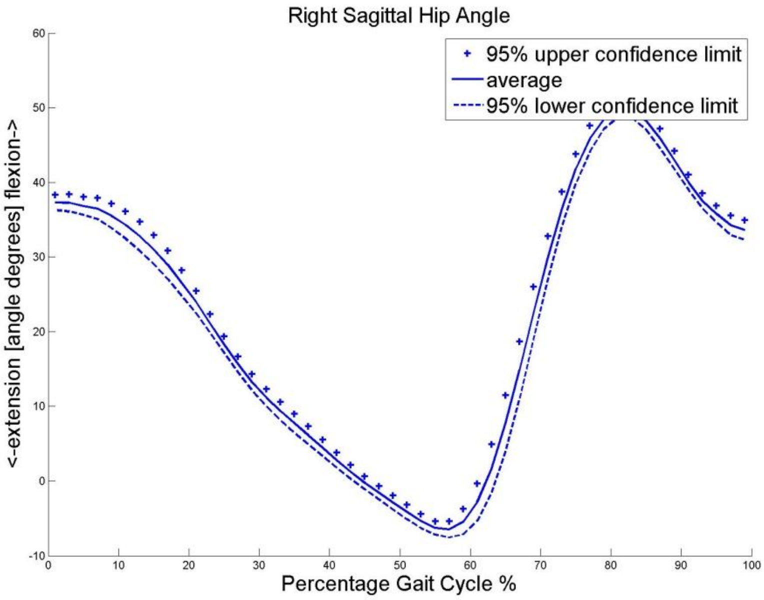


Figure 4-1 Intact limb (A) and prosthetic limb (B) hip angles during level walking on the 3R80 knee

The trajectories for both joints were similar with hip extension peaks occurring around push off and flexion motion during the swing phase. Right after heel strike the hip joint on the intact limb flexed 3°. Both joints were in extension throughout most part of of the stance phase with peak extension at about 55% of the gait cycle for the intact limb and 50% for the prosthetic limb. At that instant the hip joints had extended 40° and 35° respectively. After push-off both joints flexed, with peak flexion occurring at terminal swing, with trajectory of 50° for the intact limb and 40° for the prosthetic limb.

Figure 4-2 represents right (A) and left (B) sagittal hip angles for the RHEO knee.

(A)



(B)

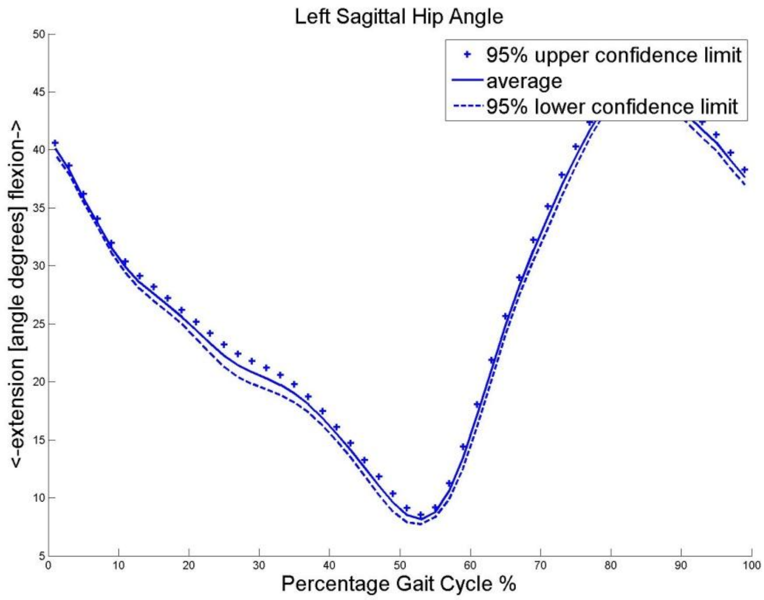


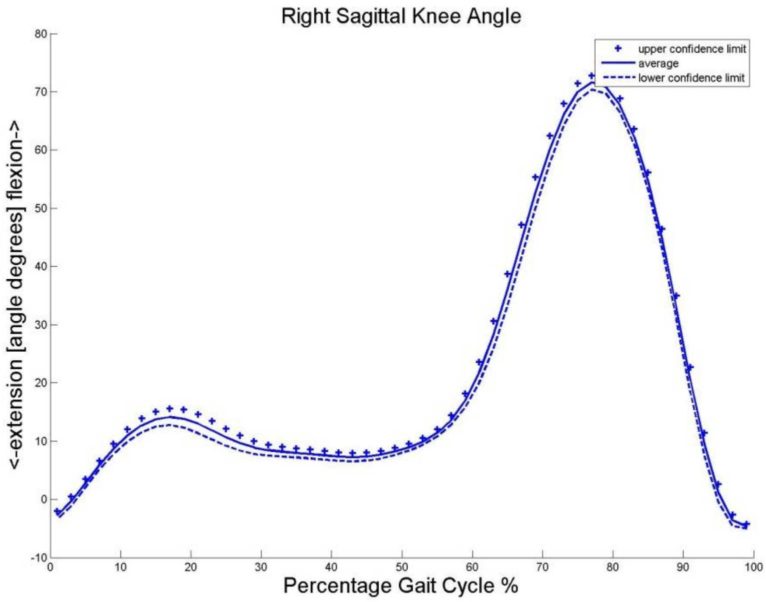
Figure 4-2 Intact limb (A) and prosthetic limb (B) hip angles during level walking on the RHEO knee

The motions of the hip joints of the intact and prosthetic limbs were similar, extension during the stance phase and flexion during the swing phase. Both joints were in extension throughout most of the stance phase with peak extension occurring at about 58% of the gait cycle for the intact limb and 52% for the prosthetic limb. At that instant the hip had extended 46° and 37° respectively. After push-off both joints flexed, with peak flexion occurring at terminal swing, with trajectory of 58° for the intact limb and 38° for the prosthetic limb.

4.1.2 Joint angles - Knee – 3R80 & RHEO

Figure 4-3, (A) and (B) represent the right and the left sagittal knee angle patterns for the 3R80.

(A)



(B)

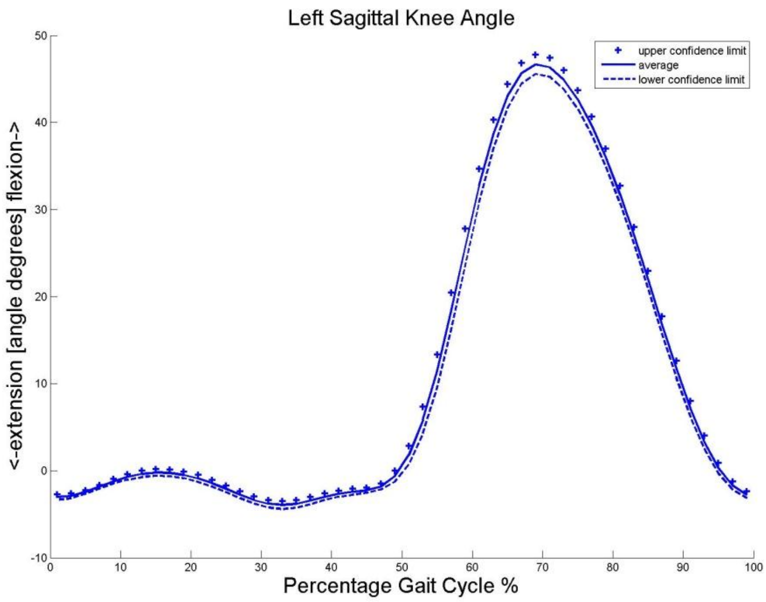
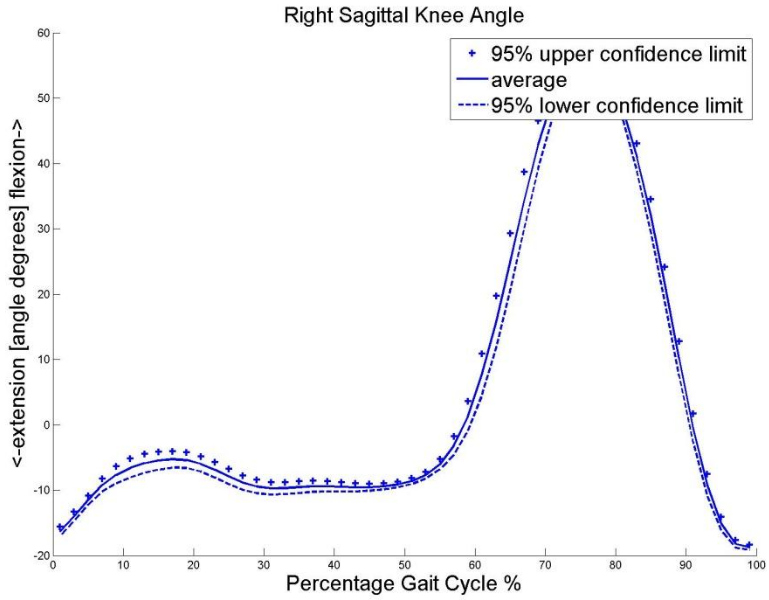


Figure 4-3 Intact limb (A) and prosthetic limb (B) knee angles during level walking on the 3R80 knee

The intact knee (A) showed close to normal stance phase knee flexion, with a trajectory of 15°, compared to normal value of 20° (figure 2-13). The motion during the stance phase was dissimilar for the prosthetic knee; there was a significantly decreased knee flexion during the stance phase, with only 4° flexion motion. For the intact limb there was a slight delayed peak knee flexion during the swing phase with flexion trajectory of 62° flexion from late stance. The prosthetic knee flexion from late stance to mid- swing was decreased when compared to normal motion and the intact limb, with flexion trajectory of 48° and normal value just over 60° (figure 2-13).

Figure 4-4 represents right (A) and left (B) sagittal knee angles pattern for the RHEO knee.

(A)



(B)

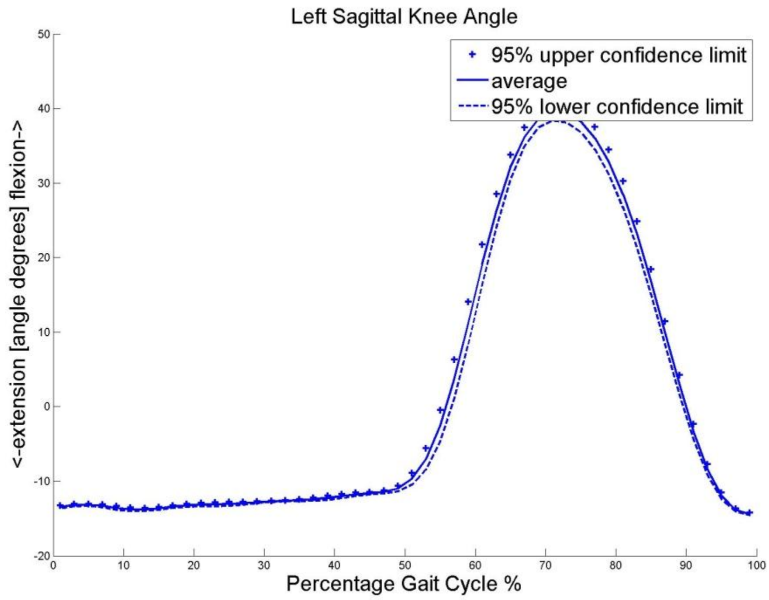


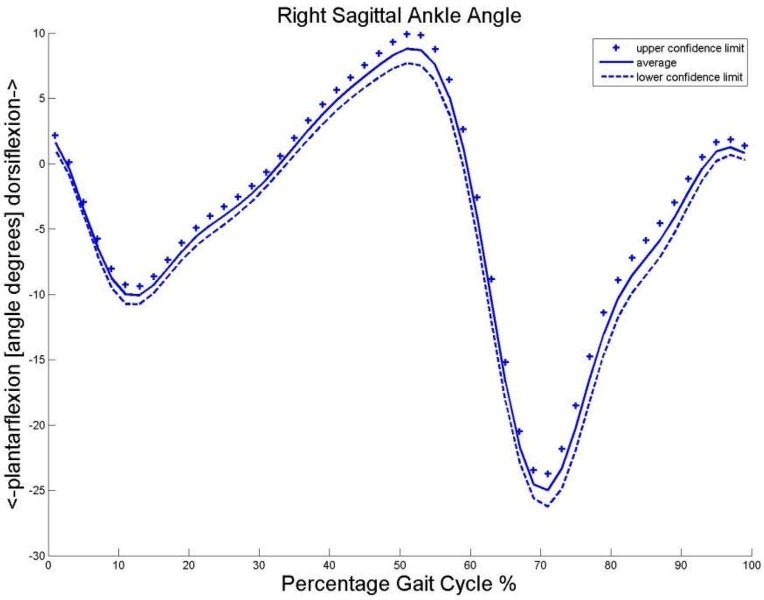
Figure 4-4 Intact limb (A) and prosthetic limb (B) knee angles during level walking on the RHEO knee

Results for the intact knee showed that during the stance phase the motion of the knee was 10° knee flexion but for the prosthetic knee there was an absence of knee stance flexion. From late stance to mid-swing the prosthetic knee had a flexion trajectory of 51° and the intact knee of 58°.

4.1.3 Joint angles - Ankle – 3R80 & RHEO

Figure 4-5 represents the intact limb (A) and the prosthetic limb (B) sagittal ankle angle patterns for the 3R80.

(A)



(B)

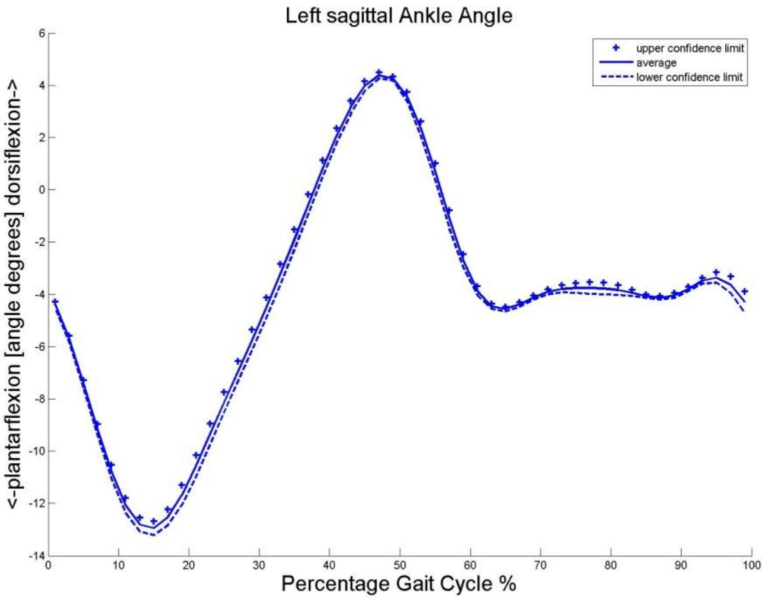
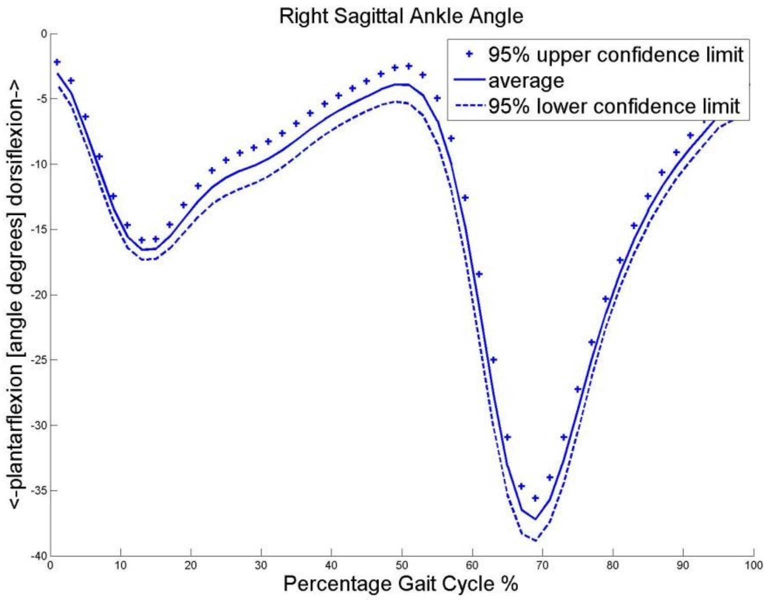


Figure 4-5 Intact limb (A) and prosthetic limb (B) ankle angles during level walking on the 3R80 knee

The angular motion of the anatomical ankle had close to normal pattern and the joint moved with normal range of motion, about 34°, which is comparable to anatomical ankle motion (figure 2-14). The range of motion of the prosthetic ankle was significantly decreased, with value of 18°. During early stance the anatomical and the prosthetic ankle had close to normal plantar flexion trajectory, 10° and 9°respectively. The main difference between the anatomical and the prosthetic ankle was seen on the trajectories during the push off phase, where normal plantar flexion trajectory is about 33°, with peak plantar flexion angle about 22° (figure2-14). The anatomical ankle had similar trajectory, 34°, with peak plantar flexion angle of 25°. For the prosthetic ankle it was different, the plantar flexion trajectory was only 8° and the peak plantar flexion angle 4°.

Figure 4-6 represents the intact limb (A) and the prosthetic limb (B) sagittal ankle angle patterns for the RHEO knee.

(A)



(B)

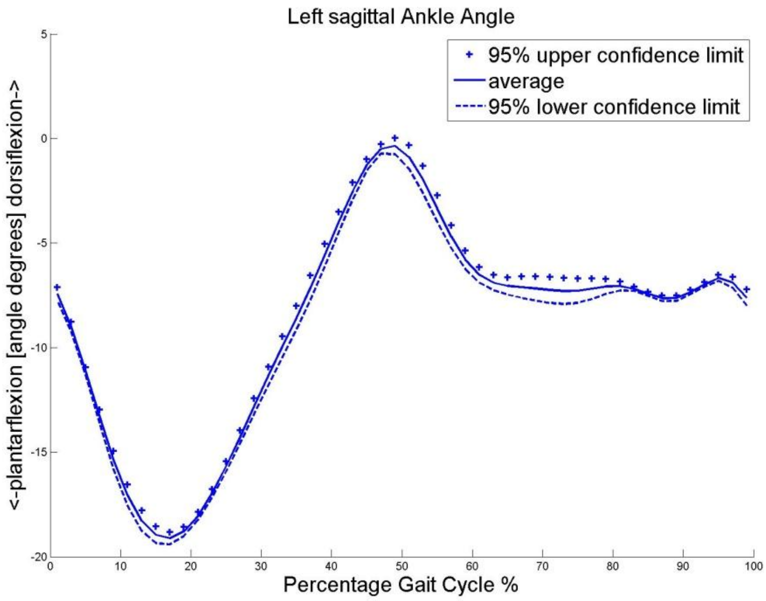


Figure 4-6 Intact limb (A) and prosthetic limb (B) ankle angles during level walking on the RHEO knee

The anatomical ankle was associated with normal angular motion, with range of motion about 31°. The range of motion of the prosthetic ankle was significantly decreased, with value of 19°. During early stance the anatomical and the prosthetic ankle had similar plantar flexion trajectories, 15° and 12° respectively. The main difference between the anatomical and the prosthetic ankles was seen during push off, where the plantar flexion trajectory for the anatomical ankle was 31° but only 7° for the prosthetic ankle.

4.1.4 Joint angles - Comparison – RHEO knee and 3R80

Figure 4-7 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip angle patterns for the RHEO knee and the 3R80 knee.

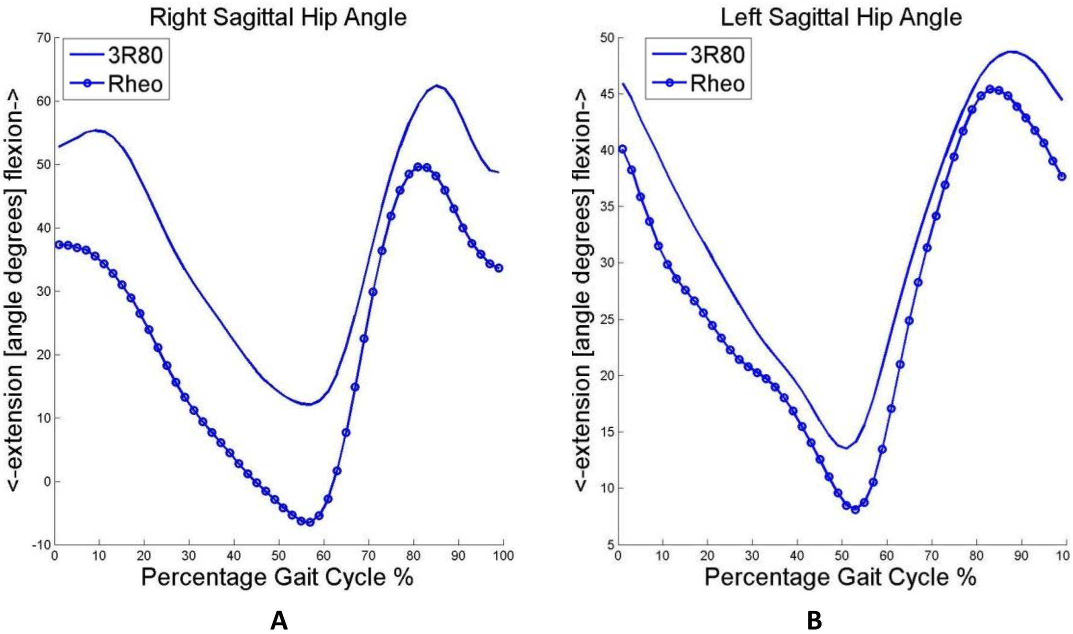


Figure 4-7 Intact limb (A) and prosthetic limb (B) hip angles during level walking on the RHEO knee and the 3R80 knee

Comparison of the results obtained for the 3R80 and the RHEO knee for the hip joint on the prosthetic side showed similar patterns of motions, however, with slightly different trajectories. During the stance phase the RHEO knee was associated with slightly greater angular motion, with extension trajectory of 37° but 35° when using 3R80 knee joint. Peak hip extension occurred at similar instant when using the knees, at push off. The flexion trajectories obtained for the knee joints were similar but the 3R80 was associated with slightly greater flexion motion, with trajectory of 40° compared with 38° for the RHEO knee.

For the hip joint on the intact side the trajectories of motion were to some extent different when using the different types of knees. The RHEO knee was associated with greater range of motion of the joint throughout the gait cycle. The trajectory for the extension motion during the stance phase was 40° for the 3R80 and 46° for the RHEO knee. The flexion trajectories during the swing phase showed more rapid hip flexion when using the RHEO knee and greater flexion motion when using the Rheo knee compared with when using the 3R80, 58° and 50° respectively.

Figure 4-8 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal knee angle patterns for the RHEO knee and the 3R80 knee.

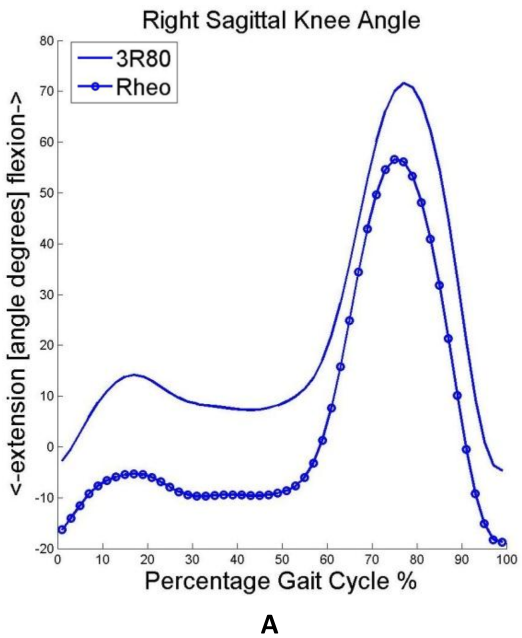


Figure 4-8 Intact limb (A) and prosthetic limb (B) knee angles during level walking on the RHEO knee and the 3R80 knee

The patterns of motions of the prosthetic knee joints were identical. During the stance phase both knees were associated with lack of knee flexion, the RHEO knee did not flex at all and the 3R80 flexed 4°. At late stance, about push off, both knees rapidly flexed, with flexion trajectories from push off to about mid-swing of 48° for the 3R80 and 51°for the RHEO knee. The knee extensions during the swing phase were similar for the joints. The results for the intact knee were comparable for the RHEO knee and the 3R80. The knee stance flexion was similar, 10° for the RHEO knee and 15° for the 3R80. The knee flexion and extension during the swing phase had slightly different flexion trajectories, 62° when using the 3R80 and 58° when using the RHEO.

Figure 4-9 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal ankle angle patterns for the RHEO knee and the 3R80 knee.

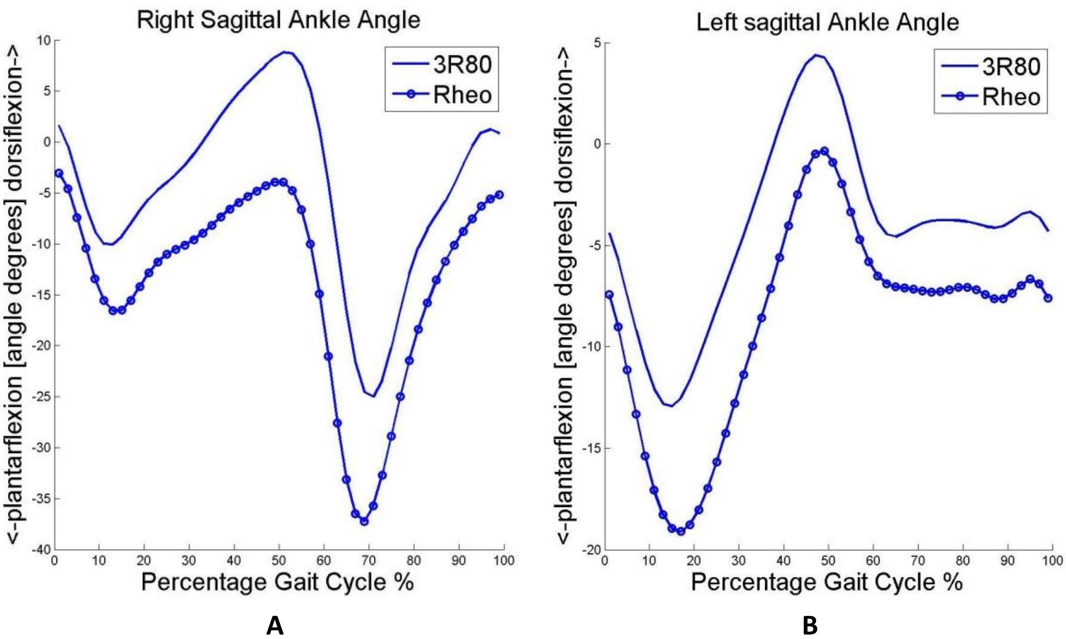


Figure 4-9 Intact limb (A) and prosthetic limb (B) ankle angles during level walking on the RHEO knee and the 3R80 knee

The range of motion of the prosthetic ankle was slightly different when using the 3R80 and the RHEO knee. During early stance the ankle was in plantar flexion with trajectory of 9° for the 3R80 and 12° for the RHEO knee. Following the peak plantar flexion the motion of the ankle switched to dorsiflexion with similar trajectories for both knees, 19° when using the RHEO knee and 18° with the 3R80. At push-off the trajectory of the plantar flexion was comparable, 8° when using the 3R80 and 7° when using the RHEO knee. The range of motion of the anatomical ankle joint was slightly different when using the different knees. During early stance the plantar flexion motion was identical. The dorsiflexion motion of the joint for the remaining of the stance phase was different for the knees. The trajectory associated with the 3R80 was 19° and the following plantar flexion trajectory was 25°. For the RHEO knee the dorsiflexion motion was decreased and the plantar flexion motion increased when compared with the 3R80, with 13° trajectory for the dorsiflexion and 31° trajectory for the plantar flexion.

