THE UNIVERSITY OF HULL

Biomechanical and psychological factors that distinguish fallers from non-fallers: A comparative study of transtibial amputees and able-bodied individuals

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by

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Abstract

Transtibial amputees are at a higher risk of falling than age-matched able-bodied individuals. In order to make recommendations for falls prevention and treatment programmes, it is important to have a sound understanding of the underlying biomechanical function in persons at high risk of falling. While biomechanical differences between fallers and non-fallers have been identified in older adults, no research to date has specifically compared amputee fallers and non-fallers. The aim of this thesis was to undertake a biomechanical comparison of amputee and able-bodied fallers and non-fallers performing activities of daily living. A secondary aim was to investigate the effect of falls on balance confidence and quality of life and to determine whether a relationship existed between functional and psychological measures according to a person's falls history. Twenty participants (11 transtibial amputees and 9 controls) took part in several studies including a kinematic and kinetic analysis of level walking, stair ascent and descent using a 3-step staircase. They also completed the Sensory Organisation Test (SOT) and Motor Control Test (MCT) on the NeuroCom EquiTest and their postural control was measured in static and dynamic conditions. Participants completed the MFES and SF-36 psychological instruments aimed at quantifying balance confidence and perceived quality of life, respectively. The first study investigated how falls were monitored by physiotherapists and the use of outcome measures in amputee rehabilitation in England. Shortcomings were identified in amputee rehabilitation in that physiotherapists did not monitor falls incidence regularly among their amputee patients and that there was no consensus on the types of recommended outcome measures. The second study explored the biomechanical differences between fallers and non-fallers during level walking and the findings indicated that the amputee fallers had a significantly larger vertical GRF with respect to body weight during loading on the affected limb ($p=0.01$) and consequently loaded their affected limb significantly more than the non-fallers ($p = 0.03$). The opposite finding was reported in the control group, where the non-fallers had significantly greater load rates compared to the fallers ($p=0.02$). The amputee fallers also had significantly different power profiles at the hip (power absorption in stance, $p=0.01$) and the ankle (power generation in pre-swing, $p=0.04$) during the transition from double to single support on the affected leg. In the third study, biomechanical differences were examined during stair ascent revealing that the fallers walked significantly faster up stairs than the non-fallers ($p=0.05$) in the amputee groups, while the opposite was observed in the control groups ($p=0.03$). Kinematic differences were revealed, such as significantly increased knee ROM in both groups of fallers when compared to their non-faller counterparts (p=0.04 and p=0.05 for the amputee and control groups, respectively). The amputee fallers had significantly larger vertical GRF peaks (p=0.01 and p=0.00, respectively), decay rate (p=0.01), ankle plantarflexor moment (p=0.01) and knee joint powers (power absorption in pre- and mid-swing, p=0.00 and p=0.01, respectively) on the intact limb compared to the non-fallers. A forth study exploring gait patterns during stair descent revealed that some amputees used a modified stepping strategy during stair locomotion by adopting a ‘step to’ pattern. The fifth study used computerised dynamic posturography with the Neurocom Equitest to understand how fallers and non-fallers maintained postural control under static and dynamic conditions. The results demonstrated that the amputee fallers scored significantly better on the equilibrium score on the SOT when visual and somatosensory input was inaccurate (p=0.05) (indicating less postural sway). The amputee fallers also bore significantly more weight through their affected limb during destabilising backwards and forwards translations, while the amputee non-fallers bore more weight through their intact limb ($p < 0.05$). The Neurocom Equitest was deemed population-specific and therefore not an appropriate diagnostic tool for identifying fallers in a community-dwelling amputee population. In the final study, relationships between functional performance tests and balance confidence and quality of life were made between the fallers and non-fallers. The psychological instruments revealed that fallers had significantly lower balance confidence on outdoor-type activities (p=0.03) and this was correlated with poor functional performance. The fallers rated their general health (p=0.01), vitality (p=0.03) and emotional role (p=0.04) significantly lower than the non-fallers. In conclusion, the novel results presented in this thesis have important implications for amputee rehabilitation and falls prevention and treatment programmes. These include identifying which muscle groups would benefit from targeted strength exercises and how this would influence gait parameters in key phases in the gait cycle during level walking, stair ascent and descent.

Keywords: amputee; falls; biomechanics; gait; stairs; posture; outcome measures; rehabilitation

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Vanicek N, Strike S, McNaughton L, Polman R. Postural responses to dynamic perturbations in amputee fallers vs. non-fallers: a comparative study with able-bodied subjects. *Archives of Physical Medicine & Rehabilitation* 2009b; 90: 1018-1025.

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Nomenclature

ABC scale	Activities-specific Balance Confidence scale
ASIS	anterior superior iliac spine
BACPAR	British Association of Chartered Physiotherapists in Amputee Rehabilitation
BOS	base of support
CDP	computerised dynamic posturography
CNS	central nervous system
COG	centre of gravity
COM	centre of mass
COP	centre of pressure
FES	Falls Efficacy Scale
GRF	ground reaction force
GRFv	ground reaction force vector
HEY	Hull & East Yorkshire
IC	initial (foot) contact
LCI	Locomotor Capabilities Index
LHJC	left hip joint centre
LREC	Local Research Ethics Commitee
MCT	Motor Control Test
MFES	Modified Falls Efficacy Scale
NASDAB	National Amputee Statistical Database
NHS	National Health Service
PSIS	posterior superior iliac spine
REC	Research Ethics Committee
RHJC	right hip joint centre
ROM	range of motion
SAFE	Survey of Falling and Fear of Falling in the Elderly
SCS	Segment Coordinate System
SD	standard deviation
SF-36	Short Form 36 Health Survey

SIGAM	Special Interest Group in Amputee Medicine
SOT	Sensory Organisation Test
TO	toe off
TUG test	Timed Up and Go test

CHAPTER ONE - INTRODUCTION

People are living longer and our population is rapidly ageing, and the financial cost of falls has been increasing. In 1999, it was estimated that the financial impact of falls on the UK government was almost £1 billion, and cost the NHS almost 60% of that amount (Scuffham *et al*., 2003). The highest costs were incurred in the age group > 75 years. Falls may have severe consequences for the person. Injuries from a fall may require hospitalisation, long-term care or even institutionalisation. The impact can also be psychological and a fall may lead to increased fear of falling and low balance confidence, activity avoidance and therefore poorer quality-of-life (Li *et al*., 2003).

Either for traumatic or vascular reasons, transtibial amputees have had their leg amputated below the knee and therefore have modified locomotor systems. A prosthesis is typically prescribed to provide the structural support needed to ambulate. However, some muscle function has been lost and somatosensory feedback from muscles and receptors has been altered. These factors predispose transtibial amputees to falling. Previous research has reported that lower-limb amputees fall more than age-matched, able-bodied individuals over a one-year period (Miller *et al*., 2001; Miller *et al*., 2003). The age-related decline in musculoskeletal function affects locomotor function and the ability to perform daily tasks safely (Mian *et al*., 2007). Gait and balance impairments, as a result of amputation or ageing, are major risk factors for falls (Mian *et al*., 2007). Locomotor dysfunction and mobility difficulties have been cited as the main risk factors for falls in older adults (American Geriatrics Society, 2001).

Transtibial amputees have altered locomotor function and must function within their new level of constraints. Effective amputee rehabilitation and exercise programmes have the potential to improve locomotor performance, reduce falls and improve general health and quality-of-life. In 2003, the British Association of Chartered Physiotherapists in Amputee Rehabilitation (BACPAR), a clinical interest group of the Chartered Society of Physiotherapists, published evidence based guidelines on physiotherapy treatment in lower-limb amputees (BACPAR, 2003). The guidelines provided insight on physiotherapy practice in this patient population, and advocated a multidisciplinary team approach to prosthetic rehabilitation, patient education, and discharge and maintenance procedures. There was little mention of falls prevention or treatment programmes and no recommendations for targeted exercises were made. The only information about falls management was that as the risk factors for falling were increased in lower-limb amputees, rehabilitation programmes should include education on falls prevention and coping strategies if a fall occurred, and that patients and/or their carers should be taught how to get up off the floor. Moreover, few recommendations were made about the informed use of outcome measures, specifically which measures should be used and how frequently and how the results could influence rehabilitation.

In 2006, the BACPAR guidelines were reviewed but the section on falls management was not modified or updated. The guidelines included limited evidence-based recommendations on exercise programme and falls management. Two points were noted on exercise programmes recommending that 1) they should target the hip flexors, hip extensors and ankle plantarflexors and that 2) they should be relevant to the patient's goals.

Based on the current BACPAR guidelines, it is not clear how falls are monitored by amputee physiotherapists and which muscles should be targeted in exercise programmes to improve locomotor function. In order to make recommendations for falls prevention and treatment programmes, it is important to have a sound understanding of the underlying biomechanical function in persons at high risk of falling. While much research has examined the musculoskeletal differences in older fallers and non-fallers, no scientific studies have made the same comparisons in lower-limb amputees.

This thesis compares the biomechanics of functional daily activities, balance confidence and perceived quality-of-life in amputee fallers and non-fallers. The overall aim of this thesis was to assist the clinical recommendations for amputee rehabilitation to reduce falls incidence and to improve falls monitoring, prevention and treatment in physiotherapy practice. This was done by undertaking a review of current amputee rehabilitation as it relates to falls monitoring and the use of outcome measures to inform amputee practice. Biomechanical analyses of level waking, stair ascent and descent, and postural control between fallers and non-fallers in transtibial amputees and able-bodied participants were then completed to determine whether biomechanical differences existed between these groups. The BACPAR guidelines advocate the use of psychological tools as important in amputee rehabilitation. Therefore, balance confidence and quality-of-life measures in fallers and non-fallers were examined using psychological instruments and their relationship with functional measures were explored. If biomechanical and psychological differences could be identified between fallers and non-fallers, then awareness could be raised among health practitioners and current clinical guidelines for amputee rehabilitation could be improved. Ultimately, the findings of this thesis aim to inform current amputee rehabilitation and physiotherapy treatment in the UK.

Thesis structure

A review of the relevant literature is undertaken in Chapter 2. This chapter begins with a review of current standards and guidelines in amputee rehabilitation. The following areas focus on locomotor function in transtibial amputees; the biomechanics of activities of daily living; falls and balance confidence. This chapter concludes with the specific aims and hypotheses of this study.

Chapter 3 presents the general methods that form this thesis. The ethical review process, participants and inclusion/exclusion criteria are explained. The general procedures that were followed for the biomechanical analyses are described and the psychological instruments are evaluated.

Chapter 4 forms the first of six studies in this thesis. An audit was conducted with lead physiotherapists in Disablement Service Centres. This study investigates how falls are monitored by physiotherapists and the use of functional and psychological outcome measures in amputee rehabilitation in England. The findings identify shortcomings in amputee rehabilitation that have important clinical implications and will serve to make recommendations in Chapter 10.

Chapter 5 is the first of four biomechanical studies in this thesis. In this chapter, a biomechanical analysis of level walking is undertaken and comparisons are made between the fallers and non-fallers. The chapter closes by discussing the biomechanical differences that were observed and how this could impact on future treatment.

Chapters 6 and 7 examine the biomechanics of stair walking during ascent and descent, respectively. The results highlight the kinematic and kinetic differences between the fallers and non-fallers. The chapter also discusses alternate stair walking patterns observed in some of the participants.

In order to understand how fallers and non-fallers maintain postural control under static and dynamic conditions, Chapter 8 explores the novel use of computerised dynamic posturography with the Neurocom Equitest in an amputee population.

Chapter 9 focuses on two psychological instruments that measure balance confidence and quality-of-life. Comparisons between the fallers and non-fallers are discussed. The relationships between functional performance tests and the two instruments are also explored.

Finally, Chapter 10 provides a summary of this thesis and limitations are explored. Recommendations for future studies are made based on the biomechanical and psychological findings from the studies. Finally, this chapter closes with suggestions for improving clinical practice that will be disseminated among amputee physiotherapists. It is anticipated the clinical recommendations will raise awareness on the biomechanical differences between fallers and non-fallers and that this will influence rehabilitation and physiotherapy treatment positively.

CHAPTER TWO - LITERATURE REVIEW

2.1. Introduction

This literature review will focus on areas relating to locomotor function in transtibial amputees. It consists of five main areas. The first part will explore the causes and characteristics of lower-limb amputation in the UK. The second part reviews falls recommendations in the clinical guidelines for amputee rehabilitation and the use of outcome measures as part of current practice. The third part explains human gait while the forth part addresses the biomechanics of activities of daily living. Three activities (level walking, stair walking and balance control) have been selected for analysis because they represent typical activities of daily living of increasing difficulty and have been linked with falls. This section will contrast typical patterns observed in able-bodied persons with transtibial amputees and will discuss similarities between the gait of transtibial amputees and older individuals. The final part will focus on falls incidence and causes in older, able-bodied people and transtibial amputees. The relationship between falling and fear of falling are also explored.

2.2. Lower-limb amputation: Causes and characteristics

An amputation is the surgical removal of a body extremity. Lower-limb amputations account for the majority of amputations in the UK every year (NASDAB, 2005). During a 12-month period in 2005/6, 4567 new referrals were made to prosthetic centres in the UK, and 91% were following a lower-limb amputation (NASDAB, 2005). The major causes for a lower-limb amputation were dysvascularity (75%), trauma (9%) and infection (7%). Other causes were related to neurological disorders or neoplasia. The lower-limb amputations related to vascular disease could be categorised into diabetes mellitus (42%), non-diabetic arteriosclerosis (29%), and other dysvascularity (29%). Lower-limb amputation can occur at various levels. In 2005/6, below the knee, (transtibial), accounted for 53% of all lower-limb amputations; above the knee, (transfemoral), accounted for 38%. Bilateral amputations were less common and only accounted for 5%; other lower-limb amputations (such as through the knee, ankle or forefoot) accounted for the remaining 4% (NASDAB, 2005/6).

The majority of individuals who experience a lower-limb amputation are over 65 years (54%) (Figure 2.1). There are also clear gender differences, with more males experiencing a lower-limb amputation than females across all ages. However, the number of females over the age of 75 years almost doubles compared to females in the younger age category (Figure 2.2).

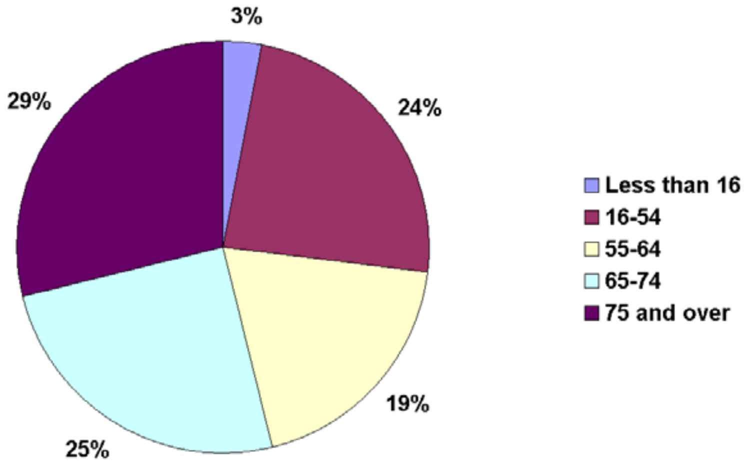


Figure 2.1. New referrals to prosthetic limb centres in the UK according to age (Adapted from NASDAB, 2005/6)

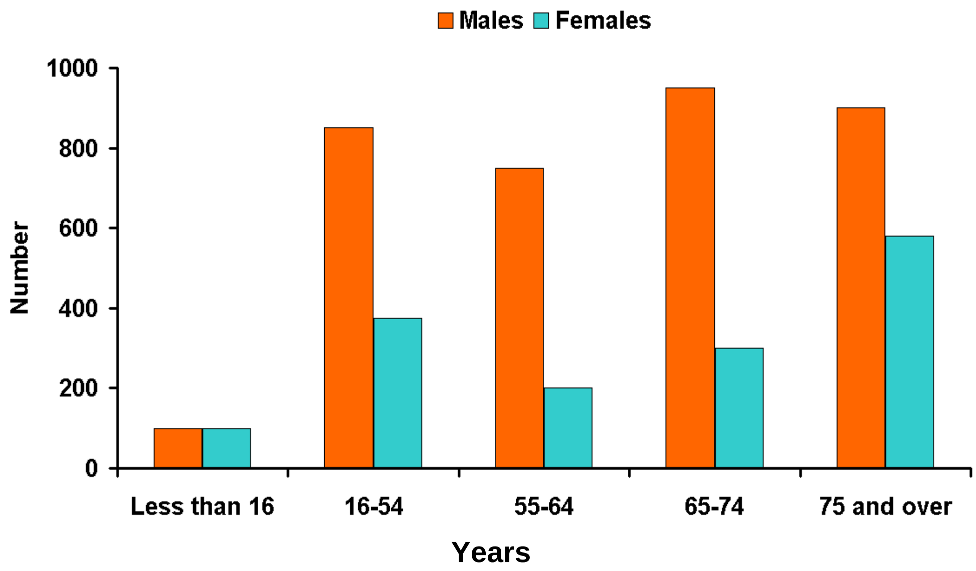


Figure 2.2. Number of new referrals to prosthetic limb centres in the UK according to age and gender (Adapted from NASDAB, 2005/6)

2.3. Current practice in amputee rehabilitation

Clinical standards for rehabilitation in the UK recommend the use of standardised outcome measures as tools to monitor the effect of rehabilitation practice (Turner-Stokes, 2001; Skinner and Turner-Stokes, 2006). Outcome measures have also been used to determine functional independence and assist in discharge planning (Bischoff *et al.*, 2003) and a potential predictor for falling (Large *et al.*, 2006). Patient focused outcome measures can help amputees achieve their desired levels of function by monitoring their performance and progress, but can also help physiotherapists to compare the efficacy of different treatment protocols.

Returning patients to a level of safe, functional mobility is one of the primary goals of the clinical guidelines for the physiotherapy management of lower-limb amputees (BACPAR, 2003). As with the causes leading to the amputation, treatment strategies are diverse and complex. In 2003, BACPAR released evidence based clinical guidelines to promote best practice in amputee rehabilitation. The following section entitled 'Coping Strategies Following Falls' was the only part of the clinical guidelines that specifically addressed falls in amputees:

"5.5.1 All parties involved with the patient should be made aware that the risk of falling is increased following lower-limb amputation.

5.5.2 Rehabilitation programmes should include education on preventing falls and coping strategies should a fall occur.

5.5.3 Instructions should be given on how to get up from the floor.

5.5.4 Advice should be given in the event that the patient is unable to rise from the floor." (BACPAR, 2003, p.32)

The main focus of the 2003 guidelines was how patient assessment should be conducted; how the aims of the prosthetic rehabilitation programme should be discussed; and provided information about patient discharge and maintenance (BACPAR, 2003). These guidelines advocated the use of subjective and objective examination and outcome measures to evaluate and record change. However, there was no information on which outcome measures were most suitable with lower-limb amputees or could quantify a patient's improvement. Furthermore, the guidelines did not provide any recommendations on which targeted exercises were important for new amputees regaining their functional mobility with a prosthesis.

In 2006, BACPAR published revised guidelines about pre- and post-operative physiotherapy management. This document defined the role of the physiotherapist in the multidisciplinary team; explained the knowledge a physiotherapist should have in understanding and influencing the outcomes of rehabilitation; described the information a physiotherapist should give to the patient and/or their carer; and defined the treatment physiotherapists should provide pre- and post-operatively. Despite numerous articles on falls, as well as recent standards set by the NHS and Department of Health, the section on falls was not updated in the updated guidelines.

The revised guidelines made one reference to exercise programmes in an amputee population, this recommendation was based on the findings from only one study by Seroussi *et al*. (1996) that investigated level gait in 8 transfemoral amputees. These authors concluded that the hip extensors and flexors and ankle plantarflexors would benefit from strengthening. As BACPAR recommended exercises based on the findings from this study only, it did not mention the importance of the knee musculature which plays a crucial role in safe level and stair walking in transtibial amputees. Therefore, a systematic review of the biomechanics literature related to amputees should be undertaken before BACPAR update their evidence-based clinical recommendations on falls and exercise.

2.4. Human gait

Human walking is the most common form of locomotion and one of the most familiar daily activities. Healthy, able-bodied individuals exhibit sterotypical “normal” patterns of motion at the ankle, knee and hip. However, transtibial amputees have had their ankle joint and musculature replaced with prosthetic components. The power-generating plantarflexor muscles and the anatomical ankle complex, that normally play an important role during the propulsion phase of walking, have been lost. While the prosthesis may provide the required structural and mechanical support to engage in bipedal locomotion, amputees inevitably have to make compensatory adjustments to their gait pattern. In order to understand the gait of transtibial amputees, it is important to understand the basic functional phases of “normal” gait.

2.4.1. Phases of the gait cycle

The gait cycle is defined as the time between two successive gait occurrences and typically begins with foot contact (termed initial contact) (Whittle, 2007). Broadly, the gait cycle (one stride) can be divided into the stance and swing phases. Stance makes up approximately 60% and swing 40% of the gait cycle (Perry, 1992). Each stance phase is made up of two double support phases (each accounting for 10% of the gait cycle) and a single support phase (40% of the gait cycle). The stance phase starts with initial contact and can be subdivided into smaller sub-phases: weight acceptance (or loading response), mid-stance, terminal stance and pre-swing. The swing phase starts with toe off and is sub-divided into initial, mid and terminal swing. Although the gait terminology for the sub-phases may vary slightly according to publications, the above terms will be used consistently in this thesis.

2.4.2. Functional tasks of the gait cycle

Functionally, the gait cycle can be divided into three primary tasks with several sub-phases that describe the patterns of motion of the lower-limb joint angles during the gait cycle.

<u>Task 1: Weight acceptance and the loading response</u>

The first task is weight acceptance and begins with initial contact. Following initial contact and the commencement of the initial double support phase, body weight is transferred onto the supporting limb. The contralateral limb is in the pre-swing phase. In the sagittal plane, the ankle begins to plantarflex during controlled lowering of the foot to the floor and the knee begins to flex to approximately 15-20° for shock absorption during the loading response. During this time, the hip angle remains relatively constant with flexion angles of approximately 30°. At initial contact, the pelvis is neutral in the sagittal and frontal planes, and rotated internally approximately 5°. When the limb is loaded, the pelvis begins to display a small anterior pelvic tilt of approximately 4°. In the frontal plane, the hip is adducted and is accompanied by a rapid contralateral pelvic drop of approximately 4°, which is decelerated by the hip abductors. In the transverse plane, the pelvis begins to rotate internally to approximately 10°.

As the foot makes contact with the ground, there is a rapid rise in vertical ground reaction force (GRF) (approaching 100% body weight). The posterior shear reaction (braking) force reaches a peak during the loading response and is equivalent to approximately 13% body weight. There is also a peak lateral reaction force that occurs in mid-loading response that is small (5% body weight) and highly variable.

When initial contact is made with the heel and the knee is almost fully extended, then the GRF vector (GRFv) is located behind the ankle joint and in front of the hip and knee joints. This would cause an initial ankle dorsiflexor, knee flexor and hip extensor moment. The resultant power absorption at the ankle joint is small and not labelled during the loading response. As the GRFv progresses anterior to the ankle, the internal ankle moment becomes plantarflexor in direction, the knee and hip moment become progressively larger (both approaching 1 Nm/kg) and extensor in direction during the loading response. The knee flexes under the control of the knee extensor muscles and results in a power absorption phase (labelled K1 – power absorption at the knee in loading) (Winter, 1987). While hip powers are considered variable with walking velocity and between strides, the hip extensors contract concentrically during the loading response to produce a small, positive power generation phase (labelled H1 – power generation at the hip in loading) (Figure 2.3).

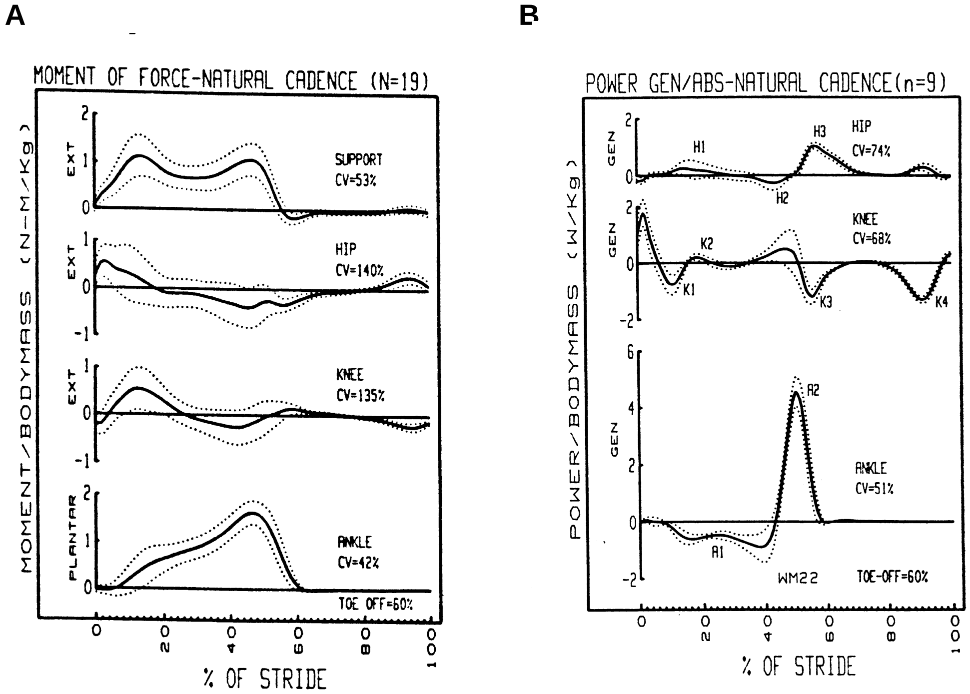


Figure 2.3. (A) Average support moment and joint moments for the hip, knee and ankle during level gait. (B) Average joint powers for the hip, knee and ankle. The gait cycle is initiated with foot contact. (from Winter, 1987).

Task 2: Single limb support

The second task is referred to as single limb support and begins when the contralateral limb is lifted into the air. In this phase, one limb supports the entire body mass during mid and terminal stance and ensures safe progression of the swinging limb. The stance limb progresses forwards over the supporting foot, causing increased ankle dorsiflexion, knee extension and hip extension.

Mid-stance occurs between 10-30% of the gait cycle when the ankle moves into dorsiflexion as the shank advances over the supporting foot and the knee and hip joints both extend. Terminal stance, from 30-50% of the gait cycle, completes the single support phase. It begins with heel rise and terminates when the contralateral limb makes foot contact with ground, signalling the start of the second double support phase. Throughout terminal stance, the body weight vector is moving anteriorly. With initiation of heel rise, there is deceleration of ankle dorsiflexion and the hip begins to extend, putting the support limb into the trail limb position, and the pelvis in anterior pelvic tilt.

As the GRFv passes in front of the ankle, the lever arm of the body vector moves under the metatarsal head, placing demands on the plantarflexor muscles to stabilise the ankle joint. This causes a peak ankle plantarflexor moment of approximately 1.5 – 2 Nm/kg in terminal stance. During mid- to terminal stance, the GRFv passes from behind, through the knee joint and then in front of the knee, resulting in a knee moment approaching zero then becoming flexor in orientation. There is a positive burst of power generation from the knee extensors (labelled K2 –power generation at the knee in mid-stance) representing 10-15% of power generation during level walking (Winter, 1987). At the hip joint, the vector is positioned through the joint and then passes close behind therefore, the hip moment remains small and flexor in orientation throughout mid- and terminal stance. As the thigh begins to rotate backwards, the hip flexors contract eccentrically, (labelled H2 – power absorption at the hip in stance. In the frontal plane from mid- to terminal stance, the thigh moves from its neutral position in mid-stance to passive abduction in terminal stance (Figure 2.3).

At a self-selected walking velocity, the vertical GRF reaches its lowest force of approximately 0.6-0.8 N/kg during mid-stance (Winter, 1987). The anterior-posterior GRF changes from posterior to anterior in direction, and the shear horizontal force becomes medial in direction throughout the remainder of the stance phase.

Task 3: Limb advancement

During the final task of the gait cycle, the trail (supporting) limb must prepare the body for swing limb advancement and adequate foot clearance during the swing phase. It begins with the pre-swing phase (50-60% of the gait cycle), which also corresponds to the second double support phase. This is the interval when the contralateral limb makes ground contact until the ipsilateral limb achieves foot off and is no longer in contact with the ground. The main role of the double support phase is to transfer body weight from the trail, supporting limb onto the lead, swinging limb.

During the pre-swing phase, the complex movements of the ankle joint and the ankle plantarflexors play an important role in the initiation of swing. The ankle joint rapidly plantarflexes to approximately 20° and the ipsilateral limb is rapidly unloaded. The plantarflexor muscles are generating power and contracting concentrically (labelled A2 – power generation at the ankle in pre-swing). This muscle action accelerates the advancement of the unloaded limb and causes the knee to begin flexing rapidly to initiate swing. During this phase, the knee begins to flex passively up to 40° preparing the limb for adequate foot clearance during swing. With the GRFv behind the knee joint, there is a small knee extensor moment. This is associated with the K3 power burst (power absorption at the knee in pre-swing), when the knee extensors absorb power. Knee flexion, rather than ankle dorsiflexion, plays an important role for ensuring adequate floor clearance during initial swing. In pre-swing, the hip reverses its motion from extension to begin flexion. With the GRFv behind the hip joint, there is a small hip flexor moment that is evident prior to foot off. The hip flexors generate a positive power burst, also called 'pull-off' burst (labelled H3 – power generation at the hip in pre-swing), which increases when the inertial load of the swinging limb increases (Winter, 1987). When body weight is transferred onto the contralateral limb, the hip adductor muscles decelerate the passive abduction of the thigh at the hip joint, which is accompanied by a pelvic drop of approximately 5°. There is a continuous hip abductor moment observed throughout the stance phase (Figure 2.3).

During the initial swing phase (60-73% of the gait cycle), the foot is lifted off the ground because of increasing knee and hip flexion. The ankle is moving from a plantarflexed position to dorsiflexion. This does not obstruct foot clearance because the limb begins swing from a trailing position. When the swing foot is parallel to the contralateral, supporting foot, it is just slightly plantarflexed. With negligible inertial properties of the foot and in the absence of the GRFv, there is reduced ankle moment and/or ankle joint power. Knee flexion continues to play an important role for foot clearance. With the toes of the foot pointing downwards, the swinging limb is functionally longer than the standing, support limb. Therefore, peak knee flexion of 60° in swing is essential for ensuring the swinging limb has sufficient ground clearance. This is particularly important for obstacle and trips avoidance. Pre-swing knee flexion of 40°, momentum of the swinging limb through rapid hip flexion and active knee flexion all contribute to achieving sufficient knee flexion (Perry, 1992).

The mid-swing phase (73-87% of the gait cycle) occurs when the swinging limb is parallel to the support limb until it is forward and the shank is vertical. In this phase, the primary objective is for limb advancement, whilst ensuring foot clearance as the contralateral limb is in single support. The joint motions that occur at the knee and hip are passive in nature. The knee begins to extend because of knee flexor muscle relaxation. The hip continues passive hip flexion thanks to momentum of the swinging limb. During the initial and mid-swing phases, the pelvis regains its neutral position.

The final sub-phase of the gait cycle and the swing phase is terminal swing. This starts when the tibia is vertical until the swinging limb is preparing for subsequent initial foot contact (87-100% of the gait cycle). The ankle angle approaches neutral to ensure safe foot placement. Limb advancement is achieved with knee extension, while hip flexion remains relatively constant during this phase. During terminal swing, the knee flexors absorb most of the power from the swinging limb resulting in a negative power burst (labelled K4 – power absorption at the knee in terminal swing). In the frontal plane, the thigh remains abducted at the hip joint throughout the majority of swing, but returns to a neutral position in preparation for the subsequent weight-bearing, stance phase. In terminal swing, the pelvis follows the swinging limb, causing it to rotate maximally forward (internally) of approximately 5°.

2.4.3. Temporal-spatial parameters

Stride characteristics represent an individual's basic walking capability. Velocity is the basic gait measurement and is the product of step length and step frequency (cadence). Walking velocity can be attained by different combinations of these two variables. However, as velocity increases, the biomechanical constraints of the musculoskeletal system limit further increases in step length and/or frequency during walking (Donker and Beek, 2002). Individuals have a preferred, spontaneous, self-selected walking speed, which also corresponds to their most metabolically efficient velocity. Average self-selected walking velocity on a level surface is 82 m/min or 1.37 m/s in adults (Perry, 1992).

2.5. Biomechanics of human motion

2.5.1. Level gait in transtibial amputees

Given their inherent lower-limb asymmetry, it has been well documented in the literature that transtibial amputees adopt an asymmetrical walking pattern (Donker and Beek, 2002; Isakov *et al.*, 1996; Jaegers *et al.*, 1995; Nolan *et al.*, 2003; Sanderson and Martin, 1997; Winter and Sienko, 1988). Gait asymmetry is often regarded as an indicator of pathological gait (Sadeghi *et al.*, 2000). For example, transtibial amputees chose to walk at a slower self-selected walking velocity compared to age-matched, able-bodied individuals (Isakov *et al.*, 1996; Jaegers *et al.*, 1995); spend less time in stance to reduce loading on the prosthesis and the affected limb (Donker and Beek, 2002; Nolan *et al.*, 2003); and take longer steps with their affected limb (Nolan *et al.*, 2003; Sanderson and Martin, 1997).

Compared to able-bodied individuals, transtibial amputees show obvious kinematic differences at the ankle joint. The prosthetic ankle does not actively plantarflex and does not achieve as much ankle dorsiflexion during mid- to terminal stance (Sanderson and Martin, 1997). Kinematic differences have also been observed at the ankle of the intact limb. Although this limb had not lost the plantarflexor musculature to generate adequate push-off power, it still displayed less ankle plantarflexion than the ankle of able-bodied individuals (Sanderson and Martin, 1997). Other notable differences have included reduced knee flexion during the loading response on the affected limb, whilst still displaying peak knee flexion of approximately 60° during mid-swing (Powers *et al.*, 1998; Sanderson and Martin, 1997). At the hip joint on the affected side, there may be less flexion in stance but increased flexion during mid- to terminal swing. This was considered a foot clearance strategy in the absence of active dorsiflexion at the prosthetic ankle/foot (Sanderson and Martin, 1997). While other gait differences, particularly at the hip and pelvis could have been identified, most authors (Powers *et al.*, 1998; Sanderson and Martin; 1997) have limited their kinematic analysis to the sagittal plane only.

Dynamic control from the lost musculature is difficult to replace. Therefore, analysis of GRFs, joint moment and power profiles provide insight into the level of internal adjustment and the active muscle groups during the different phases of the gait cycle. Previous studies of level gait with transtibial amputees have reported significantly smaller braking, propulsive and peak vertical GRFs under the affected leg compared to the intact leg and control subjects (Nolan *et al.*, 2003; Sanderson and Martin, 1997). Other studies have found the intact limb plays an important compensatory role and experiences higher a/p and vertical GRFs compared to the affected limb and control subjects (Beyaert *et al.*, 2008). Significantly smaller internal joint moments, specifically at the ankle and knee joints of the affected limb, have been reported, irrespective of walking velocity (Beyaert *et al.*, 2008; Sanderson and Martin, 1997; Winter and Sienko, 1988). Knee moments of almost zero substantially reduce the demands on the knee extensor muscles, which are typically weaker in transtibial amputees and are needed to keep the knee from collapsing (Isakov *et al.*, 1992; Sanderson and Martin, 1997). Power absorption by the knee extensors during the loading response (K1) and power generation in mid-stance (K2) are both typically absent on the affected limb in transtibial amputees (Winter and Sienko, 1988). Thus maintaining the thigh and shank in a more extended position, through the absence of knee flexion to reduce loading on the residual limb, is a gait characteristic commonly reported in transtibial gait.

More subtle differences were found in hip extensor moments and especially at the faster walking velocity. As a result of the vertical orientation of the thigh, the GRFv was located closer to the hip joint of the affected limb, resulting in reduced peak hip extensor moment during the loading response. Typically power generation by the hip extensors (H1) remains quite strong and serves as a compensatory strategy for the lack of power absorption by the ankle plantarflexors at this phase in the gait cycle (Winter and Sienko, 1988). Power generation by the hip flexors during push-off (H3) is an important mechanism by which individuals ensure adequate foot clearance in absence of sufficient ankle plantarflexion. Typically, the H3 power burst by the hip flexors is quite important in amputees.

2.5.2. Level gait in older adults

The previous section has shown that transtibial amputees have reduced joint range of motion and muscle strength in the lower-limbs. Similar observations have been reported in other studies investigating age-associated changes to the biomechanics of walking in older adults (Barak *et al.*, 2006; Kerrigan *et al.* 1998 Kerrigan *et al.*, 2000; Kerrigan *et al.*, 2001; Lee and Kerrigan, 1999). These studies have consistently reported that comfortable walking speed decreases with age. Comparing the gait of older persons with transtibial amputees may present a novel framework for understanding gait adaptations these individuals make as a result of altered musculoskeletal function.

Joint kinematic analysis has shown that the elderly groups exhibited less peak ankle plantarflexion and peak hip extension, but more anterior pelvic tilt compared to young individuals irrespective of walking speed (Kerrigan *et al.*, 1998). Furthermore, hip extension was the only joint parameter that was significantly reduced in elderly vs. young and elderly fallers vs. elderly non-fallers (Kerrigan *et al.*, 2001). When kinetic parameters were examined, ankle power generation in pre-swing (A2) was significantly reduced in the elderly compared to the young adults (Kerrigan *et al.*, 1998) and this was linked with weakened concentric ankle plantarflexor contraction. Other findings, such as reduced knee and hip extensor moments and powers, were age-associated biomechanical reductions as a consequence of slower walking speed (Kerrigan *et al.*, 1998). One study reported increased internal ankle plantarflexor, knee flexor and hip extensor moments in elderly fallers compared to elderly non-fallers walking at comfortable speeds (Lee and Kerrigan, 1999). Although conflicting with other studies (Wolfson *et al.*, 1995), Lee and Kerrigan (1999) suggested that strength reductions were not the major cause of falls during level walking. Instead they suggested rehabilitation programmes should focus on improving dynamic stability.

Using gait analysis, the biomechanics of “normal” and amputee walking have been well-documented and consistent differences have been noted between the affected and intact limbs. Some studies have also observed gait differences linked to age and falls’ history during walking. Despite similarities of physical characteristics between the elderly and transtibial amputees, it is unclear whether differences between elderly fallers and non-fallers could be extended to amputee individuals. Furthermore, while these differences are important and can make recommendations for reducing falls, it is likely that more physically demanding tasks, such as stair negotiation, would highlight other variables that could limit locomotor function. The next section will describe the biomechanics of stair walking in able-bodied individuals and transtibial amputees.

2.5.3. Stair walking

Stair walking is a challenging locomotor task that has been linked with falls, especially in an ageing population. Since the publication of McFadyen and Winter's (1988) article of an integrated biomechanical analysis of normal stair ascent and descent, more of the current research has been aimed at understanding the biomechanics of stair locomotion in those with altered locomotor function, such as the elderly and lower-limb amputees (Schmalz *et al.*, 2007; Powers and Boyd, 1997; Yack *et al.*, 1999). These analyses raise awareness of the demands placed on the musculoskeletal system and make recommendations for safer solutions to environmental factors such as visual factors and stair design for falls prevention (Beaulieu *et al.*, 2008; Cavanagh *et al.*, 1997; Hamel *et al.*, 2005; Nadeau *et al.*, 2003; Reeves *et al.*, 2008b; Simoneau *et al.*, 1991).

The aim of stair walking is to move the head arms and trunk (HAT segment) safely in both the vertical and horizontal directions. The greater support moments measured during stair walking vs. level walking indicate that it is a more challenging activity (McFadyen and Winter, 1988). Kinetic energy is converted into gravitational potential energy during the 'up' phase, while potential energy is transferred into kinetic energy during the controlled lowering motion in the 'down' phase. Therefore, stair ascent is characterised by concentric muscle activity, while stair descent is accomplished through eccentric activity.

<u>Stair ascent</u>

McFadyen and Winter (1988) described several sub-phases for the stance and swing portions of the stride cycle. During stair ascent, the gait cycle starts with initial foot contact (typically with the mid- to forefoot region) in what is termed the *weight acceptance phase*, when the ankle positions the body for the next phase (labelled A1 – power absorption at the ankle in loading). Following contralateral toe off, the *pull-up phase* is when the body's postural stability is most vulnerable as all three joint angles are flexed. At this point, the ankle, knee and hip extensor muscles need to generate sufficient muscle power to prevent the lower-limbs from collapsing (labelled A2 – power generation at the ankle in mid-stance, K1 – power generation at the knee in loading, H1 – power generation at the hip in pre-swing). Once the contralateral leg is in mid-swing, the body has been elevated one step and is continuing to the next step in the *forward continuance phase*. The ankle contributes to lift with the largest power generation burst (labelled A3 – power generation at the ankle in pre-swing) observed during the gait cycle. Swing is initiated with toe off, when the primary goal is to maintain adequate *foot* *clearance* when the foot is crossing the intermediate step. Power absorption by the hip flexors (referred to as H2) is practically not present during stair climbing (Nadeau *et al.*, 2003) and therefore was not labelled by McFadyen and Winter (1988). Foot clearance is characterised by power generation of the hip flexors (labelled H3 – power generation at the hip in initial swing) and knee flexor power generation (labelled K2 – power generation at the knee in pre-swing/initial swing). Once the swing leg has been lifted and into mid-

swing, the knee extensors contract eccentrically to limit knee flexion (labelled K3 – power absorption at the knee in initial swing) followed by power absorption of the knee flexors just prior to the subsequent foot contact (labelled K4 – power absorption at the knee in terminal swing). The cycle is completed with ipsilateral *foot placement* (Figure 2.4)

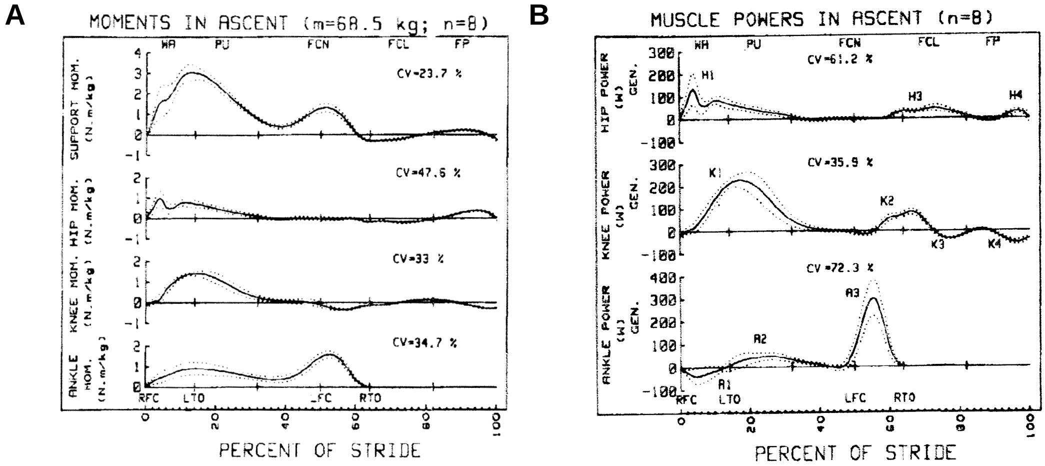


Figure 2.4. (A) Average support moment and joint moments for the hip, knee and ankle during stair ascent. (B) Average joint powers for the hip, knee and ankle. The gait cycle is initiated with foot contact. (from McFadyen and Winter, 1988).

<u>Stair descent</u>

The stance phase begins with *weight acceptance*. Similarly to stair ascent, initial contact is typically made with the mid- to forefoot with the ankle maximally plantarflexed and large power absorption at the ankle (labelled A1 – power absorption at the ankle in loading) and knee (labelled K1 – power absorption at the knee in loading). However, contrary to ascent, the body's COM is usually more centrally placed within the base of support, placing the body in a more stable position in stance in stair descent compared to ascent. When the contralateral leg toes off and single stance is initiated, the body has already been lowered one step in the *forward continuance phase*. This is followed by knee extension as seen by the knee generation power burst (labelled K2 – power generation at the knee in mid-stance). The body continues its downward path during the *controlled lowering phase* characterised by the power absorption at the knee (labelled K3 – power absorption at the knee in pre-swing) and the ankle (labelled A2 – power absorption at the knee in mid-stance). The hip flexors generate power during pre-swing when they assist in pulling the leg off the top step (labelled H1). The ankle plantarflexors contract concentrically to produce the A3 (power generation at the ankle in pre-swing) positive power burst which McFadyen and Winter (1988) believed to help control the excessive dorsiflexion observed just before toe off. When the ipsilateral leg initiates swing, the hip flexors produced a positive power burst (labelled H2 – power generation at the hip in initial swing) to pull the leg through. After toe off, knee flexion decreases in stair descent (labelled K4 – power absorption at the knee in mid-swing) because ensuring foot clearance is less critical compared to ascent. From mid-swing to foot contact, the lower-limb joints begin to extend in preparation for the next weight acceptance (Figure 2.5).

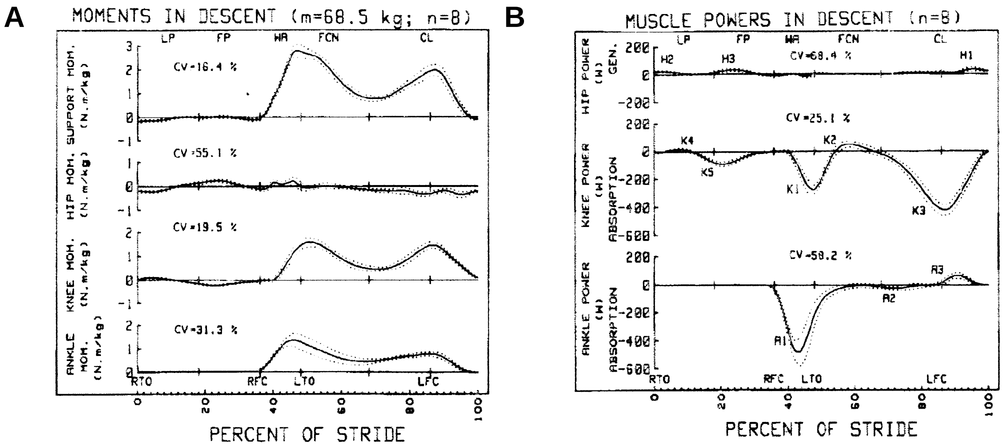


Figure 2.5. (A) Average support moment and joint moments for the hip, knee and ankle during stair descent. (B) Average joint powers for the hip, knee and ankle. The gait cycle is initiated with toe off. (from McFadyen and Winter, 1988).

2.5.4. Stair walking in transtibial amputees

It is accepted that stair walking is a more complex mechanical task compared to level walking (Beaulieu *et al.*, 2008; McFadyen and Winter, 1988; Reeves *et al.*, 2008a; Reeves *et al.*, 2008b). Both stair ascent and descent rely on muscle power particularly from the ankle plantarflexors and knee extensors. In transtibial amputees, the ankle plantarflexors are absent and the knee musculature on the affected side, particularly the extensors, is weakened (Sanderson and Martin, 1997). Stair walking places greater functional demands on the prosthesis (Schmalz *et al.*, 2007) and moves the lower-limb joints through a greater ROM than level walking (Lin *et al.*, 2005). The prosthesis may provide structural support but does not allow active plantarflexion. In amputees, the amount of energy generated at the ankle can only be equivalent to the amount of energy stored (Yack *et al.*, 1999).

During the pull-up phase, the contribution of the ankle joint and the soleus muscle is high (McFadyen and Winter, 1988). On the amputated side, this work must be divided between the knee and hip musculature with some help from the intact limb during the double support phase (Yack *et al.*, 1999). Yack *et al.* (1999) reported greater reliance on the hip extensor musculature during the weight acceptance and pull-up phases on the amputated side. They emphasised the importance of hip muscle strengthening strategies to facilitate a reciprocal stair pattern in amputees (Yack *et al.*, 1999).

There are several studies that have previously investigated the biomechanics of stair walking patterns in transtibial amputees, during stair ascent and descent over several steps (Powers and Boyd, 1997; Schmalz *et al.*, 2007; Yack *et al.*, 1999). However, few have conducted both kinematic and kinetic analyses in this population. These studies reported that transtibial amputees walked slower than control subjects. Kinematics of the amputated side revealed reduced ankle joint motion. At the knee joint, results were more confounding. Powers and Boyd (1997) reported similar knee flexion during stair ascent in amputees and controls but reduced flexion angles during descent in the amputee group; while Schmalz *et al.* (2007) reported reduced and delayed knee flexion in stance during descent. Such inconsistencies may reflect the different dimensions of the staircases used in these studies. Powers and Boyd (1997) used a 4-step staircase with a step height of 15 cm and a tread depth of 27 cm whereas Schmalz *et al.*, (2007) did not provide the step dimensions. Other kinematic data showed increased hip flexion (Powers and Boyd, 1997; Schmalz *et al.*, 2007) and anterior pelvic tilt (Powers and Boyd, 1997) during both ascent and descent when compared to controls.

The above-mentioned studies did not measure frontal plane motion, particularly hip ab/adduction and pelvic obliquity (Powers and Boyd, 1997; Schmalz *et al.*, 2007; Yack *et al.*, 1999). These movements could show compensatory mechanisms in the absence of adequate ankle motion and limited knee motion due to prosthetic component limitations. Lee and Chou (2007) reported exaggerated movements of the trunk in the frontal plane, especially during stair descent, in older adults. This was believed to indicate a greater challenge to maintaining balance (Lee and Chou 2007). Therefore, sagittal and frontal plane kinematics of the lower-limbs, pelvis and trunk are important for understanding balance strategies for those with balance deficits.

Kinetic results have shown that transtibial amputees have reduced peak vertical and decelerating GRFs on the affected leg but higher vertical GRFs on the intact leg during both stair ascent and descent (Schmalz *et al.*, 2007). Joint moments revealed a reduced knee flexor moment on the amputated side keeping the knee in a more extended position (Schmalz *et al.* 2007). Using a 3-step staircase, Yack *et al.* (1999) reported significantly reduced peak ankle moments on the amputated side during stair ascent. They also found a significantly smaller peak ankle power on the intact side compared to the control group. However, there were no significant differences in knee and hip joint moments and powers on the intact side vs. control subjects, although the intact side was significantly greater than the amputated side. Yack *et al.* (1999) concluded that the transtibial amputees employed a knee-extensor dominant strategy on the intact side and a hip-extensor dominant strategy on the amputated side, such that the hip extensors and the intact limb were the main source of power generation during stair ascent. As there were no differences in walking speed between the groups, the amputee individuals (n=5; age range 27-36 years; mean time since amputation 13.1 years – range 1.5 – 20 years) were closely matched to the controls on age (n=5; age range 24-36 years) and fitness level (Yack *et al.*, 1999). No amputees used the staircase handrails. Therefore, caution must be used when generalising the stair biomechanics patterns to the general amputee population including inexperienced amputees and older amputees, who would normally rely on handrails for support.

The research on stair negotiation in transtibial amputees has been primarily focused on improving prosthetic components and design. Some studies have investigated foot clearance (Hamel *et al*., 2005) and environmental factors (stair design, lighting factors) (Cavanagh *et al*., 1997; Hamel *et al*., 2005; Simoneau *et al*., 1991) on the biomechanics of stair walking in able-bodied individuals. However, these analyses have not measured the full 3D kinematic and kinetic parameters making it difficult to compare their findings. Moreover, no published research to date has specifically investigated the biomechanics of stair walking in amputee fallers and non-fallers despite the common occurrence of falls during stair walking (Talbot *et al.*, 2005).

2.5.5. Postural control

Falls are likely to result from inadequate postural control in both able-bodied individuals and transtibial amputees. Therefore, understanding postural strategies is important when undertaking a comprehensive analysis about falls in these populations. During quiet standing and walking, a person maintains their balance by controlling the body's COG over the base of support. Postural control necessitates the ability to accurately predict and detect balance perturbations and execute appropriate responses. A perturbation may be described as an external disturbance to the body's stability. In response, the CNS locates the body's COG and adjusts its postural responses by centrally pre-programming postural strategies or plans for action based on available information (Horak *et al.*, 1989). This information will reflect incoming somatosensory and visual input, but will also draw upon prior experience and environmental cues. The resultant 'motor synergy' reflects the motor outcome of muscle groups that function together to correct or re-adjust the COG movement (Horak *et al.*, 1989). Postural control will be compromised if some aspect of the system is disrupted. The inability to realise how far the body's COG has been displaced, or failure to produce the appropriate response due to muscle weakness or insufficient flexibility, will likely result in a fall (Judge *et al.*, 1995).

Depending on the size of the perturbation, three postural strategies can be used to reposition the COG during standing balance. The *ankle strategy* is typically used in response to slow or small perturbations and relies on distal to proximal sequential muscle activation of the ankle, thigh and trunk musculature. Postural dyscoordination, such as delayed muscle activation, may result in destabilising the knee and hip; whereas simultaneous co-activation may stiffen the joints (Horak *et al.*, 1989). The *hip strategy* is used for larger, faster perturbations and is initiated through the recruitment of proximal muscles about the hip joint. The *stepping strategy* repositions the base of support under the COG with rapid hops, steps or stumbles (Horak *et al.*, 1989). Individuals will select the most appropriate response based on prior experience (Horak *et al.*, 1989) and may show a mixture of strategies as part of the successful adaptation to a perturbation (Horak and Nashner, 1986).

In healthy individuals, postural responses were typically recorded 70-110 msec after an unexpected perturbation (Horak and Nashner, 1986). Lord *et al.* (1991) estimated the contribution of sensory input from the foot/ankle complex was 58%, visual input was 22% and vestibular input was 20% towards postural control. Increased latencies or the inability to produce a postural strategy will lead to a fall. Sufficient ankle muscle strength and flexibility are needed to execute the ankle strategy (Horak *et al.*, 1989). The ability of individuals with altered locomotor function (such as a result of ageing, trauma or disease) to execute a successful response will be adversely affected. Mackey and Robinovitch (2006) found that the older women who had previously fallen showed a decreased ability to recover their balance (reflected by decreased speed and strength of response) using an ankle strategy compared to young healthy women.

The loss of muscle and receptors from the amputated limb and the resultant lower-limb asymmetry suggests that transtibial amputees must have modified internal representations within the CNS compared to able-bodied people (Quai *et al.*, 2005). As previously shown in the review of the biomechanics of gait in the elderly and transtibial amputees, reduced musculoskeletal function at the ankle joint would have detrimental effects on gait and when maintaining dynamic postural control. Such biomechanical deficits may place greater reliance on the hip and/or stepping strategies to maintain the COG within the BOS.

Isakov *et al.* (1992) reported that transtibial amputees showed greater sway than control subjects when they were asked to stand with their eyes open or eyes closed during their rehabilitation. Similarly, Hermodsson *et al.* (1994) found that traumatic and dysvascular transtibial amputees had significantly more lateral sway during a two-legged standing test than control subjects and that vascular amputees had significantly more sway than traumatic amputees with and without visual input. Increased lateral sway has been reported in elderly persons as a possible sign of decreasing balance capacity (Baloh *et al.*, 1995). However, Hermodsson *et al.* (1994) found traumatic amputees had decreased sway in the sagittal direction and this was related to the stiffness of the prosthetic ankle/foot complex. Maintaining postural control places additional demands on muscle control and balancing activity of the contralateral limb (Isakov *et al.*, 1994). Hermodsson *et al.* (1994) concluded that standing balance differed according to cause of amputation and that the residual limb could not fully compensate for the loss of the anatomical foot as a source of somatosensory feedback. However, these authors did not comment on whether prosthetic experience could have explained the decreased postural sway in the traumatic amputees, as they had significantly longer time since amputation (mean: 23.8 years) compared to the dysvascular amputees (mean: 6 years). Isakov *et al.* (1994) believed that that well-trained transtibial amputees were able to adapt to uncomfortable balancing conditions, implying that prosthetic experience was an important factor in postural control.

Computerised dynamic posturography (CDP) using the NeuroCom EquiTest is an objective measure of the contributions of visual, vestibular and somatosensory input for postural control. This method detects postural sway by measuring shifts in the COG as a person moves within their limits of stability and can quantify postural strategies to static and dynamic perturbations. Quantitative posturography has previously been used to assess balance with elderly (Camicioli *et al.*, 1997; Horak *et al.*, 1989; Judge *et al.*, 1995) and differentiate between elderly fallers and non-fallers (Parry *et al.*, 1995; Wallmann, 2001). Judge *et al.* (1995) reported a strong association between loss of balance and postural response scores on the Sensory Organization Test (SOT), indicating the SOT was a valid balance test in older persons. A loss of balance occurred when a person's sway exceeded their limits of stability requiring them to use the stepping strategy or hold onto something for support. The authors also found that greater occurrence of balance loss was linked with poorer performance and greater difficulty with daily tasks in individuals who were otherwise still functioning independently within the community (Judge *et al.*, 1995). In a pilot study by Parry *et al.* (1995), they found the composite score of the SOT could consistently distinguish between fallers and non-fallers, suggesting it was a sensitive diagnostic measure. Similarly, Wallmann (2001) reported significantly lower balance scores in fallers vs. non-fallers indicating poor postural adaptations, especially in conditions where somatosensory and/or visual input were modified.

All these previous findings lend strong support to the use of CDP as part of a diagnositic tool for identifying fallers in older, able-bodied persons. Transtibial amputees have inherently modified sensory input, and yet clearly they are able to maintain their balance even under dynamic conditions. Although previous studies have investigated postural sway in amputees, no study to date has quantified postural control in transtibial amputees when all biological systems (somatosensory, visual and vestibular) are challenged bilaterally using CDP. Postural adaptations to dynamic situations are a vital aspect of independent daily living and warrant further exploration in transtibial amputees.

2.6. Falls

Falls are a common occurrence among older adults and the consequences can be devastating. A fall has been described as "an event which resulted in the person coming to rest inadvertently on the ground or other level, other than as a consequence of lost consciousness, a violent blow, stroke or epileptic seizure" (Askham, 1990). Falling can lead to injury, hospitalisation and long-term institutionalisation. It has been reported that over one third of adults aged over 65 years will experience at least one fall every year (Stevens, 2005; Powell and Myers, 1995; Barak *et al.*, 2006, Tinetti *et al.*, 1988). The incidence of falls increases each decade and over 40% of adults over 80 years fall each year (Powell and Myers, 1995). Some falls go unreported and therefore, the actual incidence of falls is likely to be much higher.

Much literature has been devoted to addressing falls incidence, causes and consequences (Talbot *et al.*, 2005; Ozcan *et al.*, 2005; van Dieen and Pijnappels, 2008) and guidelines for prevention and treatment particularly in older able-bodied adults (American Geriatrics Society, 2001; Rubenstein, 2006). In 2004, the National Institute for Clinical Excellence (NICE) published guidelines on reducing falls. The problem has been equally recognised by the Department of Health with the development of the Older People's National Services Framework (NSF) Standards. Standard Six is specifically related to falls and aims to "reduce the number of falls which result in serious injury and ensure effective treatment and rehabilitation for those who have fallen" (Department of Health, 2008). It is beyond the scope of this thesis to review all the current guidelines in falls prevention and treatment. However, the following references are particularly relevant to the area (American Geriatrics Society, 2001; Askam *et al*., 1990; Ozcan *et al*., 2005; Rubenstein, 2006; Stevens, 2005; Tinetti *et al*., 1988; ProFaNE - Prevention of Falls Network Europe).

Most falls are a result of intrinsic (e.g. muscle weakness, balance or gait impairments) and extrinsic factors (e.g. environmental factors such as poor lighting, obstacles) causing the person to trip or slip during their typical daily activities. Talbot *et al*. (2005) found that the largest number of falls in the older age group (>65 years) occurred when level walking indoors. This indicated that falls in older adults occurred during daily activities and without apparent environmental hazards. Talbot *et al*. (2005) listed the most frequent activities performed prior to falling: 56% of all falls occurred during ambulation (walking, turning), 9% during stair walking/curbs and 9% during transfers (getting in/out of bed/chairs/wheelchairs). These results strongly suggest that continued research on the mechanical demands of performing daily activities is necessary to make recommendations for falls prevention and rehabilitation programmes.

Falls on stairs are often more dangerous than on level ground and may lead to hospitalisation, loss of independent living and may induce a fear of falling. In 2002, 11% (approximately 300,000) of all accidents were the result of falling while walking on stairs (DTI, 2003). Muscle weakness, particularly in the lower-limb extensor muscles, and reduced joint mobility have been cited as the primary reasons for falls during stair walking (Beaulieu *et al.*, 2008; Reeves *et al.*, 2008a; Reeves *et al.*, 2008b). Experiencing a fall when descending stairs is more common than during stair ascent and the consequences are often more severe (Reeves *et al.*, 2008). Self-reported difficulties during stair ascent have been linked with poor balance and gait abnormalities. Problems with stair descent have been linked with more recurrent falls and indicated individuals had limitations when performing other types of daily activities (Verghese *et al.*, 2008).

