Muscle activation patterns during functional movements in transtibial amputees

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DEDICATION

This thesis is dedicated to my loving Gran, Lorna Watt.

Your strength and faith during adversity have taught me so much and I am truly grateful for that.

DECLARATION

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SUMMARY

Functional movement capabilities of individuals with unilateral transtibial amputations are altered due to muscle loss and prosthesis limitations compared to healthy, typical individuals. The extent of the adaptions made during functional activities is however, unknown. The purpose of the study was to gain a better understanding of unilateral transtibial amputees (UTTA) muscle activation levels during functional activities. A systematic review (article one) relating to the gait and balance of UTTA was completed. It revealed the need for research relating to muscle activation and movement strategies during functional activities. Stage two of the Van Mechelen model was addressed through biomechanical analysis during single leg balance (SLB) and sit-to-stand-to-sit (SiStSi) tasks through muscle activation and biomechanical analysis.

The study included 12 UTTA (34±10 years) and 13 able-bodied controls (CON) (34±11 years). The average time since amputation was 10±7 years. Each UTTA made use of their personal prosthesis for the observational testing. The participants were required to perform a unilateral SLB task followed by 10 continuous SiStSi movements. Muscle activation was measured for seven muscle groups using surface electromyography (EMG) together with a three dimensional biomechanical analysis.

The results of article two relates to the single leg balance activity. Significantly greater muscle activation levels were found for the lumbar erector spinae (LES), gluteus medius (Gmed), gluteus maximus (Gmax), biceps femoris (BF) and vastus lateralis (VL) ($p<0.05$) for the affected side (AF) in comparison with the unaffected side (UN) and CON. Greater hip flexion moment and concentric hip power were observed for AF ($p<0.05$) while hip and knee flexion was greater than UN and CON ($p<0.05$). No significant differences were found for the knee and ankle joint moments during SLB ($p>0.05$).

The SiStSi results are discussed in article three. Lower muscle activation levels were found for VL of AF compared to UN and CON, with greater activation levels of the tibialis anterior (TA) for UN than CON ($p<0.05$). The peak hip moment for AF during the SiSt was greater than UN and CON ($p<0.05$). Significantly greater hip power and hip flexion were identified for UTTA compared to CON ($p<0.05$), while the knee and ankle joint moments and powers were greater for UN than AF and CON ($p<0.05$). Lastly, vertical ground reaction force (vGRF) was significantly higher for UN than AF ($p<0.05$)

The main findings included greater muscle activation of the muscles surrounding the hip joint of UTTA during the SLB and the SiStSi activities. Joint overloading was noted for the UN knee as well as overcompensation by the UN ankle during the SiStSi. Lastly, asymmetry was observed in the vGRF between the AF and UN sides during the SiStSi.

OPSOMMING

Aangepaste funksionele bewegingsvermoeë kom voor by individue met unilaterale transtibiale amputasies (UTTA) wanneer hulle met gesonde tipiese persone vergelyk word. Verlies van spierfuksie en die beperkings van die prostese speel ‘n rol in kompenserende bewegings, alhoewel die mate van aanpassings tydens funksionele aktiwiteite onbekend is. Die doel van hierdie studie was om meer kennis te verkry rakende die spieraktiveringsvlakke van UTTA gedurende funksionele aktiwiteite. ‘n Sistematiese oorsig (artikel een) aangaande die looppatrone en balans van UTTA het die behoefte aan verdere navorsing met betrekking tot spieraktiveringsvlakke en bewegingstrategieë gedurende funksionele aktiwiteite in UTTA uitgelig. Fase twee van die Van Mechelen model was aangespreek deur biomeganiese analises en meting van spieraktiveringsvlakke tydens die een-been-staan en die sit-tot-staan-tot-sit (SiStSi) aktiwiteite.

Hierdie studie het 12 UTTA (34±10 jaar) en 13 tipiese persone (kontrole groep) (34±11 jaar) ingesluit. Die gemiddelde tyd sedert amputasie was 10±7 jaar. Tydens hierdie waarnemingstudie het elke UTTA het van sy eie persoonlike prostese gebruik gemaak. Daar was van elke deelnemer verwag om ‘n een-been-staan (SLB) beweging uit te voer, wat opgevolg was met 10 aaneenlopende SiStSi bewegings. Driedimensionele biomeganiese analise sowel as oppervlak elektromiografie van sewe spiergroepe was gemeet tydens hierdie bewegings.

Die resultate van die SLB aktiwiteit word in artikel twee bespreek. Dit dui op beduidend groter spieraktiveringsvlakke aan van die lumbale erector spinae (LES), gluteus medius (Gmed), gluteus maximus (Gmax), biseps femoris (BF) en die vastus lateralis (VL) ($p<0.05$) van die geaffekteerde kant van UTTA, in vergelyking met beide die ongeaffekteerde kant en die kontrole groep. Groter heupfleksor draaimoment en konsentriese heupdrywing was gevind vir die geaffekteerde kant ($p<0.05$), te same met groter heup- en kniefleksie ($p<0.05$) wanneer dit met die ongeaffekteerde kant en kontrole groep vergelyk word. Geen betekenis verskille was tydens die SLB gevind vir nóg die knie- nóg die enkelgewrig draaimoment nie.

Die bevindinge van die SiStSi word in artikel drie bespreek. Dit toon laer spieraktiveringsvlakke aan vir VL van die geaffekteerde kant in vergelyking met die ongeaffekteerde kant en die kontrole groep. Hoër spieraktivering was gevind vir die tibialis anterior (TA) van die ongeaffekteerde been van UTTA in vergelyking met die kontrole groep ($p<0.05$). Die piek heupdraaimoment vir die geaffekteerde kant van UTTA was beduidend hoër as vir die ongeaffekteerde kant en die kontrole groep tydens die SiSt ($p<0.05$). Heupdrywing en heupfleksie was beduidend meer vir UTTA as vir die kontrole groep ($p<0.05$), terwyl die knie- en enkelgewrig draaimoment van die ongeaffekteerde been groter was as die geaffekteerde been van die kontrole groep ($p<0.05$). Vertikale grondreaksiekrag was beduidend hoër vir die ongeaffekteerde been as vir die geaffekteerde been van die UTTA ($p<0.05$).

Die hoofbevindinge sluit in hoër spieraktiveringsvlakke van die spiere rondom die heupgewrig aan die geaffekteerde kant van UTTA gedurende beide die SLB en SiStSi aktiwiteite. Verder is gevind dat die kniegewrig van die ongeaffekteerde kant oorlaai word, sowel as dat die enkle-gewrig aan die ongeaffekteerde kant tot ‘n betekenisvolle mate kompenseer vir die verliese aan die geaffekteerde kant tydens die SiStSi. Asimmetriese vertikale grondreaksiekrag tussen die geaffekteerde en ongeaffekteerde kant kom voor tydens die SiStSi.

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Even though I walk through the valley of the shadow of death, I will fear no evil, for you are with me; your rod and your staff, they comfort me. – Psalm 23:4

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ABBREVIATIONS

% movement cycle	Percentage of the movement cycle
%	Percentage muscle activation
°	Degrees
3D	Three dimensional
AF	Affected side
ANOVA	Analysis of variance
BF	Bicep femoris
BoS	Base of support
CINHAL	Cumulative index of nursing and allied health literature
cm	Centimeter
COM	Centre of mass
CON	Control
COP	Centre of pressure
CV	Coefficient of variation
D	Dominant side
EMG	Electromyography
ESAR	Energy storage and return prosthesis
Gmax	Gluteus maximus
Gmed	Gluteus medius
LBP	Lower back pain
LES	Lumbar erector spinae

m	Meters
MEDLINE	Medical literature analysis and retrieval system online
MeSH	Medical subject heading
MG	Medial gastrocnemius
MTC	Minimal toe clearance
MVC	Maximal voluntary contraction
$N.kg^{-1}$	Newton per kilogram
$N.m.kg^{-1}$	Newton meter per kilogram
ND	Non-dominant side
PRISMA	Preferred reporting items for systematic review and meta-analyses
RMS	Root mean squared
ROM	Range of motion
SACH	Solid ankle cushion heel
SAFE	Stationary ankle flexible endoskeleton
SD	Standard deviation
SiSt	Sit - to - stand
SiStSi	Sit - to - stand - to - sit
SLB	Single leg balance
SPW	Self-paced walking
StSi	Stand - to - sit
TA	Tibialis anterior
TFL	Tensor fascia latae

UN	Unaffected side
UTTA	Unilateral transtibial amputees
vGRF	Vertical ground reaction force
VL	Vastus lateralis
$W.kg^{-1}$	Watts per kilogram

DEFINITIONS

Compensatory mechanism	:	Possible adaptive strategies that may be used to control a movement and maintain balance
Muscle co-ordination	:	The ability of muscles to work together to control a movement
Ankle strategy	:	The response of the ankle joint receptors to trigger an activation of the muscles surrounding the ankle to respond to the stimulus and recover balance and control the movement
Hip strategy	:	The response of the hip joint receptors to trigger an activation of the muscles surrounding the hip to respond to the stimulus and recover balance and control the movement
Knee control	:	The co-ordinated activation of the muscles acting to control the movement at the knee
Talux prosthesis	:	A multi-axial foot contributing to balance and agility as well as a more natural motion of the foot
Seattle LightFoot 2 Prosthesis	:	Split keel design with improved dynamic response and stability on uneven surfaces
Propriofoot prosthesis	:	Motor powered foot for low to moderate functional below knee amputees

Chapter 1

Background & problem statement

1.1. Introduction to transtibial amputees

Amputations of the lower limbs can be classified into two main categories namely transfemoral or transtibial amputations. These categories are then further expanded into either unilateral (affecting only one limb) or bilateral where both lower limbs are affected. A transtibial amputation can be as a result of three main causes including congenital deformity, traumatic injury and vascular irregularities (Gailey, 2008). According to Ziegler-Graham *et al*, (2008), it was estimated that vascular amputations account for 54% of lower leg amputations while traumatic amputations make up 45%. The increased prevalence of diabetes and peripheral vascular disease attribute to the higher rate of vascular amputation. Ziegler-Graham *et al.* (2008) predict that in the years to come the prevalence of lower limb amputation may continue to increase. Thus, a better understanding of the consequences of a lower limb amputation is imperative.

An amputation of a lower limb is considered a debilitating injury as it affects daily-required activities such as standing, balancing, walking and running. Apart from the physical pain and challenges experienced by an individual with a lower limb amputation, the individual endures severe psychological and emotional stress (Desmond & MacLachlan, 2006). Therefore, it can be said that a great amount of adaptation is needed in order to live life holistically without a limb. The dynamic system theory can be used to holistically evaluate the relationship between environmental factors, the tasks and the amputee to ensure that optimal movement will take place or how it may be adapted (Holt *et al.*, 2010). This will be expanded on further in this chapter.

1.1.1 Locomotion and daily functional activities

Locomotion is a daily functional need whether it is walking, running, climbing stairs, or standing on the bus. Moving from position 'A' to position 'B' needs to be performed in the most efficient manner to avoid possible injury and fatigue.

As mentioned, it is not only walking gait that is of importance but also all activities that need to be achieved on daily basis namely functional activities. In this thesis, daily functional activity includes that of single leg balance and going from a seated to standing position and vice versa. In a study by Chisholm (2015), five functional activities of daily living were used in the research of transfermoral amputees. These activities included walking gait, ascending and descending of steps, sit-to-stand and stand-to-sit movements as well as door pull. The importance of investigating daily functional activities is to add to the pool of knowledge, as well as to improve on the current rehabilitative strategies in order to improve quality of life using a holistic rehabilitation approach.

1.1.2 Prosthetic development

In recent years, there has been constant development in the design of prosthetic feet to help improve functionality and quality of life. Lower limb prostheses originally took the form of a peg shaped leg. Thereafter, the designs began to resemble armour shaped limbs. Dr Bly designed the first ‘anatomical leg’ in 1858, which allowed for eversion and inversion (Gutfleisch, 2003). It consisted of an ivory ball and socket as an ankle joint, which allowed for limited ‘ankle’ movement. Warfare resulted in a demand for technological development in lower limb prosthetics and the more modern era has called for further advancements (Gutfleisch, 2003).

Socket materials have changed to include the use of carbon fibre for its rigidity and lightweight properties while liners are now made from silicone (Selles *et al.,* 2004). The development of the ankle and foot components has seen the most improvement in recent years. What started as a solid ankle design has developed to a prosthesis that allows for ankle movement, which more closely simulates natural ankle motion (Barr *et al.*, 1992). This has resulted in designs that include varying degrees of plantar and dorsiflexion as well as mechanisms that help to control ankle movement, acting like the Achilles and Gastroc-Soleus complex (Gutfleisch, 2003).

Recent development has not only focused on microprocessor components, but also the aesthetics of the prostheses (Le & Scott-Wyard, 2015). Current trends in prosthesis design are focused on improving the appearance of the prosthesis, comfort of the prosthesis and development of individualised prostheses that are aimed at providing more functional and practical prostheses improving participation in activities of daily life (Griffet, 2016). Research also aims to help guide prosthetic selection based on the activity capacity (Agrawal *et al.,* 2013). There are many different lower limb prostheses including but not limited to the following:

- Talux prosthesis (A multi-axial foot contributing to balance and agility as well as a more natural motion of the foot),
- Seattle LightFoot2 (Split keel design with improved dynamic response and stability for navigating uneven surfaces) and the
- Propriofoot (Motor powered foot for low to moderate functional below knee amputees).

Reference will be made to these in chapter two.

1.1.3 Gait and asymmetry

It is well known that unilateral transtibial amputees (UTTA) have asymmetrical gait (Bateni & Olney, 2002; Silverman *et al.,* 2008). The asymmetries seen are not only anatomical, but also functional and affect gait, balance and co-ordination of movement (Silverman *et al.*, 2008). One of the most established reasons for this is the lack of plantar flexors in the lower limb of the affected side (Sadeghi *et al.*, 2001). Several additional factors may influence the gait of UTTA. These include stump length, type of socket, socket fit and type of prosthesis (Silverman *et al.*, 2008). As mentioned earlier UTTA have reduced ankle range of motion (ROM) on the affected side (Silverman *et al.,* 2008). This results in reduced push-off power by the prosthetic ankle during the gait cycle and therefore can result in asymmetry (Smith, 2008). Asymmetry has also been noted in the step length on the unaffected side. According to Hak *et al.*, (2014), a shorter step length on the unaffected side during initial contact aids in increasing the base of support. They concluded that this might result in functional compensation to limit the risk of falling (Hak *et al.*, 2014).

1.1.4 Balance and risk of falling

There has been an attempt to investigate various factors relating to the wellness of UTTA. One such area includes determining the risk of falls in this population (Amosun *et al.*, 2005). As mentioned one such risk of injury is osteoporosis affecting elderly individuals (Gailey, 2008). Individuals with amputations are at a higher risk of falling compared to that of a healthy able-bodied individual and the accompanying osteoporosis places them at a higher risk of fractures (Kaufman *et al.*, 2014; Rosenblatt *et al.*, 2014).

Postural asymmetries and poor general posture in UTTA may negatively affect movement potentials and place the individual at an unnecessary risk of compensatory back pain (Devan *et al.*, 2014). This is similar to what is found in the able-bodied population (Quinlan *et al.*, 2006). Knowing that UTTA are at risk for lower back injuries results in a need to improve posture and muscle recruitment to aid injury prevention. Kulkarni *et al.* (2005) determined that lower back pain is commonly experienced in lower limb amputees and suggested that future research focus on determining the compensations experienced.

Two common strategies have been discussed in literature used to control balance. These include the ankle strategy and the hip strategy. The use of a particular strategy depends on numerous factors including the type of activity, the magnitude of the external force, the type of prosthesis and the relative strength of muscles surrounding the joints (Reimann *et al*. 2017). The ankle strategy is the control or recovery of control whereby the ankle musculature work together to respond to an imposed force. This acts by inducing a ‘single segment inverted pendulum’ controlling the postural sway to recover balance (Reimann *et al.*, 2018). The hip strategy works in a similar way by activating the muscles surrounding the hip to co-ordinate a recovery action. It is generally used as a strategy when an external force is applied and the ankle strategy is not sufficient (Riemann *et al.*, 2003).

1.1.5 Risk of injury

The risk of injury for this population has been of concern. The concern stems from the compensations made in order to perform certain movements, adaptations in muscle usage and the asymmetry due to the missing lower limb (Ramstrand & Nilsson, 2009). Research has undertaken to determine the risk of osteoarthritis in amputees and they have found that transfermoral amputees are at a greater risk of developing these conditions in comparison to transtibial amputees (Kulkarni *et al.*, 2005). Unilateral transtibial amputees have a moderate predisposition to developing osteoarthritis in the unaffected leg and this may be due to asymmetries as discussed previously (Lloyd *et al.*, 2010).

Lower back pain (LBP) has also been highlighted as a risk amongst the amputee population. Kulkarni *et al.* (2005) discussed that LBP is not only a common condition experienced by general population but that 69% of transtibial amputees also suffer from it. Muscle strength imbalances can also occur in the lower limbs as well as the back to help compensate or adapt in order to perform daily movements (Silverman *et al.*, 2008).

1.1.6 Electromyography in transtibial amputees

There is limited literature describing the muscle activity in UTTA. Isakov *et al.* (2001) however investigated the knee muscle activation ratios during walking. More specifically, they identified the activation and ratios between the vastus medialis and the bicep femoris muscles. The results of this study indicated that the bicep femoris activated significantly later during the gait cycle on the affected side compared to that of the unaffected side (Isakov *et al.,* 2001). Another study investigated the muscle activation patterns of the residual limb at the stump-socket interface (Huang & Ferris, 2012). They found that there were lower levels of activations of these muscles, however there was higher variability in the muscle activation measured during walking (Huang & Ferris, 2012). Viton *et al.* (2000) determined that during a standing, side leg raise by UTTA, the gastrocnemius and tibialis anterior muscles activated together. They also found that muscle activation patterns were different in transtibial amputees in comparison to controls (Viton *et al.*, 2000). A study has also investigated the muscle activation of the hamstring and quadriceps muscles during gait and found greater activation levels on the unaffected side (Powers *et al.*, 1998). Further research is needed to understand muscle activation patterns of bilateral lower body muscles during functional activities and the influence this may have on the movement abilities of transtibial amputees.

1.2 Models and theories

1.2.1 Dynamic systems theory

The dynamics system theory has been used within the field of motor control as well as injury rehabilitation to understand and interpret the findings and the impact of these changes on the process or intervention (Wolpert *et al.*, 2001; Kvist, 2004; Kelly & Darrah, 2005; Wikstrom *et al.*, 2013). The dynamic systems theory is used to understand the relationship between three main constraints, which include the organism, the task and the environment. It is based on the premise that these constraints may change over time. As one of them changes, the influence this may have on the other two main factors needs to be considered (Holt *et al.*, 2010). In the case of UTTA, it is important that all three of these constraints are considered in order to adequately adapt the environment or movement pattern to perform a task. On the other hand, it is important to understand what the movement patterns are, depending on the demand of the task, to determine the risk of injury as well as the efficiency of the movement.

1.2.2 Van Mechelen Model

The Van Mechelen model was developed in 1987 and was designed to monitor and prevent injuries (van Mechelen, 1997). This model consists of four key stages incorporating: 1) recognising the extent of the problem, 2) determining the mechanism of the problem, 3) implementing rehabilitative steps and 4) evaluating the effectiveness of the intervention (van Mechelen, 1997). In the current study, the second stage of the model was explored through observational testing. Through the analysis of results, possible mechanisms for the problems seen were explored and reasoned.

1.3. Motivation for the study

Researchers have focused on prosthesis design, gait parameters and gait retraining for straight-line locomotion. Scant literature is available relating to the muscle activation patterns of UTTA during daily functional activities such as single leg balance, sit-to-stand and stand-to-sit. Therefore, there is a need to gain a greater understanding of UTTA muscles activation during daily functional activities. This is important in order to design rehabilitation programmes that may be most effective for these individuals.

1.4. Purpose and research questions

The purpose of the study was to gain a greater understanding of the muscle activation patterns of UTTA during movements of daily living, which could lead to information regarding muscle overloading.

For the purpose of this study, the research questions below were the key focus in the articles that are included.

1. Is there a difference in muscle activation levels between able-bodies and unilateral transtibial amputees (UTTA) during functional activities?
2. How do the joint kinetics and kinematics for the hip, knee and ankle during functional activities compare between the affected side (AF) and unaffected side (UN) of unilateral transtibial amputees and dominant side (D) and non-dominant side (ND) of controls?

Objective 1

1.1 To determine the skeletal muscle activation levels between the AF and UN of UTTA during functional movements using surface EMG placed on the vastus lateralis, bicep femoris, gluteus medius, gluteus maximus and the lower region of the lumbar erector spinae.

1.2 To determine the difference in muscle activation levels between able-bodied controls and UTTA during specific functional movements using surface EMG placed on the tibialis anterior, medial gastrocnemiums, vastus lateralis, bicep femoris, gluteus medius, gluteus maximus and the lower region of the lumbar erector spinae.

Objective 2

2.1 To compare the vertical ground reaction force (vGRF), joint moments, powers and angles acting at the hip, knee and ankle during functional activities using Vicon 3D analysis between the affected side (AF) and unaffected sides (UN) of the UTTA and the dominant side (D) and non-dominant side (ND) of the control group.

1.5. Scope of study and limitations

This study followed a descriptive, observational study design. Healthy UTTA between the age of 18 and 65 years of age were included. The UTTA had undergone a transtibial amputation at least one year previously and were able to use a walking prosthesis. An age and gender matched control group of healthy able-bodies who were free from injury and illness was included for comparative purposes.

Limitations of this study included that of a small sample size limiting the ability to apply findings to the bigger population group. The SLB and SiStSi protocols did not allow for the use of the arms. The protocol was slightly adapted so that the participants placed their hands on top of each other, just below the xiphoid process so as not to obstruct the view of any markers. The types of prostheses were not controlled for the study, as we wanted participants to be used to their own prosthesis that they were familiar with for daily activities.

1.6. Chapter overview

This thesis is structured using an article format. Three research articles (Chapters two, three and four) were prepared for possible publication in specific journals and thus followed journal specific guidelines concerning format and references styles. Consequently, reference styles are not consistent throughout this thesis

Chapter 1

Background and problem statement: In chapter 1, the topics of this thesis were introduced and the reason for research in this field was discussed. The purpose and research questions were documented. The Harvard reference style was used as per Department of Sport Science, Stellenbosch University requirements.

Chapter 2

Article 1: "*Locomotion and postural stability in unilateral transtibial amputees: A systematic review.*" This chapter consists of a systematic review article. This article examines the literature available related to the influence of unilateral transtibial amputations on the biomechanics of gait and balance. This article was prepared for possible publication in the Gait & Posture journal. The submission guidelines were observed, however for ease of reading, the tables have been kept within the text and the left and right margins have not yet been set. The reference style was set as the Elsevier – Vancouver method.

Chapter 3

Article 2: "*Muscle activation patterns during single leg balance of unilateral transtibial amputees in comparison to controls.*" Chapter 3 consists of a second article that focused on single leg balance in UTTA. This study involved an in-depth biomechanical analysis with specific focus on the muscle activation levels. The article was formatted for the Gait & posture journal however, for ease of reading the tables have been kept within the text and the left and right margins have not yet been set. The reference style was set as the Elsevier – Vancouver method.

Chapter 4

Article 3: "*Biomechanical analysis of the sit-to-stand-to-sit activity in unilateral transtibial amputees."* This chapter consists of an article related to the sit-to-stand and stand-to-sit activities. It discussed the biomechanical differences with the UTTA as well as with a control group. This article was written with the intention of submitting it to the Journal of Prosthetics and Orthotics International. The reference style selected was that of the Sage – Vancouver method.

Chapter 5

Discussion: This chapter discusses and integrates the findings of the overall study. It also consists of the limitations to the study, possibilities for future research as well as the conclusion. The Harvard reference style was used for this chapter as per the Department of Sport Science, Stellenbosch University's regulations.

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Chapter 2

Locomotion and postural stability in unilateral transtibial amputees: A systematic review

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This article was prepared for possible publication in the Gait & Posture journal. The submission guidelines were observed, however for ease of reading, the tables have been kept within the text and the left and right margins have not yet been set. The specified journal reference style used was the Elsevier – Vancouver method. We were limited to 6000 words for a review for this journal and we have currently used 5951 words. The abstract was limited to 250 words which we have used.

Abstract

INTRODUCTION: Locomotion and balance of unilateral transtibial amputees (UTTA) are influenced by numerous factors (physiological and biomechanical), which may influence functional ability. In order to implement effective rehabilitation programmes it is important to understand how the functionality of UTTA are affected. The purpose of this paper was to systematically review the available literature on the biomechanics of walking gait and postural stability in terms of muscle activation and asymmetry in UTTA.