Table 2 and 3 represent the hip, knee and ankle motion trajectories for the prosthetic (2) and the intact (3) sides that were considered relevant for the evaluation of the knees.

Table 2 Hip, knee and ankle extension and flexion trajectories for the prosthetic side during level walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**37**	**35**
Hip flexion trajectory in swing (degrees)	**38**	**40**
Knee flexion in stance (degrees)	**0**	**4**
Knee flexion trajectory in swing (degrees)	**51**	**48**
Ankle plantar flexion trajectory at push off (degrees)	**7**	**8**

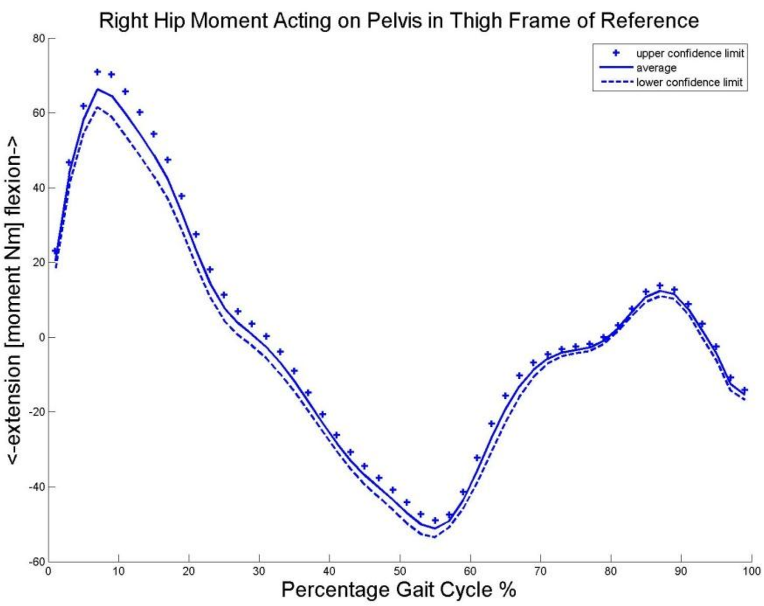
Table 3 Hip, knee and ankle extension and flexion trajectories for the intact side during level walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**46**	**40**
Hip flexion trajectory in swing (degrees)	**58**	**50**
Knee flexion in stance (degrees)	**10**	**15**
Knee flexion trajectory in swing (degrees)	**58**	**62**
Ankle plantar flexion trajectory at push off (degrees)	**31**	**25**
Ankle plantar flexion trajectory at push off (degrees)	**31**	**25**

4.1.5 Joint moments - Hip – 3R80 & RHEO

Figure 4-10 represents results for test performed on the 3R80 knee. The diagrams show results for the intact limb (A) and the prosthetic limb (B) hip moments acting on the pelvis.

(A)



(B)

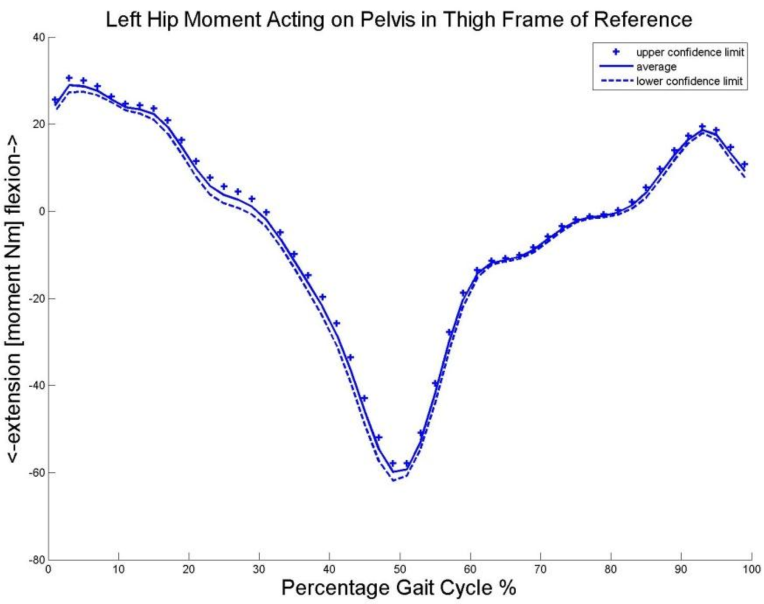
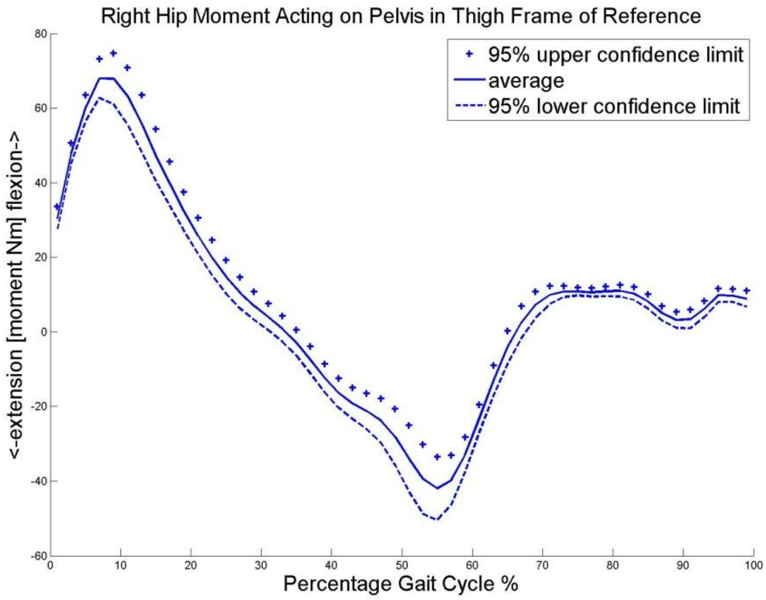


Figure 4-10 Intact limb (A) and prosthetic limb (B) hip moment during level walking on the 3R80 knee

Right after initial contact there was a flexion moment about the hip joints where the joint on the prosthetic side had a significantly lower moment value, 5 Nm compared with the intact limb moment of 45 Nm. In major part of the gait cycle, from early to terminal stance both the intact and the prosthetic limb experienced extension hip moment, however of different magnitude. About 50% of the gait cycle, around push off there was a peak extension moment about both hip joints with different trajectories, about 90 Nm about the joint on the prosthetic side but significantly larger moment about the intact side, 120 Nm, which for both sides is significantly decreased when compared to normal value of about 190 Nm (figure 2-12). The hip flexion from late stance to late swing had similar range, about 80 Nm on the prosthetic side and 73 Nm on the intact side.

Figure 4-11 represents results for test performed on the RHEO knee. The diagrams show results for the intact limb (A) and the prosthetic limb (B) hip moments acting on the pelvis.

(A)



(B)

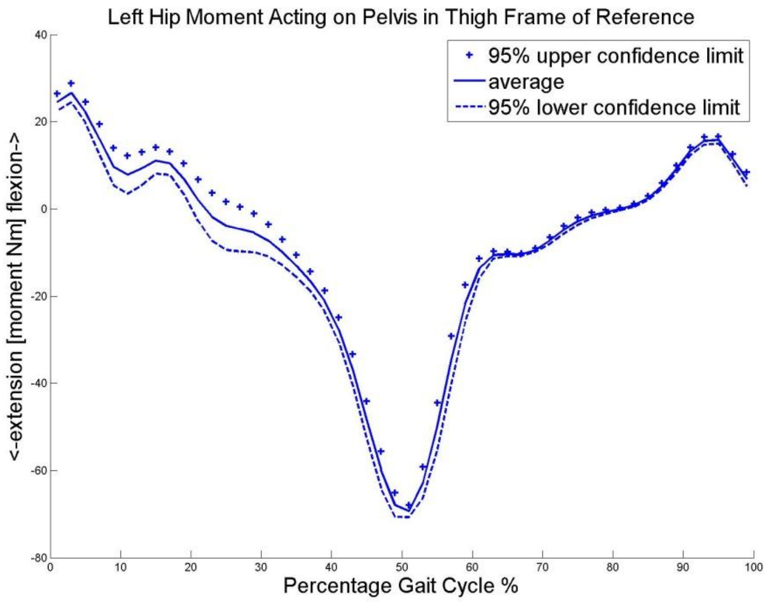


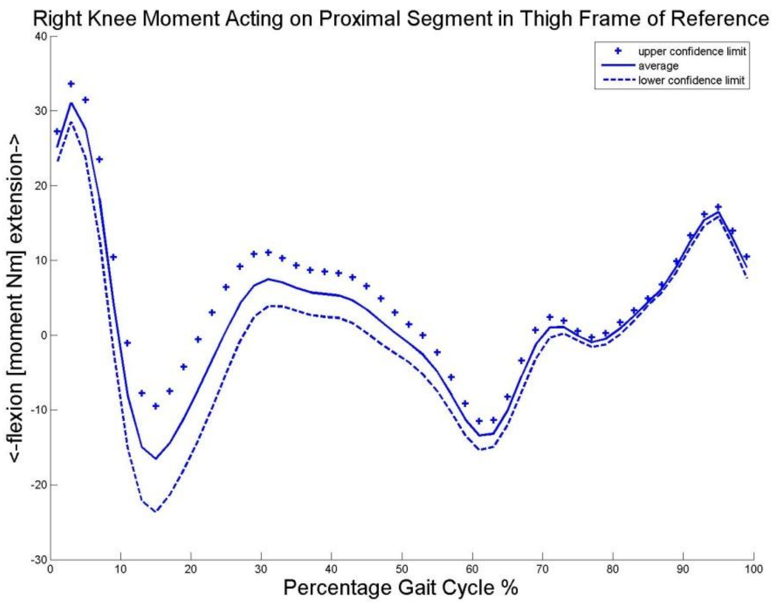
Figure 4-11 Intact limb (A) and prosthetic limb (B) hip moments during level walking on the RHEO knee

The prosthetic limb experienced decreased peak hip flexion moment right after initial contact. During early to late stance there was an extension moment about both hip joints with similar trajectories, about 98 Nm for the prosthetic limb and 110 Nm for the intact limb, which is significantly decreased when compared to normal value (figure 2-12). The hip flexion from late stance to late swing had similar range, about 80 Nm; however, the pattern between the limbs was different.

4.1.6 Joint moments - Knee – 3R80 & RHEO

Figure 4-12 represents the intact (A) and the prosthetic (B) knee joint moment curves for the test performed on the 3R80 knee.

(A)



(B)

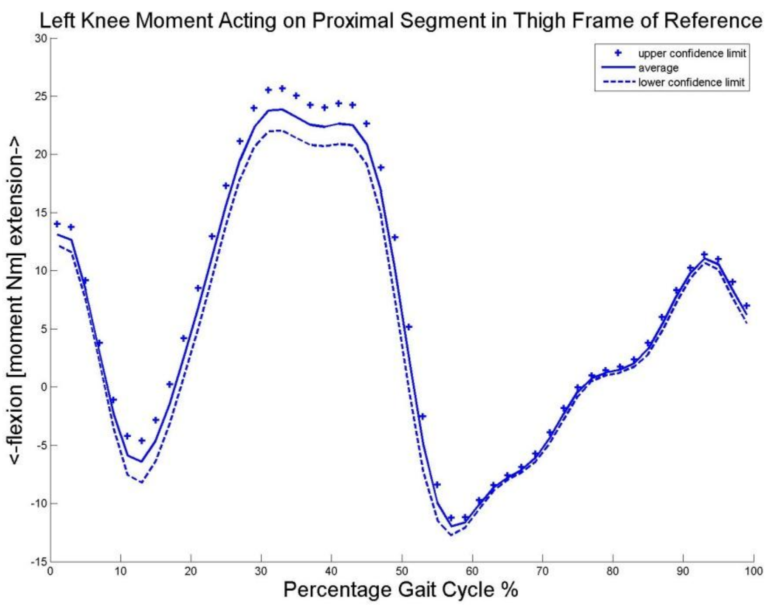
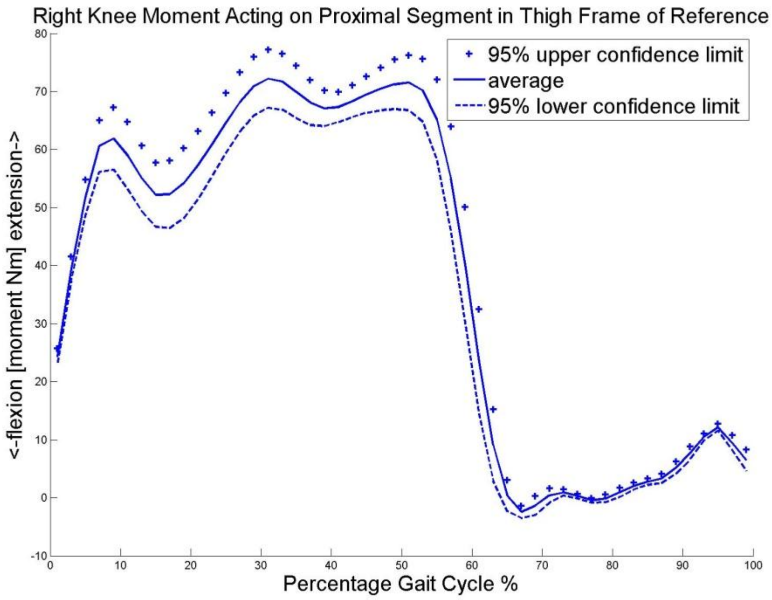


Figure 4-12 Intact limb (A) and prosthetic limb (B) knee moment during level walking on the 3R80 knee

At initial contact there was an extension moment about the intact knee, with trajectory of 5 Nm. During early stance the flexion moment about the intact knee was significantly greater than the flexion moment about the prosthetic knee, with 46 Nm and 20 Nm trajectories respectively. The prosthetic knee had greater peak knee extension moment during mid-stance, with trajectory of 30 Nm compared with the intact knee moment trajectory of 24 Nm. Late to terminal stance there was a flexion moment about both joints, with different trajectories, 36 Nm for the prosthetic knee and 20 for the intact knee, with peak occurring about 60% of the gait cycle. During the swing phase extension moment was about both joints, with slightly different extension trajectories, 24 Nm about the prosthetic knee and 20 Nm about the intact knee.

Figure 4-13 represents the intact (A) and the prosthetic (B) knee joint moment curves for the test performed on the RHEO knee.

(A)



(B)

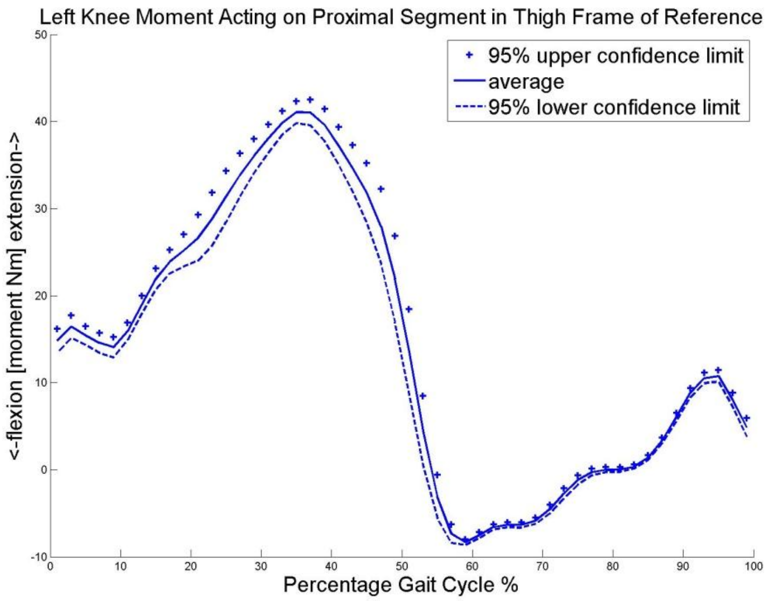


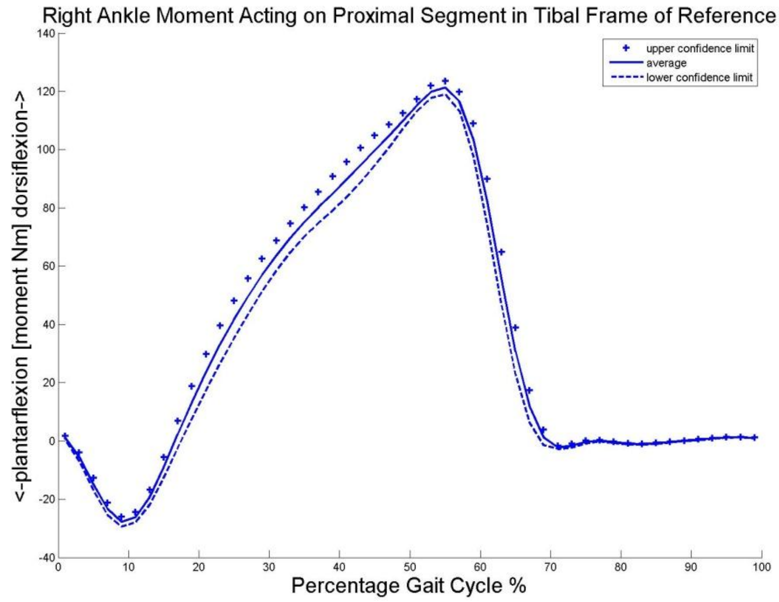
Figure 4-13 Intact limb (A) and prosthetic limb (B) knee moments during level walking on the RHEO knee

At initial contact the extension moment was considerably greater about the intact knee than the prosthetic knee, 35 Nm and 5 Nm respectively. During early stance the flexion moment about the intact knee was significantly greater than the flexion moment about the prosthetic knee joint, 10 Nm and 2 Nm and the following extension moment from early to mid-stance was very similar for both knees, with trajectory of about 26 Nm. Late to terminal stance there was a flexion moment about both joints, with 49 Nm trajectory for the prosthetic knee and 70 Nm for the intact knee. During the swing phase there was an extension moment about both joints, with very similar trajectories, about 24 Nm.

4.1.7 Joint moments - Ankle – 3R80 & RHEO

Figure 4-14 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the 3R80 knee.

(A)



(B)

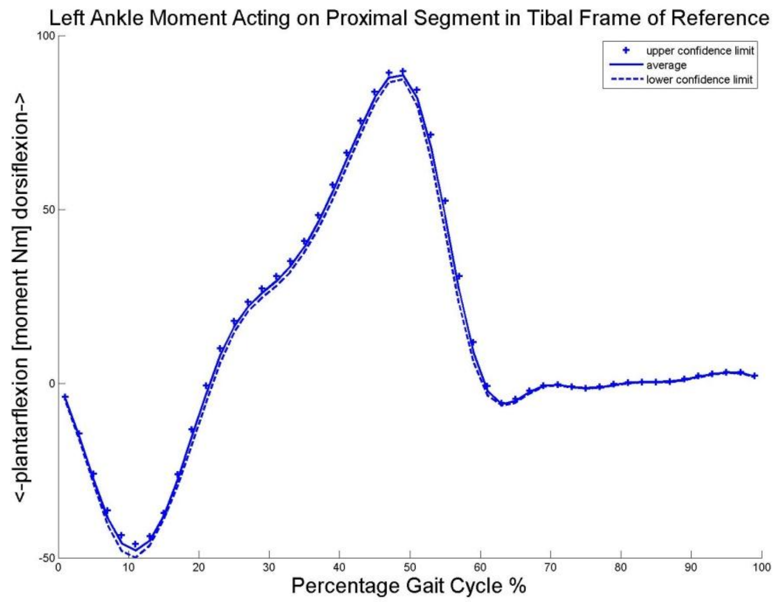
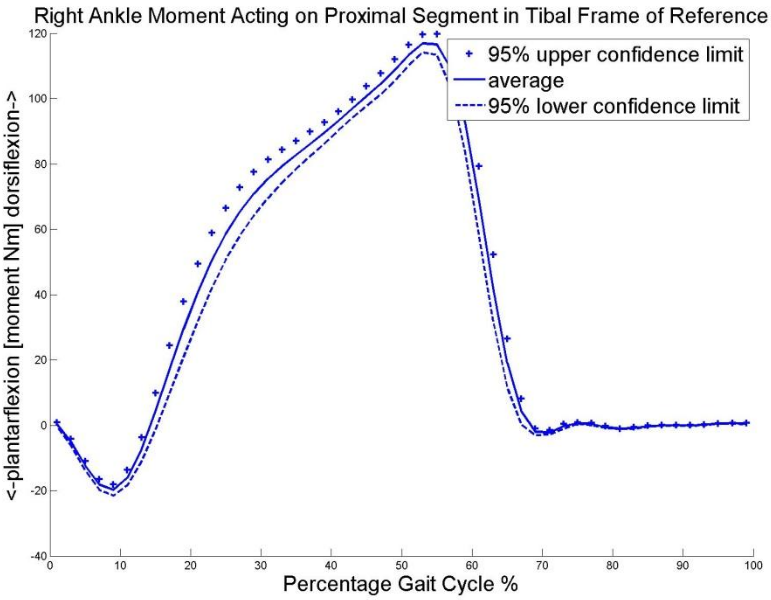


Figure 4-14 Intact limb (A) and prosthetic limb (B) ankle moments during level walking on the 3R80 knee

Both the anatomical and the prosthetic ankles demonstrated normal patterns of ankle moments. Normally peak plantar flexion moment of 15 Nm occurs at 10% of the gait cycle and the peak dorsiflexion moment, 95 Nm occurs at 50% of the gait cycle (figure 2-14). For both the anatomical and the prosthetic joints, peak plantar flexion moment about the joints was at 10% of the gait cycle, however, with different values when compared with normal, 30 Nm and 50 Nm respectively. The peak dorsiflexion moment about the anatomical ankle occurred about 50% of the gait cycle with trajectory of 150 Nm and the trajectory for the prosthetic ankle was 140 Nm, normal trajectory is about 95 Nm.

Figure 4-15 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the RHEO knee.

(A)



(B)

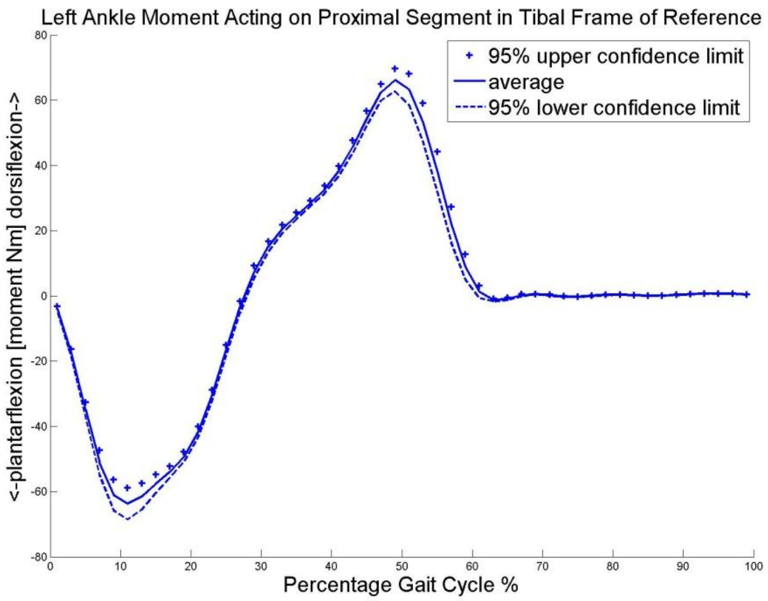


Figure 4-15 Intact limb (A) and prosthetic limb (B) ankle moments during level walking on the RHEO knee

Both the anatomical and the prosthetic ankles demonstrated normal patterns of ankle moments. For both the anatomical and the prosthetic joints peak plantar flexion moment about the joints was at 10% of the gait cycle, however, with different values when compared with normal, 20 Nm for and 60 Nm respectively. The peak dorsiflexion moment about the anatomical ankle occurred about 50% of the gait cycle with trajectory of 140 Nm and the trajectory for the prosthetic ankle was 120 Nm.

4.1.8 Joint moments - Comparison – RHEO knee & 3R80

Figure 4-16 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip moments for the RHEO knee and the 3R80 knee.

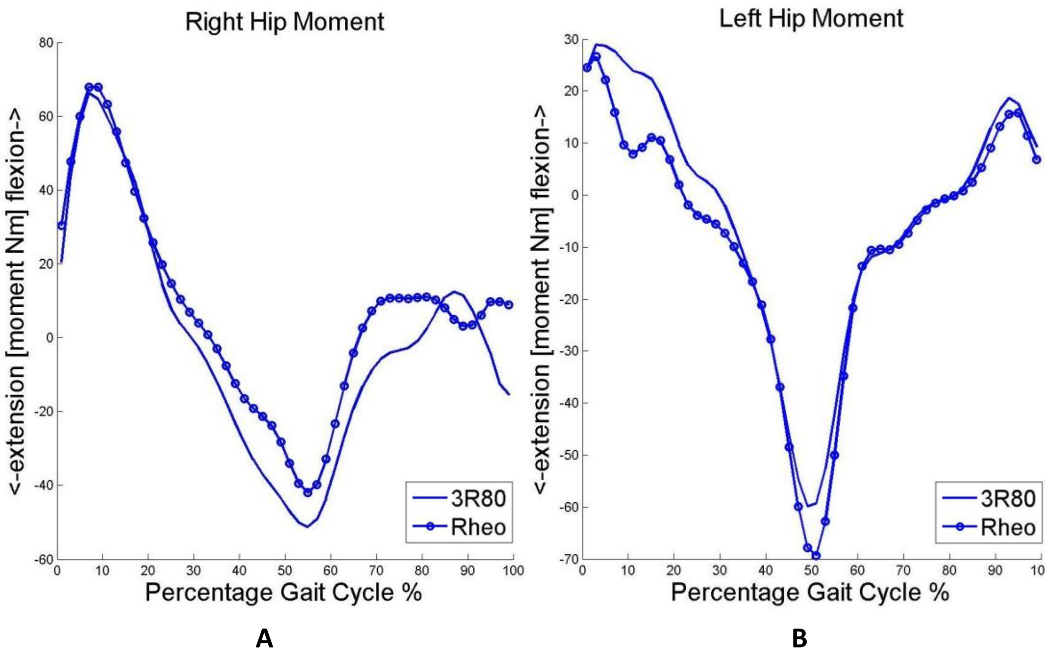


Figure 4-16 Intact limb (A) and prosthetic limb (B) hip moments during level walking on the RHEO knee and the 3R80 knee

As seen on figure 4-16 the hip moment curves obtained for the prosthetic side are very similar although using different type of prostheses. When using the RHEO knee the magnitude of the extension moment from early to late stance was slightly greater than when using the 3R80, 98 Nm and 90 Nm respectively. During the swing phase the extension and flexion moments were comparable. Similarly for the intact limb the hip moments were very similar throughout the gait cycle except at terminal stance where the 3R80 was associated with slightly greater peak extension moment.

Figure 4-17 represents the comparison on the anatomical (A) and the prosthetic (B) knee moments for the tests made on the 3R80 and the RHEO knee.

