Gait analysis of Transfemoral amputees

Effects of an adaptive microprocessor-controlled prosthetic foot and the effects of individualized training

Anna Lára Ármannsdóttir

Ritgerð til meistaragráðu

Háskóli Íslands
Læknadeild
Rannsóknarstofa í hreyfivísindum
Heilbrigðisvísindasvið

HÁSKÓLI ÍSLANDS

Göngugreining einstaklinga sem aflimaðir eru fyrir ofan hné

Samanburður á tveimur mismunandi stillingum á tölvustýrðum gervifæti og áhrif einstaklingsmiðaðrar þjálfunar

Anna Lára Ármannsdóttir

Ritgerð til meistaragráðu í hreyfivísidum

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Ágrip

Í göngu reiðir heilbrigður einstaklingur sig á samhæfða vöðvavinnu, eðlilegan liðferil og flókið samspil ólíkra lífeðlisfræðilegra þátta. Hjá einstaklingum sem misst hafa útlim fyrir ofan hné eru margir þessara þátta ekki til staðar eða á einhvern hátt skertir og endurspeglar göngulag þeirra þessa skerðingu, sem og þær takmarkanir sem hljótast af þeim stoðtækjum sem þeir nota. Breytingar í göngulagi geta svo haft neikvæð áhrif á líkamann í heild, og eru vandamál eins og verkir í baki og aukið liðslit stórt vandamál hjá einstaklingum sem misst hafa neðri útlim. Þar sem þættir eins og endurhæfing og tegund stoðtækja geta haft mikil áhrif á gæði göngu, var markmið þessarar rannsóknar að meta áhrif ákveðinnar aðlögunar í tölvustýrðum gervifæti, sem og áhrif einstaklingsmiðaðrar þjálfunar á göngu. Niðurstöður eru settar fram í tveimur hlutum I) þar sem meðaltal niðurstaðna hjá fimm einstaklingum er skoðað með hliðsjón af gögnum samanburðarhóps og II) í tveimur tilfellarannsóknum, þar sem niðurstöður þeirra einstaklinga sem fengu þjálfun er skoðaðir sérstaklega.

Aðferðir: Fjórir heilbrigðir og fimm einstaklingar sem misst hafa neðri útlim fyrir ofan hné tóku þátt í rannsókninni, og af þeim fengu tveir einstaklingar þjálfunarprógramm, sem var ákvörðun sem var byggð á niðurstöðum skoðunar sem framkvæmd var við byrjun mælinga. Allir þátttakendur notuðu sömu stoðtækin við rannsóknina, þ.e. SYMBIONIC LEG, sem samanstendur af tölvustýrðum gerviökkla og gervihné. Þegar kveikt er á þeirri stillingu sem stýrir aðlögun á gervifæti, á sér stað virk beygja í ökkla í sveiflufasa göngu, sem og aðlögun að mismunandi halla á undirlagi. Hreyfingar um liðamót við göngu á jafnsléttu og í halla voru teknar upp með þrívíddar-myndatökubúnaði með annars vegar kveikt og hins vegar slökkt á aðlögun í gervifæti. Hreyfiferlar um ökkla, hné, mjöðm og mjaðmargrind voru metnir, og til að meta virkni *m. Gluteus medius og m. Tensor fascia latae* var notast við vöðvarafrit hjá þeim einstaklingum sem greint var frá í tilfellarannsóknunum tveimur, en hjá þeim einstaklingum verða niðurstöður tveggja mælinga kynntar, þ.e. fyrir og eftir sex vikna þjálfun.

Niðurstöður og umræður: Marktækur munur var á ákveðnum þáttum í hreyfimynstri heilbrigðra einstaklinga samanborið við hreyfingar um liðamót beggja hliða hjá aflimuðum einstaklingum og var sá munur, og aðrar breytingar í hreyfimynstri, ræddar sem og mögulegar ástæður fyrir þeim. Ekki fannst marktækur munur á hreyfingum öðrum en beygju í gerviökkla þegar bornar voru saman stillingar á gervifæti sem stýra aðlöguninni. Í tilfellarannsóknunum tveimur var áhersla lögð á að skoða hreyfingar í mjöðm og mjaðmagrind en mikill breytileiki var í hreyfimynstri á milli einstaklinganna. Hjá þátttakanda 2 mátti sjá bætt hreyfimynstur í öllum mældum atriðum, bæði hvað varðaði stillingu á gervifæti og þjálfunina, sem endurspeglaðist í minnkuðum mun á milli hliða. Hjá þátttakanda 3 var einungis hægt að sjá slíka bætingu í mjaðmagrind, aflimuðu megin í sveiflufasa göngu.

Ályktanir: Bæði stillingar á gervifæti sem og þjálfun getur bætt hreyfimynstur í mjöðm og mjaðmargrind, sem endurspeglast í minnkuðum mun á milli hliða. Hins vegar þarf að hafa í huga að þegar ganga aflimaðra er skoðuð að mikill breytileiki er greinilegur þegar hver aflimaður einstaklingur er skoðaður sérstaklega, með tilliti til hreyfinga mjaðma og mjaðmargrindar, og því þarf að hafa margvíslega þætti í huga við val á stoðtækjum og æfingarprógrammi og eins þegar niðurstöður vísindarannsókna eru túlkaðar.

Abstract

An able-bodied individual relies on a well-coordinated muscle activity, intact joint structures and complex interactions of numerous physiological factors during gait in order to successfully ambulate. For transfemoral amputees these elements are often impaired or absent, and their gait pattern is reflected by those limitations, as well as by necessary restrictions provided by their prosthesis. In order to compensate for those restrictions a change in movement pattern occurs, which can negatively affect the musculoskeletal system and of the main concerns for lower limb amutees are back problems and degenerative changes in intact joints. Factors such as rehabilitation and type of prosthesis can have a great impact on the quality of gait, and hence the aim of this study was to evaluate both the effect of an adaptive mode of a prosthetic foot and the effect of individualized training on gait. Results will be presented in two sections in the thesis I) with the averaged results from five participants compared to a control group, and II) in two separate case studies for individuals receiving training.

Method: Four able bodied and five transfemoral participants were recruited for the study, two of whom received a individualized training program, based on an examination performed at the start of measurements. All participants wore the same prosthesis, the SYMBIONIC LEG which is a combination of a powered microprocessor foot and a microprocessor knee joint, which in its active mode produces ankle dorsiflexion during swing phase and adapts to inclined and declined surfaces. An eight camera motion capture system was used to capture 3D movements during level and incline walking, with the prosthetic foot in both inactive and active adaptive mode. Excursions of the ankle, knee, hip and pelvis were evalutated, and as a secondary analysis for the case studies, muscle activation was assessed, using electromyography for *m. Gluteus medius* and *m. Tensor fasia latae.* For the two participants receiving training, data from two measurements will be presented, that is, before and after the six week training period.

Results and discussion: When compared to able-bodied individuals, significant differences were observed in a number of kinematic measures between controls and both sides of the amputees. Those changes, and other deviations, were discussed and possible causes proposed. No significant differences were observed in the kinematic parameters other than the prosthetic ankle dorsiflexion when comparing the active adaptive mode to the inactive adaptive mode. For the case studies, where the focus was set on the hip and pelvis kinematics, there was considerable variability in the movement pattern of the two participants. There were indications of improvements that may have resulted from the adaptive mode and the training, as reflected by greater interlimb symmetry for all parameters evaluated for one of two participants receiving training. For the other participant only pelvic movement in the swing phase of gait indicated an increase in interlimb symmetry.

Conclusion: For a transfemoral amputee both an adaptive mode of a prosthetic foot and individualized training may improve gait as reflected by greater interlimb symmetry. However, the vast variability in the hip and pelvis kinematic pattern is apparent when each transfemoral amputee is analyzed specifically. Therefore a number of factors must be taken into account when gait is analyzed clinically, when selecting an appropriate prosthesis or designing training programs, and when interpreting research results.

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Table of Contents

List of figures

List of tables

List of abbreviations

LLA	Lower limb amputee
TFA	Transfemoral amputee
TTA	Transtibial amputee
Amp	Amputated side for the amputee
Ctrl	Control group
Rctrl	Right side for control group
Lctrl	Left side for control group
IC	Initial Contact
SW	Swing phase of the gait cycle
WA	Weight acceptance
OA	Osteoarthritis
BMI	Body Mass Index
EMG	Electromyography
MMT	Manual muscle testing
MVIC	Maximal Voluntary Isometric contraction
ROM	Range of motion

1. Introduction

When learning to walk again after amputation the gait alters notably in many ways, which can have a negative effect on other healthy joints of the body. In this thesis, the gait of transfemoral amputees (TFA) will be analyzed and contrasted with that of able-bodied controls, and interlimb differences in amputees identified. The main focus will be on kinematics of the pelvic girdle and the hip during level and incline walking, while looking at the effect of two different settings of a powered microprocessor-controlled prosthetic foot. In addition the effect of a six week specialized gait and strengthening training will be examined, where the main goal was to help the user to utilize the special functions of the prosthetic foot, along with general strengthening goals, in order to improve the gait.

Knowing and understanding the functions and movements of the remaining musculoskeletal system in a common act such as walking is essential for professionals working with TFA, whether in a rehabilitation setting or in the process of designing prostheses. Great advancements have been made over the last 20 years in the design of prostheses, with the introduction of bionic technology with artificial intelligence being the newest addition in the field of prosthetic feet and knees. Kinetics and kinematics during gait with the particular prosthetic foot used in this study has been documented among transtibial amputees (TTA) (1, 2). However, the effects of using a microprocessor-controlled prosthetic foot for an individual with a prosthetic knee are not well known. Identifying adapted motor strategies among TFA during gait with different settings of the prosthetic foot, before and after training, could help professionals get better insight into the design of the prosthetic foot as well as the relevant factors of training protocols for TFA gait.

1.1 Lower limb amputation

The most common causes for lower limb amputations are vascular diseases, including diabetes and peripheral arterial disease (≈78% of total lower limb loss in the USA in the year 2005) and trauma (≈20%) (3). Other causes are cancer and infections (3, 4). No epidemiological data exist for limb loss in Iceland, but on average there are 10-20 major amputations performed annually on a population where 80% are over the age of 60 years, and the most common causes arising from vascular diseases (Guðbjörg K. Ludviksdóttir, written information, 2014).

Major lower limb amputations are performed either below the knee (transtibial amputations), above the knee (transfemoral amputation) or through-knee (knee disarticulation), all depending on the status of the patient and the clinical judgment of the surgeon. Other less commonly performed amputations are partial-foot amputation, Syme ankle disarticulation (though-ankle), hip disarticulation and transpelvic amputation (4). At what level to amputate is a complex decision based on numerous aspects of the patient’s condition and circumstances, with an obvious goal to achieve the best possible quality of life for each patient. For an individual whose goal is to be able to walk again, amputation level can be of great importance (5), but a study by Baum et al. (2008) showed that as long as the length of the femur of the residual limb is greater than 57% of the length of the contralateral intact femur, the length does not have a significant effect on any of the temporal-spatial, kinetic or kinematic parameters examined, but these results may not apply to individuals with shorter residual limbs. They further concluded that factors such as type of rehabilitation and type of prosthesis, as well as performance of myodesis could have a greater impact on the quality of gait (6).

1.1.2 TFA functional anatomy

All participants in this study are amputated above the knee. Not only is the motor system affected when it comes to lower limb amputees (LLA), but also the sensory system. For TFA, two major joints are missing, along with their intra-articular structure and function. This affects essential proprioceptive feedback from the knee joint and ankle joint, as well as from the muscular systems originating from, and below, the hip and the pelvis. With the loss of the insertions of the adductor muscles, their effective moment arm is shorter and while the femur no longer has its normal alignment with the tibia, the result of these changes can be a more abducted residual femur compared to the sound limb. This can alter the neuromuscular function of the lower limb, and therefore, along with many other factors, greatly affect the ability to ambulate.

Apart from the apparent impairments due to the loss of the foot and ankle and their muscular system, movements in all planes around the knee and the hip are compromised. With the loss of the distal insertion of the *m. Quadriceps*, there is no direct muscle control of knee extension, and flexion of the hip is controlled mainly by the *m. Iliopsoas* and to a smaller extent, the *m. Tensor fasciae latae*. Extension of the hip is weakened with the loss of the distal insertion of the *m. Hamstrings* and the *m. Adductor magnus*, although the *m. Gluteus maximus* is intact. *M. Gluteus medius* and *m. Gluteus minimus* which control abduction are preserved in TFA but abduction of the hip is somewhat weakened by the loss of the *m. Tensor fascia latae* distal insertion. Lastly, with the loss of *m. Adductor magnus* and other muscles of the adductor group, depending on the level of amputation, adduction is weakened. In order to prevent contractures and to maintain mechanical alignment and as normal muscle function as possible, myodesis is often performed where muscles are stabilized directly to the distal end of the femur. Often the main goal of myodesis for TFA is to restore adductor strength, so the *m. Adductor magnus* is fixed (7), and sometimes the *m. Hamstrings* for extension strength (8).

1.1.3 Complications secondary to amputation

Following a lower limb amputation numerous complications can arise, with long-term detrimental effects on the musculoskeletal system, which can have an effect on mobility and quality of life. Among secondary problems commonly seen among LLA is osteoarthritis (OA) of the hip on both the intact and amputated side, with up to sixfold higher incidence in hip OA compared to age-matched healthy controls, according to Kulkarni et al. (1998). They also found that there is up to a threefold increased risk of OA for TFA compared with TTA (9, 10). Another common problem is back pain, which seems to be one of the main concerns for LLA, with a higher prevalence than in the general population (11, 12). Studies have shown a prevalence of moderate to severe back pain ranging from 48%- 69% (11-15), with a significantly higher perception of back pain after the amputation compared to before (12).

Whether this is a problem that is more likely to occur among TFA rather than TTA is not clear, as study designs have varied and studies have therefore shown different results in that matter (12-17).

The causes for the above noted secondary problems can be numerous, and of various origin, for example less mobility, more reliance on the sound leg as well as general deconditioning. More detailed factors such as socket fit and alignment, postural changes and amputation level can all lead to changes in movement patterns and force distribution on the joints. In addition there are many other complications that can interfere with the quality of gait, such as phantom pain, which is a very common problem among amputees (11), and problems with the skin on the stump. Psychological problems are also of concern, as chronic pain has been shown to have a high relevance to depressive symptoms among amputees (11).

In an attempt to evaluate mobility, quality of life and general well being of Icelandic LLA, a questionnaire was sent out to all individuals that had gone through rehabilitation at Landspitali University Hospital in Iceland in the years 2000 – 2009 and had received a prosthetic foot. The response rate was 72.3%, or 34 individuals. Regarding secondary problems, 82% (N=28) of participants had had pain in the stump in the previous four weeks and 79% (N=27) phantom pain, but results did not include data on back pain specifically. Other complaints were skin problems (53%, N=18) and pain in the intact limb (29%, N=10) (18).

1.2 Gait

Gait is a very big part of the human locomotion, and one of the biggest components in amputee rehabilitation. In order to analyze pathological gait of any kind, it is essential to know the functional requirements and movement patterns of normal gait. In that respect researchers have broken each stride of the gait, which consists of one heel strike to the consecutive heel strike by the same foot, down to two phases; the Stance phase and the Swing phase. The stride can be further divided into eight functional patterns, each of them having a functional goal, that is, to accomplish weight acceptance, single limb support and swing limb advancement. The eight functional patterns are 1) initial contact, 2) loading response, 3) mid stance, 4) terminal stance, 5) pre-swing, 6) Initial swing, 7) mid-swing and 8) terminal swing (19). One step length is the distance between one heel strike to the consecutive heel strike by the opposite foot, and step width is the distance in the medial-lateral plane. Stride length, or one gait cycle, is the distance between two consecutive heel strikes of the same foot.

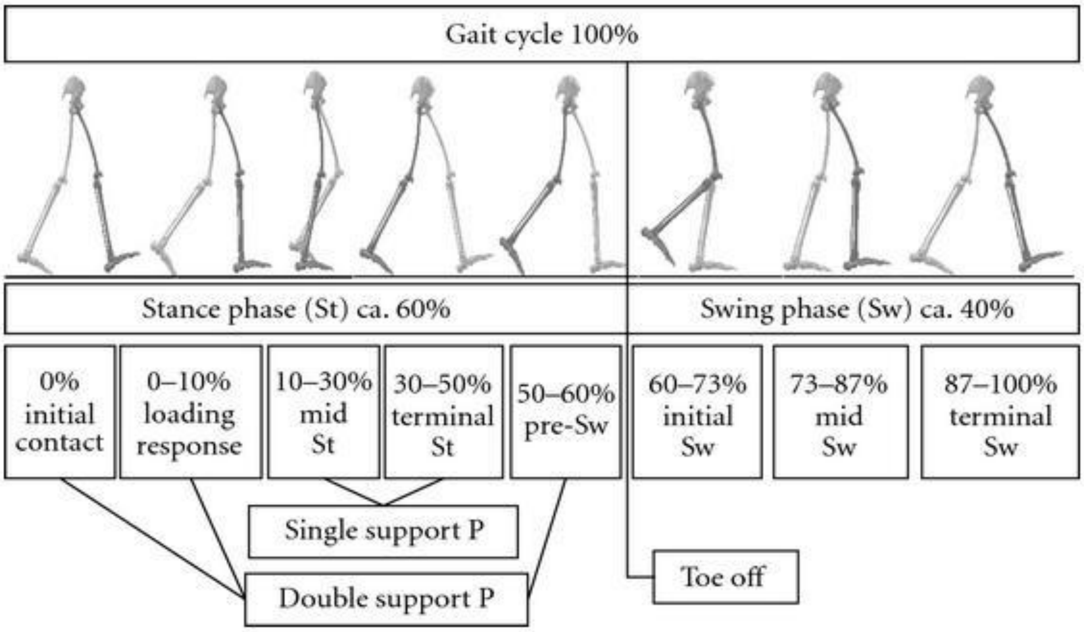

Figure 1. The gait cycle

1.2.1 Lower limb amputee gait

When learning to walk again after amputation the gait alters notably in many ways. Among apparent deviations in TFA gait are (4):

- Side-bending of the trunk, when the person leans over the amputated side during the stance phase of gait, possibly caused by weak hip abductors, faulty socket alignment or a short prosthesis;
- A wide step width, during the double support period of gait, sometimes as a result of insecurity, long prosthesis or simply due to pain or discomfort from the socket;
- An uneven step length, where the step of the amputated limb is frequently longer, possibly to compensate for insecurity of weight bearing on the prosthetic leg, or because of insufficient friction of the prosthetic knee;
- Circumduction of the amputated limb during swing phase, in an attempt to achieve a safe toe clearance, possibly to compensate for an excessively long prosthesis which can be a cause of various factors such as insufficient flexion of the knee, too small socket or excessive plantar flexion of the prosthetic ankle;
- Excessive lumbar lordosis during the stance phase of gait, possibly because of hip flexion contracture, weak hip extensors or abdominals or insufficient support from the socket.

This is not a complete list of possible compensations seen in LLA gait, rather a list of sometimes easily observable components seen by the examiner. In order to study the movements of LLA more precisely researchers typically analyze gait, either level gait (5, 20-23) or in other settings, such as stair walking, either descending or ascending (20, 24). Few studies have looked at the sit-to-stand test (25), incline walking (2, 26), circle walking (26, 27) and obstacle crossing (28). Gait can be analyzed in a variety of ways, and among LLA, 3D motion analysis, force measurements, electromyographic (EMG) recordings, various temporal-spatial parameters and oxygen consumption measurement are the most common research methods (29).

