THE UNIVERSITY OF HULL

Biomechanics and Quality of Life in Transtibial Amputees During and Following Rehabilitation: A Longitudinal Study

being a Thesis submitted for the Degree of Doctor of Philosophy in the University of Hull

by

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ABSTRACT

Following surgery, amputees must re-learn how to perform various movement tasks using altered lower limb mechanics. In order to optimise the process of re-learning these tasks and inform rehabilitation practice, an understanding of the longitudinal adaptations that occur both during and following a period of rehabilitation must be established. Scientific literature has reported the biomechanical, balance and quality of life (QOL) characteristics of transtibial amputees. However, no studies to date have outlined how these characteristics develop over time. The aim of this thesis, therefore, was to investigate the longitudinal changes that occurred in unilateral transtibial amputee movement, balance and QOL from their first treatments following amputation up to six months post-discharge from rehabilitation.

Studies one and two assessed the kinematic and psychological adaptations that occurred during the rehabilitation of 15 unilateral transtibial amputees. The amputees were randomly allocated into two groups, differing by early walking aid (EWA) used. One group used the Amputee Mobility Aid (AMA), which incorporated an articulation at the knee joint. The other group used the Pneumatic Post-Amputation Mobility Aid (PPAM) with no articulation at the knee joint. Amputee's gait and quality of life (QOL) were assessed at five standardised time points using three-dimensional motion capture and the SF-36 questionnaire, respectively. Overall, amputee's gait improved with walking velocity increasing over time ($p<0.01$). However, this did not differ between groups during EWA use, with most gait adaptations occurring upon receipt of patients' first functional prosthetic limb. Quality of life improved over time ($p=0.01$), although mental health was generally better than physical health. These results indicated that, despite increases in gait function and QOL during rehabilitation, there were no benefits of using one EWA over another.

Studies three, four and five assessed the biomechanical, balance and psychological adaptations that occurred in the six month period following discharge from rehabilitation in seven unilateral transtibial amputees. Amputee's gait and performance of activities of daily living (ADL) were assessed using three-dimensional motion capture. Balance ability and postural control were measured during the sensory organisation test (SOT) and the limits of stability (LOS) test protocols on the Neurocom Equitest®. Generic and prosthesis-related QOL and falls efficacy were assessed using the SF-36, prosthesis evaluation questionnaire (PEQ) and the modified falls efficacy scale (mFES), respectively. Amputee's gait improved over time with the intact limb experiencing greater forces (load rate, $p=0.01$; initial peak vertical ground reaction force, $p=0.04$). Amputees were able to perform ADLs safely, although they relied upon the intact limb in order to improve functioning. Overall, balance ability increased ($p=0.01$) with improved use of ankle movements during dynamic balance tasks ($p=0.02$), although amputees tended to rely heavily upon visual information. Amputees were able to improve the accuracy of movements during postural control tasks ($p<0.04$) without increasing the speed at which the tasks were completed. There were no significant psychological changes following discharge from rehabilitation. These results suggested that although transtibial amputee functioning improved following discharge from rehabilitation, inter-limb differences still remained.

In conclusion, the results from the current thesis have pertinent implications for the treatment of transtibial amputees both during and following rehabilitation. These include the identification of possible improvements to muscular strength, joint flexibility and balance training that may further improve transtibial amputee functioning.

Key Words: Amputee, Transtibial, Rehabilitation, Early Walking Aid, Gait, Balance Posture, Quality of Life.

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DISSEMINATION

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Conference Presentations

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GLOSSARY OF TERMS

AMA	Amputee Mobility Aid
BACPAR	British Association of Physiotherapists in Amputee Rehabilitation
COG	Centre of Gravity
COM	Centre of Mass
CPD	Computerised Dynamic Posturography
EWA	Early Walking Aid
Functional Prosthesis	Initial prosthesis prescribed post EWA use
GRF	Ground Reaction Force
kg	kilograms
LOS	Limits of Stability Test
m	metres
m/s	metres per second
n	number of subjects in sample
NHS	National Health Service
N	Newtons
mFES	Modified Falls Efficacy Scale
PPAM	Pneumatic Post-Amputation Aid
ROM	Range of motion
s	seconds
SD	standard deviation
SOT	Sensory Organisation Test
V	velocity
3D	Three-dimensional

1 CHAPTER ONE - INTRODUCTION

1.1 Introduction

Most individuals are able to move around in a safe, easy and energy efficient manner. The ability to execute day-to-day motor tasks such as standing, walking and negotiating stairs and obstacles is important as this ability forms an integral part of an independent lifestyle. Although many people achieve this independence reasonably well, a number of circumstances can compromise a person's motor functioning and thus, subsequently affect independence and quality of life (QOL).

Lower limb amputation results in significant physical alteration of the lower limb and presents the individual with various mechanical, physiological and psychological challenges. Following amputation, individuals must attempt to re-learn how to move within their environment using altered lower limb mechanics. This thesis focuses on the adaptations in gait, balance and QOL during rehabilitation and up to six months post-discharge from rehabilitation.

Lower limb amputation, occurring most commonly at the transtibial or below-knee level, has consequences specific to the individual and that are dependent upon pre-amputation status and physical capability. However, for many the goal is to sustain or regain a certain level of mobility. This mobility may have benefits at the individual level, in terms of independence and QOL, and at the societal level with regards to both healthcare and social costs. In order to achieve mobility following lower limb amputation, patients often follow a course of rehabilitation post-surgery. During rehabilitation patients re-learn how to walk with a prosthetic limb, whilst utilising a variety of prosthetic components. In the UK, guidelines for the post-operative rehabilitation of transtibial amputees are produced by an interest group under the jurisdiction of the Chartered Society of Physiotherapists called the British Association of Chartered Physiotherapists in Amputee Rehabilitation (BACPAR).

These guidelines recommend a variety of treatment methods and techniques during gait retraining, one of which is the use of early walking aids (EWA). Early walking aids are generic prosthetic devices aimed at encouraging early mobilisation and weight-bearing, prior to receiving a customised functional prosthesis (Scott *et al.,* 2000). During rehabilitation, transtibial amputees will initially use EWAs followed by customised functional prostheses whilst re-learning how to walk. Two EWAs are routinely used in the UK, the Amputee Mobility Aid (AMA) (Ortho Europe Ltd, Alton, UK) and the Pneumatic Post-Amputation Aid (PPAM) (Ortho Europe Ltd, Alton, UK). However, the selection and efficacy of using either EWA has not been thoroughly investigated within the literature. Once the relevant clinician deems a patient's level of mobility to be satisfactory, they are discharged from rehabilitation. Following discharge, those amputees with the ability to walk independently are likely to encounter more challenging physical tasks.

The lack of investigation into how transtibial amputees adjust to new physical and biomechanical constraints during two distinct periods of time post-surgery namely, during and immediately following rehabilitation, identifies a clear gap within the current literature. This may be due to the longitudinal study design requirements coupled with the potential difficulties of investigating a clinical patient group during these time frames. Few reports into amputees' biomechanical, balance or psychological adjustments to amputation have been made and there are currently no reports within the scientific literature of how this process of readjustment occurs or its influence on QOL (Isakov *et al.,* 1992; Brooks *et al.,* 2001; Vrieling *et al.,* 2009; Zidarov *et al.,* 2009). Therefore, the aims of this thesis were twofold. Firstly, to investigate the biomechanical movement adaptations that took place during rehabilitation along with the associated psychological changes. Secondly, this thesis aimed to investigate adaptations in biomechanical movement, balance ability and postural control along with the associated psychological changes that occurred during the six-month period following discharge from rehabilitation. During the six-months following discharge from rehabilitation, amputees are more likely to be required to independently accomplish more challenging motor tasks than previously required, thus experiencing a greater learning demand.

Understanding the longitudinal adaptations that occur will inform current and develop further rehabilitation protocols and treatments. This information may highlight areas of amputee mobility, both during and post rehabilitation that would benefit from further assessment and clinical intervention. These investigations will be of benefit to clinicians by providing them with objective information on which to base or justify clinical decision making and prosthetic prescriptions. These factors could also have a number of benefits in terms of improving cost-effective treatment and early and long-lasting mobilisation for transtibial amputees. In essence, this thesis will help inform rehabilitation practices of transtibial amputees in the UK.

1.2 Thesis Structure

The thesis begins with a comprehensive review of the pertinent literature in Chapter Two. The aim of the literature review is to present the rationale for the thesis. The review critically analyses the literature related to the biomechanics of activities of daily living (ADL), balance function, postural control, QOL and post-surgical rehabilitation of lower limb amputees. Finally, the aims, objectives ad hypotheses are presented to conclude the chapter.

A general methodology section is presented in Chapter Three. Here, details of the ethical approval and inclusion exclusion criteria are outlined. In addition, the chapter describes the experimental procedures followed and the justification and description of the biomechanical and psychological analysis tools used within the thesis.

Chapters Four and Five are the first empirical studies of the thesis, and reports the results of the longitudinal study assessing the efficacy of using different EWAs during the rehabilitation process of unilateral transtibial amputees.

Specifically, Chapter Four reports the kinematic gait adaptations during rehabilitation along with the subsequent effects on gait of using two different EWAs. Chapter Five outlines self-reported QOL during rehabilitation from a generic QOL assessment tool.

A summary of findings is then presented bringing together the information from during rehabilitation. Findings are discussed and the relationships between the biomechanical and QOL data are presented.

Chapters Six to Eight form the analyses of adaptations of movement, balance and QOL in transtibial amputees from the end of the structured rehabilitation process at one, three and six months post-discharge. Chapter Six reports the biomechanical analyses of level gait, obstacle crossing and gait when stepping from and to a new level, in order to assess participant's ability to successfully complete these ADLs. Analysis of participants' balance function and postural control is presented in Chapter Seven. This is pertinent given amputees report increased risk of falling (Miller *et al.,* 2001a). Computerised dynamic posturography (CDP) is used to assess adaptations during both static and dynamic conditions as participants adopt modified strategies for maintaining balance and explore limits of stability. Chapter Eight investigates the associated psychological changes that occur following discharge from rehabilitation. Specifically, self-reported QOL from generic and prosthesis related assessment tools are presented along with perceived falls efficacy. Changes in these self-reports over time are discussed with reference to previous findings.

A summary of analyses of amputees post-discharge from rehabilitation is presented, discussing the relationships between the various data sets presented in Chapters Six, Seven and Eight.

Finally, Chapter Nine provides an overall summary of the thesis contents. The clinical implications of the findings are outlined with regards to the management and treatment of transtibial amputees during and post-rehabilitation. The limitations of the thesis are highlighted and suggestions for future research directions are made. Concluding remarks are presented to bring the thesis to a close.

2 CHAPTER TWO - REVIEW OF LITERATURE

2.1 Introduction

Initially, the review of literature defines amputation and presents national descriptive statistics on lower limb amputation. The review then uses key reference literature to describe the biomechanics of movement in able-bodied individuals. The review then critically evaluates the pertinent literature investigating the biomechanics associated with lower limb amputee gait, activities of daily living (ADLs), balance and postural control. The literature relating to generic and prosthesis specific quality of life (QOL), as well as falls efficacy in lower limb amputees is critically analysed. The literature on the rehabilitation process, including gait re-education and early walking aids (EWA), as well as able-bodied prosthetic simulator gait is presented and critiqued. Finally, a summary of the literature is presented with the overall aim, specific objectives and hypotheses outlined.

2.2 Lower Limb Amputee Statistics in the UK

Amputation has been defined as ‘the removal of a dead, bad or useless limb’ (Kirtley, 2006, p 208) and occurs to around 5000 people each year in the UK (NASDAB, 2009). Lower limb amputations accounted for around 90% of all amputations in the UK over the last 10 years (NASDAB, 2009). The most common level of lower limb amputation is transtibial or below the knee (Table 2.1). Transtibial amputations account for approximately 53% of the total lower limb amputations each year, equating to around 2560 amputations (NASDAB, 2009). Transtibial amputations tend to occur to those aged over 55 years (74%) with the majority of individuals being male (73%) (NASDAB, 2009). Transtibial amputation can occur for a variety of reasons such as trauma e.g. motor vehicle accident or elective surgery for a predisposing condition e.g. talipes equinovarus. However, the most common cause of transtibial amputation is lower limb dysvascularity, accounting for 74% of transtibial amputations in 2006/7 (NASDAB, 2009). Lower limb dysvascularity occurs for a number of reasons including diabetes mellitus, arteriosclerosis and Buergers's disease, a progressive inflammation and thrombosis of peripheral circulatory vessels. In such cases, where a general worsening of the lower limbs vascular condition is observed, amputation is carried out to alleviate symptoms associated with and/or to prevent further deterioration of the lower limb.

Table 2.1 Number, percentage and level of lower limb amputations in the UK in 2006/7. Data reproduced from United Kingdom national statistics database (NASDAB 2009).

Amputation Level	Total Number	Percentage (%) of Total Amputations
Hemipelvectomy	14	<1
Hip Disarticulation	26	1
Transfemoral	1788	39
Knee Disarticulation	57	1
Transtibial	2411	53
Ankle Disarticulation	14	<1
Partial Foot	51	1
Lower Digits	17	<1
Double Lower Limb Amputation	196	4
Total	4574	100

2.3 Human Movement and Gait Analysis

Modern motion analysis systems allow in-depth assessment of human movement and in particular walking. The assessment of movement patterns in lower limb amputees has the potential to further understand and improve functioning. This process, termed gait analysis, has revealed a multitude of features about the cyclic nature of walking or the gait cycle. The gait cycle is defined as foot contact with the ground to the next subsequent contact with the ground, on the same foot. The divisions of the gait cycle are noted below in Figure 2.3. The gait cycle is broadly split into phases termed stance phase and swing phase, relating to periods where the foot is in contact with the ground (stance) or not (swing).

2.3.1 Functional Tasks of Gait

The stance and swing phases are associated with three functional goals that must be achieved during successful gait namely, weight acceptance, single limb support and limb advancement (Perry, 1992). These functional tasks are again subdivided into eight sub-phases that further describe the movement of the lower limbs. These functional tasks are outlined below with reference to the joint rotations (angles), ground reaction forces (GRF) and joint kinetics (moments and powers) of the lower limbs (Perry, 1992).

2.3.1.1 Weight Acceptance

Initial Contact – Loading Response

The weight acceptance task of gait forms the initial 10% of the gait cycle. Beginning with initial contact, also referred to as heel strike and foot contact, this also marks the start of double limb support where body weight is transferred from one foot to the other (Perry, 1992). Typically in able-bodied gait, the heel will contact the ground first with the ankle gradually plantarflexing in order to lower the foot to the ground until ‘foot flat’ (Figure 2.1). At the same time, the knee will go from a relatively extended position (0-3°) to a more flexed position (15-20°) in order to attenuate shock from heel strike (Figure 2.1) (Kirtley, 2006). This mechanism is termed the loading response. There is little movement in the knee and ankle joints in the frontal or transverse planes during weight acceptance. The hip joint will remain in a relatively flexed position during this period, ranging between 25 and 35° (Figure 2.1). The pelvis will also remain in a relatively fixed position in the sagittal plane although it may be internally rotated by between 2-8° and in upward obliquity (hip hike) by around 5° (Kirtley, 2006).

Following heel strike there is a great increase in vertical (Fz) GRF as the lower limb is loaded. This will typically reach around one times body weight in magnitude. Force in the anterior-posterior direction (Fy) is represented by an increasing braking force, which peaks just after weight acceptance concludes. The medial-lateral force (Fx) may be signified by an increasingly medial force, although the magnitude of this force is relatively low, usually below 5% of body weight and is highly variable between individuals (Kirtley, 2006).

The combination of the joint angles and GRF vector (GRFv) allows for the analysis of joint kinetics via a process termed inverse dynamics (Kirtley, 2006). The GRFv changes position throughout the gait cycle and at initial contact, the GRFv passes through the heel, behind the ankle and in front of the knee and hip. This results in initial hip extensor and knee flexor moments. The GRFv progresses in the anterior direction as the foot approaches foot flat. Here the hip and knee moments increase and both become extensor in direction and, as the GRFv moves anterior to the ankle, an increasing plantarflexor moment is observed (Figure 2.2) (Kirtley, 2006). At this point there is little power generated or absorbed by the ankle. The aforementioned knee flexion is controlled by the eccentric action of the knee extensors during the K1 power absorption phase (Figure 2.2). Meanwhile, the hip extensors contract concentrically to produce the H1 power generation phase (Kirtley, 2006).

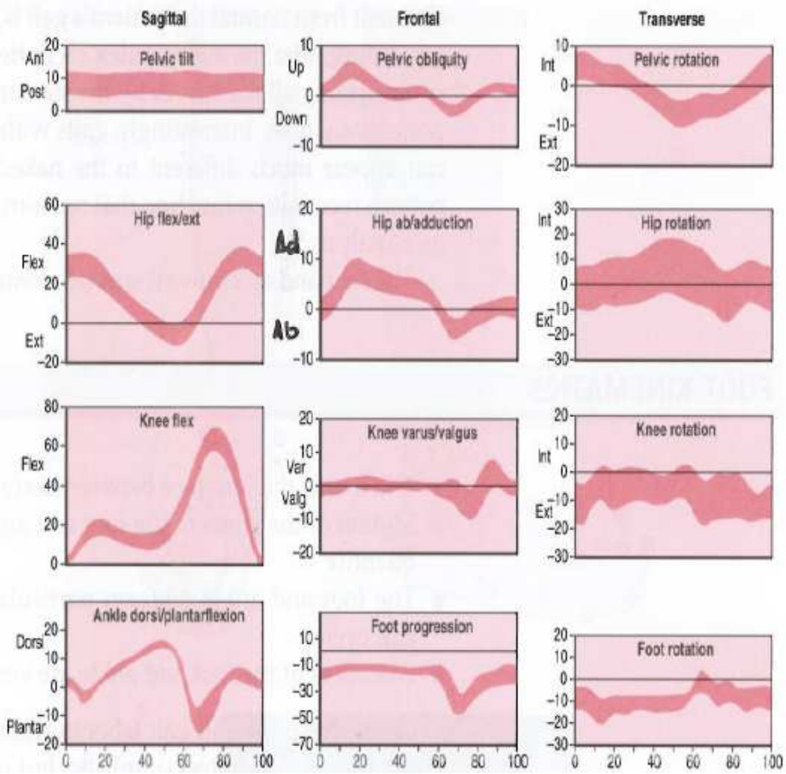


Figure 2.1 Normative 3D lower limb joint kinematics during level gait shown ± 1SD as (Image taken from Kirtley, 2006).

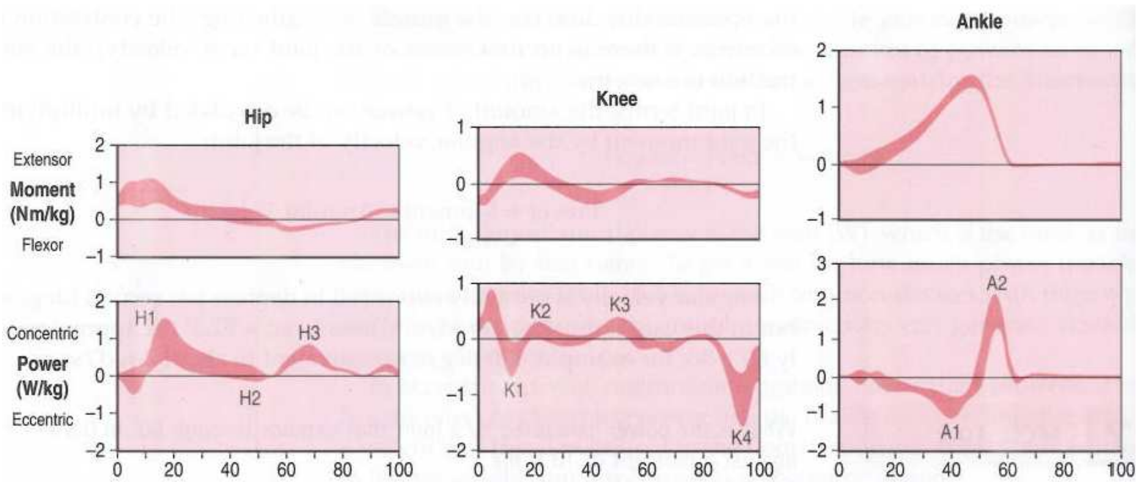


Figure 2.2 Normative sagittal plane joint moments and powers during level gait shown as ± 1SD with power bursts labelled (Image taken from Kirtley, 2006).

2.3.1.2 Single Limb Support

Mid-Stance – Terminal Stance

The single limb support task of gait occurs between 10-50% of the gait cycle (Perry, 1992). Here the limb in question supports the body whilst the opposing limb advances. Typically, the ankle joint will go from a slightly plantarflexed position to a dorsiflexed position due to the action of the shank during mid-stance progressing over the foot (Perry, 1992) (Figure 2.3). Although flexed, the knee and hip joints will begin to extend. In the frontal plane, the pelvis exhibits slight upward obliquity. Terminal stance begins when the supporting limb heel starts to rise and finishes as the opposing limb contacts the ground, also commencing the second double limb support phase (Perry, 1992). Approaching terminal stance, the ankle joint continues to dorsiflex before changing direction around 45-50% of the gait cycle in preparation for pre-swing. The knee joint reaches near to full extension (5-10°) before increasing flexion in preparation for pre-swing. The hip joint is still in flexion although it continues to extend before reaching peak extension prior to toe off (Perry, 1992).

During mid-stance, the vertical GRF peaks around 1.2-1.3 times body weight. This is followed by the vertical GRF falling below body weight (around 0.7 times body weight) during terminal stance due to the swinging action of the opposing limb reducing the whole body loading on the ground before rising towards a value equal to body weight prior to pre-swing. The Fy force changes from anterior to posterior during single limb support (Kirtley, 2006).

The GRFv passes through the hip joint resulting in relatively low joint moments and powers during mid-stance (Kirtley, 2006). As the body progresses forward during terminal stance the GRFv passes behind the hip joint. This results in a hip flexor moment and a power absorption phase (H2) as the hip flexors contract eccentrically. With regards to the knee, the GRFv goes from behind, through and then to the front of the knee joint resulting in extensor, neutral then flexor joint moments respectively. There is a power generation phase (K2) during mid-stance as the knee extensors contract concentrically (Kirtley, 2006). A large peak ankle plantarflexor moment (around 1.4Nm/kg) is observed during mid to terminal stance as the GRFv is positioned towards the front of the foot, under the metatarsal heads and passes in front of the ankle joint (Kirtley, 2006). The power absorption phase A1 is also associated with this action as the ankle plantarflexors contract eccentrically.

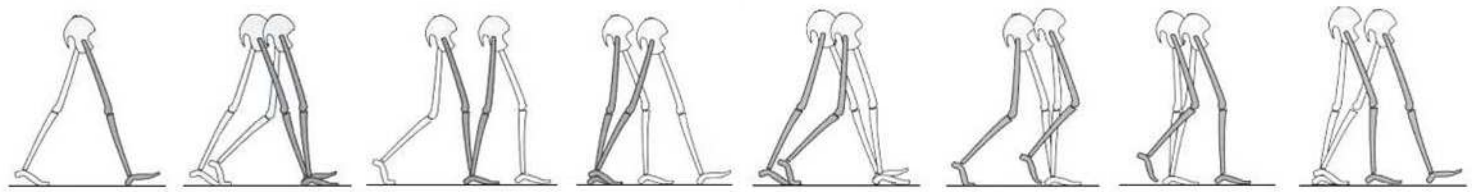
2.3.1.3 Limb Advancement

Pre-Swing – Initial-Swing – Mid-Swing – Terminal Swing

During limb advancement the supporting limb prepares to become the trail or swing limb. Limb advancement begins with pre-swing (50-60% of gait cycle) at which point the second double limb support phase ends (Perry, 1992). Here the aim is a safe transition from double limb support to single limb support. During pre-swing an extension-flexion transition is observed at the hip, with the knee also flexing to around 40° ensuring adequate foot clearance. This occurs as a result of the ankle joint plantarflexing quickly prior to initial swing in order to propel the limb forwards (Kirtley, 2006). In the transverse plane, the pelvis reaches peak external rotation of just below 10°. Following pre-swing initial swing occurs between 60-73% of the gait cycle. Here the hip and particularly the knee (peak 60°) flex in order to lift the swinging limb foot from the ground. This important mechanism assists in obstacle crossing and stepping given that the ankle joint is, at times, plantarflexed during swing phase, thus more likely to contact the ground. The ankle joint moves from peak plantarflexion and begins to dorsiflex in order to assist ground clearance until the swing limb is opposite the contralateral limb, where the ankle is very slightly plantarflexed. Mid-swing occurs between 73-87% of the gait cycle and is primarily focussed on limb advancement (Perry, 1992). This phase occurs from the point the swinging limb is parallel to the contralateral limb to when the swinging limb has advanced forward (Perry, 1992). The ankle joint is maintained in a relatively neutral position. Due to the momentum of the swinging limb and the relaxation of knee flexors (previously active during initial swing), the hip and knee joints flex and extend respectively. During mid-swing, the pelvis exhibits slight downward obliquity in the frontal plane. Finally, the swinging limb is prepared for stance during terminal swing (87-100% of the gait cycle) (Perry, 1992). During terminal swing the ankle joint position is neutral as the shank continues to advance. The knee joint moves from peak flexion and begins to extend whilst the hip joint is flexed in preparation for foot contact. Here the pelvis is internally rotated in the transverse plane (Kirtley, 2006).

The hip flexor action, coupled with the GRFv being behind the joint results in a hip flexor moment during pre-swing. There is a small knee extensor moment during this time period (K3) as the power is absorbed during pre-swing. The lack of GRFv means there are negligible joint moments for the remainder of the gait cycle (Kirtley, 2006).

The hip flexors also generate the power burst H3 as they attempt to rotate the thigh forwards by contracting concentrically in preparation for initial swing. The plantarflexor action of the ankle joint at this point produces the concentric power bursts labelled A2 (Kirtley, 2006). The lack of hip and ankle moments from initial swing to the end of the gait cycle results in no power generation or absorption at these joints during this time period. During terminal swing the knee flexors absorb power resulting in a negative power burst labelled K4 (Kirtley, 2006).



Phase	Stance Phase					Swing Phase		
Goal	Weight Acceptance		Single Limb Support		Limb Advancement			
	1	2	3	4	5	6	7	8
Sub-phase	Initial Contact	Loading Response	Mid Stance	Terminal Stance	Pre-Swing	Initial Swing	Mid Swing	Terminal Swing
Percentage Gait Cycle	0-2%	0-10%	10-30%	30-50%	50-60%	60-73%	73-87%	87-100%
Description	Instant that the foot touches ground. The aim is to position the limb to begin stance with a hell rocker.	Initial double stance period. The aim is to absorb shock, begin weight-bearing and preserve forward limb progression.	The first half of single limb support with the aim of progression over a stationary foot and to provide limb and trunk stability.	Last half of single limb support with the aim of progression past the stationary foot.	Also known as weight transfer, the aim is to prepare and position the limb for swing.	Foot lifts off in order to achieve ground clearance and advancement of the limb.	Swinging limb advances past contralateral limb to further advance gait.	Swinging limb slows as foot makes contact with the ground in order to complete limb advancement and prepare the limb for swing.

Figure 2.3 The gait cycle with phase and sub-phase divisions highlighted. Schematic representation of able-bodied kinematics and a description of the functional goals of each of the eight subdivisions are also provided. Values relate to percentage (%) of total gait cycle. Figures adapted from Perry, (1992).

2.3.2 Temporal-Spatial Variables

Movement of the lower limbs also provides information regarding the time and distance characteristics of gait. These characteristics are termed temporal-spatial (TSP) variables and provide information about walking speed, cadence, and step and stride length among others. Values for the TSPs of gait vary depending upon sex, age, height and various movement pathologies; therefore reporting of normative values is problematic and prone to variability (Kirtley, 2006). However, the literature reports that in able-bodied men, self-selected walking speed, cadence and stride length can be approximated to be around 1.3-16 m/s, 110-115 steps/min and 1.4-1.6m, respectively (Kirtley, 2006).

Gait analysis has revealed much about the functioning of able-bodied movement. However, modern day gait analysis is derived from a need to understand how pathological conditions affected human movement, as well as quantifying the effects of subsequent treatment and interventions on these conditions. Many pathological conditions affect movement and one such condition that has an obvious impact upon movement, and gait in particular, is lower limb amputation.

2.4 Lower Limb Amputee Movement Patterns

2.4.1 Biomechanics of Amputee Gait

Overall, reports from gait studies show that transtibial amputees are able to walk effectively and in some cases, with a gait not too dissimilar to able-bodied individuals (Sanderson and Martin, 1997). However, there are noted compensatory mechanisms evident in the kinetic profiles of transtibial amputees when compared to able-bodied individuals.

Perhaps one of the most obvious patient concerns following surgery, is whether or not they will be able to walk again. In terms of mobility, the physical loss of part of a limb is perhaps the most debilitating factor associated with lower limb amputation.

Functional prosthetic limbs are prescribed in order to replace the absent lower limb and can help increase patient's functioning and QOL by providing them with a means to ambulate. However, prosthetic limbs are exactly that and transtibial amputees must still walk with some mechanical constraint and a reduction in degrees of freedom. The literature has investigated the way in which amputees walk in comparison to that of able-bodied individuals (Sanderson and Martin 1997; Nolan *et al.,* 2003). This study design has been questioned in the literature suggesting that given an amputee's inherent physical asymmetries, a new asymmetrical optimum should be sought (Winter and Sienko, 1988). Although this is a valid argument, comparison to able-bodied gait allows studies to compare amputee functioning to what may be considered more optimal functioning. In addition, the restoration of symmetrical functioning is often the aim during rehabilitation.

2.4.1.1 Temporal-Spatial Variables

Amputee gait analysis is an area where scientific investigation has discovered a number of common compensatory mechanisms and features observed in transtibial amputee gait patterns.

Such features include altered temporal-spatial characteristics of amputee gait. Many studies have reported lower walking velocities, longer step length but shorter relative stance duration on the affected side and longer intact limb stance duration in amputees when compared to able-bodied individuals (Winter and Sienko, 1988; Hurley *et al.,* 1990; Perry, 1992; Sanderson and Martin, 1997; Powers *et al.,* 1998; Isakov *et al.,* 2000; Bateni and Olney 2002; Grumillier *et al.,* 2008; Vickers *et al.,* 2008; Vrieling *et al.,* 2008 Vanicek *et al.,* 2009a). The temporal-spatial asymmetry reported between the intact and affected limbs has been shown to reduce as a consequence of increased walking velocity but increase with higher prosthetic limb mass (Mattes *et al.,* 2000; Donker and Beek 2002; Nolan *et al.,* 2003). These asymmetries have been attributed to the perceived attempts of the amputee to protect their affected limb from increased forces and loading (Hurley *et al.,* 1990; Sanderson and Martin 1997; Powers *et al.,* 1998; Nolan *et al.,* 2003). Another explanation for the temporal-spatial asymmetries proposed by other studies was a lack of confidence in the ability to control the affected limb (Sanderson and Martin 1996; Sanderson and Martin 1997).

2.4.1.2 Joint Kinematics

These temporal-spatial asymmetries apparent in amputee gait are a result of altered lower limb mechanics. Assessment of the altered mechanical functioning apparent in transtibial amputees has highlighted some common kinematic compensatory mechanisms (Winter and Sienko 1988; Perry, 1992; Sanderson and Martin 1997; Powers *et al.,* 1998; Bateni and Olney 2002; Beyaert *et al.,* 2008; Grumillier *et al.,* 2008; Silverman *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a).

It has been reported that amputees display reduced hip flexion during the stance phase of gait, maintaining a more vertical orientation of the lower limb (Sanderson and Martin 1997). However, increased hip flexion from mid to terminal swing has also been observed (Sanderson and Martin 1997), a feature related to the reduced ankle function described below.

Transtibial amputees tend to reduce the range of motion (ROM) at the knee joint of the affected limb during the stance phase (Powers *et al.,* 1998; Bateni and Olney 2002; Beyaert *et al.,* 2008; Vickers *et al.,* 2008). The literature has suggested that this lack of knee ROM, particularly during weight acceptance, is a protective mechanism of the affected limb and by keeping the GRF vector closer to the knee joint, demands placed upon knee extensor musculature are reduced (Beyaert *et al.,* 2008).

The loss of calf musculature in the affected limb results in the inability to actively control the ankle joint during gait in amputees. The literature has reported a lack of dorsiflexion from mid to terminal stance, perhaps as a result of a stiff prosthetic ankle complex in the prosthetic limb (Sanderson and Martin, 1997). Reduced plantarflexion or push off during pre-swing in amputees when compared to able-bodied individuals has led to the development of energy storing prostheses which attempt to compensate for this absent mechanism (Gitter *et al.,* 1991). As the ankle joint of passive prosthetic limbs cannot actively dorsiflex during swing phase the increased hip flexion reported, could be a compensatory measure that is as an attempt to aid ground clearance (Sanderson and Martin, 1997).

Most studies investigating the kinematic profiles of transtibial amputees report some level of kinematic asymmetry between the affected and intact limbs (Winter and Sienko, 1988; Hurley *et al.*, 1990; Sanderson and Martin, 1997; Powers *et al.,* 1998; Beyaert *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). This effect may be less profound in more experienced transtibial amputees, where fewer kinematic differences were noted when compared to able-bodied gait (Sanderson and Martin, 1997). This study tested patients that were experienced in using their prosthetic limbs (range 1-22 years, $\bar{x}$ 12.1 years) and it is likely that over time these patients learnt to better manage their altered lower limb mechanics and learn to walk relatively proficiently (Sanderson and Martin, 1997).

2.4.1.3 Ground Reaction Forces

The altered kinetic function of the affected limb described below is linked to the GRFs associated with the lower limbs transtibial amputees (Sanderson and Martin, 1997; Nolan *et al.,* 2003; Beyaert *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). Studies have reported that the affected limb exhibited reduced vertical GRF and generated less propulsive impulse than the intact limb (Nolan *et al.,* 2003; Silverman *et al.,* 2008). A common explanation for these effects is an attempt to protect the surfaces of the residuum from increased loading (Hurley *et al.,* 1990; Sanderson and Martin, 1997; Jones *et al.,* 2001; Nolan *et al.,* 2003). Also, the whole body centre of mass is shifted towards the intact limb during gait due to its increased mass in comparison to the affected limb, and this may cause the GRF to be higher in the intact limb (Nolan *et al.*, 2003). However, alteration of the prosthetic limb inertial properties by attempting to match those of the intact limb has been reported to increase the energy cost of gait in amputees (Mattes *et al.*, 2000). Increased intact limb GRF could also be explained by the greater confidence of the amputee in controlling the intact limb, thus exposing it to greater forces (Sanderson and Martin, 1997). Similarly, findings have been reported in the literature where an increase in static weight-bearing and decreased perceived pain over time was observed in transtibial amputees (Jones *et al.*, 2001). This study suggested that new amputees were more cautious of weight-bearing on their affected limb, as its surfaces and constructs were not used to or designed for receiving high stump interface pressure (Jones *et al.*, 2001).

2.4.1.4 Muscle Activation

Studies have reported increased activity in musculature controlling the knee of the affected limb, via surface electromyography, in transtibial amputees compared to that in able-bodied individuals (Winter and Sienko, 1988; Powers *et al.*, 1998; Isakov *et al.*, 2000). These studies noted greater knee flexor muscle activity throughout stance phase in the transtibial amputees, with peak activity occurring during weight acceptance (Winter and Sienko, 1988; Powers *et al.*, 1998; Isakov *et al.*, 2000). Another study reported that greater knee flexor activity was a result of transtibial amputees' tendency to lean forward with the trunk, in order to aid limb progression over a solid prosthetic ankle (Powers *et al.*, 1998). These findings, coupled with the observation of increased knee extensor muscle activity during the first 40% of stride, resulted in reduced knee moments, as described below (Winter and Sienko, 1988).

The literature has also suggested that the co-contraction of knee flexors and extensors was an attempt by the amputees to stabilise the affected knee joint during stance phase

(Powers *et al.,* 1998). During the stance phase observed in able-bodied gait, foot strike is followed by plantarflexion at the ankle, resulting in 'foot flat' (Perry, 1992). As this mechanism is not possible in all prosthetic ankles, amputees may co-contract the musculature controlling the knee joint action in order to control weight acceptance whilst in stance phase on the prosthetic heel. This may provide some added stability in response to the lack of a foot flat mechanism (Powers *et al.,* 1998).

2.4.1.5 Joint Kinetics

Although Sanderson and Martin, (1997) reported similarities in the kinematic profiles of amputee and able-bodied groups, distinct kinetic differences were observed. This is the case in many investigations into transtibial amputee gait particularly with reference to the affected limb (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers *et al.,* 1998; Vickers *et al.,* 2008, Beyaert *et al.,* 2008; Vanicek *et al.,* 2009a).

The aforementioned reduced knee ROM during the affected limb stance phase is a result of the altered patterns of muscle activity described above. Essentially, this reflects the amputees maintaining the affected limb in a more extended position during the stance phase, with the absence of a knee loading response. This in turn results in a net affected limb knee moment close to zero during early to mid stance (Winter and Sienko, 1988).

A number of studies have reported this effect of decreased knee joint moments on the affected limb (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers *et al.,* 1998; Beyaert *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). A suggested explanation is that by maintaining the knee in an extended position, the demands on knee extensor musculature during stance phase are reduced due to the vertical GRF vector being closer to the knee joint centre, also preventing the knee from collapsing during stance phase (Sanderson and Martin 1997; Powers *et al.,* 1998; Vanicek *et al.,* 2009a).

The tendency in transtibial amputees to forward trunk lean has been partially corroborated by observations of increased hip flexion (Powers *et al.,* 1998; Bateni and Olney, 2002). Forward trunk lean causes a greater flexion moment at the hip, thus the hip extensors must work harder to control the hip joint, while also reducing the external knee flexion moment and subsequent demand on the hip flexor/knee extensor musculature (Powers *et al.,* 1998).

It has been reported previously that able-bodied individuals also displayed a greater support moment, defined as the combined effect of net moments about the ankle, knee and hip, compared to transtibial amputees (Sanderson and Martin, 1997). With reference to early stance phase, the magnitude of difference in support moments was greater in the affected limb than the intact limb when compared to the able-bodied group. These differences were exaggerated at higher walking velocities, although the profiles remained similar throughout (Sanderson and Martin, 1997).

In able-bodied gait, calf musculature aids limb progression and stability and has been reported to contribute up to 80% of the mechanical power (Winter, 1983). However, the lack of calf musculature in transtibial amputees results in reduced ankle power generation in the affected limb of amputees (Vickers *et al.,* 2008). Reduced power generation at the prosthetic ankle compared to the intact ankle during pre-swing have been linked to a lack of propulsive GRF in the affected limb described above (Silverman *et al.,* 2008; Vickers *et al.,* 2008).

Along with reduced knee flexion, joint moments and abnormal EMG activity in the affected limb, knee joint powers are also reduced, with reference to the power generation phase (K2) in the knee extensors following weight acceptance (Winter and Sienko 1988; Perry, 1992; Sanderson and Martin, 1997; Powers *et al.,* 1998; Beyaert *et al.,* 2008; Silverman *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). Studies reported that very little power was generated at the knee in the affected limb where it is required to prevent limb collapse during early to mid-stance and aid propulsion during late stance phase (Winter and Sienko 1988; Powers *et al.,* 1998; Beyaert *et al.,* 2008; Silverman *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). The aforementioned altered patterns of muscle activity affecting the knee joint, along with the resulting kinetic and kinematic adaptations described previously, have also been linked to the lack of function in the prosthetic ankle (Powers *et al.*, 1998).

The literature has shown that the lack of propulsion from the prosthetic ankle, whilst affecting the knee joint, also places extra demands upon the hip joint extensor muscles during stance at pre-swing, where increased power generation has been reported (Winter and Sienko, 1988). Several studies are generally in agreement that this compensatory mechanism attempts to supplement the power generation lost in the prosthetic ankle and the knee joint in the affected limb. The increased power generation at the hip is required in order to prevent the collapse of the affected limb during stance phase (Vanicek *et al.,* 2009a) as well as aiding forward propulsion of the affected limb (Winter and Sienko, 1988; Bateni and Olney 2002; Grumillier *et al.,* 2008; Silverman *et al.,* 2008; Vanicek *et al.*, 2009a). As may be expected, the compensatory mechanisms apparent in amputee gait lead to an increase in energy consumption, even at slower self-selected walking speeds with energy consumption increasing as a function of increasing walking speed (Houdijk *et al.*, 2009).

2.4.1.6 <u>Limitations to Biomechanics of Amputee Gait Literature</u>

The literature reviewed provides a very descriptive analysis of transtibial amputee gait. They also highlight a number of compensatory mechanisms that are evident during transtibial amputee gait. However, there are aspects within the literature that prevent universal application of their findings to the wider amputee population. The number and heterogeneous nature of the amputee participants reported mean that these participant groups do not accurately reflect the transtibial amputee population as a whole. Firstly, age ranges have varied between 29 to 56.6 years of age ($\bar{\times}$44.9 ± 8.3) (Hurley *et al.*, 1990; Sanderson and Martin, 1997; Powers *et al.*, 1998; Isakov *et al.*, 2000; Mattes *et al.*, 2000; Bateni and Olney, 2002; Nolan *et al.*, 2003; Royer and Wasilewski, 2006; Beyaert *et al.*, 2008; Grumillier *et al.*, 2008; Silverman *et al.*, 2008; Vrieling *et al.*, 2008; Houdijk *et al.*, 2009; Vanicek *et al.*, 2009a), with one study focussing exclusively on the older amputee ($\bar{\times}$71 years of age) (Vickers *et al.*, 2008). Secondly, causes of amputation have also varied within the literature, some participants were exclusively secondary to trauma (Sanderson and Martin, 1997; Isakov *et al.*, 2000; Bateni and Olney, 2002; Nolan *et al.*, 2003; Beyaert *et al.*, 2008; Grumillier *et al.*, 2008) or vascular disease (Powers *et al.*, 1998; Vickers *et al.*, 2008 – 1/8 due to cancer) while some participant groups were secondary due to a variety of causes (Hurley *et al.*, 1990; Mattes *et al.*, 2000; Royer and Wasilewski, 2006; Silverman *et al.*, 2008; Vrieling *et al.*, 2008; Houdijk *et al.*, 2009; Vanicek *et al.*, 2009a).

With regards to the characteristics of transtibial amputee participants, the application of findings from this younger and physically more capable population, to that of the overall population of older vascular transtibial amputees, may not be completely valid. There are a number of inconsistencies in patient characteristics associated with the studies reported above. In some cases these inconsistencies are due in part to some studies controlling certain characteristics such as time since and cause of amputation (Hurley *et al.*, 1990; Sanderson and Martin, 1997; Powers *et al.*, 1998; Isakov *et al.*, 2000; Mattes *et al.*, 2000; Bateni and Olney, 2002; Nolan *et al.*, 2003; Royer and Wasilewski, 2006; Beyaert *et al.*, 2008; Grumillier *et al.*, 2008; Vickers *et al.*, 2008; Vrieling *et al.*, 2008; Houdijk *et al.*, 2009). However, it is likely that the large and varied nature of the general transtibial amputee population, coupled with difficulties in recruiting from this population, may result in the wide-ranging reports of patient characteristics. While this is not a criticism of the literature per se, such issues make comparison of results between studies difficult. Therefore, it is important to take the patient characteristics reports into account when interpreting results from such studies.

There are also a number of technical and methodological issues associated with the studies reviewed above. For example, studies report inverse dynamics calculations such as joint moments and powers from simplified models of the lower limb using a simplified two dimensional analysis (Winter and Sienko, 1988; Bateni and Olney, 2002).

Studies have also manipulated the velocity at which participants are required to walk (Sanderson and Martin 1997; Houdijk *et al.,* 2009). This protocol may be questioned as amputees will walk at a self-selected velocity during their everyday life. Amputees secondary to vascular disease tend to have lower self-selected walking velocities than the amputees reported in studies altering gait velocity (Powers *et al.,* 1998; Vickers *et al.*, 2008). Therefore, comparison of results between these groups is problematic as they may represent slightly different sub-populations.

The cross-sectional design of many studies reviewed, fails to indicate how the compensatory mechanisms of amputee gait are established over time during and following a period of rehabilitation. Although time since amputation and subsequent practice effects have been shown to be indicative of amputee ability (Hurley *et al.,* 1990), most studies controlled this variable by testing patients exclusively ≥ one year post-amputation (Hurley *et al.,* 1990; Sanderson and Martin, 1997; Powers *et al.,* 1998; Isakov *et al.,* 2000; Mattes *et al.,* 2000; Bateni and Olney, 2002; Nolan *et al.,* 2003; Royer and Wasilewski, 2006; Beyaert *et al.,* 2008; Grumillier *et al.,* 2008; Vickers *et al.*, 2008; Vrieling *et al.,* 2008; Houdijk *et al.,* 2009). The amputees tested in previous studies may have been more accustomed to walking within their new mechanical constraints whilst using prosthetic components. Also, a range of experience in prosthetic use within the same study may mask any experience inter-participant differences in amputee gait. Therefore, it is difficult to conclude whether the reported gait asymmetries and compensatory mechanisms were indicative of typical transtibial amputee gait or whether these profiles are a result of other factors such as previous prosthetic use, age, physical ability or rehabilitation methods, among others. Results from such studies may only be specific to the experienced amputee population under investigation.

An explanation of the longitudinal gait adaptations and the factors that may influence them during gait relearning has not been reported. This is an important oversight as the development of compensatory mechanisms and associated asymmetrical gait described above have been reported to predispose this patient group to a number of further complications such as osteoarthritis (Royer and Koenig, 2005; Royer and Wasilewski, 2006) and falling (Vanicek *et al.,* 2009a). Identification and quantification of the factors associated with gait re-learning may allow clinicians involved in amputee rehabilitation to reduce compensatory gait mechanisms and the associated increased metabolic cost, by re-educating gait more effectively.

2.4.2 Biomechanics of Activities of Daily Living

Following amputation, patients will initially start to practice level gait. However, gait is performed in a number of contexts that include steps, obstacles and stairs. The scientific literature has assessed transtibial amputee movement patterns as they perform tasks of this nature, collectively known as activities of daily living (ADL).

Amputees' ability to step to and from a new level during continuous gait is an ADL that has not been investigated thoroughly. It is important to understand how amputees perform this task as it may be encountered on a regular basis, for example, when stepping to and from a roadside kerb. Although this thesis does not analyse stair climbing in transtibial amputees, this review critically analyses the stair climbing literature as it is envisaged that many commonalities will lie between the stair climbing and stepping to and from a new level.

2.4.3 Stair Negotiation

Stair negotiation is an ADL performed by people on a regular basis and for some, is a challenging physical task (McFadyen and Winter, 1988; Beaulieu *et al.*, 2008). This is supported in the literature by reports of increased lower limb joint moments and thus, increased support moments, during stair ascent and descent when compared to level gait (McFadyen and Winter, 1988; Kirtley, 2006, Beaulieu *et al.*, 2008). These studies also reported that during stair ascent concentric muscle contractions were predominately observed, whereas during stair descent, predominately eccentric muscle contractions were present (McFadyen and Winter, 1988, Beaulieu *et al.*, 2008). This confirms that not only is stair climbing a more physically demanding task than level gait, it also requires different neuromuscular functioning for ascent and descent.

The key phases of stair ascent and descent have been outlined in the literature (McFadyen and Winter, 1988). These analyses help to form a basis for comparison against results from various clinical populations.

Stair ascent begins with the weight acceptance phase as the middle to front portion of the foot contacts the step. The ankle plantarflexor muscle group then help to position the body in preparation for the next phase. The 'pull up' phase follows weight acceptance and is the main progression in moving from one step to another (McFadyen and Winter, 1988). The 'pull up' is achieved by the concentric knee extensor activity by the quadriceps (K1) (Figure 2.4) and here the largest period of instability occurs, commencing with contralateral toe off, as body weight is supported by the lower limb with hip, knee and ankle joints all in flexed positions (McFadyen and Winter, 1988). Following 'pull up', the contralateral limb is in mid-swing and the 'forward continuance' phase begins (McFadyen and Winter, 1988). Here, mostly forward motion is observed as the ankle provides a ‘push off’ similar to that observed in level gait during pre-swing (A3) (Figure 2.4) (McFadyen and Winter, 1988). As the ankle joint plantarflexes, the lower limb is prepared for the swing phase. Beginning with toe off, the aim of swing is to ensure the safe progression of the lower limb up and over to the next step whilst avoiding contacting the intermediate step (McFadyen and Winter, 1988). Plantarflexion of the ankle and flexion of the knee aid step clearance, whilst hip flexion and the action of the contralateral limb aid limb progression.

Stair descent also begins with weight acceptance, although the lateral portion of the foot is the first part to make contact. Here energy is absorbed by the knee flexors during power burst K1 and ankle plantarflexors at power burst A1 (Figure 2.4) (McFadyen and Winter, 1988). A single limb stance phase commences with contralateral limb toe off and the ‘forward continuance’ phase begins. Here, there is a slight knee extension and power generation phase by the quadriceps at power burst K2, as the body moves forward and rises slowly (Figure 2.4). From mid stance to the beginning of swing phase, the body is lowered to the next step in the ‘controlled lowering phase’ (McFadyen and Winter, 1988). During this phase, the majority of the downward progression is achieved as power is absorbed at the ankle and knee (Figure 2.4). Following this, the hip joint flexes, producing power burst H1, in order to pull the limb from the current step. Power burst H2 is observed at the start of swing phase as the hip joint pulls the limb through, knee flexion decreases and the lower limb begins to extend in preparation for the next weight acceptance phase (Figure 2.4).

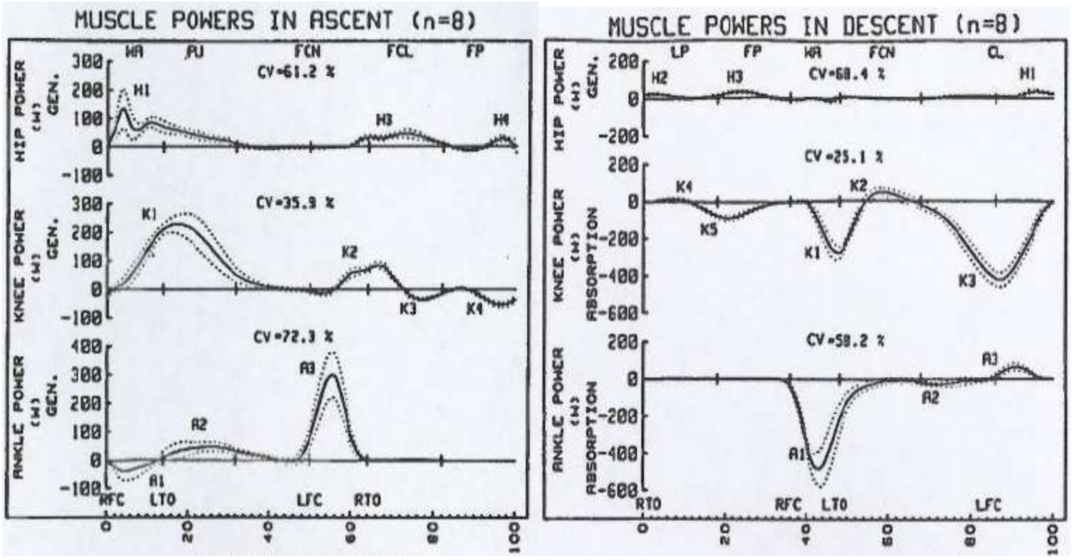


Figure 2.4 Normative lower limb joint powers at the hip, knee and ankle during stair ascent and descent (taken from McFadyen and Winter, 1988).

2.4.4 Biomechanics of Amputee Stair Negotiation

The literature has reported transtibial amputees' performance of stair negotiation with some reference to able-bodied individuals (Powers *et al.,* 1997). Previous studies have reported that transtibial amputees are able to negotiate stairs effectively however they display mechanical adaptations similar to those reported for level gait (McFadyen and Winter, 1988; Powers *et al.,* 1997; Jones *et al.,* 2006; Schmalz *et al.,* 2007; Alimusaj *et al.,* 2009). Amputees tend to negotiate stairs more slowly than able-bodied individuals, and display asymmetry in temporal-spatial variables, spending more time in stance on the intact limb compared to the affected limb (Powers *et al.,* 1997; Vanicek *et al.,* 2007).

During stair ascent, increased external hip moments on the affected side are generated during stance phase along with increased forward trunk lean in order to progress and elevate the body. At the same time, the affected limb knee joint is kept in a relatively extended position to provide stability (Powers et al., 1997; Schmalz *et al.,* 2007; Alimusaj *et al.*, 2009). The lack of active plantarflexion in the prosthetic ankle results in a less elevated centre of mass in preparation for intact limb stance phase, thus the intact limb displays increased knee flexion as a compensatory mechanism (Alimusaj *et al.*, 2009). Also, decreased clearance of the prosthetic foot during swing phase, owing to a lack of active dorsiflexion at the ankle, results in increased plantarflexion of the intact ankle (Alimusaj *et al.*, 2009).

Stair descent has been characterised in transtibial amputees by two strategies each specific to a particular limb. On the affected limb, amputees have been reported to maintain the centre of mass over the extended limb at initial contact (Jones *et al.*, 2006; Schmalz *et al.*, 2007). This has been explained as an attempt to keep the vertical GRF vector anterior to the knee joint, thus reducing the knee moment, power absorption and demand on the knee extensor musculature i.e. reduced loading of the limb (Jones *et al.*, 2006; Schmalz *et al.*, 2007; Alimusaj *et al.*, 2009). These explanations are supported by the finding of reduced vertical GRF produced by the affected limb (Schmalz *et al.*, 2007). The intact limb is characterised by a falling pattern, where the foot contacts the ground with a plantarflexed ankle due to the lack of dorsiflexion in the prosthetic ankle, resulting in greater vertical GRF (Schmalz *et al.*, 2007).

2.4.5 <u>Obstacle Crossing</u>

The negotiation of obstacles encountered during gait is an important skill required in order to avoid a trip or fall. This has led to much research on obstacle crossing focussing on high risk populations, such as the elderly, who are more likely to experience a trip or fall (Chou *et al.*, 2003; Hahn and Chou, 2004; Lowery *et al.*, 2007). The understanding of how able-bodied individuals successfully negotiate obstacles is important as it allows for direct comparison to analyses of obstacle crossing in populations that are more at risk of falling.

When compared to level gait, obstacle crossing results in a slight increase in stride length, a large increase in lead limb swing time and a reduction in double limb support time (Patla and Rietdyk, 1993). Although it is obvious that toe clearance when crossing the obstacle must be increased, literature has described two kinematic strategies of obstacle crossing that contribute to this in able-bodied individuals namely, an upward bias of limb trajectory and increased limb flexion (Patla *et al.,* 1991; Patla and Rietdyk, 1993). These two strategies are employed in unison in order to aid clearance of the obstacle, although limb flexion is the dominant contributor (Patla and Rietdyk, 1993; Chou and Draganich, 1997). At toe off, the lead limb, the first limb to cross the obstacle, is slightly more flexed than during level gait as a result of increased hip and knee flexion and ankle dorsiflexion (Patla and Rietdyk, 1993). As the lead limb toe progresses towards the obstacle, the lower limb continues to flex until it reaches a point directly above the obstacle (Figure 2.5). At this point, hip and knee flexion are increased by approximately 20º and 35º respectively, when compared to level gait, although ankle dorsiflexion is similar to that observed in level gait (Figure 2.3) (Patla and Rietdyk, 1993). A functional straightening of the limb via knee extension and ankle plantarflexion following obstacle clearance is observed as the lead limb prepares for foot contact (Austin *et al.,* 1999). The lead limb motion described above has been reported to be affected by both obstacle height and width and age (Patla and Rietdyk, 1993; Austin *et al.*, 1999; Lu et al., 2006).

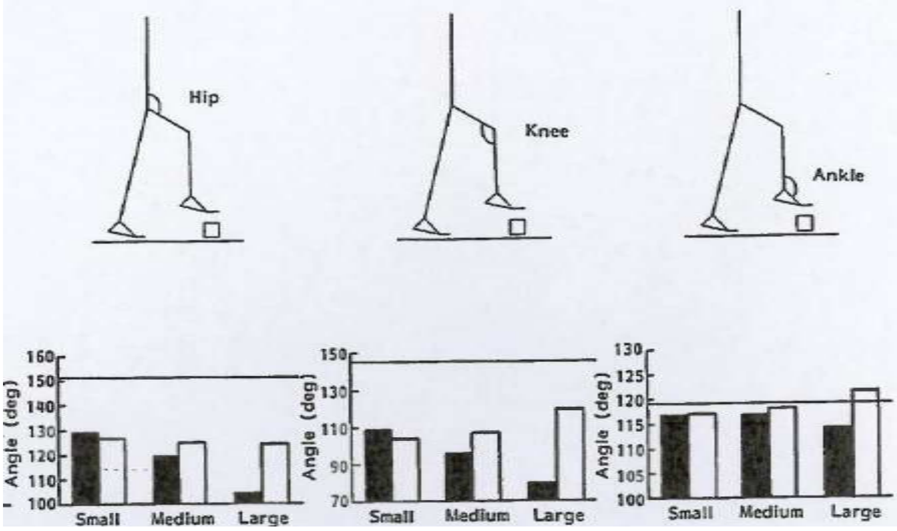


Figure 2.5 Representation of lead limb kinematics and observed values during obstacle crossing of different heights. (taken from Patla and Rietdyk, 1993).

Analysis of GRFs has reported that the trail limb, the standing limb as the lead limb crosses the obstacle, acts to slow the body's COM during obstacle crossing (Patla and Rietdyk, 1993). Both the anterior-posterior and vertical GRF and impulses were decreased during obstacle crossing when compared to level gait. In addition, increased sagittal plane joint moments were observed at the knee and hip of the trailing limb during stance phase, suggesting that increased muscular effort was required as the lead limb prepared for the swing phase (Chen and Lu, 2006). The lead limb has been reported to display reduced vertical GRF when compared to level gait, suggesting that there is a controlled lowering of the lead limb mass once it has made contact the ground following obstacle crossing (Chen and Lu, 2006).

2.4.6 Biomechanics of Amputee Obstacle Crossing

Amputees may encounter a variety of obstacles on a regular basis and as such, crossing obstacles safely is an ADL that warrants investigation to ensure safe ambulation in all environments. Previous studies have also investigated how lower limb amputees avoid or negotiate obstacles, with studies reporting that transtibial amputees were also able to negotiate obstacles of differing heights effectively (Hill *et al.,* 1997; Hill *et al.,* 1999; Hofstad *et al.,* 2006; Hofstad *et al.,* 2009; Vrieling *et al.,* 2007; Vrieling *et al.,* 2009). However, when compared to able-bodied individuals, transtibial amputees negotiated obstacles more slowly than able-bodied individuals (Vrieling *et al.,* 2007) and were less able to negotiate unexpected obstacles, especially under increasing time pressure (Hofstad *et al.,* 2006; Hofstad *et al.,* 2009). However, this effect has been seen to significantly diminish with time since amputation and perhaps, subsequent practice effects (Hofstad *et al.,* 2006). In order to negotiate an obstacle, one limb must be elevated and placed in an advanced position, thus becoming the lead limb. One study reported that transtibial amputees had no lead limb preference during obstacle crossing (Hill *et al.,* 1997) whereas more recently, an affected limb lead preference has been reported (Vrieling *et al.,* 2007) These studies highlight a lack of a clear consensus within the published literature, perhaps due to individual preferences between amputees Anecdotal reports from physiotherapists specialising in the rehabilitation of lower limb amputees in the UK report that, generally, transtibial amputees are taught to cross obstacles leading with their ‘strongest limb’ which is usually their intact limb. This may help explain the inconsistent reports of lead limb preference. When leading with the affected limb there was an increase in knee and hip flexion as a function of obstacle height compared to able-bodied individuals (Hill *et al.,* 1997). There was also an increase in intact trail limb ankle plantarflexion which had been described as a compensatory mechanism employed in order to aid toe clearance of the affected lead limb (Hill *et al.,* 1997). Some studies suggested that leading with the affected limb benefits the amputee in terms of visual feedback (Hill *et al.,* 1997; Vrieling *et al.,* 2007) and increased time to prepare the limb for stance phase (Vrieling *et al.,* 2007). However, this strategy may have been selected by amputees with reduced knee joint ROM due to the posterior shell of the prosthesis, precluding it from being a suitable trail limb (Hill *et* *al.*, 1997). This observation was not reported in a more recent study, suggesting that these amputees were capable of increased knee flexion and thus negating the need for increased intact trail limb plantarflexion (Vrieling *et al.*, 2007). A possible cause of reduced knee ROM on the affected limb has been attributed to the posterior shell of prosthesis and socket fit. This may render the affected limb an ineffective trail limb, meaning rotational work about the hip must be modulated as obstacle height increases (Hill *et al.*, 1997; Hill *et al.*, 1999). It has also been suggested that reduced knee ROM in the affected limb reflects instability in the knee musculature in preparation for the subsequent stance phase or an inability to effectively control musculature about the knee (Hill *et al.*, 1999; Hofstad *et al.*, 2006). Also, the leading limb is required to 'push off' at the end of the preceding stance phase. This propulsion is reduced in the affected limb, despite more advanced prosthetic ankle design (Hill *et al.*, 1999). Following the investigation of obstacle crossing in transtibial amputees during rehabilitation, these authors have made suggestions on how to further improve the performance of this task (Vrieling *et al.*, 2009). They found that during the course of rehabilitation, both walking velocity increased and swing phase kinematics were improved in terms of increased hip and knee flexion during swing phase (Vrieling *et al.*, 2009). The literature suggests that obstacle crossing in transtibial amputees is a more 'conscious' act than in able-bodied individuals (Hofstad *et al.*, 2009). Therefore, early introduction of more complex daily tasks (such as obstacle crossing) during rehabilitation and practicing knee flexion on the affected limb during such complex tasks, along with innovations in prosthetic design may improve amputees' ability to perform these tasks more effectively (Vrieling *et al.*, 2007; Vrieling *et al.*, 2009; Hofstad *et al.*, 2009).

2.4.6.1 Limitations to Biomechanics of Activities of Daily Living Literature

The literature investigating the performance of ADLs in transtibial amputees has shown that with practice they are able to complete these tasks, albeit with altered mechanical functioning. However, similar to studies on amputee gait, very few have focussed upon the longitudinal changes that occur in these individuals, with only one study investigating the biomechanics during rehabilitation (Vrieling *et al.,* 2009). Studies have also reported variable patient characteristics reducing the comparability of findings between studies. Conflicting reports on how amputees are able to negotiate obstacles, as well as the lack of investigation of 'stepping gait', indicates that further investigation into these tasks and how amputees learn to perform them would be beneficial to this body of literature and amputee physiotherapists.

2.4.7 Balance and Postural Control

Along with performing gait and various ADLs, transtibial amputees must learn how to control posture in order to prevent the loss of balance and a subsequent fall because of a postural disturbance. Transtibial amputees are at a disadvantage in terms of balance ability due to the loss of somatosensory input and musculoskeletal receptors in the lower limb that help in the maintenance of postural control.

Able-bodied individuals maintain balance by keeping the body's centre of pressure (COP) within the base of support (Vanicek *et al.,* 2009b; Horak *et al.,* 1989). Postural control has been reported to rely upon an individual's ability to correctly predict, detect and encode the characteristics of passive and dynamic disturbances to posture (Horak *et al.,* 1989). Proactive or reactive adjustments in able-bodied individuals are characterised by well coordinated motor patterns that take place in order to adjust the position of the body's centre of mass (COM) (Winter, 1995). This maintenance of balance and posture is achieved using three main sources of sensory feedback; somatosensory, visual and vestibular (Winter, 1995). This sensory information, in addition to previous experience, allows the body to detect any changes in the position of the COG and correct them if necessary (Horak *et al.,* 1989). Lower limb amputation has obvious effects on the functioning of the human balance system by directly altering the somatosensory feedback available to the individual.

2.4.8 Balance and Postural Control in Lower Limb Amputees

Studies investigating balance and postural control of lower limb amputees have revealed that this patient group has poorer performance when compared to able-bodied individuals (Isakov *et al.,* 1992; Isakov *et al.,* 1994; Buckley *et al.,* 2002; Vrieling *et al.,* 2008). Amputees use their intact limb as the primary means of control during static and dynamic tasks and due to the loss of somatosensory information in the affected limb, rely heavily on visual control to modulate balance and posture (Isakov *et al.,* 1992; Vanicek *et al.,* 2009b).

Studies that investigated the differences in postural sway between transtibial amputees and able-bodied individuals reported contradictory results of COP excursion (Dornan *et al.,* 1978; Vittas *et al.,* 1986; Isakov *et al.,* 1992; Isakov *et al.,* 1993; Hermodsson *et al.,* 1994; Aruin *et al.,* 1997). Some studies have reported no difference in static sway between amputee and able-bodied individuals (Dornan *et al.,* 1978; Vittas *et al.,* 1986) while other more recent studies reported poorer amputee performance, especially in vascular amputees (Isakov *et al.,* 1992; Hermodsson *et al.,* 1994; Buckley *et al.,* 2002). Many of these studies used a single force plate in order to analyse the COP trajectory and thus inferring sway. This may be problematic as it masks any compensatory mechanisms adopted by the amputees in either the intact or affected limb. Studies that have employed dual force plate instrumentation have found differences in sway, as well as weight-bearing, between the intact and affected limbs (Isakov *et al.,* 1992; Isakov *et al.,* 1994; Vrieling *et al.,* 2008). Studies investigating sway activity and changes in COP anterior-posterior excursion reported increased postural sway in amputees' affected limb when compared to the intact limb and to able-bodied individuals (Isakov *et al.,* 1994). Postural sway has been reported to reduce as a function of time across rehabilitation (Isakov *et al.,* 1992). The dependence on the intact limb during static posture has further been highlighted in amputees whilst dual tasking (Aruin *et al.,* 1997). Some studies reported that increased EMG activity on the intact limb was linked to a lateral shift of the COP towards the intact limb during standing thus placing greater demands on the intact limb musculature (Isakov *et al.,* 1994; Aruin *et al.,* 1997). Other studies have also noted the importance of visual input in the control of balance and postural stability, with reference to amputees increased reliance on this source of information during static (Isakov *et al.,* 1992) and dynamic conditions (Vanicek *et al.,* 2009b) perhaps due to the loss of somatosensory input from the affected limb (Vanicek *et al.,* 2009b). While static sway was useful in establishing the body of literature pertaining to amputee balance and postural control, the tasks employed lack ecological validity as they do not mimic real life situations closely enough. When maintaining balance, a combination of strategies are employed in order to avoid falling. When small perturbations are experienced, the ankle plantarflexors and dorsiflexors contract to control the model of balance represented as an inverted pendulum (Winter, 1995). This is known as the 'ankle strategy' (Winter, 1995). However, during larger perturbations, weakness or absence of ankle plantarflexors, necessitates the need for movements at the hip to maintain balance (Winter, 1995). Hip flexion and extension would shift the COM anteriorly and posteriorly respectively, and this is known as the 'hip strategy' (Winter, 1995). If the perturbation to balance is large enough, individuals may also be required to take a step in order to maintain balance by altering or increasing the base of support, thus maintaining the COG within its limits (Horak *et al.,* 1989). Buckley *et al.* (2002) suggested that static assessment of postural ability does not assess how participants utilise the ‘hip strategy’ when responding to larger dynamic perturbations. With this in mind, the literature has moved towards the assessment of dynamic balance and postural control (Buckley *et al.,* 2002; Vrieling *et al.,* 2008; Vanicek *et al.,* 2009b), with some studies using more advanced technological methodologies to tease out the various aspects of amputee balance performance and postural control (Vrieling *et al.,* 2008; Vanicek *et al.,* 2009b). Buckley *et al.,* (2002) employed a single force plate and custom stabilimeter methodology to assess postural control when the support surface could rotate about a single axis in either the sagittal or frontal plane. This study reported that lower limb amputees displayed poorer dynamic balance than able-bodied individuals in both axial rotations and that when vision was occluded, the lower limb amputees tended to tilt towards their affected limb in the medio-lateral direction (Buckley *et al.,* 2002). It was suggested that this effect, along with the observation of increased board-floor contact time, was an attempt by amputees to gain extra somatosensory input from the affected limb residuum (Buckley *et al.,* 2002). It was also interesting to note that the lower limb amputees in this study were highly active. Thus, the reduced static and dynamic postural control observed could be attributed to amputation and not the reduced joint mobility or muscle weakness associated with ageing or inactivity, as may be the case in vascular amputees (Buckley *et al.,* 2002). Further investigations into amputee responses to dynamic perturbations have found that the weight-bearing asymmetry reported during static posture (Isakov *et al.,* 1992; Isakov *et al.,* 1994) increased with the addition of a secondary dynamic task (Vrieling *et al.,* 2008). Another study noted an increased intact limb anterior-posterior GRF and COP excursion compared to able-bodied individuals, relating to previous reports of amputees using the intact limb as a stabilising method (Buckley *et al.,* 2002; Vrieling *et al.,* 2008). Although there was noted asymmetry in anterior-posterior GRF and COP excursion between the intact and affected limbs, the values observed for the affected limb were higher than in able-bodied individuals (Vrieling *et al.,* 2008). Similar to Buckley *et al.,* (2002) Vrieling and colleagues (2008) interpreted this effect an attempt by the amputees to gain extra somatosensory input from their affected limb.

2.4.8.1 Computerised Dynamic Posturography

Computerised dynamic posturography (CDP) is a method of assessing postural sway and balance performance during dynamic task conditions (Monsell *et al.,* 1997). Typically, strain gauge or force plate instrumentation, incorporating a means of unexpectedly perturbing the support surface, is employed along with methods of altering or isolating the somatosensory and/or visual information available to the participant (Monsell *et al.,* 1997). The Neurocom Equitest® (NeuroCom International , Inc, Clackamas, US) is one instrument that employs CDP protocols in order to assess postural control and balance function and is described in detail in Chapter Three, Section 3.5.

A study using CDP has reported the aforementioned visual dependence of transtibial amputees during static balance and whilst maintaining posture during dynamic perturbations (Vanicek *et al.,* 2009b). This study also reported the use of the ankle strategy during easier tasks and more reliance upon the hip strategy as task difficulty increased, supporting the previous suggestions for the use of dynamic assessment in this patient group (Buckley *et al.,* 2002).

Computerised dynamic posturography has also highlighted the differences in balance ability and postural control between fallers and non-fallers in transtibial amputees and able-bodied individuals (Vanicek *et al.,* 2009b). Amputee fallers reportedly relied more upon the use of the affected limb, further supporting suggestions that the intact limb plays an important role in successful balance ability (Vanicek *et al.,* 2009b). The inability to maintain balance can lead to falling and it has been reported that lower limb amputees have a higher fall rate when compared to age-matched able-bodied individuals

(Miller *et al.*, 2001a). These results re-iterate the importance of not only better understanding the way amputees achieve balance and postural stability, but also the process by which they do this following their rehabilitation. Understanding this learning process may highlight areas of amputee balance that would benefit from further clinical intervention during and after rehabilitation. This may help clinicians and health care professionals to reduce the aforementioned increased falls rate in lower limb amputees, with a potentially significant reduction in cost to the National Healthcare Service.

2.4.8.2 Limitations to Lower Limb Amputee Balance and Postural Control Literature

Studies investigating balance and postural control have provided a clear picture as to how amputees perform these tasks. However, this body of literature shares the same limitations in patient characteristics mentioned previously. Despite an early study reporting that cause of amputation should be accounted for due to differing postural characteristics (Hermodsson *et al.*, 1994), following studies have tended to test amputees secondary to a variety of causes and with varying levels of amputation. This is likely to mask the deficits in balance and postural control associated with the neurological and musculoskeletal changes apparent with different causes and levels of amputation.

Although the balance and postural control tasks employed within the literature are well validated and have a solid rationale for their use within each discrete experiment, varying methodologies make it difficult to directly compare results between studies. A degree of standardisation in testing protocols may help overcome the issue of comparability. The cross-sectional nature of many studies does not reveal how the mechanisms of maintaining balance and postural control is established in amputees, despite previously reported adaptations during rehabilitation (Isakov *et al.*, 1992). Therefore, it is important for future research to focus upon the longitudinal adaptations that occur in balance ability and postural control within this patient group. This process would aid those involved in the care and rehabilitation of amputees in developing more effective interventions targeted at improving balance.

2.5 Quality of Life in Lower Limb Amputees

Up to this point, the review of literature has focused upon the biomechanical, balance performance and postural control related aspects of the transtibial amputee. However, a transtibial amputee presents a multifaceted case, only part of which can be investigated and explained by the analysis of movement patterns. Psychological factors such as how the amputee feels about their amputation and prostheses are also important factors as general health is comprised of both physical and mental health (Ware and Gandek, 1998). One such factor that has received significant attention in the health literature is the issue of QOL. It has been suggested that in order to provide a complete assessment of the benefits of an intervention, evidence of its impact upon health related QOL must be reported (Garratt *et al.*, 2002). Despite this, health related QOL in lower limb amputees has received little attention, especially longitudinal changes during the rehabilitation process (Asano *et al.*, 2008).

Studies that have assessed QOL in lower limb amputees have used a number of instruments including the World Health Organisation Quality of Life Scale (WHOQOL) (The WHOQOL Group, 1994b) and the Medical Outcome Study Short Form-36 Questionnaire (SF-36) (Ware and Gandek, 1998). An amputee specific questionnaire, namely the Prosthesis Evaluation Questionnaire, has also been developed in order to assess prosthesis related QOL (Legro *et al.*, 1999).

There have been variable reports of QOL in lower limb amputees. It has been reported to be both equal to or higher than (Asano *et al.*, 2008; Zidarov *et al.*, 2009) as well as lower (Legro *et al.*, 1999; Pezzin *et al.*, 2000) than that reported from so called normative disease-free populations. Further to this, studies have reported that lower limb amputees tended to have better mental health compared to physical health (Legro *et al.*, 1999; Pezzin *et al.*, 2000; Van der Schans *et al.*, 2002; Asano *et al.*, 2008; Zidarov *et al.*, 2009). Factors affecting psychological health include depression, which has been reported as an important predictor of QOL (Asano *et al.*, 2008), as well as the aesthetics of the prosthesis (Legro *et al.*, 1999; Gallagher and MacLachlan, 2004). Studies employing these self-report measures have reported QOL to be highly related to both physical (Legro *et al.*, 1999) and social (Deans *et al.*, 2008) aspects of an amputee's life, as well as being closely related to the functioning of their prosthesis (Legro *et al.*, 1999; Zidarov *et al.*, 2009). Although psychological health is reported to be better than physical health in lower limb amputees, studies have reported physical health to be more closely related to overall QOL (Gallagher and MacLachlan, 2004).

Fear of falling, rather than the event of an actual fall, has been linked to reduced QOL in lower limb amputees (Miller *et al.*, 2001a,b). This finding has been attributed to lower limb amputees' expectation to fall due to their physical constraints or falling whilst attempting tasks of ever increasing difficulty (Miller *et al.*, 2001a). Despite this, no amputee specific measure of falls efficacy has been developed. However, the modified falls efficacy scale (mFES) (Hill *et al.*, 1996) has been used to assess falls efficacy within the elderly population (Delbaere *et al.*, 2009), the effect of falls efficacy on elderly gait (Chamberlin *et al.*, 2005) and improvement in fall rates via training (Vrantsidis *et al.*, 2009).

Although fear of falling is detrimental to QOL and has been seen to increase as a function of age, QOL in lower limb amputees has been reported to marginally increase with time since amputation (Asano *et al.*, 2008). Although it could be assumed that the physical gains attained through increasing prosthetic use over time would be mirrored by a greater QOL, this effect has not been observed in the literature. Some authors explained this effect through the so-called *response phenomena*, theorising that as lower limb amputees adjust their expectations over time, they converge with the reported QOL

(Zidarov *et al.,* 2009). Thus, as physical ability and subsequent psychological health improves with time, lower limb amputees' expectations are raised, which has an influence on the reported QOL. Studies assessing QOL during lower limb amputee rehabilitation suggested that this reflects an increase in QOL when compared to baseline (Zidarov *et al.,* 2009).

2.5.1.1 <u>Limitations to Quality of Life in Lower Limb Amputee Literature</u>

Studies in lower limb amputees have begun to highlight the negative effects associated with amputation on QOL (Miller *et al.,* 2001b; Asano *et al.,* 2008). Lower limb amputees tend to have better mental health than physical health, explained in the literature by the alleviation of lower limb pain pre-surgery or happiness at having survived a traumatic event (Zidarov *et al.,* 2009). However, these studies suffer from inherent inconsistency in their reports due to a number of factors, including but not limited to the use of varied self-report scales (WHOQOL, SF-36 and PEQ), causes of and time since amputation, patient numbers and low response rates. The cross-sectional design of many studies does not highlight the changes in QOL as lower limb amputees' physical ability and expectations change. Only one study has investigated QOL during rehabilitation, which is surprising given that some studies suggest a holistic approach to rehabilitation and the importance of assessing psychological health during this time period (Asano *et al.,* 2008; Zidarov *et al.,* 2009).

Along with physical adjustments, studies have reported psychological differences in lower limb amputees when compared to able-bodied individuals, specifically with regards to QOL. Although these studies have contributed significantly to our understanding of transtibial amputees, many of them do not explain how the variables on which they report were established over time. Many studies do not explain the longitudinal psychological adaptations that occur in new transtibial amputees as they learn to adjust both mentally and physically to the experience of a lower limb amputation.

2.6 Rehabilitation and Longitudinal Change in the Transtibial Amputee

Following the experience of transtibial amputation, the patient must go through a process of rehabilitation whereby they attempt to regain and re-learn the ability to complete various day-to-day tasks. Professional guidelines have been provided by the Chartered Society of Physiotherapists (www.csp.org.uk) and are authored by an interest group made up of multi-regional senior physiotherapists involved in lower limb amputee rehabilitation across the UK (Broomhead *et al.*, 2006). This interest group is called the British Association of Chartered Physiotherapists in Amputee Rehabilitation (BACPAR) (www.bacpar.org.uk). These evidence-based guidelines provide information on the pre- and post-operative physiotherapy management of lower limb amputees including, the use of various equipment and exercises.

A published text on lower limb amputee rehabilitation, designed as a handbook for both experienced and student physiotherapists, mirrors many of the recommendations presented in the BACPAR guidelines (Engstrom and Van de Ven, 1999). This text provides a more hands-on resource to amputee rehabilitation with various illustrations of recommended exercises and treatments protocols relating to ADL. Information is also provided relating to increased functioning of lower limb amputees such as car and motorcycle transport and various sporting activities. Details of prosthetic design and function are also outlined.

In the UK, lower limb amputees typically follow an individualised programme of rehabilitation prescribed by the physiotherapists and multidisciplinary team involved, based upon their experience and knowledge. This programme will differ between centres but may involve a pre-operative discussion of what the patient can expect from their rehabilitation and a meet and greet with a fellow amputee. A pre-operative meeting also allows the physiotherapist to identify any gait abnormalities already present and discuss the patient's aspirations following surgery. Patients are visited as inpatients following surgery, where they are advised on skills such as transfers, bed movements and crutch or wheelchair mobility. If appropriate, generic prosthetic device and residuum shrinker i.e. tight bandage, use commences, no less than five days post-surgery. Once discharged from inpatient care, outpatient rehabilitation may include the practice of simple tasks such as donning of prostheses and weight-bearing progressing onto tasks that aim to improve balance, core stability and the use of walking aids. Depending upon patient ability and inclination, further rehabilitation sessions may involve more complex tasks such as graded walking, walking of varying terrains, stair climbing and running. There is an ongoing assessment of the patient who may re-visit the rehabilitation team in order to learn or develop new skills. In addition to this, information such as the guidelines outlined above are followed to inform the rehabilitation procedure.

2.6.1 Early Walking Aids

Along with the aforementioned techniques and procedures, both sets of physiotherapy guidelines advocate the use of early walking aids (EWAs). Early walking aids are commonly found in UK physiotherapy departments involved with transtibial amputee rehabilitation. While re-learning how to walk with their new mechanical constraints, transtibial amputees often use EWAs during rehabilitation within a physiotherapy environment as an initial gait re-education and weight-bearing tool, prior to casting for a functional definitive prosthesis (Scott *et al.*, 2000). Early walking aids have a number of reported benefits: they have been used as early as one week post-operatively (Dickstein *et al.*, 1982) and have been shown to reduce the deterioration in physical ability (Redhead *et al.*, 1978). When utilised correctly EWAs have also been shown to reduce post-operative oedema, accelerate the healing and maturation of the residual stump

(Redhead *et al.,* 1978; Dickstein *et al.,* 1982; de Noordhout and de Brogniez, 2004) whilst reducing time in hospital and time from surgery to casting for definitive prosthesis (Scott *et al.,* 2000). Early walking aid use has also been reported to provide patients with improved psychological functioning (Engstrom and Van de Ven, 1999) and more desirable cosmetic appearance (Dickstein *et al.,* 1989). There are also economic benefits associated with EWAs in terms of both reducing therapy cost (Dickstein *et al.,* 1982) and their use as a substitute for a dedicated prosthesis (Redhead *et al.,* 1978).

There are a variety of EWAs available to lower limb amputees worldwide. Two are commonly found in physiotherapy departments in the UK and used in the rehabilitation of transtibial amputees: (A) the Pneumatic Post-Amputation Aid (PPAM Aid) (Ortho Europe Ltd, Alton, UK) and (B) the Amputee Mobility Aid (AMA) (Ortho Europe Ltd, Alton, UK). The PPAM aid is an EWA designed for use seven to ten days post surgery and is a partial weight-bearing device encompassing two pneumatic bags within a rigid frame with a rocker foot at the distal end (Scott *et al.,* 2000). The AMA, developed in 1993 after the PPAM aid's introduction, was designed to allow the biological knee to articulate freely, allowing patients to practice a more natural gait with knee flexion and extension possible in the affected limb (Scott *et al.,* 2000). Functionally, this articulation is the only difference between the two EWAs although they differ in their aesthetics and how they are donned (Scott *et al.,* 2000). Full details of these EWAs are provided in Chapter Three, Section 3.2.4. Although the use of EWAs in transtibial amputee rehabilitation has documented benefits, there has been little evidence provided as to whether one EWA is more beneficial than another.

One study employing a cross-over design endeavoured to investigate the differences in joint kinematics using electrogoniometry and stump interface pressures between PPAM Aid and AMA use (Scott *et al.,* 2000). Two groups of transtibial amputees were recruited, one group using the AMA for two weeks followed by PPAM Aid use for two weeks, the other group using the EWAs in the opposite order. When compared to the PPAM Aid, stump interface pressures observed in the AMA were found to be increased during static standing but not significantly different during walking (Scott *et al.*, 2000). In terms of joint kinematics, goniometry did not highlight any differences between the affected limb biological knee joint movement and the AMA prosthetic peak knee flexion/extension. However, there were highly variable peak knee flexion values in the affected limb between the groups, though the study did not explain at which point these peak values occurred in the gait cycle (Scott *et al.*, 2000). An explanation for the differences in stump interface pressure observations were partially accounted for by proposed measurement error (Scott *et al.*, 2000). The study reported that the AMA's lower stump socket interface surface area may have resulted in increased pressure rather than greater weight-bearing (Scott *et al.*, 2000). Interestingly, pressure (inferring weight-bearing) did not increase with time from surgery. Kinetic analysis would have supplemented these findings to provide a clearer picture of partial weight-bearing ability. The lack of differences in the affected limb knee joint kinematics and the articulated knee mechanism of the AMA suggested that these joints functioned synchronously as a single entity (Scott *et al.*, 2000). However, some of the methods employed by Scott *et al.* (2000) had limitations. The crossover design employed may have masked any learning effects associated with a particular EWA. The lack of an inter-limb comparison was also a limitation as important adaptations may have also occurred in the movement of the intact limb. This study helped us to gain insight into the clinical aspect of amputee gait and EWAs. However, from this study alone it remains unclear whether the use of one EWA is better in terms of gait re-education than the other in transtibial amputee rehabilitation.

