IMPACT OF COMBINED MICROPROCESSOR CONTROL OF THE PROSTHETIC KNEE AND ANKLE ON GAIT TERMINATION IN UNILATERAL TRANS-FEMORAL AMPUTEES

LIMB MECHANICAL WORK PERFORMED ON CENTRE OF MASS TO TERMINATE GAIT ON A DECLINED SURFACE USING LINX PROSTHETIC DEVICE

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Abstract

The major objective of this thesis was to investigate how the use of a recently developed microprocessor-controlled limb system altered the negative mechanical work done by the intact and prosthetic limb when trans-femoral amputees terminated gait. Participants terminated gait on a level surface from their self-selected walking speed and on declined surface from slow and customary speeds, using limb system prosthesis with microprocessor active or inactive. Limb negative work, determined as the integral of the negative mechanical (external) limb power during the braking phase, was compared across surface, speed and microprocessor conditions.

Halting gait was achieved predominantly from negative work done by the trailing/intact. Trailing versus leading limb mechanical work imbalance was similar to how able body individuals halted gait. Importantly, the negative limb work performed on the prosthetic side when terminating gait on declined surface was increased when the microprocessor was active for both slow and customary speeds (no difference on level surface) but no change on intact limb. This indicates the limb system's 'ramp-descent mode' effectively/dynamically altered the hydraulic resistances at the respective joints with evidence indicating changes at the ankle were the key factor for increasing the prosthetic limb negative work contribution. Findings suggest that trans-femoral amputees became more assured using their prosthetic limb to arrest body centre of mass velocity when the limb system's microprocessor was active. More generally findings suggest, trans-femoral amputees should obtain clinically significant biomechanical benefits from using a limb system prosthesis for locomotion involving adapting to their everyday walking where adaptations to an endlessly changing environment are required.

Keywords: Gait termination, Ramp descent, Trans-femoral amputee, Microprocessor, Above-knee prosthesis, Limb mechanical work

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Publications

Abdulhasan, Z.M., Buckley, J.G., Gait Termination on Declined Compared to Level Surface; Contribution of Terminating and Trailing Limb Work in Arresting Centre of Mass Velocity. (In review) This work is detailed in chapter 4.

Abdulhasan ZM, Scally AJ and Buckley JG (2018) Gait termination on a declined surface in trans-femoral amputees: impact of using microprocessor-controlled limb system. Clinical Biomechanics. 57: 35-41. This work is detailed in chapter 7.

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List of Abbreviations

AB	Able-Body Individuals
AP	Anteroposterior
BoS	Base of Support
CoM	Centre of Mass
CoP	Centre of Pressure
DoF	Degree of freedom
ES	Effect size
GCS	Global Coordinate System
GRF	Ground Reaction Force
GRV	Ground Reaction Vector
LCS	Local coordinate system
LLA	Lower Limb Amputees
MC	Microprocessor-Controlled
ML	Mediolateral
NHS	National Health Service
NASDAB	National Amputee Statistical Database
TFA	Trans-Femoral Amputee
TTA	Trans-Tibial Amputee
3D	Three-dimensional

Chapter One

Thesis Introduction

1.1 Overview

Modern lower limb prostheses are designed to replicate the function of the physiological limb in order to preserve an amputee's ability to carry out routine standing and walking tasks. Patients' medical circumstances dictate the number of joints being removed during amputation; a higher number leads to a considerable rise in disability. Thus, the level of amputation (e.g. above or below the knee) imposes fundamental design considerations which need to be addressed in order to minimise this disability. Since knee joint has a key role in locomotion, trans-femoral amputees (TFAs) are at a greater disadvantage compared to trans-tibial amputees (TTAs). Distinctly, a prosthesis for a below knee amputee consists mainly of a foot-ankle device, but an above knee amputee also requires a knee device. Foot-ankle-devices are required to provide a stable base of support and stance phase 'roll over' function. Knee devices not only need to provide stable support of body weight during stance, they are also required to offer appropriate swing phase function, to ensure the limb is able to swing forwards appropriately/optimally to start the next step.

These gait requirements that prosthetic limbs (knee and foot devices) must fulfil become more difficult if the surrounding environment is not uniformly flat. It is known that different terrains necessitate a range of biomechanical compensations to produce smooth and safe locomotion. For instance, when able-body individuals (AB) walk down-slope in comparison to level walking, extra kinetic energy accelerates the body, which unless controlled could cause balance to be lost and potential injure to occur. Thus, to remain dynamically stable, a reduction in body Centre of Mass (CoM) velocity is facilitated by eccentric muscular contraction (Chapman and Fraser 2008; Winter 2009). The angle of the surface can have a significant effect on kinematic and joint kinetic variables, with major effect on knee joint kinematics; as a primary joint used to lower the body down the slope during each step (Redfern and DiPasquale 1997; Redfern 2001; Cham and Redfern 2002).

One would expect a more complicated scenario for TFAs since having the line of action of GRV anterior to the prosthetic knee joint axis is essential to keep the knee extended during stance. But achieving this when walking down slopes can be problematic. A prosthetic foot without an ankle device provides 'simulated plantar-dorsiflexion' via deformation of the heel and forefoot keels.

On level surface the simulated plantar-dorsiflexion that occurs following foot contact allows the foot to attain a foot-flat position. However, during down-slope walking the deformation (simulated plantarflexion) is not enough to achieve foot-flat.

Difficulty in attaining foot-flat when descending slopes means that the prosthetic shank has a tendency to rotate forward during early stance, and given that body mass is posterior to the foot during this time, this creates a flexion moment at the prosthetic knee (Highsmith et al. 2014). In TTAs, knee flexion is increased down-slope compared to level surface with rigid ankle (Vickers et al. 2008; Fradet et al. 2010). Stance phase knee stability is the key factor in gait stability when the surface slopes downward, since the knee is the main joint involved in lower the CoM down-slopes when the limb is loaded (Lay et al. 2006; Lay et al. 2007). However, TFA are unable to flex the knee to help attain foot-flat so they must utilize certain adaptations to walk down-slopes. As a result, slope descent is a demanding task for TFA (Vrieling et al. 2008b).

As such, stopping down-slope for TFA would become even more difficult than walking because of the requirement to accommodate the angled surface (as highlighted above). The ramp descent mode of the limb system prosthesis is set to keep the knee stable (unflexing) during weight acceptance but then to allow controlled flexion during late stance to lower the CoM down the ramp. With the knee stable during weight acceptance this should allow the limb to do the necessary mechanical work to arrest CoM forward and downward velocity when terminating gait during ramp descent. Although, knee design and function are critical to how TFA walk down ramps it is unclear how knee design affects stopping whilst descending ramps.

For such tasks, a microprocessor-controlled (MC) prosthesis can provide adaptive function. Having an artificial limb that can ‘smartly’ (i.e. sensing, interfacing, signal processing) and ’intelligently’ (i.e. synchronised response to sensor signals for necessary dynamic adaptation) differentiate between overground and ramp movement to allow the user to carry out locomotion with pre-set modes is considered an important innovative solution for an amputee. As such, use of MC prostheses are purported to allow a more natural walking movement, minimise the likelihood of falls, and reduce the energy costs of locomotion as they are thought to require less compensatory adjustment to diverse environmental terrains.

A recently designed MC control-limb is the Linx limb system prosthesis (Chas A Blatchford & Sons Ltd) which is an integrated ankle-knee control system composing of a hydraulic knee and hydraulic ankle being controlled by a single MC in a synchronised way. The limb system allows the simultaneous communication to/from knee and ankle, controlling hydraulic resistance/damping rate, via several pre-set control algorithms. One such control algorithm is the ‘ramp descent mode’. The ramp-descent mode in brief automatically adjusts the resistance at the ankle (i.e. the plantar-flexion resistance goes to its second lowest setting and the dorsi-flexion resistance goes to its second highest) for MC foot-ankle device. According to the manufacturer- it also reduces knee resistance in late stance to an intermediate level rather than the usual pre-swing (low resistance) level which provides a ‘brake effect’ to help reduce forward (downward) momentum and thus promote better dynamic stability in late stance. This device should facilitate down-slope walking in TFA and thus it should also help TFA to terminate gait when walking down slopes.

Gait termination task is mainly performed via power absorption to arrest body momentum (Oates et al. 2005). Therefore, it has been decided to assess negative mechanical limb work as the main determinant for stopping down-slope. Thus, the general aim of this thesis was to investigate the biomechanical adaptations used by TFAs during gait termination when descending a ramp. A principle focus was to evaluate the effectiveness of using a limb system prosthesis for ramp descent and in particular terminating ramp-descent. The central question was how much external mechanical limb work was done to stop body CoM to terminate ramp descent. In order to introduce a fair and objective assessment about the limb system efficacy, it was decided to investigate its biomechanical effects compared to using a TF prosthesis with passive hydraulic knee and ankle (i.e. Linx limb system with the MC inactive, so that hydraulic resistance reverts to default levels).

This thesis aspires to promote a deep comprehension of gait termination task and present an objective scientific evaluation of the impact on gait biomechanics of using the limb system prosthesis. The first task was to establish a profile of negative mechanical work and power change on declined versus overground in AB (experimental chapter 1). In addition, investigating the negative mechanical work during gait termination relative to different walking speed prior to stop on declined surface (experimental chapter 2). In the last two experimental chapters, the limb external work methodological approach was employed in investigations/analysis/studies on TFAs gait termination using the limb system prosthesis. The intention was to increase and develop understanding of this task and provide a reference point for researchers. Subsequently, enabling future designs appropriate for above knee prosthetic limbs as well as highlighting the main differences compared to able bodied that can contribute to the improvement of rehabilitation programs.

The specific aims were to determine:

1. The percentage/relative contribution of the intact limb (penultimate step) and prosthetic limb (final step) in negative mechanical limb work done on the CoM to arrest velocity during the two steps of gait termination. The main adaptation in ankle, knee and hip joint power and work due to prosthetic/surface interaction.

2. The effect of walking speed on the measured negative limb work and particularly in which orthogonal direction this effect would mainly occur.

3. How TFA perform limb work on CoM to terminating gait on ramps versus level surface compare to AB; what are the main observed compensations, what is the main influence of ramp descent mode/MC control (if any) on gait termination task, and finally whether speed of walking prior to stopping affects the above.

1.2 Contribution to the field (rationale for this work)

Abundant research has been conducted on the gait of TFAs but most of this has focused on level ground walking. For active TFAs who use prostheses on a daily basis, the ground conditions which present increased difficulties in walking, such as slopes and stairs, are commonly experienced. To date, a limited number have investigated TFA walking down-slopes, only three published studies have investigated the gait termination of TFAs involving level surface. The outcome measures have not varied among different studies. The joint power analysis was the gold standard traditional method to investigate the strategies of lower limb amputees' locomotion when walking or stopping. However, the contribution of these studies can be limited to some extent since the lack of information regarding CoM dynamic relationship with both intact and the prosthetic limb during gait termination task. Additionally, there have been many studies that have compared different prosthetic knee or ankles/feet separately (in isolation) i.e. most of these have focused on evaluating either the knee or ankle device; the results of these studies can therefore be of limited impact for supporting prostheses selection for TFAs. All above studies will be discussed in the literature review (next chapter).

Chapter Two

Literature Review

2.1 Introduction

This chapter reviews a selection of the available literature regarding statistics on lower limb amputation in the UK and some other developed and developing countries, in terms of the main conditions associated with amputation incidences and the prevalence of trans-femoral amputation. A critical analysis of the literature on prosthetic functional development that helps TFAs tackle deficiencies in gait biomechanics is also presented in the current chapter. Followed by an overview of the prosthetic limb evaluated in this research in addition to literature investigated task was introduced. Finally, literature related to the main outcome variables being evaluated in this thesis as well as the rational to determine these variables in particular are presented and assessed.

2.2 Epidemiology of lower-limb amputation

Lower limb amputation is the surgical removal of one or both limbs performed at different levels/sites as a result of a trauma or performed due to medical necessity. Amputation surgery is done as a lifesaving procedure in the case of a traumatic injury or as preventive measure; from further complications such as inadequate treatment of diabetic and vascular conditions. Furthermore, surgical amputations can be done electively, in case of limb congenital malformations, to make it amenable to prostheses. As an attempt to restore function thereby restoring ability to walk or sometimes performed for cosmetic reasons.

UK statistical information showed that approximately 1-2% of patients with peripheral vascular disease undergo a lower limb amputation (Norgren et al. 2007), though this number increases to 5% in diabetics (VSGBI 2012) as the most common cause of lower limb amputations in the UK (NASDAB 2007). Based on only new UK referrals for prosthetics treatment, the amputee statistical database for the all UK prosthetics service centres reveals that in 2011-2012, approximately 2055 out of 5400 lower limb amputees are TFAs (i.e. 38 %). The most common site of lower limb amputation is below knee, followed by the above-knee (trans-femoral) amputation as the second largest group of amputees. Although, the incidence of the higher level of amputation is relatively small, trans-femoral amputations can account for around 40% of total referrals to a prosthetic centre (NASDAB 2007). In 2005, nine out of ten referrals following lower limb amputations were either TTAs (52%) or TFAs (38%)

(Figure 1) (Table 1), this is similar to the previous years (NASDAB 2005). In 2010 there were 4346 amputees referred for limb fitting, 1575 (36%) were TTAs (Limbless Statistics 2010). The most recent amputee statistics by United National Institute for Prosthetics & Orthotics Development (Figure 2) stated that during 2011-2012, about 55.5% of all lower limb amputations were unilateral TTAs and 38% were unilateral TFAs, and 6.4% through -hip, knee, ankle-joints amputation and partial foot or foot digits removal. This suggests there are potentially 38% (of 500) who might benefit from using a limb system above-the-knee prosthesis. Up to 5400 are the total lower limb amputees in the UK for the period from 1st April 2011 to 31st March of the following year with approximately 1000 are due to trauma, infection or neoplasia. Approximately 47% represents the percentage of lower limb amputees below 64 years old compared to 53% of amputees aged over 64 years (Limbless statistics 2012). As mentioned previously, limb loss can be caused by trauma, malignancy, and infection. However, the main cause of amputation in TF population is dysvascularity, leading to 73% of trans-femoral amputations in 2007. This population also tends to be older with 64% of TFAs over the age of 65 (NASDAB 2007).

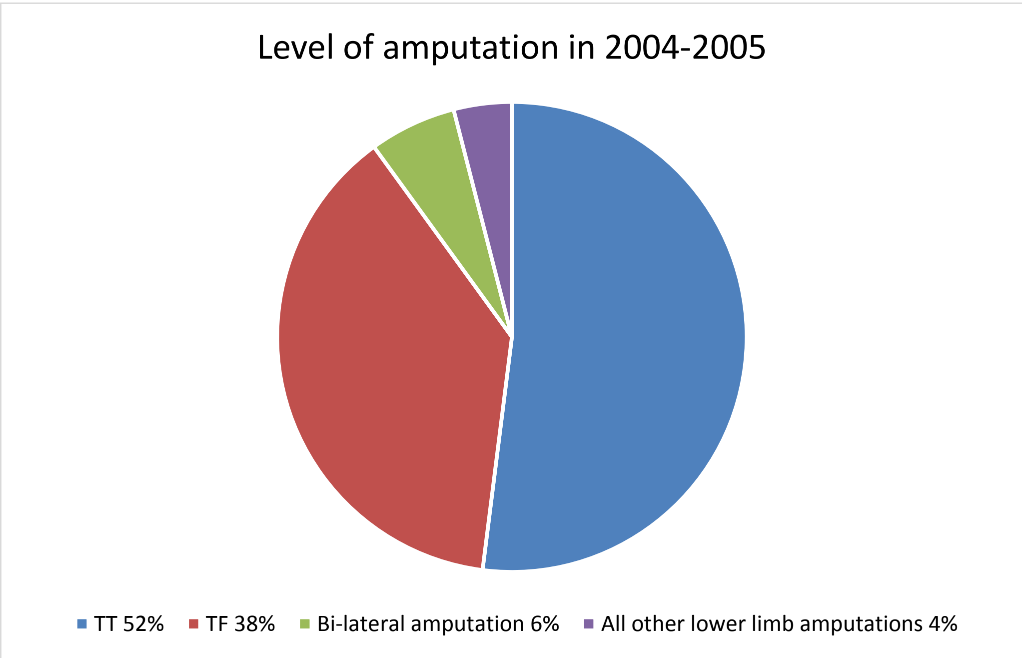


Figure 1. Percentages of lower limb amputations by level in UK, year 2005 (Limbless Statistics 2011-2012).

Table 1. Lower limb amputees in UK.

Year	All causes	Trauma	Dysvascular	others
2005	4786	426	3585	671
2012	5389	536	3091	1762

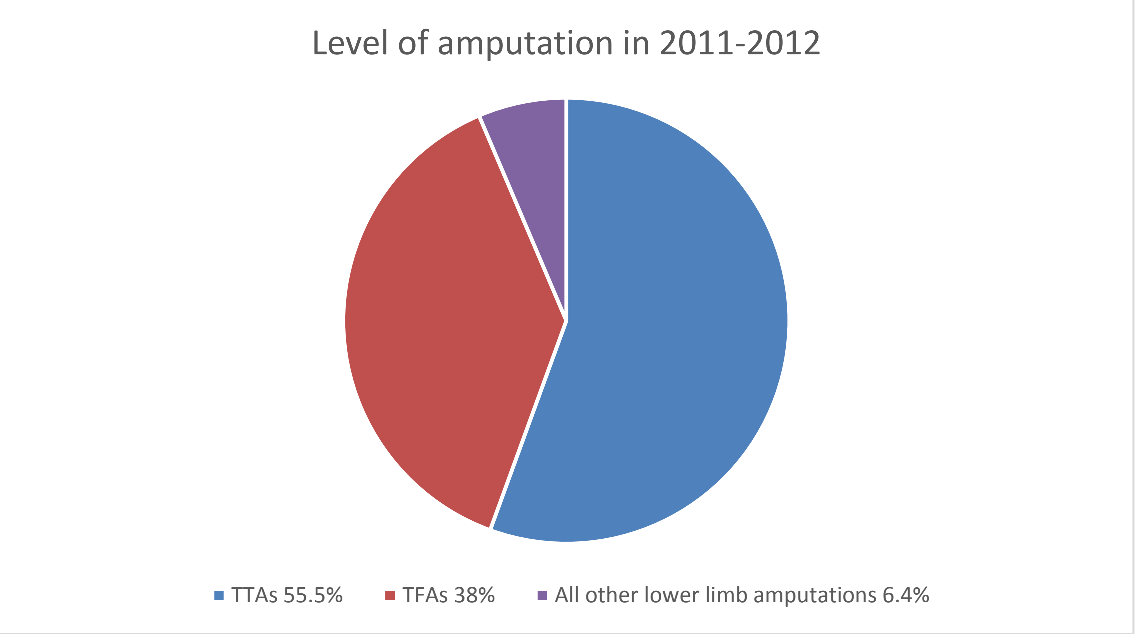


Figure 2. Percentages of lower limb amputations by level in UK, year 2012 (Limbless Statistics 2011-2012).

Worldwide, the published statistical data, reports a significant growth of the amputee population. Frequency of amputation globally involved about 200-500 million major amputation procedures are executed each year (Mathi et al. 2014). The incidence of lower limb amputation is 15 times higher in diabetic patients compared to non-diabetic patients (Diabetes UK 2012) with up to 70% dying within 5 years of the amputation surgery as a result of diabetes according to statistics on diabetes in 2012 (Diabetes UK 2012). Major limb amputations of the lower extremities account for approximately 85% of all cases of amputations. The research from the mid-1990s (Gailey et al. 1994) and (Nikitin 1996 in Pitkin, 2009) investigated lower extremity amputations; the data reported similarities to the numbers of patients experienced amputation. The US reported 311,000 amputees compared to 300,000 individuals with lower limb amputation in Russia. Interestingly, when comparing the percentages with respect to a country’s population the US stated approximately 0.11% of lower limb amputees with a greater result reported as 0.16% from within the Russian population. Less developed countries publish more generalised statistical data regarding numbers of amputees per head of population. Established from a report published by the National Sample Survey Organisation of India in 2003 an estimated 0.077% out of the all total population in India are nonspecific amputees, which when extrapolated corresponds to an excess of 950,000 individuals when projected to the current population. Utilising an earlier small study from Narang and Jape (Narang 1982), it was possible to predict a prospective percentage between categories of amputees for the population of India as a whole. The research on 14,000 amputees treated in Pune, India, and discovered that 62% were lower-limb amputees, among which 43% were unilateral above-knee amputees, 1.4% were bilateral above-knee amputees, and 2.2% were bilateral amputees with one above-knee and one below-knee amputation. Consequently, simple correlation predicts estimated values of 440,000 above knee amputees in India in (Narang 2013).

Approximately 199,000 persons in the U.S. were using an artificial limb in 1996, with the majority using an artificial leg or foot (173,000) (Data source: National Centre for Health Statistics, Disability Report). Data from the National Health Interview Survey in U.S., estimated that in 1996, approximately 1.2 million persons underwent limb amputation (Adams et al. 1999), this estimated number escalated to 1.6 million in 2005 (Ziegler-Graham et al. 2008) with approximately 1.03 million lower limb amputees. This high number reports that one in 190 Americans has undergone a limb amputation. Of these, about 45% were traumatic amputees with male amputees' number triple that of females. Amputations secondary to dysvascular disease account (54%) for most cases (Table 2) (Ziegler-Graham et al. 2008).

Table 2. Lower limb amputees: year 2005 in US.

Year	All causes in thousands	Trauma	Dysvascular	others
2005	1027	207	806	14

In support of the relatively high percentage of traumatic amputees, the US Department of Defence conveyed that in February 2008, in excess of 1000 (1,031) individuals endured amputations, of whom 730 had medical procedures involving major limb amputations (Polly et al. 2004). These statistics are biased in relation to the general population due to the nature of the injuries sustained by military personnel whilst on active service. The majority of the 1,031 total amputees, acquired their serious wound while serving in the Army and to a lesser degree in the Marines (Fischer 2008). The literature suggests that prolonged military actions have reduced the mean age of amputation incidence (Pitkin 2009). Recent statistics (April 2012) provided by the US military integration database indicated that almost 1500 wounded US service personnel required suffered major limb trauma and required an amputation procedure. Tragically, almost a third (438) experienced multiple limb loss, compared to approximately two thirds (1015) of the injured sustaining a single-limb loss. The importance of including the data as a separate set is to be mindful of the fluctuation in numbers of amputees when a country is deploying its military in hostile situations. Note: these traumatic amputations characterise more than 2% of all battlefield injuries and more than 7% of major extremity injury is associated with military service. Not all theatres of war provide the same data with respect to amputees. Specific to wars in Iraq and Afghanistan, the number of amputees is variable between studies (Petersen et al. 2007)

Bearing in mind that this number is multiplied when referring to the civilian population. Unfortunately, military actions and terrorism in Middle East brought more amputees who have not been recorded officially. Theatres of war have provided an estimated number of the amputees based on the records of Iraqi ministry of health stated that there are more than 3 million amputees; a third of them was lower limb amputees with a majority of young active amputees. Elderly or less active amputees are likely to be offered a prosthetic limb incorporating a manual locking knee with a non-articulated foot with basic level of function and shock absorption, such a prosthesis provides stability and requires slight controlling efforts. Since most recent developments in prosthetic technology are about presenting an advanced functionality level that requires a great deal of control/input from the user, it likely to be irrelevant to use them with sedentary or limited household ambulators (i.e. able to walk in the home but limited by endurance, strength or safety (walks rarely in the home/never in community). Thus a ‘high-tech’ prosthetic limb is targeting healthy young amputees where in most cases trauma is the common reason of amputation and they are otherwise healthy.

2.2.1 Key conditions associated with lower limb amputations and amputation rates

Peripheral arterial disease is the leading cause of leg amputation in developed countries and is second only to trauma in developing countries. In the UK for instance, the main causes of acquired amputation include vascular and circulatory disease as poor circulation in the limb due to arterial diseases about (70%), with more than half of all the amputations occurring among people with diabetes mellitus. In the US, of people with diabetes who have a lower extremity amputation, only 45% do not require amputation of the second leg within 2-3 years (Pandian et al. 2001). Amputation of a limb may also occur after a traumatic event (23%) or for the treatment of tumour (4%), or congenital conditions (3%). Approximately 86% limb amputees refer to amputation of the lower extremity (Dillingham et al. 2002). Slightly more than half of all lower extremity amputees are either TTAs or TFAs. With twenty-six percent of lower extremity amputees or approximately 360,000 individuals have a trans-femoral level amputation, ninety-five percent of trans-femoral amputations are associated with vascular disease and diabetes-related vascular disease. The remaining five percent of TFAs are attributable to trauma, malignancy, and congenital limb deficiencies. About 72% of transtibial amputations in the US are attributable to the vascular disease, of the remaining 18%, 7% of TTA's are the result of trauma(Dillingham et al. 2002). Dysvascular related amputation has grown in incidence and prevalence with advancing age (Dillingham et al. 2002). Regarding incidence of lower limb amputations by age, sex and diagnosis of diabetes, a study undertaken by Fosses et.al 2009 showed that, in France 2003, irrespective of the cause of amputation (i.e. with/without diabetes), LLA incidence increased with age and was almost twice as high in men compared with women(Fosse et al. 2009a). The highest incidence rate ratios between people with and without diabetes were observed in men aged 35–55 years and in women aged 30–65 years (Figure 3).

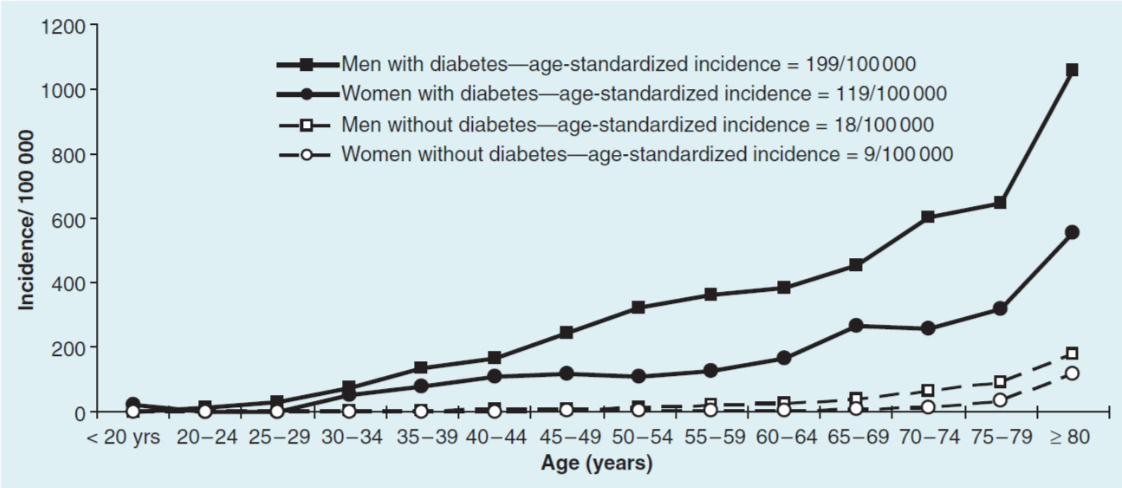


Figure 3. Incidence of lower limb amputations by age, sex and diagnosis of diabetes; France 2003 (adapted from (Fosse, 2009))

This study also showed that lower limb amputation was more frequent in men, in the elderly and in people with diabetes compared with those without diabetes. However, the characteristics of LLA are different, which may reflect differences in the pathological process that led to amputation. In people with diabetes, amputations are more often performed at a distal level, even if a second amputation occurs more frequently (16% of people with LLA have another LLA during the same year) but are often disabling: more than one-third of all LLAs affects the leg or the thigh (Fosse et al. 2009). Lower limb arterial investigation and a lower limb arterial revascularisation during 2002 or 2003 were reported in 55% and 33% of people with diabetes and LLA, respectively (Fosse et al. 2009, in France 2003). Arterial investigations related to multiple amputation reported; one amputation in 2003: 53%; two amputations: 66%; three or more amputations: 73% were undertaken; and somewhat more often when the amputation was above the toe level: toe 52%; foot 57%; below- the-knee 56%; above-the-knee 58% (Fosse et al. 2009).

Amongst the civilian population in the UK, trauma accounts for 7 to 9% of the 5000 amputation procedures performed every year (NASDAB 2007). In comparison, the USA, trauma accounts for 16 per cent of annual amputations, whereas traumatic amputees represent 45 per cent of people living with an amputation. The reason behind this unexpected discrepancy is due to the prevalence of traumatic amputation. The higher prevalence compared to vascular amputees, skews the predicted data as traumatic amputees are typically young with long life expectancy compared to vascular amputees, as traumatic amputees are typically young with long life expectancy (Perkins et al. 2012). The incidence of Western world-amputation and aetiology in developing countries are related to trauma more than vascular diseases. Traumatic amputation is the main trend in developing countries (Godlwana et al. 2008) primarily due to wars and to lesser extent due to injuries related to infrastructural challenges. Possibly due to an increasing population and rapidly expanding economy with relatively limited medical care and surgical provisions.

2.3 Prosthetic knee function and design development

Progressive development of prostheses technology from one generation to another is mirrored in the transition from simple crunch body weight support, to computerised prosthesis designed as an attempt to mimic the function of a physiological limb and restore normal gait pattern by actively sensing/analysing data on user's movement, activity, environment and terrain; providing a coordinated stream of instructions to the hydraulic/pneumatic support system. In response to load, position, and/or velocity sensors, MC knee resist or allow flexion. Most of the recently developed prosthesis allows a dynamic adaptation of the prosthetic limb under electronic control that provides natural gait and active lifestyle with a chance to perform a variety of daily activity (Kaufman et al. 2007; Sawers and Hafner 2013) . As an individual facing an irreversible loss (i.e. amputation) of naturally-accurate functioning lower limb, mobility and comfort are what matters most (Legro et al. 1999). For a TFA, the primary functions required from a prosthetic knee device are; stance phase stable support, smooth transition to and controlling of swing phase (i.e. simulate the action of muscles) (Silver-Thorn and Glaister 2009), plus, unrestricted knee flexion necessary for movement such as; sitting, bending, kneeling, etc. (Smith and Michael 2004). As a universal design for an above-knee prosthesis the components are; a socket and a liner which interface between the residuum and the prosthesis, a prosthetic knee device, shank (or shin) connecting the knee to prosthetic foot, and distally is the prosthetic foot-ankle device.