METHODS: Three databases were used for the literature search including Pubmed, CINHAL and MEDLINE. Search words used were medical subject headings (MeSH) terms including, unilateral transtibial amputees, gait, muscle activity, kinematics, balance and asymmetry. An individual two-person review process according to strict criteria was followed in the selection of articles for full review and inclusion into the study.

RESULTS: The literature search revealed 176 possible articles to be included of which 76 were included for full text review with 25 articles finally being included. The literature review indicated that walking gait of UTTA is vastly different compared to able-bodied controls. Asymmetrical gait and the associated compensations due to limb loss negatively affect the daily functionality of UTTA. Literature also indicated that prosthetic foot type, socket fit and age of the amputee may influence gait and postural stability in UTTA.

CONCLUSION: Along with the external factors that may influence the gait or postural stability of UTTA, it is clear that the loss of a limb results in asymmetrical and compensatory movement patterns. The extent of these warrants further investigation to enhance the rehabilitation process.

(250)

Key words: Unilateral transtibial amputee, gait, muscle activation, postural stability, biomechanics, asymmetry

2.1 Introduction

Unilateral transtibial amputees (UTTA) have asymmetrical locomotion and are known to be at a higher risk of falling. Considerable research has been completed on various aspects of UTTA. Complex factors may influence the locomotion of transtibial amputees and need to be considered when working with individual patients.

These factors include postural and biomechanical compensations, residual muscle mass and muscle strength, the etiology of amputation, variance in rehabilitation and componentry of the prosthesis used [1–3]. The time since amputation also needs to be considered when examining biomechanical factors of movement [4]. A period of at least one year since amputation may allow for more confident movement capabilities when performing daily functional activities with the use of a walking prosthesis [1,5].

Considering biomechanical variations in amputees during movement including, variation in step length, contact time and flight time during the gait cycle and the subsequent impact that any one of these variables may have on the individual is important to determine [6]. Biomechanical analyses investigating both kinematic and kinetic variables provide insight into the possible influence that these variables may have on the joints during locomotion [7]. Nolan et al. [8] compared transtibial amputees and able-bodied controls in terms of force asymmetries and gait parameters. It was found that regardless of the speed of walking, the unaffected limb of the UTTA experienced higher impact forces than both the affected side and non-amputee control (able-bodied) group. Furthermore, UTTA take longer to shift their weight onto the prosthetic limb while walking, furthering the burden on the unaffected side. These asymmetries and compensations should be considered with respect to possible longitudinal overload injuries.

Little is known about the nature of the difference between typical gait and the gait of UTTA but most studies have evaluated single variables or the biomechanics of single joints. A review of the influence on walking gait, postural stability, muscle activity and other factors influencing locomotion may help to summarise existing facts and reveal possible gaps in the research that need to be addressed.

Previous systematic reviews have focused on individual parameters in isolation. These include balance in lower limb amputees during quiet stance [9], effect of prosthetic mass on gait [10], movement asymmetries around the spine, pelvis and hip [11] and walking capacity [3]. To date no systematic review has focused on a combination of these factors. Therefore, the aim of this review was to systematically identify the literature available on UTTA in terms of walking gait, balance and symmetry as complex interactions as well as muscle activation using electromyography (EMG) and kinetic and kinematic parameters.

2.2 Methods

For the purpose of this systematic review locomotion referred to ambulatory movement such as walking gait, stair ambulation and the navigation of obstacles while walking. Postural stability referred to either static balance or dynamic balance during gait. Asymmetry is considered to include asymmetrical movement between the affected and unaffected side in terms of muscle strength, gait parameters and postural stability.

Search strategy

Three databases were searched: Pubmed, Medline, and CINAHL. The search strategy consisted of both medical subject headings (MESH) terms as well as alternative terms known for each of the MESH terms. The search terms included unilateral transtibial amputation or single below leg amputation AND gait or walking gait, or locomotion AND muscle activation or EMG or muscle firing patterns or neuromuscular control AND biomechanics or kinetics or kinematics AND balance or stability or postural control AND asymmetry or symmetry. Electronic alerts were set up on all searched databases to identify additional articles that may fit the inclusion criteria. Selected article reference lists were searched for possible article inclusion. The most recent search date was 22 February 2018.

Selection process

Articles were selected for this systematic review through a three stage process involving 1) a title evaluation, followed by 2) an abstract review and 3) a full text review evaluation.

Abstract and title review

The exclusion criteria of this literature review are presented in the included preferred reporting items for systematic reviews and meta-analyses (PRISMA) flowchart (figure 2.1). Articles were included for full text review if they included UTTA performing walking or balance tasks while biomechanical parameters were being investigated.

Articles were excluded during the first two rounds if they included transfemoral or bilateral transtibial amputees, if the time since amputation was less than a year, if the focus was placed on running or sport performance, if the amputation was below the ankle or of the upper limbs, if external walking aids were used, if the research paper type was a review article or letters to the editor, and finally if there were any other secondary conditions that could influence the individual during locomotion. The reference lists of included articles were cross checked for possible article inclusion.

Final review process

The full texts that were reviewed were then evaluated and were excluded if/when there was no mention of ethical clearance/approval; there was no mention of time since amputation or if time since amputation was less than one year. Full texts were also excluded if the primary focus was on the prosthesis properties.

The reviewing process was independently performed by two researchers, both of whom were fluent in the English language. Disagreements on article selection were resolved during a consensus meeting between the researchers. The review process was completed using review documents designed by Helena Von Ville of the University of Texas (helena.m.vonville@uth.tmc.edu).

2.3 Results

During the screening process a total number of 176 articles were identified of which 76 articles were selected for full text review after the title and abstract evaluation was completed. Six papers were unavailable for review after extensive attempts to access the original articles including contact with the Stellenbosch University library, emailing and requesting of articles from the authors. At the end of the full text review stage, a total of 25 articles were included while 45 were excluded having not met the final inclusion criteria.

Reliability between researchers was calculated by Cohen's Kappa inter-rater statistical analysis using 66 randomly selected titles and abstracts from the searched articles [12]. A strong inter-rater reliability (0.87) was determined between the two researchers.

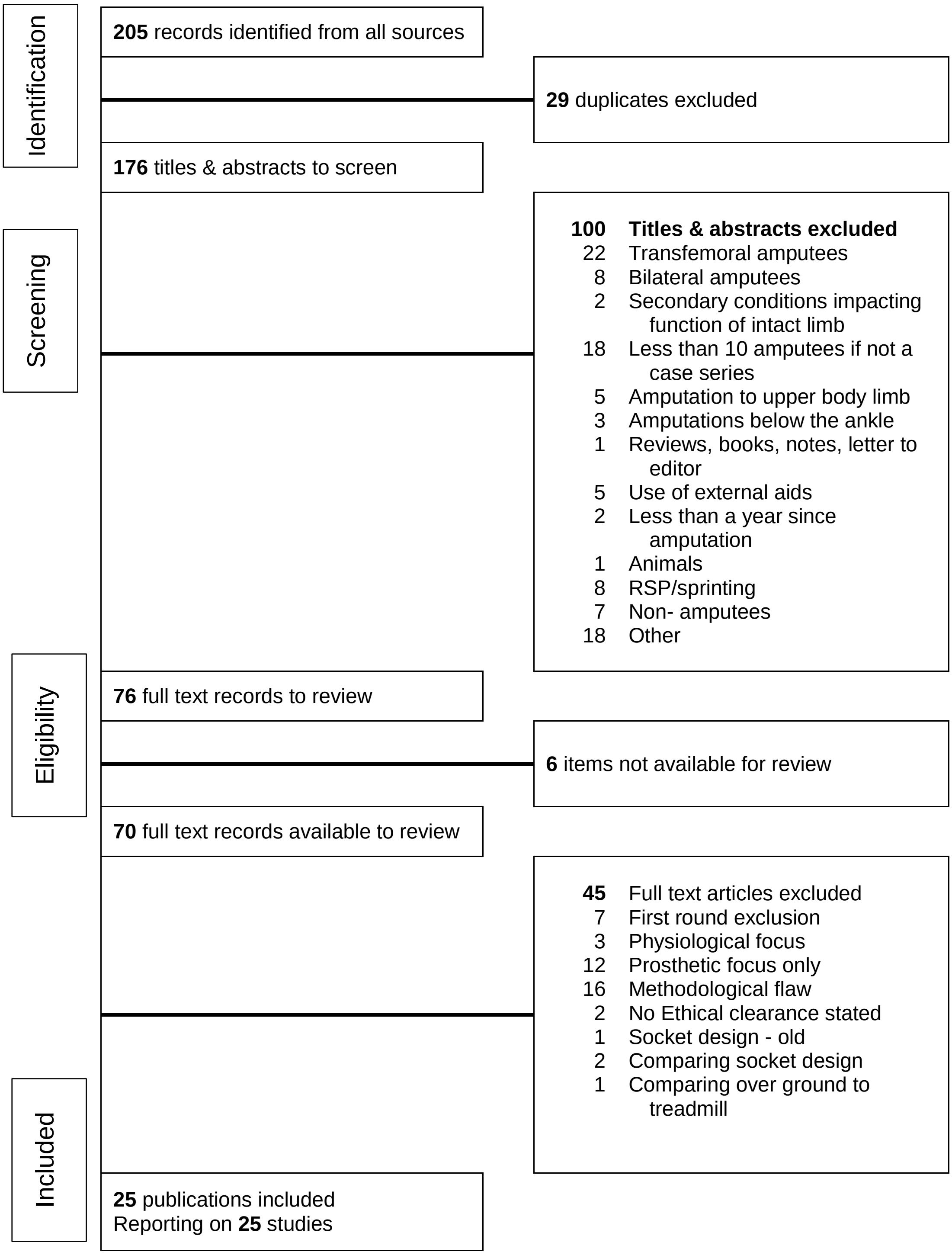

Figure 2.1 PRISMA flowchart of search results
RSP = running specific prosthesis, PRISMA = preferred reporting items for systematic reviews and meta-analyses

Of the articles (n=25) included in this systematic review, the average sample size used was 11 UTTA while the minimum and maximum were five and 25 respectively. Majority of the studies included multiple etiologies of amputation. These included traumatic (20), vascular (8), other conditions such as infection, neoplastic, congenital and immune conditions (16). The types of analysis in the studies mentioned below were three dimensional (3D) motion analysis (18), two-dimensional (2D) motion analysis (4), electromyography (EMG) (2), and other older methodologies (2).

This systematic review was then organised into four categories in order to highlight different aspects that influence locomotion and postural stability in UTTA. The first category was the biomechanics of walking gait in UTTA. Here the focus was on kinematic and kinetic characteristics and then the influence of different prostheses on gait. The second category was postural stability and risk of falls in UTTA. This was subdivided to document risk of falls and postural stability. The third category included muscle activation patterns in UTTA. The final category also included other additional factors affecting the locomotion of UTTA.

2.3. 1. Biomechanics of walking gait

2.3.1.1 Kinetic and kinematics variables

The full text review process yielded eight articles that focused on kinematic and kinetic variables during walking gait in UTTA (table 2.1). One article focused on stepping asymmetry while another determined the changes in gait during medial or lateral perturbations. Six articles examined the kinematics at the hip and knee joint while two investigated the thorax and ankle joints during gait.

Table 2.1 Summary of included articles on the biomechanics of walking gait

Authors	n	Age(yrs)	Time since amputation	Analysis	Main variables	Main finding
Bateni & Olney, 2002	5	51 ± 19	>1 yr	2D Motion Analysis	Hip, knee & ankle joint angles	Greater hip & knee flexion on AF side during stance
Isakov et al., 1996	14	41±13	17±15 yrs	Electric contact walkway system	Step length, hip and knee joint angles	Increased speed influenced hip angle (heel-strike) & knee angle during stance
Michel et al., 2004	5	43±20	>6 yrs	2D Motion Analysis	A/P COM, progression velocity	Different spatio-temporal strategies used by amputees to achieve same progression velocity by the end of the first step
Miff et al., 2005	10	54±8	4–37 yrs*	3D motion Analysis	Peak mean acceleration	Difference peak mean acceleration between step initiation using the anatomical side vs prosthesis
Molina-Rueda et al., 2013	15	56±1 4	>1 yr	3D Motion Analysis	Joint moments of the hip & knee	Decreased hip abductor moment (stance) in UTTA, decreased knee valgus moment on AF side
Molina-Rueda et al., 2014	25	50±14	>1 yr	3D Motion Analysis	Pelvic alignment (stance), hip abduction moment	AF side pelvis was higher (midstance), lower hip abduction moment on AF side compared to UN side (midstance)
Sadeghi et al., 2001	5	27±13	>5 yrs	3D Motion Analysis	Muscle power of hip, knee & ankle	Greater hip extensor power (propulsion), greater knee extensor power (absorption)
Villa et al., 2017	15	51±12	13±12 yrs	3D motion Analysis	Gait speed, step width, hip, knee & ankle joint angles	Decreased speed and increased flexion at hip & knee of the UN leg on higher side of a slope

Age given in mean years ± standard deviation; * = age given as a range
UTTA = unilateral transtibial amputee, AF = affected, UN = unaffected, COM = centre of mass, A/P = anterior/posterior

The study by Molina-Rueda [13] revealed that in the frontal plane, there is a lower internal moment at the hip joint on the affected side, compared with that of the unaffected side [13]. A previous study by Molina-Rueda [14] also found a difference in frontal plane movement during hip abduction. They found a smaller hip abductor moment on the affected side compared to that of the unaffected side during stance phase of the gait cycle. They also found a lower knee valgus moment on the affected side. It was suggested that these loading asymmetries may develop as a protective mechanism for the affected side against stump pain [14]. Sadeghi [15] confirmed this finding in their study which examined force production during gait. They found that there was a smaller force production in the ankle and knee joints of the affected side however the hip extensor moments were greater on this side compared to the unaffected limb. They suggested that this was due to the hip compensating for lower muscle power ability of the knee and ankle of the prosthesis, in order to help with stability during the transfer of body mass from one leg to another [15]. The kinematic and kinetic variations in UTTA gait was also examined by Bateni [16] where they found that there was a decrease in the power generation during push off from the affected limb, as well as more knee and hip flexion in early stance phase of the gait cycle [16]. It was suggested that these findings support the hypothesis of compensation for the lack of ability in power production of a prosthetic ankle.

Michel [17] examined the centre of mass in an anterior and posterior direction during gait initiation and determined the effect on the strategy to control velocity during walking. They found that this was in line with what has been seen in able-bodied controls, however the UTTA step execution phase duration was longer when gait was initiated with the affected limb [17].

Two studies identified the effect of change in walking speed on gait parameters. Isakov [18] examined the influence of walking speed on gait and found a significant difference in all temporal parameters and distances of both the affected and unaffected limbs. They also found that an increase in walking speed resulted in an increase in knee angles during the loading phase as well as during toe-off in the unaffected limb [18].

In a study by Miff [19] they determined the effect of walking speed on gait initiation and termination and they concluded that, regardless of the walking speed UTTA can stop within approximately two steps. This is the same for able-bodied controls. They suggested that a possible reason for this was to have an increased acceleration or deceleration time period by changing the centre of body mass (COM) in order to stop in two steps even when walking at a greater velocity [19].

Villa [20] investigated the strategies used during cross slope and level walking. They found that while walking with prosthesis on the uphill side the transtibial amputee adapts their movement strategy by increasing both hip and knee flexion to ensure ground clearance [20].

In summary many factors influence the movement strategies used depending on the specific task demands.

2.3.1.2 Types of prostheses and walking gait

The review process revealed five articles with a focus on the influence of different prostheses on gait parameters in UTTA (table 2.2). Four articles investigated the stiffness of prostheses and its effect on ankle and knee kinematics, or the influence of powered ankle prosthesis on whole body angular momentum during walking. One other article identified the loading of joints while walking using five different prostheses.

Table 2.2 Summary of included articles on the types of prostheses and their influence on gait

Authors	n	Age(yrs)	Time since amputation	Analysis	Main variable	Main finding
Agrawal et al., 2013	10	54±8	1-37 yrs*	F scan sensor	SEW, vGRF, COM	K level 2 UTTA had better gait symmetry with a more flexible prosthesis
Bateni & Olney, 2004	5	32-77*	>1 yr	2D Motion Analysis	Walking speed, stride length	UTTA walked faster when using the steel prosthesis component than titanium. Other findings were inconsistent.
D'Andrea et al., 2014	8	47±8	19±12 yrs	3D Motion Analysis	Angular momentum	Better regulation of angular momentum using a powered prosthesis compared to a passive elastic prosthesis
Segal & Klute, 2014	10	45±6	>1 yr	3D Motion Analysis	Step width	Smaller step width for UTTA with medial applied perturbation. Changes in foot stiffness did not change these findings.
Supan et al., 2010	10	34-62*	8-44 yrs*	3D Motion Analysis	Hip, knee & ankle angles, step length, GRF	Ankle ROM more similar to anatomical foot when using the Talux foot (heel height of 24mm)

Age given in mean years ± standard deviation; * = age given as a range
UTTA = unilateral transtibial amputee, SEW = external symmetry of work, vGRF = vertical ground reaction force, COM = centre of mass

These articles focused on kinetics and kinematics during walking gait in order to determine either forces acting at the joints or how different prosthetics affect the walking characteristics of this population. The majority of the articles listed above used 3D motion analysis to examine the joint kinematics and /or kinetics. While it can be acknowledged that there are differences between the affected and unaffected sides, it is crucial to understand the magnitude of these changes.

A study by Bateni [21] measured the effect of weight of prosthetic components on gait parameters whilst comparing titanium to steel prosthetic components. They did not find any significant differences affecting the walking gait of the amputees [21].

Supan [22] compared the Talux prosthetic foot to the unaffected limb where they found that the Talux was more similar to the unaffected foot in the way it responds. It is designed to act more dynamically allowing for improved plantar flexion and dorsiflexion as well as inversion and eversion. They also determined the influence of changing the heel height of the prosthesis. They found that, by changing the heel height by 24mm in the Talux foot, there were no significant changes to the alignment of the prosthesis. They suggested that this may allow an individual user to adjust the heel height to accommodate for the shoe, without changing the biomechanical loading on the body. Supan [22] also found that there was more similarity between the Talux foot and the unaffected limb during the gait cycle, compared to the FlexFoot. They concluded that the Talux foot more closely mimics the way the unaffected foot works, with more plantar and dorsiflexion possible [22].

In terms of whole body angular momentum, D'Andrea [23] determined if a powered prosthesis would help to return angular momentum to what is typically observed in able-bodied controls. They found that the UTTA using a powered prosthesis were better able to regulate force production and therefore able to better control angular momentum, within a similar range to able-bodies [23].

Agrawal [24] investigated the influence that gait training as well as the category of the prosthesis had on external work symmetry. They concluded that the design of a foot such as the Talux with its 'J' shaped ankle as well as the heel- to- toe footplate showed the largest symmetry of work for the K level 2 amputee ambulators (Individuals able to navigate most curbs, stairs and uneven surfaces) followed by the K level 3 amputees (Individuals able to navigate most environmental barriers as well as take part in sport activities) [24].

One study compared the Seattle LightFoot2 with the Highlander foot in terms of recovery after perturbation [25]. While walking, a medial or lateral perturbation was applied immediately before heel strike. They found that there were no differences between the stiff Seattle foot compared to the more compliant, Highlander foot. They therefore combined the results and compared them to a control group as well as to the unaffected limb. They found that with a lateral perturbation the initial recovery step resulted in an increased step width (30% increases) with a complete recovery of step width by the third step. The opposite was experienced with a medial perturbation (30% decreases) and the UTTA only recovered step width by the fifth step [25].

In summary the choice of prosthesis may influence the spatio-temporal variables and kinematics during walking.

2.3.2. Postural Stability and risk of falls during gait

2.3.2.1 Balance & fall risk

In the subgroup of balance and the risk of falls, three articles were identified (table 2.3). The main focus was the recovery from a possible fall and the co-ordination between the affected and unaffected legs. There were also two articles that particularly investigated the risk of falling by determining the toe clearance of different prosthetic feet. The range of motion provided by the prosthetic ankle determined the risk of falls due to tripping.

Table 2.3 Summary of included articles on balance and risk of falls

Authors	n	Age (years)	Time since amputation	Analysis	Main variables	Main finding
Curtze et al., 2010	17	55± 9	13±14 yrs	3D Motion analysis	Knee flexion angle, step length, GRF	Less knee flexion & longer step length at heel-strike when recovery step is led with the prosthetic side
Munjal & Kulkarni, 2014	21	48±13	>2 yrs	3D Motion Analysis	MTC, hip flexion angle	Improved MTC and increased hip flexion (swing) using hyA-F prosthesis
Rosenblatt et al., 2014	8	50±10	>1 yr	3D Motion Analysis	MTC	Propriofoot led to increased MTC

Age given in mean years ± standard deviation
GRF = ground reaction force, MTC = minimal toe clearance

Curtze [26] determined the ability of UTTA to recover from an evoked forward lean fall (10 degrees), compared to able-bodied controls. They compared recoveries using either the left or right foot or affected or unaffected foot. The step length and recovery time was greater and longer in the amputee group compared to the control group, however they found that the amputees were comfortable leading with either their affected or unaffected foot [26].

Munjal [27] compared the effect of non-articulating ankle prostheses compared to hydraulic ankle attachment prostheses on minimal toe clearance (MTC) and risk of tripping. They found that the hydraulic ankle attachment foot provided a greater MTC on both the affected (2.07±0.63cm vs 1.76±0.85cm) and unaffected (2.27±0.63cm vs 2.12±0.91cm) sides. They suggested that the improved ground clearance during swing phase of the gait cycle was due to the ability of the hydraulic ankle foot complex to dorsiflex the toes. This in turn reduced the risk of tripping [27].

Rosenblatt [28] studied the effect of an active dorsiflexion prosthesis compared to the participants' own prosthesis with regards to risk of tripping while walking on a treadmill. They found that the minimum toe clearance possible was higher when using the ProprioFoot (28.8±1.6mm), compared to the standard feet (17.4±1.1mm) tested. It was suggested that with an active dorsiflexion foot, the risk of tripping was reduced. They suggested however that further research is required in order to quantify the extent to which an active dorsiflexion foot may reduce risk of falls [28].

2.3.2.2 Postural stability

Three articles focused on the postural stability of the amputees (table 2.4). One study investigated the difference in balance control on the affected and unaffected side while others determined the effect of mass perturbations on balance recovery and another determined foot placement strategies. Perturbations refer to small external forces applied to the body. The direction of the force applied is usually either applied medially or laterally but can be applied in an anterior or posterior direction as well.

Table 2.4 Summary of included articles referring to postural stability

Authors	n	Age	Time since amputation	Analysis	Main variables	Main finding
Curtze et al., 2012	15	55±10	2-44 yrs*	3D Motion Analysis	GRF, knee & ankle kinematics	Ankle strategy mainly used to maintain balance with A/P perturbation in UTTA
Segal et al., 2015	14	45±16	>1 yr	3D Motion Analysis	COP, GRF, hip & ankle angular impulse	Lower ankle impulse when using stiffer prosthesis compared to more compliant prosthesis
Selles et al., 2004	10	44±12	>1 yr	3D Motion Analysis	COM, hip & knee kinetics & kinematics	Addition of weight strips to the prosthesis increased hip and knee moments

Age given in mean years ± standard deviation
COP = Center of pressure, GRF = Ground reaction force, COM = Center of mass, UTTA = unilateral transtibial amputee

In a subsequent study, Curtze [29] investigated the effect of waist perturbations, which are external forces applied to the waist, on balance control in UTTA. They evoked waist perturbations using a pulley in a medial/lateral direction as well as an anterior/posterior direction. They found that the UTTA mainly used the ankle strategy, which is the ability to change the centre of pressure under the foot by relying on the muscles around the ankle joint, to aid recovery during a fall. They also found greater ankle moments in the unaffected limb, which they suggested was a means to help compensate for the reduced ankle control of the prosthesis [29].

Selles [30] attempted to determine whether the addition of weighted strips to the prosthesis would influence kinetic or kinematic variables while walking on a treadmill. They found that the weighted strips attached to the prosthesis resulted in a greater change of kinetic variables. More specifically, they noted greater joint torques of the hip and knee, while kinematic variables such as speed and stride length remained unchanged [30].

A study by Segal [31] investigated the influence of either a medial or lateral perturbation on the coronal ankle angular impulse, coronal hip moment and centre of pressure (COP). With a medial perturbation applied they found a decrease in coronal ankle impulse compared to an undisturbed step, however it did return to the undisturbed step by the first recovery step. The opposite was true with a lateral perturbation. In fact, two recovery steps were required to return the impulse to the previously undisturbed levels. They found that there was a lower coronal hip moment for the prosthetic limb compared with the unaffected limb and the control group. It was noted that neither medial nor lateral disturbances affected the COP excursion for the prosthetic limb [31].