2.6.2 Investigation of Lower Limb Rehabilitation

Currently, the choice of EWA used for any individual during a rehabilitation programme is the decision of the relevant clinical specialist and is not evidence-based. The scientific literature has investigated various aspects of the lower limb amputee rehabilitation process. Firstly, reports have provided review-type information, similar to the BACPAR professional guidelines, with a focus on empirical literature (Esquenazi and DiGiacomo, 2001; Esquenazi and Maier, 1996). These reports outline nine stages of rehabilitation after amputation, starting with pre-operative treatment and finishing with long term follow-up and include various strength, cardiovascular, balance and prosthetic mobility training exercises (Esquenazi and DiGiacomo, 2001). Other studies focused upon the rehabilitation of the older amputee (Cutson *et al.*, 1994; Cutson and Bongiorni, 1996; Fletcher *et al.*, 2001) secondary to lower limb dysvascularity (Cutson *et al.*, 1994; Bailey and MacWhannell, 1997; Fletcher *et al.*, 2001). One study highlighted the benefits of early rehabilitation in this patient group in terms of time from surgery to prosthetic gait training, reporting reduced time from surgery to receipt of prosthesis in the early rehabilitation group (Cutson *et al.*, 1994). However, the methods by which this was achieved may have been specific to that particular clinic (Cutson *et al.*, 1994). One study focused upon the cardiac monitoring of this group, reporting that whilst exercise stress during early gait re-training was within acceptable limits, therapists should monitor amputees' ECG and heart rate during exercise to increase patient safety (Bailey and MacWhannell, 1997). Another study assessing prosthetic fitting rates reported that placing a foam rubber insert to the distal end of the patients' socket during gait training increased wound healing and stump maturation (Hallam and Jull, 1988). Interruptions in treatment and their impact on rehabilitation have been monitored with 30% of patients having rehabilitation interrupted for reasons such as stump healing (18%), acute medical illness (10%) and other causes (2%) (Meikle *et al.*, 2002). An increased incidence of interruptions was more common among women, those with vascular causes of amputation and reduced days between amputation and rehabilitation, although 79% of patients with interruptions went on to complete rehabilitation (Meikle *et al.,* 2002).

Few studies have provided quantitative biomechanical information about how transtibial amputees progress through rehabilitation. Factors such as the efficacy of falls interventions, stump injuries (Gooday and Hunter, 2004), effects of prosthetic intervention (Hallam and Jull, 1988) weight-bearing, pain, walking velocity (Jones *et al.,* 2001) and self-report scales of functional ability (Panesar *et al.,* 2001) have all received attention. However, these variables do not all directly relate to transtibial amputee movement adaptations or how they may change as a function of time. Studies have investigated obstacle crossing (Vrieling *et al.,* 2009) and postural sway (Isakov *et al.,* 1992) in lower limb amputees during rehabilitation while another study assessed GRFs in transfemoral amputees during rehabilitation (Gravel *et al.,* 1995). The study by Gravel *et al.* (1995) displayed significant increases in walking velocity and affected limb static weight-bearing along with a significant decrease in intact limb static weight-bearing. However, vertical GRF results during gait were variable, perhaps due to patients walking with the use of parallel bars (Gravel *et al.,* 1995).

Following amputation, transtibial amputees follow a course of rehabilitation from which they are discharged once a satisfactory level of functioning has been achieved as determined by the relevant clinician. After discharge, transtibial amputees will face a range of tasks of ever increasing difficulty as they attempt to continue the process of re-adjustment following amputation. So far, the literature has failed to adequately investigate these two key stages in transtibial amputees' lives and the implications this may have for the rehabilitation of transtibial amputees.

2.6.3 Prosthetic Simulator Gait

Although studies have yet to investigate the process by which transtibial amputee re-learn how to walk and perform ADLs, two studies have investigated prosthetic simulator gait, endeavouring to imitate transtibial (Vanicek *et al.*, 2007) and transfemoral amputee gait (Lemaire *et al.*, 2000). Initially Lemaire *et al.* (2000) set out to provide non-amputee health care practitioners with a real life experience of transfemoral amputee gait thus sensitising clinicians to patient experience of prosthetic gait. A custom built prosthetic simulator allowed able-bodied individuals to walk similarly to a unilateral transfemoral amputee. This study showed that non-amputee participants produced similar gait kinematic and kinetic results to that of experienced transfemoral amputees (Lemaire *et al.*, 2000). The report suggested that in the absence of a lower limb amputation, it was still possible to evaluate how individuals relearned locomotor tasks by using a prosthetic simulator. Although this study suggested that there were similarities in transfemoral amputee and transfemoral amputee prosthetic simulator gait, the process of how individuals achieved these results was not investigated.

Vanicek *et al.* (2007) investigated the kinematic adaptations in gait of able-bodied participants walking with a prosthetic simulator. The prosthetic simulator allowed able-bodied individuals to walk similarly to a unilateral transtibial amputee, without the use of the knee joint on the affected side. In addition, Vanicek *et al.* (2007) also sought to gain insight into the learning processes apparent whilst performing this novel ambulatory task. Lemaire *et al.* (2000) had failed to look at the initial stages of the gait re-education process, by allowing participants to gain a certain level of proficiency in using the prosthetic simulator during warm-up sessions prior to data collection. Vanicek *et al.* (2007) aimed to evaluate how individuals adapted their gait from the very onset of learning a novel ambulatory task. In this case, it was the first use of a prosthetic simulator. Learning a novel ambulatory task with a prosthetic simulator was achieved by monitoring kinematic changes over two visits, one week apart with walking velocity used as an overall descriptor of gait proficiency. Changes occurred in the early stages of performing this novel ambulatory task as walking velocity increased sharply. These effects were retained in the second test period where initial walking velocity was significantly higher than initial walking velocity in the first test period (Vanicek *et al.,* 2007). Learning to walk with altered lower limb mechanics took place early on in the learning process. Vanicek *et al.* (2007) also found that the intact limb played an important role in modulating walking velocity. Increases in overall walking velocity were achieved by increasing step length of the intact limb, not by increasing step length in both limbs as hypothesized. This could reflect the underlying confidence in control of the intact limb previously mentioned.

One limitation of both studies described above was the use of healthy able-bodied individuals to investigate prosthetic simulator gait. Lower limb amputees may have associated psychological health concerns that may impact upon gait functioning, an example being their physiological capacity with relation to lower limb dysvascularity.

However, scientific investigation regarding these time periods is essential as it will provide clinicians and health care professionals involved in transtibial amputee rehabilitation and treatment with evidence–based information, on which to base clinical decision making along with clinical experience.

2.7 Summary and Rationale

The literature has investigated a number of themes relating to lower limb amputees, with each theme giving rise to commonly reported findings. These reports have helped in the understanding of transtibial amputees and the challenges this population face. However, as with any scientific investigation there are various methodological limitations associated with these studies.

The scientific literature has not yet fully investigated the period of time between when an individual undergoes amputation surgery until they reach their physical potential, including a period of rehabilitation. This is understandable given the complex nature of the population in question and the time commitment required for longitudinal study designs. Nevertheless, it is important to understand how amputees adapt to the challenge of rehabilitation, the period of time following rehabilitation and the factors that may influence their progress during these timeframes. It is clear this information would have various clinical implications for the amputee and healthcare service providers.

2.8 Aim and Objectives

The overall aim of this thesis was to investigate the longitudinal changes that occurred within unilateral transtibial amputees from their first treatments following amputation up to six months post-discharge from rehabilitation.

The first objective was the assessment of the gait adaptations that occurred in unilateral transtibial amputees during rehabilitation and the effect of using different EWAs. Although, very few studies had attempted to assess these variables biomechanically, it was hypothesised that (1) during EWA use the AMA group would display a more proficient gait pattern in terms of variables such as walking velocity, when compared to the PPAM aid group as they were using an EWA with a greater functional capacity.

It was also hypothesised that (2) upon receipt of a functional prosthesis, those patients having previously used the AMA would display a greater improvement in gait parameters than those having previously used the PPAM aid as they would have been used to practicing the control of the knee joint in the affected limb. Lastly, it was hypothesised that (3) following the receipt of a functional prosthesis, until discharge from rehabilitation, the differences between patients using either the AMA or PPAM aid would diminish as both groups adapted to their new mechanical constraints.

The second objective was the assessment of changes in QOL in unilateral transtibial amputees during rehabilitation. It was hypothesised that (4) QOL would increase during the course of rehabilitation, specifically the physical health aspect of QOL, as patients mobility increased. This was based on previous findings that QOL increased with time since amputation (Asano *et al.,* 2008). It was also hypothesised that (5) patients using the AMA would display better QOL during rehabilitation as they would be able to practice a more ‘natural’ gait pattern.

The third objective was the assessment of adaptations in gait and ADL during the six-month period following discharge from rehabilitation. The literature has shown that patients with > 1 year experience of prosthetic use are likely to display increased function when compared to recent transtibial amputees.

Therefore, it was hypothesised that during the time period following discharge from rehabilitation, gait proficiency (6) and performance of ADLs such as crossing obstacles (7) and stepping to and from a new level (8) would improve in terms of walking velocity.

The fourth objective related to the assessment of balance function and postural control during the six month period following discharge from rehabilitation. It was hypothesised that (9) balance ability during dynamic perturbations would improve over time. It was hypothesised that (10) amputees would rely more heavily on visual input as shown in previous literature, with this effect diminishing over time (Isakov *et al.,* 1992; Vanicek *et al.,* 2009b). It was also hypothesised that (11) amputees’ utilisation of the hip strategy would decrease over time following discharge from rehabilitation. Lastly, it was hypothesised that (12) amputees’ ability to volitionally explore their theoretical limits of stability would increase over time.

Finally, the last objective of the current thesis was to assess changes in generic and prosthesis specific QOL and falls efficacy. Therefore, it was hypothesised that (13)

QOL would increase following discharge from rehabilitation, specifically the physical health aspect of QOL. It was also hypothesised that (14) mental health would be higher than physical health as has been reported in the literature (Legro *et al.*, 1999; Pezzin *et al.*, 2000; Van der Schans *et al.*, 2002; Asano *et al.*, 2008; Zidarov *et al.*, 2009). Lastly, it was hypothesised that (15) changes in falls efficacy would follow a similar pattern to the hypothesised changes in QOL.

3 CHAPTER THREE – GENERAL METHODS AND MATERIALS

3.1 Introduction

The current chapter presents specific details pertaining to the individuals, equipment and methodologies used. The current chapter also provides, where necessary, the rationale and justification for use of the aforementioned equipment and methodologies with reference to their previous use in the scientific literature. Equipment and methodologies that were specific to a particular study are detailed in subsequent chapters.

3.2 Patients and Participants

Individuals that participated in the current research were all unilateral transtibial amputees recruited from the Vascular Limb Unit, Hull Royal Infirmary, Hull, UK (studies one and two – referred to as patients during rehabilitation) and from the Department of Physiotherapy, Castle Hill Hospital, Cottingham, UK (studies three, four and five – referred to as participants following discharge). Specific patient demographics are detailed in each particular study, as well as details of patient's specific prosthetic components. Prior to taking part in the current research, participants were made aware as to the nature of the studies by participant information sheets (Appendix A – studies one and two, Appendix C – studies three, four and five). Signed informed consent was provided by patients to the vascular surgeon at the decision to amputate (Appendix B - studies one and two) and to the physiotherapist at discharge from rehabilitation (Appendix D - studies three, four and five). When referring to individual limbs, the term affected related to the amputated limb, with intact relating to the unamputated contralateral limb.

3.2.1 Ethical Approval

Ethical approval for all studies was sought through the National Health Services National Research Ethics Service framework. Ethical approval of studies one and two were obtained from the South Humber Research Ethics Committee (reference number: 04/Q1105/31). Ethical approval of studies three, four and five were obtained from the Hull and East Riding Research Ethics Committee (reference number: 08/H1304/10). South Humber and Hull and East Riding Research and Development Departments also granted approval once ethical approval was confirmed, including the award of honorary NHS contracts to researchers associated with each study.

3.2.2 Inclusion/Exclusion Criteria

3.2.2.1 Studies One and Two – During Rehabilitation

Inclusion criteria for studies one and two stipulated that patients were at least 18 years old, had recently experienced unilateral transtibial amputation and were due to attend Hull and East Yorkshire Hospitals NHS Trust for specialist amputee rehabilitation. Patients were also expected to receive, but had not yet received their functional prosthesis. Finally, patients were required to tolerate and use an early walking aid (EWA) and be able to walk a distance of four metres with the assistance of parallel bars under the supervision of a physiotherapist.

Patients were excluded from the studies if they were previously unable to walk due to a medical condition (e.g. rheumatoid arthritis) or had previously experienced major amputation of the contralateral limb. Patients were also excluded if they were not expecting to receive their functional prosthesis or were unable to follow instruction and/or unable to follow a programme of rehabilitation.

3.2.2.2 Studies Three, Four and Five – Post Discharge from Rehabilitation

Inclusion criteria for studies three, four and five stipulated that participants were unilateral transtibial amputees and at least 18 years of age. Participants were required to have completed specialist amputee rehabilitation within the previous four weeks prior to consenting to participate in the studies. Participants were also required to travel to the University of Hull for data collection session. Further inclusion criteria required participants to be able to use their prosthesis without pain or discomfort and complete the following tasks without the use of a walking aid: walk a distance of five metres; step over an obstacle; step onto and from a new level; and stand still for two minute intervals. Suitability of the participant's ability to complete these tasks was assessed by experienced physiotherapists commonly dealing with amputee rehabilitation.

Participants were excluded from studies three, four and five if they had any current musculoskeletal injuries or any cognitive deficits. Participants were also excluded if they were bilateral or transfemoral amputees. Lastly, participants were excluded if they did not use their prosthesis regularly or if they experienced pain or discomfort whilst doing so.

3.2.3 Prosthetic Components

Details of amputee's prosthetic components are provided in each relevant chapter. This section provides a general description of the prosthetic components used and the fitting of these prosthetic components.

3.2.4 Early Walking Aids

Early walking aids (EWA) are generic prosthetic devices used during rehabilitation for the goal of initial gait re-education and partial weight-bearing (Scott *et al.*, 2000). This section outlines details of the two EWAs assessed and their use within rehabilitation.

3.2.4.1 The Amputee Mobility Aid

The Amputee Mobility Aid (AMA) is an EWA that is specifically designed for use within the transtibial amputee population. The AMA consists of a thigh corset, uni-planar knee joint, shin tube or pylon and a solid ankle and foot complex. The patient's residuum is covered by a residuum bag, which is then placed inside the thigh corset. One unique design feature of the AMA is that it allows patients to practice flexion and extension at the knee of the affected limb via an articulated knee joint (Figure 3.1). The AMA allows for different sized thighs and taller individuals via short and standard thigh corsets and varying shin tube lengths respectively. The foot incorporated within the AMA is a solid complex, not allowing for plantar or dorsiflexion at the ankle.

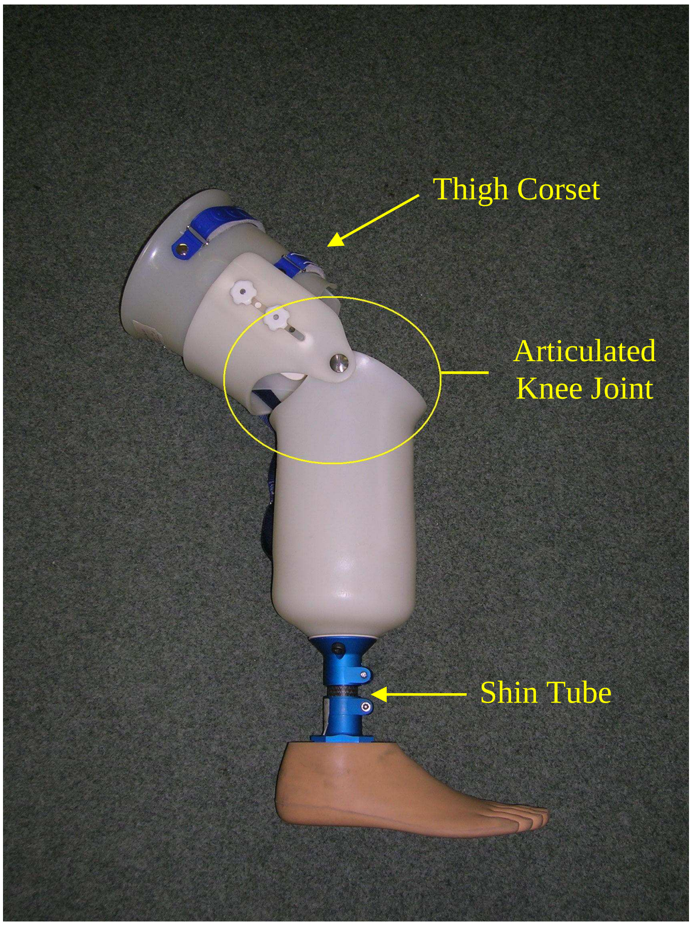


Figure 3.1 The Amputee Mobility Aid. Image used with permission (Ortho Europe Ltd, Alton, UK) (www.ortho-europe.co.uk).

Seated fitting of the AMA (Figure 3.2) initially required measurement of patients' intact limb, groin to knee and knee to floor lengths in order to adjust the thigh corset and to select the correct shin length respectively (A). A residuum bag was placed over the residuum of the amputated limb and then placed into the thigh corset of the AMA (B). The AMA was then donned by the patient (C), thigh support straps were tightened (D), followed by inflation of the residuum bag to a pressure of 40mmHg (E). The patient began partial weight-bearing between parallel bars and any adjustments could be made (F). The AMA length was adjusted by matching the thigh corset and shin length to the length of the intact limb. In both cases, fitting of the AMA and prosthetic limb length was determined by highly experienced physiotherapists prior to data collection.

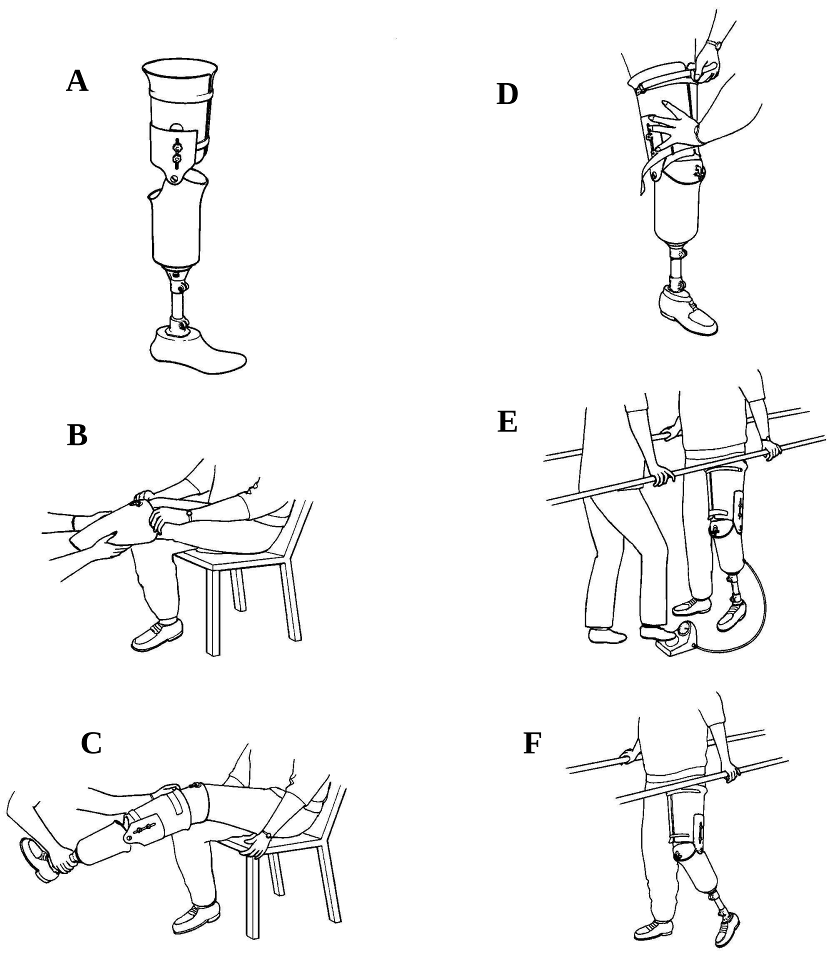


Figure 3.2 Schematic representation of the fitting procedure for the Amputee Mobility Aid. (Ortho Europe Ltd, Alton, UK) (www.ortho-europe.co.uk).

3.2.4.2 The Pneumatic Post-Amputation Aid

The Pneumatic Post-Amputation Aid (PPAM Aid) is an EWA that is designed for use within both the transtibial and transfemoral amputee population (Figure 3.3). The PPAM aid is a rigid frame structure that does not articulate at the knee or ankle. The foot is represented by a convex rocker complex at the distal end of the device. Similar to the AMA, patients' residuum were placed into an inflatable pneumatic residuum bag before being secured into the device, via the crucible strap, ready for use. The PPAM aid is adjustable for patients of different heights and an above-knee residuum bag is also available for use with transfemoral amputees (Figure 3.3).

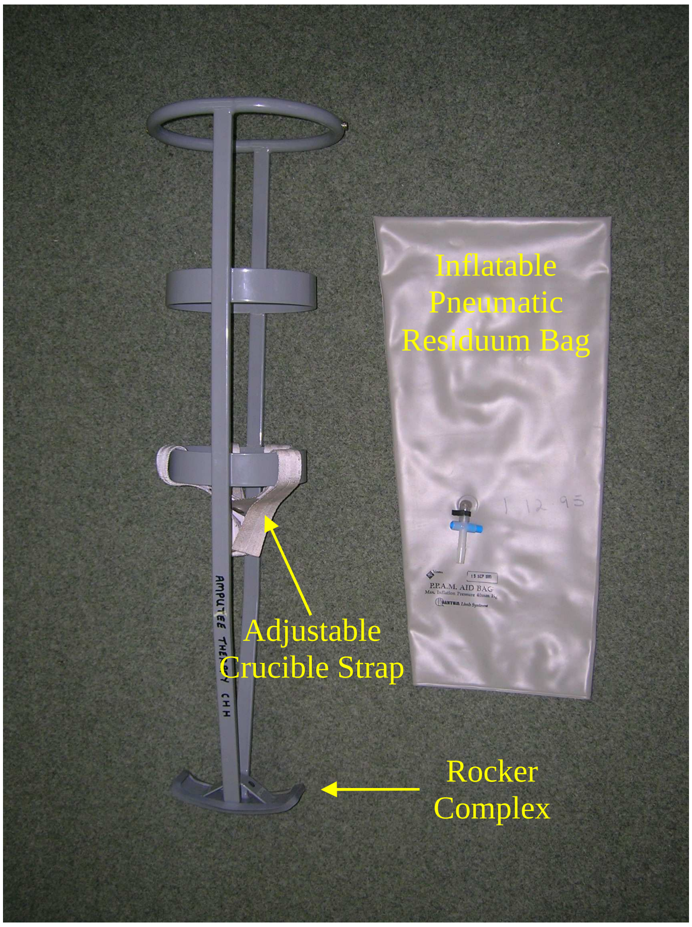


Figure 3.3 The Pneumatic Post-Amputation Aid with inflatable pneumatic residuum bag. Image used with permission (Ortho Europe, Alton, UK) (www-ortho-europe.co.uk).

Fitting of the PPAM aid (Figure 3.4) was initiated whilst patients were seated. Firstly, the residuum of the amputated limb was covered with a soft dressing and a small cushion bag was placed at the distal end of the residuum (A). The outer pneumatic bag was placed over this and covered the length of patient's affected limb, up to the level of the groin (B). The rigid frame was then placed over the outer bag and slid up to the desired length but no closer than 8cm below the top of the outer pneumatic bag (C). The pneumatic bag was inflated to a pressure of 40mmHg, while the frame was being supported (D). A crucible strap was fitted to the distal ring of the PPAM aid to give support, at this point partial weight-bearing was achieved and any adjustments made (E). The length of the PPAM aid was adjusted by sliding the rigid outer frame over the outer inflatable bag until the rocker foot was suitably positioned as decided by the relevant physiotherapist. Fitting of the PPAM aid was conducted by the physiotherapist prior to data collection.

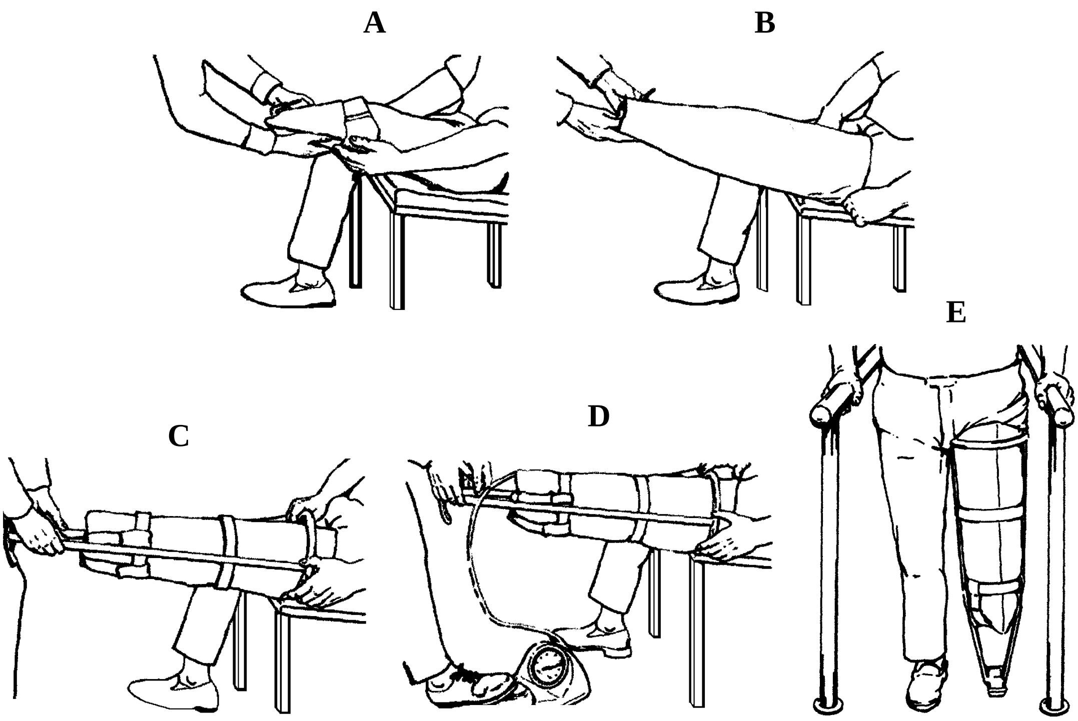


Figure 3.4 Schematic representation of the fitting procedure for the Pneumatic Post-Amputation Aid. (Ortho Europe Ltd, Alton, UK) (www.ortho-europe.co.uk).

3.2.4.3 Functional Prostheses

All amputees assessed were examined and prescribed their functional prostheses by the same consultant within the Hull Artificial Limb Unit, Hull and East Yorkshire Hospitals NHS Trust, UK. Following EWA use, patients were cast for a functional prosthesis and prescribed a prosthetic limb which was custom built to match the length of the intact limb, an example of which is shown in Figure 3.5. Typically this initial functional prosthetic limb was comprised of the same components for all patients. However, the specific needs of individual patients were taken into consideration. The functional prosthesis comprised of a custom-fitted polypropylene thermoplastic socket into which the patient's residuum was placed. The socket is lined with a rigid foam liner whilst the residuum covered with a cotton sock liner. The socket was then placed into a socket interface device located directly above the pylon. The various ankle and foot complexes available to patients were attached to the pylon as well as an optional cosmetic covering. The prescription of these components may vary due to age, weight, activity level, cost and patient preference. However, all patients from studies one and two were prescribed the same complex, with two exceptions. The ankle and foot complexes prescribed to the majority of patients in studies one and two were the Endolite Multiflex ankle and foot (Chas A Blatchford and Sons Ltd www.blatchford.co.uk). One patient was prescribed a solid ankle and cushion heel (SACH) foot (Chas A Blatchford and Sons Ltd www.blatchford.co.uk) due to a higher mass and activity level, while another (female) was prescribed an Elation Foot® (Ossur UK www.ossur.co.uk) to accommodate wearing a raised heel shoe.

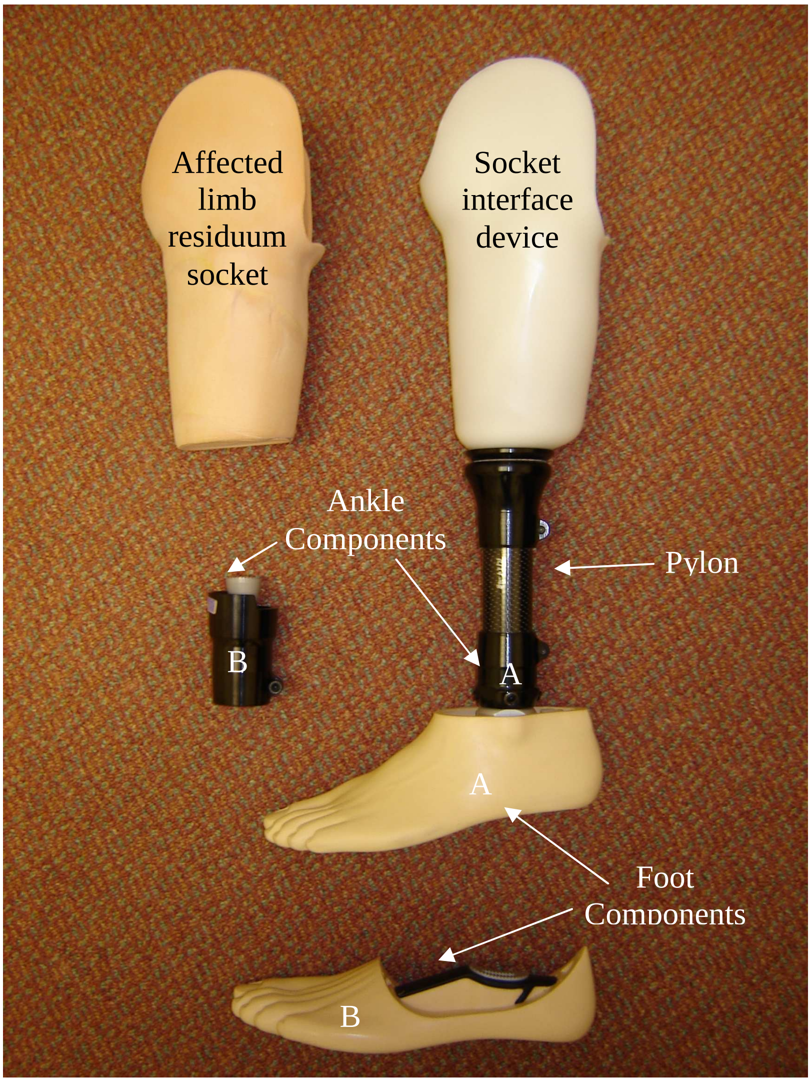


Figure 3.5 A functional prosthesis with components labelled, A – Senior and B – Multiflex ankle and feet components (Chas A Blatchford and Sons Ltd www.blatchford.co.uk.

Following the receipt of the functional prosthesis, patient's abilities were likely to change markedly over time. This led to a revision of their requirements in terms of prosthetic components, in particular for those who re-entered the workplace or continued sporting activities. Following discharge from rehabilitation, participants visited the same consultant within the Hull Artificial Limb Unit, HEY Hospitals NHS Trust, UK for these revisions. Details of changes in participant's prosthetic components following discharge from rehabilitation are detailed in Chapter Six, Section 6.2.1.

3.3 Biomechanical Data Acquisition, Processing and Analysis

3.3.1 Three-Dimensional Motion Capture

The three-dimensional (3D) motion capture system used was manufactured by Qualisys Motion Capture Systems (Qualisys, Gothenburg, Sweden). The motion capture system at the Department of Sport, Health and Exercise Science, University of Hull was made up of optoelectronic Qualisys ProReflex MCU1000 cameras, the associated data acquisition software Qualisys Track Manager version 2.2 (QTM v2.2) and all associated hardware (Qualisys, Gothenburg, Sweden). This equipment allowed for the capture of 3D movement (kinematic) data via retroreflective markers placed upon the object of interest. Two types of force plate were used, namely, a Kistler 9281B11 piezoelectric force plate (dimensions: 600x400mm) (Kistler, Winterthur, Switzerland) and an AMTI BP600600 strain gauge force plate (900x600mm) (AMTI, MA, US). These force plates are capable of measuring ground reaction forces (GRF) produced by individuals as they move over the force plates and make contact with them. The force plates measure GRFs along three axes, namely vertical (Fz), anterior-posterior (Fy) and medial-lateral (Fx).

Different combinations of camera numbers, positioning, force plates and associated equipment were employed in order to capture 3D data. The number of cameras used and their positioning is specified within the relevant methodology sections of each study.

3.3.2 Data Capture Unit Set-Up

The Qualisys ProReflex camera system is a flexible data capture system that is arranged in a serial fashion via the use of category 5 data cables as illustrated in Appendix I.

The cameras were arranged on adjustable tripods to allow for optimal and accurate viewing and re-positioning. In study one, cameras were connected to a laptop PC (Dell Latitude D800, Dell, Bracknell, UK) via a PC-S10-485 ultra serial port from which data were fed into QTM v2.2 (Qualisys, Gothenburg, Sweden).

Study three collected both kinetic and kinematic data, full hardware details are given below and in Appendix I. The analogue kinetic data signals ran from the Kistler force plate to the Kistler connection box (Kistler Type 5606A, Kistler, Winterthur, Switzerland) via connection cables (Kistler Type 1758A). The AMTI signal ran from force plates to signal amplifier units. These data were then fed into the analogue to digital (A-D) converter (Qualisys PCI-DAS6402/16, Qualisys, Gothenburg, Sweden) via coaxial cables and BNC connectors, as was kinematic data from the cameras, for synchronisation purposes. Camera one was connected to a desktop PC (Dell Optiplex GX280, Dell, Bracknell, UK) via a category five data cable while kinematic data were fed into QTM v2.2 (Qualisys, Gothenburg, Sweden). Finally, the A-D converter was connected to the desktop PC via ribbon cable with the Kistler connection box connected to the desktop PC, completing the fully synchronised unit.

3.3.3 Camera Calibration

Prior to data acquisition, the 3D volume in which the object of interest moved was calibrated. The same calibration procedure was used for all motion capture studies. In order to capture accurate and reliable 3D coordinate data, an arbitrary global or laboratory coordinate system was defined (Z - vertical, X – anterior/posterior and Y – medial/lateral). Qualisys Track Manager v2.2 uses a dynamic calibration method where an L-shaped reference structure (750 mm x 550 mm) (Figure 3.6) with retro reflective markers attached is placed in the estimated centre of the 3D volume. The marker in the corner represented the lab origin or zero point. A calibration wand is then required to carry out the calibration procedure. The calibration wand used in the current studies had markers at each endpoint of the T, an exact known distance of 749.4mm apart (Figure 3.6). The L-frame was placed in a consistent location for each calibration. Qualisys Track Manager v2.2 collected a fixed number of 1000 calibration frames over a 100-second interval in order to allow collection of the calibration frames over an extended period of time. This allowed coverage of a relatively small 3D volume of approximately $6.75m^3$ (4.5m x 1m x 1.5m) in study one and a relatively large 3D volume of approximately $60m^3$ (6m x 4m x 2.5m) in study three. Calibration quality was determined by assessing the residual error associated with each camera produced by QTM v2.2 at the end of the 100-second time interval. Residual errors were required to be below 2mm for each camera. Reports on the reliability of the data capture unit can be found in Appendix H.

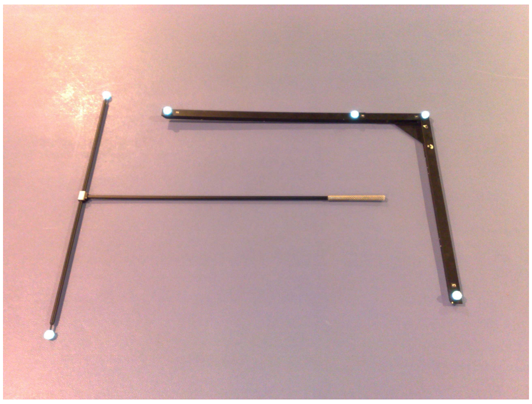


Figure 3.6 Qualisys ProReflex 3D motion capture system calibration equipment.

3.3.4 Data Acquisition

QTM software allowed for the synchronised capture of both kinematic and kinetic data as patients performed the movements assessed within each study.

Prior to acquisition of 3D data, acquisition parameters were set. These parameters were pre-determined as a workspace configuration that could be loaded, altered, saved and reloaded each time data acquisition occurred. These predetermined settings included kinematic and kinetic sampling frequencies and residual error tolerances, details of which are given in Table 3.1.

Table 3.1 Pre-determined data acquisition parameters within QTM v2.2 for studies one and three.

Parameter	Study One	Study Three
Kinematic sampling frequency (Hz)	100	100
Kinetic sampling frequency (Hz)	n/a	1000
Calibration wand size (mm)	749.4 (Medium)	749.4 (Medium)
Number of frames used in calibration	1000	1000
3D tracking parameters: Prediction error (mm)	20	20
3D tracking parameters: Max residual (mm)	5	5
Auto joining of markers (number of frames)	10	10

Spherical retro reflective markers (25mm – study one and 14mm – study three) were used in order to capture 3D kinematic data. Larger, more easily viewed markers were selected during study one as occlusion due to parallel bars occurred in the 3D volume recreated in the amputee rehabilitation room. These markers were placed upon patients lower limbs at pre-determined points of both anatomical and technical relevance, namely the six degrees of freedom (6DoF) marker model set described previously (Cappozzo *et al.,* 1995; Kalogridi *et al.,* 2006; Collins *et al.,* 2009; Buczek *et al.,* 2010). There are many marker sets currently available to researchers each with their own inherent strengths and weaknesses. The 6DoF marker model set and a rationale for its use is outlined in Section 3.3.5.

Once the camera system had been calibrated, the markers appropriately placed upon the patient and the acquisition parameters loaded, it was then possible to commence data collection. Patients were instructed as to what tasks they were required to perform, prior to 3D motion capture commencing. The length of time recording occurred for depended upon the time taken to complete each task. This varied between studies and mainly due to patient abilities. Marker trajectories were then labelled in QTM v2.2 with the assistance of the Automatic Identification of Markers (AIM) function. Trajectories were visually checked for marker switching and if necessary, edited. Files were also cropped to include only instances of the tasks being performed. These data were then exported in C3D format to the modelling software, Visual 3D (C-Motion, Rockville, US). The post processing and modelling stages of data analysis are detailed in Section 3.4.

3.3.5 Six Degrees of Freedom Marker Model Set

The six degrees of freedom (6DoF) marker model set is one of many marker models sets that are currently available to those interested in capturing and modelling human movement. The model used in this thesis consisted of 28 individual markers placed at predetermined anatomical landmarks on the lower limb as well as rigid clusters of four markers to define the static calibration file (Appendix J, Table 3.2). Due to the absence of anatomical landmarks on the prosthetic components, marker positions were estimated from anatomical landmarks on the intact limb, a procedure previously reported in the literature (Powers *et al.,*1998; Vanicek *et al.,* 2009a). Adapting the inertial properties of prosthetic limb has not been shown to adversely affect the resulting kinetic features apparent when investigating amputee movement (Miller, 1987; Czerniecki *et al.,* 1991; Sanderson and Martin, 1997; Powers *et al.,* 1998).

The 6DoF does not require any anthropometric assumptions with regards to the joint constraints between segments (i.e. thigh, shank) such as the knee, (Cappozzo *et al.,* 1995; Kirtley, 2006; Buczek *et al.,* 2010). The 6DoF marker model set defined and tracked each segment independently using rigid clusters of markers. This avoided some of the error from modelling assumptions apparent in other models (Kirtley, 2006; Collins *et al.,* 2009). The 6DoF model was able to track segments individually after the relationship between the rigid clusters (technical set) and some anatomical landmarks (anatomical set) has been defined. This involved recording a static trial with the full 28 marker set plus segment clusters (four markers per cluster) present (Appendix J), as the patient stood in the anatomical neutral position. Once this was recorded some markers were removed to perform 'dynamic' trials, those tasks which were of interest to the current thesis (Table 3.2). Following this, modelling software Visual 3D (C-Motion, Rockville, US), was used to define the relationship between the static trial and the dynamic trials. Details of this procedure are outlined below. Appendix J illustrates the placement of retroreflective markers for the 6DoF marker model set employed, with Table 3.2 detailing markers, anatomical positions and sizes.

Table 3.2 Markers employed within the six-degrees-of-freedom marker model set with associated anatomical positioning and sizes. Numbers correspond with those in Appendix J.

Marker Number	Anatomical Position	Marker Removed for Dynamic Trials	Marker Size	
			Study One	Study Three
1	Posterior Superior Iliac Spine		25mm	14mm
2	Anterior Superior Iliac Spine		25mm	14mm
3	Iliac Crest	•	25mm	14mm
4	Greater Trochanter	•	25mm	14mm
5	Thigh		Four 25mm Cluster	Four 14mm Cluster
6	Lateral and Medial Femoral Epicondyles	•	25mm	14mm
7	Shank		Four 25mm Cluster	Four 14mm Cluster
8	Distal Aspect of Lateral and Medial Malleoli	•	25mm	14mm
9	Distal Head of 1st and 5th Metatarsals		25mm	14mm
10	Proximal Head of 2nd Metatarsal		25mm	14mm
11	Dorsum of the 2nd Metatarsal		25mm	14mm
12	Posterior Aspect of Calcaneus		25mm	14mm
13	Medial and Lateral Aspects of the Calcaneus	•	25mm	14mm
14	Toe*		n/a	14mm

N.B. All markers and clusters were placed bilaterally. In the absence of anatomical landmarks, marker placement was estimated from intact limb, as described above.

*Marker placed on most anterior point of patients' foot in study three only.

The 6DoF marker model set was selected for the assessment and modelling of transtibial amputee movement for a number of reasons. Firstly, the 6DoF is a widely reported and accepted method of modelling human movement and has been shown to have good repeatability (Cappozzo *et al.,* 1995; Collins *et al.,* 2009; Kalogridi *et al.,* 2006; Buczek *et al.,* 2010). Also, assumptions are not made about joints constraints between segments when using the 6DoF marker model set. This is an important aspect when attempting to model a prosthetic limb due to the number of prosthetic components available in place of the ankle and knee of the amputated limb. This factor also allows for the visualisation of erroneous marker movement that may be hidden in other models (Kirtley, 2006).

3.4 Three-Dimensional Modelling and Signal Processing

Three-dimensional modelling was conducted using Visual 3D (C-Motion, Rockville, US). Raw data exported from QTM v2.2 in .C3D format was opened in Visual 3D for signal processing and modelling. This section outlines how the signals were processed, the data modelled, various modelling assumptions and finally, the outputs from the modelling software.

The modelling procedure involved tracking segmental movement through space via the use of rigid clusters once the segments had been defined using a static trial. In the case of the foot and pelvis, markers from the static trial were also used as tracking markers in the dynamic trials, as per the rigid clusters in the case of the thigh and shank. During the static trial the full 28 markers of the 6DoF marker model set were attached to the bony landmarks highlighted in Table 3.2. These markers identified the proximal and distal ends of segments as well as the medial and lateral aspects of each joint, with the exception of the pelvis, which is discussed in detail below. This information allowed for the computation of the segmental geometry and thus the centre of mass and radii of each segment. Table 3.3 outlines the definition of each segment with the exception of the pelvis.

Table 3.3 Segmental properties, values and definitions used during modelling in Visual 3D.

Properties		Segment		
		Thigh	**Shank**	**Foot**
Proximal Parameters	Lateral	Greater Trochanter	Lateral femoral epicondyle	Lateral malleolus
	Joint	Hip Joint	n/a	
	Medial	n/a	Medial femoral epicondyle	Medial malleolus
	Radius	Explicit from calculation of HJC	From endpoint to edge of segment geometry	
	Endpoint	Point from proximal lateral marker to end of explicit radius	Midpoint of proximal lateral and medial markers	
Distal Parameters	Lateral	Lateral femoral epicondyle	Lateral malleolus	5^{th} metatarsal head
	Joint	n/a		
	Medial	Medial Femoral Epicondyle	Medial malleolus	1^{st} metatarsal head
	Radius	From endpoint to edge of segment geometry		
	Endpoint	Midpoint of distal lateral and medial markers		
Segmental Geometry		Cone	Cone	Cone
Segmental Mass (proportion of total patient mass)		0.1	0.0465	0.0145

One assumption of the present modelling technique was that each segment was a rigid structure. This assumption was quite accurate for all lower limb segments assessed in the current studies except the foot. Although in reality the foot is not a rigid structure, the aim of the current thesis was not to assess the articulations present in the foot. Also, by modelling the foot as a rigid segment, a more accurate representation of the movements in some of the (more basic) prosthetic components used by patients may have been obtained.

When defining segments using the marker-based information above, various aspects of each segmental model can be modified. Segmental mass was estimated as a percentage of the total patient mass using regression equations (Dempster, 1955). The segmental geometry was also selected based upon previous anthropometric reports (Hanavan, 1964) with the segmental length being determined using the marker-based information. This also provided the segmental or local coordinate system (SCS, LCS) located at segment COM. This was required to analyse the motion of each segment. Inertial values of each segment were calculated using the segmental mass and geometry.

In Visual 3D, the pelvis segment (Visual 3D Pelvis) can be defined using similar procedures as the thigh, shank and foot. However, this is not recommended by the software developers as it requires the additional measurement of leg length and ASIS to greater trochanter length. The CODA pelvis was defined in order to complete the link model and used to obtain pelvic kinematics. The definition of the CODA pelvis used the right and left anterior and posterior superior iliac spines (ASIS and PSIS respectively) with the pelvis being modelled as a cylinder and its mass a proportion of total body mass of the patient (0.142). These bony landmarks are generally easier to palpate on slimmer patients. The origin of the CODA pelvis and the location of the SCS are located at the midpoint of the line between ASIS markers. From here, the hip joint centres were estimated using regression equations adapted by Visual 3D from previous experimental work (Bell *et al.*, 1989; Bell *et al.*, 1990). A virtual Visual 3D pelvis was also created in order to offset the 20 degree of anterior pelvic tilt apparent in the definition of the CODA pelvis and to calculate pelvic obliquity and rotation.

Once each segment had been defined, it was contained within a link model, whereby joints (e.g. knee) were defined between segments at the proximal end of one segment (thigh) and the distal end of another segment (shank). This procedure was saved as a model template in .mdh format. The model template is simply an ASCII file that contains information on segment definitions and participant data.

Once the static trial had been modelled the dynamic trials were then assigned to the static trial. This defined the relationship between the modelled segments in the static trial and the rigid clusters and other tracking markers present in both the static and dynamic trials.

Following the building of the model, assignment of dynamic trials to the static trial, some processing of the raw data signals was completed. Marker trajectories were initially interpolated using a cubic spline algorithm with a maximum frame gap of ten. Both the processed marker trajectories and the raw kinetic data were then filtered to remove high frequency noise using a low pass Butterworth filter with a cut-off frequency of 6Hz, as recommended in the literature (Robertson and Dowling, 2003). Once the pipeline command had been executed and the data were re-calculated, event identification was possible.

Event identification was necessary in order to normalise data to one gait cycle. In study one, kinetic data were not collected. Therefore, gait events of heel strike and toe off for both left and right feet were determined and verified visually, a procedure used previously (Vanicek *et al.,* 2007). This approach was also adopted for parts of study three, however, with the addition of kinetic data, it was possible to more accurately identify when these gait events occurred. Once all dynamic trials had gait event identification, it was then possible to present various measures as a single mean trace for that particular patient from that particular session, over one gait cycle. As well as normalising the kinetic and kinematic data, event identification also provided temporal-spatial variables such as step and stride length. All the variables provided from the processed data set were then presented in Visual 3D as a gait report.

Kinematic measures were defined in Visual3D using the relative orientation of the local coordinate systems of the two segments making up the joint and an x,y,z, cardan sequence. Details of the mathematical procedures are provided by authors of the modelling software and in the literature (Hamill and Selbie, 2004). The following joint angular position conventions were used:

	Positive	**Negative**
Sagittal Plane	Flexion	Extension
	Dorsiflexion	Plantarflexion
Frontal Plane	Adduction	Abduction
Transverse Plane	Internal Rotation	External Rotation
Pelvic Definitions:	**Positive**	**Negative**
Sagittal Plane	Anterior Tilt	Posterior Tilt
Frontal Plane	Upward Obliquity	Downward Obliquity

Joint moments (N.m/kg) normalised to mass, were defined using traditional inverse dynamics procedures in Visual 3D where a link segment model was created that initially separated each segment as rigid bodies. The following joint kinetic conventions were used:

Joint Moments	**Positive**	**Negative**
Sagittal Plane	Extensor	Flexor
	Plantarflexor	Dorsiflexor
Frontal Plane	Abductor	Adductor
Support Moment	Extensor	Flexor
Joint Powers	Generation	Absorption

Starting at the ankle joint, the moments acting upon the joint were calculated taking into account the effect of gravity on the COM, the effect of the GRF acting through the centre of pressure (COP) as well as the joint reaction force (Kirtley, 2006). Once this had been calculated at the ankle joint, joint moments for the knee and subsequently the hip were calculated using the equations below (Kirtley, 2006):

$$\text{Ankle Moment: } M_a = F_y(COP - x_f) + F_x(y_c - y_f) - R_y(x_c - x_a) - R_x(y_a - y_c) + I_f\alpha_f$$

$$\text{Knee Moment: } M_k = F_y(x_a - x_s) + F_x(y_s - x_a) - R_y(x_s - x_k) - R_x(y_k - y_s) + I_s\alpha_s - (-M_a)$$

$$\text{Hip Moment: } M_h = F_y(x_k - x_t) + F_x(y_t - x_k) - R_y(x_t - x_h) - R_x(y_h - y_t) + I_t\alpha_t - (-M_a)$$

Key: M_a = $\text{Moment}_{\text{ankle}}$

F_y = Force_y

$I\alpha$ = Moment of Inertia of segment x angular acceleration of segment

x_s = distances calculated from marker $\text{coordinates}_{\text{shank}}$

COP = Centre of Pressure

$_x$ = Horizontal

$_y$ = Vertical

$_{a,\ c,\ f,\ h,\ k,\ s,\ t}$ = ankle, COM, foot, hip, knee, shank, thigh.

The concept of support moments was presented in the literature as a general measure of muscular support in the lower limb and has been described as a useful clinical tool in gait rehabilitation (Winter, 1980; Whittlesey and Robertson, 2004). Support moments were calculated by summing the three lower limb joint moments calculated above:

$$\text{Support Moment: } M_s = \sum(M_h + M_k + M_a)$$

Joint powers (W/kg) normalised to mass, were calculated by Visual 3D after the computation of joint moments as they were required in the power calculation below:

$$\text{Joint Power: } (M_x + M_y + M_z) \text{ x } (\acute{\omega}_x + \acute{\omega}_y + \acute{\omega}_z)$$

Key: M_x = Joint $\text{Moment}_{\text{x direction}}$

$\acute{\omega}_x$ = Angular $\text{Velocity}_{\text{x direction}}$

3.5 Computerised Dynamic Posturography – The Neurocom Equitest®

This section outlines the set-up and technical specifications for computerised dynamic posturography (CPD) using the Neurocom Equitest®. Details of test protocols used are detailed in Section 7.2.

The Neurocom Equitest® is composed of a dynamic dual force plate system capable of translation in the anterior posterior plane and rotation about the sagittal plane. Two force plates measuring 23 x 46 cm are connected by a central pin joint, recording forces via four force transducers mounted symmetrically on a central plate with a fifth transducer bracketed to the central plate below the pin joint. This configuration allows for individual analysis of vertical force under the right and left feet separately. The four force transducers measure vertical forces applied to the support surface with the central transducer measuring anterior posterior shear force for both feet (Appendix M). The visual surround is capable of rotating in the sagittal plane with a maximum velocity of 15 deg/s. The force sampling frequency was set at 100 Hz. Developers of the Neurocom Equitest® provide specific guidelines pertaining to the experimental set up, participant preparation and administration of the testing procedures relating to the equipment which are outlined in Chapter Seven, Section 7.2.

3.6 Generic and Prosthesis-Related Quality of Life and Falls Efficacy – Self-Report Measures

Patient reports of quality of life (QOL) and their prostheses are important factors that determine how well an amputee adapts to the experience of amputation.

The self-report measures described in detail below aim to assess the generic and prosthesis-related QOL as well as falls efficacy reported by amputees.

3.6.1 The Medical Outcomes Study Short Form-36

3.6.1.1 Introduction

The SF-36 health survey is a generic measure of health status and is one of many tools available that allow clinicians and researchers to assess patient reported QOL. Its ability to be administered in a variety of ways (postal, phone, in person) as well as being translated into a large number of languages and region specific versions, has led to the SF-36 being widely accepted tool for the assessment of an individual's QOL. The paper based UK version of the SF-36 was used to assess changes in transtibial amputee generic QOL as they progressed through rehabilitation and six months post discharge.

3.6.1.2 Background and Development

The SF-36 was designed for a variety of uses including clinical practice and research (Ware and Sherbourne, 1992) and aimed to provide a standardised measure comparing patients with chronic health problems to those from the general population (Ware *et al.*, 2000).

The SF-36 questionnaire (Appendix E) is made up of 36 items, these items then contribute to eight scales (Table E.1), assessing different health phenomena, such as perceived well-being. These eight scales were selected from many and were the most frequently occurring concepts in health surveys (Ware and Gandek, 1998), namely Physical Functioning (PF), Role Physical (RP), Bodily Pain (BP), General Health (GH), Vitality (VT), Social Functioning (SF), Role Emotional (RE) and Mental Health (MH). These scales measure health from a subjective point of view, for example, perceived well-being. Subjective terms are assessed via self-reports of the frequency and intensity of feeling states (Table 3.4). Developers of the SF-36 argued that an individual's psychological state cannot be completely deduced from observable behaviour, thus necessitating self-report. A summary of the health phenomena assumed to be assessed by the SF-36 questionnaire is provided in Table 3.5.

The psychometric development of each scale is outlined in detail and referenced to previous research within the author guidelines on the SF-36 (Ware *et al.,* 2000). It is beyond the scope of the current thesis to determine the reliability and validity of the psychometrics of the SF-36.

Table 3.4 The eight scales of the SF-36 and the interpretation of high and low scores from each scale. (Adapted from Ware and Sherbourne, 1992).

			Interpretation of scores	
Scale	Number of items	Number of levels	High	Low
Physical Functioning (PF)	10	21	Performs all types of physical activities including the most vigorous without limitations due to health.	Limited a lot in performing all physical activities including bathing or dressing due to health.
Role Physical (RP)	4	5	No problems with work or other daily activities as a result of physical health.	Problems with work or other daily activities as a result of physical health.
Bodily Pain (BP)	2	11	No pain or limitations due to pain.	Very severe and extremely limiting pain.
General Health (GH)	5	21	Evaluates personal health as excellent.	Evaluates personal health as poor and believes it is likely to get worse
Vitality (VT)	4	21	Feels full of life and energy all of the time	Feels tired and worn out all the time.
Social Functioning (SF)	2	9	Performs normal social activities without interference due to physical or emotional problems.	Extreme and frequent interference with normal social activities due to physical or emotional problems.
Role Emotional (RE)	3	4	Feels peaceful, happy and calm all of the time.	Problems with work or other daily activities as a result of emotional problems.
Mental Health (MH)	5	26	Believes general health is much worse now than one year ago.	Believes general health is much better now than one year ago.

Table 3.5 A summary of the health phenomena assessed by the eight SF-36 scales. (Adapted from Ware *et al.*, 2000).

		Physical				Mental			
Scale	Label	Function	Well-Being	Disability	Personal Evaluation	Function	Well-Being	Disability	Personal Evaluation
Physical Functioning	PF	●							
Role Physical	RP			●					
Bodily Pain	BP		●	●					
General Health	GH				●				●
Vitality	VT		●				●		
Social Functioning	SF			●				●	
Role Emotional	RE							●	
Mental Health	MH					●	●		

3.6.1.3 Reliability, Validity and Use of the SF-36 in Empirical Literature

The SF-36 questionnaire has been reported to be both a reliable and valid tool for the assessment of QOL (Ware and Gandek, 1998). The reliability of the eight scales and two higher order dimensions of the SF-36 have been subject to both internal consistency and test-retest analysis. These studies assessed patients from a variety of disease states such as AIDS, diabetes, haemodialysis and GP practices (Ware *et al.,* 2000). Reliability coefficients from these analyses were, with a few exceptions, consistently above the recommended 0.70, mostly around 0.80 for the eight scales with the PCS and MCS displaying values exceeding 0.90 (Ware and Kosinski, 2001).

Validity of the SF-36 health survey has also received wide ranging attention. The items selected by authors of the SF-36 focus on eight health concepts from the Medical Outcome Study (Ware and Sherbourne, 1992). These items, when compared to other widely used generic health surveys, were among the eight most frequently represented health concepts (Ware and Gandek, 1998). Physical health orientated scales (Physical Functioning PF, Role Physical RP and Bodily Pain BP) have been found to be responsive to the benefits of hip replacement (Katz *et al.,* 1992), knee replacement (Kantz *et al.,* 1992) and heart valve replacement (Phillips and Lanksy, 1992). Mental health orientated scales (Mental Health MH, Role Emotional RE and Social Functioning SF) have been found to be responsive to changes in severity of depression (Beusterien *et al.,* 1996) and interpersonal therapy for depression (Coulehan *et al.,* 1997).

As well as being used to assess a wide range of disease states, the SF-36 questionnaire has also been used to specifically assess QOL in amputees of varying levels (Meikle, *et al.,* 2002; van der Schans *et al.,* 2002; Hoogendoorn and van der Werken 2001; Pezzin *et al.,* 2000), displaying its validity of use in an patient population of amputees. The current thesis deemed the SF-36 appropriate for use given the numerous reports of validity and reliability provided by authors and independent reviews, as well as its extensive use within the scientific literature. However, there is not an amputee specific version of the SF-36.

3.6.2 The Prosthesis Evaluation Questionnaire

3.6.2.1 Introduction

The Prosthesis Evaluation Questionnaire (PEQ) (Appendix F) is a measure of prosthesis related QOL, designed for use within a population of lower limb amputees. The PEQ is a self-administered questionnaire designed to be completed by the individual using a visual analogue scale with positive and negative response anchors to assess patient responses. The PEQ was employed to assess changes in transtibial amputee prosthetic function and health-related QOL from discharge from rehabilitation, up to six months post discharge.

3.6.2.2 Background and Development

The Prosthesis Evaluation Questionnaire (PEQ) was developed between 1995 and 1997 due to the lack of a specific amputation or prosthesis-related QOL measure (Legro *et al*,. 1998; www.prs-research.org). It was reported that although there were a range of measures that enabled the assessment of patients' use of prostheses, there were various issues with these measures (Legro *et al.*, 1998). Some were deemed comprehensive but too lengthy, such as the Prosthetic Profile of the Amputee (Gauthier-Gagnon and Grise 1994; Grise *et al.*, 1993) while others had issues with psychometric robustness, such as the Houghton Scale and Functional Independence Measure (Houghton *et al.*, 1989; Centre for Functional Assessment Research, 1991). Similar to the authors of the SF-36, part of the rationale for the development of the PEQ was the ever-increasing importance placed upon patient input in the delivery of health care (www.prs-research.org). The PEQ was designed for use within a rehabilitation health service research setting.

The PEQ contains 82 items or questions, 42 of these items contribute to nine independent scales (Table F.1) assessing various prosthesis specific issues with relation to QOL. The nine scales calculated within the PEQ are: Ambulation (AM), Appearance (AP), Frustration (FR), Perceived Response (PR), Residual Limb Health (RL), Social Burden (SB), Sounds (SO), Utility (UT) and Well-Being (WB). Some of these scales pertain to more generic QOL issues such as SB and WB whereas others are more lower limb amputee specific such as SO and RL. These scales were developed from an original pool of items formulated from a small group of clinicians and researchers as well as from published research, health professionals and an amputee support group (Legro *et al.,* 1998). The draft questionnaire was pilot tested with local patients before being readied for a field study (Legro *et al.,* 1998). A visual analogue scale format was selected as pilot testing revealed that the positive and negative anchors aided patients in their understanding of each item (Legro *et al.,* 1998).

3.6.2.3 <u>Reliability, Validity and Use of the PEQ in Empirical Literature</u>

Developers of the PEQ conducted a field study with a final group 92 amputees varying in level of amputation. The SF-36, The Sickness Impact Profile (SIP) and the Profile of Moods States short form (POMS-sf) questionnaires were selected against which to validate the PEQ (Legro *et al.,* 1998). Scales were developed from the test-retest data obtained from postal PEQ responses, with authors initially categorising all items by life domains before modifying the scales by reviewing the descriptive statistics, correlational and factor analyses as well as the responses to importance questions. Finally, scales were statistically tested for reliability and validity using Cronbach's alpha, Pearson product-moment correlation coefficients, intraclass correlation coefficients and principle component factor analysis using varimax rotation (Legro *et al.,* 1998). All but one of the original scales (transfers - subsequently omitted from the final version of the PEQ) were shown to be reliable as the PEQ correlated significantly with questionnaires it was compared to, suggesting it is a valid tool (Legro *et al.,* 1998). The PEQ has been used in a variety of scientific investigations pertaining to amputees of varying characteristics. One previous study used the PEQ as one comparison tool between groups of amputees using different prosthetic components (Kaufman *et al.,* 2008). One study validated the mobility scale of the PEQ (Miller *et al.,* 2001b) while others used the same scale when assessing the predictors of QOL, the development of a new functional test for lower limb amputees (Asano *et al.,* 2008; Deathe and Miller, 2005) and the influence of falling and the fear of falling on mobility in lower limb amputee mobility (Miller *et al.,* 2001a). These studies further highlight the efficacy of the PEQ and its sub-scales in assessment of prosthesis health related QOL. The current thesis deemed the PEQ appropriate for use given the reports of validity, reliability and psychometric properties provided by authors and the relevant use in the scientific literature.

3.6.3 The Modified Falls Efficacy Scale

3.6.3.1 Introduction

The Modified Falls Efficacy Scale (mFES) is a self-report measure of fear of falling or falls efficacy (Hill *et al.,* 1996) (Appendix G). Falls efficacy relates to a person self-perceived ability to complete a task without falling. The mFES is a variation on the original self-report measure (Falls Efficacy Scale) produced by Tinetti *et al.* (1990) and includes reports of outdoor activities. The mFES is primarily targeted at detecting and assessing falls efficacy in the population groups at higher risk of falling, for example, the elderly.

In this thesis, the paper-based version of the mFES was used to assess changes in falls efficacy in transtibial amputees from discharge from rehabilitation, up to six months post discharge.

3.6.3.2 Background and Development

As previously stated, the mFES is a variation of The Falls Efficacy Scale (FES) which was developed in order to provide a more sensitive measure of falls efficacy than was previously available (Hill *et al.,* 1996). The ten item FES questionnaire assessed individuals' confidence in completing everyday tasks on a ten point scale from 'not at all confident' to 'completely confident'. Authors of the mFES also allude to the potential ceiling effects associated with the exclusion of outdoor activities in the FES, thus not being able to differentiate between average and more mobile individuals. Four items assessing tasks commonly reported by fallers were added to the FES to create the mFES, the psychometric properties of each questionnaire were then contrasted within the study reported by authors (Hill *et al.*, 1996).

3.6.3.3 Reliability, Validity and Use of the mFES in Empirical Literature

The mFES was subject to analyses of reliability and validity within the development of the questionnaire itself. Modified versions of the FES have been reported to have good re-test reliability, with a lowest intraclass correlation coefficient of 0.54 for any item, the majority being considerably higher (Hill *et al.*, 1996). The validity of the mFES was highlighted by the observation of statistically different population responses in falls efficacy between those referred to a falls clinic and a control group (Hill *et al.,* 1996). A modified version of the FES has also been reported to have greater internal consistency and response variability than the original FES (Edwards and Lockett, 2008). The FES has been subject to a review article (Jorstad *et al.,* 2005). This article reported both good internal consistency (cronbach's alpha 0.90) and test re-test reliability (r=0.71), reporting the mFES to be both a valid and reliable tool.

An mFES has been reported as a tool used in empirical research into the falls within a community dwelling elderly population (Delbaere *et al.*, 2009), improvement in fall rates in the elderly via training (Vrantsidis *et al.*, 2009) and in analyses of the effect of fear of falling on gait in the elderly (Chamberlin *et al.*, 2005). The current thesis deemed the mFES appropriate for use given the reports of validity and reliability provided by authors and the relevant use in the scientific literature.

3.7 Statistical Analysis

A range of statistical models were applied to data and details of these statistical models are presented in the methods sections of the relevant studies. The majority of statistical models applied to data as well as the dependant variables analysed within these statistical models were chosen *a priori*. If the statistical model and/or the dependant variables were chosen post-hoc, then this has been reported within the statistical analysis sections of the relevant methodology sections.

Assumptions of all statistical tests were checked, where violation of these checks occurred, the appropriate non-parametric statistical test was employed. Details of each statistical model fit are detailed within the methodology section of relevant studies. The alpha level of statistical significance for all statistical analyses was fixed at $p \leq 0.05$.

4 CHAPTER FOUR – STUDY ONE. Kinematic Gait Adaptations in Transtibial Amputees During Rehabilitation.

4.1 Introduction

Previous research has not investigated the influence of different EWAs on relearning independent gait or how the prior use of an EWA affects early prosthetic gait. However, understanding how patients modify their gait as they learn to walk with a prosthesis in a rehabilitation setting could have important implications for both patients and therapists.

The aims of the current longitudinal study were three-fold. Firstly, the study investigated the gait patterns of transtibial amputees using either the AMA or PPAM aid. Secondly, the study investigated how the previous use of either EWA influenced gait as patients started to walk with their functional prostheses for the first time. Lastly, the longitudinal changes in gait that occurred from the first use of the functional prostheses to discharge from rehabilitation were investigated.

It was hypothesised that (1) during EWA use the AMA group would display a more proficient gait pattern in terms of variables such as walking velocity, when compared to the PPAM aid group as they were using an EWA with a greater functional capacity.

It was also hypothesised that (2) upon receipt of a functional prosthesis, those patients having previously used the AMA would display a more proficient gait pattern as they would have been used to practicing the control of the knee joint on the affected side. Lastly, it was hypothesised that (3) following the receipt of a functional prosthesis, until discharge from rehabilitation, the differences present between patients using either the AMA or PPAM aid would diminish as both groups adapted to their mechanical constraints.

4.2 Methods

4.2.1 Patients

Fifteen patients (12 men and 3 women) (Table 4.1) who had recently undergone transtibial amputation and were expected to receive, but had not yet received, a functional prosthesis were recruited into the study. These patients were recruited over a period between May 2005 and June 2007. Patients had the study explained to them by physiotherapists and subsequently gave written informed consent prior to data collection. Inclusion and exclusion criteria of patients in the current study have been detailed in Chapter Three, Section 3.2.2.

4.2.1.1 Prosthetic Components

Patients participated in the current study during normal rehabilitation treatment. Early walking aids were only available during physiotherapy treatment, limiting the time patients could practice walking with such devices. Once patients had received their functional prosthesis they were then assessed by physiotherapists to ensure safe mobilisation outside of the rehabilitation setting. The amount of time they used their prosthesis outside of the rehabilitation setting varied according to their needs and abilities. EWAs and functional prostheses were fitted by experienced physiotherapists according to the manufacturer's guidelines.

Table 4.1 Patient characteristics of transtibial amputees.

Group	Gender (Male/Female)	Age (years)	Height (m)	Mass (kg)	Amputated Limb (Right/Left)	Cause of Amputation	Functional Prosthesis
PPAM	F	49	1.61	93	R	Non-vascular	All patients used patella tendon bearing Endolite prostheses with a multiflex foot and ankle except; ‡SACH foot and †Elation Foot.
	M	71	1.78	71	R	Vascular	
	M	51	1.88	111	L	Non-Vascular	
	M	68	1.71	101	R	Vascular	
	M	65	1.80	95	R	Vascular	
	M	61	1.60	63	L	Vascular	
	F†	41	1.49	57	R	Non-Vascular	
Mean±SD		58.0±11.2	1.70±0.14	84.4±20.6			
AMA	F	66	1.70	75	R	Vascular	
	M	40	1.79	77	R	Non-vascular	
	M	70	1.67	72	L	Vascular	
	M	26	1.83	63	R	Non-Vascular	
	M	35	1.70	58	R	Vascular	
	M	43	1.72	81	L	Non-Vascular	
	M	57	1.77	121	R	Non-Vascular	
	M‡	62	1.87	111	L	Vascular	
Mean±SD		49.9±16.0	1.76±0.07	82.3±22.3			
All Patients		53.6±14.1	1.73±0.11	83.3±20.1			

4.2.2 Experimental Design and Protocol

Data were collected when patients attended a specialist amputee rehabilitation physiotherapy unit staffed by physiotherapists with clinical expertise in this area. The unit serves as both an in- and out-patients clinic as part of the Regional Limb Fitting service. Patients attended treatment as often as physiotherapists felt was appropriate to their stage of rehabilitation. Patients followed an individually designed programme consisting of goals negotiated and agreed with the patient. The study was a repeated measures design with randomised group allocation. Prior to data collection, patients were randomly allocated into experimental groups using the sealed envelope method; one group using the AMA (n = 8) the other using the PPAM aid (n = 7).

Patients attended a different number of gait retraining sessions as walking ability with either EWA or patients' initial functional prosthesis progressed at different rates. The majority of data were collected when patients attended as outpatients. To enable comparisons between patients, data collection sessions were standardised to five time points during their rehabilitation. Data were collected during visits one and two when patients attended the initial and final rehabilitation sessions, respectively, whilst using their specified EWA. Visit three measured patients whilst using their functional prosthesis for the first time and data were then collected two weeks later at visit four. Assessing patients at visit four allowed the measurement of gait adaptations that occurred in the short time following receipt of the functional prosthesis. The final data collection was completed when patients were discharged from rehabilitation at visit five.

Patient's height (m) and mass (kg) were recorded post-surgery using a free-standing height measure and beam column scale (Seca, Birmingham, UK). Data collection took place in the amputee physiotherapy room. An eight camera motion capture system sampled three-dimensional kinematic data at a frequency of 100 Hz using QTM software. Details of these methodologies were outlined in the Chapter Three, Sections 3.3 and 3.4.

Six wall-mounted cameras with multi-planar views and two tripod-mounted cameras with frontal plane view were set up in order to allow for a capture volume (approximately 6m^3) suitable for gait analysis. This configuration was selected given the dimensional restrictions inherent to the amputee rehabilitation room and in order to capture data between parallel bars (Figure 4.1). Data were only collected as patients walked towards the two tripod-mounted cameras.

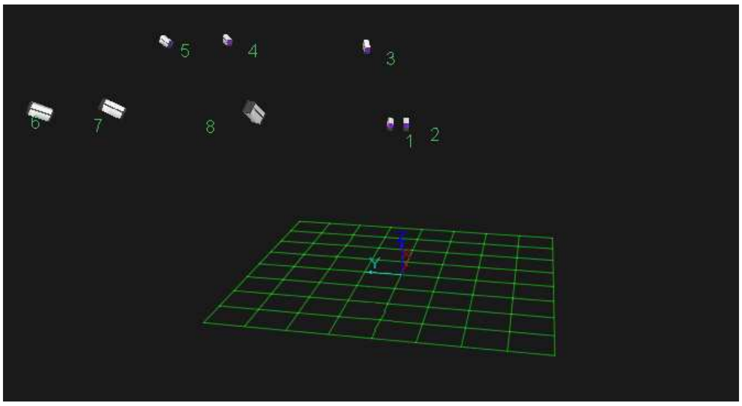


Figure 4.1 The eight camera ProReflex© system setup in the Amputee Therapies Room at Castle Hill Hospital, Hull, UK.

Patients were required to walk between parallel bars at a self-selected velocity, resting as required. A minimum of five walking trials were recorded per session. Patients wore their own comfortable, flat footwear during all data collection sessions. The PPAM aid has a convex rocker ‘foot’ at the distal end (Figure 3.3), thus patients only wore a shoe on the intact limb. A TES belt (Syncor, Dublin, Ireland) was employed in order to aid accurate three-dimensional reconstruction about the pelvis by reducing soft tissue movement. Once patients had been fitted with their specified EWA or functional prosthesis, 25 mm reflective markers were attached to specific anatomical landmarks by the same investigator according to the six degrees of freedom marker model set, described in the Chapter Three, Section 3.4. Marker placement on the affected limb was estimated from intact limb anatomical landmarks, a procedure previously reported in the literature (Powers *et al.*, 1998).

4.2.3 Data Analysis

Data frames of steady-state walking were analysed and averaged for walking trials. Gait events were identified visually from the motion capture data. Group mean (SD) temporal-spatial variables of walking velocity, step and stride length, cadence, relative double limb support and relative stance duration were calculated and normalised to the gait cycle. Walking velocity is of particular clinical relevance as improvements between 0.10 and 0.16 m/s have been used to infer clinically meaningful functional progress following hip fracture and stroke (Palombaro *et al.*, 2006; Tilson *et al.*, 2010). Kinematic data of the ankle, knee, hip and pelvis were measured in the frontal and sagittal planes and normalised to the gait cycle. Frontal plane (hip and pelvis) and sagittal plane (ankle, knee, hip and pelvis) joint angles were analysed at foot contact and toe off. Peak joint angles were also compared during the swing phase and, for the knee only, during the loading response. In order to display the effects of using either EWA when walking with a functional prosthesis during rehabilitation, data were presented from the first (visit three) to the last (visit five) use of functional prostheses.

4.2.4 Statistical Analyses

Group averaged means were used for statistical analysis. Differences in each group characteristic were analysed using an independent samples t-test. A mixed design repeated measures analysis of variance (ANOVA) was performed, Limb (affected vs. intact) * Group (AMA, PPAM) * Time (visit number), with repeated measures on the last factor. In relation to the hypotheses, this statistical model allowed for the analysis of change in both general indicators of gait progress, such as walking velocity as well as the discrete measures of joint biomechanics. In the instance of a significant time main effect or interaction effect, post-hoc comparisons were conducted using a Sidak adjustment in SPSS v.15.0 (SPSS Inc., Chicago, USA). The underlying assumption of sphericity of the data was verified and where this was violated, adjustments to the degrees of freedom following the Greenhouse-Geisser method were applied. The alpha level of statistical significance was set at $p \leq 0.05$.

4.3 Results

The mean (SD) time interval for all patients between visits one and five was 78.1 ± 25.3 days (range 40-126 days). Data for age $D(15) = 0.17$, $p=0.20$, height $D(15) = 0.13$, $p=0.20$ and mass $D(15) = 0.15$, $p=0.20$ were normally distributed as verified using the Kolmogorov-Smirnov test. Data for age $F(1,13) = 2.51$, $p=0.14$, height $F(1,13) = 4.02$, $p=0.07$ and mass $F(1,13) = 0.03$, $p=0.87$ also satisfied the requirement of homogeneity of variance as verified using Levene's test. There were no significant differences between the PPAM group and the AMA group in terms of age (years) ($p=0.28$), height (m) ($p=0.29$) or mass (kg) ($p=0.85$). There were no significant differences between the PPAM group and the AMA group in terms of total rehabilitation time (days) ($p=0.36$), time to receipt of prosthesis (days) ($p=0.25$) or the total number of physiotherapy treatments received during rehabilitation ($p=0.71$).

4.3.1 Temporal-Spatial Variables

Temporal-spatial variables across all visits are presented in Table 4.2 with complete statistical analyses provided in Table 4.3. Post-hoc comparisons for the walking velocity time main effect revealed that walking velocity increased significantly during rehabilitation, except between visits four and five ($p=0.07$). However, there were no significant differences in walking velocity between groups.

Post-hoc comparisons for the significant time by limb interaction revealed that affected limb step length was significantly longer than intact limb step length at visits one, two ($p=0.00$) and five ($p=0.02$). However, from visit three to discharge from rehabilitation, intact limb step length increased significantly ($p=0.01$), reducing between limb differences, although affected limb step length was still longer than intact limb step length at visit five. Stride length increased significantly between visits one and three, four and five (all $p<0.02$), although there were no group differences ($p=0.16$).

During visits one and two, the PPAM group displayed significantly larger between limb differences in cadence compared to the AMA group ($p=0.01$). Also increases in affected limb cadence from visits two to three and visits three to four, were significantly larger in the PPAM group compared to the AMA group ($p=0.04$).

Post-hoc comparisons for the relative stance duration three-way interaction effect showed that during visits one and two, the PPAM group showed significantly larger between limb differences than the AMA group, due to shorter relative stance duration in the affected limb ($p=0.01$). The between limb differences for the AMA group were somewhat smaller, but not significantly reduced over time, as relative stance duration decreased in both limbs. The PPAM group displayed a significant increase in affected limb relative stance duration from visit two to visit three ($p=0.01$). Relative double limb support analysis produced a significant Visit * Group interaction ($p=0.00$). This resulted from a generally linear decrease in relative double limb support in the AMA group, contrasted with inconsistent changes in the PPAM group.

Table 4.2 Mean (SD) temporal-spatial variables. Data are presented for the affected and intact limbs separately.

		Rehabilitation Session Number				
	Group	Visit One	Visit Two	Visit Three	Visit Four	Visit Five
Walking Velocity (m/s)	AMA	0.30 (0.11)	0.41 (0.17)	0.49 (0.11)	0.58 (0.12)	0.71 (0.13)
	PPAM	0.33 (0.08)	0.37 (0.04)	0.52 (0.04)	0.65 (0.09)	0.72 (0.14)
Relative Double Limb Support (%GC)	AMA	60.1 (6.1)	57.5 (9.1)	53.8 (4.5)	50.8 (4.6)	48.2 (4.3)
	PPAM	46.0 (7.8)	44.7 (2.3)	52.7 (2.0)	46.0 (4.8)	48.9 (6.3)
Stride Length (m)	AMA	0.80 (0.16)	0.86 (0.17)	0.92 (0.13)	0.98 (0.14)	1.04 (0.17)
	PPAM	0.72 (0.06)	0.74 (0.04)	0.88 (0.03)	0.92 (0.06)	0.98 (0.06)
Step Length (m)	AMA Affected	0.41 (0.11)	0.45 (0.04)	0.47 (0.04)	0.50 (0.04)	0.53 (0.03)
	PPAM Affected	0.44 (0.03)	0.43 (0.03)	0.44 (0.03)	0.47 (0.04)	0.49 (0.04)
	AMA Intact	0.35 (0.06)	0.41 (0.04)	0.45 (0.03)	0.47 (0.03)	0.51 (0.03)
	PPAM Intact	0.29 (0.05)	0.31 (0.03)	0.44 (0.03)	0.44 (0.03)	0.48 (0.03)
Cadence (Step/Min)	AMA Affected	49.6 (13.5)	53.1 (4.7)	61.1 (5.9)	71.6 (4.2)	80.8 (4.1)
	PPAM Affected	48.0 (5.7)	52.8 (4.7)	73.6 (4.3)	85.3 (4.1)	90.1 (5.6)
	AMA Intact	50.0 (4.6)	58.4 (6.9)	64.9 (4.6)	72.0 (4.2)	83.2 (5.8)
	PPAM Intact	65.9 (7.6)	69.0 (8.4)	71.6 (3.7)	84.8 (4.1)	86.3 (4.5)
Relative Stance Duration (% GC)	AMA Affected	78 (4.5)	75 (6.7)	72 (2.5)	72 (3.4)	72 (1.5)
	PPAM Affected	64 (5.4)	62 (2.6)	76 (3.1)	71 (4.5)	73 (2.9)
	AMA Intact	81 (5.2)	82 (3.5)	80 (3.2)	77 (3.7)	75 (3.2)
	PPAM Intact	83 (4.4)	82 (0.6)	79 (0.7)	76 (3.7)	75 (4.1)

Table 4.3 Statistical breakdown of temporal-spatial variables. Results are reported (F value, significance level (P) and effect size, eta^2) from the mixed design repeated measures ANOVA.

	Main Effect									Two-way Interaction									Three-way Interaction		
	Time			Limb			Group			Time * Group			Limb * Group			Time * Limb			Time * Limb * Group		
	F(4,52)	p	eta^2	F(1,13)	P	eta^2	F(1,13)	p	eta^2	F(4,52)	p	eta^2	F(1,13)	P	eta^2	F(4,52)	p	eta^2	F(4,52)	p	eta^2
Walking Velocity	44.84	.00*	.76				0.30	.59	.02	0.68	.61	.05									
Relative Double Limb Support	4.09	.01*	.24				9.87	.01*	.43	8.42	.00*	.39									
Stride Length	23.17	.00*	.64				2.21	0.16	0.15	0.48	0.62	.04									
Step Length	22.40	.00*	.63	30.88	.00*	.70	2.20	.16	.15	0.53	.71	.04	2.18	.16	.14	6.09	.00*	.32	2.08	.10	.14
Cadence	38.71	.00*	.75	8.78	.01*	.40	6.36	.03*	.33	0.58	.68	.04	1.26	.28	.09	12.01	.00*	.48	11.28	.00*	.47
Relative Stance Duration	5.20	.00*	.29	79.37	.00*	.86	4.53	.05*	.26	8.35	.00*	.39	6.53	.02*	.33	19.12	.00*	.60	18.90	.00*	.59

*Indicates statistically significant result, p≤0.05.

4.3.2 Joint Kinematics

Group mean joint kinematics from functional prosthetic gait are presented in the sagittal plane (Figure 4.2) and frontal plane (Figure 4.3). Results from statistical analyses are provided for sagittal (Table 4.4) and frontal plane (Table 4.5) joint kinematics. Significant three-way interactions were found for all sagittal plane ankle and knee angles throughout the gait cycle. However, at visit one and two the ankle joint of both EWAs and the knee joint in the PPAM aid were non-articulated. Therefore, statistically significant differences in ankle and knee joint kinematics might be expected once the patients were able to move their joints through a greater range of motion (ROM) using a functional prosthesis.

4.3.3 Sagittal Plane Kinematics

Active plantarflexion was not possible given the passive nature of the ankle-foot complex of the observed prosthetic components. The intact limb in the PPAM group achieved greater ankle plantarflexion during early stance phase and early swing phase at visit five (Figure 4.2).

The affected limb knee for both groups was generally flexed throughout stance phase at visit three, and gradually became more extended during early and late stance by discharge (Figure 4.2). Peak knee flexion in the loading response was not significantly different during rehabilitation or between groups. Throughout rehabilitation, the intact limb of the AMA and PPAM aid groups did not fully extend at the knee during mid-stance. However, peak knee flexion during the loading response occurred somewhat before in the intact limb compared to the affected limb (Figure 4.2). The reduction of peak intact knee flexion between visit three to five in the AMA group during swing phase can be observed in Figure 4.2, whereas, the PPAM group peak intact knee flexion increased during the same period.

Table 4.4 Statistical breakdown of sagittal plane kinematic gait variables at Foot Contact (FC), Peak Joint Angle During Loading Response (LR), Toe Off (TO) and Peak Joint Angle During Swing (PDS). Results are reported (F value, significance level (p) and effect size, eta^2) from the mixed design repeated measures ANOVA.