Knee design can be a simple device developed to permit a range of motion that meet the minimal requirements of swing phase, sitting, and kneeling, but provide no other functions or controls. Despite the variety of functions and features afforded by these mechanisms, it is possible to classify them on the basis of certain primary functions; knee axis of rotation (flexion/extension), and resistance to knee rotation in swing phase and/or in stance phase (Stewart and Staros 1972). According to the axis of rotation, prosthetic knees can be a single-axis knee (Figure 4) or polycentric (i.e. multi-axis) knee (Figure 4 and 5). Knee devices that are commercially available perform swing and/or stance control resistance by mechanical friction or resistance to fluid flow (i.e. pneumatic or hydraulic knee unit).

Knee mechanisms fall into different categories; constant friction, pneumatic, hydraulic, locking, polycentric etc. Generally, the most inherent stability can be provided by manual-locking knees, followed by many polycentric knees, weight-activated friction knees, constant-friction knees, and lastly, outside hinges (Silver-Thorn 2002). These designs generally provide non-adaptive resistance at the knee, thus cannot perfectly accommodate differences in walking speed and/or changes in terrain. All types mentioned above can be classified under the category of non-MC prosthesis as per sections below (section 2.3.1 to section 2.3.3). A more sophisticated controlling mechanisms of knee devices known as adaptive or MC prosthesis has the ability to adapt the character and timing of the stance and/or swing phases of gait and other activities under the MC automatic control (Kahle et al. 2008). Some MC hydraulic knee-prosthesis can provide stance yielding function at the knee along with swing-phase alterations (Zahedi et al. 2003) will be discussed in detail in section 2.3.4.

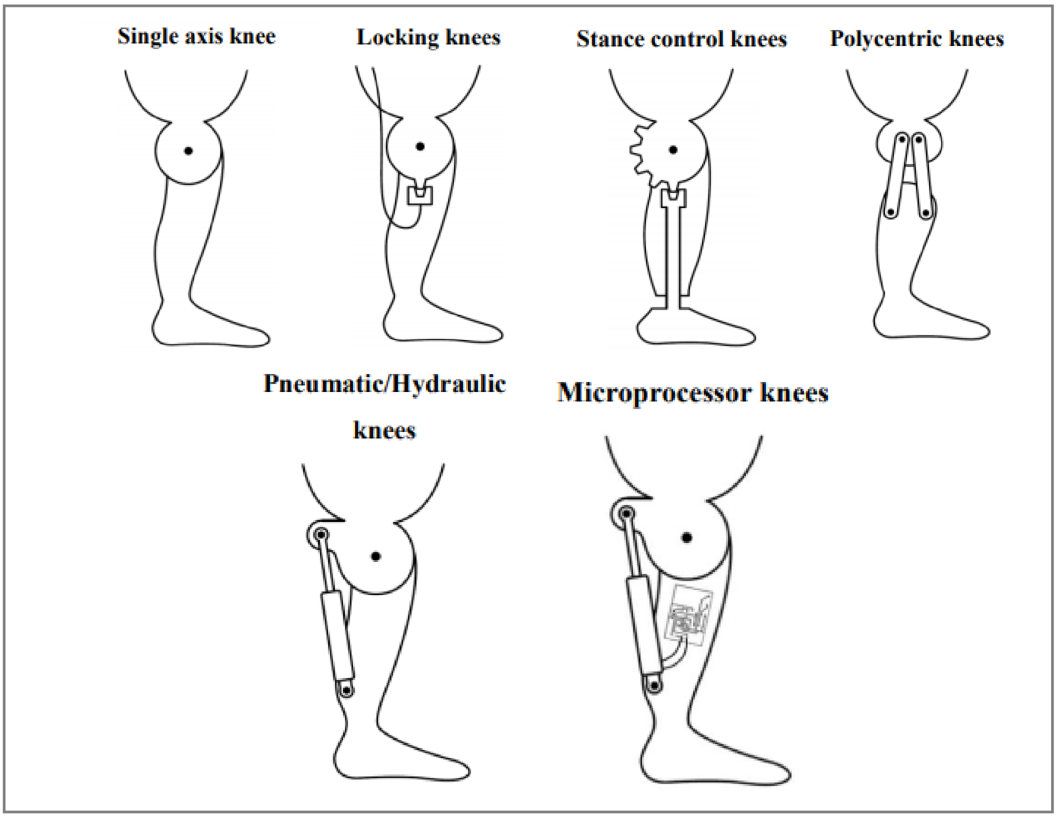


Figure 4. A mechanical prosthetic knee includes characteristics from one or more of the illustrated types of knees; single axis knees bend freely, locking knees are kept straight during walking, stance control knees are self-lock at any angle when loaded, polycentric knees implement a moving centre of rotation, and pneumatic or hydraulic knees adjust swing and stance flexion rate. (Adapted from(Lambrecht, 2008))

2.3.1 Constant friction (i.e. swing control) prostheses with locked knee

Single axis is the most basic prosthetic knee included a single pivot between the thigh/socket and shank. During swing phase, the basic hinge design with constant friction control to allow the limb to swing (Michael and Bowker 2004). With such design of prosthetic knee; the lower leg acts as a pendulum with rate of swinging limited by its length; the amputee has to walk with a fixed slow speed (Michael and Bowker 2004). The mechanical friction settings wear rapidly and must be frequently readjusted. In spite of their simplicity these knees are not a good choice for older patients in a generally weakened condition because of their relatively poor stance-phase control (Stewart and Staros 1972). Another disadvantage of single-axis, constant friction design is that appearance is normal at only one speed of walking for a given setting of friction, that amputee must be very careful in walking, especially on uneven surfaces, to avoid stumbling (Muilenburg and Wilson 1989). However, constant friction swing phase control knees are low in initial cost and their repair and maintenance are relatively simple and inexpensive (Dupes et al. 2004).

Stance stability is dependent on alignment stability (involuntary control) (Oberg 1983) and amputee's muscle contraction (voluntary control) (Muller 2016). Stance stability of the constant friction knee is obtained when the alignment of the knee axis is posterior to the vertical weight reference line drawn to the ankle from the trochanter (Radcliffe 1994). The pivot is located posterior of the load path through the leg thus, as long as the leg is straight at heel strike, the knee will not buckle during stance. Therefore, alignment of the prosthesis makes the knee axis posterior to the load line which force the knee into extension and locks it (Silver-Thorn 2002). However, such a location of knee axis result in difficulties to flex the knee under the load at push-off (Oberg 1983). Amputees using a non-locked single axis knee can control the direction of the load line as observed in the active use of the flexion-extension musculature about the hip joint of the stump (i.e. voluntary control of knee stability). The amputee may need to exert more hip extension torque to keep the knee over centre during mid to late stance. Thus, these knees require good voluntary control of the hip musculature (Silver-Thorn 2002). A TFA with a weak hip who is unable to exert the necessary muscle effort would obviously have greater difficulty in maintaining knee stability without dramatic changes in alignment stability or installing a brake type mechanism (Radcliffe 1994). Foot initial contact is the most critical period of the stance phase for knee security. Failure to fully extend the knee as a preparation to heel contact plus the lack of knee locking that counteract the flexion moment caused by the applied body weight will cause the prosthetic limb to buckle suddenly as weight is applied (Radcliffe 1994). As a result, the amputee will not be confident that the knee remains stable throughout the stance phase and gait deviation might be observed. Absence of sufficient stance control, causes lack of inherent stability (Silver-Thorn 2002) (Radcliffe 1955). Therefore, stance phase stability is considered as a key aspect of knee functional performance.

Despite of prosthetic alignment, stability control during the stance phase can be obtained using mechanical approaches including locking the knees, weight activated and certain hydraulic knee. Constant friction knee with manual locking preserves a straight position while the amputee is walking. Manually locked knee prevents rotation (i.e. flexion) at the joint during stance phase. Amputee must perform hip-hiking, vaulting, circumduction, or abduct the prosthesis to clear the floor which result in abnormal gait and believed to increase the energy cost of ambulation (Michael and Bowker 2004). Bending the knee in case of sitting requires the amputee to manually disengage the locking mechanism by pulling a lever or cable (Stewart and Staros 1972). The locked knee is recommended for the older amputees as it enables a relatively higher walking velocity and lower effort compared to unlock condition (Isakov et al. 1985) with enhanced gait stability. Nevertheless, it can result in walking speed variations, awkward stiff gait pattern with the operator manually locking/unlocking the knee (Silver-Thorn 2002); particularly for young amputees due to their healthier physical condition, their preference would be an unlocked-knee (Isakov et al. 1985).

2.3.2 Knee prostheses with stance and/or swing control

In terms of their controlling mechanism, prosthetic knee units were classified (Stewart and Staros 1972; Silver-Thorn 2002) as; non-mechanical knee control (locked knee) (as explained in the previous section), swing phase control only, stance phase control only, and swing and stance phase control (the hybrid design) (Dupes et al. 2004). The swing control component of a knee mechanism acts as a mechanical analogue of the quadriceps and hamstring muscles acting to dampen the swing of the knee at the extremes of flexion and extension (Stewart and Staros 1972). Stance phase begins with contact between the heel and the ground at a time when a large proportion of the body weight is still on the other leg and well behind the knee centre. Consequently, there is a bending moment about the knee tending to collapse it or drive it into flexion during early stance phase (Highsmith et al. 2014). Mechanisms of stance stability (manual or weight-activated locking systems) works as a substitute solution of the missing action of the quadriceps and helps to improve prosthetic limb functions hence promote stability (Stewart and Staros 1972).

Stance control knee is a single axis knee which self-locked by means of a friction brake (Muller 2016). When fully loaded with the amputee's body weight during stance, the friction brake engages, preventing the knee from buckling. The weight-activated friction brake controls knee (Figure 7) flexion at heel contact and continues throughout the stance phase via body weight application (Michael, 2004). Knee locking operates cyclically under the control of applied axial loads, inertia or other forces to prevent knee flexion (Stewart and Staros 1972; Silver-Thorn 2002). Once the load is removed, the brake disengages, and the shank swings freely. Although the knee improves safety compared to a single axis knee by locking at a range of angles, it does not allow knee flexion in terminal stance to initiate swing until the knee is fully unloaded. To flex the knee as a preparation to swing phase, the amputee must shift body weight onto the other limb to fully unload the prosthetic side which in turns release the braking mechanism so that leg swing freely (Silver-Thorn 2002). This stability mechanism comes on expense of normal cadence since the lack of knee flexion under partial weight bearing during late stance can disturb gait pattern. Excessive knee stability is a condition in which the knee of the prosthesis is so stable and resistant to flexing that it is difficult for the amputee to initiate the knee flexion required to achieve toe-off and swing of the shank. An increased energy expenditure and an unnatural swing phase of the gait cycle are the results (Bowker 1992). Stance control knees are appropriate to limited users who are capable to walk only at slow pace (Michael and Bowker 2004). A higher degree of resistance to prevent knee rotation than normally available can be achieved with yielding resistance in stance phase, designed to reduce the rate of knee rotation under load (Stewart and Staros 1972). In many cases, the limb incorporating a compressible rubber snubber at the knee that allows a slight amount of knee flexion during mid-stance so that the amputee does not have to walk over the locked knee.

In addition, stance control is also achieved by having a moving centre of rotation provides prosthetic knee stability (Michael 1999). Polycentric knees (Figure 5) incorporate a four-bar mechanism moving in the sagittal plane to attain a moving centre of rotation. The centre of rotation changes instantaneously from more proximal and posterior position, in stance phase this permits stability in early stance phase to more distal anterior position during swing phase and provides adequate knee flexion for swing phase (i.e. flexion) (Stewart and Staros 1972; Oberg 1983; Radcliffe 1994). Thus, near the straight position of the knee, the effective pivot lies well posterior of the load path for safe early stance loading then during terminal stance, the polycentric knee buckles easily to initiate swing (Radcliffe 1994). Polycentric knee designs offer biomechanical advantages of relatively compact design; minimising leg-length discrepancies for knee disarticulation and allowing weight-activated stance-control function similar to a single-axis knee during swing with a braked knee during stance.

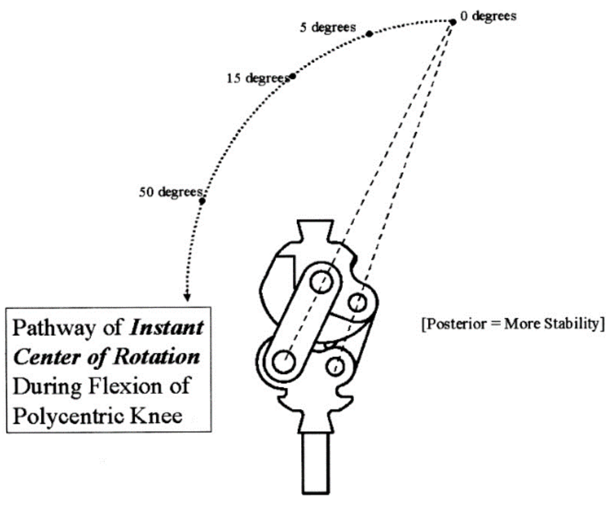


Figure 5. The four-bar polycentric knee prosthesis with instantaneous centre of rotation located at the intersection point of the virtual lines defining the anterior and posterior articulations. The dotted curve refers to the pathway of the instant centre of rotation as the knee is flexed (Adapted from (Michael, 1999)).

2.3.3 Fluid-controlled knees (stance and/or swing controlled prostheses)

Prosthetic knee devices included fluid resistance control mechanisms. Pneumatic/hydraulic control refers to pistons inside cylinders. The fluid mechanisms consist of a fluid-filled cylinder joined to the knee bolt by a piston, typically located in the posterior aspect of the shank. The contained air/oil acts as a damping medium thereby the amount of fluid being transferred between the sides of the piston determine the level of heel rise. During swing phase, the resistance to knee flexion is produced as the piston in the cylinder pushes against air or oil. The resistance to swing triggers a knee extension bias that then assists the prosthetic knee into extension. For hydraulic knees (oil as a medium), the fluid is incompressible. The resistance to piston motion results from fluid flow through one or more orifices. As such, the resistance is dependent on the fluid viscosity and density, the size and smoothness of the channel, and the speed of movement. In contrast, for pneumatic (air medium) knees, the fluid is compressible. The resistance is again due to fluid flow through the orifice(s) but is also influenced by fluid compression.

As an improvement overcame of the limitations of the previous reviewed designs (Staros 1964), fluid dynamics has been employed as a mean of controlled stability to ensure an adaptable resistance enabling amputees to walk comfortably at different speeds. Thus, a prosthesis with an advanced mode of swing-phase control, either by a pneumatic or a hydraulic knee unit, is somewhat better than a prosthetic knee that only provides constant friction (van der Linde et al. 2004). A primary limitation to the pneumatic knee is that the inability to provide sufficient resistance to vigorous activity due to the compressible nature of gas (Michael and Bowker 2004). Even though, pneumatic units tend to be lighter than hydraulic units due to air being less dense than oil; the reduced density and viscosity of the air utilised in pneumatic units provide less cadence control than hydraulic units. Additionally, fluid viscosity is influenced by temperature, hydraulic (and pneumatic) knee units may perform differently inside and outside in cold weather climates (Silver-Thorn 2002). Participants found bending mechanical swing phase knee (Otto Bock 3R20 knee) was very easy compared to pneumatic swing phase control (Tehlin knee) which led to an unsafe feeling. Furthermore, walking with Tehlin knee pneumatic swing phase control was easier and/or faster although required more energy (Boonstra et al. 1996).

Compared to the pneumatic approach, hydraulic systems provide a much higher swing phase resistance as necessary using incompressible liquid to control knee motion (i.e., high viscosity more responsive to variant cadences) and some knee types are also designed to provide hydraulic stance stability (Michael and Bowker 2004). During swing phase, normal heel rise and gradual smooth forward swing that matches the amputee walking speed are advantages provided by the hydraulic system (Dupes et al. 2004; Michael and Bowker 2004). Further, stance control is integrated to enhance the role of the hydraulic knee by providing resistance to sudden flexion that allows a slow flexion in the case of downstairs walking. Although hydraulic knees provide a more natural smooth gait (Zahedi et al. 2003), in comparison with other knee systems hydraulic units are heavier, require more maintenance, and are more expensive to purchase and maintain (Dupes et al. 2004).

2.3.4 MC knee prosthesis

MC knees devices are an advanced category of prosthetic knee components which has become more widely prescribed in the last two decades. There is a requirement for a time/function synchronised artificial limb that satisfies amputee’s ambition to restore the normal range of limb function necessitated employing the MC knee device for this purpose. A variety of MC knees are available with a control afforded by the MC differs among knees. Some MC knees regulate only the stance behaviour of the knee (e.g., OttoBock Compact); some regulate only the swing behaviour of the knee (e.g., Endolite IP+), while others regulate both the stance and swing behaviour of the knee (e.g., OttoBock C-leg, O¨ ssur Rheo). MC control involves electronic sensors that are developed to detect the movement, timing and accordingly adjusts a controlling hydraulic and/or pneumatic cylinder(s) via hydraulic servo valve and/or pneumatic needle valve (Buckley et al. 1997; Thiele et al. 2014). For every gait situation, the collected real-time data determines which setting (i.e., damping rate) are needed to be used. Therefore, the amount of effort required by amputees to control their timing will be reduced by the control MC (Hafner et al. 2007; Kahle et al. 2008; Hafner and Smith 2009), enhance a more natural gait (Dupes et al.

2004; Johansson et al. 2005), likewise influence their physical activity during daily life (Kaufman et al. 2007).

The development of MC knees started with implementing swing phase into the prosthetic knee devices before stance phase control. The first commercially released MC knee prostheses (The Blatchford Intelligent Prosthesis (IP) knee) was a swing phase control unit in the early 90s (Zahedi 1993). MC swing control knees provided a range of biomechanical advantages (beyond the scope of this thesis); improved gait pattern and influence metabolic energy expenditure than that of the conventional prosthesis (Taylor et al. 1996; Buckley et al. 1997; Chin et al. 2003). However, stability and security during stance phase are considered as an essential issue. A safe, consistent walking pattern for TFAs can be achieved by synchronising variable stance control (i.e., adaptive resistance) so that a knee joint is not vulnerable to collapse as the shank rolls over the artificial foot as being loaded. Stance phase control was also under the research scope to make stance stability available along with the swing phase control. Until the late of the 1990s, the C-Leg (Otto Bock HealthCare) was first commercially available MC knee device monitoring both swing and stance phases of gait. The design of C-Leg included a sensor network that measures the ankle moment as well as the knee joint’s angle and angular velocity 50 times/second (Burnfield et al. 2012). Thus, additional to a primary function, i.e., stability in stance, agility in swing, MC stance swing control allow continuous "fine-tune" of the prosthetic knee's intrinsic behaviours over the conventional non-MC knees.

MC prosthetic knee technology is designed to; be sensitive to the user's instantaneous mobility requirements (i.e., enable him or her to change cadence and walking speed) and adjust its behaviour (i.e., provide more or less knee-flexion resistance) in response to extrinsic conditions, thereby mimicking the action of a nondisabled knee (Sawers and Hafner 2013). The inclusion of a MC into the prosthetic knee unit is believed to provide a number of hypothetical benefits to the amputee, including improvements in balance, confidence, uneven terrain ambulation, stair descent, incline negotiation, and overall activity as well as reducing the likelihood of adverse events; stumbles and falls. Previous research demonstrated controversial evidences regarding differences across the gait biomechanics with and without MC knee prosthesis (Segal et al. 2006; Sawers and Hafner 2013). MC inclusion with the prosthetic devices provided data to suggest that automatic adjustment to fluid resistance provided more normal gait kinematics (Kirker et al. 1996), more energy- efficient gait (Taylor et al. 1996), improved gait mechanics as terminal stance can be identified early, so swing phase flexion can be initiated more easily (Schmalz et al. 2002). Furthermore, adapting flexion and extension impedance to increase swing phase speed improves walking symmetry, walking speed, and comfort (Boonstra et al. 1996). MC biomechanical benefits will be discussed in detail later in the current chapter section 2.7. MC knee controls the hydraulics during both stance and swing phase sometimes exceed the needs of amputees with a certain level of activity (Burnfield et al. 2012) such as those expected to maintain a relatively constant walking speed (e.g., K2 level walkers). Therefore, C-Leg Compact was introduced by Otto Bock HealthCare in 2002 to address the needs of deconditioned K2 level walkers. This prosthetic knee includes sensor network modulating only stance mechanics but controlling the swing phases done hydraulically. The hybrid MC knee prosthesis is designed for individuals expected to benefit from added stance phase knee stability, but whose invariable cadence limits the need for sophisticated swing control. Interestingly, TFAs exhibited significantly improved function and balance while using the stance-phase only MC knee compared to their traditional non-MC knee (Burnfield et al. 2012).

2.4 Considerations for prosthetic feet used by TFAs

The main requirement from the prosthetic foot is that providing shock absorption, weight bearing, and facilitating forward progression of the body (i.e. a rocker mechanism during stance phase (Perry and Davids 1992). Ankle-foot devices fall into a wide range of designs (Figure 6). Structurally, (according to the connection with the prosthetic shank), prosthetic feet can be divided into two main categories: those with a rigid connection to the prosthetic shank (non-articulated) and those with a hinged ankle mechanism (articulated) (Hopkins and Binder 2011). Depending on the mechanical characteristics of the different feet i.e. the presence or absence of an ankle axis in the sagittal/frontal plane and energy storing ability the non-articulated prosthetic feet were categorized into; Solid Ankle Cushioned Heel (SACH) (simulated planter/dorsi flexion motion of a rigid keel limited deformation), Elastic (flexible) Keel Foot (long elastic keel with limited tri-planar motion), Energy Storing (long elastic keel with maximised push off effect). Whereas articulated group are subdivided into; Single-Axis Foot (provides sagittal plane movement), Multi-Axis Foot with tri planer range of movement, and Dynamic-Response Foot that also allows tri planer range of movement in addition to maximised push off effect using long elastic keel (van der Linde et al. 2004; Hopkins and Binder 2011).

Prevention of knee buckling and preserving stability is a major concern in the gait of TF amputees (Cullen 1984). Knee stability originated via the adjustments done first at the foot and ankle complex (Radcliffe 1994). TFAs showed significant improvement with the hydraulic ankles/feet in the sagittal plane ankle moment. Bai suggested an enhancement of prosthetic knee stance stability when using hydraulic ankles/feet (Bai 2017). Providing a stable base of support during ambulation is the main general purpose of the feet. This is done through manipulating/redirecting the CoM so that it remains within the base of support during ambulation hence maintain gait dynamic stability (Shumway-Cook and Woollacott 2007). A dorsiflexed foot, a stiff plantar flexion bumper, or a firm heel cushion action are key design factors changing knee stability accordingly CoM dynamics. Each one of these factors promote the prolonged time of weight bearing on the heel demonstrated by TF amputees (Radcliffe 1994) and therefore less control of the CoP on body CoM and reduced dynamic stability. Prosthetic feet used for TTAs are similar in design (construction) compared to that used for TFAs (Rajťúková et al. 2014). Even so, further consideration of the prosthetic ankle mechanism (static correction, dynamic correction) should be taken into account when prostheses are prescribed for TFAs (McNealy and Gard 2008). As a general description/rule, a prosthetic foot of any type should provide three functions (Radcliffe 1955; Perry and Davids 1992); the first one is that shock absorption at heel contact that decreases impulse to the residual limb. The majority of available prosthetic knee designs do not allow amputees exhibit stance phase knee flexion (McNealy and Gard 2008). Ankle mechanisms (single-axis foot, multi-axis foot, or those with a soft heel) provides absorption of shock and dampens the torque generated at heel strike (Smith and Michael 2004). Thus, shock absorption at the prosthetic feet is crucial for TFA and can compensate for the reduced ability of shock absorption at the prosthetic knee. The second feature is stable/safe weight bearing without limb collapse by simulating the smooth transition occurred on the ball of the foot during propulsion of the physiological foot. Gait stability determined by ankle stiffness (Kamali et al. 2013) is necessary during loading body weight to avoid the collapse of the leg at dorsiflexion or heel strike; at plantarflexion or toe off. Along with the above two required features, a third essential design requirement of the prosthetic feet is assisting the transition from stance to swing phase via simulating foot rockers.

A prosthetic ankle with an improved sagittal range of motion can enhance this transition, therefore, gait comfort and functionality in persons with trans-femoral amputations (McNealy and Gard 2008) which makes this type favourable for sedentary or elderly amputees (Smith and Michael 2004). On the other hand, an articulated ankle/foot can offer a range of dynamic stability during movement activity. A single axis (sagittal plane movement) might be more stable compared to the multi-axis (3-planar movement) foot. Accordingly, less energy compared to the multi-axis foot would be required to stabilise body weight during gait (Smith and Michael 2004). The articulated foot might be more beneficial for active amputees who can control such prosthesis. However, some studies highlighted that inactive prosthetic users might also benefit from an early foot-flat mechanism to facilitate weight transfer onto their prosthesis (Goh et al. 1984; Perry et al. 1997; Postema et al. 1997). The idea is that some prosthetic feet with an ankle axis in the frontal plane, such as the single-axis Lager foot (Otto Bock), would mimic the normal roll-off motion of the ankle-foot complex in the sagittal plane, allowing an early foot-flat position and concomitant early-stance-phase stability (van der Linde et al. 2004). Increased plantar flexion is probably welcomed by the TFA since it will give more stability at heel contact (Radcliffe 1994). A prosthetic foot that achieves a quick foot flat/plantarflexion initiates knee stability (Struchkov and Buckley 2015) which in turn linked to driving residual limb.

Similar to knee prostheses, MC control was also integrated into ankle foot devices establishing another category which is the MC ankle-foot device. The controlling mechanism of the MC foot has sensors that continuously determine the angle of the terrain being walked on and the cadence of walking. This information is used to automatically alter the hydraulic damping to pre-defined settings to facilitate a more optimal foot-ground interaction for the terrain or cadence.

2.5 MC limb system prosthesis

Traditionally MC knee devices have been incorporated into a prosthetic limb containing a foot with either no ankle or an ankle device incorporating a rubber snubber providing a small range of elastic motion (Vrieling et al. 2008b; Bellmann et al. 2010; Bellmann et al. 2012; Villa et al. 2015) . TFAs showed a crucial demand for both prosthetic knee and foot functions. Ankle-foot devices that have a MC hydraulic ankle have been used during the last few years. Such devices also have adaptive resistance as highlighted in the previous section that automatically decreases the plantar-flexion resistance to help attain foot-flat following ground contact and then increases dorsi-flexion resistance to control how quickly the shank-pylon progresses over the foot. Controlling how quickly the shank-pylon progresses over the foot also controls the lowering of the CoM down the ramp. Use of these types of ankle-foot devices have been shown to facilitate/improve how TTAs walk down slopes in comparison to using the same foot-ankle device but having hydraulic resistances at the ankle at constant default levels (Fradet et al. 2010; Agrawal et al. 2015; Struchkov and Buckley 2015).

A combined and simultaneous MC control of the hydraulic resistances of the ankle and knee has become recently incorporated as an above-knee limb system prosthesis (Figure 6). The development of this new prosthetic limb system, necessitated more research on prosthetic knee and ankle together as an integrated system designed to enhance amputees' gait. This limb system - called Linx (Endolite, Chas A Blatchfords) - has several modes in which the hydraulic resistances at the knee and/or ankle are simultaneously altered in response to a change in terrain and or a change in walking speed as highlighted in the thesis introduction. A preliminary study on TFA participant investigated limb system benefits during down-slope walking compared to 'conventional prosthesis' (Tang et al. 2015) using joint limb work showed that with 'ramp descent mode' there was an increase in joints work on the prosthetic side (including the residual hip). This made the gait safer compared to MCoff condition and caused a reduction in intact limb compensatory joint work (at the knee and hip only).

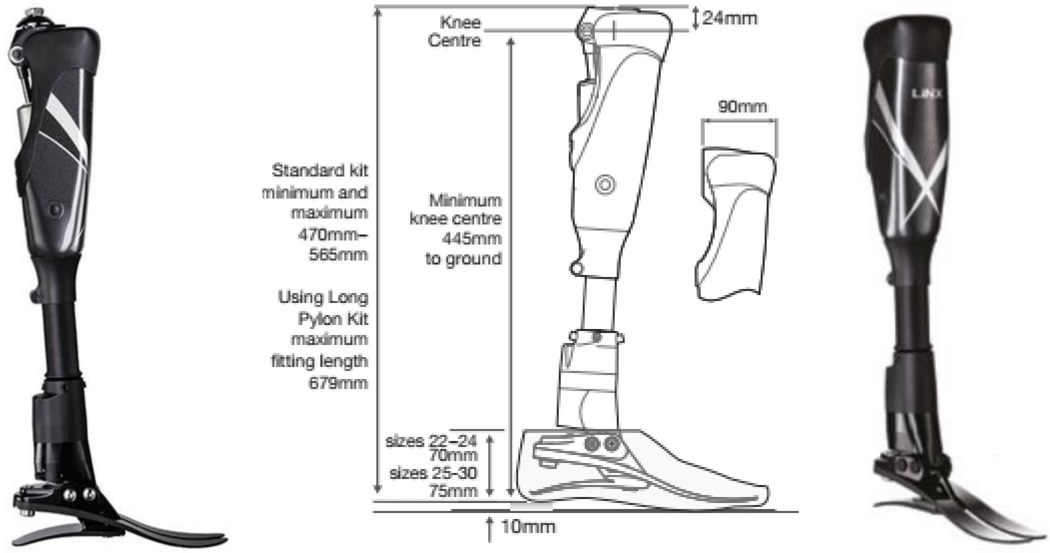


Figure 6. The Linx limb system (adapted from Endolite, Chas A Blatchfords).

2.6 Amputee gait biomechanics

Limitations in prosthesis function compel amputees to perform adjustment strategies and lead to gait deviations (Silver-Thorn and Glaister 2009) that involve the prosthetic and/or the non-affected limb. Amputees walk slower with asymmetrical temporal–spatial measures and impact forces as a result of these compensatory strategies. Gait adjustment strategies are employed as an attempt to preserve prosthetic limb stability (Gailey et al. 2008). Prosthetic limb stability can be reflected by the degree of differences/deviations in gait biomechanics for TFAs in comparison with that of AB individuals. Perhaps the most prominent gait issue/parameter thoroughly investigated is gait imbalance/asymmetry between the prosthetic and the intact side (spatiotemporal, kinematic and kinetic changes).

2.6.1 Changes in spatial and temporal gait variables

Measurement of stance as well as swing phase duration indicates the general prosthetic stability during walking (Murray et al. 1980; Silver-Thorn and Glaister 2009). Lower limb amputees typically spend more time in sound limb stance than in prosthetic limb stance (i.e. spatiotemporal asymmetry) (Silver-Thorn and Glaister 2009). More specifically, the double-limb-support phases of the intact and prosthetic side were not equal in most amputees. TFA patients walked with an asymmetrical gait pattern in which the successive stance phases were uneven; longer stance on the intact side then shorter stance on the prosthetic side (Zuniga et al. 1972; James and Oberg 1973; Murray et al. 1980). TFA individuals demonstrated an asymmetrical walking pattern when walking at self-selected walking speed (Nolan et al. 2003; Schaarschmidt et al. 2012). The prosthetic limb was found to be less speed-adapted compared to the intact limb, however, with rapid walking speed, asymmetry in temporal parameters was decreased (Nolan et al. 2003).