2.3.3. Muscle activation during locomotion

One study investigating knee muscle activation of the vastus medialis and bicep femoris during locomotion was included for the systematic review (table 2.5). In this study, electromyography of the hamstrings and quadriceps muscles was performed during walking gait in UTTA. One other article included in this review was included in the last section on other factors affecting locomotion.

Table 2.5 Summary of included articles on muscle activation in unilateral transtibial amputees

Authors	**n**	**Age (years)**	**Time since amputation**	**EMG**	**Main variable**	**Type of Prosthesis**
Isakov et al., 2001	11	37±8	5–44 yrs*	VM, BF	Ratio of muscle activation during swing and stance of gait	BF muscle activation was less on AF side than the control

Age given in mean years ± standard deviation
VM = vastus medialis, BF = bicep femoris, AF = affected

There has been limited research into the muscle activation patterns in UTTA. A study by Isakov [32] found that the vastus medialis muscle reached peak activity later during the gait cycle of the affected side (8.84±4.80%), compared to the unaffected side (6.06±6.60%). They also found that the bicep femoris on the affected limb reached peak activity at 9.81±4.8% of the gait cycle, compared to the unaffected limb which peaked at 92.43±6.60% of the gait cycle. This was recorded as a significant difference ($p<0.05$) and showed asymmetry of knee muscle timing during gait. They lastly suggested that these asymmetries may be related to the stiffness of the prosthesis [32]. As mentioned previously, Powers [33] also investigated muscle activation. However, it was with reference to stair ambulation [33].

2.3.4. Other factors influencing locomotion

A total of five studies were included in this category (table 2.6). Three of the five articles were orientated around stair ambulation. One of them focused on the influence of the Seattle foot prosthesis while the second attempted to identify the foot placement strategy of amputees compared to able-bodied controls while navigating stairs. Another study focused on the comparison of a passive and powered prosthesis for stair ambulation whilst the second study listed in table 6 investigated the effect of the bone bridge amputations (osseous bridge between the tibia and fibula) compared to traditional amputations on gait.

Table 2.6 Summary of included articles of other factors influencing locomotion

Authors	n	Age (years)	Time since amputation	Analysis	Main variable	Main finding
Buckley et al., 2013	8	46±13	15.0±12.3 yrs	3D motion Analysis	Velocity, knee flexion	UTTA decrease velocity approaching obstacles & had less knee flexion in the lead leg compared to controls
Kingsbury et al., 2014	14	24±3	>1 yr	3D Motion analysis	Walking speed, vGRF	Bone bridge group had greater vGRF on AF than traditional surgical methods
Pickle et al., 2014	9	30±6	2±1 yrs	3D analysis	GRF, angular momentum	Increased sagittal angular momentum on AF compared to controls during stair ascent
Powers et al., 1997	10	51±15	>2 yrs	3D analysis & EMG of VL, RF, GMAX, SMEMB, BFLH, BFSH	Gait velocity, muscle activity	UTTA slower velocity than controls during ascent, Greater muscle activation for all muscles measured
Ramstrand & Nilsson, 2009	10	33-67*	5–48 yrs*	3D analysis	Velocity, step width	UTTA had lower velocity during stair ascent with greater step width than controls

Age given in mean years ±Standard deviation; * = Age given as a range
GRF = ground reaction force, UTTA = unilateral transtibial amputee, AF = affected, VL = vastus lateralis, RF = rectus femoris, GMAX = gluteus maximus, SMEMB = semimembranosus, BFLH = bicep femoris long head, BFSH = bicep femoris short head

There are several other factors that may have an effect on locomotion of UTTA, such as the type of surface, obstacles or stairs and even factors relating to the type of amputation.

Kingsbury [34] examined the difference in gait after a bone bridge amputation compared with a traditional amputation. The only significant difference found was that, during fast walking, the roll-off vertical ground reaction force (vGRF) was higher in that of the bone bridge group. They suggested that this aids the stability of the individual during terminal stance of the gait cycle [34].

Three studies investigated locomotion and risk of fall while ascending or descending stairs. Powers [33] investigated the kinematic, kinetic and muscular effort required when ascending stairs with a Seattle LightFoot2. They determined that UTTA had a significantly slower ascending speed compared to that of the control group (29.6m/min vs. 33.4m/min; $p<0.05$). They also found less dorsiflexion in the affected limb, compared to both the unaffected limb as well as the control group. In terms of muscle activity or effort required, measured using indwelling electromyography it was found that the UTTA muscles measured worked harder (% maximal muscle contraction) than that of the controls based [33].

Ramstrand [35] found that not only did UTTA have a slower ascending walking velocity; they also increased their step width to help improve the base of support, in comparison to the control group. While ascending the stairs UTTA also spent a greater proportion of time in double stance compared to the control group. During stair descent, UTTA exhibited a slower velocity and an extended time in double stance of the gait cycle. In both the amputee and control group, the percentage time spent in double stance was found to be longer while ascending stairs (ascent = 15% [Control], 21% [UTTA] vs descent = 11% [Control], 15% [UTTA]) [35].

A study conducted by Pickle [36] described and compared the use of a powered prosthesis compared to a passive prosthesis, in UTTA. These results were also compared to an able-bodied control group. It was found that in the sagittal plane the UTTA group had an increased angular momentum while the prosthetic limb was in stance phase compared to the control group. They also concluded that there was no significant difference between the use of a powered prosthesis and a passive prosthesis for stair ambulation.

Buckley [4] investigated gait parameters while walking over obstacles. It was found that the UTTA walked slower than the able-bodied controls when leading with the affected limb. It was also noted that the UTTA were required to place their foot closer to the obstacle before stepping over the obstacle with a shorter step length. This also meant that a greater knee flexion was observed compared to that of the able-bodied control group. They concluded that this difference was to aid in decreasing the foot contact angle and the loading effect on the affected limb. This is important to maintain stepping balance in order to prevent falling [4].

In summary UTTA adapt their movement in order to perform different tasks and maintain balance. This has been done by decreasing speed, increasing step width and adjusting joint kinematics.

2.4 Discussion

The purpose of this systematic review was to identify literature available on UTTA in terms of muscle activation patterns and biomechanical parameters during walking gait, as well as balance and movement symmetry. External factors influencing the movement capabilities of the population during daily activities were also identified through the review.

Factors influencing movement of unilateral amputees are important to understand. While it can be acknowledged that there are differences between the affected and unaffected sides, the extent and mechanism of this is important to understand. This systematic review has highlighted that UTTA have asymmetrical biomechanics during gait. Decreased hip internal moment and abduction moment have been found during stance phase of gait while hip extensor moments were found to be greater on the affected side compared to unaffected side [13–15]. Decreased force production was observed at the ankle and knee of the affected side compared to the unaffected side [15] and decreased power production at push-off was also noted on the affected side in comparison with unaffected side [16]. Increased flexion at the hip and knee was observed on the affected side during early stance of the gait cycle [16]. Spatiotemporal differences were also noted. Step execution duration during gait initiation with the affected limb was longer [17]. Increased speed resulted in greater knee angles on the unaffected side during the loading phases as well as toe-off [18]. Another study experimented with the speed of walking of amputees and the number of steps needed to stop. It was found that by altering the acceleration or deceleration phase amputees were able to stop within two steps [19]. Lastly, it was found that when walking on a cross slope with the affected side on the upper section, greater hip and knee flexion was used to ensure ground clearance [20].

Some of these findings have been suggested to be mechanisms for the protection of the stump. Greater loading asymmetry may help to decrease pain experienced through the stump [13]. The inability of the ankle to produce power at push-off may be due to the prosthesis characteristics [16]. While some of these differences may be for protection of the joints and muscles others may be due to poor mechanics or properties of the prostheses. It is therefore important to understand if these differences may have a positive or negative influence on the movement ability UTTA.

The importance of these findings are valuable for rehabilitative purposes, aiding technique training and confidence building to be able to walk across a sloped area without the fear of falling. What is unknown is how muscles are activating to control the movements or how they are compensating and this warrants further research.

Several articles in this systematic review specifically examined the influence of different prostheses on walking gait. No significant differences were found for gait parameters when various weighted components were added to the prosthesis [21]. It was suggested that the Talux foot better simulates the ankle motion of an unaffected foot [22]. Supan [22] also found that it may be possible to adjust the heel height of the prosthesis without altering the biomechanical loading on the body [22]. Another study suggested that the Talux foot resulted in improved work asymmetry in amputees that is able or has the potential to navigate small curbs, stairs and uneven surfaces [24]. Powered prostheses were shown to better regulate the force production in amputees so that they more closely compared to able-bodied controls [23]. Literature has also observed the influence of medially or laterally directed perturbations on the Seattle LightFoot2 while walking [25]. It was found that the UTTA recovered in fewer steps with lateral perturbations compared to medial perturbations.

In summary, these articles suggest that the design of the prosthesis, to better mimic the function of an unaffected foot, may help to reduce asymmetry as well as decrease the risk of falls in this population. Improved prosthesis selection would also be an important factor to consider. Further research may be important to determine the influence of different prostheses on activities that are performed on a daily basis, other than walking.

The risk of falling is often a concern within this population and there has been research into balance recovery after an applied external perturbation while walking. What is clear is that UTTA take longer to recover and use a greater step length to do so, compared to an able bodied control group [26]. There has also been a study that investigated the risk of falling using a hydraulic ankle [27]. It was observed that with the use of the hydraulic ankle there was minimal toe clearance due to increased dorsiflexion possible at the ankle. This therefore decreased the risk of tripping that would result in a fall [27]. Further supporting this was a study investigating the use of active dorsiflexion prostheses where they found that it contributed to improved ground clearance [28]. Both studies provided evidence to support the hypothesis that active dorsiflexion ankle prostheses allow for improved ground clearance, therefore helping to reduce the risk of tripping.

The maintenance of postural stability is also important in decreasing the risk of falls as well as to prevent overcompensation or injury. Research has shown that waist perturbations result in UTTA using the ankle strategy to maintain balance. This was found to be greater on the unaffected side which was suggested to help compensate for the lack of control possible by the prosthesis [29]. Further research has shown that mass perturbations either in a medial/lateral or anterior/posterior direction resulted in greater changes in kinetic variables specifically greater moment acting at the hip and knee [30]. Decreased coronal ankle impulse was noted with medial perturbations during walking however UTTA recovered within one step while with lateral perturbations two steps were required for recovery. They also noted a lower hip coronal moment on the affected side but there was not change in the COP for either the medial or lateral perturbations [31].

This could infer that while the COP remains unchanged during the lateral perturbation the maintenance of balance or correction for loss of balance needs to be controlled for. This may be possible by either shifting weight to the unaffected side during the two recovery steps or by a change in joint kinematics such as increased trunk lean during the recovery steps. Further research would be advised to determine the difference in muscle activation and postural strategies needed for single leg balance compared to walking or other daily activities.

Overall, the above mentioned studies suggest that postural stability can be trained to help decrease the risk of falling and improve the ability to recover from a fall with either their prosthetic limb or unaffected limb.

Limited research has identified how skeletal muscles are recruited to perform various movements. What has been observed is that during gait the vastus medialis muscle peaks later on the affected side than the unaffected side and the bicep femoris muscle peaks significantly earlier on the affected [32]. The asymmetry found when the vastus medialis and bicep femoris muscles activate may be attributed to the stiffness of the prosthesis. It is unknown as to how muscles work together above and below each of the joints of the ankle, knee and hip. It would be valuable to determine muscle activation patterns for different movements to determine if there are muscle groups that have a common weakness in UTTA and how this may influence the UTTA longitudinally.

There are several other factors that may have an effect on locomotion of UTTA, such as the type of surface, obstacles or stairs and even factors relating to the type of amputation. This review of the literature has highlighted that a bone bridge amputation may contribute to greater stability during terminal stance phase [34]. It has also highlighted that UTTA have a decrease velocity when either ascending or descending stairs. More specifically the UTTA may increase the step width to increase the base of support and will spend a longer period of time in double stance phase [35]. The muscles of the lower limb of the UTTA were observed to work significantly harder than that of the control group [32]. The muscles measured were limited to that of the gluteus maximum, semitendinosus, bicep femoris long and short heads, vastus lateralis and rectus femoris [36]. No significant differences were noted between a powered prosthesis and passive prosthesis during stair ambulation. An increase in angular momentum was noted when the affected side was in stance phase. Obstacle navigation is challenging for most but the use of prosthesis may make this more challenging. UTTA were found to decrease their speed and step length when leading with their affected limb and stepped closer to the obstacle before stepping over it. Greater knee flexion angles were also measured when navigating obstacles. Due to the high risk of falling, it may be beneficially to determine if training programmes may influence the ability and confidence of the UTTA to improve the way as well as the speed at which they can navigate obstacles.

This systematic review highlighted that there are several factors that influence the biomechanics of movement in UTTA. While there was substantial evidence that there is asymmetry as well as significant differences in a number of kinematic and kinetic variables it follows mostly a reductionist approach by measuring individual variables independent from a system. The dynamic systems theory is often used in adapted movement programmes as a theory where the task, organism and environmental constraints are seen in a holistic way to develop rehabilitation programmes [37]. It may therefore be important to perform biomechanical analysis with the addition of EMG to holistically identify how the different parameters and variables influence the performance done. The importance of understanding muscle activity during the above mentioned findings would also be critical if rehabilitation methods are to be enhanced.

2.5 Clinical and research implications

The implications of gaining a better understanding of how different movements and environmental factors affect the gait characteristics and capabilities of UTTA identify research gaps and therefore allow for improved focus in future research. A more complex research design involving a comprehensive biomechanical analysis including muscle activation patterns of multiple joints and surrounding pertinent musculature during functional activities may be beneficial. A greater understanding of the biomechanical strategies used by UTTA may assist to improve functional capabilities, direct rehabilitation strategy and eventually quality of life for UTTA. Furthermore, further research may assist in the assessment and better prescription of a walking prosthesis, as well as focussed training plans to improve factors associated with daily prosthesis use. The immediate implication of the review is to design a biomechanical study investigating kinematic, kinetic variables as well as muscle activation during functional activities in UTTA.

Table 2.7 Key findings of this review and suggestions for future research

Known	Unknown
Asymmetry - vGRF, kinetic and kinematic variables (gait and stairs)	Movement strategies and muscle activation patterns used for activities of daily living
Risk of fall/poor postural stability (prosthesis)	Effectiveness of training programmes
Type of prosthesis influences amount of dorsiflexion – Impacts ground clearance	Guidelines for the prescription of prosthesis
Many factors influencing the movement capabilities of UTTA (amputation, prosthesis, rehabilitation, pain)	Rehabilitation guidelines based on prosthesis and type of amputation

vGRF = vertical ground reaction force, UTTA = unilateral transtibial amputees

2.6 Limitations

One of the limitations of this review was the difficulty in the accessibility of several articles. Unfortunately, while articles were requested via interlibrary loan or directly from the authors, not all were available. There were also several articles that had to be excluded based on lack of evidence of ERB (Ethics Review Board) approval.

2.7 Conclusion

From this review we can ascertain that the loss of a limb causes both asymmetry and results in the compensation of muscles and joint control during locomotion and can lead to overloading of the unaffected limb with possible adverse long term health outcomes. It is also clear that further research is required to gain a more holistic understanding of movement and simultaneous muscle activation patterns to explain the findings of the kinematic and kinetic research performed. Furthermore, most studies in UTTA are conducted during gait cycle. It is clear that there may be a need to further investigate other functional activities including single leg balance, sit-to-stand, stand-to-sit and stair ambulation with the use of both electromyography as well as 3D motion analysis in order to fully understand the effect of amputation on daily functionality on this population group.

2.8 Acknowledgements

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Chapter 3

Muscle activation patterns during single leg balance of unilateral transtibial amputees in comparison to controls

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This article was prepared for possible publication in the Gait & Posture journal. The submission guidelines were observed, however for ease of reading, the tables have been kept within the text and the left and right margins have not yet been set. The specified journal reference style used was the Elsevier – Vancouver method. We were limited to 3000 words by this journal and currently have used 2763 words. The abstract was limited to 250 words of which we have used 247 words.

Abstract

Background

Single leg balance (SLB) is a task that is required daily in both a static and dynamic form. Static balance in lower limb amputees is affected due to limitations of the prosthesis and compensatory biomechanics. Limited research on the mechanisms of adaption and muscle activation patterns in unilateral amputees is available. The aim was to determine the muscle activation patterns of unilateral transtibial amputees (UTTA) during SLB.

Methods

A cohort of 25 participants (12 amputees and 13 controls) was recruited. Surface electromyography (EMG) was measured for seven muscle groups together with the joint kinetics and kinematics of the hip, knee and ankle. Participants were required to perform three trials of SLB on each side.

Results

Results indicated significantly greater muscle activations ($p<0.05$) of the affected side (AF) for lumbar erector spinae (LES), vastus lateralis (VL), gluteus medius (Gmed), gluteus maximus (Gmax) and bicep femoris (BF) in comparison to the unaffected side (UN) of the UTTA as well as the non-dominant side (ND) of the controls. The LES and BF muscle activation of AF was significantly greater than the dominant side (D) of control group ($p<0.05$). Hip moments, powers and flexion angles as well as knee power and flexion angle were greater on AF than UN ($p<0.05$).

Conclusions

These findings suggest that UTTA may be predominantly using the muscles surrounding the AF hip to maintain postural control during SLB. This in conjunction with greater hip moments, powers and angles needs to be considered for the purpose of injury prevention and rehabilitation.

[247]

Keywords: Unilateral transtibial amputee, single leg balance, muscle activity, postural control

3.1 Introduction

Static and dynamic balance is required throughout daily activities, for spacial awareness and the control of movements [1]. A single leg balance (SLB) test can often provide insight into how the body maintains an upright position as well as the difference between the two sides of the body [2]. This information is invaluable for the determination of the effects during dynamic task of daily activities as well as the risk of falls [2,3].

Maintenance of balance may be controlled via different strategies including but not limited to different joint strategies and postural sway [1,4]. The purpose of these strategies is to maintain the centre of mass (COM) over the base of support [1]. The ankle strategy is most commonly used as the ankle joint is the first joint that receives feedback from the ground through foot contact [5,6]. This is however not possible in the case of UTTA [7]. In UTTA the use of a prosthesis and the loss of musculature influence the balance strategy possible [5,7].

Muscle co-ordination plays an important role in efficient joint control as well as the reduction of stress on the joints [8]. This is true for SLB in order to maintain joint stability [9]. Furthermore, SLB performance has been used as a predictor for ambulatory control and risk of falls [7]. It is therefore of importance to better understand how UTTA maintain balance.

A better understanding of the muscle activation used during SLB could potentially guide improved rehabilitation methods, which in turn may reduce the risk of falls and improve the confidence to perform daily activities [3]. While there is some literature relating to the muscle activation patterns in UTTA during SLB, it is limited in the number of muscles observed whilst the majority of research has focused on gait parameters [1,10]. Furthermore, studies have focused mainly on the changes in centre of pressure (COP) during static and dynamic balance while little is known about the muscle activation patterns that influence the static balance strategy used in UTTA [1,9].

The aim of the study was to determine muscle activation patterns in UTTA during the SLB, and compare the affected side (AF) and unaffected side (UN) of the UTTA and age matched controls (CON).

3.2 Methods

Participants

The study included 25 participants (12 UTTA & 13 CON) between the ages of 18 and 65 years. Participants in the UTTA group had undergone a transtibial amputation at least one year previously; were able to ambulate with a walking prosthesis without any assistance; were healthy and free from injury; were free from movement limiting stump sores and without any uncontrolled secondary conditions [11]. Recruitment into the study was independent of the type of prosthesis that the participant was using. The control group (CON) participants consisted of healthy age-matched individuals free of injury and illness at the time of testing. Ethical approval was obtained from the Human Research Ethics Committee 2 at Stellenbosch University (M16/08/032) and informed consent was obtained from all participants.

Procedures

Bipolar surface electromyography (EMG) was recorded using the wireless desktop DTS receiver and DTS 16 channel lossless sensors running MyoResearch 3.10 software (Noraxon, UK). Electrode placement followed the surface electromyography for the non-invasive assessment of muscles (SENIAM) guidelines. The muscles identified and tested included that of bilateral lumbar erector spinae (LES), gluteus medius (Gmed) and gluteus maximus (Gmax), bicep femoris (BF), vastus lateralis (VL) and unilaterally (unaffected side), medial gastrocnemius (MG) and tibialis anterior (TA). The control group included bilateral placement for the MG and TA. Each placement area was shaved and cleaned using ethanol. Two electrodes were placed with an electrode centre distance of 2 cm. Signal to noise ratios were measured to ensure the quality of the data before tests were performed.

Kinematic and kinetic data were collected with the use of the Vicon 3D motion analysis system (Vicon, UK) comprising eight MX-T20 cameras. Furthermore, three Bertec floor imbedded force plates were used (two FP4060-07 models and one model FP6090-15 (Bertec, USA)). Reflective markers with a 7mm diameter were placed according to a modified Helen Hayes marker set. Markers were placed on the sternoclavicular notch, Xiphoid process, 7^{th} cervical vertebrae, 10^{th} thoracic vertebrae, bilateral posterior superior iliac spine and anterior superior iliac spine, lateral and medial epicondyles of the femur, fibula head, lateral and medial malleoli. Markers were also placed at the site of the first, second and fifth metatarsal heads, as well as the heel, lateral heel and medial heel. Technical markers were placed on the flat surface of the shin and slightly distally to that. Technical thigh cluster markers were placed bilaterally consisting of a marker over the proximal lateral third of the thigh, a marker distally and posteriorly and one distally and anteriorly so as to form a non-equilateral triangle. Markers were placed on the prostheses at positions correlating to those of the unaffected side. Knee markers were found based on the estimation of the joint centre and compared to the unaffected side.

Test procedures

Participants were instructed to stand with each foot on a separate floor-imbedded force plate and to place their hands one on top of the other on their abdomen just below the Xiphoid process, to avoid covering of the markers. They were then instructed to first shift their weight to stand on their unaffected or dominant leg for up to a maximum duration of 20 seconds. The trial was initiated when the other foot left the floor and the trial ended when the foot made contact with the floor again. Three of the best trials for each participant were used for analysis purposes with trials while trials shorter than three seconds were discarded or if a breach in protocol was noted. Trials alternated between the unaffected side (UTTA) or dominant side (CON) and the affected side (UTTA) or non-dominant side (CON).

Data reduction & analysis

Raw EMG data were pre-processed using a Butterworth band pass filter with a band pass frequency of 20-500Hz. The signals were then smoothed using Root-mean-square (RMS) method with 50ms smoothing window. The filtered EMG of each muscle was normalised to a percentage of functional maximum voluntary contraction (peak dynamic activation) for that muscle. The absolute maximum activation level of each muscle over all the trials, recorded while performing the functional movements, was taken as the 100% functional maximum voluntary contraction for a particular muscle. EMG onset/offset calculations used a threshold value of 5% functional maximum voluntary contraction where the level of activation had to be above or below threshold for at least 0.05 seconds before it was considered to be on or off, respectively. The average activation levels were calculated for the EMG data.

Force plate data were filtered using a Butterworth fourth order (zero lag) low-pass filter with a cut off frequency of 100Hz. Kinematic data (Standard Plug-in-Gait model outputs) were filtered using a fourth order (zero lag) Butterworth low pass filter with a cut off frequency of 6Hz. Force plate data were registered above a threshold of 20N. Vertical GRF and joint moment and power data were normalised to body weight for comparative purposes. The means, maximum, minimum, and percentage of the movement cycle where each of these points occurred were determined for joint angles, joint moments and joint powers.

Statistical analysis

Statistica (version 13.2; Dell, USA) was used to perform a statistical analysis of the data that were visually inspected for normality. All data were found to be normally distributed. Mixed model repeated measures analysis of variance (ANOVA) were used to determine differences between the UTTA and CON group as well as each limb side (affected side (AF) and unaffected side (UN), dominant side (D) and non-dominant side (ND) respectively) for each of the different variables. Fisher's least significant differences (LSD) as a post hoc test were used in the case of statistical significance. An alpha (α) level of 0.05 was selected. Data is presented as means and standard deviations ($\bar{x} + SD$).

3.3 Results

The UTTA group consisted of nine men and three women with the average age of 34±10 years, while the CON group consisted of nine men and four women with the average age of 34±11 years. The UTTA group had an average height of 1.78±0.10m and a body mass of 73.42±16.28kg. The CON group had an average height of 1.78±0.08m and a body mass of 77.52±15.75kg. The average time since amputation was 10±7 years. Table 3.1 indicates the different prostheses used by the UTTA participants.