		Main Effect									Two-way Interaction									Three-way Interaction		
		Time			Limb			Group			Time * Group			Limb * Group			Time * Limb			Time * Limb * Group		
		F(4,52)	P	eta^2	F(1,13)	P	eta^2	F(1,13)	P	eta^2	F(4,52)	p	eta^2	F(1,13)	P	eta^2	F(4,52)	p	eta^2	F(4,52)	p	eta^2
Ankle	FC	2.74	.04*	.17	13.02	.00*	.50	1.82	.20	.12	3.58	.01*	.22	1.04	.33	.07	3.86	.01*	.23	3.23	.02*	.12
	TO	1.89	.13	.13	11.96	.00*	.48	0.03	.86	.00	0.61	.66	.05	0.01	.91	.00	6.83	.00*	.34	0.35	.84	.03
	PDS	0.96	.44	.07	11.71	.01*	.47	5.55	.04*	.30	3.62	.01*	.22	1.44	.25	.10	4.66	.00*	.26	3.00	.03*	.19
Knee	FC	2.75	.04*	.18	0.02	.88	.00	0.43	.53	.03	2.15	.09	.14	2.51	.14	.16	9.92	.00*	.43	7.81	.00*	.38
	LR	0.73	.57	.05	0.26	.61	.02	1.35	.27	.09	1.59	.19	.11	1.93	.19	.13	5.32	.00*	.29	4.70	.00*	.27
	TO	28.52	.00*	.69	146.65	.00*	.92	2.51	.14	.16	6.80	.00*	.34	58.55	.00*	.82	36.72	.00*	.74	17.69	.00*	.58
	PDS	22.26	.00*	.63	121.08	.00*	.90	5.71	.03*	.31	8.07	.00*	.38	39.93	.00*	.75	29.69	.00*	.70	17.32	.00*	.57
Hip	FC	2.36	.07	.15	1.42	.26	.10	0.10	.76	.01	0.43	.78	.03	1.42	.26	.10	4.10	.01*	.24	7.34	.00*	.36
	TO	3.80	.01*	.23	34.43	.00*	.73	0.00	.96	.00	1.33	.27	.10	20.35	.00*	.61	15.06	.00*	.54	10.75	.00*	.45
Pelvis	FC	2.21	.08	.15	35.40	.00*	.73	0.01	.85	.00	0.61	.66	.05	1.78	.21	.12	2.59	.05*	.17	1.15	.34	.08
	TO	2.77	.04*	.18	0.01	.93	.00	0.00	.98	.00	0.89	.48	.06	4.26	.06	.25	1.37	.26	.10	1.00	.42	.07
	PDS	2.00	.11	.13	6.15	.03*	.32	0.00	.98	.00	0.67	.62	.05	1.30	.28	.09	2.34	.07	.15	0.42	.80	.03

*Indicates statistically significant result, p≤0.05.

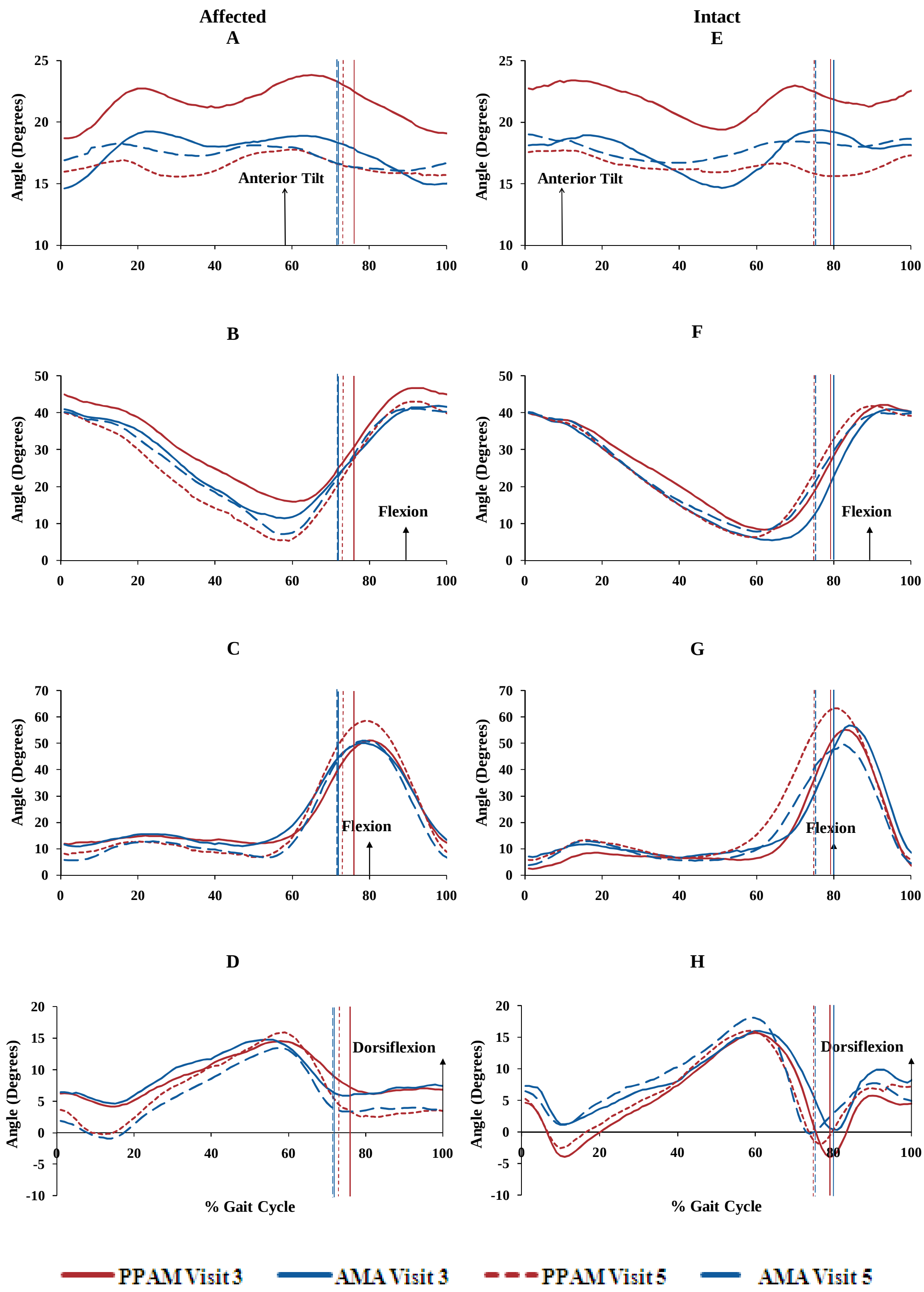


Figure 4.2 Group mean sagittal plane kinematics of the affected limb pelvis (A), hip (B), knee (C) and ankle (D) and intact limb pelvis (E), hip (F), knee (G) and ankle (H). All values in degrees (º). Time normalised to 100% of gait cycle. Vertical lines represent toe off.

Throughout rehabilitation, neither limb in either group achieved full hip extension during the gait cycle. The PPAM group displayed a larger change in affected limb hip ROM from visits three to five, almost reaching full extension at the pre-swing phase (Figure 4.2). At foot contact (p=0.02) and toe off (p=0.00), post hoc analysis revealed that the PPAM group's affected limb hip flexion significantly increased from visits two to three resulting in significant three-way interaction effects.

No significant interaction effects were found in pelvic tilt, reflecting the low magnitude of changes in pelvic motion. Pelvic tilt remained anterior in direction, although reduced pelvic ROM was observed at visit five in both affected and intact limbs (Figure 4.2).

4.3.4 Frontal Plane Kinematics

The AMA group displayed an observable reduction in intact hip abduction from mid-stance to early swing phase during visits three to five, whereas the PPAM group displayed a general increase in intact hip abduction (Figure 4.3). Post-hoc analysis revealed that PPAM group intact limb hip abduction significantly decreased between visits two and three at foot contact (p=0.00) and toe off (p=0.01), resulting in significant three-way interactions for peak hip abduction. Affected limb hip abduction generally decreased during the gait cycle in both group from visits three to five (Figure 4.3)

Table 4.5 Statistical breakdown of frontal plane kinematic gait variables at Foot Contact (FC), Toe Off (TO) and Peak Joint Angle During Swing (PDS). Results are reported (F value, significance level (p) and effect size, eta^2) from the mixed design repeated measures ANOVA.

		Main Effect									Two-way Interaction									Three-way Interaction		
		Time			Limb			Group			Time * Group			Limb * Group			Time * Limb			Time * Limb * Group		
		F(4,52)	p	eta^2	F(1,13)	P	eta^2	F(1,13)	p	eta^2	F(4,52)	p	eta^2	F(1,13)	P	eta^2	F(4,52)	p	eta^2	F(4,52)	p	eta^2
Hip	FC	1.33	.27	.09	10.35	.01*	.44	0.95	.35	.07	0.91	.46	.07	0.18	.68	.01	5.63	.00*	.30	5.44	.00*	.30
	TO	0.59	.67	.04	0.11	.74	.01	0.36	.56	.03	1.38	.25	.10	0.44	.52	.03	3.72	.01*	.22	5.26	.00*	.29
Pelvis	FC	0.66	.66	.05	0.01	.94	.00	0.00	.99	.00	1.03	.40	.07	1.11	.31	.08	2.57	.05*	.17	4.63	.00*	.26
	TO	7.09	.00*	.35	0.09	.77	.01	3.30	.09	.20	4.23	.01*	.25	2.83	.12	.18	2.66	.04*	.17	6.28	.00*	.33
	PDS	2.57	.05	.17	1.29	.28	.09	3.21	.10	.20	0.28	.89	.02	2.77	.12	.18	2.40	.06	.16	4.54	.00*	.26

*Indicates statistically significant result, $p<0.05$.

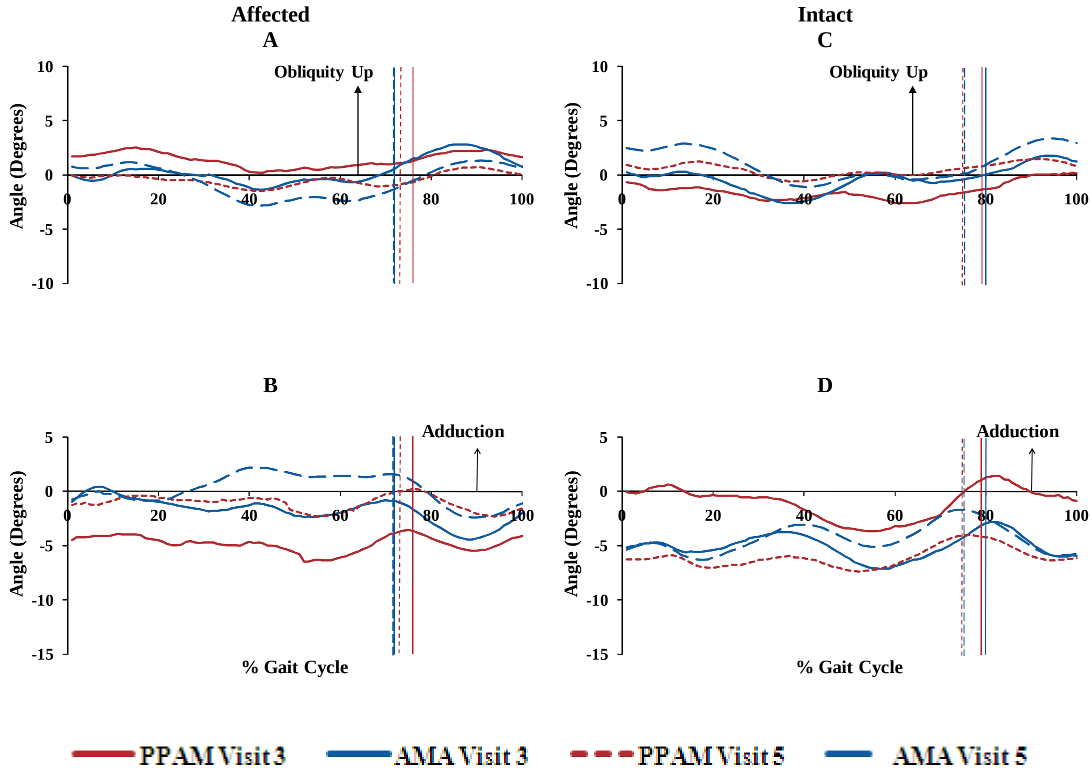


Figure 4.3 Group mean frontal plane kinematics of the affected limb pelvis (A) and hip (B) and intact limb pelvis (C) and hip (D). All values in degrees (º). Time normalised to 100% of gait cycle. Vertical lines represent toe off.

At visit five, hip-hiking had reduced on the affected side and increased on the intact side in relation to visit three, for both groups (Figure 4.3). Profiles of pelvic obliquity remained similar but changed in magnitude. However, post-hoc analysis revealed that, in the PPAM group, intact hip-hiking significantly decreased between visits two and three at foot contact ($p=0.02$) and toe off ($p=0.01$), resulting in significant three-way interactions.

4.4 Discussion

Research has shown that transtibial amputees are able to walk effectively (Sanderson and Martin, 1997; Nolan *et al.,* 2003). However, there have been no reports to date about the process by which amputees regain the ability to walk during rehabilitation or the effect of different EWAs. Therefore, the current study investigated the frontal and sagittal plane kinematic differences between transtibial amputees using an articulated (AMA) and a non-articulated (PPAM aid) EWA during gait retraining. This study also investigated how the previous use of either EWA influenced subsequent gait patterns, and if either EWA had any gait benefits during rehabilitation.

4.4.1 EWA Gait

It was hypothesised that the AMA group would display a more proficient gait pattern at this stage of rehabilitation when compared to the PPAM group. However, walking velocity increased similarly between groups as patients progressed through rehabilitation. At the end of EWA use, velocities observed in the current study (0.39 ± 0.12 m/s) were slower than previously reported in transtibial amputees, four weeks into their rehabilitation (0.51 ± 0.40 m/s) (Jones *et al.,* 2001).

The PPAM group did, however, display larger inter-limb differences in cadence at visits one and two and achieved increases in walking velocity more as a function of greater affected limb cadence at visit three. The AMA group took longer steps with both respective limbs to increase walking velocity, although stride length did not increase significantly between visits one and two. This is a novel finding and suggests that the type of EWA used during rehabilitation results in different gait adaptations but similar increases in walking velocity. The consequences of this finding are unknown and would benefit from further investigation. Similar increases in walking velocity coupled with inconsistent inter-limb differences meant that the hypothesis of improved gait function in the AMA group during EWA use was rejected.

Between-limb differences have been reported in studies of experienced amputees, as was the case in temporal-spatial variables of the current study, supporting the notion that gait asymmetry is an inherent characteristic of amputee gait (Winter and Sienko 1988; Hurley *et al.,* 1990; Sanderson and Martin, 1997; Powers *et al.,* 1998; Isakov *et al.,* 2000; Bateni and Olney, 2002; Grumillier *et al.,* 2008; Vickers *et al.,* 2008; Vrieling *et al.,* 2008; Vanicek *et al.,* 2009a). Therefore, during gait retraining and rehabilitation, achieving gait symmetry may not always be the goal. Rather, returning patients to a functionally stable and comfortable level of mobility may be more realistic. Further improvement in limb symmetry may be anticipated with additional prosthetic use post-discharge, as previous studies found that kinematic gait patterns of transtibial amputees, with more experience of walking with a prosthesis than the patients in the current study, demonstrated minimal distinguishing features from able-bodied individuals (Sanderson and Martin, 1997).

4.4.2 Transfer to Functional Prosthesis

At visit three the affected limb knee had a small ROM and was mainly flexed during weight acceptance. At visit five, there was an increase in knee ROM during weight acceptance. In both groups, the knee was more extended at initial contact, there was a greater knee flexion during the loading response with the knee then extending towards mid-stance. The greater ROM suggested that patients improved their ability to control the knee joint on the affected side. With practice, patients seemed to gain proficiency in controlling the knee musculature especially during the loading response. Despite the fact that the AMA group had more practice controlling the knee of the affected limb since visit one, the knee flexion profiles for both groups were remarkably similar at discharge, with the PPAM group showing increased knee flexion during swing phase (Figure 4.2). Patients in the current study appeared to adapt the intact limb more than the affected limb, as between limb differences were reduced during rehabilitation, especially in temporal-spatial measures. This may have reflected the amputees increased ability to adjust their intact limb during gait whilst progressively developing the control of their affected limb, an adaptation strategy that has been reported previously (Vanicek *et al.*, 2007).

Hip-hiking on the affected side reduced over time during the gait cycle, however, there were no observable differences between groups. This indicated that the amount of 'hip-hiking' measured at visit one in both groups, reduced towards discharge. This suggested a greater ability to flex and extend the affected limb knee, thus reducing the need to elevate the pelvis and flex the hip on the affected side to ensure adequate ground clearance. It was hypothesised that upon receipt of a functional prosthesis, patients having previously used the AMA would display a more proficient gait pattern compared to those having previously used the PPAM aid. However, due to a lack of clear inter-group differences this hypothesis was rejected as both groups seemed to adapt to their functional prostheses similarly.

4.4.3 Prosthetic Gait

The hypothesis that inter-group differences in gait would diminish following the receipt of a functional prosthesis to discharge from rehabilitation was accepted. The inconsistent differences noted during earlier period of rehabilitation (visits one and two) seemed to disappear upon receipt of a functional prosthesis. This was also coupled with the lack of significant group main effects. Walking velocity did not significantly increase during the latter stages of rehabilitation, reflecting a plateau in progress at discharge from physiotherapy. This indicated that physiotherapists were only discharging patients once a consistent level of mobility had been achieved. It was likely that increases of 0.41 (AMA) and 0.39 m/s (PPAM) represents highly clinically relevant increases in walking ability. Prior to discharge, patients that had the capability practised more functionally demanding tasks such as walking at different velocities, turning, stair climbing, carrying loads and walking on different terrains. Practice of such tasks, may be necessary to invoke further improvements in walking ability at discharge.

At discharge from rehabilitation, walking velocity and cadence values were still below values reported in the literature (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers *et al.,* 1998; Nolan *et al.,* 2003). However, previous studies did not investigate gait patterns of new prosthetic users (Winter and Sienko, 1988; Hurley *et al.,* 1990; Sanderson and Martin, 1997; Powers *et al.,* 1998; Nolan *et al.,* 2003) and some of the previous research investigated gait patterns from a younger, healthier population undergoing amputation following trauma, with a greater potential for speedier rehabilitation (Sanderson and Martin, 1997, Nolan *et al.,* 2003).

Both groups of patients displayed decreased affected limb stance duration. This has previously been explained as a compensatory mechanism employed by amputees in order to protect their affected limb from increased forces (Hurley *et al.,* 1990; Powers *et al.,* 1998; Nolan *et al.,* 2003), wariness in applying pressure to the affected limb and its surfaces and constructs, which are not used to or designed for receiving pressure (Jones *et al.,* 2001) and also a lack of confidence in the ability to control the affected limb (Sanderson and Martin, 1996; Sanderson and Martin, 1997). Affected limb stance duration increased during rehabilitation such that stance duration was similar between affected and intact limbs of both groups. This was mirrored by a general decrease in relative double limb support time, more markedly so in the AMA group. This suggested that patients became more comfortable and confident whilst weight-bearing on the affected limb during the course of rehabilitation.

All patients displayed a reduction in intact limb ankle plantarflexion between 50-80% of the gait cycle, compared to values reported in literature (Sanderson and Martin, 1997). Keeping the intact limb in dorsiflexion during early swing phase may assist in reducing step length and between limb asymmetry as well as aiding ground clearance. This kinematic adaptation may also explain the observed reduction in walking velocity, as plantarflexor muscle contribution was absent on the affected limb.

Hip flexion profiles revealed that across all visits, neither limb reached full extension in either group (Sanderson and Martin, 1997; Kirtley, 2006). No patient displayed a hip flexion contracture, as assessed by Thomas' test, where the patient lies supine and flexes one hip while one whilst maintaining the other in extension. However, there was an improvement in affected limb hip extension in both groups, as the hip extended more between 50 - 65% of the gait cycle between visit three and five. The lack of extension at the hip (late stance) and knee (initial contact, mid stance), as well as ankle dorsiflexion and anterior pelvic tilt gave the impression of a more flexed hip, knee and ankle gait pattern.

These findings suggest that transtibial amputees may benefit from additional home or therapy-based exercise programmes that target increasing muscle length, strength and joint mobility of the lower limb musculature. Future studies may also consider assessing muscular strength and activity during amputee gait relearning. The flexed hip, knee and ankle gait pattern and associated lowered centre of gravity coupled with lower walking velocities, could also reflect a lack of confidence in mobility of the new amputees in the current study.

4.5 Conclusion

The findings from the current study contribute to our understanding of how amputees achieve levels of gait proficiency required for independent living. Some kinematic and temporal-spatial differences were found between the two groups of transtibial amputees during EWA rehabilitation, the differences were not consistent enough to accept the first hypothesis. When patients transferred from EWA to their functional prosthesis, differences in gait between groups were still apparent. However, at discharge, both groups had improved walking performance and had reached an acceptable level of walking ability, despite very different gait patterns with the EWAs during early rehabilitation, supporting the third hypothesis. This suggests that the most significant gait adaptations occurred following receipt of a functional prosthesis. Our results did not show a clear benefit in gait patterns at discharge following use of either EWA. This may have important cost implications for the NHS given that the PPAM aid was approximately 50% cheaper to purchase than the AMA. In addition, the PPAM aid can be used during the rehabilitation of both transtibial and transfemoral amputees, whereas the AMA was designed specifically for transtibial amputees. With limitations on financial resources and the apparent lack of clear benefits of one EWA over another, this factor is likely to play an important role in physiotherapist's selection of an EWA. Increased patient numbers and kinetic analysis of amputees would help to further elicit the origin of differences observed between the AMA and PPAM groups.

4.6 Chapter Four – Study One Summary of Findings

Title: Kinematic Gait Adaptations in Transtibial Amputees During Rehabilitation

Patients: Fifteen recent transtibial amputation patients (12 men and 3 women). Mean ± SD Age 53.6 ± 14.1 years, height 1.73 ± 0.11 metres, mass 83.3 ± 20.1 kg.

Setting: Amputee rehabilitation.

Intervention: Early walking aid (EWA) – Amputee Mobility Aid (AMA) or Pneumatic Post-Amputation Aid (PPAM).

Comparison: Temporal-spatial (TSP) and kinematic variables during gait.

Main Findings:	Description
Increased walking velocity	Similar increases in walking velocity between groups. Statistically and clinically significant improvements throughout rehabilitation.
Step length and cadence	During EWA gait, AMA group took longer steps, PPAM group took faster steps.
Knee ROM	Increases in affected knee joint ROM during weight acceptance between receipt of functional prosthesis to discharge from rehabilitation. No differences in this effect between groups.
Asymmetry	Between limb differences in both TSP and kinematic variables reduced over time but were still present at discharge from rehabilitation.
Overall Summary	Different TSP and kinematic gait features were evident between groups during EWA use. Following receipt of a functional prosthesis, between group differences in gait were still present although at discharge, both groups displayed a similar level of walking ability. Our results did not show a clear benefit in gait following use of either EWA, which has significant implications to the NHS with regards to patient preference and cost.

5 CHAPTER FIVE – STUDY TWO. Changes in Self-Reported Generic Quality of Life in Transtibial Amputees During Rehabilitation.

5.1 Introduction

No study to date has documented the effect of different EWA use on transtibial amputee QOL or how QOL changes as transtibial amputees progress through rehabilitation. Understanding these relationships is important as patients' perceived QOL may affect their transition back into the workplace, engagement in physical and/or social activities and motivation to adhere to a programme of rehabilitation.

Therefore, the aims of the current study were two fold. The first aim was to investigate the changes that occurred in self-reported QOL in transtibial amputees as they progressed through rehabilitation. The second aim was to determine if and how these changes in self-reported QOL differed between patients who had previously used different types of early walking aid (EWA), namely the Post-Amputation Aid (PPAM Aid) and the Amputee Mobility Aid (AMA).

It was hypothesised that (1) QOL would increase during the course of rehabilitation, specifically the physical health aspect of QOL, as patients' mobility increased. It was also hypothesised that (2) patients using the AMA would display increased QOL during the early stages of the rehabilitation process as they would be able to practice a more natural gait pattern.

5.2 Methods

5.2.1 Patients

The patients assessed in the current study were the same patient group as in study one. Details of patient characteristics are provided in Table 4.1. Details of the inclusion and exclusion criteria have been outlined in Chapter Three, Section 3.2.2.

5.2.2 The Medical Outcomes Study Short Form-36

The medical outcomes study short form-36 (SF-36) (Appendix E) questionnaire is a multi-purpose health survey consisting of 36 items (Ware *et al.*, 2000). The SF-36 produces an eight-scale profile of health namely, Physical Functioning (PF), Role Physical (RP), Bodily Pain (BP), General Health (GH), Vitality (VT), Social Functioning (SF), Role Emotional (RE) and Mental Health (MH). It also produces summary components of physical (PCS) and mental health (MCS), as well as an overall or Total QOL score (Ware and Gandek, 1998) (Figure E.1). These scales and component summary scores can then be used for comparison against previous research findings.

5.2.3 Experimental Design and Protocol

The experimental design of the current study was consistent to that of study one. Patients were required to complete one SF-36 questionnaire at five standardised time-points (visits one to five) during their rehabilitation following amputation, typically, upon arrival to rehabilitation sessions. The reasoning for this being that discussion of health-related issues or interaction with physiotherapists or researchers may have influenced a patient's response to the questionnaire. Patients were encouraged to answer questions based upon their own interpretation and, if required, questions were repeated verbatim by the researcher or physiotherapist.

5.2.4 Data Analysis

The SF-36 scoring system is such that a higher score indicates an improved health state on that scale. For example, an individual with a bodily pain score of 84 is deemed to experience less pain than an individual scoring 19. The paper hard copies of SF-36 questionnaires were collected and scored by the same researcher and data manually inputted into a Microsoft Excel workbook (Microsoft, Reading, UK). Scoring of the SF-

36 follows a three-step procedure according to the author guidelines, item recoding, computing raw scale scores and computing transformed scale scores (Ware *et al.,* 2005). The item recoding procedure involved taking the manually inputted raw precoded data and assigning a recoded value to each item score. Once the data has been recoded a raw scale score was calculated, a simple algebraic sum of the item responses for a particular scale. Once the raw scale score had been calculated it was then transformed using the formula below:

$$\text{Transformed Scale} = \left[\frac{(\text{Actual Raw Score - Lowest Possible Raw Score})}{\text{Possible Raw Score Range}}\right] \times 100$$

Transformation of the raw scale scores to a 0-100 scale, allowed for comparison between studies and those using different or previous versions of the SF-36 questionnaire (Ware *et al.,* 2000). The transformed scores were the scores that were reported for each scale. As well as obtaining the transformed scores for each of the eight scales of the SF-36 questionnaire, it is possible to compute higher order dimension scores for Physical and Mental Health as well as an overall of Total SF-36 score. These higher order dimensions were named the Physical Component Summary (PCS) and the Mental Component Summary (MCS) and were computed as an arithmetic mean of their associated scales scores. The Total SF-36 score was the arithmetic mean of the PCS and MCS.

5.2.5 Statistical Analyses

Group averaged means for patients in the current study were used for statistical analysis. A linear mixed model analysis (LMM) was employed, Group (AMA, PPAM) * Time (Visit Number), with repeated measures on the last factor. This design allowed for the comparison of both the changes in QOL during rehabilitation and any differences present between groups (Brown and Prescott, 1999). Each feature of the design (Group and Time) was modelled as a fixed effect with the appropriate model being selected according to the lowest value for Hurvich and Tsai's Criterion (AICC). In the instance of a significant main effect or interaction effect, post-hoc comparisons were conducted using a Sidak adjustment in SPSS v.17.0 (SPSS Inc., Chicago, USA). The alpha level of statistical significance was set at $p \leq 0.05$.

5.3 Results

Results from statistical analyses are provided in Table 5.1. There were significant time main effects for the Physical Functioning (PF), Social Functioning (SF) and Role Emotional (RE) scales ($p<0.05$). Post-hoc analysis of PF results highlighted that significant increases occurred between visits one, two and three compared to four ($p<0.01$), as well as visit one – five ($p=0.04$). This increase in physical functioning, observed in Figure 5.1, was likely to be related to patients better adapting to their biomechanical constraints during rehabilitation. The post-hoc analysis of SF and RE did not reveal where the significant time main effect had occurred.

Figure 5.2 displays the changing nature of the eight scales of the SF-36 as patients progressed through rehabilitation. In general, most of the eight scales showed an observable and steady increase in SF-36 scores across visits, suggesting that QOL improved as patients progressed through rehabilitation.

Significant time main effects were observed for both physical and mental higher order components. Post-hoc analysis revealed these differences to be between visits one and four ($p=0.02$) for the PCS, visits one and five for the MCS ($p=0.03$) and between visits one and four and one and five (both $p=0.02$) for Total SF-36. A pattern of increase across visits similar to the eight scales, was observed in the PCS, MCS and Total SF-36. Figure 5.1 displays clear increases in PCS, MCS and Total SF-36, indicating that increases in both physical and mental health contributed to the improvement in overall QOL. From Figure 5.1, it was observed that MCS scores were generally higher than

PCS scores. This would suggest that mental health was a larger component of QOL for the current group of amputees.

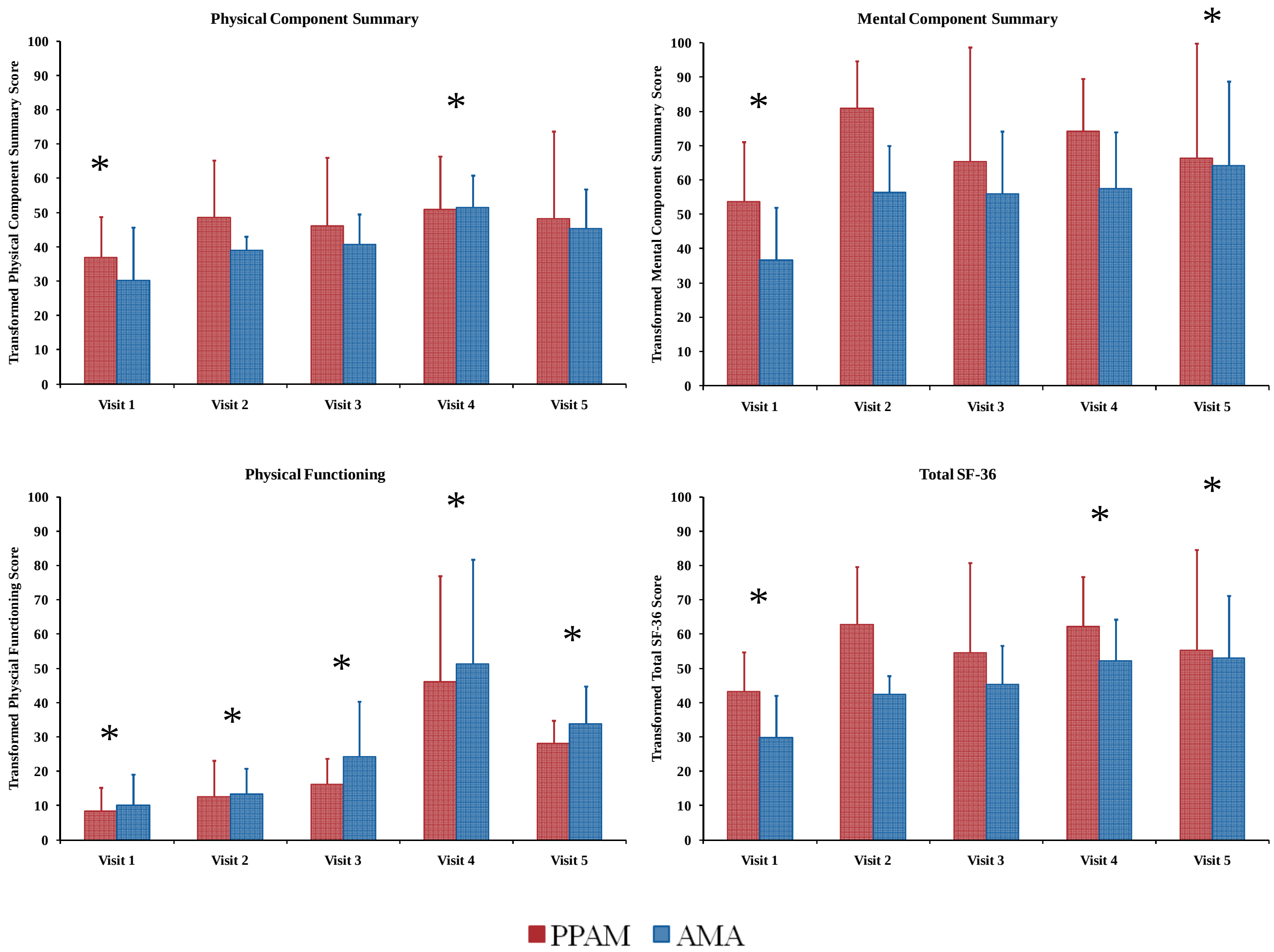


Figure 5.1 Group mean transformed scores of the Physical Component Summary (PCS), Mental Component Summary (MCS), Physical Functioning (PF) and Total SF-36 presented from visits one-five. * Indicates a significant time main effect.

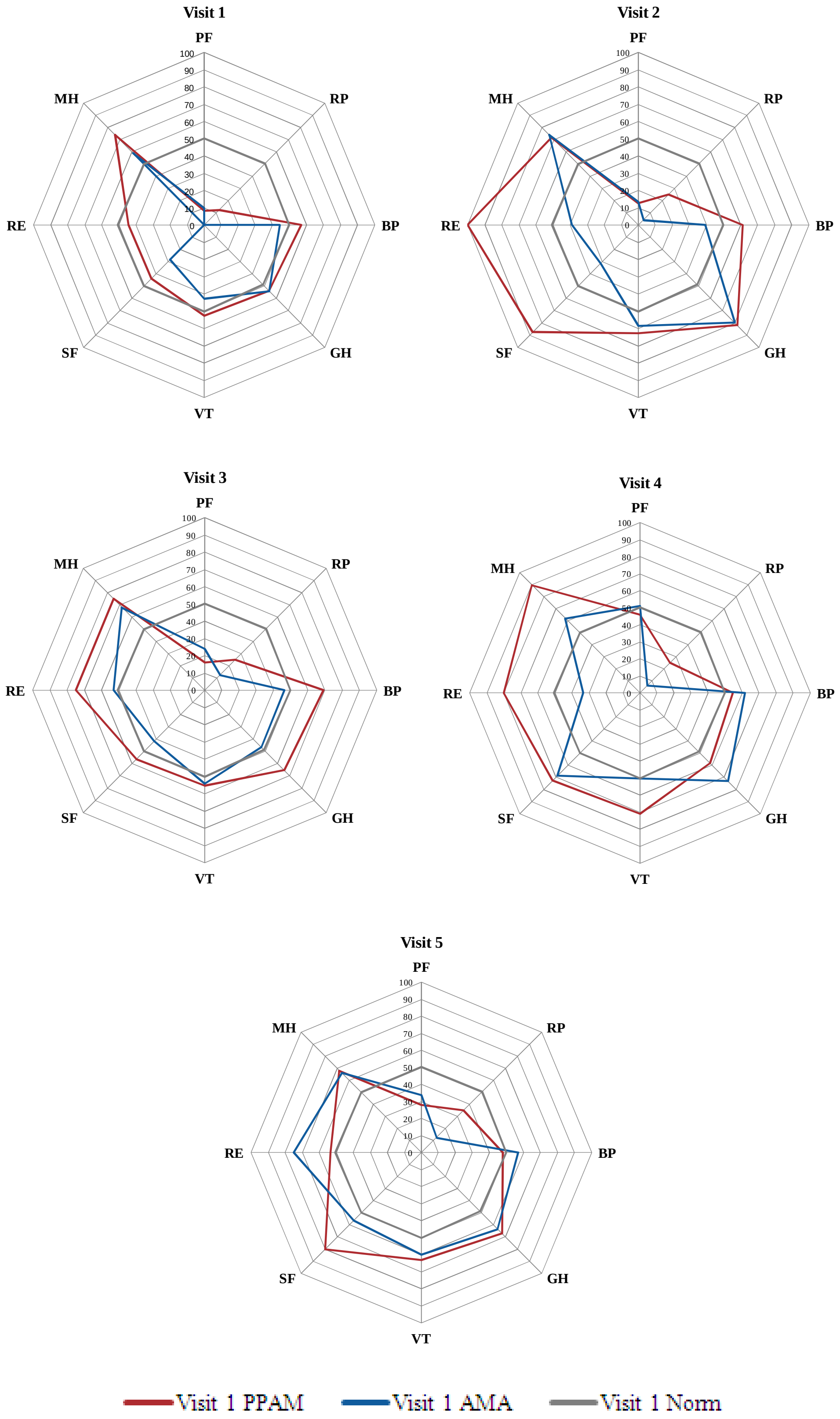


Figure 5.2 Target plots of group mean transformed scores from 8 scales of SF-36 from visits one to five. Age-matched normative data are presented to provide a visual comparison (Ware *et al.*, 2000). Scores closer to outer border of plots relate to increased QOL in that scale.

Table 5.1 Statistical breakdown of SF-36 questionnaires responses. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant main effect.

	Main Effects				Interaction Effects	
	Time		Group		Time * Group	
Item	F	P	F	P	F	P
Physical Functioning (PF)	(4,30.96) = 9.21	0.00*	(1,13.58) = 0.38	0.56	(4,30.96) = 0.09	0.98
Role Physical (RP)	(4,19.63) = 0.65	0.63	(1,7.18) = 5.02	0.06	(4, 19.63) = 0.26	0.90
Bodily Pain (BP)	(4,25.92) = 0.35	0.84	(1,11.66) = 0.33	0.58	(4,25.92) = 0.36	0.84
General Health (GH)	(4,24.05) = 0.67	0.62	(1,11.54) = 0.23	0.64	(4,24.05) = 0.26	0.90
Vitality (VT)	(4,22.74) = 2.40	0.08	(1,11.26) = 0.35	0.57	(4,22.74) = 1.99	0.13
Social Functioning (SF)	(4,26.42) = 3.32	0.03*	(1,11.01) = 3.52	0.09	(4,26.42) = 0.99	0.43
Role Emotional (RE)	(4,24.63) = 3.40	0.02*	(1,13.01) = 0.54	0.48	(4,24.63) = 1.18	0.35
Mental Health (MH)	(4,24.76) = 0.47	0.76	(1,12.22) = 0.21	0.66	(4,24.76) = 1.64	0.2
Dimension	F	P	F	P	F	P
PCS	(4,23.51) = 3.69	0.02*	(1,10.41) = 0.41	0.54	(4,23.51) = 0.25	0.91
MCS	(4,24.10) = 3.10	0.03*	(1,11.80) = 0.72	0.41	(4,24.10) = 0.40	0.81
Total SF-36	(4,24.02) = 4.28	0.01*	(1,11.18) = 1.14	0.31	(4,24.02) = 0.11	0.98

5.4 Discussion

Although the literature has reported on the QOL in transtibial amputees (Asano *et al.,* 2008; Zidarov *et al.,* 2009; Van der Schans *et al.,* 2001; Pezzin *et al.,* 2000; Legro *et al.,* 1999) fewer studies have assessed QOL during the rehabilitation of amputees (Brooks *et al.,* 2001) and none have reported the effects of using different EWAs during the rehabilitation process. Therefore, the current study investigated the changes in self-reported QOL as transtibial amputees progressed through rehabilitation. The current study also investigated how these changes in self-reported QOL differed between patients who used either the PPAM Aid or the AMA previously.

5.4.1 Physical Health Scales

Statistically significant increases in physical functioning scores were observed during rehabilitation. This partially supports the first hypothesis and suggests that patients' mobility improved across visits. Physical functioning and role physical scored lowest of all eight SF-36 scales with bodily pain remaining in comparison to normative population (Ware *et al.,* 2000). There were no significant group differences or interaction effects in scales pertaining exclusively to physical health (Table 5.1, physical functioning, role physical and bodily pain). This resulted in the second hypothesis being rejected as seemingly neither EWA produced greater benefits in terms of physical health QOL. Visual inspection of Figure 5.2 revealed that two out of the three scales relating to physical health (physical functioning and role physical) were scored lower than age-matched normative data (age range 45-54 years of age, Ware and Kosinski, 2007) with bodily pain being around the same value (Ware *et al.,* 2000). This may be expected as the amputees in the current study were still adapting to significant mechanical alterations that impacted upon their physical capabilities and mobility.

Interestingly, bodily pain was not reported to be as low as physical functioning or role physical and did not change significantly over time (Table 5.1), reflecting constant levels of bodily pain with increasing physical functioning and role physical scores. An interpretation of this finding could be that as patients progressed through rehabilitation their ability to perform physically orientated tasks increased. This is thought to be linked to increased walking speed during rehabilitation (Brooks *et al.,* 2001 and Jones *et al.,* 2001). Patients may have developed increased pain tolerance or experienced a reduction of phantom limb pain. Previous studies have reported reductions in phantom pain (Houghton *et al.,* 1994) and stump pain during weight-bearing (Jones *et al.,* 2001) following amputation. Therefore, levels of reported bodily pain may remain the same due to an increased pain tolerance being matched against an increasing physical capacity. It is not clear if an increased pain tolerance leads to an increased physical ability or vice versa however, this relationship would benefit from further investigation, perhaps incorporating analyses of physical activity and specific indices of pain. This relationship between physical capacity and pain tolerance has rehabilitation implications for those involved in the care of amputees as a focus on improving the antecedent may lead to gains in the other factor. There was also no significant group main effect for bodily pain, indicating that neither EWA was more beneficial in terms of bodily pain reported during rehabilitation. This finding supported the rejection of the second hypothesis.

At discharge from rehabilitation both groups of amputees reported physical functioning and role physical to be lower than age-matched normative data and QOL data presented for traumatic amputees a number of years following amputation (7.5 years) (Pezzin *et al.,* 2000). This suggests that patients' physical ability has the potential to improve further, even following discharge from rehabilitation.

5.4.2 Mental Health Scales

Social functioning and role emotional displayed significant time effects, generally increasing across visits, although post-hoc analyses did not reveal where these differences occurred. A possible reason for this may be the variability present in the data. This finding partially supports the first hypothesis as social functioning and role emotional related QOL improved during rehabilitation. Mental health scores from both groups of amputees in the current study remained fairly consistent throughout rehabilitation. The values observed were comparable with amputees assessed a number of years post-amputation (mean 7.5 and median 10 years respectively) (Van der Schans *et al.,* 2001; Pezzin *et al.,* 2000). This suggested that mental health in amputees remained relatively stable following discharge from rehabilitation and was not affected by changes in physical ability. Visual inspection of Figure 5.2 revealed that scales pertaining exclusively to mental health were scored higher than age-matched normative data (Ware *et al.,* 2000). Although, the PPAM group generally scored higher in these scales, there were no significant group differences or interaction effects in scales pertaining to mental health (Table 5.1, social functioning, role emotional and mental health). These findings refute the second hypothesis as scales relating to mental health were not reported to be higher during rehabilitation in the AMA group.

5.4.3 General Health and Vitality

General health and vitality do not belong to either higher order dimension as they incorporate aspects of both physical and mental health. The scales of general health and vitality remained fairly consistent throughout rehabilitation (Figure 5.2 and Table 5.1). There were no group differences in either of these scales suggesting that initial walking with either EWA did not influence patients' responses to items within each scale. Figure 5.2 shows that general health and vitality, in the current patient groups, were generally higher than in an age-matched normative group. Previous research has argued that lower limb amputation, whilst not being significantly different in terms of QOL when compared to limb salvage surgery, may be beneficial in avoiding further complications to those with severe lower limb damage requiring treatment (Hoogendoorn and Van Der Werken, 2001). This may be the case in the current patient group given the causes of amputation. Higher levels of general health and vitality may have been reported as a result of improved QOL with reference to their previous physical condition or disease state.

5.4.4 Component Summary Scores and Total SF-36 Score

The first hypothesis was supported by the observation of a significant time effect in component summary scores from both groups. This indicated that both physical and mental health improved from the start to the end of rehabilitation. This in turn led to a significant time effect for Total SF-36 score in the current patient groups. Total SF-36 score increased as a function of both improving mental and physical health. The lack of a group effect and subsequent interactions effects confirmed that neither EWA was better at increasing physical or mental health. This finding refuted the second hypothesis. To this end, it could be suggested that EWA selection can be made independent of concerns of its effects on QOL. As previously reported, significant increases in physical functioning during rehabilitation were reflected in similar results for the PCS (Ware and Kosinski, 2001).

Consistent with previous studies of lower limb amputees (Pezzin *et al.,* 2000; Smith *et al.*, 1995), mental health was significantly better than physical health in the current patient group. In the current patient group, this could be interpreted in a similar fashion to the results for general health and vitality. The event of amputation often occurs as a result of pre-operative lower limb dysvascularity which can be alleviated following various surgical procedures such as limb revascularisation (Hoogendoorn and Werker, 2001; Albers *et al.*, 1996; Thompson *et al.*, 1995). It seems in the case of amputation, although physical capacity was reduced, mental health was improved as patients may have referenced their current health (both physical and mental) to their pre-operative states, which in many cases was likely worse pre- than post-amputation.

5.5 Conclusion

The current study adds to our understanding of how QOL is affected by the event of amputation and how it changes following a course of rehabilitation. Overall increases in physical, mental and overall health lead to the first hypothesis being accepted. The current study also found that initial gait retraining in transtibial amputees using an articulated EWA (AMA) did not produce significant benefits in terms of QOL at any stage during the rehabilitation process, when compared to the use of a non-articulated EWA (PPAM Aid). This resulted in the second hypothesis being rejected and implied that a clinician's selection of an EWA can focus upon variables other than attempted gains in QOL. For the current patient group, support was found for the sensitivity of SF-36 use as similar profiles of change in sub-scales were also reported in component summary scores. Lastly, it was observed that mental health in transtibial amputees was higher than physical health, partially supporting the first hypothesis. This suggested that a rehabilitation programme focussing upon improving physical health aspects would elicit further increases in overall QOL in transtibial amputees.

5.6 Chapter Five – Study Two Summary of Findings

Title: Changes in Self-Reported Generic Quality of Life in Transtibial Amputees During Rehabilitation

Patients: Fifteen recent transtibial amputation patients (12 men and 3 women). Mean ± SD Age 53.6 ± 14.1 years, height 1.73 ± 0.11 metres, mass 83.3 ± 20.1 kg.

Setting: Amputee rehabilitation.

Intervention: Early walking aid (EWA) – Amputee Mobility Aid (AMA) or Pneumatic Post-Amputation Aid (PPAM).

Comparison: A generic quality of life measure (QOL) (SF-36).

Main Findings:	Description
Overall QOL	During rehabilitation, QOL improved in both groups.
Components of QOL	Both physical and mental health scales increased similarly between groups during rehabilitation. Mental health tended to be higher than physical health.
Group differences	Quality of life was similar between groups during rehabilitation, despite some visible differences.
Overall Summary	Overall, physical, mental and total QOL improved during rehabilitation with mental health tending to be higher than physical health. Using the AMA did not produce significant benefits in terms of QOL at any stage during the rehabilitation process. The selection of EWA may be made independent of concerns of effects on QOL. Rehabilitation focussed upon increasing physical health may elicit improvements in overall QOL.

SUMMARY – AMPUTEES DURING REHABILITATION

Studies one and two investigated the kinematic gait adaptations and self-reported QOL in two transtibial amputee groups walking with two different EWAs as they progressed through rehabilitation. These two studies have also investigated the effect of using EWAs with an articulated vs. non-articulated knee and the effect upon the aforementioned variables.

During rehabilitation, it was seen that walking proficiency improved as did self-reported QOL. Interestingly, these results did not differ according to the type of EWA that was used prior to receiving a functional prosthesis. These two studies showed that, at discharge from rehabilitation, patients walked proficiently and similarly irrespective of which EWA was used previously. This is not unusual in that patients were discharged by the same physiotherapy team once a satisfactory level of ambulation had been achieved. However, their walking performance was reduced when compared to more experienced amputees reported in the previous literature (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers *et al.,* 1998; Nolan *et al.,* 2003). Previous reports in the literature of increased mental health when compared to physical health in lower limb amputees were supported by the observations within the current studies (Legro *et al.,* 1999; Pezzin *et al.,* 2000; Van der Schans *et al.,* 2002; Asano *et al.,* 2008; Zidarov *et al.,* 2009).

These results suggested that at discharge from rehabilitation, as is required by the physiotherapy team, transtibial amputees had reached a satisfactory level of functioning that had greatly improved from the time of their first steps following amputation. The rehabilitation programme they attended had a beneficial impact on both their physical functioning when using a prosthesis, as well as QOL. However, it is clear that further mechanical adaptation must occur following discharge from rehabilitation, suggesting that the re-learning process continues post-discharge. It is also likely that changes in QOL will occur with changing physical ability, as was seen during rehabilitation.

It is not yet known what changes in biomechanics and QOL occur in the timeframe following discharge from rehabilitation, as transtibial amputees continue to adapt to their mechanical constraints.

The next series of studies aimed to address this issue with the use various biomechanical and psychometric tools. Studies investigated the biomechanics, balance performance and postural control, along with QOL during this potentially crucial period of time within the transtibial amputee re-learning process.

6 CHAPTER SIX – STUDY THREE. Biomechanical Adaptations in Gait and Activities of Daily Living of Transtibial Amputees Following Discharge from Rehabilitation.

6.1 Introduction

Previous research has not investigated the gait re-learning process that occurs immediately following discharge from rehabilitation, as amputees are faced with an ever increasing number of more complex movement tasks. Understanding how amputees adapt to movement challenges during this time period, as they learn to successfully and comfortably perform ADLs, could have important implications for both the amputee and therapists involved in amputee outpatient care and rehabilitation.

The aim of the current study was to explore the adaptations in transtibial amputees' movement patterns following discharge from rehabilitation (from discharge up to six months post-discharge) in three specific ADLs: 1) level gait, 2) level gait whilst crossing an obstacle, 3) and gait when stepping to and from a new level.

The literature has shown that amputees with > 1 year experience of prosthetic use are likely to display increased function when compared to recent transtibial amputees (Sanderson and Martin, 1997; Nolan *et al.*, 2003). In addition literature has reported that amputees are able to negotiate obstacles and stairs effectively (Hill *et al.*, 1997; Powers *et al.,* 1997). Therefore, it was hypothesised that during the time period following discharge from rehabilitation, gait proficiency (1) and performance of ADLs such as crossing obstacles (2) and stepping to and from a new level (3) would improve in terms of walking velocity.

6.2 Methods

6.2.1 Participants

Seven participants (all male) (Table 6.1) were recruited into the study between May 2008 and December 2009. These participants had previously followed a course of rehabilitation within the Department of Physiotherapy, Castle Hill Hospital, Hull and East Yorkshire NHS Trust, as outlined in Chapter Four, Section 4.2. Participants were recruited within one month of being discharged from rehabilitation consented to be contacted at the last (discharge) physiotherapy treatment. Initially, participants had the study explained to them by physiotherapists and agreed to be contacted by the principle investigator. Participants were contacted and attended data collection in the Human Performance Laboratory, at which point the study was detailed and written informed consent collected. Inclusion and exclusion criteria of participants in the current study have been described in Chapter Three, Section 3.2.2.

Table 6.1 Individual characteristics and prosthetic components of unilateral transtibial amputees.

Gender (Male/Female)	Age (years)	Height (m)	Mass (kg)	Amputated Limb (Right/Left)	Cause of Amputation	Functional Prosthetic Components	
M	44	1.77	76.5	R	Non-Vascular	Renegade Freedom Foot*	All ankle feet complexes allowed for similar axial movement with the addition of specific differences highlighted.
M	63	1.74	83.7	L	Non-Vascular	Tres Foot with torque absorber	
M	44	1.82	81.0	R	Non-Vascular	Renegade Freedom Foot*	
M	75	1.93	101.9	L	Vascular	Multiflex Ankle and Foot	
M	50	1.83	106.6	R	Vascular	Senator Freedom Foot‡	
M	41	1.92	95.4	R	Vascular	Multiflex Ankle and Foot	
M	70	1.74	96.7	R	Vascular	Multiflex Ankle and Foot	
(Mean ± SD) All Participants	56.1 ± 14.9	1.82 ± 0.08	91.7 ± 11.4				

*Shock absorbing ankle foot complex, ‡Energy returning ankle foot complex for low to moderately active participants.

All participants used the same socket interface device and pylons as outlined in Chapter Three, Section 3.2.3. Only the ankle foot complexes differed and were provided by RSL Steeper Ltd (www.rslsteeper.com).

From discharge to six months post-discharge, participants attended 9.3 ± 4.6 appointments at the Regional Limb Centre. Repairs and adjustments of the prosthesis accounted for 42% of these visits, consultant examinations 37%, fitting and delivery of a prosthetic component 18%, with castings making up 3% of the total visit number.

6.2.2 Experimental Design and Protocol

Data were collected as participants attended sessions at the Human Performance Laboratory, Department of Sport, Health and Exercise Science, University of Hull. The experimental design of current study was a longitudinal repeated measures design where participants attended a standardised number of data collection sessions at one, three and six months following discharge from rehabilitation. Two patients attended a session at twelve months post-discharge. These time points were selected in order to assess the longitudinal adaptations in movement following discharge from rehabilitation.

Participants' height (m) and mass (kg) were recorded using a free-standing height measure and beam column scale (Seca, Birmingham, UK). A ten camera motion capture system synchronised with two force plates captured 3D kinematic and kinetic data at sampling frequencies of 100 Hz and 1000 Hz respectively, using QTM software. Details of these methodologies were outlined in the Chapter Three, Section 3.3. The cameras were set up with multi-planar views in order to allow for a capture volume of approximately $80m^3$, ideal for gait analysis. This configuration was selected as it provided a large capture volume in which to capture various ADLs as well as gait related tasks (Figure 6.1).

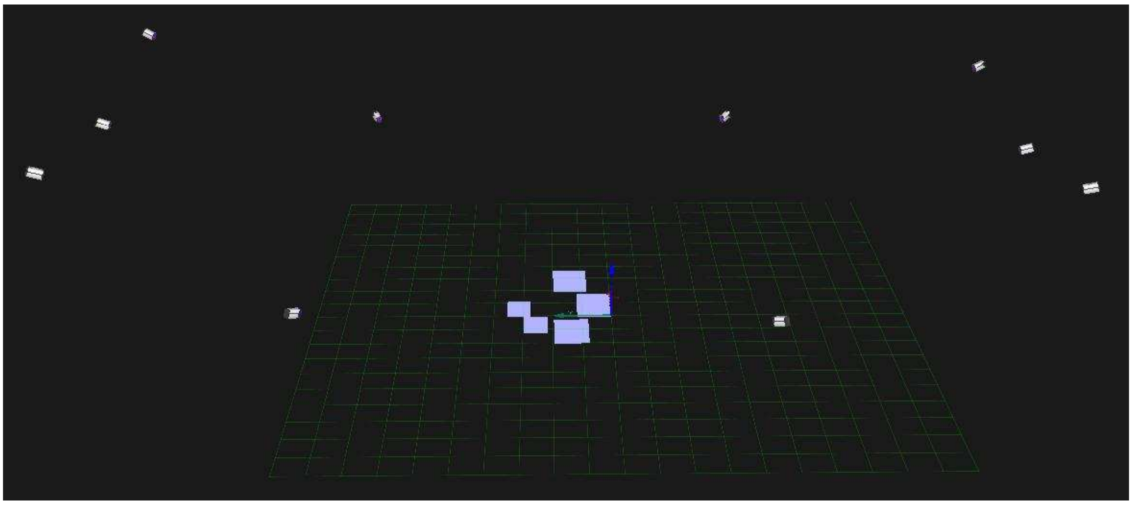


Figure 6.1 The ten camera ProReflex© system setup in the Human performance Laboratory at the Department of Sport, Health and Exercise Science, University of Hull

Participants wore their own comfortable, flat footwear during all data collection sessions. Participants were able to fit and re-adjust their own prostheses, as is the case on a daily basis, in order to gain a comfortable fit prior to the commencement of data collection. Once this was achieved, 14mm reflective markers were attached to specific anatomical landmarks by the same investigator according to the six degrees of freedom marker model set, described in Chapter Three, Section 3.3.5. Marker placement on the affected limb was estimated from intact limb anatomical landmarks, a procedure previously reported in the literature (Powers *et al.*, 1998).

Participants were required to perform a number of gait tasks and ADLs at a self selected velocity, resting as required. A minimum of five trials were recorded per task and the tasks were standardised in the following order; level gait, obstacle crossing and stepping gait. These tasks were selected as it was possible to recreate these everyday situations that participants were likely to encounter, in a controlled laboratory environment.

In order to recreate the stepping tasks, a custom raised surface walkway was constructed with a step height that approximated roadside kerbs in the UK (BS 5395-1 2000, British

Standards Institute, 2000) (Appendix K). In order to recreate obstacle crossing, an obstacle was constructed from polystyrene allowing for movement and/or breakage should participants have touched or stood on the obstacle (Appendix K). The height of the obstacle was selected in order to be higher than most objects that are likely to be on the floor in an average home, e.g. shoes, children's toys. This height also corresponded to obstacle heights previously reported (Vrieling *et al.,* 2009; Vrieling *et al.,* 2007). As previously noted, the width of the obstacle was purposefully large to prevent amputees from negotiating the obstacle by walking around it (Vrieling *et al.,* 2007; Hill *et al.,* 1997).

Performance of the obstacle crossing task required participants to walk towards, step over and walk away from an obstacle (Appendix L). During the stepping gait task participants walked towards and stepped onto the walkway, they then continued to walk, turned and then walked off of the walkway (Appendix L). This allowed for the capture of continuous gait while stepping onto and from a new level.

6.2.3 Data Analysis

Data frames of movement trials were analysed and averaged for all tasks. For ADLs, data from the transition step was analysed, as participants crossed the obstacle or stepped to or from the raised surface. The transition step represented the main functional difference between level gait and various ADLs. Movement events were identified using kinetic data and in its absence, visually from kinematic data. Temporal-spatial variables of walking velocity, step and stride length, cadence, double limb support and relative stance duration were calculated. Kinematic joint angle data from the ankle, knee, hip and pelvis were measured in the frontal and sagittal planes. In addition, the vertical displacement of the toe and heel and horizontal displacement of the toe were calculated during obstacle crossing and stepping tasks. Joint moment and power data were calculated for the ankle, knee and hip. Support moments were calculated for each limb, further information on support moments is provided in Chapter Three, Section 3.4. GRF data in the three orthogonal directions were normalised by dividing by body weight. All data were group mean (±SD) and normalised to the gait cycle for the intact and affected limbs.

6.2.4 Statistical Analysis

Group averaged means for participants in the current study were used for statistical analysis. A linear mixed model analysis (LMM) was employed, Limb (Affected, Intact) * Time (One Month, Three Months and Six Months) with repeated measures on the last factor. This design allowed for the analysis of changes in multiple gait variables hypothesised a priori (Brown and Prescott, 1999). Each feature of the design (Time and Limb) was modelled as a fixed effect with the appropriate model being selected according to the lowest value for Hurvich and Tsai's Criterion (AICC). In the instance of a significant result, post-hoc comparisons were conducted using a Sidak adjustment in SPSS v.17.0 (SPSS Inc., Chicago, USA). The alpha level of statistical significance was set at $P \leq 0.05$.

6.3 Results

Group mean (±SD) were presented from all time points following discharge from rehabilitation for all participants. Data were also presented from a 12 month visit for two participants (one and two) although these results were not analysed statistically.

6.3.1 Level Gait

6.3.1.1 Temporal-Spatial Variables

Temporal-spatial variables are presented in Table 6.2 with complete statistical analyses provided in Table 6.4. Participants walking velocity increased by 14% at six months following discharge and although this was not statistically significant, the 0.13 m/s increase between one and six months following discharge represents a clinically meaningful increase. Post-hoc comparisons for the statistically significant time effect showed increases between one month and three (p=0.04) (p=0.02) and one and six months (p=0.01) (p=0.02) in step length and stride length respectively, although no limb main effect was observed. Post-hoc comparisons for the significant time and limb main effects revealed statistically significant decreases in relative stance duration between one and three (p=0.04) and one and six months (p=0.01) with differences between the intact and affected limbs (p=0.03). There was a 4% decrease in relative double limb support time, although this was not statistically significant.

Table 6.2 Mean (SD) temporal-spatial variables of level gait. Data are presented for the affected and intact limb separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Walking Velocity (m/s)		0.93 (0.17)	1.04 (0.17)	1.06 (0.20)	1.17 (0.03)
Relative Double Limb Support (%GC)		34.65 (5.20)	31.53 (4.27)	30.78 (6.26)	24.78 (1.19)
Stride Length (m)		1.18 (0.13)	1.28 (0.18)	1.31 (0.19)	1.40 (0.07)
Step Length (m)	Affected	0.58 (0.06)	0.65 (0.10)	0.66 (0.11)	0.72 (0.00)
	Intact	0.59 (0.08)	0.63 (0.08)	0.64 (0.09)	0.67 (0.07)
Cadence (Step/Min)	Affected	94.5 (7.9)	96.8 (2.1)	96.8 (5.4)	98.6 (6.2)
	Intact	93.1 (11.0)	96.6 (6.1)	96.4 (7.4)	102.6 (1.7)
Relative Stance Duration (% GC)	Affected	67 (3.6)	67 (2.5)	64 (3.2)	61 (0.5)
	Intact	67 (3.0)	67 (3.1)	66 (3.7)	63 (1.5)

*Data from two participants, not included in statistical analyses.

6.3.1.2 Joint Kinematics

Joint kinematics are presented in Figure 6.2 (sagittal plane) and Appendix N (frontal and transverse) with complete statistical analyses provided in Tables 6.4 and 6.5.

Ankle range of motion (ROM) during stance phase was statistically lower in the prosthetic ankle than the intact ankle joint ($p<0.01$). This likely to be due to the reduced plantarflexion apparent during early stance phase on the affected side (Figure 6.2). Figure 6.2 displays the large and statistically significant difference in peak ankle plantarflexion during swing phase between limbs ($p<0.01$). This is unsurprising, given that active plantarflexion during swing phase was not possible due to the prosthetic components apparent in the prosthetic limb. Peak dorsiflexion during stance phase was similar between limbs (Figure 6.2).

The observed increase in knee ROM during loading response in the intact limb when compared to the affected limb was significant ($p<0.01$) (Figure 6.2). Slight increases in knee ROM during loading response resulted in a significant time main effect between one month and six months post-discharge ($p=0.02$). In addition, knee ROM during single limb support was also statistically greater on the intact side than the affected side ($p=0.01$). Knee flexion during swing phase was comparable between limbs.

Intact limb hip flexion seemed to increase at six months post-discharge (Figure 6.2) and with hip abduction profiles differing between limbs (Appendix N). However, no statistically significant differences were found in hip or pelvis kinematics in any plane, reflected by Figure 6.2 and Appendix N.

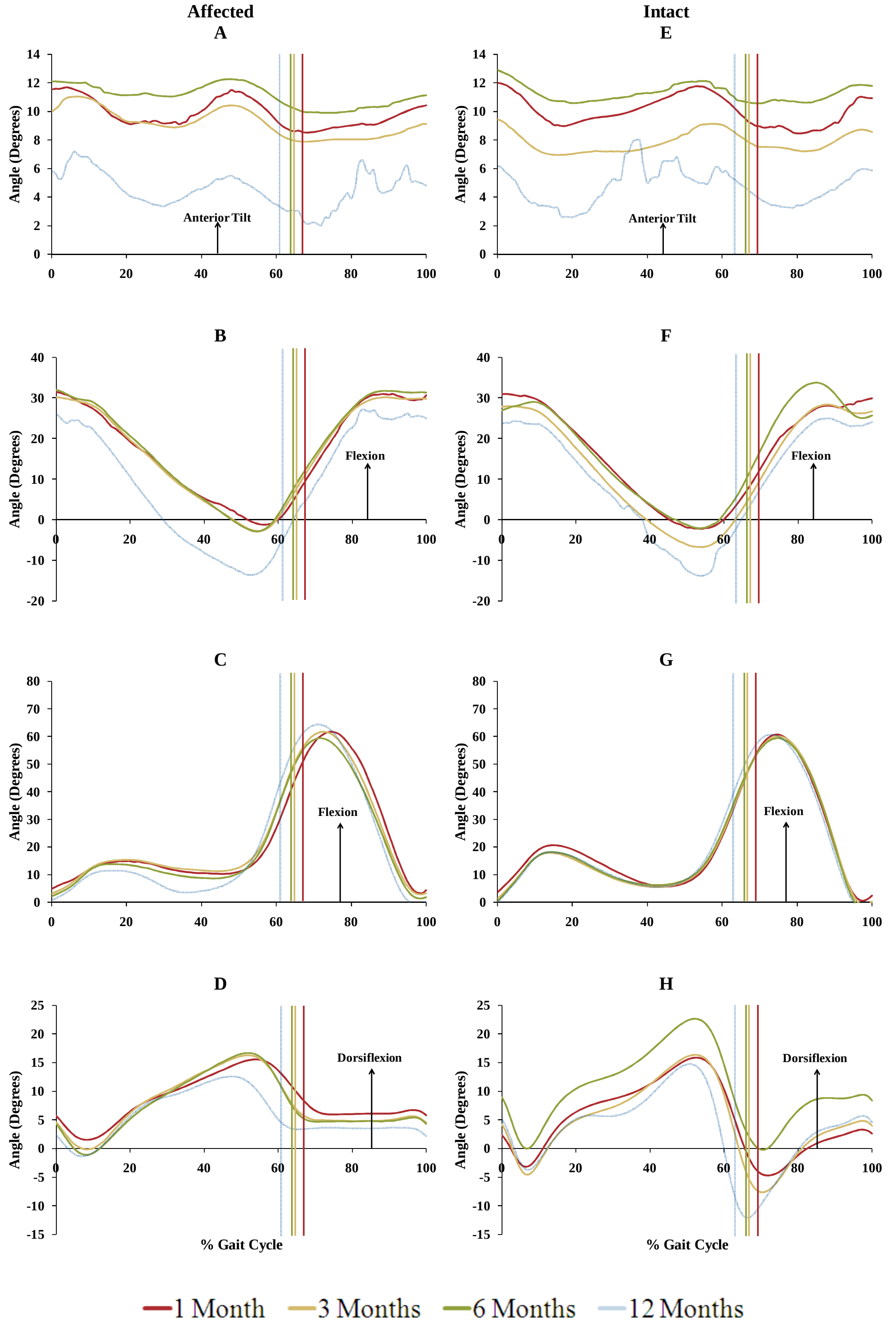


Figure 6.2 Group mean sagittal plane kinematics of the affected limb pelvis (A), hip (B), knee (C) and ankle (D) and intact limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during level gait. Vertical lines represent toe off. Data at 12 months from n=2.

6.3.1.3 Ground Reaction Forces

Loading and decay rates are presented in Table 6.3, ground reaction force (GRF) data are presented in Figure 6.3 with complete statistical analyses provided in Table 6.5.

As can be observed in Table 6.3, load rate was significantly higher in the intact limb than the affected limb (p=0.01), although loading rate did not increase significantly over time. Figure 6.3 illustrates the statistically significant increased initial peak vertical (p=0.04) and posterior (p=0.01) GRFs on the intact limb when compared to the affected limb.

Table 6.3 Mean (SD) loading and decay rate of level gait. Data are presented for the affected and intact limb separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Load Rate (BW/s)	Affected	5.0 (0.4)	5.0 (0.6)	5.6 (1.7)	5.5 (0.0)
	Intact	6.2 (1.8)	6.4 (1.2)	6.7 (1.1)	7.1 (1.0)
Decay Rate (BW/s)	Affected	5.1 (0.4)	5.4 (0.5)	5.0 (0.8)	6.5 (0.0)
	Intact	4.9 (1.3)	5.3 (1.0)	5.5 (1.7)	7.2 (0.7)

*Data from two participants, not included in statistical analyses

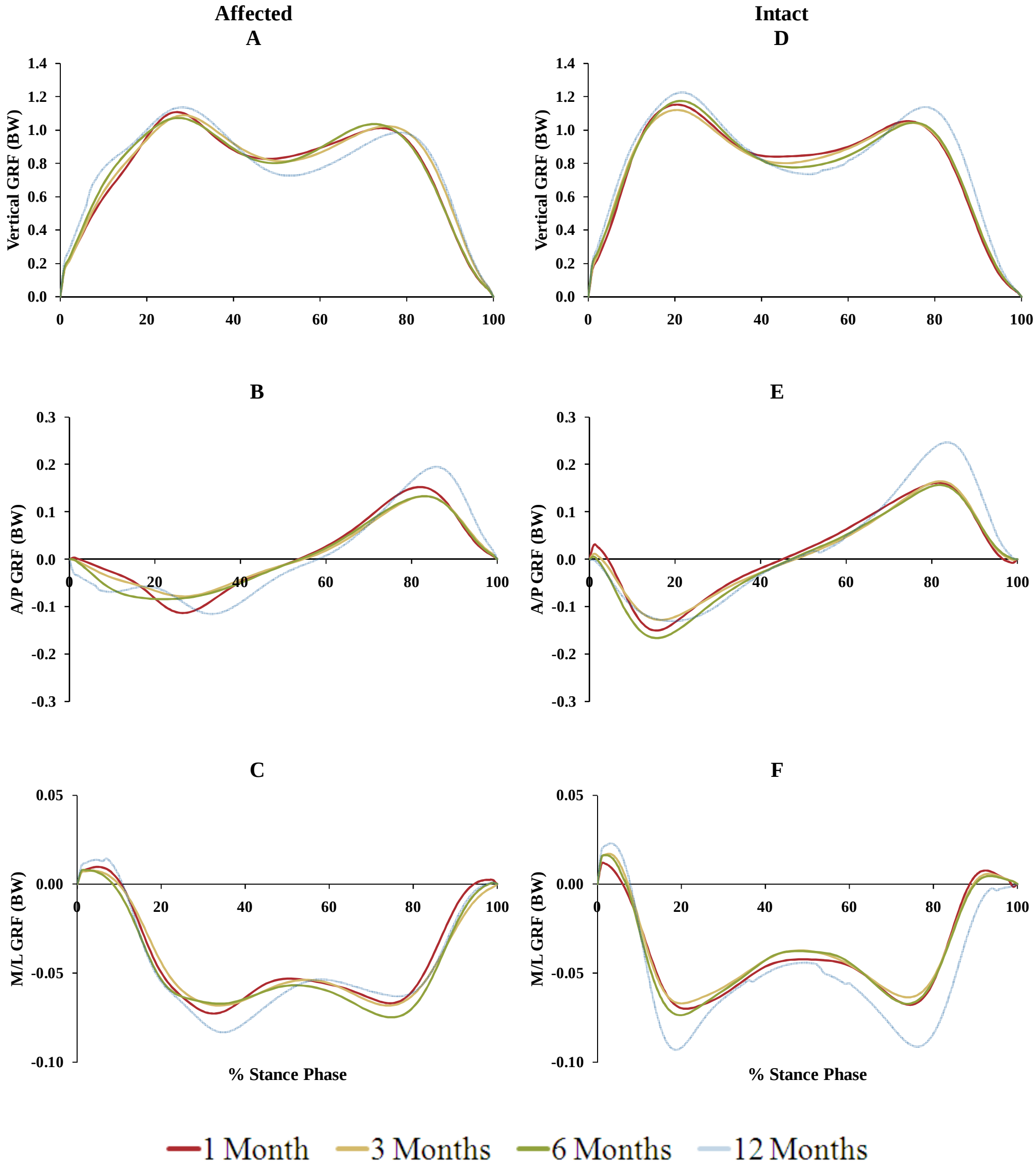


Figure 6.3 Group mean vertical, anterior-posterior and medial-lateral ground reaction forces for the affected (A, B and C) and intact (D, E and F) limbs. All data normalised and presented as times body weight (BW). Time normalised to 100% of stance phase during level gait. Data at 12 months from n=2. Vertical, anterior and lateral are positive.

6.3.1.4 Joint Kinetics

Sagittal plane and frontal plane joint moments are presented in Figure 6.4, support moments and joint powers are presented in Appendix O and Figure 6.5 respectively, with complete statistical analyses provided in Tables 6.5 and 6.6.

Ankle plantarflexor moment profiles were similar between limbs (Figure 6.4). Post-hoc analysis of peak ankle dorsiflexor moment during loading response revealed a significant increase between three and six months (p=0.02).

Intact and affected limb sagittal plane knee moment profiles followed similar trends while differing in certain peak magnitudes (Figure 6.4). A significant limb main effect highlighted the increased peak knee extensor moment during loading response in the intact limb when compared to the affected limb (p=0.02). This variable also increased significantly over time, post-hoc analyses revealing this difference to be between three and six months post-discharge (p=0.03).

Sagittal plane hip moments also displayed similar trends while differing in peak magnitudes between limbs (Figure 6.4). The main observable difference was found in peak hip flexor moment magnitude during late stance phase (Figure 6.4). Peak hip flexor moment during late stance phase was larger in the intact limb when compared to the affected limb (Figure 6.4). Coupled with the relatively larger increase across time in this variable in the affected limb, these observations resulted in a significant interaction effect (p=0.03). Figure 6.4 illustrates the significantly greater intact limb peak hip abductor moment when compared to the affected limb during both early (p=0.02) and late stance phase (p=0.03).

In terms of joint powers, most of the observed differences were at the ankle and knee joints (Figure 6.5). Post-hoc analysis of peak power absorption at the ankle joint represented by A1, revealed an increase between one and six months (p=0.02) visible in Figure 6.5. In addition, the power generation burst A2, was considerably larger in the intact limb ankle joint when compared to the affected limb (p=0.03), likely due to the limitations of the prosthetic ankle joint components.

Eccentric power absorption at K1 changed significantly over time, post-hoc analysis revealing the differences to be between three and six months (p=0.05). However, these changes were different between limbs. Although K1 magnitude seemed higher on the intact compared to the affected side (Figure 6.5), there was no significant limb main effect. Figure 6.5 highlights the increased power generation at K2 in the intact limb when compared to the affected limb although this was not statistically significant (p=0.09). Power absorption during late stance phase (K3) and early swing phase (K4) were similar between limbs and did not change significantly over time (Figure 6.5).

6.3.1.5 Data for n=2 at 12 Months Post-Discharge

In terms of temporal-spatial variables, participants at 12 months post-discharge continued to increase walking velocity, stride length, step length and cadence, while relative stance duration and double limb support decreased. This reflected an overall improvement in functioning during this time period.

Sagittal plane hip ROM increased at 12 months post-discharge, as did knee ROM during loading response in the affected limb, a reflection of increased ability in the control of these joints. Interestingly, no further increases occurred in intact limb knee ROM during loading response. Affected limb ankle ROM during stance phase seemed to reduce at 12 months post-discharge, perhaps in conjunction with the aforementioned knee adaptations.

Ground reaction forces experienced by each limb also changed between 6 and 12 months post-discharge (Figure 6.3 and Table 6.3). Load rate increased in the intact limb, with little change in the affected limb whilst decay rates increased in both limbs. Linked to the changes in load and decay rates, were increases in both initial and second peak vertical GRF in the intact limb. However, similar effects were not visible in the affected limb. Peak anterior GRF increased in both limbs at 12 months post-discharge, however the magnitude of peak posterior GRF remained relatively similar to that at six months. These between limb differences continued the pattern of increased forces being experienced by the intact limb.

Although ankle moment profiles were similar at 12 months post-discharge to those at six months post-discharge, large adaptations were observed at the knee (Figure 6.4). Peak knee flexor moment during stance phase increased at 12 months post-discharge in the affected limb. However, the intact limb knee moment profile was notably reduced at 12 months post-discharge during stance phase. Apart from an increase in hip extensor moment during loading response in both limbs, there were few changes in hip moment profiles at 12 months post-discharge.

Many of the adaptations that occurred in terms of joint powers related to power generation. In particular, ankle (A2) and hip power (H1) bursts increased in both limbs during late stance phase and loading response respectively. These variables were also greater in the intact limb when compared to the affected limb (Figure 6.5), highlighting that most power generation during gait occurred in the intact limb.

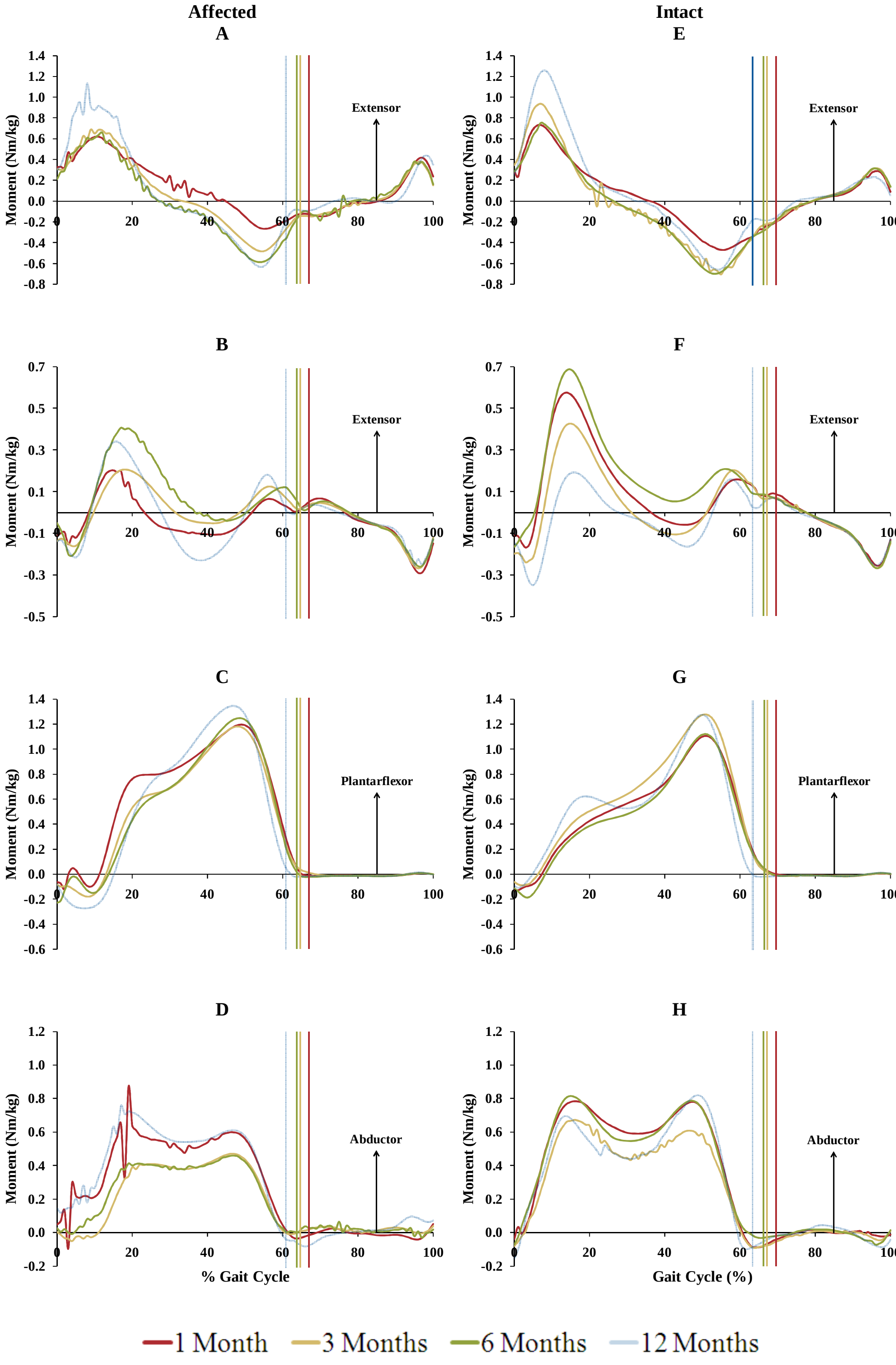


Figure 6.4 Group mean sagittal plane joint moments for the affected limb hip (A), knee (B) and ankle (C) and intact limb hip (E), knee (F) and ankle (G). Frontal plane hip moments also presented for the affected (D) and intact (H) limbs. Time normalised to 100% of gait cycle during level gait. Vertical lines represent toe off. Data at 12 months from n=2.

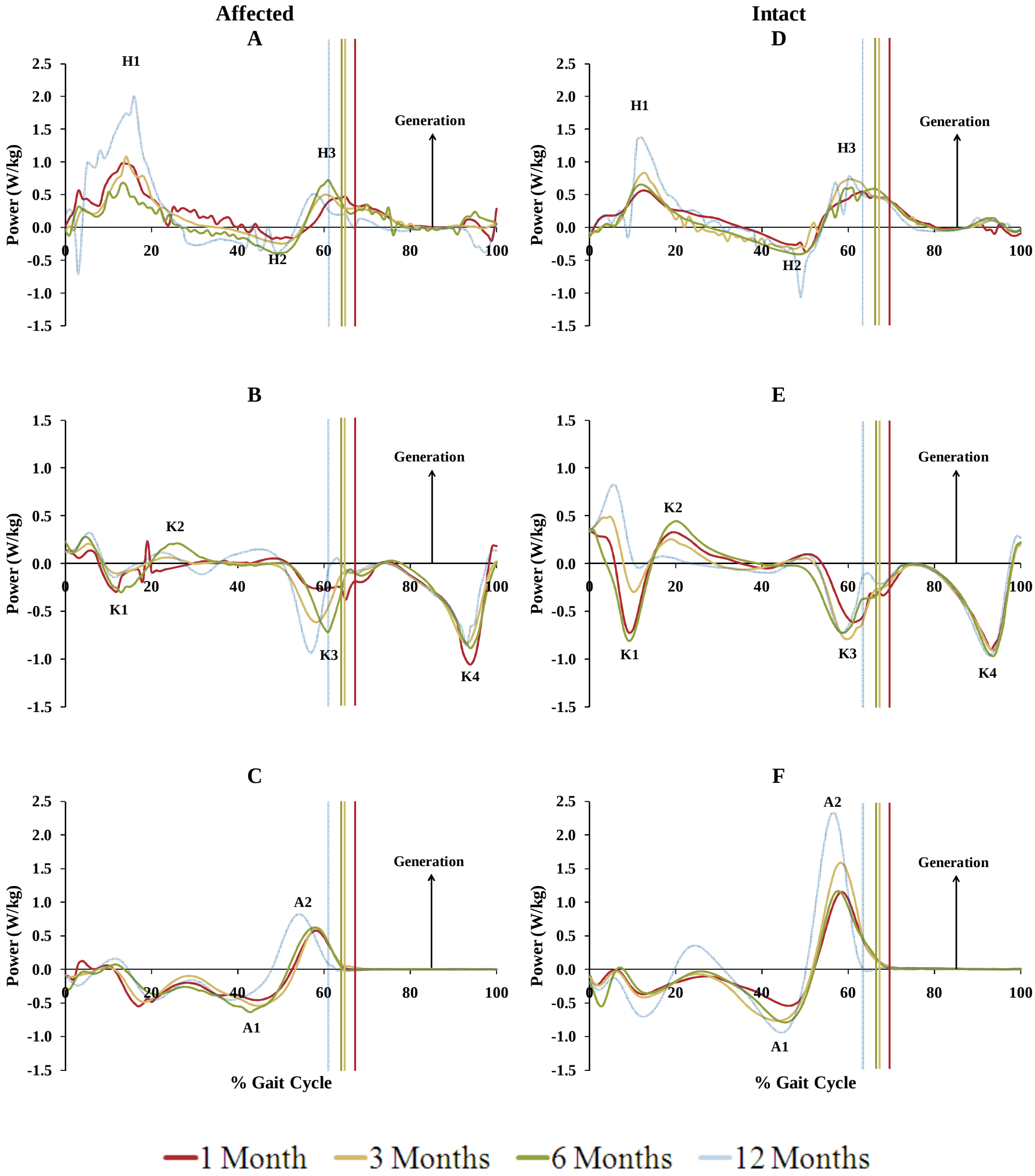


Figure 6.5 Group mean sagittal plane joint powers for the affected limb hip (A), knee (B) and ankle (C) and intact limb hip (D), knee (E) and ankle (F). Time normalised to 100% of gait cycle during level gait. Vertical lines represent toe off. Data at 12 months from n=2.