2.6.1. Falls in lower-limb amputees

Lower-limb amputees experience more falls than their age-matched, able-bodied counterparts. Kulkarni *et al.* (1996) investigated the prevalence of falls in lower-limb amputee patients in the UK. They found that 58% of respondents fell over a one year period; the prevalence was higher in transfemoral (64%) compared to transtibial (53%) amputees. The reported reasons for falling were intrinsic patient related falls (48%), prosthesis related falls (12%), environment related falls (22%) or a combination of the above (18%). Although not significantly different, falls were also more frequent in non-established (less experienced) patients (59%) compared to established patients (53%). A study by Miller *et al.* (2001) reported very similar findings in Canada: 52.4% of the amputee respondents in their survey had fallen within a 12-month period, and 75% of these fallers were recurrent fallers (fallen two or more times). These studies clearly highlight that falls are a significant problem among amputees, especially those over 65 years, and that effective falls prevention and rehabilitation strategies must form an integral part of the amputee rehabilitation programme.

In 2003, BACPAR published evidence based guidelines on physiotherapy treatment in lower-limb amputees. The guidelines provided valuable insight on physiotherapy practice in this patient population. However, the section entitled ‘Coping Strategies Following Falls’ was the only part of the clinical guidelines that specifically addressed falls in amputees (as described in this Chapter, section 2.3, page 7).

In 2006, BACPAR released a revised document on the clinical guidelines of pre and post operative physiotherapy management. The only section on falls was entitled ‘Falls management’ and did not revise the 2003 guidelines despite considerable literature in the area. In both 2003 and 2006, the guidelines’ authors only acknowledged one published study, that of Kulkarni *et al.* (1996). No specific recommendations were made on the causes, prevention or treatment of falls in lower-limb amputees. While the guidelines suggested that ‘advice should be given’, there was no information on what the advice should be.

As discussed in section 2.5.2, biomechanical differences have been indentified and compared in older fallers vs. non-fallers in able-bodied individuals and the results have made recommendations for falls exercise programmes (Kerrigan *et al.*, 1998; Kerrigan *et al.*, 2000; Kerrigan *et al.*, 2001; Lee and Kerrigan, 1999). However, there is currently no published literature that has specifically compared biomechanical parameters between amputee fallers and non-fallers to establish whether differences exist. This type of study would provide valuable information for updating BACPAR’s clinical guidelines on falls.

2.6.2. Fear of falling

Falls, their treatment and management, is a multifactorial problem influenced by physical, mechanical, economical, social and psychological factors. Many older people experience one of the main psychological difficulties associated with falling. These include a fear of falling (Hill *et al.*, 1996; Tinetti *et al.*, 1990 Lachman *et al.*, 1998); 1994), low self-confidence (Li *et al.*, 2002), and activity avoidance (Jorstad *et al.*, 2005; Miller *et al.*, 2001; Powell and Myers, 1995). From the age of 60 years, falls frequency increases with each decade of life, so too does a fear of falling, especially among women (Powell and Myers, 1995). Fear of falling has been associated with decreased mobility and activity avoidance (Myers *et al.*, 1998). Fear of falling and low falls efficacy may affect previous fallers and non-fallers alike. The incidence of fear of falling has been described over wide ranges: 12-65% in non-fallers and 29-92% in previous fallers (Jorstad *et al.*, 2005).

Fear of falling has also been found to influence temporal-spatial gait parameters such as walking speed, stride length and width, and double support times (Chamberlin *et al.*, 2005) and balance ability (Maki *et al.*, 1991). Therefore, understanding the relationships between falls, fear of falling, and functional performance has become an important area of research for improving health-related quality-of-life, physical independence and function in the elderly. Indeed, several major research councils in the UK have introduced initiatives that focus on healthy ageing and independence (New Dynamics of Ageing programme) and the relationship between biological factors and mental well-being (Lifelong Health and Wellbeing initiative) as part of the promotion of healthy ageing. These research calls acknowledge that developing recommendations for healthy ageing should be multidisciplinary.

In the falls literature, fear of falling has been measured by asking people whether they were afraid of falling, providing a yes/no response (Maki *et al.*, 1991; Miller *et al.*, 2001). This method has poor sensitivity and assumes that fear of falling is a dichotomous variable (Hill *et al.*, 1996). Other methods and scales have been developed that measure self-efficacy and balance confidence on a continuum ranging from “not at all confident” to “completely confident” (Tinetti *et al.*, 1994; Powell and Myers, 1995; Hill *et al.*, 1996). The advantage of these methods is that fear of falling may be quantified, even relatively small changes can be monitored and the effectiveness of treatment programmes can be evaluated.

A number of valid and reliable instruments have been developed to assess fear of falling, self-efficacy and confidence on performance on everyday tasks, e.g. Falls

Efficacy Scale (FES) (Tinetti *et al*., 1990), Modified Falls Efficacy Scale (MFES) (Hill *et al*., 1996); Activities-specific Balance Confidence Scale (ABC) (Powell and Myers, 1995). Fear of falling is typically measured using two approaches: single-item questions about fear of falling, such as the Survey of Activities and Fear of Fallig in the Elderly (SAFE) questionnaire (Lachman *et al*. 1998), and self-efficacy and balance confidence questions about performance of everyday tasks, such as the FES (Tinetti *et al*., 1990), MFES (Hill *et al*., 1998) and ABC-scale (Powell and Myers, 1995).

Despite considerable literature dedicated to the topic of fear of falling, it is surprising there is no one agreed definition. Many studies appear to use the terms 'fear of falling', 'falls efficacy', 'falls confidence' and 'balance confidence' interchangeably. Within the falls literature, falls efficacy and/or balance confidence relate to an individual's perceived ability to perform daily tasks without falling (Hill *et al*., 1996; Jorstad *et al*., 2005; Powell and Myers, 1995; Tinetti *et al*, 1990). Furthermore, some studies have assumed that falls efficacy and performance of functional tasks and physical function are related (Li *et al*., 2002). Throughout this thesis, the term 'balance confidence' will be used to describe the belief that one can perform daily activities without falling. An understanding of Bandura's social cognitive theory would be useful in interpreting many of the questionnaires described in this thesis and the following section provides some background information about this theoretical framework.

2.6.3. Self-efficacy – social cognitive theory

In 1977, Bandura developed a theoretical framework of self-efficacy to evaluate changes achieved in fearful and avoidant behaviour. "Perceived self-efficacy refers to beliefs in one's capabilities to organize and execute the courses of action required to produce given attainments" (Bandura, 1997; p. 3). Self-efficacy has been shown to influence a person's choice of activities and settings and their expectations and success. People who have greater perceived self-efficacy persist for longer when encountering obstacles and/or difficulties which ultimately further reinforce their sense of self-efficacy. Conversely, people who ceased their coping strategies in the face of adversity retained their low self-expectations and hence, have low self-efficacy. Therefore, self-efficacy is relevant for the study of falls and the effect of falls on quality-of-life.

Self-efficacy beliefs are formed from four main sources of information.

I. Past performance experience: This is the most important source of information because it is based on mastery (authentic) experience. Good self-efficacy is based on previous successes, whereas poor self-efficacy is caused by previous failings.

Therefore, if a person were to fall before they developed firm self-efficacy in their capability of performing an activity (e.g. an inexperienced amputee walking down stairs), then their self-efficacy would be undermined. Equally, repeated failures could cause poor self-assurance and lead to activity avoidance, which would negatively impact on quality-of-life.

II. Vicarious experience: Bandura (1997) stated that modelling is another tool for developing self-efficacy and that people can appraise their capabilities according to others' skills and ability. Visualising similar people performing activities successfully boosts efficacy beliefs. With respect to amputees and previous fallers, observing others of similar age or physical characteristics performing an activity (e.g. stair walking) that might otherwise be perceived as difficult, could increase self-efficacy if the task is performed successfully. Conversely, if the other person fails, it could undermine self-efficacy.

III. Verbal persuasion: This source reflects reinforcing others self beliefs by telling them they have the capabilities of succeeding and is most effective when the beliefs are within attainable and realistic boundaries. Verbal persuasion can be done by others through evaluative feedback or by oneself, through self-talk. The role of verbal persuasion in falls prevention and treatment may be most effective when encouraged early during rehabilitation or immediately after experiencing a fall.

IV. Physiological information: Judging one's own physiological or emotional state reflects personal efficacy, especially in activities that relate to health functioning and physical performance. Variables such as poor strength, fatigue and pain are considered indicators of physical self-efficacy. Enhancing physical characteristics is one way of improving self-efficacy (Bandura, 1997). Therefore, it is important that individuals understand how improvements in strength could positively influence their physical performance and ultimately their own self belief. This would be most beneficial in people who have reduced motor functioning as a result of age or disease.

Influenced by Bandura's theory of self-efficacy, other authors have linked decreased balance confidence with reduced performance (e.g. activity avoidance) (Miller *et al.*, 2001). Miller and colleagues (2002) cited benefits to using balance confidence as an indicator of fear of falling: 1) balance confidence could be measured on a continuum, 2) asking about confidence was considered less threatening than asking about a fear and 3) self-confidence was strongly linked with independence in activities (Tinetti *et al.*, 1994; Miller *et al.*, 2001; Miller *et al.*, 2002).

2.6.4. Balance confidence in lower-limb amputees

Compared to balance confidence in the elderly, there are fewer studies that have addressed fear of falling among community-dwelling amputees. Miller *et al.* (2001) investigated the prevalence of falls and fear of falling in a group of 435 amputees. Almost half of their respondents reported a fear of falling when asked "Are you afraid of falling?" and 76% of those who had previously fallen said they avoided activities because of their fear. However, a fear of falling was also reported among 43% of the non-fallers. A limitation of this study is that the authors established that individuals had a fear of falling but they did not establish the magnitude of this fear. That is, like other fears and anxieties, it is expected that individuals will vary in their intensity of experiencing fear of falling. These authors recommended that future research should focus on exploring whether fear of falling caused activity avoidance and thereby decreased physical function and quality-of-life or *vice versa*.

Using the same sample set, the same research group evaluated balance confidence among lower-limb amputees using the ABC scale (Miller *et al*., 2003). Participants were most confident at performing activities such as reaching at eye level, getting in and out of a car and walking around the house. They were least confident at walking on icy pavement, standing on an escalator without holding the rail and standing on a chair to reach something. The authors found that there were differences in scores between transtibial and transfemoral amputees, as well as between those who had fallen in the past year compared to those who had not fallen, but that they were not statistically significant. Moreover, they compared the results for amputees with vascular vs. non-vascular causes and found that non-vascular amputees scored significantly higher on all items of the ABC scale. However, all amputees scored lower mean scores (X = 63.8) than older people with good health (X = 88). The authors of the ABC scale suggested that a mean score of less than 80 was indicative of a need for intervention (Powell and Myers, 1995). Miller *et al*. (2003) concluded that balance confidence strongly correlated with prosthetic performance (what people did with their prosthesis), prosthetic capability (what people could do with their prosthesis) and engaging in social activities. Miller *et al*. (2003) advocated incorporating balance confidence into prosthetic rehabilitation as a method of improving quality-of-life in lower-limb amputees.

A review of the literature has found that few studies have investigated falls efficacy in lower-limb amputees specifically. Miller *et al.* (2001, 2003, 2004) have conducted research in the area, but admittedly with some limitations. They acknowledged that the ABC scale may not be a valid measure of falls efficacy in lower-limb amputees. The

ABC scale is not disease specific and, when used with lower-limb amputees, could not distinguish clinical differences according to level of amputation (Miller *et al.*, 2003). Despite these shortcomings, Miller and colleagues continued to use the ABC scale in their published research (Miller *et al.*, 2004). Although studies have reported a relationship between balance confidence and actual performance (Miller *et al.*, 2001), to date no study has been found that has actually quantified the relationships between balance confidence and physical performance on validated performance measures, such as the Timed Up and Go test (Podsialdo and Richardson, 1991) and 10m timed walk test (Watson, 2002).

2.7. Aim and objectives

The overall aim of this thesis was to assist the clinical recommendations to improve amputee rehabilitation as it relates to falls prevention and treatment. In order to achieve this aim, the following objectives were set:

- To determine whether physiotherapists were monitoring falls incidence among their patient population and whether amputee physiotherapists used functional and psychological outcome measures to inform patients' discharge and to understand how physiotherapists determined a patient's readiness for discharge. It was hypothesised that falls would not be monitored regularly in prosthetic centres. It was hypothesised that physiotherapists would use functional outcome measures more than psychological measures, but that there would be no consistency of use across England and that outcome measures would not inform a patient's discharge.

- To further our understanding of the biomechanics of transtibial amputees performing specific activities of daily living that are important for independent mobility. These included level gait, stair ascent and descent and postural control during static and dynamic conditions.

- To compare the gait patterns of fallers and non-fallers in transtibial amputees and able-bodied controls during level walking, stair ascent and descent. It was hypothesised that fallers would walk more slowly and demonstrate reduced joint mobility compared to their non-faller counterparts. It was also anticipated that the fallers would exhibit reduced lower-limb joint moments and powers and that these would be evident at critical moments in the gait cycle, such as during the transition from double to single support and when changing from level walking to stair walking and vice verse.

- To determine whether computerised dynamic posturography could be used to differentiate between fallers and non-fallers during static and dynamic conditions using the NeuroCom Smart Equitest system and to measure and quantify the postural control strategies during static and dynamic conditions in transtibial amputee and able-bodied control groups..It was hypothesised that the fallers would have increased postural sway and show greater reliance on hip strategies in response to dynamic perturbations compared to the non-fallers. It was also expected the amputee fallers would exhibit greater reliance on the affected limb by bearing more weight through this limb during backwards and forwards translations.

- To determine if fallers and non-fallers differed in their performance of basic functional tests and on balance confidence and quality of life questionnaires and to determine if a relationship existed between functional and psychological measures according to a person's falls history. It was hypothesised that previous fallers would perform the functional performance tasks more slowly and have lower balance confidence on everyday activities and quality of life scores compared to the non-fallers. It was also postulated that lower performance scores would be associated with lower balance confidence and quality-of-life scores

CHAPTER THREE – GENERAL METHODS, EQUIPMENT AND OUTCOME MEASURES

3.1. Introduction

The current chapter discusses the ethical review process, the participants and inclusion/exclusion criteria and the general procedures that were followed. The equipment and outcome measures are also included, as well as considerations for their use. Biomechanical variables are explained in this chapter. Testing protocols are outlined, for level walking, stair ascent and descent, postural stability and balance confidence. Data analysis procedures are detailed in the methods section of the associated experiment.

3.2. Ethical approval

This study was reviewed and approved by the South Humber NHS Local Research Ethics Committee (REC) in November 2005 (REC ref. number: 05/Q1105/68). The Participant Information Sheet (Appendix 1) and Informed Consent Form (Appendix 2) can be found in Appendices 1 and 2, respectively. As per LREC approval, prospective amputee participants were identified by the Prosthetics Services Manager at the Hull and East Yorkshire (HEY) NHS Trust Artificial Limb Unit. Participants had to meet the inclusion and exclusion criteria below (section 3.2.1.). Able-bodied participants were recruited from the local community by word of mouth and posters.

3.2.1. Inclusion and exclusion criteria

Amputee participants were included in this study if they had experienced a unilateral lower-limb amputation below the knee and wore a prosthesis on a daily basis without experiencing any self-rated discomfort. They must have completed a rehabilitation programme following their amputation. In order to undertake the biomechanical analysis, all participants must have been able to walk a minimum distance of 10 metres with their prosthesis 12 times, but without mobility aids (e.g. crutches), turn 180° degrees while walking, stand quietly for up to 5 minutes and walk up and down a custom-built three-step staircase with handrails. Amputees were excluded if they had a unilateral transfemoral or bilateral lower−limb amputation. All participants were excluded if they had known neurological or gait disorders, had any current or past medical diagnosis affecting balance; any known vestibular and/or neurologic disorder; were taking any medication known to affect balance and coordination; or had any current symptoms of dizziness, suffered from blindness or hearing disorders, were unable to understand verbal explanations or written information given in English or have special communication needs, were unable to follow instructions (for the safety of participants and researchers).

3.2.2. Falls history

A fall was described as an event which resulted in the person coming to rest inadvertently on the ground or other level, other than as a consequence of lost consciousness, a violent blow, stroke or epileptic seizure (Askham, 1990).All participants were classified into either the 'Non-faller' or 'Faller' (F) groups based on their falls' history. Individuals who had experienced a fall in a maximum 12-month period leading up to testing were classified as a previous 'Faller'. Individuals who had not fallen in this period were considered 'Non-fallers'. Actually, all fallers (amputee and able-bodied) had fallen within a 9-month period prior to testing. Amputee and able-bodied participants were matched according to age, but not falls history. Three amputee fallers were below the age of 50 (range 43-46 years) and it would have been difficult to find age-matched controls who had recently fallen but with no other underlying health condition.

3.3. Participants

A total of twenty participants took part in this study. Eleven (n=11) were unilateral, transtibial amputees and nine (n=9) were age-matched, able-bodied control participants. Participant characteristics are presented in Table 3.1. All participants gave written informed consent prior to testing. The original power calculation for this study was a total of 16 amputees, with 8 participants each in the faller and non-faller groups. This was based on achieving a power of 80% with $p < 0.05$. Sixteen participants were recruited from over a 4-month period. However, 5 of these participants did not fit the inclusion criteria (e.g. they could not walk without experiencing pain from the prosthesis). Therefore, a total of 11 amputees completed the protocol.

3.4. General participant preparation

Prior to testing, all participants were contacted and instructed to bring comfortable flat shoes in which they would perform the walking and postural tests. They were also asked to bring shorts and a short sleeved shirt. If someone did not have a pair of shorts, these were provided by the principal investigator.

3.5. Height and mass determination

The height of all participants was measured using a free-standing stadiometer (SECA, Germany). Transtibial amputees cannot stand on their prosthesis without shoes. Therefore, all participants were measured whilst wearing their flat shoes. Body mass was determined from the static data capture trial when the participants stood stationary on one force plate (AMTI, Massachusetts, USA). Their body weight was measured in Newtons (N) and converted into mass (kg) by dividing by 9.81 m/s^2. All participants' mass was measured when they were clothed in shorts, short sleeved shirt and shoes.

Table 3.1. Amputee and control participant characteristics

Amputee subjects	Gender	Age (yrs)	Height (cm)	Body mass (kg)	Amputated Limb	Type of prosthetic foot	Reason for amputation	Time since amputation (yrs)
Faller								
1	M	46	181	83	Left	Variflex foot	Traumatic	12.0
2	M	43	173	76	Right	Ceterus foot	Traumatic	1.2
3	M	67	168	62	Right	Multiflex foot	Traumatic	1.7
4	M	43	196	93	Left	Multiflex foot	Traumatic	4.0
5	M	65	185	92	Right	Multiflex foot	Vascular	0.8
6	M	71	165	63	Right	Multiflex foot	Vascular	1.3
Mean (SD)		**56 (13)**	**178 (12)**	**78 (13)**				**3.5 (4.3)**
Non-faller								
7	F	50	163	97	Right	Dynamic foot	Clubfoot	1.0
8	M	82	169	88	Left	Multiflex foot	Vascular	3.3
9	F	70	147	49	Left	Multiflex foot	Traumatic	22.0
10	M	26	185	63	Right	Variflex foot	Clubfoot	0.8
11	M	55	185	73	Left	Multiflex foot	Traumatic	26.0
Mean (SD)		**57 (21)**	**170 (16)**	**74 (19)**				**10.6 (12.3)**
Control subjects								
Faller								
1	F	72	157	69				
2	F	77	170	80				
3	F	65	168	74				
4	M	74	173	102				
Mean (SD)		**72 (5)**	**167 (7)**	**82 (14)**				
Non-faller								
5	F	72	155	68				
6	M	60	193	93				
7	M	58	165	73				
8	M	42	190	91				
9	M	30	187	66				
Mean (SD)		**52 (16)**	**178 (17)**	**78 (13)**				
***p* value**		**0.29**	**0.50**	**0.91**				**0.28**

p value indicates no significant differences between the four groups for age, height and body mass (one-way ANOVA) and between the two amputee groups for time since amputation (independent samples t-test)

3.6. Motion capture system

Three-dimensional kinematic data were captured using Qualisys Track Manager software (Qualisys, Gothenburg, Sweden). Ten ProReflex MCU1000 cameras (Qualisys, Gothenburg, Sweden) captured 3D marker coordinate data at 100 Hz. These cameras emit infra-red light. When the light from the camera hits the retro-reflective markers attached to the participant, light is reflected back to the cameras. The cameras were wall- and tripod-mounted at varying heights surrounding the measurement area. The measurement volume was approximately 6 m^3 for the level gait trials (Figure 3.1). The motion capture system was calibrated using a 750 mm calibration wand and large L-frame reference object identifying the lab origin.

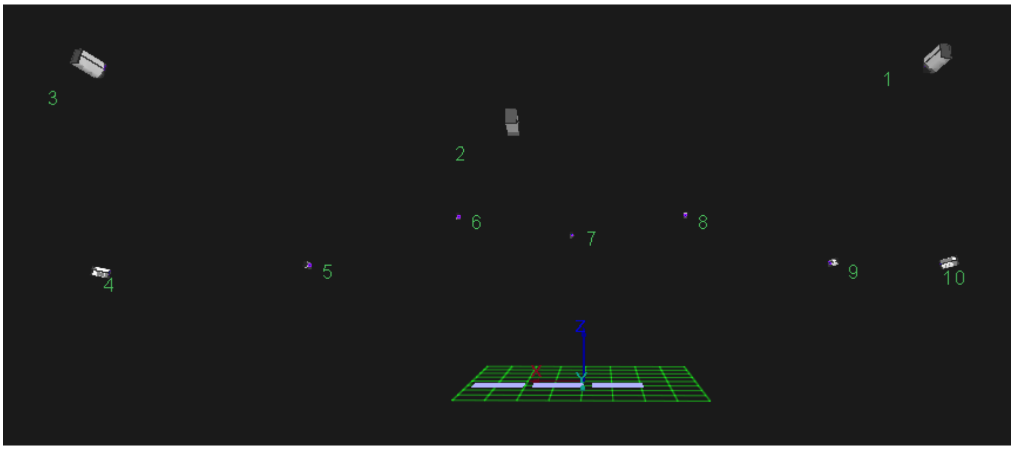


Figure 3.1. Ten-camera set-up for level walking trials. The L-frame was placed onto the middle force plate and was used to define the lab origin.

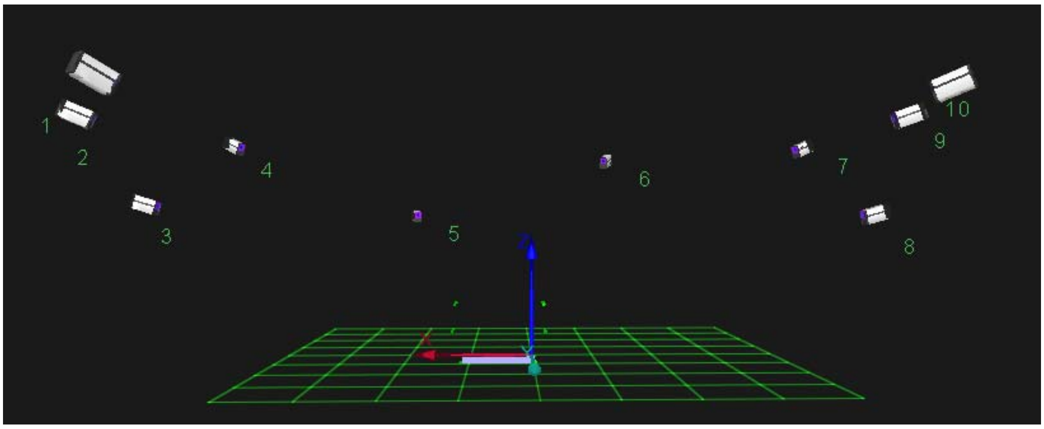


Figure 3.2. Ten-camera set-up for stair walking trials. The L-frame was placed onto the force plate on the first step and was used to define the lab origin

Ten cameras were also used to capture motion during the stair walking trials. In this case, all ten cameras were mounted on tripods in a circle surrounding the staircase. The measurement volume was approximately 4 m^3 for the stair trials (Figure 3.2). The motion capture system was calibrated using a 300 mm calibration wand and medium L-frame. The smaller calibration wand was used because the volume within the staircase was more constricted than the level walking volume and the cameras could see the smaller L-frame on the force plate within the first step.

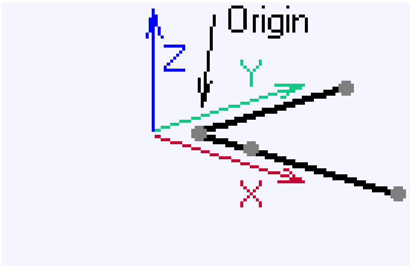
The 3D tracker parameters had a prediction error of 30 mm, maximum residual of 10 mm, acceleration factor of 50 000 mm/s^2 and noise factor of 10 mm. System reliability tests were carried out and the results showed that the system had a coefficient of variation of 0.19% over 20 trials within the 6 m^3 volume and 0.17% within the 4 m^3 volume (Appendix H).

Prior to each data capture session, the cameras and the field of view were checked. This was done to ensure no sunlight or other reflective noise existed within the measurement volume. The camera aperture was adjusted accordingly, only if necessary.

3.7. Calibration

Calibration of the measurement volume was performed prior to every data collection session and every participant. For the level walking trials, a 750 mm wand kit was used. The exact wand length was unique to each calibration structure and was specified at 749.9 mm within the QTM software. The L-frame was positioned onto the corner of the middle AMTI force plate. The coordinate system was calculated such that the positive Z-axis pointed upwards and the long arm axis of the L-frame was in the positive x-axis direction. The short arm (A) end marker was positioned 550 mm from the lab origin location, the long arm (C) end marker was 750 mm from the lab origin and the long arm middle marker (B) was 200 mm from the lab origin (Figure 3.3).

A smaller wand kit (300 mm) was used for the stair walking trials. The exact wand length was 299.5 mm. The L-frame was positioned on the force plate of the first step such that at least 5 cameras could see the markers on the L-frame and an extended calibration process was done in QTM. The L-frame dimensions were such that the short arm (A) end marker was positioned 200 mm from the lab origin location, the long arm (C) end marker was 300 mm from the lab origin and the long arm middle marker (B) was 90 mm from the lab origin.



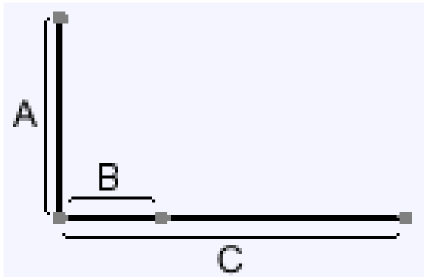


Figure 3.3. Positive directions of the L-frame were x-positive long arm, y-positive short arm, z-positive upwards. (left) Position of four markers defining the long and short arms of the L-frame (right)

During calibration, the wand was waved around as much of the measurement volume for 100 seconds so that as many cameras as possible saw the wand at all times. Calibration was considered successful when the mean residual of each camera was less than 2.0 mm.

3.8. Force data

For the level walking trials, force data were measured with three AMTI (model BP600600) force plates (Advanced Medical Technology Inc, Massachusetts, USA) that were flush with the floor and embedded into the gait walkway. The force data were synchronised with the motion capture system via a 64-channel AD board (Qualisys, Gothenburgh, Sweden) and all force data were collected in QTM software at a sampling frequency of 500 Hz. The three force plates were zeroed between each trial. Force plate locations were recorded to define the global lab coordinate system and the values were saved in the Workspace folder within the QTM software.

For the stair walking trials, force data were measured with one Kistler (model 9286AA) force plate (Kistler GmbH, Winterthur, Switzerland) with built-in charge amplifiers and embedded into the first step of the staircase. The force data were synchronised with the motion capture system via a 16-channel AD board (Qualisys, Gothenburgh, Sweden). All force data were collected in QTM software at a sampling frequency of 500 Hz. Prior to every data collection session, the location of the force plate was determined with 4 mm markers placed in each corner. These values were then saved within the Workspace folder for that participant.

3.9. Six degrees of freedom

The six degrees of freedom (6 DOF) model set-up was used to capture the three-dimensional coordinates of body segments (Buczek *et al*., 2009; Cappozzo *et al*., 1995). It is assumed that each segment has six variables that describe its position and orientation. Three variables relate to the position of the origin, and three indicate the rotation about each of the principal axes (x,y,z) of the segment. Unlike other models used in 3D motion capture of clinical gait analysis, the 6 DOF model tracks the adjacent segments to joint centres and makes no assumptions about joint constraints during dynamic motion trials (Kirtley, 2006). A set of three (or more) markers are attached to a rigid segment and track the body segment movement to specify its position and orientation. The marker set-up is a combination of markers placed on a rigid surface and markers placed on anatomical landmarks directly on the skin.

3.10. Marker placement

Following the 6 DOF, 44 passive reflective markers (14 mm) were affixed onto the ankle, knee and hip joint centres, and pelvic landmarks, using double-sided tape, according to the calibrated anatomical systems technique (CAST) principles (Cappozzo *et al*., 1995). Specific marker locations are illustrated in Figure 3.4 and described in Table 3.2. Four tracking markers arranged on rigid clusters were securely fastened around the participant's thigh and shank/prosthesis bilaterally. These tracking markers served to track body segments. Participants completed a static trial that served as a calibration file for model building. Markers that defined joint centres were then removed and participants completed the dynamic motion trials with four markers on each body segment: foot, shank, thigh (all bilaterally) and pelvis. For the dynamic walking trials, participants wore 28 reflective markers, with four markers on each body segment.

3.11. Definition of limb terminology

The following terminology was used to differentiate between the two limbs for all the gait and posture data:

'Intact' referred to the ankle, knee and hip on the limb unaffected by the amputation.

'Prosthetic' only referred to the ankle and foot complex, which had been replaced with a prosthesis, on the amputated side.

'Affected' referred to the knee and hip on the amputated side.

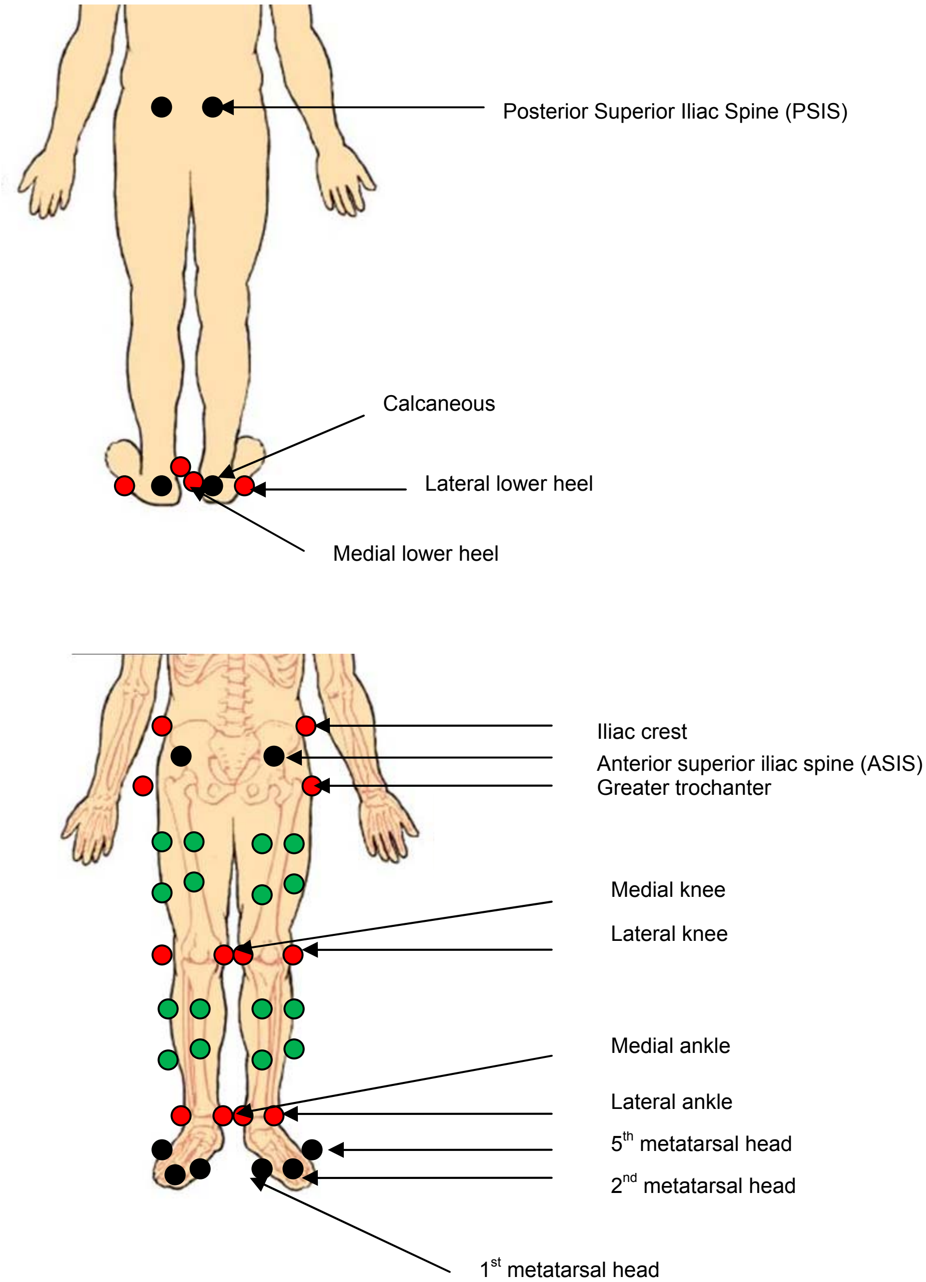


Figure 3.4. 6 DOF marker set-up. The green markers were mounted onto rigid clusters and served as tracking markers of the thigh and shank segments. The black markers were affixed onto skin (or shoes). The red markers were used to define joint centres and were used in the static calibration file only.

Table 3.2. Anatomical landmarks used to define the lower-limb segments according to the 6 DOF marker set-up. *Italicised landmarks* were used for the standing (static calibration) trial only and removed during the dynamic walking trials.

Pelvis	
PSIS	posterior superior iliac spine
ASIS	anterior superior iliac spine
GT	*greater trochanter*
IC	*iliac crest*
Femur	
TH_1	thigh tracking marker secured onto rigid cluster
TH_2	thigh tracking marker secured onto rigid cluster
TH_3	thigh tracking marker secured onto rigid cluster
TH_4	thigh tracking marker secured onto rigid cluster
LE	*lateral femoral epicondyle*
ME	*medial femoral epicondyle*
Shank	
SH_1	shank tracking marker secured onto rigid cluster
SH_2	shank tracking marker secured onto rigid cluster
SH_3	shank tracking marker secured onto rigid cluster
SH_4	shank tracking marker secured onto rigid cluster
LM	*distal aspect of lateral malleolus*
MM	*distal aspect of medial malleolus*
Foot	
1st_MT	dorsal aspect first metatarsal head
5th_MT	dorsal aspect of fifth metatarsal head
2nd_MT	dorsal aspect of second metatarsal head
CAL	upper aspect of posterior calcaneous
medial lower heel	*inferior to medial malleolus marker at height of 1st_MT marker*
lateral lower heel	*inferior to lateral malleolus marker at height of 5th_MT marker*

3.12. Three-dimensional coordinate system and marker reconstruction

The three axes were defined according to the position of the L-frame during the calibration process. The z-axis was vertical and all markers had a positive value. The x-axis was defined as the direction of progression for level walking, while the y-axis was medial/lateral and perpendicular to the x- and z-axes. In stair walking, the y-axis was the direction of progression. Once a trial was captured in QTM, the software immediately reconstructed the 3D coordinates of each marker. The coordinates were based upon the Cartesian coordinate system (x,y,z) and identified the position of the marker with respect to the lab origin.

3.13. Participant calibration – static trial

The static file defines the local coordinate system for each segment. Once the participant was prepared with the static 6 DOF marker-set, the static trial was captured. On the command ‘go’, the participant was instructed to take two or three steps and position themselves onto one of the force plates. They were requested to stand motionless with their feet shoulder-width apart and arms loosely at their side. This was to ensure the medial joint markers could be easily distinguished on the right and left sides. The static trial was captured for 5 seconds. The file was checked to ensure all markers were clearly visible. The markers where then manually identified and labelled and two frames of data were exported as a C3D file to create the model in Visual 3D. Body weight was determined by averaging the recorded (filtered) force (in Newtons) when the participant was standing motionless for 2 seconds on the force plate.

3.14. Level walking trials

Participants were asked to walk along a 10 m walkway in the laboratory at their self-selected walking speed. Kinematic and GRF data from the left and right limbs were recorded bilaterally when the participant entered into the measurement area. A total of 10 walk trials with good foot strike patterns were selected for analysis (on average, participants completed 13-15 walk trials to achieve 10 good trials). The trial was not selected for analysis if the participant deliberately changed their gait pattern to target the force plate. Participants were not given any walking instructions other than to walk at their self-selected pace without any walking aid and were allowed adequate rest if needed. If they did not make complete foot contact on the force plate, the kinetic data from that trial was discarded.

3.15. Stair walking trials

3.15.1. Stair ascent

On the command ‘go’, participants were asked to walk along a 2 m walkway, climb the 3-step staircase and continue walking once they reached the top step. There was an 80 cm long landing at the top where participants could turn around and stop. They were asked to complete the task at their self-selected pace, using their typical stair walking pattern, and could use the handrails if they felt necessary or more comfortable in doing so.

3.15.2. Stair descent

Participants were instructed to stand at the back of the landing (80 cm away from the edge of the top step) surrounded by the handrails. On the command 'go', participants walked to edge of the steps and began descending the 3-step staircase and continued level walking once they reached the floor. They were given the same instructions about walking pace, pattern, and handrail use as during the stair ascent trials. All participants completed 10 stair ascent and descent trials.

3.16. Motion data analysis and model build in Visual 3D

Once the motion data were collected in QTM, each marker was identified, labelled and checked for marker switching. An AIM model was then generated which enabled the automatic identification of markers in subsequent motion trials. The AIM model looked for constant distances between the different markers based on the first manually labelled file.

In order to recreate the body segments in the Visual3D motion analysis software (C-Motion, Rockville, MD, USA), the 3D walking files were exported from QTM into Visual 3D as C3D files. Kinematic data were interpolated using a cubic spline algorithm over no more than 10 consecutive frames. Noise introduced during the data collection process were reduced by applying a low-pass 2nd order Butterworth filter, with a cut-off frequency of 6 Hz, to the kinematic data. Kinetic data were filtered with a cut-off frequency of 15Hz.

A biomechanical 'Visual3D Hybrid 3D model' (6 DOF link model) was constructed using the average value for the marker location from the static file. This 6 DOF model consisted of a series of rigid segments representing skeletal bone structures, liked by joints.

The 6 DOF model was made up of a left and right, thigh, shank and foot. One limitation to the model used was that the foot was modelled as one rigid segment and therefore, did not distinguish between the ankle and subtalar joints. Only the pelvis was modelled as a CODA segment using the left and right ASIS and PSIS markers (Bell *et al*., 1989; Bell *et al*., 1990). A virtual lab and Visual 3D pelvis were modelled to calculate pelvic motion in 3D. The X-Y-Z Cardan sequence defined the order of rotations following the Right Hand Rule about the segment coordinate system axes.

A joint was defined as the location of the distal end of one segment and the proximal end of the adjacent segment. Applying the 6 DOF model, Visual 3D software constructed individual segments regardless of the joint constraints and considered each segment separately. The biomechanical model built in Visual3D used the following rules regarding segment definition:

- To define the proximal end of a body segment, two markers (the lateral and medial proximal end joint centre) must have been visible and identified in Visual3D
- To define the distal end of a body segment, two markers (the lateral and medial distal end joint centres) must have been visible and identified in Visual 3D
- At least 3 tracking markers for every segment must have been visible for each frame of associated motion file and identified in Visual3D. Therefore, four tracking markers mounted onto rigid clusters were used to ensure the minimum number of tracking markers were visible despite potential marker obstruction.

In Visual 3D, a segment has an inertial property and mass based on Dempster's values (Dempster, 1955). The segment's moment of inertia and location of the segment COM are calculated assuming the segment has a geometrical shape.

Anthropometric data in Visual 3D were derived from Dempster's values (Dempster, 1955) and the inverse dynamic analysis assumed that the prosthetic ankle and foot components were the same as for able-bodied individuals. Although this could have affected subsequent knee and hip joint moment and power calculations, this method has been previously reported by Vickers *et al.* (2008).

Each segment has a local coordinate system called the segment coordinate system (SCS). The SCS was computed using 4 markers, where markers were placed on the medial and lateral positions of the proximal and distal segment endpoints. The endpoints were then calculated based on the mid-point of the medial and lateral markers. The z-axis passed through both segment endpoints from distal to proximal direction. The frontal plane (x- and z-axes) was fitted using four markers (medial proximal, lateral proximal, medial distal, lateral distal). The y-axis was orthogonal to the x- and z-axes.

The pelvis segment was constructed using the CODA model (Figure 3.5) with anatomical locations of the ASIS (Anterior Superior Iliac Spine) and the PSIS (Posterior Superior Iliac Spine) bilaterally (Bell *et al*., 1989; Bell *et al*., 1990). The system coordinates of the pelvis where such that the x-plane was defined as the plane bisecting the left and right ASIS markers, and the y-plane was the plane bisecting the mid-point between the left and right PSIS markers. The z-axis was orthogonal to the x- and y-planes. The origin of the pelvis was the mid-point between the ASIS markers

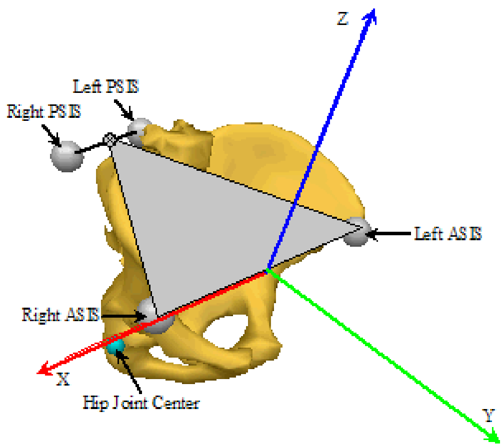


Figure 3.5. CODA pelvis. (C-Motion, Inc, 2009)

Once the CODA pelvis was created, the left and right hip joint centres (x,y,z) were calculated. These were based on the following equations:

RHJC = (0.36*ASIS_Distance,-0.19*ASIS_Distance,-0.3*ASIS_Distance)

LHJC = (-0.36*ASIS_Distance,-0.19*ASIS_Distance,-0.3*ASIS_Distance) (Bell *et al.* 1989; Bell *et al.*, 1990)

RHJC: right hip joint centre; LHJC: left hip joint centre

Another pelvis (Visual 3D pelvis) was also modelled using the iliac crest (proximal) and greater trochanter (distal) markers (Figure 3.6). The Visual 3D pelvis does not create a hip joint centre by default. It was created to calculate pelvic tilt and obliquity. The length of the segment was the distance between the proximal and distal markers. The hip angle was calculated as the angle between the thigh and pelvis; the knee angle represented the angle between the shank and thigh; and the (neutral) ankle angle was the angle between the virtual foot and shank, with the proximal segment acting as the reference segment. A virtual foot was created to define a neutral ankle angle in the standing (static) trial. The x-, y- and z-planes are shown in Figure 3.6.

The following joint angular definitions were used:

	Positive	**Negative**
Sagittal plane	Flexion	Extension
	Dorsiflexion	Plantarflexion
Frontal plane	Adduction	Abduction

The following pelvic definitions were used:

	Positive	**Negative**
Sagittal plane	Anterior tilt	Posterior tilt
Frontal plane	Obliquity up (pelvic hike)	Obliquity down (pelvic drop)

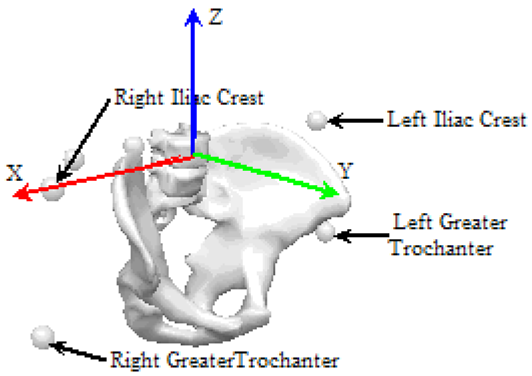


Figure 3.6. Visual 3D pelvis (C-Motion, Inc., 2009)

Using the 6 DOF model, there is no clear joint centre. The joint coordinate system is defined as the intersection between the proximal and distal segments about the joint. Visual 3D assumes the joint centres are situated at the proximal end of the segment that is positioned distal to the joint. The proximal and distal ends of adjacent body segments are 'linked' and move relative to each other rather than rotate about a central axis. For example, the knee joint centre is assumed to be at the proximal end of the shank segment.

3.17. Gait analysis

Once the model was built, gait events were determined from kinetic data and the data were normalised to the gait cycle. In the level walking and stair ascent trials, the gait cycle was initiated with initial foot contact (IC, 0%) and terminated with the subsequent foot contact for the same limb (100%). In the stair descent trials only, the gait cycle was initiated with toe off (TO, 0%) and terminated with the subsequent toe off for the same limb (100%).

A user-created gait report template was used to define and calculate temporal-spatial data, joint angles and ground reaction force data. Joint moments and powers were calculated using inverse dynamic analysis.

3.18. Joint moments and powers

Joint moments are calculated based on Newtonian laws where it is assumed the only forces acting on the system are gravity, the external ground reaction force and internal forces from muscles and tendons. Anthropometric data, such as segment COM location and inertial properties, were based on Dempster's values (Dempster, 1955). Joint moments are first calculated for the ankle joint and then resolved proximally for the knee and hip. Visual 3D uses the measured 3D marker coordinates to calculate accelerations and GRF (with COP) data to calculate internal joint moments. By convention, extensor and abductor moments were reported as positive and flexor (dorsiflexor) and adductor moments were reported as negative (Eng and Winter, 1995; Winter, 1991).

Joint reaction force (x-axis): $\boldsymbol{\Sigma F x = m a_x}$, $\boldsymbol{R_{xp} - R_{xd} = m a_x}$ **(equation 1)**

Joint reaction force (y-axis): $\boldsymbol{\Sigma F y = m a_y}$, $\boldsymbol{R_{yp} - R_{yd} = m a_y}$ **(equation 2)**

Joint reaction force (z-axis): $\boldsymbol{\Sigma F z = m a_z}$, $\boldsymbol{R_{zp} - R_{zd} - mg = m a_z}$ **(equation 3)**

Net muscle moment about the COM: $\boldsymbol{\Sigma M = I_0 \alpha}$ **(equation 4)**

where a_x, a_y, a_z represent the acceleration of the segment COM;

R_{xd}, R_{yd}, R_{zd} represent the reaction forces acting at the distal end of the segment, calculated previously by computing the proximal forces acting on the distal segment;

R_{xp}, R_{yp}, R_{dp} represent the reaction forces acting at the proximal joint (unknown variables);

α is the segmental angular acceleration in the x,y and z planes

M_d = net muscle moment acting at distal joint, calculated previously from an analysis of the proximal muscle acting on distal segments

M_p (unknown variable) = net muscle moment acting on segments at proximal joints (derived from Winter, 1991).

The muscle moment at the proximal end can only be computed once the proximal reaction forces (R_{xp}, R_{yp}, R_{dp}) have been calculated.

The following joint moment definitions were used:

	Positive	**Negative**
Sagittal plane	Extensor	Flexor
	Plantarflexor	Dorsiflexor
Frontal plane	Abductor	Adductor

Once joint moments were computed, the joint moment data and joint angular velocity were used to calculate joint powers. Joint power is the product of moment of force and angular velocity at the joint. Positive joint power indicates the moment of force is acting in the same direction as the angular velocity. The muscle is contracting concentrically and therefore generating power. Conversely, negative joint power indicates the moment of force is acting in the opposite direction to the joint motion, with the muscle absorbing power during an eccentric contraction (Winter, 1991)

Joint power:

$$\textbf{\textit{Power}} = \boldsymbol{M * \omega} \qquad \textbf{(equation 5)}$$

where M = net joint moment; ω = joint angular velocity

The following joint power definitions were used:

Positive	**Negative**
Generation	Absorption

3.19. Gait variables

The following section highlights the variables that were analysed for the level and stair walking trials. Data for the amputees were presented for the affected and intact limbs separately. Variables used to identify the biomechanical differences between amputee fallers and non-fallers included temporal-spatial parameters such as walking speed (m/s), step length (normalised to % body height) and support times (% gait cycle). Joint mobility was examined dynamically by comparing peak ankle and knee angles and range of motion in the sagittal plane; peak hip angles and range of motion in the sagittal and frontal planes (degrees).

Peak anterior-posterior and vertical GRF (N/kg) were analysed. The GRF data, joint moments and powers were normalised for body weight. The medial-lateral GRF component was not analysed due to its inherent variability. Load rate was calculated as the positive slope of the force vs. time curve from initial foot contact to the first peak vertical force; whilst decay rate was the negative slope from the second peak vertical force until toe off. Both had units of N/kg/s. Joint moment and power (especially absorption) profiles have previously revealed differences in older fallers and non-fallers (Lee and Kerrigan, 1999). In the walking experiments, lower-limb joint moments (Nm/kg) were compared in the sagittal plane to determine if fallers displayed reduced gait mechanics vs. non-fallers. Peak power bursts for each mechanical joint power phase (W/kg) were calculated for the ankle, knee and hip joints in the sagittal plane and compared between the fallers and non-fallers.

Within-subject movement variability during level walking was determined by calculating the coefficient of variation for temporal-spatial variables over ten gait cycles. Coefficient of variation was defined as the ratio of the SD to the mean value and was expressed as a percentage (%). Variables included step length and frequency and time in stance, swing and double support for both limbs.

3.20. Postural control tests

The following section explains the NeuroCom Smart Equitest system that was used to measure postural responses to dynamic perturbations. Details about the two tests that were specific to the postural experiment are given in the methods section of the associated chapter. The Smart Equitest is composed of a steel frame incorporating a dual force plate system capable of translating in the forwards-backwards direction and causing rotation about the ankle joint in the sagittal plane when the support surface is sway-referenced (Figure 3.7). Each force plate is 23 X 46cm and the two force plates are connected by a pin joint. Force is measured by four transducers, mounted symmetrically on a central plate, and a fifth transducer bracketed to the central plate below the pin joint. This allows for forces to be measured for the right and left foot, separately. The four transducers measure the vertical forces applied to the force plate, while the central transducer measures shear forces in the forwards-backwards direction parallel to the floor. The visual surround is capable of rotating in the forwards-backwards direction with a maximum velocity of 15°/s and is referenced to the smoothed center of force position (e.g. sway-referenced). The force sampling frequency was set at 100Hz.

Figure 3.7. NeuroCom Smart Equitest system with sway-referenced moving surround and support surface. The participant is wearing a harness to prevent them from falling during the tests.

The participant's data were entered (age, body height, and mass) into the NeuroCom software so that current results were compared to age-matched normative data sets of the NeuroCom software and dynamic perturbations were scaled according to a person's height. The NeuroCom normative data sets were collected for 3 age groups (20–59, 60–69, and 70–79 years) using clinically asymptomatic participants with no balance disorder. A total of 121, 26, and 29 participants were included in the normative data set for each age group, respectively. Inclusion criteria for the normative data included no medical diagnosis affecting balance (past or present); no use of medication that affects balance or the CNS; no symptoms of dizziness and/or lightheadedness; no known vestibular and/or neurologic disorder; no psychological problems, including depression; no history or unexplained falls within the previous 6 months; normal vision with or without glasses. Men and women were included in each age group. For participants in the current study who were older than 79 years, their data was compared with the 70-79 years age group.

3.21. Psychological instruments

Two psychological instruments (SF-36 and MFES) were used to measure overall quality of life and balance confidence. The other psychological instruments (FES, ABC) were discussed with physiotherapists when the use of outcome measures was investigated in current amputee physiotherapy practice.

3.21.1. Short Form (SF-36)

The SF-36 Health Survey (Appendix F) is a general health measure composed of 36 questions (Ware Jr and Sherbourne, 1992) that measures quality of life. It is a multi-item scale that measures eight different health attributes with 2 to 10 items in each: 1) physical functioning, 2) role limitations due to physical health problems, 3) bodily pain, 4) general health, 5) vitality, 6) social functioning, 7) role limitations due to emotional problems and 8) mental health (psychological well-being). The SF-36 is easy to use, can be self-administered and relatively quick. The scales use Likert's method of summated ratings. The SF-36 items cover a large spectrum of tasks that affect a person's roles, such as limitations in work or typical daily tasks; reducing the amount of time typically dedicated to work or other daily tasks and difficulty performing work or other activities.

The SF-36 evaluates *Physical functioning* across different functional levels on self-care activities, including walking moderate distances; lifting and carrying groceries; bending, kneeling and stooping; and climbing stairs. *Role limitations* items cover both physical and emotional problems. There are two SF-36 items on *Bodily pain* that evaluate the intensity of bodily pain/discomfort and investigate how pain interferes with normal activities. *General health* is measured using five items rating health on a continuum from excellent to poor. *Vitality* refers to a person's energy levels and fatigue and is measured on four items. *Social functioning* reflects a person's social activities, both in terms of frequency and enjoyment with others, and is measured on two items. *Mental health* is assessed using five items from each of the four main mental health dimensions (anxiety, depression, loss of emotional control, and psychological well-being). When administered to an older population, Lyons *et al.* (1994) found good internal consistency with Cronbach's alpha and greater than 0.8 (ranging from 0.83 to 0.94) for each health parameter. These authors also found the SF-36 was able to distinguish between older persons with vs. without markers of poor health (Lyons *et al.*, 1994).

3.21.2. Modified Falls Efficacy Scale (MFES)

The MFES (Appendix F) was developed as a more sensitive measure of fear of falling than the Falls Efficacy Scale (Hill *et al.*, 1996), originally created by Tinetti *et al.* (1990). The MFES is comprised of 14 items in total and has proven useful in detecting early stages of fear of falling in relatively active, community-dwelling older individuals. It includes ten of the original activities in the FES such as dressing, preparing meals, bathing, rising from a chair and bed, walking inside the home, reaching into cabinets, light housekeeping and doing simple shopping. The additional 4 items were added to reflect a person's confidence in performing outdoor activities, such as gardening, crossing roads and using public transport. Participants were asked to rate their confidence at performing the activities without falling on a visual analogue scale of from 0 (not at all confident) to 10 (completely confident) (Hill *et al.*, 1996). Hill *et al.* (1996) found that the mean score for healthy older people was 9.76 ± 0.32 compared with a mean score of 7.69 ± 2.21 for previous fallers. Cronbach's alpha for internal consistency was 0.95 and ICCs were high for test-retest reliability. Two Factors (indoor- vs. outdoor-type activities) were identified that could account for 75% of the sample variance. However, one limitation of the MFES, and other questionnaires assessing fear of falling, is that they cannot establish a causal effect (Hill *et al.*, 1996). It is unclear whether poor balance and previous falls lead a person to develop fear of falling and avoid "high-risk" activities or whether fear of falling leads to activity avoidance and thereby impaired balance. Moreover, high scores could reveal a ceiling effect.

3.21.3. Falls Efficacy Scale (FES)

The FES was developed by Tinetti *et al.* (1990) to measure a person's self-reported confidence in performing ten daily activities, such as getting in and out of a chair/bed, walking around the house, reaching into closets and taking a bath or shower, without falling. The FES is reportedly reliable (r = 0.71) and has good construct validity and predictive validity for measuring fear of falling. The FES does not assess a person's confidence in undertaking outdoor-type activities.

3.21.4. Activities-specific Balance Confidence (ABC) Scale

The ABC Scale (Myers *et al*., 1998) was developed from the FES (Tinetti *et al*., 1990) in an attempt to assess loss of balance confidence in older individuals with a higher level of functioning. The ABC is made up of 16 items, each scored on a 0-100% response continuum. Participants are asked to rate their confidence that they will not lose balance or become unsteady during specific activities. The ABC Scale has good reliability ($r = 0.92$, $p < 0.01$) and high internal consistency (Cronbach's alpha of .96). The ABC correlated well with the FES ($r = 0.84$, $p < 0.001$) and was found to be a better discriminator of high vs. low mobility individuals.

3.22. Functional outcome measures

The following sections describe the functional outcome measures referred to in this thesis. The Timed Up and Go (TUG) test was used to understand the relationship between function and quality of life and balance confidence. The other measures were discussed with physiotherapists during the audit. The first two measures (TUG test, and Tinetti balance test) are not disease-specific whereas the L-test, SIGAM and LCI are amputee specific.

3.22.1. Timed Up and Go (TUG) test

The TUG test is a modified version to the Get-Up and Go test developed by Mathias and colleagues (1986). The Get-Up and Go and TUG tests have been used to assess function in older people. The activity-based test incorporates basic mobility skills such as rising from a seated position, walking three metres, turning 180°, returning to the chair and resuming a seated position. The test has good agreement in time scores between-raters (ICC 0.99) and within-raters (ICC 0.99). The TUG test has good intra-rater and inter-rater reliability ($r = 0.93$ and $r = 0.96$, respectively) for older individuals with lower-limb amputation (Schoppen *et al.*, 1999).

In the current study, participants were asked to complete the TUG test three times and an average value was then calculated. A standard armchair was used, with a seat height of 46 cm and arm height of 65 cm, as recommended by Podsiadlo and Richardson (1991).

3.22.2. L-test of functional mobility

The L-test is a modified TUG test for lower-limb amputees that was developed to be functionally more demanding and, at the same time, practical. It can be easily administered by therapists in a clinical setting. The activity-based test comprises of 2 transfers and 4 turns and covers a total distance of 20 metres. It consists of rising from a seated position, walking 3 metres, turning 90°, walking 7 metres, turning 180°, returning 7 metres, turning 90° walking 3 metres and resuming a seated position. The ICC for intra-rater reliability was .97 (0.93-.098: 95% CI for agreement) and for inter-rater reliability was .96 (0.94-0.97: 95% CI for agreement). The L-test correlated well with other outcome measures: the TUG test (Pearson r = 0.93), 2-minute walk test (Pearson r = -.86), 10-metre walk test (Pearson r = 0.97), and ABC Scale (Pearson r = -.48).

3.22.3. Tinetti Balance and Gait Assessment Tool (aka Tinetti's Performance Oriented Mobility Assessment)

The Tinetti test (Tinetti, 1986) is also activity-based and administered by a therapist. The test measures a patient's balance and gait during manoeuvres that emphasise stability and gait at a preferred and rapid pace. It is scored using a 3-point ordinal scale. The maximum score is 28, a score of 19-24 presents a moderate risk of falls and a score of 19 or less represents high risk (Tinetti, Williams and Mayewski, 1986). The Tinetti test has good inter-rater reliability across many raters of varied experience (fair to excellent k coefficients (.40-1.00) across all raters for the 8-item balance test) (Cipriany-Dacko *et al.*, 1997).

3.22.4. SIGAM (Special Interest Group in Amputee Medicine) mobility grade

The SIGAM amputee mobility measure for was developed by Ryall *et al.* (1993) based on the Harold Wood Stanmore grades. The SIGAM is a single-item scale made up of six clinical grades (A-F) assessing amputee mobility, although the questionnaire itself is not observational. Amputees complete a self-report questionnaire that assists in assigning the clinical grade. The questionnaire assesses if a person wears their prosthesis, and then rates their performance indoors and outdoors with the prosthesis. The SIGAM scale has good validity and reliability of measuring mobility in lower-limb amputees. ICC alpha of 0.79 and reproducibility kappa values (0.70-1.0; overall kappa coefficient of 0.86) were generally good.

3.22.5. Locomotor Capabilities Index (LCI)

The LCI is an amputee-specific instrument that forms part of the Prosthetic Profile of the Amputee questionnaire. It consists of 14 items that measure the self-reported locomotor capabilities of the amputee wearing their prosthesis when performing basic (7 items) and more advanced (7 items) daily activities. The LCI also has good internal consistency (Cronbach's alpha of 0.95) and excellent test-retest agreements (intraclass coefficient = 0.80) on the total LCI score, indicating there is a high level of agreement on the two scores.

3.23. Statistical analysis

One of the aims of the following experiments was to compare the biomechanical and psychological differences between fallers and non-fallers. The existing literature has already documented the biomechanical characteristics that distinguish transtibial amputees from able-bodied individuals during level walking (Donker and Beek, 2002; Isakov *et al.*, 1996; Jaegers *et al.*, 1995; Nolan *et al.*, 2003; Sanderson and Martin, 1997; Winter and Sienko, 1988), stair walking (Schmalz *et al.*, 2007; Powers and Boyd, 1997; Yack *et al.*, 1999) and postural control (Hermodsson *et al.,* 1994; Isakov *et al.* 1992). Therefore, the statistical analyses that were carried out in this thesis specifically focused on the differences between amputee fallers and non-fallers and control fallers and non-fallers.