When looking at the biomechanical differences in TFA gait or LLA gait in general, comparisons are most commonly made with the gait of a healthy individual or the sound side. Numerous differences have been seen to occur for TFA during gait with various concluded causes, as study design/protocol and research questions vary. Reduced flexion and extension of the hip on the amputated side is a commonly reported finding (20, 30), and has been associated with an increased movement of the pelvis in the sagittal plane (5, 20, 30), possibly as a compensatory movement in order to maintain functional step length (30). Increased movement of the pelvis in the frontal plane (5, 20, 22) is an alteration found among TFA, often referred to as „hip hiking" seen most commonly on the amputated side, when the person raises the pelvis abnormally high in the swing phase, possibly to compensate for the lack of dorsiflexion of the ankle and achieve toe clearance, as was mentioned above. Other deviations are increased movement of the trunk in frontal, sagittal and transverse plane, which may alter the individual's global torque production, possibly having an effect on joint loading (5).

Variables derived from kinetic data comprise an integral part of understanding the strategies amputees use while ambulating and their effects on the musculoskeletal system. Main aspects of kinetic evaluation of LLA gait include joint moments, power and work. An example of a reported finding of altered kinetics in TFA gait is a significantly less concentric ankle work generated by the prosthetic ankle compared both to a healthy individual's ankle and the sound side ankle of the TFA, prior to the toe-off phase of gait (31), or when the foot plantarflexes in forward progression of the body. This decrease in work generated in turn has an effect on the motor strategies adopted by the amputee, in order to achieve the forward progression of the body and the advancement of the swing leg. Prinsen et al. (2011), in their systematic review of adaptation strategies among LLA in terms of joint power or work, reported a statistically significant increase in work (W) at the hip of the sound limb, and a tendency for more work at the hip of the amputated limb, compared to data from an able-bodied individual. These changes were very similar among TFA and TTA, however larger differences were shown for TFA (32). So in order to compensate for the decreased power of the prosthetic ankle, TFA might exhibit increased work at the hip and possibly increase the risk of joint degeneration of the hip, as has been documented to be high among TFA (10), and mentioned previously in this thesis. Another finding possibly indicating an increased effort on the sound side hip is hypertrophy of the *m.Iliopsoas* on the sound side and a corresponding atrophy on the amputated side among TFA, as measured with magnetic resonance imaging (15). Kinetic analysis in gait can give an insight into the efficiency of gait with respect to demands made of lower extremity muscles, which is of importance in LLA gait analysis in light of altered function of the remaining muscles, both of the amputated side and the sound side. The focus of this thesis, however, is on motion of and between defined segments, with reference to temporal-spatial parameters as well as muscle activation patterns during gait.

Changes in movement patterns are undoubtedly accompanied by changes in the effort of the remaining muscles on the amputated side of the TFA as well as on the sound side. The electrical activity of the muscles, as measured by electromyography (EMG), has been used extensively to assess recruitment patterns of muscles, such as signal amplitude and timing. Studies involving measures of muscle activity during TFA gait have mostly focused on the musculature around the pelvis, on both the amputated and sound limb. Hip muscles on the sound side and the amputated side in TFA have been measured active for a longer period of time than in healthy subjects (33). Bae et al. (2007) measured lower activity levels in *m. Quadriceps* and *m. Hamstring* during level walking, and at the same time, greater activity in *m. Tibialis anterior* and *m. Gastrocnemius* of the sound side of the TFA when compared with healthy individuals. During stair ascent and descent, activity in all of the aforementioned muscles was greater, excluding the *m. Hamstring* during ascent (20). Studies investigating muscle activation patterns in non- amputated persons with back pain have demonstrated prolonged activity of the *m. Erector spinae*, as did the study of Jaegers et. al, among TFA when comparing the *m. Erector spinae* activity during gait to that of able-bodied individuals (33, 34), but activity of the muscles of the back and trunk have not been extensively studied in TFA gait.

As for other physiological parameters of importance is the energy expenditure during gait, which has been shown to be higher among LLA compared to able bodied, and increasing with walking speed. Schmalz et al. found a 25% increase of oxygen consumption among TTA, and a 55-65% increase among TFA, when walking at a self-selected comfortable walking speed (35).

In their systematic review of biomechanics and physiological parameters during gait in LLA, Sagawa et al. (2011) discuss the difficulty in identifying the parameter that best serves gait analysis for LLA (29). In this study the main focus will be on parameters that have been thought to have an effect on the frequent occurrence of back pain among TFA. As mentioned above, some authors (16, 17) have reported a higher incidence of back pain among TFA than TTA. The only kinematic difference found between TFA with back pain and TFA without back pain, to the author's knowledge, is increased movement of the lumbar spine in the transverse plane (23). Back pain seen in TFA has been considered to arise from the muscles and connective tissues as a result of altered or wrong movement patterns, with gait training as an early intervention being a recommended prevention strategy (15)**.**

1.3 Effect of training among LLA

Few have studied the effect of training on pain and general function among TFA and different training protocols and outcome parameters have been used. In a study by Sjödahl et al. (2001, 2003) positive changes were seen in TFA who underwent special gait re-education program with combined methods in physiotherapy and psychological awareness for 10 months. After the training, improved movements of the pelvis in the frontal and transverse plane were seen, when compared to healthy individuals, as well as increased walking speed and less back pain (36, 37). Another study implemented a 10 week balance, co-ordination and strength training program and found an increase in hip strength and a decreased oxygen consumption (38). The authors concluded that whilst these positive changes improved the amputees' ability to run, the program's effects on gait mechanics were unknown and emphasized the need for further research in that field. Other programs such as an eight week treadmill training program has also been tested, demonstrating improvements in temporal-spatial gait symmetry at self-selected walking speed as well as decreased oxygen consumption (39).

With these results in mind a training protocol was designed for two of the participants in the current project with an even shorter training time or six weeks. A more detailed description of the training protocol and daily notes will be provided in appendix 1 (Training protocol).

1.4 Symbionic leg

All participants in this study used the same lower leg prosthesis, SYMBIONIC LEG® by Össur hf. (Reykjavik, Iceland). The bionic prosthesis is microprocessor-controlled and is a combination of Össur's RHEO KNEE® and PROPRIO FOOT®. The integrated components operate together, with built-in artificial intelligence systems and motion sensors that enable the prostheses to automatically adapt to inclined or declined surfaces during walking, by adjusting the angle of the ankle accordingly. The Proprio foot has motor-powered 4° dorsiflexion, with specialized sensors detecting the motion and powers of the joint during ambulation. Information is then sent to a motor that generates movements that helps the user with toe-clearance during the swing phase of gait. The adjustment is based on acceleration sensor data sampled with a frequency of 1600 Hz.

If the adaptive mode, i.e. the motor-powered dorsiflexion during swing phase and the adaptation to inclined or declined surfaces, is inactive or turned off, the function of prosthetic ankle is the same as in a low profile carbon fiber ankle. Studies have been made with this particular prosthetic foot on TTA during stair and ramp walking with the adaptive mode both active and inactive. During an active adaptive mode an increased knee flexion and increased knee moment on the amputated side was seen during stair walking (1) and a reduced knee extension during ramp ascend (2), which led the authors to suggest that a better function around the knee on the amputated side of TTA was obtained with the active control turned on.

A prostehtic foot commonly used by TFA is a passive carbon fiber foot. Whether an active foot like the PROPRIO FOOT can benefit TFA is not yet known. A greater understanding of the changes in movements during functional tasks like walking using an active vs. inactive adaptive mode of the prosthetic foot, will inform professionals such as physical therapists and prosthetic designers, and may ultimately improve outcome for the users.

2. Aims

The principle aim of this study is to examine the effect of an adaptive mode of a microprocessor-controlled prosthetic foot, on gait among transfemoral amputees, as well as the effect of a specific individualized training.

The specific aims are:

I. To compare participant´s amputated side to both the sound side and to healthy individuals during both level walking and incline walking. The active adaptive mode of the microprocessor controlled prosthetic foot, i.e. the motor-powered dorsiflexion during swing phase and the adaptation to inclined or declined surfaces, will be compared to the inactive adaptive mode during gait, with the main outcome measures being:
   - Temporal – spatial parameters
     - Step length on both sides
     - Step width
     - Double support time
   - Kinematics
     - Ankle movement in the sagittal plane
     - Knee movement in the sagittal plane
     - Hip movement in the sagittal plane
     - Hip movement in the frontal plane
     - Pelvic movement in the frontal plane

II. To analyze the effect of a specific individualized training and the effect of the active vs. inactive adaptive mode, in a within subject comparison, during level walking, with the main outcome measures being:
   - Temporal – spatial parameters
     - Step length on both sides
     - Step width
     - Double support time
   - Kinematics
     - Hip movement in the frontal plane
     - Pelvic movement in the frontal plane
   - Self report regarding pain, security, quality of life and prosthetic satisfaction
   - Balance
   - Electromyography
     - Activation levels

3. Methods

3.1 Participants

For this study five participants were recruited by convenience sampling, with all participants being active users of the prosthetics components from Össur (Reykjavík, Iceland). All participants were amputated above the knee, and met the following inclusions criteria: had been fitted with a prosthesis at least six months prior to the study, had to be able to ambulate without assistive device and lastly had to have BMI less than 35 kg/m^2 for the purpose of optimal EMG sampling and marker positioning. In addition four healthy individuals were recruited as control subjects, also by convenience sampling, with the aim to match the participants of the study with respect to gender, age, height and weight. Having received information regarding their role and the implementation in the study (appendix 2; Kynningar-og upplýsingarblað), participants signed an informed consent form (appendix 3; Upplýst samþykki).

The study was approved by the Faculty of Medicine at the University of Iceland, ethical approval was granted by the Bioethics committee of Iceland and an announcement was sent to the Icelandic Data Protection Authority.

3.2 Experimental procedure

3.2.1 Examination and Questionnaires

A general physical examination was performed by a physical therapist in order to identify possible factors relating to pain or reduced function in daily life. Joint range of motion (ROM) was measured, using a goniometer and specific standard muscle length/joint ROM tests, as was muscle strength, using manual muscle testing (MMT). Deviations in posture, such as a rotated pelvis or asymmetries in limb length, as can often be seen in LLA (40), among others, were noted. Balance was examined by testing the ability to stand on one leg, both on stable surface, and on a Airex balance pad (Airex AG, Switzerland) which is a pad made of a yielding foam that adds difficulty in maintaining balance. Gait was thoroughly examined, noting if there were any deviations in the gait pattern. This examination was done in part to identify participants who might benefit from a training program in order to improve their gait and to assess general function. Based on the findings of the physical examination, three participants were assigned an individualized program (appendix 1, Training protocol). The remaining two participants did not, as they showed no significant deviations in strength, gait or posture and were both elite athletes. Two measurement sessions were conducted, both before and after the six week training period. All participants were asked to attend both sessions, regardless of whether they received a training program or not.

At the start of measurements all TFA participants answered questions regarding general quality of life and satisfaction with the prosthetic limbs (appendix 4; Lífsgæði). They answered the Activities-specific Balance Confidence Scale - ABC scale, which is a measure of self-efficacy designed to assess fear of falling and is considered reliable in assessing LLA (41) (appendix 5, The Activities-specific Balance Confidence Scale – The ABC scale). Participants rated pain if any (rating based on the Numerical rating scale) (42). The “Timed up and go” (TUG) (43) test was performed, which assesses basic motor skills necessary to handle a variety of daily activities. In the TUG test the individual rises up from a chair, walks three meters, turns around and walks back three meters and sits down in the chair, with the result being time measured seconds that it takes for the individual to complete the test. The test in the current study was modified by adding an obstacle at the end of an extended five meter walkway in order to add the level of difficulty in a functionally relevant manner (44). Participants that received the training program answered all of the questionnaires and performed the TUG test, both before and after the training period.

3.2.2 Data collection and analysis

Kinematic measurements were made using an eight camera 3D motion capture system (Qualisys AB, Gothenburg, Sweden). Data were collected at 250 Hz, and marker position data were filtered with a Butterworth filter with a cut-off frequency at 15Hz. The marker setup that was chosen according to the recommendations of C-motion (Germantown, USA) (45). Thirty reflective markers were placed on the following landmarks (Figure 2 and 3; note that not all of the markers are visible in the figures):

- *First and fifth metatarsals (bilaterally)*
- *Medial and lateral malleoli (bilaterally)*
- *Heel (2 markers) (bilaterally)*
- *Medial and lateral epicondyle of femur (bilaterally)*
- *Trochanter major (bilaterally)*
- *Crista iliaca (bilaterally)*
- *Acromion (bilaterally)*
- *Anterior superior iliac spine (ASIS) (bilaterally)*
- *Posterior superior iliac spine (PSIS) (bilaterally)*
- *Sacrum*
- *Manubrium*
- *Cervical vertebrae 7 (C7)*
- *Thoracic vertebrae 10 (Th10)*
- Four marker clusters were placed at *thighs* and *shanks* bilaterally.

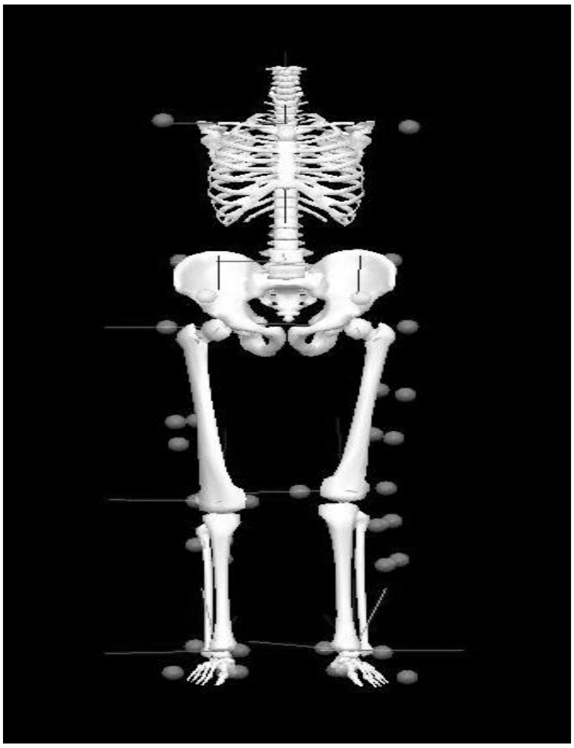

Figure 2. Marker placement, frontal view, from Visual 3D^{TM}

Figure 3. Marker placement, sagittal view

Markers on the prosthetic ankle and knee were placed at the joint center of rotation and marker placement of the metatarsals 1 and metatarsals 5 on the prosthetic foot markers were approximated according to the sound limb. These markers defined the segments from which the parameters were derived. While the design of the foot cover of the prosthetic ankle often prevented marker placement directly on the joint center of the ankle, offset measurements were made in order to adjust the data accordingly. Since force plates were not used in this study, marker data were used to label specific events in the gait cycle, i.e. „Heel Strike“ and „ Toe Off“. All markers were placed by the same individual.

Muscle activation of *m. Gluteus med* and *m. Tensor fasciae latae* were captured with wireless surface EMG (KineMyo KMS 8N, EMG Triode Electrode. Kine ehf, Reykjavik, Iceland). Electrode placement was according to SENIAM recommendations (46) and Bird et al. (2003) (47). Data was collected at 1600 Hz and the electrodes had a low-pass filter at 500 Hz and a high-pass filter at 16 Hz. For normalization of the EMG data, and for the purpose of strength measurement, participants performed maximal voluntary isometric contractions (MVIC) for the abductors of the hip. This was done by having the participants lying on the side, with a strap around the most distal part of the femur with a hand held dynamometer placed under the strap (48).

Both static and dynamic measurements were captured within a pre-calibrated area during both level walking and a 4° ramp ascent, at a self-chosen comfortable walking speed and during a steady state of gait. Measurements were made with adaptive mode of the microprocessor-controlled prosthetic foot both active and inactive. Walking trials were repeated five times on average, or as often as necessary to get five good kinematic recordings and three good EMG recordings. Caution was taken in data capturing during ramp ascent, while the adjustment time for adaptive mode of the prosthetic foot for incline walking are approximately three steps, the measurements started after that adaption time.

Kinematic calculations for analysis were made using Visual3D™ (C-motion Inc, Germantown, USA). In order to obtain the 3-dimensional motion, angles either between a) two segments or b) between a segment (pelvis) and the laboratory reference system were calculated. Data were exported and Microsoft Excel used for viewing and further mathematical calculation. Angular ROM was calculated for the ankle, knee and hip in the sagittal plane and for the hip and the pelvis in the frontal plane, for the duration of swing phase and weight acceptance, specifically. For the excursions of the hip in the sagittal plane, flexion and extension excursions were calculated. Temporal-spatial data, i.e. step length, step width and double support were also obtained from these data. Results were averaged across the trials for each condition measured.

3.3 Statistics

Mixed model analysis of variance (ANOVA) was used to evaluate each kinematic variable of both limbs (‚*limb*‘) for each setting of the prosthesis (‚*mode*‘) during level and incline walking (‚*gait condition*‘) (within-subject factors) between groups (amputated participants (Amp) and control subjects (Ctrl)). The assumption was made that controls exhibited the same kinematic patterns during the amputee active and inactive setting of the prosthesis. Alpha was set at 0.05.

4. Results and discussions

Based on the methodological grounds of this study, and the nature of this data analysis, this chapter will combine results and discussion. The chapter will be divided into three sections. After presentation of the participants' demographic characteristics, the remaining sections focus on results relating to each of the specific aims of the study:

- **I.** Gait analysis and comparisons will be made using data from the first data collection session. Contrasting the groups´ mean measures of both limbs, during level and incline walking with both active and inactive adaptive mode of the prosthetic foot.
- **II.** Two case-studies will be described, for participants 2 and 3, where comparisons will be made between both inactive and active adaptive mode of a prosthetic foot before and after training. Specific focus will be on hip and pelvis kinematics in the frontal plane.

4.1 Participants characteristics

Participants 1, 2, 3 and 5 completed both measurement sessions, while participant number 4 was unable to attend the training program as well as the second measurement due to personal reasons. Four gender-, age-, height- and weight- matched able-bodied individuals were also recruited for the study, for one measurement. Table 1 summarizes the main characteristics of the TFA participants.

Table 1. Overview of participants characteristics

Participant:	1	2	3	4	5
Gender	Male	Male	Female	Male	Male
Age (years)	32	42	38	36	24
BMI* (kg/m^2)	24,4	22,8	24,3	26,6	26,8
Time since amputation (years)	13	22	29	10	6
Cause of amputation	Tumor	Trauma	Tumor	Trauma	Tumor
Individualized training	No	Yes	Yes	Yes	No

**Body Mass Index*

4.2 Gait analysis- Comparisons between groups and conditions

4.2.1 Temporal-spatial measurements

The parameters analyzed here are the step length of each side, the step width and the time spent in double support (both the right initial double limb support time and the right terminal double limb support time). Comparison between the averaged results from the five amputees and the averaged results from the four controls can be seen in Table 2 for both level and incline walking.

Table 2. Mean (SD) temporal-spatial measurements

		Step length Amputation side - meters	Step length Sound side - meters	Step width - meters	Double support- seconds
		Results for Level walking			
Amputees	Inactive*	0.85 (0.02)	0.71 (0.01)	0.16 (0.02)	0.38 (0.02)
	Active*	0.84 (0.02)	0.72 (0.02)	0.16 (0.01)	0.38 (0.02)
		Left	*Right*		
Controls		0.77 (0.04)	0.76 (0.03)	0.14 (0.02)	0.37 (0.05)
		Results for Incline walking			
Amputees	Inactive*	0.84 (0.02)	0.72 (0.02)	0.16 (0.01)	0.41 (0.03)
	Active*	0.84 (0.02)	0.70 (0.09)	0.16 (0.01)	0.40 (0.02)
		Left	*Right*		
Controls		0.81 (0.03)	0.80 (0.03)	0.12 (0.02)	0.38 (0.05)

*Inactive and Active = represents the setting of the mode of the prosthetic foot.

There was a significant limb by group interaction for step length (p= 0.009). The step length was longer for the amputated compared to the sound side of amputees, while the step length was equal bilaterally for the control group. This difference in step length exhibited by amputees might be caused by a lack of control of the remaining musculature of the amputated limb and of the extension of the prosthetic knee, or because of insecurity during weight bearing on the amputated side. There was also a significant interaction of group by gait conditions (incline vs. level walking; p = 0.031), as amputees slightly decreased their step length during incline walking while the control group increased their step length.