Table 3.1 Types of prostheses used by participants (n=12)

Number of participants	Type of Prosthesis
3	Variflex XC, Össur
1	Variflex with EVO cat 4, Össur
1	Reflex rotate with unity, Össur
1	Elevation foot, Össur
1	Proflex XC, Össur
2	Rush high pro, Ability dynamic
1	Proflex, Össur
1	Ottobock 1D35 Dynamic motion
1	Profiled prosthesis, Össur

3.3.1 Muscle activation

In the LES muscle, AF had significantly greater activation (7.51±5.10%) than UN (2.33±1.21%, p<0.05). The AF LES also had greater activation than ND (1.82±1.32%, p<0.05) and D (2.14±0.91%, p<0.05). The AF, VL muscle activation (5.3±4.31%) was significantly greater than UN (1.96±1.40%, p<0.05). The Gmed muscle group had significantly greater activation on AF (13.32±6.58%) than UN (5.10±2.50%, p<0.05) and ND (7.45±4.38%, p<0.05). In Gmax muscles AF (5.85±4.64%) had significantly greater activation than UN (2.12±2.27%, p<0.05) and ND (1.35±0.88%, p<0.05). However, it was not significantly greater than D (2.65±1.48% p>0.05). In the BF muscle there was a significantly greater activation for AF (14.56±8.30%) compared to UN (5.83±3.80%, p<0.05) as well as ND (3.67±3.09%, p<0.05) and D (2.65±1.48%, p<0.05). No significant differences were seen in the MG or TA muscle groups between the different leg conditions (p>0.05) (figure 3.1).

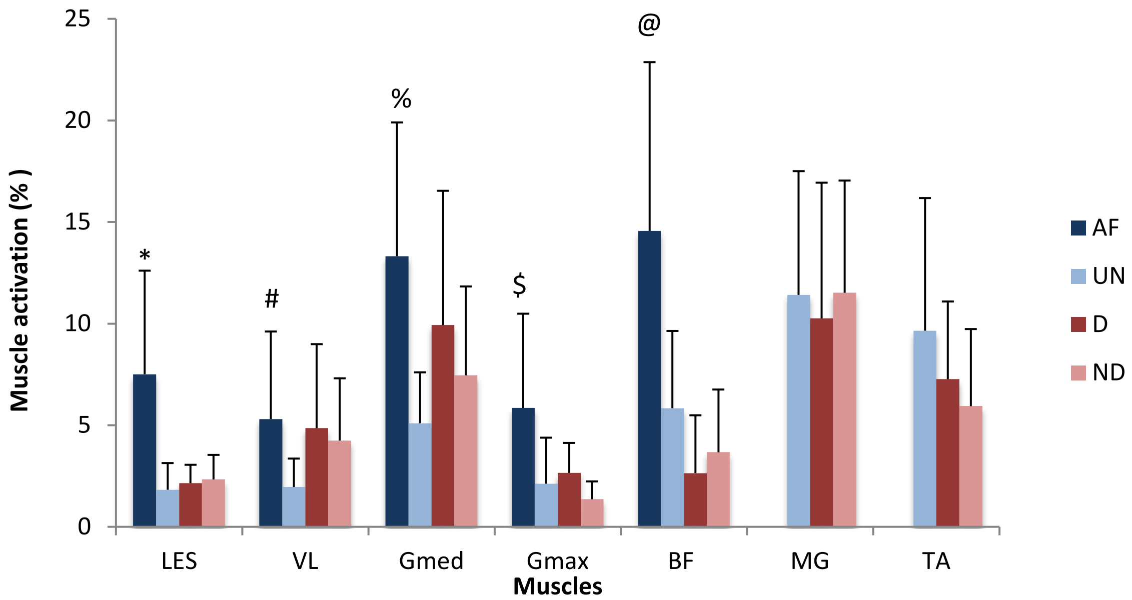

Figure 3.1 Average muscle activation (%) during SLB ($\bar{x} + SD$)
SLB = Single leg balance, AF = affected, UN = unaffected, D = dominant, ND = non-dominant, LES = lumbar erector spinae, VL = vastus lateralis, Gmed = gluteus medius, Gmax = gluteus maximus, BF = bicep femoris, MG = medial gastrocnemius, TA = tibialis anterior, SD = standard deviation, * AF significantly greater than UN, ND and D in LES (p<0.0001), # AF significantly greater the UN (p=0.0027), % AF significantly greater than UN (p=0.0001) and ND (p=0.0243),$ AF significantly greater than UN (p = 0.0043) and ND (p = 0.0095), @ AF Significantly greater the UN (p=0006), ND (p=0008) and D (p=0003)

3.3.2 Moments

A positive hip moment indicates a flexion moment while the negative indicates an extension moment. The hip moment of AF (0.15±0.23N.m.kg^{-1}) was significantly different than UN (-0.16±0.12N.m.kg^{-1}, p<0.05). It was also significantly different from ND (-0.29±0.24N.m.kg^{-1}, p<0.05) and D (-0.25±0.3N.m.kg^{-1}, p<0.05) (figure 3.2). No significant differences were observed for the knee or the ankle (p>0.05).

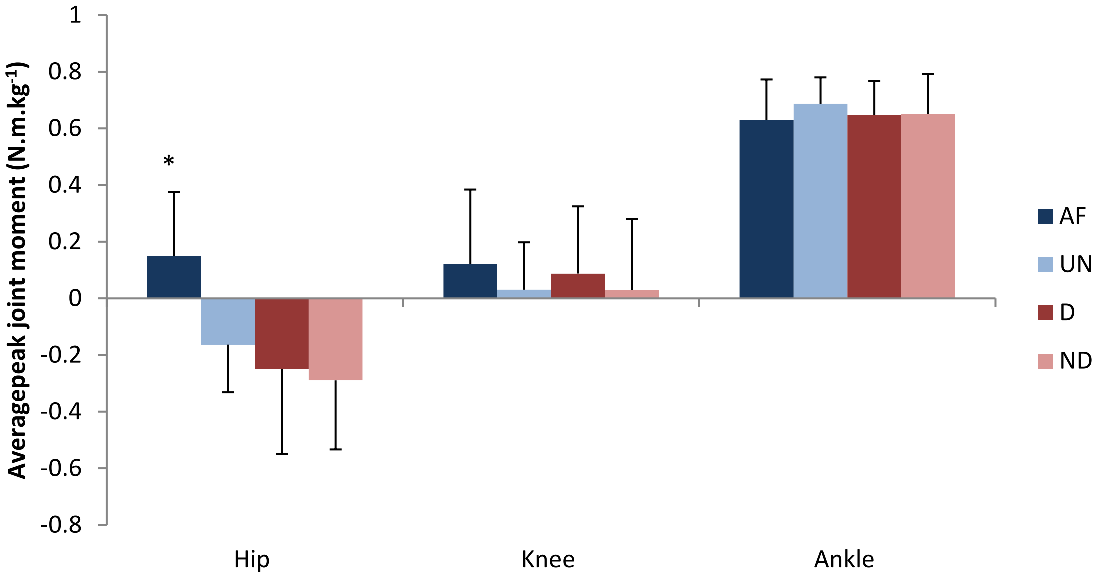

Figure 3.2 Joint moments (N.m.kg^{-1}) in the sagittal plane for the hip, knee and ankle during SLB ($\bar{x} + SD$)
SLB = Single leg balance, AF = affected, UN = unaffected, D = dominant, ND = non-dominant, SD = standard deviation, * AF significantly different than UN (p<0.0001), D (p= 0.0002) and ND (p<0.0001)

3.3.3 Power

The hip of AF (0.24±0.21W.kg^{-1}) produced significantly greater concentric power than UN (0.04±0.04W.kg^{-1}, $p<0.05$). The knee concentric power produced by AF (0.15±0.021W.kg^{-1}) was significantly greater than UN (0.02±0.02W.kg^{-1}, $p<0.05$) and ND (0.04±0.05W.kg^{-1}, $p<0.05$). No significant differences were found for the ankle ($p>0.05$) (figure 3.3).

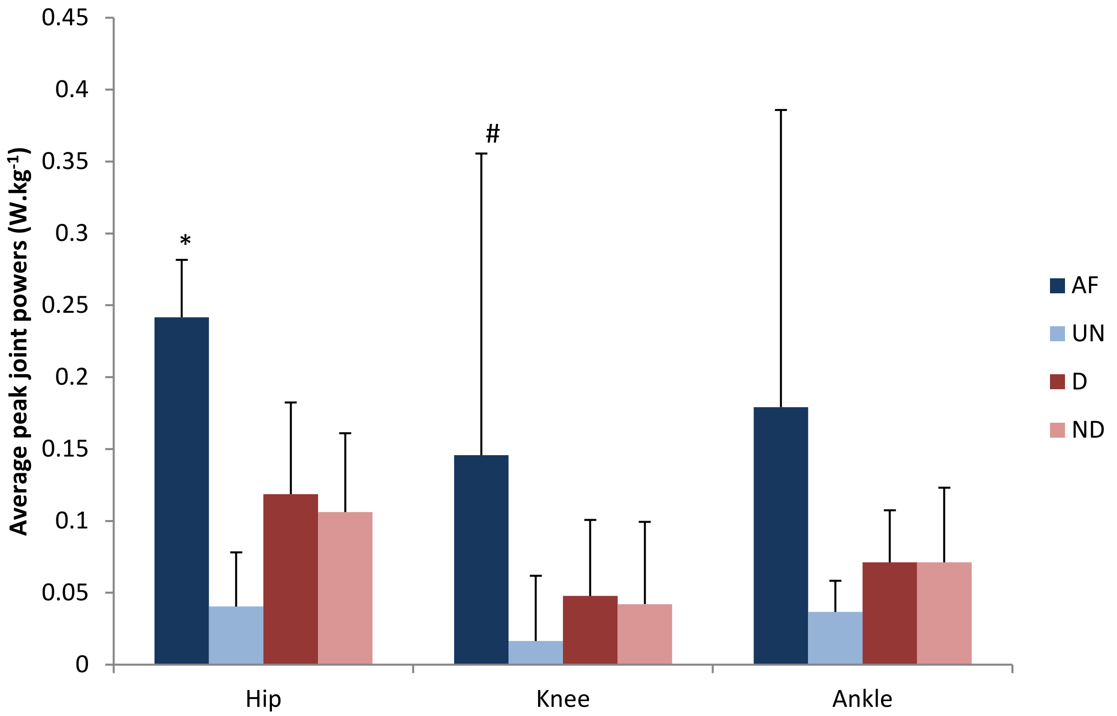

Figure 3.3 Joint power (W.kg^{-1}) for the hip, knee and ankle during SLB ($\bar{x} + SD$)
SLB = Single leg balance, AF = affected, UN = unaffected, D = dominant, ND = non-dominant, SD = standard deviation, * AF significantly greater than UN (p=0.0022), # AF significantly greater than UN (p=0.0069) and ND (p=0.0400)

3.3.4 Kinematics

A significantly greater hip flexion angle on AF (28.18±9.70°) was found compared to UN (15.41±6.84°, p<0.05). The hip flexion angle was also significantly greater than ND and D (8.96±9.84° and 11.05±10.72° respectively, p<0.05). There was a significantly greater knee flexion angle noted on AF (21.89±10.39°, p<0.05) compared to UN (11.59±4.34°, p<0.05) as well as ND and D (9.57±9.47° and 12.39±8.20° respectively, p<0.05). No significant differences were found for the ankle kinematics (p>0.05) (figure 3.4).

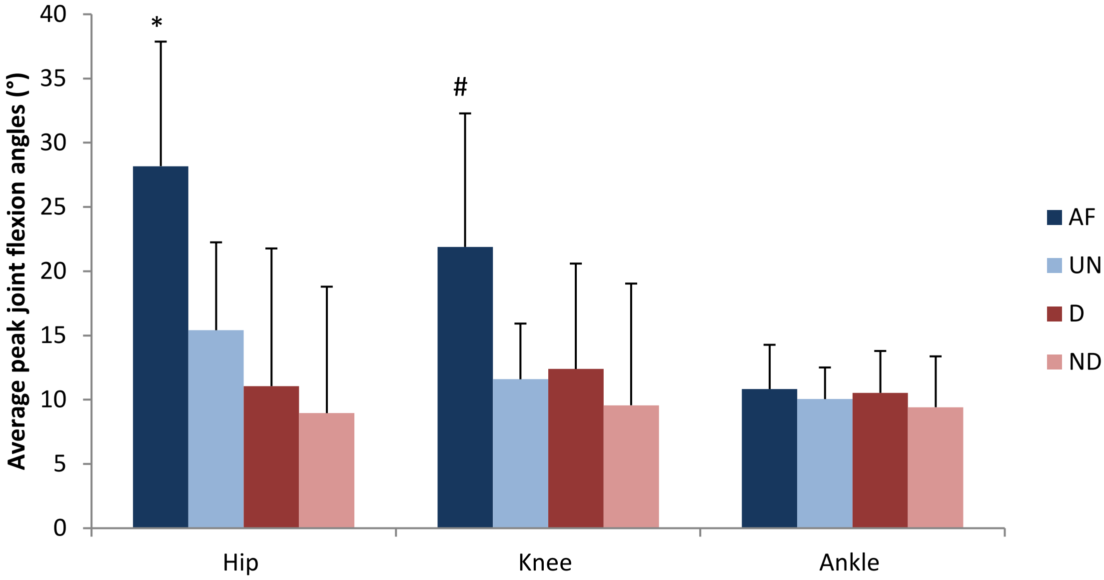

Figure 3.4 Joint flexion angles (°) for the hip, knee and ankle during SLB ($\bar{x} + SD$)

SLB = Single leg balance, AF = affected, UN = unaffected, D = dominant, ND = non-dominant, SD = standard deviation, * AF significantly greater than UN (p<0.0001), D (p<0.0001), ND (p<0.0001), # AF significantly greater than UN (p<0.0001), D (p=0.0029), ND (p=0.0003)

3.4 Discussion

The aim of this study was to determine the muscle activation of UTTA during SLB. Comparisons of muscle activation levels used between AF and UN, as well as with able-bodied CON were made. Biomechanical data with regards to joint powers, joint moments and joint angles were used to gain greater understanding of some of the strategies used by UTTA.

The main finding of the study was that the muscle activation of all the functional muscles of AF was significantly greater than all the corresponding muscles of UN. These results suggest that UTTA use different balance strategies when standing on AF compared to UN. The UN has the advantage of the use of the MG and TA muscles which are involved in the ankle strategy to maintain joint control and balance. The ankle strategy is most commonly used as the ankle joint is the first joint that receives proprioceptive feedback from the ground through foot contact [5,6]. This is however not possible in the case of prosthesis use for AF due to the lack of musculature and ankle joint control [7,12]. Previous research by Mouchino [13] and Viton [12] have investigated the standing lateral leg lift to 45° and measured the muscle activation of seven muscles including the Gmed, tensor fasciae latae, VL, MG and TA [12,13]. Both studies following this protocol found that on UN the MG and TA muscles were the primary muscles active to control balance but when standing on AF there was a shift to the TFL muscle. They suggested that balance was therefore primarily controlled through the use of a hip strategy although a knee strategy could have contributed to the overall control [12,13]. This is congruent with the results of this study. It has also previously been found that the Gmax muscle is primarily responsible for hip abduction and extension [17]. If this is true, the greater Gmax activation measured for AF during SLB may be related to the constant need to co-ordinate the muscle control around the hip joint in order to maintain balance.

In comparison to the control group, AF of the UTTA had significantly greater muscle activation for the LES, Gmed, Gmax and BF than the ND. The AF, BF was also significantly greater than D for AF. The increased muscle activation of the LES, Gmed and Gmax muscles suggest that more work is being done to control the movement around AF, hip in order to maintain balance compared to UN and CON. In an article by Van Deun [14] investigating muscle activation patterns of individuals with chronic ankle instability, significant differences were found when weight was transferred onto the injured side for a SLB compared to the uninjured side [14]. They investigated the Gmed, TFL, VL, vastus medialis obliquus, medial hamstrings, TA, peroneus longus and MG activation and timing and found that the onset of muscle activation for the chronic ankle injury group was in the muscles surrounding the hip and knee rather than at the ankle as seen in CON [14]. This is similar to what was found in our study where the muscle activation patterns were adapted to use more of a hip strategy to maintain balance when standing on AF. Due to the loss of GM activation of AF, the control strategy changes to adopt more hip control with greater activation occurring around the hip when standing on AF [9].

The biomechanical data indicated a significantly greater hip flexion moment as well as hip power for AF. Greater knee power as well as increased knee flexion was also observed for AF. No statistical significant difference was found in ankle joint angles. The biomechanical results support the theory of a predominant hip strategy being used for AF of the UTTA to maintain balance. Furthermore, this suggests that the hip strategy is utilised for the control of COM over the prosthesis during SLB [5]. A greater hip flexion moment has been shown to be associated with the use of the hip strategy as well as greater hamstring activity [4]. A previous study showed that only two UTTA were able to stand on AF leg during a SLB test while the remainder of the participants were unable to do so [15]. With the improvement of prostheses technology, balance on AF is now more possible [16]. Devan [17] mentioned that in UTTA greater hip flexion of AF has been noted during early and late stance of walking as a means to maintain the COM of their base of support (BoS) [17].

The integration of the muscle activation patterns with the kinetic and kinematic data indicates the use of the hip strategy for AF. While not specific to static SLB, Devan's [17] review supports the need for the hip and surrounding muscles to compensate for the loss of ankle muscles in UTTA.

3.5 Limitations

Optimal setup up of the prosthesis was not evaluated. In future studies, it is recommended that a detailed description of prosthesis alignments is documented for possible interpretation of data. Furthermore, marker placement on the prosthesis was based in biological estimations and due to the variation in the prostheses was not identical. There was a limitation related to marker placements on the prosthesis as they were based on biological estimations and due to the variation in prostheses were not identical. The balance ability of the UTTA was limited of AF and thus data could only be captured for a limited time, however no less than three seconds of data was captured. Lastly, the muscle activity was observed for specific muscles and due to practical limitations other muscle activations that may contribute to the movement or movement control cannot be accounted for. The study included amputees with different types of prostheses. Each participant's personal prosthesis was used as this was the prosthesis that they were familiarised with. While the inclusion of different prostheses in the study may be seen as a limitation, we specifically chose this to provide true life scenarios.

3.6 Future research

Limited knowledge is available on the possible impact of different rehabilitation protocols and exercise prescription on the possible strategies UTTA can develop during balance. Furthermore, the impact of different possible strategies on the muscle activation patterns and possible overload should be investigated. Lastly, research on the impact of various types of prostheses on SLB and the related muscle activation strategies can support prosthesis choice for an UTTA.

3.7 Conclusion

In conclusion, the results of this study contribute new findings to the field. Muscle activation of the LES, Gmed and Gmax of AF, along with AF, hip kinetic and kinematic results, indicated the use of the hip strategy for the maintenance of postural control during SLB. It indicates that during SLB the muscles surrounding the hip are working harder than UN and could result in injury or muscle imbalances around the hips. Specific training to reduce the overloading around the hip may be beneficial in improving movement and muscle symmetry for UTTA, thus further research in this important area is warranted.

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Chapter 4

Biomechanical analysis of the sit-to-stand-to-sit activity in unilateral transtibial amputees

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This article was formatted for the possible submission to the journal of Prosthetics and Orthotics International. For the ease of reading, the tables and figures have been left in the text. The journal requested reference style was that of the Sage – Vancouver method. We were limited to 3000 words by the journal and have currently used 3076 words. The abstract was limited to 200 words of which we have used 192 words.

Abstract

Background

The sit-to-stand (SiSt) and stand-to-sit (StSi) movement are repetitive daily activities which include loading of the hip, knee and ankle joints on each occasion. It is hypothesized that in amputees, compensation due to asymmetrical movements, can lead to joint overloading. This study aimed to describe the muscle activation levels and biomechanics during the SiSt and StSi movements in unilateral transtibial amputees (UTTA).

Method

Participants included 12 UTTA and 13 controls. Surface electromyography (EMG) was recorded during ten continuous sit-to-stand-to-sit repetitions. Three dimensional biomechanical data were recorded.

Results

Muscle activation was significantly greater for UTTA lumbar erector spinae ($p<0.05$) during the SiSt and tibialis anterior ($p<0.05$) during SiSt and StSi in comparison to the controls while affected side vastus lateralis activation was significantly lower than the unaffected and control ($p<0.05$). Affected hip, unaffected knee and ankle had greater joint moments and powers for SiSt and StSi ($p<0.05$).

Conclusions

Low muscle activation of the vastus lateralis on the affected side could be an indication of poor knee control during the SiSt and StSi. The AF hip and UN knee and ankle are overcompensating to perform the SiSt and StSi.

Clinical relevance

The key findings of this study indicated possible compensatory mechanisms during the SiSt and StSi. These mechanisms may use asymmetrical muscle activity and place the amputee at increased risk for joint degeneration. Rehabilitation specialists should consider specific training of daily function activities to improve symmetry, reduce compensatory patterns and enhance normal muscle and joint co-ordination.

Keywords: Amputee, sit-to-stand, stand-to-sit, muscle activity, asymmetry

4.1 Introduction

Unilateral transtibial amputees (UTTA) require extensive physical rehabilitation to help return functional movement capacity. The use of a prosthesis aims to reduce the loss of function by improving the possibility to perform daily functional tasks[1]. However, a prosthesis is unable to mirror the full biological nature of the original lower limb due to its mechanical limitations and characteristics[2]. A major concern in UTTA is that the use of a prosthesis and compensatory strategies employed by the individual may lead to over-exaggerated loading on the unaffected side (UN).

It is thought that unilateral transtibial amputees (UTTA) adapt their movement patterns when learning to ambulate with a prosthesis. New motor control patterns need to be learnt in order to adapt to the new constraints imposed by the use of a prosthesis as well as the limited residual muscle mass. The sit-to-stand (SiSt) or stand-to-sit (StSi) are activities of daily living that are more demanding on the muscles and joints than walking due to the direction of forces applied through the joints and the repetitive high loading nature of the movement pattern required[3]. Due to the SiSt being performed several times per day UTTA may suffer repetitive joint overloading when asymmetrical weight distribution is experienced each time the movement is performed[4,5].

During walking, higher ground reaction forces (GRF) have been documented on the unaffected limb compared to the affected limb of UTTA[6]. This supports the presence of asymmetry during movement which results in greater joint loading, increasing the risk for injury and long term joint conditions[7]. It has been found that continuous joint overloading may lead to joint degeneration including osteoarthritis or other musculoskeletal injuries[7–10].

Muscle activation patterns and range of motion of the joints of the lower limb have not been described together for UTTA during the SiSt or StSi. The aim of this study was to describe and compare the muscle activation levels, joint kinetics and kinematics during the SiSt and StSi between UTTA and able-bodied controls.

4.2 Methods

Participants

The study included 25 participants (12 UTTA & 13 CON) between the ages of 18 and 65 years. Participants in the UTTA group had acquired the amputation at least one year prior to testing; were able to ambulate with a walking prosthesis without any assistance; were healthy and free from injury; were free from movement limiting stump sores and without any uncontrolled secondary conditions[11]. Recruitment into the study was independent of the type of prosthesis that the participant was using. The participants of the control group (CON) consisted of healthy age-matched individuals free of injury and illness at the time of testing. Ethical approval was obtained from the Human Research Ethics Committee (Stellenbosch University) (M16/08/032) and informed consent was obtained from all participants.

Procedures

Bipolar surface electromyography (EMG) was recorded using the wireless DTS 16 channel lossless sensors running MyoResearch 3.10 software (Noraxon, UK). Electrode placement followed the surface electromyography for the non-invasive assessment of muscles (SENIAM) guidelines. The muscles measured were bilateral lumbar erector spinae (LES), gluteus medius (Gmed) and gluteus maximus (Gmax), bicep femoris (BF), vastus lateralis (VL) and unilaterally (UN side) medial gastrocnemius (MG) and tibialis anterior (TA). The CON included bilateral placement for the MG and TA. Each placement area was shaved and cleaned using ethanol. Two electrodes were placed with an electrode centre distance of 2cm. Signal to noise ratios were measured to ensure the quality of the data before tests were performed.

Kinematic and kinetic data were collected with the use of the Vicon 3D motion analysis system (Vicon, UK) comprising of eight MX-T20 cameras. Furthermore, three Bertec floor-imbedded force plates were used (two FP4060-07 models and one model FP6090-15 (Bertec, USA)). Reflective markers with a 7mm diameter were placed according to a modified Helen Hayes marker set. Technical markers were placed on the flat surface of the shin and slightly distally to that. Technical thigh cluster markers were placed bilaterally consisting of a marker over the proximal lateral third of the thigh, a marker distally and posteriorly and one distally and anteriorly so as to form a non-equilateral triangle. Markers were placed on the prostheses at positions correlating to those of the unaffected side (UN). Knee markers were found based on the estimation of the joint centre and compared to the UN.