Statistical tests used for data analysis from the experiments in this thesis were not decided *a priori*. On certain occasions, statistical analysis could not be conducted because gait patterns were so varied between subjects and group numbers were reduced. All data were initially screened for outliers and normality (Q/Q plots) and tested for homogeneity of variance using the appropriate test. In the instance of violation of homogeneity of variance, corrected statistical values were used. If significant differences were found overall, post-hoc tests were conducted, where appropriate. SPSS v15.0 (SPSS Inc., Chicago, IL) for Windows was used for all statistical analysis. Significant values were considered $p < 0.05$. The details of statistical methods used in each experiment are listed in the methods section of each associated chapter.

3.24. Conclusions

This chapter has described the tools that were used in a series of six studies to achieve the aims of this thesis, which were to understand the biomechanical and psychological differences between fallers and non-fallers in order to assist the clinical recommendations for amputee rehabilitation. These tools were selected based on a review of the current literature in the relevant areas of gait and balance, falls, rehabilitation and quality-of-life outcome measures. The next six chapters detail the application of these tools in biomechanical and psychological studies.

CHAPTER FOUR - THE USE OF OUTCOME MEASURES IN OUTPATIENT AMPUTEE REHABILITATION IN ENGLAND

4.1. Introduction

The use of subjective and objective outcome measures is recommended to monitor patient progress during amputee rehabilitation (BACPAR, 2003). Studies to date that have examined outcome measures have mostly focused on rehabilitation as a whole, and their results encompass services for disabilities such as spinal cord injury, musculoskeletal and neurorehabilitation (Turner-Stokes, 2001; Skinner and Turner-Stokes, 2006). While the underlying standards of rehabilitation services for different populations may be similar, the demands of prosthetic rehabilitation are complex. There is no consensus on the best outcome measures for the amputee population (Deathe *et al*., 2002).

There are a plethora of outcome measures, from self-report to observational, multidimensional to performance-based tools, amputee-specific to generic that are used to assess functional and mental health status in the prosthetic rehabilitation setting. However, there is no documented evidence to determine which tools are most widely used by physiotherapists in prosthetic centres, how frequently they are used and what the data are used for. Currently, clinicians are unaware how different centres monitor patient progress and how amputee treatment varies across England. The purpose of this study was to conduct an audit that would investigate if falls were monitored and outcome measures were used in amputee rehabilitation in England. The objectives were to determine whether physiotherapists were monitoring falls incidence among their patient population and whether physiotherapists used functional and psychological outcome measures to inform patients' discharge. It was hypothesised that falls would not be monitored regularly in prosthetic centres. It was also hypothesised that physiotherapists would use functional outcome measures more than psychological measures, but that there would be no consistency of use across England and that outcome measures would not inform a patient's discharge.

4.2. Methods

4.2.1. Design and sample

The audit consisted of six questions and was developed to gather data on the use of outcome measures in prosthetic rehabilitation in England. The contact details of each centre were obtained from the directory of Disablement Service Centre (DSC) limb centres in England. The website currently holds the contact details for 34 NHS prosthetic limb centres in England. In order to avoid multiple responses from each centre, the questions were addressed to the lead physiotherapist because they meet patients on a regular basis and play a key role in the patients' rehabilitation programme. The audit was administered via a semi-structured telephone interview during November 2007 to January 2008. Once the telephone interview was complete, member checking was conducted via e-mail (Lincoln, 1985). Responders were given a written report of their responses and were asked to confirm that their statements were correct and invited to make any further comments. An individual or their centre was never identified by name or location in any of the findings of this audit. The study was approved by the Departmental ethics committee.

4.2.2. Measurement

The questions were aimed at amputee rehabilitation as a whole and did not distinguish level of amputation (transtibial vs. transfemoral, unilateral vs. bilateral). All the physiotherapists interviewed worked with outpatients and they were asked to answer the questions from an outpatient perspective. The first half of the audit was to understand if, and how frequently, physiotherapists used outcome measures to inform patient outcome. The first question established whether the physiotherapists worked primarily with amputee inpatients, outpatients or both. The second question investigated whether the lead physiotherapist and their team used functional and/or psychological outcome measures as part of rehabilitation. If the respondent used an outcome measure, in the third question they were asked to indicate whether they used the following measures and how frequently: Timed Up and Go (TUG) test (Podsiadlo and Richardson, 1991), timed 10-m, 20-m, 40-m or 100-m walk test, 2-minute walk test, L test of functional mobility (Deathe and Miller, 2005), chair to stand test, SIGAM mobility grade (Ryall *et al*., 2003), Locomotor Capabilities Index (LCI) (Gautheir-Gagnon and Grisé, 1994), Tinetti Balance Assessment Tool (Tinetti *et al*., 1988) and/or any other functional outcome measure. The fourth question related to psychological outcome measures and respondents were asked to identify whether they used the following measures and how frequently: SF-12® or SF-36® (Ware *et al*., 1993), the Activities-specific Balance Confidence (ABC) scale (Powell and Myers, 1995), the Falls Efficacy Scale (FES) (Tinetti *et al*., 1990), the Modified Falls Efficacy Scale (MFES) (Hill *et al*., 1996) and/or any other outcome measure. The second half of the audit was to examine discharge criteria and whether physiotherapists monitored falls incidence. The fifth question asked physiotherapists to describe the criteria used to determine a patient's discharge from active rehabilitation. The final question asked physiotherapists to indicate whether they monitored falls among their patients and how frequently. The answers were grouped according to the main emerging themes for discharge. The telephone conversations generally took 10-15 minutes to complete. Responses to the audit were analysed using descriptive statistics, such as percentages and frequency of responses.

4.3. Results

Responses were received from 29 of 34 Disablement Service Centres in England, giving an overall response rate of 85%. Physiotherapists from the other 5 centres were unable to be reached despite several follow-up calls.

4.3.1. Discharge criteria and falls

Lead physiotherapists were asked to describe their main criteria for determining a patient's discharge. A total of 19 respondents (66%) stated that achievement of patient goals was an important factor for discharge. Other important criteria for discharge such as patient safety and function (34%) and the patient reaching a plateau in their prosthetic performance (34%) were mentioned. Three physiotherapists (10%) stated that they determined discharge by following BACPAR guidelines. Surprisingly, only one respondent (3%) actually referred to a patient's improvement on outcome measures as criteria for discharge and two respondents (7%) mentioned the patient's ability to manage falls. A total of 20 respondents (69%) reported that they would reconsider a patient's readiness for discharge from active physiotherapy treatment. Usually, this was associated with a change in the patient's circumstances, or if the physiotherapist could see that the patient was not coping with their prosthesis.

A large number of physiotherapists did not monitor falls incidence on a formal basis. Only two respondents (7%) formally monitored the number of falls within their amputee patient population as part of a DSC limb centre audit, while 9 (31%) did not ask patients about their falls history at all. However, 18 physiotherapists (62%) monitored falls on an informal basis. This referred to asking patients if they had recently fallen but not including the data as part of a greater DSC limb centre audit. Of the 18 who monitored falls informally, half only asked on the first visit and thereafter assumed the patient would inform them of any falls.

4.3.2. Functional and psychological outcome measures

A total of 22 physiotherapists (76%) who were interviewed worked in both inpatient and outpatient centres. The remainder worked in outpatient centres only. A total of only 11 respondents (38%) used some type of outcome measure to inform a patient's discharge. Those respondents all used a functional outcome measure (either performance-based or observational). However, twenty-three respondents (79%) used functional outcome measures at some point during a patient's rehabilitation treatment. Of those who used functional outcome measures, the most popular outcome measures included the SIGAM mobility grade and the LCI (Figure 4.1). At least eight physiotherapists (35%) reported using the TUG test and 4 physiotherapists (17%) used the timed 10-m walk test. Eight of 23 responders (35%) used some other functional outcome measure (Table 4.1).

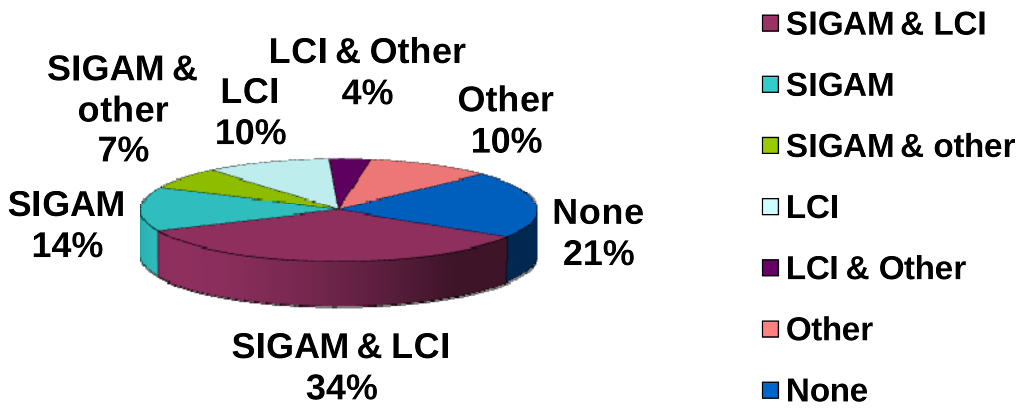


Figure 4.1. Use of functional outcome measures in amputee rehabilitation programmes in DSC limb centres in England

LCI; Locomotor Capabilities Index

There was no consensus on frequency of administration of the functional outcome measures among the physiotherapists interviewed. Overall, physiotherapists reported using the functional outcome measures at various intervals. Of the 23 respondents who indicated using some type of outcome measure during the patient's treatment, 13 (57%) respondents used the outcome measure at discharge. Frequency of use varied from every visit, to a weekly or monthly basis, to 3-month intervals.

A total of 25 of 29 respondents (86%) did not use any form of psychological outcome measure as part of their prosthetic rehabilitation treatment. Two individuals (7%) stated that they would like to measure psychological outcomes, but that they received no psychological input in their centre; another two (7%) reported that their centre was currently looking into using psychological outcome measures. Six respondents (21%) said that physiotherapists did not monitor psychological outcomes because their centre had a psychologist or counsellor who handled it. Only four respondents (14%) said that they used psychological measures, although there was no consensus how frequently these were administered. The most popular psychological measures were the SF-12® or SF-36® and the Hospital Anxiety and Depression (HAD) scale (Zigmond and Snaith, 1983). No respondents reported using any self-efficacy and/or balance confidence questionnaires.

Table 4.1. Frequency of outcome measures used in amputee rehabilitation in DSC limb centres in England. Responders could have indicated more than one response or outcome measure.

	Number of responders (n=)	**Percentage of those who used outcome measures (n/23)**	**Percentage of total responders (n/29)**
Observational measures			
SIGAM mobility grade	16	70%	55%
Berg balance scale	3	13%	10%
Tinetti balance assessment tool	1	4%	3%
Mobility performance measures			
TUG test	8	35%	28%
Timed 10-m walk test	4	17%	14%
6-min walk test	2	9%	7%
2-min walk test	1	4%	3%
L test of functional mobility	1	4%	3%
Psychological measures			
MMSE	1	4%	3%
Self-report functional measures			
LCI	14	61%	48%
SCS	2	9%	7%
Houghton scale	1	4%	3%
Self-report quality of life measures			
SF-12® or SF-36®	2	9%	7%
HAD scale	2	9%	7%
PEQ	1	4%	3%

TUG, Timed Up and Go Test; MMSE, Mini Mental State Exam; LCI, Locomotor Capabilities Index; SCS, Socket Comfort Score; SF-12® and SF-36®, Short-Form-12® and Short-Form-36® Health Surveys; HAD scale, Hospital Anxiety and Depression Scale; PEQ, Prosthetic Evaluation Questionnaire

4.4. Discussion

4.4.1. Discharge criteria and falls

The results from this audit showed that there was no clear consensus about criteria for discharge among the physiotherapists interviewed. However, the majority of responders practiced patient-focused outcome measures through the use of goal-setting and with the emphasis on safety and function. Also, the majority of respondents would reconsider discharging a patient if their circumstances had changed, emphasising the notion that amputee rehabilitation was patient-focused. However, given the recent attention and awareness of falls incidence among older individuals and lower-limb amputees, it was surprising that so few respondents specifically indicated falls prevention and management as important criteria for discharge. The BACPAR clinical guidelines discuss coping strategies following falls by encouraging patient access to falls prevention education and coping strategies if a fall were to occur.

The results from this audit also indicated that physiotherapists monitored falls incidence among amputee patients informally and infrequently. Interestingly, nine respondents (31%) assumed patients would inform the physiotherapists if they had fallen, a method that has been reportedly unreliable at detecting fallers (Hancock *et al*., 2007). In order to improve and standardise amputee treatment across centres, it should be recommended that physiotherapists ask patients if they had fallen at every visit and that this information was formally recorded within their treatment notes and accessible by the multidisciplinary team at the DSC limb centre. Understanding what activities were performed prior to the fall, whether the prosthesis was being worn, whether the patient remembered how to get up and off the floor and was able to use their falls plan, is vital. Falls education and prevention programmes can only be evaluated if falls incidence is monitored on a formal basis.

4.4.2. Functional and performance-based outcome measures

The aim of this audit was to understand the use of outcome measures in the current prosthetic rehabilitation setting, to understand how physiotherapists determined a patient's readiness for discharge and to determine how physiotherapists monitored falls incidence among their amputee patients.

Section three in the clinical guidelines provided by BACPAR (2003) addresses assessment procedures in amputee rehabilitation and states that "relevant and validated outcome measures should be used and recorded to evaluate change" (BACPAR, 2003, Pp. 24). However, the guidelines make no recommendations regarding the appropriate types of outcome measures, or how frequently they should be documented. The findings from this audit confirmed that there was no consensus among lead physiotherapists in DSC limb centres about the use of outcome measures, both in terms of identifying the most relevant tools and frequency of use. The findings showed that the SIGAM mobility grade and LCI were the most popular outcome measures. It was an encouraging result that those physiotherapists who used outcome measures were using amputee-specific tools. Compared to generic measures, disease-specific instruments were reportedly more likely to show improvements in response to treatment (Zigmond and Snaith, 1983). Deathe *et al*. (2002) reported that more than 46% of the centres they surveyed used the Functional Independence Measure (FIM™).

The FIM™ is a generic measurement tool, applicable across disabilities, and aimed at identifying the level of independence on motor and cognitive scales. However, it has reported insensitivity and high ceiling effects when used as an outcome measure in prosthetic rehabilitation. The SIGAM has reported good inter-rater reliability with Cohen's kappa values of 0.7 –1.0 and ICC α of 0.79 (Ryall *et al*., 2003). The LCI has good content and construct validity scores and good reliability scores with a Cronbach's alpha of 0.95 (Ryall *et al*., 2003). Both are amputee-specific mobility tools used to evaluate level of disability and locomotor independence when performing daily activities with a prosthesis. However, they are reportedly subject to ceiling effects and therefore, are limiting in discerning higher levels of mobility (Gauthier-Gagnon and Grisé, 2001). Also, despite being the most popular choice among physiotherapists in England, they do not actually measure the performance of daily activities, making it difficult to identify more subtle changes and improvements in mobility. Furthermore, the SIGAM primarily quantifies an individual based on their walking performance. The use of outcome measures based on performance of other functional activities is recommended.

The most popular performance-based outcome measures were the TUG and 10-m walk tests, but surprisingly the findings revealed that only 28% and 14% of all responders used them, respectively. The TUG test has been referred to as a realistic mobility assessment (Bischoff *et al*., 2003) because it incorporates activities such as transferring in and out of a chair, walking, and turning 180° necessary for independent ambulation. The TUG test is reliable inter-rater and intra-rater (ICC scores of 0.99) and correlates with patient's functional capacities balance, gait and functional abilities (Podsiadlo and Richardson, 1991). It has also been shown to predict falls in elderly community dwelling individuals by identifying those unable to complete the test, and who are also at greatest risk for falling (Large *et al*., 2006). The majority of falls in older individuals occur during ambulation (56%) and transferring (9%) (Talbot *et al*., 2005). Compared with the 10-m walk test, the TUG is a comprehensive test that does not require much space or specialised equipment, can be done in little time, is easy to administer and yet could be performed in between parallel bars or with a walking aid, while still covering a total distance of 6 metres. Therefore, it could be a more indicative functional outcome measure than just recording the time taken to walk 10 metres.

Deathe and Miller (2005) reported that the TUG test showed a ceiling effect for older people who are more physically fit and younger people with amputation. These authors developed and validated the L test of functional mobility to help determine an amputee's ability to walk with their prosthetic device during transferring (to and from a seated position), turning 90° towards both limbs and walking a total distance of 20 metres. The L test had high concurrent validity correlations with other walk test measures and the ICC for inter-rater and intra-rater reliability were 0.96 and 0.97, respectively (Deathe and Miller, 2005). The time taken to complete the L test was generally longer than the TUG test especially among the older, frailer amputee population. The authors concluded that the L test may assess transition from room ambulation to community ambulation (Deathe and Miller, 2005).

Based on BACPAR's evidence based clinical guidelines and similar research by Deathe *et al.* (2002) and Skinner and Turner-Stokes (2006), the regular use of a standardised, performance-based outcome measure as part of clinical practice in amputee rehabilitation is highly recommended. Depending on a patient's activity level, either the TUG test or the L test could be used in conjunction with amputee-specific mobility scales, such as the SIGAM grade or the LCI. Frequent monitoring of a patient's performance would allow physiotherapists to determine the impact of rehabilitation, and more importantly, would serve as an easy indicator for those patients at greatest risk of falling. It is believed the results of functional outcome measures are essential in informing patient progress and discharge.

4.4.3. Psychological outcome measures

Fear of falling has been linked with activity avoidance, limited prosthetic function and therefore decreased quality of life (Miller *et al.*, 2001). Powell and Myers (1995) argued that it was important to explore reasons for reported activity avoidance and suggested self-efficacy scales should be incorporated into the initial screening process regarding balance confidence. Understanding a patient's fear of falling would be particularly important among those at greater risk for falls. No responders in this audit used self-reported confidence scales, such as the ABC scale, the FES and the MFES and the majority were not familiar with the questionnaires. Such scales are quick and easy to administer and easy to score and interpret. Physiotherapists in DSC limb centres should become familiar with some of the balance confidence scales and use them to monitor a patient's measure of self-efficacy as their functional performance improves during amputee rehabilitation. This would be particularly relevant if the patient has recently fallen or shows a particularly high risk for falling.

The BACPAR clinical guidelines emphasise that patients should understand how and where to seek psychological advice and support following their amputation and that all advice given to the patient should be recorded. The findings from this audit showed that only 35% of physiotherapists used psychological outcome measures with their patients or referred patients to a psychologist or counsellor. Furthermore, it was found that psychological tools were not used to monitor patient progress or inform physiotherapy discharge. The majority of physiotherapists that were interviewed were unaware of questionnaire scoring systems and found it difficult to interpret results. From these results, it is not known if the DSC limb centres offered patients psychological counselling or made referrals. However, it should be recommended that physiotherapists have access to, and incorporate the use of, simple balance confidence scales into functional physiotherapy treatment.

4.5. Conclusion

Amputee rehabilitation is a complex task. Patients vary according to age, cause of amputation, general health status and psychological aspects. A treatment programme that is appropriate for one patient may not be appropriate for another patient. It is acknowledged that the suggestions may not be suitable for all patients in amputee rehabilitation. The results of this audit supported the hypothesis that physiotherapists did not monitor falls regularly and this highlighted an important shortcoming in amputee rehabilitation. Physiotherapy guidelines should make recommendations regarding a falls monitoring system in order to evaluate the effectiveness of falls education and falls prevention programmes. This study also supported the hypothesis that there was no clear consensus regarding the use of outcome measures in prosthetic rehabilitation despite the fact that clinicians, practitioners and researchers advocated their use. While performance-based measures may be more suitable among younger, traumatic amputees, as well as older, more mobile amputees, they may not be appropriate or safe for amputees who use their prosthesis for aesthetic and/or therapeutic reasons. Based on the findings of this audit, it is believed that practitioners working with amputees reach some level of agreement about the most appropriate outcome measures for patients who are expected to achieve a level of independent mobility. This would encourage the distribution of evidence-based best practice.

CHAPTER FIVE - GAIT PATTERNS IN TRANSTIBIAL AMPUTEE FALLERS VS. NON-FALLERS: BIOMECHANICAL DIFFERENCES DURING LEVEL WALKING

5.1. Introduction

Transtibial amputations account for around half of all lower-limb amputations in the USA and the UK, with the primary causes being vascular disease and trauma (Muilenburg and Wilson, 1996; NASDAB, 2007). The loss of the plantarflexor muscles, combined with muscle weakness and the mechanical limitations of the prosthetic foot, predispose transtibial amputees towards an increased risk of falling compared with age-matched, able-bodied individuals (Miller *et al*., 2001). Kulkarni *et al.* (1996) found that 60% of amputees advised that falling affected their daily life, work, leisure and confidence.

Gait disorders and an increased risk of falling in older people are linked with reduced lower-limb strength and joint range of motion, muscle tightness and postural instability (Kerrigan *et al*., 1998; Kerrigan *et al*., 2001; Lee and Kerrigan, 1999). Similar relationships may be apparent for amputees, although this has never been tested. Variability in temporal-spatial gait parameters has been linked with falls (Hausdorff *et al*., 2001; Heiderscheit, 2000) and measures of gait variability are predictive of future falls (Hausdorff *et al*., 2001).

In order to understand how gait disorders could be useful in predicting falls in lower-limb amputees, it is important to understand whether biomechanical differences exist between fallers and non-fallers. The main aim of this study was to compare the gait patterns of fallers and non-fallers during level walking in transtibial amputees and able-bodied participants. It was hypothesised that amputee fallers and control fallers would have reduced mobility compared to their non-faller counterparts. A second hypothesis was that fallers would have reduced peak joint moments and powers at the ankle, knee and hip. These hypotheses were based on previous findings that falling is associated with lower-limb weakness and joint flexibility (Kerrigan *et al*., 1998; Kerrigan *et al*., 2001; Lee and Kerrigan, 1999). A third hypothesis was that fallers would have increased variability in the temporal-spatial parameters. This hypothesis was based on research that has reported greater variability in elderly fallers (Hausdorff *et al*., 2001; Heiderscheit, 2000).

5.2. Methods

5.2.1. Participants

Eleven transtibial amputees (mean ± SD: age 56 ± 16 yr; height 1.74 ± 0.14 m; body mass 76 ± 16 kg; time since amputation 6.7 ± 9.3 yr) and nine age-matched able-bodied participants (mean ± SD: age 61 ± 16 yr; height 1.73 ± 0.14 m; body mass 80 ± 13 kg) were recruited for this study. All participants gave written informed consent to take part in this study. Individual participant details according to falls history can be found in Table 3.1.

5.2.2. Experimental protocol

Each participant completed the level walking trials on one visit to the Human Performance Laboratory at the University of Hull. All participants were made aware of the protocol and were reminded they could rest at any time. Participants changed into their shorts and reflective markers were affixed onto the lower-limbs as described in section 3.10. Data collection was completed once participants completed the walking trials described in section 3.14.

5.2.3. Data analysis

All data were analysed as explained in sections 3.16, 3.17, and 3.18. Data for the affected limb was only compared between two groups: the amputee fallers and non-fallers. Data for the intact limb was compared across four groups: amputee and control fallers and non-fallers. In the control group, the data from the left and right limbs were averaged. The variables that were selected for analysis included temporal-spatial, peak joint kinematic and joint ROM, peak GRF data, load and decay rates, peak joint moments and power bursts as described in section 3.19.

Within-subject movement variability was determined by calculating the coefficient of variation for temporal-spatial variables over ten gait cycles. Coefficient of variation was defined as the ratio of the SD to the mean value and was expressed as a percentage (%). Variables included step length and frequency and time in stance, swing and double support for both limbs.

5.2.4. Statistical analysis

For the affected limb, independent sample t-tests were used to determine if falls history had a significant effect on the specific gait variables. Levene's test for equality of variances was used to assess homogeneity. In the instance of violation of homogeneity of variance the corrected t-value was used. For intact limb variables, a one-way

ANOVA was used to compare the amputee groups and the two control groups. If significant differences were found overall, Fisher's least significant difference (LSD) post-hoc test was used to determine whether the differences existed between the amputee fallers vs. non-fallers and control fallers vs. non-fallers. No other significant differences between groups (e.g. amputee non-fallers vs. control fallers) were investigated in this study. The alpha level for significance was set a priori at 0.05.

5.3. Results

5.3.1. Participant demographics

There were no significant differences for age ($F_{(3,16)}$ = 1.36, p = 0.29), height ($F_{(3,16)}$ = 0.82, p = 0.50), or body mass ($F_{(3,16)}$ = 0.18, p = 0.91) between the groups tested. For the amputee groups, there was no significant difference in time since amputation (t_9 = 1.33; p = 0.28).

5.3.2. Temporal-spatial parameters

All participants were able to complete the level walking task without experiencing a trip, slip or fall. The temporal-spatial gait parameters are shown in Table 5.1.

Table 5.1. Amputee vs. control fallers and non-fallers mean (SD) temporal–spatial parameters during straight level walking

	Amputee				Control	
	Non-faller		Faller		Non-faller	Faller
Walking speed (m/s)	1.07 (0.20)		1.19 (0.35)		1.52 (0.32)	1.05 (0.24)
Double support (%)	30 (4)		27 (7)		21 (3)	27 (6)
	Affected	Intact	Affected	Intact	Non-faller	Faller
Step length (%BH)	37 (5)	35 (7)	38 (9)	37 (7)	45 (6)	37 (7)
Step frequency (steps/min)	104 (10)	106 (8)	105 (6)	106 (9)	111 (6)	103 (8)
Stance (%)	63 (3)	66 (3)	62 (3)	65 (4)	61 (2)	63 (3)

The one-way ANOVA found no significant differences in temporal-spatial variables except intact stance duration ($F_{(3,16)}$ = 3.12, p = 0.05). Post-hoc analysis revealed that the control non-fallers spent significantly less time in stance compared to the amputee non-fallers (p=0.01) and fallers (p=0.04).

5.3.3. Joint kinematics

The hip, knee and ankle flexion angles are shown in Figure 5.1 and hip abduction in Figure 5.2. Peak joint kinematics are reported in Table 5.2. No significant differences were found between the fallers and non-fallers for any joint flexion variable except ankle angle at toe off. The control fallers showed significantly less ankle plantarflexion at toe off compared to the control non-fallers ($p=0.01$).

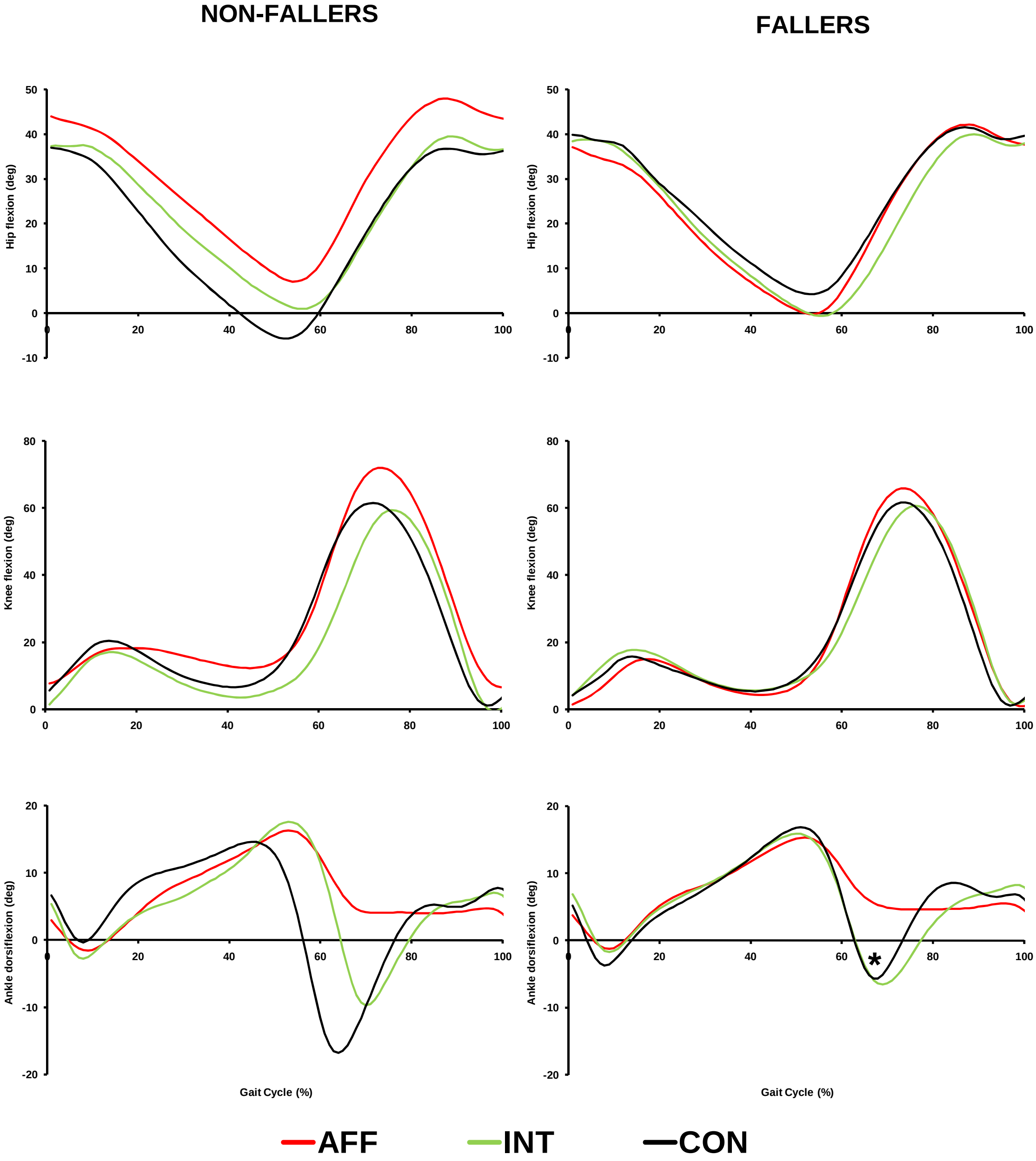


Figure 5.1. Ankle, knee and hip sagittal plane flexion angles during level walking for the fallers and non-fallers. Data are averaged according to limb: affected (AFF) (red line); intact (INT) (green line); control (CON) (black line). The gait cycle is initiated and terminated with foot contact.

Flexion and dorsiflexion are positive.

* Indicates significant difference between control fallers and non-fallers (one-way ANOVA for the intact limb)

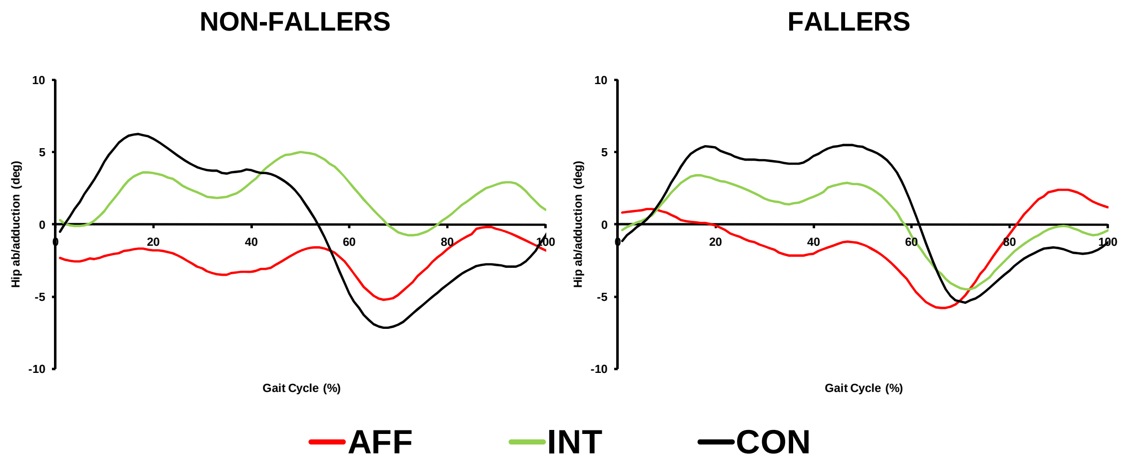


Figure 5.2. Hip ab/adduction angles during level walking for the fallers and non-fallers. Data are averaged according to limb: affected (AFF) (red line); intact (INT) (green line); control (CON) (black line). The gait cycle is initiated and terminated with foot contact. Adduction is positive.

There were no significant differences in hip ab/adduction patterns and the groups had varied profiles. The affected limb in the amputee non-fallers remained abducted throughout the gait cycle.

Table 5.2. Amputee vs. control fallers and non-fallers mean (SD) peak kinematic values (°).

	Amputee				Control	
	Non-fallers		Fallers		Non-fallers	Fallers
	Affected	Intact	Affected	Intact		
Hip adduction stance	0.4 (2.4)	5.8 (5.4)	2.3 (3.8)	4.9 (5.4)	6.3 (2.5)	6.1 (1.7)
Hip abduction swing	-6.1 (1.9)	-2.2 (5.5)	-6.0 (4.2)	-4.8 (4.8)	-7.6 (3.9)	-5.9 (1.8)
Hip ROM (frontal)	9.5 (2.7)	9.5 (2.4)	10.9 (3.5)	11.0 (4.6)	13.9 (3.7)	12.0 (0.9)
Hip extension	6.6 (8.6)	0.6 (5.5)	-0.7 (8.5)	-1.3 (8.5)	-5.8 (3.8)	4.2 (7.5)
Hip ROM (sagittal)	42.5 (6.0)	41.0 (3.7)	43.0 (6.9)	42.9 (6.4)	44.3 (5.6)	38.0 (6.2)
Knee flexion loading response	19.8 (13.5)	17.2 (3.2)	15.1 (4.6)	18.0 (4.4)	20.6 (7.0)	16.4 (5.8)
Knee flexion swing	73.2 (13.5)	59.4 (3.8)	66.5 (7.8)	61.7 (5.1)	62.1 (5.1)	62.4 (2.8)
Knee ROM (sagittal)	68.6 (12.4)	61.7 (6.5)	66.6 (11.4)	60.5 (7.9)	61.3 (6.1)	61.8 (2.4)
Ankle dorsiflexion stance	16.7 (3.9)	17.8 (2.7)	15.7 (2.3)	17.2 (4.2)	15.2 (5.8)	18.6 (3.3)
Ankle angle toe off	7.9 (3.2)	-3.9 (4.4)	8.4 (2.2)	-3.6 (8.0)	-12.9 (6.7)	-0.3 (5.9) *
Ankle plantarflexion swing	3.1 (2.9)	-10.6 (6.8)	3.7 (2.3)	-10.0 (8.5)	-17.9 (7.0)	-7.4 (9.0)
Ankle ROM (sagittal)	18.6 (3.8)	28.4 (8.5)	17.2 (3.1)	28.6 (3.9)	33.2 (4.8)	28.2 (5.3)

* Indicates significant difference between control fallers and non-fallers ($F_{(3,16)} = 3.12$, $p = 0.05$; one-way ANOVA for the intact limb)

5.3.4. GRF data

The anterior/posterior and vertical GRF components are shown in Figure 5.3 and peak forces and load and decay rates are presented in Table 5.3. There were no significant differences in either peak anterior (propulsion) or posterior (braking) GRF forces between the amputee or control fallers and non-fallers. The amputee fallers had a significantly greater first peak vertical force during loading on the affected limb (t_9 = -3.30; p=0.01) compared to the non-fallers. There were no differences in vertical force for the control groups.

Examining load and decay rates revealed that the amputee fallers loaded their affected limb significantly more than the non-fallers (t_9 = -2.60; p=0.03). A significant difference was also found for intact limb load rate ($F_{(3,16)}$ = 4.70, p = 0.02), but this time post-hoc analysis revealed that the control fallers loaded their limb significantly less than the non-fallers (p=0.01). No significant differences were found for decay rate in any of the groups.

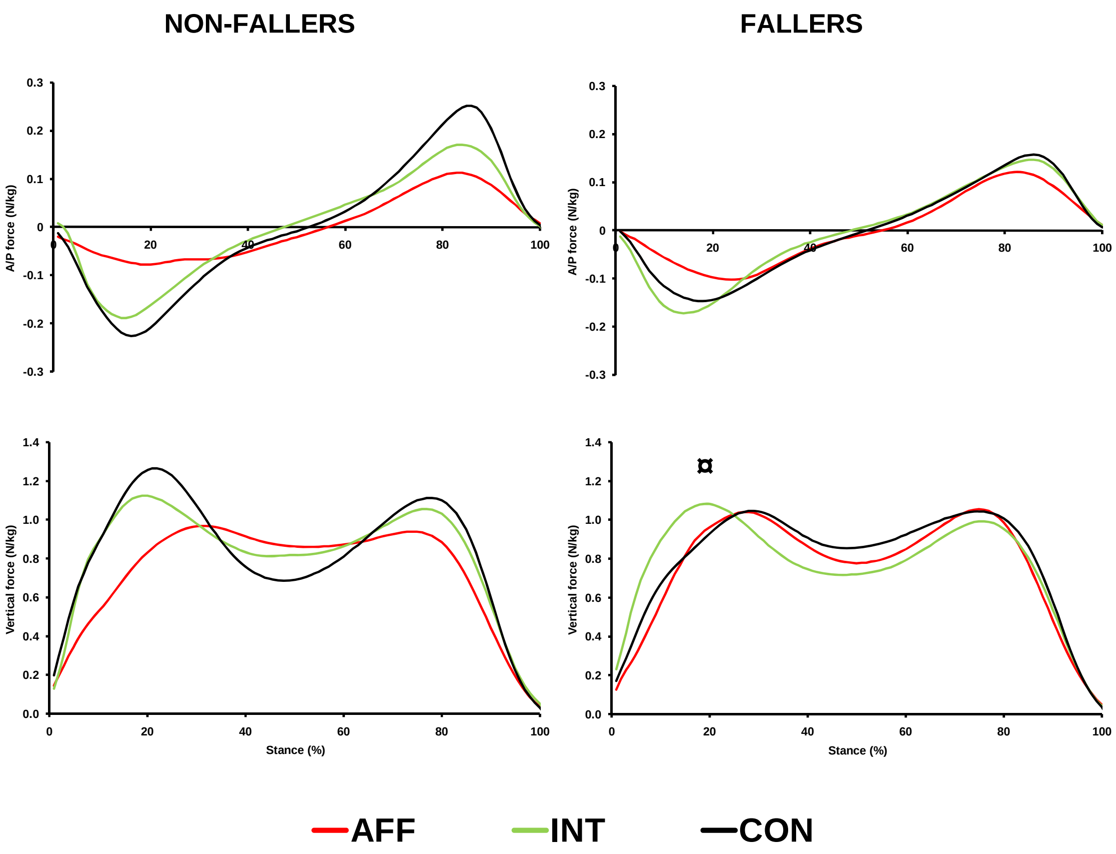


Figure 5.3. Ground reaction forces during level walking for the fallers and non-fallers. Data are averaged according to limb: affected (AFF) (red line); intact (INT) (green line); control (CON) (black line). The stance phase is initiated with initial contact and terminated with toe off.

Anterior and vertical are positive

¤ Indicates significant difference between amputee fallers and non-fallers (independent samples t-test for the affected limb)

Table 5.3. Amputee vs. control fallers and non-fallers mean (SD) peak GRF values (N/kg), load and decay rate (N/kg/s)

	Amputee				Control	
Kinetic variables	Non-fallers		Fallers		Non-fallers	Fallers
	Affected	Intact	Affected	Intact		
Posterior braking	-0.10 (0.03)	-0.20 (0.03)	-0.12 (0.03)	-0.18 (0.06)	-0.23 (0.06)	-0.15 (0.05)
Anterior propulsion	0.13 (0.02)	0.17 (0.06)	0.12 (0.03)	0.16 (0.07)	0.25 (0.06)	0.16 (0.07)
Vertical 1	1.01 (0.03)	1.14 (0.10)	1.10 (0.05) ¤	1.14 (0.28)	1.29 (0.16	1.06 (0.07)
Vertical 2	0.98 (0.08)	1.07 (0.12)	1.06 (0.08)	1.02 (0.23)	1.11 (0.07)	1.07 (0.07)
Load rate	4.54 (0.69)	8.99 (2.05)	6.67 (1.71) ¤	8.74 (1.69)	8.54 (2.03)	5.09 (0.84) *
Decay rate	-5.17 (1.09)	-5.72 (1.38)	-6.05 (1.41)	-6.28 (1.71)	-7.60 (0.83)	-6.24 (1.82)

***** Indicates significantly different from control non-fallers (one-way ANOVA for the intact limb)

¤ Indicates significantly different from amputee non-fallers (independent samples t-test for the affected limb of the amputee groups)

5.3.5. Joint moments

Joint moment profiles are shown in Figure 5.4 and peak values in Table 5.4. There were no significant differences in peak hip, knee or ankle joint moments on either the affected or intact limbs for the amputee groups. However, when the amputee fallers showed a small knee flexor moment during mid stance, the non-fallers maintained a knee extensor moment on the affected limb. The amputee non-fallers were the only group that maintained the GRF vector behind the knee during mid to terminal stance.

No differences were found for peak hip abductor moment values for the non-fallers and fallers in the amputee and control groups and the moment profiles followed typical patterns. A significant difference was found for peak hip flexor moment on the intact limb ($F_{(3,16)}$ = 5.27, p = 0.01) and post-hoc analysis showed that the control fallers had a significantly smaller hip flexor moment than the non-fallers (p = 0.01). There were no significant differences for peak ankle or knee joint moments in the control groups.

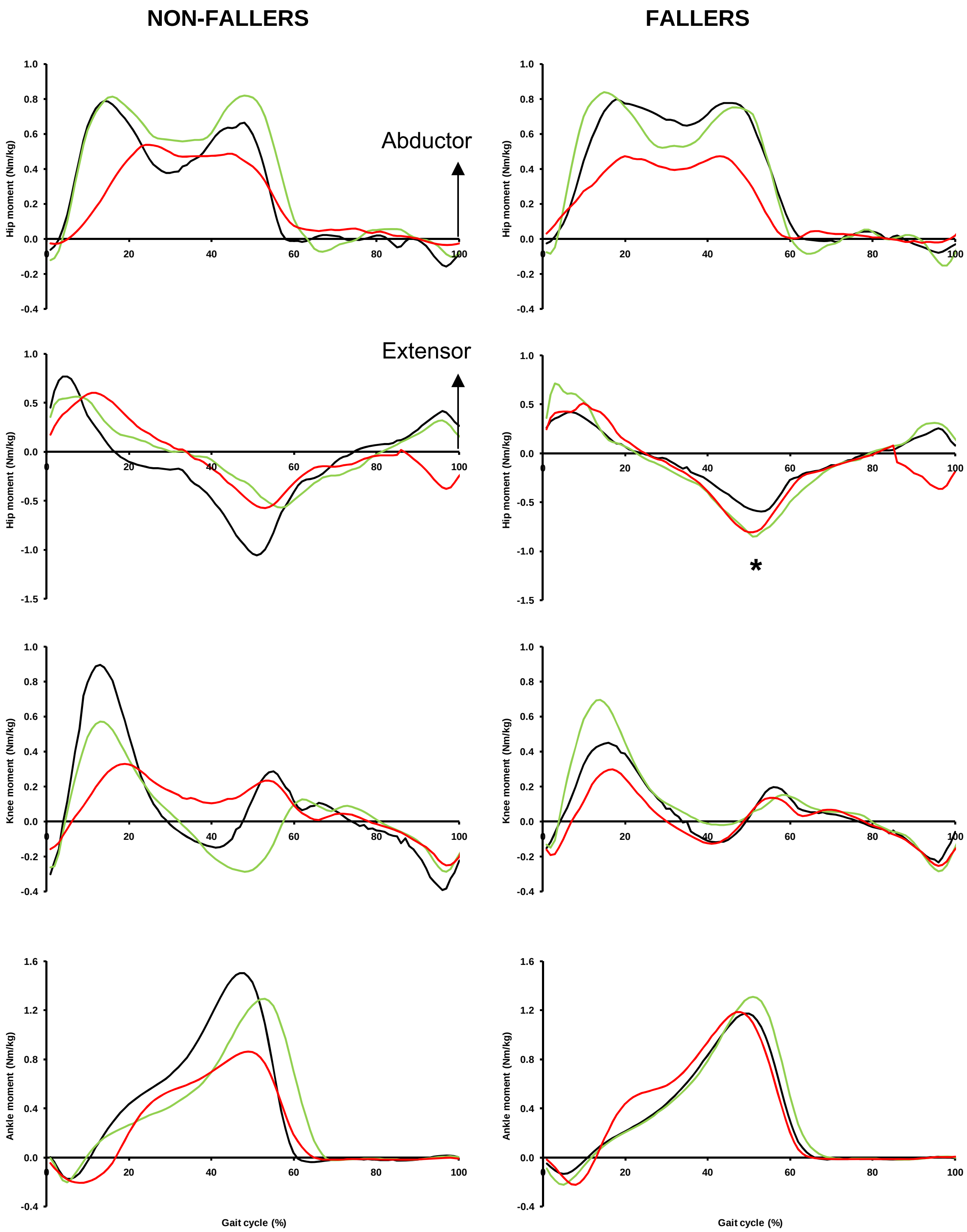


Figure 5.4. Joint flexor/extensor and hip ab/adductor moments during level walking for the fallers and non-fallers. Data are averaged according to limb: affected (AFF) (red line); intact (INT) (green line); control (CON) (black line).The gait cycle is initiated and terminated with foot contact.

Abductor, extensor and plantarflexor are positive.

***** Indicates significant difference between control faller and non-faller (one-way ANOVA for the intact limb)

Table 5.4. Amputee vs. control fallers and non-fallers mean (SD) peak joint moment values (Nm/kg).

	Amputee				Control	
Kinetic variables	Non-fallers		Fallers		Non-fallers	Fallers
	Affected	Intact	Affected	Intact		
Hip abductor moment mid-stance	0.58 (0.19)	0.94 (0.13)	0.53 (0.22)	0.92 (0.18)	0.81 (0.14)	0.84 (0.09)
Hip abductor moment pre-swing	0.58 (0.18)	0.86 (0.19)	0.49 (0.24)	0.80 (0.15)	0.68 (0.08)	0.80 (0.07)
Hip extensor moment stance	0.62 (0.40)	0.61 (0.15)	0.59 (0.20)	0.82 (0.33)	0.80 (0.36)	0.42 (0.06)
Hip flexor moment	-0.60 (0.23)	-0.65 (0.12)	-0.83 (0.37)	-0.88 (0.30)	-1.08 (0.16)	-0.60 (0.16) *
Hip extensor moment swing	0.03 (0.03)	0.33 (0.10)	0.08 (0.07)	0.35 (0.18)	0.44 (0.11)	0.27 (0.08)
Knee extensor moment mid-stance	0.39 (0.30)	0.59 (0.17)	0.32 (0.16)	0.78 (0.36)	0.91 (0.39)	0.46 (0.24)
Knee flexor moment terminal stance	-0.03 (0.13)	-0.31 (0.16)	-0.15 (0.17)	-0.15 (0.16)	-0.16 (0.14)	-0.13 (0.21)
Knee extensor moment pre-swing	0.29 (0.25)	0.15 (0.11)	0.16 (0.07)	0.30 (0.27)	0.29 (0.04)	0.22 (0.09)
Knee flexor moment terminal swing	-0.24 (0.12)	-0.29 (0.07)	-0.26 (0.13)	-0.29 (0.14)	-0.40 (0.13)	-0.24 (0.07)
Ankle plantarflexor moment	0.88 (0.39)	1.34 (0.29)	1.21 (0.28)	1.35 (0.41)	1.51 (0.30)	1.23 (0.21)

Abductor, extensor and plantarflexor are positive.

***** Indicates significant difference between control fallers and non-fallers (one-way ANOVA for the intact limb)

5.3.6. Joint powers

Joint power profiles can be seen in Figure 5.5 and peak values in Table 5.5. No significant differences were found for peak power bursts for the hip of the affected limb between the amputee fallers and non-fallers. However, there was a significant difference in peak hip joint power during terminal stance (labelled H2) ($F_{(3,16)} = 6.03$, $p = 0.01$). Post-hoc analysis revealed that the amputee fallers had significantly greater hip power absorption than the non-fallers ($p = 0.01$) and the control fallers had significantly less power absorption than the non-fallers ($p = 0.02$). No significant differences were found for peak knee joint powers between the fallers and non-fallers for the amputee and control groups. All profiles showed distinct power bursts. No significant differences were found for peak power bursts in the prosthetic ankle between the amputee fallers and non-fallers. A significant difference in ankle power absorption in terminal stance for the intact limb (labelled A1) was found across groups ($F_{(3,16)} = 2.84$, $p = 0.05$). Post-hoc analysis revealed that the amputee fallers had significantly smaller ankle power absorption during terminal stance compared to the non-fallers (-0.66 ± 0.15 W/kg and -1.23 ± 0.51 W/kg, p=0.04, respectively). No significant results were found for ankle power generation in pre-swing (labelled A2).

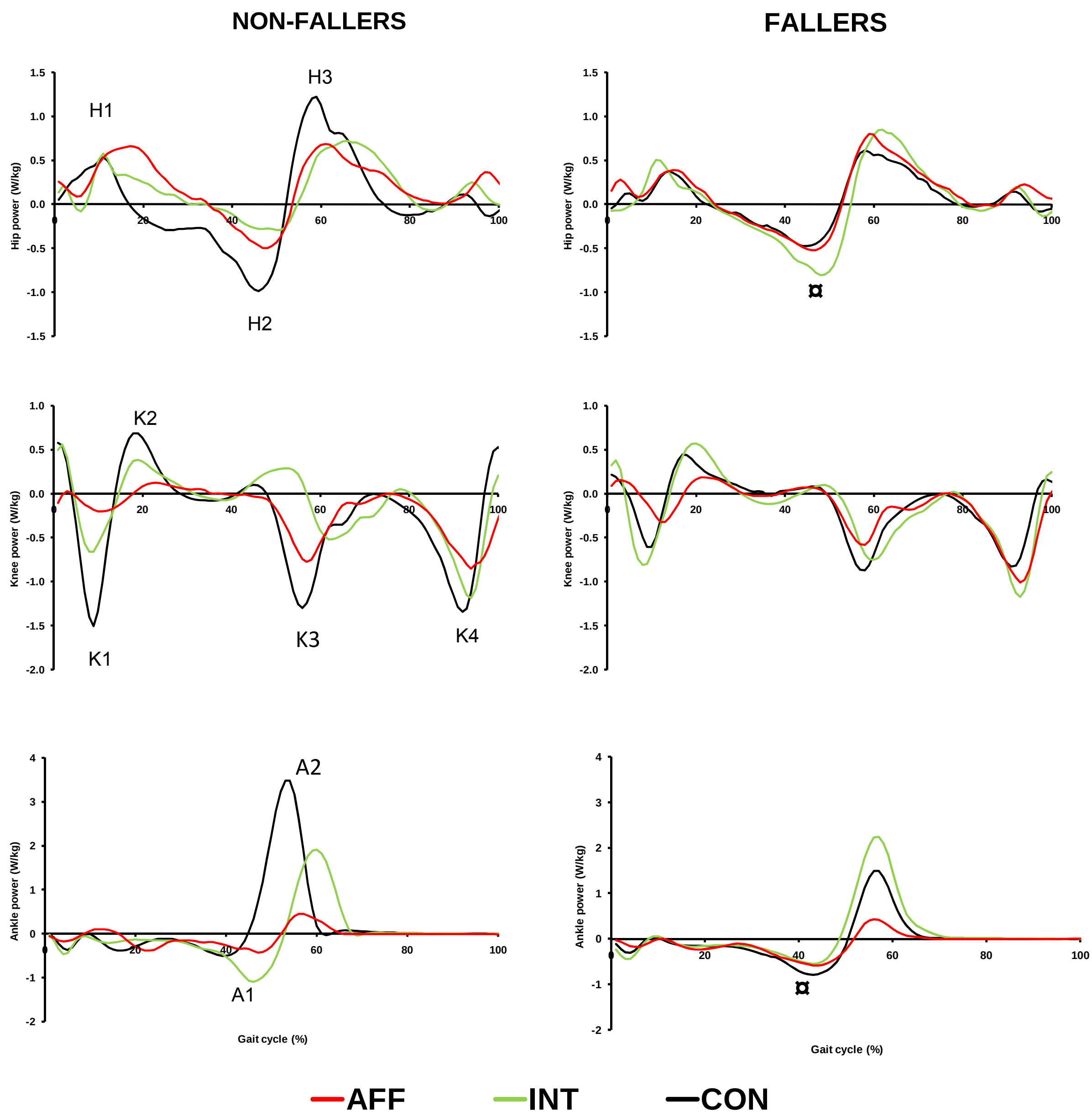


Figure 5.5. Sagittal plane joint powers during level walking for the fallers and non-fallers. Data are averaged according to limb: affected (AFF) (red line); intact (INT) (green line); control (CON) (black line).The gait cycle is initiated and terminated with foot contact.

Power generation is positive

¤ Indicates significantly different from amputee non-fallers (one-way ANOVA for the intact limb)

Table 5.5. Amputee vs. control fallers and non-fallers mean (SD) peak joint powers (W/kg).

	Amputee				Control	
Kinetic variables	Non-fallers		Fallers		Non-fallers	Fallers
	Affected	Intact	Affected	Intact		
Hip power generation loading response (H1)	0.80 (0.59)	0.65 (0.34)	0.70 (0.30)	0.61 (0.27)	0.80 (0.74)	0.38 (0.31)
Hip power absorption stance (H2)	-0.58 (0.34)	-0.35 (0.10)	-0.65 (0.25)	-0.85 (0.43) ¤	-1.02 (0.21)	-0.57 (0.12) *
Hip power generation pre-swing (H3)	0.85 (0.26)	0.84 (0.20)	1.08 (0.58)	1.02 (0.42)	1.28 (0.57)	0.59 (0.38)
Knee power absorption loading response (K1)	-0.31 (0.21)	-0.82 (0.21)	-0.41 (0.25)	-0.98 (0.75)	-1.58 (0.71)	-0.69 (0.43)
Knee power generation mid-stance (K2)	0.21 (0.20)	0.44 (0.27)	0.21 (0.16)	0.64 (0.49)	0.74 (0.51)	0.51 (0.40)
Knee power absorption pre-swing (K3)	-0.86 (0.28)	-0.66 (0.30)	-0.83 (0.43)	-0.92 (0.64)	-1.33 (0.44)	-0.95 (0.36)
Knee power absorption terminal swing (K4)	-1.16 (0.65)	-1.29 (0.45)	-1.07 (0.68)	-1.20 (0.65)	-1.42 (0.52)	-0.88 (0.28)
Ankle power absorption stance (A1)	-0.48 (0.16)	-1.23 (0.51)	-0.73 (0.33)	-0.66 (0.15) ¤	-0.61 (0.53)	-1.12 (0.43)
Ankle power generation pre-swing (A2)	0.66 (0.46)	2.30 (1.11)	0.75 (0.51)	2.70 (1.63)	3.63 (1.43)	1.89 (0.82)

Power generation is positive

***** Indicates significant difference between control fallers and non-fallers (one-way ANOVA for the intact limb)

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA for the intact limb)

5.3.7. Variability

There were no significant differences in the coefficient of variation between the fallers and non-fallers on any of the step variables except support time. The amputee fallers showed significantly more variability on time spent in swing in the affected limb than the non-fallers (t_9 = -2.52; p=0.03) (Table 5.6).

Table 5.6. Amputee vs. control fallers and non-fallers mean (SD) coefficient of variation (%).

	Amputee				Control	
	Non-fallers		Fallers		Non-fallers	Fallers
	Affected	Intact	Affected	Intact		
Step length	4.5 (1.7)	4.9 (2.6)	4.9 (2.5)	4.3 (1.9)	3.0 (1.4)	5.2 (2.0)
Step frequency	3.7 (0.7)	4.3 (1.1)	4.6 (1.9)	4.0 (1.6)	2.8 (1.0)	4.9 (2.3)
Stance	3.9 (1.3)	4.4 (1.1)	4.5 (1.5)	4.3 (1.6)	2.6 (1.1)	4.7 (2.8)
Swing	4.1 (1.5)	6.2 (1.6)	6.2 (1.2)¤	6.7 (2.7)	3.3 (1.0)	4.4 (2.1)
Double support	12.0 (2.5)	9.1 (3.9)	14.1 (7.3)	14.8 (8.1)	10.5 (1.9)	11.9 (3.6)

¤ Indicates significant difference between amputee fallers and non-fallers (independent samples t-test for the affected limb of the amputee groups)

5.4. Discussion

The aim of this study was to compare the gait patterns of fallers and non-fallers during level walking in transtibial amputees and able-bodied participants. Until now, it was unclear whether specific biomechanical variables differed between amputee fallers and non-fallers during walking, whereas the differences between able-bodied fallers and non-fallers have been well-documented (Hausdorff *et al*., 2001; Heiderscheit, 2003; Kerrigan *et al*., 1998; Kerrigan *et al*., 2001; Lee and Kerrigan, 1999). Overall, the amputee participants in this study exhibited gait patterns typically associated with pathological gait due to amputation as reported in previous research (Isakov *et al*., 1996; Nolan *et al*., 2003; Powers *et al*., 1998; Sanderson and Martin, 1997; Winter and Sienko, 1988).

5.4.1. Walking speed

Although not significantly different, the amputee fallers walked faster by taking longer and more frequent steps (Table 5.1). It is widely acknowledged that walking speed influences gait parameters and that these follow a pattern of change in response to gait speed (Winter, 1991). In the present study, walking speed was not controlled. The purpose of this study was to compare the gait patterns that best represented fallers vs. non-fallers, and walking speed could have been an important differentiating factor. Previous studies have reported significant differences in walking speed between able-bodied fallers vs. non-fallers (Kerrigan *et al*., 2001) and young vs. elderly (Kerrigan *et al*., 1998), indicating that speed is an overall descriptor of physical function and performance. Therefore, forcing an individual to walk beyond their self-selected walking speed would not have been representative of their typical gait pattern. Furthermore, as this was the first study to compare amputee fallers vs. non-fallers, it was found that the amputee fallers in this study fell within the range of walking speed reported in the amputee gait literature (1.0 - 1.2 m/s) (Jaegers *et al*., 1995; Sanderson and Martin, 1997). However, this was likely faster than amputees who are the least mobile and therefore, at the greatest risk for falling.

5.4.2. Joint mobility

The first hypothesis related to joint mobility and the results indicated that joint mobility was not a significant variable to distinguish between amputee fallers and non-fallers for straight level gait. Peak kinematic values of the prosthetic and intact ankle joints were similar during the loading and support phases for the amputee fallers and non-fallers. This was not surprising as both groups would have had reduced ROM and absent plantarflexors related to the prosthesis. Ankle motion was not considered a differentiating biomechanical variable in transtibial amputees. However, the control fallers displayed significantly reduced ankle plantarflexion at toe off compared to the control non-fallers and this was related to weak ankle plantarflexor muscle contraction (Kerrigan *et al*., 1998).

Whilst it was hypothesised that the fallers would display reduced hip range of motion in the frontal or sagittal planes, as has been reported among older able-bodied fallers, independent of speed (Kerrigan *et al*., 1998), this was not the case for the participants in this study. However, neither the amputee non-fallers nor the control fallers actually achieved full hip extension. Although none of these variables were significantly different, this was probably related to walking speed.

5.4.3. Loading and support phases

The second aim of this study was to examine the GRF, joint moments and powers to determine if there were kinetic differences between the fallers and non-fallers. The literature has found that transtibial amputees avoid weight-bearing on the residual limb by exhibiting shorter affected vs. intact support times (Donker and Beek, 2002; Nolan *et al*., 2003; Sanderson and Martin, 1997) and, as a protective mechanism, load their affected limb less (Sanderson and Martin, 1997). The findings from this research have revealed that the first vertical peak GRF and load rate on the affected limb were significantly smaller in the amputee non-fallers. Despite similar stance duration, the results showed that the fallers loaded their affected limb more than the non-fallers.

While many studies investigating falls in the elderly have collected GRF to calculate joint mechanics, few have actually presented the GRF component data. The current findings showed there were no differences between control fallers and non-fallers on peak GRF values, but that the control fallers loaded their lower-limbs significantly less than the non-fallers. A large load rate demands sufficient extensor control and strength to keep the lower-limbs from collapsing and loading is considered one of the most demanding tasks of the gait cycle (Perry, 1992). Interestingly, amputee non-fallers and control fallers showed similarities in joint mechanics related to reduced lower-limb loading but for different adaptive reasons. In the amputee, reduced loading was likely an effective mechanism for residual limb protection, whereas in the control group, it probably reflected muscle extensor weakness and compromised weight-bearing stability evidenced by the slow walking speed.