2.6.2 Changes in kinematic gait variables

For evaluating kinematics of amputees' gait, mid-stance CoM forward velocity and CoM vertical acceleration were compared for the prosthetic and intact limb. At the time of opposite initial contact, amputees seemed to have lower overall range in CoM forward velocity during prosthetic stance than during intact stance. The amputees usually exhibit a downward CoM vertical acceleration during prosthetic stance at the time of intact heel strike, rather than the upward acceleration observed during intact stance and in non-amputees (Kuo 2007). Gait deficiencies were also identified by the abnormal spatiotemporal distribution of CoP under the prosthesis (Schmid et al. 2005) which fails to move towards where it ideally should (Vrieling et al. 2008b; Vrieling et al. 2008b). Furthermore, stance phase knee flexion angle is one of the investigated factors determining gait stability. Instability of the knee and thus gait act to the created flexion moment as the body weight being posterior to the knee. If TF amputee is unconfident to put weight on the prosthetic leg, inefficient gait kinematics would be observed. To minimize the likelihood of knee buckling, TFA tends to prevent knee flexion (prolonged knee extension) in the first 30-40% of the stance phase (Ip 2007). Knee extension might give rise to the shortening of the contralateral step and an increase in the vertical displacement of the centre of gravity (Kuo 2007). One more predominant gait abnormality in the kinematics of TF amputations observed during the stance phase is ipsilateral trunk bending. This abnormality could refer to an inappropriately short prosthesis or weakness at the hip abductors on the ipsilateral side (Kishner and Monroe 2010).

2.6.3 Changes in kinetic gait variables

Kinetics of amputees' gait involves increased loading on the intact side and a decrease in the prosthetic side is a general feature of amputee's gait adaptation (Nolan and Lees 2000; Segal et al. 2006). As compensation of the functional losses, the body has to work harder i.e. the intact leg undergoes an increase in joint force moments, consequently consume higher energy (Nolan and Lees 2000). A deficit in most modern knee prostheses is that they do not flex during stance (Cochrane et al. 2001; Schaarschmidt et al. 2012) which can affect the transition of body CoM towards the intact limb (falling from relatively higher vertical position of CoM at the prosthetic side onto the intact side (Gitter, 1995; Gitter, 1995). Another potential explanation for this increased mechanical loading at the intact limb is that prosthetic limb push-off power and GRFs were reduced especially with conventional prosthetic feet (Winter and Sienko 1988).

As a TFA, the residual hip musculature is vital to normal gait and prosthetic stability. Extension and flexion moments at the residual hip were found likely to be affecting the stability of a prosthetic knee (Oberg 1983; Silver-Thorn and Glaister 2009). As the amputee is exerting a hip extension moment at heel contact and the load line moves in front of the knee joint, knee stability will be achieved. The more mechanically stable knee, the less utilizing of residual limb hip extensors responsible for maintaining voluntary stability at the knee to initiate stance phase (Stewart and Staros 1972; Oberg 1983; Radcliffe 1994; Silver-Thorn and Glaister 2009). Consideration of hip extension moment in maintaining stance stability and flexion moment in initiating swing phase can be used not only to compare the stability of two prosthetic knee designs but can also potentially be used to determine whether there is a compromise between stance phase stability and swing phase initiation effort (Schmalz et al. 2002). With an energetically passive prosthesis, TFAs have been shown to exert as much as three times the affected-side hip power and torque (Winter 1991), most likely because of the missing joints, or degrees of freedom, in the amputated leg. Kinetic and kinematic gait abnormalities are interrelated. Increased energy expenditure was associated with the increased the motion of the hips and body CoM demonstrated by TFAs (Gitter et al. 1995). Both deviations result in the increased energy cost during the motion with increased activity level of certain groups of muscles, and can cause damage to joint structures and skin.

2.6.4 Gait abnormalities during swing phase

Most gait deviation mentioned in the previous sections occurred during stance phase. Gait abnormalities associated with swing phase are distinctive from those of stance phase. According to Medscape, TFAs gait deficiencies during swing phase are limited in number (Kishner and Monroe 2010). One of these gait insufficiencies is stiff-knee gait pattern which results from excessive knee stability (locked knee) at the joint that causes difficulties in creating a flexion moment at the knee. Another deficiency occurs during swing phase is when moving the limb in a wide arc (circumduction), usually indicates poor suspension or excessive length of the prosthesis. Abnormal axis rotation at the knee that results in a whipping motion (Medial / Lateral Whip) is usually due to incorrect alignment of the prosthesis at the knee. An additional swing phase related issue encountered by the amputees and required adaptation is the toe clearance. Muscular loss and control of the prosthetic knee flexion complicate toe clearance for TFAs during level walking (Michael and Bowker 2004; Smith and Michael 2004). Several strategies are used on level surfaces to ensure toe clearance and avoid stumbling such as; vaulting of sound ankle (up on tiptoe during stance phase), increase of pelvis contralateral inclination to elevate residual hip or circumduction of residual hip (Perry and Davids 1992; Smith and Michael 2004; Schaarschmidt et al. 2012; Villa et al. 2015). Both intact and affected hips show abnormal movement; instead of negative pelvic obliquity towards the swing leg, amputee gait shows positive obliquity, indicating hip-hiking as an attempt to help foot clear the floor during swing (Michaud et al. 2000).

2.6.5 Gait biomechanics associated with different prosthetic feet

For a trans-femoral prosthesis, the most challenging period for knee control is initial contact (Smith and Michael 2004). Ankle-foot dynamics that reduce the knee flexion moment generated by GRFs at heel strike can be a significant improvement for amputees’ gait. McNealy and Gard 2008 found that the presence of the rigid ankle in the Baseline configuration and the passive function provided by the Multiflex Ankle configuration, combined with the lack of stance phase knee flexion forced the participants to spend relatively more time rotating their prosthetic legs forward until foot flat was achieved compared to the control group. A prosthetic foot design that permits true plantar flexion (i.e., closely replicate normal ankle- foot function) during loading response increases potential knee stability (Smith and Michael 2004). During walking, as foot approaches the floor with a slightly plantar flexion movement, Centre of Pressure moves quickly and smoothly forward from heel area towards the forefoot (Radcliffe 1994) hence improved CoM control can be obtained. A greater range in prosthetic ankle dorsiflexion/plantarflexion in addition to increase in first vertical GRF peak at the prosthetic side were found in comparison to the fixed ankle/foot (Bai 2017). Compared to Multiflex (conventional prosthetic foot), Vari-Flex feet (energy-storing foot) allowed faster walking and more symmetrical gait of TFAs during level ground walking at normal and fast speed conditions (Graham et al. 2007). Significant differences were found in fast walking speed and step length symmetry, prosthetic side peak dorsiflexion, and prosthetic ankle power at push-off. Articulated ankle devices help the amputee not to have the sensation of either “walking over a hill” or “lack of support during roll-over” (Radcliffe 1994). A prosthetic foot with actual moving joints achieves foot flat more rapidly than do the solid-ankle feet with the lack a moving joint, consequently, an articulated foot is therefore often preferable for the TFA (Smith and Michael 2004).

Although the increased ankle function with a hydraulically articulating versus rigid ankle joint component did not influence the yielding pattern at the knee, it reduced the joint moments at the hip of the residual limb in an individual with unilateral TF amputation (Alexander et al. 2017). This, in turn, may result in decreased energy consumption and subsequently a more efficient gait. Thus, TFAs walk with higher energy expenditure (Perry et al. 2004) and slow speed (Huang et al. 1979; Ruhe 2004) in comparison to AB individuals which can be associated with the prosthetic feet design with considerably reduced function than the physiological one (McNealy and A. Gard 2008). External work done on CoM also proved to be affected by the type of prosthetic foot articulation being used (Agrawal et al. 2009). The percentage of symmetry in the amount of work done by the intact limb and the prosthetic limb was increased during level walking at self-selected walking speeds, across four prosthetic foot designs. The CoM trajectories produced by the hydraulically articulating prosthetic feet have a distinct trough (negative displacement) followed by a peak (positive displacement), which is comparable with a typical displacement curve of an able bodied individual walking at a similar speed. While ambulating with the SACH and Seattle feet, the negative CoM displacement was not evident during the step with the intact limb. For both feet, the intact limb produced a higher CoM excursion than the prosthetic limb (Agrawal et al. 2009).

2.7 Terminating gait

One of the fundamental movements in everyday activities is gait termination. Gait termination is a process that necessitates controlling balance and arresting body forward momentum to achieve a secure transition from steady state movement to a standing stance posture (Hase and Stein 1998). Maintaining whole body balance without falling during the transition between steady state walking to standing can be challenging. High incidence of falls occur during routine locomotive movements and transitions from one posture to another, particularly in the elderly and/or those with certain neurological disorders (Fleming, 1993; Masud, 2001). CoM displacement is continuously adjusted in the horizontal direction by means of the Centre of Pressure (CoP) displacement which represent a change of instantaneous location where GRFs being acting. Gait termination needs the central nervous system to predict the future position of the whole-body CoM and control the development of the required braking forces to arrest the forward velocity of the CoM. As such it might be considered as a more challenging task than walking (Meier 2001) since it requires dissipation of kinetic energy (negative work) to achieve a stable and thus successful gait termination (Hase and Stein 1998). Gait termination studies showed that dynamic postural adjustments must bring the CoM into the stability boundaries (Remelius et al. 2008). CoM location is outside the BoS at the instance of final trailing foot toe off and prior to heel strike. But standing posture can be accomplished as long as the horizontal momentum of CoM is directed by CoP towards the BoS. Arresting forward momentum by generating enough braking impulse can be considered as a principle postural adjustment for balance preservation during gait termination in healthy individuals. For patients with cerebellar diseases, as an attempt to maintain stability during arresting forward momentum another biomechanical compensation has been shown (Conte et al. 2012). This compensation to maintain the safety margin between CoM and CoP in both the frontal and sagittal planes included; slowing the speed, increased number of steps to stop safely, and increase step width to overcome the CoM forward acceleration (Conte et al. 2012). Compared to matched healthy elderly group, elderly diabetic individuals are closer to their stability limit (i.e. less tolerate to perturbations that required quick adaptation). CoP maximum displacements relative to the motion area centre (Figure 7) and CoP trajectory area were larger, demonstrating that the elderly diabetic individuals have difficulties in precisely coordinating their final stopping phase (Meier, 2001) hence difficulties bringing their CoM into stability margin and thus have a higher risk of falling.

The shape of CoP trajectory during termination seems to be a mirror image of that of initiation (Winter, 1995) and can be divided in to specific phases as that of termination trials (Jian et al. 1993). A typical CoP trajectory for AB gait termination is presented in figure 8c. During final step when the leading left foot contacts the force platform, the CoP shows forward followed by medial displacement relative to lead foot (Jian et al. 1993; Vrieling 2008). This pattern is accompanied with forward movement and a sideways shift of the CoM towards the swing limb (Ryckewaert et al. 2014). Characterising the CoP trajectory during gait termination has been done by splitting the movement into a number of key phases (Winter 1995; Meier 2001; Vrieling 2008; Vrieling 2009; Ryckewaert et al. 2014). Firstly, the initial heel contact of the leading foot to the toe-off of the trail swing foot (CoP anterolateral displacement under the final leading stance foot). The next phase starts from trail toe-off to the heel contact of trail swing foot (the medial movement of CoP trajectory towards the trail limb forefoot area), followed by small medial backward shift to reach the bipedal stance position.

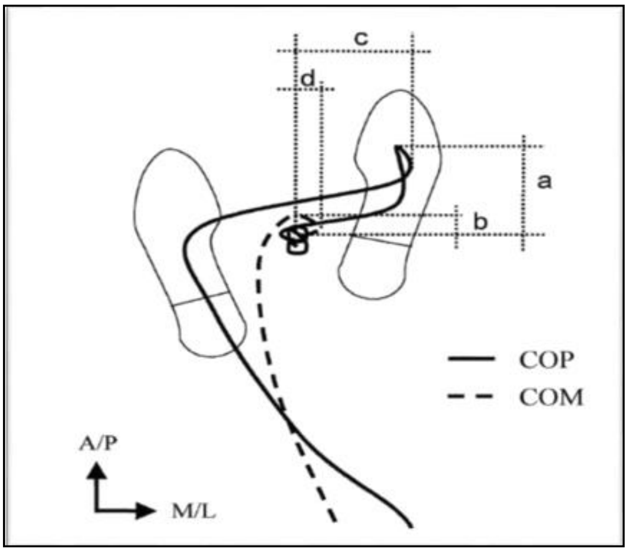


Figure 7. CoP and CoM maximum displacements with a, b stand for A/P displacements CoP and CoM respectively. c, d stand for M/L displacements of CoP and CoM respectively. Taken from (Meier 2001)

2.7.1 Amputees' gait termination

For individuals with lower limb amputation, there are a limited number of studies investigate gait termination task (Vrieling et al. 2008b; Vrieling et al. 2009; van Keeken et al. 2013; Prinsen et al. 2017). Analyzing how this task being executed compared to AB highlighted the main impairments in TFA mechanism of stopping which involved a small contribution of the prosthetic side in the production of deceleration forces necessary to halt body CoM (Vrieling et al. 2008a). These deceleration forces are normally generated by placing the CoP in front of the CoM. Lack of deceleration force of the prosthetic side was due to different factors; firstly, is the lack of ankle planter-flexion i.e. increased prosthetic ankle stiffness which inhibits a smooth anterior displacement of the CoP. Secondly, is the absence of knee flexion during loading response inhibits a posterior positioning of the CoM with respect to the CoP. Knee flexion is essentially required for power absorption, especially as the knee is the greatest contributor in gait termination negative work among the limb joints (Lynch and Robertson 2007). Lastly, the duration of single limb support of the prosthetic leg is reduced when compared to the intact leg (van Keeken et al. 2013), which limits the time in which the braking impulse can be produced. The duration of single limb support on the prosthetic leg is thought to be decreased because stability is challenged during this phase. Because of all this, individuals with an amputation heavily rely on their intact leg for the absorption of energy during gait termination (Vrieling et al. 2008b; van Keeken et al. 2013). Compared to AB, the amount of braking force for the TFAs is reduced at least to 50% and for TTAs the reduction is by 33% (Vrieling et al. 2008b). Therefore, a prolonged stance phase on the intact leading limb, loading more weight on the intact limb prior to leading the final step, and decreasing gait termination velocity can also help in providing the required braking impulse to overcome this limitation in function (Vrieling et al. 2008b).

A better understanding of the limitations in function and adjustment strategy that lower limb amputees encounter can be provided from the analysis of CoP trajectory in view of GRFs. The direction of CoP trajectory would be altered according to; limb dominance, loading preference, or even the confidence and comfort level that the amputee has. CoP shift in mediolateral and anteroposterior directions pointed out that amputees used several balance strategies to compensate for the limitations in ankle function (Vrieling et al. 2008b). For instance, during terminating with the left prosthetic limb, the trajectory of COP in one TF amputee was returned back towards the prosthetic side after intact foot contact (Figure 8B) (Vrieling et al. 2008b). This due to employing a different adjustment strategy to load the prosthetic limb during bipedal stance and bringing the CoM between the feet. GRF in A-P direction was higher in the leading intact limb than in the leading prosthetic limb (Vrieling et al. 2009) reflecting a limited ability to produce braking impulse when using the prosthetic side to lead gait termination. Amputees inability to perform plantarflexion can be observed from the incomplete forward CoP progression under the stance foot with a lower magnitude of GRF in the sagittal plane. CoP shifts/fluctuations during gait termination beneath the intact as well as the prosthetic foot have been assessed to provide insight into the postural adjustment strategies used by amputees during the rehabilitation process (Vrieling et al. 2009) or by healthy long- term amputee participants (Vrieling et al. 2008b). For individuals with a recent amputation who can be considered as inexperienced prosthetic users as they have not yet developed any adjustment strategies, CoP trajectory showed a regression in the forward shift which reflects the main deviations in gait termination when terminating with the prosthetic limb. Generally during the rehabilitation process, the CoP was held back towards the heel of the prosthetic limb rather than continue forward towards the forefoot (Vrieling et al. 2008b) (Figure 8). From the analysis of CoP pattern, it can be assumed that leading with the intact limb in termination is favourable (Vrieling et al. 2008a) because sufficient deceleration and balance control can be obtained (Vrieling et al. 2008b).

Adjusting foot placement in the A/P and M/L directions determine the change in CoP pattern versus CoM (i.e. a macro-control strategy) (Jian et al. 1993). The macro-control strategy is decided by the previous step length and step width, which defines the dynamic stability boundaries in which the CoP is allowed to travel safely for the next step. Too short a step length will not allow adequate deceleration of CoM during the next stance period, and too narrow a step width will result in too low a medial acceleration during the next single support phase (Jian et al. 1993). Evaluation of outcome variables in the step prior to the last step termination is critical, even though only 10% of the body velocity is reduced in the first step termination (i.e. the step prior to the termination step done by leading limb) (Jian et al. 1993). Leading limb contribution during the step prior to termination may compromise gait termination stability.

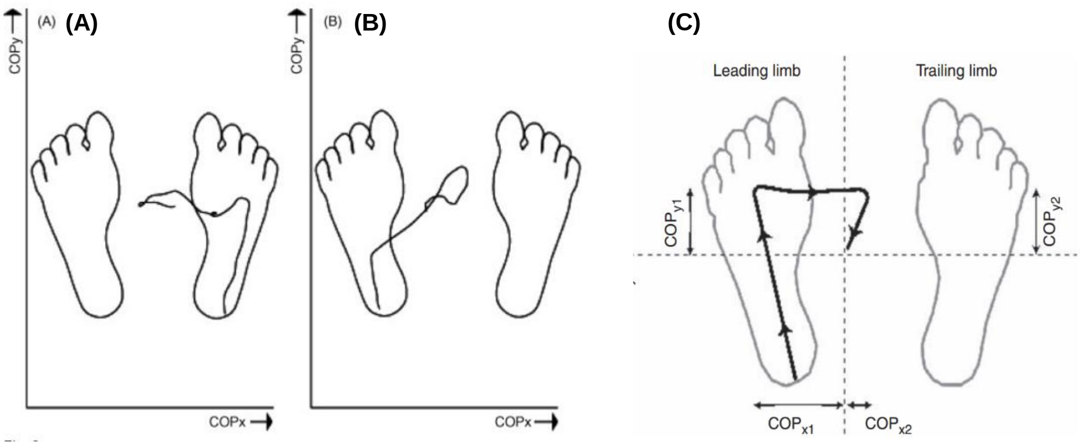


Figure 8. Examples of trajectories of the CoP in a participant of the TFA and AB group. (A) The CoP trajectory during leading with the right non-affected limb. The CoP on the leading side is shifted towards the forefoot. (B) The CoP trajectory during leading with the left prosthetic limb. During single-limb stance the CoP of the prosthetic limb does not move anteriorly. (C) Typical CoP trajectory displacement in A/P and M/L direction during gait termination og AB. Taken from (Vrieling et al. 2008b).

2.8 Mechanical work analysis and methodologies

CoM dynamics can be described as a contribution of reciprocal internal variables (i.e. muscles contraction, joint moments, and neural feedback) and external variables (e.g. external moment) to control CoM movements (Kuo 2007). Thus, another approach that provides further biomechanical insights to investigate gait pattern/mechanism is the kinetics which involve joint moment, power and work. Performing positive and negative mechanical work on the body characterises how locomotion is achieved. Historically, mechanical work analysis during locomotion was done in different studies using one of these methodologies (Winter 2009); 1) assessing the changes (absorption, generation and/or transfer) in potential or kinetic energies of body segments relative to the ground, 2) investigating the kinetic and potential energies of the major segments relative to the body's CoM and computed as the sum of the absolute changes in body segment kinetic and potential energy, 3) changes in the potential and kinetic energies of the body's CoM relative to the ground (external work done on CoM) (Donelan et al. 2002b; Kuo et al. 2005) techniques for the calculation of the positive and negative work at each joint by determining the time integral of the muscle and joint power curve.

Examining mechanical work performed led directly to fundamental insights into the mechanics and energetics of body CoM during gait. This analysis divides naturally into two parts (Willems et al. 1995). The first is related to a force external to the body that required to change the motion of the CoM i.e. the analysis of movements of the CoM relative to the surroundings. The work associated with the mechanical energy changes of the CoM is called 'external work'. Conversely, the second part, the movements of the body segments relative to the CoM are to a large extent brought about by forces internal to the body and, therefore, the work associated with the mechanical energy changes relative to the CoM is called the 'internal work' (Willems et al. 1995). Internal work is the work necessary to move the body segments relative to the body CoM and is computed as the sum of the absolute changes in body segment kinetic and potential energy (Cavagna and Kaneko 1977; Winter and Robertson 1978). Internal or joint work was conducted using analysis of the energy flow through body segments due to their relative movements to the ground (segmental power analysis) or the measurement of muscle power about a joint from the net forces and moments of forces relative to the ground i.e. inverse dynamics (Gordon et al. 1980) under a special set of assumptions of rigid bodies. Inverse dynamics does not quantify soft-tissue deformations between or within them and constantly failed to capture a significant percentage of work and especially negative work performed by body during gait (Zelik and Kuo 2010). Negative work is indeed performed by soft tissues, to a degree perhaps comparable to the joints themselves (Zelik and Kuo 2010).

The study of external work on the other hand, contributed in the identification of two generally accepted fundamental mechanisms of locomotion: first, the bouncing model of running, trotting, hopping and high-speed galloping, with its associated storage and recovery of mechanical energy by the contracting muscles and tendons (Cavagna et al. 1976; Cavagna et al. 1977) which is beyond the scope of this thesis. Second, the pendulum-like model of walking, associated with the maximum recovery of mechanical energy near the most economical speed. This approach represents a combined analysis of the kinetic-potential energy exchange moving body CoM relative to the ground (i.e. inverted pendulum directed by external force) and segments movement relative to CoM (result from internal force) (Kuo 2007). The relationship between external and internal work was investigated by (Willems et al. 1995) as an attempt to investigate any possibility of energy transfer among the limbs or between the limbs and the CoM of the whole body. Willems et al concluded that muscle–tendon work of locomotion is most accurately measured when energy transfers are only included between segments of the same limb, but not among the limbs or between the limbs and CoM of the whole body. External-internal work approach can be informative approach providing insights to; on the first-place CoM motion in space and how its displacement is controlled via the necessary energy exchange (work performed), secondly, the resulted mechanical energy changes attributed to limbs movement as a response to CoM displacement (Willems et al. 1995). Even though muscles are ultimately responsible for the mechanical work performed on the body and the major site of energy absorption (Winter 2009), developing an association between internal work and external work outcomes can be considered. The key idea of this method is to employ the inverted pendulum theory to view the stance limb and body CoM with the external work as an indication of deviations from the behaviour of a conservative pendulum (Donelan et al. 2002b).

In theory, the total energy level of the body should have the same value from any of these methods. Joint power and work technique supposed to automatically calculates any performed external work i.e. external power will be reflected in increased joint moments, which, when multiplied by the joint angular velocity, will show an increased power equal to that done externally. However, there are some limitations. These estimates provide little insight into the mechanical work generated by individual muscles during locomotor tasks of interest. Some studies suggesting inconsistency/inaccuracy of the external/internal work method in estimating musculotendinous work (Sasaki et al. 2009) or even in quantifying soft tissue work (Zelik and Kuo 2010). Zelik and Kuo considered the difference between measured joint work and the work performed on the body CoM as the possible work performed by soft tissue. Soft tissue performs significant negative work (approximately 60% of the total negative collision work and 31% of the total negative work per stride) and this amount of soft tissue work was even increased sharply with speed. Interestingly, each collision is followed by 4J of soft tissue rebound that is also not captured by joint work measures (Zelik and Kuo 2010).

2.8.1 External work

Walking is mechanically viewed as an inverted pendulum. This model of gait describes the CoM motion being continuously redirected by a conservative exchange of kinetic-potential energies (Cavagna and Kaneko 1977). Studies of able-bodied gait (Donelan 2002; Kuo 2007; Adamczyk and Kou 2009; Franz et al. 2012) highlight that during the transition from one stance leg to the next, CoM velocity must change from a forward-and-downward direction to a forward-and-upward direction (Adamczyk and Kou 2009). This redirection is achieved by the ground reaction braking/propulsive impulse i.e. negative/positive mechanical work performed on moving CoM during the step-to-step transition period, roughly corresponding to double support (Donelan et al 2002, Franz et al. 2012). During single support phases, the CoM moves in a series of arcs shaped by the stance limb acting as inverted pendulum and propelled/ guided during the previous double support phases (Donelan et al. 2002b; Donelan et al. 2002a). The majority of the mechanical work is produced or absorbed during the step-to-step transition. Energy exchange occurs during double support phase necessary for transition to each new stance limb from one inverted pendulum arc to the next (Winter 2009). Therefore, the transition phase has a great impact on the energy used in gait. Redirecting the CoM velocity requires a considerable negative work performed by the leading leg at heel contact as GRFs are applied through each limb at that moment. Energy lost due to leading limb negative work is compensated by trailing limb positive work (i.e. energy generation during push off phase).These assumptions are considered for level walking where the legs perform substantial amounts of positive and negative external work simultaneously.

Investigating the external work performed by the individual legs during down-slope walking in able body individuals Franz and colleagues revealed that the amount of negative external work performed by the limbs during double support increases as the angle of descent increases (Franz et al. 2012). When descending slopes surprisingly the trailing leg performs up to 27% of the double support negative work to walk downslope (Franz et al. 2012), which is much greater than that of double support during level walking (6%) (Donelan et al. 2012). This increase in trailing limb percentage work is associated with assisting the leading limb lowering the body's CoM with each step downslope. In addition, when walking downslopes, the perpendicular CoM velocity remains negative for a greater proportion of double support compared to level walking. The relatively large perpendicular force exerted by the trailing leg and the negative perpendicular CoM velocity is the reason the trailing leg absorbs considerable more mechanical power when walking downhill (Franz et al 2012).

As a proposed clinically friendly tool for practicing evidence-based rehabilitation, external work approach used by few studies as an intention to quantify kinetic and functional differences while using various prosthetic feet during gait (Agrawal et al. 2009), stair mobility (Agrawal et al. 2013) and during ramp ambulation (Agrawal et al. 2015). However, the conducted studies were limited to TTAs in the context of symmetry in external work by considering only the vertical component of the force and CoM. Since the GRFs during ramp negotiation are three dimensional in nature and the CoM moves in all three planes, total work during ramp ambulation could be computed using both the vertical and shear forces. Although the magnitude of shear GRFs is much lower than the vertical GRFs (Zmitrewicz et al. 2006) the contribution of work due to shear forces on the total work during ramp descent could be critical for assessing gait of higher level amputated participants like above knee amputees.

The mathematical calculations behind this method involves the usage of CoM velocity and the separate forces exerted by the leading and trailing legs "individual limbs method" (Donelan et al. 2002b). Thus, beyond the inaccuracy of the combined method, quantifying the mechanical work individually performed by each limb during gait in people with a trans-femoral amputation, taking into considerations the effect of GRFs under each limb provides useful insights to prosthetic fitting and sound limb adaptation/rehabilitation simultaneously. In addition to indicating the motion of the body's CoM, GRFs are important biomechanical measures to be analysed because amputees are at increased risk of developing joint disorders in the intact leg because of increased leg loading (Fisher and Gullickson 1978). Studies have shown the intact leg often undergoes increased anteroposterior (AP) and vertical GRFs compared with the residual leg and those observed in AB (Zmitrewicz et al. 2006).

The mathematical analysis of Individual limb (ILM) includes (Donelan et al. 2002b):

$$P_{\text{trail}} = \vec{F}_{\text{trail}} \cdot \vec{v}_{\text{com}} = F_{z,\text{trail}}\, v_{z,\text{com}} + F_{y,\text{trail}}\, v_{y,\text{com}} + F_{x,\text{trail}}\, v_{x,\text{com}},$$

$$P_{\text{lead}} = \vec{F}_{\text{lead}} \cdot \vec{v}_{\text{com}} = F_{z,\text{lead}}\, v_{z,\text{com}} + F_{y,\text{lead}}\, v_{y,\text{com}} + F_{x,\text{lead}}\, v_{x,\text{com}}.$$

$$W^{+}_{\text{trail}} = \int_{\text{POS}} P_{\text{trail}}\, dt,$$

$$W^{-}_{\text{trail}} = \int_{\text{NEG}} P_{\text{trail}}\, dt,$$

$$W^{-}_{\text{lead}} = \int_{\text{NEG}} P_{\text{lead}}\, dt.$$

$$W^{+}_{\text{lead}} = \int_{\text{POS}} P_{\text{lead}}\, dt.$$

$$W^{+}_{\text{ILM}} = W^{+}_{\text{trail}} + W^{+}_{\text{lead}},$$

$$W^{-}_{\text{ILM}} = W^{-}_{\text{trail}} + W^{-}_{\text{lead}}.$$

In these equations, F_{trail} and F_{lead} were the ground reaction forces for the trailing and leading limb, respectively. P_{trail} and P_{lead} represent the mechanical powers generated/absorbed by the trailing and leading limb, respectively.

Mechanical work from gait analysis data was evaluated to understand the causes of the variations of metabolic energy expenditure on moving CoM during gait. As previously stated, muscular work is required to redirect the CoM during double support even if the lower limb nearly behaves like a pendulum during single support phase. Thus, mechanical work is required between steps rather than within each step (Kuo et al. 2005). Work done for step-to-step transitions is a key factor reflecting the metabolic cost of walking in AB (Donelan et al. 2002a) and amputee gait (Houdijk et al. 2009). For TTAs, the increase of mechanical work needed for the transition of CoM from the affected limb to the sound limb contributed to the increased metabolic cost i.e. increase in mechanical work. The step-to-step transition is even more difficult for TFAs due to the behaviour of prosthetic knee and foot (Bonnet et al. 2014). That is the vertical displacement of the body CoM was much higher for TFAs due to the lack of knee damping. Nevertheless, (Detrembleur et al. 2005) showed that energy cost was not directly affected by CoM vertical displacement due to the conserved efficiency of the pendulum-like mechanism during the single stance phase of gait. It may be that the major part of the energy loss occurs during the step-to-step transition. Limb mechanical power at the prosthetic side was reduced compared to that of the intact side. The intact and prosthetic side presented a reduced mechanical power compared to that of AB at self-selected walking speed condition (Figure 9). An increased walking speed produced an increase of both limb work (Bonnet et al. 2014). Speed effect was less pronounced for the amputated and substantial for the intact limb. The lesser work produced by the amputated leg was associated with the increased flexion moment of the residual hip during late stance phase, which is necessary for initiating knee flexion in the affected leg (Bonnet et al. 2014).