Amputees demonstrated a significantly larger step width than controls (p=0.033), with possible causes being insecurity, too long prosthesis or discomfort from the socket. An interaction of group by gait condition was not statistically significant (p=0.088). Controls did, however, tend to decrease their step width during incline vs. level walking while this was unchanged in amputees. This trend reflects the increased step length during incline walking of controls, which may in part be achieved by a greater rotation of the pelvis in the transverse plane, thereby causing a narrowing of step width. Amputees may seek to avoid such changes during incline walking, possibly due to insecurity, hence the significantly decreased step length of the amputees.

The double limb support time was significantly longer during incline vs. level walking across both groups (p= 0.017). A longer double limb support could be an indication of insecurity.

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4.2.2 Motion capture – gait analysis

Kinematics of the ankle, knee and hip in the sagittal plane and of the hip and pelvis in the frontal plane are presented in figures 4-13. In each graph series, averaged data from the first measurement for the five transfemoral participants and the four control subjects are presented for both limbs, for the purpose of interlimb comparison between groups. Graphs will be shown as a percentage of a gait cycle, from heel strike to the consecutive heel strike of the same foot. Graphs for level walking and incline walking will be presented consecutively.

Various postural deviations can be seen among TFA, for example decreased hip extension and increased pelvic tilt in the sagittal plane on the amputated side compared to the sound side, as well as limb length discrepancies (40). During processing the data were not normalized to the participants' standing calibrations, as this would imply that all joints reached neutral (zero degrees) during standing. This, however, is not realistic for this study population and the intent was to capture possible contractures or other deviations in posture in the standing trial. The kinematic values presented are, therefore, based on calculations derived from the marker data of each session, which may include errors related to positioning of anatomical markers. Data analysis was performed with regards to movement excursions at a given phase in the gait cycle, rather than looking at an exact angle at a given event. Here the excursions are defined as the absolute value of the difference between the largest and the smallest angle at each phase.

In the graph interpretations, movement patterns of each joint will be described and compared in terms of joint excursions, with discussions thereof. One of the aims of this study was to look at the effect of an adaptive mode of the prosthetic ankle, hence the focus on the swing phase (SW) in the kinematic analysis. The SW is the last approximately 40% of the gait cycle, from toe off to initial contact (IC) on the ipsilateral limb. Weight acceptance (WA) is commonly defined as the phase from IC to toe off of the contralateral limb, or the first approximately 20% of the gait cycle. During WA important factors in determining the quality of gait occur, such as weight shift, power absorption and forward propulsion, and therefore WA was also examined in the current kinematic analysis. When applicable, other areas of the gait cycle will also be discussed. Joint excursions were calculated for each of the two parts, presented in tables below each graph.

4.2.2.1 Ankle kinematics in the sagittal plane

The graphs in figures 4 and 5 present the data for the ankle kinematics in the sagittal plane (plantarflexion – dorsiflexion) of level and incline walking, respectively.

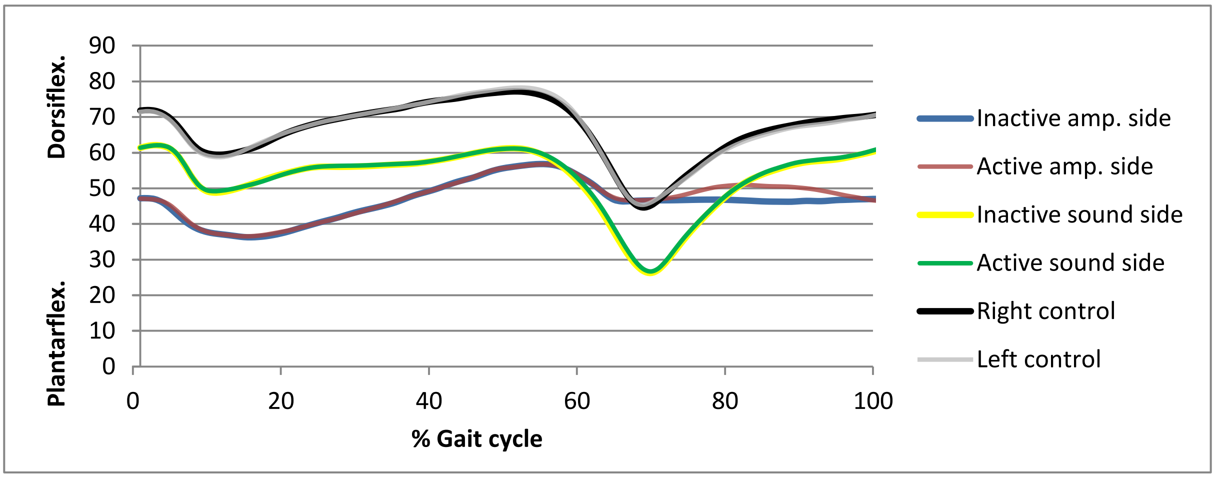

Figure 4. Mean ankle kinematics in the sagittal plane during level walking.

Table 3. Mean ankle excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	10.9°	10.6°	13.3°	12.9°	12.5°	12.9°
SW	0.5°	4.4°	31.3°	31.2°	26.2°	25.5°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

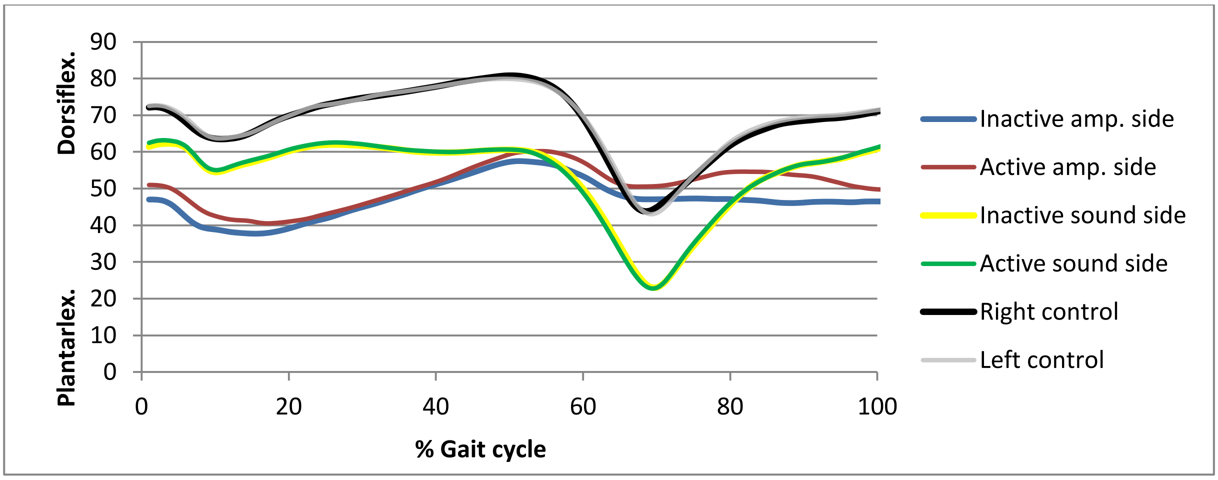

Figure 5. Mean ankle kinematics in the sagittal plane during incline walking.

Table 4. Mean ankle excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	9.3°	10.5°	7.8°	8.1°	8.8°	9.0°
SW	0.2°	4.1°	34.2°	34.2°	25.3°	26.7°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation

Weight acceptance: At IC, as the heel strikes the ground, the ankle plantarflexes while reaching foot flat. Visual inspection reveals that the ankle plantarflexion excursion rate was greater for the sound side and the control group than for the prosthetic ankles, with a greater excursion during level walking but not during incline walking, which is logical because of the 4° elevation of the surface during incline walking. This difference between gait conditions was significant (a limb by gait condition by group interaction; p = 0.004), as the mean plantarflexion excursion of the prosthetic foot did not change between level and incline walking for the amputated side, while it did for the sound side and both ankles of the control group.

While the prosthetic ankle does not have the same physiological adaptation to the incline walking as a sound ankle would, the increased dorsiflexion at IC during incline walking with the active adaptive mode reflects the function of the microprocessor control and, as a result, greater plantarflexion occurs during WA compared to the inactive adaptive mode, although this did not reach statistical significance.

During single limb support the ankle dorsiflexes while the body advances, reaching the maximal dorsiflexion right after the IC of the contralateral foot (at approximately 55% of the gait cycle). The dorsiflexion excursion of the amputees' sound side ankles differs a little from the control group ankles, and more so during incline walking and could be an attempt to ensure toe clearance of the prosthetic foot, and reduce the risk of tripping.

At the end of the single limb support the foot prepares for the swing phase by plantarflexing the ankle, using mainly the *m. Gastrocnemius* and *m. Soleus* to produce the propulsive forces needed for the advancement of the body through the gait cycle. In the absence of these muscles the plantarflexion excursion of the prosthetic foot is much smaller than for the sound side ankles and the control ankles, which inevitably will cause the amputee to compensate in various ways.

Swing phase: When looking at the data for the amputated side (red and blue lines, inactive and active adaptive mode respectively), the effect of the microprocessor control of the prosthetic foot during gait can be seen. During the swing phase of gait, which begins approximately at 65% of the gait cycle, the ankles achieves approximately 4° of dorsiflexion with the active adaptive mode (red line), while during the inactive adaptive mode (blue line) the dorsiflexion is close to 0°. During SW there was a significant interaction (limb by mode by group; p=0.014) due to greater dorsiflexion of the prosthetic ankle in amputees during active vs. inactive mode across gait conditions (Figures 4 and 5).

The excursions of the sound side ankles were slightly greater than those found in control ankles bilaterally, both during level and incline walking. During the first stages of rehabilitation, there is a tendency to keep the amputated limb length slightly shorter than the sound limb in order to achieve safe toe clearance during swing, but as confidence in walking increases for the amputee, the limb length can be increased in order to reduce postural asymmetries. The greater excursion of the sound side ankle could possibly be a result of this limb length difference.

There was also a significant interaction (limb by gait condition; p=0.033) due to slightly greater excursion on the sound/right limb during incline vs. level walking, across groups.

4.2.2.2 Knee kinematics in the sagittal plane

The graphs in figures 6 and 7 present the data for the knee kinematics in the sagittal plane (flexion – extension) for level and incline walking, respectively:

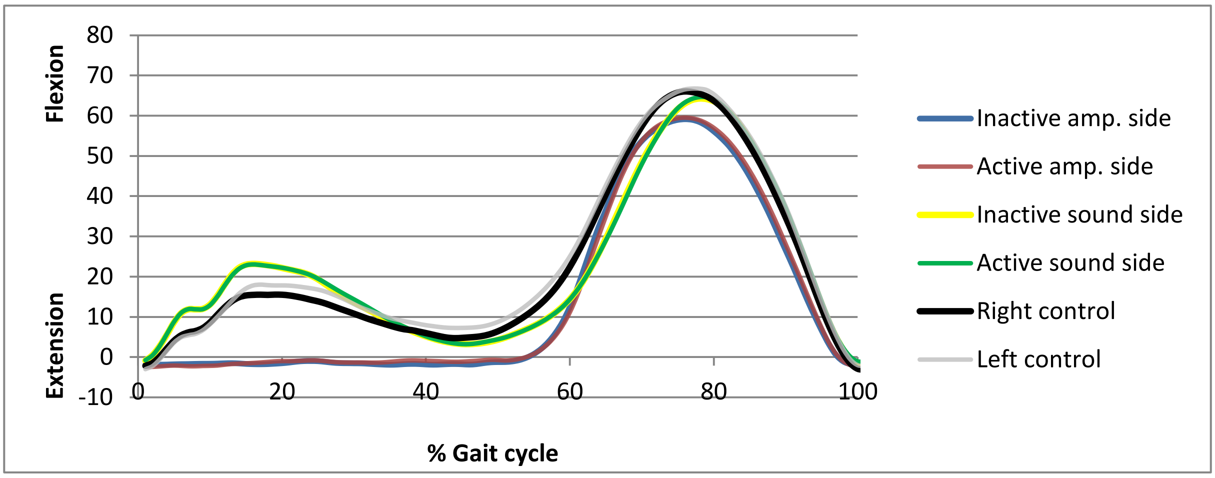

Figure 6. Mean knee kinematics in the sagittal plane during level walking.

Table 5. Mean knee excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	0.9°	1.6°	24.0°	23.8°	17.7°	21.0°
SW	60.9°	60.6°	61.0°	61.4°	61.3°	59.5°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)

SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

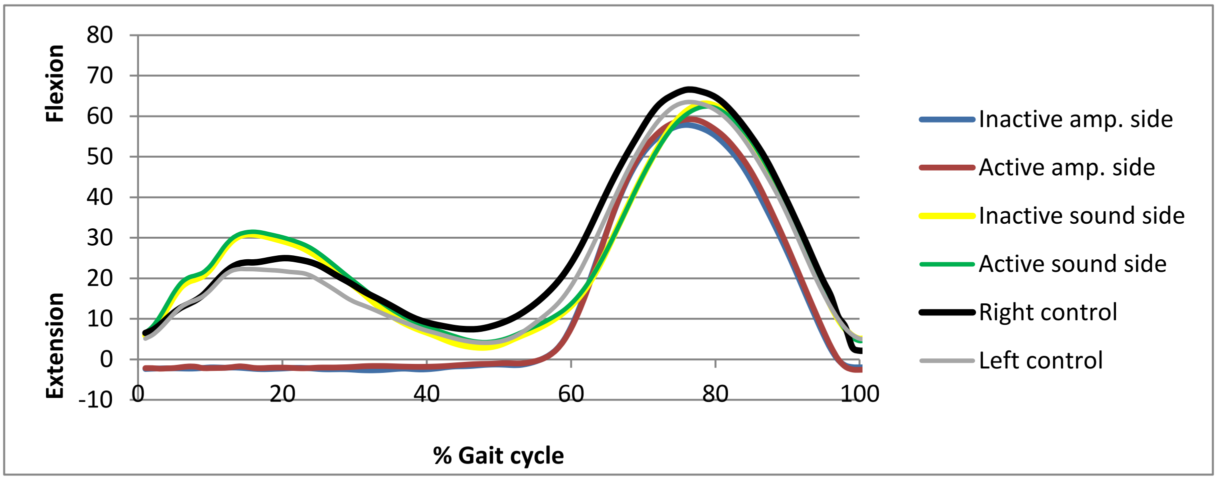

Figure 7. Mean knee kinematics in the sagittal plane during incline walking.

Table 6. Mean knee excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	0.4°	0.7°	24.9°	24.9°	17.2°	18.4°
SW	60.2°	61.0°	60.1°	58.1°	59.4°	59.1°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)

SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation:

W*eight acceptance:* Normally during WA, the knee joint flexes right after IC to reach a maximum at the end of double limb support or where the contralateral toe clears the ground. The prosthetic knees, however, exhibited close to no flexion and therefore a statistical limb by group interaction was found (p=0.001).

At the end of single stance, in preparation for the swing phase, there is, visually, a sharper rate of change from knee extension to knee flexion for the prosthetic knees than for the sound side and the control knees, i.e. they do not „give in“ in the same manner as naturally as the sound side and the control knees do. Furthermore, this change from extension to flexion occurs later in the gait cycle, compared to the sound side knees and the control knees, which may have an effect on various parameters during WA on the sound side. The amputee sound knees demonstrated around 24° knee flexion excursion during WA, compared to approximately 18°- 21° seen in control knees (Table 5). This might possibly be a result of the marked stiffness of the prosthetic knee in the late stance phase, causing this compensation of the sound side knee to occur, when the weight is transferred from the amputated side to the sound side. Previous studies have reported an increased loading on the sound side limb compared with control subjects, and a decreased loading on the amputated limb as assessed by vertical ground reaction force magnitude (49). The compensations reported here could be in accordance with those results. Another possible reason for the larger flexion excursion on the sound side compared to controls, could be a compensation caused by a slightly shorter limb length of the prosthetic side, as mentioned in the ankle sagittal graph interpretation above. .

While analyzing the data for knee kinematics, it was noticed that one of the participants exhibited approximately 4° flexion of the prosthetic knee during WA, a movement that did not appear when the data were averaged, indicating that the other participants were not utilizing this function of the prosthetic knee.

Swing phase: A greater rate of exchange from knee flexion to extension of the prosthetic knees, compared to the sound side and the control knees, is visible in figures 6 and 7. Here the excursion across conditions is very similar both during level and incline walking (Tables 5 and 6).

A noticeable difference in knee flexion at WA was seen between level and incline walking for the sound side as well as control knees during incline walking. The prosthetic knee cannot respond to the inclined surface since the center of rotation needs to be posterior to the ground reaction force vector, otherwise the prosthetic knee would buckle (50).

During SW there were no statistical differences found in knee joint excursions, indicating minimal influences of the adaptive mode of the prosthetic foot on the excursions of the prosthetic knee.

4.2.2.3 Hip kinematics in the sagittal plane

The graphs in figures 8 and 9, present the data for the hip kinematics in the sagittal plane (flexion – extension) for level and incline walking, respectively:

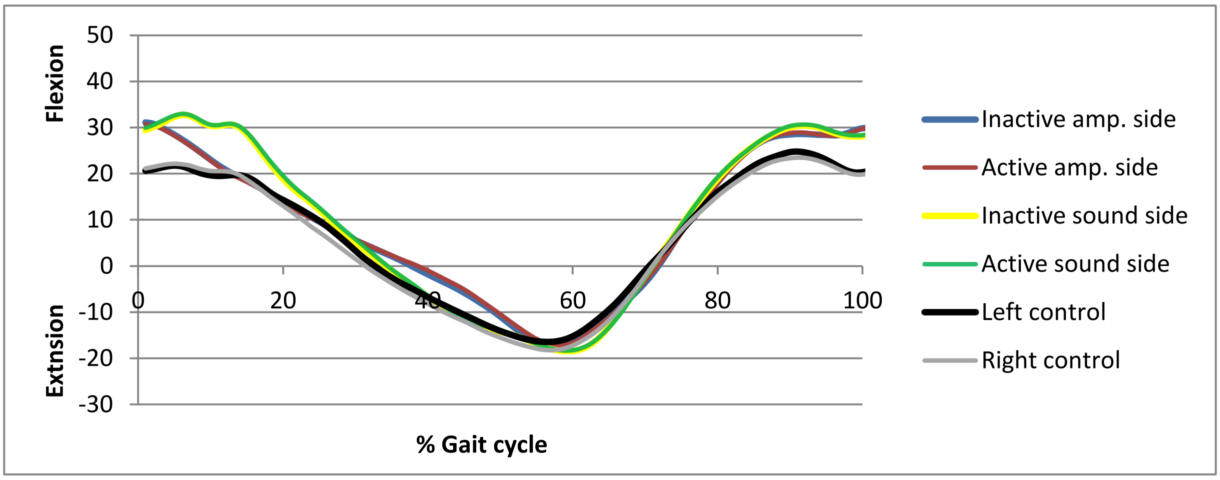

Figure 8. Mean hip kinematics in the sagittal plane during level walking.