Table 6.4 Statistical breakdown of level gait temporal-spatial variables and sagittal plane joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 8.98) = 1.82	0.22				
Stride Length	(2, 9.01) = 7.07	0.01*				
Relative Double Limb Support	(2, 9.02) = 3.64	0.07				
Step Length	(2, 11.86) = 7.09	0.01*	(1, 10.20) = 0.30	0.59	(2, 21.17) = 0.25	0.79
Cadence	(2, 13.02) = 0.52	0.61	(1, 18.36) = 0.22	0.65	(2, 22.23) = 0.04	0.96
Relative Stance Duration	(2, 11.62) = 7.81	0.01*	(1, 6.32) = 7.63	0.03*	(2, 20.37) = 0.34	0.71
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 19.69) = 0.61	0.56	(1, 7.65) = 1.73	0.23	(2, 20.74) = 1.16	0.33
Peak ankle dorsiflexion during stance	(2, 14.65) = 2.05	0.17	(1, 8.23) = 0.93	0.36	(2, 20.35) = 1.24	0.31
Peak plantarflexion during swing	(2, 8.51) = 3.19	0.09	(1, 10.27) = 45.90	<0.01*	(2, 7.95) = 1.72	0.24
Ankle range of motion during stance	(2, 11.48) = 1.57	0.25	(1, 11.23) = 19.39	<0.01*	(2, 10.46) = 0.03	0.97
Peak knee flexion during loading response	(2, 8.71) = 0.04	0.96	(1, 7.11) = 4.29	0.08	(2, 17.30) = 0.15	0.86
Peak knee flexion during swing	(2, 11.85) = 0.59	0.57	(1, 7.16) = 0.12	0.73	(2, 18.86) = 0.12	0.89
Knee range of motion during loading response	(2, 13.59) = 5.20	0.02*	(1, 7.58) = 16.59	<0.01*	(2, 20.15) = 2.07	0.15
Knee range of motion during single limb support	(2, 15.25) = 0.72	0.50	(1, 7.37) = 14.91	0.01*	(2, 20.50) = 0.34	0.72
Knee range of motion across gait cycle	(2, 15.38) = 1.60	0.23	(1, 8.95) = 0.77	0.40	(2, 20.85) = 0.32	0.73
Peak hip flexion during loading response	(2, 24.46) = 0.28	0.76	(1, 21.70) = 0.24	0.63	(2, 19.05) = 0.05	0.95
Peak hip extension during stance	(2, 24.82) = 1.01	0.38	(1, 21.92) = 1.55	0.23	(2, 20.20) = 0.43	0.66
Peak hip flexion during swing	(2, 7.31) = 1.98	0.21	(1, 20.31) = 0.01	0.91	(2, 21.03) = 0.94	0.41
Hip range of motion during single limb support	(2, 10.03) = 1.04	0.39	(1, 7.41) = 3.38	0.11	(2, 19.02) = 0.63	0.54
Hip range of motion across gait cycle	(2, 12.23) = 2.96	0.09	(1, 6.70) = 1.18	0.31	(2, 19.73) = 0.40	0.68
Pelvic range of motion during single limb support	(2, 12.55) = 1.10	0.36	(1, 8.48) = 0.03	0.86	(2, 20.24) = 0.22	0.81

Table 6.5 Statistical breakdown of level gait frontal and transverse plane kinematics, ground reaction forces and sagittal plane joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 11.65) = 0.37	0.70	(1, 7.78) = 1.77	0.22	(2, 19.60) = 0.04	0.96
Peak pelvic obliquity during swing	(2, 11.16) = 0.06	0.94	(1, 5.38) = 0.36	0.57	(2, 18.07) = 0.09	0.92
Hip rotation range of motion during single limb support	(2, 20.61) = 0.95	0.40	(1, 10.90) = 0.17	0.69	(2, 20.39) = 1.15	0.34
Pelvic rotation range of motion during single limb support	(2, 13.06) = 0.63	0.55	(1, 14.91) = 0.01	0.94	(2, 21.71) = 0.84	0.45
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 17.96) = 0.78	0.47	(1, 13.54) = 5.03	0.04*	(2, 13.09) = 0.02	0.98
Vertical GRF Fz2	(2, 17.81) = 0.39	0.68	(1, 12.95) = 2.49	0.14	(2, 12.77) = 0.81	0.47
Anterior-Posterior GRF Fy1	(2, 7.17) = 1.49	0.29	(1, 4.81) = 17.19	0.01*	(2, 12.58) = 0.16	0.86
Anterior-Posterior GRF Fy2	(2, 6.27) = 0.94	0.44	(1, 3.38) = 4.05	0.13	(2, 11.95) = 0.63	0.55
Load Rate	(2, 5.00) = 2.62	0.17	(1, 711) = 15.90	0.01*	(2, 11.09) = 0.72	0.51
Decay Rate	(2, 8.05) = 2.06	0.19	(1, 4.43) = 0.07	0.81	(2, 11.86) = 0.27	0.77
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 12.90) = 5.26	0.02*	(1, 5.34) = 0.66	0.45	(2, 19.57) = 2.30	0.13
Peak ankle plantarflexor moment during stance	(2, 11.45) = 1.00	0.40	(1, 6.34) = 0.18	0.68	(2, 15.75) = 0.38	0.69
Peak knee flexor moment during loading response	(2, 15.50) = 0.10	0.91	(1, 7.94) = 2.58	0.15	(2, 15.95) = 0.05	0.95
Peak knee extensor moment during loading response	(2, 10.53) = 5.03	0.03*	(1, 7.16) = 8.98	0.02*	(2, 14.46) = 0.81	0.46
Peak knee flexor moment during mid stance	(2, 7.33) = 2.31	0.17	(1, 6.14) = 0.00	0.97	(2, 9.51) = 2.12	0.17
Peak knee flexor moment during late stance	(2, 9.55) = 0.44	0.66	(1, 6.26) = 2.79	0.14	(2, 11.86) = 0.09	0.92
Peak knee flexor moment during swing	(2, 16.87) = 0.11	0.90	(1, 8.93) = 0.04	0.85	(2, 13.47) = 0.09	0.92
Peak hip extensor moment during early stance	(2, 16.68) = 0.54	0.59	(1, 8.98) = 1.53	0.25	(2, 15.43) = 0.05	0.95
Peak hip flexor moment during late stance	(2, 10.09) = 3.16	0.09	(1, 14.83) = 13.80	<0.01*	(2, 10.82) = 4.84	0.03*
Peak hip extensor moment during swing	(2, 7.63) = 0.28	0.77	(1, 2.82) = 0.13	0.74	(2, 13.29) = 0.12	0.89

Table 6.6 Statistical breakdown of level gait frontal plane and support moments and joint powers. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 17.31) = 2.58	0.11	(1, 13.17) = 6.78	0.02*	(2, 12.78) = 1.43	0.27
Peak hip abductor during late stance	(2, 15.18) = 1.59	0.24	(1, 6.45) = 7.44	0.03*	(2, 11.22) = 0.13	0.88
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 20.84) = 0.05	0.96	(1, 20.84) = 0.15	0.71	(2, 20.84) = 0.68	0.52
Second peak support moment	(2, 10.08) = 1.27	0.32	(1, 5.29) = 0.07	0.81	(2, 13.19) = 0.01	0.99
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 16.73) = 4.96	0.02*	(1, 8.47) = 3.93	0.08	(2, 19.43) = 0.71	0.50
A2 – Ankle power generation during pre-swing	(2, 10.44) = 1.17	0.35	(1, 4.74) = 8.72	0.03*	(2, 13.42) = 0.04	0.96
K1 – Knee power absorption during loading response	(2, 10.51) = 4.62	0.04*	(1, 6.08) = 3.45	0.11	(2, 15.19) = 1.73	0.21
K2 – Knee power generation during mid-stance	(2, 7.54) = 2.10	0.19	(1, 4.47) = 4.74	0.09	(2, 13.40) = 0.49	0.63
K3 – Knee power absorption during pre-swing	(2, 8.95) = 0.62	0.56	(1, 4.40) = 1.98	0.23	(2, 13.43) = 0.05	0.95
K4 – Knee power absorption during terminal swing	(2, 16.13) = 0.16	0.85	(1, 7.83) = 0.12	0.74	(2, 16.82) = 0.43	0.66
H1 – Hip Power generation during loading response	(2, 17.42) = 0.81	0.46	(1, 13.64) = 1.88	0.19	(2, 12.88) = 1.33	0.30
H2 – Hip power absorption during stance	(2, 15.87) = 1.57	0.24	(1, 7.54) = 3.57	0.10	(2, 16.67) = 0.23	0.79
H3 – Hip power generation during pre-swing	(2, 3.46) = 1.80	0.29	(1, 6.61) = 3.27	0.12	(2, 11.12) = 2.26	0.15

6.3.2 Discussion – Level Gait

Results from the current study such as increases in walking velocity suggested that in general, amputees' ability in performing level gait improved following discharge from rehabilitation, supporting the first (1) hypothesis.

Temporal-spatial variables improved following discharge from rehabilitation although were still reduced when compared to those reported in literature from amputees with >1 year experience in prosthetic use (Sanderson and Martin, 1997; Powers *et al.,* 1998; Bateni and Olney, 2002; Grumillier *et al.,* 2008). Temporal-spatial inter-limb asymmetry was still present with participants taking longer steps on the affected side, with a higher cadence on the intact side, an established feature of amputee gait (Winter and Sienko, 1988; Hurley *et al.,* 1990; Perry 1992; Sanderson and Martin, 1997; Powers *et al.,* 1998; Isakov *et al.,* 2000; Bateni and Olney, 2002; Grumillier *et al.,* 2008; Vickers *et al.,* 2008; Vrieling *et al.,* 2008). Literature has explained these observations as an attempt by amputees to protect the residuum from increased forces and a lack of confidence in the ability to control the affected limb (Sanderson and Martin, 1996; Sanderson and Martin, 1997; Powers *et al.,* 1998; Nolan *et al.,* 2003). Although the reduction in stance duration and double limb support observed following discharge in the current study may have reflected increasing confidence in gait stability over time, these compensatory mechanisms were present.

There were significant inter-limb differences in ankle joint kinematics, the intact limb displaying increased functioning in terms of greater joint ROM. Although this difference was likely due to the limitations associated with the prosthetic ankle components, the reduction in performance of this key joint may have been to the overall detriment of amputees' functioning. Knee ROM during loading response was greater in the intact limb, perhaps reflecting an increased ability to control the joint as the lower limb was loaded. Literature has suggested that a lack of affected knee ROM during loading response was representative of part of the aforementioned protective mechanism (Beyaert *et al.,* 2008).

This interpretation was supported by the increased load rate and peak vertical and posterior GRFs observed in the intact limb when compared to the affected limb. Literature has previously reported similar kinetic differences in experienced amputees', again highlighting the protective mechanism present in the affected limb (Powers *et al.,* 1998; Nolan *et al.,* 2003).

The reduced kinematic function of the affected limb knee joint was also reflected in the joint kinetics. Peak knee extensor moment during stance phase, particularly during loading response, was reduced in the affected limb when compared to the intact limb with similar effects previously reported in literature (Winter and Sienko, 1988; Sanderson and Martin, 1997; Powers *et al.,* 1998; Beyaert *et al.,* 2008; Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). By keeping the GRF vector closer to the knee joint, thus reducing knee extensor moment during stance phase, literature has suggested that participants reduce the demands of the quadriceps musculature whilst also preventing the knee from collapsing during stance phase (Sanderson and Martin, 1997; Powers *et al.,* 1998; Vanicek *et al.,* 2009a). In this instance, it is likely that joint reaction forces will be increased in the affected limb as the vertical GRF vector passes through the knee joint, although this has not been reported and warrants further investigation. With this in mind, increases over time in participants affected limb peak knee extensor moment suggested a gradual decline in reliance upon this strategy.

Peak hip flexor moment during late stance phase was also higher in the intact limb compared to the affected limb, however, this did increase over time in the affected limb. One interpretation of this result could be that the increased hip flexor moment aided progression of the affected limb in preparation for swing phase.

Related to this observation, was the reduced affected limb power generation (A2) at the ankle which may have necessitated the increased hip flexor moment observed in the current study and increased hip power generation reported in the literature (Winter and Sienko, 1988; Bateni and Olney, 2002; Grumillier *et al.*, 2008; Silverman *et al.*, 2008; Vanicek *et al.*, 2009). Ankle power generation plays an important role in limb progression and stability (Winter, 1983) and thus, increased intact limb ankle power generation reported in the current study played a key role in the overall improvement in walking velocity.

Similar to reports in literature, knee joint power absorption (K1) and generation (K2) during stance phase were both decreased in the affected limb when compared to the intact limb (Winter and Sienko, 1988; Perry, 1992; Sanderson and Martin, 1997; Powers *et al.*, 1998; Beyaert *et al.*, 2008; Silverman *et al.*, 2008; Vickers *et al.*, 2008; Vanicek *et al.*, 2009a). In addition to the reduced affected limb vertical GRF, knee ROM and extensor moments during stance phase were reduced. These results highlight that as amputees employ a strategy attempting to protect the affected limb, they are not fully able to utilise the affected limb to aid progression or stability during stance phase (Winter and Sienko, 1988; Powers *et al.*, 1998; Beyaert *et al.*, 2008; Silverman *et al.*, 2008; Vickers *et al.*, 2008; Vanicek *et al.*, 2009a).

Although literature reported that detriments in gait as a result of reduced affected limb ankle function placed increased demands on hip joint musculature, this was not the only effect observed in the current study (Winter and Sienko, 1988; Bateni and Olney, 2002; Grumillier *et al.*, 2008; Silverman *et al.*, 2008; Vanicek *et al.*, 2009a). Rather, there were also adaptations in affected limb knee function but far greater reliance upon the intact limb. Given that participants cited in literature tended to be of greater prosthetic experience than those in the current study, it could be hypothesised that improvements in gait function are initially obtained through increased intact limb function, with further increases a result of the combination of intact limb function and hip musculature control in the affected limb. A pertinent implication of this hypothesis is that literature has reported increased knee joint bone mineral density of the intact limb in amputees, suggesting this may lead to a higher risk of osteoarthritis and knee joint degradation (Royer and Koenig, 2005). While the intact limb plays a crucial role in the improvement of amputee functioning, care must be taken not to chronically damage the limb, negating any further progress or indeed regression. These results are relevant to those involved in the care and rehabilitation of lower limb amputees as they highlight features of less experienced amputee gait, such the lack of power absorption and generation in the affected limb. This information may help to inform and improve future rehabilitation practice which may benefit from the inclusion of targeted strengthening of the knee extensor musculature via exercises such as single limb squats, aimed at increasing eccentric and concentric knee and hip extensor strength.

Overall, gait proficiency increased as evidenced by improvements in a number of biomechanical variables such as walking velocity, therefore the experimental hypothesis (1) that gait proficiency would improve was accepted. However, the previously unknown mechanism of these increases was a novel finding, illustrating the changing pattern of adaptation in transtibial amputees.

6.3.3 Activities of Daily Living Terminology

When crossing an obstacle or stepping up/down to a new level, the first limb to approach the task e.g. to cross the obstacle, becomes the lead limb with the other becoming the trail limb. In the current study participants were free to self-select the lead limb (affected or intact) and the subsequent trail limb. The following terminology outlines the future reference to either affected or intact limb as the lead or trail limb:

Lead Affected – Lead limb is the affected limb

Trail Intact – Trail limb is the intact limb

Lead Intact – Lead limb is the intact limb

Trail Affected – Trail limb is the affected limb

6.3.4 Obstacle Crossing

6.3.4.1 Temporal-Spatial Variables

Temporal-spatial variables are presented in Table 6.7 with complete statistical analyses provided in Tables 6.10 and 6.13.

When leading with both the affected and intact limbs, participants walking velocity increased by 23.6% between one and six months post-discharge. Although this was not statistically significant, the 0.17 m/s increase between one and six months following discharge represents a highly clinically meaningful increase in walking velocity. Walking velocity was not different when leading with either limb, reflected by the lack of a limb main effect. Between one and six months post-discharge, stride length increased by 10.2 % and 13.2% when leading with the affected and intact limbs respectively, with stride length being greater in the intact limb at six month, although no main effects were found.

When leading with the affected limb, both the lead and trail limbs displayed very little change over time in relative stance duration (Table 6.7). However, the significantly larger relative stance duration in the trail limb (Table 6.7) resulted in a significant limb main effect ($p<0.01$).

Table 6.7 Mean (SD) temporal-spatial variables of obstacle crossing. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Walking Velocity (m/s)	Lead Affected	0.72 (0.25)	0.93 (0.19)	0.89 (0.20)	1.14 (0.10)
	Lead Intact	0.72 (0.15)	0.85 (0.19)	0.89 (0.20)	1.14 (0.10)
Stride Length (m)	Lead Affected	1.18 (0.17)	1.37 (0.18)	1.30 (0.19)	1.49 (0.10)
	Lead Intact	1.21 (0.10)	1.33 (0.16)	1.37 (0.20)	1.50 (0.10)
Relative Stance Duration (% GC)	Lead Affected	58 (2.0)	57 (1.9)	58 (2.1)	57 (1.2)
	Lead Intact	66 (2.1)	64 (2.4)	64 (4.1)	60 (1.1)
	Trail Intact	69 (5.1)	68 (3.4)	68 (4.0)	63 (1.8)
	Trail Affected	61 (4.6)	61 (2.5)	60 (2.4)	59 (2.2)

*Data from two participants, not included in statistical analyses.

Table 6.8 Individual participant lead limb preferences whilst crossing an obstacle.

Patient Number	One Month				Three Months				Six Months				Twelve Months			
	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference
1					1	9	10	I	0	6	6	I	4	5	9	I
2	6	0	6	A	7	0	7	A	6	0	6	A	9	1	10	A
3	0	6	6	I	0	6	6	I	0	7	7	I				
4	2	6	8	I	2	4	6	I	2	6	8	I				
5	0	6	6	I	5	5	10	No Pref	4	4	8	No Pref				
6					5	7	12	I	8	2	10	A				
7					2	4	6	I	3	5	8	I				

6.3.4.2 Lead Limb Preference

Participants lead limb preference when crossing an obstacle is presented in Table 6.8, no statistical comparisons were drawn.

The vast majority of participants across all time points displayed a level of lead limb preferences with only two exceptions (participant five). Neither the intact nor affected limbs were used definitely as the lead limb although, there was a strong general bias towards adopting the intact limb as the lead limb. (Table 6.8). Interestingly, participants tended to select one limb as the lead limb and used this strategy consistently over time.

6.3.4.3 Foot Marker Trajectories

Foot marker trajectory data are presented in Figure 6.6 with complete statistical analyses provided in Tables 6.12 and 6.15.

6.3.4.4 Lead Limb

Figure 6.6 illustrates the very consistent heel and toe trajectories of the lead limb in the six months following discharge. This observation is supported by the lack of statistically significant time main effects in peak vertical heel and toe displacements. However, there were differences in lead limb peak heel and toe displacements between affected and intact limbs (Figure 6.6). Firstly, peak vertical toe displacement was greater when leading with the affected limb when compared to leading with the intact limb ($p=0.05$). Conversely, peak vertical heel displacement was greater when leading with the intact limb when compared to leading with the affected limb ($p=0.02$).

6.3.4.5 Trail Limb

The trail limb heel and toe trajectories were less consistent over time (Figure 6.6). Although Figure 6.6 illustrates a general decrease in peak vertical heel displacement over time when trailing with the affected limb, there were no significant time main effects for peak vertical heel and toe displacements. It can also be observed that between limb differences in heel and toe displacement when trailing with the affected or intact limb were minimal (Figure 6.6).

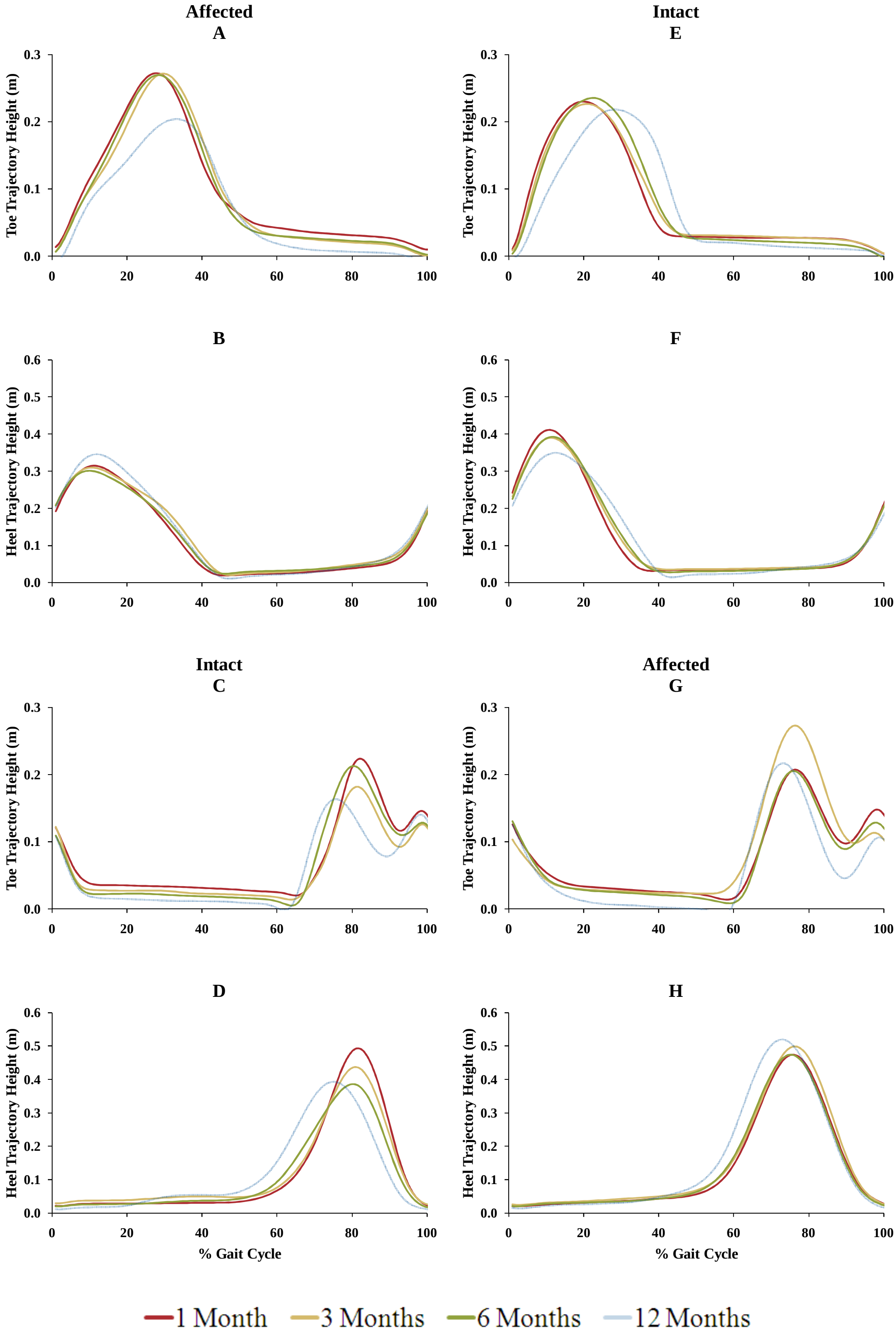


Figure 6.6 Group mean foot marker trajectories for the lead affected (toe – A, heel - B), trail intact (toe – C, heel - D), lead intact (toe – E, heel - F) and trail affected (toe – G, heel - H) limbs. Time normalised to 100% of gait cycle during obstacle crossing. Lead limb gait cycle defined from toe-off to toe-off, trail limb follows conventional definition. Data at 12 months from n=2.

6.3.4.6 Joint Kinematics

Joint kinematics are presented in Figure 6.7 (sagittal plane lead limb), Appendix N (frontal and transverse plane lead limb), Figure 6.8 (sagittal plane trail limb) and Appendix N (frontal and transverse) with complete statistical analyses provided in Tables 6.10, 6.11, 6.13 and 6.14.

6.3.4.7 Lead Limb

Sagittal plane ankle kinematic profiles when leading with the affected limb remained consistent over time (Figure 6.7). When leading with the intact limb, sagittal plane ankle kinematics maintained a relatively consistent profile, albeit at an increased magnitude at six months post-discharge (Figure 6.7). Ankle ROM was visibly increased when leading with the intact limb when compared to the affected limb, however, as with other ankle variables, there were no significant main effects (Figure 6.7).

A number of differences were observed in sagittal plane knee kinematics (Figure 6.7). Overall, knee ROM across the whole gait cycle was higher when leading with the intact limb compared to leading with the affected limb ($p=0.04$). As participants began to cross the obstacle, the knee joint reached a higher peak knee flexion during swing phase when leading with the intact limb when compared to leading with the affected limb ($p=0.03$). Once the lead limb had crossed the obstacle and landed, peak knee flexion during loading response ($p=0.04$) was increased when leading with the intact limb compared to the affected limb, which was maintained between 10-15 degrees flexion during loading response. In addition, Figure 6.10 illustrates the increased knee ROM during loading response when leading with the intact limb which resulted in a significant interaction effect ($p=0.01$). This could be a result of the reduction in knee ROM when leading with the intact limb between one and three months contrasted against the relatively unchanging knee flexion profile during loading response when leading with the affected limb.

Sagittal plane lead limb hip profiles remained relatively consistent across time in both limbs and no statistically significant main effects were observed in variables relating to sagittal plane hip variables (Figure 6.7).

Similarly, no statistical main effects were observed in sagittal plane pelvic kinematic variables, despite some visible changes in the magnitude of intact limb lead pelvic profiles over time.

Although Appendix N illustrates a visible increase in the magnitude of downward pelvic obliquity and hip abduction when leading with the intact limb, there were no statistically significant main effects found for hip and pelvic frontal and transverse plane kinematics.

6.3.4.8 Trail Limb

Trail limb sagittal plane ankle kinematics resulted in a number of statistically significant results (Table 6.11). Post-hoc analysis revealed a significant increase in trail limb ankle ROM during stance phase between one and six months ($p=0.02$) which is illustrated in Figure 6.8. In addition, Figure 6.8 displays increased ankle ROM during stance phase when trailing with the intact limb compared to the affected limb ($p=0.01$). As the trail limb crossed the obstacle during swing phase, participants displayed decreased peak plantarflexion i.e. increased peak dorsiflexion, when trailing with the intact limb compared to the affected limb ($p<0.01$). This could be due to the inability to actively control the prosthetic ankle during swing phase when trailing with the affected limb coupled with the observable increase in ankle dorsiflexion when trailing with the intact limb (Figure 6.8).

When trailing with the intact limb, sagittal plane knee joint kinematic profiles remained consistent over time (Figure 6.8). This effect was similar when trailing with the affected limb despite a non-significant increase in knee ROM during loading response (Figure 6.8). In addition, knee ROM during loading response and swing phase were visibly larger when trailing with the intact limb compared to the affected limb, although not statistically significant.

Sagittal plane hip kinematic profiles remained consistent when trailing with both limbs, however, over time they became more flexor in magnitude when trailing with the affected limb (Figure 6.8) although this was not statistically significant. When trailing with the intact limb, hip flexion during swing phase seemed to be increased when compared to trailing with the affected limb although this was not statistically significant (Figure 6.8).

Frontal plane hip and pelvic kinematics remained relatively unchanged over time when trailing with the intact limb (Appendix N). Appendix N displays an increase in pelvic obliquity and a decrease in hip abduction during when trailing with the affected limb, although no significant main effects were observed. Similarly, there were no statistically significant changes in transverse plane hip kinematics. Post-hoc analysis revealed that pelvic rotation ROM during single limb support increased between one and three months post-discharge ($p=0.02$) (Appendix N).

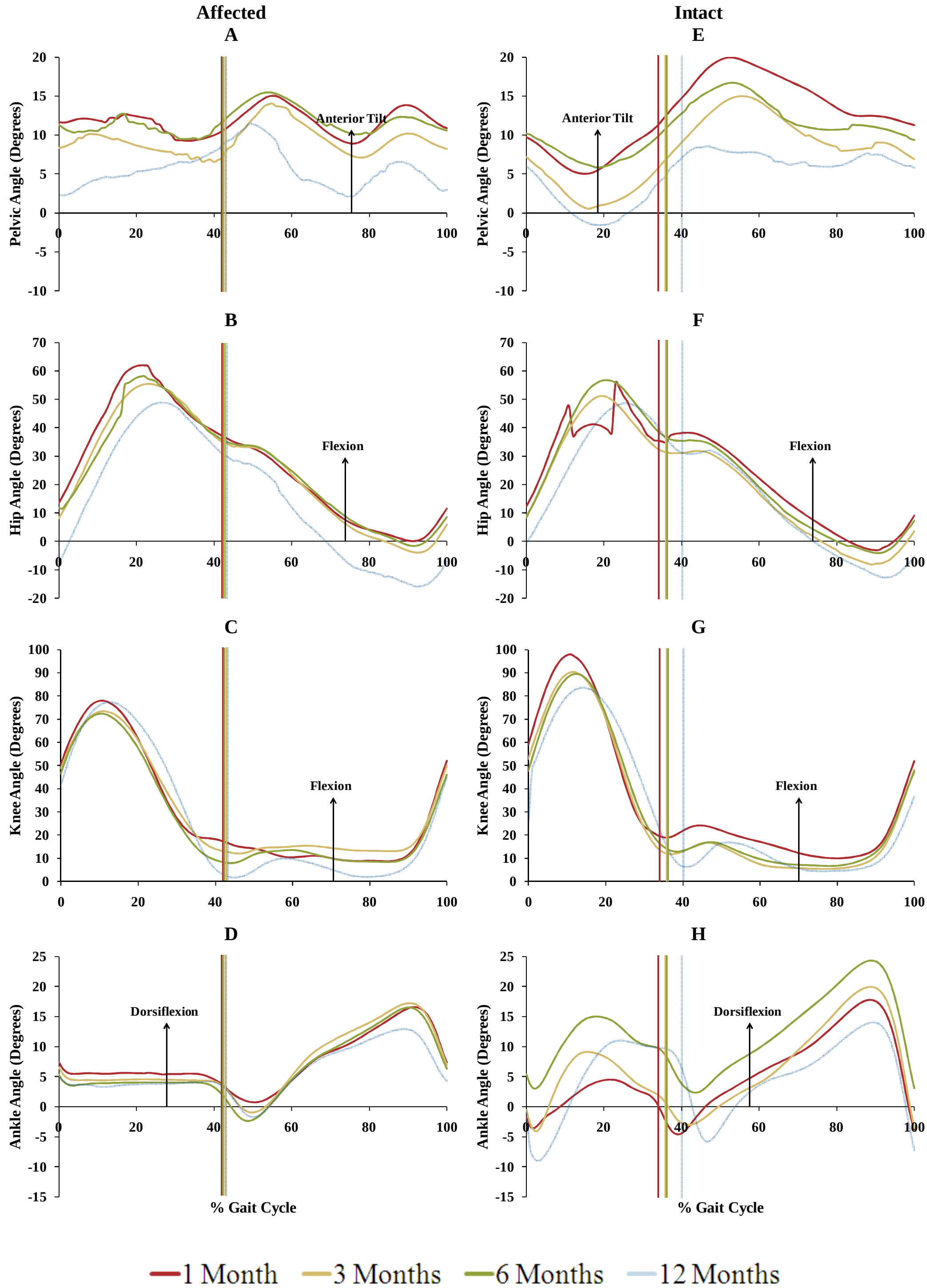


Figure 6.7 Group mean sagittal plane kinematics of the lead affected limb pelvis (A), hip (B), knee (C) and ankle (D) and lead intact limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during obstacle crossing. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

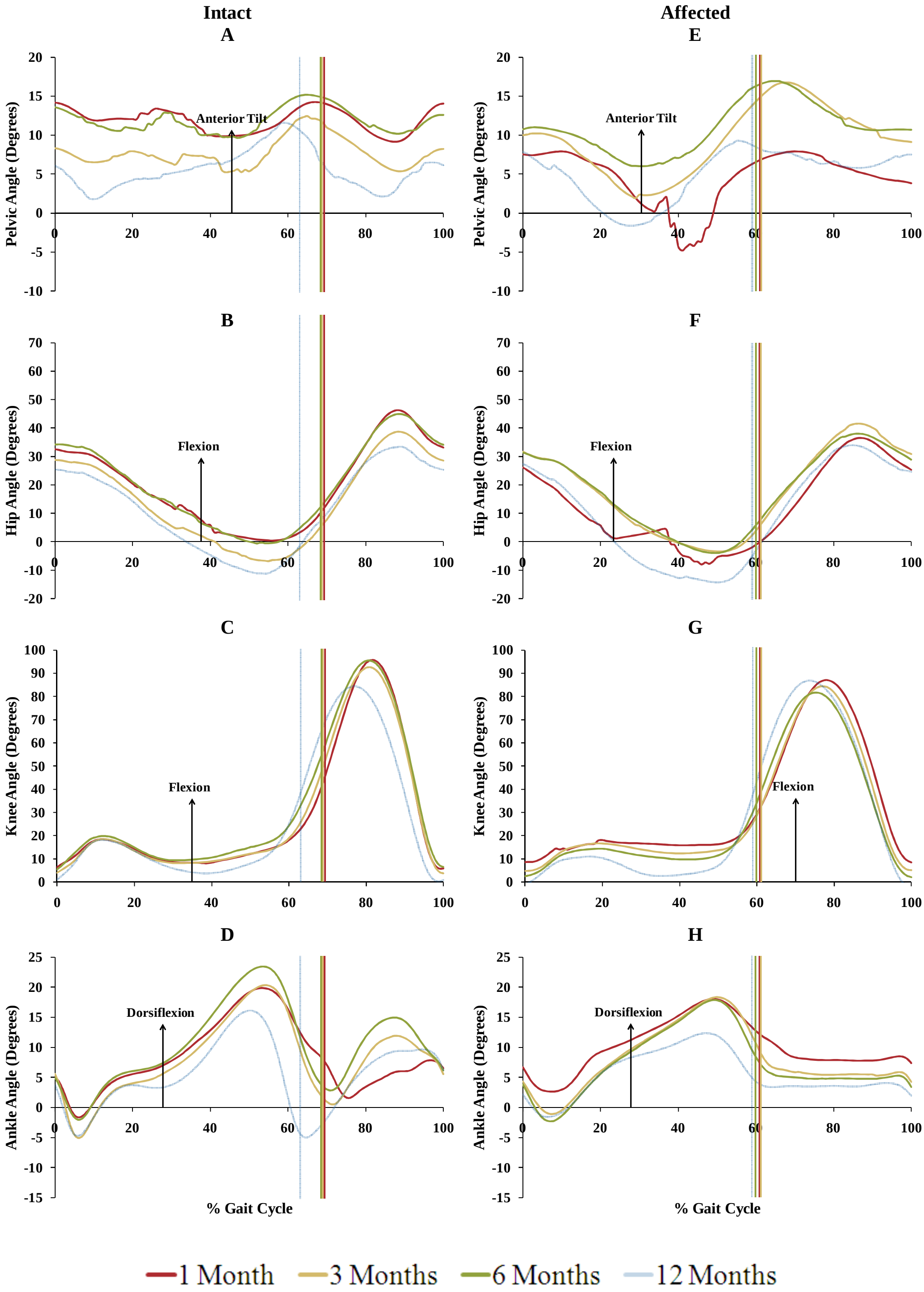


Figure 6.8 Group mean sagittal plane kinematics of the trail intact limb pelvis (A), hip (B), knee (C) and ankle (D) and trail affected limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during obstacle crossing. Vertical lines represent toe off. Data at 12 months from n=2.

6.3.4.9 Ground Reaction Forces

Loading and decay rates are presented in Table 6.9, GRF data are presented in Figure 6.9 with complete statistical analyses provided in Tables 6.11 and 6.14.

6.3.4.10 Lead Limb

Having crossed the obstacle, there were significant time main effects for the second peak vertical GRF (p=0.05) and decay rate (p=0.05), likely due to the decreases observed in these variables when leading with the affected limb. However, post-hoc analysis did not reveal where these changes occurred (Table 6.9, Figure 6.9).

A significant limb effect highlighted that loading rate upon landing was higher when leading with the intact limb compared to the affected limb (p=0.05). In addition, when pushing off following landing, Figure 6.9 highlighted the statistically significant limb main effect (p=0.03), where it can be seen that the second peak vertical GRF is higher when leading with the intact limb compared to leading with the affected limb.

6.3.4.11 Trail Limb

Both load and decay rates tended to increase over time and when trailing with the intact limb, with load rate and to a lesser extent decay rate, being higher than when trailing with the affected limb, although these effects were not statistically significant (Table 6.9). The range of anterior-posterior GRF seemed to increase over time when trailing with the intact limb although no statistically significant results were found in related variables.

Table 6.9 Mean (SD) loading and decay rate of obstacle crossing. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Load Rate (BW/s)	Lead Affected	4.8 (0.0)	4.1 (1.2)	4.1 (0.7)	8.0 (3.4)
	Lead Intact	6.1 (1.2)	6.6 (1.8)	6.7 (0.2)	-
	Trail Intact	6.5 (6.2)	7.2 (2.1)	8.9 (1.8)	9.9 (0.0)
	Trail Affected	4.6 (1.8)	4.9 (1.6)	5.5 (1.9)	-
Decay Rate (BW/s)	Lead Affected	5.7 (0.0)	5.2 (1.0)	4.1 (0.1)	6.8 (0.6)
	Lead Intact	4.6 (0.2)	5.2 (1.7)	5.1 (1.4)	-
	Trail Intact	4.9 (2.0)	6.3 (2.7)	6.4 (2.5)	8.1 (0.0)
	Trail Affected	5.0 (2.2)	5.1 (1.4)	5.4 (1.3)	-

*Data from two participants, not included in statistical analyses

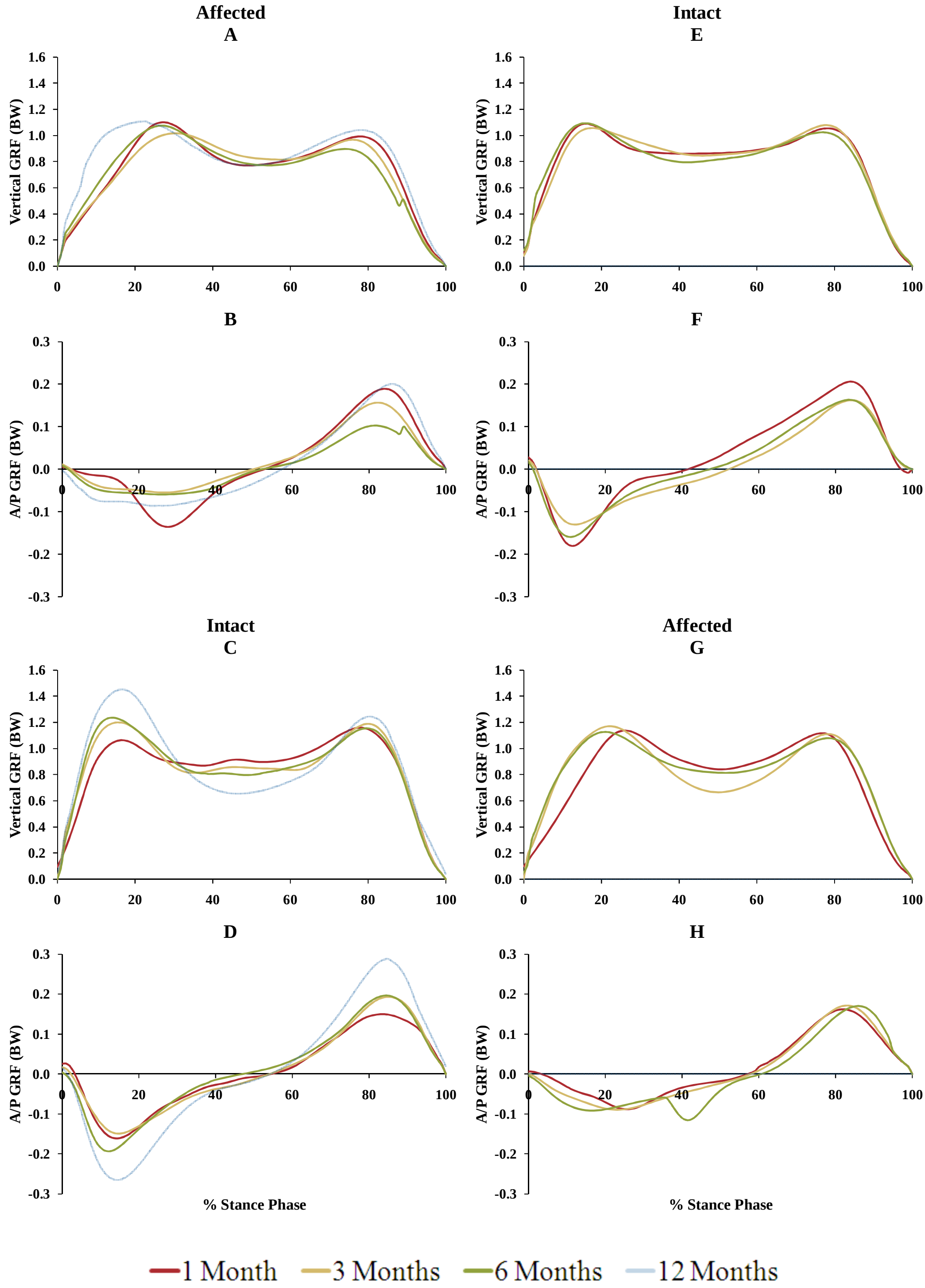


Figure 6.9 Group mean vertical and anterior-posterior ground reaction forces for the lead affected (A and B), lead intact (E and F), trail intact (C and D) and trail affected (G and H) limbs. All data normalised and presented as times body weight (BW). Time normalised to 100% of stance phase during obstacle crossing. Data at 12 months from n=2. Vertical and anterior are positive.

6.3.4.12 Joint Kinetics

Joint moments are presented in Figures 6.10 (sagittal and frontal plane lead limb) 6.11 (sagittal and frontal plane trail limb), joint powers are presented in Figures 6.12 (sagittal plane lead limb), 6.13 (sagittal plane trail limb) and support moments are presented in and Appendix O, with complete statistical analyses provided in Tables 6.11, 6.12, 6.14 and 6.15.

6.3.4.13 Lead Limb

When leading with the intact limb, ankle moment profiles remained relatively consistent over time (Figure 6.10). Although a visible reduction in the magnitude of plantarflexor moment over time when leading with the affected limb was observed, these changes were not statistically significant (Figure 6.10).

As participants foot contacted the ground following obstacle crossing, post-hoc analysis revealed a significant increase in peak knee extensor moment during loading response between three and six months ($p=0.04$) (Figure 6.10). Knee flexor moment during mid-stance seemed to reduce between one and three months, although this was not statistically significant (Figure 6.10).

When leading with the affected limb, there were few changes over time in either the profile or magnitude of sagittal plane hip moments (Figure 6.10). When leading with the intact limb, peak hip abductor moment during both early ($p=0.04$) and late ($p<0.01$) stance phase were higher than when leading with the affected limb (Figure 6.10).

Peak power absorption during stance phase (A1) was slightly higher when leading with the intact limb when compared to leading with the affected limb, although no statistically significant limb main effect was reported (Figure 6.12). Neither, peak power absorption during stance phase (A1) or peak limb power generation during late stance phase (A2) changed significantly across time (Figure 6.12). However, power burst A2 was significantly higher when leading with the intact limb than in the affected limb (p=0.01).

Both knee joint power absorption (K1) and generation (K2) during loading response did not change significantly over time or differ significantly between limbs, despite differences apparent in Figure 6.12. When leading with the intact limb, peak power absorption at the knee during late stance phase (K3) was significantly higher when compared to leading with the affected limb (p=0.05). This was also the case for peak power absorption during swing phase (K4) (p=0.01).

During swing phase, hip power profiles remained relatively unchanged across time (Figure 6.12). Power generation at the H1 power burst was visibly reduced when leading with both limbs between one and three months post-discharge, however this was not statistically significant (Figure 6.12). The concentric power generation during late stance phase, as signified by power burst H3, was significantly increased when leading with the intact limb in comparison to leading with the affected limb (p=0.05).

6.3.4.14 Trail Limb

Peak ankle plantarflexion moment during stance phase was greater when trailing with the affected limb when compared to trailing with the intact limb (Figure 6.11). Coupled with this variable being both greater (trailing intact limb) and smaller (trailing affected limb) at three months post-discharge than at one and six months post-discharge (Figure 6.11), a significant interaction effect was reported (p=0.02).

Few changes in knee moment profile were observed when trailing with the affected limb (Figure 6.11). However, a significant time main effect was observed in peak knee flexor moment during loading response (p=0.05). This may be a result of the observed decrease in knee extensor moment followed by an increase in knee flexor moment during stance phase when trailing with the intact limb (Figure 6.11).

Despite visible increases in peak hip extensor moment during loading response over time in both limbs and peak hip flexor moment during stance phase being greater when trailing with the intact limb, no statistically significant main effects observed in these variables (Figure 6.11). However, peak hip abductor moment during both early ($p=0.01$) and late stance phase ($p=0.02$) was significantly greater when trailing with the intact limb when compared to the affected limb (Figure 6.11).

Support moments were visibly greater when trailing with the intact limb compared to the affected limb, although no statistically significant limb main effects were observed (Appendix O).

Peak ankle power absorption during stance phase (A1) was greater in magnitude and displayed larger changes over time when trailing with the intact limb when compared to the affected limb (Figure 6.13), resulting in a statistically significant interaction effect ($p=0.02$).

Peak ankle power generation (A2) was greater when trailing with the intact limb when compared to the affected limb, although the magnitude of power burst A2 increased over time when trailing with the affected limb (Figure 6.13). This resulted in significant limb ($p=0.02$) and time ($p=0.05$) main effects, although post-hoc analysis did not reveal where the time main effect occurred.

Peak knee power absorption during loading response (K1) was greater when trailing with the intact limb when compared to the affected limb ($p=0.04$) (Figure 6.13). In addition, a significant time main effect was observed ($p=0.04$), although post-hoc analysis did not reveal where these differences occurred, as changes were variable over time (Figure 6.13). As can be seen in Figure 6.18, peak concentric power generation at power burst K2 was significantly greater when trailing with the intact limb compared to the affected limb ($p=0.02$). Post-hoc analysis of changes in peak power generation at power burst K3 revealed significant differences between one and three months post discharge (p=0.05). However, patterns of change were different over time, where K3 magnitude reduced when trailing with the intact limb and increased when trailing with the affected limb (Figure 6.13). Similar to the initial power absorption at K1, there were visibly large and statistically significant differences between limbs in peak power absorption at K4, where power absorption was greater when trailing with the intact limb, compared to the affected limb (p=0.01) (Figure 6.13).

Hip power profiles of trailing limbs did not produce any statistically significant main effects, reaffirming the lack of clear changes over time or difference between leading with the intact or affected limb observed in Figure 6.13.

6.3.4.15 Data for n=2 at 12 Months Post-Discharge

When crossing an obstacle, walking velocity and stride length continued to increase at 12 months post-discharge when leading with either limb. Relative stance duration decreased in both trail and lead limb irrespective of which limb was selected to lead. This reflected an overall increase in functioning during this time period. The majority of peak heel and toe displacements reduced at 12 months post-discharge when leading with both limb, perhaps as a result of patients being more able to actively control the trajectory of the foot over the obstacle, reducing over compensation. Lead limb preference did not change at 12 months post-discharge.

The trend of increased GRF variables in the intact limb when both leading and trailing, was continued at 12 months post-discharge. Peak vertical and anterior-posterior GRF were increased when trailing with the intact limb, with load and decay rates being increased at 12 months post-discharge in all limbs.

When leading with the affected limb, peak knee and hip extensor and ankle plantarflexor moments during stance phase increased at 12 months post-discharge. Trail limb peak ankle plantarflexor moments increased in both limbs at 12 months post-discharge. The magnitude of knee moment profiles reduced in both limbs although peak hip extensor moment during early stance phase remained high.

When leading with the affected limb the main increases in joint powers were observed at power bursts A2, K3 and H1 at 12 months post-discharge. These power bursts matched those observed in the intact limb at six months post-discharge, perhaps reflecting an attempt to gain inter-limb symmetry in joint kinetics. Coupled with the joint moment data, it can be seen that there is an increase in the ability of the affected limb to create and withstand joint moments and produce and absorb power. However, participants were still reliant on intact limb to manage larger joint moments and powers, to achieve the increases in temporal spatial variables.

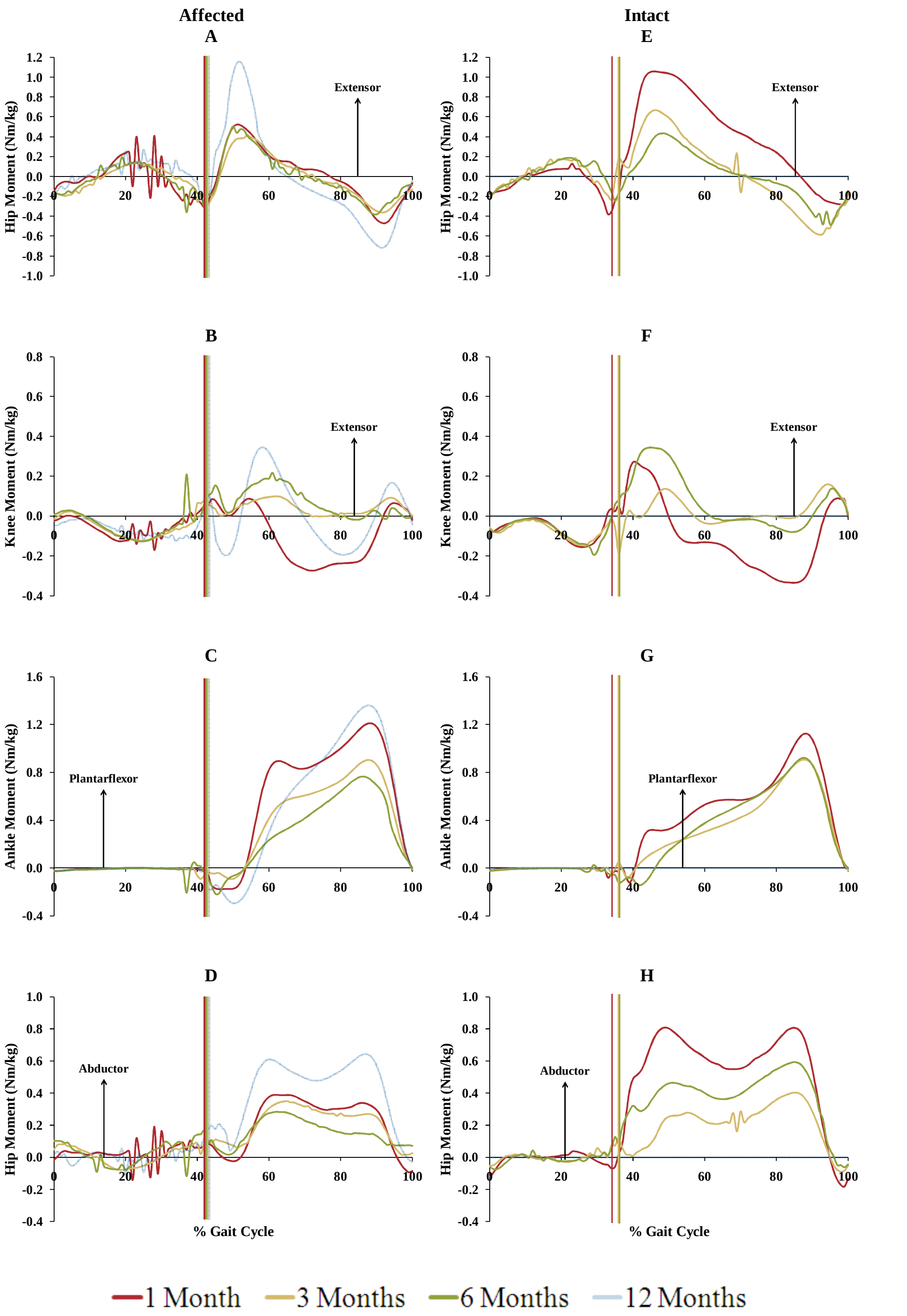


Figure 6.10 Group mean joint moments for the lead affected limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and lead intact limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during obstacle crossing. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

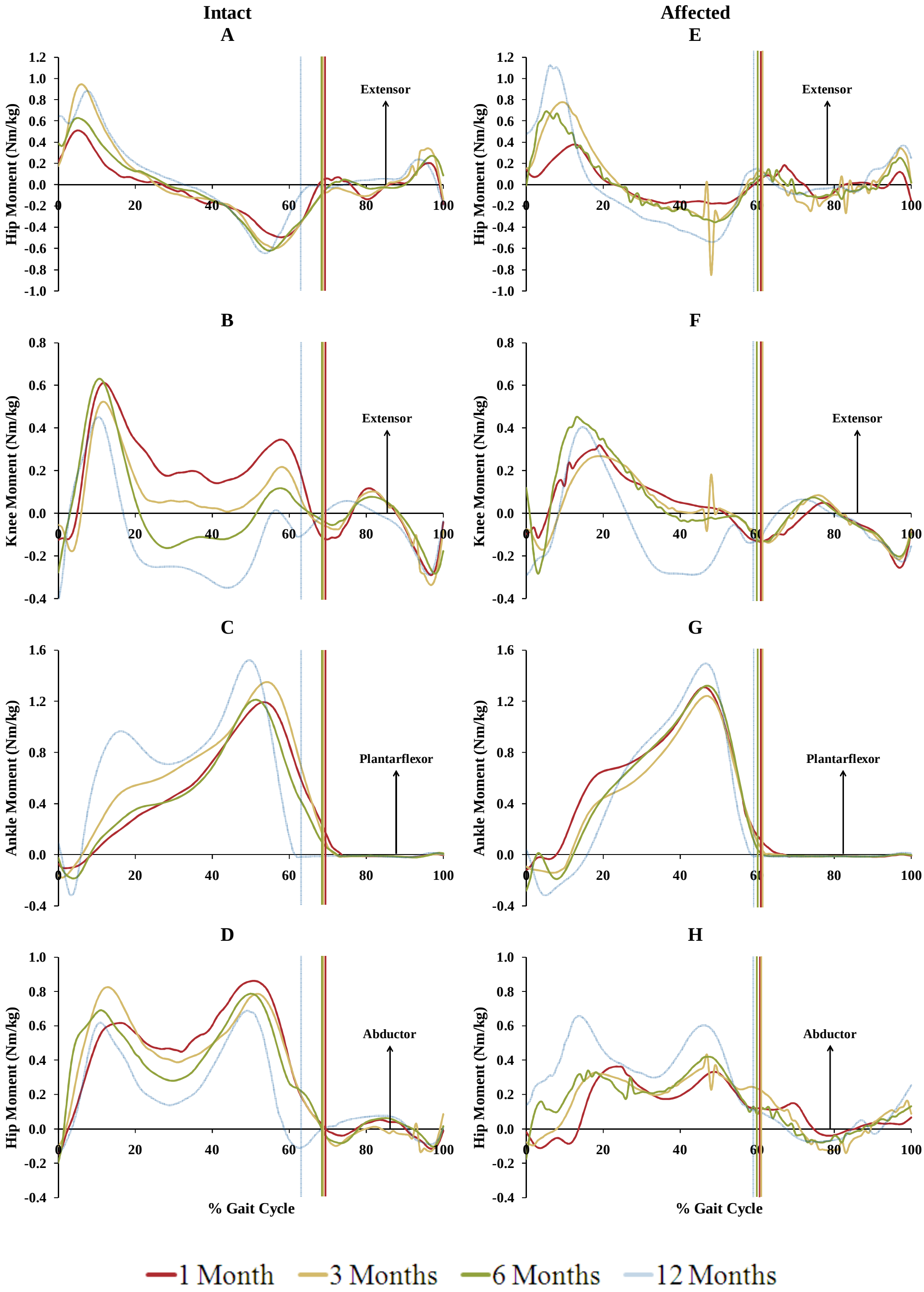


Figure 6.11 Group mean joint moments for the trail intact limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and trail affected limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during obstacle crossing. Vertical lines represent toe off. Data at 12 months from n=2.

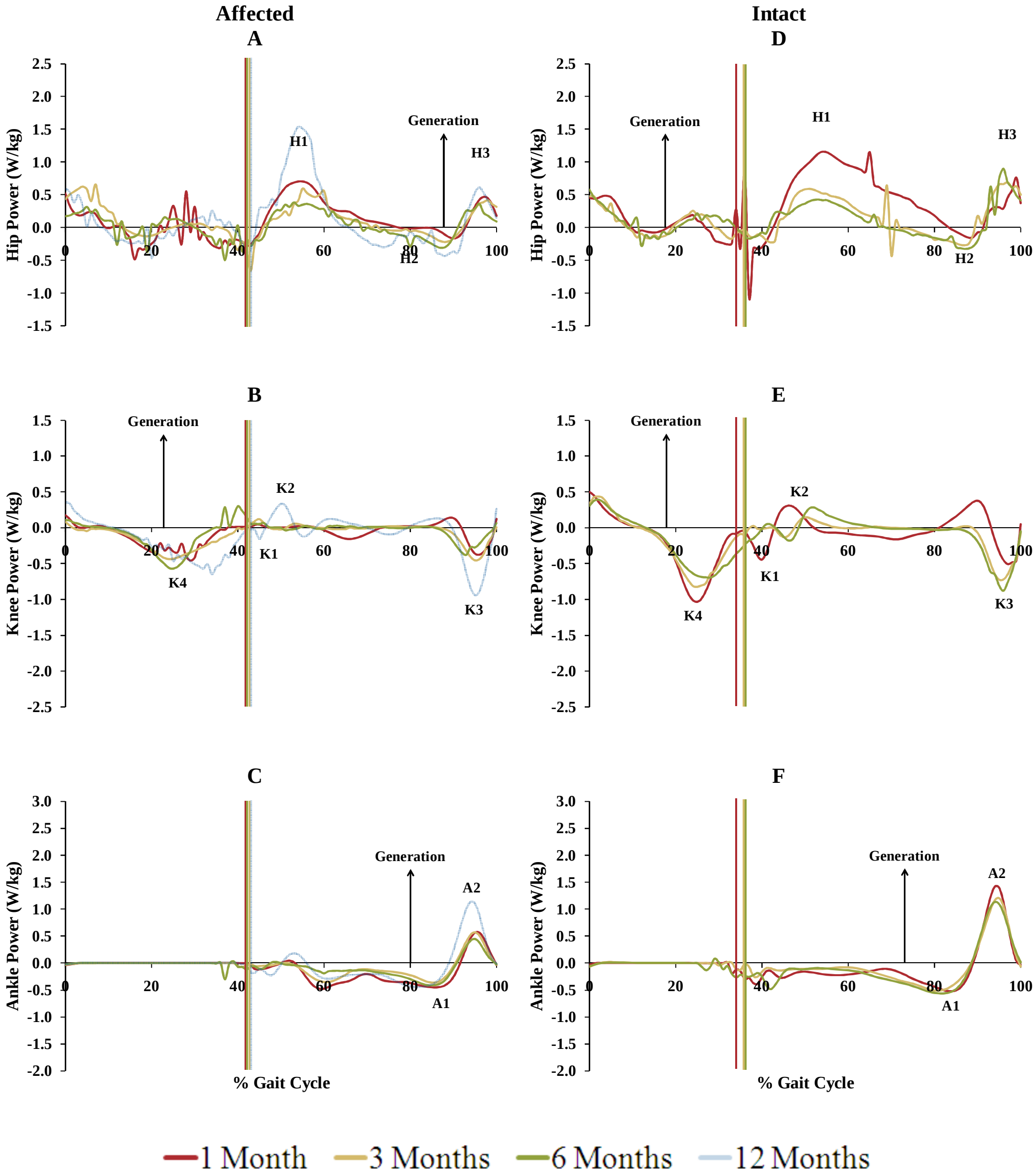


Figure 6.12 Group mean joint powers for the lead affected limb hip (A), knee (B), ankle (C) and lead intact limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during obstacle crossing. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

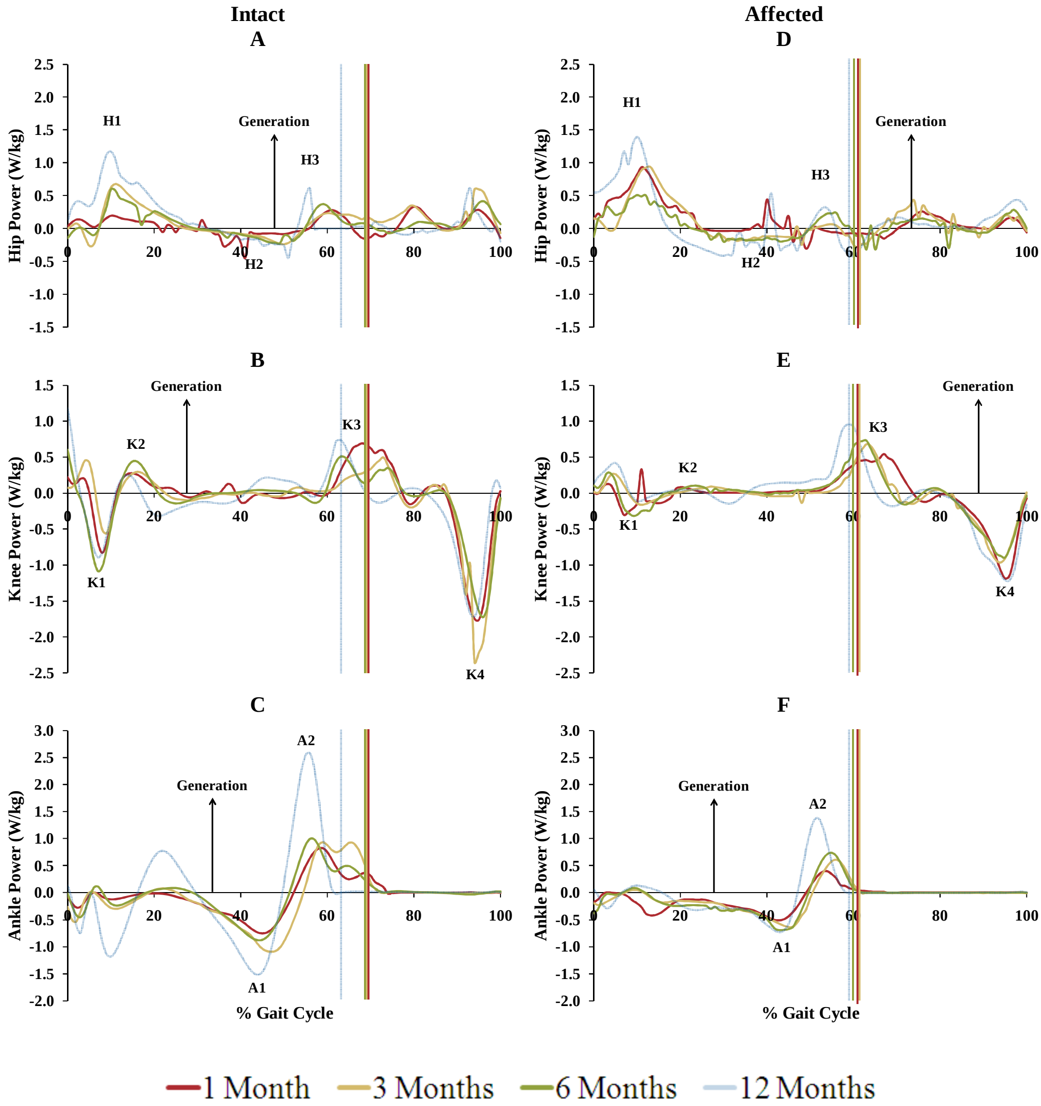


Figure 6.13 Group mean joint powers for the trail intact limb hip (A), knee (B), ankle (C) and trail affected limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during obstacle crossing. Vertical lines represent toe off. Data at 12 months from n=2.

Table 6.10 Statistical breakdown of obstacle crossing lead limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 17.48) = 1.97	0.17	(1, 17.04) = 0.30	0.59	(2, 15.65) = 1.01	0.39
Stride Length	(2, 8.22) = 2.46	0.15	(1, 16.27) = 0.08	0.78	(2, 12.98) = 2.34	0.14
Relative Stance Duration	(2, 11.35) = 1.05	0.38	(1, 6.29) = 27.44	<0.01*	(2, 14.96) = 0.54	0.59
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 9.86) = 0.26	0.78	(1, 6.96) = 0.69	0.43	(2, 13.62) = 1.23	0.32
Peak ankle dorsiflexion during stance	(2, 12.22) = 0.31	0.74	(1, 5.35) = 1.23	0.32	(2, 17.33) = 1.02	0.38
Peak plantarflexion during swing	(2, 12.68) = 0.37	0.70	(1, 5.77) = 2.12	0.20	(2, 16.78) = 1.04	0.38
Ankle range of motion during stance	(2, 10.50) = 2.62	0.12	(1, 4.48) = 5.38	0.07	(2, 12.10) = 2.01	0.18
Peak knee flexion during loading response	(2, 16.73) = 1.29	0.30	(1, 13.96) = 5.32	0.04*	(2, 12.28) = 0.84	0.46
Peak knee flexion during swing	(2, 9.67) = 0.05	0.95	(1, 5.30) = 8.35	0.03*	(2, 11.64) = 0.75	0.49
Knee range of motion during loading response	(2, 9.39) = 2.82	0.11	(1, 7.79) = 7.48	0.03*	(2, 11.70) = 7.29	0.01*
Knee range of motion during single limb support	(2, 8.69) = 0.94	0.43	(1, 2.62) = 3.74	0.16	(2, 11.25) = 3.39	0.07
Knee range of motion across gait cycle	(2, 10.07) = 0.18	0.84	(1, 3.31) = 11.95	0.04*	(2, 11.20) = 0.19	0.83
Peak hip flexion during loading response	(2, 16.58) = 0.53	0.60	(1, 14.84) = 0.02	0.88	(2, 13.03) = 0.68	0.53
Peak hip extension during stance	(2, 15.53) = 0.75	0.49	(1, 11.99) = 0.15	0.71	(2, 10.74) = 0.07	0.93
Peak hip flexion during swing	(2, 16.15) = 2.84	0.09	(1, 15.24) = 0.25	0.62	(2, 13.06) = 0.45	0.65
Hip range of motion during single limb support	(2, 8.20) = 0.54	0.60	(1, 6.95) = 1.80	0.22	(2, 13.47) = 0.93	0.91
Hip range of motion across gait cycle	(2, 8.87) = 1.62	0.25	(1, 5.29) = 0.14	0.72	(2, 11.87) = 0.11	0.90
Pelvic range of motion during single limb support	(2, 8.45) = 0.26	0.77	(1, 9.30) = 0.25	0.63	(2, 15.17) = 0.05	0.95

Table 6.11 Statistical breakdown of obstacle crossing lead limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 20.84) = 1.24	0.31	(1, 21.31) = 1.36	0.26	(2, 20.75) = 0.60	0.56
Peak pelvic obliquity during swing	(2, 17.39) = 0.65	0.54	(1, 13.30) = 1.41	0.26	(2, 11.99) = 0.72	0.51
Hip rotation range of motion during single limb support	(2, 8.97) = 0.38	0.70	(1, 6.26) = 0.05	0.83	(2, 13.87) = 0.68	0.52
Pelvic rotation range of motion during single limb support	(2, 9.22) = 1.58	0.26	(1, 6.21) = 0.49	0.51	(2, 12.72) = 1.87	0.19
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 12.74) = 0.18	0.84	(1, 10.39) = 0.20	0.66	(2, 12.84) = 0.09	0.92
Vertical GRF Fz2	(2, 5.63) = 5.29	0.05*	(1, 5.98) = 7.43	0.03*	(2, 6.06) = .1.23	0.36
Anterior-Posterior GRF Fy1	(2, 8.77) = 1.27	0.33	(1, 9.17) = 3.69	0.09	(2, 8.94) = 0.08	0.92
Anterior-Posterior GRF Fy2	(2, 9.38) = 0.74	0.50	(1, 10.14) = 0.77	0.40	(2, 8.46) = 1.02	0.40
Load Rate	(2, 9.04) = 0.23	0.80	(1, 9.95) = 4.81	0.05*	(2, 9.72) = 0.07	0.94
Decay Rate	(2, 7.96) = 4.43	0.05*	(1, 8.21) = 0.61	0.46	(2, 7.48) = 0.50	0.63
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 15.76) = 2.30	0.16	(1, 17.84) = 0.08	0.16	(2, 13.00) = 0.42	0.67
Peak ankle plantarflexor moment during stance	(2, 7.68) = 0.52	0.62	(1, 3.83) = 0.40	0.56	(2, 10.89) = 0.51	0.62
Peak knee flexor moment during loading response	(2, 12.48) = 0.03	0.97	(1, 5.86) = 0.88	0.39	(2, 15. 46) = 0.39	0.68
Peak knee extensor moment during loading response	(2, 13.24) = 5.07	0.02*	(1, 6.68) = 1.01	0.35	(2, 15.85) = 0.45	0.65
Peak knee flexor moment during mid stance	(2, 16.77) = 1.73	0.21	(1, 17.78) = 1.08	0.31	(2, 15.38) = 0.02	0.98
Peak knee flexor moment during late stance	(2, 16.27) = 0.96	0.40	(1, 17.86) = 2.33	0.15	(2, 14.01) = 0.21	0.81
Peak knee flexor moment during swing	(2, 9.10) = 2.61	0.13	(1, 9.61) = 0.09	0.77	(2, 12.46) = 0.26	0.78
Peak hip extensor moment during early stance	(2, 16.61) = 0.12	0.89	(1, 17.72) = 2.39	0.14	(2, 15.08) = 0.21	0.81
Peak hip flexor moment during late stance	(2, 13.60) = 0.87	0.44	(1, 14.44) = 1.06	0.32	(2, 12.34) = 0.02	0.98
Peak hip extensor moment during swing	(2, 9.71) = 1.18	0.35	(1, 10.21) = 0.79	0.40	(2, 12.70) = 3.24	0.07

Table 6.12 Statistical breakdown of obstacle crossing lead limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 14.69) = 2.44	0.12	(1, 16.46) = 4.79	0.04*	(2, 12.10) = 1.94	0.19
Peak hip abductor during late stance	(2, 13.97) = 0.61	0.56	(1, 15.81) = 11.47	<0.01*	(2, 11.51) = 3.17	0.08
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 8.96) = 0.23	0.80	(1, 10.70) = 2.17	0.17	(2, 13.24) = 0.35	0.71
Second peak support moment	(2, 6.48) = 0.19	0.83	(1, 6.43) = 1.51	0.26	(2, 11.36) = 1.02	0.39
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 4.14) = 0.18	0.85	(1, 7.36) = 2.29	0.17	(2, 11.90) = 0.03	0.97
A2 – Ankle power generation during pre-swing	(2, 12.98) = 0.01	0.99	(1, 14.09) = 8.00	0.01*	(2, 11.07) = 0.83	0.46
K1 – Knee power absorption during loading response	(2, 6.27) = 0.08	0.92	(1, 13.43) = 0.75	0.40	(2, 8.20) = 0.16	0.86
K2 – Knee power generation during mid-stance	(2, 8.46) = 1.43	0.29	(1, 6.95) = 0.47	0.51	(2, 10.36) = 0.01	0.95
K3 – Knee power absorption during pre-swing	(2, 11.88) = 2.45	0.13	(1, 12.74) = 4.89	0.05*	(2, 10.44) = 0.83	0.46
K4 – Knee power absorption during terminal swing	(2, 10.94) = 0.15	0.87	(1, 14.27) = 9.26	0.01*	(2, 12.90) = 0.49	0.63
H1 – Hip Power generation during loading response	(2, 14.68) = 0.37	0.70	(1, 15.31) = 1.71	0.21	(2, 12.58) = 0.82	0.46
H2 – Hip power absorption during stance	(2, 12.38) = 0.51	0.61	(1, 13.61) = 0.13	0.72	(2, 10.35) = 1.51	0.27
H3 – Hip power generation during pre-swing	(2, 9.45) = 0.06	0.94	(1, 10.75) = 4.85	0.05*	(2, 11.19) = 0.08	0.92
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 6.30) = 0.17	0.85	(1, 2.55) = 11.97	0.05*	(2, 9.66) = 0.68	0.53
Heel	(2, 5.96) = 1.64	0.27	(1, 4.81) = 10.57	0.02*	(2, 11.98) = 0.15	0.86

Table 6.13 Statistical breakdown of obstacle crossing trail limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 17.48) = 1.97	0.17	(1, 17.04) = 0.30	0.59	(2, 15.65) = 1.01	0.39
Stride Length	(2, 8.22) = 2.46	0.15	(1, 16.27) = 0.08	0.78	(2, 12.98) = 2.34	0.14
Relative Stance Duration	(2, 9.03) = 1.51	0.27	(1, 4.98) = 37.78	<0.01*	(2, 12.86) = 1.18	0.34
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 10.75) = 0.85	0.46	(1, 9.37) = 2.06	0.18	(15.27) = 0.44	0.65
Peak ankle dorsiflexion during stance	(2, 12.38) = 0.26	0.78	(1, 8.04) = 2.82	0.13	(2, 12.83) = 0.35	0.71
Peak plantarflexion during swing	(2, 11.65) = 0.89	0.44	(1, 11.87) = 13.99	<0.01*	(2, 15.38) = 1.04	0.38
Ankle range of motion during stance	(2, 10.22) = 6.05	0.02*	(1, 5.92) = 14.31	0.01*	(2, 12.90) = 0.10	0.91
Peak knee flexion during loading response	(2, 10.63) = 0.06	0.94	(1, 9.73) = 1.52	0.25	(2, 14.13) = 0.46	0.64
Peak knee flexion during swing	(2, 9.05) = 0.28	0.76	(1, 5.40) = 1.69	0.25	(2, 11.59) = 0.97	0.41
Knee range of motion during loading response	(2, 10.12) = 0.01	0.99	(1, 9.46) = 3.32	0.10	(2, 13.27) = 0.01	0.99
Knee range of motion during single limb support	(2, 10.58) = 1.37	0.30	(1, 4.23) = 3.58	0.13	(2, 12.80) = 0.40	0.68
Knee range of motion across gait cycle	(2, 9.18) = 0.28	0.76	(1, 5.27) = 1.88	0.23	(2, 11.59) = 0/08	0.94
Peak hip flexion during loading response	(2, 18.07) = 0.64	0.54	(1, 13.07) = 0.64	0.44	(2, 18.36) = 0.81	0.46
Peak hip extension during stance	(2, 13.32) = 0.65	0.54	(1, 12.81) = 0.50	0.50	(2, 16.04) = 1.28	0.31
Peak hip flexion during swing	(2, 10.65) = 0.20	0.82	(1, 11.07) = 1.02	0.33	(2, 13.40) = 0.86	0.45
Hip range of motion during single limb support	(2, 10.10) = 1.52	0.26	(1, 12.25) = 0.00	0.97	(2, 14.04) = 1.39	0.28
Hip range of motion across gait cycle	(2, 5.99) = 0.93	0.45	(1, 2.32) = 0.03	0.89	(2, 9.96) = 0.14	0.87
Pelvic range of motion during single limb support	(2, 8.46) = 0.53	0.61	(1, 5.75) = 0.08	0.79	(2, 11.78) = 0.72	0.51

Table 6.14 Statistical breakdown of obstacle crossing trail limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 9.58) = 1.95	0.19	(1, 8.92) = 0.34	0.57	(2, 14.54) = 0.72	0.50
Peak pelvic obliquity during swing	(2, 6.73) = 0.94	0.44	(1, 5.05) = 3.54	0.12	(2, 13.58) = 0.24	0.79
Hip rotation range of motion during single limb support	(2, 8.63) = 0.13	0.88	(1, 7.42) = 0.01	0.97	(2, 13.50) = 0.71	0.51
Pelvic rotation range of motion during single limb support	(2, 10.34) = 5.67	0.02*	(1, 6.31) = 1.08	0.34	(2, 13.75) = 0.78	0.48
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 4.09) = 0.62	0.58	(1, 2.08) = 1.47	0.35	(2, 6.90) = 2.76	0.13
Vertical GRF Fz2	(2, 5.73) = 2.30	0.19	(1, 3.14) = 8.86	0.06	(2, 7.43) = 2.22	0.18
Anterior-Posterior GRF Fy1	(2, 6.00) = 1.87	0.24	(1, 1.81) = 7.32	0.13	(2, 5.07) = 0.01	0.99
Anterior-Posterior GRF Fy2	(2, 6.38) = 3.91	0.08	(1, 3.00) = 0.11	0.76	(2, 8.07) = 1.83	0.22
Load Rate	(2, 4.86) = 0.80	0.50	(1, 3.12) = 3.56	0.15	(2, 7.14) = 0.06	0.95
Decay Rate	(2, 4.75) = 2.29	0.20	(1, 1.90) = 0.44	0.58	(2, 6.80) = 4.54	0.06
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 13.39) = 3.38	0.07	(1, 10.94) = 0.21	0.66	(2, 9.97) = 0.27	0.77
Peak ankle plantarflexor moment during stance	(2, 8.28) = 3.69	0.07	(1, 8.11) = 49.54	<0.01*	(2, 7.83) = 7.12	0.02*
Peak knee flexor moment during loading response	(2, 6.85) = 4.59	0.05*	(1, 6.10) = 0.10	0.77	(2, 5.34) = 1.71	0.27
Peak knee extensor moment during loading response	(2, 5.59) = 4.53	0.07	(1, 5.95) = 1.18	0.32	(2, 6.93) = 0.35	0.72
Peak knee flexor moment during mid stance	(2, 5.05) = 1.19	0.38	(1, 4.78) = 0.53	0.50	(2, 6.66) = 0.29	0.76
Peak knee flexor moment during late stance	(2, 5.63) = 1.30	0.34	(1, 5.99) = 0.44	0.53	(2, 7.00) = 2.00	0.21
Peak knee flexor moment during swing	(2, 5.57) = 1.80	0.25	(1, 5.09) = 3.85	0.11	(2, 6.48) = 2.81	0.13
Peak hip extensor moment during early stance	(2, 2.72) = 2.75	0.22	(1, 2.25) = 0.01	0.92	(2, 4.58) = 0.23	0.80
Peak hip flexor moment during late stance	(2, 7.19) = 0.63	0.56	(1, 4.42) = 0.50	0.52	(2, 4.18) =0.25	0.79
Peak hip extensor moment during swing	(2, 3.47) = 1.25	0.39	(1, 5.39) = 1.03	0.35	(2, 4.89) = 0.24	0.80

Table 6.15 Statistical breakdown of obstacle crossing trail limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 8.25) = 0.26	0.78	(1, 7.89) = 12.27	0.01*	(2, 10.43) = 1.35	0.30
Peak hip abductor during late stance	(2, 6.12) = 0.21	0.81	(1, 3.08) = 18.29	0.02*	(2, 5.37) = 0.26	0.78
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 5.97) = 1.92	0.23	(1, 3.17) = 4.41	0.12	(2, 4.36) = 1.02	0.43
Second peak support moment	(2, 1.10) = 0.30	0.99	(1, 0.01) = 0.10	0.97	(2, 1.10) = 0.12	0.99
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 5.13) = 10.12	0.02*	(1, 2.45) = 5.06	0.13	(2, 7.38) = 6.43	0.02*
A2 – Ankle power generation during pre-swing	(2, 4.40) = 6.22	0.05*	(1, 2.65) = 28.29	0.02*	(2, 6.15) = 3.08	0.12
K1 – Knee power absorption during loading response	(2, 5.71) = 5.62	0.05*	(1, 3.17) = 11.49	0.04*	(2, 5.86) = 2.51	0.16
K2 – Knee power generation during mid-stance	(2, 9.42) = 1.99	0.19	(1, 7.49) = 9.73	0.02*	(2, 6.61) = 0.16	0.85
K3 – Knee power generation during pre-swing	(2, 3.91) = 7.72	0.04*	(1, 0.81) = 8.48	0.26	(2, 5.20) = 1.14	0.39
K4 – Knee power absorption during terminal swing	(2, 7.19) = 2.01	0.20	(1, 4.55) = 21.99	0.01*	(2, 6.04) = 1.82	0.24
H1 – Hip Power generation during loading response	(2, 10.19) = 1.02	0.39	(1, 8.39) = 0.44	0.53	(2, 7.60) = 0.46	0.65
H2 – Hip power absorption during stance	(2, 7.50) = 0.42	0.67	(1, 6.09) = 0.31	0.60	(2, 5.18) = 2.54	0.17
H3 – Hip power generation during pre-swing	(2, 8.52) = 0.52	0.61	(1, 7.09) = 0.09	0.78	(2, 6.19) = 0.18	0.84
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 8.46) = 0.80	0.48	(1, 4.58) = 0.63	0.47	(2, 10.29) = 2.58	0.12
Heel	(2, 5.47) = 0.24	0.79	(1, 1.80) = 2.85	0.25	(2, 5.83) = 0.77	0.51

6.3.5 Discussion – Obstacle Crossing

Previous studies have reported that transtibial amputees were able to negotiate obstacles effectively (Hill *et al.,* 1997; Hill *et al.,* 1999; Hofstad *et al.,* 2006; Hofstad *et al.,* 2009; Vrieling *et al.,* 2007; Vrieling *et al.,* 2009). This was corroborated by the results of the current study where no trips or falls were reported.

Generally, participants selected an intact limb lead preference, whilst literature has suggested both no lead limb preference (Hill *et al.,* 1997) and an affected limb lead preference are present in lower limb amputees (Vrieling *et al.,* 2007). Although individual differences may partially account for these discrepancies, rehabilitation practice may also play a role and results from any particular study interpreted with this in mind. For example, participants in the current study were advised during rehabilitation to cross obstacles using their 'strongest' limb, which is often the intact limb.

Regardless of lead limb preference, improvements were noted in temporal-spatial variables over time (Vrieling *et al.,* 2009). This supported the second (2) hypothesis of an increase in the ability to perform obstacle crossing over time. Peak vertical toe displacement was greater when leading with the affected limb when compared to the intact limb and this could be interpreted as an overcompensation in order to avoid tripping, given that active control of the prosthetic ankle joint during swing phase was not possible.

When trailing with the intact limb, there were increases in ankle ROM during stance phase and peak dorsiflexion during swing phase. Increasing intact limb ankle ROM during stance phase, particularly ankle plantarflexion, has been described as a compensatory mechanism employed in order to aid clearance when leading with the affected limb (Hill *et al.,* 1997). Literature has reported that knee joint ROM may be reduced when leading with the affected limb, due to the posterior shell of the prostheses and socket fit, rendering it from being a suitable trail limb (Hill *et al.*, 1997; Hill *et al.*, 1999). This was the case in the current study when participants chose to lead with the affected limb, although this was not the favoured strategy. In addition, both peak ankle and knee power generation and absorption during stance phase were increased when compared to the affected limb. This suggested that greater demands were placed on the trailing intact limb musculature, which may be interpreted as a stabilisation mechanism in preparation for affected limb swing phase during obstacle crossing. An implication for amputees is that although a preferred lead limb may be selected, on occasions unexpected obstacles may be presented. For the current participant group, this may necessitate the more unfavourable or ineffective affected lead limb strategy which in turn may increase the likelihood of tripping and/or falling. Literature has reported that lower limb amputees were less able to negotiate unexpected obstacles and suggested introducing the practice of these tasks during rehabilitation, which is supported by results in the current study given the reduced affected lead limb functioning (Hofstad *et al.*, 2006; Vrieling *et al.*, 2007; Vrieling *et al.*, 2009; Hofstad *et al.*, 2009).

When leading with the intact limb, knee ROM during the gait cycle and peak knee flexion during swing phase were greater than when leading with the affected limb. In addition power absorption at the knee during swing phase (K4) was greater when leading with the intact limb. This increased joint mobility and control when crossing an obstacle may have played an important role in the selection of lead limb, perhaps as participants were more confident of avoiding contact with the obstacle with the intact limb.

Similarly, once the intact limb had crossed the obstacle and landed, increased knee ROM, load rate and peak vertical GRFs were observed when compared to the affected limb which was maintained in a position of approximately 15 degrees flexion. Literature has suggested that reduced knee ROM upon landing with the affected limb reflected instability in the knee flexors in preparation for the subsequent stance phase or an inability to effectively control musculature about the knee (Hill *et al.*, 1999; Hofstad *et al.*, 2006). This may further elicit the reasons for a lead limb preference observed in the current study, as the intact limb is more capable managing the demands during stance phase, following obstacle crossing.