Power absorption at the ankle in terminal stance (labelled A1) for the intact limb was significantly smaller in the amputee fallers vs. non-fallers (Table 5.5). This suggested that the amputee non-fallers relied on greater eccentric control of the ankle plantarflexor, primarily of the soleus muscle as it controlled the forward progression of the tibia over the ankle joint. The power absorption at the ankle also coincided with power absorption at the hip joint in terminal stance (labelled H2), which slowed the thigh into extension. This may have been an effective mechanism to slow tibial advancement over the intact stance limb, which would then allow for a safer contralateral swing phase and placement of the prosthetic foot. A small prosthetic power absorption burst (A1) was obvious in both groups. Winter and Sienko (1988) attributed this to the prosthetic foot absorbing energy as it deformed in dorsiflexion in mid- to terminal stance.

Previous studies have reported knee joint moments in transtibial amputees that were very small or close to zero, particularly during early stance, to reduce the demands on the knee extensor musculature (Sanderson and Martin, 1997; Winter and Sienko, 1988). Sufficient muscle strength in the knee extensors is needed to control the knee joint by initiating weight-bearing stability through increased knee flexion, and by reducing load rate. In the present study, there were no significant mechanical differences in joint moments or powers that differentiated the fallers from the non-fallers during the loading phase. In the amputee groups, the knee moments on the affected limb, particularly the extensor moments, were reduced compared to the intact limb.

Although not significantly different in the present study, the knee and hip extensor moments (stance) for the control fallers were approximately half that of the non-fallers. Coupled with the reduced knee power absorption (K1) and hip power generation (H1) bursts in loading, the current results suggest the control fallers showed strength deficits of the knee extensors that controlled knee flexion by eccentric contraction compared to the non-fallers. These findings were in agreement with Kerrigan *et al*. (1998), who found significant age-associated reductions in peak knee extensor moment and knee power absorption (K1) and peak hip extensor moment in elderly individuals compared to young individuals both walking at their self-selected, comfortable speeds.

Hip mechanical moment and power generation and absorption in stance (H1 and H2) are reportedly highly variable among the amputee population (Winter and Sienko, 1988). Power generation at the hip on the affected limb (H1) was concentric into mid-stance in the non-fallers. This could have reflected their hip moment that remained extensor during mid-stance (Figures 5.4 and 5.5), and may be a successful compensatory mechanism to control the collapse of the stance limb (Eng and Winter, 1995) during the support phase and the lack of energy generation necessary for propulsion due to the lack of ipsilateral plantarflexor muscles. Examining peak hip power absorption in stance (H2) again revealed similarities between amputee non-fallers and control fallers, such that the amputee non-fallers had a 59% smaller H2 power burst compared to the amputee fallers; conversely, in the control groups, the fallers had a 56% smaller H2 power burst compared to the non-fallers. Smaller H2 bursts could have reduced the demands on the hip flexor musculature to decelerate extension of the thigh as it rotated backwards in preparation for swing (Eng and Winter, 1995). Interestingly, significant differences were found between the fallers and non-fallers in both groups for hip power absorption during stance (H2) suggesting that deceleration of thigh extension could be an important variable related to falls, especially since hip powers reflect balance control of carrying the HAT segment.

5.4.4. Propulsion phase

The ankle moment plays an integral role for gait stability and propulsion, but compensations in transtibial amputees can occur at the knee and hip joints in varying contributions (Sanderson and Martin, 1997; Winter and Sienko, 1988). No differences in peak ankle moment were found for either the amputee or control groups and this likely reflected the mechanical limitations of the prosthetic ankle. One might have expected meaningful (if not significant) differences on the intact limb, and indeed the peak ankle moment was 27% lower in the amputee non-fallers. Ankle plantarflexor moments and powers were able to distinguish previously between fallers and non-fallers in control subjects irrespective of walking speed (Kerrigan *et al*., 1998), but this was not the case in the current study. Meaningful but statistically insignificant differences in the current study likely reflected small sample size and large variations in individual ankle moment profiles, particularly in the amputee groups.

On the affected limb, propulsion powers were very similar for both groups, suggesting that the lack of energy return and the mechanical characteristics of the prosthesis may influence the power profiles more than an individual's falls history. Small variations in magnitude and timing were found for each of the propulsion powers at the joints of the lower-limb on the intact limb, such that the amputee non-fallers had reduced peak power generation at the ankle (A2), power absorption at the knee (K3) and power generation at the hip (H3) in pre-swing (Figure 5.5 and Table 5.5) compared to the fallers. This possibly reflected an attempt to adopt a cautious gait pattern, with a reluctance to generate large push off forces with the intact limb for the affected limb to control during single support.

The opposite findings were observed in the control groups, such that the control fallers exhibited reduced power generation at the ankle (A2) (by 48% and 54% at the ankle and hip, respectively) compared to the non-fallers during the propulsion phase of the gait cycle. This implied the primary muscles responsible for push-off, the gastrocnemius and soleus, showed reduced concentric contraction, perhaps as a result of muscle weakness or slower walking speed. Winter (1987) cited the A2 burst as the single most important energy generation phase during the gait cycle. Although these variables were not statistically significant, the differences were considered meaningful nonetheless because they occurred during the transition from double to single support (for the amputees, transition into single support on the affected limb), which is a vulnerable time in the gait cycle, as body weight is supported by the affected limb only. No significant or marked differences between the control fallers and non-fallers were found in knee moments during the propulsion. While knee extensor weakness was associated with power absorption differences in the support phase of gait, the knee extensors, particularly the vasti muscles, are generally inactive during terminal stance and pre-swing phases. This could explain why no kinetic differences were observed between the faller and non-faller groups, as knee extensor weakness would not be obvious.

5.4.5. Variability

The third hypothesis was that fallers would have increased variability in the temporal-spatial parameters. In transtibial amputees, the mechanical limitations of the prosthetic foot reduce the degrees of freedom and variability associated with ankle joint motion. However, greater variability of stride characteristics has been considered an indicator for falling in the elderly (Hausdorff *et al*., 2001; Heiderscheit, 2000). The results revealed that the amputee fallers exhibited significantly more variability associated with swing duration on the affected limb only (Table 5.6), suggesting this was a suitable temporal-spatial characteristic for differentiating amputee fallers from non-fallers. No significant differences in variability were found for the control groups, which was in contrast to previous work on temporal-spatial variables (Hausdorff *et al*., 2001; Heiderscheit, 2000). Although the control sample set was small, it is possible that joint kinematic analysis or interlimb joint coordination could identify any differences within these groups.

A dynamical systems perspective would suggest that the presence of variability in gait characteristics indicated that the system was more flexible and therapeutic interventions in the treatment of movement disorders could be effective (Heiderscheit, 2000). By associating greater variability with falls history, amputee patients and falls prevention programmes should be encouraged to explore their movement boundaries in safe, controlled environments, such as during physiotherapy treatment. Understanding when the system was most susceptible to responding positively to physiotherapy treatment is a critical aspect to an effective rehabilitation programme.

5.5. Conclusion

In conclusion, biomechanical differences existed between amputee fallers and non-fallers and these were noted during loading of the affected limb and transition from double support to single support on the affected side. However, contrary to the hypothesis, amputee fallers did not show reduced joint mobility compared to the non-fallers. Amputee non-fallers exhibited mechanical joint patterns that were similar in shape and magnitude to control fallers. Conversely, amputee fallers showed joint mechanics similar to control non-fallers. These results have important implications for falls prevention and amputee rehabilitation. These programmes should focus on improving the strength of the knee joint musculature (particularly on the affected side for the transtibial amputees) and eccentric muscle strength of the ankle (on the intact limb for the transtibial amputees) and hip. Improvements in muscle function would help control progression and stability during locomotion

CHAPTER SIX - GAIT PATTERNS IN TRANSTIBIAL AMPUTEE FALLERS VS. NON-FALLERS: BIOMECHANICAL DIFFERENCES DURING STAIR ASCENT

6.1. Introduction

Stairs are the most common form of obstacles we encounter every day. Stair walking is recognised as a more mechanically challenging daily activity than level walking (McFadyen and Winter, 1988; Reeves *et al*., 2008a) and falls often occur during stair walking (Reeves *et al*., 2008a). During stair ascent, the lead leg performs the greatest effort in the safe negotiation from one step to the next (McFadyen and Winter, 1988). The quadriceps muscles play an important role in maintaining the body upright (McFadyen and Winter, 1988), particularly during the single support pull-up phase.

Young healthy persons can accomplish the rhythmic task of walking up stairs with relative ease. However, the neuromechanical characteristics of level gait in transtibial amputees suggest that this group of individuals may experience greater difficulties during stair ascent. Stair ascent requires that the body's COM is transported in a greater upwards vertical direction compared to level gait. Powers and Boyd (1997) reported that difficulty of stair negotiation in transtibial amputees was associated with a slower velocity and more asymmetrical gait patterns. Gait asymmetry was linked with the reduced mechanical function of the prosthetic ankle/foot components and the loss of the plantarflexors, a muscle group that is especially important during the forward continuance phase (McFadyen and Winter, 1988), and reduced strength of the knee extensor muscles.

There are currently few empirical studies that have investigated the biomechanics of stair walking in transtibial amputees (Powers and Boyd, 1999; Schmalz *et al*., 1997; Yack *et al*., 1997) and none to date that have explicitly examined whether specific lower-limb biomechanical patterns could be linked with falls in this population. This is surprising given that amputees report more falls than age-matched, able-bodied individuals and that stair walking has been particularly associated with falls. Identifying biomechanical variables that could distinguish amputee fallers from non-fallers would provide particularly valuable information for improving amputee rehabilitation practice and falls prevention programmes. The main aim of this study was to compare the gait patterns of fallers and non-fallers during stair ascent in transtibial amputees and able-bodied participants. It was hypothesised that amputee and control fallers would ascend stairs more slowly and exhibit reduced joint mobility compared to the non-fallers. This was because stair ascent places greater demands on lower-limb joint range of motion than level walking (McFadyen and Winter, 1988; Powers and Boyd, 1999; Reeves *et al*., 2008a). It was also anticipated that kinematic differences would become apparent during the transition steps from level to stair walking. A second hypothesis was that amputee fallers and control fallers would exhibit reduced GRF and altered joint moments and powers at the knee and hip compared to their non-faller counterparts. This was based on the findings that the knee extensors perform much of the work during stair ascent (Reeves *et al*., 2008b) and that transtibial amputees rely on the hip extensor muscles on the affected limb during the weight acceptance and pull-up phases (Yack *et al*., 1999).

6.2. Methods

6.2.1. Participants

Eleven transtibial amputees and nine age-matched able-bodied participants completed the stair ascent task. All participants gave written informed consent to take part in this study. Individual participant details according to falls history can be found in Table 3.1.

6.2.2. Staircase

A purpose-built 3-step wooden staircase was built for this study (Figure 6.1). The steps were 80 cm wide, with a rise of 20 cm, a tread of 25 cm, and a final tread of 80 cm. The force plate was positioned flush into the bottom step, which was independent from the remainder of the structure, and the tread was adjusted at 40 cm to accommodate the force platform. Wooden handrails were 50 cm high, attached to the main structure but not to the bottom step which housed the force plate. The staircase dimensions conformed to stair designs in public buildings and the number of steps was used previously in stair biomechanics studies (Beaulieu *et al*., 2008; Grenholm *et al*., 2009; Radtka *et al*., 2006). The structure was on wheels and could be locked securely into place.

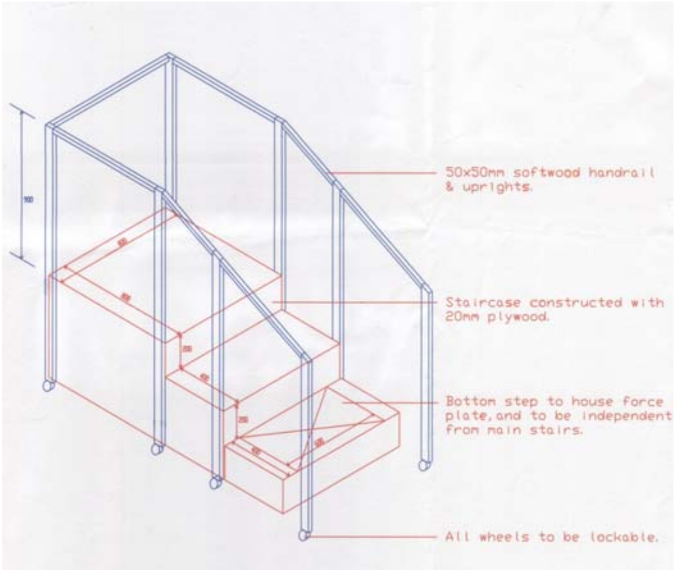


Figure 6.1. Purpose-built 3-step staircase

6.2.3. Experimental protocol

Each participant completed the stair ascent trails during their visit to the Human Performance Laboratory at the University of Hull. All participants were made aware of the protocol and were reminded they could rest at any time. Participants wore shorts and reflective markers were affixed onto the lower-limbs as described in section 3.10. Participants were asked to ascend the stairs using their preferred pattern and pace. No instructions were given about lead limb. They were told they could use the handrails if they felt necessary as described in section 3.15.1.

6.2.4. Data analysis

All data were analysed as explained in sections 3.16, 3.17, and 3.18. Data for the affected limb was only compared between two groups: the amputee fallers and non-fallers. Data for the intact limb was compared across four groups: amputee and control fallers and non-fallers. The variables that were selected for analysis included average resultant walking speed for each step and support times (single support as % of gait cycle), peak joint kinematic and joint ROM, peak GRF data, load and decay rates, peak joint moments and power bursts as described and defined in section 3.19. Step length was not analysed because the dimensions of the staircase would have controlled for this variable to some extent. Similarly, previous stair negotiation studies have not analysed step length (Beaulieu *et al.*, 2008; Hamel *et al.*, 2005; Reeves *et al.*, 2008a; Reeves *et al.*, 2008b; Schmalz *et al.*, 2007). Additionally, maximum vertical toe clearance (in cm) during the transition from floor to stair walking was calculated. This variables was selected for analysis because insufficient toe clearance would likely result in a destabilising trip and potentially lead to a fall. During the floor to stair transition, kinematic data were only presented for the swing phase. Kinetic data could only be calculated for the lead limb (beginning with foot contact from the first to the third step) because only one force plate was positioned in the first step of the staircase.

6.2.5. Lead limb preference

Not all participants used the same stair ascent strategy. The data were first examined for lead limb preference by determining if participants favoured either the affected or the intact limb when stepping onto the first step of the staircase (Table 6.1). Nine of 11 amputee participants displayed a clear intact lead limb preference (always led with this limb) while 2 participants did not exhibit a clear preference and lead with either the affected or intact limb. However, for the purpose of this analysis, the intact limb was selected as the lead limb and the affected limb as the trail limb for all participants. Two of the participants, who had an intact lead limb preference, self-selected a ‘step to’ walking pattern (Table 6.1). For the controls, three participants had no limb preference; five had a left limb preference and one a right limb preference.

The following data were selected for analysis for participants using a step over walking pattern: participants first climbed one step with the lead limb (floor to first step) and this represented the floor to stair transition. Then each limb climbed the vertical height of two steps, representative of steady state stair ascent: floor to second step (trail limb) and first step to third step (lead limb) (Figure 6.2). Kinematic comparisons could be made between the two limbs, but kinetic data were collected for the lead limb only. Finally, the trail limb climbed one step to reach the third step or landing; this was not selected for analysis. For the amputee participants who did not express a lead limb preference, the intact limb was analysed as the lead limb during ascent. For the control participants, the lead limb was analysed based on the participants’ preference. The average of both limbs was computed and analysed for the control participants who did not show a lead limb preference.

Table 6.1. Step and limb characteristics during stair ascent in the amputee and control participants

Amputee participants	Lead limb preference	Handrail use	Step pattern
Faller			
1	Intact	Light	Step over
2	Intact	Light	Step over
3	Intact	Light	Step over
4	Intact	Light	Step over
5	Intact	Reliant	Step to
6	Intact	Reliant	Step to
Non-faller			
7	Intact	Moderate	Step over
8	Intact	Moderate	Step over
9	Intact	Moderate	Step over
10	None	Light	Step over
11	None	Light	Step over
Control participants			
Faller			
1	None	Light	Step over
2	Left	Light	Step over
3	Left	Light	Step over
4	Left	Moderate	Step over
Non-faller			
5	None	Light	Step over
6	Left	No	Step over
7	Left	No	Step over
8	None	No	Step over
9	Right	No	Step over

'Light' handrail use was classified as using the handrail as a guide only (Reeves *et al.*, 2008a). In the current study, light handrail meant that participants held the handrail with one hand only.

'Moderate' handrail use occurred when participants used both arms as a guide, but did not perform a large portion of the work with their arms.

'Reliant' handrail use occurred when participants performed considerable work with their arms and, when asked, would not have felt safe without the handrails.

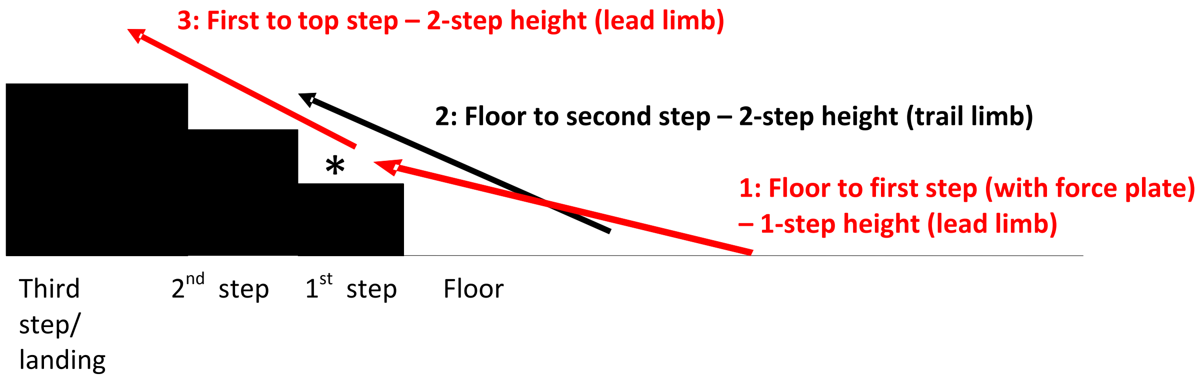


Figure 6.2. Illustration of staircase and steps selected for analysis during stair ascent. Solid red line represents lead limb. For the amputee participants, the lead limb was the intact limb. Solid black line represents the trail limb. For the amputee participants, the trail limb was the affected limb

* Indicates position of force plate

6.2.6. Statistical analysis

Statistical analysis was conducted on data from 18 participants and the amputee faller sample size was reduced from n=6 to n=4. Statistical comparisons were made according to step and not between the intact and affected limbs. A one-way ANOVA was used to compare the amputee groups and the two control groups. If significant differences were found overall, Fisher's least significant difference (LSD) post-hoc test was used to determine whether the differences existed between the amputee fallers vs. non-fallers and control fallers vs. non-fallers. No other significant differences between groups (e.g. amputee non-fallers vs. control fallers) were investigated in this study. The alpha level for significance was set a priori at 0.05.

6.3. Results

6.3.1. Participant demographics

The data here are presented for 18 participants (the two 'step to' participants were excluded from statistical analysis). There were no significant differences for age ($F_{(3,14)}$ = 1.68, p = 0.22), height ($F_{(3,14)}$ = 0.81, p = 0.51), or body mass ($F_{(3,14)}$ = 0.17, p = 0.91) between the groups tested. For the amputee groups, there was no significant difference in time since amputation (t_7 = 1.33; p = 0.89).

6.3.2. Floor to first step (intact limb)

All participants were able to complete the level walking task without experiencing a trip, slip or fall.

Table 6.2. Amputee vs. control fallers and non-fallers mean (SD) maximal toe clearance and peak kinematic values (°) in the sagittal plane during swing from floor to the first step (intact limb).

	Amputee Non-Faller (n=5)	**Amputee Faller (n=4)**	**Control Non-Faller (n=5)**	**Control Faller (n=4)**
Maximum vertical toe clearance (cm)	9.6 (1.5)	10.6 (1.5)	10.5 (0.7)	9.8 (0.8)
Hip flexion swing (°)	72.2 (4.9)	70.5 (2.5)	67.9 (7.1)	75.8 (7.2)
Knee flexion swing (°)	84.7 (3.3)	88.7 (7.9)	84.8 (5.9)	83.3 (11.8)
Ankle plantarflexion swing (°)	-4.1 (6.6)	-2.3 (2.0)	-9.7 (7.8)	-2.1 (7.5)
Ankle dorsiflexion swing (°)	17.4 (9.8)	18.4 (4.7)	15.8 (2.9)	21.7 (2.3)

Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion –ve

Maximal toe clearance and peak joint angles in swing are presented in Table 6.2 and illustrated in Figure 6.3. There were no significant differences in any of the peak values between the amputee and control fallers and non-fallers

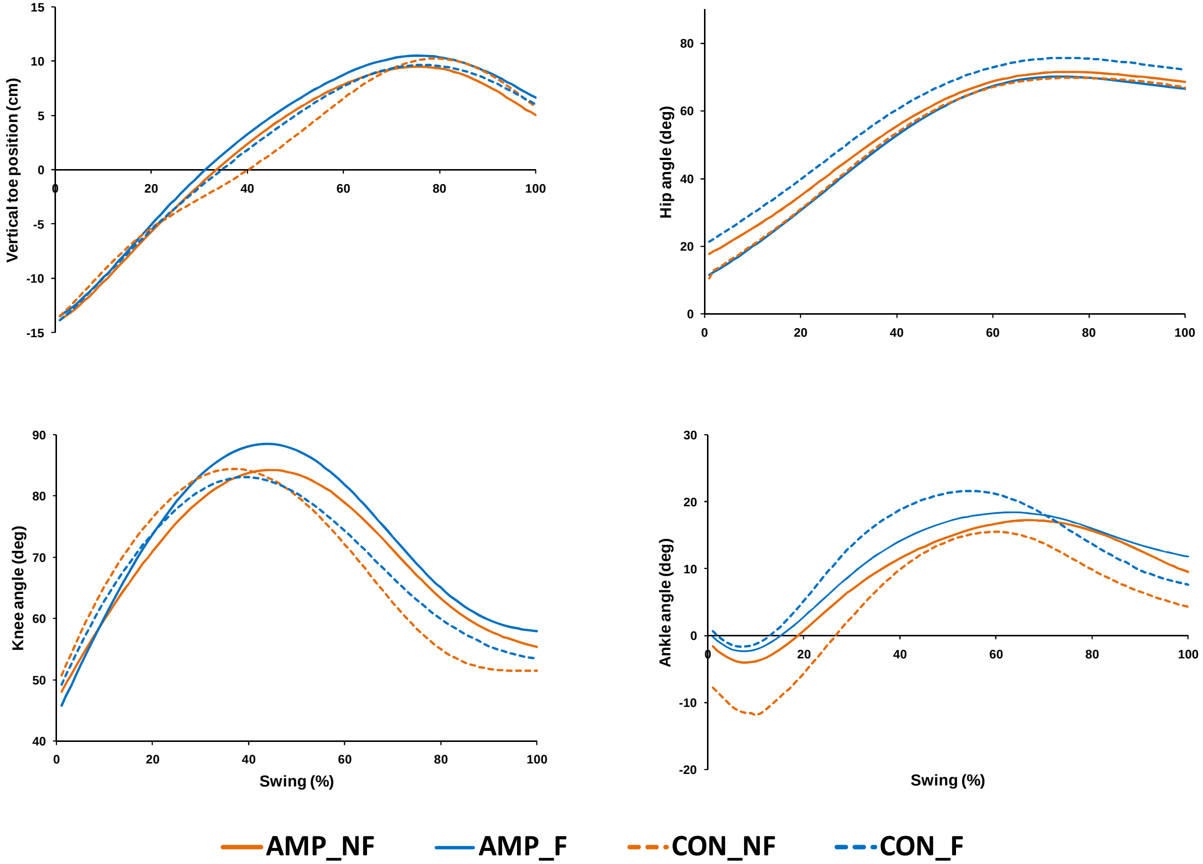


Figure 6.3. Joint flexion angles and vertical position of the toe during stair ascent from floor to the first step (intact limb) in fallers and non-fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line). The swing cycle is initiated with toe off and terminated with foot contact.

Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion –ve

6.3.3. Floor to second step (affected limb)

Temporal-spatial and peak kinematic values are shown in Table 6.3 and illustrated in Figures 6.4 and 6.5. A significant difference was found for walking speed ($F_{(3,14)} = 3.86$, $p=0.03$). Post-hoc analysis revealed that the amputee fallers moved significantly faster than the amputee non-fallers ($p=0.05$) during initial stair ascent. Conversely, the control fallers were significantly slower than the control non-fallers ($p=0.03$). Any differences in stance phase duration were not significant.

Table 6.3. Amputee vs. control fallers and non-fallers mean (SD) temporal-spatial and peak kinematic values (°) in the sagittal and frontal planes during stair ascent from floor to the second step (affected limb).

	Amputee Non-Faller (n=5)	**Amputee Faller (n=4)**	**Control Non-Faller (n=5)**	**Control Faller (n=4)**
Speed (m/s)	0.60 (0.09)	0.81 (0.10) ¤	0.86 (0.19)	0.64 (0.16) *
Stance phase (%)	56 (2)	57 (3)	59 (2)	61 (2)
Hip abduction stance (°)	-9.8 (3.9)	-9.7 (2.9)	-9.4 (3.2)	-7.2 (8.7)
Hip adduction swing (°)	16.0 (6.6)	12.3 (1.7)	2.3 (6.6)	1.5 (6.5)
Hip ROM frontal (°)	25.9 (4.9)	22.0 (3.9)	11.8 (5.6)	8.6 (2.8)
Hip extension stance (°) †	16.9 (6.1)	8.3 (4.8)	1.7 (3.6)	11.2 (10.0) *
Hip flexion swing (°)	85.3 (8.0)	81.0 (5.9)	67.1 (7.1)	77.7 (10.7)
Hip ROM sagittal (°)	68.4 (2.8)	72.7 (8.6)	65.4 (6.2)	66.4 (4.4)
Knee flexion loading (°)	19.0 (13.5)	16.3 (4.9)	22.2 (2.3)	13.6 (11.0)
Knee flexion swing (°)	80.3 (12.5)	86.0 (4.8)	89.2 (7.4)	95.4 (13.0)
Knee ROM (°)	72.4 (6.8)	82.2 (5.7) ¤	83.2 (7.8)	92.9 (5.5) *
Ankle dorsiflexion stance (°)	14.3 (4.8)	13.4 (3.6)	11.6 (5.0)	14.5 (3.4)
Ankle plantarflexion swing (°)	3.3 (2.7)	5.1 (3.2)	-19.1 (8.0)	-11.1 (3.4) *
Ankle dorsiflexion swing (°)	6.0 (2.9)	6.8 (3.3)	22.6 (7.9)	23.9 (3.7)
Ankle ROM (°)	11.0 (2.4)	8.3 (0.9)	41.7 (3.3)	35.0 (4.3) *
Pelvic tilt stance (°)	25.8 (4.5)	19.7 (2.1) ¤	17.2 (2.2)	26.3 (4.8) *
Pelvic tilt swing (°)	29.5 (4.5)	25.9 (2.2)	17.7 (3.1)	27.6 (3.9) *
Pelvic obliquity stance (°)	-12.4 (3.6)	-11.0 (2.8)	-9.6 (2.4)	-10.5 (4.4)
Pelvic obliquity swing (°)	8.1 (5.1)	10.6 (1.1)	6.8 (3.8)	7.3 (2.9)

Hip extension –ve (or minimum hip flexion +ve)†; Hip flexion +ve; Hip abduction -ve; Hip adduction +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion -ve; Anterior pelvic tilt +ve; Pelvic obliquity up (pelvic hike) +ve; Pelvic obliquity down (pelvic drop) –ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

* Indicates significant difference between control fallers and non-fallers (one-way ANOVA)

† It is noteworthy that the hip usually reached full extension during level gait and therefore was associated with a negative value. However, in stair ascent, the hip did not achieve full extension, so the positive value at this phase in the gait cycle represented the minimum amount of hip flexion

No significant differences were noted in hip abduction angles in late stance. The ANOVA results showed significant differences in peak hip adduction angles in terminal swing ($F_{(3,14)}$ = 7.10, p=0.00). However, post-hoc analysis revealed that these differences did not exist between the fallers and non-fallers in the amputee (p=0.36) or the control (p=0.83) groups. Similarly, hip ROM in the frontal plane was significantly different between groups overall, but not specifically between the amputee and control fallers vs. non-fallers (p=0.23 and p=0.32, respectively). It was beyond the scope of this thesis to discuss any results that were not between the fallers and non-fallers in either the amputee or control groups.

Significant differences existed in peak hip extension angles in pre-swing ($F_{(3,14)}$ = 4.86, p=0.02). Post-hoc analysis revealed this was only significant for the control group such that the fallers had significantly less extension at the hip joint compared to the non-fallers (p=0.04). While the ANOVA found overall significant differences in hip flexion angles in swing ($F_{(3,14)}$ = 4.62, p=0.02), these were not significant between either the fallers and non-fallers in either the amputee or control groups (p=0.45 and p=0.07, respectively). There was no difference in hip ROM.

There were no significant differences in peak knee flexion angles in loading and swing. However, the ANOVA showed differences in knee ROM ($F_{(3,14)}$ = 7.15, p=0.00) and post-hoc analysis revealed the fallers in the amputee and control groups had a significantly greater knee ROM than their non-faller counterparts (p=0.04 and p=0.05 for the amputee and control groups, respectively).

There was no difference in peak ankle dorsiflexion in stance. However, significant differences in peak plantarflexion in swing ($F_{(3,14)}$ = 24.96, p=0.00) and ankle ROM ($F_{(3,14)}$ = 147.08, p=0.00) were found. Post-hoc analysis demonstrated this difference was significant for the control groups only, such that the control fallers had less peak ankle plantarflexion and a smaller ankle ROM compared to the control non-fallers (p=0.03 and p=0.01, respectively).

Pelvic tilt profiles revealed significant differences in stance ($F_{(3,14)}$ = 7.20, p=0.00) and swing ($F_{(3,14)}$ = 10.39, p=0.00). Post-hoc analysis revealed the amputee fallers displayed significantly less anterior pelvic tilt in stance (p=0.03) but not in swing (p=0.15) compared to the amputee non-fallers. In the control groups, the opposite was true, and the fallers displayed significantly more anterior pelvic tilt in stance (p=0.00) and swing (p=0.00) compared to the non-fallers. No differences existed in pelvic obliquity profiles.

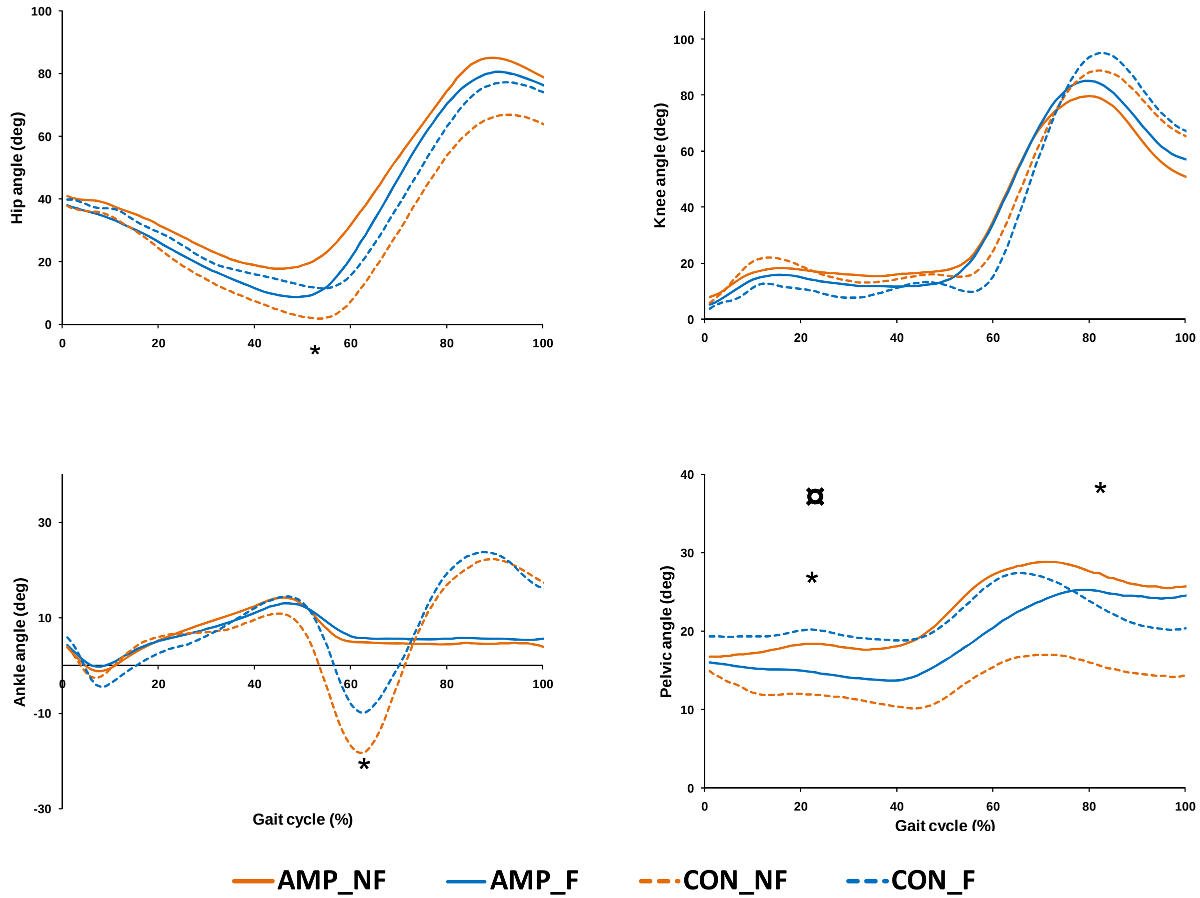


Figure 6.4. Sagittal plane angles during stair ascent from floor to the second step (affected limb) in fallers and non-fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line).The gait cycle is initiated and terminated with foot contact.

Anterior pelvic tilt +ve; Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve;

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA);* Indicates significant difference between control fallers and non-fallers (one-way ANOVA)

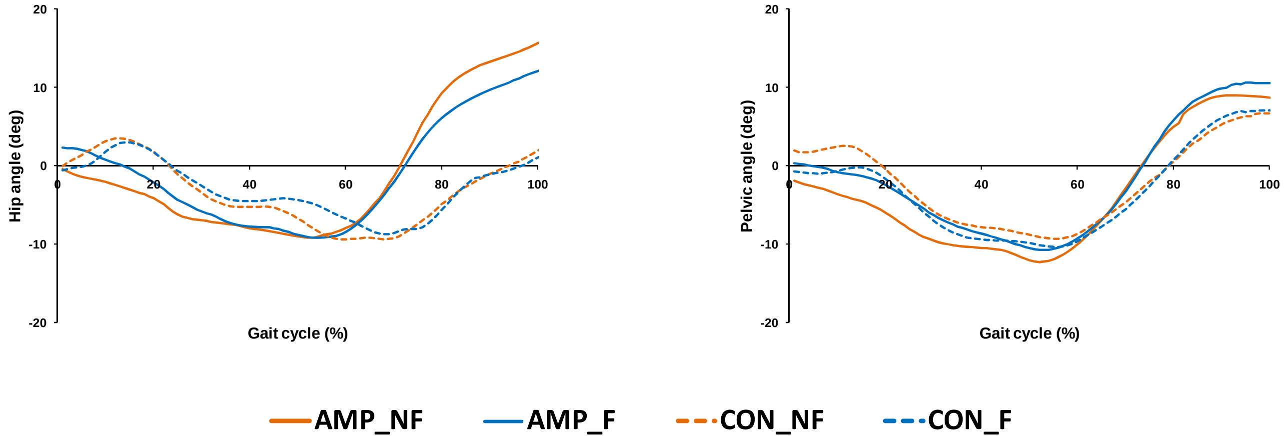


Figure 6.5. Frontal plane hip and pelvic kinematics during stair ascent from floor to the second step (affected limb) in fallers and non-fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line).The gait cycle is initiated and terminated with foot contact.

Hip adduction +ve; Pelvic obliquity up (pelvic hike) +ve

6.3.4. First to third step (intact limb)

Temporal-spatial values are shown in Table 6.4. A significant difference was found for walking speed ($F_{(3,14)}$ = 5.17, p=0.01). Post-hoc analysis revealed that the amputee fallers continued to negotiate stair ascent significantly faster than the amputee non-fallers (p=0.01). Conversely, in the control groups, the fallers were significantly slower than the non-fallers (p=0.02). Stance phase duration was not significant.

Hip, knee and ankle joint angles and pelvic profiles are illustrated in Figures 6.6 and 6.7, respectively and peak joint values can be found in Table 6.4. There were no significant differences in peak flexion (including hip ab/adduction) values or ROM at the hip, knee or ankle between the amputee and control fallers and non-fallers. However, peak anterior pelvic tilt was significantly different in mid-stance ($F_{(3,14)}$ = 6.63, p=0.00) and early swing ($F_{(3,14)}$ = 5.41, p=0.01). Post-hoc analysis revealed the control fallers exhibited significantly greater anterior tilt compared to the non-fallers (p=0.01 and p=0.02, respectively). No significant differences were noted in pelvic obliquity profiles.

Table 6.4. Amputee vs. control fallers and non-fallers mean (SD) temporal-spatial and peak kinematic values (°) in the sagittal and frontal planes during stair ascent from the first to the third step (intact limb)

	Amputee Non-Faller (n=5)	**Amputee Faller (n=4)**	**Control Non-Faller (n=5)**	**Control Faller (n=4)**
Speed (m/s)	0.46 (0.05)	0.66 (0.08) ¤	0.65 (0.12)	0.49 (0.11) *
Stance phase (%)	66 (3)	64 (2)	60 (1)	61 (3)
Hip abduction stance (°)	-13.1 (5.0)	-14.1 (3.3)	-7.0 (3.0)	-5.1 (5.0)
Hip adduction swing (°)	2.6 (3.0)	1.5 (7.5)	2.9 (6.6)	7.8 (4.9)
Hip ROM frontal (°)	15.7 (5.7)	15.6 (7.2)	9.9 (7.3)	12.9 (7.8)
Hip extension stance (°)	19.8 (3.5)	16.0 (6.3)	10.8 (4.9)	18.2 (9.8)
Hip flexion swing (°)	76.0 (5.7)	75.2 (2.1)	68.1 (7.5)	75.2 (7.6)
Hip ROM sagittal (°)	56.2 (6.0)	59.2 (6.0)	57.3 (3.8)	57.0 (9.5)
Knee flexion loading (°)	55.5 (8.1)	58.8 (5.4)	54.0 (7.5)	56.5 (8.7)
Knee flexion swing (°)	92.6 (2.3)	101.0 (6.4)	95.3 (9.9)	93.2 (11.4)
Knee ROM (°)	86.3 (5.3)	92.8 (8.3)	83.7 (7.2)	82.5 (6.4)
Ankle dorsiflexion stance (°)	15.5 (1.9)	17.0 (2.3)	14.5 (3.9)	18.0 (3.5)
Ankle plantarflexion swing (°)	-14.0 (6.1)	-19.0 (4.1)	-13.5 (10.0)	-13.5 (3.5)
Ankle dorsiflexion swing (°)	18.1 (10.2)	19.0 (4.5)	21.4 (5.6)	22.2 (5.8)
Ankle ROM (°)	34.7 (8.0)	38.6 (4.1)	35.2 (6.0)	36.8 (4.4)
Pelvic tilt stance (°)	31.3 (5.8)	28.0 (2.7)	18.3 (2.6)	28.0 (7.1) *
Pelvic tilt swing (°)	31.9 (5.9)	28.0 (3.0)	18.3 (2.4)	28.6 (9.1) *
Pelvic obliquity stance (°)	-9.9 (3.6)	-10.8 (1.0)	-7.1 (3.6)	-9.2 (4.3)
Pelvic obliquity swing (°)	9.2 (3.3)	9.5 (2.0)	8.4 (3.0)	7.8 (2.2)

Hip extension -ve; Hip flexion +ve; Hip abduction -ve; Hip adduction +ve; Knee flexion +ve; Ankle dorsiflexion; +ve; Ankle plantarflexion -ve; Anterior pelvic tilt +ve; Pelvic obliquity up (pelvic hike); +ve; Pelvic obliquity down (pelvic drop) –ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

* Indicates significant difference between control fallers and non-fallers (one-way ANOVA)

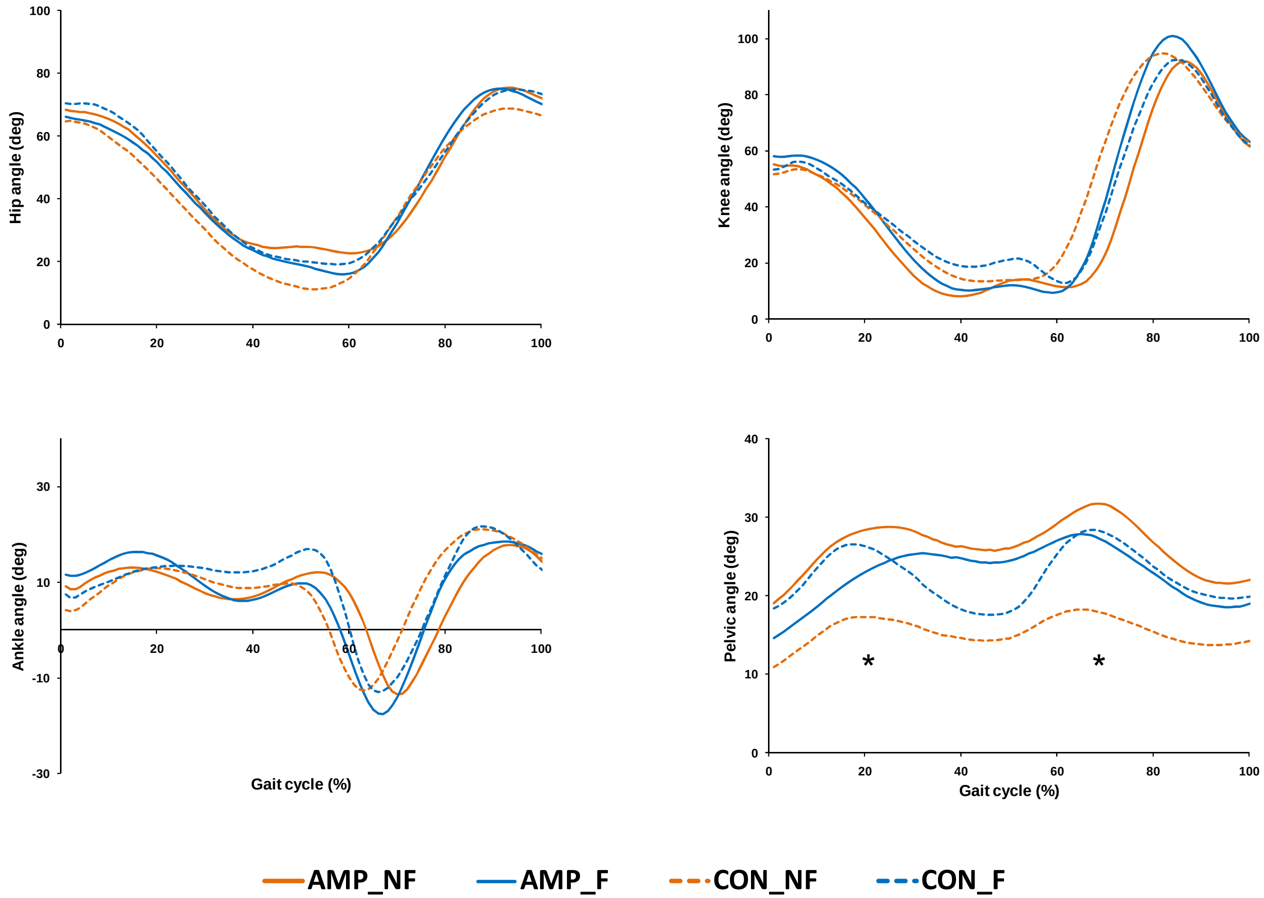


Figure 6.6. Sagittal plane angles during stair ascent from first to third step (intact limb) in the fallers and non-fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line).The gait cycle is initiated and terminated with foot contact.

Anterior tilt +ve; Flexion +ve; Dorsiflexion +ve; * Indicates significant difference between control fallers and non-fallers (one-way ANOVA)

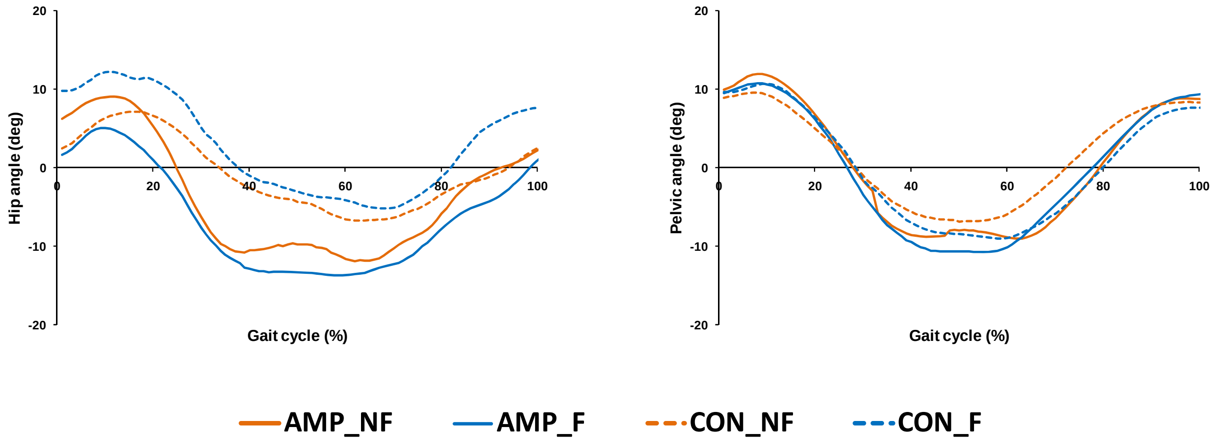


Figure 6.7. Frontal plane hip and pelvic kinematics during stair ascent from the first to the third step (intact limb) in fallers and non-fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line).The gait cycle is initiated and terminated with foot contact.

Hip adduction +ve; Pelvic hike/obliquity up +ve

Anterior/posterior and vertical GRF peak values and profiles are displayed in Table 6.5 and Figure 6.8, respectively. While no differences existed in peak braking or propulsive forces, significant differences were found in the vertical direction. The results from the ANOVA found significant results in Fz1 ($F_{(3,14)}$ = 4.43, p=0.02), Fz2 ($F_{(3,14)}$ = 5.08, p=0.01) and Fz3 ($F_{(3,14)}$ = 4.62, p=0.02) values. Post-hoc analysis revealed the significant differences were only found between the amputee groups. The amputee fallers experienced significantly larger Fz1 and Fz3 forces (p=0.01 and p=0.00, respectively) but significantly lower Fz2 forces (p=0.01) compared to the non-fallers.

Load and decay rates revealed significant results overall ($F_{(3,14)}$ = 5.40, p=0.01 and $F_{(3,14)}$ = 5.14, p=0.01, respectively). Post-hoc analysis found only the amputee fallers experienced a significantly greater decay rate compared to the amputee non-fallers (p=0.01). All other differences failed to reach statistical significance.

Table 6.5. Amputee vs. control non-fallers and fallers mean (SD) peak GRF values (N/kg), load and decay rates (N/kg/s) and joint moment values (Nm/kg) for the intact limb

	Amputee Non-Faller (n=5)	**Amputee Faller (n=4)**	**Control Non-Faller (n=5)**	**Control Faller (n=4)**
Posterior braking	-0.17 (0.06)	-0.22 (0.06)	-0.22 (0.06)	-0.14 (0.03)
Anterior propulsion	0.04 (0.02)	0.08 (0.03)	0.06 (0.02)	0.04 (0.02)
Fz1	1.10 (0.08)	1.33 (0.15) ¤	1.16 (0.11)	1.07 (0.10)
Fz2	0.79 (0.07)	0.57(0.10) ¤	0.73 (0.14)	0.81 (0.08)
Fz3	1.04 (0.07)	1.34 (0.11) ¤	1.20 (0.11)	1.22 (0.18)
Load rate	4.68 (1.08)	6.02 (1.40)	3.74 (1.15)	3.08 (0.68)
Decay rate	-5.26 (1.10)	-7.77 (0.54) ¤	-6.64 (1.06)	-8.09 (1.84)
Hip extensor moment loading	1.00 (0.20)	1.36 (0.23)	1.05 (0.40)	1.89 (0.35)
Hip flexor moment pre-swing	-0.23 (0.15)	-0.20 (0.15)	-0.30 (0.15)	-0.35 (0.18)
Knee extensor moment loading	0.97 (0.28)	1.29 (0.20)	1.05 (0.32)	0.89 (0.27)
Knee flexor moment pre-swing	-0.29 (0.10)	-0.39 (0.14)	-0.22 (0.21)	-0.08 (0.16)
Ankle plantarflexor moment loading	0.65 (0.13)	1.04 (0.58)	0.85 (0.25)	0.71 (0.12)
Ankle plantarflexor moment pre-swing	1.20 (0.18)	1.67 (0.30) ¤	1.44 (0.26)	1.29 (0.21)

Braking force -ve; Propulsion force +ve; Fz1 – first vertical peak; Fz2 – valley; Fz3 – second vertical peal; Load rate +ve; Decay rate -ve; Knee and hip extensor moments +ve; Ankle plantarflexor moment +ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

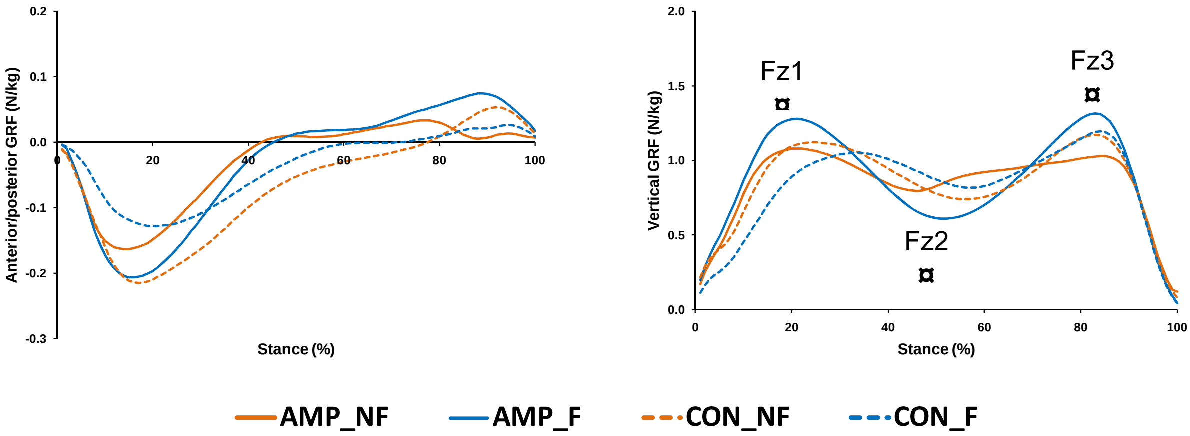


Figure 6.8. Anterior/posterior and vertical ground reaction forces (N/kg) of the intact limb during stair ascent from the first to the third step in amputee and control non-fallers and fallers. Averaged data are presented. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line).The stance phase is initiated with foot contact and terminated with toe off.

Anterior and vertical forces +ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

All joint moment profiles are presented in Figure 6.9. No differences were noted at either the hip or knee joints. There were no differences in peak ankle plantarflexor moments in the loading response, but significant findings were evident in the pre-swing phase ($F_{(3,14)}$ = 3.16, p=0.05). Post-hoc analysis showed the amputee fallers experienced a significantly greater ankle plantarflexor moment compared to the non-fallers (p=0.01), while there were no differences between the control groups.

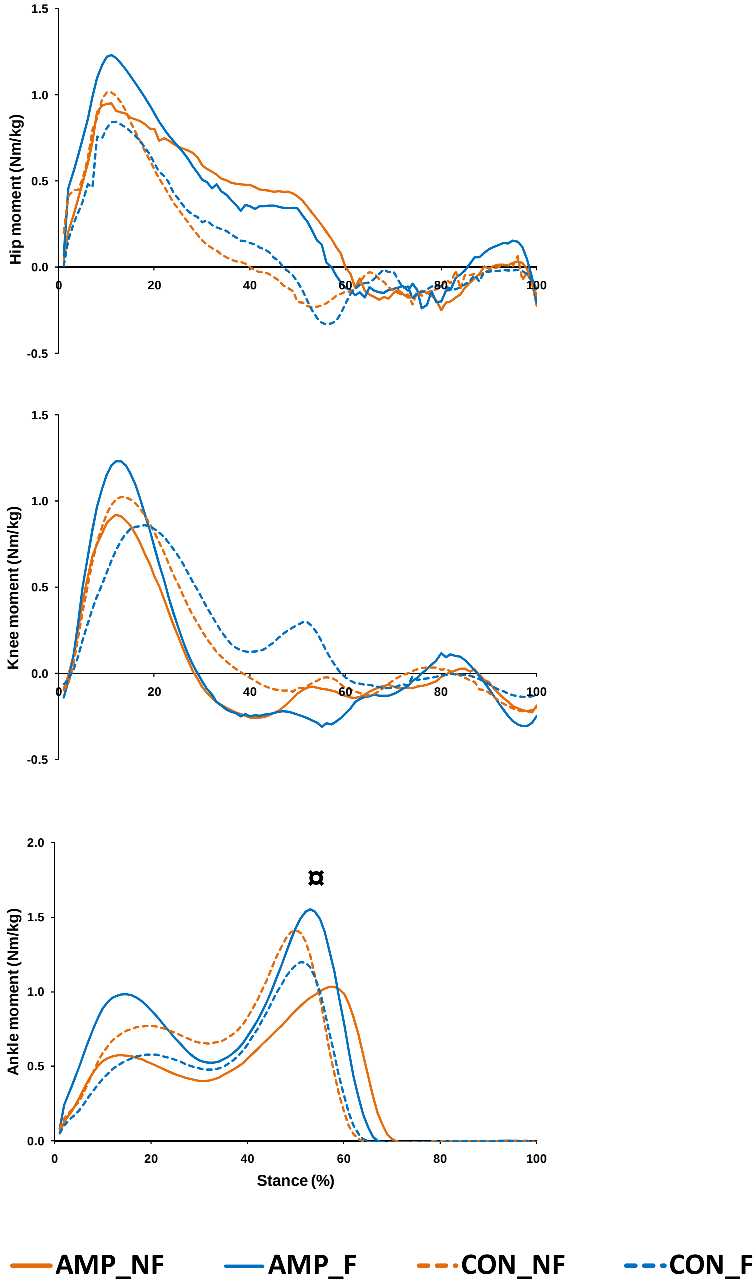


Figure 6.9. Joint flexor/extensor moments (Nm/kg) of the intact limb during stair ascent in fallers and non-fallers. Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line). The gait cycle is initiated and terminated with foot contact.

Extensor and plantarflexor moments +ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

Peak joint power values and profiles are presented in Table 6.6 and Figure 6.10, respectively. Peak power bursts for stair ascent were defined according to McFadyen & Winter (1988) and are presented in Figure 2.3 (section 2.3.3). No power differences were found at the hip. A significant difference was noted for power generation in mid-stance at the knee (labelled K2) ($F_{(3,14)}$ = 7.67, p=0.00). Post-hoc analysis revealed that the larger K2 burst in the amputee faller group (compared to the amputee non-fallers) only showed a trend towards significance (p=0.09). In the control groups, the fallers exhibited a trend towards a smaller K2 power burst compared to the control non-fallers (p=0.07). The ANOVA results for power absorption at the knee in mid- and late swing (labelled K3 and K4) were significant ($F_{(3,14)}$ = 9.11, p=0.00 and $F_{(3,14)}$ = 6.50, p=0.01) and post-hoc analysis showed that the amputee fallers experienced larger power absorption bursts (p=0.00 and p=0.01, respectively) in mid- and terminal swing compared to the amputee non-fallers. No differences were found for power absorption (A1) or power generation (A2 or A3) bursts at the ankle joint.

Table 6.6. Amputee vs. control fallers and non-fallers mean (SD) peak joint powers (W/kg) (intact limb).

	Amputee Non-Faller (n=5)	Amputee Faller (n=4)	Control Non-Faller (n=5)	Control Faller (n=4)
Hip power generation pre-swing (H1)	1.38 (0.10)	1.67 (0.30)	1.66 (0.99)	1.14 (0.40)
Hip power generation initial swing (H3)	0.69 (0.29)	0.80 (0.16)	0.52 (0.25)	0.40 (0.17)
Knee power generation loading (K1)	1.41 (0.66)	2.29 (0.52)	1.67 (0.79)	1.24 (0.47)
Knee power generation initial swing (K2)	0.56 (0.18)	0.81 (0.29)	0.14 (0.16)	0.43 (0.22)
Knee power absoprtion mid-swing (K3)	-0.06 (0.07)	-0.35 (0.13) ¤	-0.16 (0.10)	-0.04 (0.04)
Knee power absoprtion late swing (K4)	-0.62 (0.25)	-1.04 (0.15) ¤	-0.55 (0.16)	-0.39 (0.29)
Ankle power absorption loading (A1)	-0.28 (0.11)	-0.74 (0.54)	-0.55 (0.30)	-0.34 (0.25)
Ankle power generation mid-stance (A2)	0.30 (0.18)	0.85 (0.85)	0.44 (0.29)	0.21 (0.18)
Ankle power generation pre-swing (A3)	2.37 (1.08)	4.11 (1.24)	3.14 (0.94)	2.70 (0.93)

Power generation +ve; Power absorption –ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

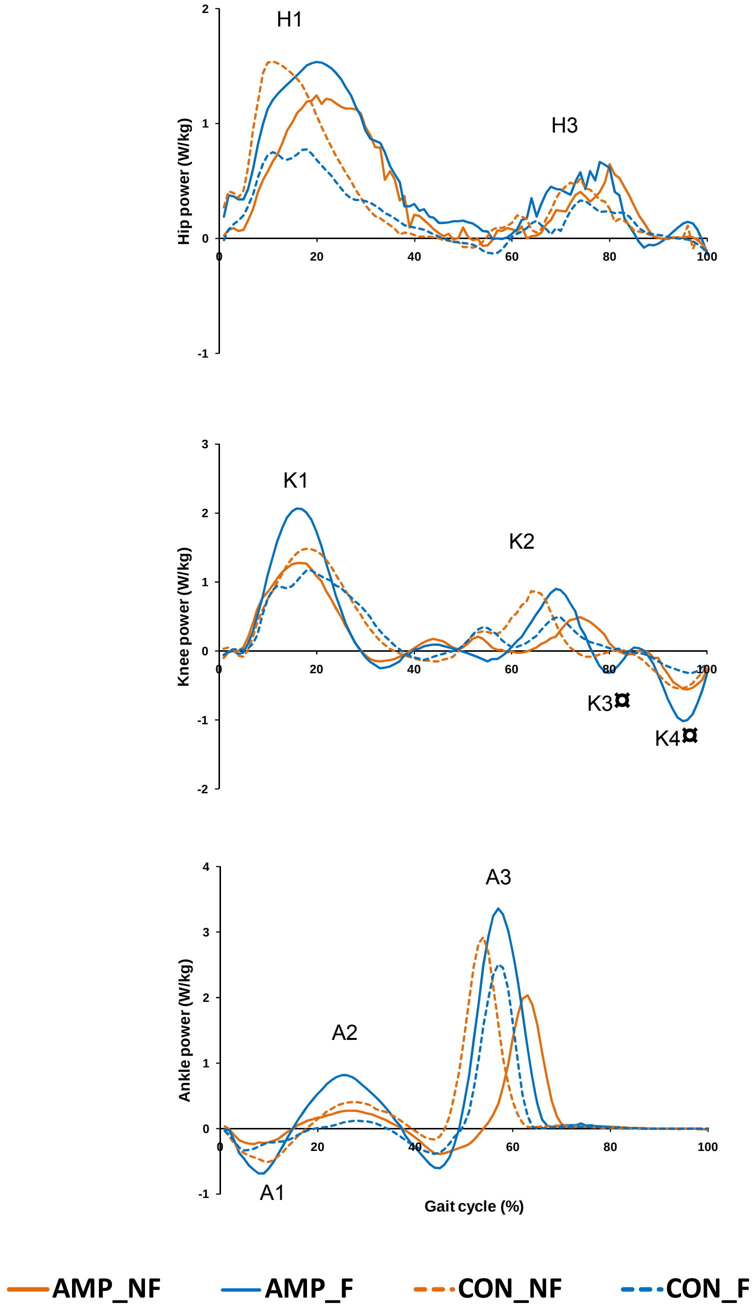


Figure 6.10. Sagittal plane joint powers (W/kg) of the intact limb during stair ascent from first to third step in fallers and non-fallers. Averaged data are presented: Amputee non-fallers (solid red line); Amputee fallers (solid blue line); Control non-fallers (dashed red line); Control fallers (dashed blue line). The gait cycle is initiated and terminated with foot contact.