Table 7. Mean hip excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
Ext	48.4°	47.9°	51.2°	51.3°	38.2°	40.4°
Flex	47.2°	47.0°	48.9°	49.0°	41.2°	41.7°

Ext = Hip extension (total hip extension during stance phase of gait, approx. 1%-60% of gait cycle)

Flex = Hip flexion (total hip flexion during swing phase of gait cycle, approx. 60%- 100% of gait cycle)

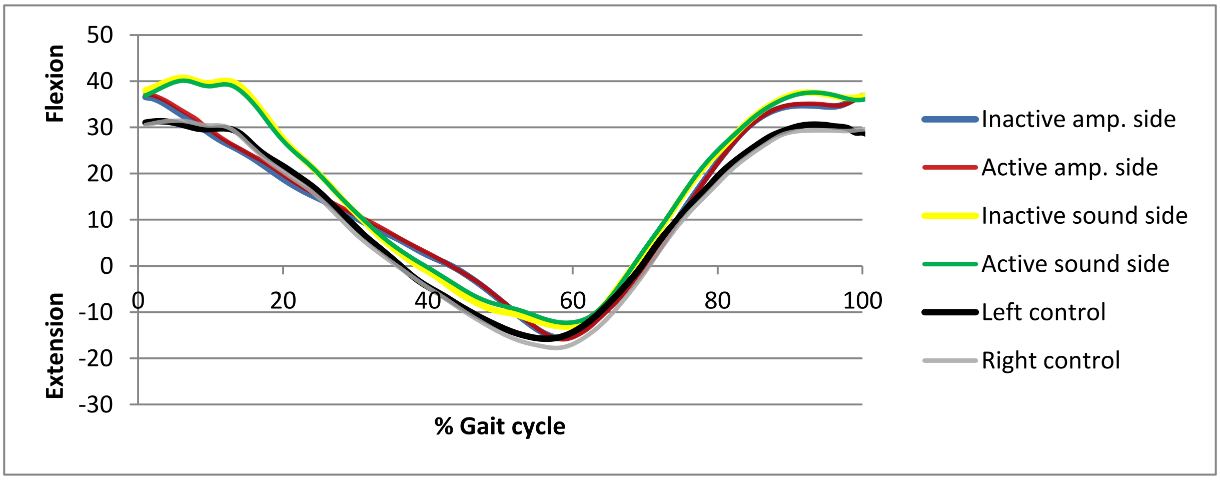

Figure 9. Mean hip kinematics in the sagittal plane during incline walking.

Table 8. Mean hip excursions in sagittal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
Ext	52.1°	53.0°	53.9°	52.3°	47.1°	49.1°
Flex	52.3°	52.5°	50.6°	49.8°	46.3°	47.3°

Ext = Hip extension (total hip extension during stance phase of gait, approx. 1%-60% of gait cycle)

Flex = Hip flexion (total hip flexion during swing phase of gait cycle, approx. 60%- 100% of gait cycle)

Graph interpretation:

Weight acceptance: A sharp change from hip flexion to extension was seen on the amputated side, right after IC, in contrast to the sound side, that initially exhibited slight hip flexion. Control group, however, maintained the same degree of flexion during those first instances of the gait cycle, prior to moving towards extension. The sharp extension of the amputated side could be due to the absence of a normal physiological response (flexion) of the knee and ankle. The knee flexion at WA usually slows down the transfer from hip flexion to extension when the trunk is pushed forward by the contralateral limb, as seen in controls. The step length on the amputated side is longer, compared to the sound side, which might also play a role in the sharp change from hip flexion to extension on the amputated side, i.e. the hip needs to move quicker into extension in order to achieve good load transfer.

The greater flexion at the beginning of the gait cycle on the sound side hips is likely associated with the increased knee flexion excursion of the sound side at WA. This might also be due to a forward trunk lean to compensate for less plantarflexion force production on the amputated side, and thereby shifting the center of mass forward. Furthermore the amputee might be using the hip to assist in the advancement of the body, causing increased loading of the sound side hip (32). The greater flexion at WA on the sound side may also result in the larger hip extension excursion seen during stance, compared both to the amputated side and control group hips. A delay in knee flexion of the prosthetic (contralateral) side at terminal stance may also play a role in the hip kinematics of the sound side, as while the prosthetic knee does not flex, the hip will need to compensate with greater extension, which might pull the pelvis into an anterior tilt, thereby increasing relative flexion measures of the sound hip. This might be interconnected to the longer step length of the prosthetic side limb presented in this study. During WA there was a significant gait condition by group interaction (p= 0,036), reflecting a difference in how the groups altered their gait patterns between incline and level walking.

Swing phase: Bilateral hip flexion excursion during swing was greater in amputees than that seen in control hip joints. Greater hip flexion of the amputated side may, in part, be caused by a lack of control of the remaining hip and thigh musculature or due to momentum caused by the weight of the prosthetic foot. The greater hip flexion for the sound side hip could be associated with a simultaneous greater hip extension of the amputated side, which may tilt the pelvis anteriorly and influence hip ROM measures. During swing there was a significant main effect of gait condition (p=0.001), reflecting generally greater hip flexion during incline vs. level gait, bilaterally, across both groups

As noted earlier in the thesis other authors have reported decreased flexion - extension of the hip of the amputated side, which is in contrast to the results of the present study. In a study by Rabufetti et al.(2005), examining hip and pelvic ROM, TFA participants demonstrated less hip extension on the amputated side at the contralateral IC and less hip flexion at ipsilateral IC, compared to the sound side hip. They also reported greater sagittal plane pelvic movement at sound side IC, a parameter that was not analyzed in the present study. The authors concluded that these compensatory movements of the hip and pelvis could be a combination of mechanical constraints, caused by a socket in which there is a direct contact to the ischial tuberosity, and the amputee's attempt to obtain a functional step length (30). In the present study the complete extension and flexion excursion were examined rather than the exact angle at a given gait event. This difference in calculation/data analysis makes it difficult to compare the results of the present study to the one by Rabuffetti et al. However, in a study by Jaeger et al. (1995) similar increases in hip flexion and extension were found as in the present study, although the authors did not provide information as to how the excursions were measured (21).

4.2.2.4.1 Hip and pelvis kinematics in the frontal plane during level walking

Because of the integral relationship of movements in the hip and pelvis, the graph interpretations of the two segments in the frontal plane (hip adduction – abduction and pelvic obliquity) will be presented consecutively, i.e. first the hip and pelvis kinematics during level walking and then the hip and pelvis kinematics during incline walking. Figures 10 and 11 will show the hip and pelvis data, respectively, during level walking and figures 12 and 13 present the same data during incline walking.

The kinematics of the hip and the pelvis are somewhat intertwined: At IC and during WA of the leading limb, the hip on the ipsilateral side adducts. This is a movement that is in part due to a pelvic drop on the contralateral side, and in part due to the weight shift onto the leading limb. At the same time a relative hip abduction on the contralateral side occurs. The adduction of the leading limb reaches a maximum approximately when the toe of the contralateral limb lifts off the ground and moves into swing phase, with a concurrent leveling off of the pelvis. The maximum pelvis lift (on the contralateral side) is approximately at mid-stance after which the pelvis then drops back down to initiate stance on the contralateral side. This way the pattern repeats itself for the consecutive stride. As an example of the relationship of the hip and pelvis kinematics is that during swing phase an increase in pelvic lift would be expected to have the effect of an increased adduction of the hip on ipsilateral side.

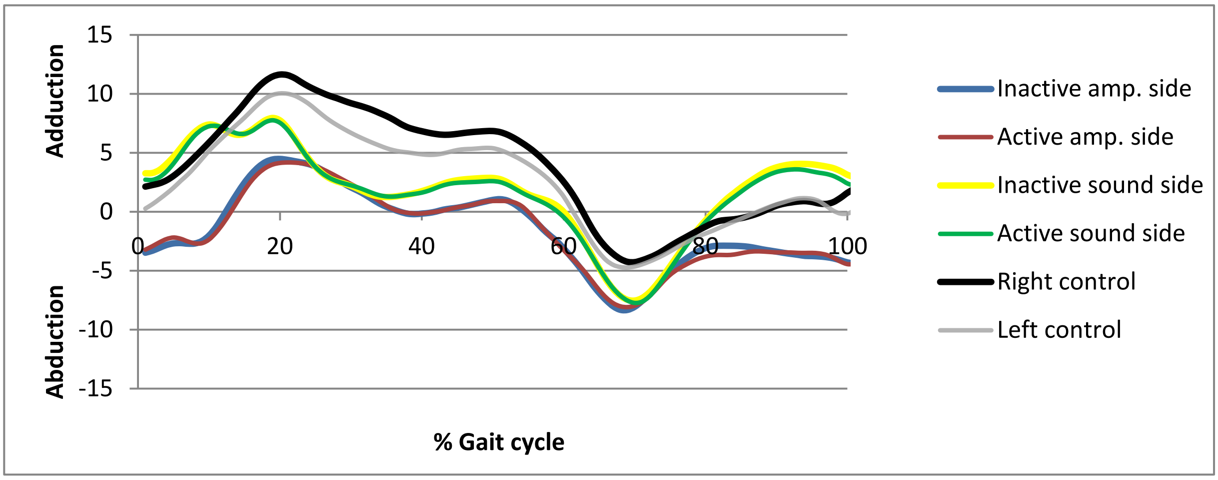

Figure 10. Mean hip kinematics in the frontal plane during level walking.

Table 9. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	8.0°	7.4°	4.7°	5.1°	9.5°	9.8°
SW	5.4°	4.7°	11.5°	11.1°	6.2°	5.9°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

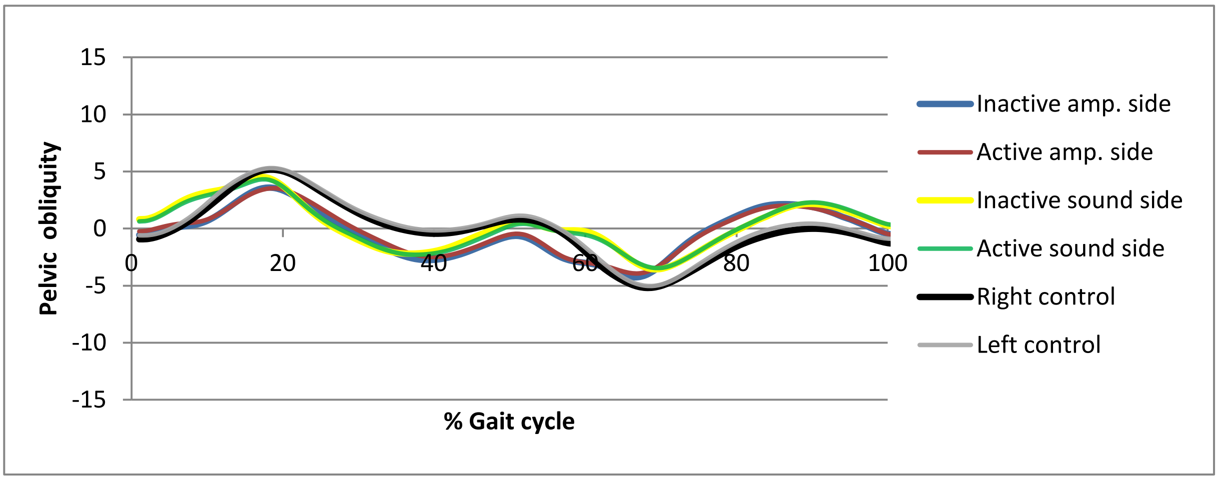

Figure 11. Mean pelvis kinematics in the frontal plane during level walking.

Table 10. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	4.2°	3.8°	3.7°	3.7°	6.1°	5.9°
SW	6.4°	5.9°	5.7°	5.7°	5.2°	5.4°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation:

Weight acceptance: The adduction movement occurs steadily in both hips of the control group (around 10°; table 9), while for amputees the sound side hip adduction movement pattern is somewhat irregular and the excursion was slightly smaller in comparison. As seen in the kinematic data of the hip in the sagittal plane presented earlier, there was an increased hip flexion at the WA for the sound side hip, suggested there as being associated with increased knee flexion on the sound side and as a compensatory mechanism caused by the absence of plantarflexion force on the amputated side. These „fluctuations“ in the current hip graph might be a result of those increased movements in the sagittal plane, as well as of the delayed flexion of the prosthetic knee in terminal stance, as described in the knee, sagittal plane graph interpretation. Simultaneously, relative movement of the pelvis, i.e. pelvis drop on the amputated side, was decreased, compared to the control group (figure 11).

The hip on the amputated side did not adduct until at approximately 10% of the gait cycle, and at the same time the pelvic drop was less, and occurred with a different pattern than for the control group. The decreased pelvic drop is in part consistent with the findings of Michaud et al. (2000) (22), who compared the prosthetic side to the sound side and found differences between the two sides. In the present study the difference in mean adduction values is slightly larger when comparing the amputated side to the control group, than to the sound side. The authors discuss possible reasons for lower pelvic drop values as being restrictions from the socket or a lateral trunk lean over the amputated side during stance phase, often seen among TFA (5, 21, 22). Lateral trunk lean can be caused by weak abductor strength, faulty socket alignment, too short prosthesis (4) or insecurity while transferring the load over the prosthesis.

Swing phase: During swing the sound side hip adduction excursion was noticeably the largest (around 11°; figure 10 / table 9). Of the known compensatory mechanisms for TFA, increased frontal plane trunk movement or lateral trunk lean (5, 21, 22) over the amputated stance limb could explain a relatively larger hip adduction excursion seen on the sound side during swing. Minimal differences were seen in value and pattern of pelvic excursions.

The amputated side had the smallest adduction excursions, and the largest pelvis lift excursion during swing, compared to the sound side hip and controls, which is in contrast to a normal kinematic pattern where an increase in hip adduction is expected as pelvis lift increases. Almost the entire hip adduction excursion on the amputated side occurs during the initial swing phase, whereafter the hips did not adduct more, as it did for the sound side and the control group hips, indicating a dysfunction in the hip/pelvis kinematics. Circumduction is a known compensation among amputees and is a likely explanation for the decreased adduction seen here for the hip on the amputated side.

The results of the adduction excursions are in context of the previously described increased step width among the amputees compared to the control group, as has been demonstrated in previous studies (21).

4.2.2.4.2 *Hip and pelvis kinematics in the frontal plane during incline walking*

The graphs in figures 12 and 13, present the data for the hip and pevis kinematics in the frontal plane for incline walking, respectively:

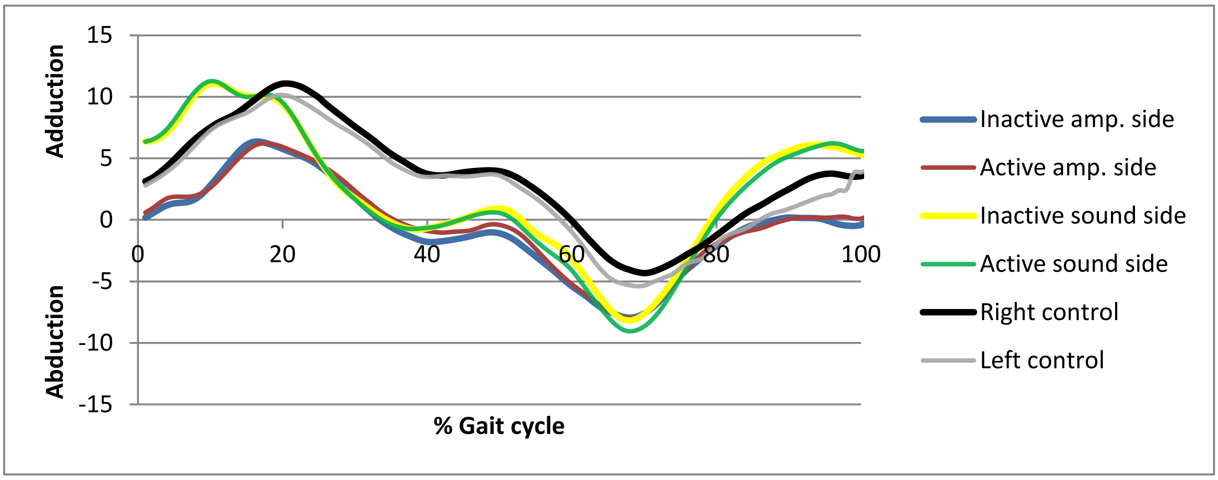

Figure 12. Mean hip kinematics in the frontal plane during incline walking.

Table 11. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	6.2°	5.6°	4.7°	4.9°	7.9°	7.4°
SW	8.1°	8.3°	14.3°	15.3°	8.0°	9.5°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

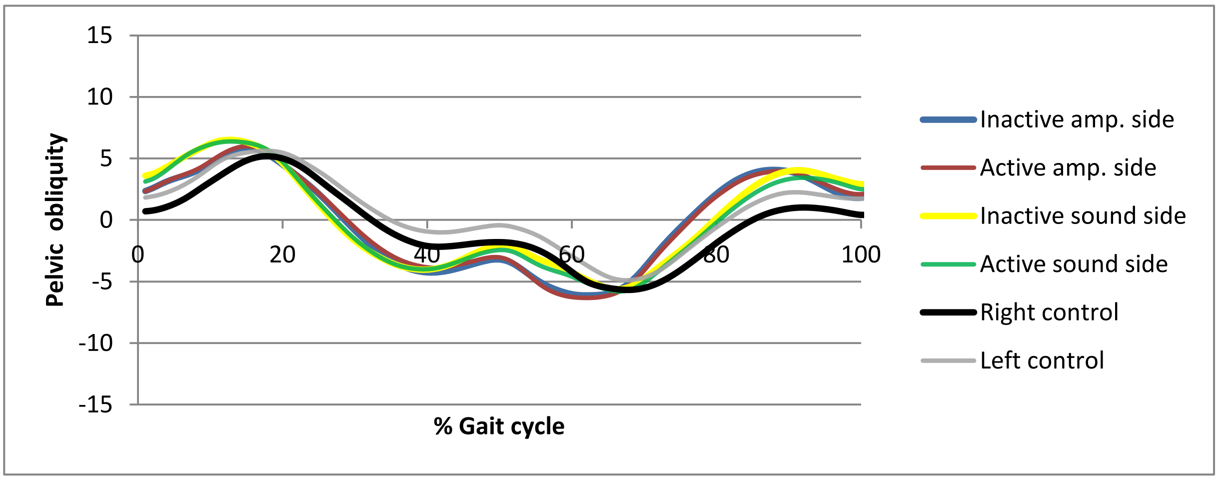

Figure 13. Mean pelvis kinematics in the frontal plane during incline walking.

Table 12. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed.

	Inact. Amp	Act. Amp	Inact. Sou	Act. Sou	Rctrl	Lctrl
WA	3.5°	3.7°	2.9°	3.3°	4.5°	3.8°
SW	10.2°	10.3°	9.7°	9.2°	6.7°	7.1°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation:

In comparison to level walking, larger excursions were found during incline walking, which is to be expected. During swing phase the interlimb and group differences in hip adduction was not as obvious as during level walking, although the sound side hip excursions were still the largest indicating a lateral trunk lean. A significant interaction (group by gait condition; p=0,001) was found for pelvic excursions, during SW due to a greater increase in excursions during incline walking compared to level walking seen in amputees (by 3,8° during inactive adaptive mode and 4,4° during active adaptive mode for the amptuees vs. 1.5° (Rctrl) -1,7° (Lctrl) in controls). Overall, hip excursions during SW were significantly greater during incline vs. level walking (by 2.7°; p=0,013). In addition, a limb by group interaction (p=0,010) reflected symmetry in hip excursions in controls (7.5° bilaterally) while the sound side hip demonstrated greater excursions than the contralateral side (14° vs. 8.5°).

4.3 Gait analysis - Two case studies

Here the two participants receiving training will be presented with a special focus on frontal plane kinematics of the hip and pelvis, and any differences found between inactive and active adaptive mode, before and after training. When interpreting these kinematic data the need for viewing hip kinematics within the context of pelvic kinematics, and vice versa, is of great importance.

In the current section the focus will be on changes for each individual over time across conditions. This is why the arrangements of graphs will be different than in section 4.2.2, where the groups were compared. There, a detailed interpretation of the movements may be seen. For the purpose of this analysis the focus will be on level walking. As a subsidiary analysis, EMG measurements were obtained from the participants receiving the training program to evaluate interlimb symmetries and possibly an effect of training. EMG activation will be presented as percentage of MVIC. Since the results presented here of the EMG activation levels are averaged activation of only three steps an interpretation of clinical significance is not applicable in this case, however this gives an insight of the individuals activation pattern.