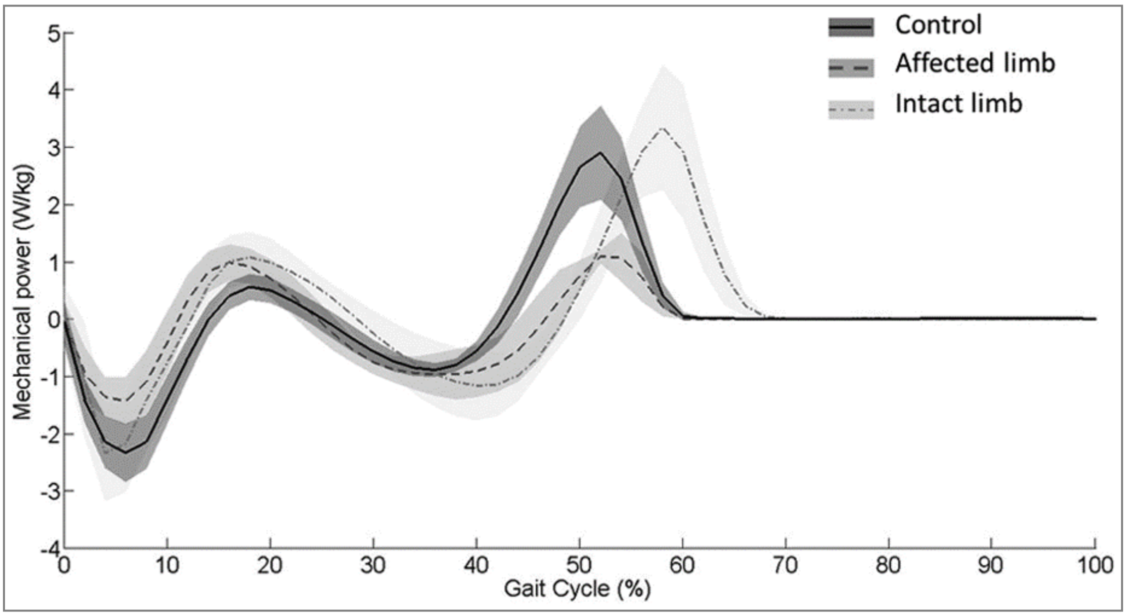


Figure 9. Mechanical power produced by affected limb (dashed line) and intact limb (dash-dotted line) of TFAs compared to AB participants (solid line) (W/kg) according to the percentage of gait cycle in the self-selected walking speed condition. Taken from (Bonnet et al. 2014).

2.8.2 Internal work (muscle and joint work)

Internal work was investigated in the literature by studying the rate of change of energy of i.e. the increase/decrease in the net flow of energy due to work done by applied forces at the segment's joints or by assessing muscle moments (Gordon et al. 1980). Negative and positive work done by lower limbs in 3 planes were investigated during level walking (Eng and Winter 1995; Allard et al. 1996). Sagittal work was the most important and accounted for 95, 75 and 88% of the power developed at the ankle, knee, and hip respectively (Eng and Winter 1995). It seems that the sagittal plane power is dominant compared to that on the transverse and frontal plane with about 50% of the negative work done by the knee (Allard et al. 1996). The power generation of the ankle and the hip contributed to the forward progression, while the absorbing activity of the knee was modulating it (Allard et al. 1996).

Compared to level walking, down-slope walking is associated with higher gait variability and so the risk of falling is potentially increased (Redfern et al. 2001). Walking on ramps requires exertion of greater forces across the hip, knee and ankle, and greater ankle motion (McIntosh et al. 2006). As shear force is directed posteriorly, the GRV directed backward to the limb, which in turn necessitate ankle dorsiflexion moment, extensor knee moment and flexor hip moment (Kirtley 2006). The effects of surface angle were seen on the calculated sagittal plane joint moments again mainly at the knee and to a lesser extent the ankle and hip (S. Redfern and DiPasquale 1997) hence joint muscle power accordingly would be altered. Further, studies investigated joint muscle actions during down-slope walking (Franz and Kram 2012) showed that knee extensor muscles of both leading and trailing limb perform the maximum negative work, to reduce the CoM velocity and achieve resistance to gravity. The knee is considered as the most adaptable joint during walking down-slope (Hansen et al. 2004). Throughout the gait cycle, Knee flexion increased from loading response to early swing. The resultant shortening of the limb lowers the body, facilitates initial contact of the other limb on the lowered surface, and reduces impact force. Moreover, knee flexion assists in rotation of the tibia in the sagittal plane and brings the body forward over the stance foot (Kuster et al. 1995; S. Redfern and DiPasquale 1997; Hansen et al 2004). Ankle dorsiflexion is increased from late stance to mid-swing, whereas hip flexion is decreased from mid-swing to early stance, which results in a pull back of the swing limb, shortening of the step length and easier positioning of the foot on the lower surface (Kuster et al. 1995; Lay et al. 2006; McIntosh et al. 2006). Further, knee extensor muscle activation (RF, VM) during the stance phase, increased with steeper descent (Franz and Kram 2012). Therefore, walking down-slope could be difficult (energy consuming) for people with muscle weakness and reduced functionality or even amputee due to inability to extend their knee down-slope. Amputees may experience limitations in function during slope walking due to the fact that the length of a prosthetic limb is usually reduced, in addition to different properties of prosthetic knees and feet compared to human joints. Consequently, TFAs and TTAs may not be able to perform the required adjustment strategies down-slope, which may cause loss of balance (Vrieling et al. 2008b).

For TFAs, the prosthetic limb shows a dramatically decreased stance flexion (i.e. power absorption), or it is often absent, in turn, result in the significant differences in the performed work compare to the intact side. On the other hand, a passive prosthetic foot does not allow a powered plantar flexion at the end of the stance phase (Au et al. 2008) producing about 20% less push off work at the prosthetic ankle compared to AB. Because of the lack of prosthetic push off and in order to control the prosthetic knee (Seroussi et al. 1996), kinetic parameters (work performed) of the intact limb joints were correspondingly modified (Bonnet et al. 2014). Intact limb adaptations were mainly observed during the double support phase of gait. A forcible overuse of intact limb compared to relatively symmetric loading ambulation of AB was demonstrated by TFAs (Nolan et al. 2003) during level gait (Murray et al. 1980) and ramp walking (Wolf et al. 2012). Particularly, work by hip moments of both limbs were greater for people with trans-femoral amputation than for AB group. As an attempt to maintain forward propulsion an increase in hip extensor activity (work) of both intact and prosthetic limb and the ankle plantar flexion activity of the intact limb were observed (Seroussi et al. 1996; Nolan and Lees 2000). Considering amputees' difficulties during level walking, the down-slope walking become even more problematic. Surface angle causes the prosthetic foot to seek foot-flat, so the forefoot will drop, the prosthetic shank will roll over/rotate forward, causing knee flexion, all due to the angle of the surface and the fact that the body mass is falling posterior to the knee, placing a flexion moment on the prosthetic knee (Highsmith et al. 2014). However, amputees would adopt further adaptations and compensation strategies to adjust their lower joint angles when walking down the slope (Chen and Gu 2013). A reduction in speed, motion range of knee and hip, and hip moments along with increased amplitude and periods of muscle activation during slope walking were the main observed compensations demonstrated by TTAs in comparison to AB people (Vickers et al. 2008). Whereas, TFAs confronted motion control challenge in lower joints during slope walking that can lead to their decrease in range of joint activity as compensation compared to AB (Chen and Gu 2013). Knee flexion in late stance created negative mechanical work, which is necessary for lowering of the body. The increased flexion in the prosthetic knee for TF group was lower than in AB and TT. However, TF were the only study group that showed an increase in the production of negative work on the prosthetic side when walking downhill compared to level walking, because in TF maximum prosthetic knee extension was not decreased in downhill walking (Vrieling et al. 2008b). In other study on joint activity (Wolf et al. 2012), comparing TF to that of the control group, intact knee showed a reduced early-stance peak knee power absorption and an increased peak during late stance (Figure 10). Besides, the ankle power generation was increased more than the control group. To maintain stable and safe gait, amputees may be keeping their CoM more posterior while leading with their prosthetic limb. The ankle power generation on the intact limb happens during double support after the prosthetic limb has made contact (Wolf et al. 2012).

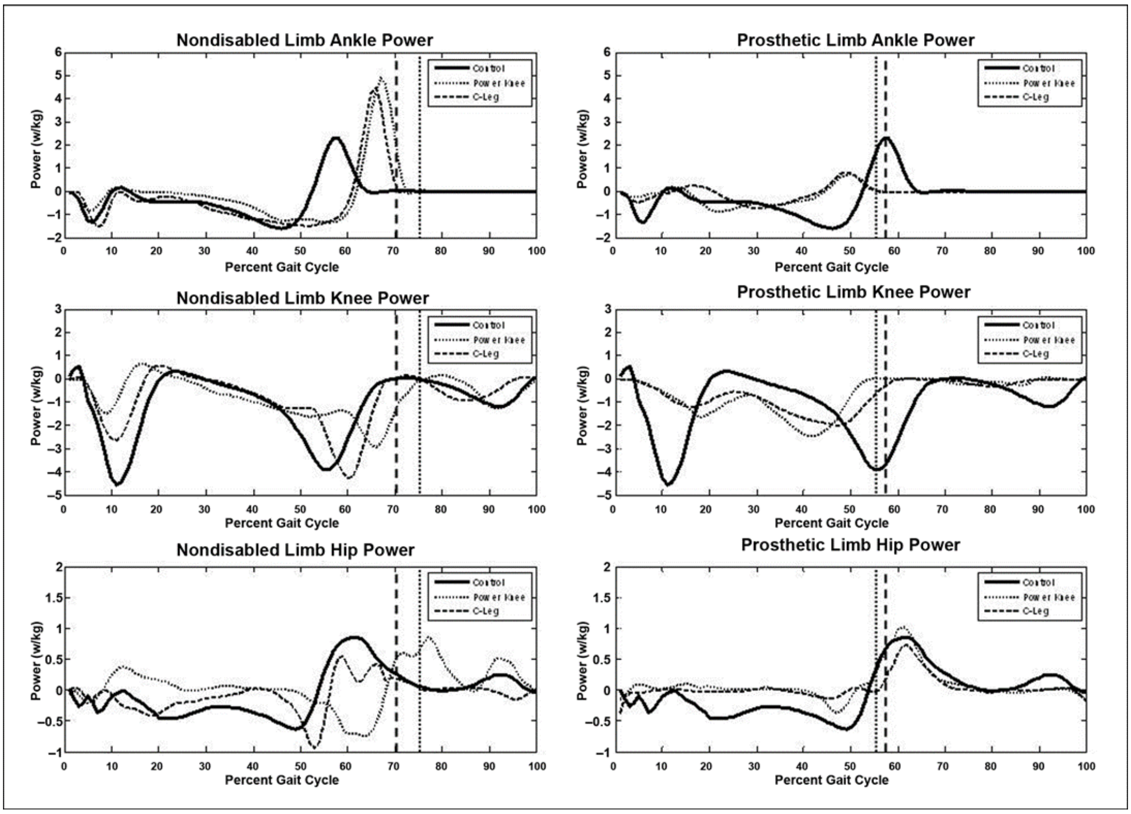


Figure 10. Ankle, knee, and hip power (W/kg) during ramp descent for trans-femoral amputates using C-Leg and Power Knee (representative data from single subject). Foot-off is represented by vertical lines.(Wolf et al. 2012)

2.8.3 Why external limb work?

Gait termination studies (Jaeger and Vanitchatchavan 1992; Jian et al. 1993; Bishop et al. 2002; Bishop et al. 2004; Lynch and Robertson 2007; Vrieling et al. 2008a; Ridge et al 2013; Ryckewaert et al. 2014) looked at assessing lower limb loading involve analysing spatiotemporal, kinematics, GRFs, joint moment, classify the power to different phases, and margin of stability. Since the lower limbs experience variable GRFs acting on body CoM during the stance period, assessment of joint power without considering body CoM motion/dynamics might not provide a complete picture of the consequences of loading throughout the entire stance period. The traditional method and power phases might be valid for the intact side joints for amputees (Winter 1988; Nolan and Lees 2000) but it might be inapplicable for the prosthetic side especially, in the situation of assessing gait with energetically passive prosthesis. The reason why is that passive prostheses are incapable of generating energy that might prevent fair comparisons of biomechanical variables between prosthetic and intact limb. Thus it is possible to quantify the mechanical work performed during gait by people with a trans-femoral amputation, using the individual limb method (Donelan et al. 2002b). The method provides information that is useful for prosthetic fitting and rehabilitation (Bonnet et al. 2014). Furthermore, non-joint work – that not captured by inverse dynamics – as an estimate of soft-tissue contributions was used to determine whether soft tissues contribute significant work to human walking. Negative work is indeed performed by soft tissues, to a degree perhaps comparable to the joints themselves (Zelik and Kuo 2010). This dissipative soft-tissue work occurs primarily during collision and increases with gait speed. Therefore, the joint work captured by rigid-body inverse dynamics may seriously underestimate the total negative work performed by the body, and perhaps even some of the positive work. Inverse dynamics consistently fails to capture a significant percentage of work, especially negative work, performed by the body during gait. The analysis of work performed on CoM makes no assumptions about rigid bodies and, therefore, captures both joint and soft-tissue work (Zelik and Kuo 2010). Although, the external work approach does not estimate individual joint contributions or work performed relative to CoM, the latter is generally considered small during stance phase (Cavagna and Kaneko 1977; Willems et al. 1995).

The technique used in determining external work depends mainly on force platforms data, which makes this approach particularly easy to implement, requires few assumptions which cause only small errors (that air resistance and skidding are negligible) and provides particularly simple to interpret, noise-free tracings (Cavagna 1975). Importantly, the mechanical work assessed is that performed by the external forces, and it is determined as the product of GRF and CoM velocity. Thus, a limitation that the external work approach can overcome is the errors due to the definition of the joint centre where joint kinetics is calculated. Any alteration in the assumed joint centre leads to change the calculated net moment ultimately affect the estimated joint power. This means the approach avoids having to make any assumptions about the interface between socket and limb, segmental rigidity and joint origin and axis definition (Kent and Franklyn-Miller 2011); as would be required if internal work estimates were determined. This was important because distal to the limb system's ankle device the 'dynamic response' prosthetic foot's heel and forefoot keels would deflect about non-defined axes and at different locations to the ankle device's articulation axis. Previous authors have highlighting that because of these issues the assessment and interpretation of 'ankle' kinetics can be problematic (Geil et al. 2000); which we wanted to avoid.

Chapter Three

General Methods

3.1 introduction

This chapter discusses the methodological details including ethical review process, participants' information and recruitment criteria in addition to the general procedures/protocols that were followed. Details of the motion capture system used, and the associated equipment/software are included along with the considerations for their use. Biomechanical variables, modelling and methodologies used to assess outcome variables are highlighted. Testing protocols for AB and TFAs are outlined; for both gait termination on level ground and ramp descent. More details regarding specific data analysis processes and the associated applied statistical analyses are provided in the methods section of each experiment chapter.

3.2 Ethical approval and participants' recruitment

Prior to conducting the current research, ethical approval was reviewed and approved by University of Bradford's Committee for Ethics in Research (ref. number E.119). All protocols were conducted in accordance with the tenets of the Declaration of Helsinki. Prior to every data collection session, each participant was made aware of the laboratory risk assessment and health and safety requirement. Each received a Participant Information Sheet (Appendix A) together with verbal information, and they were also encouraged to ask questions, so as to thoroughly understand their participation and intended research protocol. At that point the participant signed the Informed Consent Form (Appendix B).

Inclusion criteria for all participants are provided in sections 3.2.1, and 3.2.2 for able bodied and amputee participants respectively. AB were recruited from the local community of the University of Bradford. TFAs were recruited on a convenience basis from a list of TFA known by Blatchford as individuals currently using some type of MC prosthesis and willing to take part in research. All participants wore comfortable flat shoes. Lycra shorts were provided for each participant. Participant general preparation also included obtaining participant's weight using the force platform. Participant height was measured using a wall-mounted measuring rod with a single sliding calliper (H-629-1, MARSDEN Weighting Machine Group, Henley-on-Thames, UK). For amputees, height and weight were measured with them wearing the limb system prosthesis, and the footwear and clothing used during data collection. Information about participant general health status including any neurological disorders, previous lower limbs surgery/injuries in addition to any medications or alcohol being taken during the night prior to the testing day was self-reported.

3.2.1 Able-body participants

Eight healthy males (mean (SD), age 27.5 (6.93) years, height 1.77(0.067) m, mass 73.54 (10.74) kg) with no self-reported balance or gait abnormalities participated in the study.

3.2.2 Amputees' participants

Eight male TFA (Table.3) participated in the experimental study to determine the effect of speed during gait termination on declined surface (chapter 7). Only seven of them participated in the experimental study investigating the changes in gait termination for down-slope versus level surface (chapter 6) since one of the participants did not complete level surface condition (TF1 already imbedding the protocol). All the participants answered the questionnaire provided in appendix C with aid of their experienced prosthetist. Participants had undergone amputation as a result of trauma (n=7) or congenital deformity (n=1); all were unilateral TFA; using an Endolite prosthetic device (Hydraulic knee and ankle); medically fit and active to negotiate ramps and complete the protocol with relative ease; able of independent walking; all determined and classified by their rehabilitation consultant as K3 activity level according to Medicare mobility scale (will be discussed next section); and self-reported they had no balance, musculoskeletal, or residuum problems. Details of the prosthesis habitually used by participants are provided in Table 3. Five participants had habitually used a limb system prosthesis for at least two years; two of the other three were using an Orion with Elan foot and the other an Orion with Echelon VT (i.e. limbs with MC knee and MC or passive-hydraulic ankle devices). All participants used their existing prosthetic socket which was a full contact socket in which the surface of the residuum is totally in contact with the socket.

Table 3. Demographic characteristics of TFAs study participants

Participant no.	*Age*	*Height(cm)*	*Mass(kg)*	*Amputated side*	*Cause of amputation*	*Time since amputation (year)*	*Previous prosthesis* *Past year*
TF 01	39	183.5	111.8	R	Trauma	6.0	Linx
TF 02	30	177.0	78.0	R	Congenital	15.0	Linx
TF03	29	181.0	106.0	L	Trauma	7.0	Linx
TF04	57	185.0	95.4	L	Trauma	25.0	Orion2, Echelon
TF05	60	182.0	95.0	R	Trauma	9.0	Linx
TF06	62	165.0	70.0	L	Trauma	9.0	Linx
TF07	55	167.0	66.0	L	Trauma	8.0	Orion, Echelon
TF08	49	180.0	74.6	R	Trauma	21.0	Orion, Elan
Mean	47.63 (13.29)	1.78 (0.08)	87.11 (17.11)			12.5 (7.10)	

3.2.2.1 K-levels (Medicare mobility scale)

K levels is a rating system that indicates an amputee’s current and potential functional status. K levels, also called Medicare Functional Classification Levels (MFCL), was established by Medicare in 1995. The aim was to quantify the need for and the potential benefit of prosthetic devices corresponding with activity/mobility level of a patient with lower limb amputation. Medicare’s descriptions of the five (0-4) K Levels are:

Level 0

Does not have the ability or potential to ambulate or transfer safely with or without assistance and a prosthesis does not enhance their quality of life or mobility.

Level 1

Has the ability or potential to use a prosthesis for transfers or ambulation on level surfaces at fixed cadence. Typical of the limited and unlimited household ambulator.

Level 2

Has the ability or potential for ambulation with the ability to traverse low level environmental barriers such as curbs, stairs or uneven surfaces. Typical of the limited community ambulator.

Level 3

Has the ability or potential for ambulation with variable cadence. Typical of the community ambulator who has the ability to traverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic utilization beyond simple locomotion.

Level 4

Has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels. Typical of the prosthetic demands of the child, active adult, or athlete.

3.2.2.2 Limb system prosthesis technical information, preparation and adjustment

The prosthetic limb system investigated in this research was an above-knee prosthesis that has combined and simultaneous MC of the hydraulic resistances at the ankle and knee. This limb system prosthesis commercial name Linx was recently developed by Chas A Blatchford & Sons Ltd (Basingstoke, UK). As indicated within the product catalogue (https://www.blatchford.co.uk/endolite/linx/) this limb has several modes in which the hydraulic resistances at the knee and/or ankle are simultaneously altered in response to a change in terrain and or a change in walking speed. One of these modes is aimed at helping descent ramps (‘ramp descent’ mode). With this mode, the angle of terrain (by monitoring the moment applied to the shank pylon) and walking speed are continually monitored by the microprocessor via situational awareness sensors across the knee and ankle to continuously collect data on the user, activity and terrain. According to the manufacturer this data, is monitored 400 times per second and is used to continuously adjust the limb's resistance and speed for ease of use, safety and stability, mimicking a human limb. When the MC is active, it requires at least 2 walking steps down-slopes before the 'ramp descent mode' is activated.

When the ramp descent mode is activated the MC sends a message to a stepper motor within the foot which automatically adjusts the orifice opening of the hydraulic cylinder at the ankle to reduce hydraulic (plantar-flexion) resistance to facilitate attaining foot-flat more quickly and subsequently then increases hydraulic (dorsiflexion) resistance to slow/control the progression of the shank-pylon over the foot. This means the function of the foot-ankle section of the Limb System is equivalent to the *Elan* MC controlled foot-ankle device used by TTAs (Struchkov and Buckley 2016). With the ramp descent mode activated the MC also sends a message to the knee's stepper motor to reduce knee resistance in late stance to an intermediate level rather than the usual pre-swing (low resistance) level. The manufacturers indicate that the resulting higher knee/limb stiffness means the limb provides a braking effect to help reduce forward /downward momentum and promote better dynamic stability in late stance.

For both studies in chapter 6 and 7, the 'ramp descent' mode was the focus of the research. This mode was switched on (and off) via Bluetooth connection to ensure that the device was in the correct mode (MCon, MCoff). Another point to consider is that the limb system also incorporates a pneumatic cylinder at the knee in which the MC regulates to provide swing-phase control. As the focus of this thesis was on the mechanical limb work done when terminating gait, findings only provide insight into the device's stance-phase functions.

The three TFA participants who did not habitually use a limb system prosthesis were provided one for the duration of the study. This was achieved by attaching a limb system device to the participant's habitual socket; and altering the length of the shank pylon so that the limb was the same length as their habitual limb and maintaining the same limb alignment. The limb system was then set-up as per manufacturer guidelines (see Clinical Manual, http://linx.endolite.co.uk/downloads). Essentially this involved completing a calibration procedure within the limb system software, whilst the participant walked overground at their self-selected customary walking speed, then at self-selected slow and fast walking speeds. The procedure involves sequential stages so as to determine how the hydraulic (and pneumatic, knee only) resistance levels at the ankle and knee should be varied (and when they should), to optimize features such as: ‘stance release’ (knee release into free swing), overground walking at multiple speeds, intermediate knee release (prior to release into free swing, which provides a ‘brake assist’ when descending slopes), and so on. The ‘fine tuning’ procedure was also undertaken to check that ankle and knee resistances were indeed optimal. Change in impedance control parameters (e.g., stiffness and damping) during different gait phase/task of the limb system (i.e. fine-tuned) was based on observations of the participant’s gait performance and feedback until the amputee’s gait ‘‘looks good’’ i.e. participant felt that their gait was smooth and comfortable (and facilitated by the prosthesis), and there were minimum gait deviations observed by their prosthetist. The optimality criterion in this process was similar to the procedure described in (Sup IV 2009; Liu et al. 2014; Huang et al. 2016).All prosthetic adjustments, modifications and alignment were made by the same experienced prosthetist. Once the calibration procedure was finalised participants completed an accommodation period of about 20 minutes of walking over level and declined surfaces. No further adjustments were made to the device. For those who habitually used a limb system, limb alignment and the hydraulic damping levels were only checked, via the ‘fine tuning’ procedure, to ensure they were optimal. Change in impedance control parameters (e.g., stiffness and damping) during different gait phase/task of the limb system (i.e. fine-tuned) was based on observations of the participant’s gait performance and feedback until the amputee’s gait ‘‘looks good’’ i.e. participant felt that their gait was smooth and comfortable (and facilitated by the prosthesis), and there were minimum gait deviations observed by their prosthetist. Such observational criteria is similar to that used in previous research (Sup IV 2009; Liu et al. 2014; Huang et al. 2016). All prosthetic adjustments, modifications and alignment were made by the same experienced prosthetist. Once the calibration procedure was finalised participants completed an accommodation period of about 20 minutes of walking over level and declined surfaces. No further adjustments were made to the device. For those who habitually used a limb system, limb alignment and the hydraulic damping levels were only checked, via the ‘fine tuning’ procedure, to ensure they were optimal.

3.3 Motion capture system and lab set-up

The motion capture lab (Figure 11) used was equipped with a ten camera Vicon MX system (Oxford, UK) and two force platforms (Dimensions: 508*464mm each; Type: AMTI, Watertown, MA, USA). The force plates quantified GRFs along the orthogonal directions X, Y and Z. Signal conditioning and amplification of force platforms outputs (i.e. force and moment about the three orthogonal axes) were performed by two amplifiers (MSA-6 MINIAMP manufactured by AMTI) connected to the force platforms along with the computer system. The cameras, which were wall or ceiling mounted, were positioned around a walking area (5.8 m x 7.0 m x 2.8 m) with the force platforms at the centre of the lab. Set-up of the camera and force platform was undertaken using Vicon Nexus software (Version 1.8.5, Vicon MX, Oxford UK), plus other associated hardware (calibration wand and L-frame; section 3.3.1) (Vicon MX, Oxford, UK).
Prior to data collection sessions of this research, the infrared cameras were checked and adjusted (if necessary) via Vicon Nexus Software to ensure the cameras were in focus and directed appropriately to the pre-defined volume where the gait termination protocol took place. Furthermore, checking each force platform was performed to ensure both force platforms’ (with and without ramps’ blocks) accuracy in measuring GRF and CoP. Various weights of known mass were placed on each force plateform to ensure that the force platform recorded each weight accuratly. If there was a problem the scaling/correction factor was altered so the force platform gave the correct value. CoP data from both force platforms were also tested by appling force (using a screw driver) on a certain known premeasured location (on each force plateform corners) relative to lab origin (GCS). The output data from the force plateforms were then compared with the measured known location coordinates.

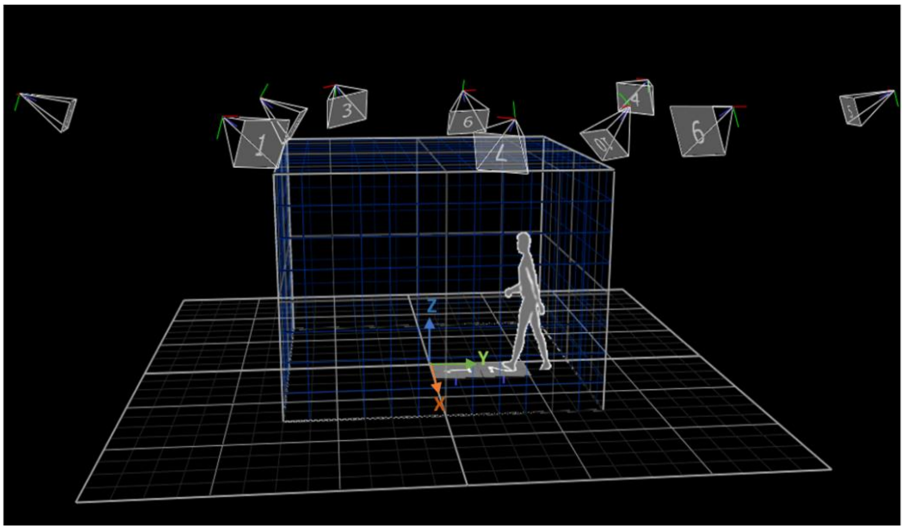


Figure 11. Adapted screen shot providing schematic of lab set-up: showing the ten-camera motion capture system and force platforms. Note the Lab origin (GCS) was located at the corner of platform 1 and participants’ direction of travel for all gait terminations was in the negative y-direction.

Participants performed gait terminations on the force platforms wore a set of retrospective markers within the capture volume monitored by Vicon nexus system/cameras. Spherical retroreflective markers were placed on each participant in order to capture 3D kinematic data while the participant stepping on the force platforms (provide kinetic data). All stopping trials were performed while participants were walking in the negative Y direction, either over the level laboratory floor or over a declined walkway. The gradient of the declined walkway was 5 deg which is recommended as a maximum angle for wheel-chair access ways (British Standards Institution 8300: 2009) as an attempt to make the outcomes more applicable to clinical gait assessment and rehabilitation programme design in the UK.

The walkway consisted of one flat section with handrail, followed by three declined sections that led onto the lab floor (Figure 12a). The guard/hand-rail and two crash mats were used to meet health and safety requirements as a part of the lab risk assessment procedure. The ramp surface was painted with anti-slip paint which provided a static coefficient of friction equal or greater than 0.62 as measured with a horizontal pull Newton meter (Super Samson; salter spring balance) by placing different weight on different shoes, pulling the shoe every time with the Newton meter then recording the first force required to move the weighted shoe. The last two declined sections were fitted around two solid wooded (chip-board) sloped blocks mounted (bolted) onto each force platform (Figure 12b). A gap of 2 mm (on each side) between the sloped blocks and the surrounding walkway ensured the force platforms only recorded GRFs when contact was made with the sloped blocks.