Power generation +ve; Power absorption –ve

¤ Indicates significant difference between amputee fallers and non-fallers (one-way ANOVA)

6.3.5. 'Step to' group (floor to step transition)

There were two amputee fallers who self-selected a 'step to' gait pattern during stair ascent. Kinematic data are presented in Table AC1 and Figure AC1 in Appendix C. There was considerable variation between the two participants, most particularly in the foot kinematics. Both individuals (AMP_F 5 and AMP_F 6) led with the intact limb at all times. One participant (AMP_F 5) even stopped and re-adjusted their pattern if they started to lead with the prosthetic foot (this happened on one occasion). Both participants had larger toe clearance with the intact foot.

While both participants exhibited greater peak hip and knee flexion angles on the lead, intact limb compared to the affected limb, between-subject values were similar. The prosthetic ankle remained dorsiflexed as determined by the mechanical properties of the artificial foot. However, variation was observed at the ankle joint of the intact limb. AMP_F5 maintained both the intact and prosthetic ankle in continuous dorsiflexion, while AMP_F 6 showed a small amount of ankle plantarflexion in early swing. Conversely, AMP_F 5 displayed greater peak dorsiflexion on the intact limb in mid-swing. The intact ankle ROM was similar for both participants. Peak hip and knee flexion was reduced by almost 20° on the affected limb.

6.3.6. 'Step to' group (first to second step)

Adopting a 'step to' stair ascent pattern was particularly reflected in the slow walking speed, which was in the range of 50% slower than the groups who adopted a step over pattern. Peak kinematic values (Table AC2) and profiles (Figures AC2 and AC3) are presented in Appendix C. Both participants chose considerably different walking speeds. AMP_F 5 and AMP_F 6 continued to display greater peak hip and knee flexion angles and ROM on the intact limb compared to the affected limb. After initial foot contact of the intact limb, both participants extended the knee slowly, which was reflected in the greater knee flexion seen during the loading response. The intact knee became fully extended in double support when the prosthetic foot made foot contact with the same step. The knee extensors played a crucial role during the single stance phase of the intact limb as they were responsible for keeping the lower-limbs from collapsing. Neither participant displayed plantarflexion of the intact foot.

6.4. Discussion

The aim of this study was to compare the gait patterns of fallers and non-fallers during stair ascent in transtibial amputees and able-bodied participants. All of the participants were able to complete the task successfully, although two amputee fallers (AMP_F 5 and AMP_F 6) self-selected a 'step to' rather than a step over walking pattern.

6.4.1. Floor to first step (lead limb - intact limb amputee groups) – swing only

A successful transition from level to stair walking is the first goal in safe stair locomotion and it requires adequate foot clearance over the nose of the step when positioning the lead foot on the first step. Previous studies have recognised that the risk of tripping is related to contact of the lead foot with the obstacle (in this case a step) during the initial swing phase (Benedetti *et al*., 2007). Both the amputee and control groups had safe toe clearance levels. There were no significant differences in peak joint angles during swing (Table 6.2). Because the lead limb was the intact limb in the amputee groups, shorter swing duration of the intact limb would also reduce the single support phase of the affected limb. The amputee non-fallers could have actively attempted to reduce single support time on the affected limb during the floor to stair walk transition and this may be a successful strategy to avoid trips occurring with the lead (intact) and trail (affected) limbs. Equally, the control fallers could simply have tried to reduce single support time as this is when the strength of the supporting leg is crucial in avoiding collapse.

6.4.2. Floor to second step (trail limb - affected limb amputee groups)

During this step, the affected limb makes the transition from floor to step and must ascend a greater vertical distance (2 steps) than the lead limb. There were no significant differences in walking speed during level walking. However, during a more mechanically challenging task, such as stair ascent, significant differences were observed between the fallers and non-fallers for both groups. These findings support the notion that more biomechanical differences between fallers and non-fallers would be observed during more complex daily activities (Lee and Chou, 2007).

The majority of significant differences in peak joint kinematics and ROM were between the control fallers and non-fallers. The ankle joint plays an integral role in stair walking, especially during pre-swing when it is responsible for lifting the body (McFadyen and Winter, 1988). It was not surprising that no differences were found between the amputee groups, as the prosthetic ankle is aligned in dorsiflexion and did not move into plantarflexion and ROM is limited and disadvantaged by the prosthetic components

(Powers and Boyd, 1999). The significantly reduced ankle plantarflexion observed between the control fallers and non-fallers supports previous research that the ankle operates close to its maximal limits during stair walking (Reeves *et al*., 2008) and that plantarflexor strength is significantly reduced in older vs. younger individuals and particularly elderly fallers (Kerrigan *et al*., 1999).

Amputee and control fallers had a significantly larger knee ROM than their non-faller counterparts. Stair ascent demands greater joint mobility and flexibility than level walking (McFadyen and Winter, 1988; Reeves *et al*., 2008a) and therefore exhibiting larger ROM could be viewed as a safe strategy and would not likely cause a fall. Since there is no published literature to date on the kinematic differences between amputee fallers and non-fallers during stair walking, it is not clear whether the greater ROM was a compensatory, more cautious strategy as a result of a previous fall. For example, the participants may have believed that a more extended knee during loading and more flexed knee in swing would reduce the likelihood of a future fall during stair ascent. Knee ROM on the affected side will vary according to biological factors and prosthetic socket fit. While the current study did not measure knee ROM passively, it is acknowledged this could have influenced the results.

Compared to level walking, the hip did not reach full extension for any of the groups (e.g. hip extension angles remained positive). Increased hip ROM during stair ascent is related to increased hip flexion during swing when compared to level walking. The amputee non-fallers showed the least hip extension, despite this result not being significantly different with the amputee fallers. The control fallers also experienced greater hip flexion compared to the non-fallers, which is consistent with other studies (Benedetti *et al*., 2007). Hip flexion is influenced by anterior pelvic tilt. The results revealed significantly more anterior pelvic tilt in the corresponding groups that showed significantly more hip flexion. Greater forward trunk lean could be related to participants looking down at the feet (Benedetti *et al*., 2007). Indeed, this was quite possible among the amputee groups, who have lost the proprioceptive feedback from the prosthetic ankle and foot. Increased use of visual information may be considered a good adaptive strategy for controlling foot position and avoiding a trip during swing. Hip and trunk flexion could also be related to handrail use. The majority of amputee non-fallers used the handrail to some level, whereas several amputee fallers used the handrails only lightly (see Table 6.1). Reeves *et al*. (2008b) reported that the use of handrails had little effect on joint kinematics but improved dynamic postural stability. Although evidence of a 'handrail effect' may have been apparent in the hip joint kinematics, these authors did not report trunk or pelvic kinematics (Reeves *et al*., 2008b). In the current study, the use of handrails could be considered a safe strategy, as it would provide an additional point of contact during the more vulnerable single support phase of the affected limb.

Interestingly, there were no significant differences in hip and pelvic frontal plane kinematics. Increased frontal plane motion has been linked with greater challenges in postural control during stair walking (Lee and Chou, 2007) and therefore one might have expected differences between fallers and non-fallers.

Although it is beyond the scope of this thesis to compare amputees with control participants, there has been no data examining frontal plane hip and pelvic kinematics during stair walking in amputees and controls. It is worth noting how the amputees (fallers and non-fallers) had greater hip adduction on the affected side during swing just prior to subsequent foot contact and this was related to the greater pelvic hike (pelvic obliquity up) on the ipsilateral side when compared to the controls (Table 6.3). It was also considered a compensatory strategy for the reduced knee ROM on the affected side (when compared to the controls) and lack of active plantarflexion used to lift the prosthesis into swing.

6.4.3. First to third step (lead limb – intact limb amputee groups)

Walking speed continued to decrease as participants continued to ascend the stairs. The same pattern of significant results between the fallers and non-fallers was observed as during the floor to 2nd step. Walking speed from the first to third step was similar to previous results investigating transtibial amputees during stair ascent (Powers and Boyd, 1997). Those authors measured walking speeds of 0.49 m/s and 0.56 m/s on a 4-step staircase in transtibial amputees and control groups, respectively. These observations continue to support the findings that the amputee fallers had comparable gait kinematics as the control non-fallers; conversely, the amputee non-fallers had very similar results to the control fallers.

It was interesting to note that fewer significant differences in peak joint kinematics were found from the first to the third step. This suggested that, in the amputee group, differences in the affected but not the intact limb distinguished between fallers and non-fallers. It was unusual that the same kinematic differences as moving from the floor to the second step were not observed in the control group. The only difference between the control groups was that the fallers continued to exhibit significantly more anterior pelvic tilt. This was attributed to similar reasons as discussed in the section during the step from floor to 2nd step.

These findings provide strong evidence to suggest that the joint kinematics between fallers and non-fallers (amputees and controls) differ in the two steps that transition from level walking on the floor to stair walking: floor to first step and floor to second step. The results of Hamel *et al*. (2005) support a similar notion, but were measured during stair descent. Using a 7-step staircase, these authors reported that minimum foot clearance was always greatest on the bottom step, when transitioning from stair to level walking, and that the elderly participants had significantly lower minimum foot clearance compared to the young participants. It was unclear whether there were other differences as these authors did not measure joint kinetics. It appears that people who are most vulnerable to falling during stair ascent need to become particularly aware of the lower-limb kinematics during the transition phases between level and stair walking. This includes using handrails and vision in the absence of somatosensory feedback for correct foot placement.

Furthermore, it has been found that 'light' handrail use has little effects on the joint kinematic parameters but that it affects kinetic data to some level by redistributing the joint moments across the ankle and knee joints (Reeves *et al*., 2008b). Therefore, some caution should be used when interpreting the kinetic results as handrail use varied across participants. On the other hand, one could argue that the kinetic data would be a true representation of the participants' stair walking biomechanics.

An important finding was observed for vertical GRF, where the amputee fallers displayed significantly larger peak forces (first and second peak values, Fz1 and Fz3, respectively) and reduced mid-stance force (valley, Fz2), and larger decay rate on the intact limb compared to the non-fallers. No differences were observed between the control groups. For the amputee non-fallers, the Fz1 value was very similar to that reported by Schmalz *et al*. (2007) for the intact limb in transtibial amputees (1.16 N/kg). Furthermore, the peak Fz1 force in the amputee fallers was even higher than the control non-fallers in the current study. This was a meaningful result and consistent with the previous study (Chapter Five) that demonstrated that amputee fallers experienced larger peak vertical GRFs on the affected limb during level walking. Larger vertical forces experienced in the intact leg were reflective of the amputee fallers' significantly faster walking speed, which may be considered a risk for falls.

The smaller valley peak (Fz2) suggested the affected limb was moving rapidly in swing as the force plate was significantly unloaded during intact single support (Perry, 1992). A larger decay rate on the intact limb indicates the affected limb would have to control larger forces when in single support. This finding could be linked with an increased risk of falling as the GRFs and load rate of the affected limb, as it made initial contact on the step above, would be larger if the affected limb moved rapidly in swing. It also highlights the need for strong extensor muscles in the intact limb when in single support. However, an instrumented staircase, with force plates on at least two subsequent steps, would be needed to confirm this observation.

Peak joint moments were consistently highest in the amputee fallers group, with values considerably greater than those reported for the sound limb in transtibial amputees during stair ascent (Schmalz *et al*., 2007). In the current study, the only significant difference for joint moments between the amputee groups was measured at the ankle joint in pre-swing. However, the amputee fallers had a larger (albeit non-significant) ankle plantarflexor moment in loading suggesting that they made contact with the more distal aspect of the forefoot in the weight acceptance phase, which is when the ankle positions the body for the pull-up phase. As a result, an important difference between the amputee fallers and non-fallers was that the non-fallers were able to maintain the GRF vector closer to the ankle and knee joints. Thus, the intact limb was in a stable position when the prosthetic foot was initiating swing (pull-up phase) during what is normally the greatest point of instability (McFadyen and Winter, 1988). If the prosthetic foot were to make contact on the nose of the intermediate step, the intact leg might be more successful at recovering when in a stable position.

The knee and hip joint moment profiles were very similar between fallers and non-fallers in each group. Visual inspection of the knee joint moment graph (Figure 6.9) shows the control fallers maintained a knee extensor moment throughout stance, while all other groups displayed a flexor moment during mid- to late stance (forward continuance phase). Towards the end of the pull-up phase, all three lower-limb joint angles are flexed. When the GRF vector is positioned behind the knee joint, the knee extensors must be strong to prevent the knee from collapsing. In those individuals prone to knee extensor weakness, this stair walking strategy would place the person in a vulnerable position.

During stair ascent, joint power profiles showed the distinct power bursts as described by McFadyen and Winter (1988). The largest power burst for all groups was measured at the ankle joint in pre-swing (forward continuance phase). McFadyen and Winter (1988) provided evidence that ankle energy generation was not the main source of forward progression during stair ascent. Concentric work of the ankle plantarflexors was responsible for lifting the body overtop the contralateral supporting leg and only some translation.

The amputee groups had to make compensatory adjustments for the missing plantarflexor muscles. The current results were consistent with the previous literature. Powers and Boyd (1997) found that the joint moments and power profiles were very similar in the intact leg of their transtibial amputees and the non-amputee controls. They cited the main difference at the ankle, where the intact leg generated more power to propel the trunk upwards and overtop the contralateral (affected) leg and reported a dominant knee extensor strategy for the intact limb. Schmalz *et al*. (2007) reported that the ankle and knee joints of the intact limb compensate for the mechanical limitations of the prosthetic foot and ankle. Specifically, they reported increased plantarflexion in the pre-swing phase and rapid knee extension during mid-stance (pull-up phase).

According to McFadyen and Winter (1988), evidence of the triphasic power burst at the knee (labelled K2, K3 and K4) is a crucial aspect to controlled stair ascent. Interestingly, the amputee fallers showed a trend towards larger power generation in mid-stance (K2) compared to the non-fallers, which is when the hamstring muscles become active to flex the knee at toe-off, followed by significantly larger power absorption bursts K3 and K4 in mid- to late swing (eccentric control of the rectus femoris to reverse knee flexion) (McFadyen and Winter, 1988). This suggests that the triphasic muscle action (hamstring power generation – rectus femoris power absorption – hamstring power absorption), together with power generation at the hip in initial swing (H3), was important during the swing phase. In the current study, the amputee fallers experienced significantly larger power absorption at the knee in mid-swing (K4) compared to the non-fallers and both amputee groups had larger K4 bursts compared to the control groups. This finding is consistent with one other study that reported activation of the hamstrings in late swing on the intact leg in amputees but not in controls (Schmalz *et al*., 2007).

McFadyen and Winter (1988) believed the hip joint moment and power profiles exhibited less stereotypical patterns than the ankle and knee. This was because the hip musculature was responsible for controlling the head, arms and trunk (HAT) segment and there was greater between-subject variability in how people carried the passenger unit during stair ascent. Exaggerated movements of the trunk in the sagittal and frontal planes have been observed in older adults (Lee and Chou, 2007). They believed this was indicative of postural disturbances.

Previous studies have reported hip-extensor dominant strategies on the affected limb (Powers and Boyd, 1997; Schmalz *et al*., 2007) with a tendency towards reduced peak values on the intact limb (Schmalz *et al*., 2007). Conversely, Powers and Boyd (1997) reported similar work values at the knee and hip on the intact side compared to controls. Because the hip extensors play an important role in raising body weight, Powers and Boyd (1997) believed hip extensor weakness may explain the inability to walk with a reciprocal gait pattern (e.g. step over pattern). In the current study, the H1 power burst was comparable for both amputee groups compared to healthy adults (1.79 W/kg) (Nadeau *et al*., 2003). The results from the current study appear to agree with Powers and Boyd (1997) but contradict Schmalz *et al*. (2007). However, without kinetic data for the affected limb, it is uncertain whether the intact limb showed reduced hip extensor activity.

6.4.4. 'Step to' group

Transtibial amputees are taught to lead with their intact limb during stair ascent because of the greater joint ROM demanded by the lead limb. It is probable the two amputee participants adopted a 'step to' gait pattern because it reduced the mechanical demands on the lower-limbs. Considerably less control of the prosthetic ankle was required and therefore it performed below its functional limits (e.g. the prosthetic ankle did not move through the ROM observed during level walking). The ROM of the intact ankle was also reduced in a 'step to' gait pattern because the ankle plantarflexor muscles did not need transport the body such a large vertical distance.

Unfortunately, kinetic data could not be analysed on those people who adopted a 'step to' gait pattern. However, based on the kinematic findings, it may be concluded that the 'step to' group adopted this pattern due to a combination of factors, including reduced prosthetic ankle ROM, insufficient knee extensor strength on the affected limb and overall reduced ROM of the knee and hip on the affected side, which may all predispose a transtibial amputee to adopt a 'step to' gait pattern during stair ascent. It is also worth noting that the 'step to' group was heavily reliant on handrail support with both arms during stair ascent. This would have reflected an effort to reduce the strength demands on the extensor muscle groups of the lower-limbs, as well as increase the base of support. Individuals with a fear of falling or low falls efficacy may also adopt this slower and more cautious stair walking strategy.

6.5. Conclusions

In summary, stair ascent was a more difficult task than level walking and the findings supported the hypothesis that stair ascent placed more functional demands on the lower-limbs. This was evidenced by the slower walking speed, greater range of motion at the ankle, knee and hip joints, larger GRFs, joint moments and powers. The results supported the hypothesis that the fallers would walk more slowly and exhibit reduced

ROM, but only for the control group. The contrary was noted in the amputee groups, where the fallers in fact walked more quickly and did not show any differences in joint ROM. The results also demonstrated that kinematic differences existed between amputee fallers vs. non-fallers on the transition of the affected (trail) leg from level to stair walking. Similarly, in the control groups, many of the kinematic differences between the fallers and non-fallers were observed in all three lower-limb joints during the transition of the trail limb from the floor onto the second step. These results supported the first hypothesis. Contrary to the second hypothesis, the kinetic results demonstrated that the amputee fallers exhibited larger GRF and knee power bursts compared to the non-fallers. This was related to the faster walking speed in the amputee fallers. However, no kinetic differences were noted at the hip joint or between the control groups. The amputee fallers demonstrated similar walking patterns to the control non-fallers, whilst the amputee non-fallers appeared to walk more cautiously. Some participants adopted safe stair walking strategies, such as handrail use and ‘step to’ gait patterns.

CHAPTER SEVEN - GAIT PATTERNS IN TRANSTIBIAL AMPUTEE FALLERS VS. NON-FALLERS: BIOMECHANICAL DIFFERENCES DURING STAIR DESCENT

7.1. Introduction

Like stair ascent, walking down stairs involves the rhythmic shift of body weight in the vertical and horizontal directions. Stair descent is characterised by large eccentric forces from the ankle plantarflexors and knee extensors during the weight acceptance (loading) and controlled lowering (pre-swing) phases (McFadyen and Winter, 1988). The controlled lowering phase is accomplished through large eccentric muscle forces, particularly about the, knee, and corresponds to a phase in the gait cycle when failure could result in a fall (Beaulieu *et al*., 2008).Accidents that occur during stair negotiation are more likely to occur during stair descent than ascent (Reeves *et al*., 2008b; Svanstrom, 1974). Difficulties descending stairs have been linked with poor balance and gait abnormalities in non-disabled older adults (Verghese *et al*., 2008). Reeves *et al*. (2008b) have shown that older adults function close to their biomechanical limits during stair descent. Combined with reduced joint ROM and muscle strength, dynamic stability during stair descent is compromised in the elderly.

Compared to the able-bodied population, there are fewer studies that have conducted biomechanical investigations in transtibial amputees descending stairs and the mechanical adaptations they make during this more complex task is not as well understood. Schmalz *et al*. (2007) reported that transtibial amputees maintain the knee extended on the affected side for a longer period of time to compensate for the loss of the dorsiflexor and plantarflexor muscle groups. They also noted that the amputees 'fall' onto the intact leg and the authors considered this a compensatory movement related to the excessive loading at the ankle and knee joints.

There is a lack of research into stair descent in transtibial amputees and especially in relation to falls. The aim of this study was to compare the gait patterns of fallers and non-fallers during stair descent in transtibial amputees and able-bodied participants. It was hypothesised that amputee and control fallers would descend stairs more slowly than the non-fallers. It was also anticipated the fallers would exhibit reduced joint mobility, compared to the non-fallers and that this would be especially evident at the intact ankle and knee on the affected limb in the amputee groups. This was because stair descent places greater demands on the ankle and knee joints and musculature than level walking or stair ascent (McFadyen and Winter, 1988; Powers and Boyd, 1999; Reeves *et al*., 2008a). A second hypothesis was that amputee fallers and control fallers would exhibit reduced joint moment and power at the ankle and the knee. This was based on the findings that the knee extensors perform important eccentric work during the controlled lowering phase (labelled K3), mid-stance phase during stair descent and failure of adequate power absorption at the knee joint could result in failure to keep the body upright (Beaulieu *et al*., 2008; McFadyen and Winter, 1988). Insufficient knee extensor eccentric strength during the single support phase was considered a major risk factor for falls. No differences were expected in hip joint moments or powers between the fallers and non-fallers because these parameters are relatively small and highly variable during stair descent (Beaulieu *et al*., 2008).

7.2. Methods

7.2.1. Participants

Eleven transtibial amputees and nine age-matched able-bodied participants completed the stair ascent task. All participants gave written informed consent to take part in this study. Individual participant details according to falls history can be found in Table 3.1.

7.2.2. Staircase

The purpose-built 3-step wooden staircase that was built for this study is described in section 6.2.2.

7.2.3. Experimental protocol

Each participant completed the stair descent trials as part of the stair locomotion task described in section 6.2.3. Participants stood on the landing and were asked to descend the stairs using their preferred pattern and pace and continue level walking once they reached the ground. Because the landing was 80 cm deep, participants took at least one step to reach the edge of the top step before initiating stair descent. No instructions were given about lead limb. They were told they could use the handrails if they felt necessary. Data collection was completed once participants completed the stair ascent trials described in section 3.15.2.

7.2.4. Data analysis

All data were analysed as explained in sections 3.16, 3.17, and 3.18. Unlike in stair ascent, the gait cycle for all stair descent trials was initiated and terminated with toe off. This was to ensure two full strides could be compared. The GRF data were presented for the stance phase (foot contact to toe off) only. Data for the affected limb was only compared between two groups: the amputee fallers and non-fallers. Data for the intact limb was compared across four groups: amputee and control fallers and non-fallers. The variables that were selected for analysis included temporal-spatial, peak joint kinematic and joint ROM, peak GRF data, load and decay rates, peak joint moments and power bursts as described in section 3.19. Kinetic data could only be calculated for the lead limb from the top step to the second step.

7.2.5. Lead limb preference

As in stair ascent, not all participants used the same stair descent strategy. The data were first examined for lead limb preference (Table 7.1). During stair descent 9 of 11 amputees displayed a preference for leading with the affected limb. Two participants did not exhibit a clear preference. One participant who had no lead limb preference in stair descent also had no limb preference in stair ascent. For the controls, three participants had no limb preference; five had a left limb preference and one a right limb preference.

Table 7.1. Step and limb characteristics during stair descent in the amputee and control participants

Amputee participants	Lead limb preference	Handrail use	Step pattern
Faller			
1	Affected	Light	Step over
2	Affected	Light	Step over
3	Affected	Light	Step over
4	Affected	Light	Step over
5	Affected	Reliant	Step to
6	Affected	Reliant	Step to
Non-faller			
7	Affected	Moderate	Step over
8	Intact	Moderate	Step to
9	Affected	Moderate	Step to
10	None	Light	Step over
11	Affected	Light	Step over
Control participants			
Faller			
1	None	Light	Step over
2	None	Light	Step over
3	None	Light	Step over
4	Right	Moderate	Step over
Non-faller			
5	None	Light	Step over
6	Right	No	Step over
7	Right	No	Step over
8	Left	No	Step over
9	Right	No	Step over

'Light' handrail use was classified as using the handrail as a guide only (Reeves *et al.*, 2008a). In the current study, light handrail meant that participants held the handrail with one hand only.

'Moderate' handrail use occurred when participants used both arms as a guide, but did not perform a large portion of the work with their arms.

'Reliant' handrail use occurred when participants performed considerable work with their arms and, when asked, would not have felt safe without the handrails.

Each limb descended the vertical height of two steps, representative of steady state stair descent: top to second step (trail limb) and first 'step to' floor (lead limb) (Figure 7.1). Kinematic comparisons could be made between the two limbs, but kinetic data were collected for the lead limb only. For the amputee participants who did not express a lead limb preference, the affected limb was analysed as the lead limb during descent. For the control participants, the lead limb was analysed based on the participants' preference. The average of both limbs was computed and analysed for the control participants who did not show a lead limb preference.

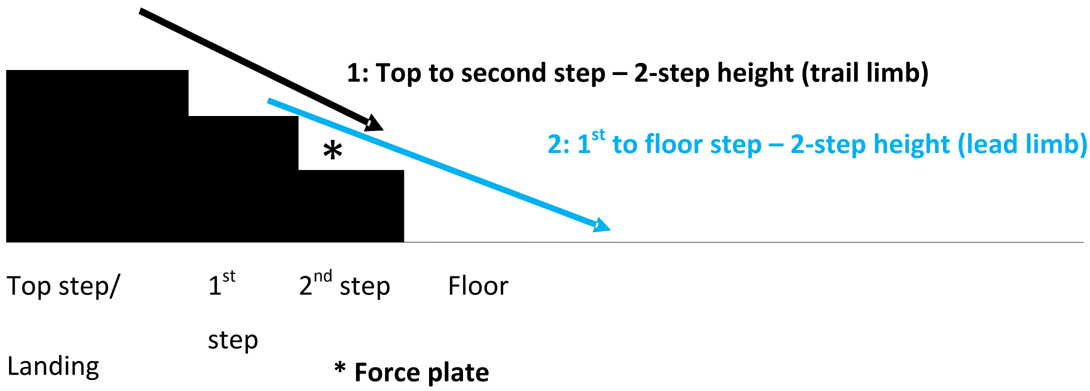


Figure 7.1 Illustration of staircase and steps selected for analysis during stair descent. Solid blue line represents lead limb. For the amputee participants, the lead limb was the affected limb. Solid black line represents the trail limb. For the amputee participants, the trail limb was the intact limb

* Indicates position of force plate

7.2.6. Statistical analysis

Kinetic data could not be analysed for four amputee participants because both feet made contact with the force plate. Moreover, one amputee non-faller did not display an affected lead limb preference and thus, no kinetic data were collected for the intact limb. This reduced the amputee non-faller group from n=5 to n=2 for kinetic variables. Therefore, it was not deemed appropriate to conduct statistical analysis on any of the stair descent data. Given that the amputee participants displayed rather unique stair descent strategies, the following results section uses descriptive statistics to compare the groups and illustrate the individual and average data for the amputee participants

7.3. Results

7.3.1. Temporal-spatial data

The data here are presented for 16 participants who used a step over or alternate step pattern. The control non-fallers walked quickest with a walking speed of 0.77 m/s, while the amputee non-fallers were the slowest at 0.50 m/s. The amputee fallers walked 44% (0.22 m/s) quicker than the amputee non-fallers. The opposite was found in the control group where the control non-fallers walked 45% (0.24 m/s) quicker than the control fallers (Table 7.2). In the amputee groups, intact stance duration was longer than stance on the affected limb. Stance duration was relatively similar for all groups and ranged from 57-63% of the gait cycle.

Table 7.2. Amputee vs. control fallers and non-fallers mean (SD) temporal-spatial and peak kinematic values (°) during stair descent (sagittal plane). Affected limb data represents first step to floor; intact limb data represents top to second step.

	Amputee Non-Faller (n=3)		Amputee Faller (n=4)		Control Non-Faller (n=5)	Control Faller (n=4)
	Affected	**Intact**	**Affected**	**Intact**		
Speed (m/s)	0.50 (0.06)		0.72 (0.12)		0.77 (0.17)	0.53 (0.10)
Stance phase (%)	58 (1)	63 (4)	57 (2)	60 (2)	59 (4)	57 (1)
Hip angle toe off (°)	49.8 (15.7)	54.2 (4.5)	40.7 (6.2)	37.3 (7.2)	30.5 (7.4)	39.4 (8.5)
Hip flexion swing (°)	51.0 (17.0)	58.3 (6.3)	49.5 (7.3)	47.7 (4.9)	40.9 (8.1)	50.0 (9.4)
Hip angle foot contact (°)	33.8 (13.6)	31.2 (1.6)	25.6 (3.9)	23.5 (5.1)	23.3 (4.9)	29.0 (8.8)
Hip extension stance (°)	19.7 (13.6)	19.3 (6.1)	-0.7 (2.9)	10.5 (5.8)	1.6 (4.1)	12.3 (10.8)
Hip ROM (°)	31.2 (8.3)	39.1 (6.6)	50.2 (9.1)	37.2 (7.5)	39.3 (5.3)	37.6 (6.4)
Knee angle toe off (°)	87.8 (5.9)	86.2 (2.4)	89.0 (3.9)	88.5 (6.4)	90.0 (8.1)	93.6 (3.5)
Knee flexion swing (°)	88.1 (5.6)	86.7 (1.5)	92.0 (5.0)	92.9 (5.6)	93.7 (10.1)	97.1 (6.8)
Knee angle foot contact (°)	12.2 (6.6)	4.6 (1.3)	6.4 (5.2)	8.4 (4.1)	10.5 (4.0)	14.2 (6.7)
Knee ROM (°)	78.8 (4.1)	83.9 (2.8)	86.9 (7.5)	87.1 (5.8)	85.9 (7.8)	86.0 (4.5)
Ankle angle toe off (°)	4.7 (2.9)	2.7 (10.9)	6.3 (3.7)	10.3 (5.0)	14.2 (8.1)	19.1 (6.0)
Ankle plantarflexion swing (°)	3.8 (2.9)	-19.2 (8.7)	5.0 (3.5)	-23.6 (4.8)	-15.9 (10.1)	-15.8 (4.3)
Ankle angle foot contact (°)	3.8 (2.9)	-17.7 (8.5)	5.6 (4.2)	-20.5 (3.5)	-12.1 (10.0)	-9.8 (4.9)
Ankledorsiflexion stance (°)	15.7 (3.1)	25.6 (12.3)	15.8 (2.5)	29.2 (8.9)	25.7 (7.9)	28.4 (6.4)
Ankle ROM (°)	12.0 (3.0)	44.8 (20.4)	10.8 (1.1)	52.8 (6.0)	41.7 (3.6)	44.2 (4.2)
Pelvic tilt toe off (°)	20.8 (1.0)	24.2 (3.4)	15.3 (0.9)	14.7 (1.0)	9.6 (4.0)	16.4 (5.0)
Pelvic tilt swing (°)	23.4 (1.0)	26.2 (4.1)	15.7 (1.2)	20.0 (3.7)	11.8 (4.4)	16.9 (4.9)
Pelvic tilt foot contact (°)	21.8 (0.5)	23.1 (3.0)	14.6 (2.2)	18.2 (1.8)	11.3 (4.8)	14.8 (4.4)
Pelvic tilt stance (°)	22.6 (1.1)	24.1 (2.2)	16.6 (2.5)	18.2 (1.8)	13.6 (4.4)	19.1 (3.5)

Hip extension -ve; Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion -ve; Anterior pelvic tilt +ve

† It is noteworthy that the prosthetic ankle is aligned in dorsiflexion and thus did not achieve plantarflexion during swing in stair descent

7.3.2. Joint kinematics

Peak sagittal plane joint kinematics are presented in Table 7.2., joint profiles are displayed in Figures 7.2 to 7.4 and pelvic tilt in Figure 7.5. The hip angle profiles showed high variability in the amputee non-faller group, especially for the affected limb and the intact limb during stance. Notable differences were also found during peak hip extension during controlled lowering on the affected side. While the amputee fallers displayed full hip extension, the amputee non-fallers showed almost 20° of flexion. The opposite trends were found for the control groups, where the fallers maintained the hip more flexed throughout and never exhibited full hip extension.

Less obvious kinematic differences were found at the knee joint. All groups initiated toe off with the knee considerably flexed. For the amputee non-fallers, the range of motion of the knee on the affected side was reduced by over 5° compared to the intact side. For the control groups, the knee flexion profiles and peak values were almost identical.

Amputee participants generally initiated swing with the ankle in dorsiflexion at toe off. On the intact side, the amputee fallers showed the greatest degree of ankle plantarflexion and, on average 23% (-4.4°), more than the amputee non-fallers. The prosthetic ankle remained dorsiflexed throughout the gait cycle. Compared to the intact ankle, peak prosthetic dorsiflexion during controlled lowering was 85% (13.4°) and 63% (9.9°) less for the amputee fallers and non-fallers, respectively. The control fallers and non-fallers had very similar ankle angle profiles and peak values throughout the gait cycle.

Peak anterior pelvic tilt tended to occur during mid-swing. The amputee fallers exhibited on average at least 5° less anterior pelvic tilt compared to the non-fallers. Conversely, among the control groups, it was the fallers who displayed more forward pelvic tilt.

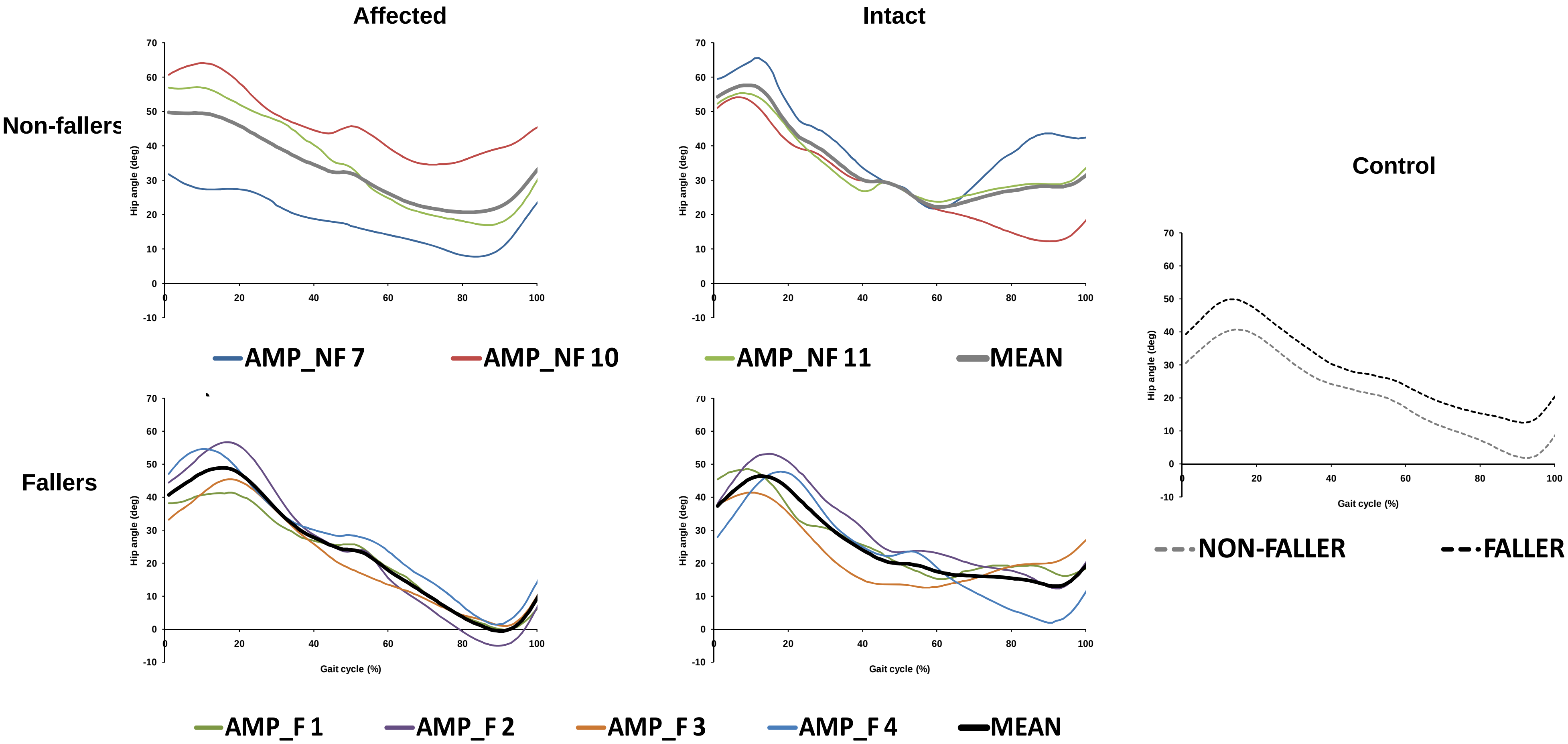


Figure 7.2. Hip flexion angles during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Flexion +ve

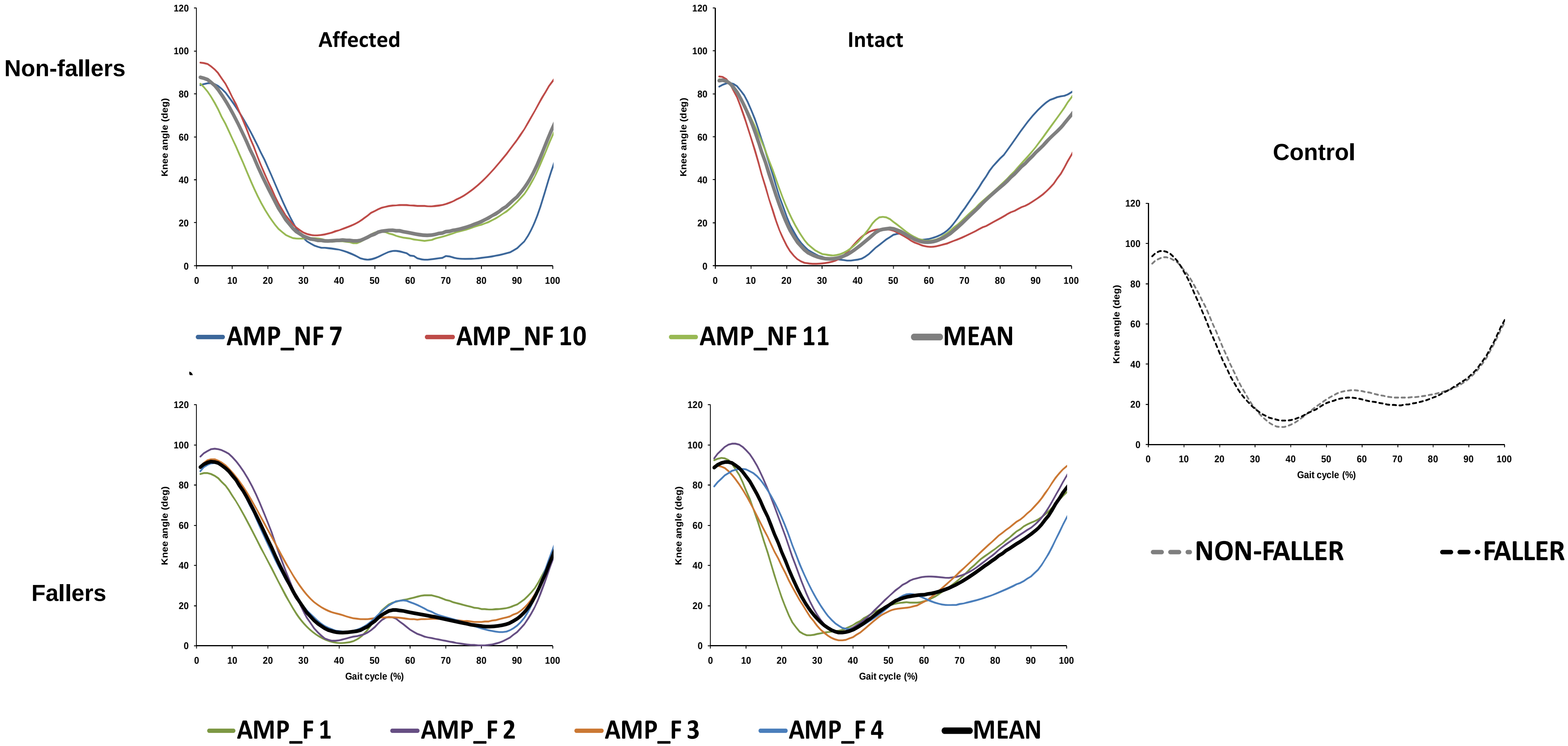


Figure 7.3. Knee flexion angles during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Flexion +ve

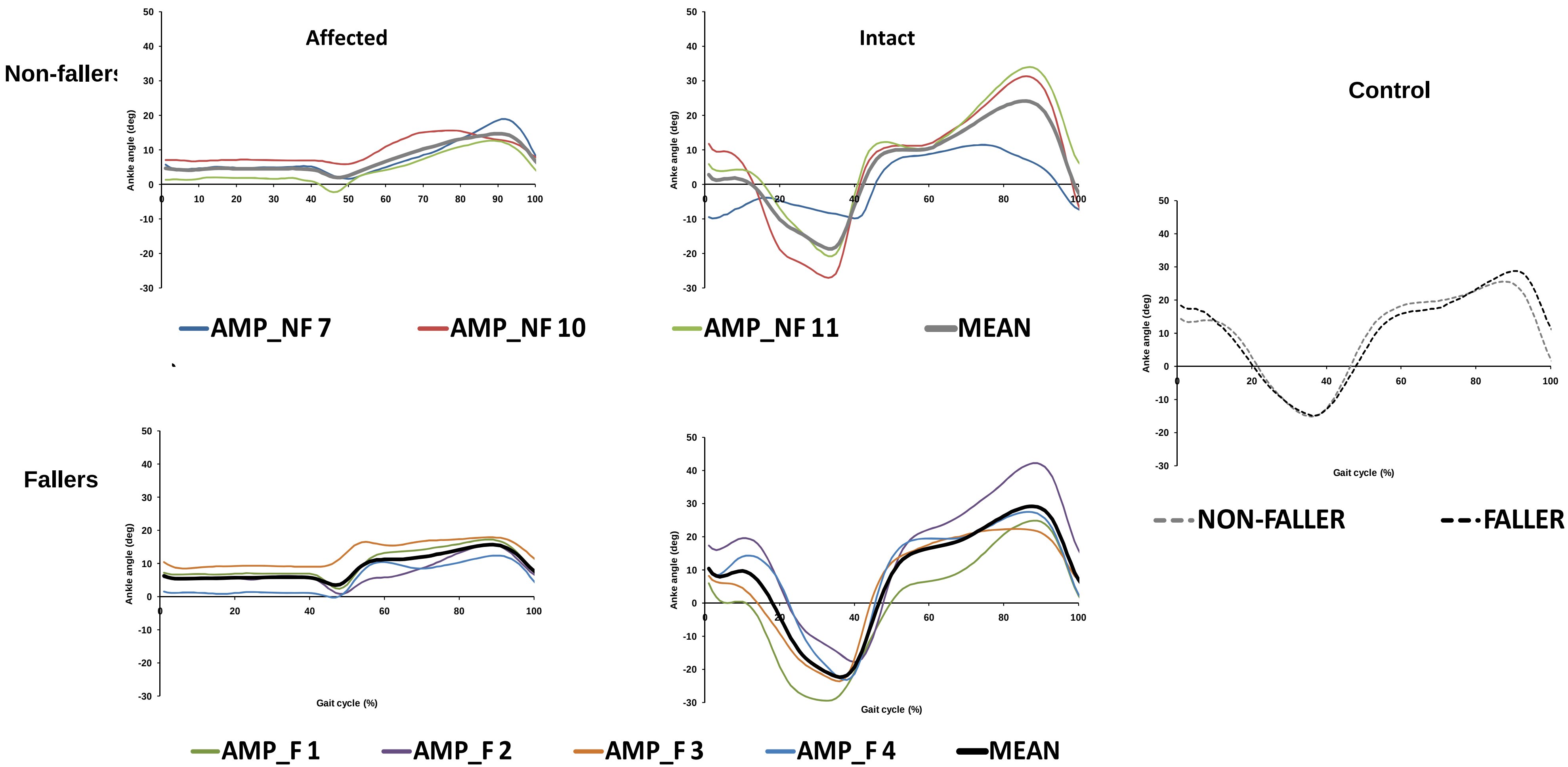


Figure 7.4. Ankle dorsiflexion angles during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Dorsiflexion +ve

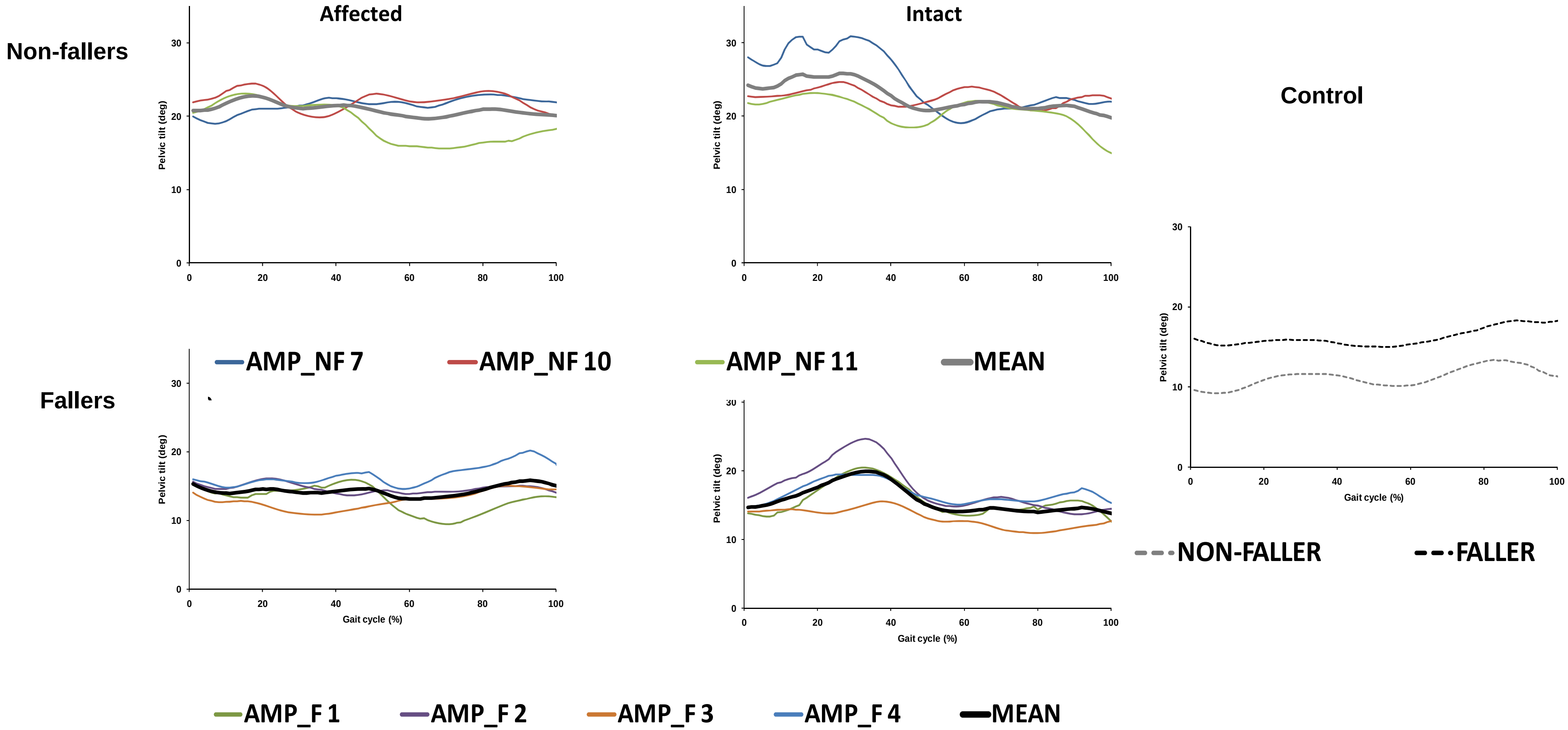


Figure 7.5. Pelvic tilt during stair descent in transtibial amputees for the fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Anterior tilt +ve

Peak frontal plane hip and pelvic kinematics are presented in Table 7.3. and illustrated in Figures 7.6 to 7.7. Participants displayed some hip adduction at toe off, followed by increasing hip abduction. For the amputee groups, the intact limb exhibited greater hip abduction during swing as the pelvis on the contralateral (affected) side was up (pelvic hike). There was considerable between-subject variance, especially on the affected side in the non-faller group and the intact side of the faller group. The control groups exhibited similar frontal plane hip profiles, except that the fallers had 88% (29°) greater hip adduction in late stance. Average frontal plane hip ROM on the intact side was the same for both amputee groups. In the control groups, the fallers displayed 16% (1.6°) greater frontal plane hip ROM than the non-fallers.

All groups initiated toe off with the pelvis up (pelvic hike), with greater hike on the affected side for both amputee groups. From toe off through swing, pelvic hike changed to pelvic drop as the swing leg was preparing to make foot contact with the step below. Both amputee groups exhibited very similar frontal plane pelvic ROM. Pelvic obliquity profiles, peak values and ROM were very similar between the control fallers and non-fallers.

Table 7.3. Amputee vs. control fallers and non-fallers mean (SD) temporal-spatial and peak kinematic values (frontal plane) during stair descent (°).Affected limb data represents first 'step to' floor; intact limb data represents top to second step

	Amputee Non-Faller (n=3)		Amputee Faller (n=4)		Control Non-Faller (n=5)	Control Faller (n=4)
	Affected	Intact	Affected	Intact		
Hip angle toe off (°)	2.7 (10.7)	2.5 (4.6)	1.6 (3.6)	-1.9 (4.4)	-2.4 (2.9)	-1.8 (5.6)
Hip abduction swing (°)	-6.1 (2.3)	-10.7 (0.3)	-6.4 (3.7)	-13.7 (4.5)	-5.9 (2.1)	-4.8 (6.2)
Hip angle foot contact (°)	-5.5 (3.0)	-10.7 (0.3)	-6.3 (3.5)	-13.7 (4.3)	-4.5 (2.0)	-3.5 (5.7)
Hip adduction stance (°)	5.0 (7.0)	6.4 (2.2)	0.0 (1.8)	3.6 (2.8)	3.3 (1.1)	6.2 (2.9)
Hip frontal ROM (°)	13.2 (9.1)	17.3 (1.8)	10.1 (1.7)	17.3 (5.0)	9.6 (2.0)	11.2 (3.6)
Pelvic obliquity toe off (°)	7.6 (4.5)	2.6 (1.3)	7.8 (3.3)	3.7 (2.0)	2.8 (1.4)	3.0 (1.1)
Pelvic obliquity foot contact (°)	-3.7 (1.6)	-7.4 (2.8)	-4.4 (2.0)	-5.6 (3.2)	-2.8 (0.7)	-3.6 (0.5)
Pelvic obliquity down stance (°)	-5.1 (3.1)	-8.5 (4.1)	-5.5 (1.5)	-7.4 (3.1)	-3.1 (0.7)	-4.1 (0.9)
Pelvic obliquity up stance (°)	-0.8 (4.0)	3.1 (2.9)	-0.6 (2.1)	4.2 (2.3)	0.5 (0.9)	1.0 (0.4)
Pelvic frontal ROM (°)	13.0 (7.5)	12.3 (5.9)	13.4 (4.8)	12.7 (5.0)	6.1 (1.9)	7.4 (1.7)

Hip abduction -ve; Hip adduction +ve; Pelvic obliquity down (pelvic drop) -ve; Pelvic obliquity up (pelvic hike) +ve

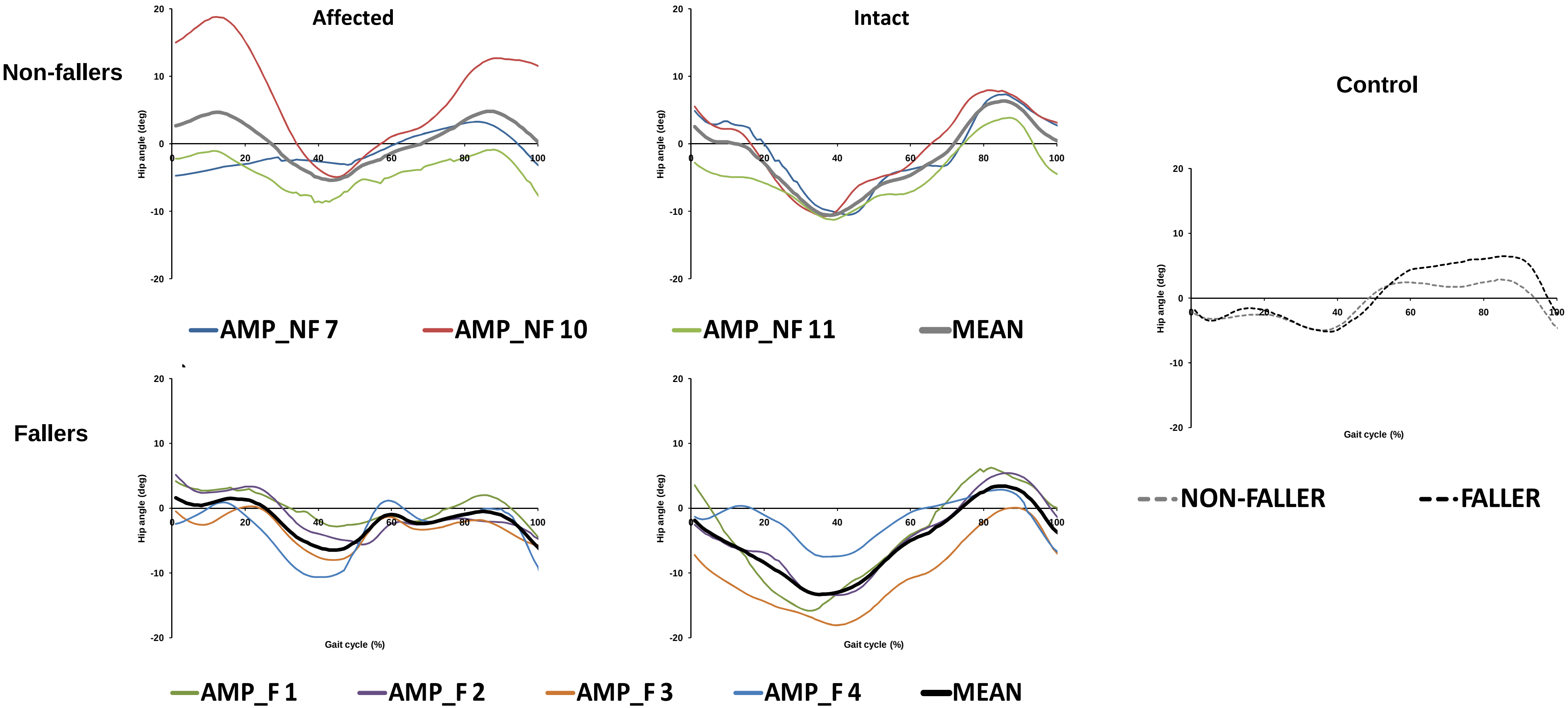


Figure 7.6. Frontal plane hip kinematics during stair descent in fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Adduction +ve

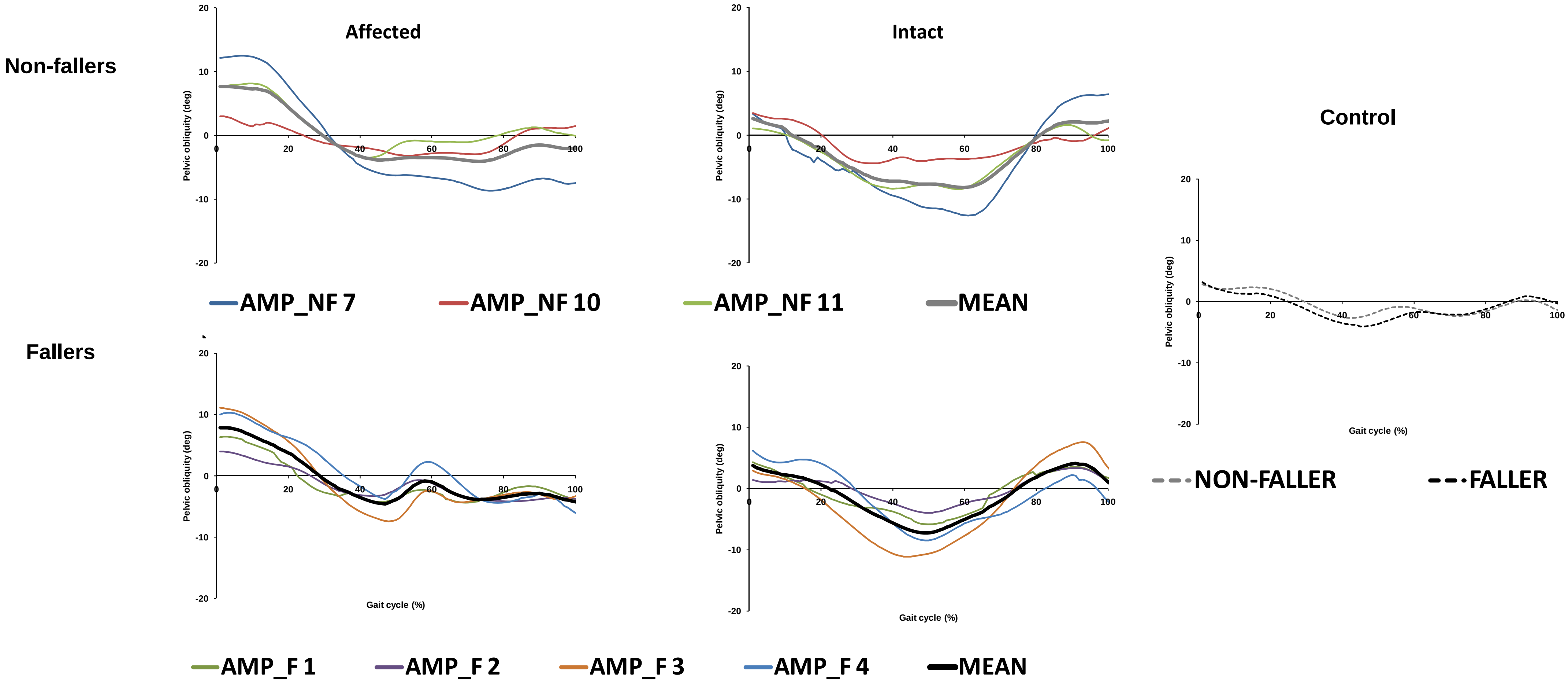


Figure 7.7. Frontal plane pelvic kinematics during stair descent in fallers and non-fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Pelvic hike/up +ve

7.3.3. GRF data

Peak GRF and joint moment values are shown in Table 7.4.GRF profiles for the intact limb of the amputee groups are plotted with the control groups in Figure 7.8.

Peak braking forces were similar between the amputee non-fallers and fallers. The most noteworthy differences were found for peak propulsive forces produced by the intact limb. These were two and half times larger among the amputee fallers compared to the non-fallers. The opposite was true for the control groups, although the differences were more modest. The control non-fallers had only 44% greater propulsive forces.

Peak vertical GRF during stair descent occurred during the first 18% of the stance phase during weight acceptance. The largest vertical forces were measured for the amputee fallers, which was 14% (0.4 N/kg) greater than the non-fallers. In the control groups, the difference in peak vertical GRF was negligible.