This hypothesis was corroborated by increased intact limb peak knee extensor moment during loading response following obstacle crossing. Additionally, power generation (A2, H3) and absorption (K3) during stance phase were greater when leading with the intact limb in comparison to the affected limb.

Although the selection of a lead limb preference may be due to the increased ability to ‘push off’ at the end of the preceding stance phase, when compared to the affected limb (Hill *et al.*, 1999), results from the current study suggest that the role of the intact limb having crossed the obstacle is also important. These results suggest that participants may have selected a lead limb preference for two reasons. Firstly, the greater control possible when crossing the obstacle as seen in the joint kinematics. Secondly, the ability to maintain relatively high joint moments and generate and absorb power in the stance phase limb during the subsequent stance phase following obstacle crossing. These factors have implications for those involved in the care and rehabilitation of transtibial amputees in that by increasing affected limb knee and hip joint ROM through stretching exercises of the hip flexors, amputees ability to cross obstacles when leading with the affected limb may improve. Combined with the practice of obstacle crossing during rehabilitation, this may reduce the lead limb preference observed following discharge from rehabilitation and increase amputees ability to avoid unexpected obstacles and subsequent falls by increasing versatility.

Despite a dependence on the intact limb that did not reduce over time, obstacle crossing in the current participant group improved. Participants were able to perform the task more quickly and with a sufficient degree of functioning, therefore the second hypothesis of an improvement in the ability to perform obstacle crossing (2) was accepted.

6.3.6 Stepping Gait

6.3.7 Temporal-Spatial Variables

6.3.7.1 Stepping Down Gait

Temporal-spatial variables are presented in Table 6.16 with complete statistical analyses provided in Tables 6.20 and 6.23.

Table 6.16 highlights the increases walking velocity when leading with both the affected limb (36%) and the intact limb (24%). Post-hoc analysis revealed these increases to be significant between one and six months post-discharge (p=0.04). In addition, these increases of 0.26 and 0.19 m/s when leading with the affected and intact limbs respectively, also represent a highly clinically meaningful increase in walking velocity. Stride length increased significantly following discharge when leading with both the affected (17%) and intact limbs (20%) (Table 6.16). Post-hoc analysis revealed these increases to be between one and six months post-discharge (p=0.01). Lead limb relative stance duration remained relatively unchanged across time (Table 6.16). However, the reduction in trail limb relative stance duration, particularly when trailing with the affected limb, resulted in a significant time main effect. Post-hoc analysis revealed these differences to be between one and three (p=0.04) and one and six months (p=0.01) post-discharge. In addition, relative stance duration was significantly reduced when trailing with the affected limb when compared to trailing with the intact limb (p=0.01).

6.3.7.2 Stepping Up Gait

Temporal-spatial variables are presented in Table 6.17 with complete statistical analyses provided in Tables 6.27 and 6.30.

When leading with the affected limb, there were no observable increases in walking velocity and, despite an increase of 22% when leading with the intact limb, no significant time main effect was reported (Table 6.17). However, the 0.17 m/s increase in walking velocity when leading with the intact limb represents a clinically meaningful increase. Walking velocity was also comparable at six months post-discharge irrespective of lead limb selected (Table 6.17). Similar trends were noted in stride length where increase of 6% and 14% when leading with the affected and intact limbs respectively, did not induce a significant time main effect (Table 6.17). Equally, there were no visible or statistically significant between limb differences in stride length when leading with either limb (Table 6.17).

Lead limb relative stance duration did not change significantly over time, although this was significantly higher when leading with the intact limb when compared to the affected limb ($p=0.02$). Trail limb relative stance duration when trailing with the intact limb was significantly greater than trailing with the affected limb ($p=0.05$) although no significant time effect was reported.

Table 6.16 Mean (SD) temporal-spatial variables of stepping down gait. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Walking Velocity (m/s)	Lead Affected	0.72 (0.18)	0.88 (0.16)	0.98 (0.13)	1.19 (0.0)
	Lead Intact	0.79 (0.0)	0.96 (0.22)	0.98 (0.19)	-
Stride Length (m)	Lead Affected	1.06 (0.13)	1.17 (0.16)	1.24 (0.12)	1.40 (0.10)
	Lead Intact	1.05 (0.10)	1.25 (0.24)	1.26 (0.14)	-
Relative Stance Duration (% GC)	Lead Affected	58 (4.1)	58 (2.7)	57 (2.0)	55 (1.3)
	Lead Intact	60 (8.4)	60 (2.7)	59 (4.3)	-
	Trail Intact	73 (3.2)	71 (3.1)	70 (3.6)	66 (0.2)
	Trail Affected	71 (1.3)	66 (2.1)	66 (2.0)	-

***Data from two participants, not included in statistical analyses**

Table 6.17 Mean (SD) temporal-spatial characteristics of stepping up gait. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Walking Velocity (m/s)	Lead Affected	0.94 (0.0)	1.01 (0.10)	0.94 (0.13)	1.21 (0.0)
	Lead Intact	0.76 (0.14)	0.92 (0.13)	0.93 (0.16)	1.22 (0.0)
Stride Length (m)	Lead Affected	1.21 (0.0)	1.34 (0.13)	1.29 (0.10)	1.45 (0.0)
	Lead Intact	1.08 (0.11)	1.27 (0.20)	1.23 (0.12)	1.48 (0.0)
Relative Stance Duration (% GC)	Lead Affected	63 (0.0)	63 (2.0)	64 (2.5)	60 (1.0)
	Lead Intact	70 (3.8)	68 (3.0)	68 (3.1)	65 (0.0)
	Trail Intact	63 (0.0)	63 (1.7)	64 (3.8)	61 (2.8)
	Trail Affected	62 (2.4)	60 (1.8)	59 (2.8)	58 (-)

*Data from two participants, not included in statistical analyses.

6.3.8 Lead Limb Preference

Participants lead limb preferences during gait when stepping up to and from a new level are presented in Table 6.18, no statistical comparisons were drawn.

6.3.8.1 Stepping Down Gait

As is observed in Table 6.18, participants favoured an affected limb lead preference. At one and three months post-discharge, participants displayed a strong bias towards an affected limb lead preference, with only two exceptions at three months displaying no lead limb preference (participants one and six). At six months post-discharge, these exceptions displayed an intact limb lead preference, with only two more changing from an affected limb to an intact limb lead preference (Table 6.18).

6.3.8.2 Stepping Up Gait

During stepping up gait, participants generally displayed an intact limb lead preference (Table 6.18). One participant maintained an affected limb lead preference up to six months post-discharge (participant two), with one (participant one) and two (participants six and seven) displaying an affected limb lead preference at three and six months respectively. Generally, once a lead limb preference had been selected, participants tended to employ this strategy consistently (Table 6.18).

Table 6.18 Individual participant lead limb preferences during gait whilst stepping up to and down from a new level.

Patient Number	One Month				Three Months				Six Months				Twelve Months			
	Stepping Up															
	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference
1					4	2	6	A	0	7	7	I	1	4	5	I
2	8	0	8	A	6	0	6	A	7	0	7	A	6	0	6	A
3	1	4	5	I	0	6	6	I	1	7	8	I				
4	0	6	6	I	0	4	4	I	3	4	7	I				
5	0	6	6	I	2	4	6	I	1	5	6	I				
6					3	4	7	I	6	1	7	A				
7					1	4	5	I	3	2	5	A				
	Stepping Down															
	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference	Affected	Intact	Total	Preference
1					3	3	6	No Pref	3	4	7	I	6	0	6	A
2	6	0	6	A	6	0	6	A	5	2	7	A	6	0	6	A
3	4	1	5	A	6	1	7	A	6	2	8	A				
4	6	0	6	A	3	1	4	A	0	7	7	I				
5	5	1	6	A	6	0	6	A	3	3	6	No Pref				
6					4	4	8	No Pref	2	5	7	I				
7					4	1	5	A	4	0	4	A				

6.3.9 Foot Marker Trajectories

Foot marker trajectory data are presented in Figures 6.14 (stepping down gait) and 6.15 (stepping up gait) with complete statistical analyses provided in Tables 6.22, 6.25, 6.29 and 6.32.

6.3.9.1 Stepping Down Gait

6.3.9.2 Lead Limb

Figure 6.14 illustrates the consistent toe trajectories of the lead limb in the six months following discharge from rehabilitation, which resulted in no significant main effects. When leading with the affected limb, there was virtually no change in heel trajectory over time. Coupled with the increased peak heel trajectory displacement and the changes over time when leading with the intact limb (Figure 6.14) a significant interaction effect was observed (p=0.01).

6.3.9.3 Trail Limb

During stance phase, heel and toe trajectories when leading with either limb were very consistent over time (Figure 6.14). There were small visible changes in both the magnitude and timing of toe and heel displacements during swing phase, however, no statistically significant main effects were obtained (Figure 6.14).

6.3.9.4 Stepping Up Gait

6.3.9.5 Lead Limb

Lead limb peak toe trajectory displacement visibly increased between three and six months when leading with the affected limb, whereas the opposite was true of peak heel displacement when leading with the affected limb, although no statistically significant results were found (Figure 6.15). There were no observable differences or statistically significant main effects in peak heel or toe trajectories when leading with the intact limb.

6.3.9.6 Trail Limb

When trailing with either limb, toe trajectory profiles displayed some variation over time, although no statistically significant main effects were reported (Figure 6.15). Similarly, when examining heel trajectories, although peak heel trajectory displacement seemed greater when trailing with the affected limb when compared to trailing with the intact limb, no significant limb main effect was observed (Figure 6.15).

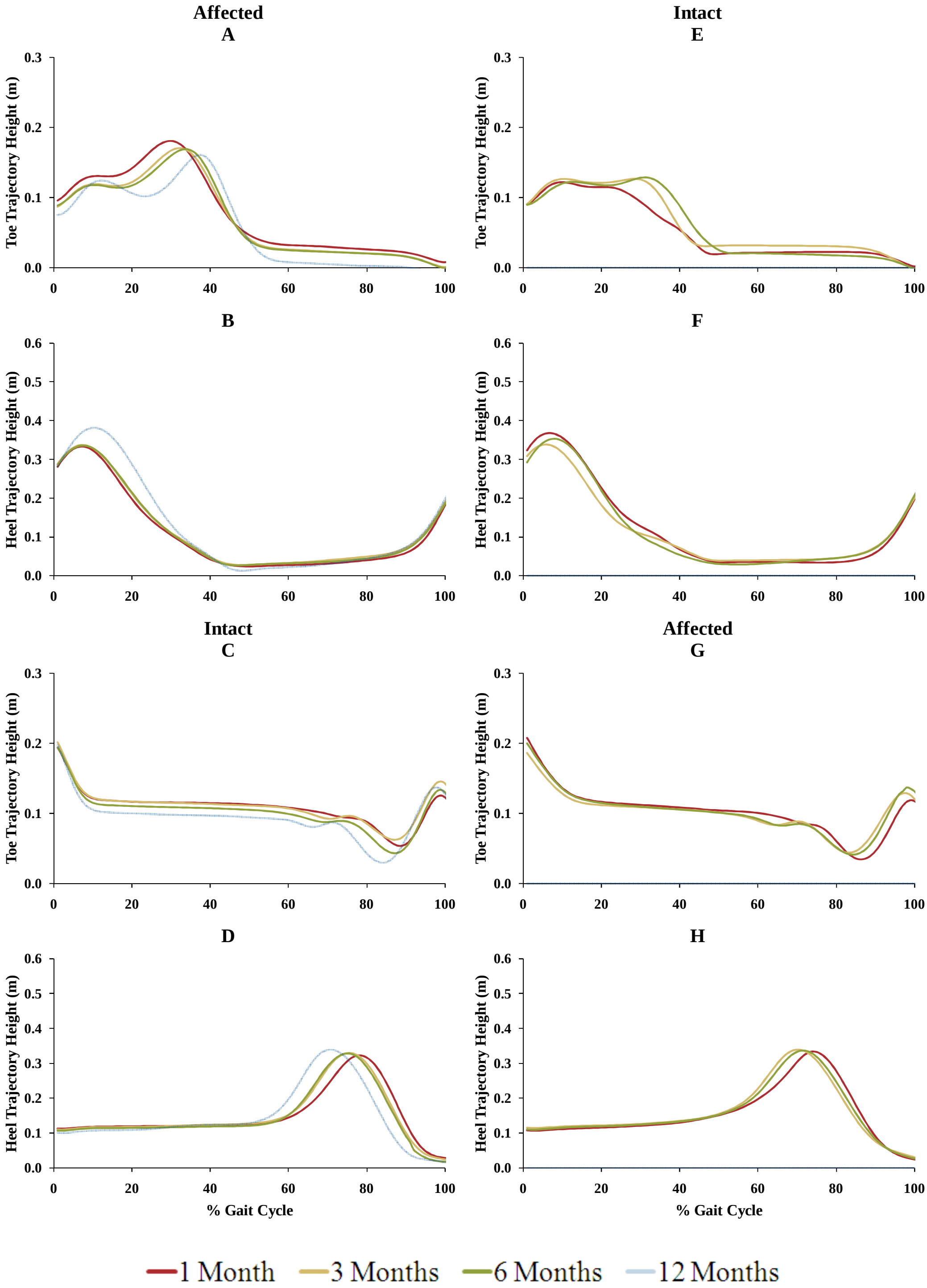


Figure 6.14 Group mean stepping down gait foot marker trajectories for the lead affected (toe – A, heel - B), trail intact (toe – C, heel - D), lead intact (toe – E, heel - F) and trail affected (toe – G, heel - H) limbs. Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off, trail limb follows conventional definition. Data at 12 months from n=2.

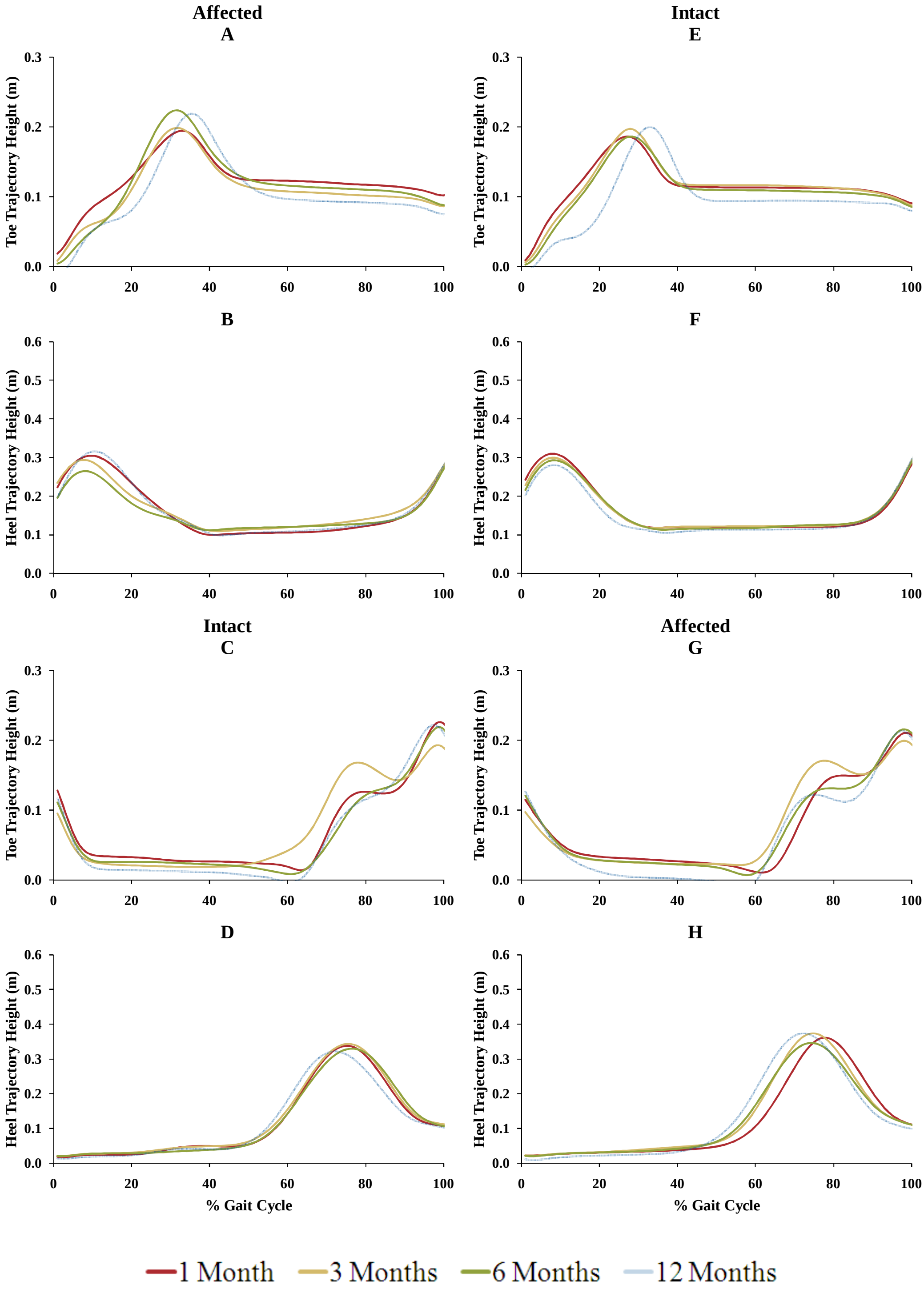


Figure 6.15 Group mean stepping up gait foot marker trajectories for the lead affected (toe – A, heel - B), trail intact (toe – C, heel - D), lead intact (toe – E, heel - F) and trail affected (toe – G, heel - H) limbs. Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off, trail limb follows conventional definition. Data at 12 months from n=2.

6.3.10 Joint Kinematics – Stepping Down Gait

Joint kinematics are presented in Figures 6.16 (sagittal plane lead limb), Appendix N (frontal and transverse plane lead limb), 6.17 (sagittal plane trail limb) and Appendix N (frontal and transverse) with complete statistical analyses provided in Tables 6.20, 6.21, 6.23 and 6.24.

6.3.10.1 Lead Limb

Figure 6.16 illustrates distinct between limbs differences in ankle kinematics over time. Firstly, peak ankle plantarflexion during loading response was greater when leading with the affected limb when compared to the intact limb ($p<0.01$). Peak ankle dorsiflexion during stance phase was consistent over time when leading with the affected limb (Figure 6.16). In comparison, when leading with the intact limb this variable was increased at one and six months but reduced at three months, resulting in a significant interaction effect ($p<0.01$). Similarly, ankle ROM during stance phase when leading with the affected limb was relatively unchanged over time (Figure 6.16), the same variable being increased at one and six months but reduced at three months when leading with the intact limb, resulting in a significant interaction effect ($p<0.01$). A significant limb effect indicated that, peak ankle plantarflexion during swing phase was significantly higher when leading with the intact limb when compared to the affected limb ($p<0.01$).

Although peak knee flexion during loading response seemed to reduce over time when leading with the intact limb, no significant time main effect was observed (Figure 6.16). However, both peak knee flexion ($p=0.01$) and knee ROM ($p=0.01$) during loading response were significantly higher when leading with the intact limb in comparison to the affected limb. In addition, knee ROM during single limb support was also greater when leading with the intact limb in comparison to the affected limb ($p=0.01$). Although an increase in hip flexion during stance phase was observed when leading with the intact limb, no significant main effects were reported for variables relating to sagittal plane hip kinematics (Figure 6.16).

When leading with both the affected and intact limbs sagittal plane pelvic kinematics generally remained in anterior tilt (Figure 6.16). A significant limb main effect was observed for sagittal plane pelvic ROM during single limb support ($p=0.01$), likely due to the large differences observed at one month post-discharge (Figure 6.16).

Relatively small changes were observed in frontal and transverse plane kinematics of the pelvis and hip (Appendix N). However, post-hoc analysis of pelvic ROM of motion during single limb support revealed significant differences between one and three ($p=0.02$) and one and six ($p=0.02$) months post-discharge, likely due to change observed when leading with the intact limb (Appendix N).

6.3.10.2 Trail Limb

Both ankle ROM ($p<0.01$) and peak ankle dorsiflexion ($p<0.01$) during stance phase were significantly higher when trailing with the intact limb when compared to the affected limb (Figure 6.17).

Similarly, knee ROM during single limb support was greater when trailing with the intact limb when compared to trailing with the affected limb ($p=0.05$). Knee ROM during loading response when trailing with the intact limb remained relatively unchanged over time, whilst being greater in magnitude when compared to the affected limb (Figure 6.17). Coupled with the reduction observed in this variable when leading with the affected limb, a significant interaction effect was observed ($p=0.03$). Although the magnitude of sagittal plane knee profiles observably reduced over time when trailing with the affected limb, these changes were not statistically significant (Figure 6.17). Sagittal plane hip kinematics remained consistent across time and were comparable between limbs (Figure 6.17). With the exception of a visible increase in anterior pelvic tilt when trailing with the affected limb, there were few changes in sagittal plane pelvis kinematics and no statistically significant main effects were reported (Figure 6.17).

Post-hoc analysis of peak pelvic obliquity during swing phase revealed a significant difference between one and three months post-discharge (p=0.03). This may have been a reflection of the reduction in the variable noted when trailing with the intact limb (Appendix N). No further significant main effects were observed for pelvis or hip kinematics in the frontal or transverse plane (Appendix N).

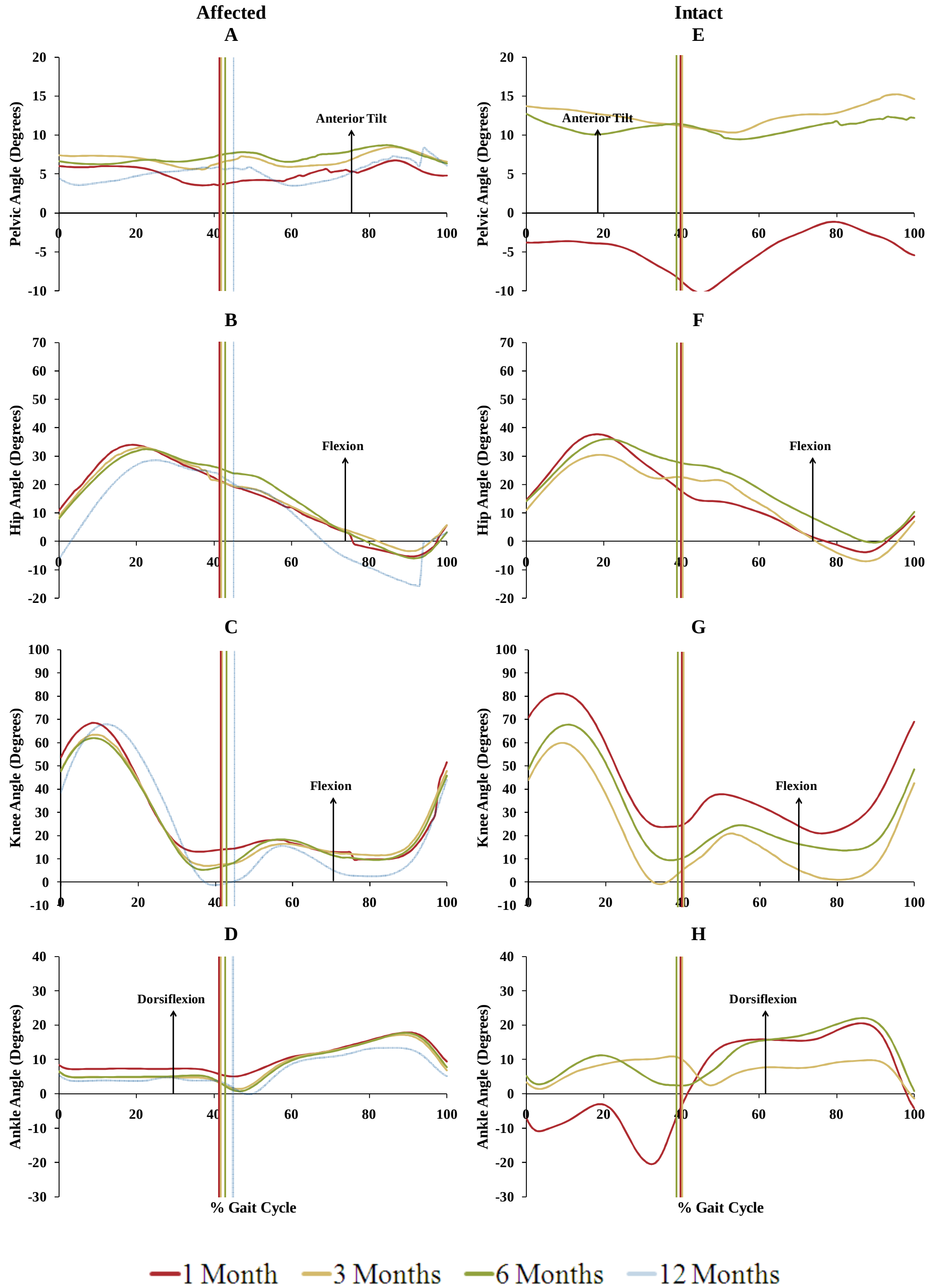


Figure 6.16 Group mean sagittal plane kinematics of the lead affected limb pelvis (A), hip (B), knee (C) and ankle (D) and lead intact limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during stepping down gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

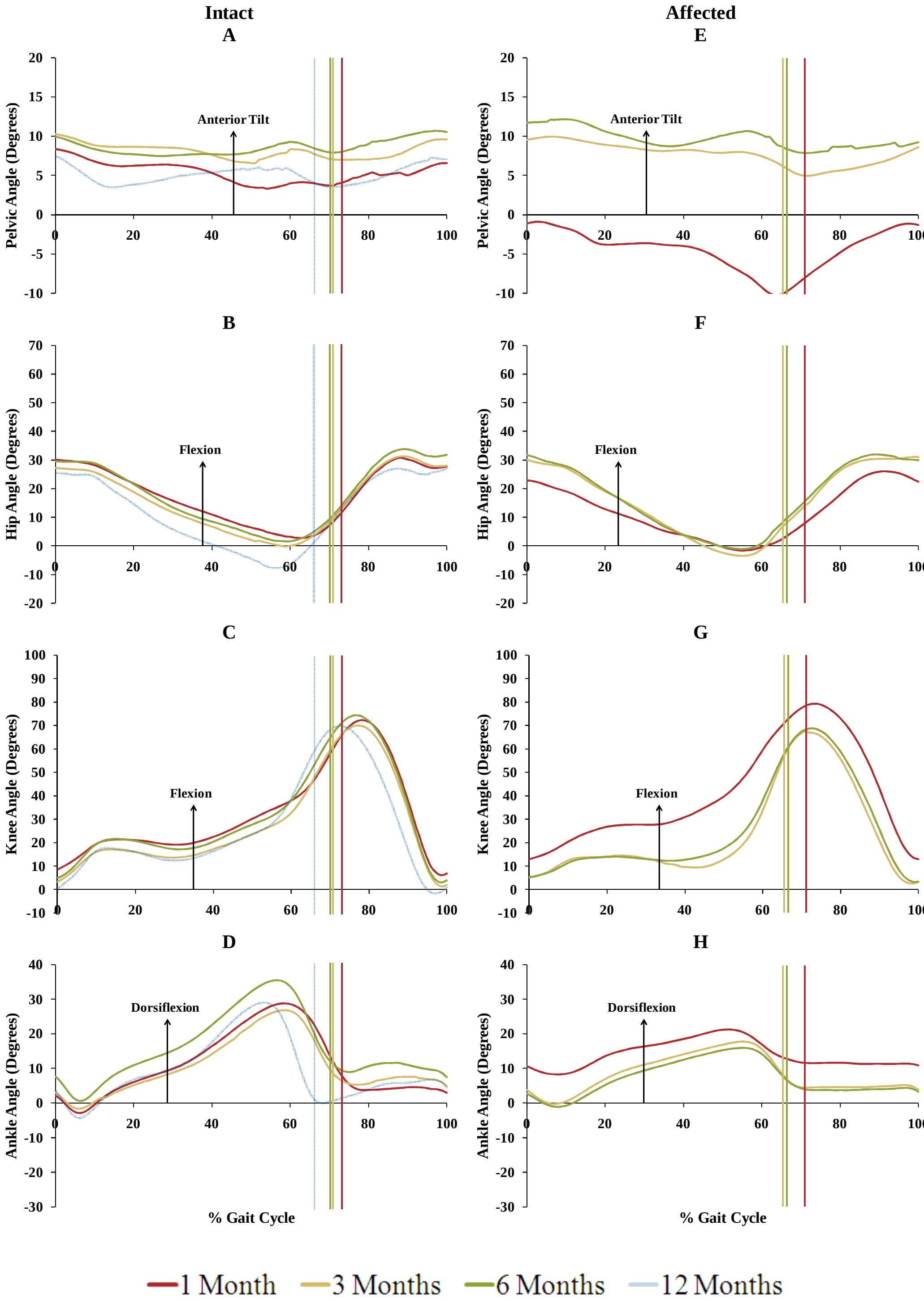


Figure 6.17 Group mean sagittal plane kinematics of the trail intact limb pelvis (A), hip (B), knee (C) and ankle (D) and trail affected limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during stepping down gait. Vertical lines represent toe off. Data at 12 months from n=2.

6.3.11 Ground Reaction Forces – Stepping Down Gait

Loading and decay rates are presented in Table 6.19, GRF data are presented in Figure 6.18 with complete statistical analyses provided in Tables 6.21 and 6.24.

6.3.11.1 Lead Limb

Upon landing when stepping down during gait, load rate ($p=0.02$), initial peak vertical GRF ($p=0.05$) and peak posterior GRF ($p<0.01$) were significantly higher when leading with the intact limb in comparison with leading with the affected limb (Table 6.19, Figure 6.18).

A significant time effect was reported for peak anterior GRF (Figure 6.18), post-hoc analysis revealing a significant increase between one and six months post-discharge ($p=0.02$).

6.3.11.2 Trail Limb

Similar effects were observed in trail limb GRF analyses, where decay rate ($p=0.01$), second peak vertical GRF ($p=0.03$) and peak anterior GRF ($p=0.01$) were significantly greater when trailing with the intact limb when compared to trailing with the affected limb (Table 6.19, Figure 6.18).

A significant interaction effect was reported for peak posterior GRF ($p=0.01$) due to longitudinal changes and no discernable limb effect (Figure 6.18).

Table 6.19 Mean (SD) loading and decay rate of stepping down gait. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Load Rate (BW/s)	Lead Affected	5.1 (1.5)	6.5 (1.9)	7.8 (3.1)	7.9 (0.0)
	Lead Intact	8.7 (4.8)	11.6 (3.4)	11.0 (4.0)	-
	Trail Intact	3.5 (1.8)	5.0 (1.5)	5.2 (1.2)	7.8 (0.0)
	Trail Affected	4.1 (0.6)	4.1 (0.1)	4.3 (0.7)	-
Decay Rate (BW/s)	Lead Affected	4.5 (0.4)	4.8 (1.2)	5.2 (0.9)	7.5 (0.0)
	Lead Intact	4.0 (0.9)	4.4 (0.7)	6.7 (2.0)	-
	Trail Intact	4.1 (1.2)	5.3 (1.2)	5.8 (1.3)	7.2 (0.0)
	Trail Affected	3.0 (0.7)	3.9 (1.3)	4.4 (0.5)	-

*Data from two participants, not included in statistical analyses

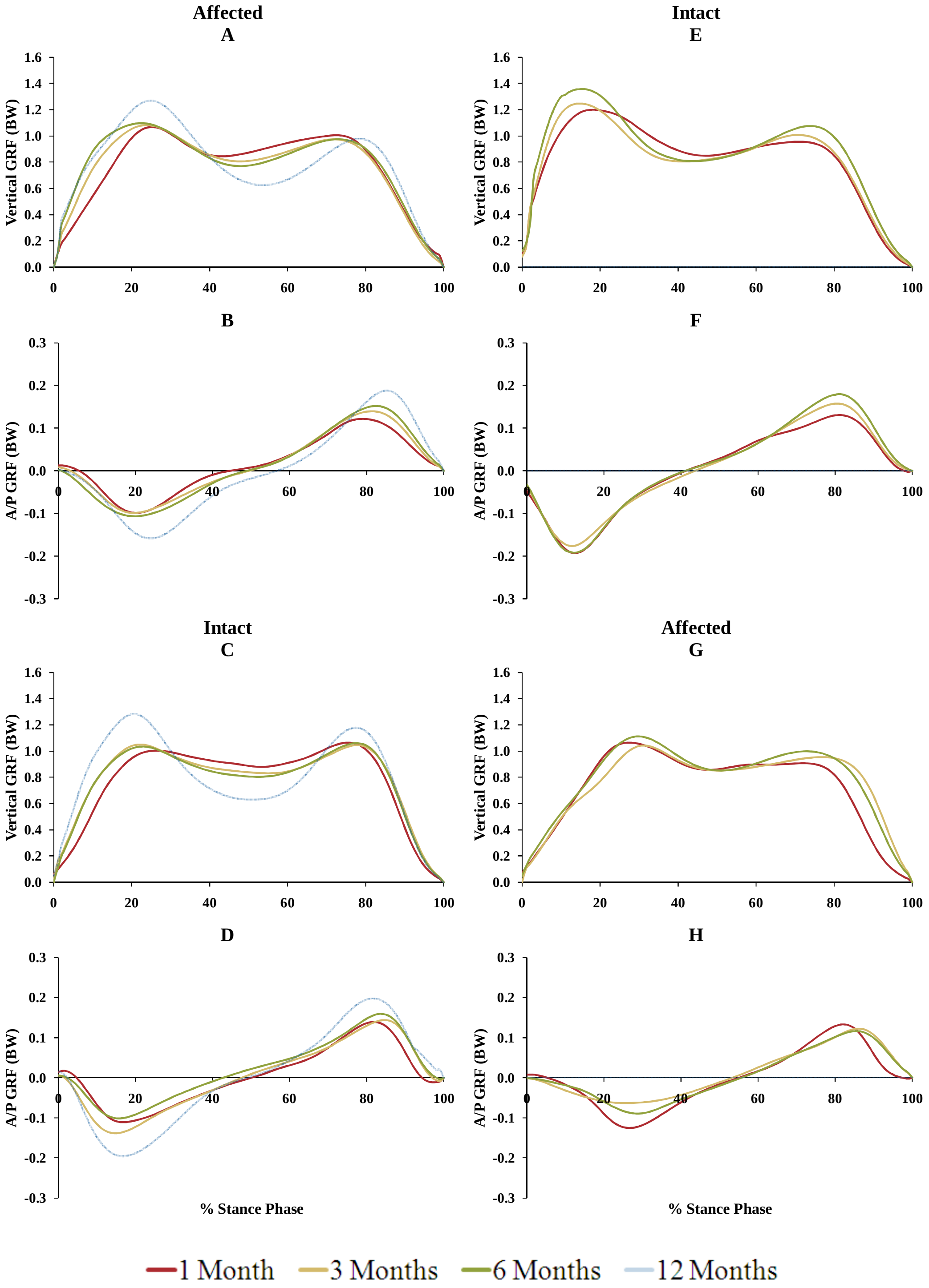


Figure 6.18 Group mean vertical and anterior-posterior ground reaction forces for the lead affected (A and B), lead intact (E and F), trail intact (C and D) and trail affected (G and H) limbs. All data normalised and presented as times body weight (BW). Time normalised to 100% of stance phase during stepping down gait. Data at 12 months from n=2. Vertical and anterior are positive.

6.3.12 Joint Kinetics – Stepping Down Gait

Joint moments are presented in Figures 6.19 (sagittal and frontal plane lead limb) 6.20 (sagittal and frontal plane trail limb), joint powers are presented in Figures 6.21 (sagittal plane lead limb), 6.22 (sagittal plane trail limb) and support moments are presented in Appendix O with complete statistical analyses provided in Tables 6.21, 6.22, 6.24 and 6.25.

6.3.12.1 Lead Limb

Ankle joint moment profiles reduced in magnitude over time when leading with both limbs and were generally larger when leading with the intact limb, although no statistically significant main effects were reported (Figure 6.19).

Peak knee extensor moment during loading response increased over time when leading with both limbs, post-hoc analysis revealing these increases to be between three and six months post-discharge (p=0.03). In addition, peak knee extensor moment during loading response was increased when leading with the intact when compared to leading with the affected limb (p=0.01). Peak knee flexor moment during swing phase was also significantly greater when leading with the intact limb in comparison to the affected limb (p=0.05) (Figure 6.19).

Sagittal plane hip moment profiles were relatively similar across time when leading with the affected limb, observable changes were apparent when leading with the intact limb although no statistically significant main effects were reported (Figure 6.19).

Significant interaction effects were observed as peak hip abductor moment during both early (p=0.01) and late stance phase (p=0.02) changed over time, the magnitude of change being larger when leading with the intact limb (Figure 6.19).

As can be seen in Appendix O, initial peak support moment reduce significantly over time in both limbs but more markedly when leading with the intact limb, post-hoc analysis revealing these differences between one and six months (p=0.04). A similar effect was noted in the second peak support moment which resulted in a significant interaction effect (p=0.03).

Both peak power absorption (A1) and generation (A2) at the ankle were observably increased when leading with the intact limb compared to the affected limb, although no statistically significant limb effect was reported (Figure 6.21).

Similarly, power bursts K1, K2 and K3, were observably larger when leading with the intact limb in comparison to the affected limb, though no statistically significant limb effects were reported (Figure 6.21). However, peak knee power absorption during swing phase (K4) was significantly greater when leading with the intact limb compared to the affected limb (p=0.01).

Lead limb hip power profiles were similar between limbs during swing phase (Figure 6.21). Despite visible between limb differences in hip power profiles during stance phase and changes in magnitude when leading with the intact limb, no significant main effects were reported (Figure 6.21).

6.3.12.2 Trail Limb

Peak ankle dorsiflexor moment during loading response increased significantly over time (p=0.05), although post hoc analysis did not reveal where this difference occurred (Figure 6.20). In addition, peak ankle plantarflexor moment during stance phase was observably larger when trailing with the intact limb when compared to trailing with the affected limb, although this was not statistically significant (Figure 6.20).

Peak knee flexor moment increased over time when trailing with the intact limb, decreased over time when leading with the affected limb and was increased in magnitude in the intact limb at six months post-discharge resulting in a significant interaction effect (p=0.05). Peak knee extensor moments were visibly increased when trailing with the intact limb although no statistically significant limb effect was reported (Figure 6.20).

Post-hoc analysis revealed that peak hip flexor moment during late stance phase significantly increased in both limbs between one and three (p=0.04) and one and six months (p=0.05) post discharge (Figure 6.20). In addition, this variable was significantly increased when trailing with the intact limb in comparison to the affected limb (p=0.01).

The reduction over time in peak hip abductor moment during early stance phase was greater when trailing with the intact limb in comparison to trailing with the affected limb, resulting in a significant interaction effect (p=0.03). Peak hip abductor moment during late stance phase reduced significantly over time when trailing with both limbs, post-hoc analysis not revealing where the differences occurred (p=0.02). In addition, this variable was significantly increased when trailing with the intact limb in comparison to the affected limb (p=0.05).

Both initial and second peak support moments reduced over time, although post-hoc analysis only revealed a significant difference for the latter between one and three months post-discharge (p=0.04) (Appendix O).

Significant limb effects were observed for both peak power absorption (A1) (p=0.01) and generation (A2) (p=0.04), these power bursts being increased when trailing with the intact limb in comparison to the affected limb (Figure 6.22).

Peak knee power generation during stance phase (K2) was also greater when trailing with the intact limb in comparison to the affected limb (p=0.05). Peak knee power absorption during swing phase (K4) reduced over time when trailing with the affected limb, an increase followed by a decrease being noted when trailing with the intact limb resulting in a significant interaction effect (p=0.03) (Figure 6.22).

Peak power absorption (H2) increased significantly between one and three months post-discharge ($p=0.04$). A significant time main effect was also reported for power burst H3 ($p=0.05$), although post-hoc analysis did not reveal where the significant increases occurred.

6.3.12.3 Data for n=2 at 12 Months Post-Discharge

Participants maintained an affected limb lead preference at 12 months post-discharge. Walking velocity and stride length increased with relative stance duration decreasing over time between 6 and 12 months post-discharge. Peak heel and toe vertical trajectories reduced and increased respectively at 12 months post-discharge. There were few changes in trail limb vertical heel and toe trajectories.

There were no large changes in joint kinematics when leading with the affected limb at 12 months post-discharge, although knee ROM during loading response seemed to increase, perhaps reflecting better control of the knee having stepped down to a new level.

Load and decay rates increased in both the lead and trail limb when leading with the affected limb at 12 months post-discharge. Similarly, initial and second peak vertical GRFs and peak anterior and posterior GRFs, increased during this time period, suggesting that participants were more capable to experience greater forces and under greater loading/unloading conditions.

These observations were linked to the increases in peak knee extensor and ankle plantarflexor moments during stance phase when leading with the affected limb apparent at 12 months post-discharge. Trail limb joint moments remained similar between 6 and 12 months post-discharge, although an increase was noted in hip flexor moment during late stance phase.

The aforementioned, increased lead limb moments were also reflected in the joint power analysis at 12 months post-discharge. Peak power bursts A2, K1, K2, and K3 were increased at 12 months post-discharged, although affected limb power generation and absorption was not as great as in the lead intact limb at six months. Intact limb trail joint power bursts increased unanimously with the exception of K4, further highlighting the inter-limb differences when acting as the trail limb.

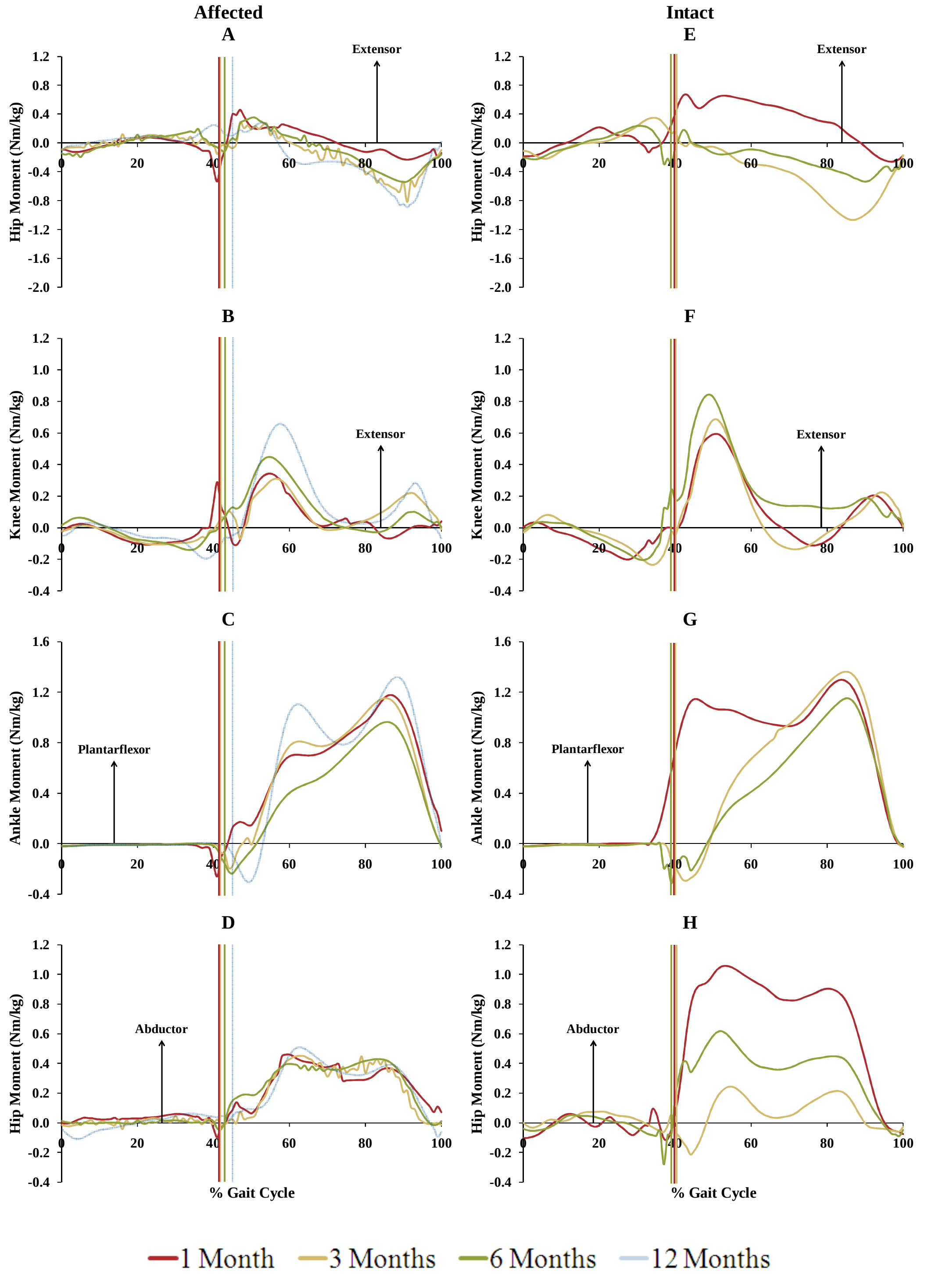


Figure 6.19 Group mean joint moments for the lead affected limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and lead intact limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during stepping down gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

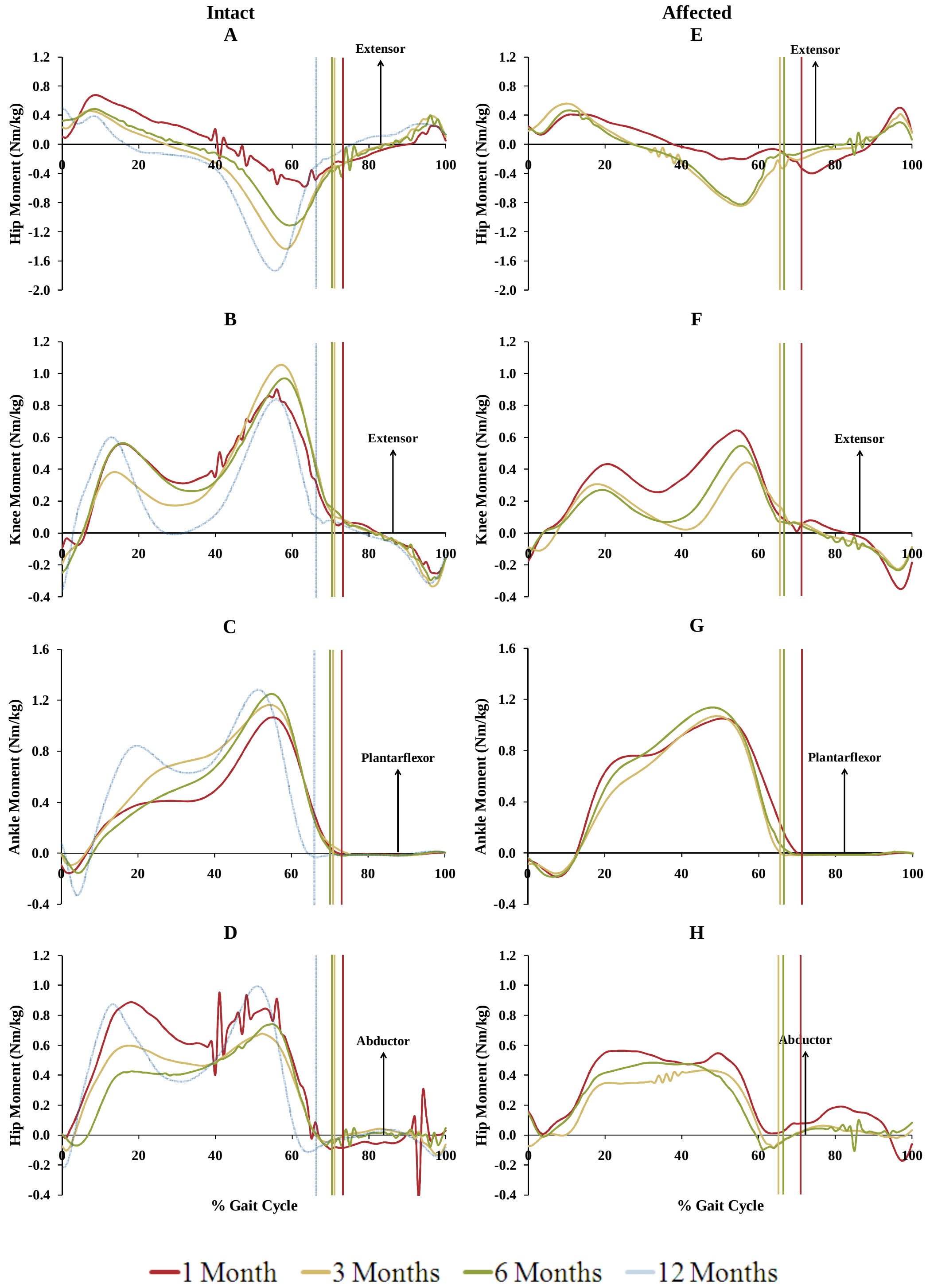


Figure 6.20 Group mean joint moments for the trail intact limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and trail affected limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during stepping down gait. Vertical lines represent toe off. Data at 12 months from n=2.

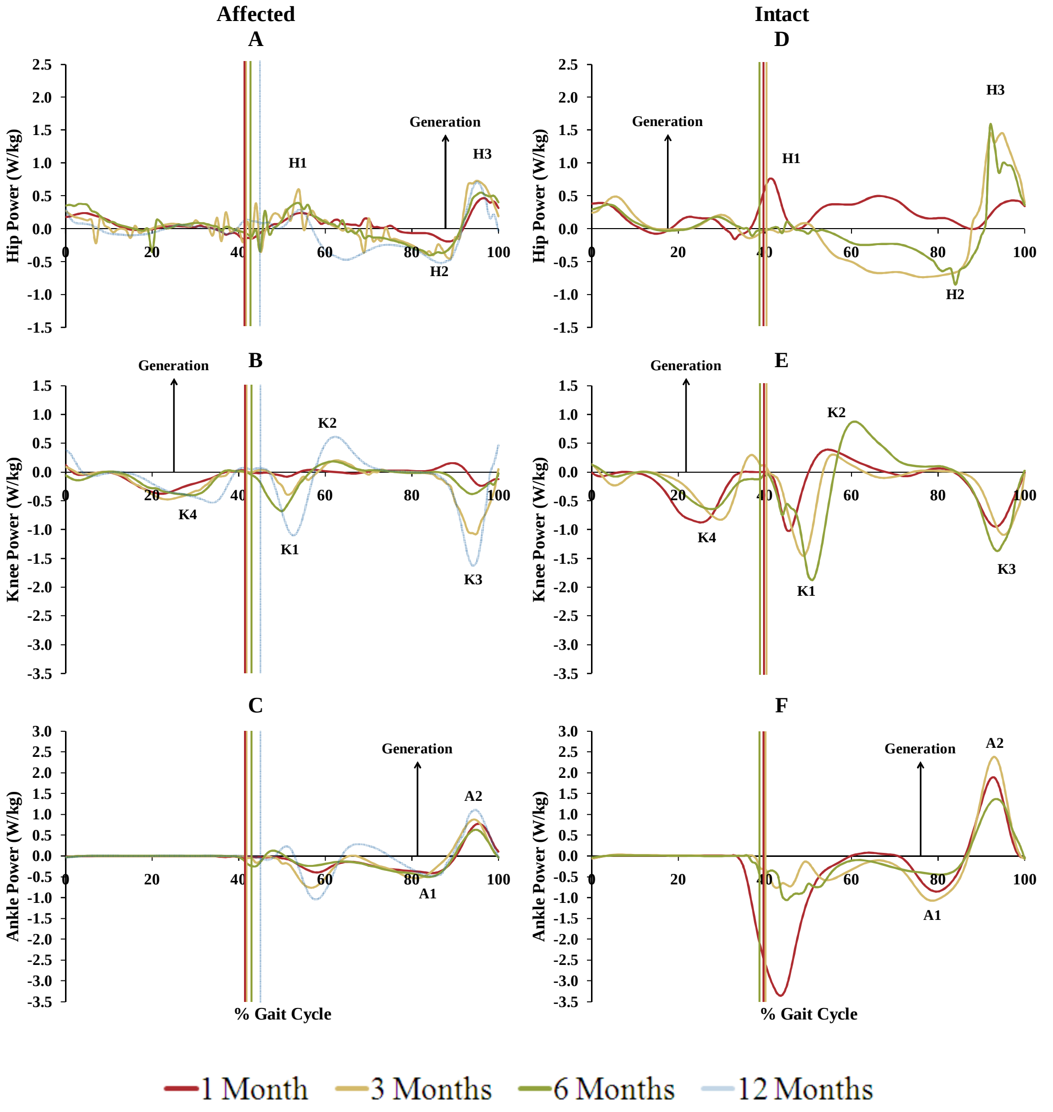


Figure 6.21 Group mean joint powers for the lead affected limb hip (A), knee (B), ankle (C) and lead intact limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during stepping down gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

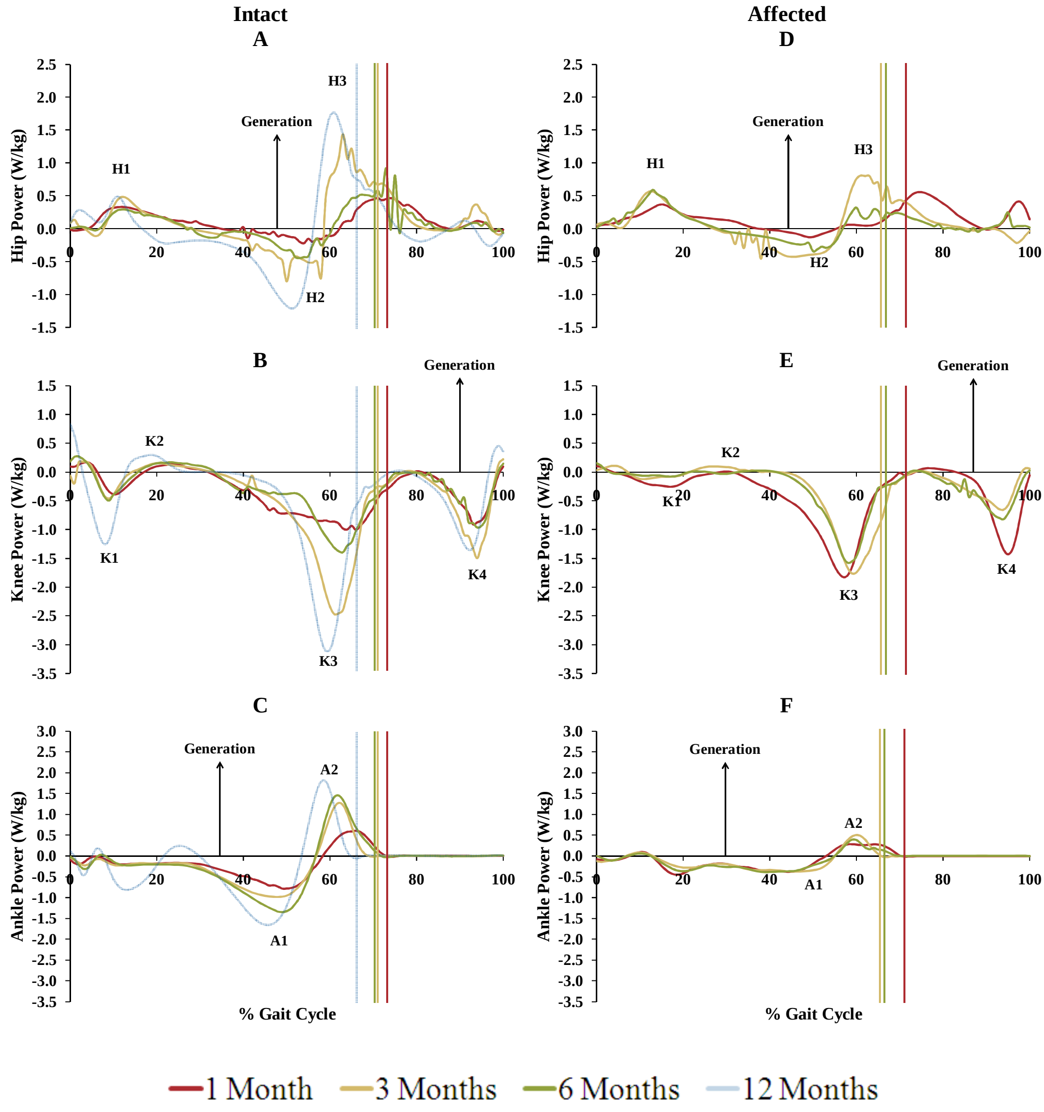


Figure 6.22 Group mean joint powers for the trail intact limb hip (A), knee (B), ankle (C) and trail affected limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during stepping down gait. Vertical lines represent toe off. Data at 12 months from n=2.

Table 6.20 Statistical breakdown of stepping down gait lead limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 7.09) = 7.71	0.02*	(1, 4.30) = 0.03	0.86	(2, 12.42) = 0.69	0.52
Stride Length	(2, 9.19) = 9.10	0.01*	(1, 6.79) = 0.03	0.87	(2, 14.11) = 0.88	0.44
Relative Stance Duration	(2, 9.10) = 0.69	0.53	(1, 7.30) = 1.49	0.26	(2, 14.33) = 0.25	0.78
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 9.84) = 1.74	0.23	(1, 15.57) = 8.97	0.01*	(2, 9.56) = 3.47	0.07
Peak ankle dorsiflexion during stance	(2, 10.20) = 18.76	<0.01*	(1, 15.04) = 4.82	0.04*	(2, 9.70) = 16.23	<0.01*
Peak plantarflexion during swing	(2, 9.43) = 3.12	0.09	(1, 15.76) = 20.61	<0.01*	(2, 9.31) = 3.81	0.06
Ankle range of motion during stance	(2, 10.47) = 21.39	<0.01*	(1, 15.93) = 16.23	<0.01*	(2, 10.18) = 21.10	<0.01*
Peak knee flexion during loading response	(2, 10.02) = 1.05	0.39	(1, 12.02) = 11.04	0.01*	(2, 11.49) = 1.02	0.39
Peak knee flexion during swing	(2, 12.10) = 2.72	0.11	(1, 12.21) = 2.72	0.13	(2, 13.27) = 1.29	0.31
Knee range of motion during loading response	(2, 10.15) = 3.03	0.09	(1, 15.87) = 8.81	0.01*	(2, 9.94) = 2.92	0.10
Knee range of motion during single limb support	(2, 8.94) = 1.68	0.24	(1, 12.20) = 11.08	0.01*	(2, 10.33) = 2.08	0.17
Knee range of motion across gait cycle	(2, 9.01) = 0.18	0.84	(1, 11.59) = 2.80	0.12	(2, 8.72) = 0.21	0.81
Peak hip flexion during loading response	(2, 15.98) = 1.98	0.17	(1, 17.80) = 0.01	0.97	(2, 17.29) = 0.39	0.68
Peak hip extension during stance	(2, 13.65) = 0.11	0.89	(1, 17.88) = 0.28	0.60	(2, 15.75) = 0.56	0.58
Peak hip flexion during swing	(2, 16.23) = 0.78	0.48	(1, 17.89) = 0.50	0.49	(2, 16.71) = 0.81	0.46
Hip range of motion during single limb support	(2, 9.11) = 1.41	0.29	(1, 7.72) = 2.91	0.13	(2, 10.99) = 0.48	0.63
Hip range of motion across gait cycle	(2, 17.84) = 0.58	0.57	(1, 17.31) = 0.01	0.92	(2, 16.15) = 0.17	0.85
Pelvic range of motion during single limb support	(2, 16.00) = 2.69	0.10	(1, 18.38) = 17.09	<0.01*	(2, 17.16) = 2.09	0.15

Table 6.21 Statistical breakdown of stepping down gait lead limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 9.92) = 0.38	0.69	(1, 18.89) = 0.76	0.40	(2, 11.01) = 2.86	0.10
Peak pelvic obliquity during swing	(2, 10.36) = 4.85	0.03*	(1, 6.37) = 4.15	0.09	(2, 11.28) = 1.84	0.20
Hip rotation range of motion during single limb support	(2, 12.39) = 0.32	0.73	(1, 12.32) = 0.01	0.99	(2, 13.65) = 0.80	0.47
Pelvic rotation range of motion during single limb support	(2, 12.76) = 7.83	0.01*	(1, 9.71) = 2.22	0.17	(2, 12.04) = 8.81	<0.01*
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 6.36) = 3.21	0.11	(1, 4.96) = 6.92	0.05*	(2, 9.43) = 0.08	0.92
Vertical GRF Fz2	(2, 7.76) = 0.95	0.43	(1, 5.79) = 1.29	0.30	(2, 10.55) = 0.41	0.67
Anterior-Posterior GRF Fy1	(2, 11.15) = 2.42	0.13	(1, 5.84) = 62.15	<0.01*	(2, 13.88) = 0.36	0.71
Anterior-Posterior GRF Fy2	(2, 9.55) = 6.82	0.01*	(1, 6.80) = 3.50	0.11	(2, 13.36) = 1.53	0.25
Load Rate	(2, 7.69) = 1.08	0.39	(1, 7.00) = 8.77	0.02*	(2, 11.90) = 0.15	0.86
Decay Rate	(2, 9.03) = 4.00	0.06	(1, 5.73) = 0.07	0.81	(2, 12.95) = 1.13	0.35
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 8.33) = 1.17	0.36	(1, 5.11) = 0.18	0.69	(2, 5.73) = 2.35	0.18
Peak ankle plantarflexor moment during stance	(2, 11.58) = 1.34	0.30	(1, 5.61) = 2.41	0.18	(2, 6.74) = 0.09	0.91
Peak knee flexor moment during loading response	(2, 9.39) = 0.26	0.78	(1, 7.10) = 0.10	0.77	(2, 8.68) = 0.41	0.68
Peak knee extensor moment during loading response	(2, 3.05) = 16.05	0.02*	(1, 1.76) = 104.25	0.01*	(2, 8.03) = 3.73	0.07
Peak knee flexor moment during mid stance	(2, 5.23) = 4.30	0.08	(1, 4.48) = 0.33	0.60	(2, 4.24) = 3.99	0.11
Peak knee flexor moment during late stance	(2, 8.63) = 0.68	0.53	(1, 5.85) = 2.32	0.18	(2, 6.14) = 0.32	0.74
Peak knee flexor moment during swing	(2, 4.30) = 2.03	0.24	(1, 4.03) = 7.98	0.05*	(2, 3.77) = 4.08	0.11
Peak hip extensor moment during early stance	(2, 5.14) = 0.49	0.64	(1, 7.29) = 0.07	0.80	(2, 10.43) = 0.02	0.98
Peak hip flexor moment during late stance	(2, 8.60) = 1.05	0.39	(1, 7.17) = 0.02	0.89	(2, 8.10) = 0.76	0.50
Peak hip extensor moment during swing	(2, 3.56) = 0.13	0.88	(1, 4.66) = 0.13	0.73	(2, 4.91) = 0.46	0.66

Table 6.22 Statistical breakdown of stepping down gait lead limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 10.47) = 10.45	<0.01*	(1, 10.09) = 6.31	0.03*	(2, 10.94) = 8.11	0.01*
Peak hip abductor during late stance	(2, 10.37) = 6.14	0.02*	(1, 5.01) = 0.06	0.82	(2, 5.94) = 8.26	0.02*
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 6.98) = 5.06	0.04*	(1, 4.59) = 1.78	0.25	(2, 5.91) = 3.50	0.10
Second peak support moment	(2, 11.41) = 12.80	<0.01*	(1, 11.41) = 2.58	0.14	(2, 11.41) = 5.19	0.03*
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 3.16) = 3.34	0.15	(1, 6.60) = 2.71	0.15	(2, 3.16) = 3.26	0.15
A2 – Ankle power generation during pre-swing	(2, 3.38) = 0.05	0.95	(1, 6.63) = 3.80	010	(2, 3.51) = 0.56	0.62
K1 – Knee power absorption during loading response	(2, 5.59) = 8.13	0.23	(1, 6.43) = 1.80	0.23	(2, 5.59) = 1.22	0.23
K2 – Knee power generation during mid-stance	(2, 0.03) = 0.64	0.82	(1, 0.10) = 1.41	0.82	(2, 3.02) = 0.72	0.56
K3 – Knee power absorption during pre-swing	(2, 6.12) = 0.17	0.85	(1, 6.93) = 1.40	0.28	(2, 9.48) = 0.15	0.86
K4 – Knee power absorption during terminal swing	(2, 8.82) = 0.21	0.81	(1, 6.10) = 13.01	0.01*	(2, 6.30) = 0.77	0.50
H1 – Hip Power generation during loading response	(2, 2.69) = 0.75	0.55	(1, 4.59) = 0.48	0.52	(2, 3.55) = 3.13	0.17
H2 – Hip power absorption during stance	(2, 10.30) = 3.42	0.07	(1, 6.93) = 0.87	0.38	(2, 7.42) = 1.83	0.23
H3 – Hip power generation during pre-swing	(2, 5.88) = 1.21	0.24	(1, 6.28) = 1.70	0.24	(2, 5.88) = 1.11	0.24
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 8.10) = 0.32	0.74	(1, 7.47) = 3.91	0.09	(2, 10.15) = 0.35	0.71
Heel	(2, 7.97) = 7.27	0.02*	(1, 2.11) = 3.88	0.18	(2, 9.58) = 8.85	0.01*

Table 6.23 Statistical breakdown of stepping down gait trail limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables						
Walking Velocity	(2, 7.09) = 7.71	0.02*	(1, 4.30) = 0.03	0.86	(2, 12.42) = 0.69	0.52
Stride Length	(2, 9.19) = 9.10	0.01*	(1, 6.79) = 0.03	0.87	(2, 14.11) = 0.88	0.44
Relative Stance Duration	(2, 11.31) = 7.03	0.01*	(1, 7.37) = 11.56	0.01*	(2, 14.42) = 0.31	0.74
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 5.62) = 0.21	0.81	(1, 5.41) = 0.95	0.37	(2, 11.77) = 1.37	0.29
Peak ankle dorsiflexion during stance	(2, 21.22) =1.62	0.22	(1, 13.76) = 25.32	<0.01*	(2, 16.69) = 1.21	0.32
Peak plantarflexion during swing	(2, 21.44) = 1.07	0.36	(1, 15.21) = 0.52	0.48	(2, 17.96) = 2.05	0.16
Ankle range of motion during stance	(2, 13.09) = 1.78	0.21	(1, 12.95) = 26.12	<0.01*	(2, 18.90) = 0.08	0.93
Peak knee flexion during loading response	(2, 11.20) = 2.43	0.13	(1, 10.52) = 1.33	0.27	(2, 14.51) = 0.80	0.47
Peak knee flexion during swing	(2, 18.44) = 1.80	0.19	(1, 14.96) = 0.96	0.34	(2, 17.76) = 1.05	0.37
Knee range of motion during loading response	(2, 9.22) = 5.32	0.03*	(1, 6.33) = 2.80	0.14	(2, 11.53) = 4.70	0.03*
Knee range of motion during single limb support	(2, 17.62) = 2.94	0.08	(1, 11.81) = 4.75	0.05*	(2, 16.55) = 3.03	0.08
Knee range of motion across gait cycle	(2, 18.46) = 1.22	0.32	(1, 13.70) = 2.17	0.16	(2, 16.86) = 0.90	0.42
Peak hip flexion during loading response	(2, 13.24) = 2.07	0.17	(1, 13.49) = 0.81	0.38	(2, 16.95) = 0.84	0.45
Peak hip extension during stance	(2, 8.63) = 0.40	0.68	(1, 5.69) = 0.52	0.50	(2, 12.55) = 0.30	0.75
Peak hip flexion during swing	(2, 12.34) = 1.60	0.24	(1, 12.07) = 1.09	0.32	(2, 15.81) = 0.38	0.69
Hip range of motion during single limb support	(2, 11.01) = 3.78	0.06	(1, 9.64) = 0.01	0.92	(2, 13.54) = 1.24	0.32
Hip range of motion across gait cycle	(2, 10.73) = 2.07	0.17	(1, 12.10) = 0.38	0.55	(2, 15.48) = 0.65	0.53
Pelvic range of motion during single limb support	(2, 9.66) = 3.90	0.06	(1, 7.15) = 0.81	0.40	(2, 12.69) = 1.58	0.24

Table 6.24 Statistical breakdown of stepping down gait trail limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 19.32) = 0.20	0.82	(1, 19.32) = 3.42	0.08	(2, 19.32) = 1.02	0.38
Peak pelvic obliquity during swing	(2, 15.29) = 5.21	0.02*	(1, 12.91) = 0.02	0.90	(2, 17.30) = 0.81	0.46
Hip rotation range of motion during single limb support	(2, 14.74) = 0.06	0.94	(1, 12.09) = 0.16	0.70	(2, 17.14) = 0.42	0.66
Pelvic rotation range of motion during single limb support	(2, 10.27) = 0.49	0.63	(1, 4.98) = 0.01	0.94	(2, 10.60) = 1.41	0.29
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 12.70) = 1.47	0.27	(1, 10.55) = 0.58	0.46	(2, 9.56) = 0.30	0.75
Vertical GRF Fz2	(2, 12.86) = 1.37	0.29	(1, 10.33) = 6.04	0.03*	(2, 9.38) = 0.20	0.83
Anterior-Posterior GRF Fy1	(2, 5.77) = 12.26	0.01*	(1, 5.00) = 2.48	0.18	(2, 7.94) = 8.77	0.01*
Anterior-Posterior GRF Fy2	(2, 4.37) = 1.48	0.32	(1, 6.98) = 11.37	0.01*	(2, 7.19) = 1.18	0.36
Load Rate	(2, 7.89) = 1.11	0.38	(1, 3.94) = 3.00	0.16	(2, 6.40) = 3.76	0.08
Decay Rate	(2, 8.75) = 0.08	0.93	(1, 6.48) = 17.02	0.01*	(2, 8.01) = 0.85	0.46
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 3.11) = 9.49	0.05*	(1, 5.38) = 0.02	0.89	(2, 4.33) = 2.35	0.20
Peak ankle plantarflexor moment during stance	(2, 4.75) = 1.90	0.25	(1, 6.35) = 0.79	0.41	(2, 7.79) = 0.01	0.99
Peak knee flexor moment during loading response	(2, 2.22) = 6.04	0.13	(1, 2.07) = 6.83	0.12	(2, 2.69) = 11.32	0.05*
Peak knee extensor moment during loading response	(2, 8.14) = 0.91	0.44	(1, 5.67) = 3.76	0.10	(2, 5.94) = 2.02	0.22
Peak knee flexor moment during mid stance	(2, 2.94) = 0.37	0.72	(1, 2.66) = 1.60	0.31	(2, 3.34) = 1.26	0.39
Peak knee flexor moment during late stance	(2, 4.17) = 1.90	0.26	(1, 5.56) = 12.41	0.01*	(2, 6.81) = 0.05	0.95
Peak knee flexor moment during swing	(2, 2.61) = 3.34	0.19	(1, 5.40) = 1.18	0.32	(2, 5.12) = 4.14	0.09
Peak hip extensor moment during early stance	(2, 5.98) = 76.44	0.64	(1, 4.95) = 0.25	0.64	(2, 6.83) = 0.20	0.83
Peak hip flexor moment during late stance	(2, 7.23) = 5.36	0.04*	(1, 5.29) = 13.36	0.01*	(2, 5.38) = 0.81	0.49
Peak hip extensor moment during swing	(2, 2.56) = 1.69	0.34	(1, 5.58) = 0.63	0.46	(2, 4.82) = 2.85	0.15

Table 6.25 Statistical breakdown of stepping down gait trail limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 3.15) = 17.14	0.02*	(1, 4.54) = 9.22	0.03*	(2, 5.54) = 7.50	0.03*
Peak hip abductor during late stance	(2, 2.43) = 38.00	0.02*	(1, 4.34) = 7.83	0.05*	(2, 2.16) = 9.47	0.09
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 2.37) = 4.93	0.14	(1, 4.99) = 0.01	0.94	(2, 4.35) = 0.54	0.62
Second peak support moment	(2, 2.58) = 18.09	0.03*	(1, 6.27) = 0.58	0.48	(2, 3.01) = 0.56	0.62
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 8.51) = 2.38	0.15	(1, 7.72) = 13.98	0.01*	(2, 12.91) = 0.34	0.72
A2 – Ankle power generation during pre-swing	(2, 3.67) = 0.97	0.46	(1, 5.26) = 7.53	0.04*	(2, 5.01) = 0.10	0.90
K1 – Knee power absorption during loading response	(2, 8.11) = 0.08	0.93	(1, 5.77) = 4.84	0.07	(2, 10.57) = 1.62	0.24
K2 – Knee power generation during mid-stance	(2, 3.03) = 5.56	0.10	(1, 5.32) = 6.44	0.05*	(2, 3.23) = 0.91	0.49
K3 – Knee power absorption during pre-swing	(2, 1.49) = 1.92	0.52	(1, 4.89) = 0.48	0.52	(2, 5.08) = 1.08	0.41
K4 – Knee power absorption during terminal swing	(2, 16.73) = 8.04	<0.01*	(1, 2.80) = 7.64	0.08	(2, 4.25) = 10.01	0.03*
H1 – Hip Power generation during loading response	(2, 3.53) = 0.94	0.47	(1, 5.43) = 1.56	0.26	(2, 6.09) = 0.73	0.52
H2 – Hip power absorption during stance	(2, 12.90) = 4.00	0.04*	(1, 6.82) = 3.54	0.10	(2, 8.39) = 0.35	0.71
H3 – Hip power generation during pre-swing	(2, 3.03) = 9.00	0.05*	(1, 4.18) = 3.48	0.13	(2, 4.14) = 4.97	0.08
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 8.31) = 2.60	0.13	(1, 7.24) = 0.04	0.85	(2, 11.63) = 3.14	0.08
Heel	(2, 15.40) = 0.83	0.46	(1, 12.00) = 0.06	0.81	(2, 11.24) = 0.10	0.91

6.3.13 Discussion – Stepping Down Gait

Literature has reported that amputees display inter-limb asymmetry in relative stance duration when negotiating stairs, with increased relative stance duration on the intact limb (Powers *et al.*, 1997; Vanicek *et al.*, 2007).When stepping down to a new level, walking velocity and stride length increased over time, with relative stance duration decreasing in the affected limb when acting as the trail limb. This suggested an increase in stepping down gait functioning and thus supporting the third (3) hypothesis of an improvement in the ability to perform stepping down gait over time.

An affected lead limb preference was observed initially, although this diminished over time at six months post-discharge. In the current participant group, this is unsurprising as during rehabilitation this strategy was advocated. However, the reduction of bias in this lead limb preference could reflect an underlying shift in stepping down gait ability.

Although not explicitly the same task, literature has reported two prevalent strategies when descending stairs (Jones *et al.*, 2006; Schmalz *et al.*, 2007; Alimusaj *et al.*, 2009). When leading with the affected limb, amputees tended to maintain an extended lead limb in an attempt to reduce the demands on the knee extensor musculature (Jones *et al.*, 2006; Schmalz *et al.*, 2007; Alimusaj *et al.*, 2009). When leading with the intact limb, ankle plantarflexion was increased due to a lack of dorsiflexion during stance phase in the affected trail limb (Schmalz *et al.*, 2007).

Characteristics of these strategies were present during stepping down gait. Firstly, when leading with the intact limb, increased lead ankle plantarflexion was observed coupled with lower ankle ROM and peak dorsiflexion during stance phase in the trailing affected limb. This lack of mobility in the prosthetic ankle necessitated participants to plantarflex the ankle of the leading intact limb in order to ‘fall’ onto the stance limb (Schmalz *et al.*, 2007). The intact limb knee joint also displayed both increased peak knee flexion and ROM during loading response along with increased load rate, peak vertical and posterior GRFs. As a result, peak knee extensor moment was increased in comparison to the affected limb and this also increased over time. Ankle and knee joint power bursts during stance phase (A1, A2, and K1-4) were also elevated. These results suggested that there were large demands placed on the knee extensor and ankle plantarflexor musculature in order to lower the body in a controlled fashion. This strategy was not adopted until later in the six month period post-discharge, perhaps due to the increased muscular demands. As previously stated, the adoption of this strategy may have signified an increase in stepping down gait ability over time. Participants lower limb knee extensor and ankle plantarflexor musculature may have become more accustomed to managing the strength requirements during stance phase in the intact limb when acting as the lead limb.

Participants in the current study initially tended to lead with the affected limb, the knee joint maintained in a more extended position, with reduced GRFs and subsequent joint moments. These results corroborated previous reports of this strategy in stair descent (Jones *et al.,* 2006; Schmalz *et al.,* 2007; Alimusaj *et al.*, 2009). However, walking velocity also increased when using this strategy and given the apparent reduced functioning of the affected limb, gait adaptations could be hypothesised to be a result of intact limb function. This hypothesis received support in the form of increased decay rate, second peak vertical and anterior (propulsive) GRF in the intact limb when acting as the trail limb in preparation for swing phase. In addition, peak ankle plantarflexor moment, hip flexor moment and ankle (A2) and hip (H3) power generation during late stance phase were increased when compared to the affected limb and increased in the six months following discharge. This suggested an increase in the propulsive mechanism of the intact limb when acting as a trail limb. One interpretation of these results was that amputees were comfortable propelling the intact lead limb forwards, while in stance on a relatively ‘rigid’ affected trail limb. Upon lead limb contact, the intact limb may have been more able to cope with the increased load as the whole body centre of mass (COM) is lowered when compared to the affected limb.

The reduced kinetic functioning of the affected limb during stepping down gait has implications for transtibial amputee rehabilitation. Similar to level gait, attempts during rehabilitation to increase the eccentric strength of knee extensor musculature may increases affected limb ability to lower the whole body COM prior to intact limb foot contact.

The improvements in walking velocity, coupled with the adaptations present in a number of biomechanical variables, supported the third hypothesis (3) of an improvement in the ability to step from a new level during gait.

6.3.14 Joint Kinematics – Stepping Up Gait

Joint kinematics are presented in Figures 6.23 (sagittal plane lead limb), Appendix N (frontal and transverse plane lead limb), 6.24 (sagittal plane trail limb) and Appendix N (frontal and transverse) with complete statistical analyses provided in Tables 6.27, 6.28, 6.30 and 6.31.

6.3.14.1 Lead Limb

Ankle ROM during stance phase was significantly greater when leading with the intact limb in comparison to leading with the affected limb ($p=0.02$) (Figure 6.23).

Peak knee flexion during loading response ($p<0.01$), knee ROM during single limb support ($p=0.01$) and peak knee flexion during swing phase ($p<0.01$) were all significantly greater when leading with the intact limb in comparison to the affected limb, with these effects being reflected in Figure 6.23.

Sagittal plane hip joint ROM during single limb support was significantly greater when leading with the intact limb ($p=0.04$). Although there were small increase and decreases in the magnitude of sagittal plane hip kinematic profiles when leading with the affected and intact limbs respectively, no significant time main effect was observed (Figure 6.23). Similarly, there were no reported significant main effects relating to sagittal plane pelvic kinematic variables (Figure 6.23).

Pelvic rotation ROM during single limb support increased significantly over time, although post-hoc analysis did not reveal where the differences occurred (p=0.04).

6.3.14.2 Trail Limb

Sagittal plane ankle kinematics were relatively consistent over time when trailing with the affected limb when compared to the intact limb (Figure 6.24). However, peak plantarflexion during swing phase was greater when trailing with the intact limb in comparison to trailing with the affected limb (p=0.01).

Knee joint ROM during loading response (p=0.03) and single limb support (p=0.02) were greater when trailing with the intact limb as opposed to trailing with the affected limb (Figure 6.24). Despite an observable decrease in the magnitude of hip extension when trailing with the intact limb and an increase in hip ROM when trailing with the affected limb over time, there were no statistically significant main effects reported (Figure 6.24).

There were no reported significant main effects relating to sagittal plane pelvic kinematic variables, despite some observable changes across time (Figure 6.24).

Peak hip abduction during swing phase was significantly greater when trailing with the affected limb when compared to the intact limb (p=0.05). A significant time main effect was observed for pelvic rotation ROM during single limb support (p=0.05), perhaps due to the reduction visible when trailing with the affected limb, however, post-hoc analysis did not reveal where the differences occurred (Appendix N).