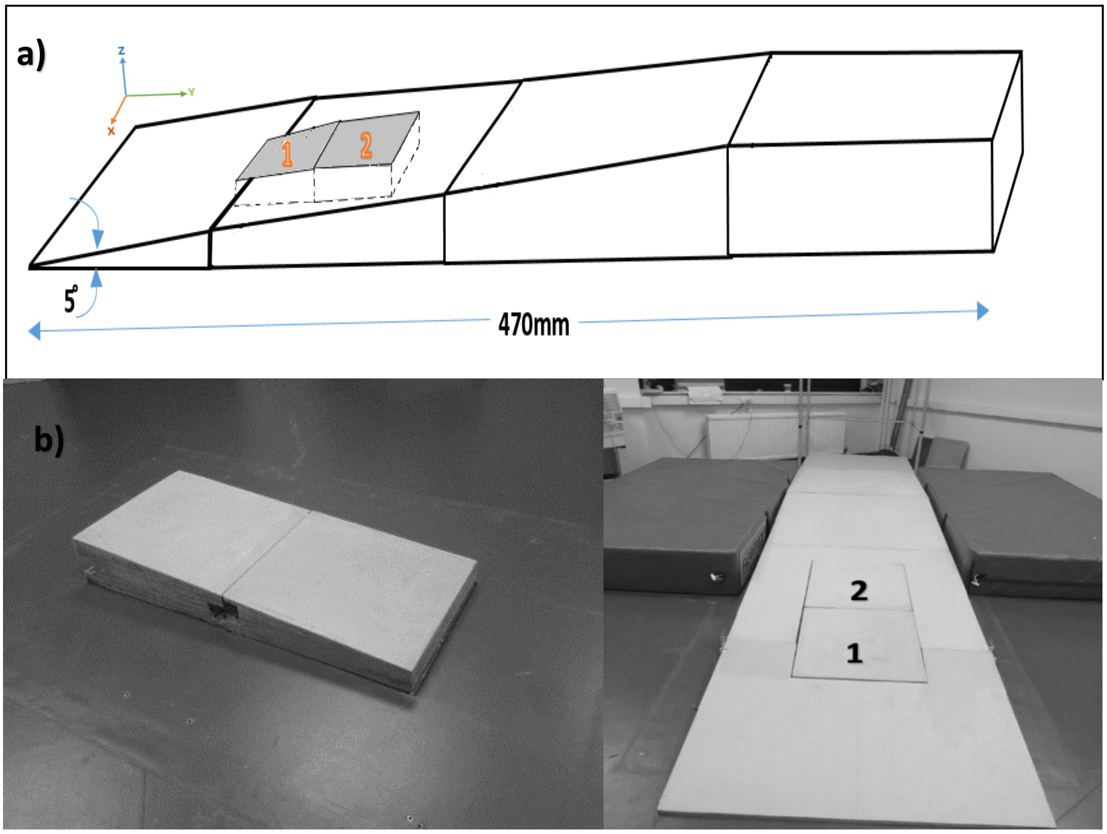


Figure 12 a) Custom ramp design. 1 and 2 grey squares are the two inclined solid blocks secured to the underlying force platforms 1 and 2 respectively. b) Ramp sections with guardrail and crashmats on both sides.

3.3.1 Calibration of motion capture system

Prior to each data collection day, the motion capture system was calibrated. The calibration process started by defining a capture volume where the data collection took place of approximately (3 m length, 2 m width and 2.5 m height). The calibration was achieved by means of calibration tools consisting the L-frame (500*470mm) (Figure 13 a), and a hand moved 3-markers calibration wand (390*240mm) (Figure 13 b). The wand was moved continuously within the 3D capture volume, covering as much area of the volume as possible for approximately 30 seconds to define the volume. The L-frame was placed on the right-hand corner, of the force platform 1 and ensured that it was level with the floor and aligned with sides of the platform (Figure 13 a). Thus, the location and orientation of the origin of laboratory global coordinate system (GCS) was set within the capture volume. This process also involved positioning and orientating of each camera relative to this GCS. The origin of GCS was set at this corner (Figure 13 a) so that the positive Z referred to vertical upward, positive Y was pointing posteriorly relative to the direction of travel, and positive X was pointing leftwards when moving in the negative Y direction (see Figure 11). The reconstruction error for each camera as well as the average for all cameras was accepted if it was less than +/-0.5 mm. The final step of calibration was to reset the force platform amplifiers to zero while there was no load on the force platforms. This step was repeated when the ramp was used. The sampling rate was set at 200 Hz for camera motion capture system and the two floor-mounted force platforms.

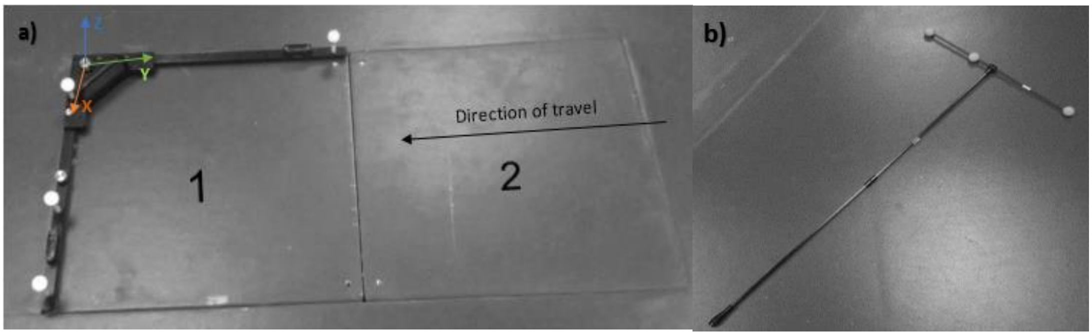


Figure 13. Vicon system calibration tools; a) L-frame (left panel) placed on the corner of force platform 1 level with floor, X, Y and Z indicate Lab coordinate system and b) a hand moved calibration wand with 3 markers (right panel). Participants' direction of travel is represented by the arrow.

3.4 Motion analysis in 3-dimensional space; six degrees of freedom background and corresponding markers placement

In general, the motion of rigid segments in space can be fully described by measuring three translations plus rotational movement (i.e. six degree of freedom movement) with respect to the GCS coordinate system. The description of the 6 variables (rotational translational components) of motion between two segments can be obtained by applying 6DoF modelling (Cappozzo et al. 1995). Measuring all six degree of freedoms requires segmental independency that may be achieved when there are at least three non-collinear markers unique to each segment (Cappozzo et al. 1995; Marika et al. 2006). So that these non- collinear markers can be used to define the embedded local coordinate system (LCS). LCS origin and axis strictly attached within each of the corresponding body segment and therefore able to move with the body.

The relative positioning of LCS axes with respect to the GCS defines the orientation of a rigid body or segment. The instantaneous position and orientation of all segments' LCSs with respect to the GCS are used to calculate the whole-body motion in real time and space by means of transformation and rotational transformation matrix (details of the procedure and mathematical background can be found in research method of biomechanics chapter 2). Transformation between LCS and GCS also includes kinetic data measured by each force platforms about the LCSs (Figure 14). When the force plates work together with the motion capture device, each force platform local kinetic data are transferred to a GCS. Translational and rotational transformation are required to transfer the relative positions of the LCSs of two force plates relative to the GCS of 3D motion capture system in the movement analysis laboratory. The position of the origin of the LCS about the GCS computed from the global coordinates of the four corners of each force plate (known during the setting up of the laboratory after the force plates been mounted) are used in transformation matrices. In view of that, movement of local coordinate systems as segments move through space during gait will be tracked by the motion capture system. Following calibration of motion capture system in which both kinematic and kinetic data are synchronised, markers placement, capturing static trial, functional joint centre trials and lastly data collection session can be started. These background theories will be applied in data processing and biomechanical modelling section 3.6.

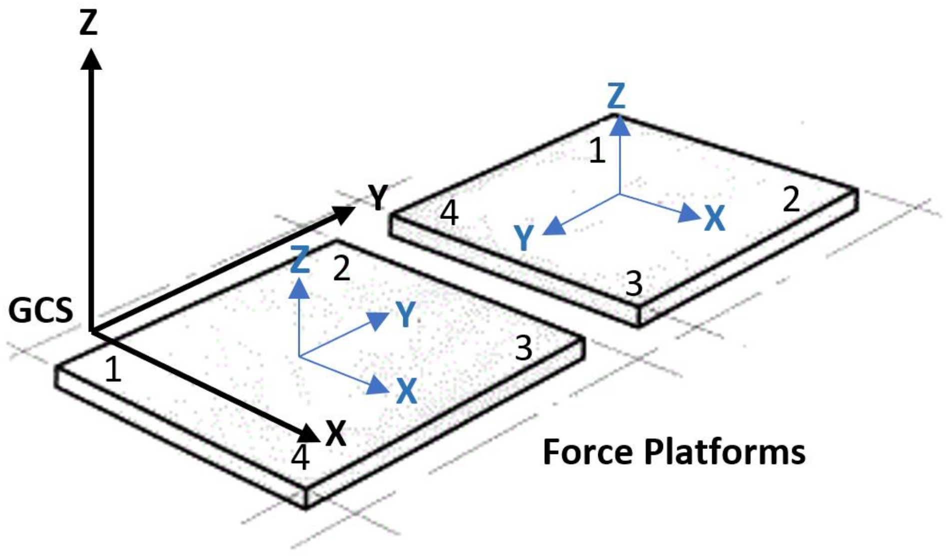


Figure 14. Schematic drawing of the two force plates and the corresponding LCSs (in blue colour) relative to GCS orthogonal directions (in black).

A 6DoF marker set approach (biomechanical model) was used in this thesis. Placement of markers was undertaken by the same researcher/investigator. Each body segment was tracked using at least 4 markers placed on it. Each segment's proximal and distal ends were anatomically defined by placing markers on specific anatomical landmarks. Therefore, a set of 54 retro-reflective markers were placed on each participant. All markers were 12.5 mm diameter except clusters markers and head band which were 14 mm diameter. 20 tracking markers (placed in clusters of 4, on semi-rigid curved plates) and 34 anatomical markers were used on predetermined anatomical landmarks as follows:

For AB; greater trochanter, iliac crest (i.e. a landmark along the Iliac crest vertically above the trochanter when standing), medial and lateral femoral condyles, medial and lateral malleoli, and bilaterally on posterior aspect of calcaneus, superior aspects of first and fifth metatarsal phalangeal heads, distal end of second toe, and pragmatically on the lateral and medial aspects of the proximal and distal mid-foot (Figure 15). Markers were also placed on acromion processes, the sternal notch, xiphoid process, and vertebrae C7 and T8. A head band was used to mount 4 head markers (with anterior markers located approximately over the left/right temples and posterior ones placed on the back of the head roughly in a horizontal plane with the front head markers). In addition, 4-marker clusters (a semi-rigid plate with four non-collinear markers glued on) were placed on the lateral aspects of thighs and shanks on the pelvis at the sacrum (Figure 15 a,b). The participant was then asked to stand within the calibrated volume with their hands facing forward for 3 secs to record a static standing trial. Afterward six movement trials were captured to define functional joint centres. These movement trials involved the lower limb being moved (waggled) in the following manner. 10 s of repeated hip range of movement: flex/extend, adduct/abduct, internal / external rotation; 10 s of repeated knee flexion- extension; and 10 s of repeated planter/dorsiflexion (Schwartz and Rozumalski 2005).

Similar marker placement was also used with TFAs, except some changes due to the absence of anatomical landmarks on the prosthetic side (Figure 16). The prosthetic ankle was defined midway between the markers placed on both sides of the pylon at the same height as the corresponding markers on the intact ankle (Powers et al. 1998; Silver-Thorn and Glaister 2009; De Asha et al. 2014). The prosthetic knee was defined by the makers placed bilaterally at the axis of the knee hinge as the most representative description of the joint centre and anatomically comparable to the intact side (Kent and Franklyn-Miller 2011). A functional joint centre approach was used to determine the residual hip joint centre using data collected in a limb ‘waggling’ trial (as per the process defined above; (Schwartz 2005).

After defining the functional joint centres, the anatomical/calibration markers on medial and lateral sides of the femoral condyles, and malleoli and markers on the acromions or corresponding locations for TFAs were detached before collection of dynamic trials.

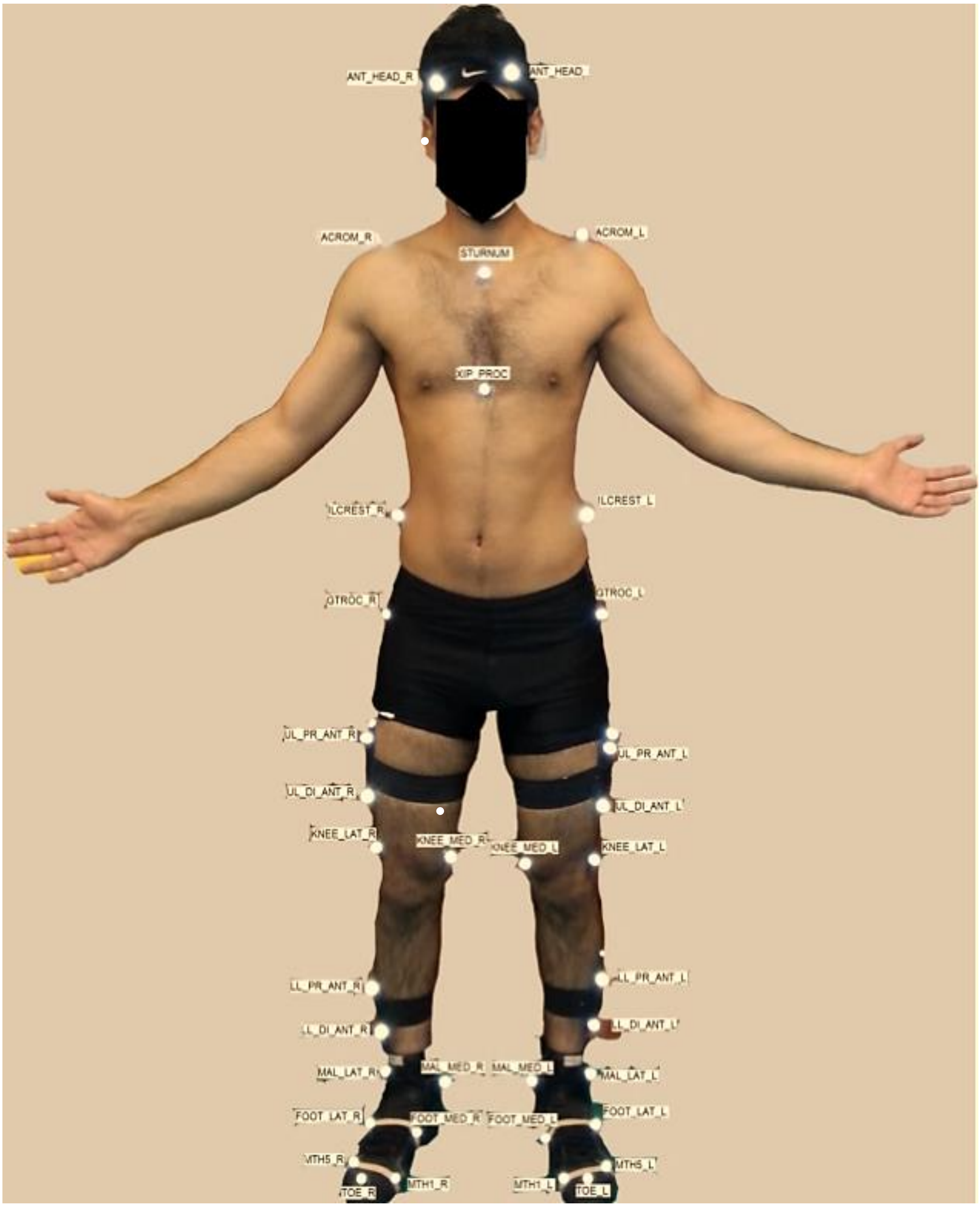


Figure 15 a. Positions of the 6DoF marker set (front) (taken from university lab documents).

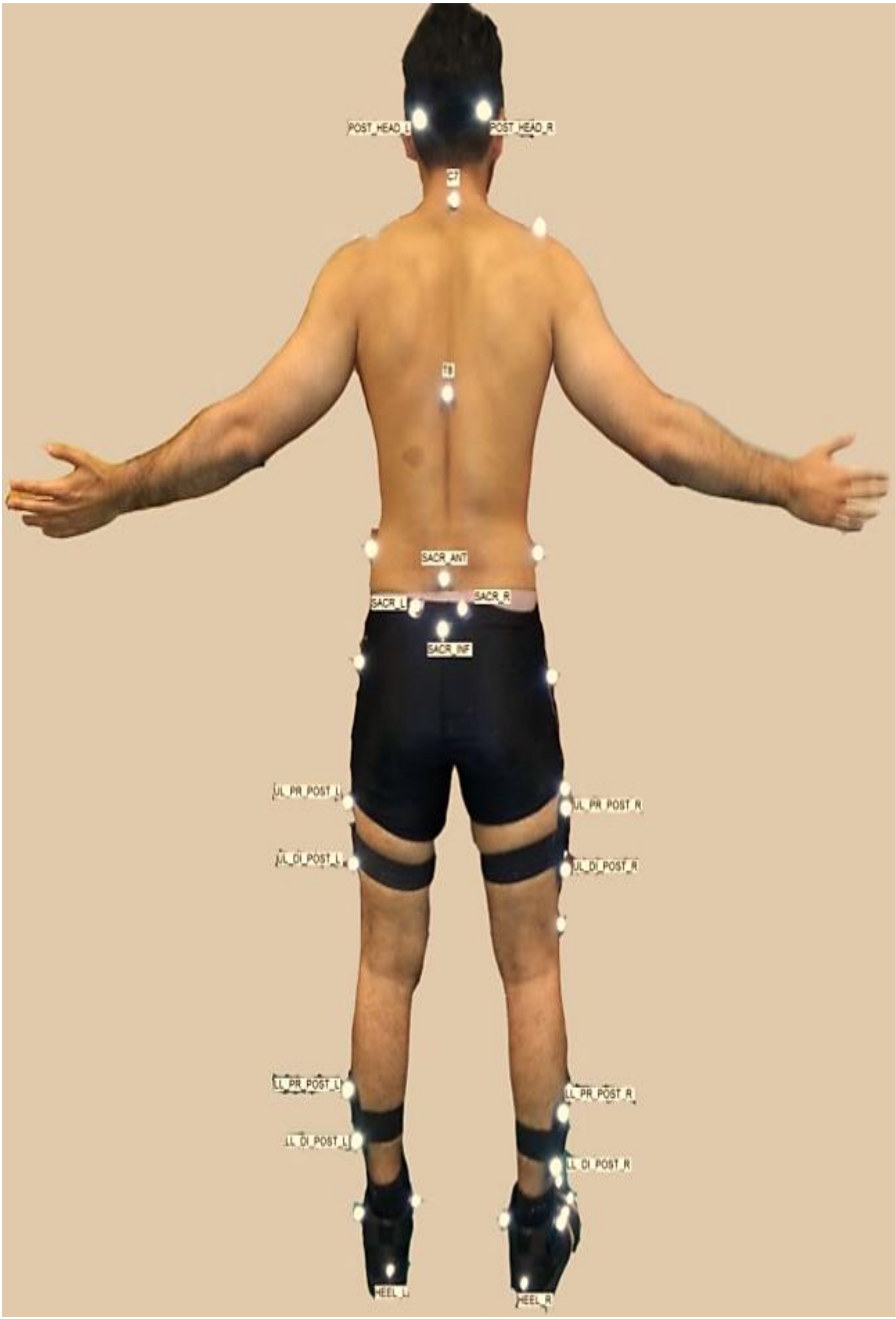


Figure 15 b. Positions of the 6DoF marker set (back) (taken from university Lab documents).

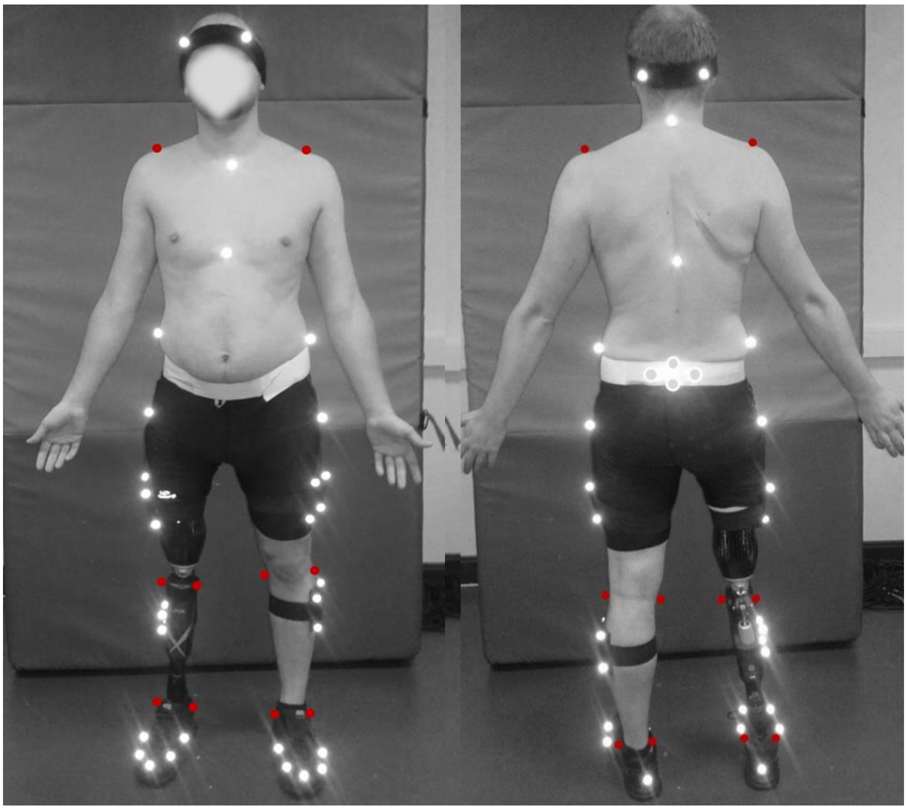


Figure 16. position of markers placed on a TFA participant with red markers on the acromions, medial and lateral femoral condyles, medial and lateral malleoli to define joint axis.

3.5 Data collection and protocol

After removing the calibration markers, practice trials were performed. These practice trials were used to select an appropriate starting location from the force platforms, to ensure the participant landed on them without adjustment of the step length. The required distance from the force plates allowed 4 walking steps to be completed, before contacting force platform 2 with the trailing limb. Previous research has indicated 4 walking steps is enough to achieve steady-state gait (Jian et al. 1993; Wearing et al. 1999; Bishop et al. 2002; van Keeken et al. 2013).

Trials were conducted in blocks (i.e. surfaces/speeds blocks). Starting either on the level section at the top end of the ramp or on the level floor, participants were asked to walk (down the ramp over level floor) at their self-selected speed, and to terminate gait on the limb they indicated was their preferred (i.e. the limb they would use to kick a ball) in case of AB, which for all participants was their right limb. For amputees, the prosthetic limb was the leading limb performing terminations. The instruction given was “Walk forward and then stop with both your feet in the second square and stand still facing forward waiting for further instruction”. For the ramp block, participants were instructed to; “Walk down the ramp and then stop with both your feet in the second square and stand still facing forward waiting for further instruction". A minimum of five ‘clean’ trials per condition (surface, speed and MC condition) were collected. To be accepted as a clean trial, the gait terminations needed to occur with the final two-foot contacts landing wholly within the bounds of the two-adjacent force-platforms or the sloped blocks above the platforms, i.e. trailing/intact foot landing within bounds of platform 2, terminating/prosthetic foot landing within bounds of platform 1. Participants were offered adequate rest if required.

3.6 Data processing and biomechanical modelling

With all the markers being placed while standing stationary in an anatomical neutral position are showed in the static standing trial (Figure 17 a) for each participant. Tracking and calibration markers were labelled manually using Vicon Nexus 1.8.5 software (Figure 17 b). After visual checking to ensure correct labelling for static trials was done without missing any marker, static trial was saved as one frame used later during the modelling process.

Each file of movement trials was cropped at the point the participant enters the capture volume just prior to foot contact (by 50 frames) on the first force platform until step off the second force platform. Each movement trial was labelled and gap filling of marker trajectories was completed by the spline fill (for gaps less than 10 frames) or by pattern filling which used the trajectory of an adjacent marker (on same segment as missing marker) as the source to fill the missing part of the marker trajectory. The C3D file was then exported to Visual 3D software (Version 5.02.27 C-Motion, Germantown, MD, USA) where all further processing and modelling took place.

In Visual 3D, implementing/embedding the six degrees of freedom (6DoF model) was done using the static standing trial (Figure 17 c); The functional joint centres (Virtual landmarks) for all of the intact joints and residual hip were created using the functional joint limb ‘waggling’ trials that defined the joint axes. For prosthetic limb joints centres were located on the mid-line of the prosthetic pylon at the same height as the functional joint centre on the contralateral, intact limb. For both AB and TFA participants, the 6DoF model was composed of nine linked segments (head, thorax/abdomen, pelvis, thighs, shanks and feet, Figure 17c). The distal end of one segment is connected to the proximal end of the adjacent via a linkage (mechanical connection to form a joint. As mentioned in the previous section each segment has a local coordinate system called the segment local coordinate system (LCS). For each segment the LCS was defined at its proximal joint centre e.g. the origin of the shank segment was at the knee joint centre. A distal end was defined as the midpoint between the two distal landmarks; i.e. the distal end of the shank segment was at the midpoint between markers corresponding to the lateral and medial malleoli. A line joining the proximal joint centre and distal end point defined the Z (longitudinal) axis of the segment. The frontal plane of the segment which defined the X axis was computed as the plane through the proximal and distal lateral and medial landmarks (De Asha et al. 2012). The location of the whole-body CoM was determined within Visual 3D as the weighted average of the nine tracked segments. The segment's moment of inertia and location of the segment CoM are calculated using the assumption of a geometrical shape for each segment having an inertial property and mass based on Dempsters' values (Dempster 1955). Anthropometric data were derived from Dempsters' values, and the inverse dynamic analysis assumed that the prosthetic knee, ankle and foot components were the same as for AB. Although this could have affected subsequent hip joint moment and power calculations, this method has been previously reported by (Vickers et al. 2008) for TTA. Furthermore, external work as the main outcome variable would not be affected by such assumption. As the kinetics of amputee movement has been shown not to be significantly affected by including modified inertial properties of the prosthetic limb (Miller 1987; Sanderson and Martin 1997; Powers et al. 1998) mass and CoM location for segments of the prosthetic limb were determined in the same way as the intact side. Kinematics and kinetics were filtered using a fourth order, zero-lag Butterworth filter with 6 Hz and 20Hz cut-off respectively. By assigning each block's dimensions and corners to the corresponding force platform, the original force and CoP signals on the physical surface of each force platform was translated on to the sloped force structures which defined the surface of the ramp's blocks (further details section 3.6.1 Force Structures).

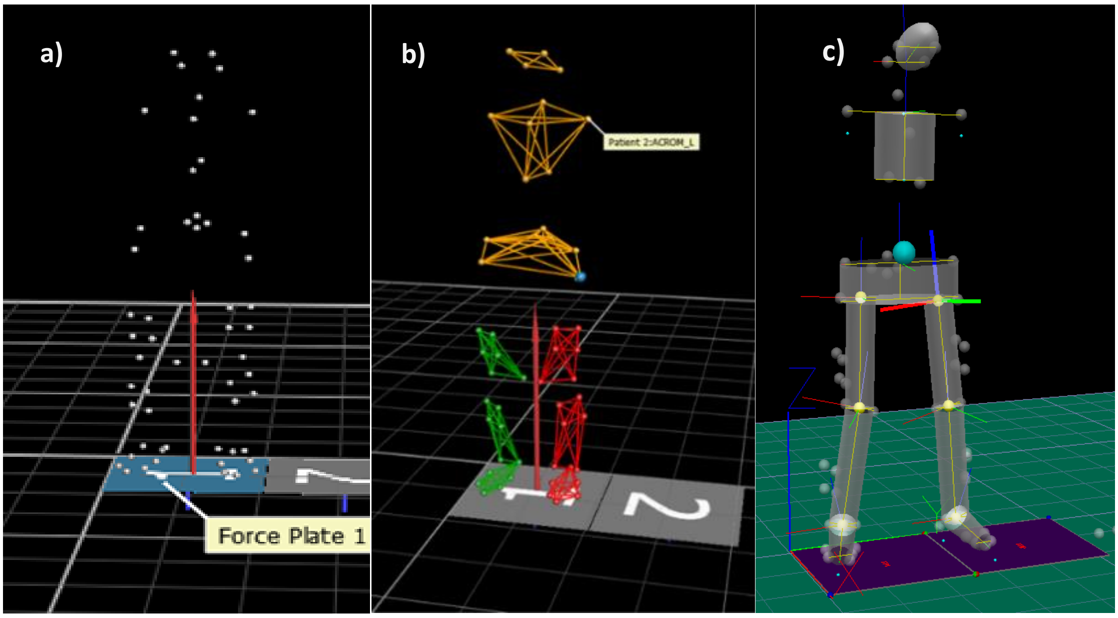


Figure 17. Reconstructed markers from static standing trial (from different data) a) before and b) after labelling in Vicon Nexus c) Representation (within Visual 3D) of Six degrees of freedom kinematic model with embedded local coordinate system (LCS) defined at each segment's proximal joint centre.

3.6.1 Force structure

When ramp blocks were placed on the top of the force platforms, the four corners of the top surface were detected in the same way as for force plates. The original GRF vector was transferred collinearly from the level ground force plate to the top surface. Thus, the coordinates of the CoP in the GCS were at the intersection point of GRF vector and the top surface. Ramp blocks as super-structures were not affecting the force platform parameters. When the structure (i.e. block) was used on the force platform, the intersection of the force occurred with the new structure's surface. The CoPz is translated along the Force Vector to the new surface. If a force is applied to the block above the force platform, the underneath physical force platform recorded GRFs. The new vector representing the force from the structure force should be collinear with the force platform's recorded GRFs and intersects the force structure (as per study trial Figure 18). If the original vector is corrected appropriately, the new vector would be coincident and collinear with the force plate one. Force structures representing the sloped block surfaces were created in visual 3D. (For further details see: https://www.c-motion.com/v3dwiki/index.php?title=Force_Structures). For each force platform the option “add a new force structure” was chosen to represent the sloped block above each force platform. By adding blocks dimensions as a block structure above the force platform, GRF and Centre of Pressure data were transferred to the ramp surface.