Table 7.4. Amputee vs. control fallers and non-fallers mean (SD) peak GRF values (N/kg); load and decay rate (N/kg/s) and joint moment values (Nm/kg) for the intact (trail) limb

	Amputee Non-Faller (n=2)	Amputee Faller (n=4)	Control Non-Faller (n=5)	Control Faller (n=4)
Posterior braking GRF	-0.12 (0.00)	-0.16 (0.02)	-0.11 (0.03)	-0.14 (0.11)
Anterior propulsive GRF	0.09 (0.01)	0.23 (0.09)	0.26 (0.07)	0.18 (0.08)
Vertical GRF	1.25 (0.02)	1.65 (0.31)	1.45 (0.11)	1.38 (0.21)
Load rate	12.19 (1.64)	9.87 (2.04)	8.00 (1.49)	11.60 (5.43)
Decay rate	-2.12 (0.59)	-4.09 (0.97)	-3.71 (1.01)	-2.99 (1.39)
Hip flexor moment controlled lowering	-0.64 (0.17)	-1.06 (0.28)	-0.97 (0.22)	-0.64 (0.25)
Knee extensor moment weight acceptance	0.65 (0.36)	0.89 (0.32)	1.34 (0.39)	1.11 (0.29)
Knee extensor moment controlled lowering	1.44 (0.09)	1.84 (0.53)	1.31 (0.40)	1.28 (0.22)
Ankle plantarflexor moment weight acceptance	0.69 (0.27)	1.32 (0.43)	1.03 (0.25)	0.72 (0.24)
Ankle plantarflexor moment controlled lowering	0.80 (0.31)	1.17 (0.31)	1.07 (0.26)	0.91 (0.16)

Braking force -ve; Propulsion force +ve; Vertical GRF - first vertical peak; Load rate +ve; Decay rate -ve; Hip flexor moment -ve; Knee extensor moment +ve; Ankle plantarflexor moment +ve

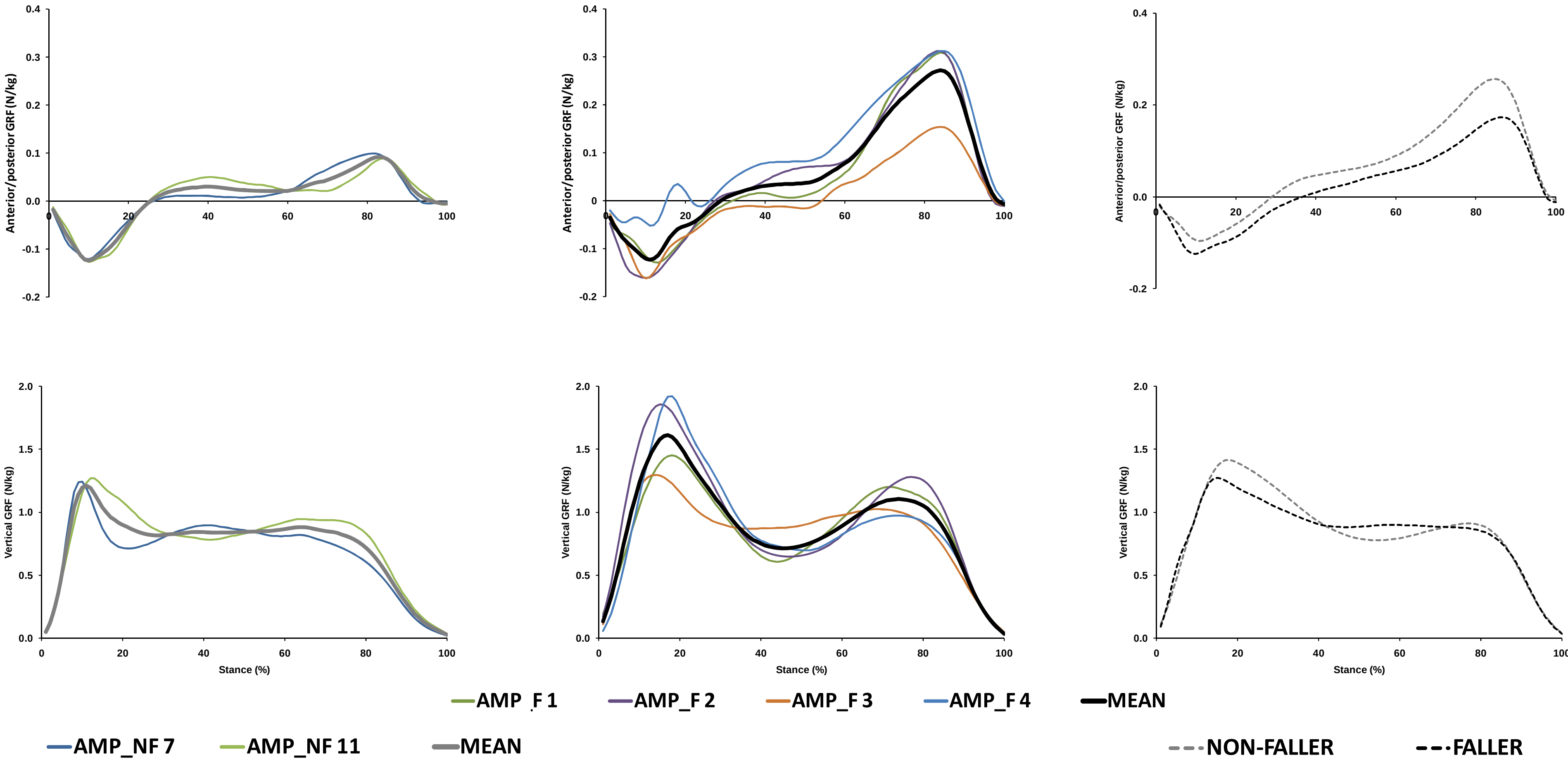


Figure 7.8. Ground reaction forces (N/kg) of the intact limb during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The stance phase is initiated with foot contact and terminated with toe off. Anterior and vertical +ve

7.3.4. Joint kinetics

Joint moment profiles for the intact limb of the amputee groups are plotted with the control groups in Figure 7.9. According to McFadyen and Winter (1988), the hip moment is small and variable during stair descent as illustrated in Figure 2.4 (section 2.3.3). Visual inspection of the current data revealed a distinct hip flexor moment during controlled lowering and this was chosen for analysis. This was because the GRFv was positioned behind the hip joint. Peak hip flexor moment was 66% (-0.42 Nm/kg) larger among the amputee fallers compared to non-fallers. The relative difference between the control groups was smaller, with the control non-fallers displaying 52% (-0.33 Nm/kg) greater hip flexor moment.

The amputee participants continued to show large between-subject variability in the internal knee joint moments. The amputee fallers exhibited higher knee joint moment profiles compared to the non-fallers. Both amputee groups had larger peak extensor moments during controlled lowering compared to the control groups and the largest extensor moment was measured in the amputee fallers (28% or 0.4 Nm/kg greater than the amputee non-fallers).

All the participants initiated the weight acceptance phase with the mid-foot or forefoot regions of the intact foot. The internal ankle moment was always plantarflexor in orientation. The peak ankle moment during the weight acceptance was greatest among the amputee fallers and was almost double compared to the non-fallers (on average 91% or 0.63 Nm/kg greater). The amputee fallers still displayed larger ankle moments compared to the non-fallers during controlled lowering, although the difference was more modest (46% or 0.37 Nm/kg). The relative differences between the control groups were smaller, with the control non-fallers exhibiting only 43% (0.31 Nm/kg) greater ankle moments in weight acceptance and 18% (0.16 Nm/kg) during controlled lowering.

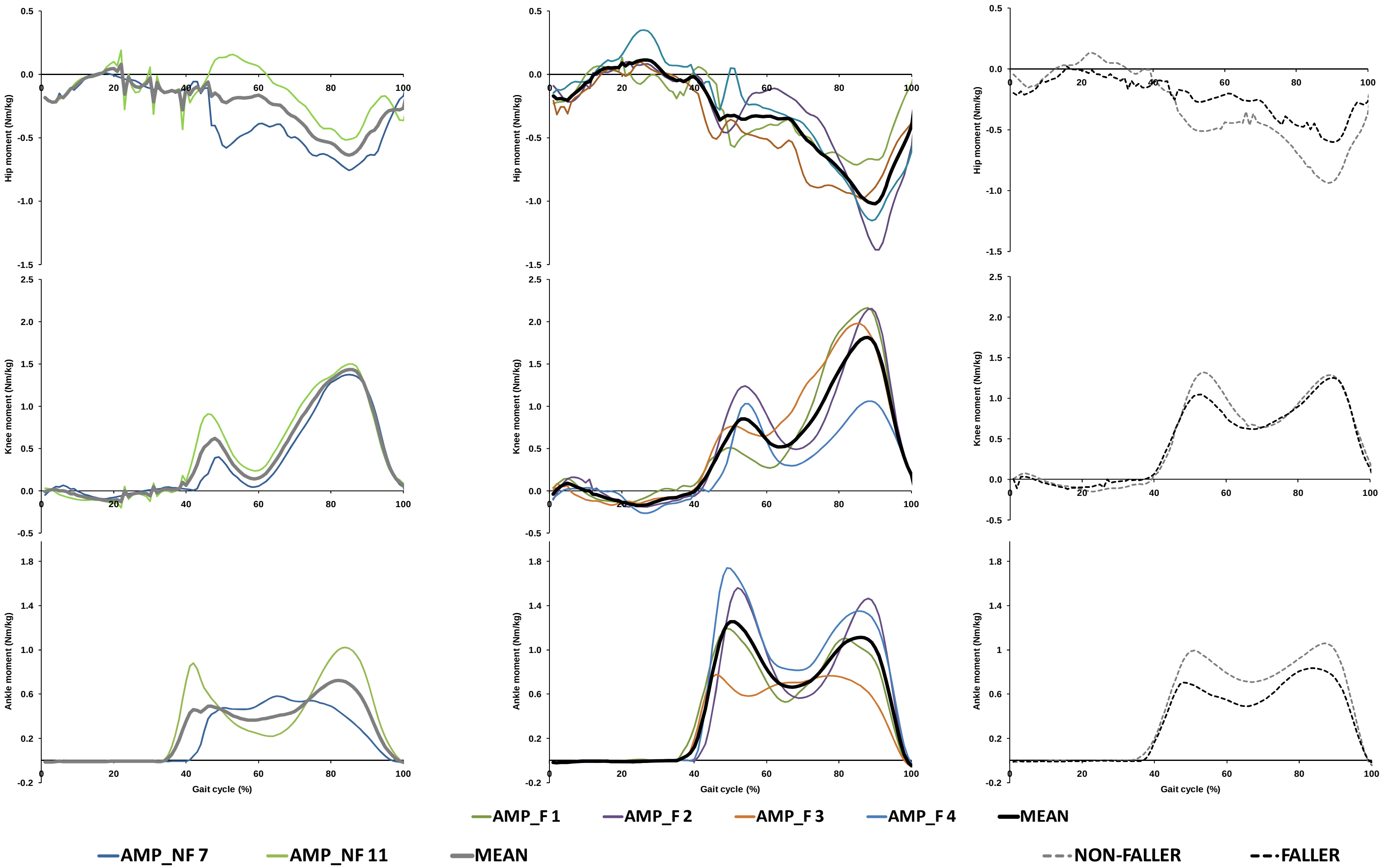


Figure 7.9. Joint flexor/extensor moments (Nm/kg) of the intact limb during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups. Extensor and plantarflexor +ve

Peak joint powers are reported in Table 7.5. Joint power profiles are displayed in Figure 7.10. Compared to the ankle and knee joint power profiles, the hip powers were small across groups. The most distinct power generation burst at the hip occurred during the controlled lowering/pre-swing phase (labelled H1) especially for the amputee fallers, although there was considerable variance between the participants in this group. A small power generation burst in early swing (labelled H2) was seen in all groups. Both faller groups had greater H2 peak values compared to their non-faller counterparts.

The knee power profiles were virtually identical for the amputee non-fallers. However, AMP_NF 7 did not display a distinct power generation burst in mid-stance (labelled K2). The same did not hold true for the amputee fallers, who showed rather more variance in the relative timing and magnitude of the knee power bursts. Knee joint power generation (K2) was also absent in the amputee faller group, with the exception of AMP_F 4. Contrary to the hypothesis, there were no observable differences in the average knee power profiles (shapes) during the controlled lowering phase on the intact limb in the amputee groups. Control non-fallers and fallers displayed similar knee power profiles, however the non-fallers had a 43 % (-0.30 W/kg) greater power absorption burst at the knee in the pre-swing/controlled lowering phase (labelled K3). This was an important finding in relation to falls because insufficient knee extensor eccentric strength could lead to the collapse of the supporting limb.

The amputee fallers and control groups show the typical ankle power phases, it is difficult to generalise the findings for the amputee non-fallers as only AMP_NF 11 showed distinct ankle power absorption and generation bursts (labelled A1, A2 and A3). Two of the four amputee fallers had a very large power absorption burst in the weight acceptance/loading phase (labelled A1), averaging over -7 W/kg. The relative timing of the three ankle power bursts was consistent among most amputee and control participants.

Table 7.5. Amputee vs. control fallers and non-fallers mean (SD) peak joint powers (W/kg)

	Amputee Non-Faller (n=2)	Amputee Faller (n=4)	Control Non-Faller (n=5)	Control Faller (n=4)
Hip power generation pre-swing (H1)	0.33 (0.00)	1.12 (0.72)	0.92 (0.40)	0.57 (0.40)
Hip power generation initial swing (H2)	0.16 (0.02)	0.33 (0.12)	0.26 (0.15)	0.34 (0.25)
Knee power absorption loading (K1)	-1.03 (0.68)	-1.48 (0.82)	-2.22 (0.94)	-2.08 (0.74)
Knee power generation mid-stance (K2)	0.45 (0.50)	0.22 (0.35)	0.56 (0.35)	0.68 (0.43)
Knee power absoprtion pre-swing (K3)	-3.63 (0.28)	-4.21 (0.81)	-3.95 (1.12)	-3.21 (0.70)
Knee power absoprtion mid-swing (K4)	-0.70 (0.18)	-1.24 (0.38)	-0.99 (0.35)	-0.69 (0.31)
Ankle power absorption loading (A1)	-2.24 (1.94)	-5.06 (2.43)	-3.49 (1.05)	-2.06 (1.39)
Ankle power absorption mid-stance (A2)	-0.61 (0.64)	-1.16 (0.57)	-1.01 (0.63)	-1.00 (0.43)
Ankle power generation pre-swing (A3)	0.76 (0.72)	1.99 (1.23)	2.09 (1.19)	1.05 (0.58)

Power generation +ve; Power absorption -ve

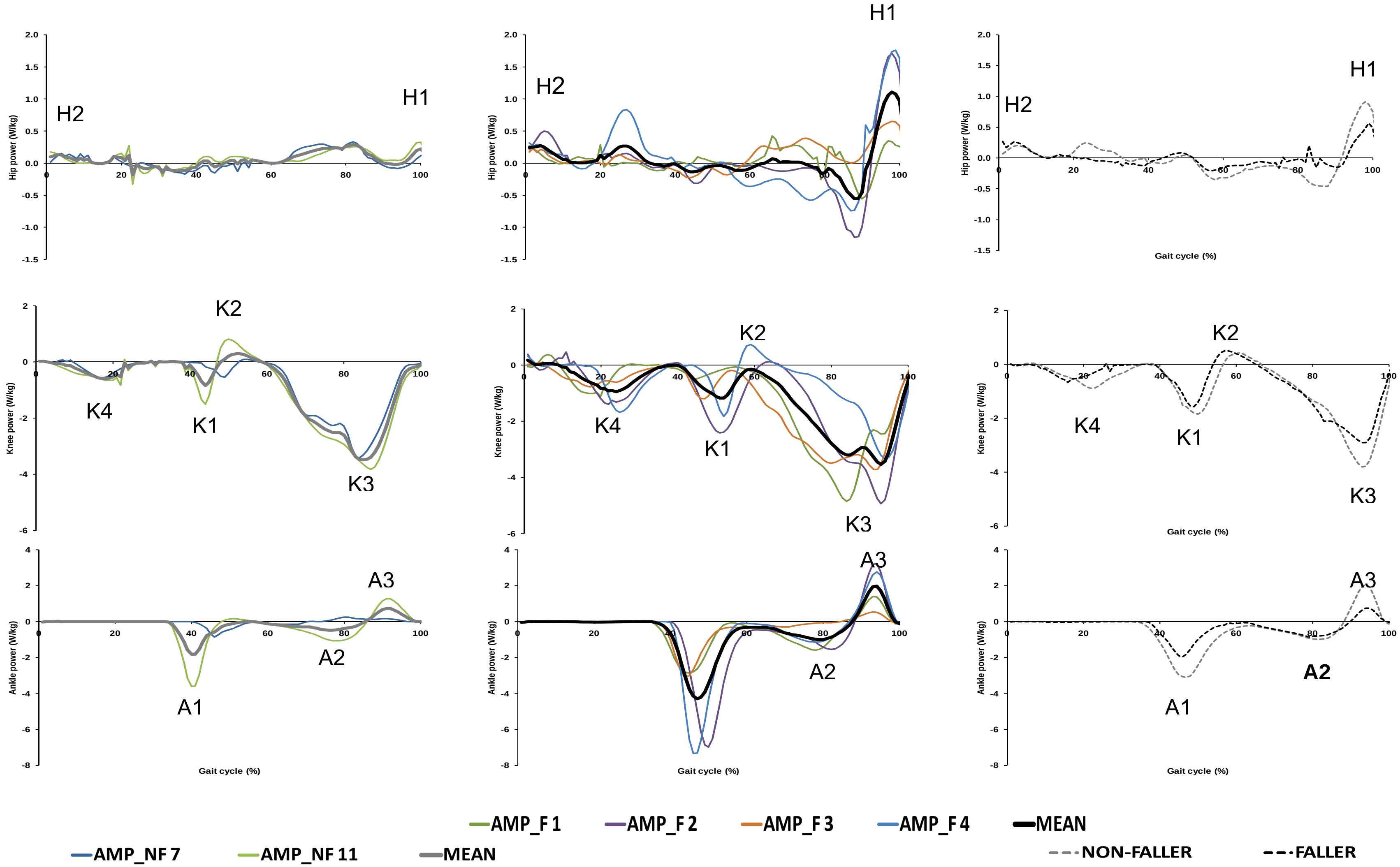


Figure 7.10. Sagittal plane joint powers (W/kg) of the intact limb during stair descent for the fallers and non-fallers. Individual participant data are presented with the averaged data: Non-fallers (solid grey line); Fallers (solid black line). Control data are presented separately: Control non-fallers (dashed grey line); Control fallers (dashed black line). The gait cycle is initiated and terminated with toe off. On average, swing represents approximately the first 40% of the gait cycle for all groups.

Power generation +ve; Power absorption –ve

7.3.5. ‘Step to’ group - (first to second step)

Two amputee fallers (AMP_F 5 and 6) and one amputee non-faller (AMP_NF 9) adopted a ‘step to’ gait pattern during stair descent. These participants always led with their affected limb. One amputee non-faller (AMP_NF 8) had an intact limb preference and started stair descent with a step over gait pattern. However, on the last step, they switched to a ‘step to’ pattern. Temporal-spatial and sagittal peak kinematic values are presented in Table AD1 and Figures AD1-AD4 in Appendix D. The non-fallers walked 42% (0.10 m/s) quicker than the fallers. The fallers spent longer in stance: 34% and 7% for the affected and intact limbs, respectively.

Peak hip flexion values and profiles were similar for the non-fallers and fallers. Neither group reached full hip extension on either the affected or intact limbs. However, the non-fallers had 60% (10.3°) greater extension at the hip joint compared to the fallers. Participants in both groups extended their hip more rapidly on the intact side. Large variability was measured on the affected side of the non-faller group.

The knee on the affected side was 71% (31.1°) more flexed at toe off and throughout swing for the non-fallers compared to the fallers. The non-fallers had 47% (18.5°) larger knee ROM on the affected side and 6% (4°) greater on the intact side. There was considerable variability, as evidenced by the large SD for the knee, on the affected side of the non-fallers.

In the ‘step to’ groups, the prosthetic and intact ankles remained in dorsiflexion during the whole gait cycle. The non-fallers exhibited 47% (5.1°) greater prosthetic ankle dorsiflexion during controlled lowering compared to the fallers. This reflected the non-fallers’ 72% (4.9°) larger ROM on the prosthetic side compared to the fallers. There were few noteworthy differences in pelvic tilt profiles between the groups other than peak anterior tilt on the intact side occurred during swing for the non-fallers and early stance for the fallers.

Frontal plane peak hip and pelvic values are presented in Table AD2 and Figures AD5-AD6 in Appendix D. Hip adduction profiles were varied with little difference between the non-fallers and fallers. On the intact side, the hip was adducted, especially among the non-fallers (54% or 3.1° more than the fallers), while the hip was abducted on the affected side during most of the stance phase. For both groups, hip ROM in the frontal plane was larger on the affected side compared to the intact side. The non-fallers maintained pelvic hike on the affected side throughout the entire gait cycle, whereas the fallers exhibited pelvic drop during the first half of stance. Pelvic ROM was 21% (2°) larger on the affected side and 35% (2.7°) smaller on the intact side for the non-fallers.

7.4. Discussion

The aim of this study was to compare the gait patterns of fallers and non-fallers during stair descent in transtibial amputees and able-bodied participants. All of the participants were able to complete the task successfully, although four amputees (2 fallers and 2 non-fallers) self-selected a ‘step to’ rather than an alternate step over walking pattern.

7.4.1. Temporal-spatial

The first hypothesis related to walking speed and joint mobility and the results indicated that walking speed was reduced during stair descent supporting the notion that it was a more mechanically complex task than either level walking or stair ascent. All groups walked approximately 0.1 m/s slower during stair descent compared to stair ascent. As walking speed is considered a good indicator of physical function and mobility, the mechanical challenge of stair descent is emphasised by a slower velocity (Powers and Boyd, 1997).

Few published studies report speed during stair walking. Powers and Boyd (1997) reported velocities of 0.49 m/s and 0.59 m/s for their amputee and control participants performing a stair descent task, respectively. In contrast, Hamel *et al*. (2005) found that 0.65 m/s was a comfortable, self-selected speed for otherwise healthy older adults and controlled for speed in their study. This was in line with Lee and Chou (2007) who found stair descent velocities of 0.60 m/s and 0.66 m/s for elderly and young, respectively.

The current results suggest the amputee non-fallers and control fallers may have put themselves at risk for falling at higher speeds. The results indicate that the amputee fallers walked at a similar pace to the control non-fallers and conversely, that the amputee non-fallers were similar to the control fallers. The intact limb spent on average 5% longer in stance compared to the affected limb in the amputee non-fallers. It was probable that the amputee non-fallers and control fallers adopted a slower, more cautious stair descent pattern because falling down stairs can have more severe consequences than falling during stair climbing. The present results concur that stair descent presents a more mechanically challenging task than stair ascent or level walking.

7.4.2. Joint kinematics

The ankle joint plays a crucial role during weight acceptance, demanding eccentric control by the ankle plantarflexors when initial contact is made with the forefoot. Ankle plantarflexion assists in lengthening the leg in preparation for contact with the step below. This facilitates smoother movement of the CoM in the vertical and horizontal directions. Previous studies investigating stair descent in amputees have not specifically examined pelvic hike or drop (Schmalz *et al*., 2007; Power and Boyd, 1997). In the current study, amputee participants showed somewhat exaggerated pelvic range of motion particularly in the frontal plane. Dropping the pelvis prior to initial contact served to lower the whole leg and was a compensatory strategy for the lack of plantarflexion. This also contributed to hip abduction during the swing phase.

Peak ankle joint kinematics were similar to those reported by Powers and Boyd (1997) for transtibial amputees. Peak dorsiflexion in stance was limited by the prosthetic ankle. Knee flexion could have been inhibited as socket fit tends to be high posteriorly (Hill *et al*., 1997). The amputee non-fallers showed a tendency to 'throw' their prosthetic foot down onto the next step compared to the fallers. This was evident with more hip flexion at toe off and throughout swing, thus lifting the whole leg into the air for stair clearance. Similar observations were reported in transtibial amputees when crossing obstacles with their prosthesis as the lead limb (Hill *et al*., 1997).

On the other hand, the amputee fallers used a 'roll over' technique, similar to that reported in TFA, which reduces ankle joint moments (Schmalz *et al*., 2007). Rolling over the edge of the step is a considerably more mechanically challenging task. It requires proper positioning of the prosthetic foot onto the nose of the step and rolling over, while in single support (labelled K3 power burst). It is possible that the amputee fallers placed themselves at greater risk of falling by relying on eccentric control of the knee extensors during this controlled lowering phase. Reeves *et al*. (2008b) demonstrated that handrail use could redistribute some of the work onto the arms and partially unload the legs, thereby reducing the demands on the knee extensors. The results suggest that improving knee extensor eccentric strength and using a handrail could reduce the potential for falls in transtibial amputees.

Knee kinematics were similar between fallers and non-fallers in both the amputee and control groups. The only noteworthy difference was smaller knee range of motion on the affected side in the amputee non-fallers (78.8 ± 4.1°) compared to the fallers (86.9± 7.5°). This reflected a combination of greater knee flexion at initial contact (because the limb was being 'thrown' over the step) and possibly differences in prosthetic socket fit restricting peak flexion. There were considerable between-subject differences, particularly among the amputee non-fallers. Together with small sample sizes, this made it difficult to generalise these findings the wider amputee population.

There were larger differences between all of the groups when hip kinematics were examined. The affected limb of the amputee fallers was fully extended in stance (-0.7 ± 2.9°) and displayed larger range of motion compared to the non-fallers. This was related to the fact that the affected limb was measured from the first step to the floor. Initiating and terminating the gait cycle with toe off meant that the stance phase of the affected limb was analysed when the foot was already on the ground and about to start level walking. The amount of average hip extension was the same as previously measured during level walking in the same group (Table 5.2). The amputee fallers displayed the same hip extension during level walking and transitioning from stair descent to level walking. Staircase dimensions did not restrict foot position during the transition from stair to level walking. Kerrigan *et al*. (1998) reported reduced hip extension and shorter step lengths in the elderly. When the amputee fallers took a longer step with the affected limb, the hip would have been more extended. On the other hand, the amputee non-fallers likely would have been unable to 'throw' their prosthesis far over the nose of the step (in the horizontal direction), resulting in a shorter step length and reduced hip extension.

Control fallers and non-fallers mainly exhibited differences at the hip joint. Although the range of motion was very similar, the non-fallers consistently exhibited on average 10° more extension and the fallers 10° more flexion (Table 7.2). These findings were consistent with the falls literature in elderly fallers and non-fallers (Kerrigan *et al*., 2001). These authors reported that reduced hip extension (by approximately 6.5°) was the only kinematic variable that consistently distinguished fallers from non-fallers, at both comfortable and fast walking speeds. It is reasonable to assume that functional limitations would remain or may even be more exaggerated during more challenging tasks. Results showed that the control fallers had approximately 10° less hip extension during level walking and stair descent. This finding has implications for falls prevention programmes and therapies as Kerrigan *et al* (2000) suggested that reduced hip extension could specifically be targeted to improve walking performance.

7.4.3. GRF data

GRF data were based on a very small sample size (e.g. amputee non-fallers = 2) and this discussion will only provide some insight about the kinetic differences between fallers and non-fallers. Schmalz *et al*. (2007) noted larger peak deceleration and smaller acceleration forces on the intact limb, while the reverse was true for the affected limb. In the current study, peak braking force of the intact limb in the amputee fallers was the same as that reported by Schmalz *et al*. (2007) (Table 7.4). However, peak propulsive force was much larger, even when compared to the able-bodied stair literature (Reiner *et al*., 2002), and approached values more frequently measured during level walking (0.2 N/kg). As this represented the transition from stair to level walking, large propulsive forces could be potentially destabilising. A fall occurs when the joints are unable to generate and maintain sufficient support (Beaulieu *et al*., 2008). There may be no handrails on level ground near a staircase, so a trip or stumble could easily lead to a fall during single support on the affected side.

Differences between the control fallers and non-fallers were consistent with previous findings by Reeves *et al*. (2008a) comparing young vs. elderly. They reported propulsive forces of 0.12 N/kg and 0.10 N/kg for the young and elderly, respectively. Schmalz *et al*. (2007) also measured 0.12 N/kg for the controls. The current results continue to demonstrate that amputee fallers exhibited similar biomechanical characteristics to control non-fallers and that amputee non-fallers were more similar to control fallers.

In the amputee fallers, vertical GRF experienced on the intact limb was considerably higher compared to results reported by Schmalz *et al*. (2007) and Reiner *et al*. (2002) in amputees and controls, respectively. There was also considerable variability between participants, with some measuring peak forces of almost 2.0 N/kg. However, an interesting difference was that the amputee non-fallers loaded their intact limb more than the fallers. Although the results were only based on the data from two individuals, higher loading was reflected by initial foot contact occurring with the knee joint almost fully extended, despite smaller peak vertical forces. Even the control groups had higher peak vertical forces compared to the literature (Reiner *et al*., 2002; Reeves *et al*., 2008a). Variations in results could reflect stair dimensions, as landing from a higher step would generate larger peak forces.

7.4.4. Joint kinetics

The second hypothesis related to reduced joint mechanics. There is very little published data presenting joint moment and power profiles in transtibial amputees during stair descent. All joint moment profiles of the intact limb were consistently smaller in the amputee non-fallers vs. fallers. Ankle moment profiles varied between the two participants in the non-faller group, such that AMP_NF 11 did not display a typical double peak profile. Although this person made initial contact with the forefoot, they maintained the GRF vector close to the ankle joint by delaying knee flexion in mid- to late stance. While the amputee fallers experienced larger joint moments than the non-fallers, they were similar in shape and magnitude to those reported by Schmalz *et al*. (2007) for the ankle and knee joints. However, the hip joint moments were remarkably different for both the amputee fallers and non-fallers compared to data presented by Schmalz *et al*. (2007). McFadyen and Winter (1988) recognised large discrepancy in hip moments because there is greater variation in how individuals carry the passenger (HAT) unit up and down stairs. In the current study, the hip moment of the intact limb closely resembled that hip moment of the affected limb measured by Schmalz *et al*. (2007).

Although GRFs were not measured for the affected limb in the current study, previous research showed reduced ankle, knee and hip moments on the affected side (Schmalz *et al*., 2007) that were very similar to the intact joint moment profiles in the amputee non-fallers in this study. One possible explanation for the low joint moments on the intact limb and differences between the amputee groups is that the non-fallers may have tried to maintain better between-limb symmetry. Overloading one side of the body could result in long-term musculoskeletal problems. Lewallen *et al*. (1986) reported that large joint forces in the intact limb may lead to premature diseases, such as degenerative arthritis. Stair walking places greater demands on joints and muscles of the lower-limbs than level walking (Lin *et al*., 2005; Reeves *et al*., 2008a; McFadyen and Winter 1988; Schmalz *et al*., 2007; Reiner *et al*., 2002). In individuals with asymmetrical musculoskeletal function, it is possible that more challenging activities would predispose the contralateral, intact leg to overloading. Indeed Schmalz *et al*. (2007) found greater loading of the intact limb during stair descent than ascent. These authors suggested improved prosthetic design and greater prosthetic ankle mobility could have positive effects on loading patterns of the intact leg during stair walking. However, it is unclear whether such modifications would challenge stability.

In the control groups, joint moment profiles were very similar in shape with peak values occurring at virtually identical times during the gait cycle. Overall, joint moments were smaller for the fallers. This was attributed to slower stair walking speed and reduced GRFs, not specifically reduced joint mobility. It is acknowledged that handrail use by some participants in the current study may have influenced the joint moments at the ankle and knee. Reeves *et al*. (2008b) found redistribution of ankle and knee joint moments such that the former increased and the latter decreased when using handrails. However, handrail use was considered a good strategy to assist balance control.

Equally, few studies have analysed joint power data in transtibial amputees during stair walking. Yack *et al*. (1999) reported less energy generation on the affected side during stair ascent. However, no published studies have been found that report joint powers during stair descent in transtibial amputees. Equally, falls literature has not specifically compared joint powers in stair descent in able-bodied participants (e.g. healthy vs. fallers). There is a significant gap in the biomechanics of stair descent literature, particularly kinetic variables such as joint moments and powers, especially since falling during stair descent is more common than ascent.

In the present study, all groups exhibited distinct power bursts at the ankle and knee as described by McFadyen and Winter (1988), with hip powers being more variable. As joint powers were calculated using joint moment and kinematic data, they showed considerable within-group variability (Table 7.5) and peak values should be interpreted with much caution given the very small sample.

Overall, the amputee fallers had larger peak power bursts compared to the non-fallers. The A1 power burst of the intact limb represents power absorption by the plantarflexors in weight acceptance/loading. This phase corresponds to the end of controlled lowering/pre-swing on the contralateral affected leg, when large eccentric power burst of the knee extensors would be expected (McFadyen and Winter, 1988; Beaulieu *et al*., 2008). As previously discussed (see section 2.5.3), the knee absorption power burst (K3) demands sufficient eccentric control of the knee extensors and thus represents the most vulnerable point in the gait cycle during stair descent.

Power absorption at the ankle (A1) could be reduced if initial contact is made with the ankle in a more neutral position, thereby requiring less eccentric control by the plantarflexors. With the ankle in a neutral position in weight acceptance, the foot became flatter on the step and the GRF vector was kept closer to the ankle axis of rotation. In turn this increased the base of support and reduced the ankle joint moment. Although Reeves *et al*. (2008b) did not measure joint powers, they reported that elderly maintained the foot flatter on the step for longer as a postural strategy in the absence of handrails. A larger base of support of the intact limb could provide the postural stability needed to reduce the important knee power absorption (K3) burst in the contralateral affected leg during controlled lowering. This may be an adaptive postural strategy for amputees who have reduced eccentric control of the knee extensors on the affected side. It could also provide additional stability to those transtibial amputees who use the roll over technique previously described (see section 7.4.2)

Another possibility for the reduced joint powers in the amputee non-fallers could reflect their attempt at maintaining some mechanical symmetry between the affected and intact limbs. While this study cannot compare the joint power profiles with the current published literature, it has shown reduced moments in the amputee non-fallers. Therefore, it would be reasonable to assume that joint powers would also be reduced on the prosthetic side, particularly at the ankle.

In the control groups, the control fallers had reduced ankle power absorption in weight acceptance/loading (A1) and knee power absorption in controlled lowering/pre-swing (K3) compared to the non-fallers. This was linked with reduced ankle joint moments in the fallers in these phases. These findings provide further evidence that amputee non-fallers and control fallers show similar mechanical constraints during stair walking.

Hip powers were varied and did not show distinct power bursts across all groups. The largest burst was power generation in pre-swing (H1), when the hip flexors contracted to pull the leg off the top step, and was important for ground clearance. It is difficult to draw conclusions based on the current sample size and because hip moment and powers reflect individual trunk movements (McFadyen and Winter, 1988).

7.4.5. ‘Step to’ group

Transtibial amputees are taught to lead with their prosthesis/affected limb during stair descent. This is because the trail limb must flex at the knee to ensure safe lowering during the controlled lowering phase (lead limb swing phase, trial limb single support phase) and move through a greater knee ROM. Depending on prosthetic fit, the height of the prosthetic socket behind the knee could limit joint flexion. Therefore, the knee on the affected side typically has a smaller ROM. This was certainly the case in the amputee fallers, who had substantially less knee flexion. Although modifiable, if prosthetic socket fit was limiting knee ROM, then that could have a detrimental effect on stair locomotion.

The ‘step to’ participants most likely adopted this gait pattern because of functional and strength limitations at the knee of both limbs. The time spent in single support on the affected limb was reduced and the knee was maintained almost completely extended. The controlled lowering phase, the most vulnerable instant during stair descent, was substantially shorter for the intact (trail) limb and virtually absent for the affected (lead) limb.

The main distinguishing characteristics between the amputee fallers and non-fallers was reduced ROM at the ankle and knee joints. Although reduced joint mobility was a characteristic of the 'step to' gait pattern, a certain range of motion would still be necessary to negotiate stair descent safely. The inability to achieve this may be considered a risk factor for falling. Exercise programmes aimed at improving knee extensor strength and knee joint mobility on the affected side, in those individuals adopting the 'step to' gait pattern, would be encouraged. Equally, prosthetic socket fit should be checked to ensure it allows adequate knee flexion.

7.5. Conclusion

This biomechanical analysis in amputee fallers and non-fallers provided some initial evidence that these two groups adopted different patterns during stair descent. Contrary to the predictions, the amputee fallers walked faster than the non-fallers and exhibited larger ROM at the hip and knee joints on the affected leg and the intact ankle joint. Notably, the non-fallers appeared to 'throw' their prosthesis over the edge of the step, whilst the fallers employed the more difficult 'roll over' technique, requiring adequate strength and control of the knee extensor musculature. The amputee fallers also showed larger propulsive forces during the transition from stair to level walking, providing further evidence that the transition steps between stair and level walking can differentiate fallers from the non-fallers. This would have important implications for amputee rehabilitation. The kinetic results did not support the second hypothesis that the fallers would have reduced joint moments and powers at the ankle and knee. In fact, these were larger in magnitude, but similar in timing. In the control groups, the fallers showed reduced joint moments and powers, meaning that any neuromuscular failure (particularly eccentric contraction of the knee extensor muscles during controlled lowering) keeping the body upright could have severe consequences and this finding agreed with the hypothesis. More participants adopted a 'step to' gait pattern in stair descent than ascent and this reduced the demands on joint mobility. The controlled lowering phase was missing on the affected limb.

CHAPTER EIGHT - POSTURAL RESPONSES TO DYNAMIC PERTURBATIONS IN FALLERS VS. NON-FALLERS

8.1. Introduction

People who have had a transtibial amputation exhibit similar characteristics compared with the elderly, such as muscle weakness and postural instability, that predispose them towards an increased risk of falling (Isakov *et al*., 1992). Transtibial amputees have lost afferent information from receptors in the muscles, tendons, and skin of the foot and lower-limb and consequently they have altered somatosensory input. If mechanisms responsible for integrating information from different sources are affected by disease, then excessive sway or falls can occur, especially in unusual sensory environments (Horak *et al*., 1989). As such, poor balance and falls incidence among this group have become a growing concern for rehabilitation specialists.

Quantitative posturography (CDP) has previously been used to assess balance with elderly (Camiciolo *et al*., 1997; Horak *et al*., 1989; Judge *et al*., 1995; Nardone *et al*. 2000) and differentiate between elderly fallers and non-fallers (Parry *et al*., 1995; Wallmann, 2001). Some studies have measured standing balance in transtibial amputees (Isakov *et al*., 1992; Isakov *et al*., 1994; Hermodsson *et al*., 1994) and postural sway (Fernie *et al*., 1978). Until now, no study has quantified postural control in transtibial amputees using CDP and it is unclear whether dynamic posturography could be used to distinguish between these groups. The aim of the current study was to investigate whether CDP could be used to differentiate between fallers and non-fallers during static and dynamic conditions using the NeuroCom Smart Equitest system and to measure and quantify the postural control strategies in the amputee and control groups. It was hypothesised that fallers would have increased postural sway on tests that particularly challenged the somatosensory system and show greater reliance on hip strategies in response to dynamic perturbations. It was also anticipated the amputee fallers would exhibit greater reliance on the affected limb by bearing more weight through this limb during backwards and forwards translations.

8.2. Methods

8.2.1. Participants

Nine transtibial amputees (mean ± SD: age 59 ± 14 yr; height 1.72 ± 0.14 m; body mass 76 ± 16 kg; time since amputation 8.1 ± 9.7 yr) and nine age-matched able-bodied participants (mean ± SD: age 61 ± 16 yr; height 1.73 ± 0.14 m; body mass 80 ± 13 kg) were recruited for this study. All participants gave written informed consent to take part in this study. Individual participant details according to falls history can be found in Table 3.1. Amputee participants 5 (faller) and 10 (non-faller) did not consent to take part in this study and their data are not included. Everyone satisfied the inclusion and exclusion criteria, had normal or corrected vision and were able to stand quietly without assistive devices (e.g. stick).

8.2.2. Experimental Protocol

The NeuroCom Smart Equitest system was used to measure postural responses to dynamic perturbations during the sensory organisation test (SOT) (Figure 8.1) and motor control test (MCT). Participants' height with shoes (cm) and body mass (kg) were measured (SECA GmbH & Co. KG, Germany) and this information was entered in the NeuroCom software. Participants were helped to step onto the platform base, where they faced the visual surround. They wore comfortable loose clothing and flat shoes. Each foot was positioned on one force plate such that the medial malleolus and lateral aspect of the calcaneous were aligned with the appropriate markers according to the manufacturer's instructions. Foot position was scaled according to body height (short – 76-140 cm; medium – 141-165 cm; large 166-203 cm). Participants were instructed to stand upright, arms at their side, facing the visual surround and to maintain their balance throughout the tests. All participants wore a safety harness that permitted postural sway beyond their limits of stability (4° posteriorly, 8.5° anteriorly, 6° laterally) but prevented the participant from actually falling. Sway angle was defined as the angle between a vertical line originating from the centre of area of foot support and a second line originating from the same point to the person's COG position. Normal anterior to posterior range of sway is typically 12.5° without experiencing a loss of balance (Nashner *et al.*, 1989).

8.2.3. Sensory Organisation Test

The order of testing during the SOT was standardised for all participants. The participant's postural sway was measured during three, 20-second trials over 6 conditions (Figure 8.1).

Condition 1: sway was measured in a static condition with uncompromised visual, vestibular and somatosensory feedback (eyes open, fixed surround, and fixed surface). This condition served to establish a baseline level.

Condition 2: sway was measured in a static condition with eyes closed.

Condition 3: sway was measured when visual cues were inaccurate (sway-referenced moving surround).

Condition 4: sway was measured under dynamic conditions when somatosensory cues were inaccurate (sway-referenced moving support surface).

Condition 5: sway was measured when visual cues were removed and somatosensory information was inaccurate (eyes closed, sway-referenced moving support surface). Condition 6: sway was measured when visual and somatosensory cues were inaccurate (sway-referenced moving surround and moving support surface).

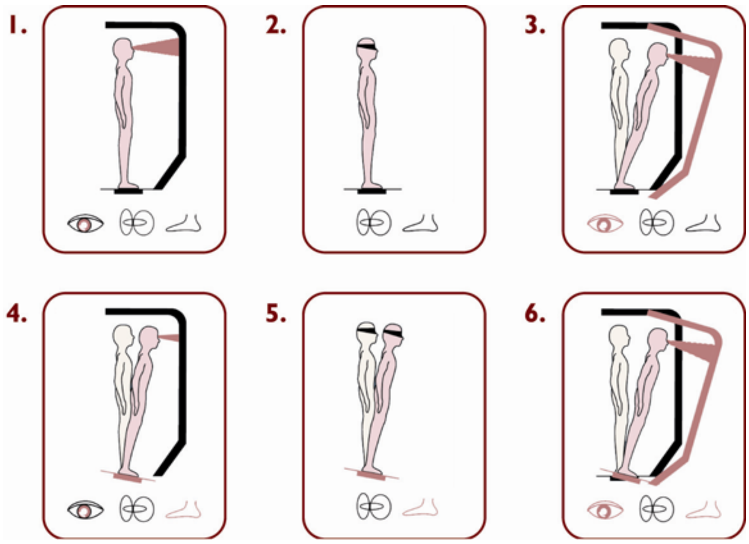


Figure 8.1. Sensory Organisation Test (SOT) Six Conditions, used courtesy of NeuroCom International, Inc. The black symbols indicate the sensory systems that were being challenged. The red symbols indicate the sensory systems that were receiving modified information through sway-referenced surround or support surfaces.

(used Courtesy of NeuroCom International Inc.)

Intraclass correlation coefficients for the average of 3 trials on the SOT ranged from 0.26 in Condition 3 to 0.68 in Condition 5 (Ford-Smith *et al.*, 1995). Conditions 1 to 2 (static balance) and Conditions 4 to 6 (dynamic balance) had fair to good reliability. The SOT composite score showed good reliability with an ICC of 0.66 (Ford-Smith *et al.*, 1995).

8.2.4. Motor Control Test

In the MCT, the participant's automatic postural reactions were measured in response to support surface translations and the order of testing was standardised. Participants maintained their eyes open and the surround remained stationary throughout the MCT. The MCT consisted of 6 conditions: graded backward (3) and forward (3) translations. The translations were scaled according to the person's height but durations were the same for everyone. Small, medium, and large translations produced 0.7° sway for 250ms, 1.8° sway for 300ms, and 3.2° sway for 400ms, respectively. Small translations represented threshold stimuli, large translations produced a maximal response and medium translations were midway between the small and large. Each translation occurred at a constant velocity and therefore transferred constant forward or backward angular momentum to the participant's body.

8.2.5. Dependent Variables

Dependent variables for the SOT included equilibrium and strategy scores during each trial and condition. Equilibrium scores were calculated by comparing the angular difference between the participant's measured maximum anterior to posterior COG displacement to the theoretical sway stability limit of 12.5° (Appendix E). Equilibrium scores ranged from 100 (little if any shear force and perfect stability) to 0 as it approached the participant's limits of stability. Loss of balance (score of 0) occurred when a person's sway had reached and exceeded their stability limits and they took a step or required support. The composite score was calculated by averaging the scores from conditions 1 and 2 and adding these 2 scores to the equilibrium scores from each trial of conditions 3, 4, 5, and 6, then dividing the total amount by the number of trials. The data were compared relative to the previously described normative data. The data for the participant, who was older than 79 years, was compared against the oldest category. In order to compare participants in different age groups, the values were normalised and expressed relative to the age-matched data. Measured equilibrium scores that were greater than the NeuroCom normative data were positive, whereas lower scores were negative. An average value was calculated for the 3 trials in each condition.

When the body's COG is accelerated, the body exerts shear forces onto the support surface. Sway movements about the ankles are low frequency (<0.5Hz) and generate small shear forces. Movements about the hips generate larger shear forces and are a higher frequency (>1Hz) (NeuroCom, 2004). Strategy scores were computed by comparing the peak-to-peak amplitude of the horizontal shear force with the maximal possible shear of 11.4kg (25lbs) (Appendix E). The maximal possible shear force represents the measured difference between the largest and smallest shear force generated by healthy participants using a hip strategy only to balance on a narrow beam (NeuroCom, 2004). Strategy scores ranged from 100 (full ankle strategy) to 0 (full hip strategy) but these were not compared against normative data. Scores in between 100 and 0 reflected the combination of both strategies. There was no NeuroCom normative data for strategy scores.

The latency scores from the MCT were determined by differentiation of force plate data from each foot. The participant's latency data were normalised to the age-matched values in the same manner as for the equilibrium score. Postural response latency was the elapsed time (in ms) between the onset of support surface translation until the point when the person actively resisted the induced sway. An active force response for each limb had to be registered in order to measure a latency score. The active force response is typically twice as large as the angular momentum due to postural sway, but in the opposite direction. If participants could not generate an active force response with the affected limb, or the change in active force was too small to be detected, then response latency could not be measured for that limb.

Weight symmetry (%) during the backwards and forwards translations of the MCT measured the distribution of total body weight over each limb and did not rely on the participant producing an adequate active force response. In order to interpret the data for the amputee groups, the data were scaled such that 0% represented perfect symmetry between the affected and intact limbs, –100% represented total intact limb and 100% represented total affected limb. For the control groups, –100% represented total left limb, 0% indicated perfect symmetry, and 100% represented total right limb. Because small translations served as threshold stimuli, only medium and large translations were analyzed.

8.2.6. Statistical Analysis

A one-way ANOVA was used to determine if falls' history had an effect on postural variables. If significant statistical differences were found between the tested groups, the least significant difference post hoc analysis was conducted to determine where the statistical difference existed.

8.3. Results

No significant differences were observed between any of the groups on age, height, body mass, and time since amputation.

8.3.1. Sensory Organisation Test

One amputee non-faller and two control fallers experienced one loss of balance during condition 6. Therefore, they scored 0 for that trial.

The equilibrium scores were greater than the age-matched normative value for all the groups, indicating overall good balance (Figure 8.2). No significant differences between either the amputee or control fallers vs. non-fallers were found for postural sway during conditions 1 to 5. In condition 6, a significant difference was noted ($F_{3,14}$=4.53, p=0.02) and post hoc analysis showed the amputee fallers scored significantly higher than the non-fallers (p=0.05), indicating better postural stability. Overall, a significant difference was found for composite score ($F_{3,14}$=4.78, p=0.02); however, post hoc analysis revealed that this difference did not exist between the fallers and non-fallers within the amputee or control groups.

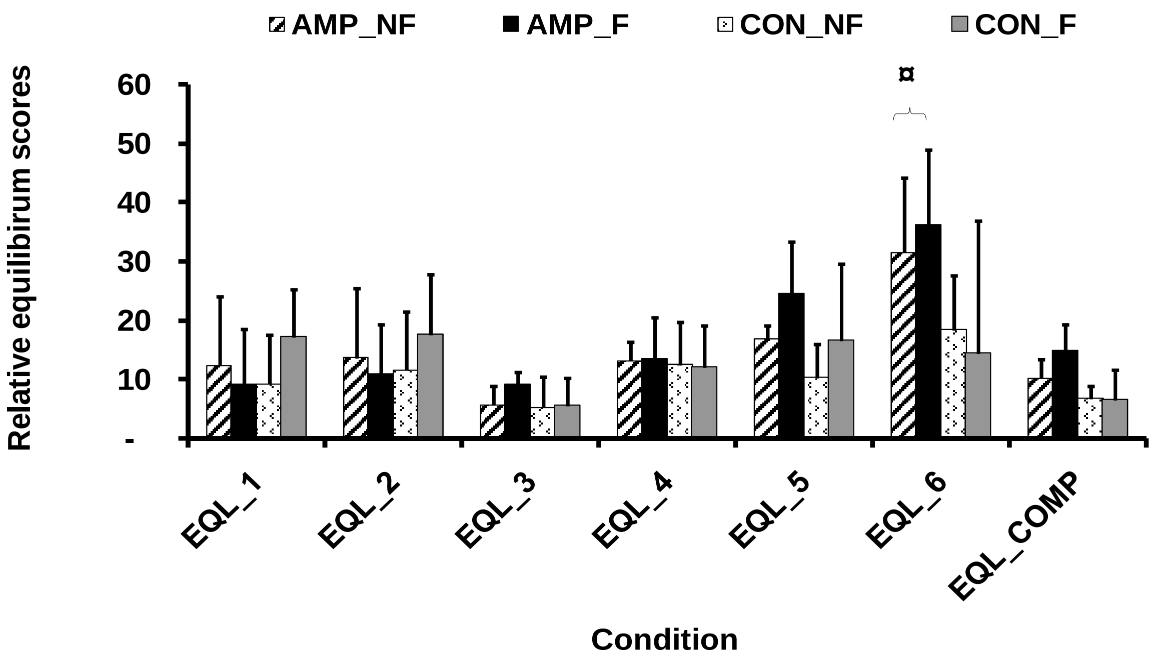


Figure 8.2. Relative equilibrium scores for the SOT. Positive scores indicate the absolute equilibrium score was greater than the age-matched, normative data, suggesting overall good balance.

Abbreviations: AMP_F, amputee fallers; AMP_NF, amputee non-fallers; CON_F, control fallers; CON_NF, control non-fallers.

¤ Significant difference between amputee fallers and non-fallers.

All groups predominantly used ankle strategies during static balance (with and without vision) and sway-referenced surround conditions (Figure 8.3). Reliance on the hip strategy increased as the tasks become more complex and anterior-posterior sway increased (lower equilibrium scores). There were no significant differences in strategy scores for conditions 1 to 5. A significant difference was found for condition 6 ($F_{3,14}$=5.88, p=0.01) and post hoc analysis showed that the control fallers had a significant lower strategy score compared to the control non-fallers (p=0.01), indicating a greater reliance on the hip strategy.

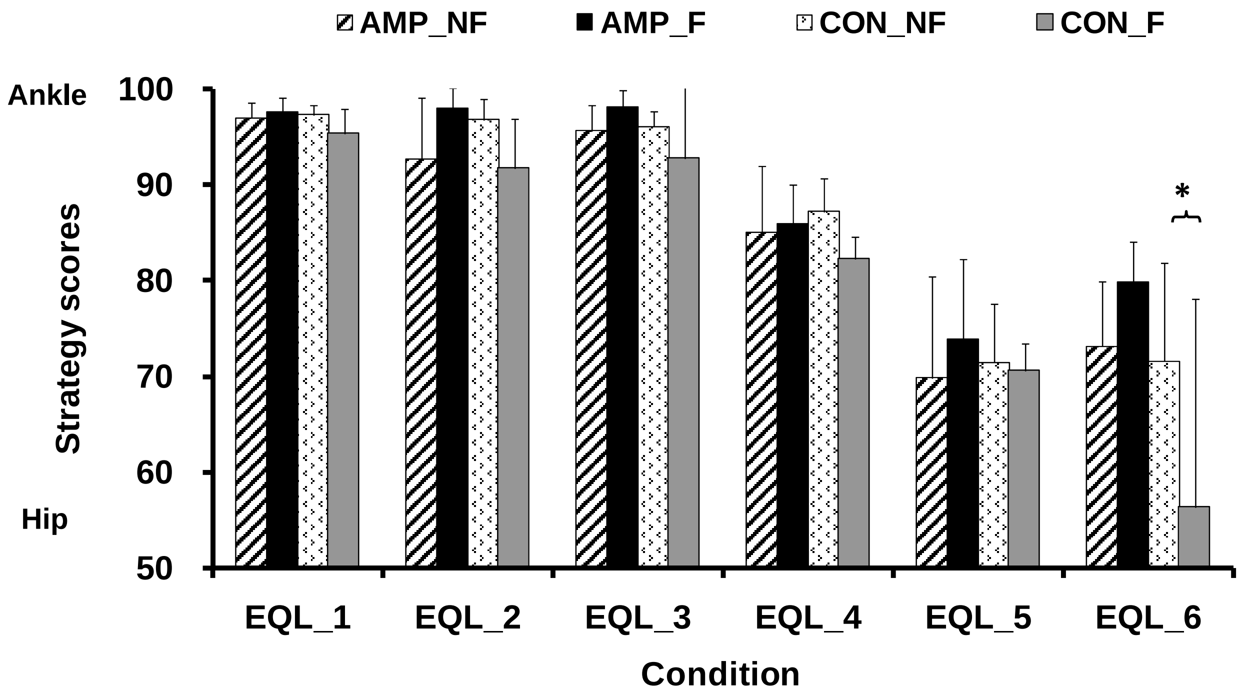


Figure 8.3. Strategy scores for the SOT. Scores approaching 100 indicate predominant use of the ankle strategy, while low scores near zero indicate total hip strategy. Scores in between indicate use of both ankle and hip strategies.

AMP_F, amputee fallers; AMP_NF, amputee non-fallers; CON_F, control fallers; CON_NF, control non-fallers.

***** Significant difference between control fallers and non-fallers.

8.3.2. Motor Control Test

The relative response latency for the intact limb, compared to normative data, is presented in Figure 8.4. The majority of amputee participants did not produce an active force response and so did not register a response latency for the affected limb. Thus, only the intact limb was used for analysis for the amputee groups. For the control groups, the average response latency for the right and left limb was used

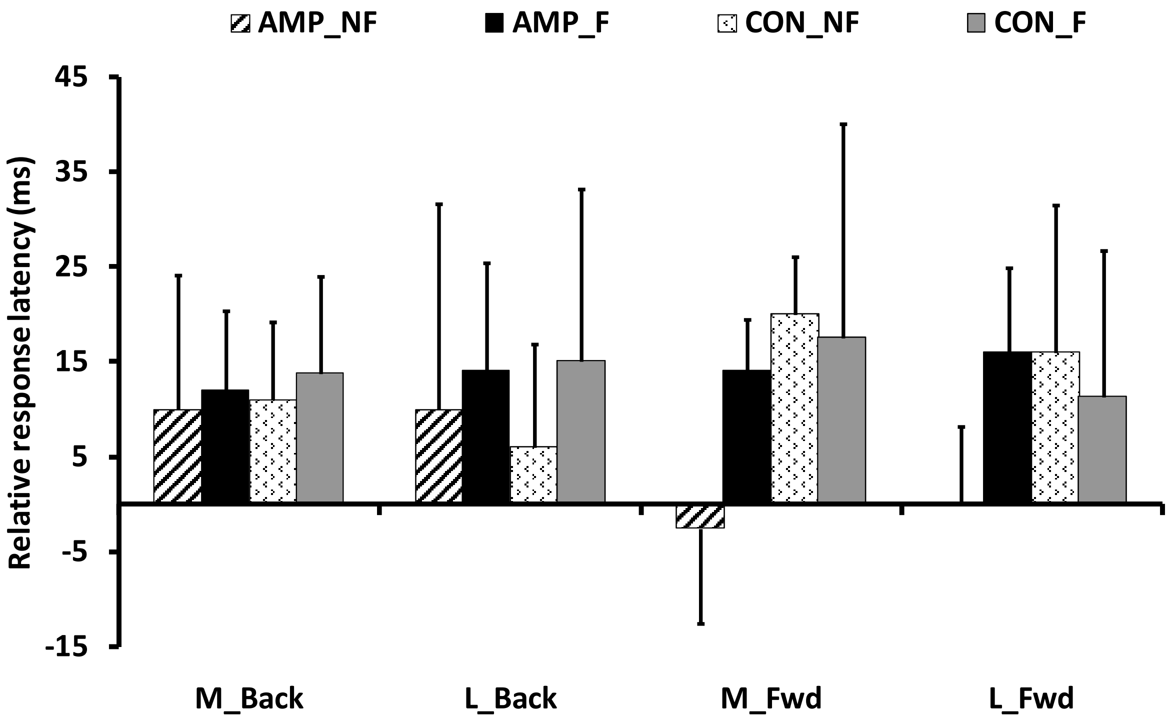


Figure 8.4. Mean (SD) relative response latency (ms) for the MCT. Positive values indicate the response latency was faster and negative values indicate the response latency was slower than the age-matched normative data.

AMP_F, amputee fallers; AMP_NF, amputee non-fallers; CON_F, control fallers; CON_NF, control non-fallers.

No significant differences were found for medium and large relative latency scores in the backwards or forwards direction. As evidenced by the large SD, variability of latency scores was high. Of all the trials, only the amputee non-faller group had a slower postural response compared to age-matched normative data for medium translations in the forwards direction.

Weight symmetry results are presented in Figure 8.5. Significant differences were found for medium and large forward ($F_{3,14}$=3.66, p=0.04 and $F_{3,14}$=3.98, p=0.03, respectively) and backward ($F_{3,14}$=5.70, p=0.01 and $F_{3,14}$=4.26, p=0.03) translations. Post hoc analysis revealed that the amputee fallers bore significantly more weight through the affected limb during all four conditions. There were no significant differences between the fallers and non-fallers in the control group.

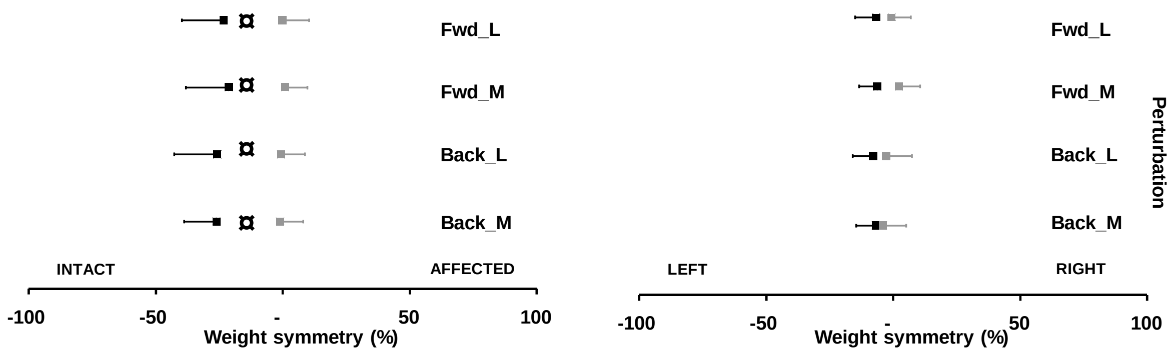


Figure 8.5. Weight symmetry during medium and large backwards and forwards translation for the amputee and control groups. Black shapes (■) represent non-fallers; grey shapes (■) represent fallers. A response strength of 0% indicates perfect symmetry between the 2 limbs. For the amputees, negative values indicate greater intact limb and positive values indicate greater affected limb response strength.

¤ Significant difference between amputee fallers and non-fallers

8.4. Discussion

The aim of the current study was to investigate whether CDP could be used to differentiate between fallers and non-fallers during static and dynamic conditions using the NeuroCom Smart Equitest system. This study is the first to objectively measure postural control in transtibial amputees using CDP with the NeuroCom Smart Equitest system. While previous studies have compared the balance of healthy elderly and young individuals (Camiciolo *et al*., 1997; Horak *et al*., 1989; Judge *et al*., 1995; Nardone *et al*. 2000) and elderly fallers and non-fallers (Parry *et al*., 1995; Wallmann, 2001), little has been documented about how people who have inherently altered neuromuscular and mechanical systems maintain balance in challenging, dynamic conditions.

8.4.1. Sensory Organisation Test

In the current study, 3 of the 18 participants experienced one loss of balance on condition 6 that would have resulted in a fall. Previous studies have reported that a loss of balance reflects the risk of falls in typical daily activities (Camiciolo *et al*., 1997; Judge *et al*., 1995). Failure to produce adequate shear force in trials with a loss of balance implies that participants could not generate sufficient ankle moments or had a response delay to body tilt (Judge *et al*., 1995). In the findings, there were no significant differences between control fallers and non-fallers despite the fact that two of the control fallers had a loss of balance. This could have reflected the relatively small sample size and high variability, although previous studies have noted that loss of balance during dynamic posturography trials did not exactly match falls history (Camicioli *et al*., 1997). One amputee non-faller had a loss of balance during condition 6. This suggests that the participant may have learned to adapt to the capabilities of the prosthesis during dynamic activities such as walking, while quiet standing in a challenging environment could have been more unfamiliar.