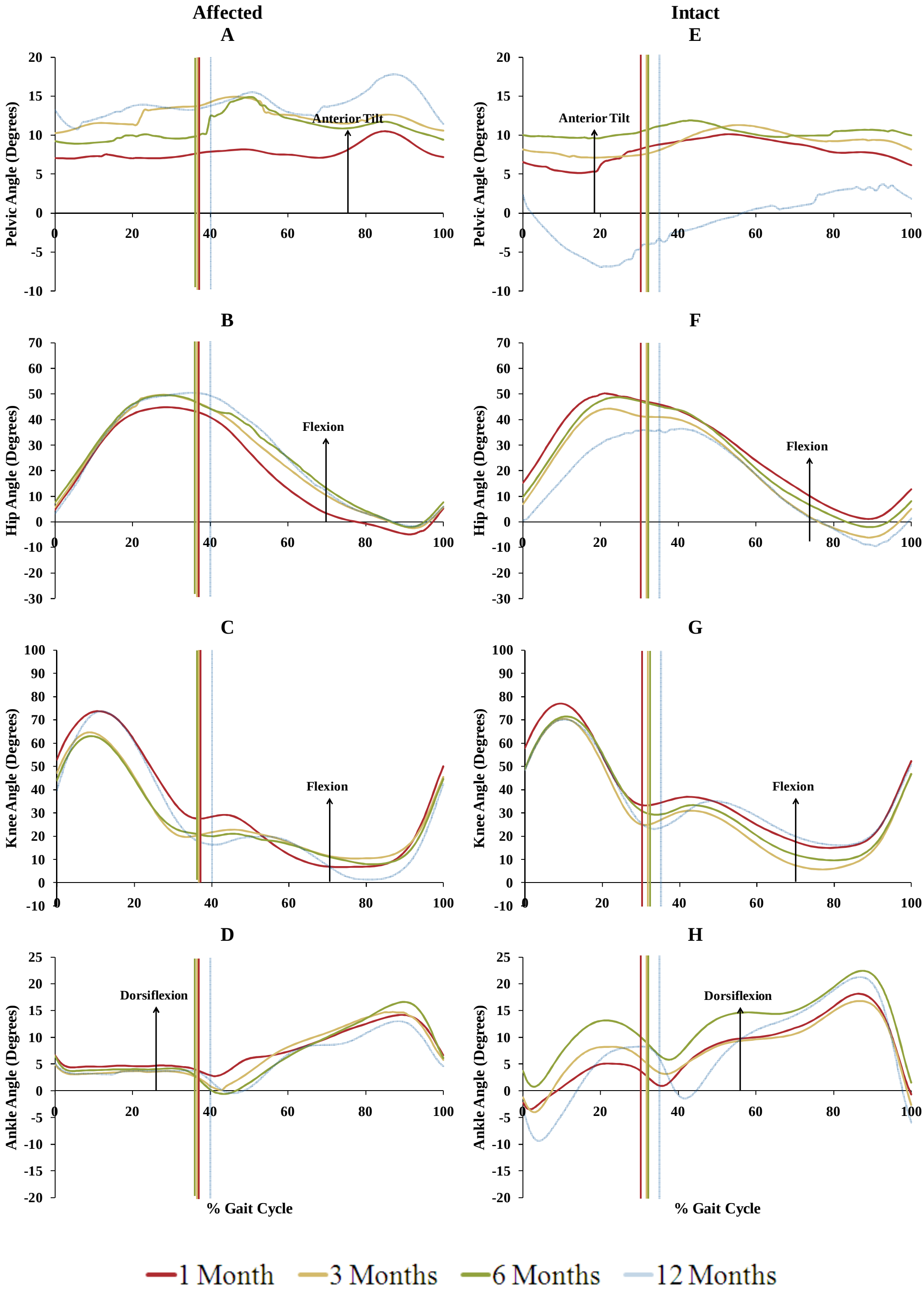


Figure 6.23 Group mean sagittal plane kinematics of the lead affected limb pelvis (A), hip (B), knee (C) and ankle (D) and lead intact limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during stepping up gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

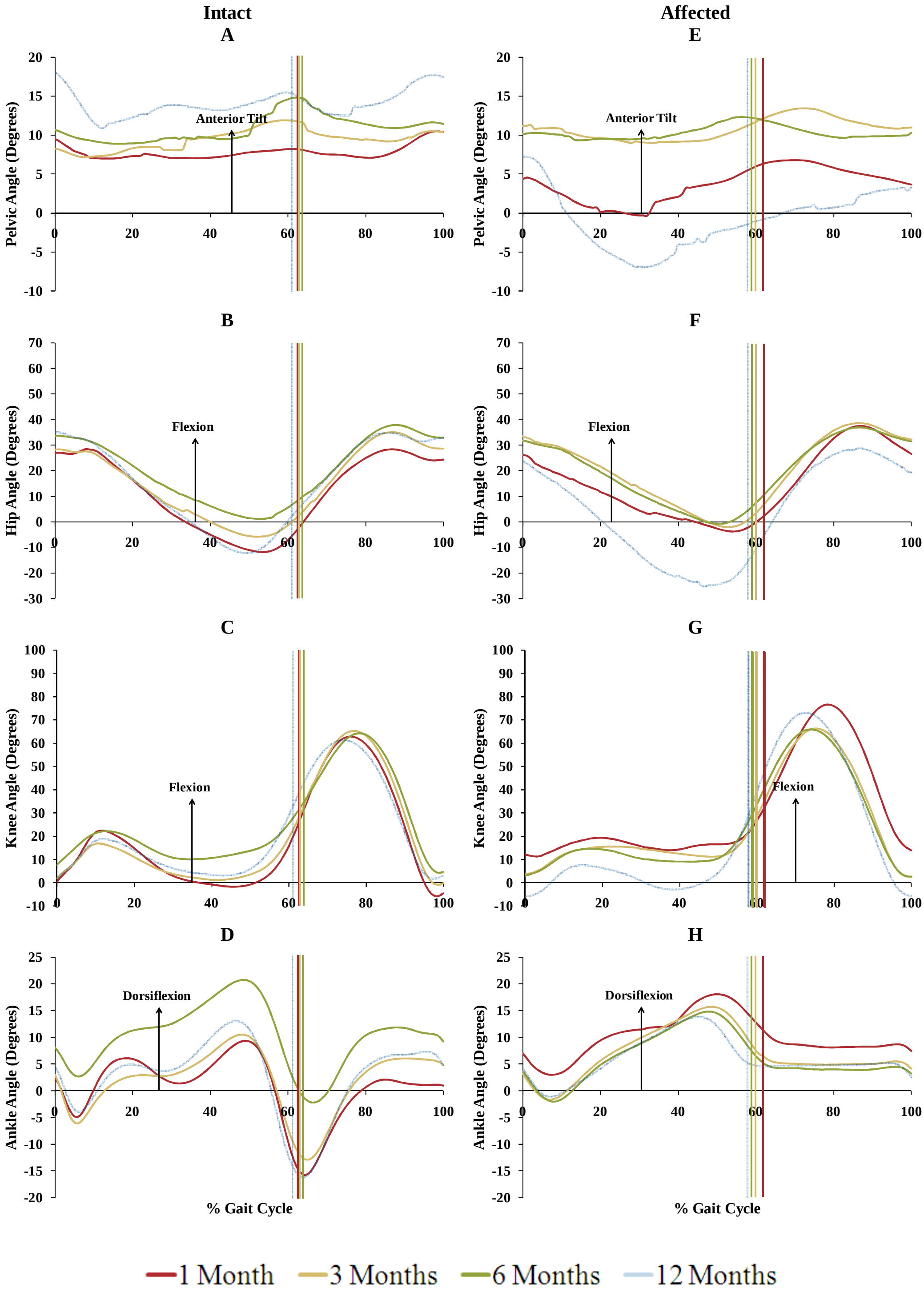


Figure 6.24 Group mean sagittal plane kinematics of the trail intact limb pelvis (A), hip (B), knee (C) and ankle (D) and trail affected limb pelvis (E), hip (F), knee (G) and ankle (H). Time normalised to 100% of gait cycle during stepping up gait. Vertical lines represent toe off. Data at 12 months from n=2.

6.3.15 Ground Reaction Forces – Stepping Up Gait

Loading and decay rates are presented in Table 6.26, GRF data are presented in Figure 6.25 with complete statistical analyses provided in Tables 6.28 and 6.31.

6.3.15.1 Lead Limb

Both loading and decay rates were observably higher when leading and trailing with the intact limb when compared to the affected limb, however no significant main effects were reported (Table 6.26).

Initial peak vertical GRF observably increased over time and was greater when leading with the intact limb in comparison to the affected limb although this did not result in any significant main effects (Figure 6.25). Peak posterior GRF was significantly greater when leading with the intact limb than when leading with the affected limb (p=0.01) (Figure 6.25). A significant interaction effect occurred (p=0.04) as a result of the combination of increasing second peak vertical GRF when leading with the intact limb, with increased magnitude observed when leading with the affected limb (Figure 6.25).

6.3.15.2 Trail Limb

Analysis of load rate produced a significant interaction effect, with load rate being increased when trailing with the intact limb at one and three months post-discharge, with more similar loading rates observed at six months (p=0.03) (Table 6.26). In addition, load rate increased over time when trailing with the affected limb. Decay rate increased significantly over time when trailing with the affected limb (p=0.05) although post-hoc analysis did not reveal where the differences occurred.

Second peak vertical GRF was significantly higher when trailing with the intact limb (p=0.03), although there was no statistically significant longitudinal change in either limb (Figure 6.25). A similar effect was noted in peak posterior GRF coupled with relatively larger changes over time, resulting in a significant interaction effect (p=0.05) (Figure 6.25).

Table 6.26 Mean (SD) loading and decay rate of stepping up gait. Data are presented for the affected, intact, lead and trail limbs separately.

	Limb	One Month	Three Months	Six Months	Twelve Months*
Load Rate (BW/s)	Lead Affected	3.9 (0.0)	4.4 (0.4)	4.0 (1.7)	5.2 (0.0)
	Lead Intact	5.0 (1.9)	6.0 (1.6)	6.3 (1.8)	6.6 (0.0)
	Trail Intact	8.0 (0.0)	8.1 (1.7)	6.8 (1.8)	9.8 (0.0)
	Affected Trail	4.6 (1.3)	4.7 (0.9)	5.3 (1.5)	8.0 (0.0)
Decay Rate (BW/s)	Lead Affected	3.6 (0.0)	4.9 (0.9)	5.1 (0.7)	7.3 (0.0)
	Lead Intact	4.2 (0.7)	4.7 (0.7)	4.7 (1.6)	6.7 (0.0)
	Trail Intact	8.6 (0.0)	7.1 (2.7)	5.8 (1.8)	10.5 (0.0)
	Affected Trail	5.1 (0.9)	5.3 (0.8)	6.1 (1.8)	6.4 (0.0)

*Data from two participants, not included in statistical analyses

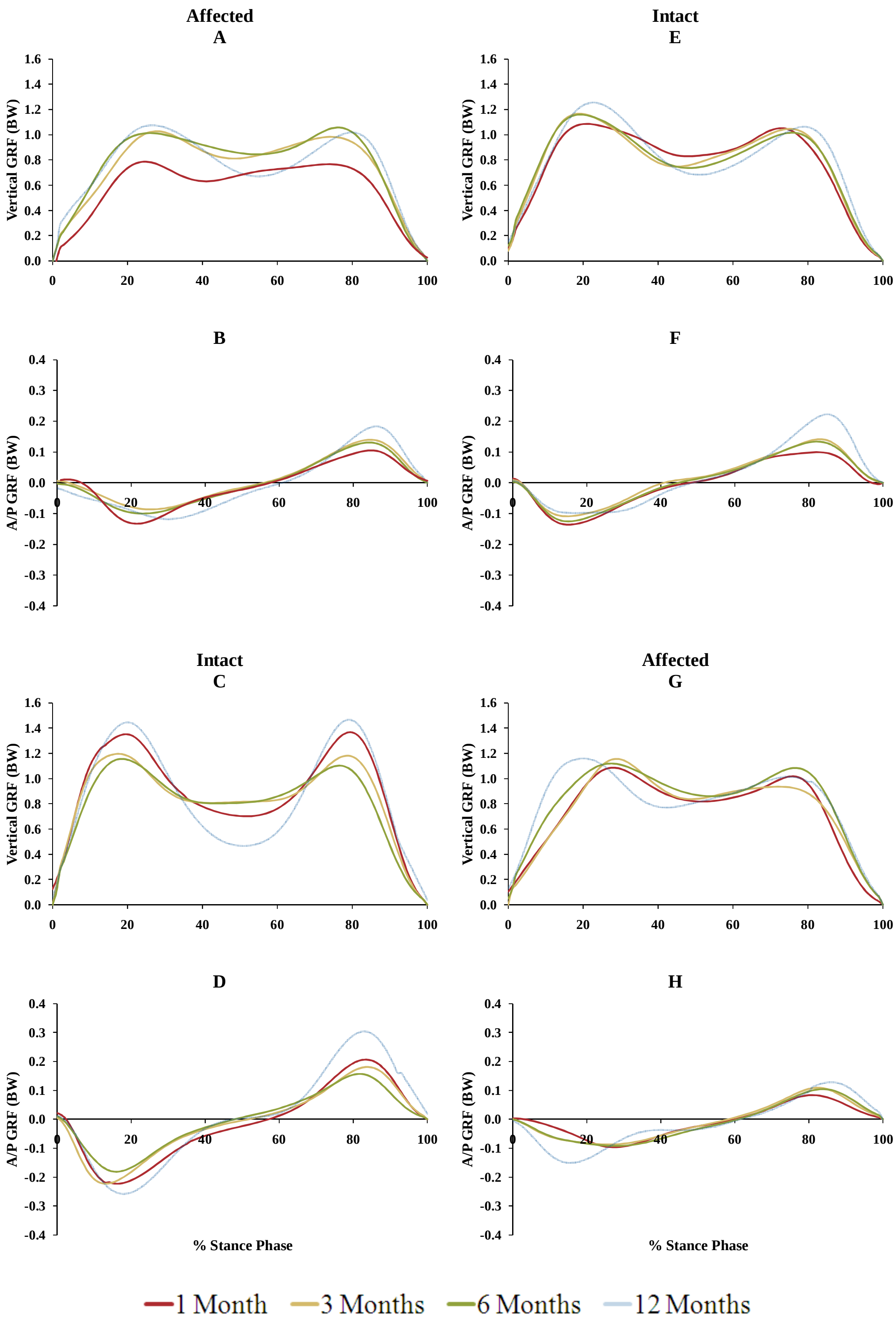


Figure 6.25 Group mean vertical and anterior-posterior ground reaction forces for the lead affected (A and B), lead intact (E and F), trail intact (C and D) and trail affected (G and H) limbs. All data normalised and presented as times body weight (BW). Time normalised to 100% of stance phase during stepping up gait. Data at 12 months from n=2. Vertical and anterior are positive.

6.3.16 Joint Kinetics – Stepping Up Gait

Joint moments are presented in Figures 6.26 (sagittal and frontal plane lead limb) 6.27 (sagittal and frontal plane trail limb), joint powers are presented in Figures 6.28 (sagittal plane lead limb), 6.29 (sagittal plane trail limb) and support moments are presented in Appendix O with complete statistical analyses provided in Tables 6.28, 6.29, 6.31 and 6.32.

6.3.16.1 Lead Limb

Knee extensor moment during loading response ($p<0.01$) was greater when leading with the intact limb with peak knee flexor moment during late stance phase ($p=0.03$) being increased when leading with the affected limb (Figure 6.26).

Peak hip flexor moment during late stance phase was significantly greater when leading with the intact limb in comparison to the affected limb ($p=0.01$) and increased between one and three ($p=0.01$) months and one and six ($p=0.03$) months post-discharge in both limbs.

Peak hip abductor moment during early stance phase was greater when leading with the intact limb in comparison to the affected limb ($p<0.01$). A similar effect was noted for initial peak support moment, which was also greater when leading with the intact limb ($p<0.01$) (Appendix O).

Peak power generation at the ankle joint (A2) was significantly greater when leading with the intact limb when compared to leading with the affected limb ($p=0.02$).

The magnitude of peak knee power generation during stance phase (K2) ($p<0.01$) and peak knee power absorption during swing phase (K4) ($p<0.01$) were greater when leading with the intact limb when compared to the affected limb (Figure 6.28). Peak power absorption during late stance phase in (K3) increased gradually over time when leading with the intact limb (Figure 6.28). In addition, power burst K3 was generally increased when leading with the intact limb in comparison to the affected limb, resulting in a significant interaction effect (p=0.01).

6.3.16.2 Trail Limb

Despite some observable changes in ankle moment profiles no statistically significant main effects were observed (Figure 6.27).

Peak knee extensor moment during loading response was significantly higher when trailing with the intact limb in comparison to the affected limb (p=0.01). A similar difference was noted for knee flexor moments during late stance phase, although no statistical limb effect was reported (Figure 6.27). Knee moment during mid-stance increased over time when trailing with the intact limb and remained relatively consistent when trailing with the affected limb, resulting in a significant interaction effect (p=0.01).

Peak hip flexor moment during stance phase was consistent over time in both limbs therefore no statistically significant main effects were observed (Figure 6.27).

Peak hip abductor moment during both early (p<0.01) and late (p=0.03) stance phase were increased when trailing with the intact limb (Figure 6.27).

Peak power absorption at the ankle (A1) was greater when trailing with the intact limb when compared to the affected limb, although this was not statistically significant (Figure 6.29). Similarly, peak power generation at power burst A2 was greater when trailing with the intact limb in comparison the affected limb (p=0.02).

The magnitude of peak knee power absorption during early stance phase (K1) (p=0.05) and peak knee power generation during stance phase (K2) (p=0.01) were significantly greater when trailing with the intact limb in comparison to the affected limb (Figure 6.29).

Peak knee power absorption during late stance phase (K3) did not change noticeably over time when trailing with the affected limb. However, an increase followed by a decrease was noted in the same variable when trailing with the intact limb, resulting in a significant time main effect between three and six months post-discharge (p=0.03).

Although there were some observable changes over time in hip power profiles, no significant time or limb main effects were reported (Figure 6.29).

6.3.16.3 Data for n=2 at 12 Months Post-Discharge

Participants maintained the same lead limb preference observed at 12 months post-discharge as was selected at 6 months post-discharge. Walking velocity and stride length increased from 6 to 12 months post-discharge when leading with both limbs. Relative stance duration decreased regardless of limb or role. Vertical heel and toe trajectories did not alter greatly at 12 months post-discharge in either limb when performing either the lead or trail role.

When leading with both the intact and affected limbs at 12 months post-discharge, sagittal and frontal plane joint kinematics remained within the ranges observed between one and six months post-discharge. A similar pattern was reported when trailing with the intact limb although there were some observable increases in knee ROM during loading response and peak hip extension during stance phase when trailing with the affected limb.

Both load and decay rates were increased when leading with the affected limb at 12 months post-discharge. Similarly, all peak vertical and anterior-posterior GRFs, with the exception of the second peak vertical GRF when leading with the affected limb, were increased at 12 months post-discharge.

Similar to the changes noted in stepping down gait, peak knee extensor and ankle plantarflexor moment during stance phase increased at 12 months post-discharge when leading with the affected limb.

This was reflected in a similar way in joint powers observed, with power bursts A2, K1, K2 and K3 all increasing at 12 months post-discharge. An identical pattern of increases were observed when trailing with the intact limb with the addition of increases in power bursts A1, H2 and H3. Again, the noted increased kinetic functioning of the intact limb highlights the importance of this limb in the successful completion of ADLs living in transtibial amputees.

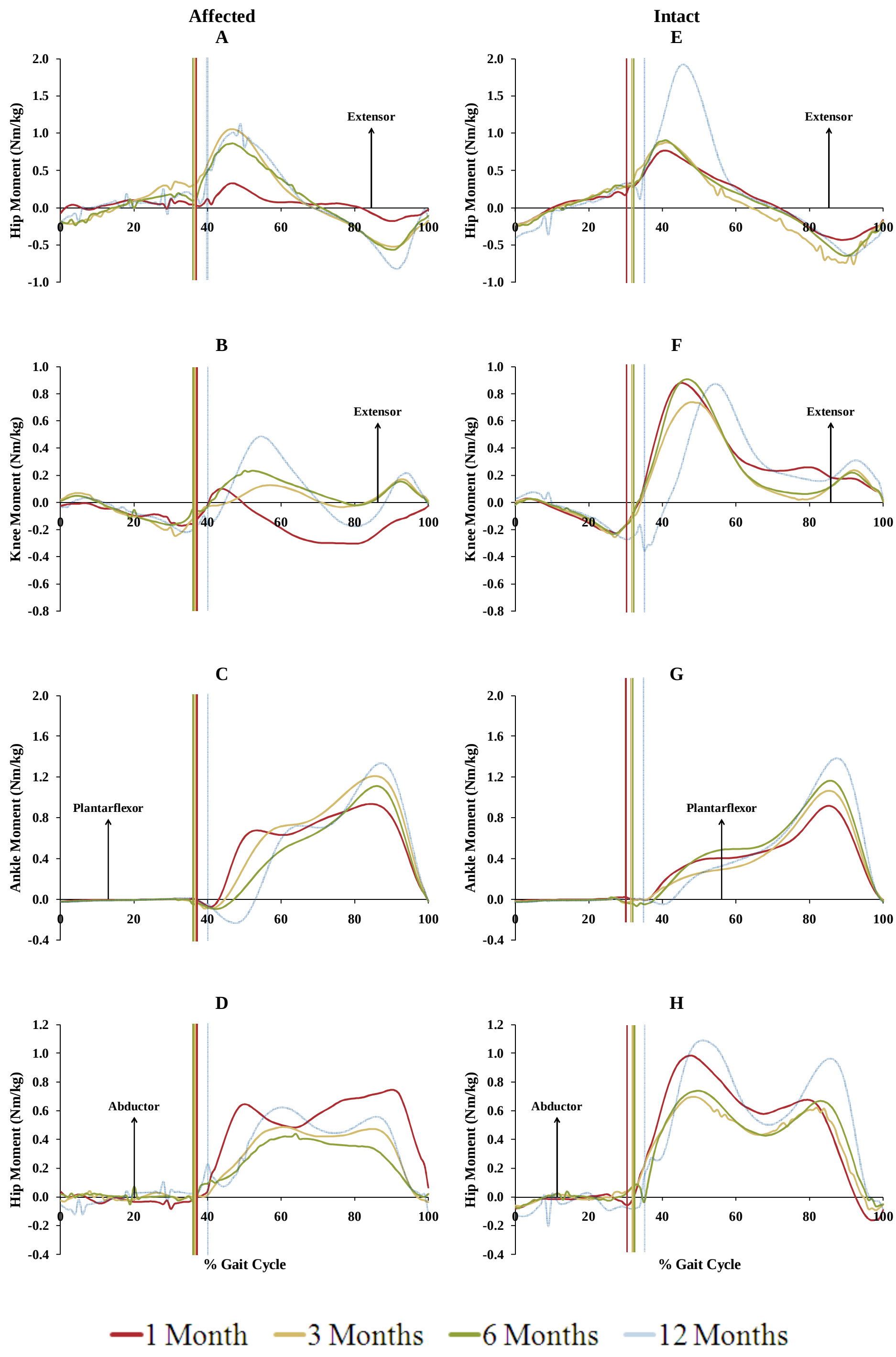


Figure 6.26 Group mean joint moments for the lead affected limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and lead intact limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during stepping up gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

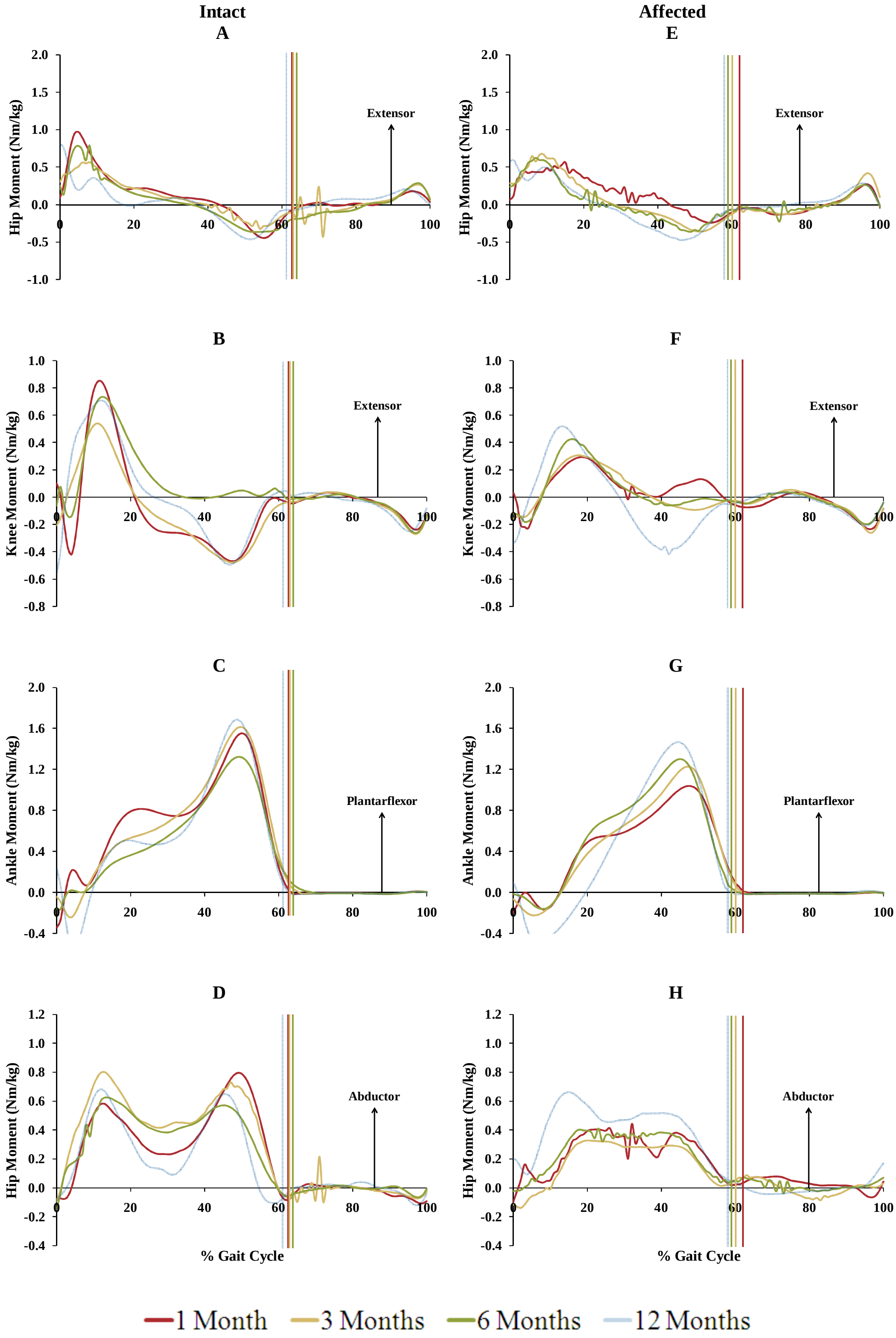


Figure 6.27 Group mean joint moments for the trail intact limb hip (A), knee (B), ankle (C) (sagittal plane) and hip (D) (frontal plane) and trail affected limb hip (E), knee (F), ankle (G) (sagittal plane) and hip (H) (frontal plane). Time normalised to 100% of gait cycle during stepping up gait. Vertical lines represent toe off. Data at 12 months from n=2.

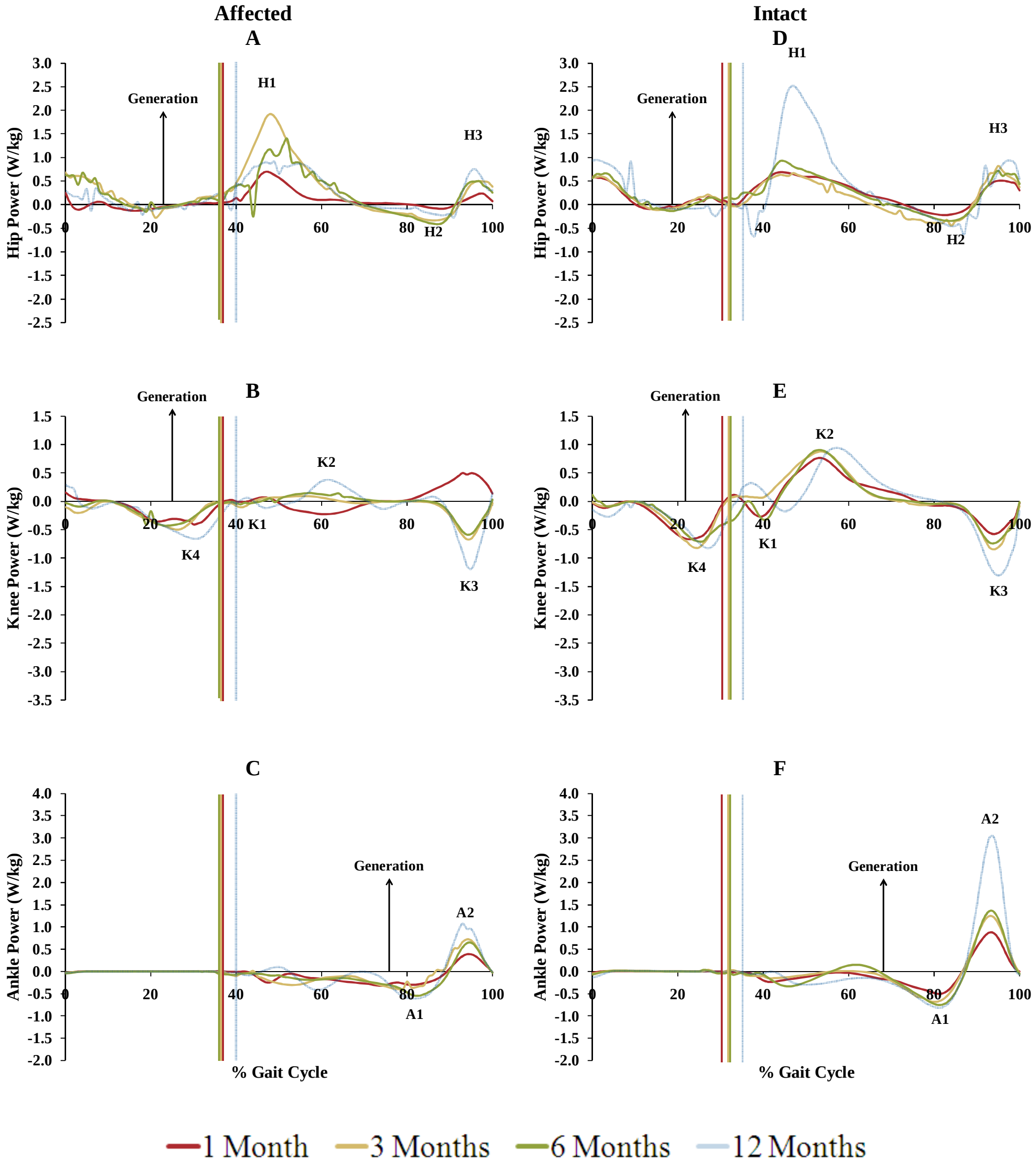


Figure 6.28 Group mean joint powers for the lead affected limb hip (A), knee (B), ankle (C) and lead intact limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during stepping up gait. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

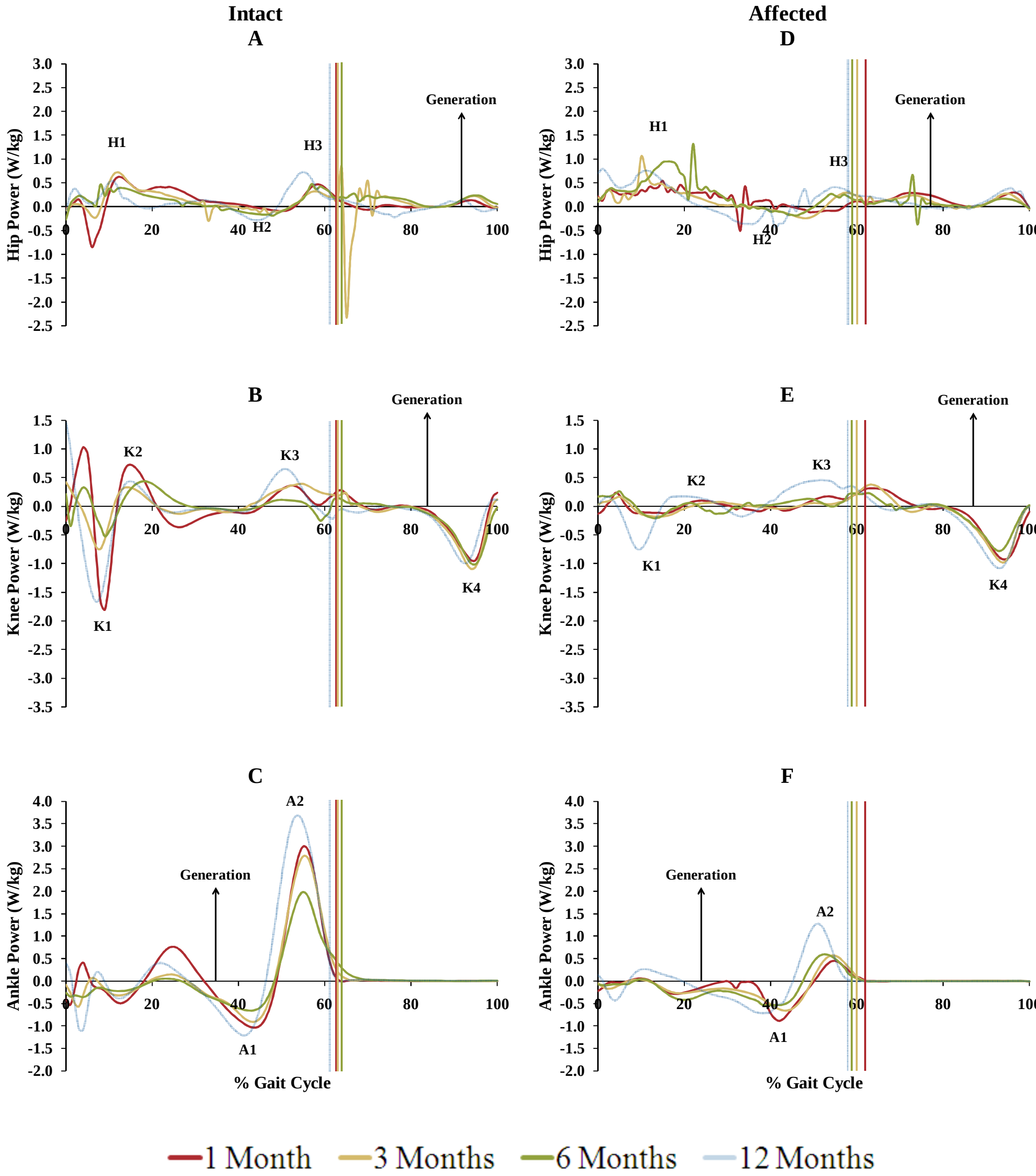


Figure 6.29 Group mean joint powers for the trail intact limb hip (A), knee (B), ankle (C) and trail affected limb hip (D), knee (E), ankle (F). Time normalised to 100% of gait cycle during stepping up gait. Vertical lines represent toe off. Data at 12 months from n=2.

Table 6.27 Statistical breakdown of stepping up gait lead limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 16.85) = 2.19	0.14	(1, 18.56) = 0.45	0.51	(2, 15.48) = 0.09	0.91
Stride Length	(2, 7.31) = 2.43	0.16	(1, 13.15) = 2.16	0.17	(2, 12.79) = 0.84	0.46
Relative Stance Duration	(2, 9.57) = 0.74	0.50	(1, 7.02) = 10.36	0.02*	(2, 10.75) = 0.41	0.67
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 9.24) = 0.09	0.92	(1, 10.28) = 0.42	0.53	(2, 14.16) = 1.03	0.38
Peak ankle dorsiflexion during stance	(2, 18.19) = 0.82	0.46	(1, 20.80) = 2.24	0.15	(2, 16.39) = 0.21	0.81
Peak plantarflexion during swing	(2, 8.59) = 0.48	0.63	(1, 6.52) = 5.47	0.06	(2, 10.44) = 1.70	0.23
Ankle range of motion during stance	(2, 12.94) = 1.16	0.35	(1, 13.12) = 7.36	0.02*	(2, 14.92) = 0.16	0.86
Peak knee flexion during loading response	(2, 10.77) = 0.07	0.93	(1, 11.77) = 14.69	<0.01*	(2, 12.44) = 0.33	0.73
Peak knee flexion during swing	(2, 8.50) = 0.12	0.89	(1, 7.17) = 6.10	0.04*	(2, 9.88) = 0.22	0.81
Knee range of motion during loading response	(2, 13.71) = 0.98	0.40	(1, 12.15) = 0.06	0.82	(2, 15.54) = 1.50	0.26
Knee range of motion during single limb support	(2, 8.77) = 1.17	0.36	(1, 6.21) = 14.79	0.01*	(2, 9.59) = 0.57	0.58
Knee range of motion across gait cycle	(2, 9.71) = 0.44	0.66	(1, 7.56) = 4.26	0.08	(2, 10.73) = 0.83	0.46
Peak hip flexion during loading response	(2, 16.42) = 0.17	0.84	(1, 20.00) = 0.58	0.45	(2, 14.67) = 0.47	0.63
Peak hip extension during stance	(2, 15.76) = 0.19	0.83	(1, 20.00) = 0.05	0.83	(2, 14.92) = 0.58	0.57
Peak hip flexion during swing	(2, 5.56) = 0.11	0.90	(1, 19.25) = 0.06	0.81	(2, 14.41) = 0.61	0.56
Hip range of motion during single limb support	(2, 11.97) = 2.33	0.14	(1, 16.04) = 15.12	<0.01*	(2, 14.32) = 2.19	0.15
Hip range of motion across gait cycle	(2, 8.77) = 2.65	0.13	(1, 6.02) = 0.48	0.51	(2, 9.10) = 2.61	0.13
Pelvic range of motion during single limb support	(2, 12.54) = 0.12	0.88	(1, 18.39) = 0.10	0.76	(2, 16.05) = 0.31	0.74

Table 6.28 Statistical breakdown of stepping up gait lead limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 19.45) = 0.02	0.98	(1, 17.37) = 3.97	0.06	(2, 19.46) = 1.95	0.17
Peak pelvic obliquity during swing	(2, 15.92) = 0.11	0.90	(1, 20.00) = 0.09	0.77	(2, 13.97) = 0.51	0.61
Hip rotation range of motion during single limb support	(2, 9.65) = 0.30	0.75	(1, 8.36) = 0.24	0.64	(2, 11.19) = 0.03	0.97
Pelvic rotation range of motion during single limb support	(2, 7.69) = 5.20	0.04*	(1, 14.24) = 1.71	0.21	(2, 12.61) = 0.55	0.59
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 9.34) = 0.21	0.82	(1, 6.46) = 5.13	0.06	(2, 9.20) = 2.00	0.19
Vertical GRF Fz2	(2, 8.08) = 0.48	0.64	(1, 5.30) = 0.20	0.67	(2, 8.42) = 5.68	0.04*
Anterior-Posterior GRF Fy1	(2, 9.00) = 1.01	0.40	(1, 8.08) = 10.04	0.01*	(2, 10.97) = 0.01	0.99
Anterior-Posterior GRF Fy2	(2, 7.01) = 1.62	0.26	(1, 3.68) = 2.52	0.19	(2, 9.40) = 0.64	0.45
Load Rate	(2, 5.60) = 0.04	0.96	(1, 1.17) = 11.50	0.15	(2, 10.76) = 0.01	0.91
Decay Rate	(2, 6.77) = 0.46	0.65	(1, 6.80) = 0.01	0.98	(2, 7.92) = 1.80	0.22
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 16.53) = 1.63	0.23	(1, 8.15) = 3.23	0.11	(2, 18.79) = 0.30	0.75
Peak ankle plantarflexor moment during stance	(2, 12.51) = 1.93	0.19	(1, 8.84) = 0.51	0.49	(2, 13.28) = 1.51	0.26
Peak knee flexor moment during loading response	(2, 17.66) = 0.09	0.91	(1, 10.48) = 2.78	0.13	(2, 17.93) = 0.44	0.65
Peak knee extensor moment during loading response	(2, 10.39) = 3.87	0.06	(1, 8.15) = 39.83	<0.01*	(2, 11.09) = 1.92	0.19
Peak knee flexor moment during mid stance	(2, 11.83) = 0.49	0.62	(1, 11.05) = 3.70	0.08	(2, 14.07) = 1.67	0.22
Peak knee flexor moment during late stance	(2, 11.26) = 1.70	0.23	(1 7.75) = 7.16	0.03*	(2, 11.98) = 3.26	0.07
Peak knee flexor moment during swing	(2, 19.42) = 1.52	0.24	(1, 15.84) = 3.83	0.07	(2, 19.36) = 0.28	0.76
Peak hip extensor moment during early stance	(2, 9.51) = 1.79	0.22	(1, 18.24) = 0.99	0.33	(2, 15.65) = 0.25	0.78
Peak hip flexor moment during late stance	(2, 10.07) = 7.08	0.01*	(1, 8.87) = 10.46	0.01*	(2, 11.24) = 3.69	0.06
Peak hip extensor moment during swing	(2, 11.37) = 1.19	0.34	(1, 10.96) = 0.87	0.37	(2, 14.11) = 0.21	0.82

Table 6.29 Statistical breakdown of stepping up gait lead limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 22.25) = 1.12	0.34	(1, 22.25) = 11.45	<0.01*	(2, 22.25) = 0.04	0.96
Peak hip abductor during late stance	(2, 17.49) = 0.80	0.47	(1, 19.97) = 2.38	0.14	(2, 16.40) = 0.77	0.48
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 9.73) = 2.04	0.18	(1, 12.61) = 12.08	<0.01*	(2, 11.51) = 1.53	0.26
Second peak support moment	(2, 10.89) = 0.30	0.75	(1, 15.51) = 0.53	0.48	(2, 14.91) = 0.38	0.69
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 13.04) = 1.94	0.18	(1, 7.81) = 2.63	0.14	(2, 13.56) = 0.10	0.91
A2 – Ankle power generation during pre-swing	(2, 9.71) = 1.92	0.20	(1, 6.33) = 8.70	0.02*	(2, 10.08) = 2.70	0.12
K1 – Knee power absorption during loading response	(2, 7.80) = 2.36	0.16	(1, 6.12) = 0.39	0.56	(2, 8.54) = 0.72	0.51
K2 – Knee power generation during mid-stance	(2, 10.09) = 0.34	0.72	(1, 8.21) = 28.31	<0.01*	(2, 11.06) = 0.31	0.74
K3 – Knee power absorption during pre-swing	(2, 8.28) = 9.07	0.01*	(1, 6.46) = 5.48	0.06	(2, 8.912) = 8.03	0.01*
K4 – Knee power absorption during terminal swing	(2, 12.31) = 0.32	0.73	(1, 19.77) = 12.31	<0.01*	(2, 16.37) = 0.06	0.94
H1 – Hip Power generation during loading response	(2, 19.34) = 3.52	0.05	(1, 15.61) = 2.08	0.17	(2, 18.59) = 2.36	0.12
H2 – Hip power absorption during stance	(2, 6.87) = 1.64	0.26	(1, 9.45) = 0.05	0.82	(2, 13.30) = 2.31	0.14
H3 – Hip power generation during pre-swing	(2, 10.72) = 2.55	0.12	(1, 12.29) = 4.45	0.06	(2, 14.00) = 0.06	0.94
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 11.86) = 0.74	0.50	(1, 12.29) = 2.78	0.12	(2, 14.09) = 0.78	0.48
Heel	(2, 15.90) = 0.27	0.77	(1, 11.52) = 1.70	0.22	(2, 15.99) = 0.58	0.57

Table 6.30 Statistical breakdown of stepping up gait trail limb temporal-spatial variables and joint kinematics. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Temporal-spatial Variables	F	P	F	P	F	P
Walking Velocity	(2, 16.85) = 2.19	0.14	(1, 18.56) = 0.45	0.51	(2, 15.48) = 0.09	0.91
Stride Length	(2, 7.31) = 2.43	0.16	(1, 13.15) = 2.16	0.17	(2, 12.79) = 0.84	0.46
Relative Stance Duration	(2, 8.99) = 1.01	0.40	(1, 6.37) = 5.96	0.05*	(2, 9.44) = 0.50	0.62
Sagittal Plane Joint Kinematics	F	P	F	P	F	P
Peak ankle plantarflexion during loading response	(2, 8.63) = 1.48	0.28	(1, 6.03) = 0.06	0.82	(2, 8.97) = 1.63	0.25
Peak ankle dorsiflexion during stance	(2, 9.07) = 2.54	0.13	(1, 19.48) = 0.30	0.59	(2, 15.07) = 3.24	0.07
Peak plantarflexion during swing	(2, 7.36) = 1.60	0.27	(1, 6.55) = 11.65	0.01*	(2, 8.18) = 1.25	0.34
Ankle range of motion during stance	(2, 11.48) = 1.64	0.24	(1, 11.63) = 2.34	0.15	(2, 13.89) = 0.88	0.44
Peak knee flexion during loading response	(2, 10.34) = 1.28	0.32	(1, 13.35) = 0.70	0.42	(2, 11.78) = 2.16	0.16
Peak knee flexion during swing	(2, 7.52) = 0.21	0.81	(1, 7.09) = 3.19	0.12	(2, 8.02) = 3.86	0.07
Knee range of motion during loading response	(2, 8.65) = 0.32	0.74	(1, 7.81) = 7.61	0.03*	(2, 9.42) = 0.80	0.48
Knee range of motion during single limb support	(2, 7.43) = 1.34	0.32	(1, 6.90) = 8.46	0.02*	(2, 7.88) = 1.66	0.25
Knee range of motion across gait cycle	(2, 6.60) = 0.41	0.68	(1, 12.80) = 0.22	0.65	(2, 9.04) = 0.16	0.85
Peak hip flexion during loading response	(2, 9.03) = 1.51	0.27	(1, 14.85) = 0.18	0.68	(2, 11.73) = 0.82	0.47
Peak hip extension during stance	(2, 7.46) = 2.29	0.17	(1, 8.33) = 0.51	0.50	(2, 8.51) = 0.82	0.47
Peak hip flexion during swing	(2, 7.56) = 1.70	0.25	(1, 11.50) = 2.11	0.17	(2, 8.77) = 1.16	0.36
Hip range of motion during single limb support	(2, 9.88) = 1.08	0.38	(1, 10.81) = 0.39	0.55	(2, 10.77) = 1.12	0.36
Hip range of motion across gait cycle	(2, 8.70) = 3.16	0.09	(1, 9.65) = 1.54	0.24	(2, 9.70) = 0.34	0.72
Pelvic range of motion during single limb support	(2, 17.10) = 0.89	0.43	(1, 18.93) = 0.01	0.98	(2, 17.43) = 1.07	0.36

Table 6.31 Statistical breakdown of stepping up gait trail limb joint kinematics, ground reaction forces and joint moments. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal and Transverse Plane Joint Kinematics	F	P	F	P	F	P
Peak hip abduction during swing	(2, 9.48) = 0.54	0.60	(1, 18.15) = 4.60	0.05*	(2, 17.50) = 1.19	0.33
Peak pelvic obliquity during swing	(2, 9.61) = 0.98	0.41	(1, 18.44) = 0.87	0.36	(2, 15.97) = 2.03	0.16
Hip rotation range of motion during single limb support	(2, 16.07) = 1.85	0.19	(1, 18.91) = 0.20	0.66	(2, 14.84) = 1.97	0.18
Pelvic rotation range of motion during single limb support	(2, 16.96) = 3.75	0.05*	(1, 18.73) = 0.03	0.87	(2, 15.86) = 2.27	0.14
Ground Reaction Forces	F	P	F	P	F	P
Vertical GRF Fz1	(2, 7.31) = 1.05	0.40	(1, 7.46) = 1.01	0.35	(2, 8.14) = 1.86	0.22
Vertical GRF Fz2	(2, 7.21) = 0.63	0.56	(1, 5.79) = 8.75	0.03*	(2, 9.30) = 1.52	0.27
Anterior-Posterior GRF Fy1	(2, 5.99) = 0.71	0.53	(1, 8.66) = 103.95	<0.01*	(2, 7.56) = 4.83	0.05*
Anterior-Posterior GRF Fy2	(2, 7.20) = 3.23	0.10	(1, 4.63) = 5.81	0.07	(2, 7.41) = 0.38	0.70
Load Rate	(2, 7.55) = 0.18	0.84	(1, 6.26) = 9.26	0.02*	(2, 7.90) = 6.08	0.03*
Decay Rate	(2, 9.45) = 4.07	0.05*	(1, 9.62) = 0.23	0.64	(2, 10.35) = 3.51	0.07
Sagittal Plane Joint Moments	F	P	F	P	F	P
Peak ankle dorsiflexor moment during loading response	(2, 17.51) = 0.02	0.98	(1, 14.14) = 2.78	0.12	(2, 17.49) = 0.06	0.95
Peak ankle plantarflexor moment during stance	(2, 7.00) = 3.02	0.11	(1, 7.45) = 1.11	0.33	(2, 7.55) = 0.45	0.65
Peak knee flexor moment during loading response	(2, 13.57) = 0.24	0.79	(1, 15.75) = 2.21	0.16	(2, 12.68) = 0.47	0.63
Peak knee extensor moment during loading response	(2, 13.61) = 3.61	0.06	(1, 8.12) = 13.69	0.01*	(2, 14.23) = 1.05	0.38
Peak knee flexor moment during mid stance	(2, 7.28) = 21.24	<0.01*	(1, 6.45) = 3.04	0.13	(2, 7.54) = 10.51	0.01*
Peak knee flexor moment during late stance	(2, 10.40) = 2.98	0.10	(1, 8.23) = 0.36	0.57	(2, 10.78) = 2.09	0.17
Peak knee flexor moment during swing	(2, 12.56) = 0.42	0.66	(1, 14.37) = 0.40	0.54	(2, 11.41) = 1.73	0.22
Peak hip extensor moment during early stance	(2, 7.57) = 2.50	0.15	(1, 7.97) = 0.03	0.87	(2, 7.89) = 2.89	0.12
Peak hip flexor moment during late stance	(2, 6.21) = 1.83	0.24	(1, 7.22) = 0.02	0.88	(2, 6.70) = 0.02	0.99
Peak hip extensor moment during swing	(2, 12.28) = 1.20	0.33	(1, 14.72) = 0.19	0.67	(2, 10.91) = 0.37	0.70

Table 6.32 Statistical breakdown of stepping up gait trail limb joint and support moments, joint powers and foot trajectories. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a statistically significant result.

	Main Effects				Interaction Effects	
	Time		Limb		Time * Limb	
Frontal Plane Joint Moments	F	P	F	P	F	P
Peak hip abductor moment during early stance	(2, 15.20) = 0.51	0.61	(1, 9.91) = 16.97	<0.01*	(2, 15.10) = 1.62	0.23
Peak hip abductor during late stance	(2, 12.15) = 0.35	0.71	(1, 13.96) = 5.49	0.03*	(2, 10.99) = 1.27	0.32
Support Moments	F	P	F	P	F	P
Initial peak support moment	(2, 8.43) = 0.94	0.43	(1, 12.10) = 0.99	0.34	(2, 9.84) = 1.31	0.31
Second peak support moment	(2, 6.18) = 0.14	0.87	(1, 10.88) = 0.88	0.37	(2, 6.73) = 0.35	0.72
Joint Powers	F	P	F	P	F	P
A1 – Ankle power absorption during stance	(2, 11.68) = 0.36	0.70	(1, 6.40) = 0.04	0.85	(2, 13.14) = 1.42	0.28
A2 – Ankle power generation during pre-swing	(2, 5.66) = 3.91	0.09	(1, 5.08) = 11.58	0.02*	(2, 6.11) = 2.03	0.21
K1 – Knee power absorption during loading response	(2, 6.83) = 0.34	0.72	(1, 13.72) = 4.59	0.05*	(2, 9.43) = 0.30	0.75
K2 – Knee power generation during mid-stance	(2, 2.57) = 4.31	0.15	(1, 10.06) = 18.19	<0.01*	(2, 7.65) = 1.31	0.33
K3 – Knee power generation during pre-swing	(2, 7.52) = 7.78	0.02*	(1, 9.69) = 4.33	0.07	(2, 7.95) = 1.62	0.26
K4 – Knee power absorption during terminal swing	(2, 12.35) = 0.47	0.64	(1, 13.73) = 1.58	0.23	(2, 11.44) = 1.50	0.27
H1 – Hip Power generation during loading response	(2, 6.89) = 0.59	0.58	(1, 15.08) = 0.74	0.40	(2, 10.16) = 0.10	0.91
H2 – Hip power absorption during stance	(2, 11.70) = 1.28	0.32	(1, 13.89) = 0.63	0.44	(2, 10.54) = 0.49	0.62
H3 – Hip power generation during pre-swing	(2, 12.18) = 0.09	0.92	(1, 14.94) = 0.60	0.45	(2, 11.06) = 0.26	0.78
Foot Trajectories	F	P	F	P	F	P
Toe	(2, 2.69) = 1.25	0.41	(1, 6.85) = 0.01	0.96	(2, 7.27) = 0.10	0.90
Heel	(2, 6.71) = 0.29	0.76	(1, 6.37) = 1.44	0.27	(2, 7.12) = 0.31	0.74

6.3.17 Discussion – Stepping Up Gait

When stepping up to a new level, participants increased walking velocity when leading with the intact limb, although no increase was noted over time when leading with the affected limb. In addition, participants tended to select an intact limb lead preference, indicating that this strategy was the most beneficial in terms of stepping up gait performance. Peak vertical heel and toe displacements remained consistent over time when leading with the intact limb, again signifying a stable movement pattern.

The lack of active plantarflexion in the prosthetic ankle when stepping up to a new level resulted in adaptations in the intact limb when acting as the lead limb. Intact limb peak knee flexion during swing phase was increased, this was likely to be a strategy used to aid intact limb toe clearance of the step as has been reported during amputee stair ascent (Alimusaj *et al.*, 2009).

Interestingly, the majority of differences occurred during stance phase once the intact lead limb had stepped up to a new level. In terms of joint kinematics, ankle, knee and hip ROM during stance phase were increased when compared to the affected limb, when performing the same role. Load rate and peak posterior GRF were also increased, along with knee extensor moment and support moment during early stance phase. Peak power absorption (K1) and generation (K2) at the knee also increased over time. These results indicated that the knee extensor musculature was required to contract eccentrically and then concentrically following heel strike in order to raise the whole body COM. Later in stance phase, peak hip flexor moment as well as peak power generation at the ankle (A2) and hip (H3) were increased in the intact limb in order to maintain progression and in preparation for swing phase. This mechanism of utilising the intact limb to negotiate the step and continue progression during stance phase provides a logical explanation for the increases in velocity reported when leading with the intact limb and provides support for the third (3) hypothesis. In addition, it could be suggested that this was a key reasoning behind the selection of the intact limb as the lead limb. Given the assumed reduction in affected limb ability to raise the whole body COM as effectively as the intact limb, participants may benefit from increased affected limb knee and hip extensor strength during activities such as stepping up gait and stair ascent. Adaptations occurring when performing these tasks pre and post strength training warrant further investigation. In addition, it could be hypothesised that rehabilitation of transtibial amputees may be further improved with the inclusion of such strength training exercises including single limb raises and squats. These activities are aimed at increasing the affected limb concentric muscle strength and subsequent power generation at the hip and knee and may improve affected limb ability to raise the whole body COM during stepping up gait and stair ascent. When stepping up and leading with the affected limb, this would allow amputees to utilise this limb more effectively thus changing the lead limb preference and reducing the burden on the intact limb

Adaptations in intact limb function when acting as the trail limb during stepping up gait and the associated increases in walking velocity supported the third hypothesis (3) of an improvement in the ability to step to a new level during gait.

6.3.18 Discussion - Participants at 12 Months Post-Discharge

In the two participants assessed, the pattern of improvement observed during the initial six month period following discharge from rehabilitation continued up to 12 months post-discharge. A common feature across all tasks were the continued increases in temporal-spatial variables, with increases noted in walking velocity and stride length and reductions in relative stance duration. At this point in time, participants functioning in terms of temporal-spatial variables approached those observed in more experienced amputees reported in literature (Sanderson and Martin, 1997; Powers *et al.,* 1998; Bateni and Olney, 2002; Grumillier *et al.,* 2008). This suggested that although significant adaptations had occurred in the year post-discharge from rehabilitation, further increases in functioning may have been possible.

During obstacle crossing and both modes of stepping gait, participants retained their lead limb preferences. This suggested that once selected, a strategy was maintained and utilised regularly. Participants were able to perform the ADLs effectively without tripping or falling. However, if participants had been presented with an unexpected obstacle or task necessitating the non-preferred lead limb and subsequent motor pattern, the risk of tripping or falling may have increased.

In the six month period following discharge, participants were reliant upon the observed increased functioning of the intact limb to induce overall improvement in gait and ADLs. This effect was prevalent at 12 months post-discharge where peak vertical GRF and loading rates were still increased in the intact limb, compared to the affected limb, regardless of the role performed.

There were some noted improvements in the joint kinetics of the affected limb when performing obstacle crossing and stepping gait. In particular, power generation at the ankle (A2) and power bursts at the knee (K1-3) were increased when leading with the affected limb. It could be hypothesised that some of the improvements seen in the performance of level gait and ADLs a year post-discharge, were due to the adaptations in function of the affected limb. However, the functioning of the intact limb also improved at 12 months post-discharge, maintaining the inter limb differences. Literature has reported that there were few kinematic differences between amputees' affected and intact limbs although kinetic differences were reported (Sanderson and Martin, 1997). This statement is supported by the results of the current study, where it is clear that the role of the intact limb was integral to the overall functioning of transtibial amputee movement.

6.4 Conclusion

The literature has reported on transtibial amputees' ability to perform level gait and a variety of ADLs (Hill *et al.,* 1997; Powers *et al.,* 1997; Sanderson and Martin, 1997; Nolan *et al.*, 2003). However, the longitudinal adaptations that occur in the performance of these everyday tasks have not been investigated.

Therefore, the current study explored the biomechanical adaptations in unilateral transtibial amputees' movement patterns when performing level gait and ADLs in the time period following discharge from rehabilitation. This was achieved using three-dimensional motion capture and customised equipment aimed at recreating three ADL.

Results from the current study highlighted a number of adaptations that occurred in level gait, obstacle crossing and stepping gait during the time period following discharge from rehabilitation.

Firstly, there were positive adaptations in level gait kinematics and kinetics that saw walking velocity increase over time, confirming the first (1) hypothesis. However, although the functioning of the affected limb improved, a clear inter-limb asymmetry was noted in terms of GRFs, and joint kinetics, with the intact limb performing a more crucial role in increasing overall gait performance.

Participants were able to cross an obstacle effectively and generally selected an intact lead limb strategy. Across time, the speed at which participants completed the task increased, supporting the second (2) hypothesis. Previously unreported, the current study detailed the adaptations of the intact limb and its key role during the subsequent stance phase following obstacle crossing which was a novel finding. This indicated that when crossing an obstacle, the lead limb must be able to manage the controlled loading during the stance phase and throughout single limb support. This lead limb function during single limb support is vital given that the contralateral is in swing phase and any instability in the stance limb may disrupt the movement, perhaps leading to a trip or fall.

Increased functioning in terms of limb progression during swing phase and stabilising movement during stance phase, suggests that amputees were dependent upon the intact limb to induce overall improvement in obstacle crossing.

There were longitudinal adaptations in stepping down gait that led to an overall improvement in performance, partially supporting the third (3) hypothesis. Participants in the current study tended to lead with the affected limb, with adaptations similar to reports from stair descent, although this preference diminished over time (Jones *et al.,* 2006; Schmalz *et al.*, 2007; Alimusaj *et al.*, 2009). Results from the current study suggested that participants were more comfortable lowering the whole body COM during stance phase on the intact limb. Similarly, the propulsive mechanism required to progress the trail limb was greater in the intact limb, when compared to the affected limb. These factors seemed to dictate lead limb preference, although as this effect reduced over time, it could be concluded that affected limb function when acting as the trail limb improved over time.

Longitudinal adaptations were also noted in stepping up gait, with some characteristics that were indicative of stair ascent. Increases in the speed to task completion when stepping up coupled with the adaptations noted during stepping down gait resulted in the third (3) hypothesis being accepted.

The lead limb preference observed during stepping up gait was a result of the increased ability of the intact limb knee and hip extensor musculature to generate power in order to raise the whole body mass.

Despite the low participant numbers present in the current study, a number of key recommendations could be made. Firstly, affected limb function in terms of joint kinetics was clearly inferior to that of the intact limb. Attempts to rectify this inter-limb asymmetry via improved prosthetic components and rehabilitation techniques focussed on improving knee and hip extensor strength, may improve transtibial amputee performance in the first year post-discharge. This warrants further investigation as the observed lead limb preferences may be reduced, improving amputees' ability to perform motor tasks under unexpected or unusual circumstances, thus reducing the risk of injury or falling. Another pertinent factor deserved of attention is the role of the intact limb and its importance when performing everyday tasks. Although the intact limb played a key role in increasing functioning during the current study, the burden placed on the intact limb may result in early limb degradation and perhaps reduced function (Royer and Koenig, 2005). The effects of the aforementioned attempts to increase affected limb function may reduce this dependence and subsequent chronic limb degradation.

It is not clear if the protective mechanism of the affected limb previously reported and evident in the current study, was a conscious strategy employed by amputees or an unavoidable consequence of transtibial amputation. Future research should focus on addressing this issue with a view of improving affected limb function where possible.

6.5 Chapter Six – Study Three Summary of Findings

Title: Biomechanical Adaptations in Gait and Activities of Daily Living of Transtibial Amputees Following Discharge from Rehabilitation

Patients: Seven transtibial amputees (all men) recently discharged from rehabilitation. Mean ± SD Age 56.1 ± 14.9 years, height 1.82 ± 0.08 metres, mass 91.7 ± 11.4 kg.

Setting: Human performance laboratory.

Intervention: No intervention.

Comparison: Biomechanical variables during level gait and activities of daily living (ADL) in the six month period following discharge from rehabilitation.

Main Findings:	Description
Walking velocity	The speed at which amputees were able to perform level gait and ADLs increased over time.
Kinetic assymetry	The intact limb was more able to absorb and generate joint powers when compared to the affected limb and this asymmetry was still present at six months post-discharge. The intact limb contributed heavily to the increased performance of level gait and ADLs.
Lead limb preference	Amputees generally selected a consistent lead limb preference, leading with the intact limb when crossing obstacles and stepping up and leading with the affected limb when stepping down.
Overall Summary	Overall, amputees ability to perform level gait and ADLs improved in the six months following discharge from rehabilitation. The intact limb played a key role during the successful completion of these tasks. These results were similar to those previously reported from related ADLs and have important implications for clinicians. Rehabilitation or home-based therapy protocols that include targeted improvement of the concentric and eccentric functioning of the affected limb knee extensors may further improve performance of the aforementioned tasks.

7 CHAPTER SEVEN – STUDY FOUR. Adaptations in Balance Function and Postural Control in Transtibial Amputees Following Discharge from Rehabilitation.

7.1 <u>Introduction</u>

There has been little longitudinal research into the adaptations in balance ability and postural control in transtibial amputees over time. Understanding these adaptations could have important implications for the participant and therapists with particular reference to falls prevention.

The aims of the current study were fourfold, with a number of variables being assessed during the six month period following discharge from rehabilitation. Assessments were made using computerised dynamic posturography (CDP) via the Sensory Organisation Test (SOT) and Limits of Stability test (LOS) protocols on the Neurocom Equitest®. Firstly, the study investigated the adaptations in participants' ability to maintain balance whilst experiencing ever increasing dynamic perturbations during the SOT protocol. Secondly, the study investigated changes in participants' reliance upon visual, vestibular and somatosensory sources of information during the SOT protocol. Thirdly, the study investigated the adaptation of 'ankle' and 'hip' strategies use by participants during the SOT protocol. Lastly, adaptations in participants' ability to volitionally alter their COG trajectory towards pre-determined positions were assessed during the LOS test protocol.

Postural sway has been reported to reduce in amputees during rehabilitation (Isakov *et al.,* 1992). Therefore, it was hypothesised that (1) following discharge from rehabilitation balance ability, as measured by equilibrium scores from the SOT protocol, would increase over time. Amputees have been reported to be most reliant upon visual sources of information (Buckley *et al.,* 2002; Vanicek *et al.,* 2009b). Therefore, it was hypothesised that (2) amputees would be most reliant upon visual information as measured by the sensory analysis tool within the SOT protocol. It was also hypothesised that (3) participants' utilisation of the 'hip strategy', as measured by the strategy analysis tool within the SOT protocol, would decrease following discharge from rehabilitation as movements about the intact limb ankle were adapted to counter dynamic perturbations. Lastly, it was hypothesised that (4) participants' ability to volitionally explore their theoretical LOS would increase over time following discharge from rehabilitation, as measured by the COG excursion characteristics from the LOS test protocol.

7.2 Methods

7.2.1 Participants

The participants assessed in the current study were the same group as in study three. Details of participant characteristics are provided in Chapter Six, Table 6.1. Details of the inclusion and exclusion criteria have been outlined in Chapter Three, Section 3.2.2.

7.2.2 Computerised Dynamic Posturography

Computerised dynamic posturography is a quantitative technique for the measurement of upright balance function under a number of controlled conditions that attempt to simulate real life (Nashner, 1997). The Neurocom Equitest® (NeuroCom International, Inc, Clackamas, US) was used to assess balance function during dynamic perturbations in the SOT and postural control during the LOS test.

7.2.3 The Sensory Organisation Test

The SOT was used to assess participants' balance performance, use of sensory information and balance strategies. These analyses were conducted as participants experienced perturbations to somatosensory and visual inputs, via sway referencing, during a sequence of tasks graded in difficulty (Nashner, 1997). Sway-referencing provided the participants with inaccurate somatosensory and/or visual information by perturbing the support surface and/or visual surround, respectively.

The standardised order of the SOT contains eighteen trials of 20 seconds in length comprised of three consecutive trials of six test conditions. During conditions one and two of the SOT, the support surface and surround were stable with the participant's eyes open and closed respectively, providing a baseline measure of balance ability (Figure 7.1). In condition three, the support surface was stable whereas the surround was sway-referenced and may tilt. In the final three conditions, the support surface was sway-referenced with the eyes open and surround fixed (condition four), eyes closed (condition five) and the eyes open with the surround also sway-referenced (conditions six).

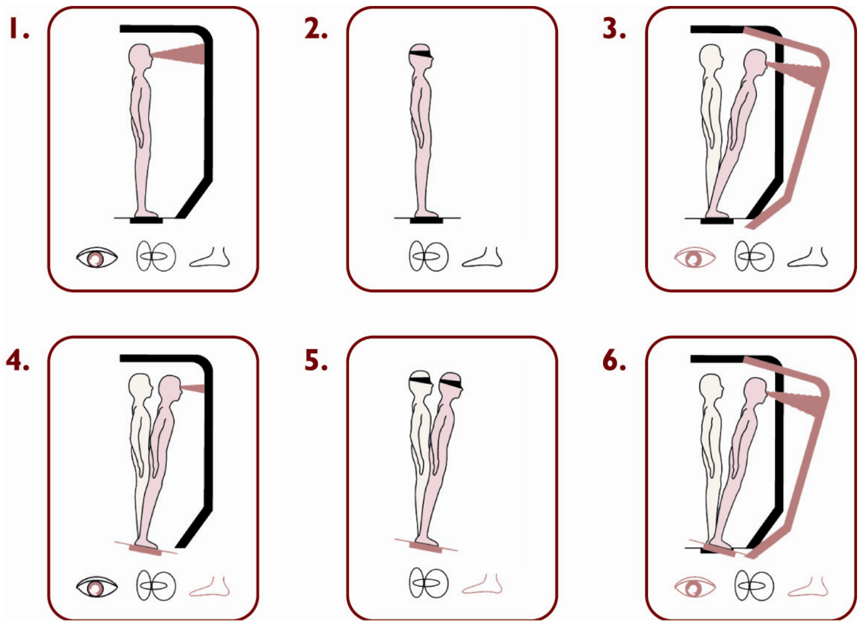


Figure 7.1 Visual representation of the six testing conditions of the sensory organisation test (SOT). Image courtesy of Neurocom International Inc, Data Interpretation Manual.

The SOT test-retest reliability rated from poor to good although significant learning effects have been reported as well as some issues with test sensitivity (Ford-Smith *et al.*, 1995; Leitner *et al.*, 2009). The SOT validity has been outlined, distinguishing differences in balance function between control groups and balance disorder, chronic low back pain and diabetes mellitus/neuropathy populations (El Kahky *et al.,* 2000; Leitner *et al.,* 2009; Emam *et al.,* 2009). The SOT has also been used to validate a less well known measure of balance performance (Broglio *et al.*, 2009).

The SOT was deemed an appropriate test protocol as it is commonly used in the assessment of clinical populations, as well as allowing comparison of amputees balance performance in the current study to those previously reported (Vanicek *et al.,* 2009b). The SOT has also been used in the assessment of postural control in amputees (Vanicek *et al.*, 2009b). Lastly, the detailed information produced from the SOT allows for an in depth investigation into transtibial amputee balance function.

7.2.4 The Limits of Stability Test

The SOT measures balance function in response to, among other things, dynamic perturbations that unexpectedly disrupt the balance system. Dynamic perturbations may not always been encountered by participants and in fact actively avoided, such as standing while riding a bus.

The LOS was used to measure participant’s ability to voluntarily move their centre of gravity (COG). This was achieved via a visual representation of the participant’s COG on a screen that was altered by adapting posture. The LOS requires participants to move their COG to eight pre-determined positions as quickly and as accurately as possible (Figure 7.2). The eight pre-determined positions are representative of an individuals’ 100% limit of stability based upon their height (Wallmann, 2001). Assuming that the body acts as an inverted pendulum with rotation about the ankle, this relates to the amount of movement possible before the COG position necessitates adjustment of the base of support.

Participants were required to hover the visual target over the starting point ('S') (Figure 7.2). Participants then responded to the onset of a visual cue (countdown timer) by moving the cross towards and hovering over or close to the intended target until the trial ended after an eight-second period. The sequence of targets was completed in a standardised clockwise direction starting with position one (Figure 7.2).

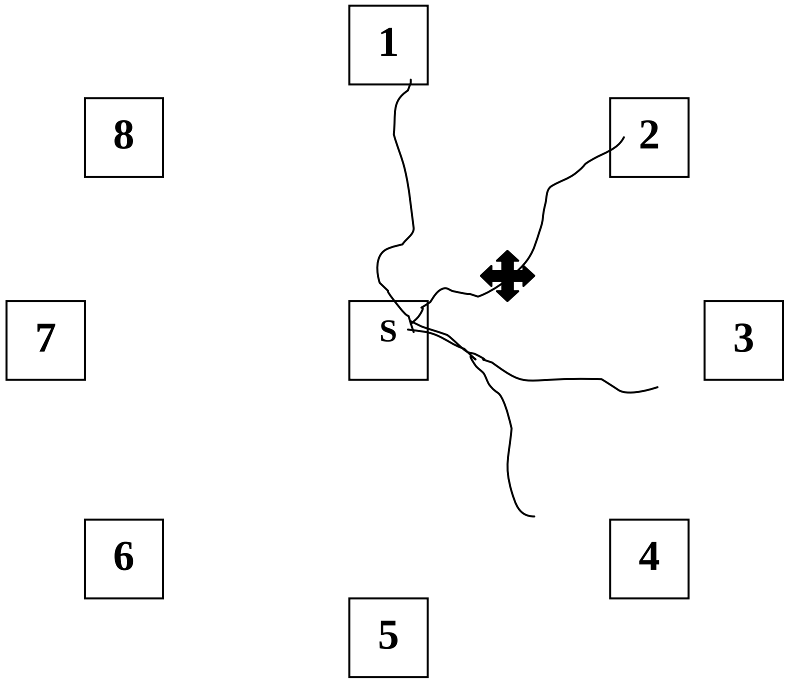


Figure 7.2 Schematic representation of the test protocol of the limits of stability (LOS) test. Directions defined: 1 – Forward, 2 – Affected Forward, 3 – Affected, 4 - Affected Back, 5 – Back, 6 – Intact Back, 7 – Intact and 8 – Intact Forward.

The LOS has been reported as a reliable tool with test re-test reliability being rated from moderate to high for all variables measured across multiple evaluations as well as being consistent and reliable within a population of fallers (Clark *et al.*, 1997; Clark and Rose, 2001). In addition, a variation of the LOS test protocol reported highly reliable results when used to assess a group of stroke participants (Liston and Brouwer, 1996). The LOS has also been shown to be a valid tool, being used to assess postural control in elderly adults, elderly fallers, stroke participants as well as the effects of a balance function intervention programme in the elderly (Clark *et al.*, 1997; Rose and Clark 2000; Clark and Rose 2001).

The LOS was deemed appropriate for use in the current study given its good validity and reliability in the assessment of balance function. In addition, the inclusion of a volitional postural control measure was important in order to assess participants with a more tentative approach to exploring their balancing ability.

7.2.5 Experimental Design and Protocol

The experimental design of the current study was identical to that of study three.

Initially participants' height (m) was recorded using a free-standing height measure (Seca, Birmingham, UK) and entered into the Neurocom Equitest®. Participants were required to complete the SOT protocol followed by the LOS protocol. This standardised order was selected so that the task difficultly was low to start and became progressively more difficult, as recommended by developers of the Neurocom Equitest® (Nashner, 1997). Participants wore their own comfortable, flat footwear during all data collection sessions and were able to fit and re-adjust their own prostheses in order to ensure a comfortable fit. Participants were fitted into an overhead safety harness that prevented them from actually falling whilst allowing them freedom to adjust posture accordingly. The malleoli of the intact limb and prosthetic ankle joint on the affected limb were aligned with the anterior/posterior axis of rotation of the platform. During the SOT, participants were instructed to stand upright and if they reached out to touch the surround or stepped out of position then this was marked as a 'fall' and the trial scored zero (Nashner, 1997). Participants were informed not to move their feet during the LOS unless they felt it necessary to avoid falling.

Once participants had been briefed and prepared, the testing protocols commenced. During administration of the testing protocols, participants were observed for obvious signs of fatigue or above normal levels of instability and rest periods were allowed, although no participant required any intervention in the current study.

7.2.6 Data Analysis

7.2.6.1 The SOT Outcome Measures

For each 20 second SOT trial condition, equilibrium scores were calculated and related to the observed anterior-posterior COG excursion contrasted against a maximal theoretical limit of stability of 12.5^{o} sway, calculated using the participant's height. Increased sway amplitude i.e. increased postural adjustment and shear force production, resulted in a lower equilibrium score being produced on a scale of 0 (poor balance) to 100 (perfect balance). A composite equilibrium score was also produced, providing an overall indication of balance ability. The composite equilibrium score is the arithmetic mean of the condition one mean, condition two mean and each score from conditions three, four, five and six. This score is weighted more heavily towards more complex tasks as sensory balance deficits are deemed to be more easily detected under more challenging conditions (Nashner, 1997). Data were referenced against age-matched disease free normative values provided by the developers (Nashner, 1997).

Strategy analysis during the SOT assessed the amplitude and frequency of shear forces produced in order to move the bodies COG during balance maintenance, inferring the extent to which the 'ankle' or 'hip' strategy was utilised. Reduced amplitude, low frequency shear forces produced by movements about the ankle inferred ankle strategy use with higher frequency and larger amplitude shear forces caused by hip movements inferring hip strategy use. The ankle and hip strategy analysis was combined with and plotted against the corresponding equilibrium score for each trial to produce the strategy analysis. A higher score related to increased bias towards ankle strategy use with lower scores relating to hip strategy use, on a scale of 0 to 100.

The SOT sensory analysis calculated the extent to which amputees relied upon visual, somatosensory or vestibular information to maintain balance and whether there was a reliance upon visual information (preference), even when this information was inaccurate. Increased scores related to improved ability in utilising somatosensory, visual or vestibular information and to a decreased reliance on visual cues (preference). Sensory analyses was used as an heuristic tool as no direct measure of input was recorded and they are calculated using the following ratios of equilibrium scores from specific pairs of sensory test conditions:

Somatosensory	$\frac{\text{Condition Two}}{\text{Condition One}}$	Participants ability to use input from somatosensory system to maintain balance
Visual	$\frac{\text{Condition Four}}{\text{Condition One}}$	Participants ability to use input from visual system to maintain balance
Vestibular	$\frac{\text{Condition Five}}{\text{Condition One}}$	Participants ability to use input from vestibular system to maintain balance
Preference	$\frac{\text{Condition Three + Six}}{\text{Condition Two + Five}}$	Degree to which participant relies on visual information to maintain balance, even when the information is incorrect

7.2.6.2 The LOS Outcome Measures

A number of temporal and spatial variables were derived from the LOS protocol for each of the eight target directions (Figure 7.2)

Reaction time (RT) is the measure in seconds between the onset of the visual cue, to the initiation of movement, measured by COG excursion. Movement velocity (MVL) was measured in degrees per second (deg/sec) relating to the angular velocity participants moved or leaned towards the intended target. Maximum COG (MXE) and endpoint COG (EPE) excursions are measures of the observed percentage (%) COG excursion contrasted against a theoretical maximum based upon the theoretical limit of stability. Directional Control (DCL) is a measure of the observed percentage (%) movement in the intended direction i.e. towards the pre-determined target, versus any other erroneous movement.

7.2.7 Statistical Analysis

A linear mixed model analysis (LMM) was employed, with repeated measures on the factor Time (One Month, Three Months and Six Months). This design allowed for the analysis of changes in multiple balance ability and postural control variables hypothesised a priori (Brown and Prescott, 1999). Each feature of the design (Time) was modelled as a fixed effect with the appropriate model being selected according to the lowest value for Hurvich and Tsai's Criterion (AICC). In the instance of a significant main effect or interaction effect, post-hoc comparisons were conducted using a Sidak adjustment in SPSS v.17.0 (SPSS Inc., Chicago, USA). The alpha level of statistical significance was set at $P \leq 0.05$.

7.3 Results

Group mean (±SD) data are presented from all time points following discharge from rehabilitation for all participants. Data are also presented from a twelve month visit for two participants (one and two) although these results were not analysed statistically.

7.3.1 Sensory Organisation Test

Equilibrium, composite equilibrium and strategy scores from each condition are presented in Figures 7.3 and 7.4 respectively. Statistical analysis of these variables are summarised in Tables 7.1. Sensory analysis results are presented in Figure 7.5, with statistical analysis provided in Table 7.2.

Post-hoc analysis of composite equilibrium scores indicated that balance ability improved by 15.2% between one and six months ($p=0.01$) post-discharge. Visual inspection of Figure 7.3 shows that participants balance ability was better when compared to age-matched, normative data at three months and six months post-discharge in composite equilibrium scores as well as conditions one, four, five and six. Equilibrium scores tended to decrease with task difficulty, reflecting increased anterior-posterior sway during more complex task conditions. Post-hoc comparisons of the significant time effect found in condition two ($p<0.01$) indicated that balance ability improved by 9.8% (79.5 to 87.3) between one month and six months ($p=0.02$). Statistically significant improvements were observed in condition three between one and six months ($p=0.05$) (20.3% increase, 72.3 to 87.0). Condition four produced a significant time effect ($p=0.04$), however post-hoc analysis revealed this initial decline (4.3%) between one and three months ($p<0.05$) was not present between one and six months ($p=0.20$) or from three to six months ($p=0.74$). Equilibrium scores from conditions five and six increased by 29.6% (58.7 to 76.1) and 32.6% (56.4 to 74.8) respectively. This indicated that balance during more challenging perturbations improved over time, post-hoc analysis of condition six revealing this difference to be between one and three months ($p=0.02$) and one and six months ($p=0.01$).

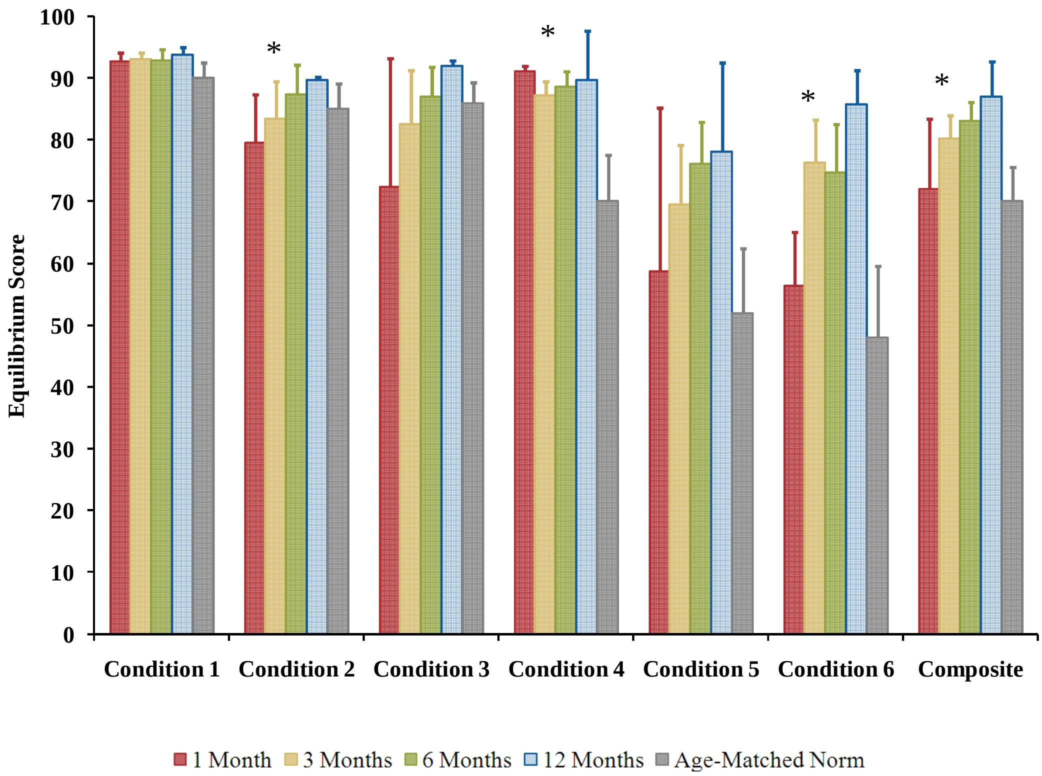


Figure 7.3 Group mean (SD) equilibrium scores from the SOT test protocol. Increased scores relate to improved performance. Data at 12 months from n=2. *Indicate a significant main effect.

From Figure 7.4 it can be observed that as task difficulty increased, use of the hip strategy increased, although this effect tended to reduce over time with statistically significant changes reported in the more dynamic task conditions. During condition one, participants primarily relied on an ankle strategy and this did not alter significantly in the six months following discharge (p=0.55). In condition four, although the ankle strategy was employed to a lesser extent than during condition one, there was no change in strategy score across time (p=0.91). There was an observable increase in the use of the ankle strategy in conditions two (79.3 to 87.4 – 10.2%), three (74.0 to 89.9 – 21.4%) and five (61.8 to 72.9 - 18%) between one and six months post-discharge. However, this effect was only significant in condition five (p<0.01), post-hoc analysis revealing the differences between one and six months post-discharge (p<0.01). The largest increase was observed in the most challenging task conditions present in condition six where the use of the ankle strategy increased dramatically over time between one month (40.5) and three (to 69.8 - 72.3%) (p=0.02) and six months (to 70.6 - 74.3%) (p=0.01) post discharge.

Over time, participants in the current study seemed to become more able to utilise somatosensory and vestibular input in order to maintain balance, with scores increasing by 9.7% and 34.1% respectively between one and six months post-discharge. However, these increases were only significant for somatosensory input (p=0.01). Post-hoc analysis revealing these differences between one and six months post discharge (85.7 to 94.0) (p<0.01). Results for vestibular input failed to reach significance (p=0.07). However, the adaptations in the use of somatosensory information led to participants gaining relative parity with age-matched normative data. Utilisation of visual input to maintain balance did not change over time (p=0.13) however, as can be seen from Figure 7.5, participants seemed to utilise visual information more than somatosensory or vestibular information. Participants also seemed to rely upon visual input more than age-matched normative data. There was no change over time in participants ability to assess the accuracy of visual information (p=0.21), as displayed by the preference analysis.

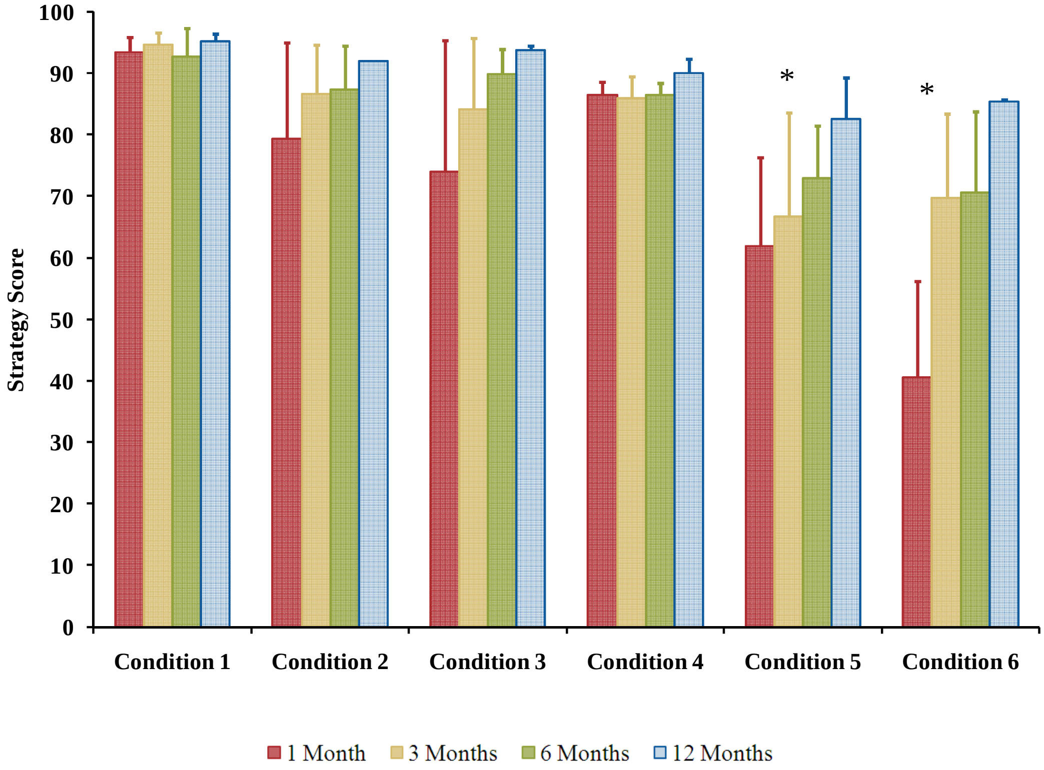


Figure 7.4 Group mean (SD) strategy scores from the SOT test protocol. Increased scores relate to increased reliance upon the ankle strategy. Data at 12 months from n=2. *Indicates a significant main effect.

Table 7.1 Statistical analysis of SOT equilibrium and strategy scores. Results are reported, F value and significance level (p) from the linear mixed model. *Indicates a significant main effect.

SOT Variable		
Equilibrium Scores	F	P
Condition One	(2, 7.02) = 0.08	0.93
Condition Two	(2, 8.48) = 13.53	*<0.01
Condition Three	(2, 17.08) = 3.48	0.05*
Condition Four	(2, 3.89) = 8.33	*0.04
Condition Five	(2, 12.08) = 2.64	0.11
Condition Six	(2, 4.23) = 18.69	*0.01
Composite Score	(2, 9.09) = 6.39	*0.02

Strategy Scores	F	P
Condition One	(2, 7.62) = 0.64	0.55
Condition Two	(2, 7.26) = 2.81	0.13
Condition Three	(2, 19.77) = 3.25	0.06
Condition Four	(2, 8.27) = 0.10	0.91
Condition Five	(2, 6.09) = 13.20	*0.00
Condition Six	(2, 6.36) = 15.53	*<0.01

Table 7.2 Statistical analysis of SOT sensory analysis scores. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a significant main effect.