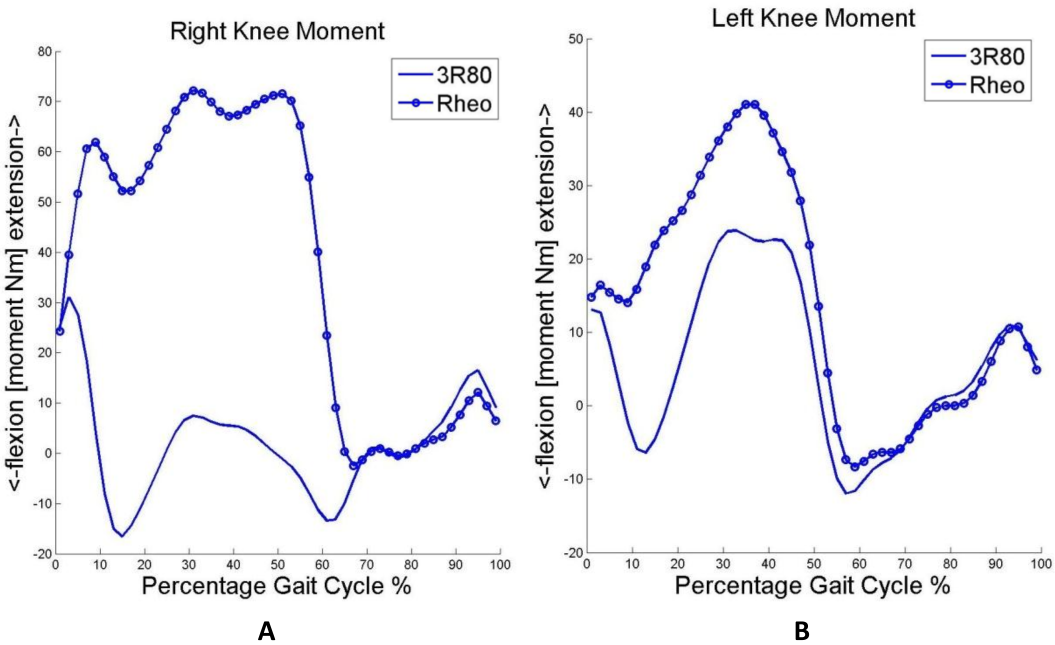


Figure 4-17 Intact limb (A) and prosthetic limb (B) knee moments during level walking on the RHEO knee and the 3R80 knee

The curves for the intact knee moments were very different when using the RHEO knee and the 3R80. Right after initial contact there was an extension moment about the knee in both cases but of different magnitude. When using the 3R80 there was a small peak moment of 5 Nm. The extension moment about the knee when using the RHEO knee was of greater magnitude, 35 Nm. The following flexion moment trajectory was considerably greater when using the 3R80 compared with the RHEO knee, 46 Nm and 10 Nm respectively. From late to terminal stance there was a flexion moment about the knees, the trajectories were very different, 71 Nm for the RHEO and 19 Nm for the 3R80 knee joint. The moments during the swing phase were identical. The curves obtained for the prosthetic knee moments were significantly different across the knee joints. During early stance there was a flexion moment about both joints, of considerably greater magnitude about the 3R80 knee, 20 Nm compared with 2 Nm about the RHEO. From early stance to toe-off there were extension moments about the joints with similar trajectories, except slightly different peaks where the peak extension moment for the 3R80 lasted longer. The flexion moment about the joints in late stance had different trajectories, 49 Nm for the RHEO knee and 36 Nm for the 3R80. During the swing phase the moment curves were identical.

Figure 4-18 represents the comparison on the anatomical (A) and the prosthetic (B) ankle moments for the tests made on the 3R80 and the RHEO knee.

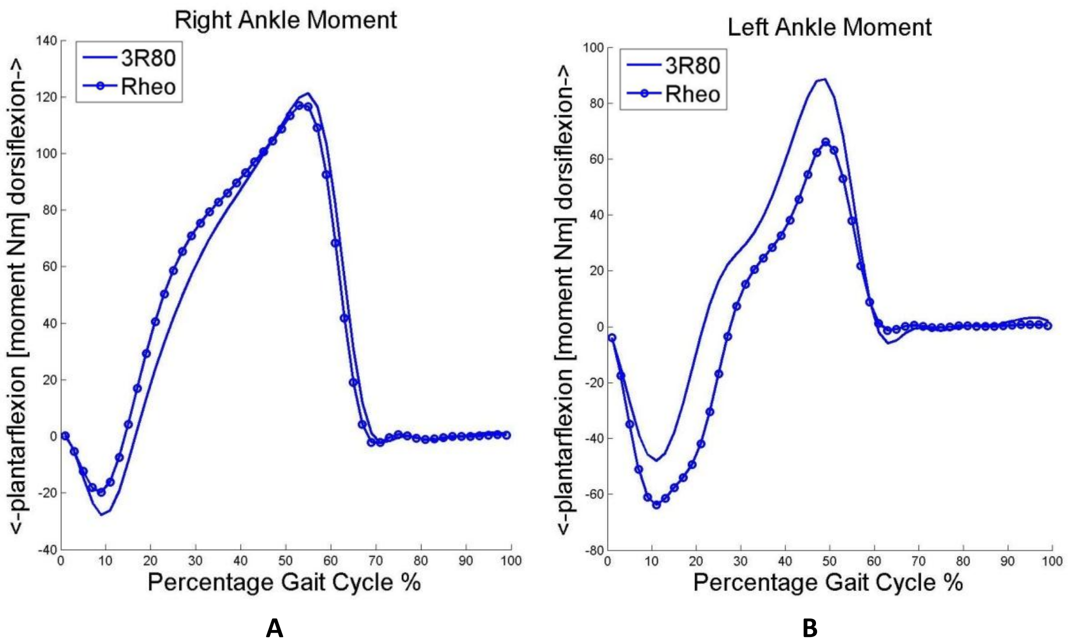


Figure 4-18 Intact limb (A) and prosthetic limb (B) ankle moments during level walking on the RHEO knee and the 3R80 knee

For the anatomical ankle the moments obtained for the different knee types were identical, with same pattern and equal peak moments. The moments about the prosthetic ankle were very similar when using the two different knees. During early stance the peak plantar flexion moment about the joint was slightly greater when using the RHEO knee and the peak dorsiflexion moment during late stance was slightly greater when using the 3R80.

Table 4 and 5 represent the hip and knee moment trajectories for the prosthetic (4) and the intact (5) sides that were considered relevant for the evaluation of the knees.

Table 4 Hip and knee extension and flexion moments of the prosthetic limb during level walking

Parameter	RHEO Knee	3R80 Knee
Hip extension moment trajectory at push off (Nm)	**98**	**90**
Knee flexion moment trajectory in early stance (Nm)	**2**	**20**
Knee extension moment trajectory in early to mid-stance (Nm)	**25**	**29**

Table 5 Hip and knee extension and flexion moments of the intact limb during level walking

Parameter	RHEO Knee	3R80 Knee
Hip extension moment trajectory at push off (Nm)	**110**	**120**
Knee extension moment trajectory at initial contact (Nm)	**35**	**5**
Knee flexion moment trajectory in early to mid-stance (Nm)	**10**	**46**

4.2 Discussion for level walking

4.2.1 RHEO knee and 3R80

The results for the angular motion of the hip joints when using the 3R80 and the RHEO knee showed that the hip joints had close to normal motion pattern when compared with intact hip motion in gait (figure 2-12). However, peak values for flexion and extension are different and peaks occurred at slightly different instances during the gait cycle.

The difference in angular motion of the hip joints on the prosthetic and the intact sides consisted of smaller range of motion for the joint on the prosthetic side, both when using the 3R80 and the RHEO knee. That is contradictory to reports by Jaegers et al (1995), which reported wider range of motion for the hip of the prosthetic limb compared with intact individuals and the intact limb (chapter 2.6.2.1). The decreased range of motion obtained for the prosthetic limb indicates uneven step length with the prosthetic and the intact limbs. The subject was likely taking smaller steps with the prosthetic limb, that indicates that he was not fully relying on either of the prostheses and taking full normal steps with the prosthetic limb. When using the 3R80 knee, the peak hip extension was reached slightly later on the prosthetic side, which could indicate that the subject was compensating for the decreased step length of the prosthetic limb by increasing the step length of the intact limb. Seroussi et al (1996) discussed that due to the absence of prosthetic knee flexion during stance a lack of hip flexion in early stance will occur (chapter 2.6.2.1).

Comparison on the curve obtained for the intact and the prosthetic limbs and to intact individuals (figure 2-12) shows abrupt transition from hip extension to hip flexion around push off both for the 3R80 and the RHEO knee. Similar results were reported by Jeagers et al (1995) (chapter 2.6.2.1). The sharp transition from hip extension to flexion of the prosthetic limb indicates that when using both types of knees the subject used the hip to initiate knee extension prior to the swing phase. However, the general pattern of hip flexion and extension was similar and relatively close to normal pattern. That is agreeing with results reported by Jeagers et al (1995) (chapter 2.6.2.1).

Results for the intact knee showed a delayed peak knee flexion during the swing phase when using the 3R80, similar results have been reported by Seroussi et al (1996) (chapter 2.6.2.1). and Farahmand et al (2006). However, in general when using the 3R80 and the RHEO knee, the intact knee motion patterns were close to normal. That is agreeing with results reported by Jaegers et al (1995) and Seroussi et al (1996) (chapter 2.6.2.1).

The prosthetic knee joint started to flex at about 50% of the stride, which is very close to what is seen from intact individuals (figure 2-13). This result is contradictory to reports by Jaegers et al (1995), that reported a delayed prosthetic knee flexion for the amputee subjects or just before toe off (chapter 2.6.2.1). During the stance phase there was an absence of knee flexion for both prosthetic knees, as the knees were locked in extension. Inadequate prosthetic knee flexion during early stance is well reported in the literature (figure 2-16, chapter 2.6.2.1).) (Farahmand, 2006; Blumentritt, 1998; Seroussi, 1996; Jaegers, 1995).

The absence of the prosthetic knee flexion during the stance phase indicates that the subject was not fully loading the prostheses. It is likely that this is a habit that the subject has established, his way to stabilise the prosthetic knee joint could be by using hip extensors to lock the knee in extended position during stance phase and that way prevent it from buckling under his weight. The 3R80 knee joint utilises a compressible elastic bumper and the primary stance phase knee flexion of the knee joint is approximately four degrees. That indicates that voluntarily knee flexion was absent as for the RHEO knee and these four degrees of knee flexion most likely due to the knee mechanism. It is unlikely that adjustments to the knee flexion resistance of the knee joint would impact the stance knee flexion, as the 3R80 knee joint is the prosthesis that the subject uses on daily basis, therefore he is used to it and adjustments optimised for him.

The peak flexion angle during the swing phase associated with the RHEO knee was close to normal, 51°. The flexion angle associated with the 3R80 was less than normal with peak value of 48° compared with normal value of 55° (figure 2-13), these results are agreeing with reports in the literature (figure 2-16, chapter 2.6.2.1) (Murray, 1980; Blumentritt, 1998; Jaegers, 1995)).

Results for the prosthetic ankle joint showed a significantly smaller than normal peak plantar flexion angle during push off for both knees and has also been reported in the literature (chapter 2.6.2.1). Seroussi et al (1996) concluded that the decreased plantar flexion of the ankle joint results in decreased energy generation by the ankle, thus likely to cause other compensatory movements. Considering the amount of work the anatomical ankle does it is expected that other joints will compensate for the significantly decreased energy generation by the prosthetic ankle joint. The plantar and dorsiflexion movements of the prosthetic ankle during the swing phase are most likely due to measurement errors as movements of the prosthetic ankle during the swing phase are not expected (Farahmand, 2006).

The carbon fibre Echelon foot used for the present study utilises hydraulics that control plantar and dorsiflexion movement. The Echelon foot is considered as one of the best prosthetic foot for active amputees. The alignment of the foot and resistance was optimised for the subject; therefore problems such as misalignment and wrong level of resistance should not be causing the subject problems to move the prosthetic ankle. The decreased movement of the prosthetic ankle is reasonable considering the absence of anatomical ankle, muscle actions and the loss of proprioception.

Generally, for both knees; the pattern and range of motion of the intact limb was close to joint motions exhibited by intact individuals. However the joint motions of the prosthetic limb was different from the intact limb and intact individuals, where the limb exhibited decreased angular motion. The decreased motion of the prosthetic limb is understandable, due to lack of muscular function and proprioception and the absence of anatomical joints. Apart from anatomical causes, the confidence and skills of the subject has a great impact. As seen on the results for the present study the subject is a very able walker and does not exhibit great gait deviations considering his level of amputation.

The hip moment pattern throughout the gait cycle is close to normal for both limbs when using both knees, with however different values for peak flexion and extension moments. For both the RHEO and the 3R80 the extension and flexion peak moments about the hip joints of both limbs are decreased when compared with peak moment about a hip joint of an intact individual (figure 2-12). Both knees were associated with lower hip extension moment about the hip joint on the prosthetic side, which indicates that when using both knee types the subject was relying more heavily on the intact limb and compensating for the prosthetic limb. These results correspond to results reported by Seroussi et al (Seroussi, 1996) that reported greater hip extension moment on the intact side than the prosthetic side (chapter 2.6.2.1). Agreeing with Seroussi et al (Seroussi, 1996) the results obtained for both knees showed a greater hip flexion moment about the hip on the prosthetic side in late stance.

For the knee joint moments of both the prosthetic and the intact knee, the moment curves had a shifting pattern between knee extension and flexion moments. The RHEO knee was associated with considerably different moment pattern about the prosthetic joint than the intact knee. About the RHEO knee there was an extension moment through majority of the stance phase, as the knee was locked in extension different from the intact knee. For the 3R80, at initial contact and early stance both the prosthetic and the intact knee experienced extension moment, where the 3R80 knee had significantly lower peak value than the intact knee. The prosthetic knee extension moment obtained is not consistent with the results reported by Seroussi et al (1996) (chapter 2.6.2.1), which reported permanent flexion moment early to mid-stance but agreeing with results reported by Fahramand et al (2006).

During early stance the intact and the prosthetic knee moments shifted to flexion moment when using the 3R80, where both limbs experienced decreased peak knee flexion moment compared to normal, especially the prosthetic knee joint. The flexion moment of the intact knee obtained contradicts results reported by previous references Farahmand et al (2006) and Seroussi et al (1996) where both groups reported significantly higher peak flexion moment value for the intact limb during early stance (chapter 2.6.2.1). In late stance and in swing phase the 3R80, the RHEO knee and the intact knee were associated with extension moment, with lower peak values than normal.

The ankle joint moment patterns obtained are close to normal when using both knees, however, with plantar and dorsiflexion peaks occurring at slightly different instances during the gait cycle and with different peak values. For the prosthetic ankle the plantar flexion peak moment during early stance was increased compared to normal peak value and the intact ankle when using both knees. That is contradictory to equivalent data in the literature, that have reported significantly decreased plantar flexion moment for the prosthetic ankle compared with normal and intact limbs Seroussi et al (1996) (Yang, 1991) (chapter 2.6.2.1). The intact limb had also greater plantar flexion moment peak value than normal (figure 2-14) when using both knees and that corresponds to results reported by Seroussi et al (1996) (chapter 2.6.2.1). but contradicts reports by Farahmand et al (2006).

4.2.2 Discussion about the comparison of both knees for level walking

The results obtained for the hip joint on the intact side did not show great difference in hip motion when using the RHEO knee or the 3R80. The difference in range of motion was 6° (46°-40°) in hip extension and 8° (99°-90) in hip flexion where the RHEO knee was associated with greater flexion and extension trajectories than the 3R80.

The greater flexion and extension motion indicate that when using the RHEO knee the step length with the intact limb was greater and as a result there is a greater range of motion of the hip joint. Increasing the step length of the intact limb could be a way for the subject to compensate for the prosthetic limb´s step length and compensate for the lack of energy generation at toe-off by the prosthetic ankle.

The RHEO knee was associated with slightly more rapid transition from extension to flexion in terminal stance, which could indicate that the subject was relying more on the intact limb when using the RHEO knee and as a result he drove the limb forward with rapid flexion movement to start next gait cycle. The results for the hip joint on the intact side indicate that when using the RHEO knee more compensatory movements were needed, which could partly be explained by the limited time the subject was given to get used to the knee joint.

The results for the angular motion of the hip joint on the prosthetic side did not reveal any major differences when using the two types of knees, where the hip flexion and extension motions were very similar for both knees. Similarly as for the joint on the intact side the RHEO knee was associated with slightly more rapid hip flexion in terminal stance and beginning of swing. The results for the motion of the hip joint on the prosthetic side indicate that the subject demonstrated similar hip flexion to initiate the prosthetic knee flexion in late stance phase for both knee types.

The results for the angular motion of the prosthetic knee joints were surprising, as it was expected that the RHEO knee would be associated with increased stance phase knee flexion angle when compared with the 3R80. Results revealed that was not the case, as no voluntary knee flexion was seen when using either the 3R80 or the RHEO knee throughout the stance phase. That is consistent with results reported by Segal et al (2006) (chapter 2.10.2). The small four degree flexion motion seen on the 3R80 curve is due to the mechanism of the knee joint. The results indicate that the subject was not able to demonstrate more natural knee motion when using the RHEO knee, however, the limited adjustment time give for the RHEO knee likely influenced the performance of the subject. Additionally the subject is a long-term user of the 3R80 knee and might have found it difficult to break the habit of keeping the prosthetic knee extended throughout the stance phase.

The transition from extension into flexion about toe off was very similar for both knees, although the resistance within the knees is controlled with different techniques the subject demonstrated similar pattern when using both joints. The RHEO knee demonstrated greater peak knee flexion angle during the swing phase. That is contradictory to reports by Johansson et al (2005) and Segal et al (2006), which reported greater knee extension angle for a mechanical knee joint (chapter 2.10.1). The reason for the greater peak flexion angle associated with the RHEO knee could be related to different walking speed when using the RHEO knee and the 3R80. Bellman et al (2010) reported increased peak knee flexion angle with increased walking speed, for both microprocessor controlled knees and mechanical knees (chapter 2.10.3).

The results indicate that when using the RHEO knee the subject was able to more naturally flex the knee joint during the swing phase and therefore the gait during the swing phase was more symmetric during the swing phase for the intact and the prosthetic knee. The reason could have been the different level of resistance. Perhaps the technology utilised by the microprocessor control of the RHEO knee has the advantage of being able to switch more rapidly from high level of resistance during the stance phase to lower resistance during the swing phase. The resistance provided by the RHEO knee enabled the subject to more naturally flex the joint and possible the resistance provided by the 3R80 was slightly higher. Despite that the peak flexion angle of the RHEO knee was greater it was considerably smaller than the target flexion angle setting of the knee, which was 65°. No general difference was seen in flexion and extension speeds between the prosthetic knees, results indicate that the knee joints were equally rapid to move into flexion and extension although utilising different technology.

The plantar and dorsiflexion motions of the prosthetic ankle were similar when using both knees. The RHEO knee was associated with similar peak plantar flexion angle compared with the 3R80, that is consistent with reports by Johansson et al (2005) (2.10.1). Similarly for both knees there was a lack of prosthetic ankle plantar flexion in terminal stance and as a result a great difference must have been in power generation in push off by the prosthetic ankle and the anatomical ankle, which exhibited normal plantar flexion motion at push off when using both knees. The results indicate that the different types of prosthetic knees do not impact the motion of either the prosthetic or the anatomical ankle to any extent.

It should be noted that the clinical specialist prosthetist from Össur UK that was present during the preparation of the study, claimed that the Echelon foot was not the recommended prosthetic foot to use in combination with the RHEO knee (Hirons, 2012). The reason is that the Echelon foot utilises hydraulic resistance which is not ideal in combination with the RHEO knee. The recommended foot to use with the RHEO knee is the Flex-Foot and is manufactured by Össur. The manufacturer claims that the combination of the RHEO knee and the Flex-Foot will provide the best function of the RHEO knee (Hirons, 2012). For this study it was decided to use the same prosthetic foot throughout the study, which was done to be able to evaluate the impact on the gait that was due to different knee technologies. However, possibly the subject would have performed better using the RHEO knee in combination with the Flex-Foot prosthetic foot.

The moments about both hip joints were very similar when using the 3R80 and the RHEO knee. For the joint on the prosthetic side the RHEO knee was associated with slightly greater magnitude of peak hip moment in late stance, these results are contradictory to reports by Johansson et al (2005) (chapter 2.10.1). The greater hip extension moment associated with the RHEO knee could indicate that when walking on the RHEO knee greater force was needed to extend the hip and the knee and to drive the head, arms and trunk segments forward. However, the magnitudes of the hip moments, intact and prosthetic sides, were more similar when using the RHEO knee, which could result in more normal gait. The moments about the hip joint on the intact side were similar when using the knees, apart from greater peak extension moment associated with the 3R80 knee.

The moments about the prosthetic knee joints were considerably different from initial contact to late stance, that is inconsistent with reports by Segal et al (2006) that studied and compared a microprocessor controlled knee and a mechanical knee and reported no significant difference between the knee moments (chapter 2.10.2). The reason for the greater flexion moment about the 3R80 knee in early stance is due to the mechanical properties of the knee, as the knee flexes. The small flexion moment about the RHEO knee during stance was expected as the knee did not flex during the stance phase. The following extension moment about both joints was similar; however, the RHEO knee was associated with sharper transition from extension to flexion moment in mid-stance. The reason could be greater angular velocity of the RHEO knee; Johansson et al (2005) reported significantly greater angular velocity of the RHEO knee when compared with a mechanical knee joint. No major differences were observed between the knee moments in terminal stance and in the swing phase. The peak extension moments at toe-off were similar for both joints, apart from the peak moment lasting longer about the 3R80. That is inconsistent to Johansson et al (2005), which reported significantly greater peak extension moment about a mechanical knee joint when compared with the RHEO knee (2.10.2). During the swing phase the moments about the RHEO knee and the 3R80 were identical, that is contradictory to reports by Johansson et al (Johansson, 2005), which reported significantly lower peak knee flexion moment in terminal swing for the RHEO knee when compared with hydraulic mechanical knee(2.10.2).

The intact knee moments were very different when using the RHEO knee and the 3R80. During the stance phase there was a considerably greater extension moment about the intact knee when using the RHEO knee and greater flexion moment when using the 3R80. The greater flexion moment about the intact knee when using the 3R80 is due to the increased knee flexion when compared with the intact limb when using the RHEO knee. The greater extension moment about the intact limb at initial contact that is associated with the RHEO knee could indicate greater impact at heel trike. That would indicate uneven weight bearing of the limbs, where the subject loads the intact limb more heavily. As the results show, the moments about the prosthetic knees are more similar than about the intact knee joint. The reason is likely due to the fact that the anatomical joint is influenced by active muscles and the subject compensates in different ways for the different prosthetic knee joints. The movements and moments about the prosthetic knees are more predictable. The results for the knee moments indicate that the net intact limb knee moment did not decrease when using the RHEO knee as was expected.

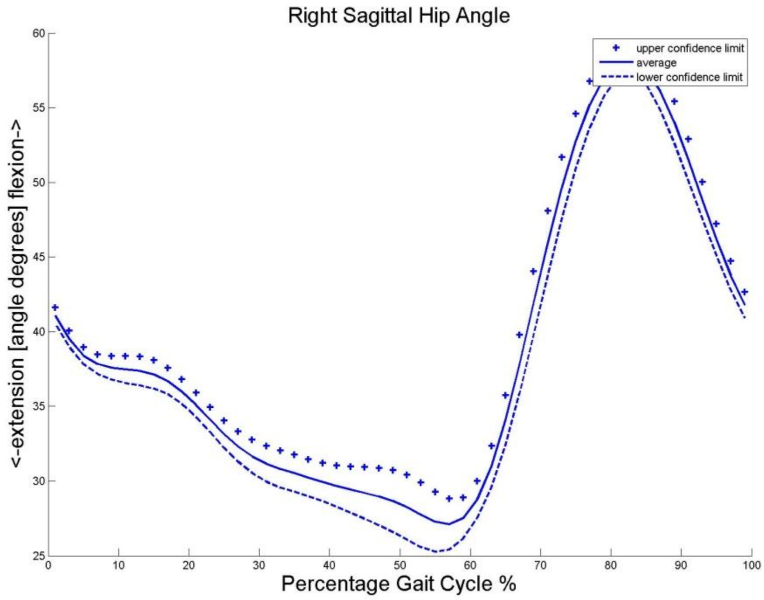
The results for the intact ankle showed that the moment about the joint was identical when using the 3R80 and the RHEO knee. Similarly for the prosthetic ankle the different knee types did not influence the joint to any extent.

4.3 Downhill walking

4.3.1 Joint angles – Hip – 3R80 & RHEO

Figure 4-19 represents the anatomical (A) and the prosthetic (B) hip angle curves for the tests made on the 3R80 knee.

(A)



(B)

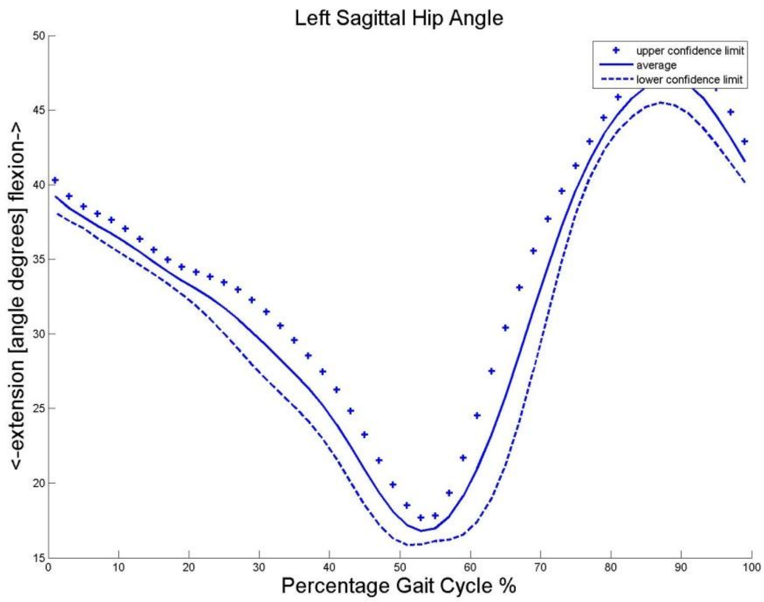
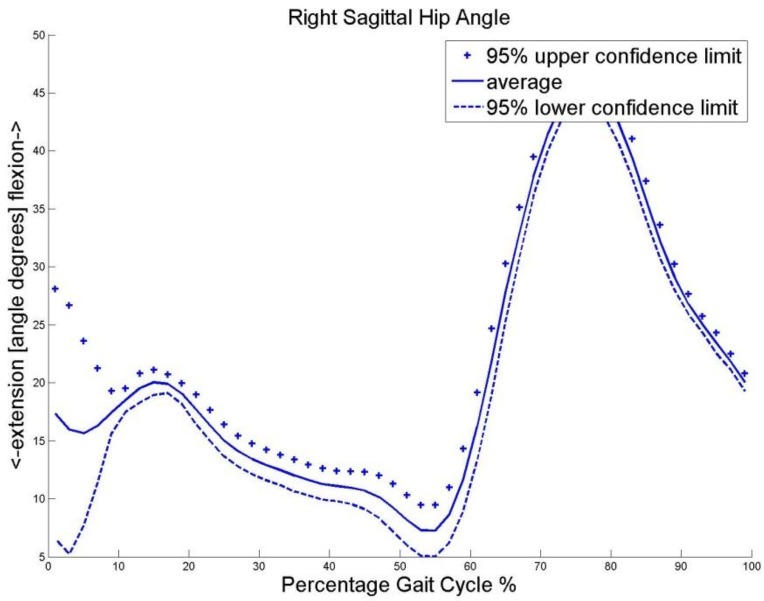


Figure 4-19 Intact limb (A) and prosthetic limb (B) hip angles during downhill walking on the 3R80 knee

During downhill walking the motion patterns of the hip joints were similar with comparable trajectories of hip flexion and extension. At initial contact and throughout the stance phase the motion of both joints was extension with peak extension occurring at late stance, the extension trajectories for the joints were 23° for the joint on the prosthetic side and 20° for the intact side. At terminal stance the hip movement changed to hip flexion with flexion trajectory of 30° for the intact limb and 44° for the prosthetic limb.