The results revealed that the amputee fallers scored significantly higher compared to the non-fallers in condition 6 suggesting that the non-fallers may rely more heavily on visual cues even if they are inaccurate. This could be an important finding, as the non-fallers may be at greater risk of experiencing a future fall in situations with reduced visual input. This finding is in agreement with a previous study that reported that lower-limb amputees compensate for proprioceptive deficits by increasing their dependence on visual input (Fernie and Holliday, 1978). Although it was not the focus of the current study to compare amputee and control participants, it was surprising to note that the amputee groups had higher relative equilibrium scores in condition 6 compared to the control participants. Previous research reported that traumatic amputees had reduced sway in the sagittal direction compared to healthy individuals (Hermodsson *et al*., 1994). These authors suggested that reduced sway may be caused by the relatively stiff ankle of the prosthesis and the fact that the amputees could rely on slight muscle movements of the intact limb to maintain postural control (Hermodsson *et al*., 1994). This may explain why the amputees, who mainly had an amputation due to trauma, had higher equilibrium scores in more challenging conditions.

A relationship between loss of balance on the SOT and poor balance performance has been observed previously (Judge *et al*., 1995; Parry *et al*., 1995). However, the use of the SOT to distinguish amputee and control fallers from non-fallers and to predict future falls may be questioned. Previous studies have attributed the lack of association of a person's falls history with the SOT to the heterogeneous nature of falls; they can occur during challenging or simpler tasks (Judge *et al*., 1995). The findings from the current study suggest that community-dwelling transtibial amputees and able-bodied participants are unlikely to fall during quiet standing, even when visual input is reduced, and that static balance is not a sufficiently challenging task to distinguish between fallers and non-fallers. It may be that the more active amputees are at risk of falling when performing challenging tasks that the frailer amputees, with greater balance impairments, would likely not perform. Alternatively, a fall could result from the inability of the locomotor system to execute a successful response during a simple task in both the amputee and control groups. The current results support previous research (Judge *et al*., 1995) that loss of balance during the SOT has the potential for identifying people who have difficulty with activities of daily living, ambulation, and general mobility task, but who are still functioning independently. Therefore, the SOT may be better used as a form of risk assessment rather than diagnostic tool among community-dwelling amputee and control fallers and non-fallers.

No significant differences were found between the amputee or control fallers and non-fallers in the composite score of the SOT. Furthermore, the group composite scores were all better compared to the normative data. This is in contrast to Parry *et al*. (1995) who found that the composite score of the SOT totally distinguished between a fallers group and an active elders group. These authors reported that the composite score was 100% sensitive and therefore a good diagnostic test for identifying fallers as having impaired balance. The findings suggest that the SOT is population specific and may not be an appropriate diagnostic test for reliably identifying fallers among transtibial amputees or distinguishing between community-dwelling control fallers and non-fallers.

Another aim of this study was to measure and quantify the postural control strategies in transtibial amputees and able-bodied controls. All groups scored high on the strategy scores for conditions 1 and 2, indicating that fallers and non-fallers were able to maintain balance by predominantly using an ankle strategy. Transtibial amputees are unable to generate an ankle strategy, and therefore must rely entirely on the hip or trunk strategies to maintain balance (Viton *et al*., 2000). As the strategy score is computed from the overall shear force measured by the central transducer, it cannot be analysed for each limb separately. It is possible that the overall ankle strategy was measured for the intact limb, while the affected limb was unresponsive. This could have resulted in overall low frequency shear forces, suggesting the use of the ankle strategy. However, these results support previous research that the intact limb plays a critical role in maintaining postural control in amputees (Aruin *et al*., 1997).

Even with inaccurate visual information (condition 3), amputee and control participants relied on the ankle strategy. Previous studies have found that the ankle strategy is successful for recovery of postural sway within 8° of forwards and 4° of backwards sway (Horak *et al*., 1989). In conditions 4 and 5, when somatosensory information was inaccurate, strategy scores were lower compared to conditions 1 through 3 indicating that amputee and control fallers and non-fallers used a combination of ankle and hip strategies but were still able to prevent a loss of balance. In condition 6, the control fallers scored significantly lower than the non-fallers, with two participants recording a loss of balance on one trial in this condition. These results indicate that people, who use a predominantly hip or stepping strategy when visual and somatosensory cues are inaccurate, have a higher risk of falling and this supported the hypothesis of this study.

8.4.2. Motor Control Test

No one experienced a loss of balance during the MCT. All groups had faster response latencies compared with the NeuroCom normative data during the backwards translations. Only the amputee non-fallers showed slower or comparable latency responses compared with the normative data (Figure 8.4). For the majority of amputee participants, the affected limb did not generate a sufficiently large active force response needed to measure latencies. Therefore, the results from the current study do not further our understanding about the postural response time to perturbations of the affected limb. Although this information would be valuable in understanding the time lapse between efferent input and afferent output during displacement of the affected limb, the findings indicate the MCT is not a suitable test to understand response latencies in transtibial amputees.

Weight symmetry between the affected and intact limbs consistently distinguished between amputee fallers and non-fallers. The amputee fallers bore significantly more weight through their affected limb during forward and backward translations compared to the non-fallers which supported the hypothesis. Conversely, the fallers had an overall weight distribution closer to 0%, indicating better inter-limb symmetry. Matching postural response to disturbance and having the physical ability to execute the correct response safely, quickly and effectively is imperative for maintaining balance (Horak *et al*., 1989), especially in more challenging environments. It is possible that, in an attempt to maintain inter-limb symmetry, the fallers were relying too heavily on the affected limb to correct for postural disturbances during rapid, dynamic movements. Unlike the control groups, that showed good symmetry between both limbs, the affected limb is less flexible, has diminished strength and has inherently altered somatosensory input compared to the intact limb. Greater reliance on the intact limb, through rapid motion of the COP towards the intact limb in response to perturbations, appeared to be a successful strategy among the amputee non-fallers. Prosthetic fitting may have influenced the findings. However, none of the participants experienced pain while standing or walking and were generally content with their prosthesis. This indicated that prosthetic alignment was satisfactory (Isakov *et al*., 1994) and likely did not affect the weight symmetry results.

Conversely, in the control group, weight symmetry did not differentiate between the fallers and non-fallers. This suggests that able-bodied people, with no apparent physiologic, mechanical, or muscular asymmetry in the lower-limbs, can detect a disturbance and execute an appropriate response with both limbs. The ability to maintain weight symmetry during a postural disturbance is likely more critical among able-bodied people than among amputees. Furthermore, weight symmetry could be a discerning variable in determining a person's ability to recover from a forwards or backwards translation than a person's falls history alone.

These findings could have important implications for amputee rehabilitation and gait retraining. Exercises that train the neuromuscular system to move the COP rapidly towards the intact limb in response to postural disturbances could help mimic challenging situations and yet be performed in a safe rehabilitation environment. Furthermore, measuring weight distribution in amputees during quiet standing and some dynamic movements could be done with the use of simple scales and presents a cost effective method of evaluating weight symmetry. These results also have implications for elderly falls treatment and prevention programs. These programs should encourage people to challenge the biological systems responsible for balance, particularly the somatosensory system, by performing tests on different support surfaces. Additionally, amputees and the elderly should explore their postural limits of stability and practice selecting the appropriate responses to destabilizing forces performed in a safe, controlled environment.

8.5. Conclusions

Understanding effective postural responses is important for falls prevention and treatment programs in amputees and the elderly. The current study has shown that the SOT and MCT on the NeuroCom Equitest may be population specific and thus may not be suitable diagnostic tests for reliably identifying fallers among transtibial amputees. However, a loss of balance during the SOT and MCT may identify amputees and control fallers who have difficulty in performing more challenging tasks, but who are otherwise independent. Contrary to the hypothesis, the amputee fallers scored better on Condition 6 of the SOT, when somatosensory and visual input was inaccurate, indicating less postural sway. In the able-bodied control group, the fallers evidenced greater reliance on the hip strategy to control posture in more challenging dynamic conditions and this result supported the hypothesis. As hypothesised, weight symmetry during dynamic movements was an important differentiating variable between the amputee fallers and non-fallers. These findings have revealed that amputee and control fallers can prevent a fall during challenging static and dynamic conditions by adapting their neuromuscular responses.

CHAPTER NINE - BALANCE CONFIDENCE, QUALITY-OF-LIFE AND FUNCTIONAL PERFORMANCE IN FALLERS VS. NON-FALLERS

9.1. Introduction

The relationship between falls and function has been well-documented (Kerrigan *et al*., 1998; Kerrigan *et al*., 2000; Mian *et al*., 2007; Tinetti, 1986; Tinetti *et al*., 1988). Falls often lead to activity avoidance, loss of independence and mobility. Another consequence of falling is that the person may develop low falls efficacy or balance confidence. Poor balance confidence has been recognised as one of the most important predictors of fall severity in older adults (Bishop *et al*., 2007). However, balance confidence is modifiable and it is important that it is recognised as an important factor in falls prevention and treatment programmes.

To date, one group of authors have focused on balance confidence (Miller *et al*., 2004) and fear of falling (Miller *et al*., 2001; Miller *et al*., 2003) specifically among lower-limb amputees. While they have reported low balance confidence (using the ABC scale) and fear of falling among this population, individuals were not actually distinguished according to their falls history. Having fallen is likely to influence a person's balance confidence. The terms 'balance confidence', 'falls efficacy' and 'fear of falling' are used interchangeably in the published literature and also in this thesis. There is insufficient empirical research into balance confidence and functional performance in amputee fallers vs. non-fallers. The first aim of this study was to determine if fallers and non-fallers differed in their time to complete functional performance tests, and had different scores for balance confidence (MFES) and quality-of-life (SF-36). The second aim was to determine if there was a relationship between the functional tests and the psychological measures (balance confidence and quality-of-life). The third aim was to understand if there were any relationships between balance confidence (MFES) and quality-of-life (SF-36) when categorised according to a person's falls history. It was hypothesised that previous fallers would perform the functional performance tasks more slowly and have lower balance confidence on everyday activities and quality-of-life scores compared to the non-fallers. With regards to the second aim, it was also anticipated that lower performance scores would be associated with lower balance confidence and quality-of-life scores. The third hypothesis was that low quality-of-life scores on the SF-36 would be related to poor balance confidence as measured with the MFES.

9.2. Methods

9.2.1. Participants

Eleven transtibial amputees and nine age-matched able-bodied participants completed two physical performance tasks (TUG and 10-metre walk tests) and two psychometric measures (the SF-36® Health Survey and Modified Falls Efficacy Scale). All participants gave written informed consent to take part in this study. Participant details according to falls history can be found in Table 3.1.

9.2.2. Functional performance tasks

Timed Up and Go (TUG) test: The TUG test is a modified version to the Get-Up and Go test developed by Mathias and colleagues (1986). The Get-Up and Go and TUG tests have been used to assess function in frail, elderly people. The tests incorporate basic mobility skills such as rising from a seated position, walking three metres, turning 180°, returning to the chair and resuming a seated position. The test has good agreement in time scores between-raters (ICC 0.99) and within-raters (ICC 0.99). The TUG test has good intra-rater and inter-rater reliability ($r = 0.93$ and $r = 0.96$, respectively) for older individuals with lower-limb amputation and correlated well with the Groningen Activity Restriction Scale (GARS) (Spearman correlation coefficient 0.39, $p < 0.03$) and the Sickness Impact Profile, 68-item version (SIP68) (Spearman correlation coefficient, 0.40), suggesting adequate concurrent validity (Schoppen *et al.*, 1999)

The TUG test is considered a useful measurement tool because it describes a realistic mobility assessment including potential fall situations, such as transfer in an out of a chair, walking and turning. Furthermore, Podisaldo and Richardon (1991) suggested that TUG scores reflected a person's basic mobility skills and could be used as a screening tool, by indicating the patient's level of functional mobility, or a descriptive tool, depicting the person's level of functional capacity. Individuals, who could complete the test in less than 20 seconds were independent for basic transfers, could negotiate stairs and could go out independently. Those who took 30 seconds or longer were more dependent and needed assistance in accomplishing simple tasks like getting in and out of a chair and/or tub/shower and no one in this group could go out alone (Podisaldo and Richardon, 1991).

In the current study, participants were asked to complete the TUG test three times and an average value was then calculated. A standard armchair was used, with a seat height of 46 cm and arm height of 65 cm, as recommended by Podsiadlo and Richardson (1991).

Timed 10-metre walk test: Timed walk tests are simple and easy test to administer in a variety of settings and can provide clinical information about normal and pathological gait abilities. The 10-metre walk test records the time taken to cover 10 metres, which is considered the minimum functionally significant distance to achieve independent ambulation (Watson, 2002). Furthermore, the test can be modified by increasing walking distance (20, 40, 100 metres) or by asking participants to vary their walking speed selectively (Watson, 2002). Good levels of inter-rater reliability were found for young healthy individuals (-0.38 to +0.38 seconds, 95% CI of agreement) (Watson, 2002) and individuals with pathological gait due to neurological trauma (-0.36 to +0.49 seconds, 95% CI of agreement) (Colleen *et al*., 1990). Thompson *et al*. (2008) reported test-retest ICC of 0.81 (95% CI 0.65-0.90) for children with cerebral palsy.

In the current study, participants were asked to walk along gait walkway in the laboratory at their self-selected pace. The starting and finishing positions over 10 metres were marked on the floor and the investigator used a stopwatch to time how long it took the participant to cover that distance. Gait initiation and termination were not included and only steady state walking was measured.

9.2.3. Psychological instruments

Two psychological instruments were used to measure overall quality-of-life and balance confidence.

Short Form (SF-36): The SF-36 Health Survey (Appendix F) is a general health (quality-of-life) measure composed of 36 questions (Ware and Sherbourne, 1992). It is a multi-item scale that measures eight different health attributes with 2 to 10 items in each: 1) physical functioning, 2) role limitations due to physical health problems, 3) bodily pain, 4) general health, 5) vitality, 6) social functioning, 7) role limitations due to emotional problems and 8) mental health (psychological well-being). The SF-36 is easy to use, can be self-administered and relatively quick to complete. The scales use Likert's method of summated ratings (1932). The SF-36 items cover a large spectrum of tasks that affect a person's roles, such as limitations in work or typical daily tasks; reducing the amount of time typically dedicated to work or other daily tasks and difficulty performing work or other activities. Although disease-specific versions of the SF-36 exist (e.g. Cancer, Heart Disease), an amputation-specific version of the SF-36 has not been developed. Therefore, the generic inventory was used in this study.

The SF-36 evaluates *Physical functioning* across different functional levels on self-care activities, including walking moderate distances; lifting and carrying groceries; bending, kneeling and stooping; and climbing stairs. *Role limitations* items cover both physical and emotional problems. There are two SF-36 items on *Bodily pain* that evaluate the intensity of bodily pain/discomfort and investigate how pain interferes with normal activities. *General health* is measured using five items rating health on a continuum from as excellent to poor. *Vitality* refers to a person's energy levels and fatigue and is measured on four items. *Social functioning* reflects a person's social activities, both in terms of frequency and enjoyment with others, and is measured on two items. *Mental health* is assessed using five items from each of the four main mental health dimensions (anxiety, depression, loss of emotional control, and psychological well-being). When administered to an elderly population, Lyons *et al.* (1994) found good internal consistency with Cronbach's alpha greater than 0.8 (ranging from 0.83 to 0.94) for each health parameter. These authors also found the SF-36 was able to distinguish between elderly persons with and. without markers of poor health (Lyons *et al.*, 1994).

Physical and mental component scales (PCS and MCS, respectively) on the SF-36 were calculated using the norm-based scoring method developed by Ware (1994) (see Appendix G). In this method, the means and standard deviations of the 1998 U.S. population and the 1990 factor score coefficients are combined. The physical scale uses scores from Physical functioning, Role-Physical, Bodily Pain and General Health, whereas the mental scale uses scores from Vitality, Social functioning, Role-Emotional and Mental Health. In the current study, the PCS and MCS were calculated for the amputee and control data separately, and were compared with "Norms for limitations in the use of an arm(s) or leg(s): General U.S. population". Limitation in use of arm(s)/leg(s) were defined as the self-reported limitations of use of an arm or leg that was missing, paralysed or weakened (Ware and Kosinski, 2001).

Modified Falls Efficacy Scale (MFES): The MFES (Appendix F) was developed as a more sensitive measure of fear of falling than the Falls Efficacy Scale, originally created by Tinetti *et al*. (1990). The MFES is comprised of 14 items in total and has proven useful in detecting early stages of fear of falling in relatively active, community-dwelling elderly individuals. It includes ten of the original activities in the FES such as dressing, preparing meals, bathing, rising from a chair and bed, walking inside the home, reaching into cabinets, light housekeeping and doing simple shopping. The additional 4 items were added to reflect a person's confidence in performing outdoor-activities, such as gardening, crossing roads and using public transport. Participants were asked to rate their confidence at performing the activities without falling on a visual analogue scale from 0 (not at all confident) to 10 (completely confident) (Hill *et al*., 1996). Hill *et al*. (1996) found that the mean score for healthy older people was 9.76 ± 0.32 compared with a mean score of 7.69 ± 2.21 for previous fallers (this also indicates a skew in the data). Cronbach's alpha for internal consistency was 0.95 and ICCs were high for test-retest reliability (0.93). Two Factors (indoor- vs. outdoor-type activities) were identified that could account for 75% of the sample variance. Factor 1 consisted of 10 indoor-type activities. Factor 2 consisted of 6 outdoor-type activities. Two activities (reaching and using steps) were included in both Factors (Hill *et al*., 1996).

9.2.4. Statistical analysis

To determine whether falls history had a significant effect on TUG and 10-metre walk times, a one-way ANOVA was used to compare the amputee groups and the two control groups. If significant differences were found overall, Fisher's least significant difference (LSD) post-hoc test was used to determine whether the differences existed between the amputee fallers vs. non-fallers and control fallers vs. non-fallers. If no significant differences were found, between the groups on functional performance tests, then the data were grouped according to falls history only. This was done because having fallen previously was considered to have a greater influence on balance confidence than having an amputation. Independent sample t-tests were used to determine if falls history had a significant effect on balance confidence between fallers and non-fallers. Levene's test for equality of variances was used to assess homogeneity. In the instance of violation of homogeneity of variance the corrected t-value was used. To assess the relationship between functional tasks and psychological data, Pearson's Product Moment Correlation Coefficients were used. Effect sizes were categorised as follows: low $r < 0.3$, moderate $0.3 < r < 0.5$ and high $r < 0.5$ (Cohen, 1992). The alpha level for significance was set a priori at 0.05.

9.3. Results

9.3.1. Functional performance tasks

All participants were able to complete the TUG and 10-metre walk tests successfully without falling. The results are presented in Table 9.1. The one-way ANOVA found no significant differences on TUG time ($F_{(3,16)} = 0.92$, $p = 0.46$) or 10-metre walk time ($F_{(3,16)} = 1.36$, $p = 0.29$) between the four groups. An independent t-test was conducted to see whether falls history had an effect on functional tests. No significant results were found for TUG ($t_{18} = -1.57$; $p = 0.13$) or 10-metre walk time ($t_{18} = -1.02$; $p = 0.32$).

Table 9.1. Mean (SD) TUG and 10m walk times (s) presented for the amputees, controls and overall for the fallers and non-fallers

	Amputee		Control		Overall	
	Non-faller	**Faller**	**Non-faller**	**Faller**	**Non-faller**	**Faller**
TUG time (s)	11.5 (1.5)	13.1 (5.3)	9.9 (2.5)	13.1 (2.7)	10.7 (2.1)	13.1 (4.3)
10m walk time (s)	9.9 (2.0)	9.5 (3.0)	7.3 (2.2)	10.2 (2.4)	8.6 (2.4)	9.8 (2.7)

TUG: Timed Up and Go test

As expected, strong relationships (at $p < 0.01$ level) were found for performance variables in both the fallers and the non-fallers. Overall for the non-fallers, significant positive correlations were found for TUG tests and the 10-metre walk test ($r = 0.84$). For the fallers, strong positive correlations were found between TUG tests and the 10-metre walk test ($r = 0.79$).

9.3.2. Psychological data

Results from the SF-36 and MFES questionnaires are presented in Figures 9.1 and 9.2, respectively. Overall, non-fallers scored higher on all items on the SF-36 survey and significant differences were found between the groups for some SF-36 items. The fallers rated their general health ($t_{12.37} = 3.15$; $p = 0.01$), vitality ($t_{18} = 2.39$; $p = 0.03$) and emotional role ($t_9 = 2.45$; $p = 0.04$) significantly lower than the non-fallers. The results also showed the non-fallers had a trend towards better physical role compared to the fallers ($p = 0.06$).

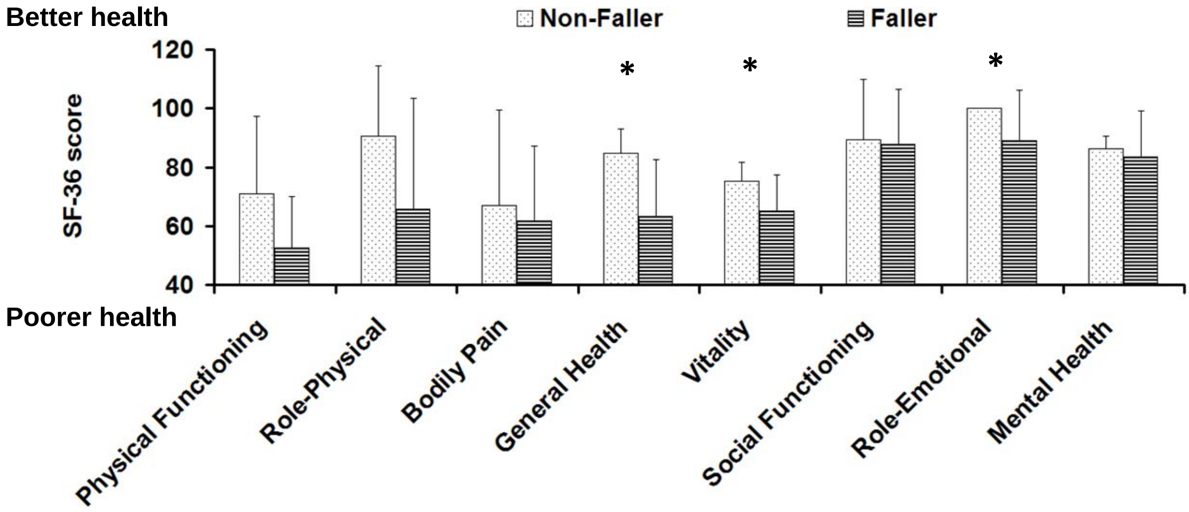


Figure 9.1. Mean (SD) SF-36 scores for all 8 items for the fallers and non-fallers. Both groups contain data from amputees and controls.

***** Indicates significant difference between fallers and non-fallers (independent samples t-test)

Examining the different MFES activities individually revealed that the fallers scored significantly lower on tasks such as light gardening/hanging out the washing (t_{18} = 2.13; p = 0.05) and using steps (t_{18} = 2.38; p = 0.03) (Figure 9.2).The differences in mean MFES scores between the fallers and non-fallers just failed to reach significance (t_{18} = 1.90; p = 0.07). No significant difference existed on indoor Factor 1 activities (t_{18} = 1.47; p = 0.16) but the fallers had significantly lower outdoor MFES Factor 2 activity scores compared to the non-fallers (t_{18} = 2.35; p = 0.03) (Figure 9.3).

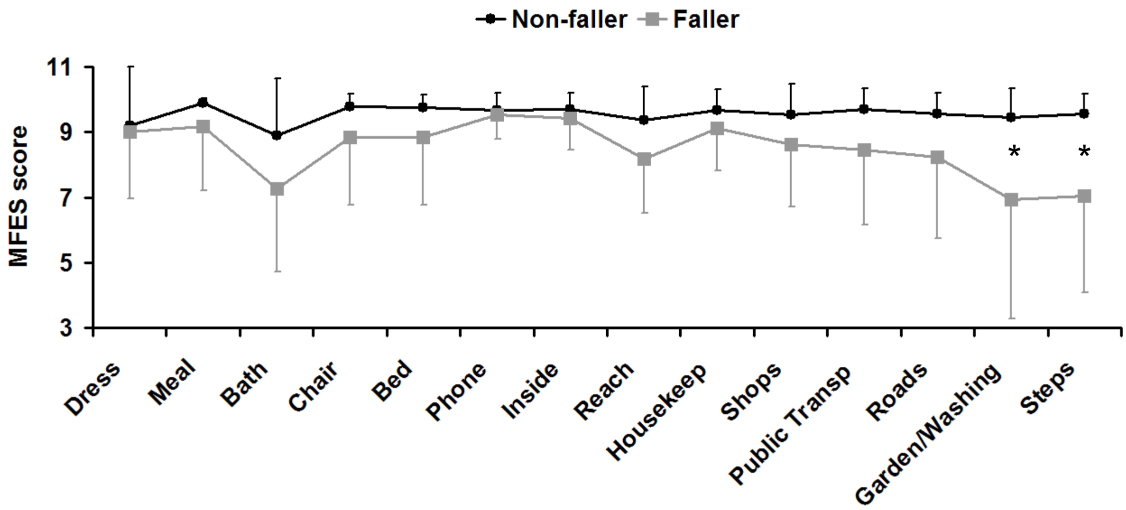


Figure 9.2. Mean (SD) MFES scores of all 14 items for the fallers and non-fallers. Both groups contain data from amputees and controls.

* Indicates significant difference between fallers and non-fallers (independent samples t-test)

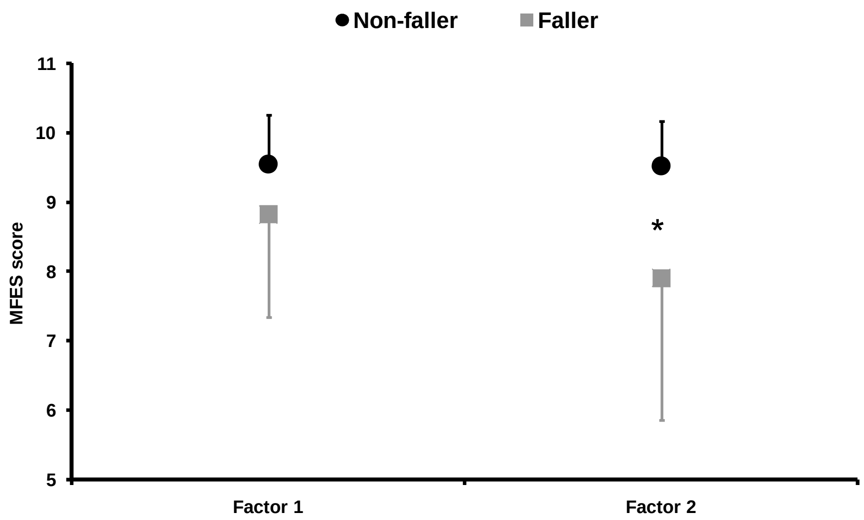


Figure 9.3. Mean (SD) self-rated confidence at performing indoor- (Factor 1) vs. outdoor-type (Factor 2) activities on the MFES. Both groups contain data from amputees and controls.

* Indicates significant difference between fallers and non-fallers (independent samples t-test)

9.3.3. Relationships between functional tasks and psychological components

The correlation matrices for the non-fallers, fallers and all participants combined can be found in Table 9.2. Relationships between SF-36 items that were significantly different between the non-fallers and fallers, were further examined in relation to performance variables (TUG test and 10-metre walk test). In the fallers, general health was negatively correlated with functional performance on the TUG test ($r = -0.66$; $p < 0.05$) and the 10-metre walk test ($r = -0.69$; $p < 0.05$), whereas no significant relationships were found for the non-fallers. No significant relationships were found for either vitality or emotional role and functional performance in either group. For all participants, physical function was negatively correlated with the 10-metre walk test ($r = -0.58$; $p < 0.01$) and general health was negatively correlated with functional performance on the TUG test ($r = -0.54$; $p < 0.05$) and the 10-metre walk test ($r = -0.45$; $p < 0.05$).

9.3.4. Relationships between balance confidence (MFES) and physical performance

The relationship between balance confidence and functional performance was measured by analysing Factors 1 and 2 of the MFES with TUG test and the 10-metre walk test. In the fallers, balance confidence on indoor-type activities (Factor 1) was not significantly correlated with the TUG test ($r = -0.16$) or 10-metre walk time ($r = -0.60$). Outdoor-type activities (Factor 2) were negatively correlated with 10-metre walk times ($r = -0.78$; $p < 0.01$), but no significant relationship was found with the TUG test ($r = -0.42$). Similarly, the non-fallers did not show any significant relationship between their balance confidence and physical performance on any of the activities on the MFES scale. Balance confidence on Factor 1 activities were not significantly correlated with TUG test ($r = -0.27$) or 10-metre walk time ($r = -0.18$). On Factor 2 activities, there were no significant relationships with either TUG test ($r = -0.45$) or 10-metre walk time ($r = -0.31$). For all participants, balance confidence on indoor-type activities was negatively correlated with 10-metre walk test performance ($r = -0.49$; $p < 0.05$) and balance confidence on outdoor-type activities was negatively correlated with TUG test ($r = -0.51$, $p < 0.05$) and 10-metre walk time ($r = -0.64$, $p < 0.01$).

9.3.5. Relationships between balance confidence (MFES) and quality-of-life (SF-36)

Only two significant relationships were found for balance confidence and quality-of-life in the fallers group. Balance confidence on indoor-type activities (Factor 1) was positively correlated with vitality ($r = 0.77$; $p < 0.01$), whereas balance confidence on outdoor-type activities (Factor 2) was positively correlated with physical function ($r = 0.77$; $p < 0.05$) and general health ($r = 0.64$; $p < 0.05$). In the non-fallers, no significant correlations existed between balance confidence as measured by the MFES and quality-of-life measured using the SF-36. For all participants, physical function, general health and vitality were positively correlated with balance confidence on indoor-type ($r = 0.47$; $p < 0.05$, $r = 0.52$; $p < 0.05$ and $r = 0.67$; $p < 0.01$, respectively) and outdoor-type activities ($r = 0.60$; $p < 0.01$, $r = 0.67$; $p < 0.01$ and $r = 0.60$; $p < 0.01$, respectively), Emotional and mental health were positively correlated with both Factor scores on the MFES and these relationships were all significant at the $p < 0.05$ level.

9.3.6. Comparison of PCS and MCS with norms for limitations in the use of an arm(s) or leg(s)

PCS and MCS values can be found in Table 9.3. A one-sample t-test was used to determine whether the amputees in the current study were significantly different with the U.S. data ($p < 0.05$). No significant difference was found for the PCS component ($t_{10} = 1.03$; $p = 0.33$) or age ($t_{10} = 0.95$; $p = 0.37$). However, the amputees in the current study scored significantly higher on the MCS component ($t_{10} = 7.13$; $p = 0.00$) compared to the US population data.

Table 9.2. Pearson correlation matrices on functional tests (TUG time, 10m walk test), MFES (Factors 1 and 2) and SF-36 categories for the non-fallers, fallers and all participants combined

	Non-fallers (n=10)		Fallers (n=10)		All participants (n=20)	
Correlations	**r value**	***p* value**	**r value**	***p* value**	**r value**	***p* value**
TUG / SF36 physical function	-0.42	0.23	-0.33	0.36	-0.41	0.08
TUG / SF36 physical role	0.09	0.81	-0.26	0.47	-0.30	0.21
TUG / SF36 body pain	-0.27	0.46	0.29	0.42	0.02	0.94
TUG / SF general health	0.57	0.08	-0.66	0.04	-0.54	0.02
TUG / SF vitality	-0.25	0.50	-0.19	0.60	-0.33	0.15
TUG / SF social fucntion	-0.02	0.95	-0.61	0.06	-0.37	0.11
TUG / emotional role	could not be computed †		-0.10	0.79	-0.24	0.30
TUG / mental health	-0.14	0.71	-0.40	0.92	-0.12	0.63
10 m WT / SF36 physical function	-0.61	0.06	-0.51	0.14	-0.58	0.01
10 m WT / SF36 physical role	-0.11	0.76	-0.11	0.77	-0.19	0.41
10 m WT / SF36 body pain	-0.50	0.14	0.20	0.58	-0.19	0.42
10 m WT / SF general health	0.27	0.44	-0.69	0.03	-0.45	0.05
10 m WT / SF vitality	0.01	0.99	-0.43	0.21	-0.36	0.12
10 m WT / SF social function	-0.34	0.34	-0.35	0.32	-0.35	0.13
10 m WT / emotional role	could not be computed †		-0.04	0.91	-0.14	0.55
10 m WT / mental health	0.23	0.53	-0.22	0.53	-0.16	0.50
TUG / MFES Factor 1	-0.27	0.45	-0.16	-0.66	-0.27	0.24
TUG / MFES Factor 2	-0.45	0.19	-0.42	0.23	-0.51	0.02
10 m WT / MFES Factor 1	-0.18	0.63	-0.60	0.07	-0.49	0.03
10 m WT / MFES Factor 2	-0.31	0.38	-0.78	0.01	-0.64	0.00
MFES Factor 1 / SF physical function	0.38	0.27	0.51	0.14	0.47	0.04
MFES Factor 1 / SF physical role	-0.03	0.93	0.17	0.63	0.25	0.30
MFES Factor 1 / SF body pain	0.34	0.33	0.44	0.21	0.38	0.10
MFES Factor 1 / SF general health	-0.01	0.99	0.51	0.13	0.52	0.02
MFES Factor 1 / SF vitality	0.01	0.98	0.77	0.01	0.67	0.00
MFES Factor 1 / social function	-0.05	0.90	0.13	0.72	0.80	0.74
MFES Factor 1 / emotional role	could not be computed †		0.44	0.20	0.49	0.03
MFES Factor 1 / mental health	-0.19	0.59	0.52	0.12	0.46	0.04
MFES Factor 2 / SF physical function	0.41	0.24	0.77	0.01	0.60	0.01
MFES Factor 2 / SF physical role	-0.04	0.92	0.19	0.59	0.32	0.16
MFES Factor 2 / SF body pain	0.35	0.32	0.32	0.37	0.30	0.21
MFES Factor 2 / SF general health	-0.10	0.79	0.64	0.05	0.67	0.00
MFES Factor 2 / SF vitality	0.08	0.83	0.55	0.10	0.60	0.01
MFES Factor 2 / social function	-0.06	0.88	0.38	0.27	0.23	0.32
MFES Factor 2 / emotional role	could not be computed †		0.35	0.33	0.49	0.03
MFES Factor 2 / mental health	-0.15	0.69	0.48	0.16	0.47	0.04

TUG: Timed Up and Go test

MFES: Modified Falls Efficacy Scale

Shaded areas indicate significant correlations at p < 0.01 and p < 0.05 levels

† The relationship of SF-36 category ‘emotional role’ could not be computed with any of the functional or MFES parameters because all non-fallers had exactly the same score in this category.

Table 9.3. PCS and MCS scores for the US population with limited arm/leg use and lower-limb amputees in the current study

	PCS	MCS	Mean age (yrs)	N
General US scores	37.7	45.9	51.5	263
Current study	40.6	57.6 ‡	56.2	11

‡ Indicates significant difference between current study and general US scores

9.4. Discussion

9.4.1. Functional performance tests

It is widely recognised that functional performance declines with increasing age (Kerrigan *et al*., 1998, 2000; Mian *et al*., 2007). The first hypothesis related to performance of functional tasks and that the fallers would perform tasks more slowly. The TUG scores from the current study were not significantly different between the fallers and non-fallers. This was despite the fact the fallers performed the TUG test 2.4 seconds faster overall, which could be considered a clinically meaningful difference. The lack of statistical significance could be attributed to the large SD in the amputee and control fallers (Table 9.1), and overall small sample size. Furthermore, the results showed that TUG scores did not differ between community-dwelling fallers, as the average time to complete the test was the same for all the fallers (Table 9.1). Podsiadlo and Richardson (1991) stated that medically stable patients did not vary in their TUG scores over time. The TUG test did not appear to be sensitive enough to distinguish between community-dwelling, independent amputee and control fallers in the current study.

Large *et al*. (2006) suggested that it was a person's ability or inability to complete the TUG test that was the most important indicator for stratifying patients according to their risk of falls and concluded that excluding patients who were unable to complete the task failed to detect those at the highest risk for falls. The inclusion criteria of the current study stipulated that participants were able to perform the sit-to-stand task independently. It was likely that participants, who were at the greatest risk of falling, were excluded from the current study because of their inability to rise from a seated position independently. Based on these findings, it is recommended that older people and lower-limb amputees with a falls history attending outpatient treatment are asked to complete the TUG test in a safe, controlled environment. This would have several advantages, including the identification of clinical change, and could be monitored according to the person's falls history. Conversely, active amputees would benefit from performing more sensitive measures, such as the L test (Deathe and Miller, 2005).

The strong correlations found between TUG time and 10-metre walk time have been previously reported in other studies (Podsiadlo and Richardson, 1991). The results showed that speed was an overall good descriptor of functional performance but the results did not support the hypothesis that fallers would perform more slowly than the non-fallers. Similar findings have been noted previously as Deathe and Miller (2005) found the TUG test (6-metre total walk distance) showed ceiling effects, especially for older, relatively fit people and younger people with amputation. These authors suggested that younger or more active amputees may benefit from performing a task requiring a higher level of skill, such as the L test of functional mobility (Deathe and Miller, 2005). The L test is similar to the TUG test and was designed to reflect a higher level of skill. In addition to sit-to-stand transfers, the participant performs four turns to both the left and right sides and walks a greater distance (20m in L test vs. 6m in the TUG test). Deathe and Miller (2005) reported the ceiling effect of the L test was minimised compared to the TUG test. As the functional demands are greater with the L test, it is possible that such a test would have discriminated between the fallers and non-fallers in the current study.

9.4.2. Psychological data

The first hypothesis also stated the fallers would have lower balance confidence on everyday activities and quality-of-life scores than the non-fallers. The health concepts included in the SF-36 represent health issues that have been shown to be most affected by disease and treatment (Ware and Gandek, 1998). The results have shown that general health (general measures), vitality (general measure) and emotional role (both mental components) were overall health-related factors that distinguished between people based on their falls history. Of the eight SF-36 scales, the fallers scored significantly lower on general health, vitality and emotion compared to the non-fallers. Ware and Konsinski (2001) explained that general health and vitality were considered general measures of health and correlated with both the physical and mental component scales (PCS and MCS), whereas emotional role had stronger loadings and correlated more highly on the mental scale (MCS).

The mean MFES score was not able to distinguish between perceived balance confidence in the fallers vs. non-fallers. This suggested that, like the TUG test, the MFES may show a ceiling effect in community-dwelling fallers, because it measures balance confidence on relatively non-threatening activities. This was in contrast to Hill *et al*. (1998) who reported that the healthy older group scored significantly higher than the falls and balance clinic group, with bathing and reaching scoring the lowest from the indoor-type activities. According to TUG scores, needing help to get in and out of the bath/shower typically reflect low basic mobility skills (Podsiadlo and Richardson, 1991). However, in the current study, the fallers scored significantly lower than the non-fallers on balance confidence for outdoor-type activities (Factor 2) but not indoor-type activities (Factor 1). This supports previous research that people who have fallen and/or have balance deficits find it more difficult to perform outdoor-activities and, according to self-efficacy theory, may avoid those types of activities (Hill *et al*., 1998). This observation was also in agreement with other studies that have investigated activity restriction and avoidance in fallers (Miller *et al*., 2001; Rubenstein, 2006). These findings suggest that low balance confidence on outdoor-type activities could restrict social functioning, because using public transport, crossing roads and using steps in and out of the house are often prerequisite for social activities.

9.4.3. Relationships between functional tasks and psychological components

The second hypothesis stated that poor functional performance would be related to low balance confidence and quality-of-life scores. General health was the only health concept on the SF-36 survey that significantly negatively correlated with functional performance on the TUG and 10-metre walk tests in the fallers and the relationships generally presented moderate to large effect sizes ($r < 0.5$) in this group. General health was measured by asking respondents five questions where they rated their overall health (three questions), tendency to get sick (one question) and health expectations (one question). The results suggested that, 25% of the variance in functional performance could be explained by perceived general health. General health could be a good indicator of function as those who rated their health as being poor (low score) also took the longest to complete the functional tasks (long TUG and 10-metre walk times). These findings support the notion that perceived general health is related to overall performance on common daily tasks, such as walking, transfers in/out of a chair and turning.

Balance confidence scores on outdoor-type activities (Factor 2) were related to 10-metre walk time, but not TUG time in the fallers. This was probably reflected by the nature of the walking tasks. In the home and during indoor-type activities, one rarely walks at a fast speed, whereas outdoor-type activities typically involve walking over longer distances (e.g. walking to the bus, walking around shops). Those individuals who had lower balance confidence on outdoor-activities (poorer scores on MFES

Factor 2) were those who walked slower (10-metre walk time). The correlations between balance confidence and walking speed (10-metre walk time) showed strong relationships and large effect sizes in the fallers and for all participants, while the effect sizes were only small to moderate in the non-fallers. This finding further supports the notion that walking speed is a good overall descriptor of functional mobility and that outdoor-activities are considered higher falls risks. It must be noted that a limitation to the MFES is that it does not account for seasonal changes or different walking surfaces. The participants all completed the MFES during a relatively warm time of year. Had they been asked to rate their balance confidence, especially on Factor 2 activities, during a particularly snowy winter, the results could have been considerably different.

9.4.4. Relationships between balance confidence and quality-of-life

The third aim investigated the relationships between the SF-36 and the MFES and it was hypothesised that low quality-of-life scores would be related to poor balance confidence. Physical functioning and vitality showed the largest effect sizes on balance confidence and quality-of-life parameters for the fallers and all participants together. Vitality reflected a person's overall energy levels and this was the only health attribute on the SF-36 that positively correlated with balance confidence on indoor-activities in the fallers group. This probably referred to the notion that adequate vitality was more important for fallers than non-fallers in performing daily tasks confidently without falling. Being tired or worn out could potentially be a risk factor for falling in individuals who are already at a higher risk for falls. Not surprisingly, physical functioning and general health were positively related with balance confidence on outdoor-type activities in the fallers. Despite the small sample size, many of the correlations between balance confidence and quality-of-life parameters were moderate to strong in the fallers. This finding suggested that the use of self-report questionnaires, such as the MFES and SF-36, could provide a good description of how falling had an impact on a person's perceived health and function. Conversely, the magnitude of the correlations was low for almost all of these variables and no significant relationships were found between balance confidence and quality-of-life in the non-fallers. Individuals who presented a greater risk of falling (e.g. those who were institutionalised or in a wheelchair or amputees with limited functional use of their prosthesis), but were excluded from the study because of the inclusion criteria, would likely have scored even lower than the current fallers on both the MFES and quality-of-life questionnaires.

9.4.5. Comparison of PCS and MCS with norms for limitations in the use of an arm(s) or leg(s)

The data in the current study were compared with normative data for individuals who had limited use in an arm or leg, which included lower-limb amputees. There is no norm-based data of only lower-limb amputees using the SF-36 questionnaire. Although the number of amputee participants in the current study (n=11) was considerably smaller compared with the SF-36 data (n=236), there were some similarities, including age. In general, individuals with manual or locomotive dysfunction scored lower on the physical component scale than the mental scale. This was not surprising since manual or locomotive dysfunction would have detrimental effects on the physical ability to perform many daily activities, such as walking, carrying shopping, vigorous and moderate daily activities. These findings also suggest that individuals with limited function of an arm or leg may adopt coping strategies that benefits their mental health. It was unclear why the participants in the current study scored significantly higher on the MCS compared to the US population data. The MCS score for the US population represented both upper and lower-limb amputees and it is unclear which proportion of the sample set each represented and the level of their amputation (e.g. a transfemoral amputee would have greater physical limitations than a transtibial amputee). Conversely, the participants in the current study were exclusively lower-limb amputees. Purely speculatively, cultural differences could have affected the results. Alternatively, amputee treatment in the UK may address mental health more than in the US. However, more realistically the disproportionate sample sizes were the principal reason for the differences.

9.5. Conclusion

The findings from this study did not support the first hypothesis by demonstrating that there were no differences in functional performance times or balance confidence between community-dwelling fallers and non-fallers. The TUG test, 10m walk test and MFES showed ceiling effects and were not sensitive enough to differentiate between independently living fallers and non-fallers. This may limit the use of these tests with younger traumatic amputees or older, relatively fit adults. Rather, the inability to complete these tests could provide a better indication of functional performance. Perceived quality-of-life was related to performance on daily tasks such as walking, turning and transferring from a seated to a standing position (TUG test), and therefore was considered a good indicator of overall function. The findings from this study also supported the hypothesis that low scores of the functional tests were correlated with lower quality-of-life scores. The SF-36 results revealed that the consequences of falling may have more negative effects on mental, rather than physical health. In the fallers, low balance confidence on outdoor-type activities was highly correlated with poor functional performance. Three of the eight health attributes on the SF-36 were correlated with balance confidence in the fallers, whereas there were no relationships between the two psychological questionnaires in the non-fallers.

CHAPTER TEN – SUMMARY, LIMITATIONS, RECOMMENDATIONS FOR FUTURE RESEARCH AND CLINICAL IMPLICATIONS

10.1. Summary

The age-related changes and effects of falling on the biomechanics of daily activities and quality of life in older people have been receiving widespread attention over the past 20 years (Kerrigan *et al*., 1998; Kerrigan *et al*., 2000; Kerrigan *et al*., 2001; Lee *et al*., 1999; Lee & Chou, 2007; Podisadlo & Richardson, 1991; Powell & Myers, 1995; Tinetti *et al*., 1988). The consequences of having a lower-limb amputation on a person's ability to walk, perform activities of daily living and maintain balance have been well-established from both a biomechanical (Aruin *et al*., 1997; Isakov *et al*., 1992; McFadyen & Winter, 1988; Sanderson & Martin, 1997; Schamlz *et al*., 1997; Winter & Sienko, 1988) and psychosocial perspective (Miller *et al*., 2001; Miller *et al*., 2003; Miller *et al*., 2004). However, this is the first study to have undertaken a holistic approach to amputee function by specifically investigating the biomechanical and psychological differences in amputee fallers vs. non-fallers and making recommendations for amputee rehabilitation.

As the overall aim of this thesis was to assist the clinical recommendations for amputee rehabilitation (described in this Chapter, section 10.4), Chapter 4 investigated current practice by amputee physiotherapists. The results revealed that the monitoring of falls among amputee patients was, at best, inconsistent and, at worst, not part of a patient's regular treatment. Therefore, important inconsistencies and shortcomings were identified. The findings also showed that there was lack of agreement about outcome measures used in amputee rehabilitation and inconsistency in frequency of use.

The biomechanical data from this research have shown that amputee fallers exhibit differences compared to amputee non-fallers during level walking, stair ascent and descent, and when adopting postural strategies in dynamic situations. More complex tasks than level walking, such as stair locomotion, elicited even more biomechanical differences between the two groups. Most of the results demonstrated that amputee fallers walked faster than amputee non-fallers such that their walking velocity closely approached that of the control non-fallers. On the other hand, the amputee non-fallers appeared to adopt a more cautious gait pattern. Chapter 5 showed that while level walking did not elicit kinematic differences at the ankle, knee or hip between the amputee fallers and non-fallers, pelvic motion and kinetic differences were observed. Some kinetic variables (e.g. peak vertical force, load and decay rates) distinguished between the amputee fallers and non-fallers during both level and stair walking. More importantly, these differences occurred during critical moments in the gait cycle. Specifically, kinetic differences between the amputee fallers and non-fallers occurred during the transition from double support to single support on the affected side during level walking, the single support pull-up phase during stair ascent and controlled lowering phase during stair descent. In the control groups, the biomechanical differences during level and stair walking were in agreement with the published literature. Chapters 6 and 7 revealed one important difference within the amputee groups and that was that some amputee participants self-selected a different stepping strategy during both stair ascent and descent. This in itself was probably a very good indicator of these individuals' physical function and balance confidence during more biomechanically challenging tasks.

The current research is the first to measure postural responses to dynamic perturbations in an amputee population using the NeuroCom Equitest. Chapter 8 showed that the NeuroCom may not be a suitable diagnostic tool for lower-limb amputees because of their inability to generate an ankle strategy and active force responses on the affected side. However, a loss of balance as indicated by a stepping strategy, could serve to identify those individuals who are at the greatest risk of falling under dynamic conditions. One of the most interesting and novel findings was that the amputee non-fallers bore more body weight under their intact limb during forward and backward translations. This was considered a successful postural strategy.

With regards to the relationship between falls history and balance confidence, the MFES instrument revealed that previous fallers showed poorer balance confidence on outdoor-type activities. As going outside is an integral part of leading an independent and social lifestyle it is possible that low balance confidence would ultimately lead to activity avoidance and lower life quality. The results from the SF-36 instrument in Chapter 9 suggested that perceived general health was an overall good indicator of physical function.

10.2. Clinical implications

The overall aim of this thesis was to assist the clinical recommendations for improving amputee rehabilitation and physiotherapy treatment based on the biomechanical and psychological findings.

10.2.1. Falls and outcome measures

- Physiotherapists should ask amputee patients if they have fallen on every visit. This information, including circumstances of the fall, should be formally recorded in the patient's records. Physiotherapists should adapt a patient's rehabilitation programme if the patient has fallen. Physiotherapists should be encouraged to share this information with the patient's GP and the prosthetic centre.

- Greater awareness must be raised about the different outcome measures (functional and psychological) that can be incorporated into amputee rehabilitation. Physiotherapists in the UK are actively encouraged to reach a consensus about the use of appropriate outcome measures in amputee rehabilitation.
- Physiotherapists should also agree on how frequently these measures should be administered and how the information would inform treatment goals.

10.2.2. Level walking

- Transtibial amputees should improve eccentric control of the ankle plantarflexor muscles on the intact limb to control the forward progression of the tibia over the ankle joint during terminal stance. This could facilitate a safer contralateral swing phase with the affected leg and foot placement.
- It is not recommended that the intact limb generates large push off forces in pre-swing because the affected limb would need to control larger accelerations during single support on the affected side.
- Knee extensor eccentric strength on the affected limb should be improved to prevent the knee from collapsing during the loading response.
- Transtibial amputees should concentrate on increasing hip flexor eccentric control on both limbs to slow deceleration of thigh extension during late stance.

10.2.3. Stair ascent

- Trantibial amputees should practice prosthetic foot placement in front of the bottom step to determine how much propulsive force needs to be generated by the intact (lead) limb during the floor to stair transition (single support). In this case, the affected limb is typically the trail limb
- Light handrail use is encouraged during stair locomotion, especially during single support on the affected side, because it provides an additional contact point
- Greater pelvic hike on the ipsilateral side may be considered a compensatory strategy for the reduced knee ROM on the affected side and lack of active plantarflexion used to lift the prosthesis into swing (forward continuance phase).
- The transition from floor to stair walking was identified as a critical transition among

those who are at greatest risk of falling. Amputees have absence of somatosensory feedback from the prosthetic foot and ankle. Therefore they should be encouraged to take adequate time, use visual input for foot placement and clearance and use handrails to increase postural stability.

- Large vertical forces on the intact limb may be indicative of a high risk of falling
- Transtibial amputees should be taught to maintain the GRF vector close to the ankle and knee joint centres on the intact limb to increase postural stability when the affected limb is initiating swing. If the affected leg were to trip on the intermediate step above, the intact limb could be more successful at recovering from the trip.
- Amputees should practice the triphasic muscle action at the knee (hamstring concentric contractions, rectus femoris eccentric contractions, hamstring concentric activity) together with hip flexor concentric activity during the swing phase of stair ascent. This would assist sufficient foot clearance and subsequent foot placement.

10.2.4. Stair descent

- Amputees should improve knee extensor eccentric strength on the affected side during the single support, controlled lowering phase.
- By increasing hip flexion at toe off and throughout swing on the affected side, the whole leg is lifted into the air. Amputees appear to 'throw' their prosthesis down onto the next step for stair clearance. With support from handrails, this may be a successful strategy for those who have trouble using the roll-over technique when descending stairs.
- In order to increase hip extension in the pre-swing phase, amputees would benefit from stretching the hip flexor muscles.
- If the prosthetic foot is on the floor and the intact limb is transitioning from stair to floor, amputees should be taught to control the propulsive forces of the intact limb. Without handrails on the floor, the person must rely on knee extensor strength to keep the affected limb from collapsing in single support.
- The intact foot could be encouraged to remain flatter for longer after foot contact to increase the BOS and potentially reduce the demands on the knee extensors of the affected limb during the difficult controlled lowering phase.

10.2.5. Postural control

- Amputees should practice postural control during quiet standing under more challenging conditions, such as with eyes closed (no vision), or when standing on different support surfaces (moving, flexible or uneven surfaces)

- Amputees should avoid increasing their dependence on purely visual input. Interpreting somatosensory feedback from the stump, knee joint and knee musculature should provide important sensory information in the absences of the biological ankle and foot complex and plantarflexor muscles.
- Amputees should practice their ability to move their COP rapidly towards the intact limb in dynamic conditions (e.g. being lightly pushed from behind or the side). The intact limb would have greater strength and flexibility than the affected limb.
- Weight distribution under the affected and intact limbs could be evaluated in a rehabilitation setting with the use of simple scales

10.2.6. Functional performance and balance confidence

- The use of a falls efficacy / balance confidence instrument is strongly recommended in amputee rehabilitation. The MFES presents a quick, cost-effective method of measuring and monitoring an amputee's balance confidence in indoor- and outdoor-type activities. The results can be easily calculated and interpreted.
- Functional performance tests such as the TUG test, L test or the 10m walk test should be used regularly to monitor progress and inform the process of discharge from active rehabilitation. New measures may be needed to monitor function in the more able and active amputees (e.g. traumatic amputees) if a ceiling effect is observed with the existing measures.

10.3. Limitations

Several limitations of the audit that was undertaken in Chapter 4 should be acknowledged. Only the lead physiotherapists in the DSC limb centres in England were asked to complete the audit. It is understood that the lead physiotherapist sets the clinical guidelines within the treatment centre. However, it is not known whether other physiotherapists and/or more junior members of staff implement the same standards of practice as the lead physiotherapist. The DSC limb centre involves a multidisciplinary team, including prosthetists, physiotherapists and occupational therapists. It was not specifically enquired which DSC limb centres employed occupational therapists and whether these clinicians ever used functional outcome measures with the patients. However, assessing outcome measures and falls in lower-limb amputees fits within the clinical guidelines set out by BACPAR and the Chartered Society of Physiotherapy. These guidelines state that physiotherapists should be aware of the activities of other members of the multidisciplinary team. It is believed that physiotherapists would have informed us if other team members were actively involved in a patient's functional rehabilitation.

The results obtained and conclusions drawn from the biomechanics research were inevitably related to the amputee participants that were tested and therefore, certain limitations must be acknowledged. Transtibial amputees were selected based on the inclusion criteria that they could perform the functional tasks independently and without a walking aid (e.g. crutch, stick). Therefore, the very nature of these criteria indicated that the amputee participants could function independently and did not represent the frailest and most vulnerable amputees. As falls incidence has been linked with older adults, and the majority of transtibial amputations occur as a result of dysvascularity in older age, the original inclusion criteria of this research was to only test amputees who had an amputation for vascular reasons. However, the participants that were identified as vascular patients rarely fulfilled the other inclusion criteria (see section 3.2.1). This revealed that many vascular, transtibial amputees that had a prosthesis did not use it on a daily basis, experienced pain in their contralateral limb, could not function without assistive devices and primarily had a prosthesis for cosmetic, not functional, reasons. Therefore, the decision was made to extend participant recruitment to vascular and traumatic amputees who had fallen in the previous 12 months. The sample size in the current research was small and between-subject variability was large, to the point that statistical analysis could not be undertaken in Chapter 7 because some participants adopted alternate movement strategies, thereby reducing group sizes. The participants in the amputee faller vs. non-faller groups were not matched according to their cause of amputation, time since amputation or age. Therefore, caution must be taken when generalising the current findings to the wider amputee and falls populations.

Another limitation was that the results were unable to differentiate between cause and effect of the movement strategies and falls history. By using a cross-sectional design, it remains unclear whether the fallers adapted their movement strategies as a consequence of their fall or whether the movement strategies were the cause of a fall. Many of the biomechanical differences occurred during critical moments in the gait cycle and vulnerability related to muscle weakness or joint stiffness at this time could cause a fall. However, without conducting a longitudinal study, the exact cause and effect relationship remains inconclusive and should be acknowledged when interpreting the data. The functional and psychological correlations that were investigated in Chapter 9 do not infer causality. It is possible that other variables that were not considered affected those relationships.

Although similar to other stair biomechanics studies, the stair walking task was collected using a 3-step staircase with one force plate. As the majority of amputee participants showed a lead limb preference during both stair ascent and descent, then the position of the force plate restricted the kinetic data that were collected to the intact limb only. It is very likely that important biomechanical differences would have been noted on the affected side and that this would have differentiated further between amputee fallers and non-fallers. As the staircase only had 3 steps, only two steps were considered steady-state stair walking. A staircase with at least 5 steps and multiple force plates could have revealed somewhat different results. Staircase dimension should be considered when interpreting the results.

Many studies investigating stair motion using able-bodied individuals do not include a handrail in their staircase construction and thereby exclude the effects of limb unloading on kinetic data. Given the risk that the participants were more likely to fall during stair walking than level walking, and that handrail use may be integral to safe stair negotiation in amputees, the custom-built staircase included a handrail on both sides. This in itself was not a limitation to the study, but rather the effect of handrail use on kinetic data was the limiting factor. Many participants used the handrail and without force transducers, it was unclear how much body weight was actually transmitted through the handrail. Other studies that have investigated handrail use in stair locomotion told participants to use the handrail as a guide only and avoid using the arms to perform work through pulling or pushing their body weight (Reeves *et al*., 2008b). Therefore, it is acknowledged that if participants in this study used the handrails heavily, the kinetic data were 'corrupt' to some level. Conversely, it could be argued that the kinetic profiles represented the participants' true stair negotiation patterns and is therefore more ecologically valid.

Transtibial amputees form a specific population that is inherently highly variable and not all amputees had the same prostheses or ankle-foot components. This would have affected the biomechanical variables, particularly the ankle powers. Furthermore, the inertial properties of the prostheses were not measured or adjusted in the inverse dynamic analysis and therefore, the underlying assumption of the biomechanical model in Visual 3D was that the prosthetic ankle and foot behaved the same as the anatomical ankle and foot. In fact, the rigid model of the foot would be better suited for the prosthetic foot than the intact anatomical foot. Although, this method has been previously reported by Vickers *et al*., (2008) it should be acknowledged as a limitation in the walking data.

The calculation of the equilibrium scores in the SOT on the NeuroCom assumes that an individual's theoretical sway stability limit is 12.5° (see Appendix E for the equilibrium score equation). Equilibrium scores can range from 100 (perfect score indicating no movement) to zero (indicating a fall, loss of balance or stepping strategy). However, the NeuroCom does not take into account an individual's limits of stability on the SOT. If a participant had limits of stability exceeding 12.5°, then they would theoretically obtain a negative score. Therefore, some caution should be taken when interpreting the equilibrium scores if participants have limits of stability that exceed the theoretical maximum discussed by Nashner *et al*. (1989).

10.4. Future recommendations

As there is no consensus among physiotherapists in England on monitoring falls in amputees and the use of outcome measures in amputee rehabilitation, a nationwide audit on falls frequency and outcome measures in DSC across the UK should be conducted. This would facilitate communication within the BACPAR community and the results should be disseminated and discussed at relevant BACPAR meetings. Such research would raise awareness among amputee physiotherapists on existing outcome measures, facilitate better falls monitoring, and encourage greater transparency in patient treatment. Ideally, more falls programmes should be developed and future biomechanical analyses could evaluate their effectiveness.

Future research would benefit from undertaking a detailed biomechanical investigation that relates to falls in vascular and traumatic transtibial amputees separately. This is because the confounding comorbidities that are associated with vascular patients (e.g. polypharmacy, hypertension, obesity, diabetes and other vascular problems in the contralateral intact limb) are different to confounding factors in the traumatic population. Equally, similar studies should be undertaken with transfemoral amputees. As this is the first study to compare amputee fallers vs. non-fallers a cross-sectional design was used. However, to establish a causal relationship, longitudinal studies should be undertaken.

As evidenced by the self-selected strategies during stair ascent and descent, future research should establish the biomechanical differences between alternative locomotor strategies (e.g. step over vs. 'step to' stair walking technique) in both the amputee and elderly populations. The 'step to' pattern may present a safer alternative, with less demand placed on the knee extensor muscles, especially during stair descent. Backwards stair descent has recently been investigated in young healthy adults (Beaulieu *et al*., 2008) and future studies should extend this to the amputee and older populations.

Reduced joint flexibility has previously been linked with falls in able-bodied individuals and many of the hypotheses in this thesis related to joint mobility. It is acknowledged that joint ROM was not measured passively or actively, and was only inferred from the joint kinematic data during level and stair walking. Therefore, future studies should measure lower-limb joint flexibility or, in the case of transtibial amputees, on both the affected and intact limbs. Prosthetic componentry and stiffness were not controlled for in this study and it is possible that prosthetic ankle-foot settings may contribute to identifying fallers and non-fallers. Future research may explore the relationship between prosthetic functionality and alignment and falls risk in lower-limb amputees.

The current study investigated the biomechanics of level and stair walking and postural strategies. However, there are many other activities of daily living that are more mechanically challenging tasks than level walking, such as stepping over obstacles, recovering from a slip, turning, and transferring from a seated to a standing position and vice versa. While these activities have been investigated using amputee and control participants, no studies to date have specifically compared amputee fallers vs. non-fallers.