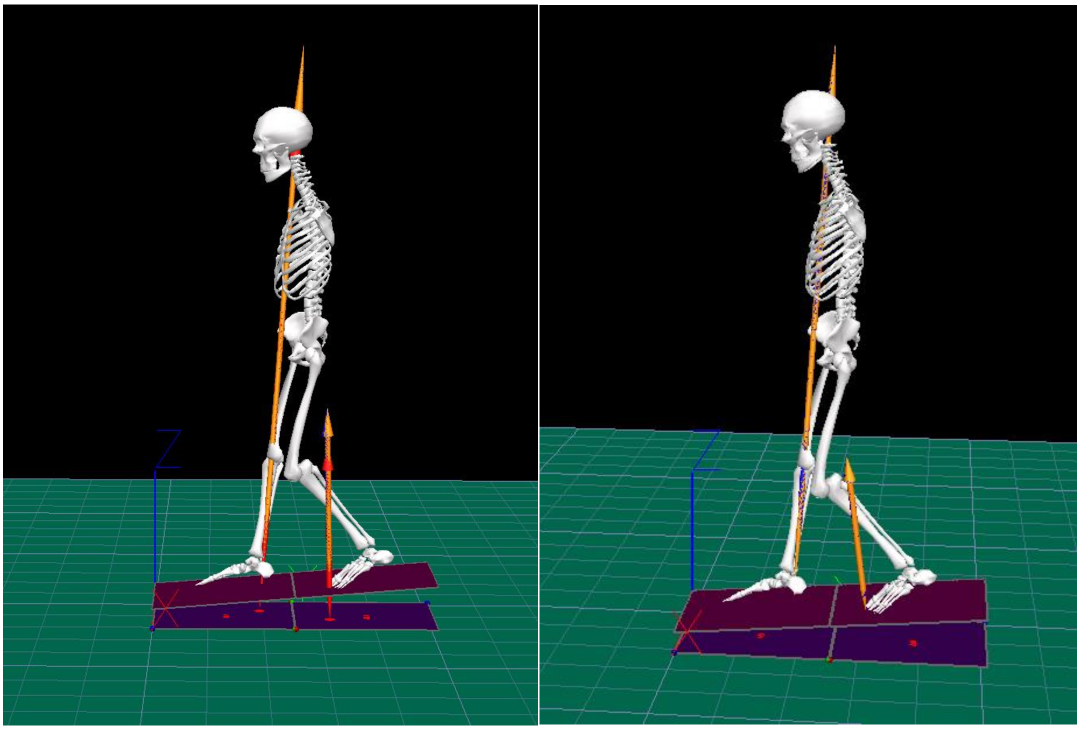


Figure 18. Screen shot from Visual3D motion analysis software (C-Motion, Germantown, MD, USA) illustrated the original force vectors being transferred to the top surface of ramp (sloped blocks). The left picture showed the transferred GRF vectors are in yellow colinear with the original GRF vectors are in red colour.

3.7 Data analysis and outcome variables

To divide the two-step termination into key periods, five events were created: trailing-limb initial foot contact, terminating-limb initial foot contact, trailing-limb toe-off, trailing-limb final foot contact and stable bipedal stance. The vertical component of GRFs was used to identify the first three events of gait termination. Visual 3D places an event label at the frame where a signal crosses (above 50N for contact and below 50N for toe-off). The specified value (i.e. 50N threshold) was in accordance with published work (Neckel et al. 2008; Franz and Kram 2012; Schaarschmidt et al. 2012; Gouwanda et al. 2016; Martelli et al. 2017). The moment when trailing-limb final foot contact next to the right foot was defined as the moment the CoPx reached the highest velocity when moving from under the single limb stance foot to a point between the two feet (van Keeken et al. 2013). Pilot work indicated that 0.2 m/s CoPx coincided with the beginning of limb-loading/contact of the trailing limb on platform 1 (i.e. as the trailing limb is loaded the CoP rapidly moves medially from the terminating limb towards the trailing limb).

Finally, stable bipedal stance was defined (after contact of the ipsilateral (trailing) foot) using the A-P force component with threshold (-5N) where the negative sign is counter to the direction of walking, i.e. negative Y within lab coordinate system.

The 'braking-phase' (Brk_T) duration for each limb was determined as the time period between foot contact up to contralateral-limb foot contact. Using the approach described by Donelan and colleagues (Donelan et al. 2002b), global limb mechanical power was determined as the sum of limb mechanical powers in each orthogonal direction. Limb power in each direction was calculated as the dot product of the respective GRF component and the corresponding component of the CoM velocity. Limb negative work, was determined as the time integral of negative global limb mechanical power during the braking phase of each limb. Limb directional-power: dot product of GRF under each limb and velocity of the CoM; determined separately for the directions parallel, perpendicular and mediolateral to the surface (Donelan et al. 2002b; Franz and Kram 2012). This meant that for the gait descent trials the GCS was tilted by 5 deg to correspond with the ramp inclination angle and hence provide the orthogonal directions as illustrated in (Figure 11 and 12 a):

$$P_{perp} = F_z \,.VCoM_{perp} \quad (1)$$

$$P_{par} = F_y \,.VCoM_{par} \quad (2)$$

$$P_{ML} = F_x \,.VCoM_{ML} \quad (3)$$

Where Fx, Fy, and Fz are the GRF under each limb in the side-to-side, parallel (A-P direction), and perpendicular directions respectively, and VCoM ($_{ML/perp/par}$) are the CoM velocities in the side-to-side (M-L), perpendicular (perp) and parallel (par) directions respectively.

Limb total power (normalised to body mass): summation of parallel, perpendicular and mediolateral power:

$$P_{tot} = P_{perp} + P_{par} + P_{ML} \quad (4)$$

Limb negative directional-work: time integral of negative power in each orthogonal direction, restricted to braking phase of each limb:

$$\text{Limb}W_{perp\ (-ve)} = {}_{BrkT}\int P_{perp}\, dt \quad (5)$$
$$\text{Limb}W_{par\ (-ve)} = {}_{BrkT}\int P_{par}\, dt \quad (6)$$
$$\text{Limb}W_{ML\ (-ve)} = {}_{BrkT}\int P_{ML}\, dt \quad (7)$$

Limb negative total work: time integral of negative total power, restricted to braking phase of each limb:

$$\text{Limb}W_{tot\ (-ve)} = {}_{BrkT}\int P_{tot}\, dt \quad (8)$$

Joint [muscle] power: dot product of joint sagittal moment (M, normalized to body mass) and angular velocity of joint (ω) (Quanbury et al. 1975):

$$P_j = M.\omega \quad (9)$$

Joint negative work: integral of negative joint power, restricted to braking phase for each limb:

$$W_{j\ (-ve)} = {}_{BrkT}\int P_j\, dt \quad (10)$$

For each participant, outcomes variables were determined for each limb for each trial and then averaged across trials to give mean values for each surface condition (level, ramp) or speed condition (slow customary and fast) per participant. All statistical analyses were performed using Statistica (StatSoft, Inc., Tulsa, OK, USA). Repeated measures analysis of variance (ANOVA) analyses were undertaken to compare the outcome variables with limb, surface and speed conditions as repeated factors. Post-hoc analyses were undertaken using Tukey HSD tests. The specific details of which outcome variables were assessed via ANOVA in each experimental chapter are considered within the methodology section of corresponding chapters. The alpha level of statistical significance for all statistical analyses was fixed at $p \leq 0.05$.

Additionally, to understand how the mechanical work done by the terminating (prosthetic) and trailing (intact) limbs compares to that in able-body individuals, group ensemble-average mechanical power profiles (± SD band) for each limb were plotted alongside ensemble average mechanical power profiles (± SD band) for a group of able-body individuals. A comparison of the current data to these previous data allow a subjective evaluation of how TFA produce the negative mechanical limb work to terminate gait on a ramp in comparison to how able-body individuals do.

Chapter Four

Gait Termination on Declined Compared to Level Surface; Contribution of Terminating and Trailing Limb Work in Arresting Centre of Mass Velocity

4.1 Introduction

Gait termination in AB group was explored throughout the current chapter as a ‘normative database’ study. It was initially necessary to evaluate AB gait termination on a declined surface compared to level ground. The Equality Act 2010 required public buildings to provide ramps in substitution for single step access to enable access for disabled citizens. Ramps considered as an everyday terrain associated with both home and general commuting provides an obstacle to challenge body eccentric work; the eccentric muscle contraction brakes the forward momentum generated due to gravity as the person descents. The human body uses the same muscles when walking up (muscle shortening) or down-slope i.e. muscle lengthening – generating potential energy stored in elastic cartilage and ligament tissues with excess energy dissipated as heat. To minimise injury and remain in control during a ramp decent the human body can reduce the gain in kinetic energy by eccentric muscular contraction. Consequently, walking down a slope has a greater chance to cause injury (Redfern and DiPasquale 1997; Cham and Redfern 2002). Sloped surfaces were recognised as environmental hazards that challenge locomotor behavior (Khandoker et al. 2010), walking down ramps specifically increased the risk of individuals falling compared to level surfaces (Redfern et al. 2001). In 2002, Cham and Redfern reported how gait is altered when walking down ramps, illustrating that peak normal GRFs increase by about 11% for a 5 deg ramp, whereas peak horizontal GRF in the direction opposite the direction of walking increase by about 66% for a 5 deg ramp. Ultimately, the GRF generated reflect the work done on the whole-body CoM by the lower-limbs (so called external mechanical work) (Donelan et al. 2002a). The effects of angle of the surface were seen on the calculated sagittal plane joint moments mainly at the knee, and to a lesser extent the ankle and hip (S. Redfern and DiPasquale 1997). In walking down a slope the shear force is directed posteriorly, causing the GRV to be directed backwards, which in turn necessitate dorsiflexor ankle moment, extensor knee moment and flexor hip moment (Kirtley 2006). Unlike the overground gait, the negative limb work performed by the trailing limb was increased with steeper downhill grade when walking downhill (Franz et al. 2012).

Walking on a level surface the limb initially performs negative external mechanical work on the CoM following foot contact so as to arrest the downward trajectory of the CoM and redirect it upwards (Donelan et al. 2002). During the same period, the contralateral (trailing) limb achieves positive mechanical work in order to assist with the transfer the CoM on to the leading limb (Donelan et al. 2002). However, the limbs when walking down ramps do additional negative mechanical work (power absorption) (Lay et al. 2007), and remarkably even during the double support phase when the CoM is being transferred from the trailing to the terminating limb, principally negative mechanical work is performed by the trailing limb (Franz et al. 2012). When terminating gait, the mechanical work done is likely to be primarily negative to fully halt the CoM velocity. When descending a ramp, arresting CoM velocity included the lowering of CoM and subsequently would require even greater negative mechanical work compared to terminating gait on a level surface. The present study explores these assumptions. The specific aims were to determine the negative external mechanical work performed by each limb when terminating gait and evaluate how such work is affected by a change in surface angle from level to declined.

In satisfying this aim the negative external mechanical work done by each limb in both the parallel, perpendicular and mediolateral directions, along with the overall negative work done by each limb were determined. A gait termination was accomplished over two steps (Jaeger and Vanitchatchavan 1992; Jian et al. 1993; Wearing et al. 1999). Based on previous research (Bishop et al. 2002; Donelan et al. 2002a; Franz et al. 2012), a hypothesis was developed suggesting that during the two steps of gait termination, the limb that initiates the final step (terminating limb) will perform more negative work (power absorption) compared to that done by the contralateral trailing limb. Typically, the increased work will be in the parallel direction due to having to arrest CoM forward velocity. In addition, the negative work done by the terminating limb will increase further when terminating gait on a declined surface, and this additional increase in work will be predominantly in the perpendicular direction due to having to lower the CoM down the ramp whilst also arresting CoM forwards velocity. An external mechanical work applied by the legs on (CoM) is necessary for the conservative exchange of kinetic/ potential energies of the CoM during single support (Adamczyk and Kuo 2009; Franz et al. 2012).

Stopping gait requires deceleration of the body forward velocity and associated control of body momentum (i.e. CoM sagittal plain velocity) to safely bring the body within a stable base of support. Such control is achieved by coordinated muscle action to generate the necessary internal muscle and/or joint moments and powers. As joint moments and powers for stopping on a ramp have not been reported in the literature yet, the objective of this study was also to quantify the changes in kinetics of leading and trailing limb for stopping on a sloped surface versus level surface in correlation with CoM dynamics/ control. Therefore, this chapter provided an insight of how AB control stopping gait differently on sloped surfaces relative to overground walking and provide a set of data for comparison with other populations (amputees) that are more vulnerable to falling.

4.2 Methods

4.2.1 Participants

Eight healthy males (mean (SD), age 27.5 (6.93) years, height 1.77(0.067) m, mass 73.54 (10.74) kg) with no self-reported balance or gait abnormalities participated in the study, all giving written informed consent (Appendix B).

4.2.2 Experimental protocol

Participants completed gait terminations in two blocks: with block order counterbalanced across participants. In one block, gait terminations were performed on the declined ramp and in the other they were completed over the laboratory floor. Each block included 10 repetitions. Gait terminations occurred 4 walking steps away from the starting location, which was adjusted for each participant so that gait terminations occurred with the final two-foot contacts landing wholly within the bounds of the two adjacent two-adjacent force-platforms or the sloped blocks above the platforms, i.e. left foot landing within bounds of platform 2, right foot landing within bounds of platform 1. Further details can be found in chapter 3 section 3.5.

4.2.3 Data acquisition and processing

Retro-reflective markers were placed locations mentioned in the method chapter (section 3.4). Labelling and gap filling were done using Vicon Nexus

1.8.5 software. Data were subsequently exported in C3D format to Visual3D software (Version 5.02.27 C-Motion, Germantown, MD, USA) where all further processing took place.

A six degrees of freedom nine segment model (Cappozzo et al. 1995) was created for each participant. Joint centres were determined using a functional joint centre approach using data from the limb 'waggling' trials (Schwartz and Rozumalski 2005). For ramp trials, a force structure representing the dimensions and location of each sloped surface was created above each force platform (method section 3.6.1).

4.2.4 Data analysis

Section 3.7 in the method chapter explain more details regarding gait termination events and the main outcome variables being assessed. Outcomes variables were determined for each limb for each trial and then averaged across trials to give mean values for each surface condition (level, declined) per participant.

4.2.5 Statistical analysis

Limb directional- and limb total- negative work were compared using repeated measures analysis of variance (ANOVA) with surface condition (level, declined) and limb (trail, term) as repeated factors. Joint negative work was compared using repeated measures ANOVA with surface condition (level, declined), limb (trail, term), and joint (ankle, knee, hip) as repeated factors. Post-hoc analyses were undertaken using Tukey HSD tests. Statistical analyses were performed using Statistica (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

As highlighted above, the analysis focussed on comparing mean outcome variables (calculated across the 10 repetitions) between surface conditions and between limbs. To confirm that it was justified in assessing mean outcome variables (rather than for example, comparing the minimum or maximum), A preliminary statistical analysis was used to assess if there were any differences across repetitions (e.g. learning and/or trial/fatigue effects). For this analysis, repetition was included as a repeated measures factor within the ANOVA model. This analysis indicated there were no main ($p > 0.84$) or interaction effects ($p > 0.91$) for repetition for any of the outcome variables.

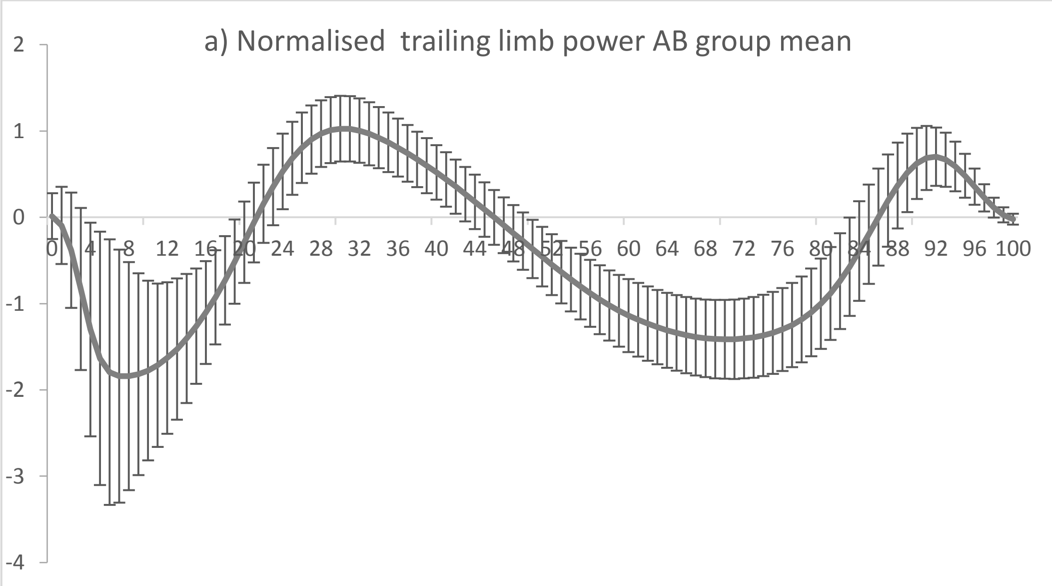
4.3 Results

Group ensemble average directional- and total- power profiles for each limb for ramp and level trials are presented in Figure 20. Group mean (±SD) directional- and total- negative work values for each limb for ramp and level trials are depicted in Figure 21. Group ensemble average joint power profiles for the ankle, knee and hip for each limb for ramp and level trials are presented in Figure 22. Group mean (±SD) joint negative work values for each limb for ramp and level trials are depicted in Figure 23.

Group average walking speed was 1.14 (0.16) and 1.08 (0.27) m/s for level and declined surface respectively ($p = 0.69$).

It is noteworthy to highlight that power profiles for each group are presented as group ensembled-average (e.g. Figure 19a). The group ensemble-average for each condition was computed from each participant's ten trial ensemble average for that condition. Prior to computing ensemble-averages limb powers were divided by body mass and were normalised to 100% of braking phase.

Time normalisation introduces an aspect of temporal 'blurring/smudging', which had the effect of lowering the power peaks (Figure 19 a) compared to non-time normalised signal (figure 19 b). This is because the events/peaks do not occur at the same time in different trials and/or for the different participants. So, a systematic underestimation of peak values might have occurred. However, this could be a problem only if there is large variability in event timing across trials; and Figure 19b would suggest this was not the case. It is worth emphasising that the limb work data used for statistical analyses was that calculated for each individual trial of each participant (and was calculated in absolute time terms; area under the power curve).



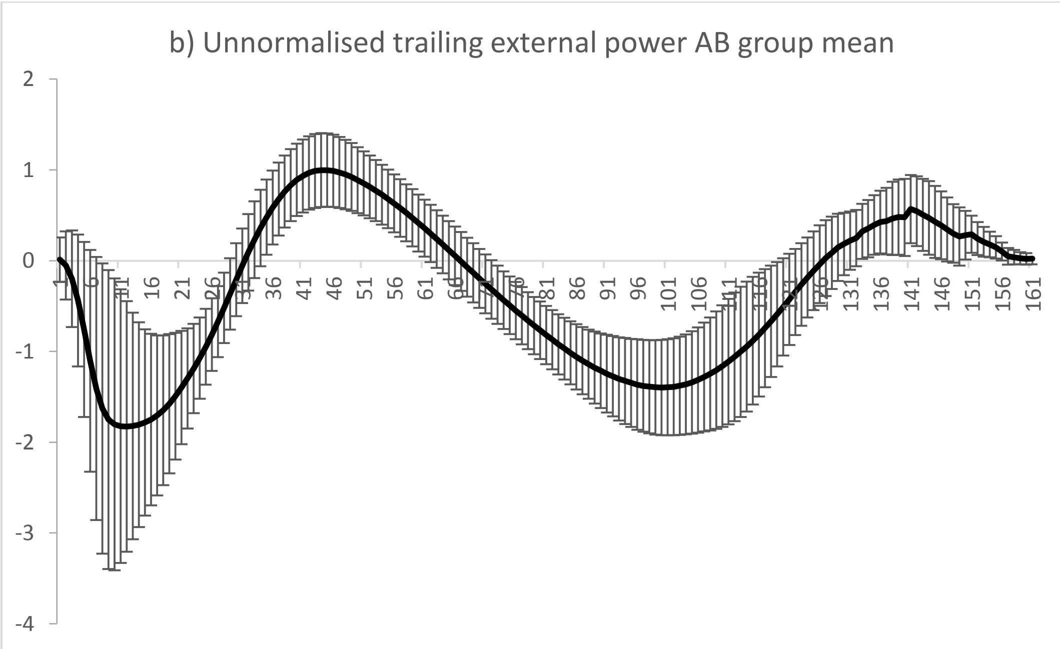


Figure 19. Group ensemble mean (SD) of external power for the trailing limb on declined surface a) normalised to 100% of braking phase and b) unnormalized power profile.

4.3.1 Limb work

Limb negative total work was significantly affected by limb (p<0.001) but not by surface (p=0.34): the interaction between terms approached significance (p=0.055). Limb negative total work was greater for the trailing-limb (-0.386, -0.451 J/kg) for level and decline surface respectively) compared to terminating-limb (-0.193, -0.160 J/kg) for level and decline surface respectively). The trend interaction between limb and surface indicated that trailing-limb negative work slightly increased on declined surface whilst the contribution of the terminating-limb decreased slightly; however post-hoc analyses indicated differences were non-significant.

4.3.2 Directional work

Limb directional negative work in all three orthogonal directions was significantly affected by limb ($p<0.03$). More negative work was done by the trailing- compared to terminating- limb in the perpendicular and parallel-AP directions but in the ML direction less work was done by the trailing- compared to terminating- limb. Limb perpendicular negative work was also significantly affected by surface ($p= 0.0004$) and by a limb by surface interaction ($p=0.025$): but there were no differences between surface conditions or limb by surface interactions in the other two directions ($P>0.37$). More limb perpendicular negative work was done for declined compared to level surface, and the limb-by-surface interaction indicated this increase was mainly due to more work being done on the declined surface by the trailing-limb ($p=0.006$) with little difference between surface conditions for the terminating-limb ($p=0.77$).

4.3.3 Joint work

Joint negative work was significantly affected by limb, by surface and by joint ($p<0.001$). There was also limb by surface, limb by joint and surface by joint ($p<0.001$) interactions, and a significant three-way interaction ($p<0.001$). Joint negative work was greater for the trailing- compared to terminating- limb and increased on both limbs for declined compared to level surface but post-hoc analyses of the limb-by-surface interaction indicated only increases for the trailing-limb were significant ($p<0.001$). More work was done at the ankle compared to knee which in turn was greater than work done at the hip; however, post-hoc analyses of the limb-by-joint interaction indicated differences between joints were only significant for the trailing-limb ($p<0.001$). The surface-by-joint interaction indicated the increased negative joint work done on declined surface was mainly due to increased knee work ($p<0.001$) with negligible increases at the ankle or hip ($p>0.46$). Post-hoc analysis of the three-way interaction did not reveal anything different to what the three 2-way interactions (detailed above) indicated.

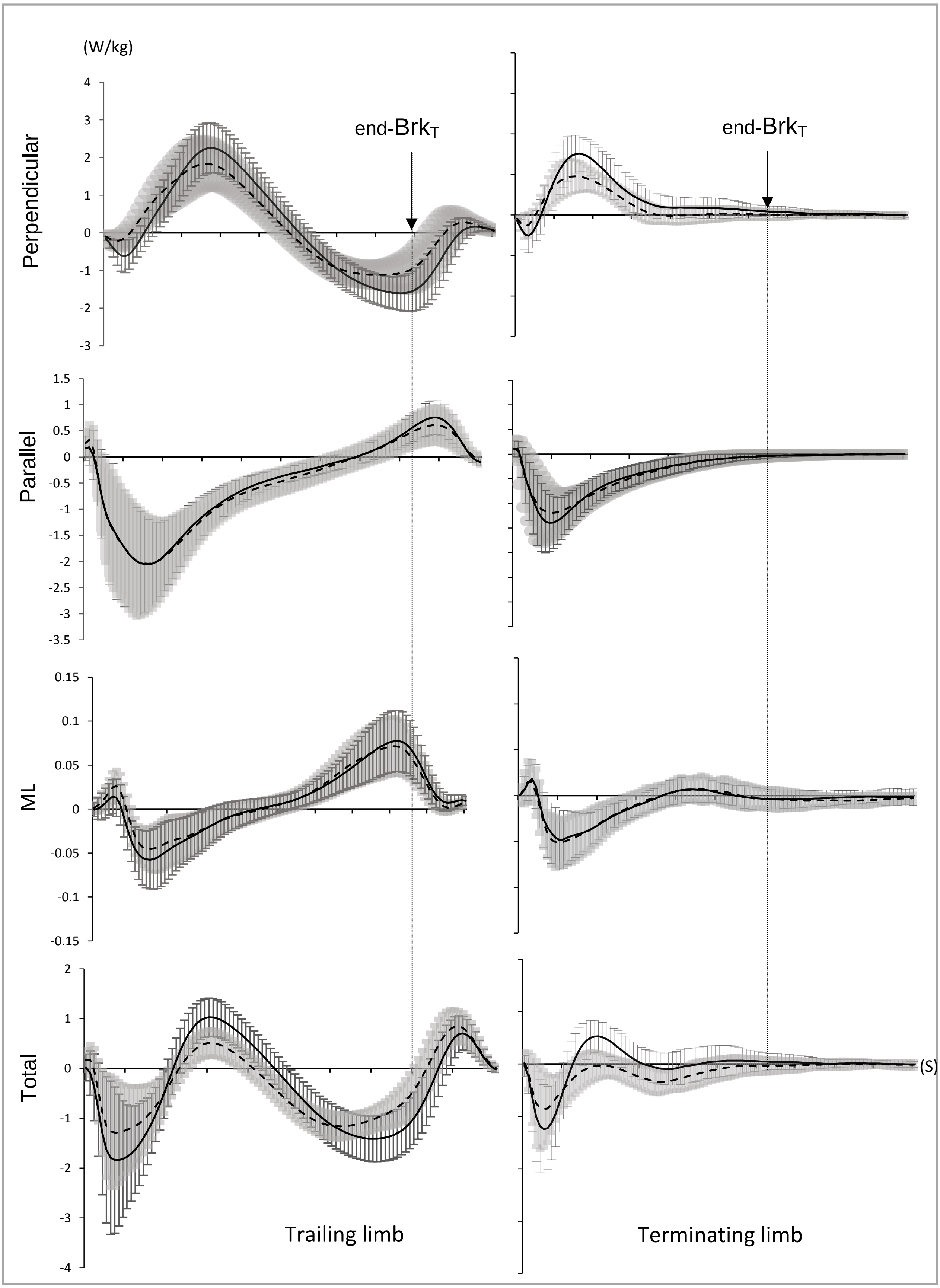


Figure 20. Group ensemble mean(+/-SD band) limb directional- and total-power profiles (W/kg) for the trailing- and terminating- limbs. Data are plotted for stance phase of each limb with end of braking-phase indicated by vertical line. Bold line = declined surface; dashed line = level surface.

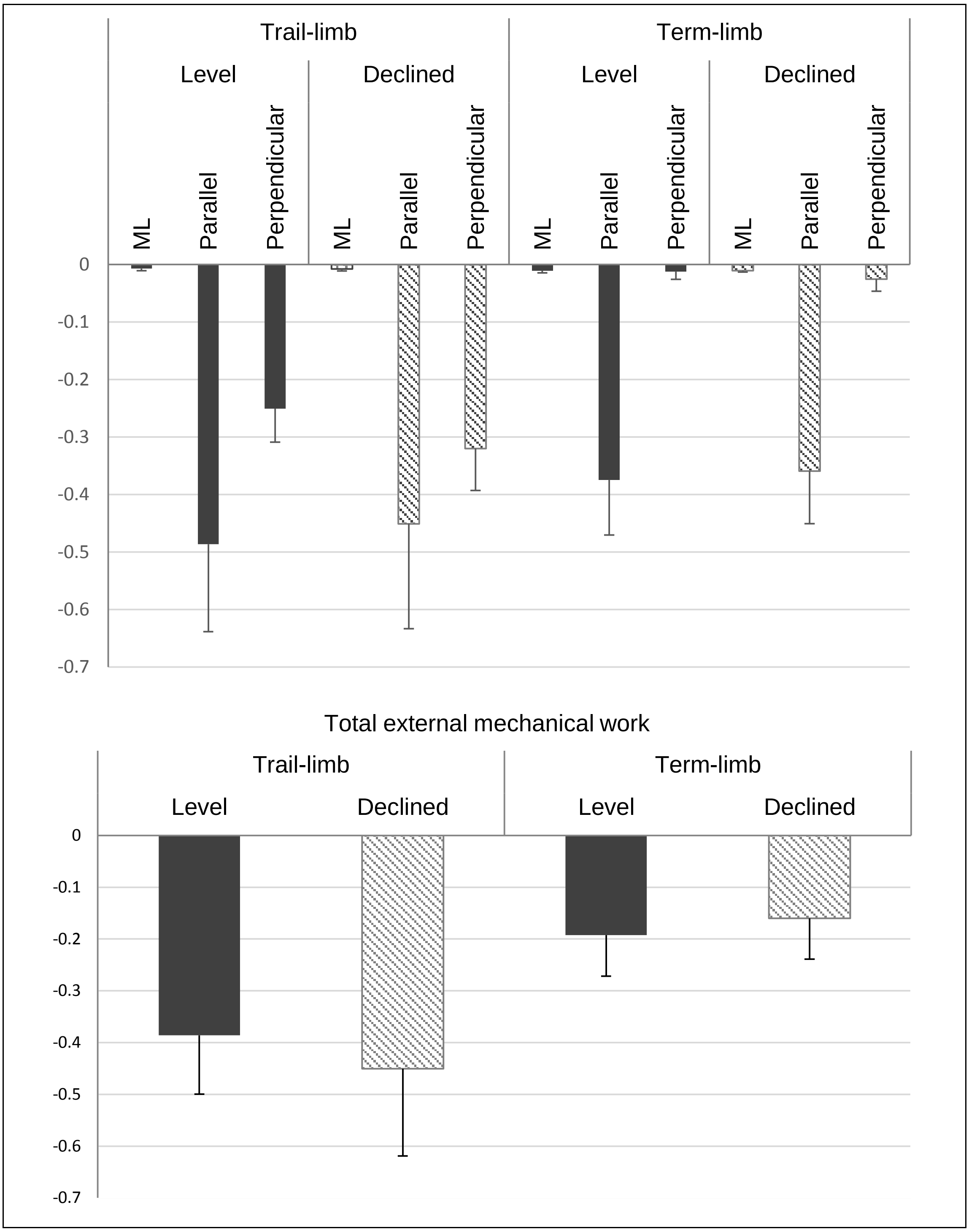


Figure 21. Group mean negative mechanical (external) limb work (J/kg) in all three orthogonal planes (ML, Parallel and Perpendicular) and total limb work for the trailing- and terminating- limbs. Hashed bars = declined surface; solid bars = level surface.