Figure 4-20 represents the anatomical (A) and the prosthetic (B) hip angle curves for the tests made on the RHEO knee.

(A)



(B)

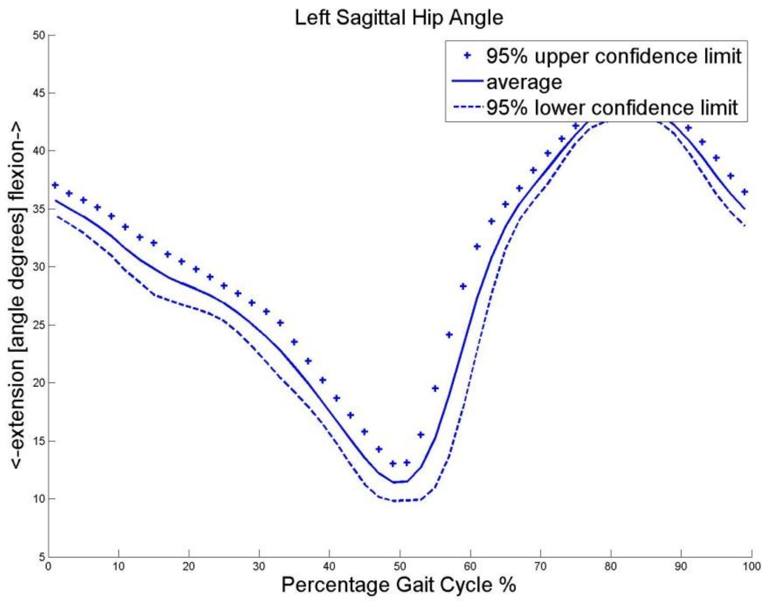


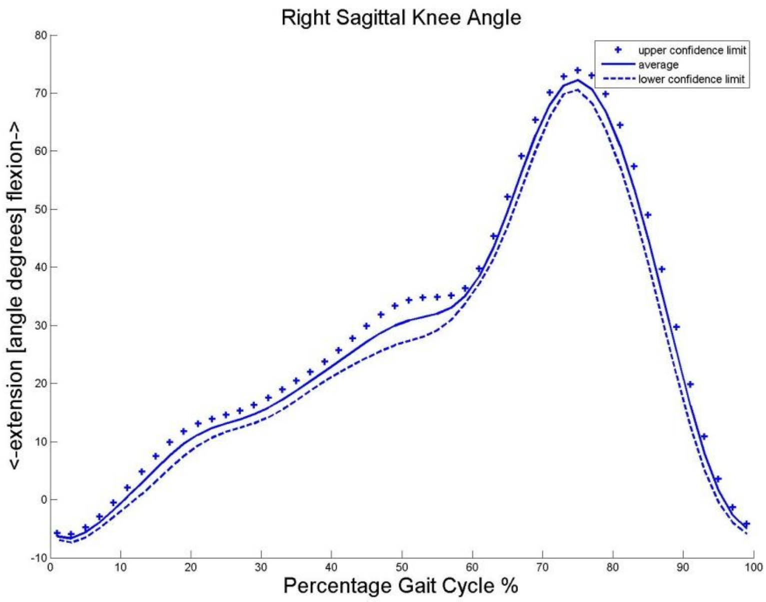
Figure 4-20 Intact limb (A) and prosthetic limb (B) hip angles during downhill walking on the RHEO knee

The motion curves obtained for the joints are slightly different. The joint on the prosthetic side is in extension from initial contact to late stance, with extension peak at 50% of the gait cycle and total extension trajectory of 34°. The hip joint on the intact side has a small flexion peak in early stance, 5°, and then switches to extension motion that peaks at about 55% of the gait cycle with 12° extension trajectory. At terminal stance the hip motion is flexion and the trajectory obtained for both joints is similar, 40° for the intact limb and 44° for the prosthetic limb.

4.3.2 Joint angles – Knee – 3R80 & RHEO

Figure 4-21 represents the anatomical (A) and the prosthetic (B) knee angle curves for the tests made on the 3R80 knee.

(A)



(B)

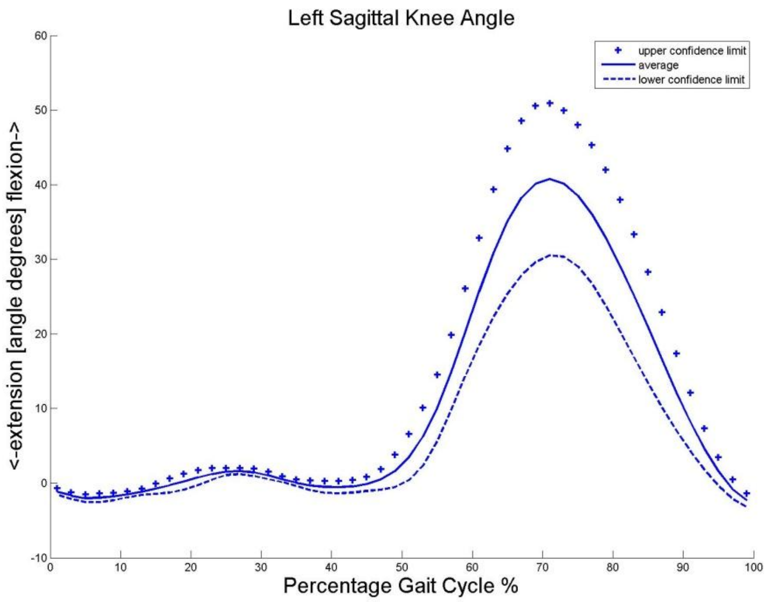
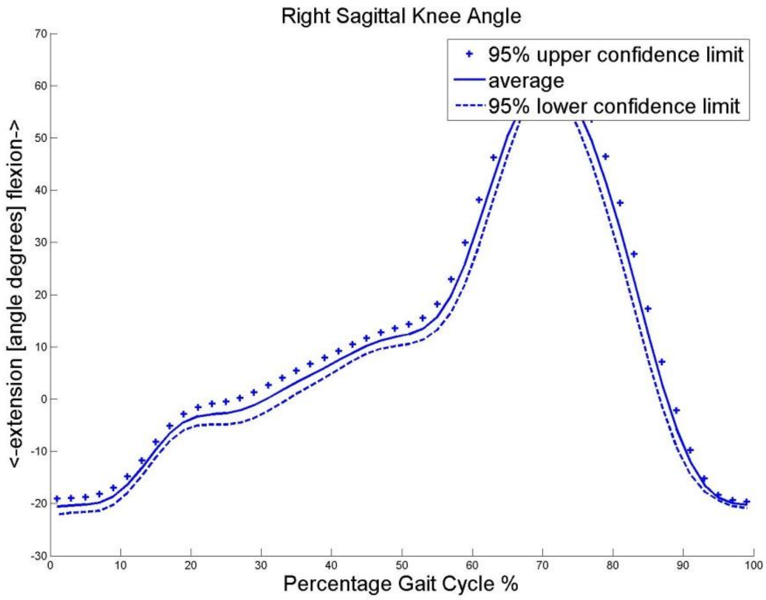


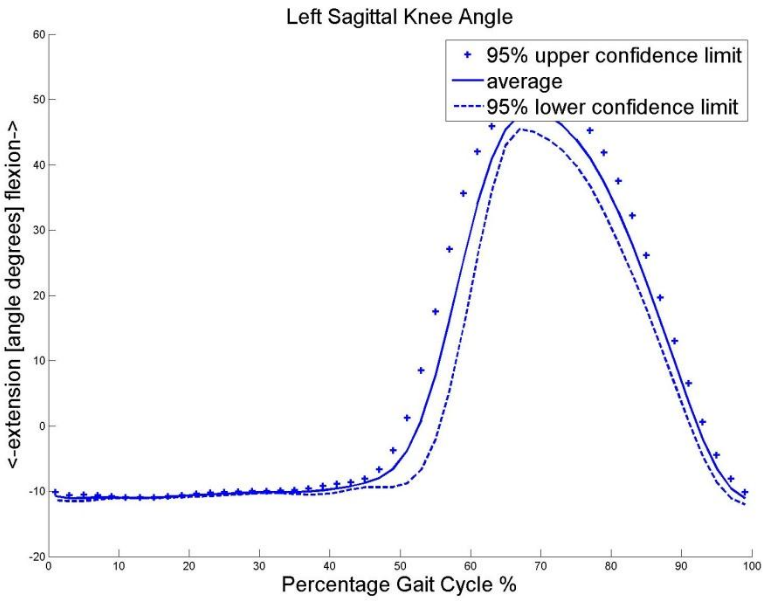
Figure 4-21 Intact limb (A) and prosthetic limb (B) knee angles during downhill walking on the 3R80 knee

The angular motions of the knee joints of the prosthetic and the intact limbs were dissimilar. The intact limb showed significantly increased knee flexion compared with when walking on level ground, from the initial contact and throughout the stance phase the knee was flexed. During mid-swing, peak knee flexion was reached, with flexion trajectory of 75°. The prosthetic knee exhibited similar lack of knee flexion during the stance phase during downhill walking as when the subject was walking on level ground. During the stance phase the knee flexed approximately 1°, and peak knee flexion during the swing phase was significantly decreased when compared with the intact limb, with peak flexion of about 40°.

Figure 4-22 represents the anatomical (A) and the prosthetic (B) knee angle curves for the tests made on the RHEO knee.

(A)





(B)

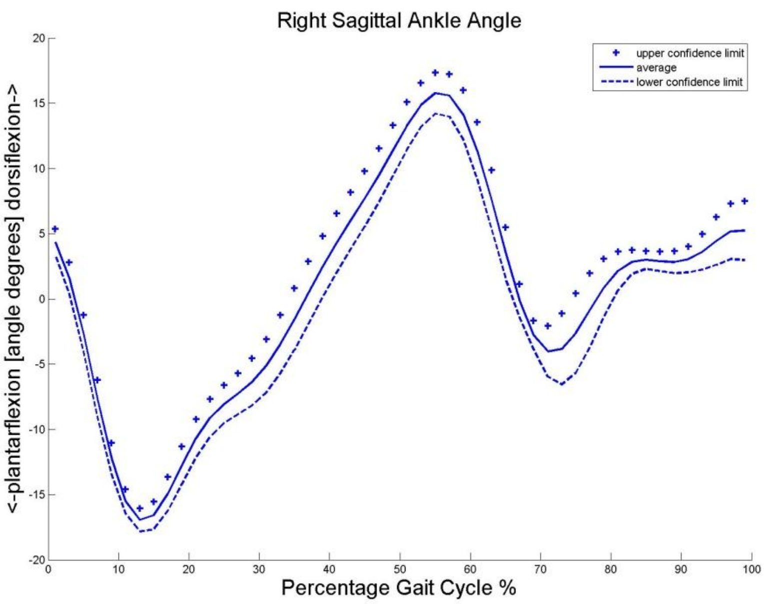
Figure 4-22 Intact limb (A) and prosthetic limb (B) knee angles during downhill walking on the RHEO knee

The angular motions of the prosthetic and anatomical knee joints were dissimilar during the stance phase. From initial contact and throughout the stance phase the intact knee was flexed with peak flexion angle occurring early swing of 80°. There was a total absence of prosthetic knee stance flexion and the peak prosthetic knee flexion during the swing phase was also significantly decreased when compared with the intact limb, with flexion trajectory of 59°.

4.3.3 Joint angles – Ankle – 3R80 & RHEO

Figure 4-23 represents the anatomical (A) and the prosthetic (B) ankle angle curves for the tests made on the 3R80 knee.

(A)



(B)

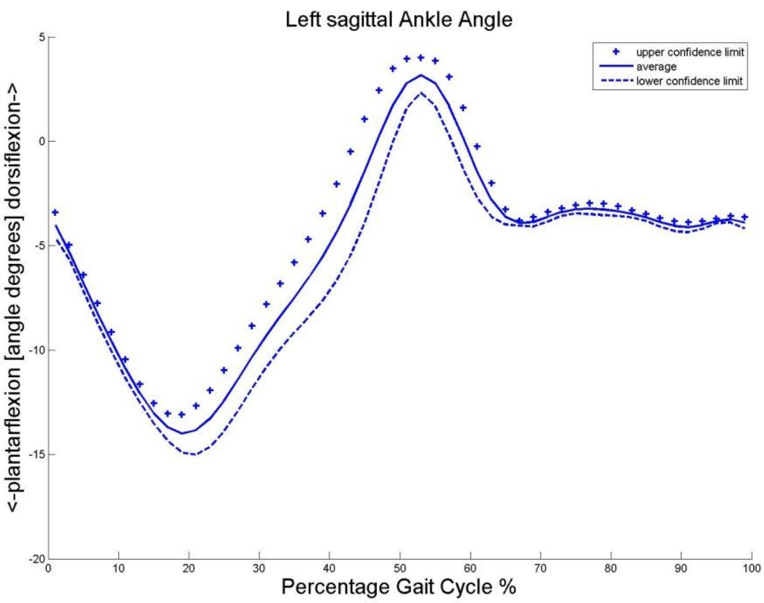
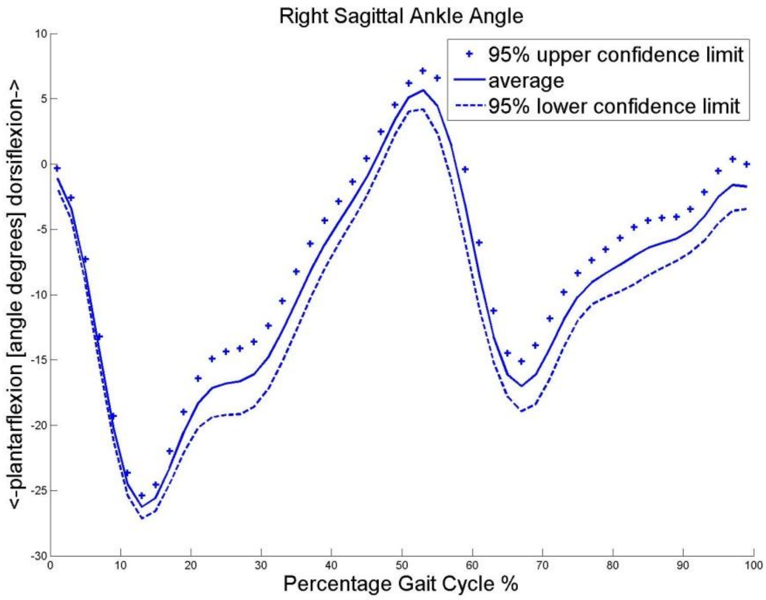


Figure 4-23 Intact limb (A) and prosthetic limb (B) ankle angles during downhill walking on the 3R80 knee

The anatomical ankle and the prosthetic ankle exhibited similar patterns of motion, however, with different peak plantar and dorsiflexion angles. During early stance the anatomical ankle exhibited peak plantar flexion of 23° but the prosthetic ankle showed significantly decreased peak plantar flexion, 10°. From terminal stance to early swing the anatomical ankle was in plantar flexion, with trajectory of 20°. The plantar flexion motion of the prosthetic ankle was considerably smaller than of the anatomical joint, with trajectory of only 7°.

Figure 4-24 represents the anatomical (A) and the prosthetic (B) ankle angle curves for the tests made on the RHEO knee.

(A)



(B)

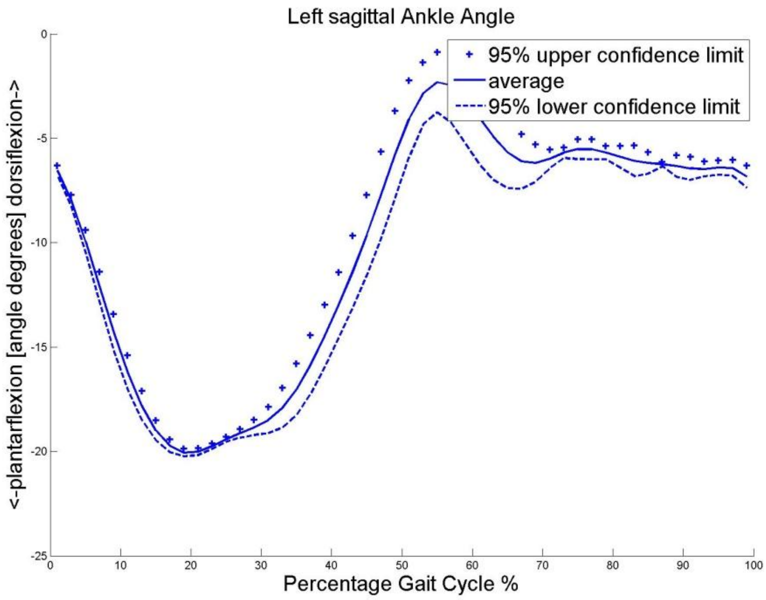


Figure 4-24 Intact limb (A) and prosthetic limb (B) ankle angles during downhill walking on the RHEO knee

The motions of the anatomical and the prosthetic ankles had similar patterns of plantar and dorsiflexion motion, however, with slightly different trajectories during early and mid-stance and early swing. During early stance the anatomical ankle exhibited peak plantar flexion of 26° but the prosthetic ankle was associated with considerably decreased peak plantar flexion, a trajectory of 13°. During the swing phase there was an absence of prosthetic ankle plantar flexion while the anatomical ankle was in 5° plantar flexion.

4.3.4 Joint angles - Comparison – RHEO knee & 3R80

Joint angles

Figure 4-25 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip angles for the RHEO knee and the 3R80 knee.

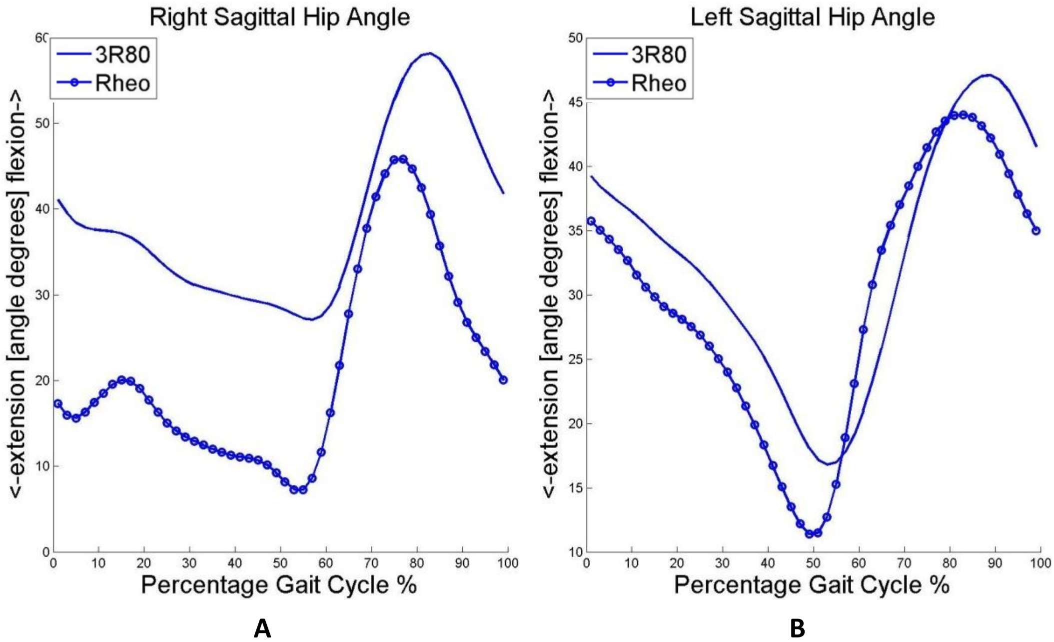


Figure 4-25 Intact limb (A) and prosthetic limb (B) hip angles during downhill walking on the RHEO knee and the 3R80 knee

The motion curves obtained for the hip joint on the intact side were very similar for the RHEO knee and the 3R80. Throughout the stance phase the joint was extending, apart from a small flexion peak occurring in early stance when using the RHEO knee. The total extension trajectories from initial contact to late stance were 20° for the 3R80 and 12° for the RHEO knee. The flexion motion from terminal stance to early swing had trajectory of 40° for the RHEO knee and peak extension occurring slightly earlier than for the 3R80, which was associated with extension trajectory of 30°. The motion pattern of the hip joint on the prosthetic side was very similar for both knees. From initial contact the hip was extending with trajectories of 34° for the RHEO knee and 23° for the 3R80. From late stance to mid swing the hip joint was flexing, with similar trajectories for both knees, about 44°.

Figure 4-26 represents the comparison on the anatomical (A) and the prosthetic (B) knee angles for the tests made on the 3R80 and the RHEO knee.

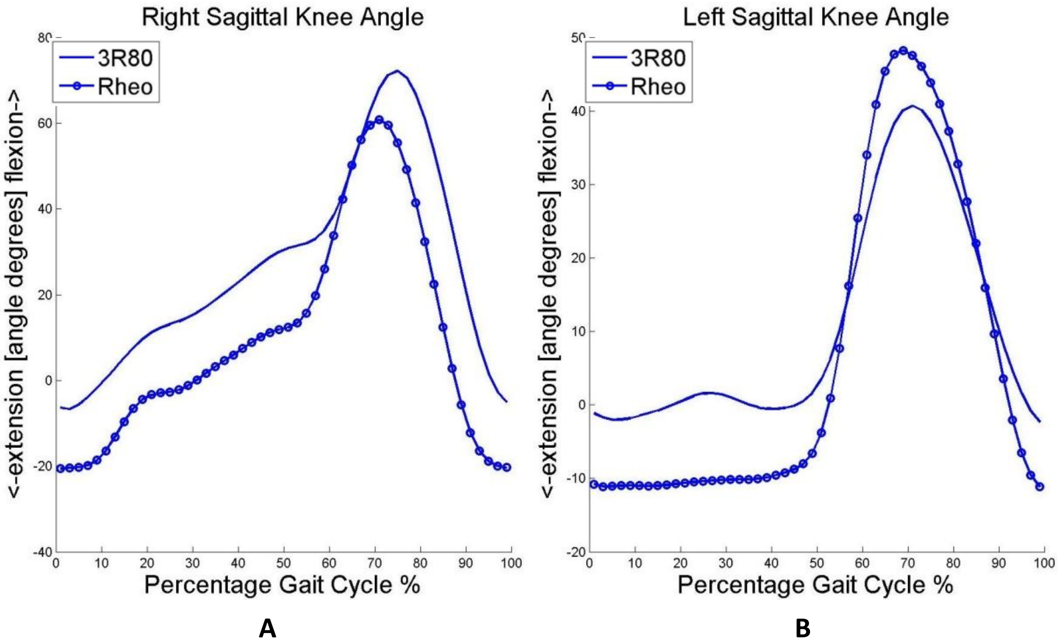


Figure 4-26 Intact limb (A) and prosthetic limb (B) knee angles during downhill walking on the RHEO knee and the 3R80 knee

The angular motion of the intact knee was similar for both knees. From initial contact and throughout the stance phase the knee was flexed with trajectory of 30° for the 3R80 and 35° for the RHEO knee. In the swing phase the RHEO knee was associated with peak flexion angle earlier, about 72% of the gait cycle compared with about 78% for the 3R80. The flexion trajectories were 50° for the RHEO knee and 40° for the 3R80. The motions of the prosthetic knees were comparable throughout the gait cycle. Neither of the joints flexed during the stance phase but had slightly different flexion trajectories in the swing phase, where the RHEO knee was associated with greater flexion trajectory, 59° compared with 40° for the 3R80.

Figure 4-27 represents the comparison on the anatomical (A) and the prosthetic (B) ankle angles for the tests made on the 3R80 and the RHEO knee.

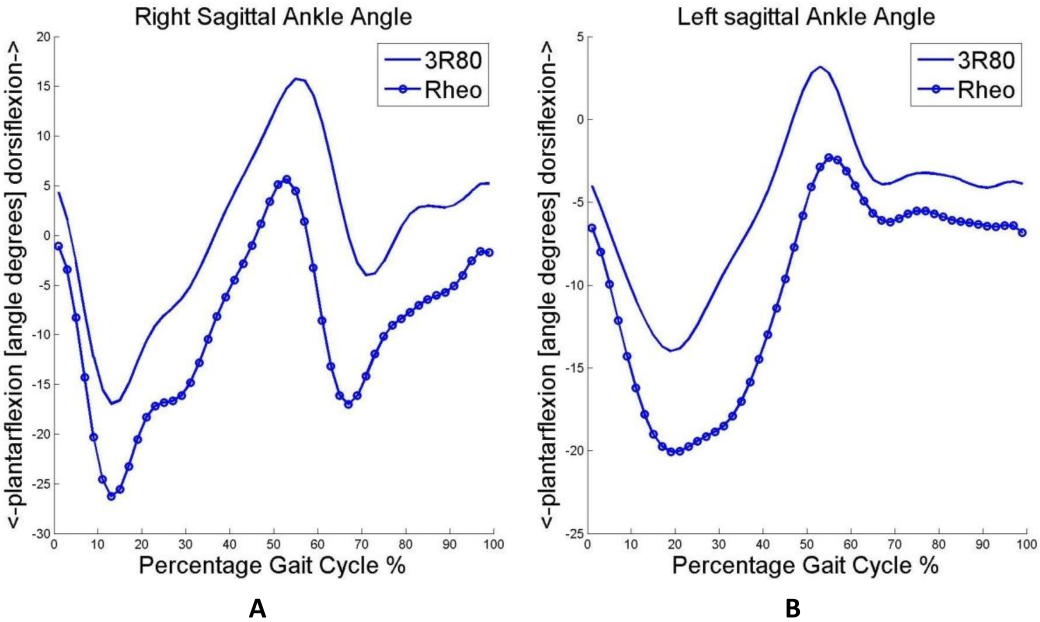


Figure 4-27 Intact limb (A) and prosthetic limb (B) ankle angles during downhill walking on the RHEO knee and the 3R80 knee

The motion of the anatomical ankle was identical when using both knees, with similar plantar flexion trajectory from initial contact to early stance and dorsiflexion from early to terminal stance. The prosthetic ankle also had similar motion for both knees; however, the RHEO knee was associated with slightly slower transition from plantar flexion to dorsiflexion in early stance.

Table 6 and 7 represent the hip and knee motion trajectories for the prosthetic (6) and the intact (7) sides that were considered relevant for the evaluation of the knees.

Table 6 Hip and knee extension and flexion motions of the prosthetic limb during downhill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**34**	**23**
Hip flexion trajectory in swing (degrees)	**44**	**44**
Knee flexion in stance (degrees)	**0**	**1**
Knee flexion trajectory in swing (degrees)	**59**	**40**

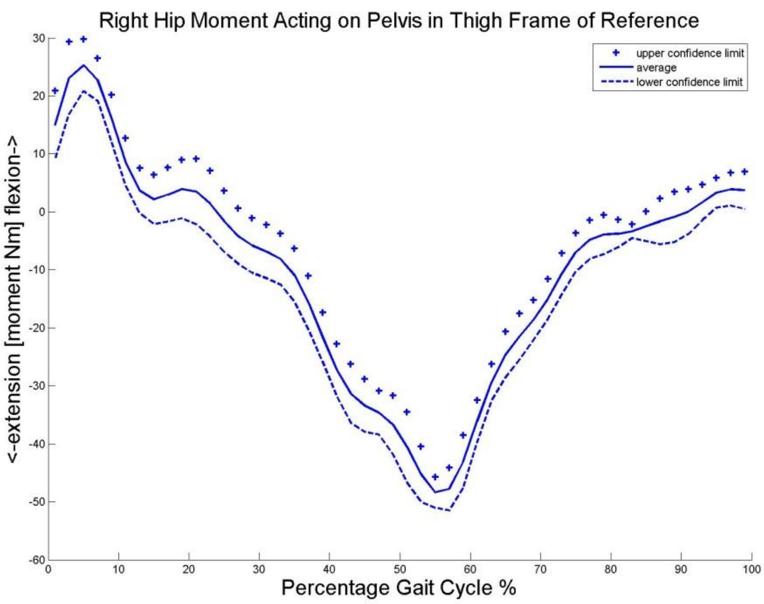
Table 7 Hip and knee extension and flexion motions of the intact limb during downhill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**12**	**20**
Hip flexion trajectory in swing (degrees)	**40**	**30**
Knee flexion in stance (degrees)	**35**	**30**
Knee flexion trajectory in swing (degrees)	**50**	**40**

4.3.5 Joint moments – Hip

Figure 4-28 represents the anatomical (A) and the prosthetic (B) hip moment curves for the tests made on the 3R80 knee.