Other studies that test whether postural strategies are learned in response to having fallen, or whether effective strategies could be learned as preventative measures through practice and exercises could be undertaken within a rehabilitation setting. Longitudinal studies that measure and record falls incidence and track rehabilitation practice would be fundamental at evaluating treatment programmes and the findings from this thesis. It is important that future studies undertaken in the area of amputee rehabilitation, particularly those that relate to falls, build upon those made in this thesis. This would be especially useful if researchers from different backgrounds engaged with the BACPAR community.

10.5. Conclusions

This thesis contributes to an important and novel body of knowledge that is fundamentally multidisciplinary, by focusing on key research areas in health such as falls and rehabilitation. From the outset of this project in 2005, there was no scientific literature that specifically investigated the relationship between falls history, biomechanics and psychological well-being in an amputee population. Presently, the only published literature on this subject has stemmed from this thesis.

This thesis has demonstrated that amputee and able-bodied fallers and non-fallers exhibit different biomechanical and psychological characteristics during daily activities. Furthermore, this thesis has identified shortcomings in current amputee rehabilitation, proposed new guidelines for rehabilitation and suggested that the current guidelines on falls prevention and treatment in lower-limb amputees should be seriously investigated.

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APPENDIX A- PATIENT/PARTICIPANT INFORMATION LEAFLET

How do transtibial amputees perform activities of daily living? A biomechanical investigation

You are invited to take part in a research study. In order to make an informed decision, you are asked to read the following documents. Please do not hesitate to ask the Investigator for any clarification or further information at any point in time. Feel free to take the time to discuss it with your family, friends and GP. You will be given as much time as you want to make a decision.

What is the purpose of this study?

Lower-limb amputees may experience a fall while wearing their prosthesis and, as a result, may develop a fear of falling. Falls are associated with decreased independence and mobility and a lower quality of life. Virtually no studies have investigated the internal factors that may distinguish how transtibial amputees perform activities of daily living. Therefore, the goal of this project is to examine transtibial amputees performing typical daily tasks, such as walking, performing a 180° turn, rising from a seated position, returning to a seated position and stair climbing.

Why have I been invited?

You have been invited to participate in this study because you received your prosthesis within the past 6 to 12 months.

What will happen if I decide to take part?

If you decide to take part in this study, you will be invited to the Human Performance Laboratory in the Department of Sport, Health and Exercise Science at the University of Hull. You will be asked to bring a pair of shorts, comfortable shoes (no high heels!) and t-shirt. Shorts will be provided if you do not have a pair. If you do not have your own transportation, it will be arranged for you by the University.

When you arrive, you will be asked to change into shorts and a t-shirt and reflective markers and surface electrodes will be placed onto the surface of your skin using sticky tape. Reflective markers are small and spherical and placed onto your skin with tape; they are used to record the movement of your limbs. EMG electrodes are used to monitor the electrical activity your muscles produce when they contract. The electrodes will not cause any unpleasant sensations in your muscles. There are generally no adverse reactions or discomfort involved with using sticky tape.

You will then be asked to perform several activities of daily living: walking along a walkway in the laboratory, performing a 180° turn; rising from a seated position, walking three metres and turning around to sit down again and climbing up and down a two-step staircase. Special digital motion cameras will be used to track the movements of your limbs. These cameras only see the reflective markers and therefore, your anonymity is completely guaranteed. The floor surface in the laboratory is non-slippery and the custom-built staircase has handrails on both sides and a non-slippery surface. Finally, you will complete several short tests on a sophisticated measuring platform that measures your balance control under test conditions designed to reflect the challenges of daily life.

Additionally, you will be requested to complete questionnaires asking you to rate your perceived fear of falling, balance confidence, ability to use your prosthesis and quality of life.

What do I have to do?

To take part in this study, you will need to visit the Human Performance Laboratory at the University of Hull on one occasion for approximately 2-3 hours. You will have reflective markers and surface electrodes placed onto the surface of your skin. You will then proceed through a series of short tests designed according to you typical daily activities, such as walking short distances, rising from a chair, and using a two-step staircase. You may always rest in between, as you feel necessary. Finally, you will be fitted with a security harness around your torso and waist that is attached to a safety bar on the Neurocom Equitest® that measures your balance control under moving test conditions. By wearing a harness, the performance environment in which you will undertake the testing protocol is much safer.

In between testing, you will be asked to complete questionnaires with the researcher that rate your perceived fear of falling, balance confidence, ability to use your prosthesis and quality of life.

Do I have to take part?

Participation in this study is entirely voluntary.

You may refuse to participate or withdraw from the study at any time. You do not need to tell the researchers why you do not want to take part. If you choose to withdraw or not to participate, your decision will in no way affect your future treatment. It may be that the investigator or sponsor of the study consider that it is in your interests to withdraw you or stop the study altogether.

Are there any risks involved?

Although rare, one possible side effect would be a skin reaction to the adhesive tape used to affix the surface electrodes and reflective markers onto your skin. Your skin will be checked upon removal and, if any reaction occurred, appropriate treatment would be recommended.

Appropriate safety measures will be taken at all times, You may feel that you are going to fall on the Equitest®...but the safety harness will hold you up!!! The Equitest® was designed for use in many types of medical disciplines including geriatrics and physical rehabilitation and used in the management strategies of a wide range of physical disorders.

Are there any costs involved?

No

Confidentiality

Any information regarding your identification resulting from this study will be kept strictly confidential. All documents will be identified by code number and kept in a locked and secured filing cabinet. You will be assigned a number so that no data can be linked to you directly.

Data obtained from your visit to the Human Performance Laboratory will be recorded on a Case Record Form (CRF). Your information will be stored on a computer in the laboratory and will require an access password. The video cameras will not identify you in any way, your anonymity is completely guaranteed.

All information in your notes and CRF will be treated in strict confidence. A copy of this Informed Consent Form will be kept with the CRF and you will be given a copy.

The information from this study will be retained by the University of Hull until the data are analysed and for 2 years after the end of the study.

By signing the attached consent form, you give permission for the above to occur.

If you agree to participate in this study, your General Practitioner will be informed, unless you state otherwise.

Your rights

Your participation in this study is entirely voluntary and refusal will not affect any other medical treatment. You may, without reason, refuse to take part in the trial, and this will not, in any way, affect your continuing treatment.

Who is organising the research?

The study is being organised by the Department of Sport, Health and Exercise Science at the University of Hull.

Trial-related injury

In the unlikely event that you suffer from injury or illness as a result of participation in this study, indemnity will be provided by the Hull and East Yorkshire hospitals NHS Trust. Compensation will be by the usual NHS procedures.

If you were to suffer from illness or injury during the study, or have any questions about the research study, please contact Natalie Vanicek at the Department of Sport, Health and Exercise Science, University of Hull on 01482 466212.

Thank you.

APPENDIX B - INFORMED CONSENT

How do transtibial amputees perform activities of daily living? A biomechanical investigation

NAME OF LOCAL LEAD RESEARCHER: Natalie Vanicek

PARTICIPANT ID: ______________________

Please initial box

1 I confirm that I have read and understand the information sheet dated ……….. (version ……….) for the above study and have had the opportunity to ask questions.

2 I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected.

3 I understand that sections of any of my medical notes relating to my taking part in the study may be looked at by responsible individuals from <*Company name/sponsor*> or from the appropriate regulatory authority(ies). I give permission for these individuals to have access to my records.

4 I agree to take part in the above study. ☐

Name of Participant (BLOCK CAPITALS)	Date	Signature
Name of Person taking consent	Date	Signature
Researcher/witness	Date	Signature

1 copy for participant; 1 for researcher; 1 to be kept with Local NHS Trust notes

APPENDIX C- STAIR ASCENT – ‘STEP TO’ GROUP

Appendix C.1. ‘Step to’ group (intact limb - lead limb floor to first step)

Table AC.1. Amputee ‘step to’ group mean (SD) foot position and peak kinematic values (°) in the sagittal plane during swing from floor to the first step.

	AMP_F 5		AMP_F 6	
	Affected	**Intact**	**Affected**	**Intact**
Maximum vertical toe clearance (cm)	12.4	13.5	6.6	9.0
Hip flexion swing (°)	51.9 (5.4)	70.0 (1.6)	57.4 (2.4)	70.8 (1.5)
Knee flexion swing (°)	67.3 (4.3)	88.5 (2.2)	62.3 (3.0)	84.7 (4.4)
Ankle plantarflexion swing (°)	4.8 (0.7)	5.2 (1.6)	5.2 (0.8)	-3.3 (4.4)
Ankle dorsiflexion swing (°)	6.5 (1.2)	16.4 (1.2)	6.2 (0.8)	9.4 (1.0)

Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion -ve

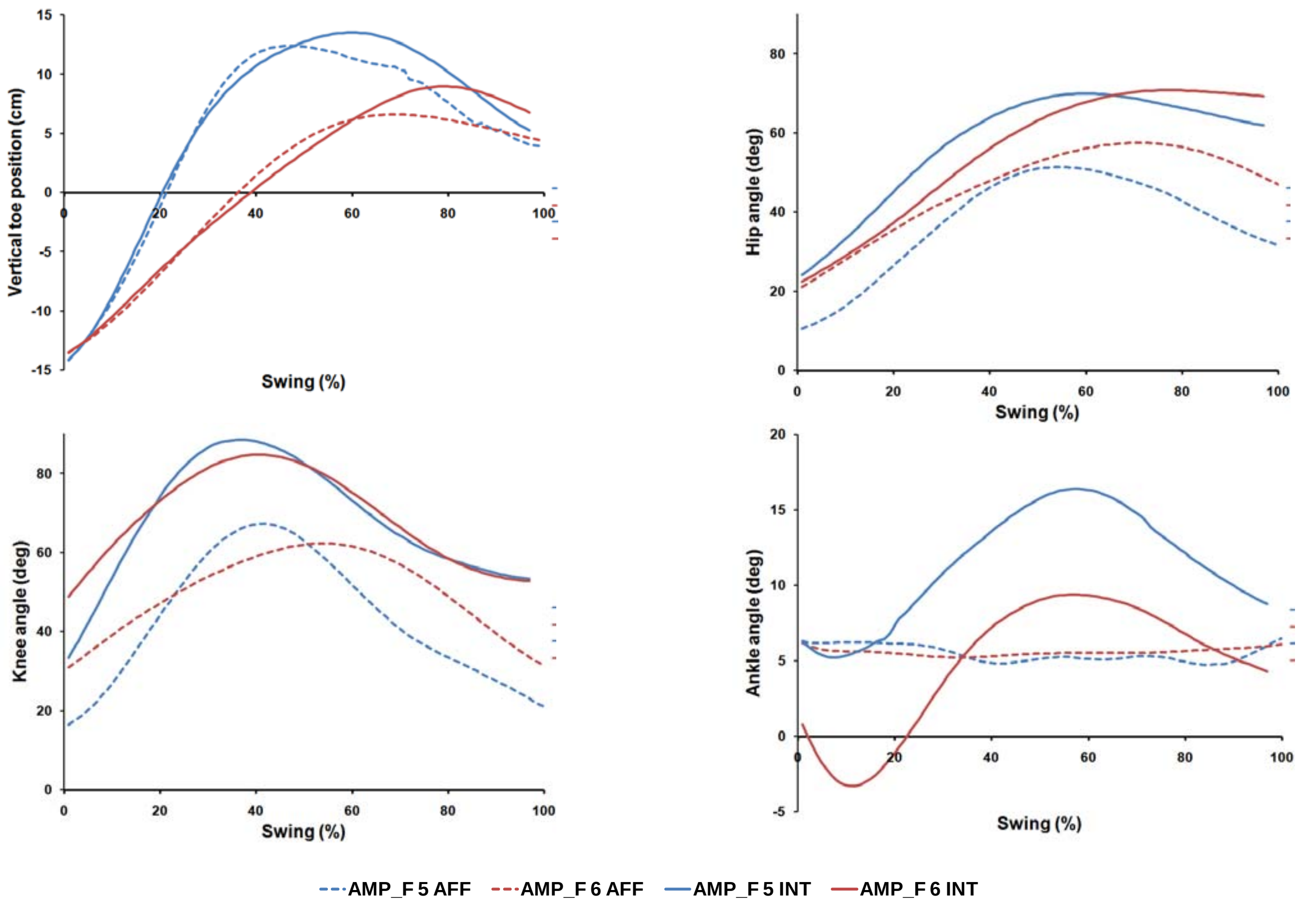

Figure AC.1. Joint flexion angles and vertical position of the toe during swing ascent from floor to the first step in the amputee 'step to' group. Averaged individual data are presented. Participant 14 - affected leg (dashed blue line); Participant 15 – affected leg (dashed red line); Participant 14 – intact leg (solid blue line); Participant 15 – intact leg (solid red line).The swing cycle is initiated with toe off and terminated with foot contact.

Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion –ve

Appendix C.2. 'Step to' group (intact limb - lead limb first to second step)

Table AC.2. Amputee 'step to' group mean (SD) temporal-spatial and peak kinematic values (°) during stair ascent from the first to the second step.

	AMP_F 5		AMP_F 6	
	Affected	**Intact**	**Affected**	**Intact**
Speed (m/s)	0.19		0.32	
Stance phase (%)	65 (8)	77 (10)	60 (4)	74 (8)
Hip abduction stance (°)	-19.5 (1.7)	-6.5 (1.8)	-4.6 (1.1)	-5.4 (1.6)
Hip adduction swing (°)	-1.0 (1.7)	7.8 (2.1)	11.4 (0.7)	1.5 (0.5)
Hip ROM frontal (°)	18.5	14.3	16.0	6.9
Hip extension stance (°)	7.6 (2.2)	19.4 (3.9)	14.1 (1.9)	24.5 (2.4)
Hip flexion swing (°)	49.4 (7.1)	70.6 (2.4)	55.6 (1.7)	73.1 (1.7)
Hip ROM sagittal (°)	41.8	51.1	41.5	48.5
Knee flexion loading (°)	23.3 (4.6)	55.6 (1.9)	29.3 (2.5)	55.1 (4.4)
Knee flexion swing (°)	64.3 (2.4)	90.2 (2.8)	60.5 (1.7)	94.5 (1.7)
Knee ROM (°)	57.9	82.1	47.3	88.0
Ankle dorsiflexion stance (°)	11.2 (1.0)	14.1 (2.6)	15.7 (1.4)	9.0 (4.5)
Ankle plantarflexion swing (°)	4.3 (0.5)	2.7 (1.8)	5.0 (0.9)	1.1 (1.2)
Ankle dorsiflexion swing (°)	13.3 (0.8)	15.6 (0.5)	15.8 (1.4)	10.7 (0.6)
Ankle ROM (°)	9.0	13.0	10.9	9.6
Pelvic tilt stance (°)	22.9 (2.1)	25.5 (1.4)	24.2 (2.2)	27.4 (2.7)
Pelvic tilt swing (°)	24.2 (4.7)	22.9 (2.3)	27.2 (1.6)	25.2 (2.8)
Pelvic obliquity stance (°)	-14.6 (1.9)	-5.7 (2.5)	-8.1 (1.1)	-1.4 (1.3)
Pelvic obliquity swing (°)	5.3 (1.5)	14.9 (1.6)	2.6 (0.8)	8.0 (1.1)

Hip extension –ve (or minimum hip flexion +ve)†; Hip flexion +ve; Hip abduction -ve; Hip adduction +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion -ve; Anterior pelvic tilt +ve; Pelvic obliquity up (pelvic hike) +ve; Pelvic obliquity down (pelvic drop) –ve

† It is noteworthy that the hip usually reached full extension during level gait and therefore was associated with a negative value. However, in stair ascent, the hip did not achieve full extension and therefore the positive value at this phase in the gait cycle represented the minimum amount of hip flexion

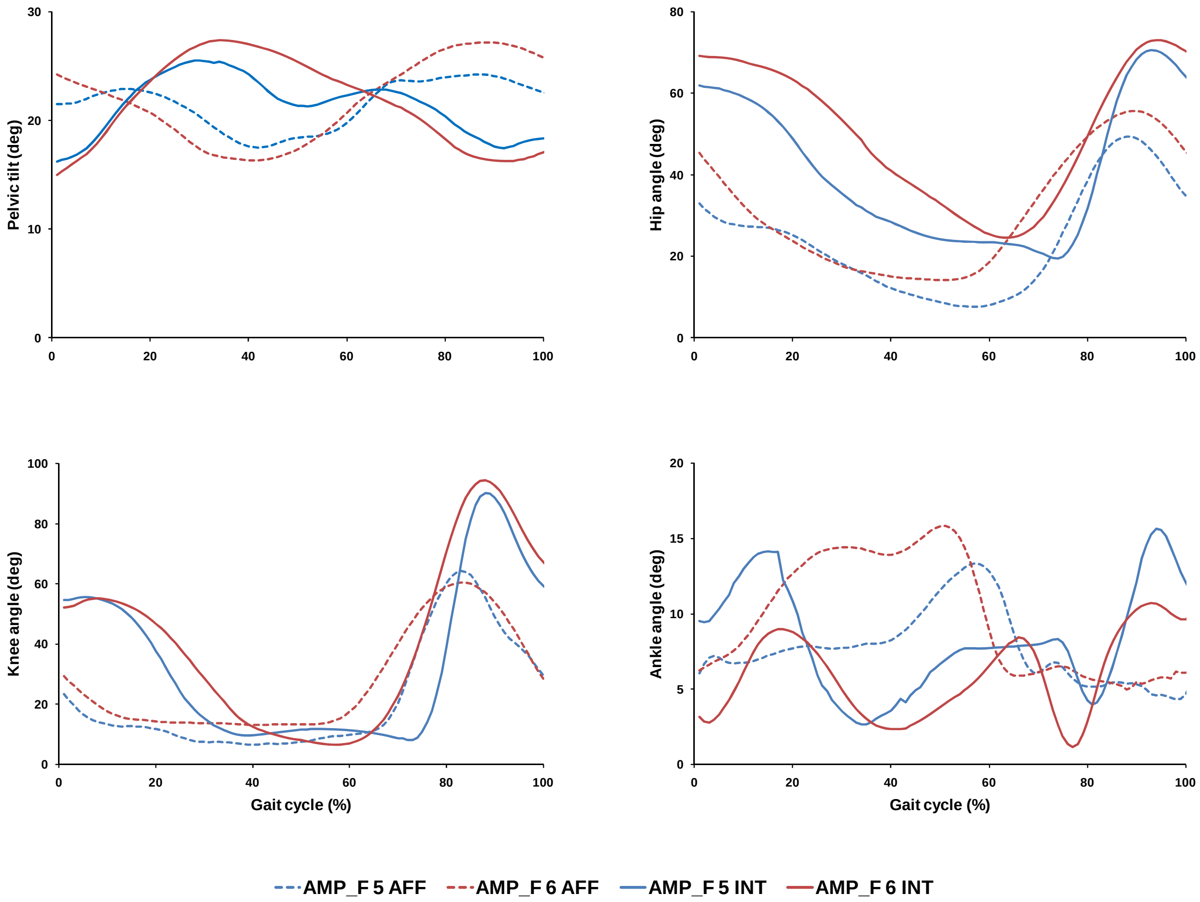

Figure AC.2. Sagittal plane angles during stair ascent from floor to the first step in the amputee ‘step to’ group. Averaged individual data are presented. AMP_F 5 - affected leg (dashed blue line); AMP_F 6 – affected leg (dashed red line); AMP_F 5 – intact leg (solid blue line); AMP_F 6 – intact leg (solid red line).The gait cycle is initiated and terminated with foot contact.

Anterior pelvic tilt +ve; Hip flexion +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion –ve

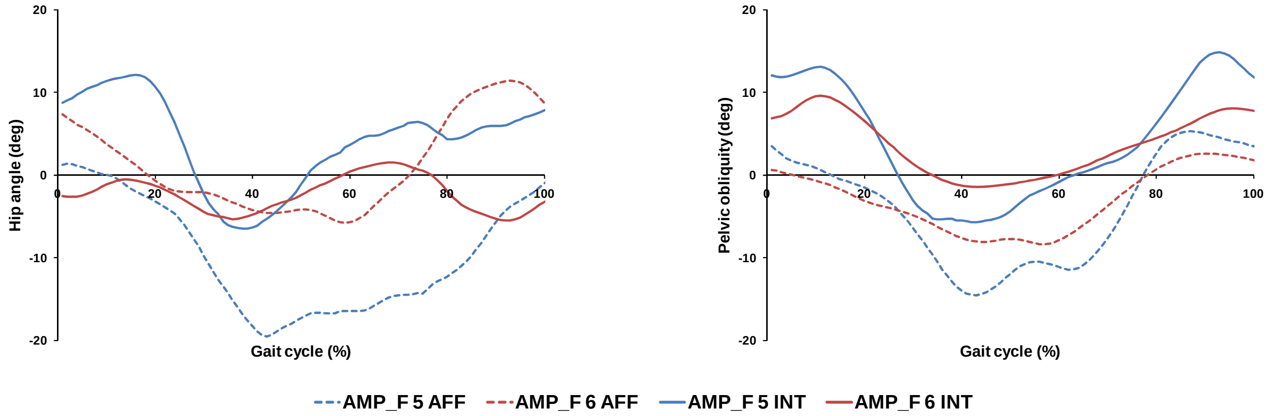

Figure AC.3. Frontal plane hip and pelvic kinematics during stair ascent in the amputee 'step to' group. Averaged individual data are presented. AMP_F 5 - affected leg (dashed blue line); AMP_F 6 – affected leg (dashed red line); AMP_F 5 – intact leg (solid blue line); AMP_F 6 – intact leg (solid red line).The gait cycle is initiated and terminated with foot contact.

Hip adduction +ve; Pelvic hike/obliquity up +ve

APPENDIX D - STAIR DESCENT – 'STEP TO' GROUP

Appendix D.1. 'Step to' group - (first to second step)

Table AD.1. Amputee 'step to' group mean (SD) temporal-spatial and peak kinematic values (°) during stair descent from the first to the second step

	Amputee Non-Faller (n=2)		Amputee Faller (n=2)	
	Affected	**Intact**	**Affected**	**Intact**
Speed (m/s)	0.34 (0.10)		0.24 (0.08)	
Stance phase (%)	44 (2)	76 (1)	59 (8)	81 (2)
Hip angle toe off (°)	44.8 (15.3)	42.3 (3.5)	44.7 (0.3)	46.3 (0.7)
Hip flexion swing (°)	53.4 (18.7)	48.3 (2.2)	51.2 (1.2)	52.7 (5.0)
Hip angle foot contact (°)	33.2 (12.8)	31.3 (4.7)	30.0 (5.5)	33.6 (3.9)
Hip extension stance (°)	21.9 (14.1)	17.2 (6.8)	19.6 (1.7)	27.5 (1.7)
Hip ROM (°)	31.5 (4.6)	31.1 (4.7)	31.7 (3.0)	25.2 (3.3)
Knee angle toe off (°)	74.6 (42.8)	88.6 (5.3)	43.5 (7.5)	78.9 (1.6)
Knee flexion swing (°)	77.7 (44.5)	90.1 (4.0)	48.7 (5.7)	79.5 (0.8)
Knee angle foot contact (°)	21.9 (19.7)	25.6 (5.0)	16.6 (6.6)	17.9 (2.8)
Knee ROM (°)	57.7 (26.3)	72.3 (3.8)	39.2 (3.2)	68.3 (2.5)
Ankle angle toe off (°)	7.4 (7.7)	20.1 (0.0)	5.6 (3.0)	7.5 (11.5)
Ankle plantarflexion swing (°)	4.2 (5.2)	6.7 (1.2)	4.0 (1.1)	0.7 (6.5)
Ankle angle foot contact (°)	6.5 (4.6)	7.1 (0.6)	4.8 (0.7)	4.1 (1.7)
Ankledorsiflexion stance (°)	15.9 (6.6)	40.4 (3.0)	10.8 (2.3)	30.8 (14.3)
Ankle ROM (°)	11.7 (1.4)	33.7 (4.2)	6.8 (1.2)	30.1 (7.9)
Pelvic tilt toe off (°)	18.6 (2.9)	19.2 (0.4)	22.9 (0.6)	19.5 (0.7)
Pelvic tilt swing (°)	20.4 (0.3)	20.2 (0.2)	22.9 (0.6)	21.6 (0.2)
Pelvic tilt foot contact (°)	19.6 (0.8)	18.0 (2.3)	19.1 (2.3)	20.9 (1.2)
Pelvic tilt stance (°)	20.0 (0.5)	13.3 (1.7)	23.3 (0.6)	14.6 (0.8)

Hip extension -ve (or minimum hip flexion +ve)†; Hip flexion +ve; Hip abduction -ve; Hip adduction +ve; Knee flexion +ve; Ankle dorsiflexion +ve; Ankle plantarflexion -ve; Anterior pelvic tilt +ve; Pelvic obliquity up (pelvic hike) +ve; Pelvic obliquity down (pelvic drop) –ve

† It is noteworthy that the hip usually reached full extension during level gait and therefore was associated with a negative value. However, in stair ascent, the hip did not achieve full extension and therefore the positive value at this phase in the gait cycle represented the minimum amount of hip flexion

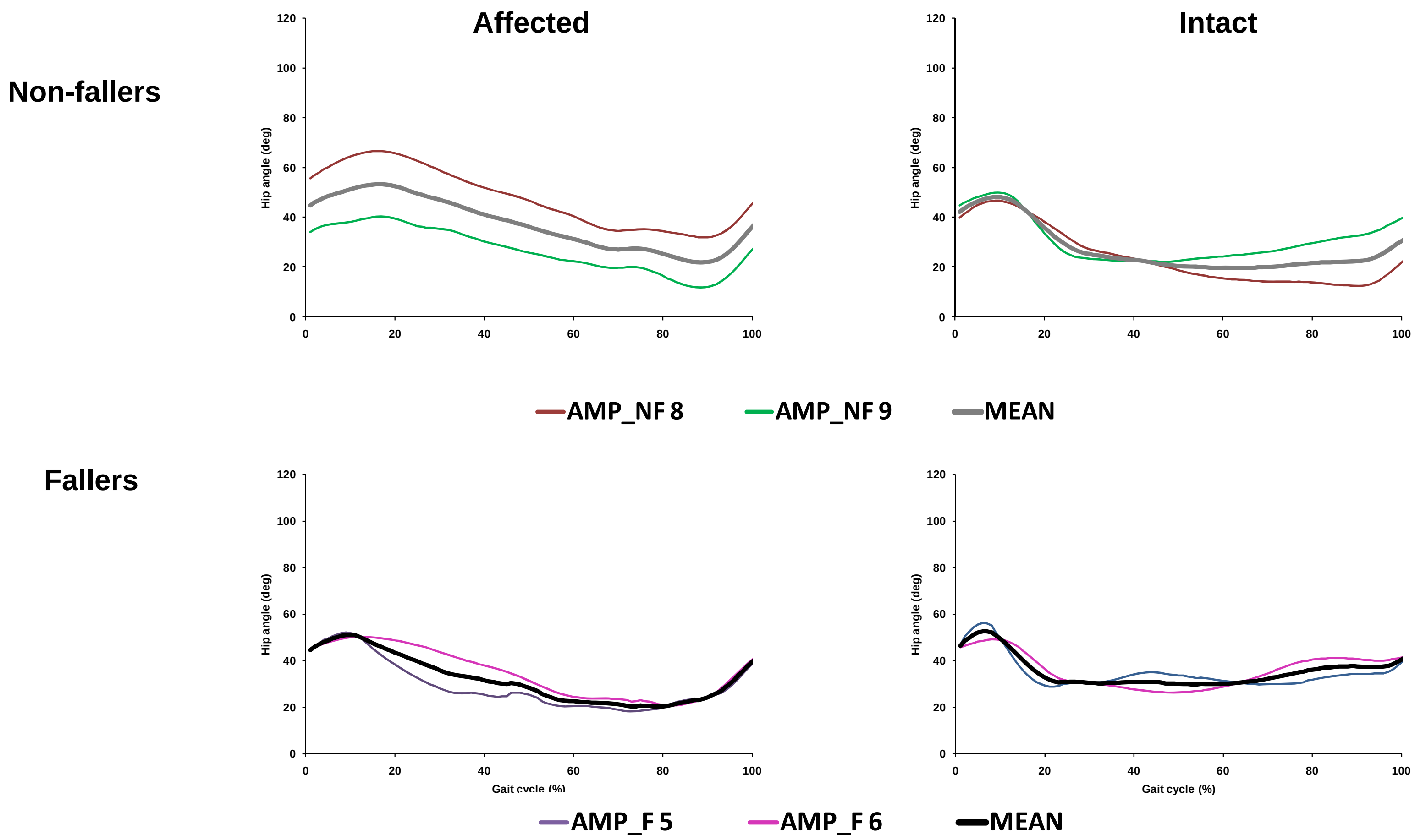

Figure AD.1. Hip sagittal plane flexion angles during stair descent in the amputee 'step to' group for the non-fallers and fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off.

Hip flexion +ve

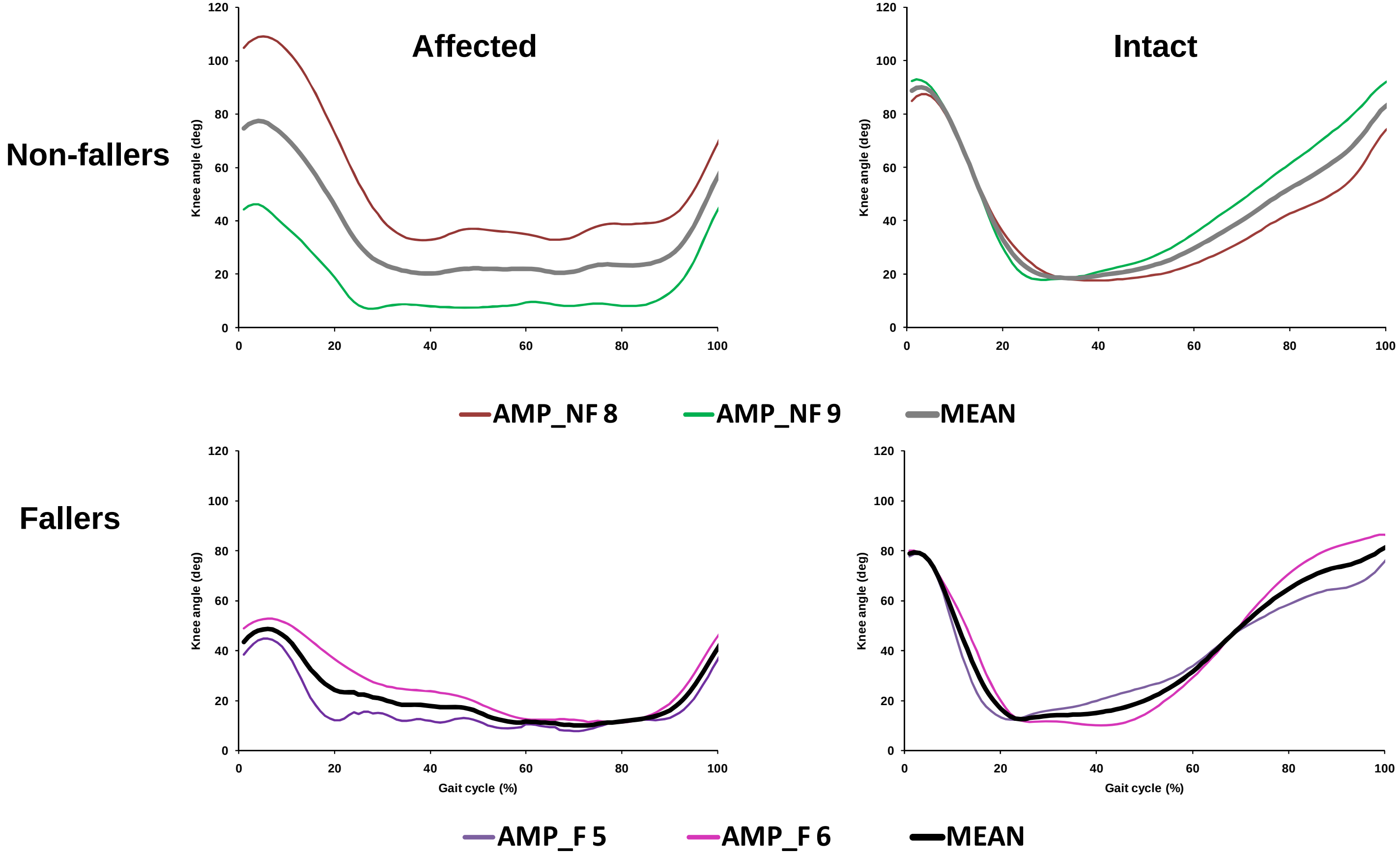

Figure AD.2. Knee sagittal plane flexion angles during stair descent in the amputee ‘step to’ group for the non-fallers and fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off.

Knee flexion +ve

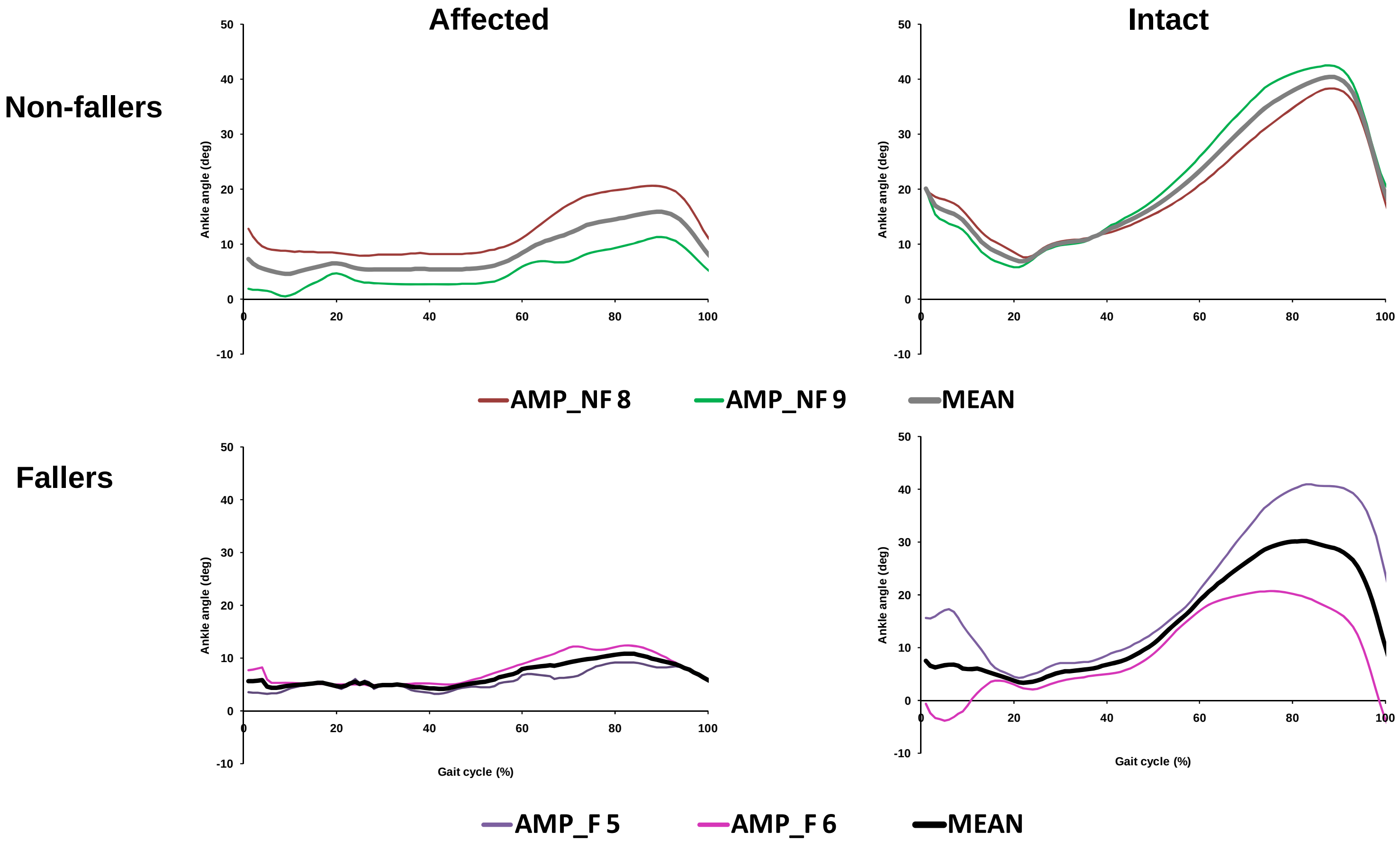

Figure AD.3. Ankle sagittal plane flexion angles during stair descent in the amputee 'step to' group for the non-fallers and fallers. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off.

Ankle dorsiflexion +ve

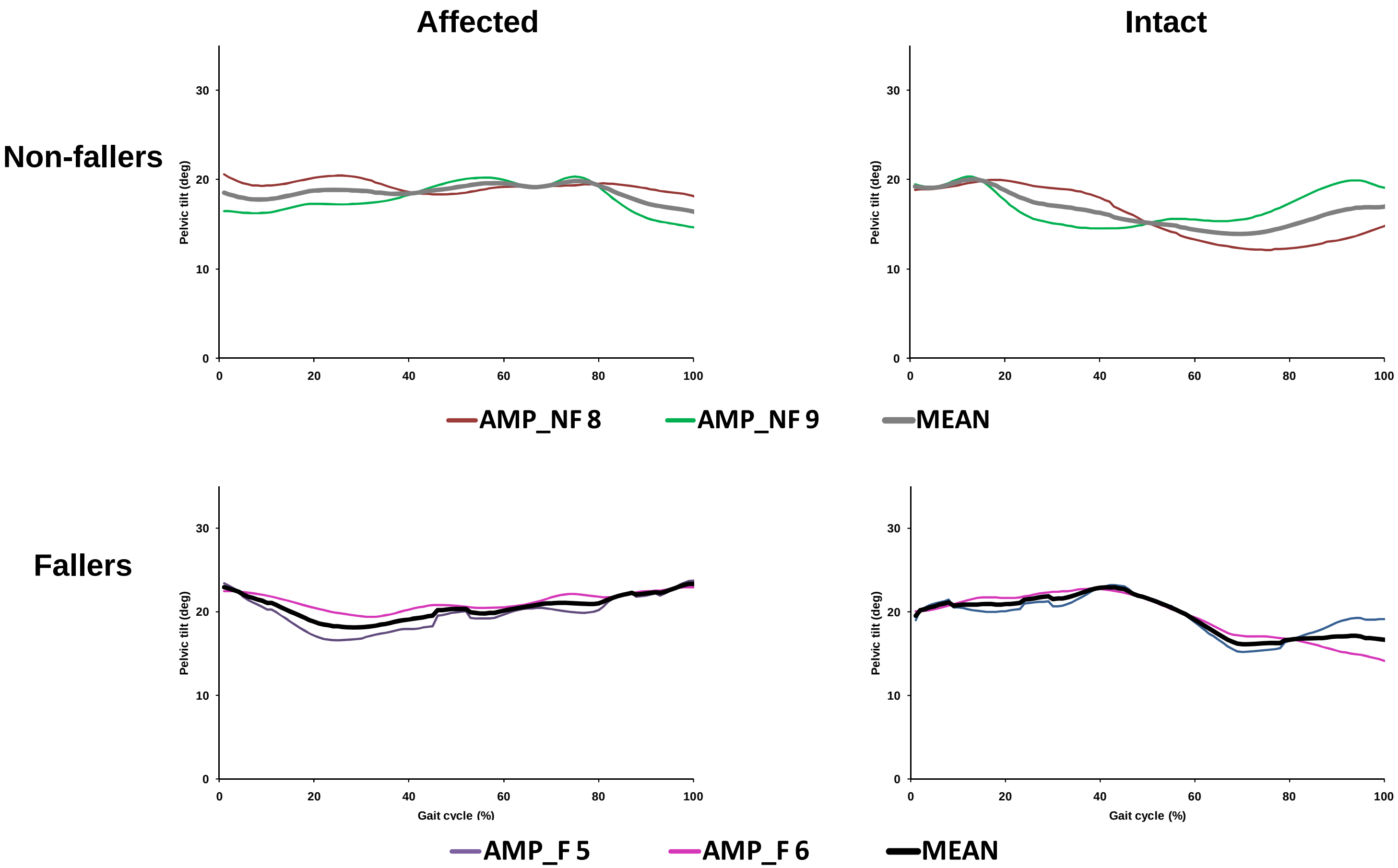

Figure AD.4. Pelvic tilt during stair descent in the amputee ‘step to’ groups. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off

Anterior pelvic tilt +ve

Table AD.2. Amputee 'step to' groups mean (SD) temporal-spatial and peak kinematic values (frontal plane) during stair descent (°).

	Amputee Non-Faller (n=2)		Amputee Faller (n=2)	
	Affected	**Intact**	**Affected**	**Intact**
Hip angle toe off (°)	4.3 (3.6)	1.7 (1.6)	2.2 (6.7)	2.3 (5.9)
Hip abduction swing (°)	-8.3 (3.7)	-1.3 (0.7)	-8.8 (4.3)	1.0 (4.8)
Hip angle foot contact (°)	-8.3 (3.7)	3.3 (4.2)	-8.2 (4.8)	4.7 (2.5)
Hip adduction stance (°)	0.1 (1.2)	8.8 (4.0)	4.6 (6.0)	5.7 (3.9)
Hip frontal ROM (°)	13.5 (0.9)	10.1 (3.3)	14.0 (0.7)	9.9 (4.7)
Pelvic obliquity toe off (°)	2.9 (5.1)	7.3 (0.6)	2.8 (0.8)	5.8 (3.6)
Pelvic obliquity foot contact (°)	-8.5 (1.8)	4.5 (1.9)	-4.8 (1.7)	2.0 (1.2)
Pelvic obliquity down stance (°)	-8.6 (1.8)	1.7 (0.7)	-5.4 (2.4)	-4.3 (0.4)
Pelvic obliquity up stance (°)	-1.6 (0.6)	10.0 (0.9)	4.1 (0.4)	5.1 (2.0)
Pelvic frontal ROM (°)	11.6 (3.3)	8.3 (1.6)	9.6 (2.2)	11.0 (3.8)

Hip adduction +ve; Hip abduction –ve; Pelvic obliquity up (pelvic hike) +ve; Pelvic obliquity down (pelvic drop) –ve

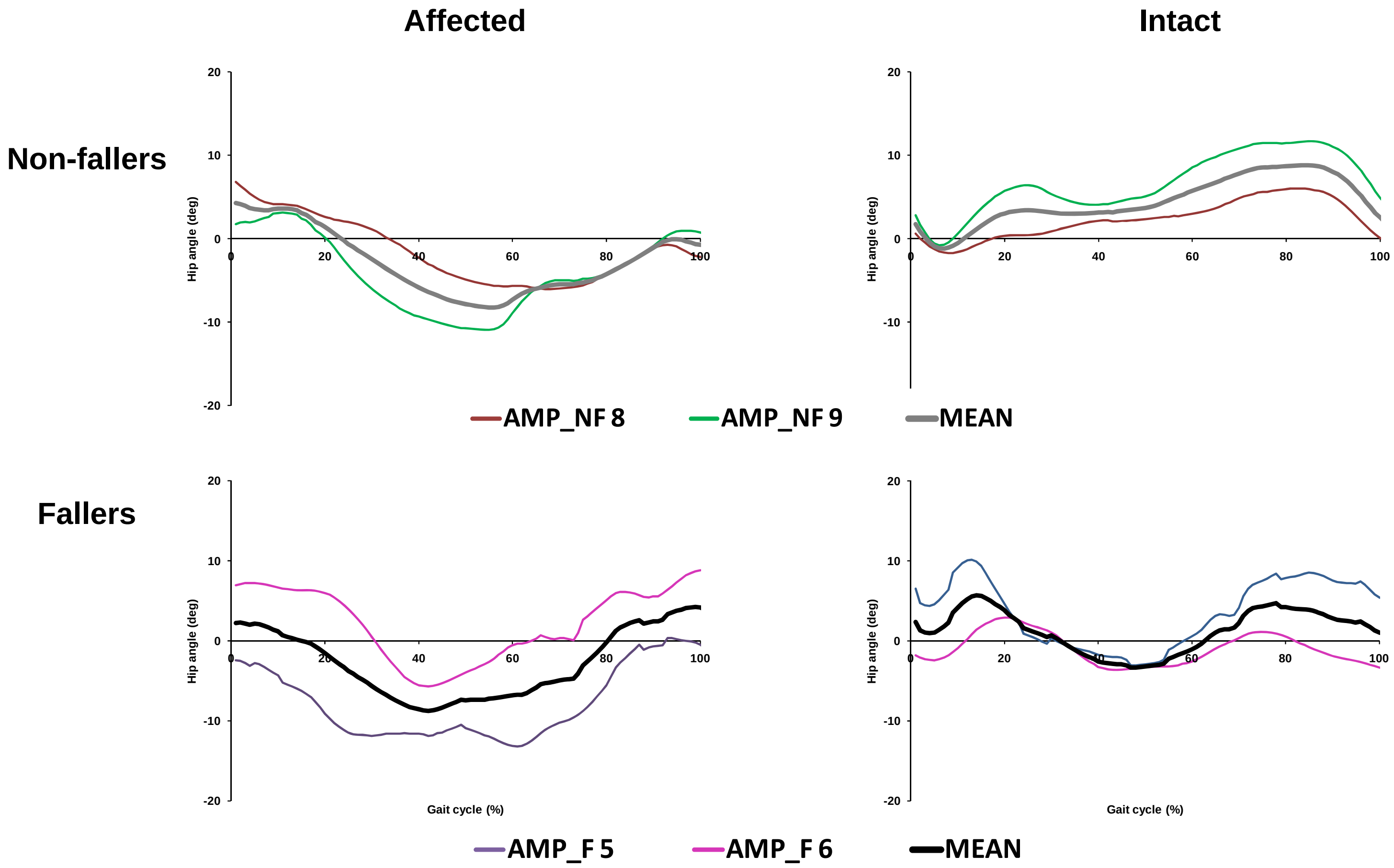

Figure AD.5. Hip adduction during stair descent in the amputee 'step to' group s. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off

Hip adduction +ve

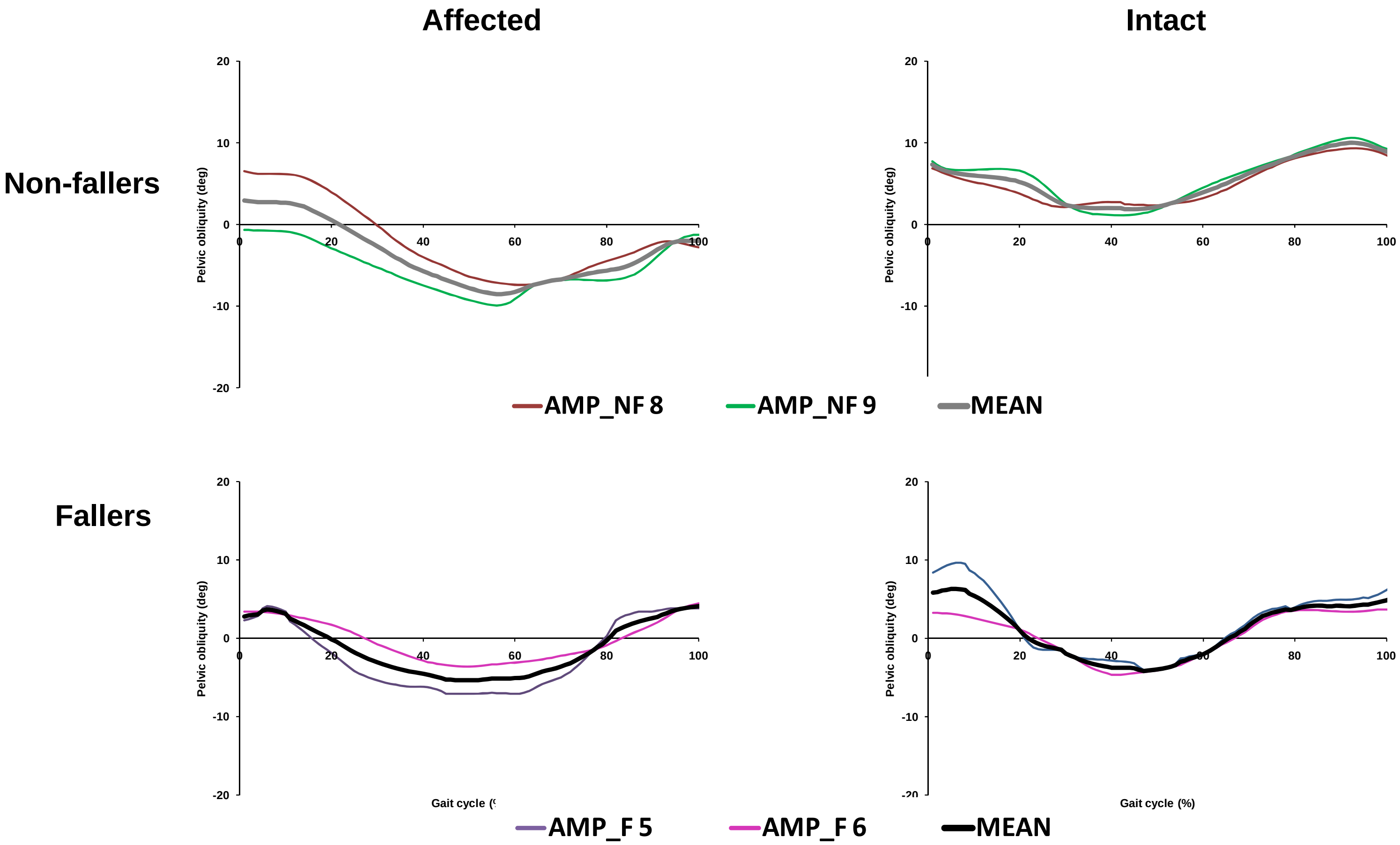

Figure AD.6. Pelvic obliquity during stair descent in the amputee ‘step to’ group s. Individual participant data are presented with the averaged data according to limb: Non-fallers (solid grey line); Fallers (solid black line). The gait cycle is initiated and terminated with toe off

Pelvic hike/up +ve

APPENDIX E – NEUROCOM FORMULAE

NeuroCom software Equilibrium score and Weight symmetry formulae:

Equilibirum score = $\frac{12.5° - (\theta\ max - \theta\ min)}{12.5°} * 100$

where 12.5° is the normal limit of the anterior-posterior sway angle range

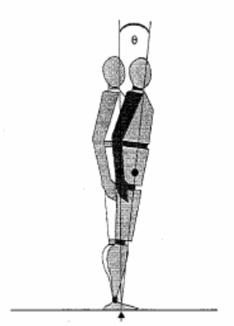

Figure PO-3 Center of Gravity Sway Angle θ

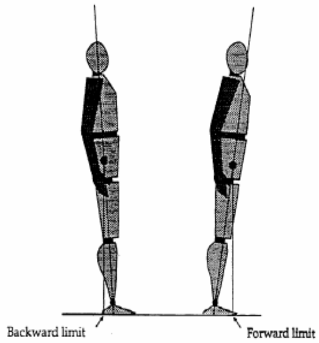

Figure PO-4 Limits of stability

(Neurocom International Inc., 2004)

Strategy score = $[1 - \frac{SH_{max} - SH_{min}}{25}] * 100$

Where 25 lbs is the difference between maximum and minimum shear force generated by able-bodied individuals who only used the hip strategy to maintain balance on a narrow beam (Nashner, unpublished data; cited in NeuroCom International Inc., 2004)

Weight symmetry = $\frac{RF + RR}{LF + LR + RF + RR} * 200$

Abbreviations: SH: Shear load cell; RF: Right front load cell; RR: Right rear load cell; LF: Left front load cell; LR: Left rear load cell

APPENDIX F - MODIFIED FALLS EFFICACY SCALE (Hill *et al*., 1996) AND SHORT-FORM (SF-36) (Ware and Sherbourne, 1992)

Instructions

As you read each statement, remember there is no right or wrong answer. Just think about how confident you are to execute each activity without falling. Do this by making a mark through the line anywhere along the line from 'not-confident / not sure at all' (score of 0) to 'completely confident / completely sure' (score of 10).

How confident/sure are you that you do each of the activities without falling:

(1) Get dressed and undressed

Not Confident At All	Fairly Confident	Completely Confident

(2) Prepare a simple meal

Not Confident At All	Fairly Confident	Completely Confident

(3) Take a bath or a shower

Not Confident At All	Fairly Confident	Completely Confident

(4) Get in/out of a chair

Not Confident At All	Fairly Confident	Completely Confident

(5) Get in/out of bed

Not Confident At All	Fairly Confident	Completely Confident

(6) Answer the door or the telephone

Not Confident At All	Fairly Confident	Completely Confident

(7) Walk around the inside of your house

Not Confident At All	Fairly Confident	Completely Confident

(8) Reach into cabinets or closet

Not Confident At All	Fairly Confident	Completely Confident

(9) Light housekeeping

Not Confident At All	Fairly Confident	Completely Confident

(10) Simple shopping

Not Confident	Fairly	Completely
At All	Confident	Confident

(11) Using public transport

Not Confident	Fairly	Completely
At All	Confident	Confident

(12) Crossing roads

Not Confident	Fairly	Completely
At All	Confident	Confident

(13) Light gardening or hanging out the washing (rate most commonly performed of these activities)

Not Confident	Fairly	Completely
At All	Confident	Confident

(14) Using front or rear steps at home

Not Confident	Fairly	Completely
At All	Confident	Confident

APPENDIX G - STANDARDISATION OF PCS AND MCS SCALES FROM THE SF-36 HEALTH SURVEY (derived from Ware Jr and Kosinski, 2001)

Step1: Formulae for z-score standardisation of SF-36 scales

PF_Z = (PF -82.96845) / 23.83795

RP_Z = (RP – 77.93107) / 35.34865

BP_Z = (BP – 70.22865) / 23.35310

GH_Z = (GH – 70.10060) / 21.35900

VT_Z = (VT – 56.99917) / 21.12677

SF_Z = (SF – 83.56494) / 23.02758

RE_Z = (RE – 83.10276) / 31.64149

MH_Z = (MH – 75.21913) / 17.60698

Step 2: Formulae for aggregating scales in estimating aggregate physical and mental component scores

AGG_PHYS = (PF_Z*.42402) + (RP_Z*.35119) + (BP_Z*.31754) + (GH_Z*.24954) + (VT_Z*.02877) + (SF_Z*-.00753) + (RE_Z*-.19206) + (MH_Z*-.22069)

AGG_MENT = (PF_Z*-.22999) + (RP_Z*-.12329) + (BP_Z*-.09731) + (GH_Z*-.01571) + (VT_Z*.23534) + (SF_Z*.26876) + (RE_Z*.43407) + (MH_Z*.48581)

Step 3: Formulae for t-score transformation of component scores

Transformed physical (PCS) = 50 + (AGG_PHYS * 10)

Transformed mental (MCS) = 50 + (AGG_MENT * 10)

Abbreviations

PF: Physical function; RP: Role – Physical; BP: Bodily pain; GH: General health; VT: Vitality; SF: Social functioning; RE: Role – Emotional; MH: Mental health

APPENDIX H - ACCURACY AND RELIABILITY OF BIOMECHANICAL EQUIPMENT USED IN LEVEL AND STAIR GAIT

In order to determine how accurate and reliable the Qualisys motion capture system was, distance and angular tests were undertaken. Ten cameras captured raw kinematic data at 100 Hz. The system was calibrated as normal. For the distance trials, a wand of a known length was moved throughout the measurement volume 10 times for 10 seconds each. The distance between two markers (14mm diameter) was calculated. The known distance between markers on the first (large) wand was 749.4 mm, and the second (small) was 500.0 mm. Neither of these wands had been used to calibrate the camera system.

For the large wand (Tables A.I.1), the mean (± SD) difference between the measured and known length was 0.18 ± 0.17 mm and the RMS was 0.24. The coefficient of variation was 0.93. However, if the coefficient of variation was calculated for the absolute mean difference, it was 0.64.

Table A.I.1. Qualisys measured wand length (749.4 mm) (captured on 03/10/2006)

Trial	Measured wand length (mm)	Difference (mm)	Absolute difference (mm)
1	749.4	0.0	0.0
2	749.7	0.3	0.3
3	749.3	-0.1	0.1
4	749.7	0.3	0.3
5	749.6	0.2	0.2
6	749.5	0.1	0.1
7	749.7	0.3	0.3
8	749.7	0.3	0.3
9	749.8	0.4	0.4
10	749.5	0.1	0.1
Mean	749.6	0.2	0.2
SD	0.2	0.2	0.1

For the small wand (Table A.I.2.), the mean (±SD) difference between the measured and known length was -0.06 ± 0.31 mm and the RMS was 0.30. The coefficient of variation was -4.76. However, if the coefficient of variation was calculated for the absolute mean difference, it was 0.61.

Table A.I.2. Qualisys measured wand length (500.0 mm) (captured on 03/10/2006)

Trial	Measured wand length (mm)	Difference (mm)	Absolute difference (mm)
1	499.6	-0.4	0.4
2	499.8	-0.2	0.2
3	500.2	0.2	0.2
4	499.7	-0.3	0.3
5	500.0	0.0	0.0
6	499.8	-0.2	0.2
7	500.4	0.4	0.4
8	499.9	-0.1	0.1
9	499.5	-0.5	0.5
10	500.3	0.3	0.3
Mean	499.9	-0.1	0.3
SD	0.3	0.3	0.2

For the angular trials, three reflective markers (14 mm diameter) were placed onto a plastic goniometer. One marker was placed on each of the two arms of the goniometer and one marker at the vertex. The goniometer was adjusted at the following known angles: 45°, 90° and 180° and was moved throughout the measurement volume 10 times for 10 seconds each.

For the 45° angle (Table A.I.3), the mean (± SD) difference between the measured and known angles was 0.30 ± 0.05° and the RMS was 0.31. The coefficient of variation was 0.15 .

Table A.I.3. Qualisys measured angles (45°) (captured on 03/10/2006)

Trial	Measured angle (°)	Difference (°)	Absolute difference (°)
1	45.4	0.4	0.4
2	45.4	0.4	0.4
3	45.4	0.4	0.4
4	45.3	0.3	0.3
5	45.3	0.3	0.3
6	45.3	0.3	0.3
7	45.3	0.3	0.3
8	45.3	0.3	0.3
9	45.3	0.3	0.3
10	45.3	0.3	0.3
Mean	45.3	0.3	0.3
SD	0.0	0.0	0.0

For the 90° angle (Table A.I.4), the mean (± SD) difference between the measured and known angles was 0.33 ± 0.30° and the RMS was 0.33. The coefficient of variation was 0.09.

Table A.I.4. Qualisys measured angles (90°) (captured on 03/10/2006)

Trial	Measured angle (°)	Difference (°)	Absolute difference (°)
1	90.4	0.4	0.4
2	90.3	0.3	0.3
3	90.3	0.3	0.3
4	90.3	0.3	0.3
5	90.4	0.4	0.4
6	90.3	0.3	0.3
7	90.3	0.3	0.3
8	90.4	0.3	0.3
9	90.3	0.3	0.3
10	90.4	0.3	0.3
Mean	90.3	0.3	0.3
SD	0.0	0.0	0.0

For the 180° angle (Table A.I.5), the mean (± SD) difference between the measured and known angles was -0.54 ± 0.30° and the RMS was 0.61. The coefficient of variation was -0.54, respectively

Table A.I.5. Qualisys measured angles (180°) (captured on 03/10/2006)

Trial	Measured angle (°)	Difference (°)	Absolute difference (°)
1	179.5	-0.5	0.5
2	179.7	-0.3	0.3
3	179.3	-0.7	0.7
4	179.5	-0.5	0.5
5	179.1	-0.9	0.9
6	179.9	-0.1	0.1
7	179.1	-0.9	0.9
8	179.3	-0.7	0.7
9	179.2	-0.8	0.8
10	179.9	-0.1	0.1
Mean	179.5	-0.5	0.5
SD	0.3	0.3	0.3

In order to determine the accuracy of the Kistler and AMTI force plates, static weights were placed onto the respective force plates and the vertical force was recorded. The following known weights were used: 49.1 N (5 kg), 98.1 N (10 kg), 245.3 N (25 kg), 392.4 N (40kg). The weight of 392.4 N was achieved by placing the previous 3 known weights onto the force plates. The force for each weight was calculated 10 times for 10 seconds each.

Table A.I.6 indicates the mean (± SD) measured load for the Kistler and AMTI force plates (measured on 03/10/2006).

Applied load	**49.1 N**	**98.1 N**	**245.3 N**	**392.4 N**
Mean meaured load (Kistler)	49.4 ± 0.4	98.4 ± 1.1	244.1 ± 0.4	394.5 ± 0.2
RMS	0.37	0.32	0.34	0.41
Mean measured load (AMTI)	49.0 N ± 0.7	98.1 ± 0.1	242.1 ± 0.1	389.0 ± 1.6
RMS	0.28	0.32	0.31	0.33