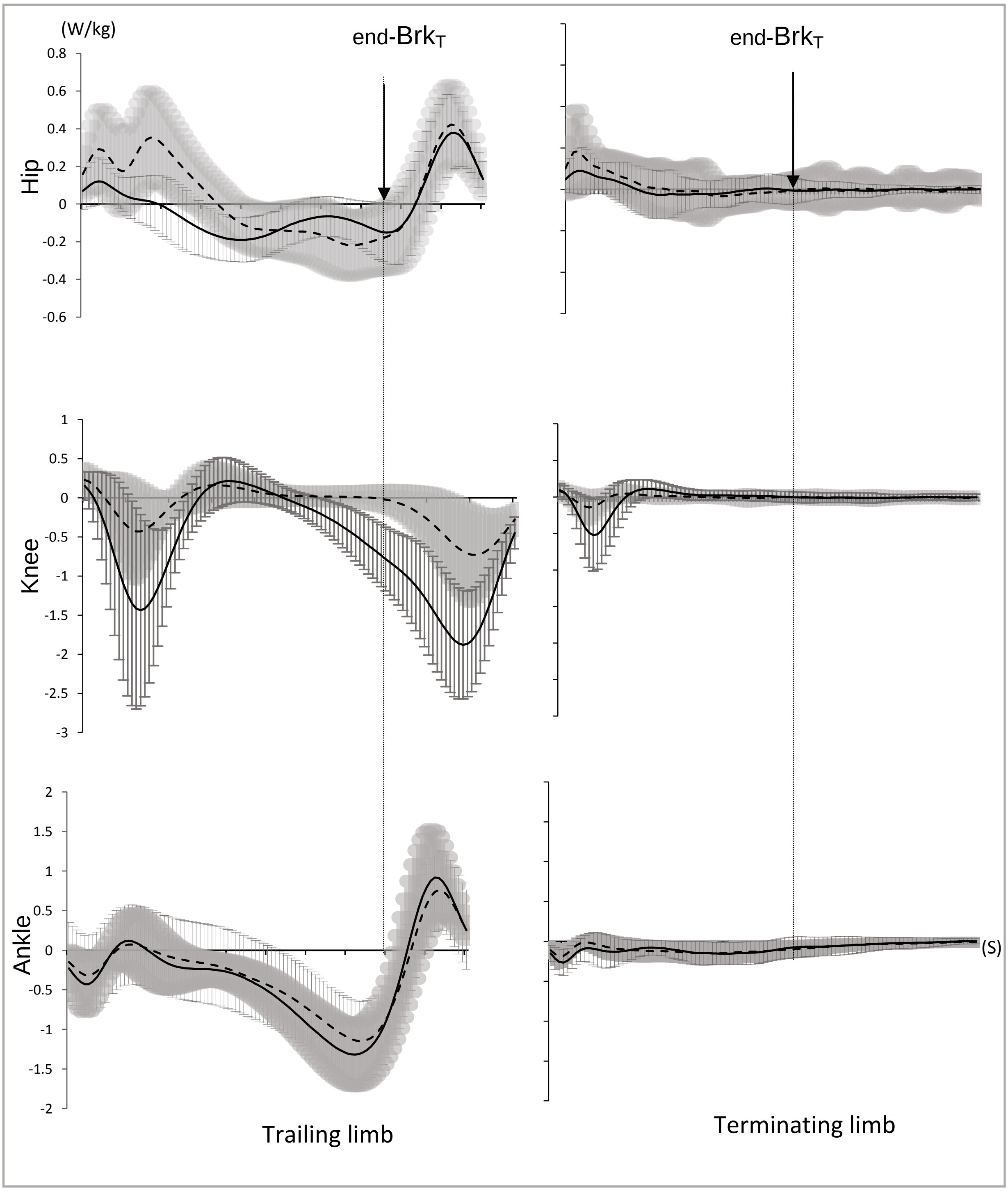


Figure 22. Group ensemble mean joint power profiles for the hip, knee and ankle joints (W/kg) of the trailing- and terminating- limbs. Data are plotted for stance phase of each limb with end of braking-phase indicated by vertical line. Bold line = declined surface; dashed line = level surface.

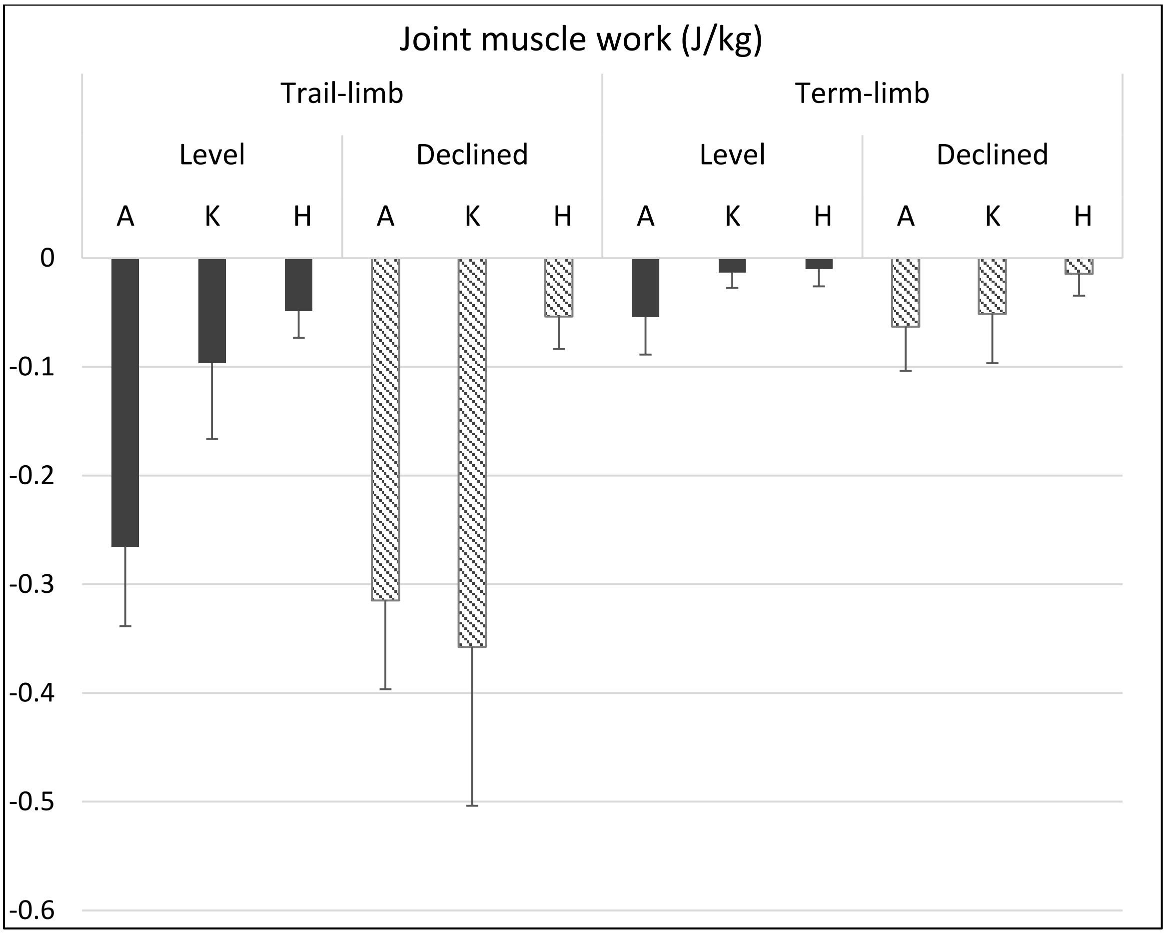


Figure 23. Group mean negative joint (internal) work for ankle, knee and hip for the trailing- and terminating- limbs (J/kg). Hashed bars = declined surface; solid bars = level surface.

Table 4. Negative mechanical limb work done by terminating limb and trailing limb, to terminate gait on levelled and declined surfaces.

	Terminating(J/kg)		*Trailing*(J/kg)	
participant	Levelled	Declined	Levelled	Declined
P01	-0.202	-0.207	-0.425	-0.359
P02	-0.255	-0.239	-0.382	-0.411
P03	-0.343	-0.266	-0.630	-0.823
P04	-0.116	-0.049	-0.336	-0.404
P05	-0.134	-0.092	-0.243	-0.279
P06	-0.141	-0.144	-0.356	-0.375
P07	-0.222	-0.195	-0.317	-0.553
P08	-0.128	-0.087	-0.401	-0.402
Mean	-0.193	-0.160	-0.386	-0.451

4.4 Discussion

The present study determined the external negative mechanical work done by each limb to terminate gait and how such work was affected by a change in surface angle from level to declined.

4.4.1 Limb work

Counter to what was hypothesised, results indicate that the mechanical work done to halt gait was done mainly by the trailing limb irrespective of surface angle. Specifically, the trailing limb did 67% of the overall negative work undertaken by both limbs to terminate gait on the level; and this increased to 74% in ramp trials. This means the limb that gait was terminated on only did 33% and 26% of the overall negative work in level and ramp trials respectively.

Previous research has shown that the terminating limb contributes considerably more of the braking (A/P GRF) force compared to the trailing limb when terminating gait on the level (Bishop et al. 2002; Lynch and Robertson 2007). In the present study A/P GRFs were similarly greater for the terminating compared to trailing limb for termination on both the level and decline. During planned/predicted stopping there is advanced information available regarding the distance and conditions of the future stopping location which can be used to determine the most efficient stopping strategy (Wearing et al. 1999; Bishop et al. 2002; Bishop et al. 2004). In such 'predicted' gait terminations the CoM velocity undergoes preparatory braking during the first step when it loses around 10% of forward speed before it undergoes rapid braking during the final step (Jian et al. 1993). As calculation of the external limb work takes in to account not only the magnitude of the GRFs but also the magnitude of the instantaneous CoM velocity, a rapidly reducing CoM velocity during the final step would result in considerably reduced limb work. This explains why in the present study the limb work done by the terminating limb ('final step') was considerably less than that done by the trailing limb ('first step') even though the braking forces were greater for the terminating compared to trailing limb.
During level walking, the total limb power (combined parallel, perpendicular and mediolateral) has a negative followed by positive phase (Donelan et al. 2002a),

and when walking down-slope, the total limb positive work reduces, and the negative work increases (and during upslope walking the total limb positive work increases while the negative work reduces) (Franz et al. 2012). In the present study, although the total limb power during the braking phase for both limbs was predominantly negative, there was a period of positive total limb power on both limbs (though such was negligible for the terminating limb in level trials) during the early-to-mid part of the braking phase).

4.4.2 Directional work

The greater negative mechanical work done by the trailing limb, was due to higher magnitude negative limb power (compared to terminating limb) in the parallel direction throughout the braking phase, and to a period of negative limb power in the perpendicular direction during the latter part of the braking phase (Figure 20 and Figure 21). The higher magnitude negative limb power in the parallel direction throughout the braking phase reflects the increased negative mechanical work done by the trailing limb to arrest the CoM velocity. The increased negative limb power in the perpendicular direction for ramp compared to level trials would have been a result of the CoM being lowered more/further in such trails. There was no negative limb power in the perpendicular direction evident for the terminating limb, apart from a very short period immediately following foot contact. This would be expected as the CoM was halted on this limb rather than being lowered/transferred to the next step. The mechanical limb total power profile for the trailing limb for terminations on both level and decline is similar (in shape and magnitude) to that reported for constant speed walking on a decline (Franz et al. 2012).

Previous research has indicated that positive limb power at this period in stance (i.e. double-support period) is due to the ‘push’ from the contralateral limb, i.e. transition from one stance limb inverted pendulum to the next (Donelan et al. 2002a). As evident in the present study this ‘push’ was a result of positive limb power in the perpendicular direction (i.e. upwards ‘push’, Figure 19). The amount of positive limb work done in the perpendicular direction was reduced for the terminating- compared to trailing- limb, which indicates the transfer of bodyweight onto the terminating limb occurred with a reduced upwards ‘push’ from the contralateral (trailing) limb. For the trailing limb, there was also a short period in late stance (following contralateral limb foot contact) when limb power in the parallel direction became positive. This period of positive limb power likely acted to help transfer the body CoM forwards onto the terminating limb. No such positive parallel power was evident for the terminating limb, which would be expected because the CoM was halted on this limb rather than being transferred forwards onto the contralateral limb. There was also positive limb power evident in the mediolateral direction on the trailing limb during the latter part of the braking phase. This positive power would have occurred because as the CoM was being transferred forward onto the contralateral limb it would have also moved slightly sideways (rightwards) from the trailing (left) limb towards the terminating (right) limb.

4.4.3 Joint work

The greater negative mechanical work done by the trailing compared to terminating limb was associated with greater amounts of negative work at the joints on the trailing limb, particularly at the ankle during the latter part of the braking phase (Figure 22). Negative ankle joint work done in the latter part of the braking phase reflects the control exerted on shank to govern how quickly it rotated forwards over the planted foot during single-limb support, which in turn governed how quickly the CoM progressed over the planted foot. At the onset of trailing limb single-support the CoM would be forward of the ankle and thus would begin to ‘fall’ (inverted pendulum). As highlighted above, negative limb power in the perpendicular direction also increased during the latter part of single-limb support (Figure 22). This suggests that there is an association between limb negative power in the perpendicular direction and negative ankle joint power, and thus negative ankle work (power absorption) acts to control the lowering of the CoM during single-limb support (i.e. acts to control inverted pendulum).

For gait terminations on the declined surface, knee negative joint work markedly increased (p=0.0005), particularly on the trailing limb. This suggests that kinetic adaptations at the knee are important in controlling the increased CoM lowering required for ramp descent. This finding is in agreement with studies showing that during down-slope walking, peak power absorption increases markedly at the knee joint compared to that for when walking on the level (Kuster et al.

1995; Lay et al. 2007). In the present study, it is worth noting that the increase in trailing limb negative knee joint work for ramp compared to level trials was due to increased knee negative power during both the initial and final part of stance (Figure 23). The increase in knee negative joint power during initial stance corresponds with a period of negative limb power in the parallel direction (Figure 19). This suggests that negative knee joint work is predominant in reducing CoM forwards velocity during limb loading (weight acceptance period), particularly so for gait terminations on a ramp; as evidenced by the negative knee joint power during this period increasing in magnitude compared to level trials. The period of increased knee negative joint power during late stance 'peaked' in magnitude after contralateral limb contact and appeared not to be temporally associated with any directional component of limb power. This suggests that the knee was flexing compliantly (no effect on the CoM) during late stance. Such compliant flexion may have occurred to ensure there was a minimal increase in CoM height, as it was transferred from the trailing to the terminating limb.

The results presented provide an understanding of the mechanical (external) limb work in all three orthogonal directions as well as the overall limb work, and an understanding of the joints (internal) work done in achieving such mechanical limb work. However, because only sagittal plane joint work was computed, this may have underestimated the total joint work done. We investigated only sagittal plane joint work because we reasoned that since the task of terminating gait predominantly involves arresting CoM forward velocity, this would mainly be achieved via joint power absorption in the sagittal plane (Allard et al. 1996). The relatively small magnitude of mechanical limb power evident in the mediolateral direction indicates that this was indeed the case. Furthermore, frontal and transverse plane joint powers are known to be highly variable (Eng and Winter 1995) and thus their interpretation would likely be problematic. Another limitation was that the two force platforms used to collect GRF data were located next to each other with minimum spacing between them, and thus this may have affected certain participant's stopping strategy more than others, e.g. taller participants may have had to 'chop' their intended foot placements to ensure they were within the bounds of the platforms.

However, none of the eight participants were observed to have such difficulties, and all appeared to carry out the stopping task in an apparently natural manner.

4.5 Conclusion

In conclusion, this study indicates that during the two locomotor steps of gait termination, the limb that gait is terminated on only does 33% and 26% of the overall negative mechanical work done by both limbs on the CoM when terminating gait on the level and declined surface respectively. In other words, the trailing limb does the majority of mechanical work in arresting CoM forwards velocity. Negative joint work was also greater for the trailing compared to terminating limb, with negative ankle work in late stance being the foremost contributor. The increased trailing-limb negative ankle work was associated with an increase in negative limb work in the perpendicular direction, highlighting the ankle's role in slowing rotation of the limb (and thus CoM) over the planted foot (i.e. controlling inverted pendulum). Negative joint work increased on both limbs for declined compared to level surface, particularly so at the knee; indicating kinetic adaptations at the knee are important in controlling the increased CoM lowering required for ramp descent. A peak in negative knee joint power in early stance was associated with a peak in negative limb power in the parallel direction, highlighting the knee's role in slowing CoM forwards velocity during weight acceptance onto the limb.

Chapter Five

Contribution of Terminating and Trailing Limb Work in Arresting Centre of Mass Velocity during Gait Termination on Declined; Effect of Speed

5.1 Introduction

Compared to walking on the level, walking down-slopes is considered as a more difficult task as it requires additional muscle actions to lower body CoM (Lay et al. 2007; Franz and Kram 2012), more external negative limb work (Franz et al. 2012), and increased peak of negative joint power (Schwameder et al. 2005). Consequently, down-slopes walking has a greater risk of slip-related falls (Redfern et al. 2001). As highlighted in the proceeding chapter, a more neuromotor challenging task than down-slope walking is to terminate gait when down-slope walking. Coordinating how CoM velocity is arrested and controlled to a stable base of support requires the limbs to perform negative work on CoM. Presumably, with a higher speed of walking prior to gait termination the need to increase the amount of negative work to perform such coordination of CoM and halt gait would increase.

Studies on AB steady (constant) locomotion showed that at a given speed no net mechanical work was observed i.e. internal joint work is equal to external limb work. However, moving at faster speed demands more positive and negative internal joints work per stride (Farris and Sawicki 2011). The normalized temporal patterns of leg muscle activity found to remain fairly stable but with increasing amplitudes (Hof et al. 2002). This was not the case when walking down-slope since only knee extensor muscle activities increased to walk down-slope with change in muscle activations as the grade became greater with faster walking speed (Franz and Kram 2012). It is indefinite/uncertain if the same trend generally occurs in limb work when terminating gait from different walking speeds down-slope. The mechanical work done to stop CoM velocity is likely to be mostly negative when terminating gait (chapter 4). Arresting CoM velocity, a combined with lowering it on declined surface during gait termination from customary speed required even greater negative mechanical work on a declined compared to that on the level surface as highlighted in the previous chapter. Therefore, a greater negative mechanical work would be expected as the speed of walking increases for gait termination on the declined surface, but it is uncertain if this increase tends to be equal at both terminating and trailing limbs.

Although examining the effects of increased walking velocity on how gait termination is performed was considered by (Bishop et al. 2002), stopping on a declined surface with different walking speeds is not reported yet. The purpose of the present study was to determine whether mechanical work done by each trailing/terminating limb was synchronised with speed during terminating gait on the declined surface and whether directional work components at each limb would be affected by speed of stopping, thus provide an insight into how the speed of walking has an impact on the work done to halt CoM velocity on the declined surface and particularly in which direction. The specific aim was to provide further understanding of how stopping is controlled and performed when descending slopes. Understanding how AB stop has implications for the impaired populations i.e. amputees in which this thesis particularly interested in. This could help provide an insight into factors affecting TFAs gait termination strategy (in the next two chapters).

5.2 Methods

5.2.1 Participants

Same AB group was participated in the previous study conducted in chapter 4 completed gait terminations on a declined ramp (5 deg) at self-selected slow, customary and fast walking speeds.

5.2.2 Experimental protocol

Participants walked wearing their own shoes and completed the self-selected customary speed trial first. Fast and slow block order were counterbalanced across participants. The participants were asked to complete termination from walking faster than they would normally walk. For slow speed condition, they were asked to walk slower than they normally walk. For each walking speed, participants walk down the ramp, the trailing foot being landed on first platform, then participants lead stopping with terminating foot on the second force platform facing forward with both feet within the area of the ‘square’ and hold that position until informed to move.

5.2.3 Data acquisition and processing

For each participant, terminating and trailing limb data from ten trials at each speed were analysed. All participants terminated gait with their preferred-limb (limb they would use to kick a ball), which in all participants was the right limb.

For further details about data acquisition and processing of measured variable are provided in section 3.6 method chapter 3

5.2.4 Data analysis

Section 3.7 method chapter explain more details regarding outcome variables being assessed. External negative mechanical work (total and work component in each of the orthogonal directions) were determined for each limb for each trial and then averaged across trials to give mean values for each speed condition (slow, customary and fast) per participant.
Walking speed was determined as the peak CoM forward velocity at the instant of contralateral foot contact (i.e. at the start of the two-step gait termination).
Time of stopping was determined as the time-period between contralateral foot contact up to instant of final bipedal stance.

5.2.5 Statistical analysis

Limb directional- and limb total- negative work plus, time taken for stopping, walking speed prior to gait termination, and duration of braking phase were compared using repeated measures analysis of variance (ANOVA) with speed condition (slow, customary, and fast) and limb (terminating, trailing) as repeated factors. Post-hoc analyses were undertaken using Tukey HSD tests. The alpha level was set at 0.05.

5.3 Results

Group ensemble average directional- and total- power profiles for each limb across the three speeds are presented in Figure 24. Group mean (±SD) directional- and total- negative work values for each limb for the three speeds are shown in Figure 25. Group average Terminating and Trailing limb work as a percentage of total negative mechanical work (%) done by both limbs during gait terminations from slow, customary and fast speeds is presented in Figure 26.

Speed of walking prior to terminating gait was 0.86 (0.21), 1.15 (0.18) and 1.52 (0.20) m/s for slow customary and fast speeds respectively, and each speed was significantly different to other two ($p<0.01$). Time taken to stop significantly decreased with speed ($p<0.01$); 1.38 (0.19), 1.06 (0.07) and 0.85 (0.09) s for slow customary and fast speeds respectively.

5.3.1 Limb work

The power profiles of both terminating and trailing limbs (Figure 24) were considerably increased in magnitude and decreased in period as walking speed increased. Both terminating and trailing limbs showed that the duration of the braking phase (single support phase) decreased ($p<0.01$) as the walking speed increased.
Limb negative work (Figure 25 a) increased with speed for both terminating and trailing limb (effect of speed, $p<0.001$. Terminating limb negative work (Figure 25a and 26) was significantly smaller compared to that done by the trailing limb (i.e. effect of limb $p<0.01$, 0.14, -0.29, -0.38 J/kg for slow, customary and fast speeds respectively). A trend interaction ($p<0.001$) between limb and speed indicated that both terminating and trailing limb negative work increased with speeds, but a greater speed effect was observed on the trailing limb.

5.3.2 Directional work

The magnitude of limb power and work in the parallel and perpendicular direction increased with walking speed but power in the ML direction reduced with speed (Figure 24, 25b). Irrespective of speed, negative mechanical work in the parallel direction for both limbs provided the largest contributions to limb negative work. Followed by the second largest contributor being the perpendicular work component and the least being ML work (Figure 25b). More negative work was done by the trailing- compared to terminating- limb in the parallel-AP and perpendicular directions however in the ML direction less work was done by the trailing- compared to terminating- limb.

Limb perpendicular negative work was significantly affected by speed ($p=0.0017$) and limb ($p<0.001$) but there was no interaction between terms ($p=0.097$). Limb parallel negative work increased significantly as speed increased ($p<0.001$) but there was no main effect of limb ($p=0.073$). However, a limb by speed interaction ($p<0.001$) indicated that as speed increased from slow to fast, parallel negative work done increased on the trailing limb but was unchanged on the terminating limb. Limb negative work in ML direction was also significantly affected by speed ($p<0.001$) and by limb ($p=0.036$), compared to the terminating limb less negative work was done by the trailing limb in ML direction. The trend interaction ($p= 0.0085$) indicating that negative work done by the trailing limb in the ML direction was decreased as the speed increased ($p< 0.05$, $p<0.001$ for customary and fast speed respectively).

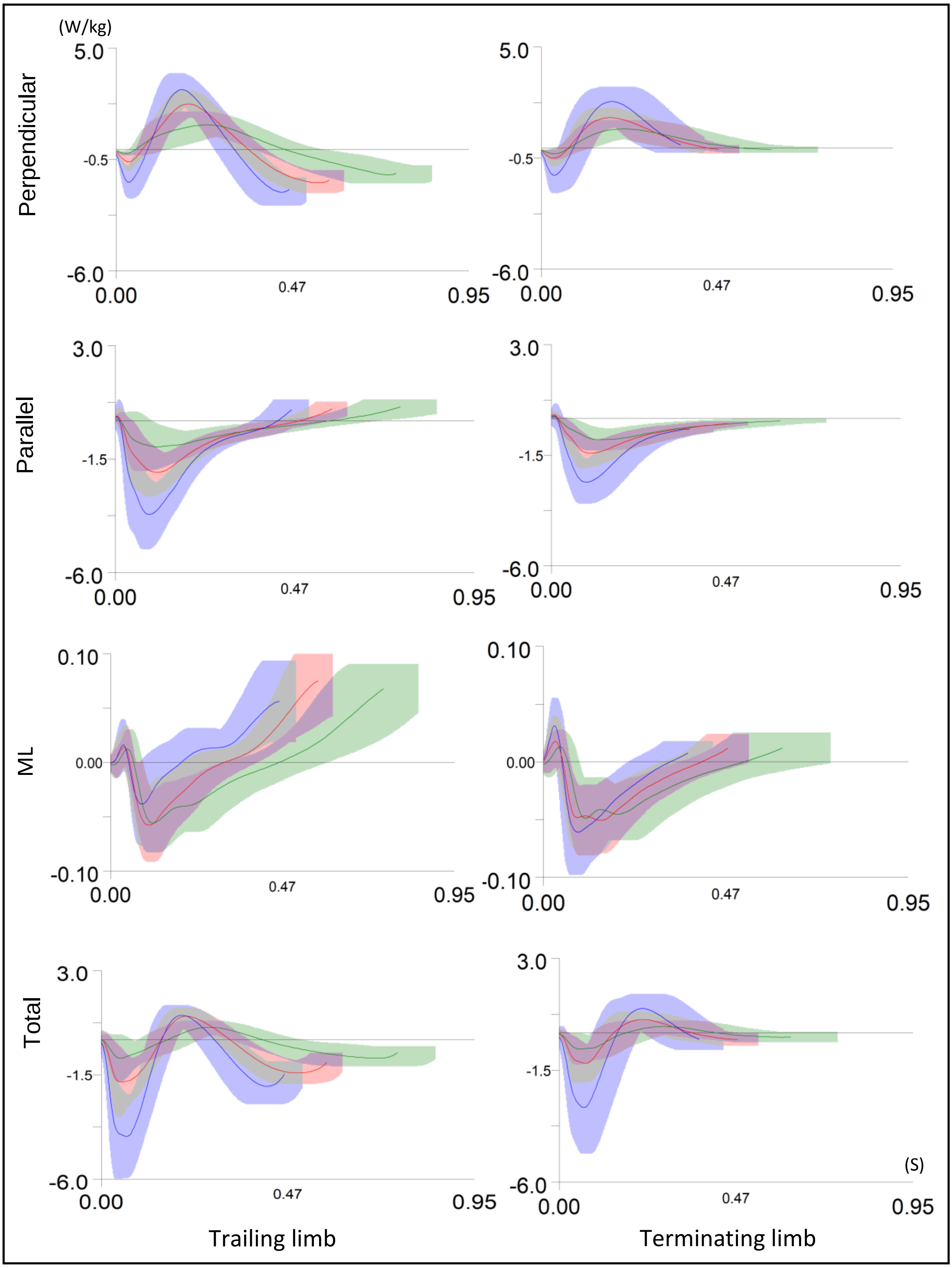


Figure 24. Group ensemble mean (+/-SD band) limb power profiles (bottom)- and directional perpendicular (top), parallel, ML components (W/kg) for the trailing and terminating limbs. Data are plotted for corresponding braking-phase of each limb. Blue line = fast speed; red line = customary speed and green line=slow speed.

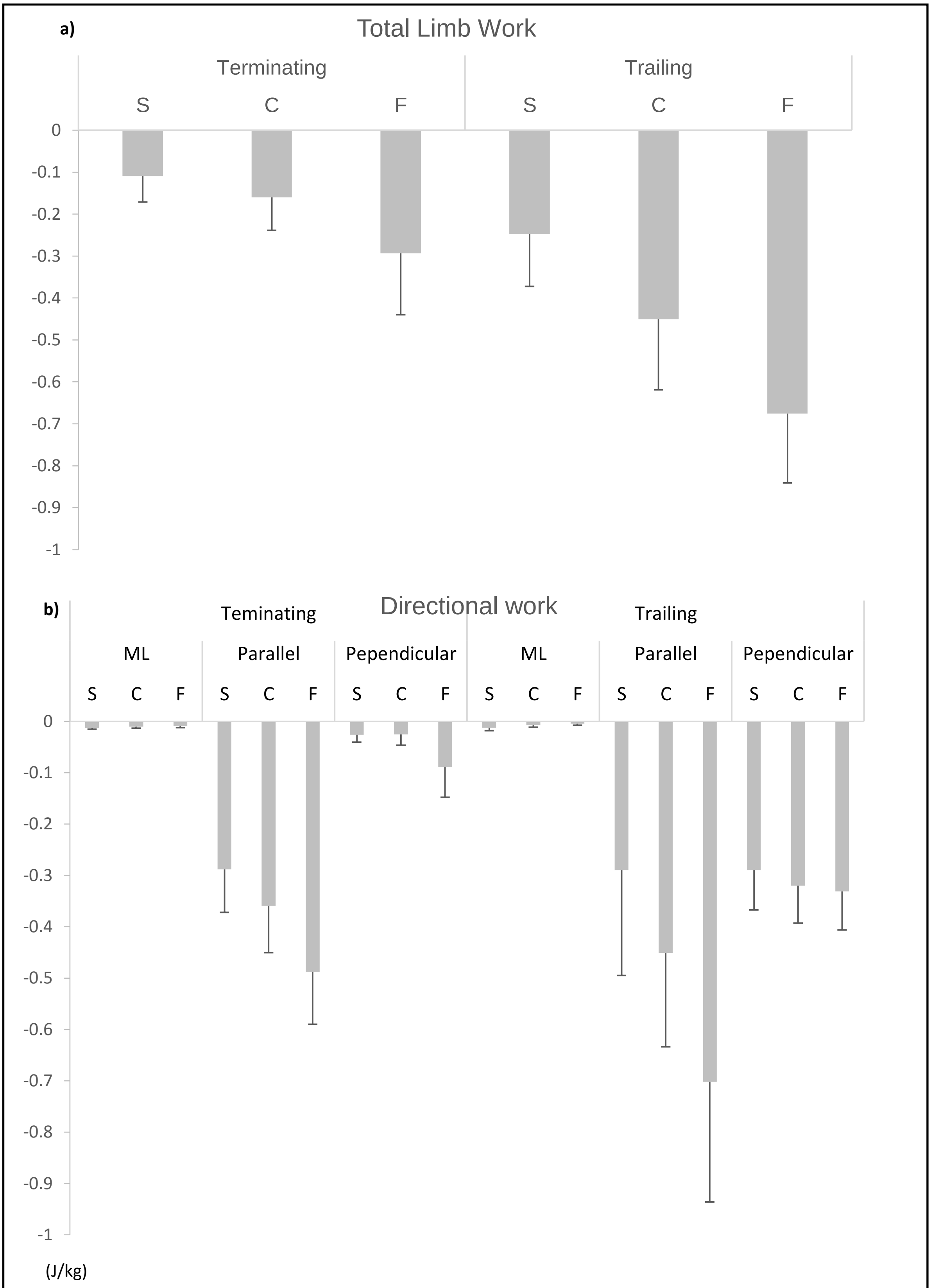


Figure 25. a) Group mean (+SD) limb negative work (J/kg) and b) negetive work in all three orthogonal planes (ML, Parallel and Perpendicular) for the terminating- and trailing- limbs at slow (S), customary (C) and fast (F) speed.