(A)



(B)

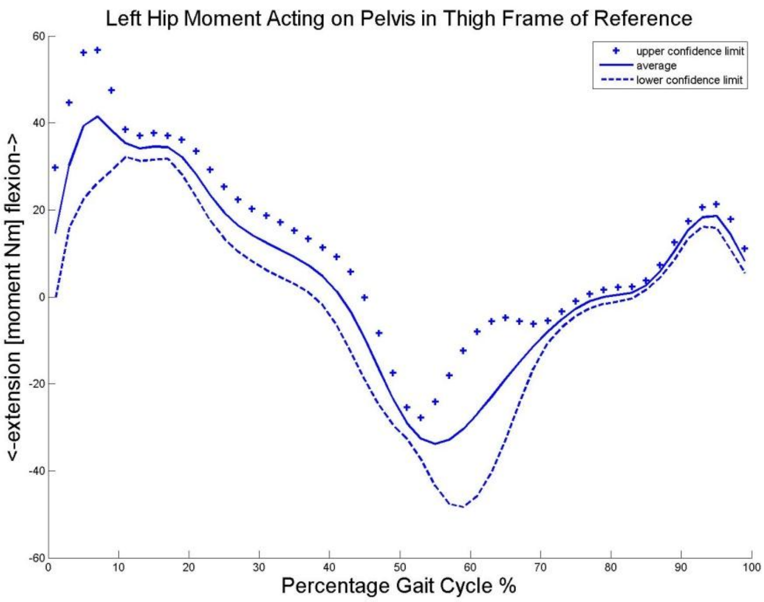
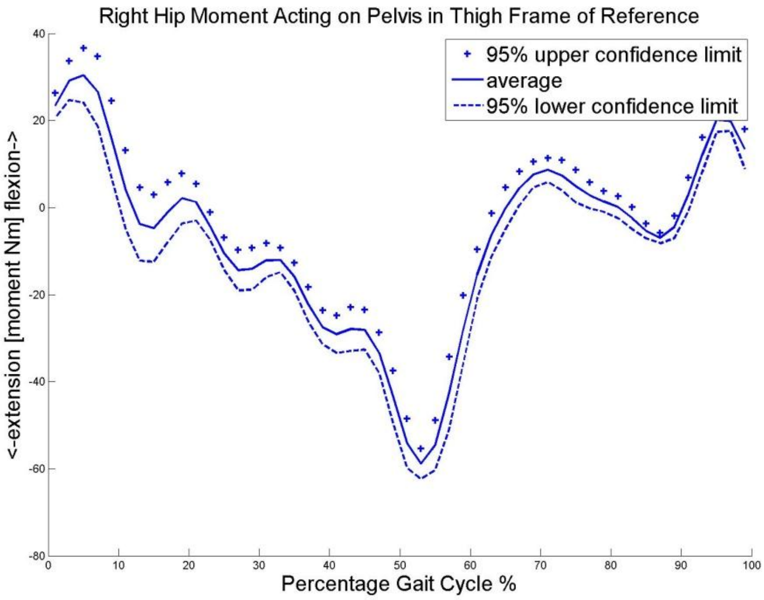


Figure 4-28 Intact limb (A) and prosthetic limb (B) hip moments during downhill walking on the 3R80 knee

The moment patterns obtained for the hip joints of the subject is similar but with different peak flexion and extension moment values. At initial contact there was a flexion moment about both joints, with trajectories of 24 Nm for the intact side and 25 Nm for the prosthetic side. From early to terminal stance there was an extension moment about both hip joints, with total trajectories of 75 Nm for both sides. The peak extension moment at push off was greater about the joint on the intact side, 50 Nm compared with 35 Nm about the joint on the prosthetic side. After peak extension moment was reached the moments about the joints switched to flexion moment that peaked during the swing phase.

Figure 4-29 represents the anatomical (A) and the prosthetic (B) hip moment curves for the tests made on the REHO knee.

(A)



(B)

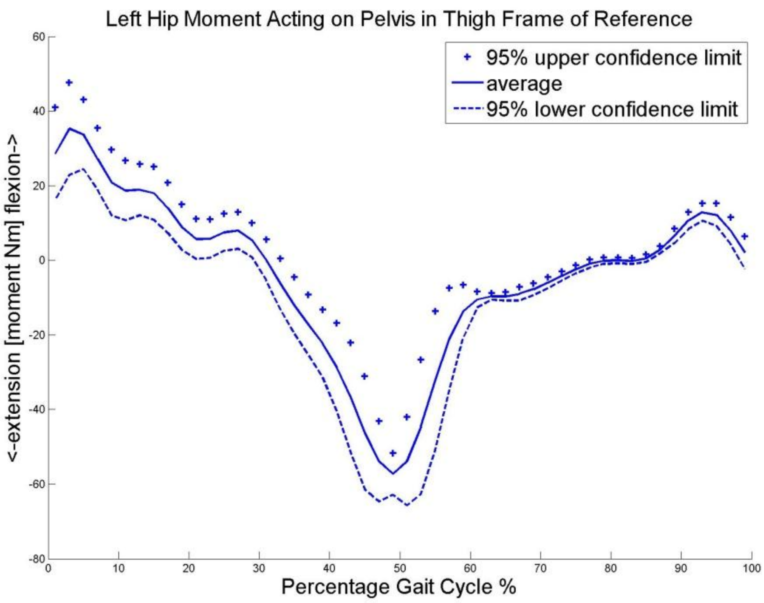


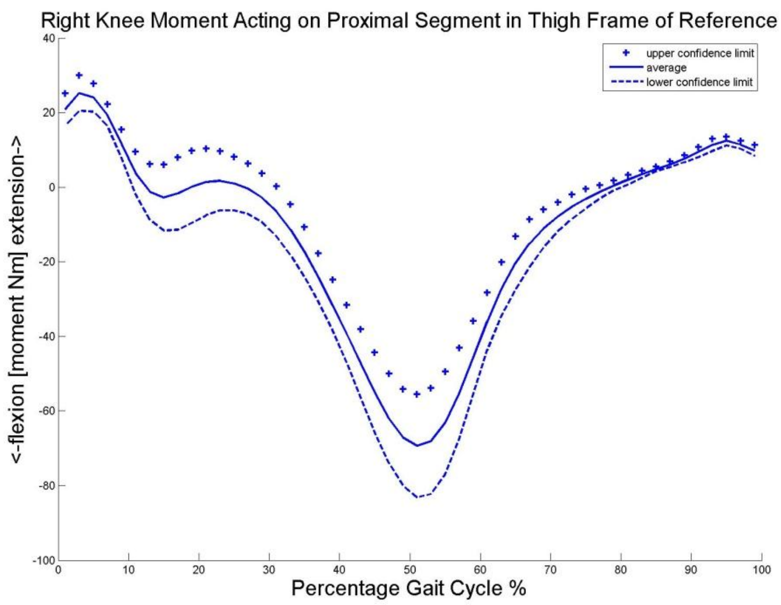
Figure 4-29 Intact limb (A) and prosthetic limb (B) hip moments during downhill walking on the RHEO knee

The moment pattern obtained for the hip joints of the subject has a fluctuating pattern throughout the gait cycle. Right after initial contact there was a flexion moment about both of the joints, 7 Nm on the prosthetic side and 10 Nm on the intact side. From early to late stance there was an extension moment about the joints, with fluctuating trajectory and few small extension peaks, especially about the intact side. The trajectories for the joints were 94 Nm for the prosthetic limb and 85 Nm for the intact limb. From late stance to late swing there was a flexion moment about the joint on the prosthetic side, with trajectory of 70 Nm. For the intact limb there was a flexion moment about the joint from late stance to early swing, with trajectory of 70 Nm and from early to mid-swing there was an extension moment with trajectory of 15 Nm.

4.3.6 Joint moments – Knee

Figure 4-30 represents the anatomical (A) and the prosthetic (B) knee moment curves for the tests made on the 3R80 knee.

(A)



(B)

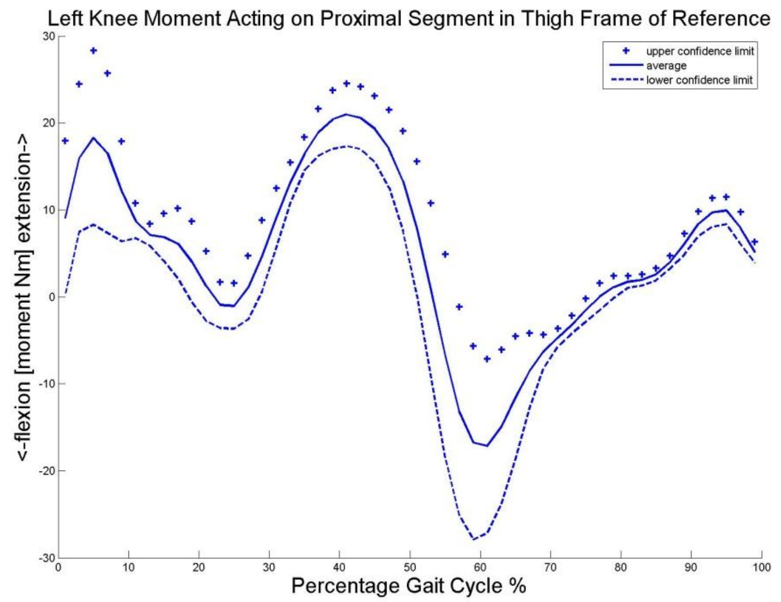


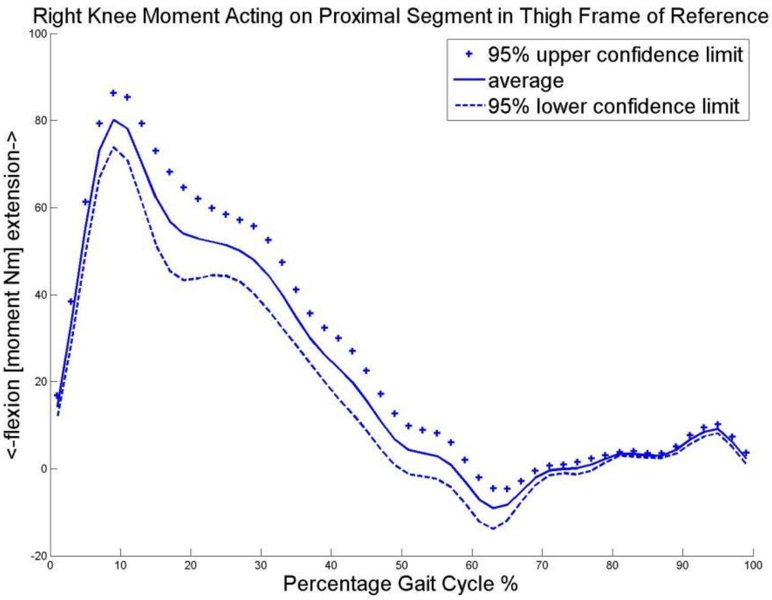
Figure 4-30 Intact limb (A) and prosthetic limb (B) knee moments during downhill walking on the 3R80 knee

The curves for the knee joint moments for the limbs are very different as seen when they are compared. The moment curve for the prosthetic knee has a fluctuating pattern of flexion and extension moment throughout the gait cycle. At initial contact there was an extension moment about the prosthetic knee joint with trajectory of 18 Nm, with peak at early stance. Another extension peak moment occurred during late stance with value of 20 Nm. At terminal stance peak flexion moment with trajectory of 40 Nm occurred and then during the swing phase there was an extension moment about the knee.

The intact knee showed a different pattern of flexion and extension moments. Mid to late stance there was a flexion moment about the knee with a small extension peak in early stance. The flexion trajectory from early to terminal stance was 90 Nm. During late stance there was a flexion moment about the knee until swing phase it switched to extension moment.

Figure 4-31 represents the anatomical (A) and the prosthetic (B) knee moment curves for the tests made on the RHEO knee.

(A)



(B)

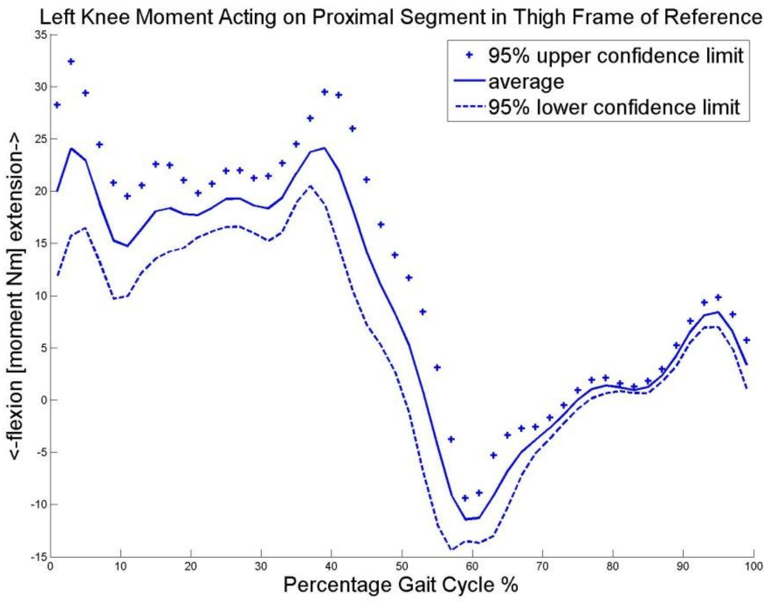


Figure 4-31 Intact limb (A) and prosthetic limb (B) knee moments during downhill walking on the RHEO knee

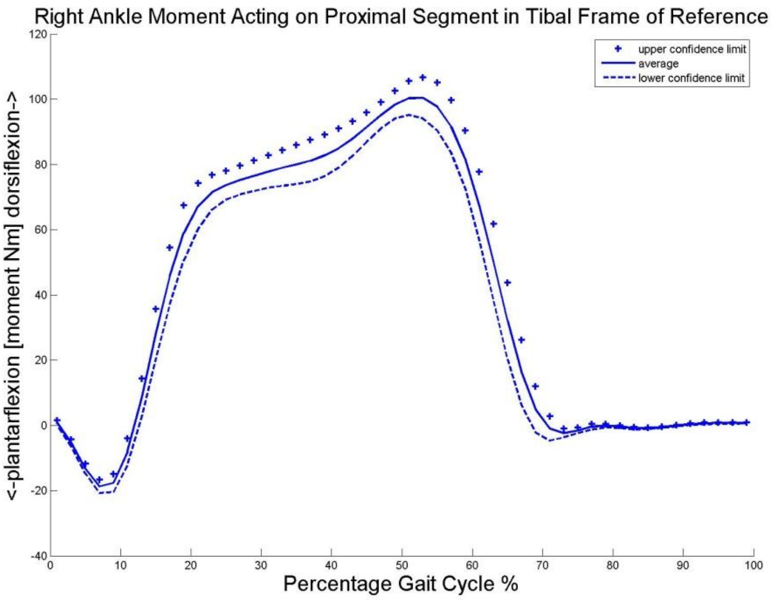
The curves obtained for the knee joint moments for the limbs have a different pattern of extension and flexion moments about the knee joints. At initial contact there was an extension moment about the prosthetic knee joint with 4 Nm peak occurring at early stance. Another small extension peak moment occurred in late stance with trajectory of 4

Nm. At terminal stance peak flexion moment of 9 Nm occurred and then during the swing phase there was an extension moment about the knee.

The knee joint on the intact side had a different pattern of flexion and extension moments. At initial contact there was an extension moment about the knee, with trajectory of 68 Nm. From early to terminal stance there was a flexion moment about the joint, with trajectory of 90 Nm.

4.3.7 Joint moments – Ankle

Figure 4-32 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the 3R80 knee.



(B)

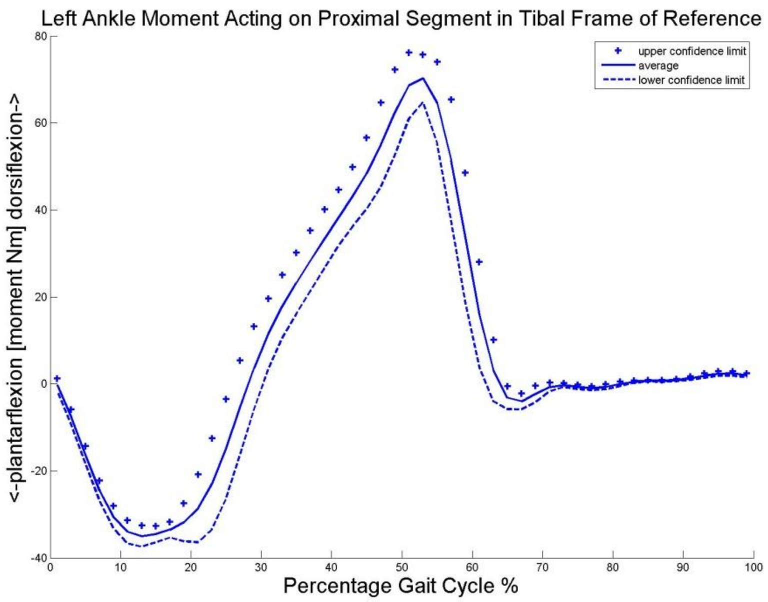
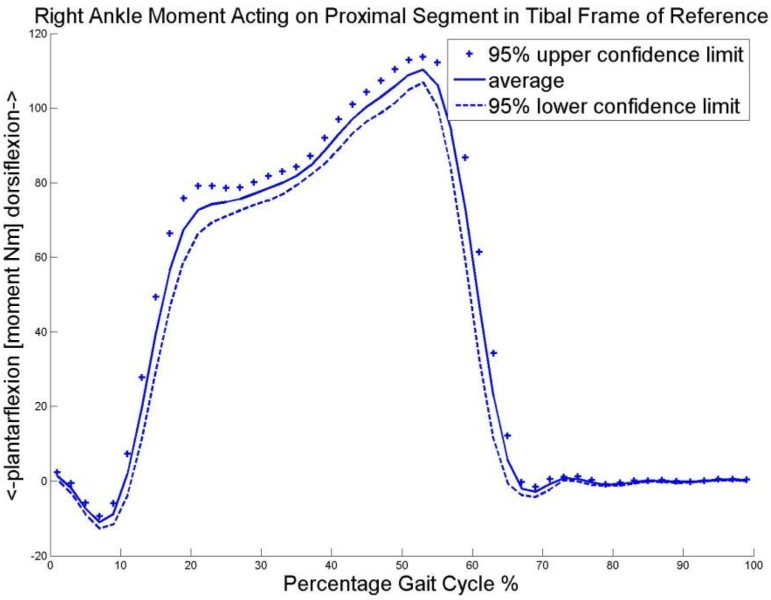


Figure 4-32 Intact limb (A) and prosthetic limb (B) ankle moments during downhill walking on the 3R80 knee

The ankle joint moments had different curves for the anatomical and the prosthetic joints. At initial contact and in early stance there was a plantar flexion moment about both joints. The plantar flexion moment about the prosthetic ankle in early stance was of significantly greater magnitude, with peak value of 35 Nm compared with peak value of 20 Nm for the anatomical ankle. After the plantar flexion peak, the joint moment switched to dorsiflexion moment for both joints with slightly different trajectories, 120 Nm for the anatomical ankle and 50 Nm for the prosthetic ankle. Around push-off the dorsiflexion moment about the joints peaked, with different peak values for the anatomical and the prosthetic joints, 100 Nm and 70 Nm respectively. During the swing phase there was a plantar flexion moment about both joints that fell down to zero at the end of the gait cycle.

Figure 4-33 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the RHEO knee.

(A)



(B(

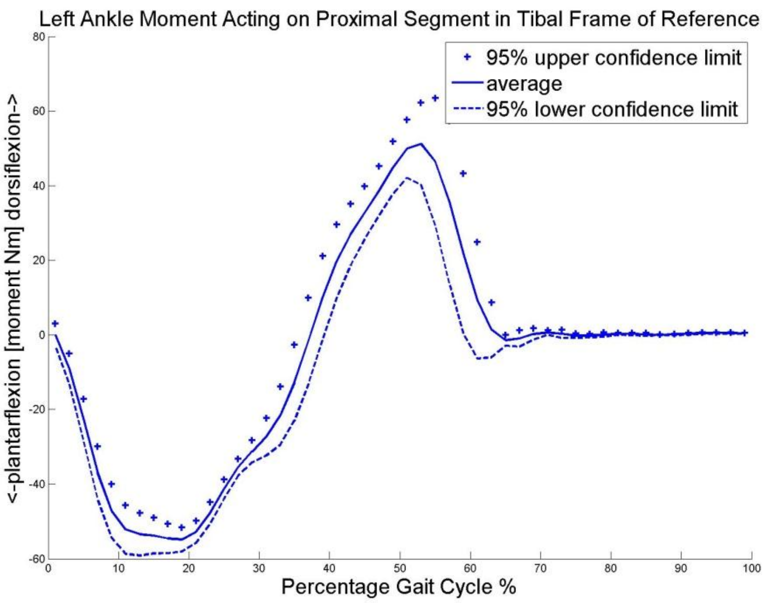


Figure 4-33 Intact limb (A) and prosthetic limb (B) ankle moments during downhill walking on the RHEO knee

At initial contact and early stance there was a plantar flexion moment about both ankle joints. The peak plantar flexion moment during early stance was significantly greater about the prosthetic ankle, with peak value of 55 Nm compared with peak value of 10 Nm for the anatomical ankle. After the plantar flexion peak, the joint moment switched to dorsiflexion moment for both joints with slightly different curve. Around push-off the dorsiflexion moment about the joints peaked, with different trajectories for the anatomical and the prosthetic joints, 120 Nm and 100 Nm respectively. During the swing phase there was a plantar flexion moment about both joints that fell down to zero at the end of the gait cycle.

4.3.8 Joint moments - Comparison – RHEO knee & 3R80

Figure 4-34 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip moments for the RHEO knee and the 3R80 knee.

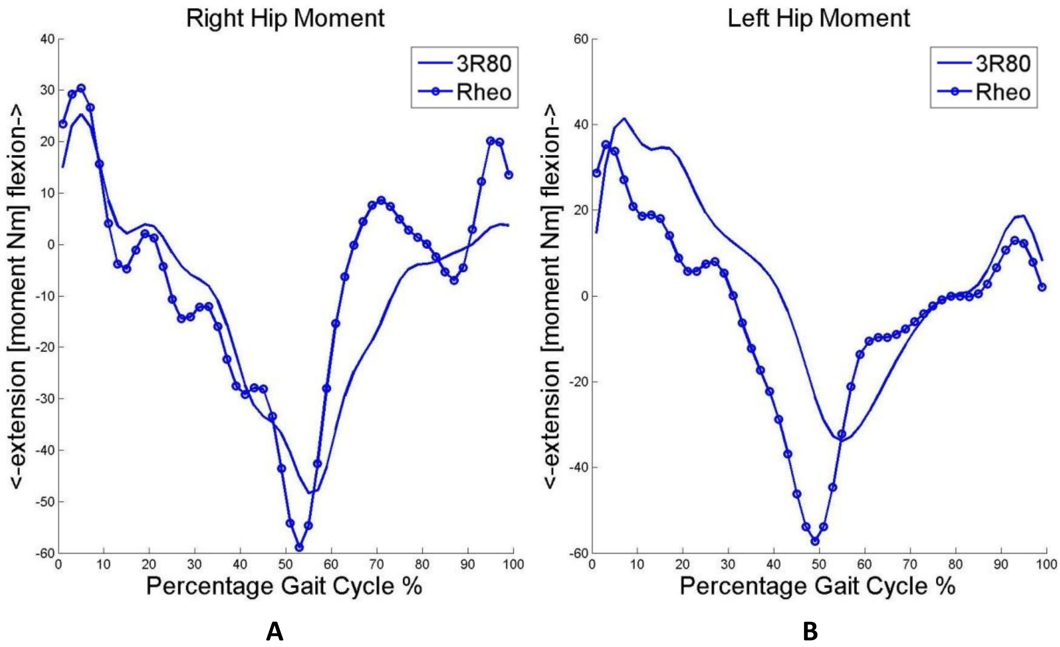
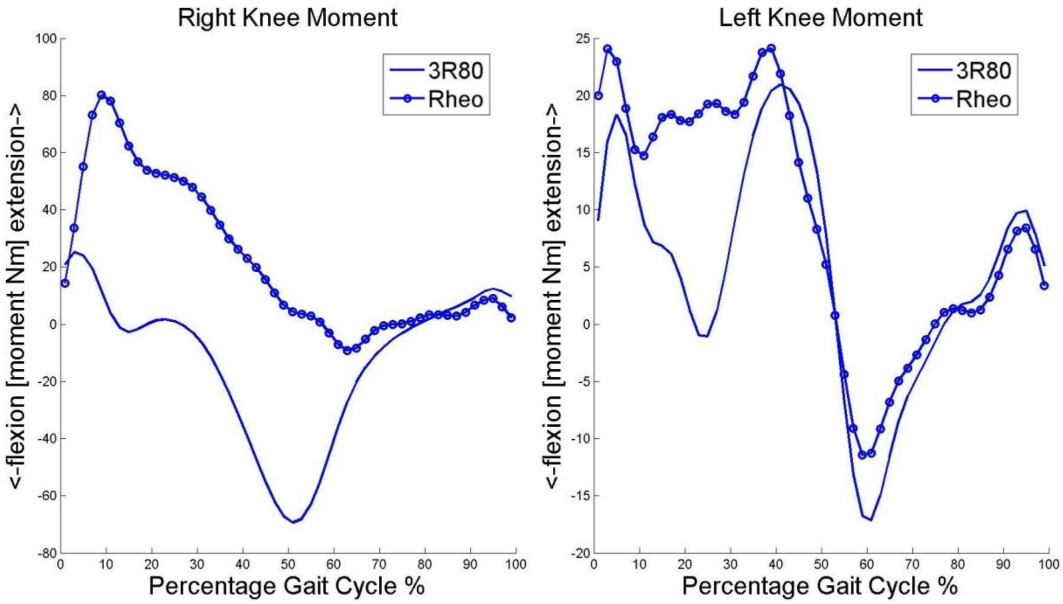


Figure 4-34 Intact limb (A) and prosthetic limb (B) hip moments during downhill walking on the RHEO knee and the 3R80 knee

The moment patterns obtained for the hip joints on the intact side when using the two types of prostheses was similar, however the curve for the RHEO knee has more fluctuating pattern. When using both prostheses peak hip extension occurs at similar instant, for the RHEO knee the peak occurs slightly earlier and the extension trajectory is slightly larger than for the 3R80, 90 Nm and 75 Nm respectively. After the peak hip extension moment a flexion moment is about both joints and the RHEO knee is associated with more rapid transition from hip extension to flexion moment and the hip flexion moment increased rapidly from terminal stance to early swing. In mid swing there was an extension moment and again flexion moment from late to terminal swing. When using the 3R80 the hip joint moment did not have as fluctuating pattern. From terminal stance to terminal swing the moment steadily increased.