Test procedures

Participants were asked to perform 10 continuous sit-to-stands-to-sits with their hands crossed over their abdomen just below the Xiphoid process, at a self-selected yet controlled and even pace. The analysis of the SiSt and StSi were based on the four phases as described by Schenkman[12]. The end of the StSi was recognised when the vGRF was at its minimum[13].

Data reduction & analysis

Raw EMG data were pre-processed using a Butterworth band pass filter with a band pass frequency of 20-500Hz. The signals were then smoothed using Root-mean-square (RMS) method with 50ms smoothing window. The filtered EMG of each muscle was normalised to a percentage of functional maximum voluntary contraction (peak dynamic activation) for that muscle. The absolute maximum activation level of each muscle over all the trials, recorded while performing the functional movements, was taken as the 100% functional maximum voluntary contraction for a particular muscle. EMG onset/offset calculations used a threshold value of 5% functional maximum voluntary contraction where the level of activation had to be above or below threshold for at least 0.05 seconds before it was considered to be on or off, respectively. The average and average peak activation levels were calculated for the EMG data.

Force plate data were filtered using a Butterworth fourth order (zero lag) low-pass filter with a cut off frequency of 100Hz. Kinematic data (Standard Plug-in-Gait model outputs) were filtered using a fourth order (zero lag) Butterworth low pass filter with a cut off frequency of 6Hz. Force plate data were registered above a threshold of 20N. Vertical GRF and joint moment and power data were normalised to body weight for comparative purposes. The means, maximum, minimum, and percentage of the movement cycle where each of these points occurred were determined for joint angles, joint moments and joint powers.

Statistical analysis

Statistica (version 13.2; Dell, USA) was used to perform a statistical analysis of the data that were visually inspected for normality. All data were found to be normally distributed. Mixed model repeated measures analysis of variance (ANOVA) were used to determine differences between the UTTA and CON group as well as each limb side (affected side (AF) and unaffected side (UN), dominant side (D) and non-dominant side (ND) respectively) for each of the different variables. Fisher's least significant differences (LSD) as a post hoc test were used in the case of statistical significance. An alpha (α) level of 0.05 was selected. Data is presented as means and standard deviations ($\bar{x} + SD$).

4.3 Results

The UTTA group consisted of nine men and three women with an average age of 34±10 years, while the CON group consisted of nine men and four women with an average age of 34±11 years. The UTTA group had an average height of 1.78 ± 0.10m (range 1.55–1.93m) and a body mass of 73.42±16.28kg (range 51.1–106.4kg). The CON group had an average height of 1.78±0.08m (1.62–1.93m) and a body mass of 77.52±15.75kg (41.8–112.6kg). The average time since amputation was 10±7 years (range 1–26 years). Table 4.1 indicates the different prostheses used by the UTTA participants.

Table 4.1 Types of prostheses used by participants (n=12)

Number of participants	Type of Prosthesis
3	Variflex XC, Össur
1	Variflex with EVO cat 4, Össur
1	Reflex rotate with unity, Össur
1	Elevation foot, Össur
1	Proflex XC, Össur
2	Rush high pro, Ability dynamic
1	Proflex, Össur
1	Ottobock 1D35 Dynamic motion
1	Profiled prosthesis, Össur

4.3.1 Muscle activation

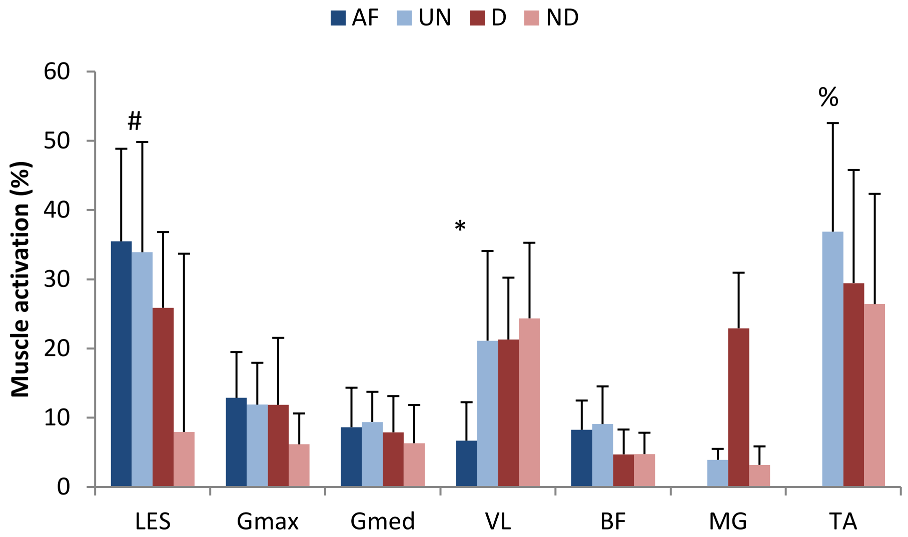
The peak normalised muscle activation during the SiSt for LES muscles was greater for AF and UN (15.98±5.93%, 15.97±6.57% respectively) than D and ND (12.72±5.93%, 4.2±12.34% respectively, $p<0.05$). VL muscle activation was significantly lower for AF (3.09±2.46%, $p<0.05$) than UN, D and ND. Peak Gmax activity was significantly higher on AF (12.87±6.62%) than the ND (6.16±4.47%, $p<0.05$). TA of UN had significantly greater activation (19.53±20.87%) compared to ND (14.34±10.57%, $p<0.05$) (figure 4.1a).

The average and peak normalised muscle activation during the last phase of standing up (30–60% of movement) identified significantly lower activation levels in LES and VL for both AF (5.72±3.28%, 5.02±5.21% respectively) and

UN (6.6±3.09%, 5.86±3.93%) compared to CON ($p<0.05$). The peak Gmax was significantly lower for UN (11.75±7.86%) compared to D (19.37±10.29%, $p<0.05$). The activation for the TA was also significantly lower (17.14±12.09%) for UN compared to D (29.59±16.44%, $p<0.05$) and ND (25.32±15.01%, $p<0.05$).

During the last phase of sitting down during the StSi (last 25% of movement) muscle activation of VL for AF (0.95±0.92%) was significantly lower than ND (11.96±6.12%, $p<0.05$). The BF for UN (9.38±9.57%) showed higher average activation levels compared to D (5.58±7.25%, $p<0.05$). The TA had a significantly higher average activation for UN (14.17±10.91%) compared to ND (8.36±6.61%, $p<0.05$) but there was no interaction effect with D. The same results were found in the peak activation levels over the last 25% of the movement (figure 4.1b).

a)



b)

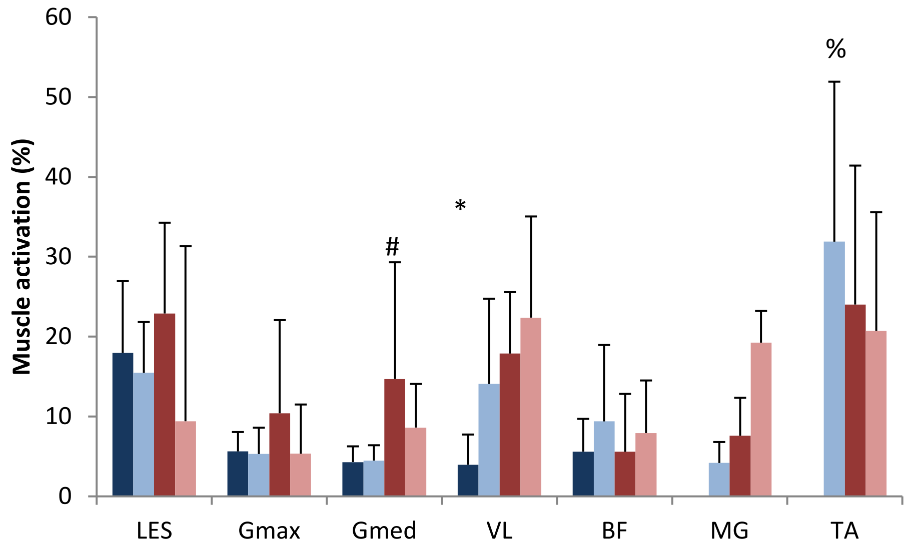


Figure 4.1 Peak averaged, normalised muscle activation (%) for the (a) SiSt and the (b) StSi ($\bar{x} + SD$)

SiSt = sit-to-stand, StSi = stand-to-sit, AF = affected, UN = unaffected, D = dominant, ND = non-dominant, LES = lumbar erector spinae, VL = vastus lateralis, Gmed = gluteus medius, Gmax = gluteus maximus, BF = bicep femoris, MG = medial gastrocnemius, TA = tibialis anterior, SD = standard deviation

a) * significantly lower VL activation on the AF than all other groups (p <0.0001), # significantly greater activation of the AF and UN LES than D and ND (p<0.01), % significantly greater TA activation on UN compared to ND (p<0.05),

b) *significantly lower activation of the VL on AF compared to all other groups (p<0.0001), % significantly greater TA activation on the UN than ND (p<0.05), # Gmed D significantly greater than AF and UN (p<0.05)

4.3.2 Kinetics and Kinematics

4.3.2.1 Vertical ground reaction force (vGRF)

The peak vGRF recorded during the SiSt was significantly greater ($p<0.05$) for UN compared to AF, ND and D. Peak vGRF for AF was reached between 38-48% of the movement cycle compared to 15-21% for UN as well as ND and D. The StSi phase results mirrored those of the SiSt (table 4.2). Peak vGRF of 65.81±5.95% $N.kg^{-1}$ for UN was significantly greater than all other sides ($p<0.05$) (table 4.2).

Table 4.2 Peak vGRF ($N.kg^{-1}$) and occurrence during movement cycle for the SiSt and StSi ($\bar{x} + SD$)

	Peak SiSt ($N.kg^{-1}$)	% Movement	Peak StSi ($N.kg^{-1}$)	% Movement
AF	52.29±5.45	43±5#	49.84±4.53	51±1#
UN	69.90±5.77*	18±3	65.81±5.95*	74±4
D	57.40±3.86	18±2	55.80±3.79	71±7
ND	56.34±3.26	18±3	54.84±3.32	73±3

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, StSi = stand-to-sit, SD = standard deviation; * UN significantly greater than UN, D and ND during SiSt and StSi ($p<0.05$), # AF reach peak GRF significantly later than UN, D and ND ($p<0.05$)

4.3.2.2 Hip joint

During the SiSt the maximum hip flexion angle measured for AF (94.41±10.97°) was significantly greater than UN (91.41±11.97°, $p<0.05$) and ND (85.58±8.58°, $p<0.05$). The above mentioned peak flexion was reached significantly later in the movement cycle for AF (20±3%). While standing upright the D and ND (6.64±7.27°, 4.89±7.26°) had greater hip extension than the AF (14.03±9.36°, $p<0.05$). The peak hip moment during the SiSt was greater for AF (0.87±0.20$N.m.kg^{-1}$) and UN (0.90±0.22$N.m.kg^{-1}$) compared to ND (0.69±0.23$N.m.kg^{-1}$, $p<0.05$) and D (0.70±0.21$N.m.kg^{-1}$, $p<0.05$). A significantly higher concentric hip power of AF (1.19±0.31$W.kg^{-1}$) was found compared to D and ND (0.86±0.29$W.kg^{-1}$, 0.87±0.33$W.kg^{-1}$, $p<0.05$) (figure 4.2).

The StSi elicited a peak flexion of 94.64±10.44° in AF which was significantly higher than all other sides, and was reached significantly later in the movement (80±4% vs 77±3%) for the other conditions ($p<0.05$). Significantly greater eccentric hip power on the AF (-0.97±0.32$W.kg^{-1}$) than UN (-0.65±0.42$W.kg^{-1}$, $p<0.05$) was found.

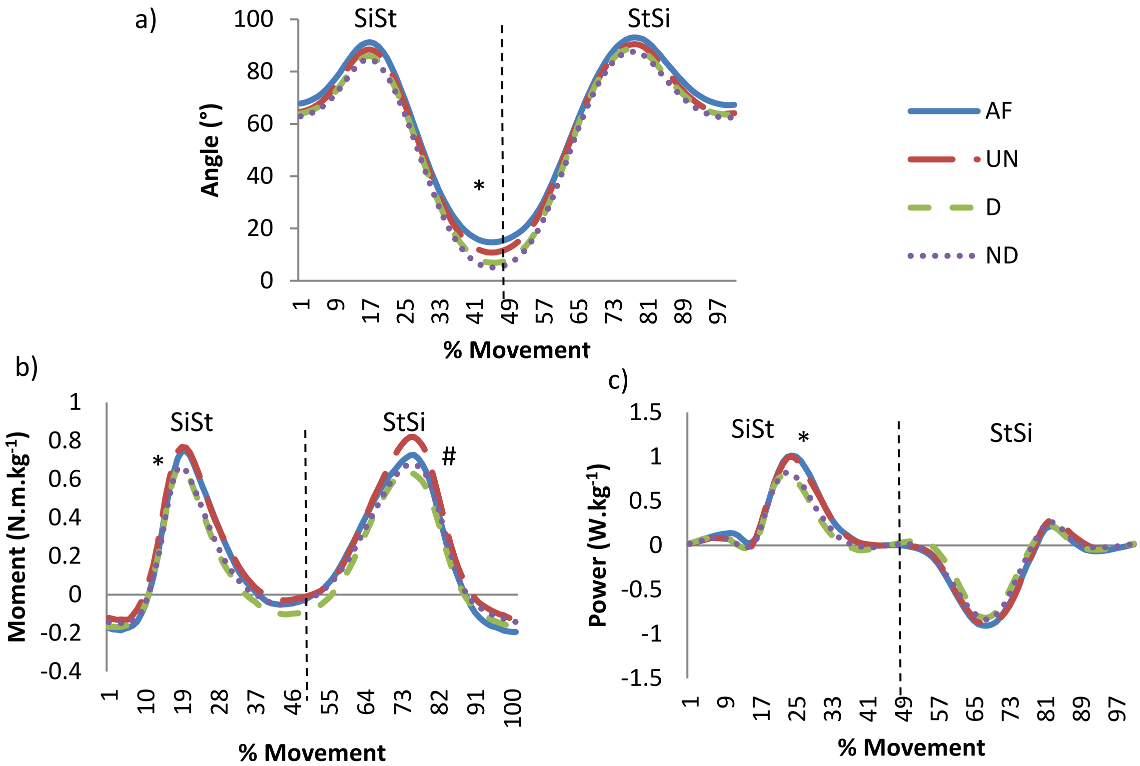


Figure 4.2 Average hip joint a) angles (°), b) moments (N.m.kg^{-1}), and c) power (W.kg^{-1}) during the SiStSi $(\bar{x} + SD)$

SiSt = Sit-to-sand, StSi = Stand-to-sit, Dotted line indicates the midpoint between SiSt and StSi, a)* AF and UN significantly less hip extension (p =0.005), b)* AF and UN greater than D and ND (p<0.03). # UN significantly greater than D (p<0.05), c)* AF and UN significantly greater than D and ND (p<0.04)

4.3.2.3 Knee

No significant differences were found pertaining to the knee flexion/extension range of motion during the SiSt (figure 4.3). However significantly less extension was performed by AF (11.68±5.65°) as well as UN (9.07±6.10°, p<0.05) whilst standing upright (midpoint of the movement) in comparison to D (4.39±4.44°, p<0.05) and ND (3.29±50°, p<0.05). Peak extension for AF was reached significantly later (46±3% of movement) in comparison to UN and CON, which reached maximum extension at 42±3% (UN) and 43±4% (CON) of the movement, respectively (p<0.05).

Significantly lower joint moments were found for AF during both the SiSt and StSi activities (0.30±0.14N.m.kg^{-1} and 0.21±0.00N.m.kg^{-1} respectively, p<0.05) than all other groups. The peak joint moment was reached significantly earlier in the cycle for AF compared to UN during the SiSt (p<0.05) but significantly later during the StSi (72±10%, p<0.05). The affected side produced significantly lower peak power at the knee joint (0.42±0.26 W.kg^{-1}) than UN (1.33±0.34W.kg^{-1}, p<0.05), D (1.32±0.33W.kg^{-1}, p<0.05) and ND (1.28±0.32W.kg^{-1}, p<0.05). The same was true for the StSi phase of the movement where AF produced significantly less eccentric knee joint power (-0.23±0.15 W.kg^{-1}, p<0.05) than UN (-0.86±0.26W.kg^{-1}, p<0.05), D (-1.01±0.29W.kg^{-1}, p<0.05) and ND (0.99±0.24W.kg^{-1}, p<0.05). The eccentric power that it was able to produce occurred later in the cycle compared to all other sides (70±5%).

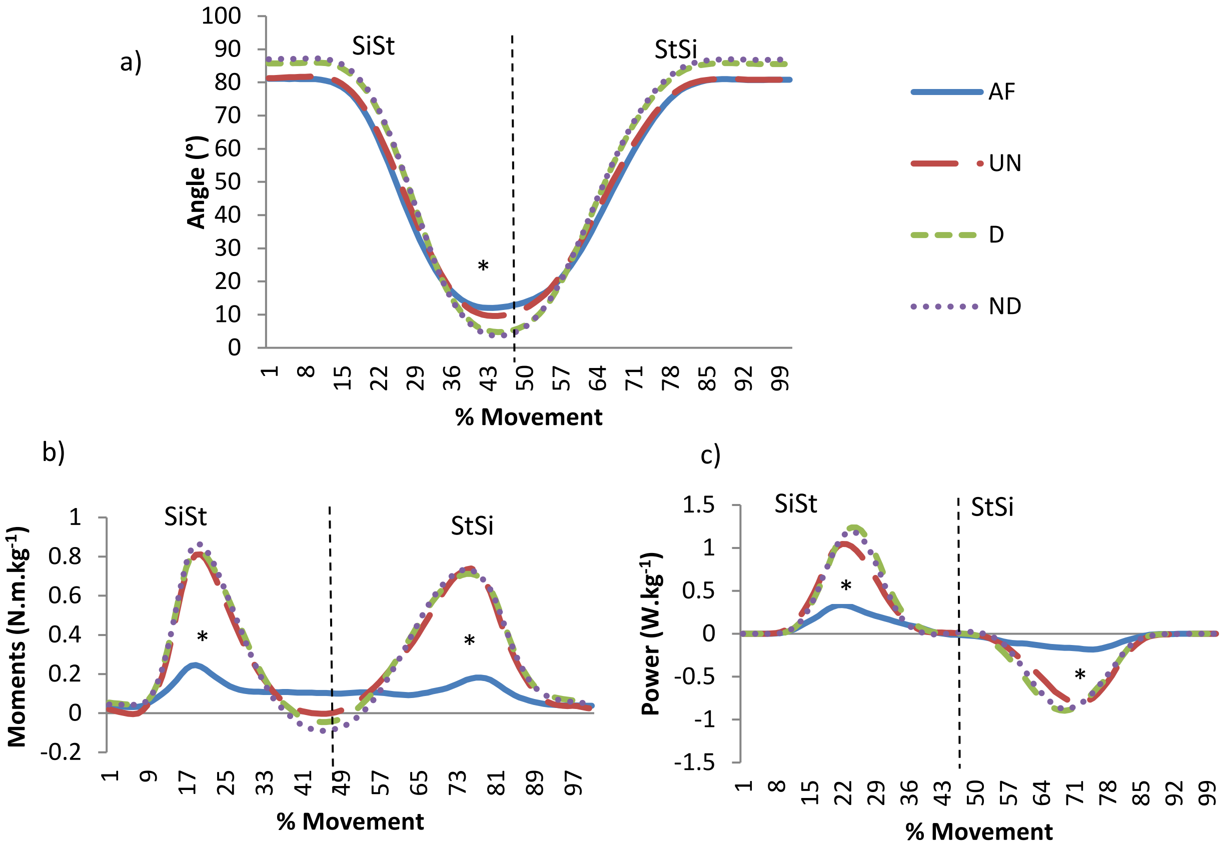


Figure 4.3 Average knee joint a) angles (°), b) moments (N.m.kg^{-1}) and c) powers (W.kg^{-1}) during the SiStSi ($\bar{x} + SD$)
SiStSi = Sit-to-stand-to-sit, SiSt = Sit-to-stand, StSi = Stand-to-sit, Dotted line depicts the point at which the participants were standing upright; a) * Affected and unaffected side significantly smaller than other groups (p<0.05), b & c) * Affected side significantly smaller than all other groups (p<0.05).

4.3.2.4 Ankle

Significantly less dorsiflexion (10.57±2.68°) was measured for AF ankle joint compared to UN (18.88±6.06°) as well as D (20.98±2.49°) and ND (22.15±3.11°) during the SiSt. Maximum dorsiflexion for AF was reached later in the cycle, 23±8 % compared to all other sides (20±3%, p<0.05). Significantly less dorsiflexion was obtained in UN in comparison to ND during the SiSt. The AF side ankle joint obtained significantly less dorsiflexion during the StSi (10.08±2.88°).

The peak ankle moment during the SiSt was significantly higher ($p < 0.05$) in AF (0.46±0.13N.m.kg^{-1}) compared to both UN (0.37±0.09N.m.kg^{-1}, $p<0.05$) and ND (0.37±0.08N.m.kg^{-1}, $p<0.05$). The peak ankle moment of AF was reached significantly earlier in the cycle (24±12%, $p<0.05$). The peak moment for AF during the StSi was reached significantly later compared to the rest of the groups (67±12%, $p<0.05$).

Mean power of the ankle was significantly lower in AF compared to UN ($p<0.05$). The UTTA group produced significantly less peak power at the ankle joint ($p<0.05$) during the SiSt with AF and UN producing 0.07±0.04W.kg^{-1} and 0.09±0.07W.kg^{-1} respectively with D and ND 0.14±0.07W.kg^{-1} and 0.16±0.05W.kg^{-1} respectively ($p<0.05$). Eccentric power is produced by AF (figure 4.4c) before it produced the expected concentric power during the SiSt. Conversely, during the StSi, AF ankle produced concentric power before it could produce eccentric power, which in turn then exhibited higher power. Eccentric power was produced predominantly for UN, D and ND during the StSi (figure 4.4c).

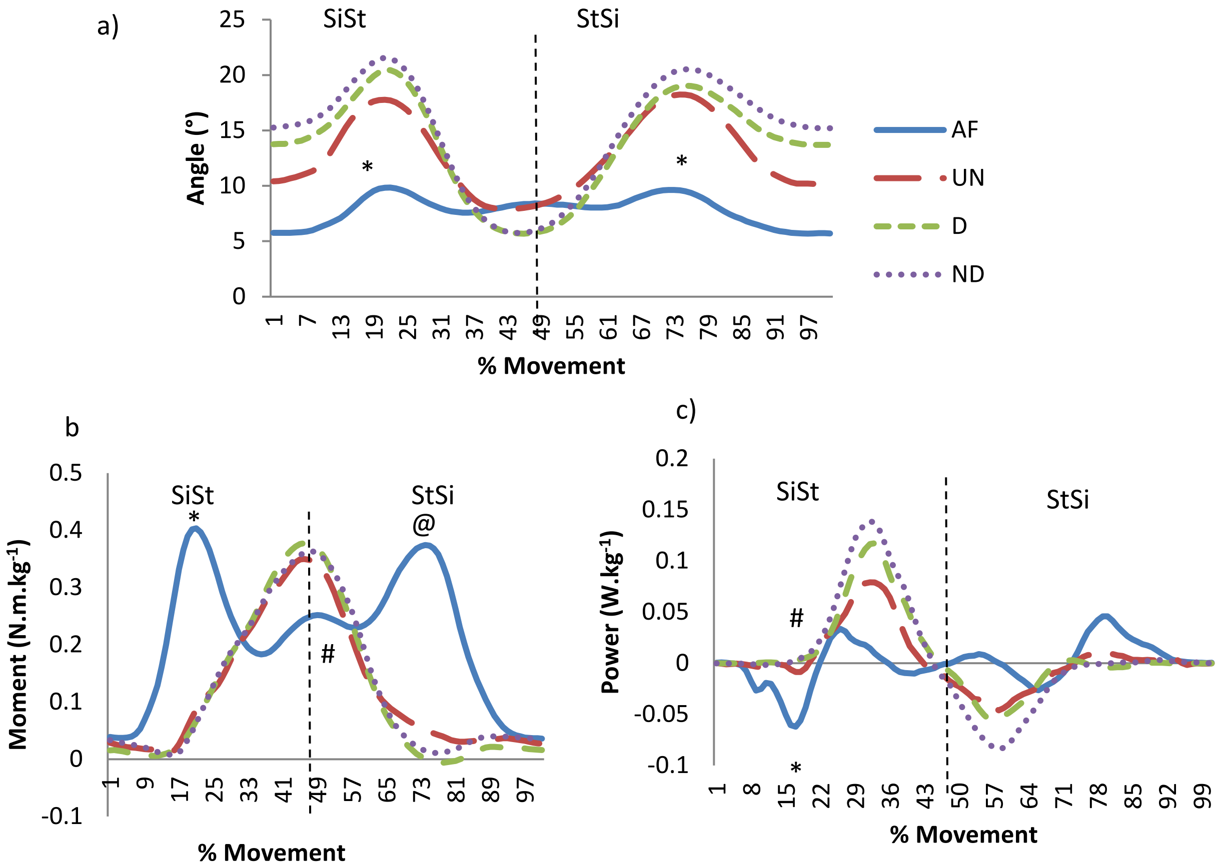


Figure 4.4 Average ankle joint plantar/dorsiflexion a) angles (°), b) moments (N.m.kg^{-1}) and c) power (W.kg^{-1}) during SiStSi ($\bar{x} + SD$)

SiStSi = sit-to-stand-to-sit, SiSt = Sit-to-stand, StSi = Stand-to-sit, , Dotted line depicts the point at which the participants were standing upright; a) * Significantly less plantar dorsiflexion on AF side compared to all other groups, b) * AF significantly greater than the UN ($p=0.005$) and ND ($p=0.027$), # Significantly lower than the ND ($p=0.044$) and D ($p = 0.038$), @ AF significantly greater than UN ($p=0.013$) and peak reached significantly later than all other groups ($p<0.0001$), c) # AF less than ND ($p=0.002$) and D ($p=0.0123$), UN less than ND ($p=0.0154$), * AF side produced eccentric power significantly earlier than all other groups (p<0001).