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4.3.1 Case study; participant 2

Participant 2 is a 42 years old male, with a BMI of 22.8 and had, at the time of measurements, been amputated for 22 years, with the cause of amputation being trauma. He had 1 ½ years of experience walking on the Rheo knee but received the Proprio foot shortly before measurements. On average he used his prostheses for more than 15 hours every day, and on average he exercised for 10-14 hours every week.

A detailed training description may be found in appendix 1, including information regarding exercises, gait deviations identified and any changes made to the prosthetic components. Participant 2 had nine sessions with the physical therapist, in addition to home exercises.

4.3.1.1 Questionnaires

Detailed information regarding the questionnaires and specific tests are presented in appendices 4 and 5. In table 13 information is gathered from the questionnaire regarding general health and prosthesis satisfaction, the results of the „Timed up and go“ test (TUG) and results from the ABC scale. In table 14 the results from the pain rating scale are presented, which was a questionnaire regarding evaluation of pain in certain areas of the body, with the pain ratings based on the Numerical rating scale.

Table 13. General health and prosthesis satisfaction, „Timed up and go“, ABC

	Prosthesis functionality	General physical health	Vitality	Discour-aged by amputation	Self-rated security	Timed up and go (sec.)	ABC (%)
Before training	5	8	8	3	8	10.9	93
After training	3	7	9	2	9	9.5	89

ABC = Activities-specific Balance Confidence Scale

There were improvements in all questions in the general health and prosthesis satisfaction questionnaire. There was also a 12.8% improvement in performance during the TUG, but a 4% decrease in the rated self-efficacy, as measured with the ABC scale. Possible reasons for that could be the fact that the participant had received the Proprio foot only shortly before measurements and was potentially not yet fully accustomed to the new prosthetic foot. Another reason could be that because of possible gait alterations and prosthetic alignment changes during the course of the training period, he might not be as secure in various environments as he was before the training period.

Table 14. Pain rating

	Cervical spine	Thoracic spine	Lumbar spine	Hip sound side	Hip amputated side	Knee sound side
Before training	1	0	3	1	0	0
After training	0	0	2	0	0	0

Perceived pain decreased after the training period, in all of the areas that the individual felt pain at the beginning of the training period.

4.3.1.2 Temporal- spatial measurements

Table 15. Temporal-spatial measurements

		Step length Amputation side - meters	Step length Sound side - meters	Step width – meters	Double support-seconds
Before training	Inactive*	0.76 (0.015)	0.68 (0.011)	0.16 (0.009)	0.42 (0.017)
	Active*	0.75 (0.03)	0.66 (0.023)	0.16 (0.009)	0.43 (0.028)
After training	Inactive*	0.74 (0.013)	0.68 (0.038)	0.15 (0.013)	0.44 (0.023)
	Active*	0.75 (0.018)	0.65 (0.026)	0.15 (0.014)	0.44 (0.019)

* Inactive and Active = represents the setting of the mode of the prosthetic foot.

Step length on the amputated side decreased over time, from 0.76 m in the inactive mode before training to 0.75 m in the active mode after training. Step length on the sound side decreased as well from 0.68 m in the inactive mode before training to 0.65 m in the active mode after training. Step width decreased, with changes only seen after the training. Time in double support increased. Gait speed was not measured, but one of the focus points for this participant in training was to shorten his step length, especially on the amputated side, which could have led to a decreased walking speed, and a resulting increased double support time. See table 15 for detailed measurements.

4.3.1.3 Motion capture- gait analysis

As noted before the arrangement of the graphs and their data will be different from the ones in section 4.2.2. The graphs in figures 14 and 15 present the data for the hip kinematics in the frontal plane (adduction – abduction), for the amputated side and the sound side, respectively, for comparison of interlimb symmetry. In figures 16 and 17 the data for the pelvic kinematics in the frontal plane (pelvic obliquity) will be presented, also for the amputated side and the sound side, respectively. In each graph the series represent averaged data from the trials in all conditions, and in comparison averaged data from the control group will be presented. Interpretation of the graphs, collectively, will follow the pelvis graphs.

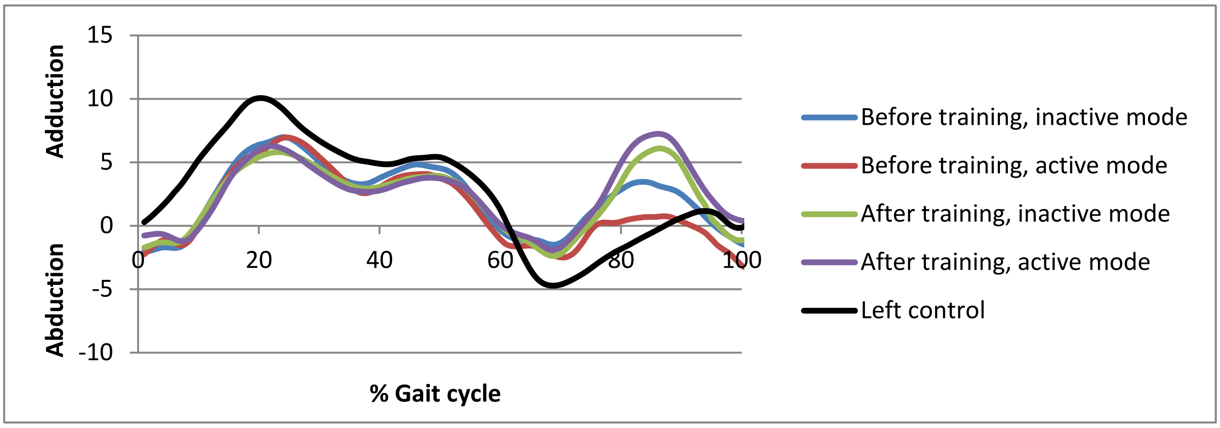

Figure 14. Mean hip kinematics, amputated side and control's left side, in the frontal plane during level walking, during all conditions.

Table 16. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Lctrl
WA	9.0°	9.2°	7.5°	7.5°	9.8°
SW	4.9°	3.2°	8.4°	9.1°	5.9°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)

SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

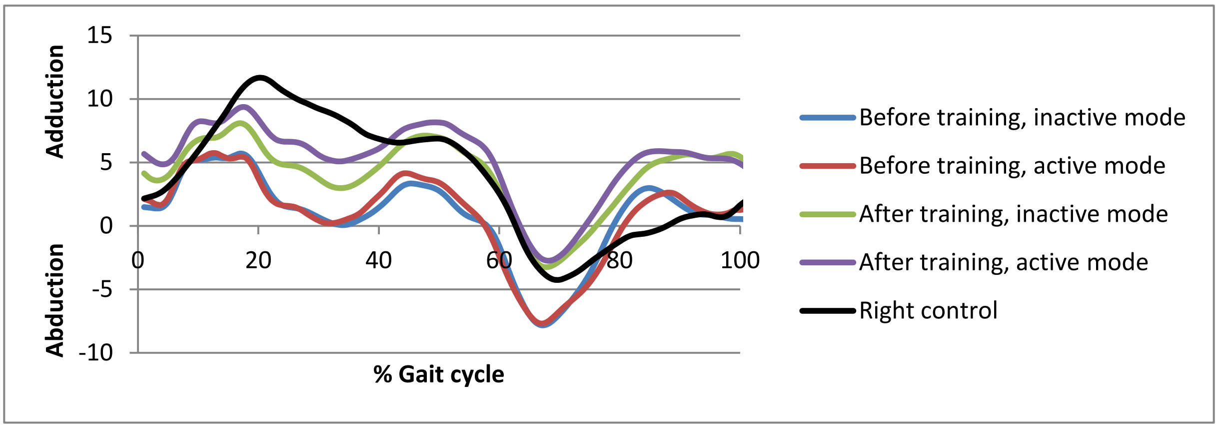

Figure 15. Mean hip kinematics, sound side and control's right side, in the frontal plane during level walking, during all conditions.

Table 17. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Rctrl
WA	4.3°	4.0°	4.5°	4.5°	9.5°
SW	10.8°	10.3°	8.9°	8.6°	6.2°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)

SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

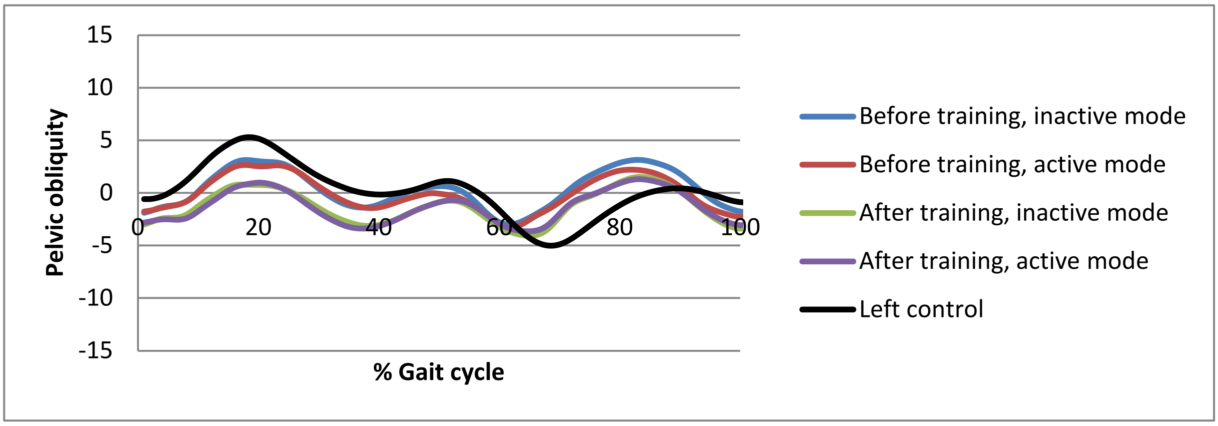

Figure 16. Mean pelvis kinematics, amputated side and control's left side, in the frontal plane during level walking, during all condition.

Table 18. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Lctrl
WA	5.0°	4.4°	3.8°	3.8°	5.9°
SW	6.1°	5.6°	5.6°	5.0°	5.4°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

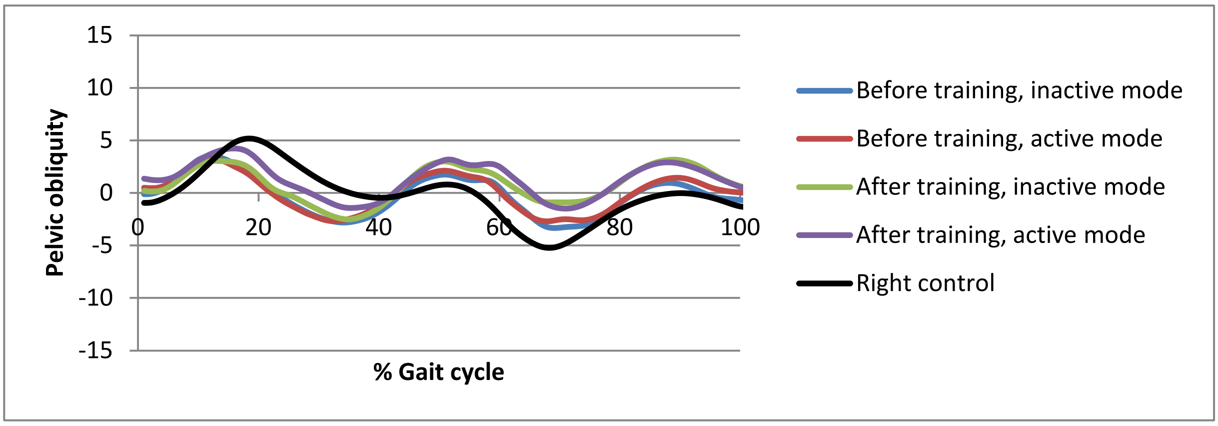

Figure 17. Mean pelvis kinematics, sound side and control's right side, in the frontal plane during level walking, during all condition.

Table 19. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Rctrl
WA	3.6°	2.7°	2.9°	3.0°	6.1°
SW	4.3°	4.1°	4.1°	4.4°	5.2°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation

Weight acceptance: Adduction excursions of the hip of the amputated side decreased from 9.2° during active adaptive mode before training to 7.5° during both conditions after training, while on the sound side the excursion ranged from 4.0°- 4.5° across conditions. Right and left hip excursions for the control group were 9.5° and 9.8° (table 16 and 17).

As hip adduction decreased, a concurrent decrease in pelvic movements was seen as well. Before training, the pelvic drop on the sound side (at IC on the amputated side) was 5.0° with the inactive adaptive mode and 4.4° with the active adaptive mode, and 3.8° after training for both conditions. For IC on the sound side, the pelvis drop on the amputated side ranged from 2.7° to 3.6° across conditions and for the control group the excursions were 5.9° - 6.1° (tables 18 and 19). These changes in hip and pelvic excursions after training suggest greater interlimb symmetry, although the difference between the amputation side hip and the control hip is now greater.

Swing phase: Before training the hip adduction excursions of the amputated side were 4.9° during the inactive adaptive mode and 3.2° during the active adaptive mode. After the training the hip adduction excursions were 8.4° during inactive adaptive mode and 9.1° during active adaptive mode. Meanwhile hip excursions on the sound side ranged from 8.6° to 10.8° across conditions, and for the control, the excursions were 5.9°- 6.2° (table 16 and 17). This indicates increased symmetry between sides and a decreased circumduction which is a common compensational movement among TFA.

For the pelvis the changes were not as extensive after training, but the pelvis lift on the amputated side went from 6.1° during inactive mode to 5.6° during active mode before training, and further down to 5.0° after the training with the active mode, while the sound side excursions ranged from 4.1°- 4.4° and among the control group the excursion was 5.2°-5.4° (table 18 and 19) When looking at interlimb symmetry during the swing phase of gait, greater symmetry was seen after the training. Hip adduction excursion of the amputated hip increased from 3.2° to 9.1, while adduction of the sound side decreased from 10.8 to 8.6. Meanwhile the participant‘s step width decreased from 0.163 m (before training, inactive mode) to 0.148 (after training, active mode). These increases in symmetry cannot be explained by changes in pelvic movements alone, although the pelvic lift on the amputated side did decrease to a value closer to both the sound side and the control group.

Discussions and clinical relevance

For this participant, several adjustments were made to the prosthetic components during the training period, of which a lengthening of 6 mm of the prosthetic leg was probably the most influential. Detailed information regarding other changes made is provided in appendix 1.

Changes observed after the training indicated improved interlimb symmetry. This was seen both for hip and pelvis movements in the frontal plane, with an interlimb difference of the hips being 7.1° before training and 0.5° after training, and the interlimb difference of the pelvis being 1.5° before training and 0.8° after training. There was an increase in adduction of the hip, but also a decrease in ipsilateral pelvic lift, which is not the expected kinematic pattern (decreased pelvis lift would generally be expected to be coupled with decreased hip adduction). This may indicate a positive effect of the active adaptive mode (decreased compensatory circumduction), in particular when combined with specific training

4.3.1.3 Electromyography

As a secondary analysis to the kinematic data, an EMG recording was done to gain insight into the activity of the pelvic muscles. The average activation of *m. Gluteus medius* and *m. Tensor fascia latae*, during stance phase of incline walking with active adaptive mode, is presented in figures 18 and 19, respectively. The signal was normalized to the signal obtained during MVIC. For comparison, data from one gender matched able-bodied participant in the study is presented as well.

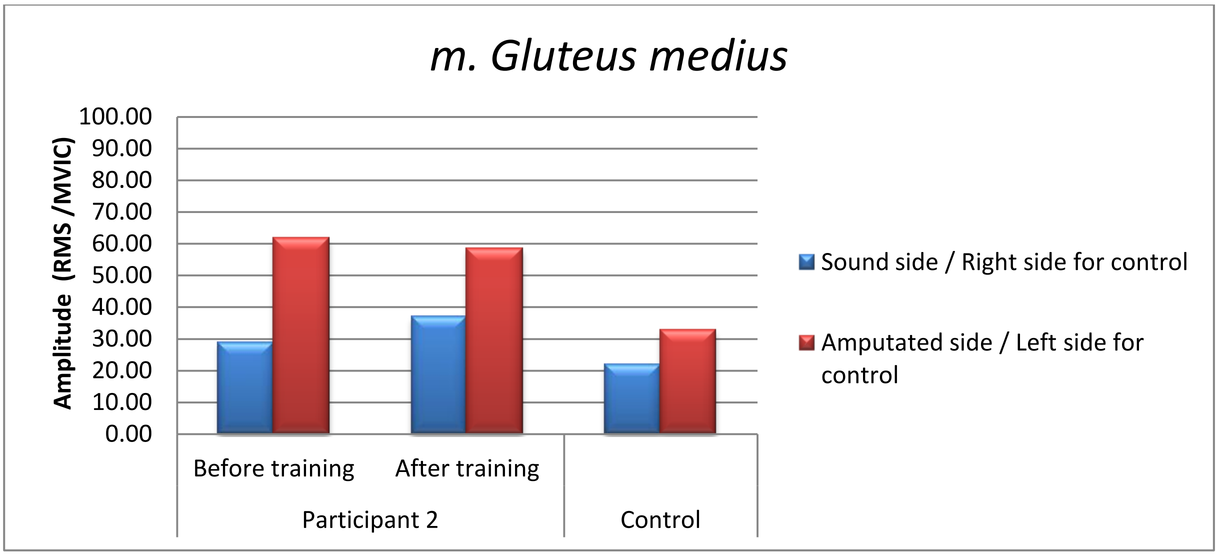

Figure 18. ***M. Gluteus medius*** **mean amplitude of the standardized RMS of EMG measurements**

The participant had relatively greater activation of *m. Gluteus medius* of the amputated side compared to the right side, both before and after training. There was also slight asymmetry seen in the signal of the control subject but not to the same extent, and there were generally less activation levels during stance. Improvements in interlimb symmetry were seen after training, with a slight decrease in activation on the amputated side and greater activation contralaterally.

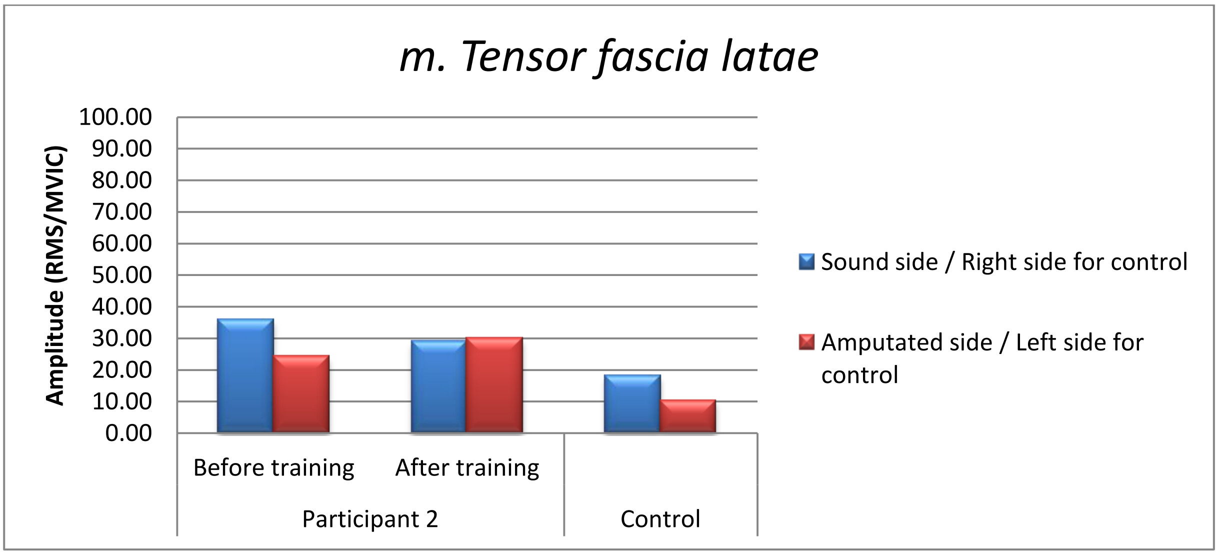

Figure 19. ***M. Tensor fascia latae*** **mean amplitude of the standardized RMS of EMG measurements**

Symmetry was also improved after the training in activation levels of *m. Tensor fascia latae*. For this individual the difference between sides was not as apparent for *m. Tensor fasica latae* as it was for *m. Gluteus medius*. The control subject generally demonstrated lower activation levels than the amputee, although interlimb differences were seen.