The hip moment patterns on the prosthetic side were similar when using both knees apart from different extension trajectories from early to late swing, where the RHEO knee was associated with more rapid change from hip extension to hip flexion and increased extension moment when compared with the 3R80, 94 Nm and 75 Nm.

Figure 4-35 represents the comparison on the anatomical (A) and the prosthetic (B) knee moments for the tests made on the 3R80 and the RHEO knee.

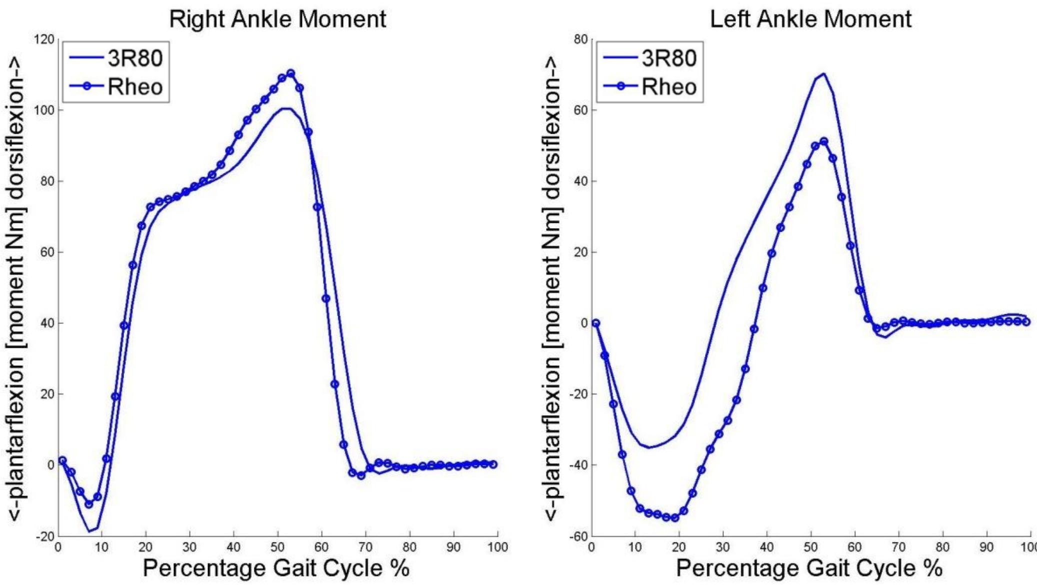


A B

Figure 4-35 Intact limb (A) and prosthetic limb (B) knee moments during downhill walking on the RHEO knee and the 3R80 knee

The curves representing the moments for the intact knee joint when using the two types of prostheses had similar pattern of flexion and extension moments, however, with different peak values. When using the RHEO knee the extension moment at initial contact to early stance is of greater magnitude. The extension moments from early to terminal stance was identical when using both knees but when using the 3R80 there was a sharp change from flexion to extension moment at about 50% of the gait cycle, about push off. When using the RHEO knee the transition from extension to flexion moment was sharp but the extension moment peak was very small. The moments about the prosthetic knee joints were different from early to late stance. At initial contact both joints experienced extension moment, of great magnitude about the 3R80. For the RHEO knee the moment pattern from early to late stance had a fluctuating pattern, with small flexion and extension peaks. For the 3R80 there was a considerably great flexion moment about the joint during the stance phase. During the swing phase the moments about the knee were identical.

Figure 4-36 represents the comparison on the anatomical (A) and the prosthetic (B) ankle moments for the tests made on the 3R80 and the RHEO knee.



A B

Figure 4-36 Intact limb (A) and prosthetic limb (B) ankle moments during downhill walking on the RHEO knee and the 3R80 knee

The moment patterns obtained for the anatomical ankle joint when using the two types of prostheses were very similar. The moments about the prosthetic ankle were also similar when using the two knees. The difference was that the RHEO knee was associated with greater plantar flexion moment during early stance phase.

Table 8 and 9 represent the hip and knee moment trajectories for the prosthetic (8) and the intact (9) sides that were considered relevant for the evaluation of the knees.

Table 8 Hip and knee extension and flexion moments of the prosthetic limb during downhill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension moment trajectory at push off (Nm)	**94**	**75**
Knee extension moment trajectory in early stance (Nm)	**4**	**18**
Knee flexion moment trajectory in early stance (Nm)	**10**	**20**

Table 9 Hip and knee extension and flexion moments of the intact limb during downhill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension moment trajectory at push off (Nm)	**90**	**75**
Knee extension moment trajectory in early stance (Nm)	**68**	**5**
Knee flexion moment trajectory from early to terminal stance (Nm)	**90**	**90**

4.4 Discussion for downhill walking

4.4.1 RHEO knee and 3R80

Since the 3R80 knee joint does not utilise any adjustment settings to adjust the knee to slope walking, compensation movements were expected. During downhill walking the motions of the hip joints were dissimilar when using both the 3R80 and the RHEO knee. Both knees were associated with greater peak values of extension and flexion for the prosthetic limb. That is not agreeing with results from Vrieling et al (2008) (Chapter 2.8) that reported decreased hip flexion of the prosthetic limb during the swing phase. The curves obtained for the hip joint on the prosthetic side showed greatly more abrupt transition from extension to flexion for both knees. That indicates that during downhill walking the subject rapidly started to flex the hip right after full extension to facilitate knee extension and secure that the prosthetic knee joints were fully extended when prosthetic foot was placed on the ground.

When using both knees the intact knee joint exhibited increased knee flexion during the stance when compared with level walking, however peak knee flexion value during mid-swing is similar for both environments. Similar results have been reported by Kuster et al (1995) (Chapter 2.8). As for the prosthetic knee joints the subject did not show stance phase knee flexion, similar results have been reported by Vrieling et al (2008) (Chapter 2.8). The subject kept the knees locked through the loading period of the gait cycle, similarly as he was walking on the level ground. When walking on declining surface the risk of the knee buckling is increased, to prevent that from happening the subject keeps the knee joint locked in extended position.

For both knees the results for the prosthetic ankle during downhill walking were similar to results obtained for level walking, as in both circumstances there is significantly decreased plantar-flexion. Apart from different peak values and lack of plantar flexion of the prosthetic ankle, the patterns obtained for the ankle joints are similar.

The results for the hip joint moments showed a great difference between the joint moments when using both knees. During early stance there was a greater flexion moment about the hip joint on the prosthetic side but in late stance there was a greater extension moment about the hip on the intact side, which applies for both the RHEO knee and the 3R80. Vrieling et al (2008) also reported greater hip extension on the intact side during downhill walking (Chapter 2.8) and the author of the present study agrees to Vriealing et al (2008) conclusion that the reason is most likely smaller step length on the prosthetic side.

The results for the knee joint moments showed a significant difference between the intact and the prosthetic knee during downhill walking for both knees. As the prosthetic knee joints were locked in extension throughout the stance phase contrary to the intact knee. The moments about the joints were different. When walking downhill the plantar-flexion moment at push-off is much less than during level walking, the reason is the force required at push-off during downhill walking is much less compared to walking on level ground. Similar results have been reported for intact individuals by Kuster et al (1995) (Chapter 2.8). The design of the Echelon foot utilises hydraulic control, the design of the foot aims to provide self-alignment and adapt foot positioning to varied terrain. The flexibility of the Echelon foot enabled the subject to achieve the dorsiflexion of the foot during mid to late stance.

4.4.2 Discussion about the comparison of both knees for downhill walking

The motion of the hip joint on the intact side was similar when using the 3R80 and the RHEO knee. As for the level walking the RHEO knee was associated with slightly more rapid transition from hip extension to flexion and greater flexion trajectory. This could indicate that the subject was relying more on the intact limb than the prosthetic limb, possible the subject was not as confident when walking and using the RHEO knee as the 3R80. With time and adjustment the subject would possibly be able to demonstrate more natural gait when using the RHEO knee. The level of resistance provided by the knee depends on the axial load on the prosthesis, thus inadequate load on the knee results in improper resistance

(Hirons, 2012). When using the RHEO knee the user must be confident in loading the prosthesis to benefit from the variable resistance provided.

The angular motion of the hip joint on the prosthetic side was also similar for both knees. In early stance a small flexion peak was associated with the RHEO knee, this flexion motion is difficult to justify. As for the joint on the intact side the RHEO knee was associated with more rapid transition from hip extension to flexion on the prosthetic side.

The prosthetic knee joints exhibited identical pattern of motion, and when using neither of the joints the subject was able to flex the knees to cushion the impact when the heel contacted the descending ramp. When walking downhill there is undeniable increased risk of the knee buckling when weight bearing but by keeping the knee locked in extension the subject prevents that. Despite different design of the 3R80 and the RHEO knee the knees demonstrate identical pattern of motion and were equally rapid in extension and flexion motion. The intact knee demonstrated similar pattern of motion when using both knee types, however, when using the RHEO knee the intact knee reached peak knee flexion in mid swing slightly earlier. That could be due to increased hip extension motion on the prosthetic side when using the RHEO knee.

The motions of the prosthetic ankle were similar when using both knee types. In early stance the RHEO knee was associated with more slowly transition from plantar flexion into dorsiflexion. This could indicate smoother transition from plantar flexion to dorsiflexion when using the RHEO knee. For the rest of the gait cycle the motion of the joint was identical for both knees which indicates that overall the use of microprocessor controlled knee did not really impact the dynamic properties of the prosthetic and the anatomical ankle joints.

The difference in the moments about the hip on the prosthetic side involved considerably faster transition from extension moment to flexion moment about toe off for the RHEO knee and the flexion and extension moments associated with RHEO knee were of greater magnitude. That could indicate that when using the RHEO knee the subject was using and loading the prosthetic limb more naturally then when walking on the 3R80, however, it could also indicate that more effort was needed to stabilise and control the prosthetic limb when using the RHEO. The extension moment about the hip joint of the intact limb had fluctuating pattern when using the RHEO knee. These moment fluctuations are likely due to some compensatory movements. As for the hip joint on the prosthetic side the RHEO knee was associated with more rapid transition from extension to flexion moment and the flexion moment in early swing was greater when using the RHEO knee.

The moments about the prosthetic knee joints were greatly different from early to late stance. The moment pattern for the RHEO knee has more fluctuating pattern and these moment fluctuations were likely needed to stabilise the knee during the stance phase until late stance where the knee flexed as it was unloaded before swing. As for the level walking the flexion moment during stance was greater about the 3R80 due to design of the knee. That is inconsistent with reports by Bellman et al (2010), which reported greater prosthetic knee flexion moment in stance phase about a microprocessor controlled knee than mechanical knee joints during downhill walking (chapter 2.10.3). The extension moments about the knees were identical. The moments about the intact knee were considerably different. The much greater extension moment about the intact knee, associated with the RHEO knee, at initial contact is difficult to justify. Possibly when using the RHEO knee the initial impact was harsher, due to uneven weight bearing of the prosthetic and the intact limb. From mid to late stance the 3R80 knee was associated with greater increase in the magnitude of the knee flexion moment when compared with the RHEO knee. This could be an indication of more even weight bearing of the limbs when using the 3R80 knee.

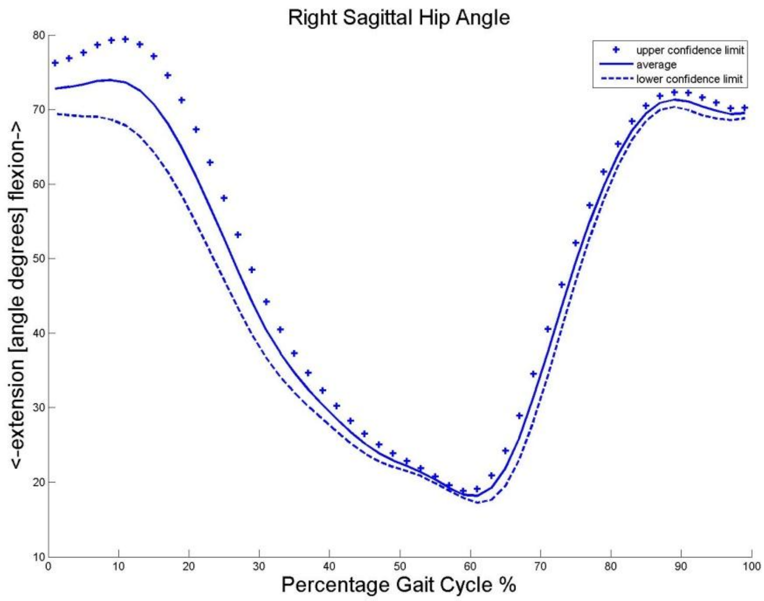
The results for the intact ankle showed that the moment about the joint was identical when using the 3R80 and the RHEO knee. Similarly for the prosthetic ankle the different knee types did not influence the joint to any extent.

4.5 Uphill walking

4.5.1 Joint angles - Hip – 3R80 & RHEO

Figure 4-37 represents the anatomical (A) and the prosthetic (B) hip angle curves for the tests made on the 3R80 knee.

(A)



(B)

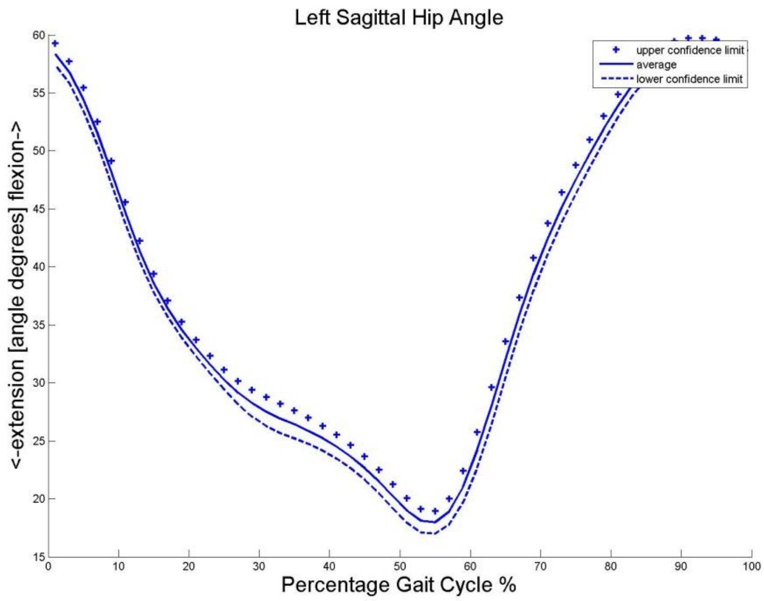
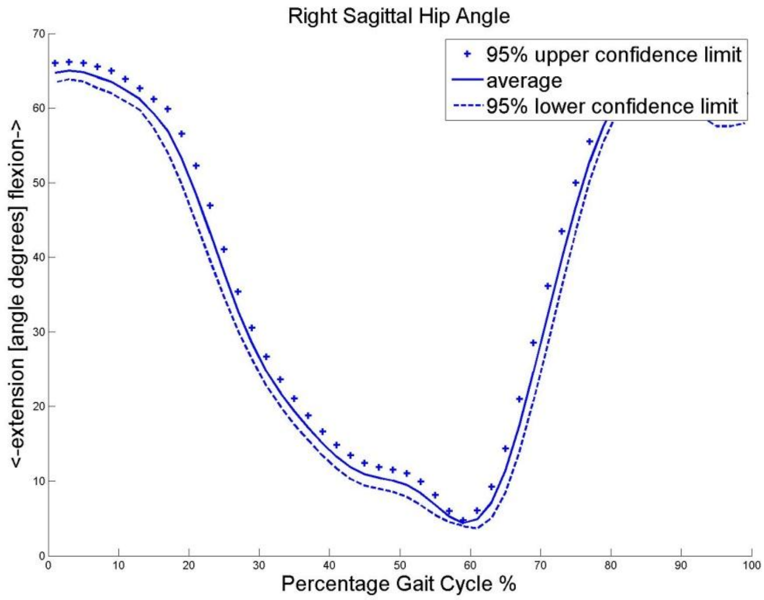


Figure 4-37 Intact limb (A) and prosthetic limb (B) hip angles during uphill walking on the 3R80 knee

During uphill walking the motion of the hip joints of the prosthetic and the intact limbs had similar patterns of extension and flexion, however, with different peak values. Both joints experienced hip extension during early stance and at terminal stance, where peak extension was reached with trajectory of 55° for the intact limb and 43° for the prosthetic limb. At terminal stance the hip motion changed to hip flexion and the prosthetic limb exhibited decreased hip flexion motion during the swing phase compared with the intact limb, with trajectory of 45° and 54° respectively.

Figure 4-38 represents the anatomical (A) and the prosthetic (B) hip angle curves for the tests made on the REHO knee.

(A)



(B)

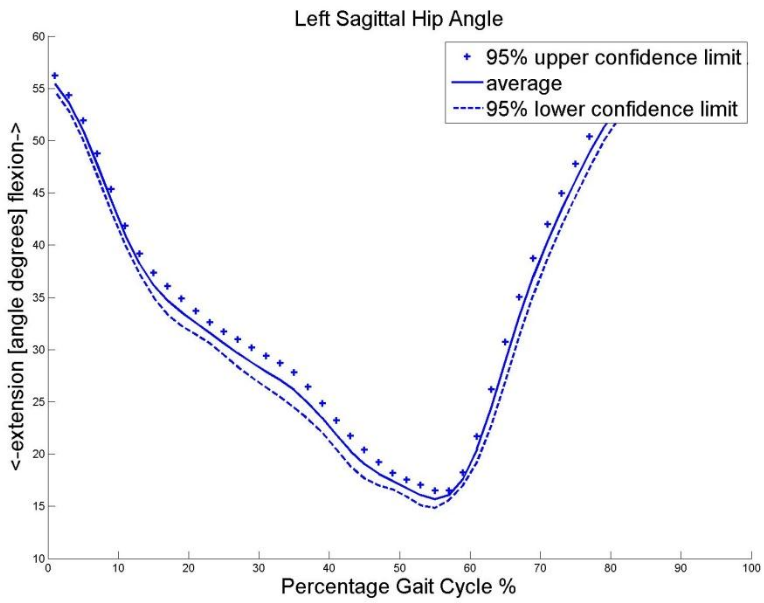


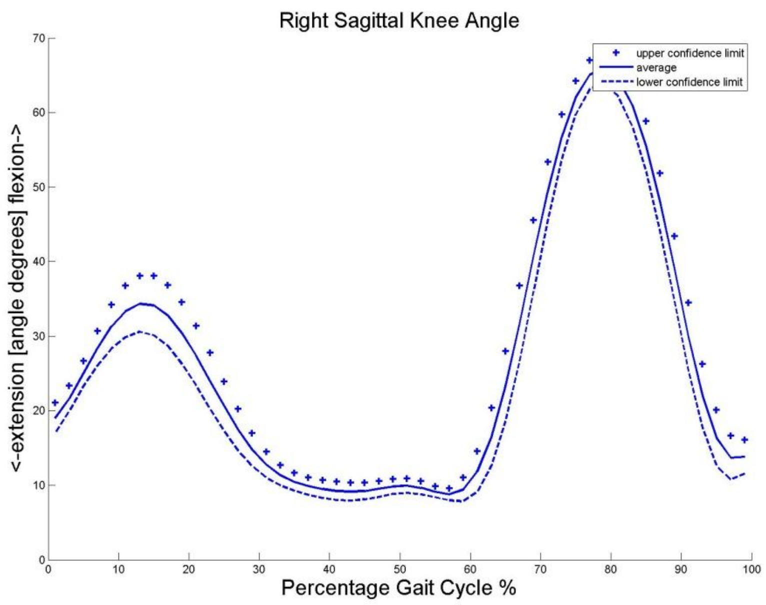
Figure 4-38 Intact limb (A) and prosthetic limb (B) hip angles during uphill walking on the RHEO knee

The motion of the hip joints on the prosthetic and the intact sides had same pattern of extension and flexion with slightly different trajectories during early stance. Both joints experienced hip extension during early stance with peak at terminal stance, with total trajectory of 60° for the intact limb and 40° for the prosthetic limb. At terminal stance the hip movement changed to hip flexion and the prosthetic limb exhibited decreased hip flexion motion during the swing phase compared with the intact limb, with trajectory of 43° and 60° respectively.

4.5.2 Joint angles – Knee – 3R80 & RHEO

Figure 4-39 represents the anatomical (A) and the prosthetic (B) knee angle curves for the tests made on the 3R80 knee.

(A)



(B)

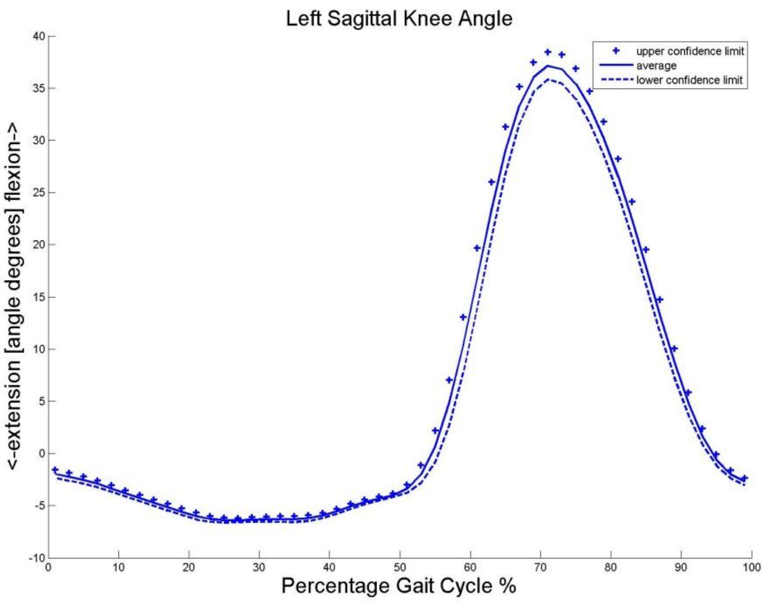
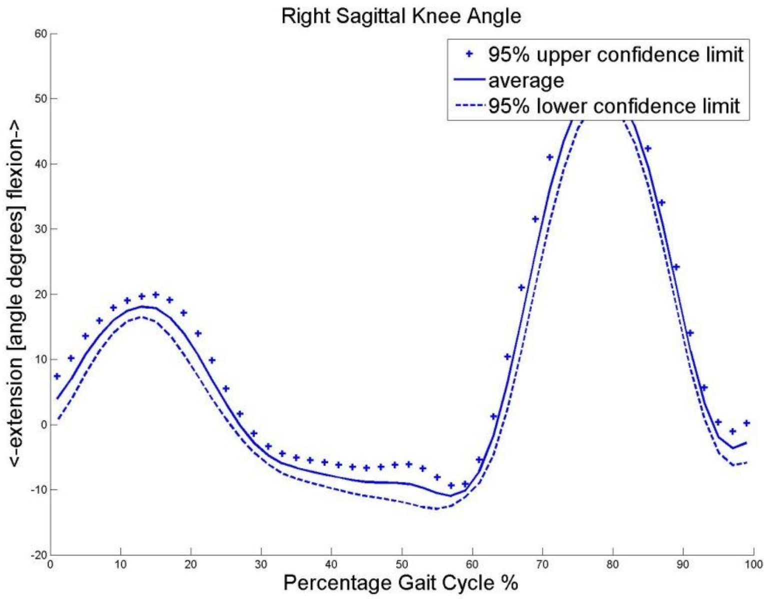


Figure 4-39 Intact limb (A) and prosthetic limb (B) knee angles during uphill walking on the 3R80 knee

The angular motions of the intact and the prosthetic knee joints were different. For the prosthetic knee there was a total absence of flexion throughout the stance phase, and the knee joint was in hyperextension until in the swing phase. In mid-swing, peak knee flexion was reached, 37°. Different curve was obtained for the extension and flexion of the intact knee. Right after initial contact and in early stance the knee was in flexion with peak knee flexion in early swing of 15°. During mid-stance the knee was in extension and terminal stance the joint moved into flexion again. The peak knee flexion occurred in mid-swing, about 55°.

Figure 4-40 represents the anatomical (A) and the prosthetic (B) knee angle curves for the tests made on the REHO knee.

(A)



(B)

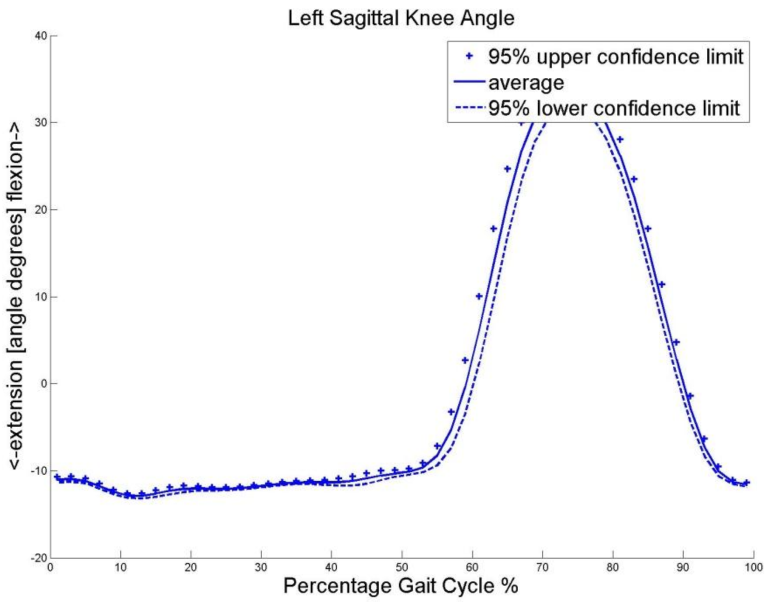


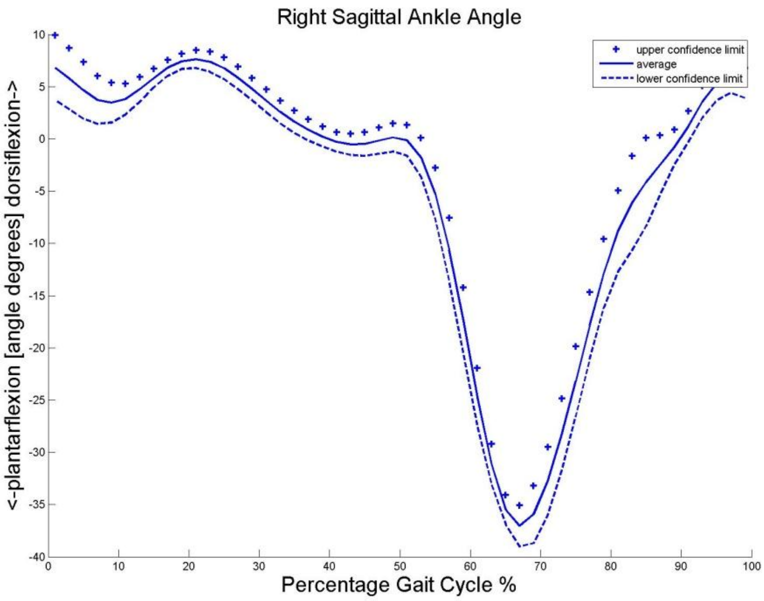
Figure 4-40 Intact limb (A) and prosthetic limb (B) knee angles during uphill walking on the RHEO knee

During the stance phase there was no prosthetic knee stance flexion and the joint was in extended position until the swing phase. In mid-swing peak knee flexion angle was reached, with trajectory of 45°. The trajectory of extension and flexion of the intact knee was different. At initial contact and early stance the knee was in flexion with peak knee flexion angle in early swing of 15°. During mid-stance the knee was in 10° extended position and at terminal stance the joint moved into flexion again. The peak knee flexion angle was in mid-swing of 62°.