4.4 Discussion

The aim of this study was to identify the muscle activation levels during the SiSt and StSi between AF and UN of UTTA and in comparison to CON. The hip, knee and ankle joint and their surrounding musculature were identified for analysis purposes.

Muscle activation levels during the SiStSi were significantly lower for the AF VL than UN, D and ND while the TA of UN was higher than the CON. The LES muscle activation for AF and UN was higher than the CON during the SiSt. During the StSi Gmed activation was greater for D than AF and UN. The lower VL activation of AF may relate to the lack in eccentric muscle control possible in this population[14]. It may also relate to the asymmetrical loading to the unaffected side and therefore less VL activation is required on the AF at that point. The TA of UN however, showed greater activation levels during the SiSt and StSi and this is likely due to the asymmetrical shift in body mass through the heel, to help maintain balance and control. According to Papa[15], the TA muscle is important for the initiation of the SiSt movement, contributing to postural stability[16]. The TA of UN may compensate for the lack of TA of AF as the prosthetic limb would not be able to control the movement in the same way. A study by Cheng[17] reported low TA activation of AF in stroke patients during the SiSt while significantly higher activations were recorded on the UN[17]. This study supports the findings of our study in terms of TA activation.

Significantly higher vGRF was experienced by UN than AF, D and ND during SiSt and StSi. These results indicate asymmetrical movement and hence possible increased joint loading in UN. The results of this study are congruent with that of Agrawal[4,13] where they also documented asymmetry during the SiSt and StSi movement with the UTTA shifting more body mass onto UN[4]. It was also noted that the weight acceptance on AF was later than that of the control group during the SiSt and StSi and could be explained by the trust in the prosthesis by the UTTA as well as the prostheses characteristics.

Results pertaining to the joint kinetics and kinematics indicated that hip flexion of AF was greater during the SiSt and StSi as well as while in an upright position than UN, D and ND. The peak hip moments were greater in UTTA than CON. The mechanism for the hip moments could be related to the amount of trunk lean or hip flexion used to compensate for the lack of stability[18]. Significantly higher average hip power was found in UTTA, compared with CON during the SiSt and StSi. Greater peak hip power was also found during the SiSt. As one of the compensatory mechanisms UTTA may employ the hip strategy in order to better control the movement while maintaining balance[15]. A hip strategy has also been noted in a population consisting of mild Parkinson's disease[18]. In the study mentioned increased hip flexion was used as a compensatory mechanism to counter for overall muscle weakness and poor postural stability[18]. Overall, greater hip flexion is maintained, which in turn creates greater hip moments. In this case, AF hip of the UTTA appears to be working harder, producing more power through the hip to possibly compensate for muscle loss of lower limb.

The peak knee moments and powers of AF during the SiSt and StSi were significantly smaller than UN and CON which indicates a lack of knee control. This may lead to the movement control of AF to take place elsewhere. Due to the prosthesis, the range of motion at the ankle is limited. Agrawal[13] speculated that because the prosthesis is selected for walking symmetry it may not be able to fulfil the function that is required to complete other functional activities. While Agrawal's[13] findings were not based on the SiStSi this study is still in agreement with our study, with respect to the limited range of motion found for the ankle aspect of the prosthesis. Despite limited ROM in AF ankle, it is able to withstand greater moments on AF as well as a double peak being measured during the SiStSi. When considering this with the limited ROM possible on AF it was noted that the heel of the prosthesis lifts up, during the start of the SiSt and at the end of the StSi, creating the impression that the lever arm has changed and therefore has a greater moment The AF ankle was also unable to produce as much power as the UN and CON. When trying to understand the power production by the ankle joints, it can be observed that the prosthetic ankle does not act in the same way that the biological ankle would work as it has reduced plantar and dorsiflexion range of motion than the biological ankle[13]. The results showed that AF produced power in a different way to that of UN, D and ND during both the SiSt and StSi as initially an eccentric power was produced. The UN ankle was however compensating for the AF side in terms of ankle joint kinetics.

In summary it appears that AF and UN hip and UN knee and ankle in terms of joint moments and powers are compensating during the SiSt and StSi in order to control the movements. The increased muscle activation of the muscles surrounding the AF hip during the SiSt and the TA muscle for UN side during the StSi support the findings of the kinetic and kinematic results.

4.5 Summary of findings

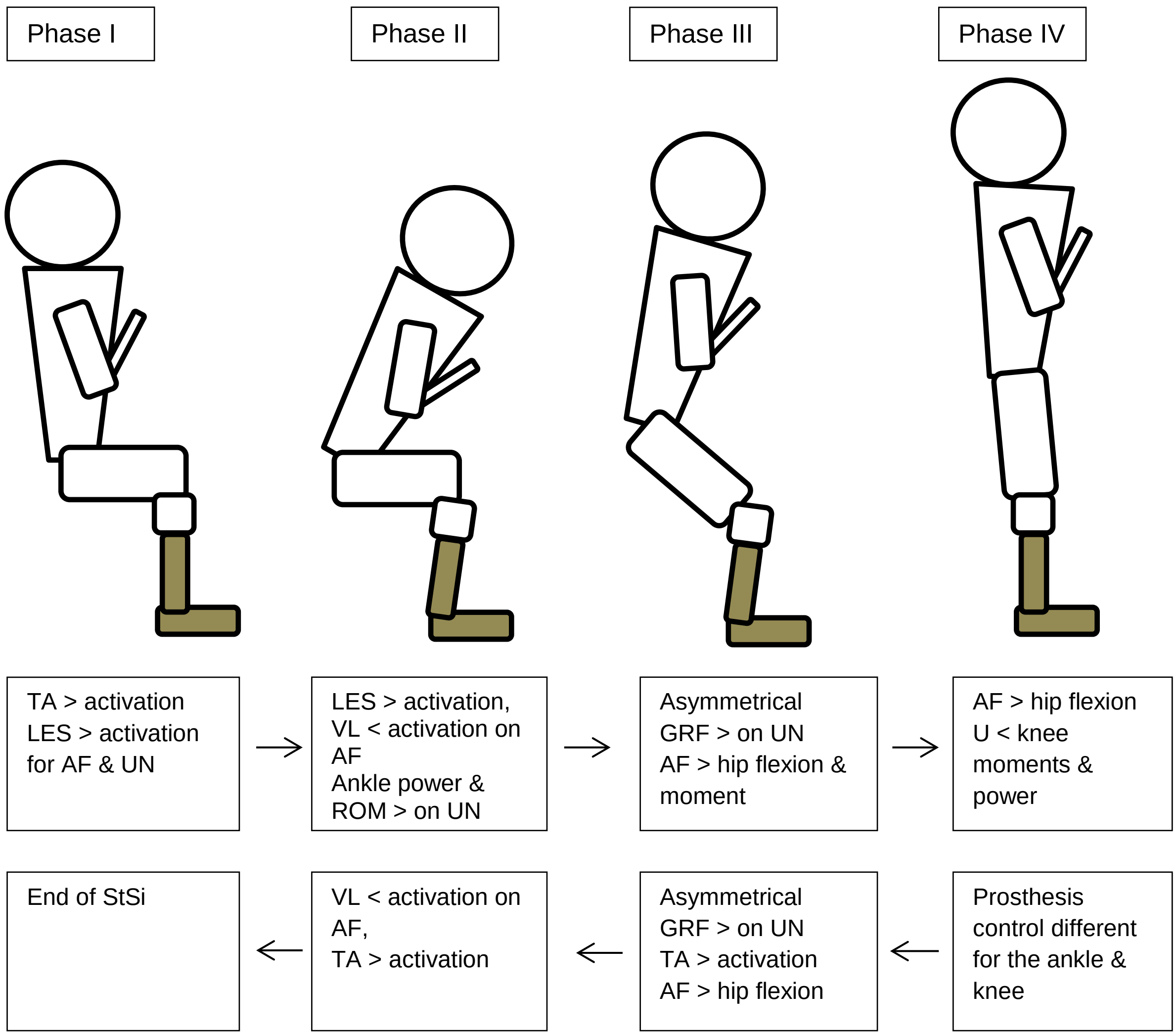


Figure 4.5 Key findings of AF compared to UN for the SiSt and StSi
SiSt = Sit-to-sand, StSi = Stand-to-sit, AF = affected, UN = unaffected, LES = lumbar erector spinae, VL = vastus lateralis, TA = tibialis anterior, > =greater, < = less

4.6 Limitations

Prosthetic setup of the participant's prosthesis was used as determined by their prosthetist, and alignment was not accessed. Marker placements on the prosthesis were based on biological estimations and due to the variation in prostheses were not identical. While the speed of the activity was not specifically controlled the participants were cued to maintain a controlled movement. Lastly, the muscle activation was measured for specific muscles and due to practical limitations we cannot account other muscle activations that may have contributed to the movement control.

4.7 Future research

In future, it may be valuable to explore possible exercise protocols that may improve the movement biomechanics of the SiSt and StSi. Research towards the development of variables important to evaluate during the use of the SiStSi movement as a screening tool for prosthesis selection and setup is recommended.

4.8 Conclusion

The results of the study showed that there are specific compensation patterns present during the SiStSi movement. These include the LES of the lower back region, as the muscles and joints are not effectively able to control the movements required during the activity to the same extent as seen in the control group. These findings suggest that when considering injury prevention, two things need to be considered. Firstly, the functionality of the prosthesis solution and secondly the extent to which the movement response can be trained. This challenges the dynamic systems of the integration of technology and the biological demands to prevent long term injuries in UTTA.

4.9 References

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Chapter 5

Discussion

Functional activities form an important part of daily life. Previous literature has focused on gait in unilateral transtibial amputees (UTTA) as well as the influence of different prostheses on gait (Silverman *et al.*, 2008; Hak *et al.*, 2014). Limited research has investigated the biomechanical influence on UTTA during functional activities and there is a need to better understand the movement patterns used to improve the rehabilitation process of UTTA (Agrawal *et al.*, 2011). The purpose of this study was to gain a better understanding of UTTA muscle activation levels and joint kinematics and kinetics during two functional activities that are performed on a daily basis. The systematic review (chapter 2) allowed for an in depth understanding of the information pertaining to the topic of biomechanical aspects of gait and balance in unilateral transtibial amputees. Two functional activities, single leg balance (SLB) and sit-to-stand-to-sit (SiStSi), were selected for three dimensional biomechanical analyses to address the research questions below.

5.1. Research questions

Question 1. Is there a difference in muscle activation levels between able-bodies and unilateral transtibial amputees (UTTA) during functional activities?

This question was answered through chapter 3 and chapter 4 of this thesis. In chapter 3, SLB was investigated while chapter 4 investigated SiStSi. Differences were found in both activities between the affected side (AF) and the unaffected side (UN) of UTTA as well as in comparison with the dominant (D) and non-dominant side (ND) of the control group (CON).

Objective 1

1.1 To determine the skeletal muscle activation levels between the AF and UN of UTTA during functional movements using surface EMG placed on the vastus lateralis, bicep femoris, gluteus medius, gluteus maximus and the lower region of the lumbar erector spinae.

In Chapter 3, the muscle activation levels were higher for AF than UN for the LES, Gmed, Gmax, BF and VL muscles.

In Chapter 4 (SiStSi) the VL for AF had lower muscle activation at the initiation of the SiSt as well as during the StSi than UN.

1.2 To determine the difference in muscle activation levels between able-bodied controls and UTTA during specific functional movements using surface EMG placed on the tibialis anterior, medial gastrocnemiums, vastus lateralis, bicep femoris, gluteus medius, gluteus maximus and the lower region of the lumbar erector spinae.

The SLB article in Chapter 3 found that the muscle activation levels were higher in the LES and BF muscles of AF compared to ND and D. The Gmed and Gmax activation for AF was greater than ND. No differences were noted for the TA or MG muscles.

In the SiStSi article (Chapter 4) several differences between UTTA and the control group were found. The LES muscles for AF and UN had higher levels of activation than D and ND during SiSt. The VL muscle activation was lower for AF than, ND and D during SiSt as well as StSi. No significant difference was found for the VL activation of the UN compared to the ND and D. Lastly, the TA muscle activation was greater for UN than ND and D during the SiSt and StSi.

Question 2. How do the joint kinetics and kinematics for the hip, knee and ankle during functional activities compare between the affected side (AF) and unaffected side (UN) of unilateral transtibial amputees and dominant side (D) and non-dominant side (ND) of controls?

The second research question was answered by Chapter 3 and 4 of this thesis. The joint kinetics investigated included ground reaction force, joint moments and joint powers while the kinematics included joint angles. During both the SLB (Chapter 3) and the SiStSi (Chapter 4), differences were found for all three variables of joint kinetics. The joint kinematics results indicated differences in the hip and ankle during the SiStSi. Less knee extension was achieved for AF and UN when an upright position was reached.

Objective 2

2.1 To compare the vertical ground reaction force (vGRF) and joint moments, powers and angles acting at the hip, knee and ankle during functional activities using Vicon 3D analysis between the affected side (AF) and unaffected side (UN) of the UTTA and the dominant side (D) and non-dominant side (ND) of the control group.

Chapter 3 & 4 assisted in achieving this objective. During the SLB the hip joint moment of AF showed a flexion moment while UN indicated extension moments. There were no worthwhile differences reported for the knee or ankle joints. The AF hip produced more concentric power than UN, D and ND. The AF knee produced more power than UN and ND. The hip and knee on AF had greater flexion than UN, ND and D.

The affected side knee experienced lower moments during the SiSt and StSi than UN, ND and D. The timing of the peak moment was earlier for AF compared to UN during the SiSt and later during the StSi. Ankle joint moments were higher for AF compared to UN and ND. The peak moment for AF was reached earlier than the other sides during the SiSt and later in the StSi than the other groups. Hip power was greater for both sides of UTTA than control group during the SiSt and StSi. The knee and ankle powers were greatest for UN, while AF produced the least power in comparison with CON during the SiSt and StSi. Greater hip flexion was noted for AF and UN compared to ND and D during the SiSt and StSi. The SiSt and StSi showed greater vGRF on UN than AF. The SiSt peak hip moment was larger for AF and UN than D and ND.

The results answering the two research questions highlighted four main findings. 1) Hip strategy for AF during SLB, and for both AF and UN during SiStSi, 2) knee increased loading on the UN during SiSt and StSi and 3) ankle increased loading of the UN during SiStSi and 4) asymmetrical loading (vGRF) during the SiStSi.

There are two primary mechanisms involved in the maintenance of balance during walking and other functional activities, namely the galvanic vestibular stimulation system and the neural control mechanism (Reimann *et al.*, 2017). When specifically considering the neural control mechanism, there are several strategies that the body can make use of within this mechanism to maintain balance (Reimann *et al.*, 2017). The possible mechanisms and their constituent strategies are discussed below for the muscle activation levels, joint kinetics and kinematics findings from a holisitic dynamic systems perspective.

5.2. Mechanisms for strategies used by UTTA

Single leg balance and the SiStSi are both activities which require muscle and joint co-ordination to maintain centre of mass (COM) over the given base of support (BoS) to efficiently execute tasks (Ku *et al.*, 2014). While these activities may require different strategies and movement patterns they both require postural control as well as muscle and joint co-ordination. Skeletal muscle functions to create stability around associated joints to maintain joint stability for postural control as well as to initiated controlled movement of the body (Blackburn *et al.*, 2000; Prilutsky, 2000). In order for this to take place, joint proprioceptors need to react to stimuli, resulting in joint movement through controlled co-ordination between the joint and its surrounding musculature (Prilutsky, 2000). Asymmetry was specifically found in SiStSi (chapter 3) but other biomechanical differences such as muscle activation in bilateral muscles and joints moments and powers also highlight asymmetry during SLB (chapter 4).

Researchers have shown that lower limb muscle strength imbalances, loading asymmetries towards the unaffected side and repetitive daily activities may lead to early onset of joint degeneration or osteoarthritis in joints of the unaffected side as well as the remaining joints of the affected side (Royer & Koenig, 2005; Gailey, 2008; Lloyd *et al.*, 2010; Agrawal *et al.*, 2011). Agrawal *et al.* (2011) mentioned concern for muscular injury or secondary conditions should poor technique be used when performing the SiSt activity multiple times per day. A conceptual framework suggested by Rimmer *et al.* (2011) when working with individuals with disabilities, urge identification of the primary, secondary and associated conditions that result from the impairment. When interpreting this framework for UTTA, the loss of the ankle and supporting muscles is the primary condition. The associated conditions are the possible muscle recruitment and adaptive movement strategies. The systematic review indicated that limited research is available in this area. The secondary conditions associated with lower limb loss are possible long term injuries due to compensatory mechanisms. However, currently there is little to no longitudinal literature available for this population.

The dynamic systems theory can also be used to understand the results of the current study with a holistic approach. This theory takes into consideration the task, the environment and the organism and it is the relationship between all three that results in a movement pattern (Holt *et al.*, 2010). Based on the dynamic systems theory, the section hereafter will discuss the results of the study in an integrating fashion.

5.2.1 Hip strategy

The greater muscle activation for the hip joint muscles (LES, Gmed and Gmax) for AF during SLB suggests the use of the hip strategy to maintain balance. Higher activity was also noted for AF in the BF and VL muscles which could contribute to the hip strategy. The muscles are however most active when the hip is at peak flexion and the hip moment for AF is at its highest. This facilitates control of the movement and maintains balance or position of COM while employing the hip strategy (Papa & Cappozzo, 2000).

More hip flexion was noted throughout the SiStSi for AF which supports the greater hip moments. This may result in the reliance of UN foot to provide feedback up the complex kinetic chain, to assist in proprioception on AF. Therefore, the role of proprioception has to be fulfilled by the knee joint, the first joint complex on AF. These findings support that of Chrisholm (2015), with specific reference to the asymmetry seen during the activity as well as the range of motion (ROM) for the joints measured. Therefore, with greater trunk lean or hip flexion, a greater hip joint moment is experienced.

As mentioned previously, the hip moments for AF and UN were greater than the control during the SiSt and StSi. The LES muscles on the AF and UN had greater muscle activation than the CON while the Gmed and Gmax had similar activation levels across the groups during the SiSt. The fact that the hips are the first bilateral joints that are able to do work could give insight to these findings. The joint kinematic results highlighted that AF hip and knee remained in a more flexed position during the SLB. This suggests that for UTTA, specifically AF, may make use of the hip strategy to maintain balance (Ku *et al.*, 2014).

5.2.2 Knee increased loading

While no specific differences were noted suggesting knee joint increased loading during the SLB, the SiSt and StSi highlighted lower activation levels for the VL of AF compared to UN. During the StSi AF and UN had lower activation of the Gmed than D. The knee moments and powers were higher for UN during the SiSt and StSi. This is a possible indication that the UN knee is compensating. This can be supported by Papa & Cappozzo (2000), where they investigated the asymmetry during the SiSt movement in elderly individuals who also suffered from knee joint degeneration. While this study did not include UTTA, the principle of compensation and asymmetry due to joint degeneration may be comparable to the UTTA population.

5.2.3 Ankle increased loading

The SiSt and StSi indicated higher activation levels of the TA for UN. This formed part of the strategy used to control the COM over the base of support suggesting asymmetrical loading of the muscles surrounding UN ankle joint. This asymmetrical loading through the UN ankle may increase the long term risk for injury and joint degeneration. The ankle joint is the first joint in the kinetic chain, and receives proprioceptive feedback through the foot’s contact with the ground, however AF is unable to do this. A study by Curtze *et al*. (2012) found that when balancing on two feet with perturbations applied in a forward/ backward direction, an ankle strategy was used. However, the UTTA group compensated for the AF limb by increasing the moment around the UN ankle (Curtze *et al.*, 2012).

5.2.4 Asymmetry

As mentioned, loading asymmetry (vGRF) was specifically found during the SiStSi of UTTA with a shift in body mass more to UN. The peak vGRF seen for UN is perhaps due to a compensation of weight shifting due to the lack of proprioception possible by the prosthesis. Furthermore, this compensation may be linked to the lack of "trust" in the prosthesis and or the strength of AF to bear the same body mass. The UTTA response is to place more weight on UN where there are greater feedback opportunities and therefore a better ability to control weight distribution/ CoM in order to reduce the risk of falling (Agrawal *et al.*, 2011). As part of a systematic review, Gailey (2008) reported a tendency to "rely" on UN during the SiStSi. The repetitiveness of the task throughout a day with this asymmetry of the movements could place significantly greater strain on UN musculoskeletal structures which may influence the risk of degenerative joint conditions (Gailey, 2008).

5.3. Dynamic systems theory

The three domains of the dynamic systems theory include the organism, task and the environment. In the case of UTTA functional abilities are influenced due to the loss of a limb (organism) and impact the movement capabilities (the task) that are possible (Holt *et al.*, 2010). A prosthesis could be viewed as an environmental constraint as it is an external aid that facilitates locomotion for UTTA. The mechanical characteristics of the different prostheses also need to be considered together with its specific movement mechanism. While some studies have shown that the prosthesis design more closely mimics the way a biological foot responds (Supan *et al.*, 2010), others show that they are better suited for straight line walking (Viton *et al.,* 2000). Technology has allowed for the design of some more microprocessor driven or bionic lower limb prostheses of which the design aims to respond to the movement requirements (Thomas *et al.*, 2000). This can enhance the movement strategy and decrease the compensation, but more holistic research in this regard is needed. Lastly, the task constraints also have an impact on whether or not the UTTA will be able to perform it. The SiSt and StSi activity requires more force or work from the UTTA to perform the task and in this case the UTTA cannot perform the activity without compensatory mechanisms and asymmetry in loading patterns.

Possible improvement in muscle and joint co-ordination during SLB may assist with gait re-training techniques and improved awareness of joint orientation relative to the CoM (Jones *et al.*, 1997). Therefore, improving symmetry of movement by focusing on technique and strengthening of weaker muscles may be beneficial for UTTA. Correct training of motor patterns from early on during the rehabilitation phase may also benefit the overall ability of the UTTA to perform functional activities.

5.4. Van Mechelen model

The study design was hinged around the Van Mechelen model which proposes a four stage approach to monitor and prevent injuries. As mentioned in chapter 1, this study dealt with stage two which aimed to determine the possible mechanisms of the problems (van Mechelen, 1997). We determined that there are different possible mechanisms used by the UTTA to maintain balance and control during SLB and SiStSi. As mentioned during the SLB the UTTA adopted a hip strategy to main balance when standing on AF. While during the SiSt and StSi they tended to shift their weight onto UN and with increased hip flexion and hip joint moments utilised the hip strategy to maintain balance. It was noted that the use of a prosthesis influenced the control of the movements. Overall, we identified asymmetry in muscle activation patterns and biomechanical aspects of AF hip and UN knee and ankle which may increase the risk of early onset of joint degeneration. These findings are supported by Royer & Koenig (2005), Gailey (2008), Lloyd *et al*. (2010), and Agrawal *et al.* (2011) with respect to the asymmetry found and the link to an increase risk of joint degeneration. Recommendations are that the last two stages of the Van Mechelen model should be addressed in future studies.