SOT Variable		
Sensory Analysis Scores	F	P
Somatosensory	(2, 19.22) = 6.88	*0.01
Visual	(2, 9.49) = 2.54	0.13
Vestibular	(2, 12.91) = 3.28	0.07
Preference	(2, 5.07) = 2.21	0.21

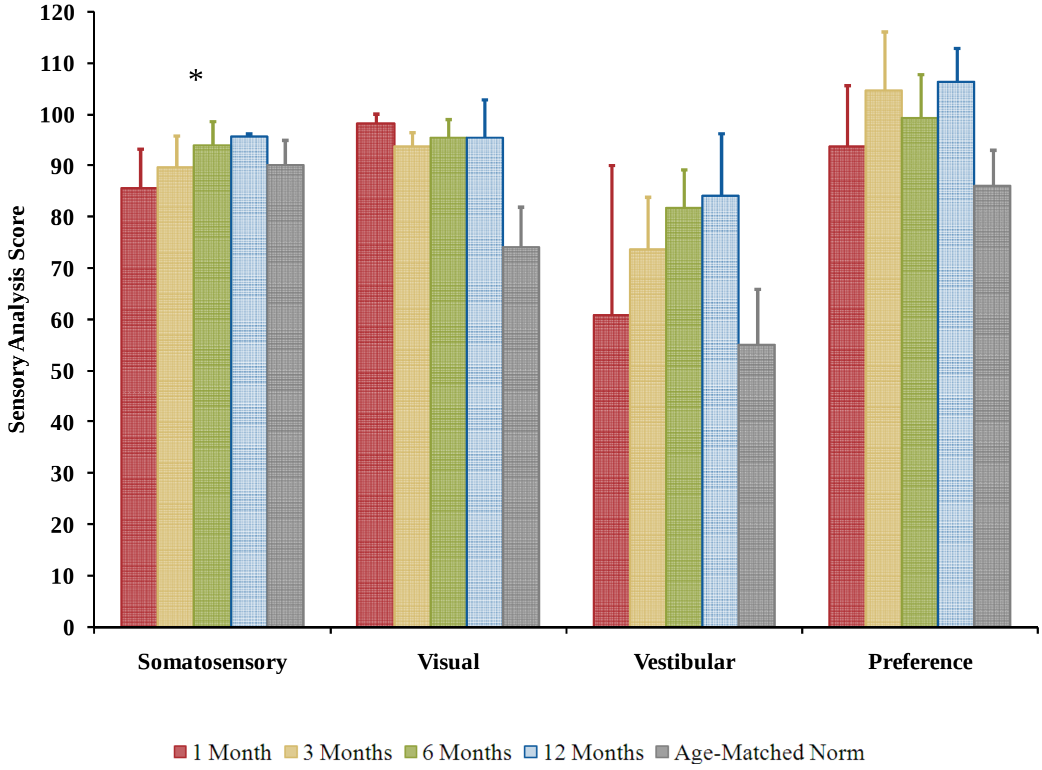


Figure 7.5 Group mean (±SD) sensory scores from the SOT test protocol. Increased scores relate to improved ability in utilising a particular input (somatosensory, visual and vestibular) and a decreased reliance on visual cues in maintaining balance (preference). Data at 12 months from n=2. *Indicates a significant main effect.

7.3.2 Limits of Stability Test

Reaction time, movement velocity, endpoint and maximal COG excursion and directional control scores are presented for each of the eight target directions in Figure 7.6. Statistical analysis of these variables is presented in Table 7.3.

Reaction time decreased in the intact direction (0.70 seconds), intact forward (0.60 seconds), forward (0.46 seconds) and affected back (0.52 seconds) directions between one and six months although these were not statistically significant. However, reaction time increased in the affected forward (0.42 seconds), affected (0.27 seconds), back (0.67 seconds) and intact back directions (0.67 seconds), with the backwards direction increasing significantly (p=0.03). Although no statistical comparisons were drawn,

Figure 7.6 illustrated that reaction time was generally greater on the intact limb than on the affected limb at one month post-discharge, although this effect diminished over time. Figure 7.6 also illustrates that participants seemed to have greater reaction times in all directions when compared to age-matched normative data.

Changes in movement velocity were variable over time, although a significant decrease was observed in the affected back direction (0.53 degrees/second) ($p=0.02$) between one and six month post discharge ($p<0.05$). This suggests that participants were not able to modulate the speed at which they leaned towards an intended target. Although no statistical comparisons were drawn, it can be seen from Figure 7.6 that movement velocity was faster in the medio-lateral directions than in the anterior-posterior directions. Also, Figure 7.6 shows that participants in the current study moved towards intended targets more slowly than individuals presented in the age-matched normative data.

Both endpoint and maximal COG excursion increased following discharge from rehabilitation. Post-hoc analysis showed that increases in endpoint COG excursion were significant in the intact forward direction ($p=0.01$) and increased by 77.2% between one and three months ($p=0.02$) and by 78.8% between one and six months ($p=0.02$) post-discharge. With regards to maximal COG excursion, a statistically significant increase of 16.2% was noted in the affected forward direction during the six month period following discharge, although post-hoc analysis did not reveal where differences occurred ($p=0.03$). Although no statistical comparisons were drawn, Figure 7.6 illustrates that participants were better able to explore their LOS on the intact side, especially with the addition of an anterior (intact forward) or posterior (intact back) component. However, performance was still reduced when compared to age-matched normative data.

Scores for directional control improved over time in all directions except for intact and intact back. Figure 7.6 illustrates the significant improvements in affected forward ($p=0.04$), intact forward ($p=0.01$) and back ($p<0.01$) directions. Post-hoc analysis revealed these improvements to be between one and three months post discharge for back (12.3%) ($p<0.01$) and intact forward (44.4%) ($p=0.02$) directions and between one and six months for the intact forward direction (45.1%) ($p<0.01$). Post-hoc analyses did not reveal where differences occurred in the affected forward direction. There were also observable increases in directional control of 30.2% and 72.0% in the affected and affected back directions respectively, between one and six months following discharge, although these were not statistically significant.

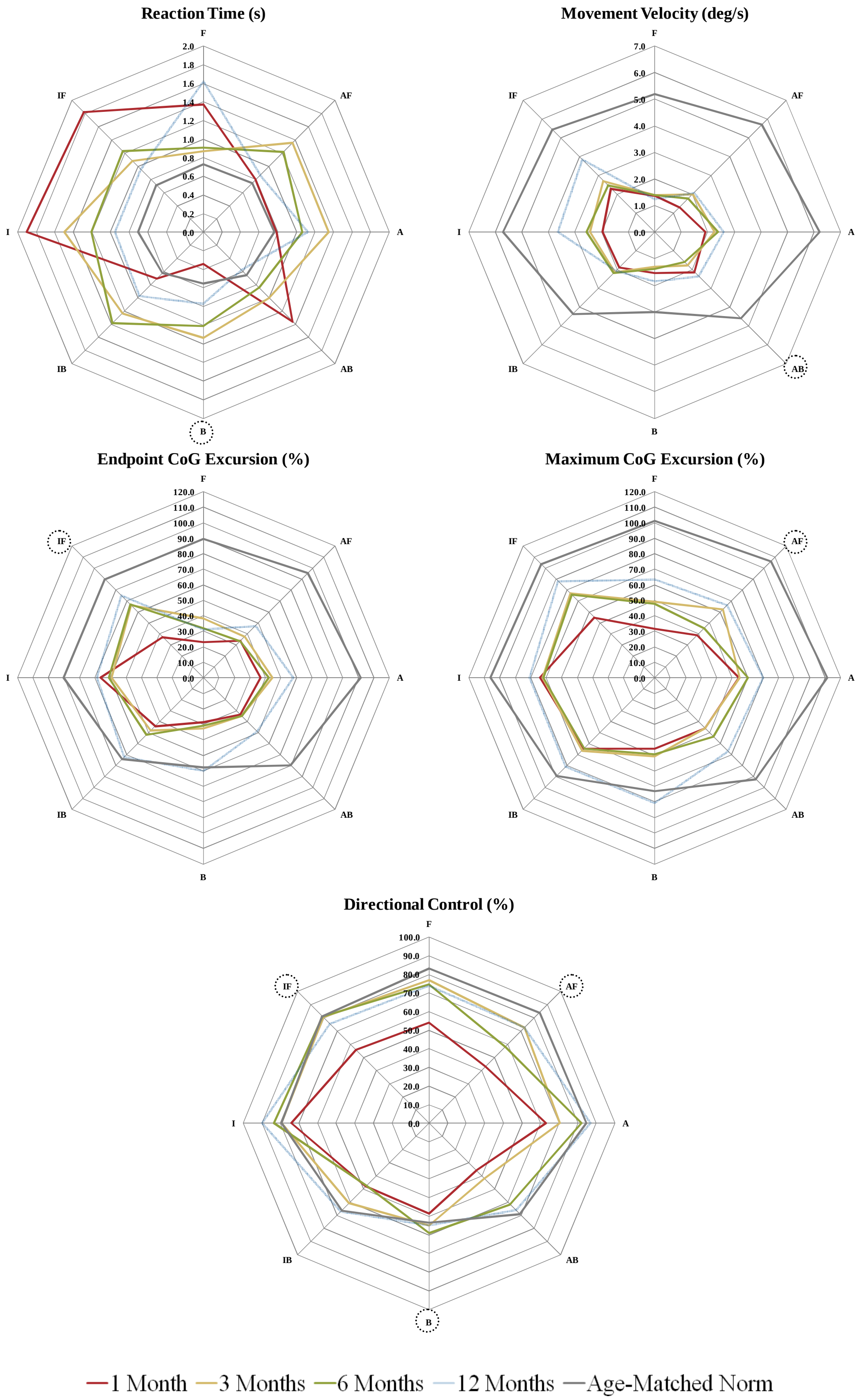


Figure 7.6 Target plots of group mean scores from LOS test protocol. Scores closer to outer border indicate increased performance with the exception of reaction time where scores closer to centre indicate increased performance. Data at 12 months from n=2. Circled directions produced a significant main effect.

Table 7.3 Statistical analysis of variable scores from the LOS test protocol. Results are reported (F value and significance level (p) from the linear mixed model. *Indicates a significant main effect.

Direction	Forward		Affected Limb Forward		Affected Limb		Affected Limb Backwards		Backwards		Intact Limb Backwards		Intact Limb		Intact Limb Forwards	
LOS Variable	F	P	F	P	F	p	F	p	F	p	F	p	F	p	F	p
Reaction Time	(2, 6.10) = 1.16	0.38	(2, 4.00) = 1.92	0.26	(2, 2.65) = 6.43	0.1	(2, 6.76) = 2.07	0.20	(2, 10.06) = 5.31	*0.03	(2, 9.71) = 1.73	0.23	(2, 3.54) = 0.87	0.49	(2, 3.94) = 5.92	0.07
Movement Velocity	(2, 6.34) = 0.83	0.48	(2, 3.93) = 1.22	0.39	(2, 7.52) = 0.38	0.70	(2, 5.82) = 8.83	*0.02	(2, 4.86) = 0.26	0.78	(2, 3.10) = 0.08	0.92	(2, 4.94) = 0.49	0.64	(2, 7.22) = 0.22	0.81
Endpoint COG Excursion	(2, 32.16) = 0.57	0.57	(2, 21.74) = 0.1	0.91	(2, 4.36) = 0.24	0.80	(2, 2.64) = 0.08	0.93	(2, 9.47) = 0.28	0.77	(2, 18.25) = 0.12	0.89	(2, 7.54) = 0.28	0.76	(2, 8.67) = 7.68	*0.01
Maximum COG Excursion	(2, 5.79) = 0.18	0.84	(2, 14.23) = 4.64	*0.03	(2, 4.06) = 0.68	0.56	(2, 5.59) = 1.00	0.42	(2, 2.62) = 0.13	0.89	(2, 9.73) = 0.21	0.81	(2, 14.26) = 0.14	0.87	(2, 2.62) = 3.45	0.19
Directional Control	(2, 5.83) = 2.43	0.17	(2, 8.77) = 4.69	*0.04	(2, 19.76) = 1.94	0.17	(2, 7.06) = 3.17	0.10	(2, 13.28) = 8.44	*<0.01	(2, 6.11) = 1.76	0.25	(2, 2.19) = 3.14	0.23	(2, 8.73) = 8.71	*<0.01

7.3.3 Data for n=2 at 12 Months Post-Discharge

With regards to participants' performance during the SOT, the trend of increasing equilibrium scores continued at 12 months post-discharge from rehabilitation. However, scores from conditions one and four were similar to those observed in the group analyses at six months suggesting that a ceiling effect had been reached. Similar trends of improvement were noted in the strategy analyses, with increased ankle strategy use at 12 months post-discharge. Interestingly there was also an improvement in ankle strategy use in condition four, despite no performance improvement as illustrated by the equilibrium score during that condition. The use of somatosensory and vestibular information in balance maintenance continued to improve after six months post-discharge although participants still heavily relied on visual information.

Analyses from the LOS test protocol indicated that participant's reaction time reduced in all directions with forwards being the only major exception. The speed at which participants moved towards targets at 12 months post-discharge was slightly better in affected, affected back and back directions and markedly better in the intact and intact forwards directions. Endpoint and maximum COG excursion continued to improve after six months post-discharge with the exception of forward direction endpoint COG excursion. Lastly, directional control remained relatively similar to performance at six months post-discharge. In general, participant's performance during the LOS test was still below that of age-matched normative data, with the exception of directional control.

7.4 Discussion

The adaptations in amputee balance ability and postural control are time following discharge from rehabilitation, have not been investigated. Therefore, the current study assessed a number of aspects of balance performance and postural control during the six month period following discharge from rehabilitation. This was achieved using CDP and in particular the SOT and LOS test protocols.

7.4.1 Sensory Organisation Test

Results from the current study suggested that in general, amputees' balance ability in response to dynamic perturbations improved following discharge from rehabilitation. The greatest change in equilibrium scores occurred during the most challenging test condition (condition six) confirming the first hypothesis of increased equilibrium scores from the SOT protocol following discharge from rehabilitation. These results follow on from previous reports of increased balance function during rehabilitation, thus suggesting that the adaptation of balance function is an ongoing process that continues until at least six month post-discharge (Isakov *et al.*, 1992). The combination of improved balance ability during highly dynamic perturbations over time (condition six), the lack of significant change during the static balance task (condition one), and the increased A-P sway represented by lower equilibrium scores as the SOT increased in difficulty, could have important implications for transtibial amputees. These results suggest that following discharge from rehabilitation, participants may benefit from practising balance tasks whereby balance is dynamically perturbed as these highly challenging task conditions may elicit further or more rapid increases in overall balance ability. Such tasks may include balance whilst on uneven or varied terrain (e.g. wobble board), with different frictional properties, on surfaces that are made up of interchangeable material and density. The addition of dual tasking has been shown to further perturb balance and may more accurately reflect a real life situation, such as maintaining balance whilst completing a household activities such as cleaning and cooking (Aruin *et al.*, 1997).

The significant decrease in reliance on the hip strategy during more dynamic task conditions as a function of time, confirmed the third hypothesis regarding reduced hip strategy use over time. Adequate joint flexibility and muscle strength are reportedly important in order to respond to postural perturbations effectively (Horak *et al.,* 1989). In addition, the literature has postulated that amputees may use the more rigid prosthetic ankle mechanism to maintain balance, thus reducing the biomechanical degrees of freedom required to control the lower limb (Hermodsson *et al.,* 1994). This suggests that any further balance training or prosthetic prescription should be mindful of the prosthetic ankle joint function in order to improve overall balance function. A previous study with transtibial amputees reported the increased use of the ankle strategy during easier task conditions, with increasing hip strategy use as task difficulty increased (Vanicek *et al.,* 2009b). This was also observed in the current study. These results support the rationale for the use of dynamic balance assessment in this population group in order to investigate balance function comprehensively (Buckley *et al.,* 2002).

Interestingly, the use of the ankle strategy during condition four, where accurate visual information was provided during support surface perturbation (inaccurate somatosensory information) (Figure 7.1), did not change significantly over time. This suggests that participant's may have prioritised accurate visual information over the perturbed somatosensory information, which is supported by the suggestion that in unusual sensory environments, the most reliable source of sensory information, in this case vision, may be selected (Horak *et al.,* 1989).

Previous reports have illustrated amputees reliance upon visual input during both static (Isakov *et al.,* 1992) and dynamic conditions (Vanicek *et al.,* 2009b). Results from the current study concur with these reports, as the sensory analysis displayed an overall heightened use of visual input when compared to somatosensory or vestibular input. This trend did not change over time, reflected in the lack of a time main effect for visual input, supporting the second hypothesis that participants would be most reliant upon visual information.

Despite the perceived reliance upon visual input to maintain balance during the SOT in the current study and in previous reports of decreased balance function in transtibial amputees, one study reported that some aspects of an amputee's balance ability were better when compared to age-matched normative data (Vanicek *et al.,* 2009b). This effect has been attributed to the low control demands of the stiff prosthetic ankle-foot complex limb (Hermodsson *et al.,* 1994). Results from the current study revealed a significant increase in somatosensory input use, which may be linked to the overall increase in balance performance. It must be stated that the observed increase in somatosensory information use could also be attributed to the intact limb, as previous studies have reported increased weight bearing on the intact limb during dynamic balance (Vanicek *et al.,* 2009b). However, despite the loss of somatosensory information from the lower limb following amputation, it could be hypothesised that increases in the use of somatosensory input originates from the affected limb. Previous literature provides an insight into this hypothesis, reporting that transtibial amputees increased affected limb board-floor contact time in an attempt to gain extra somatosensory input during a dynamic uniaxial balance task (Buckley *et al.*, 2002).

In addition, this hypothesis has an interesting link to the scenario where the event of an actual fall was not strongly linked to the fear of falling, as amputees may expect to fall whilst attempting complex motor tasks (Miller *et al.*, 2001a). In addition, when compared to amputee non-fallers during a dynamic translator balance task, amputee fallers have been shown to weight-bear more on the affected limb than intact limb (Vanicek *et al.*, 2009b). This suggests that safely increasing an amputee's ability in utilising the somatosensory input from the affected limb, without increasing falls risk, may aid the development of balance ability.

7.4.2 Limits of Stability Test

Results from the LOS test protocol represent the volitional aspect of postural control in the current participant group. Reaction time in the backwards direction increased significantly over time and overall, reaction time was increased when compared to age-matched reference data. This may reflect participants' reluctance or inability to quickly initiate movement due to decreased afferent somatosensory input or fear of falling (Miller *et al.,* 2001a). This observation is matched by the lack of statistically significant increases in movement velocity in all directions except the affected backwards direction. In addition, movement velocity was also consistently reduced when compared to age-matched normative data. Interestingly, movement velocity was generally faster in the M-L directions than the A-P directions, perhaps reflecting an unwillingness to lean forwards or backwards quickly. This may be due to a number of reasons including; reduced theoretical M-L LOS negating the postural control requirement in these directions, fear of falling being greater in the A-P direction than the M-L direction, relative lower limb muscle strength controlling M-L movement or prosthetic fitting. Although reports of these affects are unknown, they would benefit from further investigation. When combined, these results suggest that transtibial amputees did not modulate how they reacted to movement stimulus or the speed at which they moved in the six months following discharge from rehabilitation. This is a novel finding as various more reactive measures of balance ability produced from the SOT protocol were subject to change. However, when volitionally required to stress the postural control system, participants seemed more reluctant or unable to do so.

Although participants reacted to the onset of stimulus slowly and did not move towards the intended target quickly, significant adaptations were noted in the accuracy of these movements. Directional control improved significantly in the affected forward, intact forward and backwards directions with large and perhaps clinically meaningful increases in the affected and affected backwards directions. These results suggest that there was a trade off in volitional exploration of LOS. Although participants did not modulate their reaction time or movement velocity, the control and accuracy of these movements was increased, particularly on the affected limb and in the backwards direction. This hints at a speed-accuracy trade off that has been well reported in the motor control literature and the effect of which warrants further investigation (Fitts, 1954; Plamondon and Alimi, 1997; Danion *et al.,* 1999). It could be hypothesised that with greater experience, the speed of movement are also increased, following the initial improvement in movement accuracy.

The combination of these findings is also related to the increases noted in both the endpoint and maximum COG excursion. The significant increases reported from the affected forward and intact forward directions indicated that participants got closer to their theoretical maximum LOS with increased accuracy. Lower limb amputees' dependence upon the intact limb during dual tasking in static posture has been reported in the literature (Aruin *et al.*, 1997). Similarly, participants in the current study were not able to get as close to their theoretical maximum COG excursion when leaning towards the affected limb in comparison to the intact limb. A study assessing postural sway, utilising dual-force plate methodologies reported increased sway associated with the affected limb in comparison to the intact limb (Isakov *et al.,* 1994). Computerised dynamic posturography utilising the SOT protocol reported that amputee fallers have relied more upon the use of the affected limb to maintain balance under dynamic perturbation (Vanicek *et al.*, 2009b). Previous reports of affected limb function during balance tasks, coupled with the observed affected limb adaptations reported in the current study, may have important implications for transtibial amputee postural control. It could be hypothesised that the level of postural control associated with affected limb necessitates the use of the intact limb in successful postural control. However, everyday circumstances may necessitate a level of affected limb use during balance beyond amputees preferred volitional level. As postural sway has been reported to reduce as a function of time across rehabilitation, it could be suggested that activities practicing the volitional use of the affected limb during postural control may be beneficial (Isakov *et al.,* 1992). There are contemporary low cost tools such as the Nintendo Wii ™ utilising similar COG excursion assessments, as seen in the LOS, that have been reported to increase balance function in various other clinical populations (Deutsch *et al.,* 2008; Brown *et al.,* 2009). This reasoning corroborates the findings from the SOT protocol and the proposed need for increased affected limb function during dynamic balance tasks to increase overall balance function.

Although amputees were more able to explore their theoretical LOS as a function of time, many of these results were not statistically significant therefore, the hypothesis that participants' ability in this task would increase over time was rejected.

7.4.3 Participants at 12 Months Post-Discharge

In the two participants assessed, balance ability continued to improve after six months post-discharge, particularly during balance tasks that incorporated dynamic perturbations. In addition to this, ankle strategy use increased, even during more static balance tasks. This suggests that amputees may continue to improve balance by further modification of the ankle strategy use. This may help to explain the further increases in balance ability, as participants were still heavily reliant upon visual information.

Measures from the LOS test protocol indicated that the volitional aspect of postural control improved up to six months post-discharge. The temporal components improved with the spatial components remaining relatively stable and roughly equal to performance noted from an age-matched control population. This suggested that the hypothesised speed-accuracy trade off observed at six months post-discharge continued to develop at twelve months post-discharge.

7.5 Conclusion

In conclusion, results from the current study indicated that overall balance ability during dynamic perturbation improved in the time period following discharge from rehabilitation in unilateral transtibial amputees, confirming the first hypothesis. However, these individuals were heavily reliant upon vision in order to maintain balance, supporting the second hypothesis. Increased use of the ankle strategy validated the third hypothesis and, along with perceived attempts to increase somatosensory input from the affected limb, may have explained the improvements in overall balance function. Following discharge from rehabilitation, amputees were seemingly able to increase the spatial aspects of volitional exploration of their theoretical LOS and did so with more accuracy. However, the first hypothesis was rejected as the temporal aspects, namely reaction time and movement velocity, did not display any adaptation suggesting a speed-accuracy trade off effect.

Although low participant numbers may have influenced the statistical power of the current study, there are recommendations that could be made using the current data set. It could be suggested that further practice of balance ability and postural control should focus upon improving affected limb function. In addition, practice of balance tasks with reduced visual information provided may reduce amputee’s overreliance upon this source of information. Performing volitional postural movements under increasing time pressure may also improve postural control in terms of amputee’s ability to react and respond to unexpected perturbations. As mentioned previously, there are currently low cost tools that could be employed as an intervention to achieve some of these suggestions. Future research quantifying the effect of these interventions and their impact on subsequent falls rate, balance confidence and QOL, among other variables, would be of use to clinicians involved in the care of transtibial amputees.

7.6 Chapter Seven – Study Four Summary of Findings

Title: Adaptations in Balance Function and Postural Control in Transtibial Amputees Following Discharge from Rehabilitation

Patients: Seven transtibial amputees (all men) recently discharged from rehabilitation. Mean ± SD Age 56.1 ± 14.9 years, height 1.82 ± 0.08 metres, mass 91.7 ± 11.4 kg.

Setting: Human performance laboratory.

Intervention: No intervention.

Comparison: Scores from the sensory organisation test (SOT) and limits of stability test (LOS) protocols using the Neurocom Equitest in the six month period following discharge from rehabilitation

Main Findings:	Description
Balance ability	Balance ability improved over time, particularly in the more challenging task conditions. Amputees increased the use of the ankle strategy to maintain balance.
Vision	Amputees were most reliant upon vision, even when visual information was inaccurate
Postural movement	The spatial and accuracy components of postural movements improved over time, although the temporal aspects of these movements did not, suggesting a speed-accuracy trade off effect.
Overall Summary	Balance and postural control improved during the six month period following discharge from rehabilitation. However, amputees were heavily reliant upon visual information in order to maintain balance, which may be a problematic strategy given the typical age of the population group. Reaction to stimulus and the speed to postural movements did not improve over time which suggested that amputees may not be very well equipped to react to unexpected perturbations. Further practice of balance tasks with reduced visual information may reduce amputee overreliance upon this source of information. Performing volitional postural movements under increasing time pressure may also improve postural control in terms of amputees ability to react and respond to unexpected perturbations. Low cost tools are available that could be employed as an intervention to achieve these adaptations. Research quantifying the effect of such interventions on balance, falls rate, balance confidence and QOL would be of use to clinicians involved in the care of transtibial amputees.

8 CHAPTER EIGHT – STUDY FIVE. Changes in Generic and Prosthesis Related Quality of Life and Falls Efficacy in Transtibial Amputees Following Discharge from Rehabilitation.

8.1 Introduction

Although a profile of lower limb amputee QOL has been presented, the literature has not extensively investigated this area of research. Understanding changes that occur over time following discharge from rehabilitation in amputees' QOL is important. This may have long term implications with regards to mobility and social re-integration as well as participation in future physical activity and employment.

Therefore, the aims of the current study were three fold. The first aim was to investigate the psychological changes that occurred in both generic and prosthesis related self-reported QOL in transtibial amputees up to six months following discharge from rehabilitation. The second was to investigate the differences between mental and physical health during that same time frame. Lastly, the third aim of the current study was to investigate the changes in falls efficacy following discharge from rehabilitation and the link between falls efficacy and measures of QOL.

It was hypothesised that (1) QOL would increase following discharge from rehabilitation, specifically the physical health aspect of QOL, as participants achieved further increases in mobility. Despite these hypothesised improvements, it was also hypothesised that (2) mental health would be reported to be higher than physical health, as has been reported previously (Legro *et al.,* 1999; Pezzin *et al.,* 2000; Van der Schans *et al.,* 2002; Asano *et al.,* 2008; Zidarov *et al.,* 2009). Lastly, it was hypothesised that (3) changes in falls efficacy would follow a similar pattern to the hypothesised changes in QOL.

8.2 Methods

8.2.1 Participants

The participants assessed in the current study were the same group reported in study three thus, details of participant characteristics are provided in Six, Table 6.1. Details of the inclusion and exclusion criteria have been outlined in Chapter Three (Section 3.2.2).

8.2.2 The Medical Outcomes Study Short Form-36

The SF-36 questionnaire used in the current study was identical to that used within study two (Appendix E) and is described in detail in Chapter Three (Section 3.6.1).

8.2.3 The Prosthesis Evaluation Questionnaire

The Prosthesis Evaluation Questionnaire (PEQ) is a measure of prosthesis related QOL (Legro *et al.,* 1998). The PEQ consists of 82 items, 42 of these items produce a nine-scale profile of health namely, Ambulation, Appearance, Frustration, Perceived Response, Residual Limb Health, Social Burden, Sounds, Utility and Well Being. The scales are independent thus can be assessed in isolation. The PEQ is described in detail in Chapter Three (Section 3.6.2).

8.2.4 The Modified Falls Efficacy Scale

The modified falls efficacy scale (mFES) is a self-report measure of fear of falling or falls efficacy (Hill *et al.,* 1996). The mFES consists of 14 items aimed at assessing falls efficacy during both indoor and outdoor activities. Examples of the ten items assessing indoor activities include getting dressed and bathing with crossing roads and using public transport examples of the four outdoor activities assessed. The mFES is described in detail in Chapter Three (3.6.3).

8.2.5 Experimental Design and Protocol

The experimental design of the current study was identical to that of study three. Participants were required to complete an SF-36 questionnaire, PEQ and mFES questionnaire at data collection sessions at one, three and six months following discharge from rehabilitation. Questionnaires were completed upon arrival at the Human Performance Laboratory, Department of Sport, Health and Exercise Science, University of Hull and prior to completing the movement and balance tasks outlined in studies three and four. The rationale for this ordering in the protocol was outlined in study two. Participants were encouraged to respond to questions based upon their own interpretation and if required, questions were repeated verbatim by the researcher.

8.2.6 Data Analysis

Analysis of the SF-36 questionnaire has been described in detail in study two.

The paper hard copies of PEQ and mFES questionnaires were collected and scored by the same researcher and raw data manually inputted into a Microsoft Excel workbook (Microsoft, Reading, UK).

Scale scores for the PEQ were calculated using the arithmetic mean of the item scores contained within the relevant scale. At least half of the items within a specific scale must be answered to retrieve a valid scale score. As scales were individually validated and tested for reliability, each scale can be used and interpreted individually. Appendix F, Table F.1 provides details of the item content and the scale to which they contribute. The scoring system of the PEQ is such that a higher score indicates a more positive score.

The scoring system of the mFES is such that a higher score indicates lower fear of falling. The overall mFES score was calculated as the arithmetic mean of all 14 item scores. The arithmetic mean of relevant items were used to calculate Factor One (indoor activities), Factor Two (outdoor activities) and total or overall mFES scores.

8.2.7 Statistical Analysis

A linear mixed model analysis (LMM) was employed, with repeated measures on one factor, Time (One Month, Three Months and Six Months). This design allowed for the analysis of changes in multiple measures of QOL and falls efficacy hypothesised a priori (Brown and Prescott, 1999). Each feature of the design (Time) was modelled as a fixed effect with the appropriate model being selected according to the lowest value for Hurvich and Tsai's Criterion (AICC). In the instance of a significant main effect or interaction effect, post-hoc comparisons were conducted using a Sidak adjustment in SPSS v.17.0 (SPSS Inc., Chicago, USA). The alpha level of statistical significance was set at $P \leq 0.05$.

8.3 Results

Group mean (±SD) data were presented from all time points following discharge from rehabilitation for all participants. Data were also presented from a twelve month visit for two participants (one and two) although these results were not analysed statistically. All statistical analyses are presented in Table 8.1.

8.3.1 SF-36

Group mean scale scores, component summary scores and total SF-36 are presented in Figures 8.1 and 8.2, respectively. Although reports from scales at six months tended to be slightly higher than at one and three months, only role emotional was close to producing a significant time effect with scores increasing two-fold ($p=0.07$). Figure 8.1 shows that when compared to age-matched normative data, amputees in the current study reported higher QOL, with the exception of role physical.

With regards to the component summary scores, the MCS increased by 14.3% between one and six months post-discharge, although this was not statistically significant. However, scores from the PCS were lower than the MCS and did not change significantly over time (p=0.60). Total SF-36 scores did not significantly increase over time (p=0.30).

8.3.2 PEQ

Group mean scale scores from the PEQ are presented in Figures 8.3. Figure 8.3 displays the increases in scores for scales pertaining to participants' prostheses between one and six months post-discharge from rehabilitation (Utility - 21.2%, Sounds - 49.0%, Frustration - 24.0% and Appearance - 21.8%), although these were not significant. The perceived reaction of close family members and friends (Perceived Response) was not reported to have changed significantly over time (p=0.80) and was consistently the most positive score for participants in the current study. There were no significant changes on the remaining scales of the PEQ.

8.3.3 mFES

Group mean overall, Factor One and Factor Two mFES scores are presented in Figure 8.4. Overall falls efficacy did not change over time (p=0.25). Further analysis highlighted that this trend was not task specific as no significant changes were observed over time for indoor (Factor One) (p=0.27) or outdoor tasks (Factor Two) (p=0.18). This suggested that participants' confidence in executing ADLs without falling was similar as time passed following discharge from rehabilitation. In addition, this effect was similar as participants attempted both indoor and outdoor tasks.

8.3.4 Data for n=2 at 12 Months Post-Discharge

At twelve months following discharge from rehabilitation, there was a vast improvement in all SF-36 scale scores both in comparison to the score reported at six months post-discharge and age-matched normative reference data. There was also a noted increase in PCS and MCS scores and thus, total SF-36 score. In addition, mental health and physical health seemed to contribute equally to overall QOL with the discrepancy seen at six months post-discharge diminishing.

Increases were also noted in most PEQ scales scores with two exceptions, appearance and perceived response, that were similar to scores reported at six months post-discharge.

Finally, overall falls efficacy improved markedly from six months post-discharge as a result of increasing scores in both Factor 1 and Factor 2 activities.

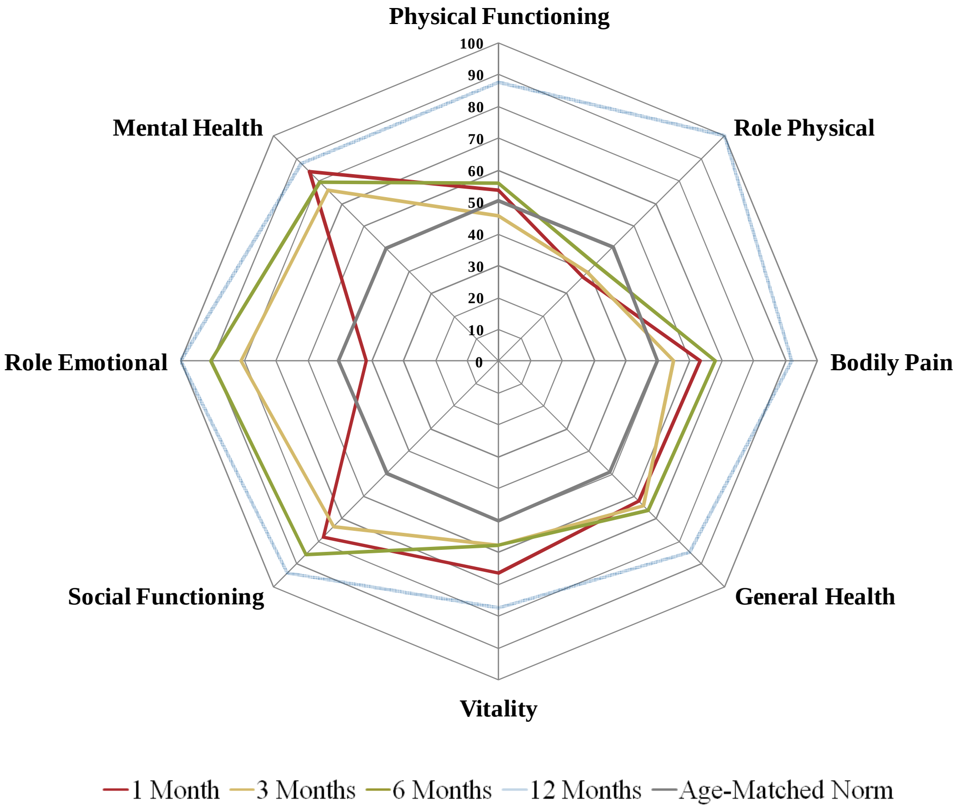


Figure 8.1 Target plot of group mean transformed scores from eight scales of SF-36. Age matched normative data are presented to provide a visual comparison (Ware *et al.*, 2000). Scores closer to the outer border of the plot relate to increased QOL in that scale. Data at 12 months from n=2.

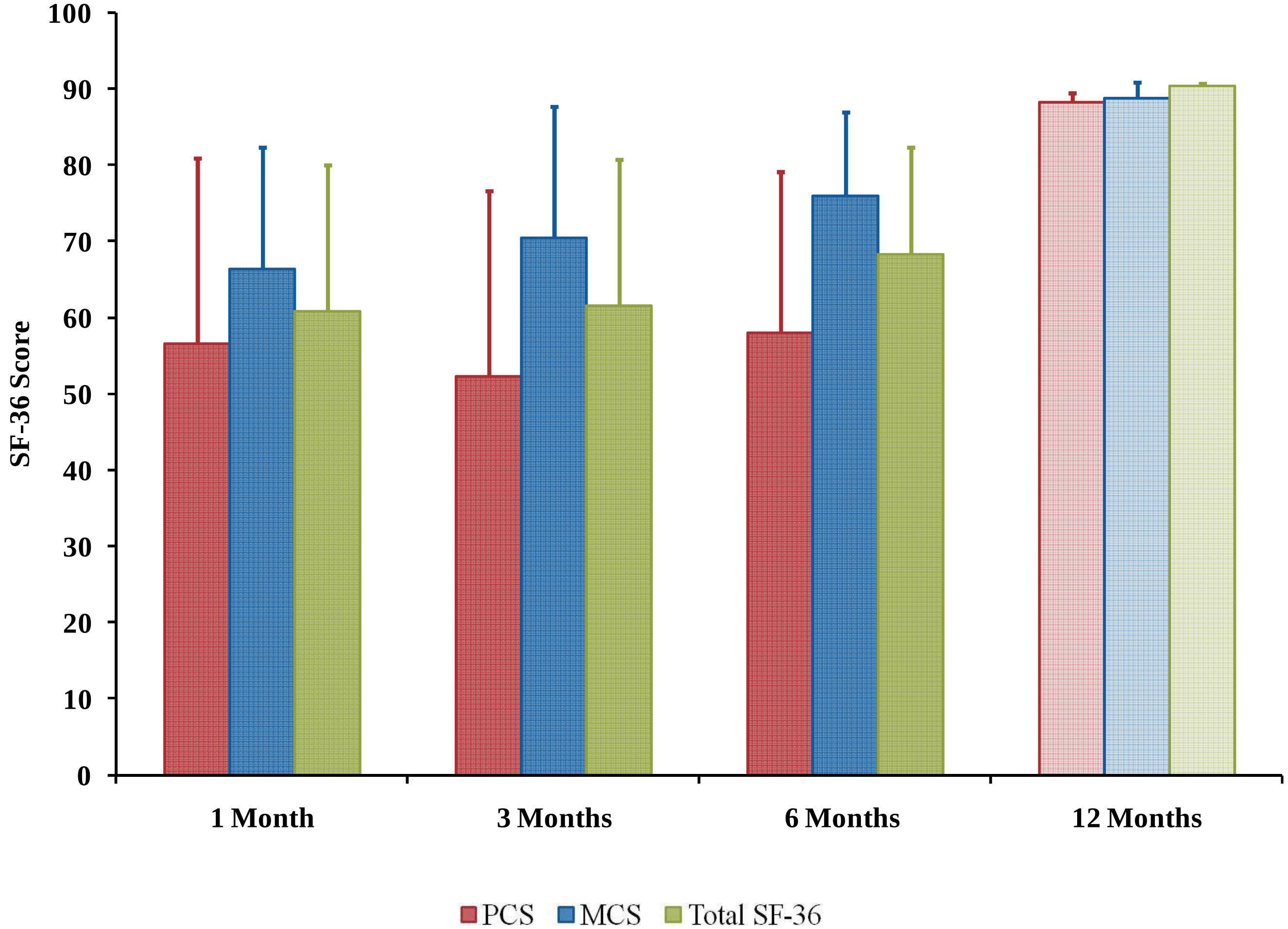


Figure 8.2 Group mean (±SD) Physical Component and Mental Component Summary scores and Total SF-36 score. Higher scores relate to increased QOL. Data at 12 months from n=2.

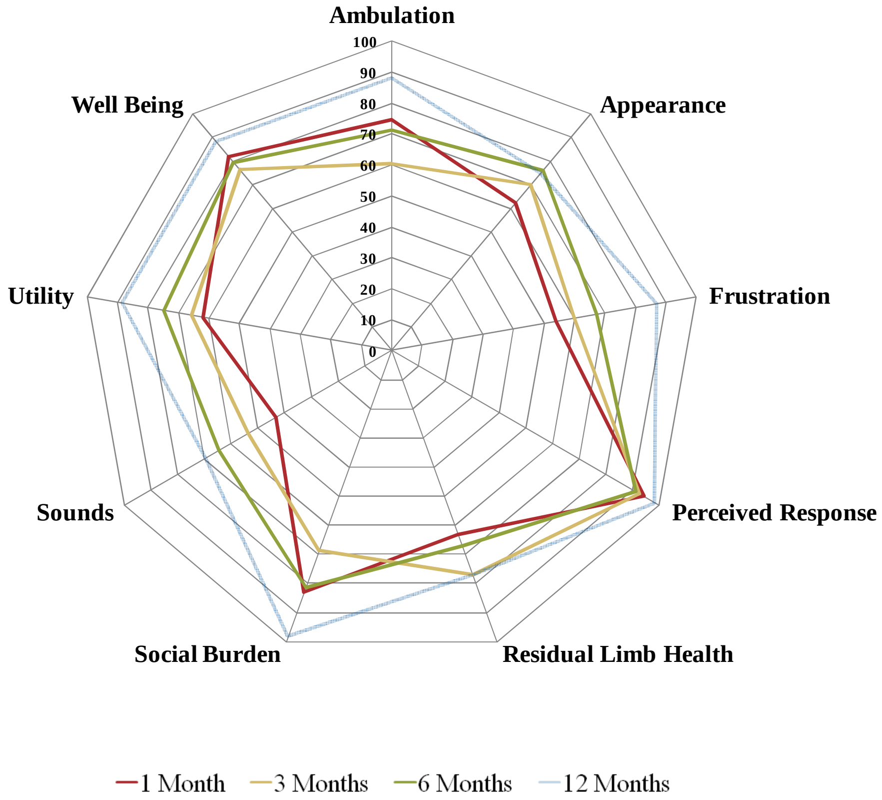


Figure 8.3 Target plot of group mean scores for the nine scales of the PEQ. Scores closer to the outer border of the plot relate to increased QOL in that scale. Scores closer to outer border of plot relate to a more positive response. Data at 12 months from n=2.

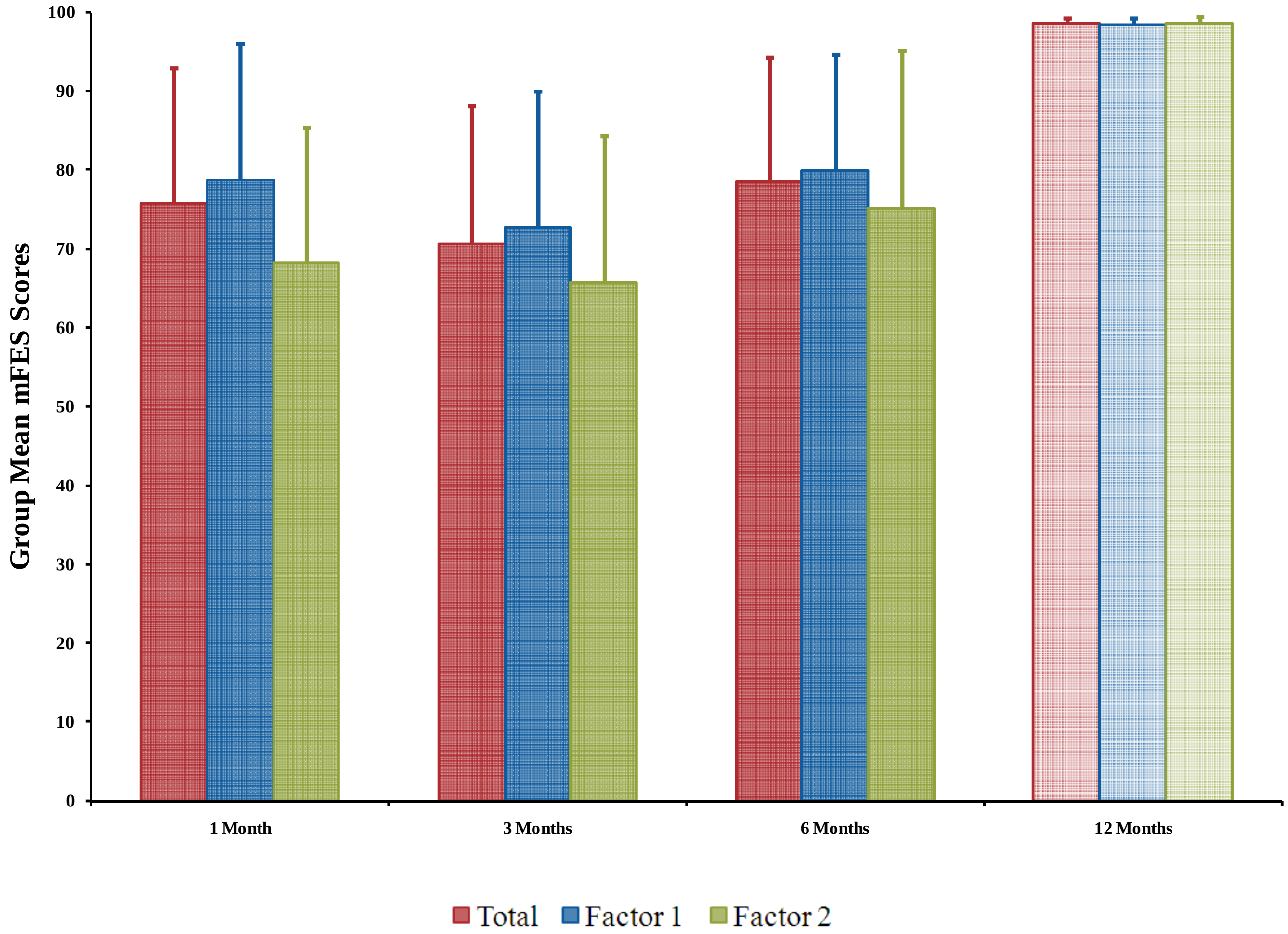


Figure 8.4 Group mean (±SD) total mFES, Factor One and Factor Two scores. Higher scores relate to increased falls efficacy. Data at 12 months from n=2.

Table 8.1 Statistical breakdown of SF-36, PEQ and mFES questionnaire responses. Results are reported (F value and significance level (P) from the linear mixed model.

SF-36	Time	
	F	P
Physical Functioning	(2, 6.15) = 2.26	0.18
Role Physical	(2, 8.08) = 0.25	0.79
Bodily Pain	(2, 3.78) = 2.40	0.21
General Health	(2, 10.91) = 0.98	0.41
Vitality	(2, 2.59) = 0.86	0.52
Social Functioning	(2, 30.32) = 2.37	0.11
Role Emotional	(2,9.98) = 3.39	0.07
Mental Health	(2, 10.90) = 0.42	0.67
Physical Component Summary Score	(2, 4.43) = 0.58	0.60
Mental Component Summary Score	(2, 4.95) = 2.10	0.22
Total SF-36	(2, 3.47) = 1.74	0.30
PEQ	**Time**	
	F	P
Ambulation	(2, 7.46) = 2.14	0.19
Appearance	(2, 8.11) = 4.24	0.06
Frustration	(2, 4.82) = 1.90	0.25
Perceived Response	(2, 13.85) = 0.22	0.80
Residual Limb Health	(2, 3.30) = 3.18	0.17
Social Burden	(2, 12.20) = 3.73	0.06
Sounds	(2, 3.74) = 1.43	0.35
Utility	(2, 4.39) = 1.93	0.25
Well Being	(2, 10.10) = 0.49	0.63
mFES	**Time**	
	F	P
Factor One	(2, 9.99) = 1.521	0.27
Factor two	(2, 29.52) = 1.84	0.18
Total mFES	(2, 9.07) = 1.60	0.25

8.4 Discussion

Scientific literature has reported various aspects of QOL in transtibial amputees (Asano *et al.,* 2008; Zidarov *et al.,* 2009; Van der Schans *et al.,* 2001; Pezzin *et al.,* 2000; Legro *et al.,* 1999). However, longitudinal assessment of how QOL develops following discharge from rehabilitation has not been investigated.

The current study had three aims, the first was to investigate the psychological changes that occurred in self-reported QOL during the six month period following discharge from rehabilitation. The second aim was to investigate the differences between mental and physical health during that same time frame. The third aim of the current study was to investigate the changes in falls efficacy following discharge from rehabilitation and the link between falls efficacy and measures of QOL.

Although QOL, as measured with the SF-36 seemed to improve in the six month period following amputation, none of the observable changes resulted in a statistically significant result. This offered support for the rejection of the first hypothesis, as did the highly insignificant result from the PCS score analysis. Findings from studies one and two along coupled with previous investigation of QOL in lower limb amputees during rehabilitation, reported a positive link between walking ability and QOL although it seemed this trend did not continue post-discharge from rehabilitation (Brooks *et al.,* 2001). Mental health, as represented by the MCS score, displayed what was likely a clinically significant improvement over time, albeit not statistically significant. However, MCS scores were generally higher than PCS scores, partially supporting the acceptance of the second hypothesis. This is also in agreement with previous reports of increased mental health when compared to physical health in lower limb amputees (Legro *et al.,* 1999; Pezzin *et al.,* 2000; Van der Schans *et al.,* 2002; Asano *et al.,* 2008; Zidarov *et al.,* 2009). In addition, scores from SF-36 scales pertaining to mental health were generally higher than those reported in age-matched normative data. The findings from the current study agree with previous reports (Asano *et al.,* 2008; Zidarov *et al.,* 2009), although contrasting reports show that this view in by no means comprehensive and would benefit from further investigation (Van der Schans *et al.,* 2001; Pezzin *et al.,* 2000; Legro *et al.,* 1999). Interestingly, the lack of statistically significant improvements in self-reported physical health may provide some support for the response phenomena hypothesis previously reported (Zidarov *et al.,* 2009). Gains in physical functioning following discharge from rehabilitation were observed in study three therefore, expectations with regards to future improvements may have been heightened. Thus, when reporting upon their physical health, participants may have reflected upon their current level, in relation to a level they were aiming to achieve. Even with improvements in physical health, the status quo may not have matched an individual's expectation, thus the self-reported physical health remains unchanged. Another interpretation could be acceptance on the part of the amputee that their physical functioning is decreased when compared to an able-bodied individual, as questions related to the general health (GH) scale required the amputees to reference their health state to other people. Thus reports of physical health are reduced, although mental health increases as the social and psychological impact of amputation decreases.

The lack of statistical significance observed from the SF-36 analyses could be hypothesised as being the result of a lack of sensitivity in the measurement tool. However, similar results were reported from the population specific questionnaire, the PEQ, where despite visible changes in scales, no statistically significant results were reported. Scales pertaining to amputees' prostheses tended to show greater improvement following discharge. This may be expected due to the stabilisation of the condition of the residuum, coupled with further adjustment of the prosthetic components and socket.

The perceived response of 'significant others' was consistently the most positive response score from the PEQ and did not change significantly over time, indicating that participants had good support from family and friends. Investigation of the effect of perceived response on reports of mental health would be interesting, as it may reveal this to be an important factor affecting lower limb amputees' mental health following discharge from rehabilitation. This would have implications for amputees that may not have the perceived social support observed in the current study group.

Overall falls efficacy did not change significantly over time and in this respect, matched results reported from QOL assessments. This partially supports acceptance of the third hypothesis and previous reports stating that falls efficacy is linked to QOL (Miller *et al.*, 2001b). Further to this, it could be hypothesised that falls efficacy is specifically linked to QOL in a physical sense, as previous reports have assessed QOL using the mobility subscale of the PEQ (Miller *et al.*, 2001b). Further analyses of participants' falls efficacy whilst undertaking indoor (Factor One) and outdoor (Factor Two) activities also displayed no significant changes over time with no discernable differences observed between the two factors. This may be indicative of participants improved mobility observed in study three and suggests that neither factor has an increased contribution to overall falls efficacy than the other. In addition, the current study reported that the pattern of falls efficacy to be linked more closely to the PCS score than the MCS score. However, this relationship would benefit from further detailed investigation.

8.4.1 Participants at 12 Months Post-Discharge

Data from two participants indicated an improvement in generic QOL between six and twelve months. This suggested that this may be an important period in an amputee's life following discharge from rehabilitation. Here, a noted improvement in physical health score seemed to be a significant contributor to overall QOL and may reflect potential physical gains that occurred from six months post-discharge. Reports of increases in generic QOL were matched by those from prosthesis related QOL which also improved from six months post-discharge, with two exceptions. Participant's perception of the appearance of the prosthesis did not change during this time period, perhaps as a result of consistent prosthetic components or amputees coming to terms with what their prosthesis looks like. Also, the highly scored perceived response of 'significant others' remained high up to one year following discharge, indicating the importance of social support during this time. Falls efficacy seemed to dramatically improve from six months post-discharge, perhaps again due to any physical gains during this time period. Similar to earlier reports during the year following discharge, there were no differences in falls efficacy when performing indoor vs. outdoor activities. This suggested that task difficulty, rather than the context in which the task is performed may be the pertinent factor for these participants.

8.5 Conclusion

The current study has provided an insight into how QOL develops once an individual is discharged from a programme of lower limb amputee rehabilitation. Despite observable and perhaps clinically meaningful changes in QOL, results from the current study indicated that, in general, QOL did not increase significantly over time. Therefore the first hypothesis was not supported. However, mental health was increased in comparison to physical health, as has been reported previously. This further supported the second hypothesis. These results suggest that, similar to study two, increases in physical health over time would be required to elicit further increases in overall QOL. Changes in overall falls efficacy was seen to be more closely linked to physical health than mental health and this would suggest that further increases in physical health over time may aid falls efficacy. However, this link was not clear enough to fully support the acceptance of the third hypothesis. Changes in overall falls efficacy were mirrored by the changes observed in falls efficacy during indoor (Factor One) and outdoor (Factor Two) activities. Neither factor seemed to contribute more than the other to overall falls efficacy. A lack of statistical power may have been the cause of the lack of significant findings. Therefore, studies employing increasing participant numbers may add weight to the results reported in the current study.

8.6 Chapter Eight – Study Five Summary of Findings

Title: Changes in Generic and Prosthesis Related Quality of Life and Falls Efficacy in Transtibial Amputees Following Discharge from Rehabilitation

Patients: Seven transtibial amputees (all men) recently discharged from rehabilitation. Mean ± SD Age 56.1 ± 14.9 years, height 1.82 ± 0.08 metres, mass 91.7 ± 11.4 kg.

Setting: Human performance laboratory.

Intervention: No intervention.

Comparison: Generic (SF-36) and prosthesis specific (PEQ) quality of life (QOL) and falls efficacy (mFES) in the six month period following discharge from rehabilitation.

Main Findings:	Description
Overall QOL	No statistically significant but perhaps clinically meaningful improvements in both generic and prosthesis related QOL.
Mental vs. physical health	Mental health was greater than physical health following discharge from rehabilitation.
Falls efficacy	No significant changes in falls efficacy were noted following discharge from rehabilitation.
Overall Summary	Observable and perhaps clinically meaningful increases in QOL were reported. Mental health was increased in comparison to physical health. The support of close family members was a key determinant of prosthesis related QOL. Changes in overall falls efficacy was seen to be more closely linked to physical health than mental health. This is relevant for clinicians as results suggested that further increases in physical health over time would be required to elicit further increases in overall QOL and falls efficacy.

SUMMARY – AMPUTEES POST REHABILITATION

Studies three, four and five investigated the adaptations in transtibial amputee level gait, performance of activities of daily living (ADL), balance ability and postural control as well as changes in quality of life (QOL) and falls efficacy. These participants were assessed in a six month period, following discharge from inpatient rehabilitation.

Results revealed that adaptations in level gait biomechanics occurred over time. However, despite increased affected limb function, inter-limb asymmetry was present in terms of joint kinetics, as reported in the literature (Sanderson and Martin, 1997; Vanicek *et al.,* 2007; Vanicek *et al.,* 2010).

All participants were able to cross the obstacle effectively (Hill *et al.,* 1997; Hill *et al.,* 1999; Hofstad *et al.,* 2006; Hofstad *et al.,* 2009; Vrieling *et al.,* 2007; Vrieling *et al.,* 2009). Participant's intact lead limb preference suggested that this limb was most beneficial to improving function both in terms control during swing phase and also during stance phase having made contact with the ground after crossing the obstacle.

Improvements were also reported in stepping down during gait, where participants utilised the intact limb during stance phase to lower the whole body centre of mass (COM) in preparation for affected limb stance phase and to propel the limb forward during swing, as has been reported during stair descent (Jones *et al.,* 2006; Schmalz *et al.,* 2007; Alimusaj *et al.,* 2009). This lead limb preference changed over time, which suggested that affected limb function improved as participants became more able to lower the whole body COM using the affected limb. During stepping up gait the intact limb lead preference enabled participants to use the intact limb to lift the COM to the raised surface and control the limb during swing to avoid tripping (Alimusaj *et al.,* 2009).

Participants overall balance ability during dynamic perturbation improved over time. However, similar to reports in literature, this balance ability was heavily reliant upon visual information (Isakov *et al.,* 1992; Vanicek *et al.,* 2009b). Participants achieved this by increasing the use of the ankle movements (ankle strategy) along with perceived attempts to increase somatosensory input from the affected limb.

In terms of postural control, participants were able to increase the spatial excursions of centre of gravity position (COG) and did so with more accuracy over time. However, temporal measures did not display any adaptation and hinted at a speed-accuracy trade off.

Participants QOL did not increase significantly over time, although mental health was increased in comparison to physical health, as has been reported previously (Legro *et al.,* 1999; Pezzin *et al.,* 2000; Van der Schans *et al.,* 2002; Asano *et al.,* 2008; Zidarov *et al.,* 2009). Changes in overall falls efficacy was more closely linked to physical health than mental health.

The results from these studies suggested that following discharge from rehabilitation, transtibial amputees had been able to further increase their physical functioning, balance ability and postural control. However, one common aspect in the performance of these tasks was the reliance on the intact limb to improve functioning. Although this may have been a necessary measure in order to improve function initially, better prosthetic components and rehabilitation techniques may reduce the long-term demands placed on the intact limb and the possible subsequent chronic limb degradation.

The lack of improvement in QOL over time may have reflected the ever increasing expectations and changing goals amputees had following discharge from rehabilitation. As falls efficacy was linked to self-reported physical health, improvements in physical functioning may aid transtibial amputees falls efficacy and wider psychological health.

9 CHAPTER NINE – SUMMARY, CLINICAL IMPLICATIONS, LIMITATIONS, FUTURE DIRECTIONS and CONCLUSIONS

9.1 Summary

The effect of lower limb amputation on an individual's gait, performance of activities of daily living (ADL), balance and postural control are well reported in the literature (McFadyen and Winter, 1988; Winter and Sienko, 1988; Powers *et al.,* 1997; Sanderson and Martin, 1997; Hill *et al.,* 1999; Vrieling *et al.,* 2008; Hofstad *et al.,* 2009; Vanicek *et al.,* 2009a; Vanicek *et al.,* 2009b; Vrieling *et al.,* 2009). In addition, the literature has also investigated quality of life (QOL) and falls efficacy in lower limb amputees (Legro *et al.,* 1999; Pezzin *et al.,* 2000; Asano *et al.,* 2008; Zidarov *et al.,* 2009).

However, the current study is the first to specifically investigate the biomechanical, balance and psychological adaptations that occur both during and following inpatient rehabilitation, with the implications for amputee rehabilitation outlined.

The overall aim of the current thesis was to investigate the longitudinal changes that occurred within unilateral transtibial amputees from their first treatments following amputation up to six months post-discharge from rehabilitation.

When re-learning how to walk during rehabilitation, two early walking aids (EWA) are routinely used in the UK. Chapter Four aimed to investigate the efficacy of transtibial amputees using an articulated vs. non-articulated EWA, along with the associated gait adaptations. During rehabilitation, patient's gait improved, although neither EWA proved to be beneficial, with most gait adaptations occurring upon receipt of patients' first functional prosthesis.

During the same time period, Chapter Five aimed to assess the changes in QOL and the subsequent effects of using different EWAs. Although QOL improved, mental health was better than physical health and there were no benefits of using one EWA over another. Results from Chapters Four and Five suggested that clinicians could select EWAs, without concern for subsequent gait ability or detrimental effects on QOL.

Following discharge from rehabilitation, lower limb amputees are likely to face more physically demanding tasks therefore, Chapters Six to Eight aimed to investigate the biomechanical, balance and QOL adaptations that occurred over a six month period post-discharge from rehabilitation. The biomechanical data reported that amputees increased functioning during this time period however, were heavily reliant upon the kinetic function of the intact limb to perform tasks successfully, particularly power generation at the ankle and power generation and absorption at the knee. The changes in lead limb preference during some ADLs were coupled with improvements in affected limb function, highlighting that over time, the affected limb contribution to overall functioning was increased.

Assessment of balance ability and postural control during the same time period in Chapter Seven found that amputees were able to maintain balance effectively, although were reliant upon visual information. Balance ability improved across time, with results suggesting that these changes were due to increasing the somatosensory information from the intact limb and better use of an ankle strategy during dynamic perturbations. Another interesting effect reported during Chapter Seven was that, when required to volitionally move the COG, participants increased the maximum excursion possible and accuracy of movements. However, the speed at which the task was performed did not change, hinting at a speed-accuracy trade off.

The tendency to rely upon the intact limb during gait, balance and ADLs, during the early stages following discharge from rehabilitation, further highlighted the need to improve affected limb function in order increase overall ability when performing these tasks.

Given the improved gait and balance function, it could be expected that increases in QOL and falls efficacy would occur, however results from Chapter Eight did not reveal such changes. This supported the hypothesis of a response phenomena, meaning amputees were expectant of further increases in functioning with reference to their current status.

9.2 Clinical Implications

The aims of the thesis were related to the investigation of the longitudinal biomechanical, balance and psychological adaptations that occurred in transtibial amputees. With this in mind, the following clinical recommendations are made based on the data presented.

9.2.1 Level Gait

- Initially, the goal of rehabilitation should shift the focus away from achieving symmetry and rather focus upon functional ability given that asymmetry seems to be an inherent feature of amputee movement that reduces over time.
- As neither AMA nor PPAM aid use during rehabilitation proved to be more beneficial in terms of gait or QOL, the selection process of an EWA should consider prioritising patient preference and cost-benefit to the NHS.
- Clinicians should consider prescribing additional home or therapy-based exercise programmes containing stretching exercises that target increasing muscle length and joint mobility, particularly in the affected limb, in order to increase joint range of motion (ROM).
- Continual assessment of muscular strength during rehabilitation may help to identify individual requirements.

- Targeted strengthening of the knee extensor musculature via exercise such as single limb squats using the affected limb should occur. Increased eccentric knee extensor strength may aid the control of the knee between the transition from single to double limb support particularly during loading response. Increased concentric knee extensor strength may aid knee power generation during mid-stance, thus reducing the kinetic asymmetry present.
- During rehabilitation, clinicians and consultants should consider early prescription of the functional prosthesis given that the most significant gait adaptations occurred upon receipt.

9.2.2 Obstacle Crossing

- Prosthetists should consider socket fit and the posterior shell of the functional prosthesis when prescribing limbs. This has been shown to be a limiting factor in affected limb knee ROM when crossing obstacles, increasing the risk of tripping and/or falling.
- Practice of obstacle crossing during rehabilitation is advocated, particularly leading with the non-preferred limb. The development of a lead limb preference enables amputees to cross obstacles effectively however, an unexpected obstacle may require the use of the non-preferred lead limb and subsequent movement pattern.
- Practice of crossing obstacles of varying dimensions and characteristics as well as expected and unexpected obstacles may further reduce the likelihood of tripping and/or falling.
- Increasing affected limb knee and hip joint ROM via stretching of the hip flexors will aid toe and heel clearance during swing phase when crossing obstacles.

9.2.3 Stepping Gait

- When stepping down to a new level, an affected limb lead preference is beneficial. Amputees are able to reduce the demands on musculature controlling the affected limb knee during stance while propelling the intact limb forwards during swing. In addition, the intact limb is able to manage the demands of lowering the body during stance. It is likely that if the lead limb makes contact with the ground or step during swing, the intact limb may be more able to recover than the affected limb.
- Caution must be taken when using the intact limb to increase stepping down gait velocity via propulsion of the intact lead limb during swing. Unless adequate control of the standing affected limb is achieved via knee extensor strength, there may be a risk that the limb collapses.
- Attempts should be made to increase affected limb power absorption at the hip and knee during single limb support via eccentric muscle training exercises such as single limb squats. This would allow the affected limb to act more effectively during stance phase when required to act as the trail limb when stepping down.
- Attempts should be made to increase affected limb power generation at the hip and knee during single limb support via exercises such as single limb raises and squats. When stepping up and leading with the affected limb, this would allow amputees to utilise this limb more effectively thus changing the lead limb preference and reducing the burden on the intact limb.

9.2.4 Balance Ability and Postural Control

- Practice of balance during dynamic perturbations may induce increases in overall balance ability. Such tasks may include balancing whilst on uneven surfaces, with varying frictional properties and made from materials of varying densities.

- Practicing balance tasks under dual tasking conditions may induce further increases in overall balance ability. This is more likely to reflect a real life situation, such as maintaining balance whilst completing a household activity.
- Safe practice of balance under reduced or no vision conditions may benefit overall balance ability, as amputees' dependence on this source of information to maintain balance is reduced. This may encourage greater use of somatosensory information from the residuum or increased sensitivity to vestibular information.
- Increasing joint flexibility and lower limb muscle strength may allow amputees to respond to dynamic perturbations more effectively.
- Amputees should be encouraged not to rely more heavily upon the affected limb than the intact limb in order to maintain balance. However, safely increasing amputees' ability to utilise the affected limb to maintain balance may benefit overall balance ability.
- Practice in volitionally displacing the centre of gravity (COG) may increase amputees' postural control and the speed at which control is regained following a perturbation. Regular use of a low-cost gaming console may induce these improvements.

9.2.5 Quality of Life and Falls Efficacy

- Clinicians are encouraged to regularly monitor QOL and falls related information. This would allow the rehabilitation team to identify if and when any further treatment interventions are required.
- The use of a population specific QOL questionnaire that is easily administered and interpreted is encouraged. This may aid clinicians to regularly monitor

changes in QOL both during and post-rehabilitation and tailor treatment accordingly.

9.3 Limitations

The inclusion and exclusion criteria set in all studies required amputees to have a certain level of functioning. By definition, these individuals may have been more physically able than other transtibial amputees. Therefore data from the current thesis particularly biomechanical data, may not have been completely representative of the wider transtibial amputee population and must be interpreted with this in mind. In favour of the current thesis were the ages of the amputees, representing individuals from the most common age group to experience transtibial amputation. Amputees in the current thesis were required to perform tasks without the use of walking aids e.g. walking sticks. While this was the case during data collection, amputees may have used walking aids outside of the research setting. If this was the case then results obtained within the empirical studies may not have represented amputees' typical movement patterns. Prosthetic components were not specifically controlled for as amputees attended the same prosthetic fitting clinic where very similar prosthetic limbs were prescribed. However, the few exceptions present may have influenced the data reported, with specific reference to ankle power generation and absorption. In addition, the inertial properties and modelling of the affected limb were not adjusted to take into account the altered mass of the prosthetic limbs or ankle and foot function. Although these must be acknowledged as limitations in the current thesis, this approach has been previously reported in the literature (Vickers *et al.,* 2008; Vanicek *et al.,* 2009a). It could also be argued that the modelling of the foot as a rigid segment in the current thesis was an accurate representation of the prosthetic feet observed.

All volunteers that participated in the current thesis completed their rehabilitation at the same centre. Therefore, it could be assumed that a level of parity was achieved in terms of the rehabilitation experienced. However, treatment is likely to differ between centres and not all centres will have facilities similar to those experienced by the amputees in the current thesis, in terms of equipment and personnel available. To an extent, the results reported and the recommendations made are specific to centres similar to the one attended by amputees in the current thesis. Cause of amputation was not controlled for within the current thesis. This may have been a confounding variable given that amputees secondary to vascular disease may have been less physically able than amputees secondary to trauma. It is probable that this lack of control will have affected the homogeneity of the groups of amputees assessed in the current thesis. Therefore, comparing results from the current thesis to those reported from studies exclusively investigating amputees secondary to trauma, may not be completely valid. However, the lack of control for cause of amputation is a common feature in transtibial amputee research and is likely to be a result of the difficulties in recruiting suitable volunteers from this population.

In the current thesis, the number of amputees taking part in the empirical studies was relatively low. This has an obvious impact on the statistical power of the studies, confirmed by some relatively large mean increases without the observation of statistical significance. In addition, this is likely to affect how confidently the results from the current thesis can be generalised and whether the amputees investigated in the current thesis were representative of the wider unilateral transtibial amputee population.

Assessment of amputee gait during rehabilitation increased the ecological validity of the results, however, maintaining a controlled environment was more difficult. Results from laboratory based studies possessed this control but may have lacked ecological validity. Each approach has strengths and weaknesses although both approaches are required to gain both a realistic and causative understanding of amputee movement. When assessing obstacle crossing in amputees, only one obstacle height was used. Although this was an ethical and safety requirement, amputees are likely to face obstacles of varying dimensions in everyday life, thus results from the current thesis would be applicable to amputees crossing obstacles of similar dimensions.

Balance assessment during the SOT and LOS test protocols was referenced against a theoretical maximum sway possible of 12.5 degrees (Nashner, 1997). Scores were reported on a scale of 0-100 (SOT) and as a percentage (%) of this theoretical maximum. However, if amputees' actual maximum sway was higher or lower than that set by the test protocol then the scores would require appropriate adjustment. Without this adjustment, inter-group and individual comparisons must be made with caution.

9.4 Future Directions

Future research would benefit from further consideration of lower limb amputee characteristics. Increasing participant numbers would provide studies with a more representative sample of the overall population thus increasing statistical power. Multi-centre recruitment may aid both sample sizes and also negate the effects of centre-specific treatment. Separating amputees by cause of amputation would also provide a valuable insight into the specific adaptations that may occur, given any variability in physical capacity from both an intra and inter individual perspective. Although transtibial amputees represent the most common level of lower limb amputation, investigation into amputees at the transfemoral level may improve the understanding of movement patterns in this population.

Although the current thesis compared the effects of using EWAs, the most relevant gait adaptations occurred upon receipt of an initial functional prosthesis. Future research may consider including a further group who are cast for and receive an initial functional prosthesis earlier in rehabilitation. This might increase the speed at which amputees progress through rehabilitation.

In terms of improving function in transtibial amputees, the focus of future research should be centred on the affected limb. Although the strength and flexibility of the affected limb were not directly measured, the biomechanical data from the current thesis suggested that these factors were reduced in the affected limb, when compared to the intact limb. Quantifying the effects of strength and flexibility training in affected limb musculature on the performance of gait, balance and the performance of ADLs would provide clinicians with information that may lead to more targeted treatment. It is not yet clear if amputees actively reduce the use of the affected limb or whether this is an unavoidable consequence of amputation. Studies investigating changes in the pain tolerances of amputees may help to clarify the suggestion of a protective mechanism with regards to the affected limb. Future research should also investigate amputee's ability to perform other ADLs such as turning and transitioning from sitting to standing, in order to identify where possible detriments in function may lie.

Similarly, future studies assessing balance ability and postural control in amputees should consider a number of interventions. Assessing the effects of an intervention incorporating practicing balance under challenging conditions on uneven surfaces of variable density and with altered visual conditions, may inform the practice of balance training in lower limb amputees. Also, there are a number of commercially available computer consoles that are designed to improve balance. These consoles tend to utilise a visual representation of an individual's COG as they perform a number of tasks designed to stress that individual's balance system. Future studies assessing the effects of using these consoles both during and after rehabilitation on transtibial amputee balance ability, postural control, falls efficacy and falls rate would have wide-ranging implications. These studies would provide clinicians with another tool by which to assess and improve amputee's balance performance during rehabilitation. Also, such consoles could be used by amputees to maintain balance ability having been discharged from rehabilitation. Assuming positive effects of console use, they have the potential to provide the NHS with large cost savings by reducing falls and fall related injuries.

The prosthesis evaluation questionnaire (PEQ) allows for the assessment of QOL in lower limb amputees. However, research should focus on the development of a shorter and more easily administered test instrument that may increase the levels of monitoring of QOL in lower limb amputees.

9.5 Conclusions

The current thesis provides an important addition to the currently available research by focussing upon the longitudinal biomechanical and psychological adaptations that occur in transtibial amputees. Currently, there are no reports in the scientific literature of these adaptations with the only published literature stemming from this thesis.

The current thesis has highlighted the progress in transtibial amputee's function during and following rehabilitation, the associated psychological changes with the integral role of the intact limb during gait and balance detailed. Based upon these results, a number of recommendations have been made regarding the treatment of transtibial amputees both during and following rehabilitation. In addition, further research directions have been suggested that will add to the greater understanding of how transtibial amputees move and the interventions that may further improve everyday function whilst reducing the risk of injury.

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APPENDIX A – Participant Information Sheet for studies one and two

PARTICIPANT INFORMATION SHEET

Comparison of Early Walking Aids

We wish to invite you to take part in a research study. Before you decide whether to do so, please read the following information carefully and discuss it with friends, relatives and your GP if you wish. Please ask if there is anything that is not clear or if you would like more information. You will be given as much time as you want to make a decision.

What is the purpose of this study?

Physiotherapists in the UK routinely use Early Walking Aids (EWAs) to help train people with a lower limb amputation to walk again. Of the two EWA's most commonly used in the UK one has a movable knee and the other does not and there is no evidence to say if one is better than the other. The study is to find out if a training leg with a movable knee has any benefits. In addition we want to see if there is any difference in quality of life between the two EWA's.

Why have I been invited?

You have been invited because you may receive an artificial limb to walk with in the future and would be expected to use an EWA as part of your normal rehabilitation. The EWA's we are studying are used with people who have had an amputation below the knee. 26 people will be recruited for the study

What will happen

You will attend for physiotherapy as usual. When you are ready to start to use an EWA you will be randomly selected to use either the one with the movable knee or one where the knee does not move. During your rehabilitation your walking will be timed over a 10-meter distance on five separate occasions.

In addition you would need to complete a questionnaire before your surgery (if possible) then 4 and 12 weeks after your amputation.

What do I have to do?

You would need to participate in a rehabilitation programme that would be the same if you did not take part in the study. In addition, on five separate occasions you would need to have your walking timed. You would also be required to complete the same questionnaire on three different occasions.

Do I have to take part?

Only if you want to. Participation is voluntary, you may refuse to participate or withdraw from the study at any time. But please let us know if you are unable fully to take part, as doing only parts of the study, rather than all of it, will affect the value of the research. You do not need to tell us 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?

No risks have been identified

Are there any costs involved?

No

Confidentiality

In order to meet legal obligations, a member of the research group may inspect your hospital records. Details of your treatment and your past relevant medical history as required for the study, will be recorded on a Case Record Form (CRF) the information from which will be entered onto computer in the Sports Science Department of the University of Hull. A CRF includes all information collected in the course of the research study. This information will be retained by research group and may be passed on to the authorised regulatory authorities.

The records will identify you only by a number (not your hospital number) and your initials. 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 until the data are analysed and for 2 years after the end of the study

In order to ensure that medical staff not involved with the study are aware of your participation in it, an alert notice will be attached to the cover of your hospital notes.

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 giving 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 Hull and East Yorkshire hospitals NHS Trust.

Trial-related injury

If 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 suffer from illness or injury during the study, or have any questions about the research study, please contact Amanda Hancock at Physiotherapy Department, Castle Hill hospital, Cottingham on 01482 875875 ext 3164.

APPENDIX B – Informed Consent Form for studies one and two

INFORMED CONSENT

Comparison of Early Walking Aids

Protocol number R0081

NAME OF LOCAL LEAD RESEARCHER:

SUBJECT ID or HOSPITAL NO:

Please initial box

1 I confirm that I have read and understand the information sheet dated 11.01.08 (version 6) 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 Hull and East Yorkshire Hospitals NHS Trust 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 Subject (BLOCK CAPITALS)	Date	Signature
Name of Person taking consent	Date	Signature
Researcher/witness	Date	Signature

APPENDIX C – Participant Information Sheet for studies three, four and five

PARTICIPANT INFORMATION SHEET

Gait and Balance in Unilateral Transtibial Amputees

We would like to invite you to take part in a research study. Before you decide, we would like you to understand why the research is being done and what it would involve for you.

Please take time to carefully read the following information and talk to others about the study if you wish.

If you are currently taking part in another research project then it is not suitable for you to volunteer for this one. Please inform Lynne Smith if this is the case.

Part 1 will tell you about the purpose of the study and what will happen if you decide to take part.

Part 2 gives you more detailed information about the conduct of the study.

Please ask us if there is anything that is not clear or if you would like more information.

We would like to know if you would like to take part in this research study. You have up to 3 weeks after being discharged to decide whether or not you would like to take part.

PART 1

What is the purpose of this study?

Lower limb amputees undertake physiotherapy treatment after surgery. It is known that following physiotherapy treatment and with practice, amputees are able to walk and move around their community. Many studies of lower limb amputees have assessed amputees with many years of experience of using their prosthesis. It is not yet known how lower limb amputees learn to walk and move around their community and if there are ways of helping them learn to do so. Therefore, the aim of this research study is to assess how the progress of walking and balance change in transtibial amputees over a one-year period.

Why have I been invited?

You have been invited to take part in this study as you have recently completed your course of physiotherapy treatment.

Do I have to take part?

No. Participation in this study is entirely voluntary.

If you do decide to take part in this study, you will be free to stop taking part at anytime without giving reason. This will not affect your care, your future treatment or your legal rights in any way.

What will happen if I decide to take part?

If you decide to take part in the study then great! You will then be invited to the Biomechanics Laboratory, at the University of Hull (don't worry, we don't wear white lab coats!). If you do not have your own transportation the University will be able to arrange some for you. You will be asked to bring along a pair of shorts, a t-shirt or vest and some comfortable shoes you can walk in, no high heels please!!! If you do not have shorts, they will be provided for you.

When you arrive, you will be asked to change into your shorts and t-shirt.

Reflective markers will be placed on your skin with double sided sticky tape. The markers are about the size of a marble, made of polystyrene and covered in reflective tape.

Once these markers are in place you will be asked to do some simple everyday tasks as follows:

- Walk in a straight line for 5 metres, turn around and walk back. You will do this up to 15 times.
- Walk in a straight line for 5 metres and step onto a raised surface, as though you are stepping onto a kerb. You will also do this up to 15 times.
- Walk in a straight line and step over and obstacle, a similar height to a kerb. This will be done up to 15 times.

The reflective markers are used to see how the limbs move while you are performing these tasks using motion capture cameras that see the light from the markers only. As the cameras do not see the person your identity is fully protected.

You will also be asked to stand on a special balance platform that can measure how you respond to movement underfoot:

- You will also be asked to stand still on a balance platform whilst the platform is stationary and also whilst it moves around. You will always wear a safety harness so that you will not fall.

Finally, you will be asked to fill out three questionnaires that may take you a small amount of time. These questions ask you about your quality of life, balance confidence and the use of your prosthesis.

Are there any costs involved?

No. The University will reimburse any costs that you incur as a result of travelling to the University at a standard University rate of 40p per mile travelled if coming by car. Your fare will be reimbursed if you come by train or taxi.

What do I have to do?

In order to take part in this study you will need to visit the Biomechanics Laboratory on four occasions at certain times during a 12 month period. This will be arranged between you and Mr Cleveland Barnett, who is organising the study. When you arrive the procedure is as described above, where you will perform certain walking and balance tasks whilst your movement is captured via reflective markers placed on your skin. You will also be asked to complete a number of questionnaires during each visit to the Biomechanics Laboratory. You may choose to rest whenever you wish. Each visit should last between 2 and 3 hours in total.

Please Inform Lynne Smith, Physiotherapist at Castle Hill Hospital 01482875875 ext 3164, if you are taking part in any other research studies. If you are taking part in any other research projects then it is not suitable for you to take part in this one.

Are there any risks involved?

It is extremely rare but one possible side effect of sticky tape being placed on the skin is a skin reaction to the tape. Your skin will be checked when the markers have been removed and, if there has been any reaction, appropriate treatment would be recommended.

The correct health and safety measures are taken at all times in the Biomechanics Laboratory. On the balance platform you may feel as though you are going to fall….however the safety harness you are strapped into will prevent this!!! Whilst performing the walking tasks you will not be asked to perform any tasks you feel are not within your capabilities.

What happens when the research study stops?

The results from the study will be published in scientific and amputee therapy publications as well as being submitted for an educational qualification. You will not be identified in any of this material to preserve your confidentiality. You may request a copy of any published results from Mr Cleveland Barnett.

What if there is a problem?

Any complaint about the way you have been dealt with during the study or any possible harm you might suffer will be addressed. Please contact Vicki Russell, Limb Unit Manger (01482 211143) if this is the case. Also, you may wish to contact Nina Dunham, Research and Development Manager (01482 623206) for independent advice on taking part in this study.

If the information in **Part 1** has interested you and you are considering taking part in the study, please read on to **Part 2** for additional details.

PART 2

Confidentiality

All information and data from the study will be kept strictly confidential. Your name and details will not be disclosed at any time and you will be assigned a code number to identify you in the study. All data and information will be kept on record electronically on a password protected computer and in locked filling cabinets.

Mr Cleveland Barnett has responsibility to safeguard the data and information and only those individuals involved with the study will have access to these sources.

All data and information will be kept at the University of Hull for the duration of the study, which concludes on 31/10/2009, although you will not be involved for that amount of time.

Please be aware that, when giving consent to participate, you are agreeing with the conditions outlined above.

Your Rights

Your participation in this study is voluntary. You are allowed to withdraw from the study at any time without reason. This will not affect any future treatment, or any legal rights. Withdrawal is totally without prejudice.

For more advice on the project please contact Mr Cleveland Barnett, 01482465106 or email C.Barnett@hull.ac.uk.

For any impartial advice on taking part in a research study please contact Nina Dunham, Research and Development Manager (01482 623206).

Trial-Related Injury

It is unlikely that you will experience an injury or illness as a result of taking part in this research study. However, indemnity is provided by the University of Hull and any compensation will be as per the University's usual standards. For more information please contact Mr Cleveland Barnett.

Who is organising the study?

Mr Cleveland Barnett, Department of Sport, Health and Exercise Science.

Thank you for your time and I look forward to speaking to you soon.

Mr Cleveland Barnett

Department of Sport, Health and Exercise Science

The University of Hull

APPENDIX D – Informed Consent Form for studies three, four and five

Centre Number: Department of Sport, Health and Exercise Science, University of Hull

Study Number: 08/H1304/10

Patient Identification Number for this trial:

INFORMED CONSENT FORM

<u>Gait and Balance in Unilateral Transtibial Amputees</u>

<u>Mr Cleveland Barnett</u>

Please Initial In the Box

1. I confirm that I have read and understand the information sheet dated…………………………...
(version 1.1) for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily. ☐

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 relevant sections of my medical notes and data collected during the study may be looked at by individuals from the University of Hull, Sport Health and Exercise Department and the Physiotherapy Department, Castle Hill Hospital, from regulatory authorities or from the NHS Trust, where it is relevant to my taking part in this research. I give permission for these individuals to have access to my records. ☐

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

______________________	______________________	______________
Name of Patient	Date	Signature

______________________	______________________	______________
Name of Person Taking Consent	Date	Signature

APPENDIX E – Short-Form 36 Questionnaire (Ware and Sherbourne 1992)

INSTRUCTIONS: This set of questions asks for your views about your health. This information will help keep track of how you feel and how well you are able to do your usual activities. Answer every question by marking the answer as indicated. If you are unsure about how to answer a question please give the best answer you can.