4.3.2 Case study; participant 3

Participant 3 is a 38 years old female, and at the start of the measurements she had a BMI of 24.3, but during the six weeks of training she lost 1.5 kg, so she had a BMI of 23.8 at the second measurement. Time of amputation was at the age of nine, with the cause of amputation being tumor. She had six years of experience walking on the Rheo knee but received the Proprio foot shortly before measurements. On average she used her prostheses for more than 15 hours every day, and on average she exercised for 0-4 hours every week, in the four weeks prior to the first measurement but more than 15 hours in the four weeks prior to the second measurement.

A detailed training description will be provided in appendix 1 with information regarding the exercises, gait deviation and changes made to the prosthetic components. Participant 3 had 11 sessions with the physical therapist, in addition to home exercises.

4.3.2.1 Questionnaires

Detailed information regarding the questionnaires and specific tests are presented in appendices 4 and 5. In table 20 information is gathered from the questionnaire regarding general health and prosthesis satisfaction, the results of the „Timed up and go“(TUG) test and results from the ABC scale. In table 21 are the results from the pain rating scale, which was a questionnaire regarding evaluation of pain in certain areas of the body, with the pain ratings based on the Numerical rating scale.

Table 20. General health and prosthesis satisfaction, „Timed up and go“, ABC

	Prosthesis functionality	General physical health	Vitality	Discour-aged by amputation	Self-rated security	Timed up and go (sec.)	ABC (%)
Before training	6	7	7	1	1	9.9	89
After training	9	8	9	0	2	11.7	98

ABC = Activities-specific Balance Confidence Scale

In the questionnaire of general health and satisfaction there were some changes seen. For prosthesis functionality, the participants’ satisfaction decreased substantially. The reason for this dissatisfaction was due to a loose socket because of a weight loss during the training period, and difficulties in adjustments to that. There were improvements in self-rating efficacy as measured with the ABC scale. However the time it took completing the TUG test increased 18% after the training, possibly because of the aforementioned problems with the prosthesis.

Table 21. Pain rating

	Cervical spine	Thoracic spine	Lumbar spine	Hip sound side	Hip amputated side	Knee sound side
Before training	2	2	3	2	0	2
After training	0	0	0	0	0	0

Before training the participant had perceived pain in all but one area in the list, but no pain was felt at the end of the training period.

4.3.2.2 Temporal – spatial measurements

Table 22. Temporal – spatial measurements

		Step length Amputation side - meters	Step length Sound side - meters	Step width - meters	Double support- seconds
Before training	Inactive*	0.80 (0.012)	0.73 (0.01)	0.14 (0.016)	0.36 (0.012)
	Active*	0.78 (0.029)	0.76 (0.02)	0.14 (0.018)	0.35 (0.023)
After training	Inactive*	0.83 (0.008)	0.74 (0.024)	0.13 (0.013)	0.43 (0.026)
	Active*	0.83 (0.018)	0.76 (0.025)	0.13 (0.012)	0.41 (0.024)

* Inactive and Active = represents the setting of the mode of the prosthetic foot.

The step length on the amputated side increased, and so did the step length on the sound side, despite an attempt during training to focus on decreasing the step length. Step width decreased after the training, but double support time increased.

4.3.2.2 Motion capture- gait analysis

As noted before the arrangement of the graphs and their data will be different from the ones in section 4.2.2. The graphs in figures 20 and 21, present the data for the hip kinematics in the frontal plane (adduction – abduction), for the amputated side and the sound side, respectively for comparison of an interlimb symmetry. In figures 22 and 23 the data for the pelvic kinematics in the frontal plane (pelvic obliquity) will be presented, also for the amputated side and the sound side, respectively. Each graph in the series represents the averaged data from the trials in all conditions, and in comparison, averaged data from the control group will be presented. Interpretation of the graphs collectively will follow the pelvis graphs.

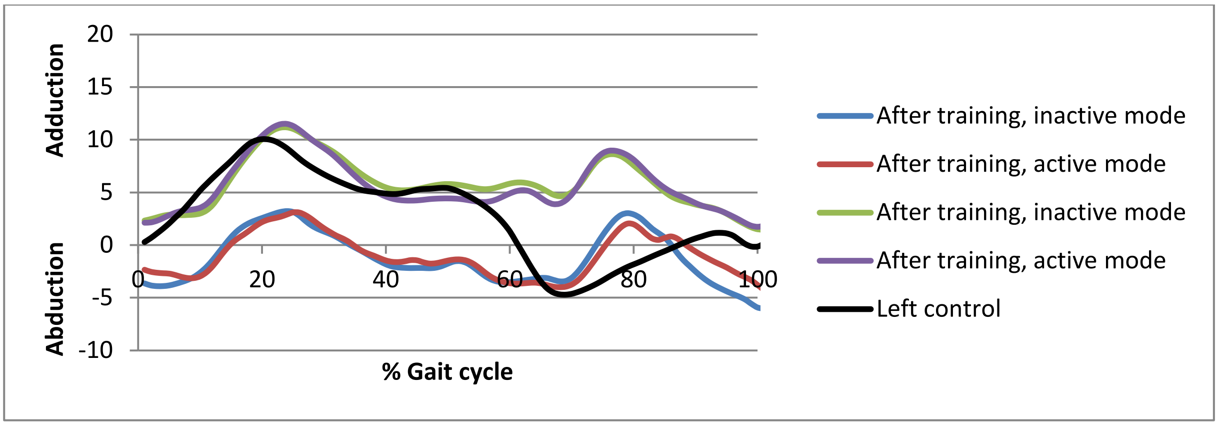

Figure 20. Mean hip kinematics, amputated side and control's left side, in the frontal plane during level walking, during all conditions.

Table 23. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Lctrl
WA	7.1°	6.2°	8.8°	9.4°	9.8°
SW	6.6°	6.0°	4.0°	5.1°	5.9°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

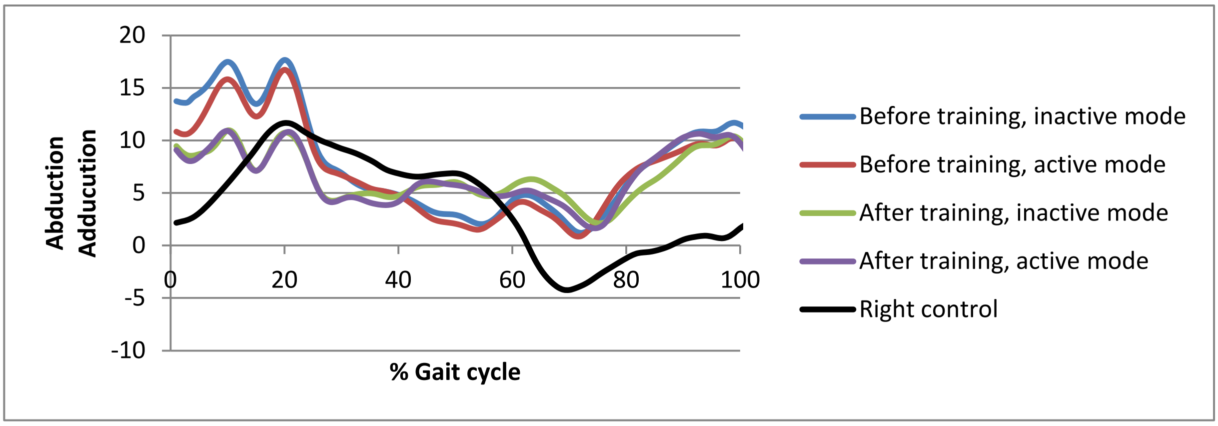

Figure 21. Mean hip kinematics, sound side and control's right side, in the frontal plane during level walking, during all conditions.

Table 24. Mean hip excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Rctrl
WA	4.1°	6.1°	2.4°	2.8°	9.5°
SW	10.5°	9.3°	8.3°	9.0°	6.2°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

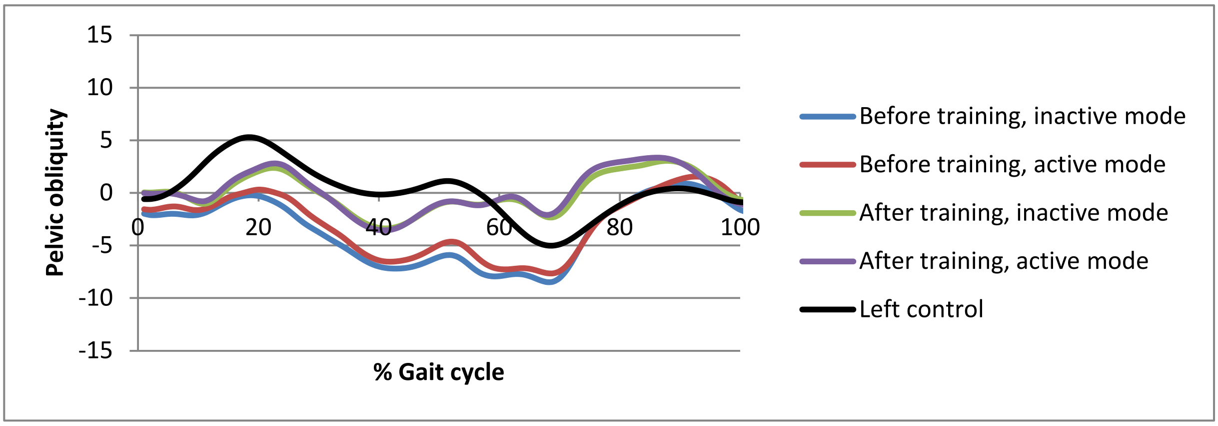

Figure 22. Mean pelvis kinematics, amputated side and control's left side, in the frontal plane during level walking, during all conditions.

Table 25. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Lctrl
WA	1.9°	1.9°	2.3°	2.9°	5.9°
SW	9.4°	9.2°	5.3°	5.4°	5.4°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

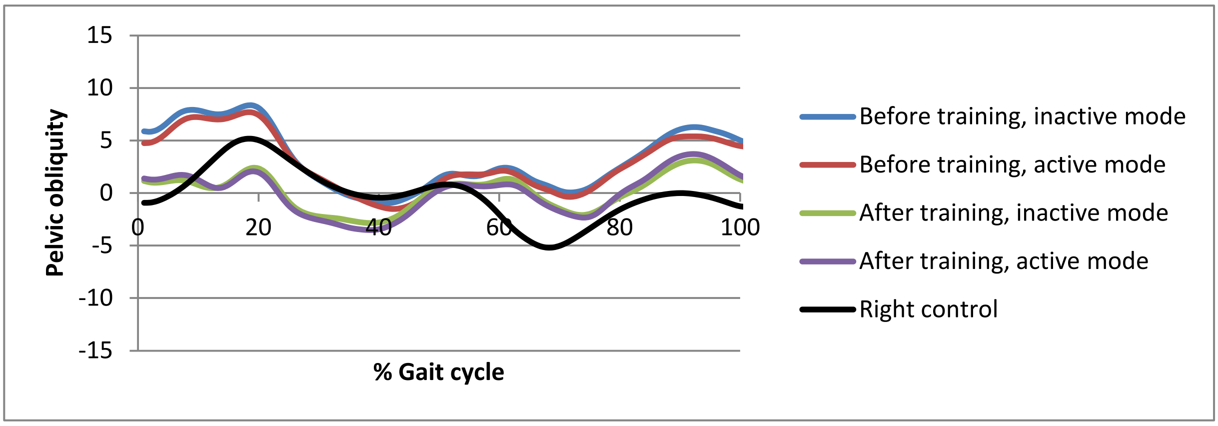

Figure 23. Mean pelvis kinematics, sound side and control's right side, in the frontal plane during level walking, during all conditions.

Table 26. Mean pelvis excursions in frontal plane during each part of the gait cycle analyzed.

	Before - Inact.	Before - Act.	After – Inact.	After – Act.	Rctrl
WA	2.5°	2.9°	1.4°	0.8°	6.1°
SW	6.2°	5.8°	5.2°	6.0°	5.2°

WA = weight acceptance (from IC to contralateral toe off, approximately 1%-20% of gait cycle)
SW = Swing phase (from toe off to ipsilateral heel strike, approximately 60%-100% of gait cycle)

Graph interpretation:

Weight acceptance: Adduction excursions of the hip of the amputated side increased from 7.1° during inactive adaptive mode before training to 9.4° during active adaptive mode after training, and on the sound side hip the excursion decreased from 4.1° during inactive adaptive mode before training to 2.8° during active adaptive mode after training. For the control group the excursions were 9.5°- 9.8° (tables 23 and 24).

Meanwhile the excursion of the pelvis at the IC of the amputated limb increased, from 1.9° during inactive adaptive mode before training, to 2.9° during active adaptive mode after training, and at the sound side IC the excursions of the pelvis decreased from 2.5° before training during inactive adaptive mode to a 0.8° after the training during active adaptive mode. For the control group the excursions were 5.9°- 6,1°(tables 25 and 26).

The movement pattern of the sound side hip differed considerably from the control hips, a pattern that remains after training. Possible reasons for this pattern have been discussed in the graph interpretations for the hip frontal plane movements being compensational mechanisms because of increased sound knee flexion and hip flexion during WA, or because of late flexion of the prosthetic knee in terminal stance. Corresponding to these hip movements patterns in the sound side hip were a different movement patterns for the pelvis, during sound side IC. The changes in excursion values during WA for both limbs suggest a decrease in the interlimb symmetry, both for the hip and the pelvis kinematics which are different results from what was seen for participant 2.

Swing phase: Adduction excursions of the amputated hip was decreased after the training period, from a 6.6° with the inactive adaptive mode before training to a 5.1° after the training during the active adaptive mode. For the sound side hip the excursions decreased from 10.5° adduction before training during the inactive adaptive mode to a 9.0° adduction after training during the active adaptive mode. Control hip adduction excursions were 5.9°- 6.2° (table 23 and 24)

Meanwhile, as might be expected in view of a normal kinematic pattern for the hip and the pelvis, the pelvis excursions decreased as well, from a 9.4° pelvic lift during the inactive adaptive mode before training to a 5.4° pelvic lift after the training during active adaptive mode on the amputated side. At the sound side there were minimal excursions changes observed or 6.2° before training during inactive adaptive mode and 6.0° after training during active adaptive mode. For the control group the excursions were 5.2°- 5.4° (table 25 and 26).

The movement pattern of the hip adduction during SW for the amputated limb was very different from the pattern seen for the control group, with a visually sharper rate of change from abduction to adduction right after the toe clears the ground and then again a sharp rate of change from adduction to abduction (figure 20). The pelvis did not exhibit the same amount of difference in pattern during the swing phase so these differences in the hip movement pattern are influenced by other aspects than the pelvic movements. During single limb support there was a lack of the normal abduction at the WA of the contralateral limb, a pattern seen both for the amputated side and sound side hip, which might be a result of a lack of weight shift between sides, which consequently had an effect on the kinematic pattern of the swing phase, on both sides. The kinematic pattern of the sound side hip and pelvis was closer to what can be seen by the control group, and with minimal changes in excursions.

During the swing phase there were no big changes regarding the interlimb symmetry for the hip but for the pelvis there was an increase in interlimb symmetry.

Discussions and clinical relevance:

For this participant the kinematic patterns were very different from the control group, and as well from the patterns seen for participant 2 in the previous case study. For the present case study there were changes in excursions that indicated a decrease in interlimb symmetry during the WA. During the swing phase, before training, there was quite a big asymmetry for the hip kinematics that remained after the training, so the changes seen in the pelvic kinematics, indicating an increase in interlimb symmetry, must reflect some other changes, not observed in the kinematic profiles here. In spite of this there was a decrease in the step width.

There are a few possible reasons for the different movement patterns seen for this participant. She was amputated at the age of 9, which could have influenced the growth of the residual limb and had an effect on surrounding anatomical structures. One obvious deviation in posture for this participant was an anteriorly tilted pelvic girdle, and more so on the side of amputation. This anatomical difference among others, quality of early rehabilitation and type of prosthesis in childhood could all possibly have affected the way the participant developed her gait through the years. As for reasons more modifiable, there was a problem with the socket for the duration of the training period, as it was too loose, and therefore did not support the remaining stump well enough. As the participant lost weight during the training period the socket became even bigger, causing a discomfort for the participant. It is likely that this might have had an influence on the outcome of measurements after the training.

4.3.1.3 Electromyography

As a secondary analysis to the kinematic data, an EMG recording was done, to gain insight into the activity of the pelvic muscles. In figures 24 and 25 the average activation of *m. Gluteus medius* and *m. Tensor fascia latae*, respectively, during stance phase of incline walking as a percentage of the MVIC, is presented. For comparison data from one gender- matched able-bodied participant in the study is presented as well.

It should be noted that because participant 3 had a much higher percentage of activation in *m. Gluteus medius* than participant 2, the axes on graphs for this participant are of a different size than for participant 2, for visual purposes.

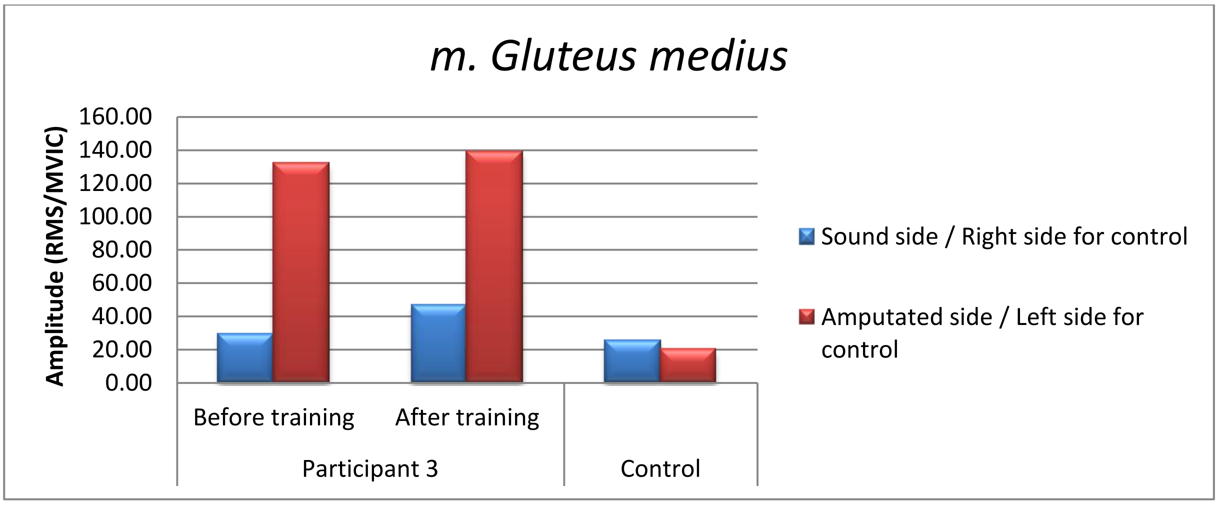

Figure 24. ***M. Gluteus medius*** **mean amplitude of the standardized RMS of EMG measurements**

What is most noticeable here is the very high percentage of increased activation of the *m. Gluteus medius* of the amputated side compared to the right side, or 132% and 139% of maximal voluntary contraction, before and after training, respectively, with barely no change in interlimb symmetry after training.