5.5. Future research / clinical understanding

There is limited research pertaining to the biomechanical demands of different daily functional activities within this population group. A greater understanding in the muscle sequencing of UTTA during functional activities and work related physical demands is necessary to support rehabilitation techniques. There is also scope to better understand how the prostheses designs influence daily activities and if the alignment can be adjusted to better suit specific activities. Longitudinal studies that track the biomechanical adaptations over time, with possible related compensatory injuries will also add to the body of knowledge. The above may help to guide more specific guidelines and protocols being developed for the testing and exercise prescription for the rehabilitation of UTTA. It could also aid integration back to work and involvement in recreational sports and possible help in the long run to decrease the amount of secondary conditions associated with UTTA.

5.6. Study limitations

This study chose not to control for the type of prosthesis used as it was the intention to observe daily functional activities on a prosthesis that the participant is accustomed to. While this can be seen as a limitation we also saw it as a way to observe real life scenarios. The prostheses used by the participants were all similar in that they had similar range of motion and were neither fixed at the ankle nor were they powered prostheses. Marker placements for the prostheses were based on the anatomical landmarks as far as possible. The most challenging to find was the “knee joint center”. This was carefully estimated with knee flexion and the height was compared to the unaffected leg. Extra markers were used to help more accurately calculate the joint centres in the Vicon system. Some movement is possible at the stump socket interface and so rotational forces may have been affected. We therefore did not specifically look at the rotational forces at the knee at this point. Data were collected once-off from each individual and therefore limited time was given for familiarisation of the tests. The tests however were all activities that the participants perform on a daily basis and to that end they were comfortable with the tests before starting. Surface EMG data were filtered and reduced based on resting levels as well as isometric contractions. We were not able to perform true maximal voluntary contractions and used functional maximal voluntary contractions instead. We understand this may limit the interpretation of the EMG data. Muscle activation data were also limited to the number of muscles that were included. We unfortunately cannot account for other muscles activating during the activities and causing cross-talk. The SiStSi task used a backless wooden chair that was at a fixed height and therefore the starting hip flexion angle may have varied slightly on initiation. Initiation of the movement therefore used both a change in trunk and hip flexion as well as a change in vGRF. Speed of the SiStSi task was not controlled as the objective was to allow the participant to perform the movement closely to the speed they would use at home. Participants were cued to control the movement and not rush it as it would not be timed. For the SLB task participants were cued to keep their hands on their abdomen so as to avoid markers being covered and to isolate the trunk and lower leg. This may have influenced the balance strategy used on a daily basis. The alignment of the prostheses were not recorded or altered for the testing. The alignment setup was that which the participants were familiar with. The coefficient of variance (Addendum 6) was calculated for the data and large values for calculated for ND of the LES muscle during SiSt and StSi. Large CV's were also calculate for the Gmax and BF for UTTA and controls during balance. SLB moments, powers and angles also revealed large CV's for some UTTA and CON variables. Previous research has also found large variation with in the variables mentioned. This could be influenced by the prosthesis, muscle strength, the method used to calculate muscle activity and the fact that we only investigated kinetics and kinematics in one plane.

5.7. Conclusion

It can be concluded that UTTA have considerable biomechanical differences during the functional activities mentioned earlier. These findings suggest that over time there is risk of overloading both muscles and joints which could have long term, detrimental health concerns such as osteoarthritis, muscular injury or lower back pain. There would be great value in improving strength, muscle and joint co-ordination and technique of activities in order to decrease the asymmetrical joint loading and muscle activations. This may improve efficacy of movements and lower the risk of degenerative conditions. Rimmer & Rowland (2008) discuss the importance of empowering the person and promoting an inclusive environment. It is possible that if UTTA can improve their movement capabilities through rehabilitation and be afforded the opportunity to use a prosthesis that complements their needs, they will have more motivation and confidence to take part in more daily activities and possibly even recreational activities (Hutzler, 2008). Overall, the health and quality of life of UTTA is important and all factors need to be considered to know where adaptations can be made to enhance this.

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Addendum 1

Ethical clearance

UNIVERSITEIT•STELLENBOSCH•UNIVERSITY
jou kennisvennoot • your knowledge partner

Ethics Letter

30-June-2017

Ethics Reference #: M16/08/032

Title: The effect of a novel energy storage and return foot prosthesis (pro-flex foot) on biomechanics, function and lifestyle patterns, compared to the gold standard energy storage and return foot prosthesis (Vari-flex foot) and a conventional solid ankle cushioned foot prosthesis

Dear Prof Wayne Derman,

Your Letter of Amendment 1 dated 08 March 2017 and the response to modifications dated 27 June 2017 refer.
The Health Research Ethics Committee (HREC) reviewed and approved the amended documentation through an expedited review process.

The following amendments were approved:

1. Updated Protocol, Dated 27 June 2017
2. Participant Information Leaflet and Consent Form, Dated 27 June 2017.

Where to submit any documentation

Kindly submit **ONE HARD COPY** to Elvira Rohland, RDSD, Room 5007, Teaching Building, and **ONE ELECTRONIC COPY** to ethics@sun.ac.za .

Please remember to use your **protocol number (M16/08/032)** on any documents or correspondence with the HREC concerning your research protocol.

Federal Wide Assurance Number: 00001372
Institutional Review Board (IRB) Number: IRB0005240 for HREC1
Institutional Review Board (IRB) Number: IRB0005239 for HREC2

The Health Research Ethics Committee complies with the SA National Health Act No. 61 of 2003 as it pertains to health research and the United States Code of Federal Regulations Title 45 Part 46. This committee abides by the ethical norms and principles for research, established by the Declaration of Helsinki and the South African Medical Research Council Guidelines as well as the Guidelines for Ethical Research: Principles, Structures and Processes 2015 (Departement of Health).

Yours sincerely,

Francis Masiye,
HREC Coordinator,
Health Research Ethics Committee 2.

STELLENBOSCH UNIVERSITY
Health Research Ethics Committee
30 JUN 2017
STELLENBOSCH UNIVERSITEIT
Gesondheidsnavorsing Etiekkomittee

Fakulteit Geneeskunde en Gesondheidswetenskappe • Faculty of Medicine and Health Sciences

Afdeling Navorsingsontwikkeling en -Steun • Research Development and Support Division

Posbus/PO Box 241 • Cape Town 8000 • Suid-Afrika/South Africa
Tel: +27 (0) 21 938 9677

Addendum 2

Informed consent
(Experimental group)

PARTICIPANT INFORMATION LEAFLET AND CONSENT FORM

TITLE OF THE RESEARCH PROJECT:

The effect of a novel energy storage and return foot prosthesis (Pro-flex foot) on biomechanics, function and lifestyle patterns, compared to the gold standard energy storage and return foot prosthesis (Vari-flex foot) and a conventional solid ankle cushioned foot prosthesis

REFERENCE NUMBER: M16/08/032

PRINCIPAL INVESTIGATOR: Professor Wayne Derman

ADDRESS:
ISEM, Clinical building, Tygerberg campus, Stellenbosch University

CONTACT NUMBER: 0725984275

You are being invited to take part in a research study. Please take some time to read the information presented here, which will explain the details of this project. Please ask the researcher any questions about any part of this project that you do not fully understand. It is very important that you are fully satisfied that you clearly understand what this research entails and how you could be involved. Also, your participation is **entirely voluntary** and you are free to decline to participate. If you say no, this will not affect you negatively in any way whatsoever. You are also free to withdraw from the study at any point, even if you do agree to take part.

This study has been approved by the **Health Research Ethics Committee at Stellenbosch University** and will be conducted according to the ethical guidelines and principles of the international Declaration of Helsinki, South African Guidelines for Good Clinical Practice and the Medical Research Council (MRC) Ethical Guidelines for Research.

What is this research study all about?

- The purpose of this study is to gain a greater understanding of the muscle activation patterns of single below knee amputees in comparison to able – bodied individuals during movements of daily living. This could lead to more information regarding muscle overloading. In other terms we would like to see how the muscles “switch on” when performing activities such as walking up and down stairs, sitting or standing up and general walking and how this may differ between the two groups (amputees and able – bodied group).

- This study will take place in the Human Motion Analysis Unit at Stellenbosch University, Tygerberg campus. We would like to recruit 20 participants as part of our experimental group
- As a participant you would be asked to attend a once off, 2 hour testing session
- All volunteers will be screened according to the inclusion and exclusion criteria. After the participant has been screened and included the following procedures will occur:
- The testing session will begin with anthropometrical measurements. This will include the measuring of height, body mass and a posture analysis. Photographs will be taken for the posture analysis for analysis on the computer. In terms of privacy and confidentiality, participant's rights will be respected. All anthropometrical and marker placement will therefore take place in a private room.
- Participants will then be prepared for the biomechanical testing. This entails markers being placed on several anatomical land marks and muscles of the body. The participant will be well informed as to when and which muscle the researcher will be placing the marker on. Marker placement will take place in a private room.
- None of the tests that will be completed will be invasive or case harm to you. Marker placement will require the researcher being in your personal space but procedures will be clearly communicated to you.

- The testing session will require you to perform four (4) different activities several times. The tests will include:
  - 1. Single leg balance on each leg
  - 2. Sit – to – stand – to - sit movement
  - 3. Walking gait tests across 10m
  - 4. Stair climbing of 7 stairs (ascending and descending)
- During these tests EMG will measure the muscle activation while 3D motion analysis data will also be collected.

What will your responsibilities be?

- Should you wish to participate in this study, your responsibility will be to participate in a single testing session on the testing dates communicated to you.
- It is asked of you that you be as honest and open with the researchers as possible.
- You will be asked to transport yourself to the testing session.
- In the case of any adverse events that may prevent your participation, it is your responsibility to inform the researcher as soon as possible so that the necessary adjustments and arrangements to help you may be possible.

Will you benefit from taking part in this research?

- As a participant in the study you will receive a summary report of your results and should any discrepancies be noted advice by a Biokineticist (Sarah Arnold) will be offered.
- Participation in the study will provide us with a better understanding of muscle activation and joint movements when performing daily activities and may reveal areas of weakness or muscle compensation that requires improvement. Regular training to improve technique and muscle strength/ balance has the potential to decrease your risk of excessive joint loading and or overuse injuries.
- From a research perspective the project will allow us to determine what the muscle activation patterns are in single leg below knee amputees and how it compares to that of able bodies. This will help us to improve the knowledge base on below knee amputees and hopefully lead to improved rehabilitation programs to reduce the risk for compensatory injuries.

Are there in risks involved in your taking part in this research?

- Although you will only be performing daily functional activities you may experience mild skin irritations due to the tape used for application of the markers and the electrodes used for EMG. Care will be taken to clean the skin both before and after testing to help limit this. A soothing cream will also be made available during testing to aid the skin irritation.

If you do not agree to take part, what alternatives do you have?

- If you do not agree to take part in the study then on your request you can be placed onto the database to be informed of future studies that you wish to take part in.

Who will have access to your medical records?

- Access to medical records will not be required. However, any information collected in connection with this study and that could be identified with you will remain confidential and will be disclosed only with your permission or as required by law.
- All personal information will be kept confidential and will be de-linked in the transfer from data sheets to the password protected computer. Participants will be assigned a numbered code as their only identifier and any information obtained will only be made accessible to the researchers of the study. No names will be used in the publications of any works as part of the study in order to maintain complete anonymity of the participants.

What will happen in the unlikely event of some form injury occurring as a direct result of your taking part in this research study?

- While this study is minimal risk, should an injury occur as a direct result of taking part in the study, this study is covered by Stellenbosch University's no fault study insurance. Should any adverse events occur then the research team will contact the insurers to arrange for compensation of any medical expenses incurred where applicable.
- The research team consists of a first aider qualified in Basic Life Support and the use of an AED, and will be present at each testing session. The main researcher is also a qualified Biokineticist and there will be a qualified Physiotherapist on the premises.
- The testing is also situated in the vicinity of Tygerberg Hospital.

Will you be paid to take part in this study and are there any costs involved?

- Participants will not receive payment for participation. Each participant will however receive reimbursement for inconvenience and time taken to participate.

- Participants will receive R100 as reimbursement and every effort will be made to provide transport to get to the testing venue.

Is there any thing else that you should know or do?

- You can contact Miss Sarah Arnold at 0725984275 or Dr Suzanne Ferreira at 0218082742 if you have any further queries or encounter any problems.
- You can contact the Health Research Ethics Committee at 021-938 9207 if you have any concerns or complaints that have not been adequately addressed by the researcher.
  - You will receive a copy of this information and consent form for your own records.

Declaration by participant

By signing below, I ………………………………..………… agree to take part in a research study entitled "The effect of a novel energy storage and return foot prosthesis (Pro-flex foot) on biomechanics, function and lifestyle patterns, compared to the gold standard energy storage and return foot prosthesis (Vari-flex foot) and a conventional solid ankle cushioned foot prosthesis."

I declare that:

- I have read or had read to me this information and consent form and it is written in a language with which I am fluent and comfortable.
- I have had a chance to ask questions and all my questions have been adequately answered.
- I understand that taking part in this study is **voluntary** and I have not been pressurised to take part.
- I may choose to leave the study at any time and will not be penalised or prejudiced in any way.
- I may be asked to leave the study before it has finished, if the study doctor or researcher feels it is in my best interests, or if I do not follow the study plan, as agreed to.

Signed at (*place*) ....…........…………….. on (*date*) …………....……….. 2017.

....

Signature of participant **Signature of witness**

Declaration by investigator

I *(name)* ……………………………………………..………. declare that:

- I explained the information in this document to …………………………………..
- I encouraged him/her to ask questions and took adequate time to answer them.
- I am satisfied that he/she adequately understands all aspects of the research, as discussed above
- I did/did not use a interpreter. (*If a interpreter is used then the interpreter must sign the declaration below.*

Signed at (*place*) ....…........…………….. on (*date*) …………....……….. 2017.

....

Signature of investigator **Signature of witness**

Addendum 3

Informed consent
(Control group)

PARTICIPANT INFORMATION LEAFLET AND CONSENT FORM

TITLE OF THE RESEARCH PROJECT:

The effect of a novel energy storage and return foot prosthesis (Pro-flex foot) on biomechanics, function and lifestyle patterns, compared to the gold standard energy storage and return foot prosthesis (Vari-flex foot) and a conventional solid ankle cushioned foot prosthesis

REFERENCE NUMBER: M16/08/032

PRINCIPAL INVESTIGATOR: Professor Wayne Derman

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CONTACT NUMBER: 0725984275

You are being invited to take part in a research study. Please take some time to read the information presented here, which will explain the details of this project. Please ask the researcher any questions about any part of this project that you do not fully understand. It is very important that you are fully satisfied that you clearly understand what this research entails and how you could be involved. Also, your participation is **entirely voluntary** and you are free to decline to participate. If you say no, this will not affect you negatively in any way whatsoever. You are also free to withdraw from the study at any point, even if you do agree to take part.

This study has been approved by the **Health Research Ethics Committee at Stellenbosch University** and will be conducted according to the ethical guidelines and principles of the international Declaration of Helsinki, South African Guidelines for Good Clinical Practice and the Medical Research Council (MRC) Ethical Guidelines for Research.

What is this research study all about?

- This study will take place in the Human Motion Analysis Unit at Stellenbosch University, Tygerberg campus. We would like to recruit 20 participants as part of our able – bodied control group
- The purpose of this study is to gain a greater understanding of the muscle activation patterns of single below knee amputees in comparison to able –

bodied individuals during movements of daily living. This could lead to more information regarding muscle overloading. In other terms we would like to see how the muscles “switch on” when performing activities such as walking up and down stairs, sitting or standing up and general walking and how this may differ between the two groups (amputees and able – bodied group).

- All volunteers will be screened according to the inclusion and exclusion criteria. After the participant has been screened and included the following procedures will occur:
- The testing session will begin with anthropometrical measurements. This will include the measuring of height, body mass and a posture analysis. Photographs will be taken for the posture analysis for analysis on the computer. In terms of privacy and confidentiality, participant’s rights will be respected. All anthropometrical and marker placement will therefore take place in a private room.
- Participants will then be prepared for the biomechanical testing. This entails markers being placed on several anatomical land marks and muscles of the body. The participant will be well informed as to when and which muscle the researcher will be placing the marker on. Marker placement will take place in a private room.
- None of the tests that’s will be completed will be invasive or case harm to you. Marker placement will require the researcher being in your personal space but procedures will be clearly communicated to you.

- The testing session will require you to perform four (4) different activities several times. The tests will include:
    - 1. Single leg balance on each leg
    - 2. Sit – to – stand – to - sit movement
    - 3. Walking gait tests across 10m
    - 4. Stair climbing of 7 stairs (ascending and descending)
- During these tests EMG will measure the muscle activation while 3D motion analysis data will also be collected.

What will your responsibilities be?

- Should you wish to participate in this study, your responsibility will be to participate in a single testing session on the testing dates communicated to you.
- It is asked of you that you be as honest and open with the researchers as possible.
- You will be asked to transport yourself to the testing session.
- In the case of any adverse events that may prevent your participation, it is your responsibility to inform the researcher as soon as possible so that the necessary adjustments and arrangements to help you may be possible.

Will you benefit from taking part in this research?

- Participation in the study will provide us with a better understanding of muscle activation and joint movements when performing daily activities and may reveal areas of weakness or muscle compensation that requires improvement. Regular training to improve technique and muscle strength/ balance has the potential to decrease your risk of excessive joint loading and or overuse injuries.
- From a research perspective the project will allow us to determine what the muscle activation patterns are in single leg below knee amputees and how it compares to that of able bodies. This will help us to improve the knowledge base on below knee amputees and hopefully lead to rehabilitation programs to reduce the risk for compensatory injuries.

Are there in risks involved in your taking part in this research?

- Although you will only be performing daily functional activities you may experience mild skin irritations due to the tape used for application of the markers and the electrodes used for EMG. Care will be taken to clean the skin both before and after testing to help limit this. A soothing cream will also be made available during testing to aid the skin irritation.

If you do not agree to take part, what alternatives do you have?

- If you do not agree to take part in the study then on your request you can be placed onto the database to be informed of future studies that you wish to take part in.

Who will have access to your medical records?

- Access to medical records will not be required. However, any information shared in connection with this study and that could be identified with you will remain confidential and will be disclosed only with your permission or as required by law.

What will happen in the unlikely event of some form injury occurring as a direct result of your taking part in this research study?

- While this study is minimal risk, should an injury occur as a direct result of taking part in the study, this study is covered by Stellenbosch University's no fault study insurance. Should any adverse events occur then the research team will contact the insurers to arrange for compensation of any medical expenses incurred where applicable.
- The research team consists of a first aider qualified in Basic Life Support and the use of an AED, and will be present at each testing session. The main researcher is also a qualified Biokineticist and there will be qualified Physiotherapist on the premises.
- The testing is also situated in the vicinity of Tygerberg Hospital.

Will you be paid to take part in this study and are there any costs involved?

- Participants will not receive payment for participation. Each participant will however receive reimbursement for travel, inconvenience and time taken to participate. The value of the reimbursement will be up to the maximum value of R200 per participant. In addition each participant will receive a summary report of their result and should any discrepancies be noted advice by a Biokineticist (Sarah Arnold) will be offered.

Is there any thing else that you should know or do?

- You can contact Miss Sarah Arnold at 0725984275 or Dr Suzanne Ferreira at 0218082742 if you have any further queries or encounter any problems.

- You can contact the Health Research Ethics Committee at 021-938 9207 if you have any concerns or complaints that have not been adequately addressed by the researcher.

- You will receive a copy of this information and consent form for your own records.

Declaration by participant

By signing below, I …………………………………..………… agree to take part in a research study entitled "The effect of a novel energy storage and return foot prosthesis (Pro-flex foot) on biomechanics, function and lifestyle patterns, compared to the gold standard energy storage and return foot prosthesis (Vari-flex foot) and a conventional solid ankle cushioned foot prosthesis."

I declare that:

- I have read or had read to me this information and consent form and it is written in a language with which I am fluent and comfortable.
- I have had a chance to ask questions and all my questions have been adequately answered.
- I understand that taking part in this study is **voluntary** and I have not been pressurised to take part.
- I may choose to leave the study at any time and will not be penalised or prejudiced in any way.
- I may be asked to leave the study before it has finished, if the study doctor or researcher feels it is in my best interests, or if I do not follow the study plan, as agreed to.

Signed at (*place*) ....…........…………….. on (*date*) …………....……….. 2017.

....	....
Signature of participant	**Signature of witness**

Declaration by investigator

I *(name)* ……………………………………………..……… declare that:

- I explained the information in this document to …………………………………..
- I encouraged him/her to ask questions and took adequate time to answer them.
- I am satisfied that he/she adequately understands all aspects of the research, as discussed above
- I did/did not use a interpreter. (*If a interpreter is used then the interpreter must sign the declaration below.*

Signed at (*place*) ....…........……………….. on (*date*) …………....……….. 2017.

....

Signature of investigator **Signature of witness**

Declaration by interpreter

I *(name)* ……………………………………………..……… declare that:

- I assisted the investigator (*name*) ………………………………………. to explain the information in this document to (*name of participant*) ……………..…………………………….. using the language medium of Afrikaans/Xhosa.
- We encouraged him/her to ask questions and took adequate time to answer them.
- I conveyed a factually correct version of what was related to me.
- I am satisfied that the participant fully understands the content of this informed consent document and has had all his/her question satisfactorily answered.

Signed at (*place*) ....…........……………….. on (*date*) …………....………………..

....

Signature of interpreter **Signature of witness**

Addendum 4

Gait and Posture guidelines

Gait & Posture 37 (2013) I–IV

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www . elsevier . com/locate/gaitpost

Instructions to Authors

These instructions for authors, along with information about copyright, can also be found on the Internet: access under http://www.elsevier.com/locate/gaitpost

Authors should submit papers online via http://ees. elsevier.com/gaipos. This is the Elsevier web-based submission and review system. You will find full instructions located on this site in the Tutorial for Authors. Please follow the guidelines to prepare and upload your article. Once the uploading is done, the system automatically creates an electronic pdf which is used for reviewing. All correspondence, including notification of the Editor's decision and requests for revisions, will be managed via this system.

Preparation of the Manuscript

1. Article types accepted are: Original Article (Full Paper or Short Communication), Review Article, Technical Note, Book Review. Word limits including the abstract are as follows: Full paper 3,000 words plus no more than 5 figures/tables in total; Short Paper or Techni-cal Note 1,200 words plus no more than 3 figures/tables in total. If the Editor feels that a paper submitted as a Full Paper would be more appropriate for the Short Com-munications Section, then a shortened version will be re-quested. References should be limited to 30 for Full Papers, 15 for Short Papers and 10 for Technical Notes. An abstract not exceeding one paragraph of 250 words should appear at the beginning of each Article. The recommended word limit for Review Papers is 6,000 words. Authors must state the number of words when submitting.
2. All publications will be in English. Authors whose 'first' language is not English should arrange for their manuscripts to be written in idiomatic English **before** submission. A concise style avoiding jargon is preferred.
3. Authors should supply up to five keywords that may be modified by the Editors.
4. Acknowledgements should be included after the end of the Discussion and just before the References. Include external sources of support.
5. The text should be ready for setting in type and should be **carefully checked** for errors. Scripts should be typed double-spaced on one side of the paper only. Please do not underline anything, leave wide margins and number every sheet.
6. All illustrations should accompany the typescript, **but not** be inserted in the text. Refer to photographs, charts, and diagrams as 'figures' and number consecutively in order of appearance in the text. Substantive captions for each figure explaining the major point or points should be typed on a separate sheet.
7. Tables should be presented on separate sheets of paper and labelled consecutively but the captions should accompany the table.

Summary of Overall Arrangement of Manuscripts

You should arrange your contribution in the following order:

1. A cover page with complete details of the title, the source, and the authors full contact details.
2. An abstract outlining the purpose, scope and conclusions of the paper.
3. The text suitably divided under headings. (frequently Introduction, Material or Patients, Methods, Results, Discussion will prove satisfactory)
4. Acknowledgements (if any).
5. References.
6. Tables with captions (each on a separate sheet).
7. Captions to illustrations (grouped on a separate sheet or sheets).
8. Illustrations, each on a separate sheet containing no text.
9. Supplementary data.

Illustrations

Authors are required to provide electronic versions of their illustrations. Information relating to the preferred formats for artwork may be found at http://authors.elsevier.com and also on the online submission homepage http://ees.elsevier. com/ gaipos.

Photographs:

Colour reproduction is available if the author is willing to bear the additional printing costs. Patients must not be identifiable from a photograph. If it is essential for a patient's face to be shown, written permission must be obtained from the patient and submitted with the manuscript.