1. In general, would you say your health is:

1. Excellent 2. Very good 3. Good 4. Fair 5. Poor

2. <u>Compared to ONE YEAR AGO</u>, how would you rate your health in general <u>NOW</u>?

1. MUCH BETTER than one year ago.
2. Somewhat BETTER now than one year ago.
3. About the SAME as one year ago.
4. Somewhat WORSE now than one year ago.
5. MUCH WORSE now than one year ago.

3. The following items are about activities you might do during a typical day. **Does your health now limit you** in these activities? If so, how much?

Activities	**1. Yes, Limited A Lot**	**2. Yes, Limited A Little**	**3. No, Not Limited At All**
a) **Vigorous activities,** such as running, lifting heavy objects, participating in strenuous sports?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
b) **Moderate activities,** such as moving a table, pushing a vacuum cleaner, bowling, or playing golf?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
c) Lifting or carrying groceries?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
d) Climbing **several flights** of stairs?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
e) Climbing **one** flight of stairs?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
f) Bending, kneeing or stooping?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
g) Walking **more than a mile**?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
h) Walking **several** blocks?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
i) Walking **one** block?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all
j) Bathing or dressing yourself?	1. Yes, limited a lot	2. Yes, limited a little	3. No, not limited at all

4. During the **past 4 weeks**, have you had any of the following problems with your work or other regular activities *as a result of your physical health*?

	Yes	No
a) Cut down on the **amount of time** you spent on work or other activities?	1. yes	2. No
b) **Accomplished less** than you would like?	1. yes	2. No
c) Were limited in the **kind** of work or other activities?	1. yes	2. No
d) Had **difficulty** performing the work or other activities (for example it took extra effort)?	1. yes	2. No

5. During the **past 4 weeks**, have you had any of the following problems with your work or other regular daily activities **as a result of any emotional problems** (such as feeling depressed or anxious)?

	Yes	No
a) Cut down on the **amount of time** you spent on work or other activities?	1. yes	2. No
b) **Accomplished less** than you would like?	1. yes	2. No
c) Didn't do work or other activities as **carefully** as usual?	1. yes	2. No

*6. During the **past 4 weeks**, to what extent has your physical health or emotional problems interfered with your normal social activities with family, friends, neighbours, or groups?*

1. Not at all 2. Slightly 3. Moderately 4. Quite a bit 5. Extremely

7. How much **bodily pain** have you had during the **past 4 weeks**?

1. None 2. Very mild 3. Mild 4. Moderate 5. Severe 6. Very severe

8. During the **past 4 weeks**, how much did **pain** interfere with your normal work (including both work outside the home and housework)?

1. Not at all 2. A little bit 3. Moderately 4. Quite a bit 5. Extremely

9. These questions are about how you feel and how things have been with you ***during the past 4 weeks****. For each question, please give the one answer that comes closest to the way you have been feeling. How much of the time during the* ***past 4 week*** *…*

	1. All of the time	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
a) Did you feel full of pep?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
b) Have you been a very nervous person?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
c) Have you felt so down in the dumps that nothing could cheer you up?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
d) Have you felt calm and peaceful?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
e) Did you have a lot of energy?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
f) Have you felt downhearted and blue?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
g) Do you feel worn out?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
h) Have you been a happy person?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*
i) Did you feel tired?	*1. All of the time*	*2. Most of the time*	*3. A good bit of the time*	*4. Some of the time*	*5. A little of the time*	*6. None of the time*

*10. During the **past 4 weeks**, how much of the time has your **<u>physical health</u> or <u>emotional problems</u>** interfered with your social activities (like visiting with friends, relatives, etc.)?*

1. All of the time
2. Most of the time.
3. Some of the time
4. A little of the time.
5. None of the time.

*11. How TRUE or FALSE is **<u>each</u>** of the following statements for you?*

	1. Definitely true	*2. Mostly true*	*3. Don't know*	*4. Mostly false*	*5. Definitely false*
a) I seem to get sick a little easier than other people?	*1. Definitely true*	*2. Mostly true*	*3. Don't know*	*4. Mostly false*	*5. Definitely false*
b) I am as healthy as anybody I know?	*1. Definitely true*	*2. Mostly true*	*3. Don't know*	*4. Mostly false*	*5. Definitely false*
c) I expect my health to get worse?	*1. Definitely true*	*2. Mostly true*	*3. Don't know*	*4. Mostly false*	*5. Definitely false*
d) My health is excellent?	*1. Definitely true*	*2. Mostly true*	*3. Don't know*	*4. Mostly false*	*5. Definitely false*

ITEMS	SCALES	DIMENSIONS		TOTAL
3. Vigorous activities	Scale 1. Physical Functioning (PF)	Dimension A. PHYSICAL HEALTH		TOTAL SF-36
4. Moderate activities				
5. Lift, carry groceries				
6. Climb several flights				
7. Climb one flight				
8. Bend, kneel				
9. Walk mile				
10. Walk half a mile				
11. Walk 100 yards				
12. bathe, dress				
13. Cut down time	Scale 2. Role-Physical (RP)			
14. Accomplished less				
15. limited in kind				
16. Had difficulty				
21. Pain magnitude	Scale 3. Bodily Pain (BP)			
22. Pain interfere				
1.General health rating	Scale 4. General Health (GH)		Dimension B. MENTAL HEALTH	
36. Excellent				
34. As healthy as anyone				
33. Ill easier				
35. Health worse				
23. Full of life	Scale 5. Vitality (VT)			
27. Energy				
29. Worn out				
31. Tired				
32. Social extent	Scale 6. Social Functioning (SF)			
20. Social time				
17. Cut down time	Scale 7. Role Emotional (RE)			
18. Accomplished less				
19. Not careful				
24. Nervous	Scale 8. Mental Health (MH)			
25. Down in dumps				
26. Peaceful				
28. Low/sad				
30. Happy				
2. Change in reported health				

Figure E.0.1 Abbreviated 36 items of the SF-36 questionnaire with associated eight scales and two dimensions. Adapted from Kalantar-Zadeh *et al.*, (2001).

APPENDIX F – Prosthesis Evaluation Questionnaire (Legro *et al.*, 1998).

Table F.0.1 Abbreviated content of 42 items of the PEQ with the associated nine scales. *Item scored as 100 if box checked, **Item scored as 'no response' if box checked.

SCALE NAME	ITEM NUMBER	ITEM CONTENT
Ambulation (AM)	13A	Rate your ability to walk when using your prosthesis.
	13B	Rate your ability to walk in close spaces using your prosthesis.
	13C	Rate your ability to walk up stairs when using your prosthesis.
	13D	Rate how you felt about being able to walk down stairs when using...
	14E	Rate your ability to walk up a steep hill when using your prosthesis.
	14F	Rate your ability to walk down a steep hill when using your prosthesis.
	14G	Rate your ability to walk on sidewalks and streets when using your...
	14H	Rate your ability to walk on slippery surfaces (e.g. wet tile, snow...
Appearance (AP)	3J	Rate how your prosthesis has looked.
	3M	Rate the damage done to your clothing by your prosthesis.
	3N**	Rate the damage done to your prosthesis cover.
	4O	Rate your ability to wear the shoes (different heights, styles) you...
	4P	Rate how limited your choice of clothing was because of your...
Frustration (FR)	10B	How frequently were you frustrated with your prosthesis.
	10C*	If you were frustrated with your prosthesis at any time over the past...
Perceived Response (PR)	10A	Rate how often the desire to avoid stranger's reactions to your...
	11D**	Rate how your partner has responded to your prosthesis.
	11E**	Rate how this response has affected your relationship.
	11G**	Rate how Family Member #1 has responded to your prosthesis
	12H**	Rate how Family Member #2 has responded to your prosthesis
Residual Limb Health (RL)	4Q	Rate how much you sweat inside your prosthesis (in the sock, liner...
	4R	Rate how smelly your prosthesis was at its worst
	4S	Rate how much of the time your residual limb was swollen to the...
	5T*	Rate any rash(es) that you got on your residual limb
	5U*	Rate any ingrown hairs (pimples) that you got on your residual limb.
	5V*	Rate any blisters or sores that you got on your residual limb.
Social Burden (SB)	12I**	Rate how much of a burden your prosthesis has been on your partner...
	12J	Rate how much having your prosthesis has hindered you socially.
	12K**	Rate your ability to take care of someone else, (e.g. your partner...
Sounds (SO)	3K	Rate how often your prosthesis made squeaking, clicking or belching...
	3L*	If it made any sounds in the past four weeks, rate how bothersome...
Utility (UT)	1B	Rate the fit of your prosthesis.
	1C	Rate the weight of your prosthesis.
	1D	Rate your comfort whilst standing when using your prosthesis.
	2E	Rate your comfort whilst sitting when using your prosthesis.
	2F	Rate how often you felt off balance while using your prosthesis.
	2G	Rate how much energy it took to use your prosthesis for as long as...
	2H	Rate the feel, such as the temperature and texture of the prosthesis...
	2I	Rate the ease of putting on (donning) your prosthesis.
Well-Being (WB)	16C	Rate how satisfied you have been with how things have worked out...
	16D	How would you rate your quality of life.

Instructions

As you read each question, remember there is no right or wrong answer. Just think of YOUR OWN OPINION on the topic and make a mark THROUGH the line anywhere along the line from one end to the other to show us your opinion.

If you use different prostheses for different activities, please choose the ONE you use more often and answer all the questions as though you were using that prosthesis.

Example

How important is it to you to have coffee in the morning?

NOT AT ALL EXTREMELY IMPORTANT

Over **the past four weeks, rate your morning coffee.**

TERRIBLE **EXCELLENT**

OR check __I haven't drunk coffee in the morning in the past four weeks.

This example shows that the person who answered these questions feels that having coffee in the morning is important to him. He also thinks the coffee he has had lately has not been very good.

If he hadn't drunk any coffee in the last four weeks, he would have put a check by that statement instead of putting a mark on the line between TERRIBLE and EXCELLENT.
As in this example, make a mark across the line rather than using an X or an O. Please answer all the questions.

Support for development of the PEQ was provided by the U.S. Department of Veterans Affairs.

These first questions are about YOUR PROSTHESIS.

A. Over the past four weeks, rate how happy you have been with your current prosthesis.

EXTREMELY UNHAPPY EXTREMELY HAPPY

B. Over the past four weeks, rate the fit of your prosthesis.

TERRIBLE EXCELLENT

C. Over the past four weeks, rate the weight of your prosthesis.

TERRIBLE EXCELLENT

D. Over the past four weeks, rate your comfort while standing *when using your prosthesis.*

TERRIBLE EXCELLENT

E. Over the past four weeks, rate your comfort while sitting *when using your prosthesis.*

TERRIBLE EXCELLENT

F. Over the past four weeks, rate how often you felt off balance *while using your prosthesis.*

ALL THE TIME NOT AT ALL

G. Over the past four weeks, rate how much energy it took to use your prosthesis for as long as you needed it.

COMPLETELY EXHAUSTING NONE AT ALL

H. Over the past four weeks, rate the feel (such as the temperature and texture) of the prosthesis (sock, liner, socket) on your residual limb (stump).

WORST POSSIBLE BEST POSSIBLE

I. Over the past four weeks, rate the ease of putting on (donning) your prosthesis.

TERRIBLE EXCELLENT

J. Over the past four weeks, rate how your prosthesis has looked.

TERRIBLE EXCELLENT

K. Over the past four weeks, rate how often your prosthesis made squeaking, clicking, or belching sounds.

ALWAYS NEVER

L. *If it made any sounds* in the past four weeks, rate how bothersome these sounds were to you.

EXTREMELY BOTHERSOME NOT AT ALL

OR check ____It made no sounds.

M. Over the past four weeks, rate the damage done to your clothing by your prosthesis.

EXTENSIVE DAMAGE NONE

N. Over the past four weeks, rate the damage done to your prosthesis cover.

EXTENSIVE DAMAGE NONE

OR check ____There is no cover on my prosthesis.

O. Over the past four weeks, rate your ability to wear the shoes (different heights, styles) you prefer.

CANNOT NO PROBLEM

P. Over the past four weeks, rate how limited your choice of clothing was because of your prosthesis.

WORST POSSIBLE NOT AT ALL

Q. Over the past four weeks, rate how much you sweat inside your prosthesis (in the sock, liner, socket).

EXTREME AMOUNT NOT AT ALL

R. Over the past four weeks, rate how smelly your prosthesis was at its worst.

EXTREMELY SMELLY NOT AT ALL

S. Over the past four weeks, rate how much of the time your residual limb was swollen to the point of changing the fit of your prosthesis.

ALL THE TIME NEVER

T. Over the past four weeks, rate any rash(es) that you got on your residual limb.

EXTREMELY BOTHERSOME NOT AT ALL

OR check __I had no rashes on my residual limb in the last month.

U. Over the past four weeks, rate any ingrown hairs (pimples) that were on your residual limb.

EXTREMELY BOTHERSOME NOT AT ALL

OR check __I had no ingrown hairs on my residual limb in the last month.

V. Over the past four weeks, rate any blisters or sores that you got on your residual limb.

EXTREMELY BOTHERSOME NOT AT ALL

OR check __I had no blisters or sores on my residual limb in the last month.

Group 2

The *next section covers very SPECIFIC BODILY SENSATIONS. Here are our definitions:*

1. *SENSATIONS* are feelings like "pressure", "tickle" or a sense of position or location, such as the toes being curled. Amputees have described sensations in their missing (phantom) limb such as "the feeling that my (missing) foot is wrapped in cotton."

2. *PAIN* is a more extreme sensation described by terms such as "shooting", "searing", "stabbing", "sharp", or "ache".

3. *PHANTOM LIMB* refers to the part that is missing. People have reported feeling sensations and/or pain in the part of the limb that has been amputated — that is, in their phantom limb.

4.*RESIDUAL LIMB (STUMP)* refers to the portion of your amputated limb that is still physically present.

REGARDING SENSATIONS IN YOUR PHANTOM LIMB

A. Over the past four weeks, rate how often you have been aware of non-painful sensations in your phantom limb.
a. ____never
b. _____only once or twice
c. ____a few times (about once/week)
d. _____fairly often (2-3 times/week)
e. ____very often (4-6 times/week)
f. _____several times every day
g. _____all the time or almost all the time

B. If you had non-painful sensations in your phantom limb during the past month, rate how intense they were on average.

EXTREMELY INTENSE EXTREMELY MILD

OR check ___I did not have non-painful sensations in my phantom limb.

C. Over the past month, how bothersome were these sensations in your phantom limb?

ALL THE TIME NEVER

OR check_I did not have non-painful sensations in my phantom limb.

REGARDING PAIN IN YOUR PHANTOM LIMB

D. Over the past four weeks, rate how often you had pain in your phantom limb.
a. ____never
b. _____only once or twice
c. ____a few times (about once/week)
d. ____fairly often (2-3 times/week)
e. ____very often (4-6 times/week)
f. _____several times every day
g. ____all the time or almost all the time

E. How long does your phantom limb pain usually last?
a. _____I have none
b. _____a few seconds
c. _____a few minutes
d. _____several minutes to an hour e. _____several hours
f. _____a day or two
g. ____more than two days

F. If you had any pain in your phantom limb this past month, rate how intense it was on average.

EXTREMELY INTENSE EXTREMELY MILD

OR check ___I did not have any pain in my phantom limb.

G. In the past four weeks how bothersome was the pain in your phantom limb?

EXTREMELY BOTHERSOME EXTREMELY MILD

OR check ____I did not have any pain in my phantom limb.

REGARDING PAIN IN YOUR RESIDUAL LIMB (STUMP)

H. Over the past four weeks, rate how often you had pain in your residual limb.
a. ____never
b. _____only once or twice
c. _____a few times (about once/week)
d. ____fairly often (2-3 times/week)
e. ____very often (4-6 times/week)
f. _____several times every day
g. _____all the time or almost all the time

I. If you had any pain in your residual limb over the past four weeks, rate how intense it was on average.

EXTREMELY INTENSE EXTREMELY MILD

OR check ____I did not have any pain in my residual limb.

J. OVER THE past four weeks how bothersome was the pain in your residual limb?

EXTREMELY BOTHERSOME NOT AT ALL

OR check ____I did not have any pain in my residual limb.

REGARDING PAIN IN YOUR OTHER (NON-AMPUTATED) LEG OR FOOT

K. Over the past four weeks, rate how often you had pain in your other leg or foot.
a. ____never
b. _____only once or twice
c. _____a few times (about once/week)
d. _____fairly often (2-3 times/week)
e. _____very often (4-6 times/week)
f. _____several times every day
g. _____all the time or almost all the time

L. If you had any pain in your other leg or foot over the past four weeks, rate how intense it was on average.

EXTREMELY INTENSE EXTREMELY MILD

OR check ____I had no pain in my other leg or foot.

M. OVER THE past four weeks how bothersome was the pain in your other leg or foot?

EXTREMELY BOTHERSOME NOT AT ALL

OR check ____I had no pain in my other leg or foot.

REGARDING BACK PAIN

N. Over the past four weeks, rate how often you experienced back pain.
a. _____never
b. _____only once or twice
c. _____a few times (about once/week)
d. _____fairly often (2-3 times/week)
e. _____very often (4-6 times/week)
f. _____several times every day
g. _____all the time or almost all the tune

O. If you had any back pain over the past four weeks, rate how intense it was on average.

EXTREMELY INTENSE EXTREMELY MILD

OR check ____I had no back pain.

P. OVER THE past four weeks how bothersome was the back pain?

EXTREMELY BOTHERSOME NOT AT ALL

OR check ____I had no back pain.

Group 3

This section is about some of the SOCIAL AND EMOTIONAL ASPECTS OF USING A PROSTHESIS.

A. Over the past four weeks, rate how often the desire to avoid strangers' reactions to your prosthesis made you avoid doing something you otherwise would have done.

ALL THE TIME NEVER

B. Over the past four weeks, rate how frequently you were frustrated with your prosthesis.

ALL THE TIME NEVER

C. If you were frustrated with your prosthesis at any time over the past month, think of the most frustrating event and rate how you felt at that tune.

EXTREMELY FRUSTRATED NOT AT ALL

OR check ___I have not been frustrated with my prosthesis.

We understand that sometimes you will have both positive and negative experiences with those close to you. Please try to answer these questions considering all the reactions you have had.

D. Over the past four weeks, rate how your partner has responded to your prosthesis

VERY POORLY VERY WELL

OR check __I don't have a partner.

E. Over the past four weeks, rate how this response has affected your relationship.

VERY BADLY VERY WELL

OR check ____I don't have a partner.

F. Think of two close family members (other than your partner) and write down their relationship to you, like mother or son.

#1 ____ #2 ____

OR check ___I don't have any close family members.

G. Over the past four weeks, rate how Family Member #1 has responded to your prosthesis

VERY POORLY VERY WELL

OR check ___I don't have close family members.

H. Over the past four weeks, rate how Family Member #2 has responded to your prosthesis.

VERY POORLY VERY WELL

OR check ____I don't have a second close family member.

I. Over the past four weeks, rate how much a burden your prosthesis has been on your partner or family members.

EXTREMELY BURDENSOME NOT AT ALL

OR check ____I don't have a partner or family members.

J. Over the past four weeks, rate how much having your prosthesis has <u>hindered</u> you socially.

A GREAT DEAL NOT AT ALL

K. Over the past four weeks, rate your ability to take care of someone else, (e.g. your partner, a child, or a friend).

CANNOT NO PROBLEM

OR check ___I don't take care of someone else.

Group 4

This section is about YOUR ABILITY TO MOVE AROUND.

A. Over the past four weeks, rate your ability to walk *when using your prosthesis.*

CANNOT NO PROBLEM

B. Over the past four weeks, rate your ability to walk hi close spaces *when using your prosthesis.*

CANNOT NO PROBLEM

C. Over the past four weeks, rate your ability to walk up stairs *when using your prosthesis.*

CANNOT NO PROBLEM

D. Over the past four weeks, rate how you have felt about being able to walk down stairs *when using your prosthesis.*

CANNOT NO PROBLEM

E. Over the past four weeks, rate your ability to walk up a steep hill *when using your prosthesis.*

CANNOT NO PROBLEM

F. Over the past four weeks, rate your ability to walk down a steep hill *when using your prosthesis.*

CANNOT NO PROBLEM

G. Over the past four weeks, rate your ability to walk on sidewalks and streets *when using your prosthesis.*

CANNOT NO PROBLEM

H. Over the past four weeks, rate your ability to walk on slippery surfaces (e.g. wet tile, snow, a rainy street, or a boat deck) *when using your prosthesis.*

CANNOT NO PROBLEM

I. Over the past four weeks, rate your ability to get in and out of a car *when using your prosthesis.*

CANNOT NO PROBLEM

J. Over the past four weeks, rate your ability to sit down and get up from a chair with a high seat (e.g., a dining chair, a kitchen chair, an office chair).

CANNOT NO PROBLEM

K. Over the past four weeks, rate your ability to sit down and get up from a low or soft chair (e.g. an easy chair or deep sofa).

CANNOT NO PROBLEM

L. Over the past four weeks, rate your ability to sit down and get up from the toilet.

CANNOT NO PROBLEM

M. Over the past four weeks, rate your ability to shower or bathe safely.

CANNOT NO PROBLEM

The following section asks about YOUR SATISFACTION WITH PARTICULAR SITUATIONS given that you have an amputation.

A. Over the past four weeks, rate how satisfied you have been with your prosthesis.

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

B. Over the past four weeks, rate how satisfied you have been with how you are walking.

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

C. Over the past four weeks, rate how satisfied you have been with how things have worked out since your amputation.

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

D. Over the past four weeks, how would you rate your quality of life?

WORST POSSIBLE LIFE BEST POSSIBLE LIFE

E. How satisfied are you with the person who fit your current prosthesis?

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

F. How satisfied are you with the training you have received on using your current prosthesis?

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

OR check _ I have not had any training with my current prosthesis.

G. Overall, how satisfied are you with the gait and prosthetic training you have received since your amputation.

EXTREMELY DISSATISFIED EXTREMELY SATISFIED

OR check _ I have not had any training since my amputation.

Group 6

This next section asks you to rate your ability TO DO YOUR DAILY ACTIVITIES when you are having problems with your prosthesis.

A. When the fit of my prosthesis is poor, I will get...

NOTHING DONE EVERYTHING DONE

B. When the comfort of my prosthesis is poor, I will get...

NOTHING DONE EVERYTHING DONE

C. Without my prosthesis, I will get...

NOTHING DONE EVERYTHING DONE

Group 7

This last section asks you to rate HOW IMPORTANT different aspects (or qualities) of your prosthesis are to you.

A. How important is it that the weight of your prosthesis feel right?

NOT AT ALL EXTREMELY IMPORTANT

B. How important is the ease of putting on (donning) your prosthesis?

NOT AT ALL EXTREMELY IMPORTANT

C. How important is the appearance of your prosthesis (how it looks)?

NOT AT ALL EXTREMELY IMPORTANT

D. How important is it to you to be able to wear different kinds of shoes (heights or styles)?

NOT AT ALL EXTREMELY IMPORTANT

E. How important is it that your prosthesis' covering is durable (cannot be torn, dented, easily scratched, or discolored)?

NOT AT ALL EXTREMELY IMPORTANT

OR check __There is no covering on my prosthesis.

F. How bothersome is it when you sweat a lot inside your prosthesis (in the sock, liner, socket)?

EXTREMELY BOTHERSOME NOT AT ALL

G. How bothersome to you is swelling in your residual limb (stump)?

EXTREMELY BOTHERSOME NOT AT ALL

H. How important is it to avoid having any ingrown hairs (pimples) on your residual limb (stump)?

NOT AT ALL EXTREMELY IMPORTANT

I. How bothersome is it to see people looking at you and your prosthesis?

EXTREMELY BOTHERSOME NOT AT ALL

J. How important is being able to walk up a steep hill?

NOT AT ALL EXTREMELY IMPORTANT

Final Notes

A. If any of the following have happened in the past four weeks, please check off and give a brief description:

_ a serious medical problem (yours)

_ a noticeable change in pain

_ a serious personal problem (yours)

_ a serious problem in the family

_ some other big change has occurred in your life

If you checked any of the five previous items, please give a brief description.

B. Please share with us anything else about you or your prosthesis that you think would be helpful for us to know (continue on the back of this page if you need more space).

THANK YOU VERY MUCH!

Acknowledgement: Roorda LD, Roebroeck ME, Lankhorst GJ, van Tilburg T, Bouter LM. Measuring functional limitations in rising and sitting down: Development of a questionnaire. Arch Phys Med Rehabil 1996;77;663-669 for their influence on questions 4-J, 4-K, and 4-L.

APPENDIX G – Modified Falls Efficacy Scale (Hill *et al.*, 1996)

Modified Falls Efficacy Scale

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 At All	Fairly Confident	Completely Confident

(11) Using public transport

Not Confident At All	Fairly Confident	Completely Confident

(12) Crossing roads

Not Confident At All	Fairly Confident	Completely Confident

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

Not Confident At All	Fairly Confident	Completely Confident

(14) Using front or rear steps at home

Not Confident At All	Fairly Confident	Completely Confident

APPENDIX H – Reliability and Accuracy of the three-dimensional motion capture system used in studies one and three

The reliability and accuracy of the Qualisys motion capture system, along with associated force plates was tested using distance, angular and loading protocols. The motion capture system was set-up and calibrated as described previously. A 10-camera system captured raw kinematic data at 100Hz.

Distance trials involved moving two 14mm markers through the calibrated volume for ten seconds per trial and repeated for ten trials. The markers were attached to calibration wands and separated by known distances of 299.5 (small) and 749.9mm (large). Neither wands were previously used to calibrate the motion capture system.

The mean (±SD) difference between the recorded and known distance for the large wand was -0.2 ± 2.6mm with a root mean square (RMS) of 0.26. The coefficient of variation (CV) was 0.35.

Table H 0.1 Recorded distance of known large (749.9mm) wand length.

Trial	Recorded wand length (mm)	Difference to known length (mm)	Absolute difference (mm)
1	750.6	0.7	0.7
2	750.6	0.7	0.7
3	750.7	0.8	0.7
4	742.5	-7.4	7.4
5	750.6	0.7	0.7
6	750.6	0.7	0.7
7	748.9	-1.0	1.0
8	750.8	0.9	0.9
9	750.3	0.4	0.4
10	750.8	0.9	0.9
Mean	749.7	-0.2	0.2
SD	2.6	2.6	2.6

The mean (±SD) difference between the recorded and known distance for the small wand was 0.0 ± 0.7mm with an RMS of 0.19. The coefficient of variation (CV) was 0.23.

Table H.0.2 Recorded distance of known small (299.5mm) wand length.

Trial	Recorded wand length (mm)	Difference to known length (mm)	Absolute difference (mm)
1	299.8	0.3	0.3
2	299.8	0.3	0.3
3	299.7	0.2	0.2
4	299.9	0.4	0.4
5	299.5	0.0	0.0
6	299.8	0.3	0.3
7	299.6	0.1	0.1
8	299.7	0.2	0.2
9	299.6	0.1	0.1
10	299.5	0.0	0.0
Mean	299.5	0.0	0.0
SD	0.7	0.7	0.7

Angular trials involved attaching three 14mm reflective markers to a plastic goniometer, one at each distal arm and one at the vertex. The goniometer was set at three pre-defined angles, 25, 45 and 90 degrees and moved through the calibrated volume ten times per angle for ten seconds per trial.

The mean (±SD) difference between the recorded and known 25 degree angle was -0.1 ± 0.1 degrees with an RMS of 0.10. The coefficient of variation (CV) was 0.40.

Table H.0.3 Recorded angle of goniometer set at pre-defined angle of 25 degrees.

Trial	Recorded angle (degrees)	Difference to known angle (degrees)	Absolute difference (degrees)
1	24.9	0.1	0.1
2	24.9	0.1	0.1
3	24.8	0.2	0.2
4	24.9	0.1	0.1
5	24.9	0.1	0.1
6	24.9	0.1	0.1
7	25.0	0.0	0.0
8	24.9	0.1	0.1
9	24.9	0.1	0.1
10	24.9	0.1	0.1
Mean	24.9	-0.1	0.1
SD	0.1	0.1	0.1

The mean (±SD) difference between the recorded and known 45 degree angle was 0.1 ± 0.1 degrees with an RMS of 0.11. The coefficient of variation (CV) was 0.22.

Table H.0.4 Recorded angle of goniometer set at pre-defined angle of 45 degrees.

Trial	Recorded angle (degrees)	Difference to known angle (degrees)	Absolute difference (degrees)
1	45.0	0.0	0.0
2	45.1	0.1	0.1
3	45.1	0.1	0.1
4	45.1	0.1	0.1
5	44.9	-0.1	0.1
6	45.0	0.0	0.0
7	45.2	0.2	0.2
8	45.3	0.3	0.3
9	45.2	0.2	0.2
10	45.2	0.2	0.2
Mean	45.1	0.1	0.1
SD	0.1	0.1	0.1

The mean (±SD) difference between the recorded and known 90 degree angle was 0.1 ± 0.1 degrees with an RMS of 0.30. The coefficient of variation (CV) was 0.11.

Table H.0.5 Recorded angle of goniometer set at pre-defined angle of 90 degrees.

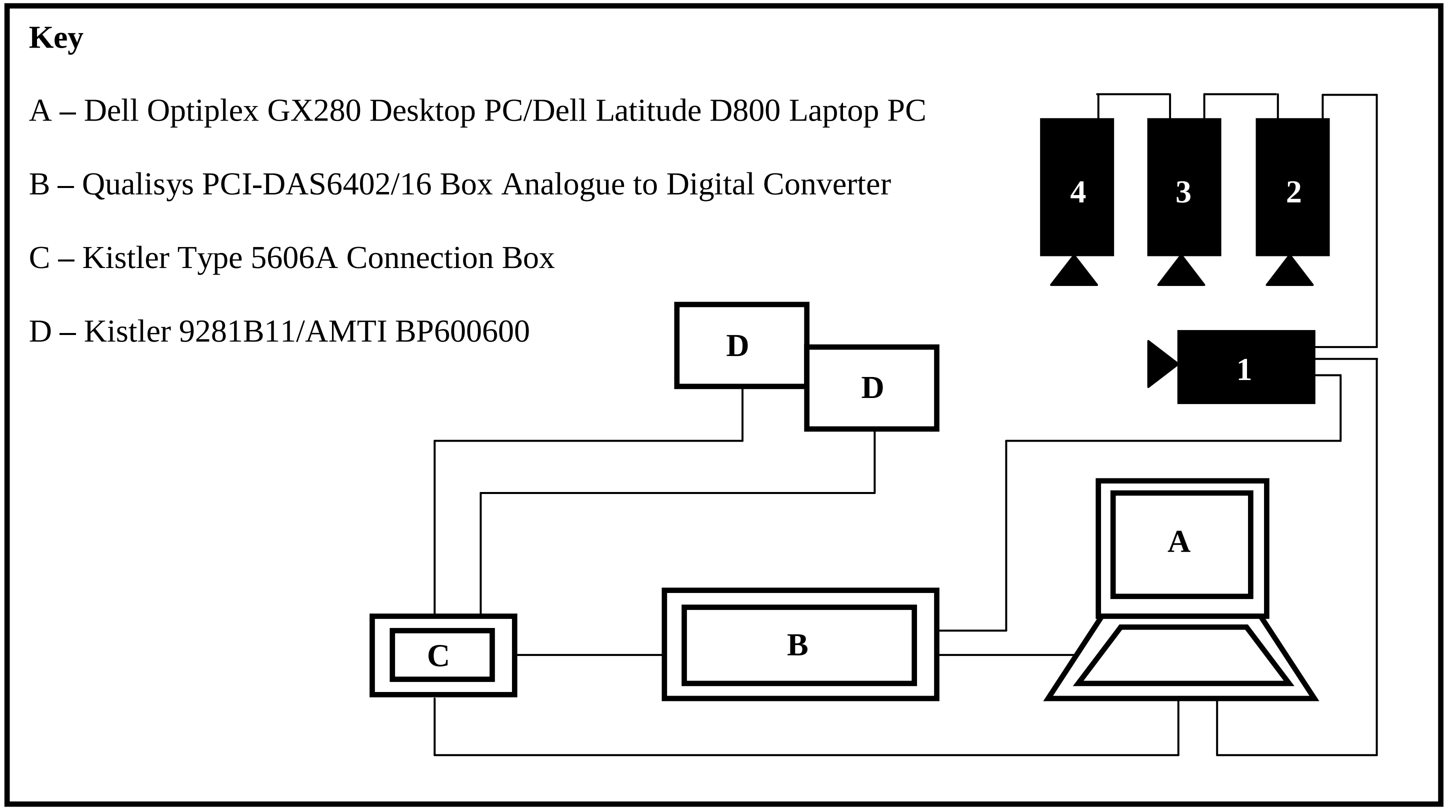
Trial	Recorded angle (degrees)	Difference to known angle (degrees)	Absolute difference (degrees)
1	89.6	-0.4	0.4
2	89.7	-0.3	0.3
3	89.7	-0.3	0.3
4	89.6	-0.4	0.4
5	89.8	-0.2	0.2
6	89.9	-0.1	0.1
7	89.7	-0.3	0.3
8	89.8	-0.2	0.2
9	89.7	-0.3	0.3
10	89.5	-0.5	0.5
Mean	89.7	-0.3	0.3
SD	0.1	0.1	0.1

The accuracy of the Kistler (Kistler 9281B11) and AMTI (AMTI BP600600) force plates was determined by statically loading the force plates with known weights. The vertical GRF was recorded for ten seconds while each respective force plate was loaded with the following weights; 245.3N (25kg), 490.5N (50kg), 735.8N (75kg) and 981.0N (100kg) force plate. This was repeated for ten trials per weight.

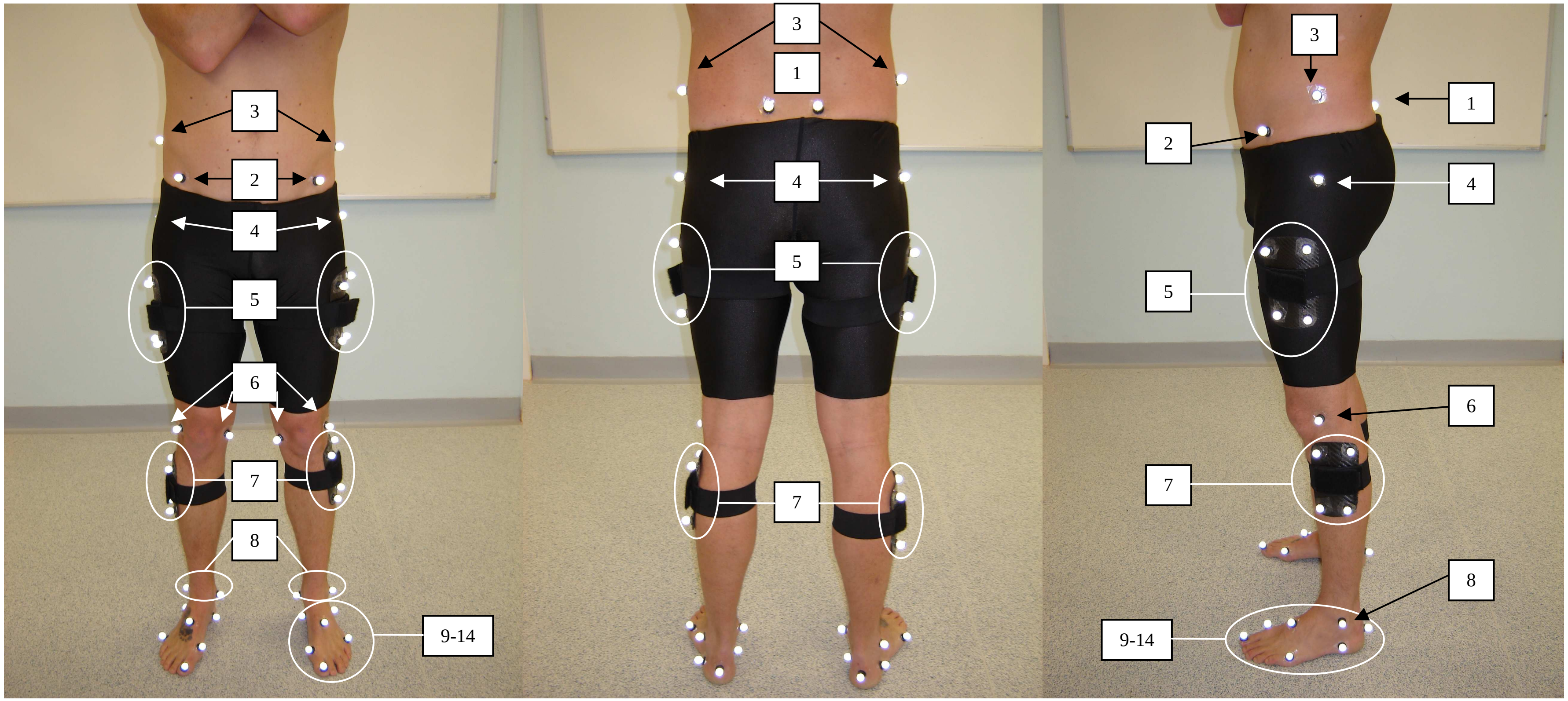
Table H.0.6 Recorded loads from Kistler and AMTI force plates when loaded with known static weights.

Known load	245.3N	490.5N	735.8N	981.0N
Mean (±SD) recorded load (N) (Kistler)	238.8±0.5N	481.3±3.2N	721.8±7.7N	968.1±10.2N
RMS	<0.01	0.01	0.04	0.05
CV	0.21	0.66	1.06	1.05
Mean (±SD) recorded load (N) (AMTI)	244.6±0.4	479.5±0.5	717.3±0.4	950.6±0.4
RMS	<0.01	<0.01	<0.01	<0.01
CV	0.16	0.10	0.06	0.04

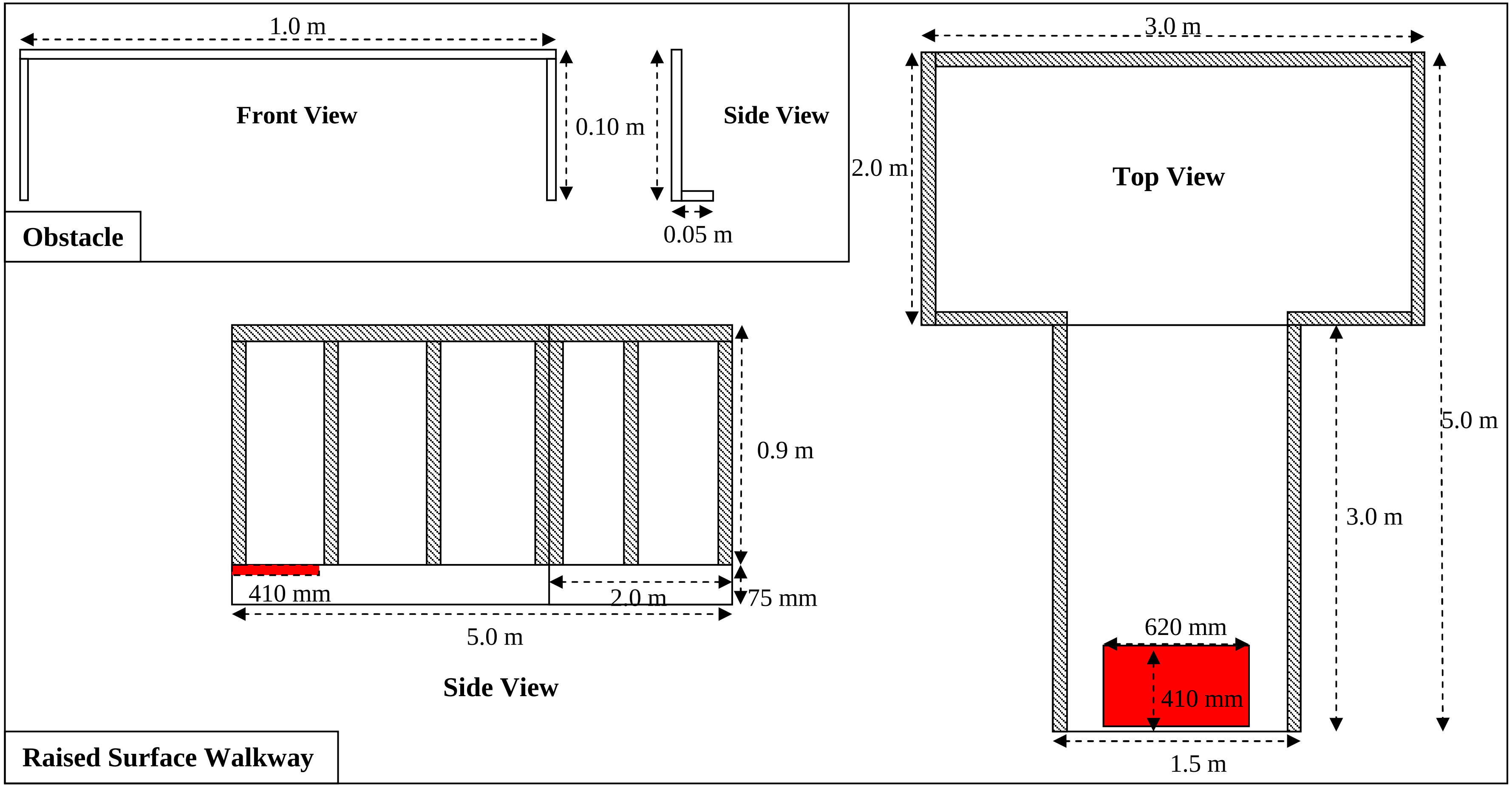
APPENDIX I - Schematic illustration of hardware set-up for studies one and three. Study one incorporates Qualisys ProReflex camera system only, with study three utilising all associated hardware.



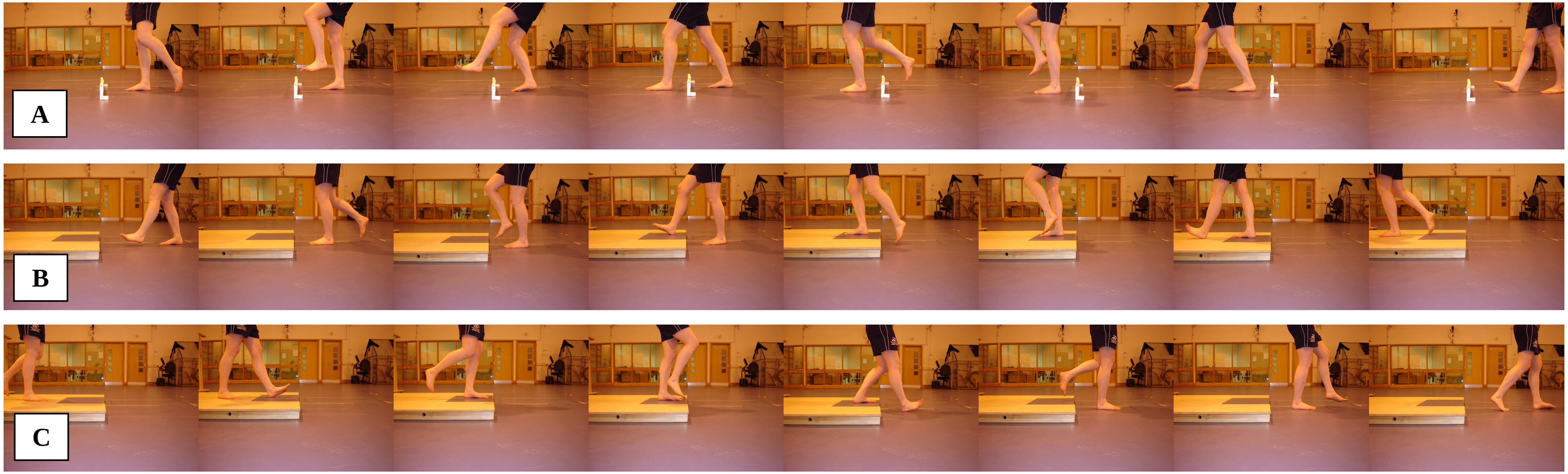
APPENDIX J - The six-degree-of-freedom marker model set. Numbers correspond with the details in Table 3.2.



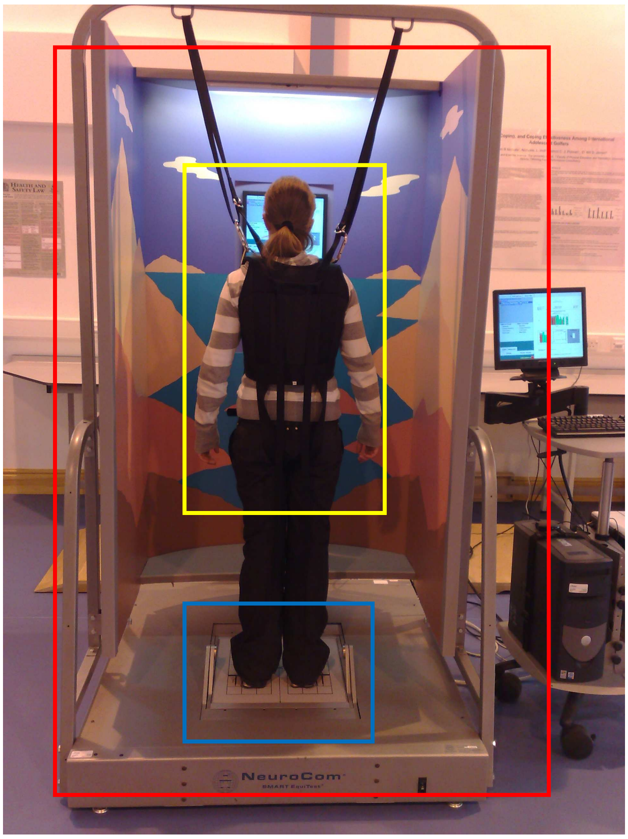
APPENDIX K - Dimensions of raised surface walkway and obstacle (inset) used in the current study.



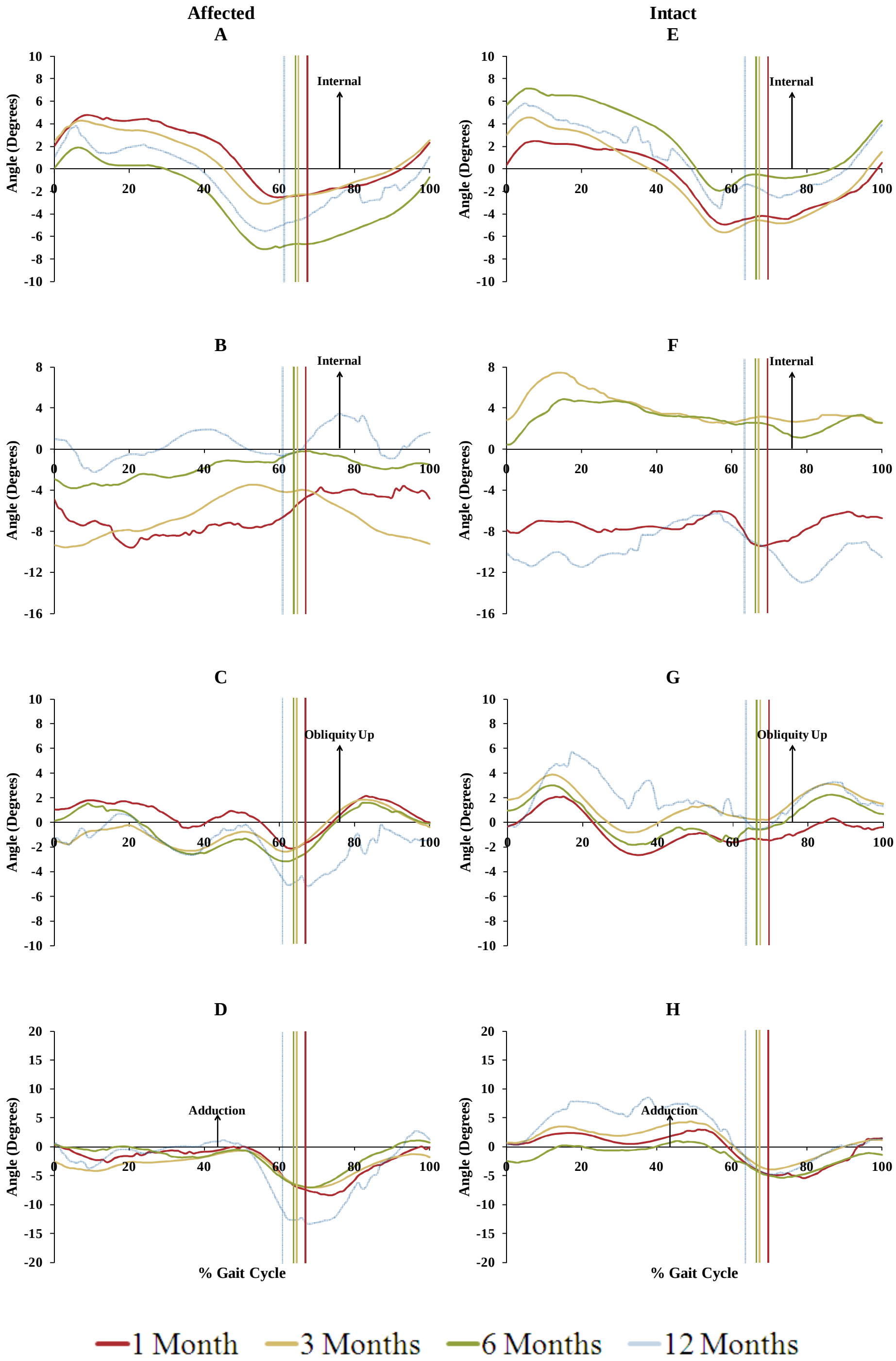
APPENDIX L - Sequential diagrams of the performance of the obstacle crossing (A), stepping up gait (B) and stepping down gait (C).



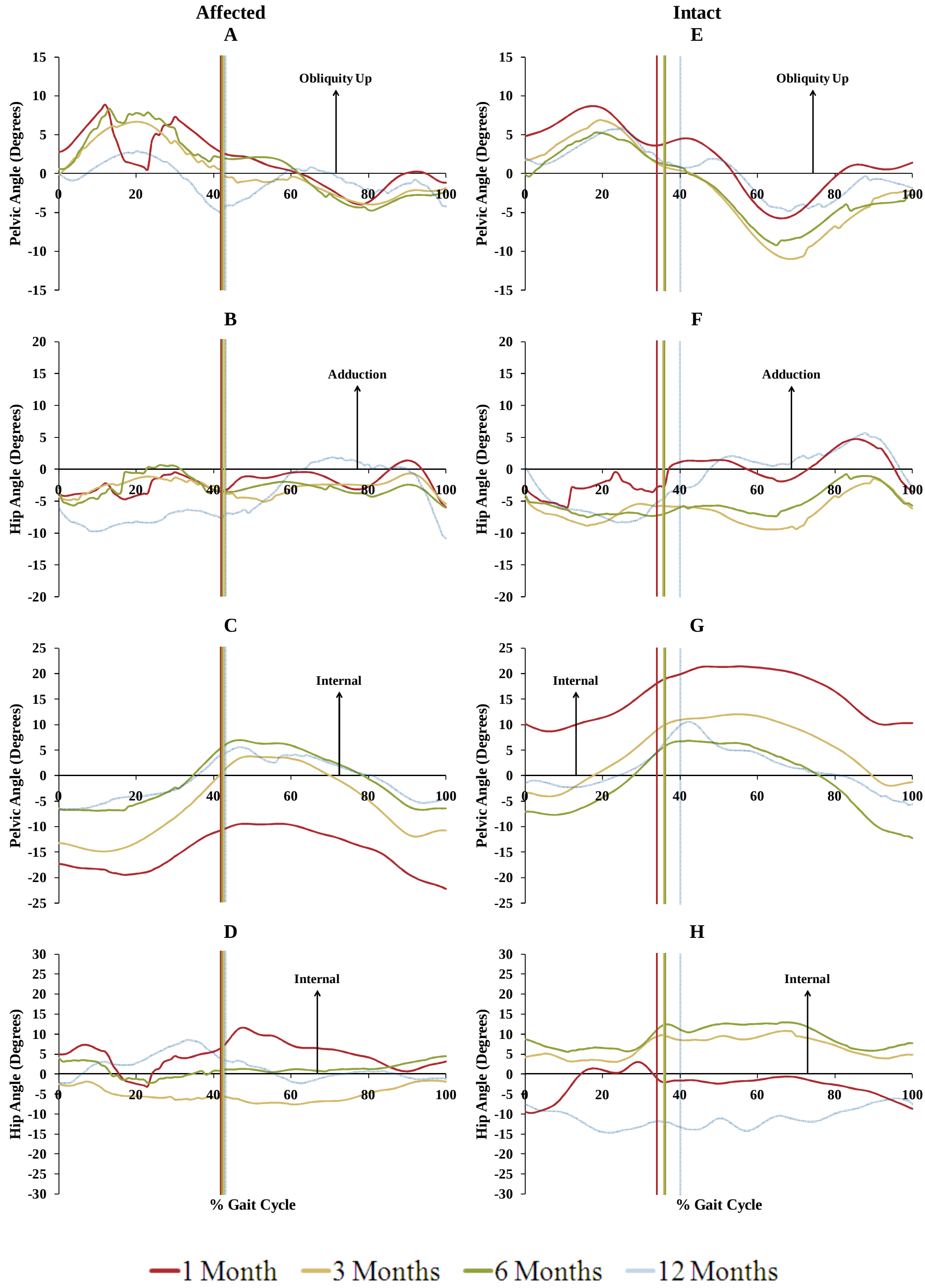
APPENDIX M - Neurocom Equitest® (NeuroCom International , Inc, Clackamas, US) with visual surround (red), platform (blue) and support harness (yellow) components highlighted. The visual display was consistently present to help explain each test protocol to patients.



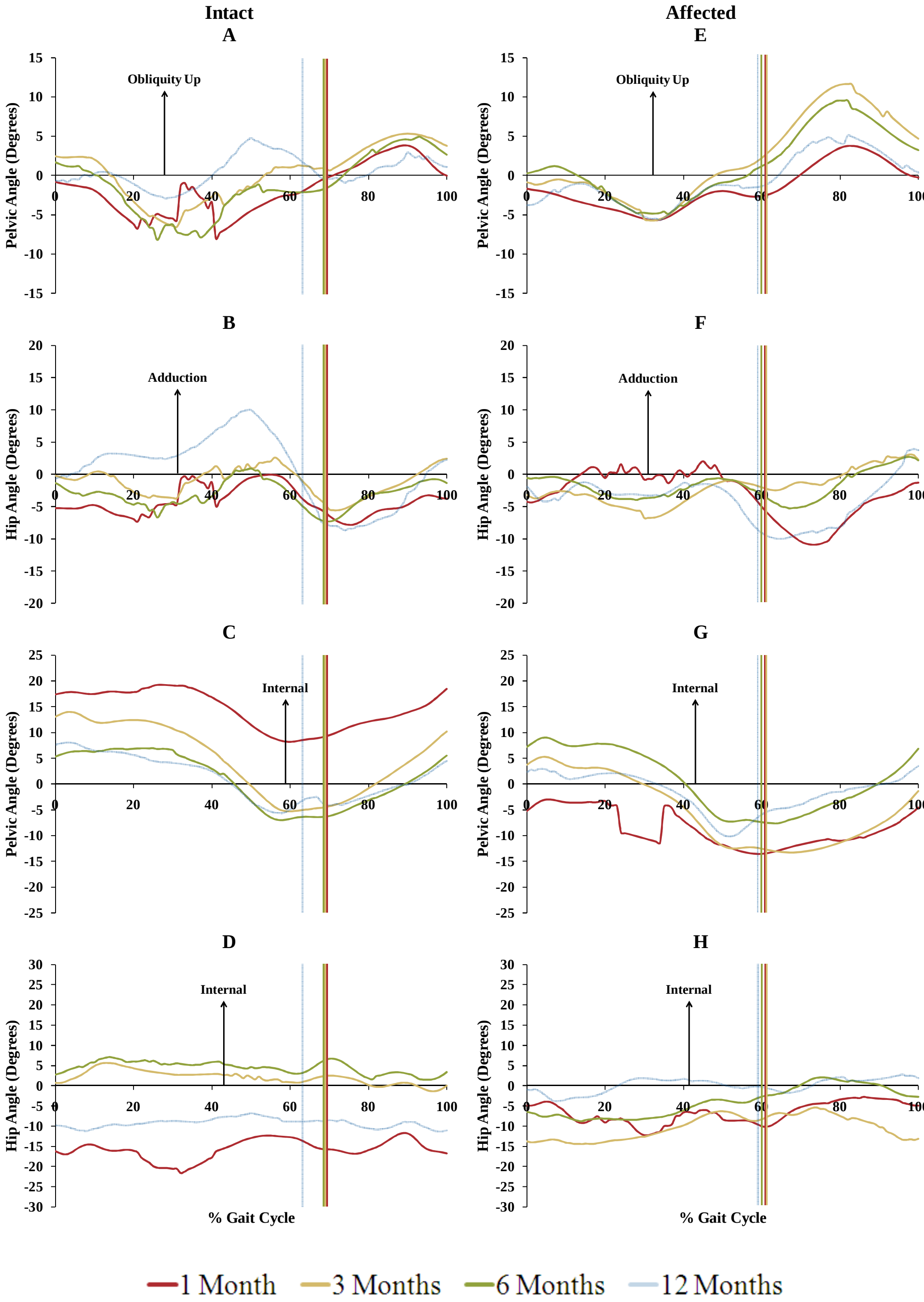
APPENDIX N – Frontal and transverse plane joint kinematic data.



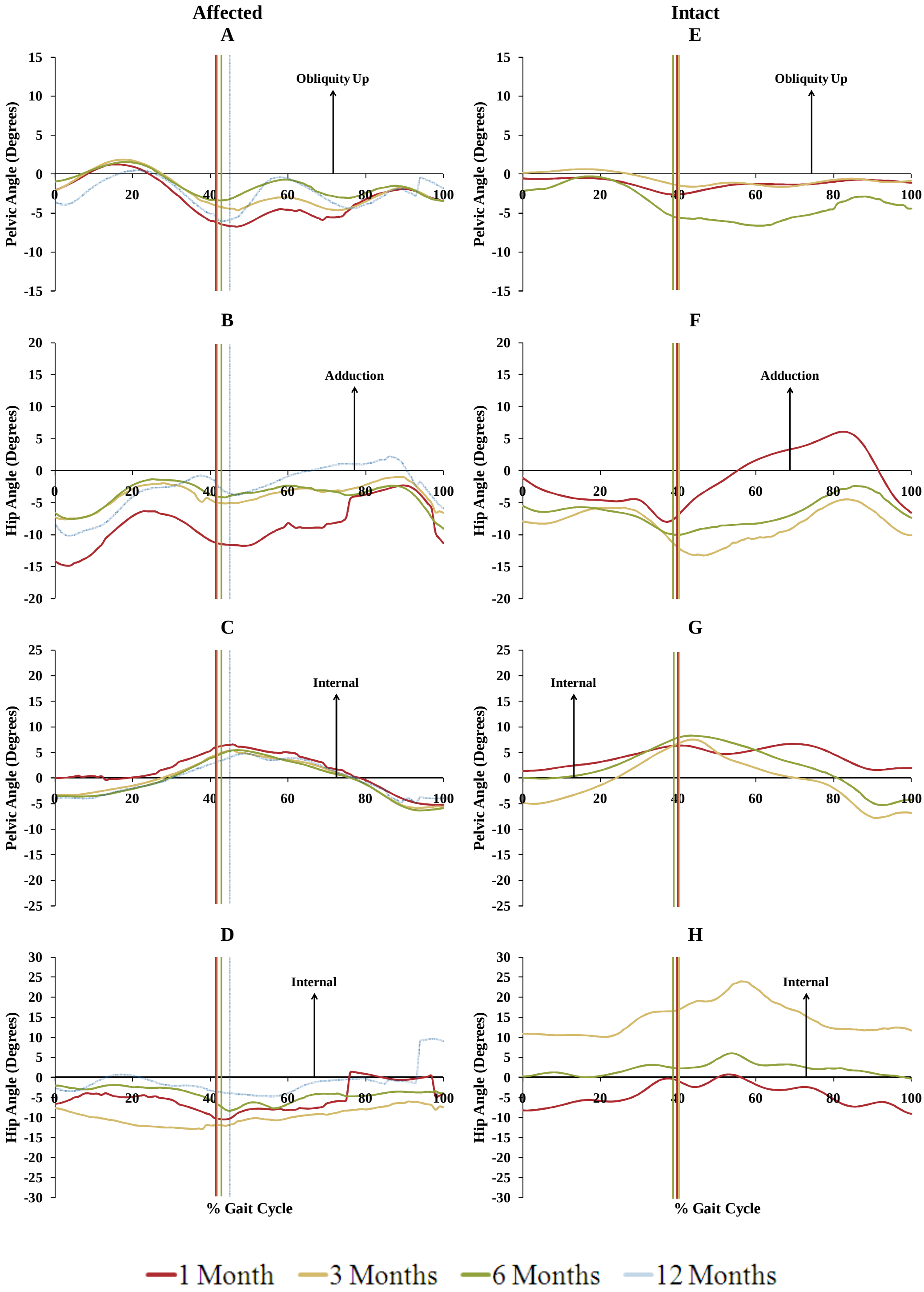
Level Gait - Group mean transverse plane kinematics of the affected limb pelvis (A) and hip (B) and intact limb pelvis (E) and hip (F). Group mean frontal plane kinematics of the affected limb pelvis (C) and hip (D) and intact limb pelvis (G) and hip (H). Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.



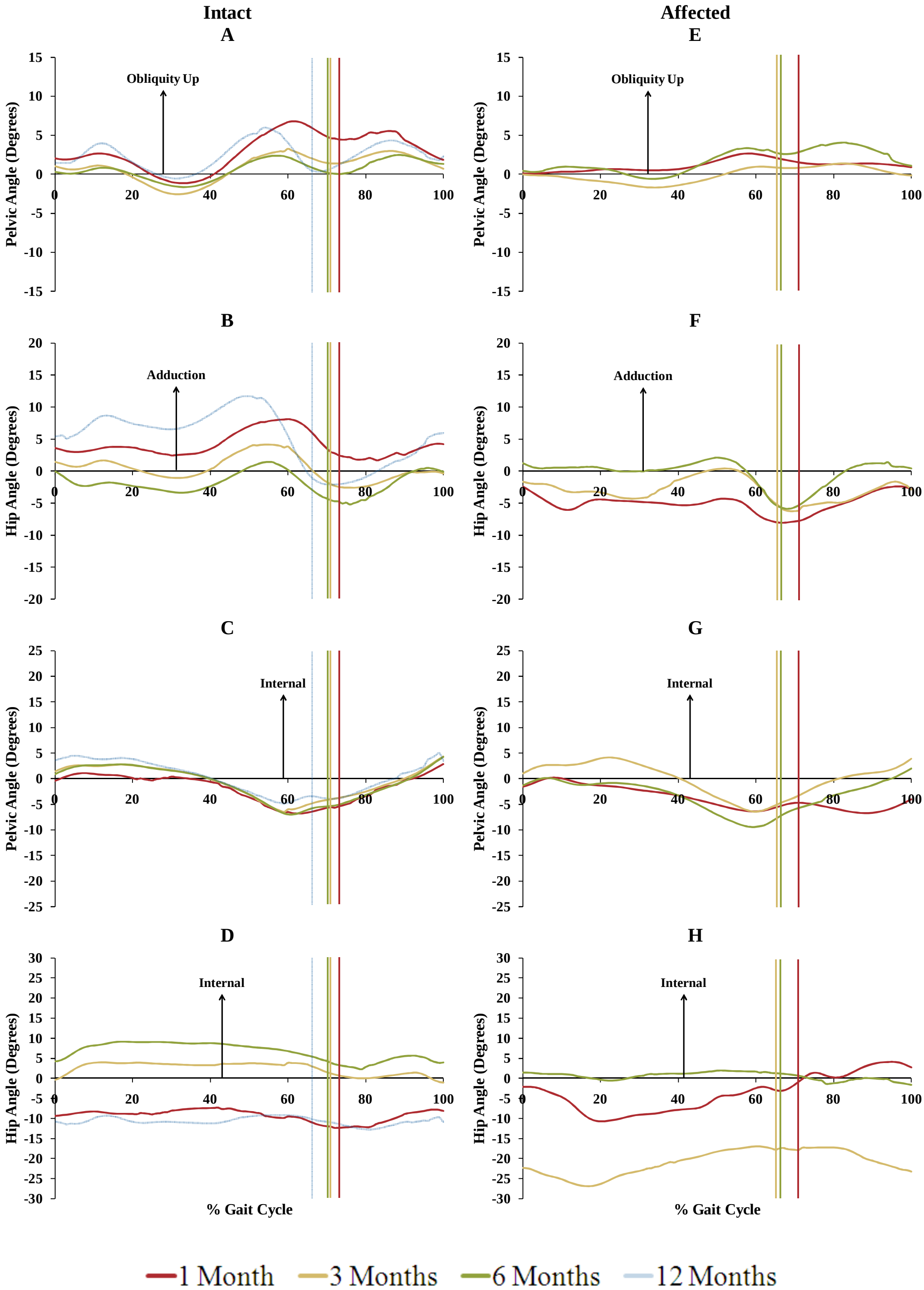
Obstacle Crossing - Group mean kinematics of the lead affected limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and lead intact limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.



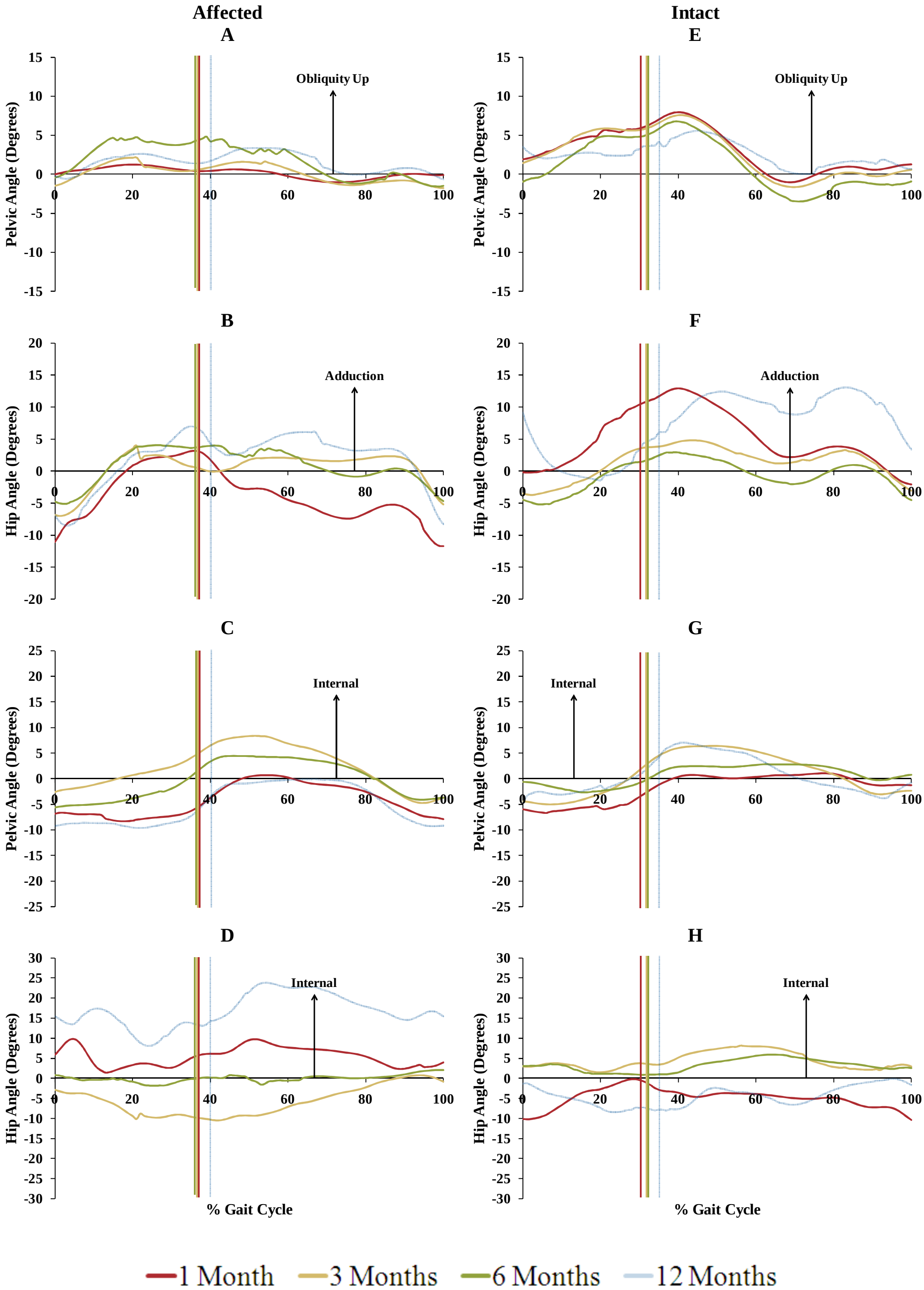
Obstacle Crossing - Group mean kinematics of the trail intact limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and trail affected limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.



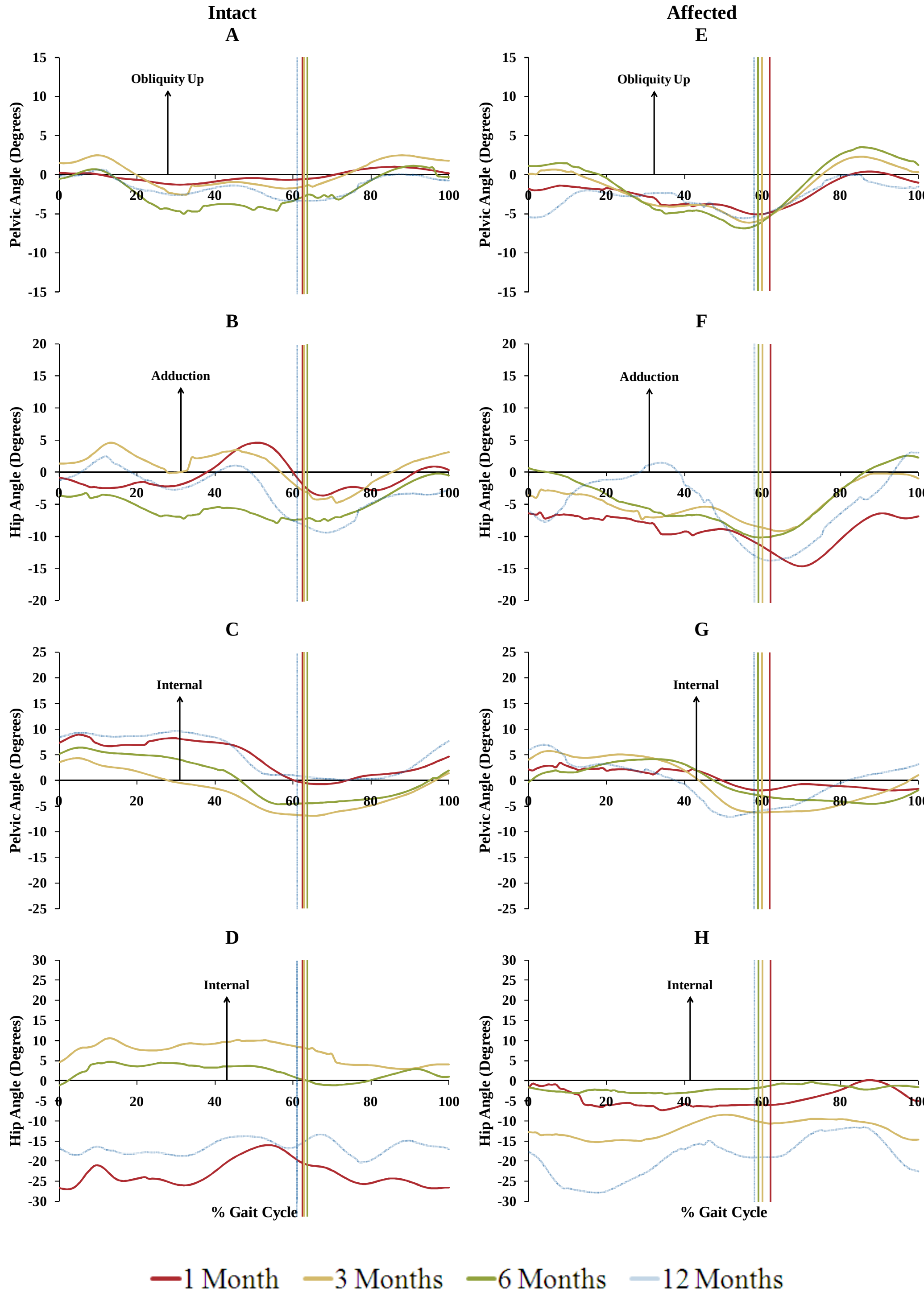
Stepping Down Gait - Group mean kinematics of the lead affected limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and lead intact limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.



Stepping Down Gait - Group mean kinematics of the trail intact limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and trail affected limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.

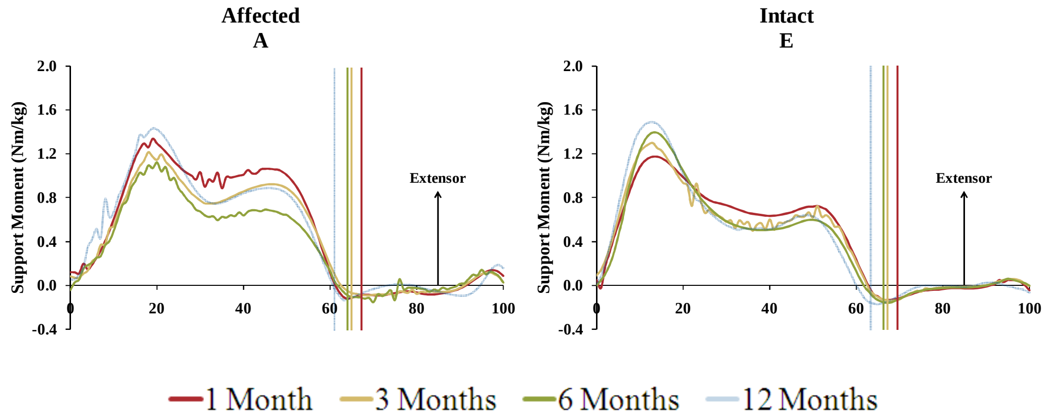


Stepping Up Gait - Group mean kinematics of the lead affected limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and lead intact limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.

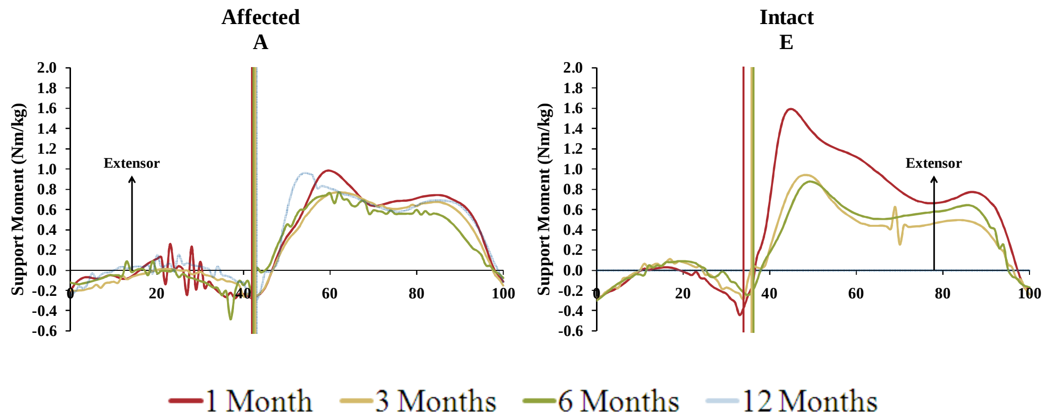


Stepping Up Gait - Group mean kinematics of the trail intact limb pelvis (A), hip (B) (frontal plane), pelvis (C) and hip (D) (transverse plane) and trail affected limb pelvis (E), hip (F) (frontal plane), pelvis (G) and hip (H) (transverse plane). Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.

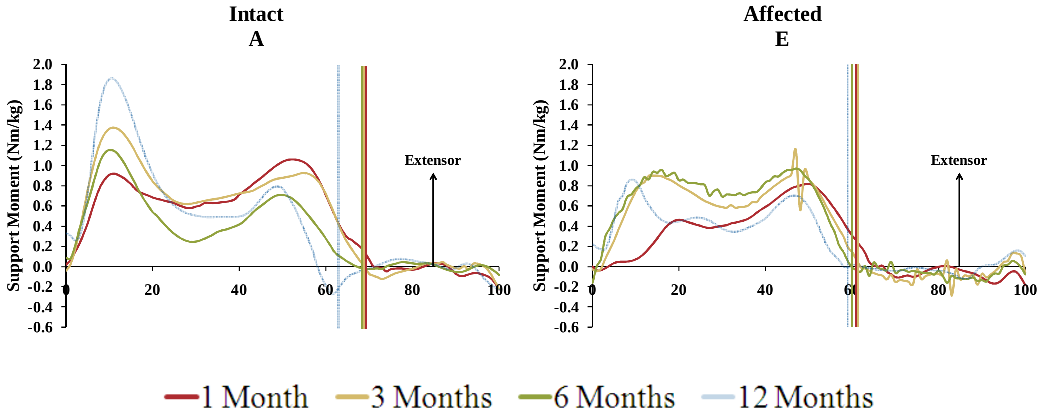
APPENDIX O – Saggital plane support moment data



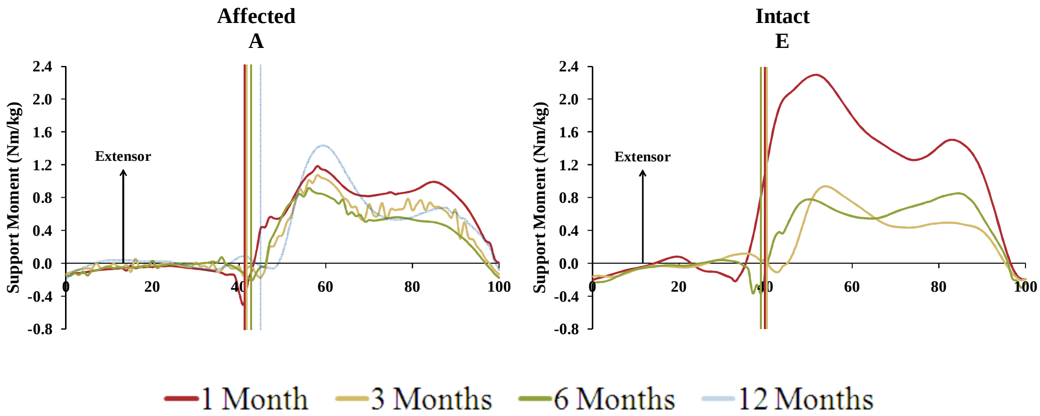
Level Gait - Group mean support moments for the affected (A) and intact limbs (E). Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.



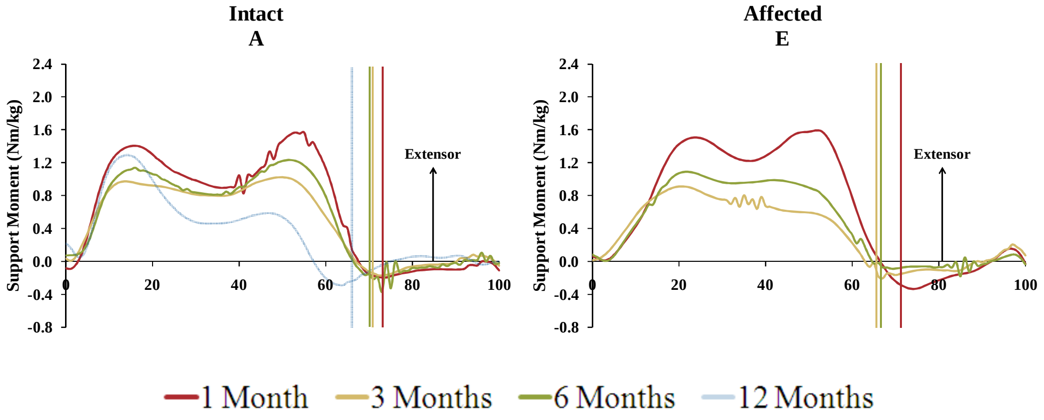
Obstacle Crossing - Group mean support moments for lead affected (A) and lead intact (E) limbs. Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.



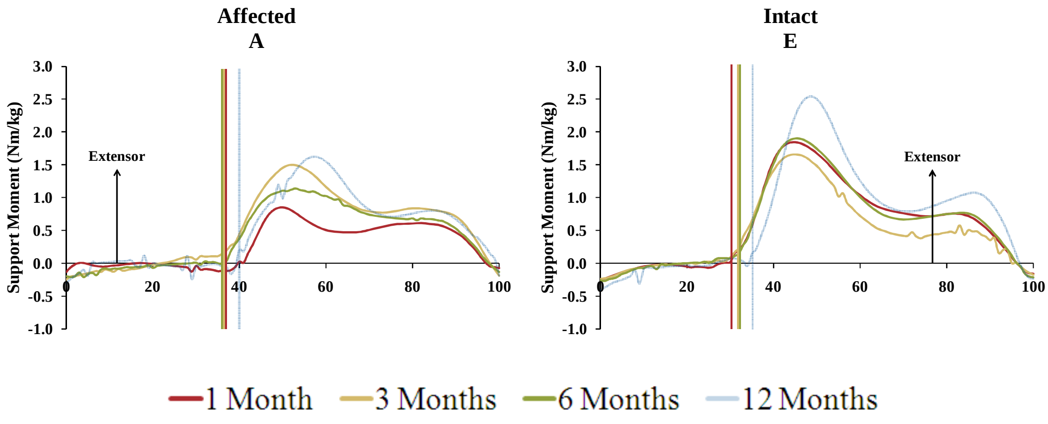
Obstacle Crossing - Group mean support moments for trail intact (A) and trail affected (E) limbs. Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.



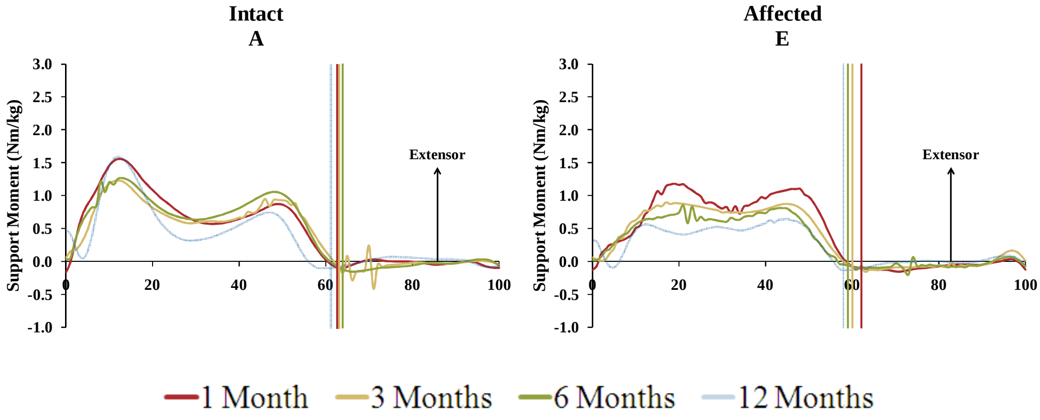
Stepping Down Gait - Group mean support moments for lead affected (A) and lead intact (E) limbs. Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.



Stepping Down Gait - Group mean support moments for trail intact (A) and trail affected (E) limbs. Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.



Stepping Up Gait - Group mean support moments for lead affected (A) and lead intact (E) limbs. Time normalised to 100% of gait cycle. Lead limb gait cycle defined from toe-off to toe-off. Vertical lines represent foot contact. Data at 12 months from n=2.



Stepping Up Gait - Group mean support moments for trail intact (A) and trail affected (E) limbs. Time normalised to 100% of gait cycle. Vertical lines represent toe off. Data at 12 months from n=2.