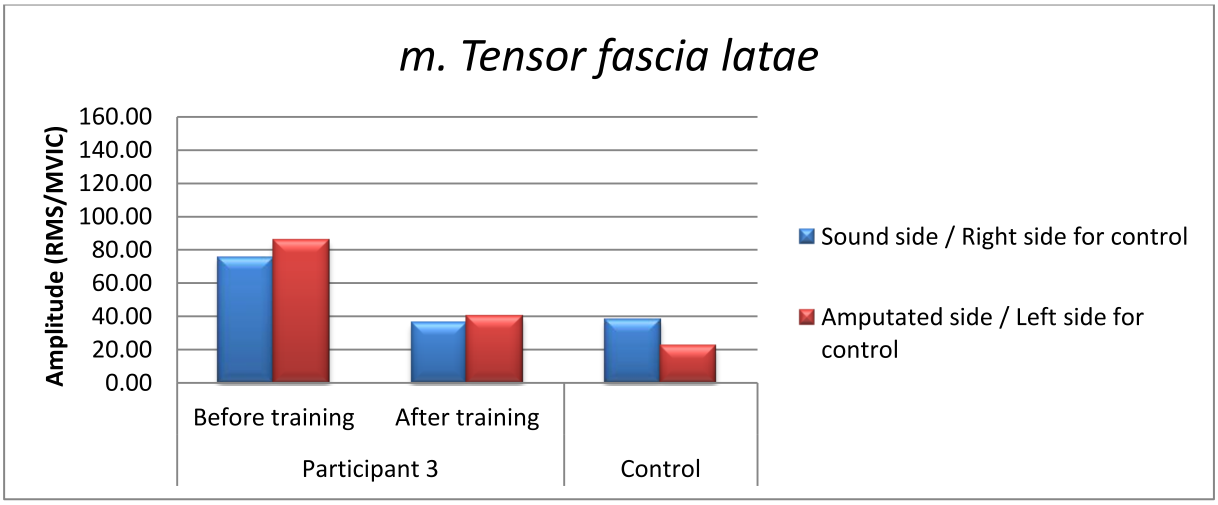

Figure 25. ***M. Tensor fascia latae*** **mean amplitude of the standardized RMS of EMG measurements**

For the *m. Tensor fascia latae* the activity of both sides decreased after training but the interlimb symmetry remained similar. Here the interlimb symmetry of the control is not as much as the amputees.

5. Summary and conclusion

The aim of this study was to evaluate effects of an adaptive mode of a microprocessor-controlled foot, on gait patterns of individuals amputated above knee, and the effects of individualized training on gait. Changes in various parameters were analyzed and presented in two parts of the study:

I. Gait analysis - comparison of kinematics and temporal-spatial parameters between groups and gait conditions, analyzing the effect of an adaptive mode.

II. Two case studies, analyzing the effect of training and adaptive mode on hip and pelvis kinematics and a subsidiary analysis of muscle activation patterns.

5.1 Gait analysis – Comparison between groups and conditions

The amputees' gait, in terms of temporal-spatial measurements, was characterized by a longer step length on the amputated side, compared to the sound side, while the control group had a near even step length. The amputees also had wider step width and a slightly longer double support time, compared to the control group. During transition from level walking to incline walking there was no change in the step length and step width for the amputees, compared to an increased step and decreased step width seen by the controls, possibly due to insecurity of the amputees while the base of support decreases with decreased step width.

Weight acceptance: During WA, or the first circa 20-25% of the gait cycle, for the amputated side, visually the rate of plantarflexion of the prosthetic ankle was slower when approaching foot flat, there was no flexion in the prosthetic knee and there was a visually sharp rate of change from flexion to extension on the hip of the amputated side, possibly, in part, because of the absent normal physiological response to load transfer at IC. Hip adduction on the amputated side was decreased, as was contralateral pelvic drop, at the WA of the amputated limb, when compared to the controls.

For the sound side the ankle had a movement pattern close to that demonstrated by controls, while knee and hip flexion excursion were greater than compared to controls, possibly because of the delayed knee flexion of the prosthetic side during terminal stance, although the flexion measures may also have been influenced by an anteriorly tilted pelvis during contralateral hip extension. Sound side hip adduction was decreased as well as the pelvic drop at the WA of the sound side limb, compared to the controls and the amputated side.

Swing phase: During the swing phase or the last approximately 60% of the gait cycle, the difference between the inactive and active mode of the prosthetic ankle is clear, as the ankle dorsiflexes during the swing phase with the active dorsiflexion. The pre-swing knee flexion of the prosthetic knee occurs later in the gait cycle than at the sound side knee and the control group knee. Bilateral hip flexion for amputees was greater than seen in controls. Adduction of the hip on the amputated side was considerably smaller compared to both the sound side and the controls, while the pelvis lift during swing was larger during amputation limb swing, compared to both the sound side and the controls.

For the sound side ankle dorsiflexion was greater during swing, compared to the controls. The knee excursions for the sound side were similar to that demonstrated by controls and as mentioned above, the hip flexion was greater than seen for the controls, bilaterally. Adduction of the sound side hip was considerably larger than seen for the amputated side and the controls but pelvis lift was similar across conditions and groups.

Through the mechanism of the kinetic chain that the body is exposed to in a functional task like gait, function and alignment of one segment is bound to have either an isolated or extensive effect on another segment, which then, subsequently as in a chain reaction, affects other segments or movements. Possible causes for the observed differences in the movement patterns were discussed mainly from a physical therapist's perspective. Prosthetic malalignments or other factors related to settings of the prosthetic components were not discussed in depth, but are recognized as important considerations, and hence the importance of cooperation of both professions – the physical therapist and the prosthetist. Proposed physiological/anatomical causes for compensational mechanisms seen by the amputee were; absence of a normal response of the prosthetic knee and foot at WA, the absence of plantarflexion muscle force of the prosthetic foot at toe clearance, an inability of the hip and pelvic musculature to control complex movements, as well as causes related to insecurity or instability of any kind. Possible causes related to the prosthetic components are, among others; the necessary restrictions caused by the essential mechanical stability of the prosthetic knee, too short prosthesis and/or socket fit.

5.2 Gait analysis - Two case studies

What stands out when the results of the two case studies are summarized, is the difference in the kinematic pattern among them regarding the hip and pelvis movements in the frontal plane. While the kinematic pattern of participant 2 gave good indications that both the active adaptive mode of the prosthetic foot and the specialized training had an effect leading to an increased interlimb symmetry, participant 3 exhibited different changes across condition, with an increased interlimb symmetry for the pelvis parameters in swing phase but decreased interlimb symmetry for the other parameters evaluated. The marked interlimb asymmetry for participant 3 were observed in EMG recordings, with a much higher activation amplitude for the *m. Gluteus medius* on the amputated side compared to the sound side. However, improvements were seen in the questionnaires and pain rating scales, for both participants, which indicate positive changes during the treatment period.

A factor that must be included, in a comparison of two individuals, is their clinical history. An example is the fact that participant 2 was an adult with a cause of amputation being trauma, whereas participant 3 was 9 years old when amputated because of a tumor, which inevitably has an effect on the way an individual develops a complex movement as gait, given the possible influence the amputation must have on normal development of a child´s growth.

The focus of the case studies was on the hip and pelvis kinematics. For participant 2 the increased adduction of the hip on the amputated side during swing phase, which led to an improved interlimb symmetry, could not be accounted for with increased pelvis lift, as might be expected when the normal hip-pelvis kinematic interaction is assumed. Rather a decrease in the pelvic lift was observed and this result highlights the fact that hip kinematics are interpreted more accurately if one is aware of concurrent pelvis motion. On the other hand, participant 3 demonstrated a decrease in the pelvic lift during swing phase, with no observable change in hip adduction over time. This demonstrates that in order to explain certain changes in joint angles, multiplanar observation of different joints and segments is important. Kinematics, as well as the kinetic factors, need to be considered in order to understand the different effect one movement can have on another.

Abductor strength was measured simultaneously to the MVIC measurements. For both amputees the abductor muscle group exhibited less strength on the amputated side. The activation pattern for participant 2 is different from participant 3, both in the percentage of MVIC and the proportional activation *of m.Gluteus medius* and *m. Tensor fascia latae*. demonstrate the different movement strategies between the amputees, probably caused by numbers of factors related to difference in motor control after amputation, muscle strength, gait technique, clinical history etc. This difference is however also noticeable for the control group, but not to the same extent. Most noticeable is the excessive activation of *m. Gluteus medius* for participant 3. This deviation might be due to the loose socket mentioned before, as there is less stability for the stump inside the socket, giving it less support, which might result in the increased activation of the hip musculature. Individuals with weaker gluteal muscles have been shown to exhibit greater gluteal activity, as measured by surface EMG, than those with stronger gluteal muscles (51), which is in accordance with the measurements of the current study.

5.3 Clinical implications - determinants of gait

The importance of a decrease in pelvis lift (or increased pelvic drop), as was seen for both participants after the training in the case studies, does not only relate to the interlimb symmetry. *Pelvic drop* is one of the six determinants of gait, proposed by Saunders (1953) which represent the adjustments made by the pelvis, hips, knees and ankles that keep the movement of the center of mass to a minimum, and therefore keeping the energy expenditure to a minimum (52). As oxygen consumption has been shown to be increased among TFA (35) a decrease in pelvic lift seems an important parameter to address in the rehabilitation of an amputee.

Another gait determinant is the *knee flexion* during stance phase, which is also a parameter which deviates substantially among TFA compared to able-bodied, as can be seen in section 4.2 of this thesis, and is largely influenced by the type of prosthetic knee, or, in the case of the microprocessor-controlled knee that all participants in this study used, the ability of the amputee to fully utilize the functional properties of the knee. In section 4.2 of the thesis, it was reported that only one participant of the study exhibited a slight knee flexion at WA. This participant did not receive a training program, based on the findings of the physical examination and the fact that he had a very good gait technique, was an elite athlete, and was experienced in walking with different prosthetic components. Of the five amputee participants, this participant was the only one demonstrating an ability to utilize this particular prosthetic knee’s function of allowing a certain stance knee flexion. This has obvious implications for other joint motions and may possibly affect energy expenditure, as knee flexion is one of the determinants of gait, hence the importance of sufficient gait training when receiving a sophisticated prosthetic knee like a microprocessor knee joint.

Other determinants of gait are; *pelvic rotation*, not examined in the current study but a large factor in the training protocol; *Foot and knee interactions*, for example the normal rapid plantarflexion at WA, associated with the initiation of knee flexion, an interaction which contributes to maintain the center of mass. This interaction is seen to be deviant in the gait of TFA, likely associated to a necessary restriction of the prosthetic foot, in order to avoid a buckle of the prosthetic knee. Lastly, the *lateral displacement of the pelvis*, controlled in a way with a physiological valgus of the knee, or a tibiofemoral angle, which is absent in the prosthetic knee and ankle. Therefore, being able to positively influence, either with training or type of prosthetic components, modifiable determinants of gait must be considered a desirable result with respect to energy consumption, but also considering interlimb symmetry, discussed in the next section.

5.4 Gait - research methods

In order to achieve successful forward progression during walking, a precise control of acceleration and deceleration during the limb advancement is needed, while maintaining sufficient stability on the weight-bearing limb. Able-bodied individuals rely on well coordinated muscle activation patterns and intact joint structures of the lower limbs, factors that are partly absent for the TFA, so some kind of compensations are needed to complete a successful stride. As described in this thesis, amputees exhibit numerous differences in joint kinematics compared to able-bodied individuals, in addition to interlimb differences. Gait, in its largest context, is often very individualized. However, the main characteristic of normal gait is its obvious symmetry between the two body sides, and a deviation from that pattern would in cases be considered a dysfunction, even pathological to some extent. This is why, in gait rehabilitation of any kind, the obvious goal is to achieve symmetry whenever possible. In the kinematic data presented in this study, especially in the two case studies, large differences were seen between the TFA and an able-bodied individual. As described before in the thesis, after an amputation, the individual does not have the same anatomical prerequisites as the able-bodied individual to complete, for example, the task of walking. So from a clinical point of view, when assessing and training the gait of a TFA, interlimb symmetry may seem to some extent be regarded as an optimal end result rather than trying to achieve a movement pattern that has been defined as “normal”. This is a topic that is probably debatable, as can be seen by the numerous studies applied to compare a movement pattern seen in an able-bodied individual to a movement pattern seen in a pathological condition. Indeed, a broad knowledge of what is considered normal is essential, in order to understand a function that in any way deviates from the norm.

Clinical gait analysis, as described in this study is an efficient way of retrieving detailed information regarding various movement patterns, which is valuable when it comes to designing or improving prosthetic components, analyzing gait deviations and designing treatment plans. However it is evident, that the way a TFA compensates for a given impairment can be very different between two persons, caused by very different reasons, in terms of hip and pelvis kinematic patterns. Therefore further research with bigger cohorts seems to be warranted, with the aim of examining the variability of various kinematic patterns among TFA, in order to be able to interpret the results of different studies.

5.5 Study limitations

The small number of participants is an obvious limitation of this study, which does not allow strong statistical conclusions. However, when working with such a specific population as is done in this study, individual differences become apparent. This kind of methodology may therefore also be considered of importance, in order to demonstrate differences in compensatory mechanisms between individuals, reflecting diverse kinematic solutions.

Another limitation to the study is the way the treatment was implemented, that is, the individualized approach to the treatment plan, rather than a standardized treatment protocol. This might have been a greater limitation in another methodological setting, but as this part of this study was presented as two case studies, the ability to identify factors that could be relevant to the differences seen between the two measurements is greater. Major factors thought to have an effect during the training period were muscle strength improvements, technical improvements related to the gait training, improved body awareness, alignment of prosthetic components and weight loss seen by participant 3. To what extent each and every factor affected the differences seen between the two measurements is unknown, but this reflects a typical clinical setting, where the cooperation between health care professionals like physical therapists, prosthetists and physicians is of great importance.

The participants receiving the training were given approximately one day to get accustomed to the new microprocessor controlled prosthetic ankle. This might be considered a limitation to the study, but this provided insight to the immediate effect of having an active prosthetic ankle versus the effect of training. An interesting additional comparison would have been to have yet another group also receiving the same prosthetic ankle for the first time but no training.

In conclusion, as is evident by the analysis of the two case studies, individual differences among TFA can be marked, and must be taken into account in gait analysis. Active dorsiflexion of a prosthetic foot, as well as a 6 week training program can have positive effects on the interlimb symmetry for TFA regarding the frontal plane kinematics of the hip and pelvis. No significant changes were seen between the active and inactive adaptive mode of the prosthetic foot on the parameters analyzed. However, various differences in the kinematic patterns were observed between able-bodied individuals and the amputated side and sound side of the TFA participants. The outcome measures best suited for assessment of TFA ideal gait have yet to be identified, as is evident in light of numerous studies with different parameters examined and sometimes different ways of data analysis and interpretation. This elucidates the need for ongoing studies with multiplanar kinematic and kinetic analysis of different joints in order to acquire knowledge of TFA gait deviations of clinical importance. That being said, one must not forget the subjective evaluation of the amputee during different stages of training and prosthetic fitting. Therefore, a comprehensive and multidisciplinary approach in the rehabilitation setting is of importance. A major trauma like amputation alters a person’s life in many ways, from physical and psychological aspects, to social and economical aspects. In this paper the physical aspect was dealt with and analyzed in part, but all of these aspects must play different roles in the way persons carry themselves following an amputation.

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Fylgiskjöl

Appendix 1: Training protocol

Appendix 2: Kynningar- og upplýsingabréf

Appendix 3: Upplýst samþykki

Appendix 4: Spurningarlisti - Lífsgæði, þægindi gerviliða, öryggi.

Appendix 5: The Activities-specific Balance Confidence Scale – The ABC scale

Appendix 1: Training protocol

After physical examination it was concluded that four participants would possibly benefit from physical therapy. Following is a description of the characteristics, main findings of the physical examination and daily notes for the physical therapy, for those participants that finished the study. Therapy sessions were based on the findings of the physical examination, where emphasis was put on deficiencies such as muscle weakness or gait deviations, as well as on basic routine exercises that were performed every session, consisting of functional training such as weight transfer exercises of the center of mass(COM) over the base of support (BOS), both on the floor as well as on a balance mat, transfer and control of the sound leg up to a stool, and strengthening exercises such as exercises with an elastic band and exercises on a mat.

Participant no. 2

Physical examination:

Range of motion: OK for all joints tested, Thomast test left side 180°. Hip extesion 15° bilaterally.

Strength: 5+ for the muscles tested, although endurance was not good

Balance: Not good for the left (amputated leg)

Posture: Left crista iliaca ca. 5 mm lower. A small anterior tilt on the pelvis on the left side. Left scapula slightly lower. Head forward tilted

Gait:

1. Step width: Abducted stride at heel strike with prostheic leg. (Possible causes: imbalance on the prosthetic leg, habit, too much weight on the sound leg, too weak muscles on the amputated side)
2. Step length: Longer stride on prosthetic side.
3. Toe load: OK
4. Knee flexion: Decreased knee flexion on the prosthetic side.
5. Pelvic rotation: Decreased pelvic rotation
6. Trunk rotation: Trunk rotation OK. Lateral lean over the prosthetic leg at stance phase. Too much outward motion of the left hand.

Daily notes:

Session 1 (28/3)

- Physical examination
- Home exercises taught

Session 2 (30/3)

- Overview of home exercises
- The basics of the Center of Mass (COM) and Base of Support (BOS) and exercises related to that.
- More home exercises

Session 3: (4/4)

- Overview of home exercises
- Exercises at walking bars- stool stepping exercise
- Exercises on mat with an exercise ball
- First gait training guidelines: Shorten the step with the amputated limb, started working on the pelvic rotation, talk about the toeload.

Session 4 (11/4)

- Overview of home exercises
- Basic routine exercises
- Extra exercises on the mat, with an elastic band around the knees
- Squats with support from the hands (on a climbing ladder)
- Gait training: pelvic rotation exercises with an elastic band and notes on the right hand movement.

Session 5 (18/4)

- Overview of home exercises
- Basic routine exercises
- Gait training: notes on the step length of both the right and left foot, gait training with the elastic band, notes on the movement of the thorax, notes on the pressure of the toe at the end of stance phase of the amputated leg.
- Extra exercises with the exercise ball
- Exercises that put too much pressure on the stump, like when the weight is fully on the amputated leg in a standing position, gives him pain in the stump.
- Notes on how to notice the muscles working in the socket, how the body weight transfers over the amputated leg

Session 6 (20/4)

- Unable to come

Session 7 (25/4)

- Overview of home exercises
- Basic routine exercises
- Lunges
- Gait training, with emphasis on the notes and guidelines given before
- **Prosthetic change: 2° lateral rotation of the ankle joint**

Session 8 (27/4)

- Unable to come

Session 9 (2/5)

- Overview of home exercises
- Basic routine exercises
- **Prosthetic change: lateral rotation of the knee joint and medial rotation of the ankle joint. 4 mm increase in height**

Session 10 (4/5)

- Overview of home exercises
- Basic routine exercises
- Gait training, with emphasis on the notes and guidelines given before

Session 11 (9/5)

- Overview of home exercises
- Basic routine exercises
- Gait training, with emphasis on the notes and guidelines given before
- **Prosthetic change: 2 mm increase in height: All in all a 6 mm increase in height**

Session 12 (11/5)

- Unable to come

Participant no. 3

Physical examination:

Range of motion: OK for all joints tested. Thomas test 170°. Hip extension 20° bilateral

Strength:5+ for all muscles tested. Decreased endurance for left hip musculature.

Balance: Not good for the left (amputated leg)

Posture: Left crista iliaca ca 3 mm lower. Anterior tilt of the pelvis on left side, ca 3-5 mm. Left scapula lower, ca 2 mm. ASIS left side lower when lying supine. Excessive lumbar lordosis and knee hyperextesion. Scoliosis of the spine.