References

Indicate references to the literature in the text by superior Arabic numerals that run consecutively through the paper in order of their appearance. Where you cite a reference more than once in the text, use the same number each time.

doi:10.1016/S0966-6362(12)00423-7

References should take the following form:

1. Amis AA, Dawkins GPC. Functional anatomy of the anterior cruciate ligament. J Bone Joint Surg [Br] 1991; 73B:260–267.
2. Insall JN. Surgery of the Knee. New York: Churchill Living-stone; 1984.
3. Shumway-Cook A, Woollacott M. Motor Control: The-ory and Practical Applications. Baltimore: Williams and Wilkins; 1995.

Please ensure that references are complete, i.e. that they in-clude, where relevant, author's name, article or book title, vol-ume and issue number, publisher, year and page reference *and* comply with the reference style of Gait & Posture. Only salient and significant references should be included.

What Information to Include with the Manuscript

1. Having read the criteria for submissions, authors should specify in their letter of transmittal whether they are submitting their work as an Original Article (Full Paper or Short Communication), Review Article, Technical Note, or Book Review. Emphasis will be placed upon originality of concept and execution. Only papers not previously published will be accepted. Comments re-garding articles published in the Journal are solicited and should be sent as "Letter to the Editor". Such Let-ters are subject to editorial review. They should be brief and succinct. When a published article is subjected to comment or criticism, the authors of that article will be invited to write a letter or reply.

2. A letter of transmittal must include the statement, "Each of the authors has read and concurs with the content in the final manuscript. The material within has not been and will not be submitted for publication elsewhere except as an abstract." The letter of transmittal must be from all co-au-thors. All authors should have made substantial contribu-tions to all of the following: (1) the conception and design of the study, or acquisition of data, or analysis and interpre-tation of data, (2) drafting the article or revising it critically for important intellectual content, (3) final approval of the version to be submitted.
3. Acknowledgement of other contributors. All contribu-tors who do not meet the criteria for authorship as de-fined above should be listed in an acknowledgements section. Examples of those who might be acknowledged include a person who provided purely technical help, writ-ing assistance, or a department chair who provided only general support. Authors should disclose whether they had any writing assistance and identify the entity that paid for this assistance.
4. Work on human beings that is submitted to *Gait & Pos-ture* should comply with the principles laid down in the Declaration of Helsinki; Recommendations guid-ing physicians in biomedical research involving human subjects. Adopted by the 18th World Medical Assembly, Helsinki, Finland, June 1964, amended by the 29th World Medical Assembly, Tokyo, Japan, October 1975, the 35th World Medical Assembly, Venice, Italy, October 1983, and the 41st World Medical Assembly, Hong Kong, September 1989. The manuscript should contain a statement that the work has been approved by the appropriate ethical com-mittees related to the institution(s) in which it was per-formed and that subjects gave informed consent to the work. Studies involving experiments with animals must state that their care was in accordance with institution guidelines. Patients' and volunteers' names, initials, and hospital numbers should not be used.
5. At the end of the text, under a subheading "Conflict of interest statement" all authors must disclose any financial and personal relationships with other people or organisations that could inappropriately influence (bias) their work. Examples of potential conflicts of interest include employment, consultancies, stock ownership, honoraria, paid expert testimony, patent applications/ registrations, and grants or other funding.
6. All sources of funding should be declared as an acknowledgement at the end of the text. Authors should declare the role of study sponsors, if any, in the study de-sign, in the collection, analysis and interpretation of data; in the writing of the manuscript; and in the decision to submit the manuscript for publication. If the study spon-sors had no such involvement, the authors should so state.
7. Authors are encouraged to suggest referees although the choice is left to the Editors. If you do, please supply their postal address and email address, if known to you.
8. Please note that papers are subject to double-blind review. Therefore any information that reveals where and by whom the study was undertaken must be removed from the manuscript.

Randomised Controlled Trials

All randomised controlled trials submitted for publication in *Gait & Posture* should include a completed Consolidated Stand-ards of Reporting Trials (CONSORT) flow chart. Please refer to the CONSORT statement website at http://www.consort-state-ment.org for more information. The Journal has adopted the proposal from the International Committee of Medical Journal Editors (ICMJE) which require, as a condition of consideration for publication of clinical trials, registration in a public trials registry. Trials must register at or before the onset of patient enrolment. The clinical trial registration number should be in-cluded at the end of the abstract of the article. For this purpose, a clinical trial is defined as any research project that prospec-tively assigns human subjects to intervention or comparison groups to study the cause-and-effect relationship between a medical intervention and a health outcome. Studies designed for other purposes, such as to study pharmacokinetics or major toxicity (e.g. phase I trials) would be exempt. Further informa-tion can be found at www.icmje.org.

Technical Notes

1. Technical Notes should be 1,200 words in length at most.
2. There should be no more than 3 figures/tables in total, and no more than 10 key references.
3. A technical note should explain or report on the development of a technical device or method that is specifically new or novel but is not a commercial product development.
4. Format of the Technical Note: 1. Introduction, 2. Technique description, 3. Discussion focusing on the role of the tech-nique, especially advantages and liabilities related to other options.
5. On acceptance of the Technical Note, authors will be re-quested to disclose any possible conflicts of interest.

Letter to Editors

Comments regarding articles published in the Journal are solicited and should be sent as ''Letter to the Editors''. Such Letters are subject to editorial review. They should be brief and succinct. When a published article is subjected to comment or criticism, the authors of that article will be invited to write a letter or reply.

Review and Publication Process

1. You will receive an acknowledgement of receipt of the manuscript by the Editorial Office before the manu-script is sent to referees. Please contact the appropriate Editor-in-Chief if you do not receive an acknowledge-ment.

Following assessment one of the following will happen:

A: The paper will be accepted directly. The corresponding author will be notified of acceptance. The Editor-in-Chief will send the accepted paper to Elsevier for publication.

B: The paper will be accepted subject to minor amendments. The corrections should be made and the paper returned to the Editor-in-Chief for checking. Once the paper is accepted it will be sent to production.

C: The paper will be rejected but resubmission invited after a major revision. A complete resubmission is required as the paper will be re-evaluated by referees and assessment will start again.

D: The paper will be rejected outright as being unsuitable for publication in Gait and Posture.

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4. An order form for reprints will accompany the proofs.

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Addendum 5

Prosthetics and Orthotics International guidelines

Prosthetics and Orthotics International

2017 Impact Factor: 1.097
2017 Ranking: 58/77 in Orthopedics | 48/65 in Rehabilitation (SCI)
Source: Journal Citation Reports®, 2018 release, a Clarivate Analytics product; Indexed in PubMed: MEDLINE

Published in Association with International Society for Prosthetics and Orthotics

1. What do we publish?

1.1 Aims & Scope

Before submitting your manuscript to *Prosthetics and Orthotics International*, please ensure you have read the Aims & Scope.

1.2 Article Types

Prosthetics and Orthotics International is a peer-reviewed journal welcoming the following article types for publication.

1.2.1 Original Research Report

This form of manuscript presents new clinical and experimental findings that advance the clinical and theoretical fields of orthotics, prosthetics, and related areas. The main text of the manuscript should not exceed 3,000 words (please refer to Table 1). The abstract for a Research Report has a 200 word limit and should contain the following subheadings: Study Design; Background; Objectives; Methods; Results; and Conclusions. Please state the word count on a line below the abstract.

Under the abstract, a Clinical Relevance statement of no more than 50 words should be provided. This should provide information on the potential application or impact of the study regarding clinical practice. The author(s) may also make reference to how the findings of the study contribute to the overall understanding of the topic. Please state the word count on a line below the statement.

The main body of an Original Research Report should include the following sections: Background; Methods; Results; Discussion; Conclusion; and References.

The Background should introduce the topic area etc. and provide a succinct review of the relevant literature. This section concludes with the objectives/purpose of the study and by stating a hypothesis. The Methods section provides information on the design of the study (i.e. retrospective, randomised controlled trial). Suggested details include the study population, inclusion and exclusion criteria, ethical approval (consent), equipment, procedures, methods of analysis, and data analysis. The Results section must contain a detailed presentation of the data analysed and must refer to supporting information presented within tables and figures and their related legends. The Discussion should be concise and focus on your main findings - was the hypothesis supported or rejected? Within this section authors should also compare and contrast the findings of their study with the existing literature. The strengths and limitations of the study should also be addressed along with suggestions for further research. Where possible, emphasis should be placed on the application to clinical practice. The Conclusion provides a different or novel view of the problem you outlined in your Background. Conclusions made should be supported by your findings. Please state the word count after the Conclusion.

1.2.2 Review

There are various types of reviews that can be submitted to *Prosthetics and Orthotics International.* A traditional Literature Review provides a critical synthesis of orthotics, prosthetics or related topics. The limit for this type of article is 5,000 words and the length of the work submitted should be stated prior to the reference section. The abstract should not exceed 200 words and should be structured using the following subheadings: Study Design (i.e. Literature Review); Background; Objectives; Methods; Results; Conclusions; and References. Please state the word count on a line below the abstract.

A Clinical Relevance statement of no more than 50 words should also be provided. This should inform the reader of the potential application to or impact on clinical practice. The author(s) may also make reference to how the findings of the current review contribute to the overall understanding of the topic. The word count should be provided on a line below the statement.

The journal also welcomes submissions based on Systematic Reviews and Meta-analyses. The guidelines for the abstract are the same as those for the Literature Review. Authors should note that the overall maximum word counts for Systematic Reviews and Meta-analyses are 4,000 and 3,500 words respectively (please refer to Table 1). The Abstract and Clinical Relevance statements for both sets of reviews are identical to those of the literature review.

1.2.3 Technical Note

A Technical Note is a technical-research based commentary which preferably addresses a current issue. The limit for a technical note is 1,200 - 1,500 words. Specifically the article should provide a succinct and balanced summary of equipment, procedures, or particular (innovative) technological approaches used in the field. The abstract should adhere to a 150 word limit, and should include the following 3 subheadings: Background and Aim; Technique; and Discussion. A Clinical Relevance statement should also be provided under the abstract and should not exceed 35 words. This should provide a key point regarding the potential application to or impact on clinical practice. The word counts for the abstract and Clinical Relevance statement respectively should be provided on a line below each section.

The subheadings within the main text should follow those of the abstract: Background and Aim; Technique; and Discussion, followed by an additional ‘Key Points’ section summarising the work in three to four bullet points (see Table 1). The word count for the main text should be stated beneath this section, prior to the References section.

1.2.4 Clinical Note

Clinical Notes are limited to 1,200 - 1,500 words and provide a clinical-research based commentary which preferably addresses a current issue. Articles should give a succinct and balanced summary of a particular (innovative) approach or procedure that enhances clinical practice and understanding. The abstract is also structured with a 150 word limit and should include the following three subheadings: Background and Aim; Technique; and Discussion. A Clinical Relevance statement of no more than 35 words should also be provided underneath the abstract. This should provide a key point on the potential application to or impact on clinical practice. Word counts for the abstract and Clinical Relevance statement should be provided on a line below their respective sections.

The subheadings within the main text should follow those of the abstract: Background and Aim; Technique; and Discussion, followed by an additional ‘Key Points’ section summarising the work in three to four bullet points (see Table 1). The word count for the main text should be stated beneath this section, prior to the References section.

1.2.5 Case Report

Case Reports may include a single Case Study or a Case Series (group of patients). These should be between 1,500 and 2,000 words. They should be based on an intervention or interesting observation of a unique clinical case. The report should include a structured abstract of 150 words using the following subheadings: Background (include aim/purpose); Case Description and Methods; Findings; and Outcomes and Conclusion. As with other articles, a Clinical Relevance statement of no more than 35 words should be provided under the abstract. This statement may include the potential impact on clinical practice and recognition of the key features of the unique case. The word counts for the abstract and Clinical Relevance statement should be provided on a line below their respective sections.

Within the main text the following subheadings should be used: Background (including aim); Case Description and Methods; Findings and Outcomes; and Discussion and Conclusion. Information presented with the Case Description should include patient characteristics, assessment, diagnosis (differential diagnosis if required), methods of assessment, and management strategies employed (please refer to Table 1). This information however will be individual to each Case Report. The word count for the main text should be stated after the Conclusion, prior to the References section.

Authors are required to ensure the anonymity of the participant(s) (names, uniquely identifying personal descriptors, detailed family trees, and geographic location should not be included). Please check with your Institutional or Local Ethics Committee regarding specific requirements for consent on Case Reports. Normally, ethical approval should be sought and adhered to as outlined for 'Original Research Reports' above. There should be an introductory section as summarized above, followed by the presentation of the case. Where applicable, the steps undertaken to address the clinical problem should be explained in detail.

1.2.6 Expert Clinical Viewpoint

This type of article is by invitation only from the Editor-in-Chief. The content, commentary, and nature of the article may follow a particular article published in the same issue or it could be presented in the form of a 'current concept' approach of key topic/area(s).

1.2.7 Letter to the Editor

Readers' letters to the Editor-in-Chief are welcomed and should address issues raised by published articles or report significant new findings that merit rapid dissemination. The decision to publish is made by the Editor-in-Chief. To submit a letter, please email the Editor-in-Chief with text attached. Please note that all letters are copyedited prior to publication (please refer to Table 1).

1.2.8 Book Review

A list of up-to-date books for review is available from the SAGE Ltd website (please refer to Table 1).

Table 1.* Overview of recommended maximums for manuscript submission to *Prosthetics and Orthotics International.

Artcle Type	**Word Limit**	**References**	**Figures** (*Parts to figure- i.e. A and B*)	**Tables**
Original Research Report	3,000	35	5(10)	3
Review				
Traditional review	5,000	100	7(15)	5
Systematic review	4,000	60	5(10)	3
Meta-analysis	3,500	60	5(10)	3
Case Report (Series)	1,250- 1,500	12	2(4)	1
Technical Note	1,500	15	3(6)	2

Clinical Note	1,500	15	3(6)	2
Letter to the Editor	500	3	1(2)	1
Book Review	750	0	0	0

1.3 Writing your paper

The SAGE Author Gateway has some general advice and on how to get published, plus links to further resources.

1.3.1 Make your article discoverable

When writing up your paper, think about how you can make it discoverable. The title, keywords and abstract are key to ensuring readers find your article through search engines such as Google. For information and guidance on how best to title your article, write your abstract and select your keywords, have a look at this page on the Gateway: How to Help Readers Find Your Article Online.

Back to top

2. Editorial policies

2.1 Peer review policy

Prosthetic and Orthotics International operates a strictly anonymous, double-blinded peer review process in which the reviewer's name is withheld from the author, and the author's name from the reviewer. Please supply any personal acknowledgements separately to the main text to facilitate anonymous peer review.

All manuscripts are reviewed as rapidly as possible, and an editorial decision is generally reached within 4-6 weeks of submission.

2.2 Authorship

Papers should only be submitted for consideration once consent is given by all contributing authors. Those submitting papers should carefully check that all those whose work contributed to the paper are acknowledged as contributing authors.

The list of authors should include all those who can legitimately claim authorship. This is all those who:

1.
    1. Made a substantial contribution to the concept or design of the work; or acquisition, analysis or interpretation of data,
    2. Drafted the article or revised it critically for important intellectual content,
    3. Approved the version to be published,
    4. Each author should have participated sufficiently in the work to take public responsibility for appropriate portions of the content.

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Addendum 6

Coefficient of variation tables

Table A6.1 Average muscle activation (%) and coefficient of variation (%) during the SiSt

Muscle	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
TA			36.88±15.68	42.53	26.42±16.34	61.86	29.45±15.92	54.06
MG			3.93±1.58	40.30	22.93±8.02	34.96	3.19±2.68	83.84
LES	35.48 ±13.37	37.70	33.91±15.92	46.95	25.88±10.94	42.28	7.94±25.75	324.48
Gmax	12.87±6.62	51.45	11.89±6.04	50.82	11.86±9.68	81.59	6.16±4.47	72.48
BF	8.26±4.23	51.26	9.08±5.45	60.05	4.72±3.58	75.80	4.75±3.08	64.81
Gmed	8.63±5.71	66.06	9.39±4.35	46.35	7.90±5.23	66.17	6.32±5.51	87.19
VL	6.68±5.57	83.28	21.13±12.95	61.26	21.30±8.93	41.95	24.36±10.91	44.78

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, LES = lumbar erector spinae, VL = vastus lateralis, Gmed = gluteus medius, Gmax = gluteus maximus, BF = bicep femoris, MG = medial gastrocnemius, TA = tibialis anterior, SD = standard deviation, CV = Coefficient of variation

Table A6.2 Average muscle activation (%) and coefficient of variation (%) during the StSi

Muscle	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
TA			31.88±20.03	62.82	20.73±14.84	71.59	24.02±17.40	72.44
MG			4.18±2.63	62.90	19.25±3.99	20.74	7.60±4.74	62.40
LES	17.97±8.99	50.02	15.47±6.37	41.17	22.89±11.37	49.66	9.41±21.91	232.89
Gmax	5.63±2.43	43.17	5.30±3.30	62.27	10.40±11.66	112.14	5.34±6.16	115.44
BF	5.61±4.09	73.01	9.38±9.57	102.03	5.58±7.25	129.76	7.92±6.59	83.14
Gmed	4.29±1.97	45.89	4.48±1.91	42.71	14.70±14.60	99.34	8.62±5.46	63.39
VL	3.96±3.78	95.59	14.07±10.68	75.95	17.88±7.69	43.03	22.38±12.66	56.59

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, StSi = stand-to-sit, LES = lumbar erector spinae, VL = vastus lateralis, Gmed = gluteus medius, Gmax = gluteus maximus, BF = bicep femoris, MG = medial gastrocnemius, TA = tibialis anterior, SD = standard deviation, CV = Coefficient of variation

Table A6.3 Average muscle activation (%) and coefficient of variation (%) during SLB

Muscle	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
LES	7.51±5.10	67.87	1.82±1.32	72.27	2.14±0.91	42.28	2.33±1.21	51.66
VL	5.30±4.31	81.33	1.96±1.40	71.41	4.86±4.13	85.15	4.25±3.06	72.07
Gmed	13.32±6.58	49.45	5.10±2.50	49.12	9.93±6.61	66.59	7.45±4.38	58.71
Gmax	5.85±4.64	79.32	2.12±2.27	106.93	2.65±1.48	55.66	1.35±0.88	65.04
BF	14.56±8.31	57.04	5.84±3.80	65.08	2.65±2.84	107.56	3.67±3.09	84.18
MG			11.41±6.09	53.39	10.26±6.68	65.10	11.52±5.52	47.96
TA			9.65±6.53	67.64	7.26±3.83	52.72	5.94±3.79	63.88

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SLB = single leg balance, LES = lumbar erector spinae, VL = vastus lateralis, Gmed = gluteus medius, Gmax = gluteus maximus, BF = bicep femoris, MG = medial gastrocnemius, TA = tibialis anterior, SD = standard deviation, CV = Coefficient of variation

Table A6.4 Peak average joint moments (N.m.kg^{-1}) and coefficient of variation (%) in the frontal plane for the hip, knee and ankle during SLB

Variable	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip flexion moment	0.23±0.24	102.04	-0.21±0.15	70.99	-0.18±0.31	172.17	-0.23±0.25	111.87
Knee flexion moment	0.19±0.26	138.84	0.17±0.15	89.01	0.27±0.16	58.04	0.22±0.23	105.45
Ankle flexion moment	0.76±0.18	23.88	0.79±0.10	12.79	0.77±0.11	14.39	0.76±0.12	16.19

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SLB = single leg balance, SD = standard deviation, CV = Coefficient of variation

Table A6.5 Peak average joint power (W.kg^{-1}) and coefficient of variation (%) for the hip, knee and ankle during SiSt and StSi

Joint	Variable	AF		UN		D		ND	
		$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip	Peak power SiSt	1.19±0.31	26.28	0.78±0.50	63.75	0.86±0.29	33.80	0.87±0.33	38.20
	Peak power StSi	-0.97±0.32	33.23	-0.65±0.42	64.00	-0.88±0.35	39.53	-0.90±0.37	40.41
Knee	Peak power SiSt	0.42±0.26	65.53	1.33±0.34	25.58	1.32±0.33	24.72	1.28±0.32	25.02
	Peak power StSi	-0.23±0.15	62.25	-0.86±0.26	30.43	-1.01±0.29	28.40	-0.99±0.24	23.75
Ankle	Peak power SiSt	0.07± 0.04	58.73	0.09±0.07	76.11	0.14±0.07	47.84	0.16±0.05	34.78
	Peak power StSi	-0.11±0.07	61.69	-0.07±0.05	66.76	-0.08±0.03	36.23	-0.10±0.03	32.37

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SLB = single leg balance, SD = standard deviation, CV = Coefficient of variation

Table A6.6 Peak average joint flexion angles (°) and coefficient of variation (%) for the hip, knee and ankle during SiSt and StSi

Joint	Variable	AF		UN		D		ND	
		$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip	SiSt	94.41±10.97	11.62	91.47±11.67	12.75	86.75±7.79	8.97	85.58±8.58	10.02
	StSi	94.64±10.44	11.03	92.05±10.51	11.41	89.98±7.40	8.23	88.80±7.74	8.71
Knee	SiSt	81.61±7.66	9.38	82.33±6.81	8.27	86.24±6.88	7.97	87.48±7.36	8.41
	StSi	81.52±7.80	9.57	81.63±6.84	8.37	86.28±6.77	7.85	87.39±7.02	8.03
Ankle	SiSt	10.57±2.68	25.32	18.88±6.06	32.07	20.98±2.49	11.89	22.15±3.11	14.03
	StSi	10.08±2.88	28.59	18.59±6.73	36.17	19.47±2.52	12.93	20.85±3.30	15.83

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, StSi = stand-to-sit, SD = standard deviation, CV = Coefficient of variation

Table A6.7 Average joint flexion angles (°) and coefficient of variation (%) for the hip, knee and ankle during SLB

Joint	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip	28.18±9.70	34.41	15.41±6.84	44.40	11.05±10.72	96.97	8.95±9.84	109.94
Knee	21.89±10.39	47.47	11.59±4.34	37.40	12.39±8.20	66.20	9.57±9.47	98.97
Ankle	10.83±3.45	31.87	10.05±2.46	24.48	10.53±3.26	30.99	9.42±3.96	42.05

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SLB = single leg balance

Table A6.8 Peak average joint power (W.kg^{-1}) and coefficient of variation (%) for the hip, knee and ankle during SLB

Joint	AF	CV	UN	CV	D	CV	ND	CV
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip	0.024±0.21	86.81	0.04±0.05	112.25	0.12±0.05	44.61	0.11±0.06	53.86
Knee	0.15±0.21	141.80	0.02±0.02	131.85	0.05±0.04	75.87	0.04±0.05	123.24
ankle	0.18±0.18	100.54	0.04±0.04	116.15	0.07±0.03	48.61	0.07±0.05	47.06

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, StSi = stand-to-sit, SD = standard deviation, CV = Coefficient of variation

Table A6.9 Peak average vertical ground reaction force (N.kg^{-1}) and coefficient of variation (%) during SiSt and StSi

	AF		UN		D		ND	
	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
SiSt	52.29±5.45	10.41	69.90±5.77	8.25	57.40±3.86	6.72	56.34±3.26	5.79
StSi	49.84±4.53	9.08	65.81±5.95	9.04	55.80±3.79	6.80	54.84±3.32	6.06

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, StSi = stand-to-sit, SD = standard deviation, CV = Coefficient of variation

Table A6.10 Peak average joint moments ($N.m.kg^{-1}$) and coefficient of variation (%) in the frontal plane for the hip, knee and ankle during SiSt and StSi

Joint	Variable	AF		UN		D		ND	
		$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV	$\bar{x}$ ± SD	CV
Hip	SiSt	0.87±0.20	22.54	0.90±0.22	24.76	0.70±0.21	29.81	0.69±0.23	33.77
	StSi	0.78±0.21	27.36	0.89±0.19	21.95	0.68±0.20	28.67	0.73±0.21	28.70
Knee	SiSt	0.30±0.14	46.70	0.95±0.22	23.06	0.86±0.15	17.67	0.88±0.18	19.84
	StSi	0.21±0.10	46.15	0.79±0.22	27.63	0.79±0.13	16.27	0.76±0.14	17.87
Ankle	SiSt	0.46±0.13	28.13	0.37±0.09	24.65	0.39±0.06	16.45	0.37±0.08	21.31
	StSi	0.42±0.14	33.88	0.31±0.11	36.18	0.34±0.08	23.27	0.34±0.09	25.72

AF = affected, UN = unaffected, ND = non-dominant, D = dominant, SiSt = sit-to-stand, StSi = stand-to-sit, SD = standard deviation, CV = Coefficient of variation