4.5.3 Joint angles – Ankle – 3R80 & RHEO

Figure 4-41 represents the anatomical (A) and the prosthetic (B) ankle angle curves for the tests made on the 3R80 knee.

(A)



(B)

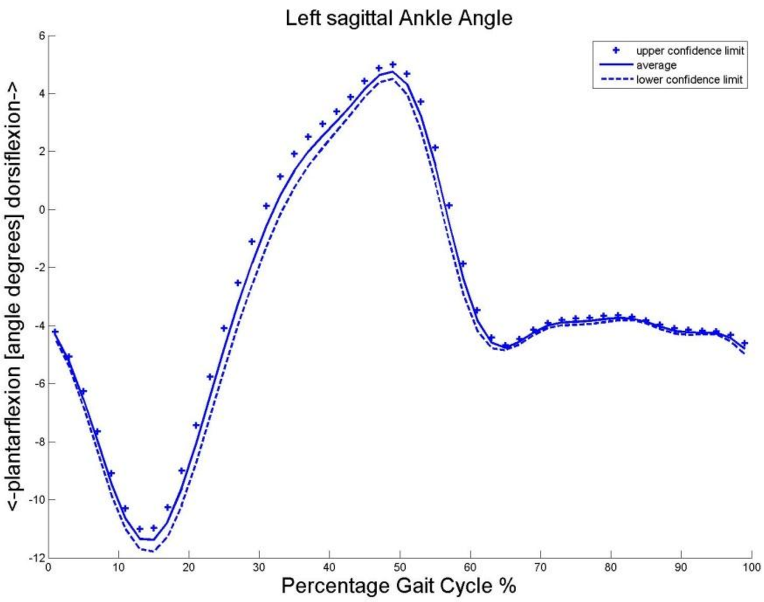


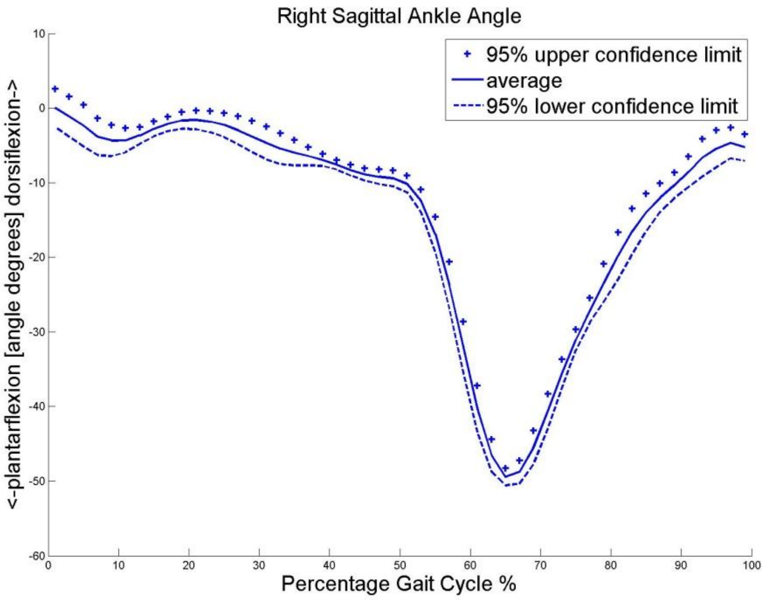
Figure 4-41 Intact limb (A) and prosthetic limb (B) ankle angles during uphill walking on the 3R80 knee

The anatomical ankle and the prosthetic ankle exhibited very different patterns of movement, with dissimilar plantar and dorsiflexion peaks occurring at different instances during the gait cycle. The angular motion of the anatomical ankle had a fluctuating pattern from initial contact to late stance, with small plantar and dorsiflexion peaks. At toe off the joint was in plantar flexion and peak plantar flexion occurs in swing with trajectory of 37°.

The angular motion of the prosthetic ankle was dissimilar to the motion of the anatomical joint, with major differences during terminal stance and in the swing phase. At initial contact and in early stance the prosthetic ankle exhibited plantar flexion, with peak value at early stance and trajectory of 11°. The prosthetic ankle movement switched to dorsiflexion and peak dorsiflexion occurred at toe off, with trajectory of 5°. At terminal stance the ankle was in plantar flexion, with significantly decreased peak value compared with the anatomical ankle, 5° and 37° respectively. During the swing phase, angular movement of the prosthetic ankle joint was negligible.

Figure 4-42 represents the anatomical (A) and the prosthetic (B) ankle angle curves for the tests made on the RHEO knee.

(A)



(B)

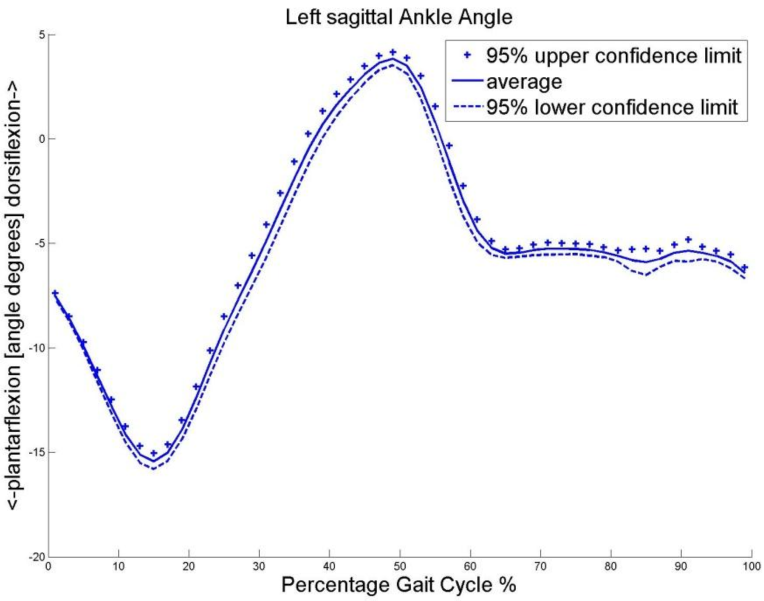


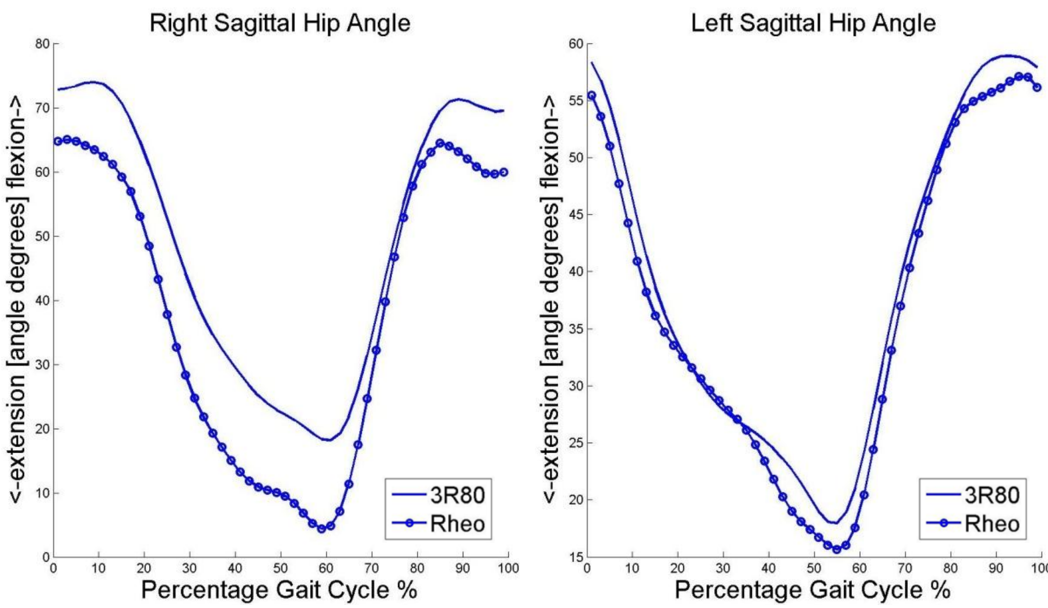
Figure 4-42 Intact limb (A) and prosthetic limb (B) ankle angles during uphill walking on the RHEO knee

The plantar and dorsiflexion motions of the anatomical and the prosthetic ankles were very different. The angular motion of the anatomical ankle from initial contact to late stance was mainly plantar flexion with trajectory of 48°. At toe off the joint was in plantar flexion and peak plantar flexion angle is in early swing with peak value of 40°.

The angular motion of the prosthetic ankle was dissimilar, especially during terminal stance and in the swing phase. At initial contact and in early stance the prosthetic ankle exhibited plantar flexion motion with trajectory of 11°. The prosthetic ankle movement switched to dorsiflexion and peak dorsiflexion occurred at toe off, with peak value of 8°. At terminal stance the prosthetic ankle was in plantar flexed position, with significantly decreased peak angle compared with the anatomical ankle, 6° and 48° respectively.

4.5.4 Joint angles Comparison – RHEO knee & 3R80

Figure 4-43 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip angles for the RHEO knee and the 3R80 knee.



A **B**

Figure 4-43 Intact limb (A) and prosthetic limb (B) hip angles during uphill walking on the RHEO knee and the 3R80 knee

The motion patterns obtained for the hip joints on the intact side when using the two types of prostheses were similar. Using both prostheses peak hip extension occurs at similar instant, when using the RHEO knee the peak occurs slightly earlier, about 60% of the gait cycle and the extension trajectory is slightly larger than for the 3R80, 60° and 55° respectively. In early swing the hip joint flexes more rapidly when using the RHEO knee. For the hip joint on the prosthetic side the motion patterns are identical.

Figure 4-44 represents the comparison on the anatomical (A) and the prosthetic (B) knee angles for the tests made on the 3R80 and the RHEO knee.

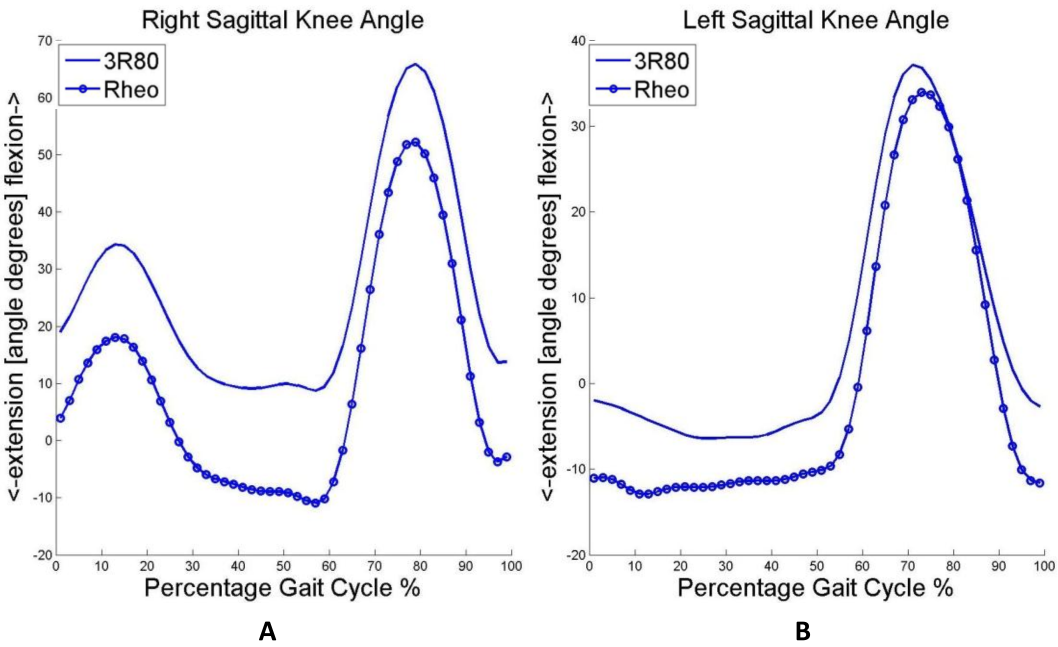


Figure 4-44 Intact limb (A) and prosthetic limb (B) knee angles during uphill walking on the RHEO knee and the 3R80 knee

The curves representing the motion of the intact knee when using the 3R80 and the RHEO knee joints are very similar, with equal flexion and extension trajectories throughout the gait cycle. The curves for the angular motion of the prosthetic knee joints are also very similar. During the stance phase neither of the joints flexed and the flexion and extension curves from terminal stance to terminal swing are similar. The peak knee flexion angle in mid swing was slightly larger for the RHEO knee compared with the 3R80, with trajectories of 45° and 37° respectively.

Figure 4-45 represents the comparison on the anatomical (A) and the prosthetic (B) ankle angles for the tests made on the 3R80 and the RHEO knee.

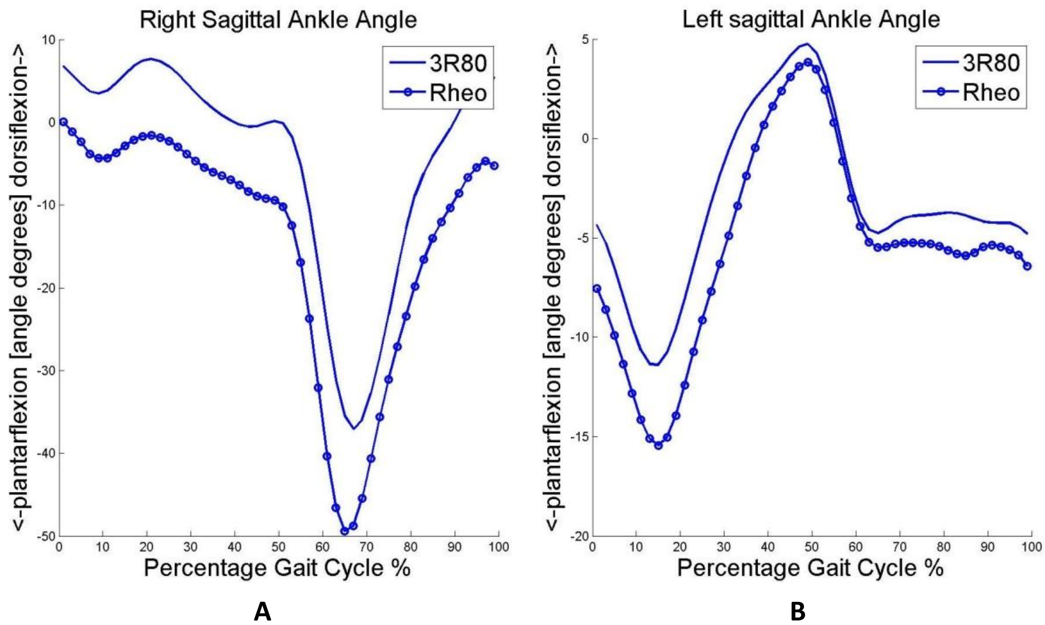


Figure 4-45 Intact limb (A) and prosthetic limb (B) ankle angles during uphill walking on the RHEO knee and the 3R80 knee

The angular motions of the anatomical and the prosthetic ankled were similar when using the different types of prostheses, with plantar flexion and dorsiflexion trajectories that were comparable.

Table 10 and 11 represent the hip, knee and ankle motion trajectories for the prosthetic (10) and the intact (11) sides that were considered relevant for the evaluation of the knees.

Table 10 Hip, knee and ankle extension and flexion motions of the prosthetic limb during uphill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**40**	**43**
Hip flexion trajectory in swing (degrees)	**43**	**45**
Knee flexion in stance (degrees)	**0**	**0**
Knee flexion trajectory in swing (degrees)	**45**	**37**
Ankle plantar flexion trajectory in terminal stance (degrees)	**0**	**0**

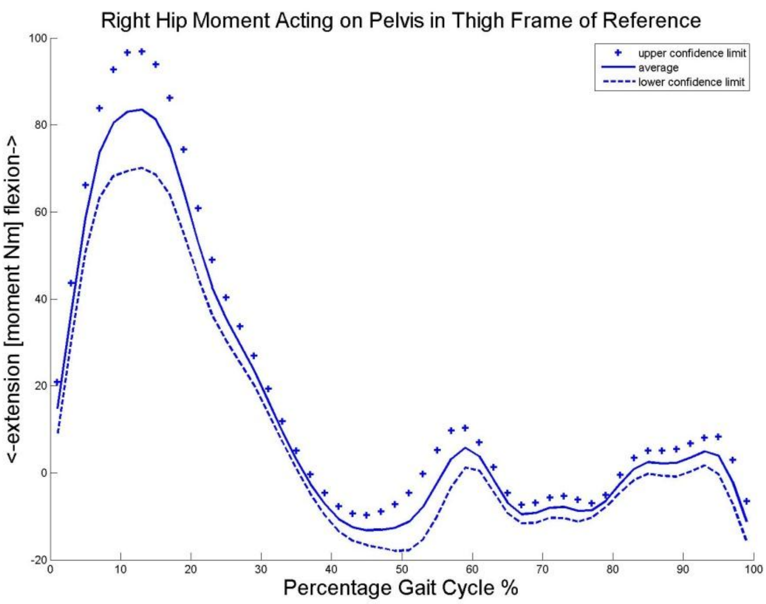
Table 11 Hip, knee and ankle extension and flexion motions of the intact limb during uphill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension trajectory in stance (degrees)	**60**	**55**
Hip flexion trajectory in swing (degrees)	**60**	**54**
Knee flexion in stance (degrees)	**15**	**15**
Knee flexion trajectory in swing (degrees)	**62**	**55**
Ankle plantar flexion trajectory late stance to early swing (degrees)	**40**	**37**

4.5.5 Joint moments - Hip

Figure 4-46 represents the anatomical (A) and the prosthetic (B) hip moment curves for the tests made on the 3R80 knee.

(A)



(B)

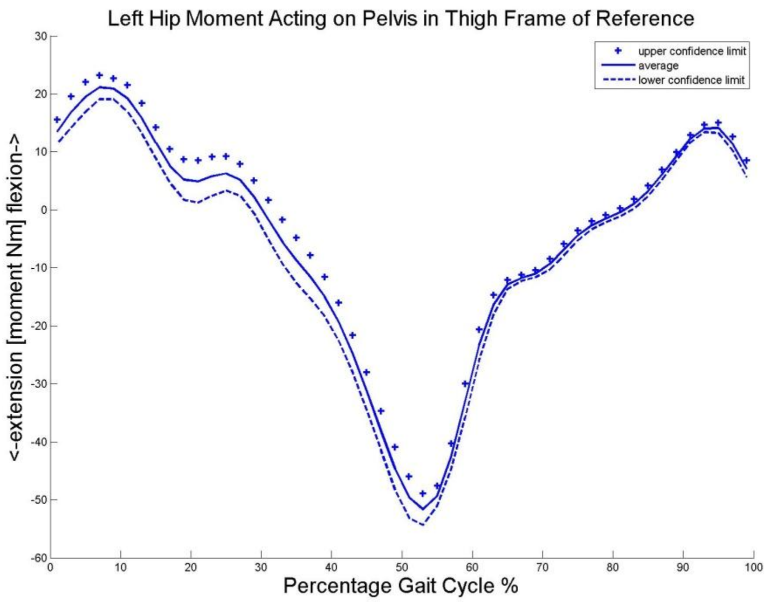


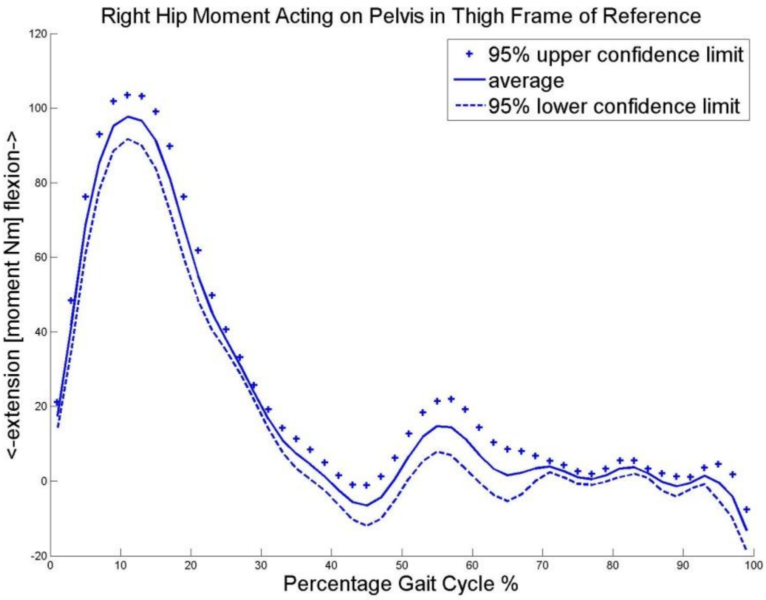
Figure 4-46 Intact limb (A) and prosthetic limb (B) hip moments during uphill walking on the 3R80 knee

The moment patterns obtained for the hip joints during uphill walking were very different for the two limbs. At initial contact and early stance there was a flexion moment about both hip joints, where peak flexion moment occurred at early stance, with trajectories of 10 Nm for the prosthetic limb and 75 Nm for the intact limb. There was an extension moment about the joint on the intact side until late stance, with trajectory of 95 Nm. In late stance the moment changed to extension moment and peaked at terminal stance with trajectory of 15 Nm. In the swing phase the hip moment had a fluctuating pattern with small flexion and extension peaks.

There was also a flexion moment about the hip on the prosthetic side at initial contact, with trajectory of 10 Nm. In early stance the moment switched to extension moment and there was an extension moment about the joint until terminal stance, with trajectory of 70 Nm. At terminal stance and throughout the swing phase the hip moment was flexion, with trajectory of 62 Nm and in terminal swing the moment fell down to zero.

Figure 4-47 represents the anatomical (A) and the prosthetic (B) hip moment curves for the tests made on the RHEO knee.

(A)



(B)

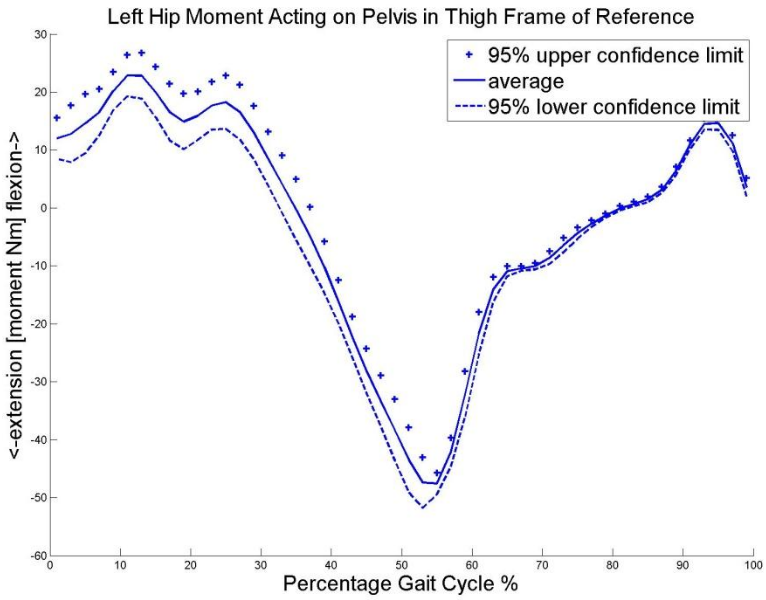


Figure 4-47 Intact limb (A) and prosthetic limb (B) hip moments during uphill walking on the RHEO knee

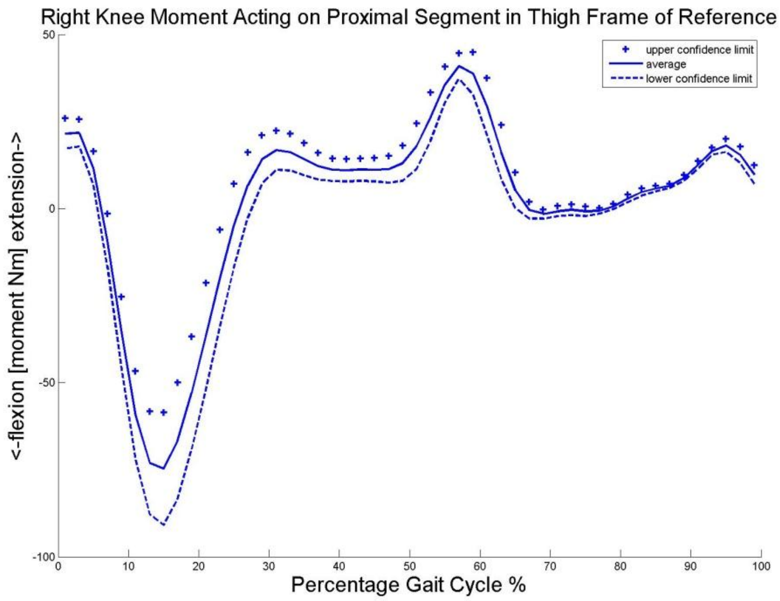
The moment patterns obtained for the hip joints during uphill walking was very different for the two limbs. At initial contact and early stance there was a flexion moment about both hip joints, where peak flexion moment occurred at early stance, with trajectories of 80 Nm for the intact limb and 11 Nm for the prosthetic limb. At mid-stance the moment about the hip on the intact side changed to extension moment and peaked at push off, with trajectory of 105 Nm. In the swing phase the hip moment had a fluctuating pattern with small flexion and extension peaks.

There was also a flexion moment about the hip joint on the prosthetic side at initial contact. The moment switched to extension moment at about 30% of the gait cycle, which peaked in late stance, at toe-off, with trajectory of 65 Nm. At terminal stance and throughout the swing phase the hip moment was flexion, with peak flexion moment value of 15 Nm.

4.5.6 Joint moments – Knee

Figure 4-48 represents the anatomical (A) and the prosthetic (B) knee moment curves for the tests made on the 3R80 knee.

(A)



(B)

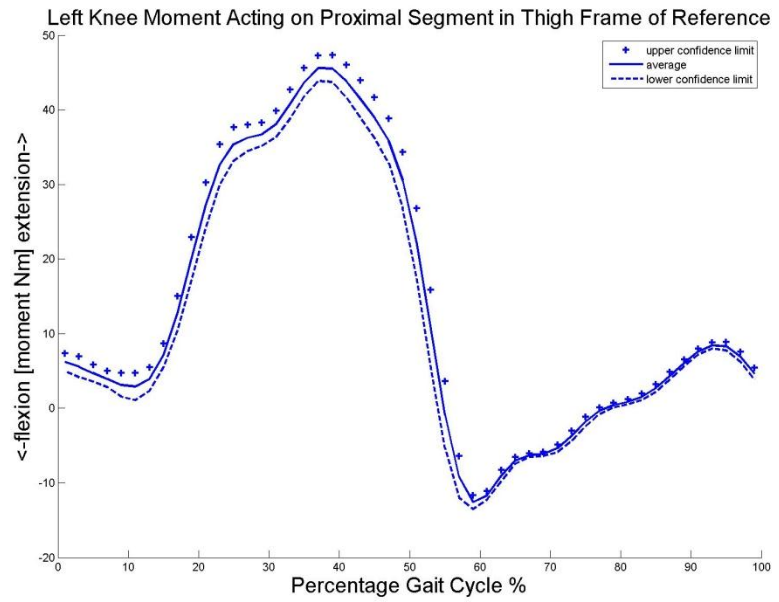


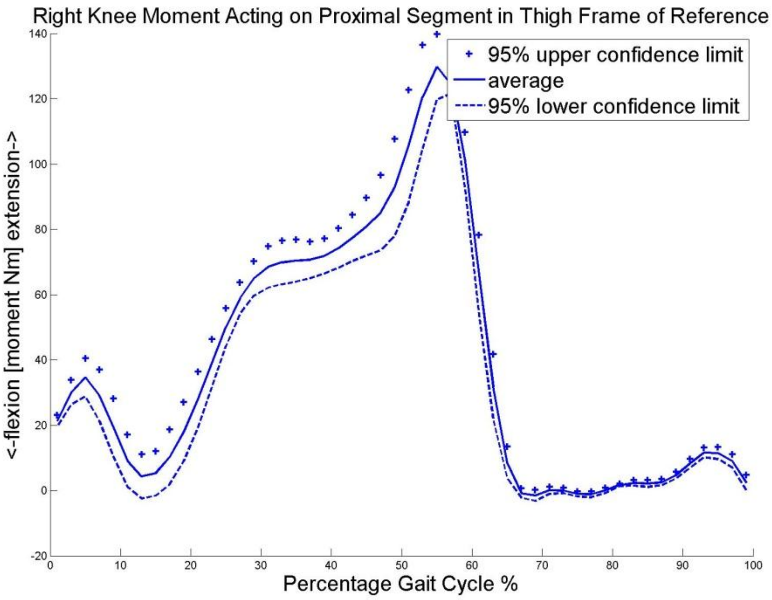
Figure 4-48 Intact limb (A) and prosthetic limb (B) knee moments during uphill walking on the 3R80 knee

The knee joint moment curves obtained for the joints were significantly different, especially during early and terminal stance. The intact knee experienced flexion moment in early stance, with trajectory of 95 Nm. After that the moment switched to extension moment that peaked at mid stance, with trajectory of 90 Nm. At toe-off there was an extension moment about the knee, with peak value at terminal stance. During early swing the intact knee experienced low magnitude extension moment.