Gait:

1. Step width:OK
2. Step length: Longer stride with the prosthetic leg. Lateral lean over the prosthetic leg (Possible causes: improper lateral wall support of the socket)
3. Toe load: Load OK, lateral whip of the heel.
4. Knee flexion:Decreased knee flexion
5. Pelvic rotation:Decreased pelvic rotation. Excessive anterior tilt of the pelvis on the left side.
6. Trunk rotation: Excessive rotation to the right of the spine at stride phase of the left leg (possible causes: improper lateral wall support of the socket, weak abduction strength)

Daily notes:

Session 1 (28/3)

- Physical examination

- Home exercises taught

Session 2 (30/3)

- Unable to come

Session 3: (4/4)

- Overview of home exercises
- The basics of the Center of Mass (COM) and Base of Support (BOS) and exercises related to that.
- More home exercises
- Exercises at walking bars- stool stepping exercise
- Exercises with the elastic band
- Exercises on mat with an exercise ball
- First gait training guidelines: Shorten the step with the amputated limb, started working on the pelvic rotation and notes on the anterior tilt of the pelvis- talk about the importance of the core muscles in relation to the anterior tilt.

Session 4 (11/4)

- Overview of home exercises
- Basic routine exercises
- More home exercises
- Lunges- focus on the anterior tilt

Session 5 (18/4)

- Overview of home exercises
- Basic routine exercises
- Extra exercises on the mat
- Extra exercises with the weigh transfer
- Squats with support from the hands (on a climbing ladder)
- Gait training: pelvic rotation exercises with an elastic band and notes on the toe load.
- Stair stepping exercises
- Extra gluteus medius exercises on a brick
- More home exercises- Goes to the gym 3-4 times a week, where she finishes the exercises

Session 6 (20/4)

- Overview of home exercises
- Basic routine exercises
- Similar training as last time
- More notes on the core muscles. Notes on how to notice the muscles working in the socket, how the body weight transfers over the amputated leg

- **Prosthetic change: Increased valgus (3/4)**

Session 7 (25/4)

- Overview of home exercises
- Basic routine exercises
- Lunges and step-up exercises, focus on the core muscles to prevent the anterior tilt of the pelvis
- Gait training, with emphasis on the notes and guidelines given before

Session 8 (27/4)

- Overview of home exercises
- Basic routine exercises
- More gluteus medius exercises- sidelying and standing
- Gait training, with emphasis on the notes and guidelines given before

Session 9 (2/5)

- Overview of home exercises
- Basic routine exercises
- Gait training, with emphasis on the notes and guidelines given before

Session 10 (4/5)

- Overview of home exercises
- Basic routine exercises
- Gluteal exercises
- Always difficult to do the abduction exercises with the elastic band
- Noted that when walking on the treadmill with very little support with the right hand on the bars the rotation of the spine decreased.
- Gait training, with emphasis on the notes and guidelines given before
- **Prosthetic change: Pads in the socket, due to loose socket**

Session 11 (9/5)

- Overview of home exercises
- Basic routine exercises
- Gait training, with emphasis on the notes and guidelines given before
- Felt better with the pads (has lost weight during this study so the socket is too big for her now)

Session 12 (11/5)

- Overview of home exercises
- Basic routine exercises
- Static lunges

Appendix 2: Kynningar og upplýsingabréf

Hreyfigreining göngu og færnimiðaðra hreyfinga hjá einstaklingum aflimuðum fyrir ofan hné. Samanburður á tveimur mismunandi stillingum á tölvustýrðum gervifæti bæði fyrir og eftir sérhæfða þjálfun.

Kynningar-og upplýsingabréf

Með þessu kynningar- og upplýsingabréfi óskum við eftir þátttöku þinni í rannsókn á ofangreindri rannsókn okkar, sem er meistaraverkefni við Læknadeild Háskóla Íslands, í samstarfi við Össur stoðtækjafyrirtæki.

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Markmið rannsóknar: Markmið rannsóknarinnar er tvíþætt; annars vegar að skoða áhrif mismunandi stillinga á tölvustýrðum gervifæti (stífur (óvirkur) ökkli / rafknúinn (virkur) ökkli) á hreyfimynstur og vöðvavirkni við göngu og færnimiðaðar hreyfingar, og hins vegar að skoða áhrif sérhæfðar þjálfunar á sömu þætti.

Þátttakendur: Þátttakendur í rannsókninni verða einstaklingar sem eru aflimaðir fyrir ofan hné og eru skjólstæðingar Össur stoðtækjafyrirtækis (rannsóknarhópur). Í rannsókninni verður notað Rheo gervihnéð og Proprio gervifóturinn. Þessir tveir gerviliðir eru tölvustýrðir. Þeir þátttakendur sem ekki þá þegar nota þessar vörur munu fá þær til afnota a.m.k. 4 vikum áður en að mælingar og þjálfun hefst og fram að lokum rannsóknar. Einnig verður gerð ein mæling á sambærilegum hóp einstaklinga með svipuð sérkenni (þ.e. kyn, aldur, hæð og þyngd), sem ekki hafa misst ganglim og hafa engin vandamál sem hafa áhrif á göngu, til samanburðar (viðmiðunarhópur).

Framkvæmd: Mælingar og þjálfun mun fara fram að Grjóthálsi 1-5, í húsakynnum stoðtækjaþjónustu Össurar. Mælingarnar verða tvær talsins hjá rannsóknarhóp og ein hjá viðmiðunarhóp. Skoðaðar verða hreyfingar í mjöðmum, mjaðmagrind, baki og efri búk með þrívíddar-myndatökubúnaði, þar sem myndatakan fylgist með sérstökum endurskinskúlum sem, við úrvinnslu gagna, meta afstöðu líkamshluta í hreyfingunni. Einnig verður mæld vöðvavirkni ýmissa vöðva í kringum mjaðmagrind og á baki með yfirborðs-vöðvarafritsmælum til þess að sjá hvernig vöðvar bregðast við virkum ökkla sem þessum sem notaður er í rannsókninni. Vöðvarafritsmælarnir eru límdir á húð, sem og flestar endurskinskúlurnar, en sumar kúlurnar verða límdar beint á fatnað eða skó. Því er ráðlagður klæðnaður í rannsókninni stuttbuxur og stuttermabolur.

Í fyrstu komu fá allir þátttakendur nákvæma skoðun hjá sjúkraþjálfara. Byggt á þeirri skoðun setur sjúkraþjálfari upp þjálfunarprógramm fyrir þátttakendur í rannsóknarhópnum, en þátttakendur í viðmiðunarhópi fá upplýsingar um eigin liðleika, vöðvastyrk og vöðvalengd. Þátttakendur í rannsóknarhóp svara einnig spurningalista varðandi almenna færni og stoðkerfisverki. Fyrri mæling rannsóknarhópsins er gerð þennan dag og framkvæma þátttakendur eftirfarandi atriði; a) ganga á þægilegum hraða, b) ganga upp og niður halla og c) ganga yfir hindrun. Rannsóknarhópur gerir þessi atriði bæði með slökkt og kveikt á tölvustýringu á gervifæti. Seinni mæling hjá rannsóknarhóp fer fram eftir 6 vikna æfingatímabil, þar sem einstaklingar gera bæði æfingar heima við og hitta sjúkraþjálfara 1-3x í viku, eftir þörfum. Hver tími með sjúkraþjálfara mun taka um 30 – 45 mínútur. Fyrri mæling og skoðun mun taka 1-2 klst. og seinni mæling um 1 klst. Mæling viðmiðunarhópsins verður gerð á þessu 6 vikna tímabili. Stefnt er að því að hefja rannsóknina í febrúar/mars 2012.

Ávinningur/ áhætta af þátttöku: Ávinningur þátttakenda er fólgin í því að allir fá nákvæma skoðun frá sjúkraþjálfara sem metur þá þætti sem gætu verið að hamla einstaklingnum í göngu eða færni og munu þátttakendur í rannsóknarhóp í framhaldi af þeirri skoðun fá einstaklingsmiðaða sjúkraþjálfun sem fer fram bæði með sjúkraþjálfara og heima við. Ekki er talin vera nein áhætta af rannsókninni þar sem mælingar eru gerðar í umhverfi sem er öruggt og þjálfun undir eftirliti sjúkraþjálfara. Sú þjálfun sem fer fram heima við eru æfingar sem þátttakandur munu fara fyrst yfir með sjúkraþjálfara og þær ekki gerðar nema sýnt sé að einstaklingur ráði fullkomlega við þær án aðstoðar. Sumir þátttakendur munu þurfa að aðlagast nýjum gervilið fyrir rannsóknina, og komi upp vafi um að sá gerviliður sem viðkomandi fær henti þátttakanda, verður viðkomandi dregin úr rannsókninni. Þeir sem eru með sérlega viðkvæma húð gætu fengið roða í húð eftir lím á elektróðum og endurskinskúlum, en slíkt er afar sjaldgæft. Þátttaka í verkefninu er án endurgjalds.

Tryggingar og trúnaður: Til að gæta persónuöryggis verða allar mælingar nafnlausar og mun hver þátttakandi fá handahófskennt númer sem notað verður við úrvinnslu gagna. Nöfn eða aðrar persónugreinanlegar upplýsingar munu hvergi koma fram. Við þrívíddarmyndatöku verða settar endurskinskúlur á ákveðna staði á líkamanum og nemur myndatökubúnaðurinn aðeins það endurskin en engin persónugreinanleg gögn. Yfirborðs-vöðvarafritsmælar nema einungis virkni vöðva. Spurningalistar verða geymdir í læstum skápum og tölvugögn í tölvum læstum með lykilorðum. Eftir að niðurstöður hafa verið birtar verður öllum rannsóknargögnum eytt. Meðan á rannsókninni stendur munu rannsakendur annast eftirlit með heilsu og líðan þátttakenda í rannsókninni. Eins og komið hefur fram er áhætta í rannsókninni hverfandi, en þó verða þátttakendur tryggðir fyrir skakkaföllum sem gætu orðið ef þeir t.d. hrasa í göngu.

Vísindalegt gildi rannsóknarinnar: Ekki er ljóst hvort eða hversu vel tölvustýrður ökkli eins og Proprio fóturinn nýtist þeim einstaklingum sem aflimaðir hafa verið fyrir ofan hné, hvað varðar vöðvavinnu og hreyfingar um ganglimi og mjóbak. Auk þess að skoða hreyfimynstur þeirra sem ganga með bæði kveikt og slökkt á tölvustýringu gervifótarins, verða einnig skoðuð áhrif sértækrar þjálfunar á þessa þætti. Það er von rannsakenda að með því að skilja vel virkni vöðva og hreyfingu liða í daglegum athöfnum aflimaðra einstaklinga við mismunandi aðstæður (virkur ökkli/óvirkur ökkli), geti fagaðilar, sem annaðhvort sjá um þjálfun aflimaðra einstaklinga eða um hönnun gervifóta-og hnjáa, nýtt sér þá þekkingu til að tryggja bestu mögulegu útkomu fyrir notendur hvað varðar færni til athafna daglegs lífs, og þar með aukin lífsgæði. Sótt hefur verið um leyfi til Vísindasiðanefndar og var rannsóknin einnig tilkynnt til Persónuverndar.

Upplýst samþykki þarf að undirrita við upphaf rannsóknarinnar. Væntanlegum þátttakendum er frjálst að hafna þátttöku eða hætta í rannsókninni á hvaða stigi sem er, án útskýringa og geta þeir hvenær sem er snúið sér til ábyrgðarmanns, rannsakanda, eða Vísindasiðanefndar hafi þeir einhverjar spurningar.

Með fyrirfram þökk;

Kristín Briem, ábyrgðarmaður

Anna Lára Ármannsdóttir, sjúkraþjálfari

Appendix 3: Upplýst samþykki

Hreyfigreining göngu og færnimiðaðra hreyfinga hjá einstaklingum aflimuðum fyrir ofan hné. Samanburður á tveimur mismunandi stillingum á tölvustýrðum gervifæti bæði fyrir og eftir sérhæfða þjálfun.

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Með undirskrift samþykkir undirritaður að taka þátt í rannsókn þar sem að skoðuð verða annars vegar áhrif mismunandi stillinga á tölvustýrðum gervifæti (stífur (óvirkur) ökkli / rafknúinn (virkur) ökkli) á hreyfimunstur og vöðvavirkni við göngu og færnimiðaðar hreyfingar, og hins vegar áhrif sérhæfðar þjálfunar á sömu þætti. Undirritaður staðfestir hér með undirskrift sinni að hafa lesið þær upplýsingar um rannóknina sem honum voru afhentar, og fengið fullnægjandi svör og útskýringar á einstökum þáttum hennar, og geri sér grein fyrir hlutverki sínu í rannsókninni.

Undrrituðum verið skýrt frá fyrirkomulagi trygginga fyrir þátttakendur rannsóknarinnar. Upplýsingabréf og samþykki fyrir þessari rannsókn eru í tvíriti og þátttakandi mun halda eftir eintaki af hvoru tveggja.

<u>Undirritaður tekur þátt í rannsókninni af fúsum og frjálsum vilja og er ljóst að þó hann hafi skrifað undir þá geti hann hætt þátttöku hvenær sem er, án útskýringa og án þess að sú ákvörðun hafi nokkur áhrif á þá þjónustu sem hann á rétt á.</u> Undirrituðum er ljóst að öllum gögnum verður eytt að rannsókninni lokinni.

______________________________ ______________________________

Staður og dagsetning Nafn þátttakanda

Undirritaður, stafsmaður rannsóknarinnar, staðfestir hér með að hafa veitt upplýsingar um eðli og tilgang hennar, í samræmi við lög og reglur um vísindarannsóknir.

____

Nafn þess sem leggur samþykkisyfirlýsinguna fyrir

Appendix 4: Spurningarlisti – Lífsgæði, þægindi gerviliða, öryggi.

Vinsamlegast merktu inn á mælistikuna hvernig þú metur eftirfarandi spurningar, ***<u>miðað við sl. 4 vikur.</u>***

Hve ánægð/ur ertu með virkni og þægindi gerviliða þinna?

0	1	2	3	4	5	6	7	8	9	10

Mjög ánægð/ur **Mjög óánægð/ur**

Hvernig hefur almenn líkamleg líðan þín verið?

0	1	2	3	4	5	6	7	8	9	10

Mjög slæm **Mjög góð**

Hefur þér fundist þú vera þróttmikil/l og virk/ur eða dauf/ur og svifasein/n?

0	1	2	3	4	5	6	7	8	9	10

Mjðg dauf/ur og svifaseinn **Mjög þróttmikil/l og virk/ur**

Hefur aflimunin komið í veg fyrir að þú getir sinnt vinnu, skóla, áhugamáli eða heimilisstörfum?

0	1	2	3	4	5	6	7	8	9	10

Aldrei **Mjög oft**

Hversu öruggur hefur þér fundist þú vera í almennri hreyfingu á gerviliðum þínum?

0	1	2	3	4	5	6	7	8	9	10

Mjög örugg/ur **Mjög óörugg/ur**

Appendix 5: The Activities-specific Balance Confidence Scale – The ABC scale

A-Ö jafnvægiskvarðinn

Veldu viðeigandi tölur, af þessum prósentukvarða, sem lýsa sjálfsöryggi þínu í eftirfarandi athöfnum.

0%	10	20	30	40	50	60	70	80	90	100%
Ekkert öryggi										Fullkomlega örugg/-ur

"Hversu örugg(ur) ert þú um að halda jafnvægi og vera stöðug(ur) þegar þú..."

1. gengur um húsið?________%
2. gengur upp eða niður stiga? ________%
3. beygir þig niður og tekur upp inniskó sem liggur fremst á botninum inni í fataskáp? ________%
4. teygir þig eftir lítilli niðursuðudós á hillu í augnhæð? ________%
5. stendur á tám og teygir þig eftir einhverju fyrir ofan höfuð? _____%
6. stendur á stól og teygir þig eftir einhverju? ________%
7. sópar gólfið? ________%
8. gengur út að bíl sem er lagt í innkeyrsluna? ________%
9. ferð inní eða útúr bíl? ________%
10. gengur þvert yfir bílastæði í áttina að verslunarmiðstöð eða búð? ________%
11. gengur upp eða niður halla? ________%
12. gengur um troðfulla verslunarmiðstöð þar sem fólk gengur hratt framhjá þér? ________%
13. lendir í því að fólk rekur sig utan í þig á göngu um verslunarmiðstöðina? ________%
14. ferð í eða úr rúllustiga og heldur í handrið? ________%
15. ferð í eða úr rúllustiga, með fangið fullt af varningi, þannig að þú getur ekki haldið í handrið? ________%
16. gengur úti á ísilagðri gangstétt? ________%

Samtals úr spurningum 1-16 = _________ Stig.

Samtals stig / 16 = ___________ A-Ö stig.

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Jafnvægiskvarði tengdur athöfnum og öryggistilfinningu:
A-Ö jafnvægiskvarðinn
(The Activities-specific Balance Confidence Scale – The ABC scale)

Lýsing:
A-Ö kvarðinn er spurningalisti, ætlaður til að meta öryggi eldra fólks til að athafna sig í daglegu lífi án þess að missa jafnvægi eða detta. Líklegt er að sjálfsöryggi fólks sé minna þegar það á við jafnvægisvandamál að stríða. Einstaklingar geta einnig haft óhóflegt sjálfsöryggi sem samræmist ekki raunverulegri getu þeirra til að halda jafnvægi. Þeir einstaklingar geta óvitandi komið sér í hættulegar aðstæður.

Framkvæmd
Hægt er að láta þátttakendur fylla A-Ö spurningalistann út sjálfa, taka við þá viðtal á staðnum eða símleiðis. Stækka þarf letrið á eyðublaðinu ef þátttakendur eiga að fylla það út sjálfir en stækkuð útgáfa af mælikvarðanum, á sérstöku spjaldi, er gagnleg þegar viðtal er tekið. Það er mikilvægt að spyrillinn endurtaki setninguna "Hversu örugg/-ur ert þú um að þú getir haldið jafnvægi og verið stöðug/-ur þegar þú..." a.m.k. við aðra hverja spurningu á listanum. Grennslast þarf fyrir um skilning hvers þátttakanda á leiðbeiningunum og hvort hann á í erfiðleikum með að svara tilteknum spurningum.

Leiðbeiningar til þátttakenda
"Nú áttu að svara nokkrum spurningum sem tengjast því hversu örugg(ur) þú ert um að geta framkvæmt eftirfarandi athafnir án þess að missa jafnvægið eða verða óstöðug(ur). Það gerir þú með því að velja eina prósentutölu á kvarðanum frá 0% upp í 100%. Ef viðkomandi athöfn er eitthvað sem þú gerir ekki á þessu tímaskeiði ævinnar, reyndu þá að ímynda þér hversu örugg(ur) þú værir ef að þú yrðir að framkvæma hana. Ef þú ert vön/vanur að nota gönguhjálpartæki, eða halda í einhvern, þegar þú framkvæmir athöfnina þá áttu að miða við að þú hefðir þann stuðning þegar þú metur sjálfsöryggi þitt. Láttu vita ef þú ert í vafa um hvernig svara skuli einhverri af spurningunum."

Leiðbeiningar fyrir stigagjöf
Til að fá A-Ö-stig fyrir hvern þátttakanda þarf að leggja saman stigin (möguleg spönn frá 0 upp í 1600) og deila í með 16 (eða fjölda svara). Ef viðkomandi hefur ekki einhlýtt svar við spurningum númer 2, 9, 11, 14 og/eða 15 (mismunandi stigagjöf fyrir "upp" og "niður", "inní" og "útúr" eða "í" og "úr") þarf að skipta viðeigandi spurningu/-m í tvennt og gefa aðskilin stig fyrir hvorn hluta fyrir sig. Nota skal lægri stigagjöfina af þessum tveimur (sá hluti mun hafa áhrif á athöfnina í heild sinni, t.d. líkurnar á að viðkomandi gangi stiga). Reikna má út heildarstig ef a.m.k. 12 spurningum er svarað. Athugið að þegar spurningalistinn er lagður fyrir fólk sem býr í heitu loftslagi má sleppa spurningu númer 16 (ísilögð gangstétt) án þess að innri áreiðanleiki (alpha) minnki umtalsvert.

Heimildir:

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