The moment about the prosthetic knee was significantly different. There was an absence of prosthetic knee flexion during the stance phase, thus the knee experienced extension moment throughout the mid stance with trajectory of 40 Nm. In late stance the moment changed to flexion moment, with peak at terminal stance and trajectory of 55 Nm. During the swing phase the extension moment curve had a fluctuating pattern that lasted throughout the swing phase.

Figure 4-49 represents the anatomical (A) and the prosthetic (B) knee moment curves for the tests made on the REHO knee.

(A)



(B)

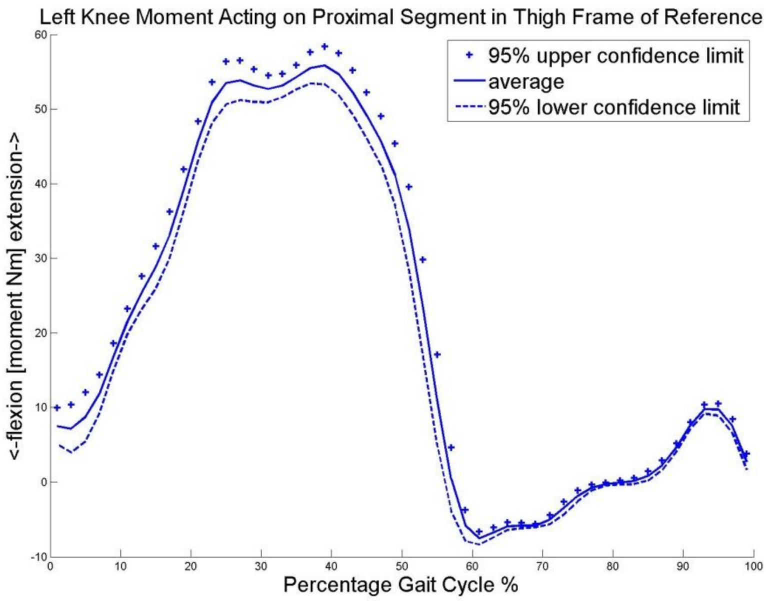


Figure 4-49 Intact limb (A) and prosthetic limb (B) knee moments during uphill walking on the RHEO knee

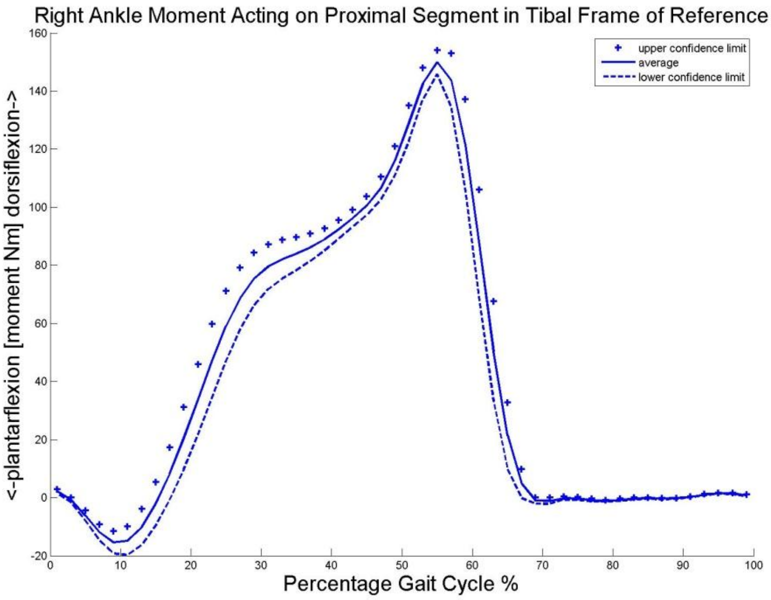
The moment curves obtained for the knee joints were significantly different, especially during early and terminal stance. In early stance there was a flexion moment about the intact knee, with trajectory of 35 Nm. The moment then switched to extension moment that peaked at toe off, with trajectory of 130 Nm. During early swing there was a flexion moment about the knee, the moment fell down to zero during mid-swing and at terminal swing there was a small extension moment peak.

The moment experienced by the prosthetic knee was considerably different, especially during the stance phase. Early to mid-stance there was an extension moment about the knee, with trajectory of 50 Nm. In late stance the moment changed to flexion moment, with peak at terminal stance and trajectory of 63 Nm. During the swing phase the extension moment curve had a fluctuating pattern with trajectory of 19 Nm.

4.5.7 Joint moments – Ankle

Figure 4-50 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the 3R80 knee.

(A)



(B)

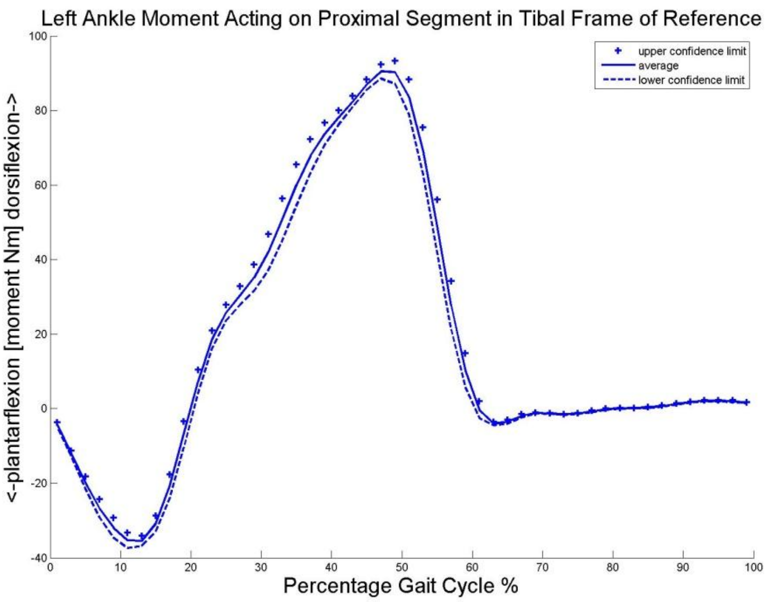
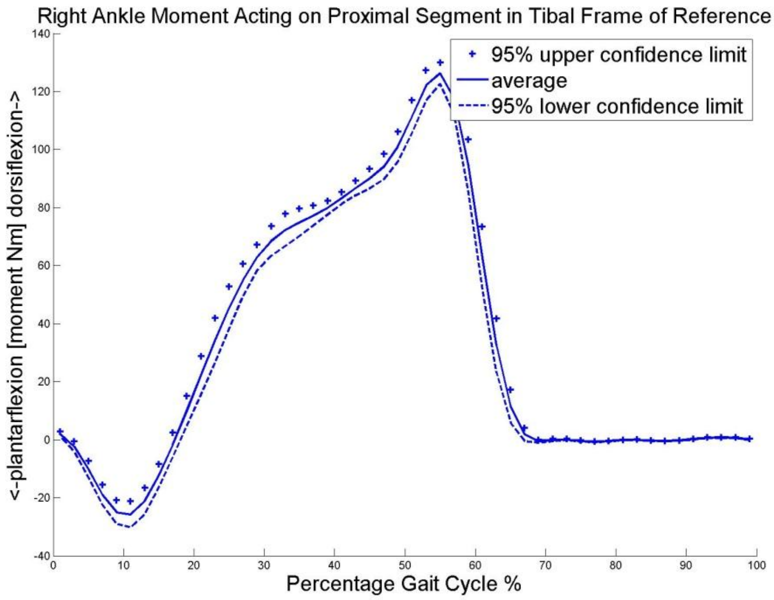


Figure 4-50 Intact limb (A) and prosthetic limb (B) ankle moments during uphill walking on the 3R80 knee

The ankle joint moments obtained had similar curves for the anatomical and the prosthetic joints, where both joints experienced plantar flexion moment at initial contact and early stance. The peak plantar flexion moment during early stance was significantly greater for the prosthetic ankle, with trajectory of 35 Nm compared with 19 Nm for the anatomical ankle. After the plantar flexion peak the joint moment switched to dorsiflexion moment for both joints with slightly different curves. Both ankles experienced peak dorsiflexion moment around push-off, with different peak values for the anatomical and the prosthetic joints, the trajectories were 150 Nm and 125 Nm respectively. At terminal stance both joints experienced plantar flexion moment.

Figure 4-51 represents the anatomical (A) and the prosthetic (B) ankle moment curves for the tests made on the RHEO knee.

(A)



(B)

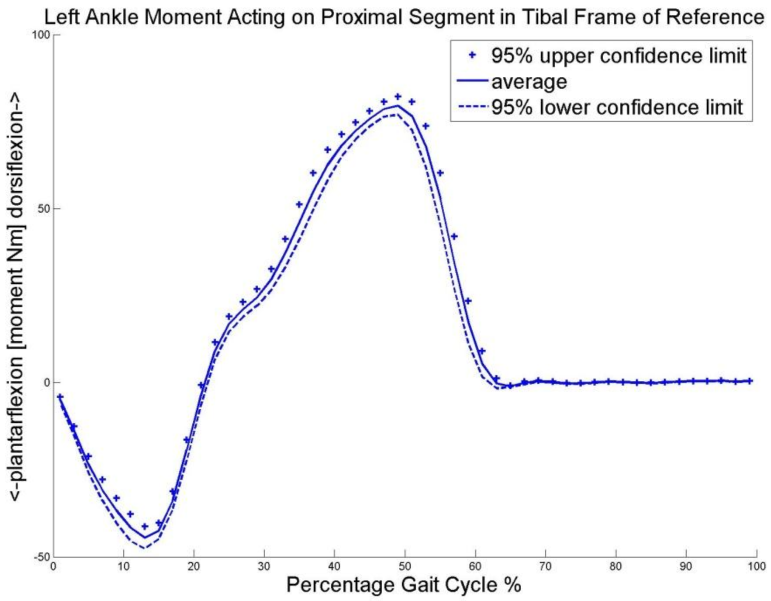


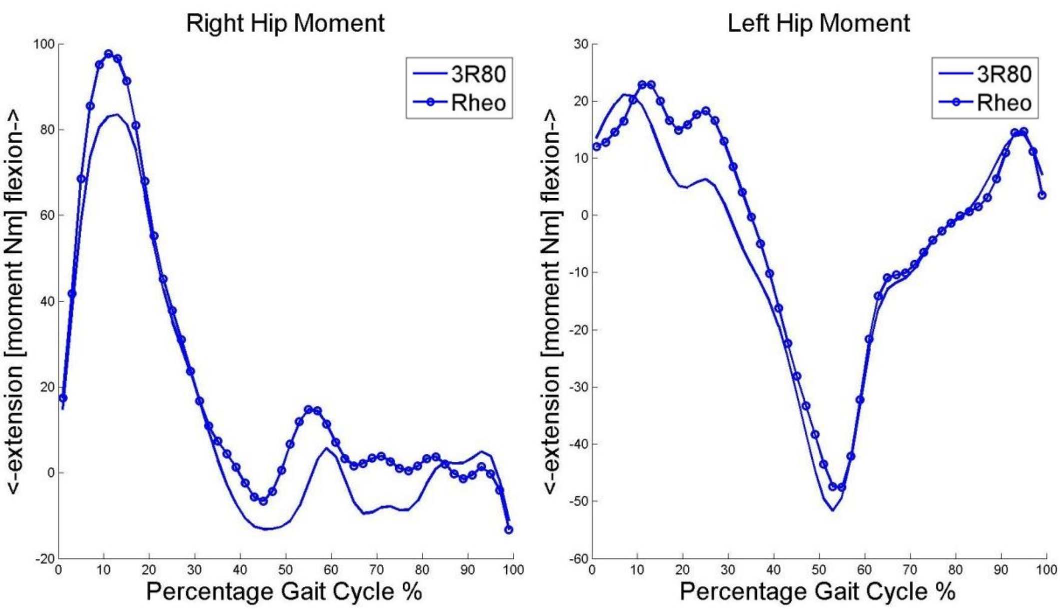
Figure 4-51 Intact limb (A) and prosthetic limb (B) ankle moments during uphill walking on the RHEO knee

The ankle joint moments had similar patterns for the anatomical and the prosthetic joints, however, with different peak values. Both joints experienced plantar flexion moment at initial contact and early stance. The peak plantar flexion moment in early stance was significantly greater for the prosthetic ankle, with trajectory of 40 Nm compared with 25

Nm for the anatomical ankle. After the plantar flexion peak the joint moment switched to dorsiflexion moment for both joints with slightly different curve. Both ankles experienced peak dorsiflexion moment around push-off, with different trajectories for the anatomical and the prosthetic joints, 150 Nm and 120 Nm respectively. At terminal stance both joints experienced plantar flexion moment that gradually fell down to zero during the swing phase.

4.5.8 Joint moments – Comparison - RHEO knee & 3R80

Figure 4-52 represents comparison of the intact limb (A) and the prosthetic limb (B) sagittal hip moments for the RHEO knee and the 3R80 knee.



A B

Figure 4-52 Intact limb (A) and prosthetic limb (B) hip moments during uphill walking on the RHEO knee and the 3R80 knee

The moments about the hip on the intact side were similar when using the RHEO knee and the 3R80. The Flexion moment in early stance was of slightly greater magnitude when using the RHEO knee. In late stance there was a sharper transition from extension moment to flexion moment associated with the REHO knee. The moments about the hip joint on the prosthetic side were very similar throughout the gait cycle when using the 3R80 and the RHEO knee.

Figure 4-53 represents the comparison on the anatomical (A) and the prosthetic (B) knee moments for the tests made on the 3R80 and the RHEO knee.

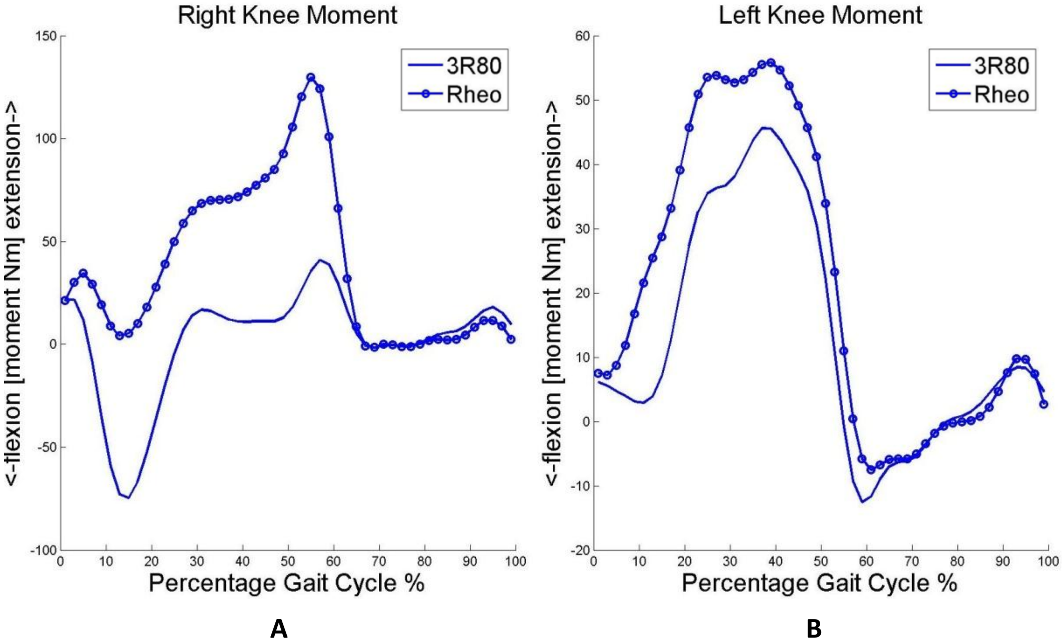


Figure 4-53 Intact limb (A) and prosthetic limb (B) knee moments during uphill walking on the RHEO knee and the 3R80 knee

The prosthetic knee moments were slightly different during the stance phase. The RHEO knee was associated with greater extension moment from early to mid-stance, 50 Nm compared with 40 Nm for the 3R80. The following extension moment, from mid to terminal stance was also greater for the RHEO knee, with trajectory of 63 Nm compared with 55 Nm for the 3R80 knee. During the swing phase the moments about the knees were equal.

For the intact knee the magnitude of the flexion and extension moments were considerably different for the knees. The flexion moment in early stance was greater for the 3R80, with trajectory of 95 Nm compared with 35 Nm for the RHEO knee. The following extension moment had different trajectories for the knees, the 3R80 was associated with a sharp transition from flexion to extension then the moment increased further with another flexion peak at terminal stance. The total extension trajectory from early to terminal stance was 115 Nm. The RHEO knee was associated with more incremental extension moment from early to terminal stance, with trajectory of 125 Nm

Figure 4-54 represents the comparison on the anatomical (A) and the prosthetic (B) ankle moments for the tests made on the 3R80 and the RHEO knee.

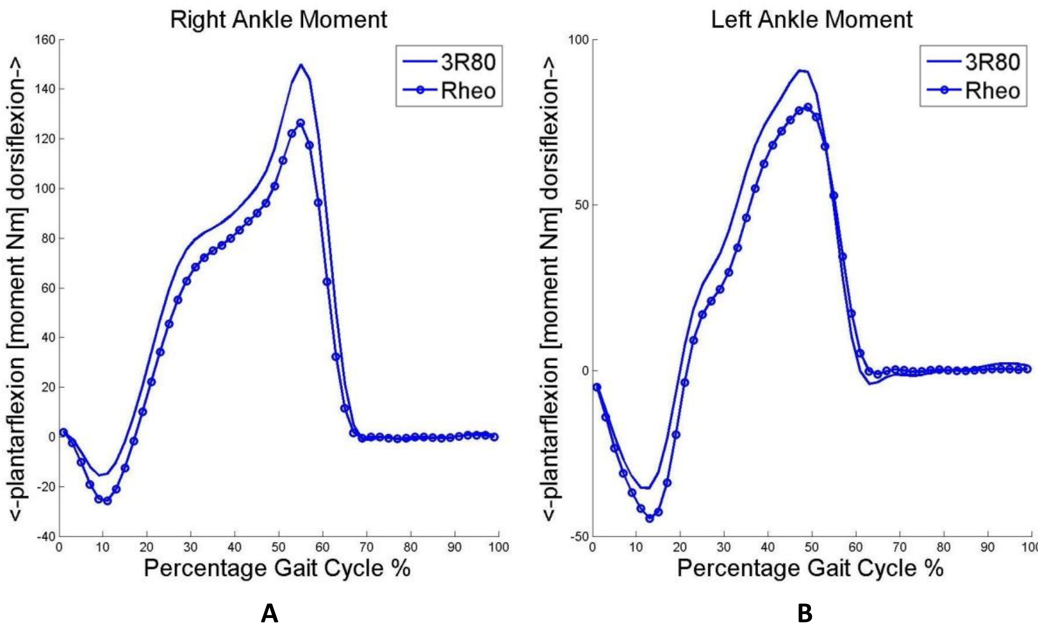


Figure 4-54 Intact limb (A) and prosthetic limb (B) ankle moments during uphill walking on the RHEO knee and the 3R80 knee

The ankle joint moments about the anatomical and the prosthetic ankles were very similar when using the different knees. As seen on figure the curves are almost adjacent throughout the gait cycle.

Table 12 and 13 represent the hip and knee moment trajectories for the prosthetic (12) and the intact (13) sides that were considered relevant for the evaluation of the knees.

Table 12 Hip and knee extension and flexion moments of the prosthetic limb during uphill walking

Parameter	RHEO Knee	3R80 Knee
Hip flexion moment trajectory at initial contact (Nm)	**11**	**10**
Hip extension moment trajectory from early to late stance (Nm)	**65**	**70**
Knee flexion moment trajectory in early stance (Nm)	**50**	**40**

Table 13 Hip and knee extension and flexion moments of the intact limb during uphill walking

Parameter	RHEO Knee	3R80 Knee
Hip extension moment trajectory at push off (Nm)	**105**	**95**
Knee extension moment trajectory in early stance (Nm)	**130**	**90**

4.6 Discussion for uphill walking

4.6.1 RHEO knee and 3R80

During uphill walking, where the foot has to be placed on the inclined terrain, increased hip and knee joint flexion and ankle joint dorsiflexion during the swing phase and at initial contact were expected. The 3R80 and the RHEO knee were associated with decreased range of motion, flexion and extension, of the hip joint on the prosthetic side compared with the intact limb. These results are according to reports by Vrieling et al (2008) (chapter 2.8). The decreased range of motion of the prosthetic limb is likely explained by smaller step length with the limb, as considerably more force is needed at push-off during uphill walking and with the lack of power generation from the prosthetic ankle the step length likely decreased.

The intact knee angular motion curves obtained when using the 3R80 and the RHEO knee showed a very different pattern compared with the prosthetic knee motions. Results showed an absence of prosthetic knee flexion throughout the stance and the prosthetic knees were both hyper extended. The absence of the knee flexion of the prosthetic knee will likely cause other compensation movements in the intact limb. Vrieling et al (2008) reported compensation movements such as increased flexion of the anatomical knee and increased plantar flexion of the anatomical ankle at toe-off (chapter 2.8). The 3R80 and the RHEO knee were found to be associated with considerably greater knee flexion of the intact knee and plantar flexion of the anatomical ankle. The author of the current study agrees with Vrieling et al (2008) suggestions, that these compensatory movements could help in loading of the prosthetic limb, as it is more difficult to transfer body weight on the prosthetic side due to the increased height while walking uphill (Vrieling, 2008) (chapter 2.8).

Both prosthetic knees were associated with significantly greater flexion moment about the hip joint of the intact limb during stance phase. During the swing phase it was reversed, the prosthetic limb experienced increased hip flexion moment when using both knees. These results indicate that in stance phase the subject was compensating for the prosthetic limb with the intact limb and in the swing phase exaggerated hip flexion was needed to initiate knee extension to be able to place the prosthetic foot on the inclining surface and with the knee in extended position. Similarly as for the downhill and level walking the subject was unable to flex the prosthetic knee joints during the stance phase. These results are according to reports by Vrieling et al (2008) (chapter 2.8). The curves obtained for the prosthetic and the anatomical ankle joints during uphill walking, using the 3R80 and the RHEO knee, were similar.

4.6.2 Discussion about the comparison of both knees for uphill walking

The hip motions on the prosthetic and the intact sides were close to being the same when using the 3R80 and the RHEO knee, hence the results indicate that the different techniques did not affect the angular motion of the hips greatly.

The knee motions for both the intact and the prosthetic knees were as well similar. Neither of the prosthetic joints flexed during the weight bearing phase, throughout the stance phase both prosthetic knees were locked in extended positions. The RHEO knee was associated with slightly greater peak flexion angle in the swing phase. That could indicate that the subject was able to flex the RHEO knee more naturally. The motions of the anatomical and the prosthetic ankles were not influenced to any extent with the use of different knee devices.

The results for the angular motions of the hip, knee and ankle joints during uphill walking indicate that despite the microprocessor technology utilised by the RHEO knee that did not influence the gait pattern majorly. The subject was able to demonstrate similar motion when using the 3R80 knee although the joint does not have any adjustment options for uphill walking and is not able to respond in the same way to changes in terrain as the RHEO knee.

The results for the hip moments did not show any major differences in either of the joints when using the 3R80 or the RHEO knee. For the hip joint on the intact side the RHEO knee was associated with higher flexion moment in early stance. That could indicate that at initial contact there was more effort needed to stabilise the RHEO knee to secure that the knee was in fully extended position when prosthetic foot was placed on the inclining surface.

The prosthetic knee moments were slightly different during the uphill walking. The RHEO knee was associated with greater extension moment from early to late stance. That indicates that when the subject was walking up the ramp he was loading the RHEO knee more naturally as he transferred his weight over the limb when placing the prosthetic foot on the inclining ramp. The greater flexion moment about the intact knee from initial contact to early stance associated with the 3R80 is likely due to the uneven weight bearing of the limbs when using the knee. By compensating for the decreased weight bearing on the prosthetic limb more load is on the intact knee. The results for the knee moments indicate that during uphill walking the subject was able to load the RHEO knee more naturally. The moments about the anatomical and the prosthetic ankles were identical for the 3R80 and the RHEO knee.

5 Future works

The data from the study of walking up stair was considered not usable due to inadequate test environment, a further investigation of the performance of the different prosthetic knees technology is needed, where sufficient number of steps is provided.

Further investigation is needed where higher number of subjects would participate and the subjects provided with more time to adjust to both knee types, especially to gain more confidence using the RHEO knee, also for the knee to gather information about the subject´s gait.

6 Conclusions

This study is focused on evaluation of the performance of the RHEO knee. In the study, the RHEO knee is tested and compared to another type of prosthetics knee, 3R80. The study is aimed at providing independent evaluation of the performance of the RHEO knee that can be used to estimate whether it is more feasible option to use rather than other simpler and cheaper prosthetics knees. This is performed by testing the knees in different environments, using state-of-the-art technology and by analysing relevant parameters.

Even though the results from stair walking were not considered usable, the study can be considered successful since all objectives were accomplished. The study presents results for level, uphill and downhill walking.

The results from this study show that for level walking there was a difference in joint angles across the knees, whereas the RHEO knee is associated with more compensatory movements in hip motion of the intact limb. In swing phase the RHEO knee was associated with closer to normal knee joint flexion angle. The variance in hip moments was greater when using the 3R80, thus, it can be assumed that the RHEO supports more symmetric gait with more equally weight bearing of the limbs.

From the results for the downhill walking it is difficult to justify better performance of either of the knees. However, the 3R80 knee shows more promising results regarding symmetric weight bearing of the limbs. Walking down the ramp requires greater skills and more confidence than walking on level ground. The fact that the subject was reasonable unused to the RHEO knee could possibly have had impact on his performance. When using the RHEO knee the intact limb was compensating for the prosthetic limb, with increased weight bearing.

The results for the uphill walking indicate that the RHEO knee was associated with more natural gait. The subject was able to demonstrate closer to normal knee flexion during the swing phase and as walking up the slope, use and load the prosthetic limb more naturally.

Although the results did not show clear advantage and better performance of the RHEO knee throughout the study, the feedback from the subject was very positive. The subject is used to participating in studies involving different types of knee prostheses. His opinion was that the RHEO knee was among the best prosthesis he has used; his favourite feature was the smoothness of the knee joint throughout the gait cycle.

7 References

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