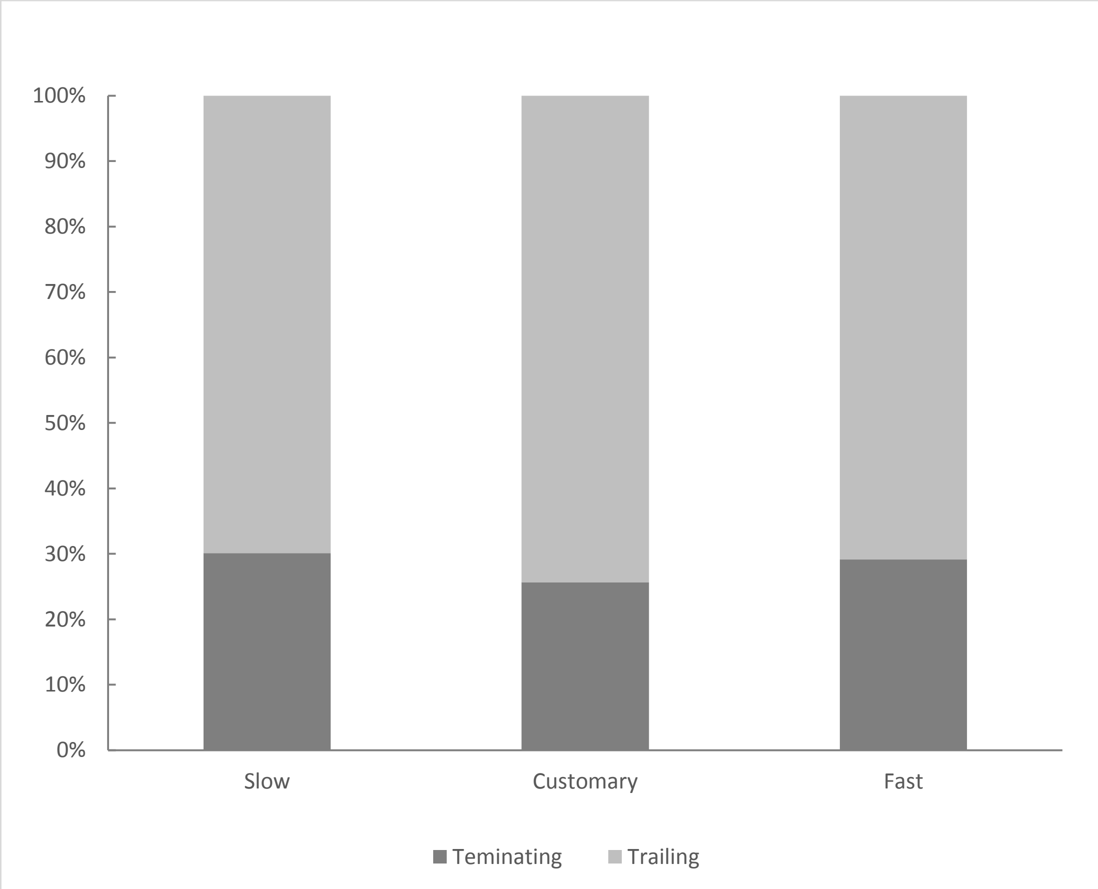


Figure 26. Group limb work as a percentage of the total negative mechanical work done (%) during gait terminations on declined surface from three speeds for the Terminating limb (dark grey shading) and the Trailing/intact limb (light grey shading).

Table 5. Negative mechanical limb work done by terminating limb and trailing limb, to terminate gait from walking at slow, customary and speeds.

	Terminating(J/kg)			*Trailing(J/kg)*		
Participant	SLOW	CUST	FAST	SLOW	CUST	FAST
P01	-0.068	-0.207	-0.389	-0.141	-0.359	-0.563
P02	-0.187	-0.239	-0.488	-0.188	-0.411	-0.702
P03	-0.182	-0.266	-0.396	-0.499	-0.823	-1.020
P04	-0.032	-0.049	-0.104	-0.207	-0.404	-0.617
P05	-0.092	-0.092	-0.190	-0.203	-0.279	-0.629
P06	-0.065	-0.144	-0.391	-0.170	-0.375	-0.503
P07	-0.176	-0.195	-0.289	-0.383	-0.553	-0.795
P08	-0.072	-0.087	-0.102	-0.191	-0.402	-0.574
Mean	-0.109	-0.160	-0.294	-0.248	-0.451	-0.675

5.4 Discussion

The present study determined how the external mechanical work done by the trailing and terminating limbs to terminate gait from down-slope walking was affected by walking speed. On both limbs the mechanical work done to halt gait increased with walking speed, but the work done was mainly by the trailing limb, with 69%, 74%, and 70% of overall negative work undertaken by both limbs at slow, customary, and fast speeds (Figure 26). Previous research has shown that negative mechanical work and power in general increased as walking speed increased (Teixeira-Salmela et al. 2008; Browning et al. 2009; Franz and Kram 2012). The current study indicated that at higher speed of walking prior to gait termination the need for trailing limb negative work to halt CoM for stopping was increased.

The greater negative mechanical work done by the trailing limb was primarily due to higher magnitude negative limb power in the parallel direction compared to that of the terminating limb (Figure 25) highlighting that halting body CoM velocity was performed mainly by the trailing limb parallel work even at different speed prior to stop. The period of negative limb power in the perpendicular direction during the latter part of the braking phase (Figure 24) was the second-place contributor to trailing limb work. The magnitude of directional power increased across speed because both CoM and GRF were increased in the corresponding trails. The greater magnitude negative limb power in the parallel direction reflects the increased negative mechanical work done by the trailing limb to arrest the CoM velocity (chapter 4). This work found to be increased with speed in the current study. The increased negative power in the perpendicular direction performed by the trailing limb as the speed increased might be for the reason of assisting the terminating limb lowering CoM (already elevated more with higher speed) during latter part of braking phase (chapter 4).

Interestingly, more work in the ML direction was done by the terminating compared to trailing limb. This trend was stronger with increased walking speed prior to stopping. The increase in ML work by the terminating limb is likely related to the act of placing the trailing foot next to the terminating foot to initiate final bipedal stance. Consequently, the CoM forward velocity is halted and transferred laterally relative to the terminating limb in order to ensure the CoM lies between the feet during final bipedal stance. Thus, the higher walking speed prior to gait termination the greater magnitude of negative work will be absorbed by the terminating limb in ML direction required to move CoM above the stance terminating foot. It is worth to note that the whole-body will tend to fall laterally towards the trailing-limb side during the execution of final step (as trailing limb is swung forwards). This may be why many TFA prefer to terminate gait on their prosthesis – so they can compensate for sideways ‘falling’ with their intact limb.

This study showed that the time taken to perform two steps of gait termination was significantly decreased with speed indicating that braking phase period (double and single support phase) was reduced as the speed increased. The time duration of the stance phase and double support phase reported in the literature (Figueiredo et al. 2011) showed a decrease value with the increase of walking speed. This behaviour seems to be consistent in the same way in case of gait termination. The reduced braking phase duration would cause relatively reduced dynamic stability of gait which perhaps related to a compensation effect of reduced ML negative work with speed (Figure 25b). The reduced duration of stance phase might be associated with increasing intensity of muscle activation (Chiu and Wang 2007). Bishop found that the sequence of muscle activation was not different between normal walking and planned stopping trials. However, the duration of muscle activity was longer during planned stopping across all velocities (Bishop et al. 2002).

For the three speeds, there was no period of negative limb power during the latter part of the braking phase in the parallel and perpendicular direction demonstrated by the terminating limb consistent with pervious findings in chapter 4. This would be expected as the participant planned to stop; even with higher speed the participant was aware of the time and location to stop so that the CoM was halted and elevated on the terminating limb then transferred towards the trailing limb via positive work especially at fast speed, thus, there was no need for negative work. Although gait termination involved increases in negative power with speed, there was also an effect of speed on the positive total limb power on both limbs during the early-to-mid part of the braking phase which was also increased with higher speed. This might be related to body CoM excursions/fluctuations in the vertical direction or even the vertical force peaks as the force impact also increased with speed (Figueiredo et al. 2011).

The interaction between limb and speed indicated that trailing-limb negative work highly increased with speeds whilst the contribution of the terminating-limb also increased; post-hoc analyses indicated these increases were significant. The study quantified individual limb contributions to total mechanical work done for a range of walking speeds. It was hypothesized that faster locomotor speeds prior to stop would result in proportionally increasing negative work for both limbs. The data provided strong support for this hypothesis within gait termination.

5.5 Conclusion

In conclusion, stopping forward progression requires reduction of the forward momentum of the CoM. So that CoM will be slowed, the dynamic base of support prepared and CoM controlled about this dynamic base of support to remain upright. Limb negative work required to bring the CoM to a complete and controlled stop is dependent on speed of walking prior to stop. In this chapter limb negative work engaged to do this, was increased with speed and was predominantly attributed to trailing limb.

Chapter Six

External Mechanical Work during Gait Termination on Declined versus Level Surface in Trans-Femoral Amputees: Effect of Limb System Prosthesis

6.1 Introduction

Experimental chapter 4.0 brought to light the impact of two interrelated factors on the accomplishment of gait termination in AB. The first was the contribution of the terminating and trailing limbs on the work performed in arresting CoM forwards velocity – on level surface. The second was how a change in surface angle from level to decline affected both the terminating and trailing limbs negative work. Compared to terminating gait on level surface, it was found that trailing limb negative work was increased when terminating ramp-descent. Trailing limb increased work on declined surface was mainly attributed to the knee joint highlighting the importance of this joint to ramp descent. TFAs may have a greater reliance on the trailing limb compared to AB since TFA cannot rely on the prosthetic knee as it does not flex.

As presented in different studies reviewed in the literature (Redfern and DiPasquale 1997; Lay et al. 2006; Franz and Kram 2012), it can be deemed that ramp added an additional burden on users' intact limb. The increase of trailing limb load/ usage was observed not only during walking but also during gait termination according to what was found by previous chapters (4 and 5) in this thesis. However, the characteristics of a limb system (detailed in method chapter section 3.2.2.2) can facilitate performing gait termination. The current study was primarily aimed at determining whether a limb system affects the negative limb mechanical work done on CoM during gait termination in TFAs. A secondary aim was to determine the changes that occur to such work as a result of a change in surface angle from level to declined surface. In satisfying the above aims the study investigated the external mechanical negative work done by the prosthetic and intact limbs when TFA terminated gait (with prosthetic limb being the terminating limb) from down-slope and level walking. It was hypothesised that the use of a limb system prosthesis with ramp descent mode would lead to more negative work (power absorption) being done by the prosthetic limb when the device's MC was active compared to inactive. Hence intact limb work can be reduced when terminating gait on the ramp compared to a level surface.

6.2 Methods

6.2.1 Participants

Seven male physically active TFAs (mean (SD), age 48.8 (13.5) years, height 1.77 (0.077) m, mass 83.6 (15.15) kg), time since amputation 13.4 (7.1), participated in this study. Further details regarding participants' preparation are given in the method chapter (sections 3.2.2)

6.2.2 Data collection and processing

Participants were asked to complete repeated trials involving walking at self-selected walking speed on level and declined walkways (method chapter section 3.3). Trials at each surface condition were repeated 10 times with a block of 5 trials completed with the MC active (MCon) and 5 trials with it inactive (MCoff). Terminations were completed for both surface condition with the terminating limb always being the prosthetic limb. The level surface condition was always first and MC conditions (MCon/MCoff) were counterbalanced across participants. Further details regarding the experimental aspects, including data collection and protocol etc. can be found in section (3.5)

6.2.3 Data analysis

Limb external mechanical power was determined as the sum of limb mechanical powers in each orthogonal direction (Donelan et al. 2002b). Briefly, 'braking-phase' was determined as the period between foot initial contact up to contralateral-limb foot contact and limb negative work, was determined as the time integral of negative mechanical power during the braking phase of each limb. Further details regarding how the above outcome variables were determined are provided in section 3.7

Note, the limb system's 'ramp descent' mode is programmed to alter the knee resistance to an intermediate level during the late stance period of a normal gait cycle. Hence, this functional change would not be activated when terminating gait. To confirm this, it was decided to determine if there was any difference across MC conditions in knee angular displacement for the prosthetic (terminating) limb. The results showed that knee was remained fully extended irrespective of MC condition.

6.2.4 Statistical analysis

Repeated measure analysis of variance (ANOVA) and percent differences were used to highlight differences in outcome variables between limbs and between MC conditions. Limb external negative work was compared using repeated measures ANOVA with surface condition (level, declined) and limb (Intact, prosthesis) as repeated factors. Post-hoc analyses were undertaken using Tukey HSD tests. Statistical analyses were performed using Statistica (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

In order to understand how the work done by the terminating (prosthetic) and trailing (intact) limbs compares to that in able-body individuals, group ensemble-average power profiles (±SD band) for each limb were plotted alongside ensemble average power profiles (±SD band) for a group of able-body individuals (level, declined trials, Figure 27). The data for able-body individiuals are from chapter 4. A comparison of the current data to these data allowed us to subjectively evaluate how TFA terminate gait on a ramp in comparison to how able-body individuals do.

Furthermore, to confirm that the hydraulic resistances did indeed change on declined surface in the way that the manufacturer claimed (explained previously method section 3.2.2.2), the following factors were evaluated (via t-test) when the MCon compared to MCoff on declined only ; a) if foot-flat (tracking the toe marker vertical velocity (Struchkov and Buckley 2015) was obtained sooner, b) whether CoM dynamic control was improved i.e. CoP being more anterior to CoM as suggested enhanced dynamic stability (Jian et al. 1993; Winter et al. 1998; Hof et al. 2005; Vrieling et al. 2008a), c) if the shank's forward rotation (i.e. assessing CoP forward velocity after foot flat) over the foot was slowed, when the device was active, d) finally, to understand the underlining reasons of any increase in prosthetic limb negative work investigation of the CoM velocity and forces (peaks) was also performed.

6.3 Results

TFA group ensemble average mechanical limb power profiles (W/kg) for terminating (prosthetic) and trailing (intact) limb on level and declined surface for the MCon and MCoff conditions are presented in Figure 27a. AB group ensemble average power profiles for terminating and trailing limb for level and declined surface trials are presented in Figure 27b. Group mean (±SD) negative limb work values (J/kg) for each limb for both TFA and AB on level and declined surface are depicted in Figure 28. Group average limb work as a percentage of total negative mechanical work (%) done on CoM during gait terminations on level and declined surfaces for the terminating (prosthetic) limb and the trailing (intact) limbs are presented in Figure 29.

6.3.1 Prosthetic limb

Negative mechanical work done was significantly affected by surface ($p<0.05$) but was unaffected by the MC condition ($p=0.15$), however, there was an interaction between terms ($p=0.04$). Post-hoc analysis revealed that this interaction indicated prosthetic limb negative work was greater for the MCon compared to MCoff condition but only on the declined surface (Figure 28).

Analysis of the CoM velocity and forces (peaks) was done to understand the underlying reasons of this increase in prosthetic limb negative work. The analysis CoM velocity and forces showed that increased work was attributed to a significant increase in both force and CoMv across MC conditions but only in A/P direction (CoMvy differences: MCon = 0.68, MCoff = 0.65 m/s; $p=0.027$ and GRFy differences: MCon = 1.30, MCoff = 1.15 N/kg; $p=0.038$). The horizontal work (negative work in parallel direction) was also significant across MC condition $p=0.00069$. Foot-flat was obtained sooner when the MC was active (MCon=0.235, MCoff=0.267 s; $p=0.010$). CoP velocity under the prosthetic limb was faster (MCon=0.348, MCoff =0.234 m/s; $p=0.050$) plus, more anterior displacement of CoP also was observed on declined (MCon=0.051, MCoff=0.034 m; ($p=0.060$: trend only). The time from foot-flat to contralateral foot contact was longer (MCon=0.282, MCoff=0.230 s; $p=0.07$, trend only) for the MCon compared to MCoff condition.

In percentage terms (Figure 29), the negative mechanical work done by the prosthetic limb on the declined surface contributed about 3% more to the total work done by both limbs for MCon compared to MCoff condition. On level surface, there was a reduction in negative work contribution of the prosthetic limb to the total limb work by 1% for MCon compare to MCoff condition.

6.3.2 Intact limb

Irrespective of MC condition, intact limb power during later part of the braking phase was greater when terminating gait on declined compared to level surface (Figure 27). As a result, negative mechanical work done was significantly greater when terminating gait on declined compared to level surface (Figure 28) ($p=0.0044$). However, the negative mechanical work done was unaffected by the MC condition ($p=0.89$) and there was no interaction between terms ($p=0.23$).

In percentage terms (Figure 29), intact limb negative work contribution to the total limb work done by both limbs was increased by 13% for declined compared to level surface at MCoff condition. For MCon, intact limb negative work contribution to the total limb work done by both limbs was increased by 9% for declined compared to level surface.

6.3.3 Comparison to able-body individuals

The power profiles of the terminating (prosthetic) limb on both surfaces were similar (shape and size) to that of the AB group (Figure 27a and b). The power profile of the intact limb on declined surface, highlights the negative power during late stance was greater (magnitude and period) than that of the trailing limb of AB (Figure 27a and b). Except for a brief period- trailing limb power was negative, particularly on the declined surface, for almost the entire braking phase: with the positive period being noticeably smaller (shorter, smaller magnitude) than that in able bodied individuals. Differences of limbs' power profiles (peaks and magnitudes) across surfaces were more pronounced for able bodied than that of TFA, except a short period during late stance of the intact limb when the power profile on declined exceed that on level surface.

In AB, the negative work done by the terminating limb to halt gait on level surface, contributed around 33% of the total negative work done by both limbs and this contribution decreased to 26% on declined surface. For TFAs, the terminating (prosthetic) limb contributed to the total negative work done by both limbs; on the level surface was 29.2% and 28% for when the MC was inactive and active respectively. On declined surface, the prosthetic limb contribution to the total negative work done by both limbs decreased to 16.2% and 19% when the MC was inactive and active respectively.

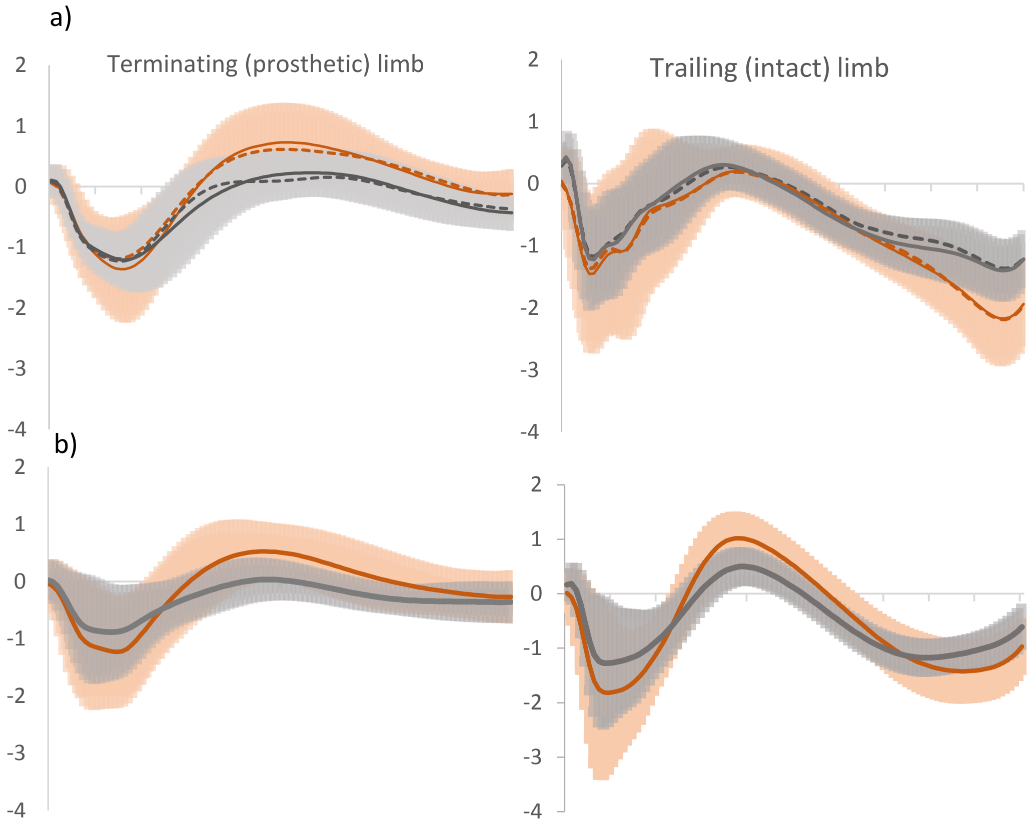


Figure 27. a) TFA group ensemble mean limb mechanical power (W/kg) during gait terminations on level and declined surface for the terminating/prosthetic (left panels) limb and the trailing/intact limb (right panels). Black bold line= *MC*on; dashed line= *MC*off. b) For comparison, group ensemble average power profiles of AB group (chapter 4) for terminating and trailing limb for level (grey line) and declined surface (black line) trials. Grey band = group SD for AB on level surface, orange band= group SD for AB on declined surface. Both groups' data are plotted for the braking phase of each limb.

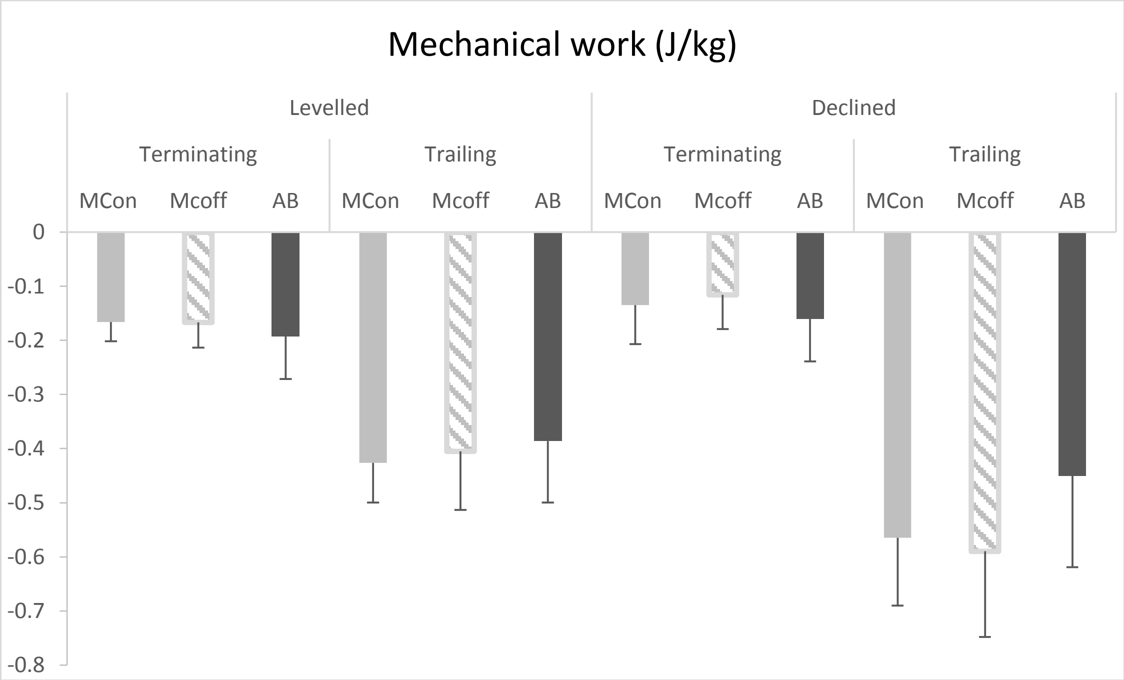


Figure 28. TFA group mean (+SD) limb negative mechanical work done (J/kg) during gait terminations on level and declined surface for the terminating/prosthetic limb and the trailing/intact limbs. AB data, from chapter 4, are shown for comparison.

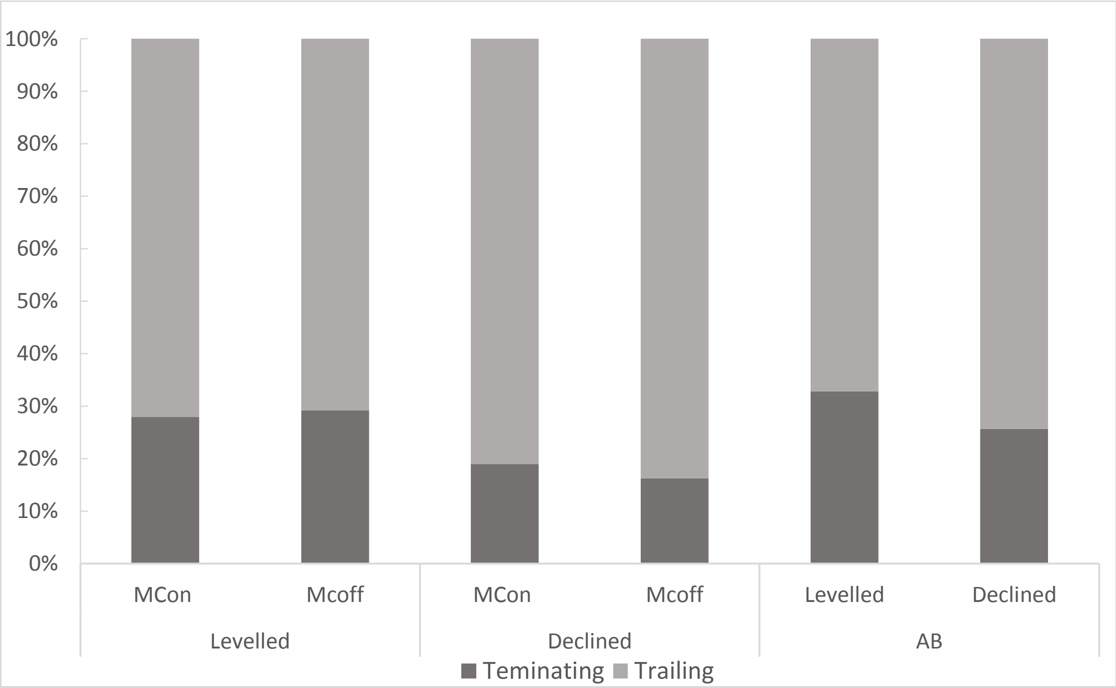


Figure 29. TFA group limb work as a percentage of the total negative mechanical work done by both limbs (%) during gait terminations on level and declined surface for the terminating/prosthetic limb (dark grey shading) and the trailing/intact limb (light grey shading). AB data from chapter 4 are shown for comparison.

Table 6. Negative mechanical limb work done by terminating (prosthetic) limb and trailing (intact) limb, to terminate gait on levelled and declined surfaces at MC on and off.

	Levelled		*Declined*		*Levelled*		*Declined*	
	Prosthetic side (J/kg)				Intact side (J/kg)			
Participant	On	Off	On	Off	On	Off	On	Off
TF02	-0.200	-0.212	-0.154	-0.154	-0.503	-0.541	-0.652	-0.719
TF03	-0.170	-0.154	-0.122	-0.097	-0.475	-0.487	-0.642	-0.707
TF04	-0.164	-0.152	-0.115	-0.094	-0.472	-0.375	-0.725	-0.811
TF05	-0.144	-0.112	-0.125	-0.098	-0.385	-0.282	-0.365	-0.392
TF06	-0.105	-0.113	-0.058	-0.048	-0.287	-0.257	-0.441	-0.433
TF07	-0.166	-0.196	-0.086	-0.077	-0.449	-0.491	-0.579	-0.512
TF08	-0.212	-0.228	-0.283	-0.240	-0.412	-0.399	-0.549	-0.555
Mean	-0.166	-0.167	-0.135	-0.116	-0.426	-0.405	-0.565	-0.590

6.4 Discussion

6.4.1 Effect of MC

As hypothesised, the limb system prosthesis' ramp descent mode lead to a significant increase the negative mechanical work (power absorption) performed by the prosthetic limb when TFA terminated gait on the ramp. Importantly, the increase in work done when the MC was active occurred only on declined, i.e. there was no difference between MC conditions on level surface. The significantly increased negative limb work on declined for the MCon compared to MCoff condition, suggests that when the MC was active the limb system resistances were altered in the manner congruent with the device's 'ramp descent' mode. Analysis of both foot-flat attainment and CoP trajectory highlights that the MC changed resistances as predicted. MC active condition might influence the separation between CoM and CoP in which deceleration forces are normally generated (van Keeken et al. 2013). Accordingly, increased dynamic stability enhanced TFA confidence and reliance (Hof et al. 2005; Prinsen et al. 2017) which helped in more horizontal work in the prosthetic side to perform halting CoM velocity.

The time from foot-flat to contralateral foot contact was longer for the MCon compared to MCoff condition. This finding in particular indicates that forwards shank rotation over the foot during single-limb support must have been slowed when the MC was active. One would expect that shank rotation attributed only to the hydraulic ankle since there is no knee flexion. Although, this is not certain since this study did not directly analyse planter flexion (i.e. foot angular velocity), attaining foot-flat was quicker and CoP forward velocity being increased when MCon suggesting a major contribution of limb system negative work came from hydraulic ankle rather than the knee.

All above findings bring to light the following points; by having a prosthesis that automatically reduces ankle resistance at initial contact to expedite attainment of foot-flat, consequently facilitate smooth progression of CoP anteriorly (i.e. CoP was faster and more anterior when MCon) for more braking control. Then increases ankle resistance to slow how quickly the shank (and thus CoM) rotates forward over the prosthetic foot, participants were able to do more mechanical limb work with their prosthesis which might improve confidence and user satisfaction.

It is worth mentioning that results indicate no effect of limb system MC on level surface, i.e. there was no difference in limb negative work between MC conditions on level surface. This means limb system worked as a conventional hydraulic prosthesis when Mcoff with a pre-set single mode at the knee and ankle. This finding lends support to a study showed that there was no influence of MC condition of the Rheo Knee on intact limb contribution to the total deceleration impulse (combined deceleration impulse of the terminating and trailing leg) when TFA terminating gait on the level surface (Prinsen et al. 2017). Previous studies found that when completing gait termination on level surface using the prosthetic limb, the mechanism of placing the CoP in front of the CoM by which braking forces are produced is impaired (Vrieling et al. 2008a; Prinsen et al. 2017). Further, the MC involvement did not improve it ($p=0.11$) (Prinsen et al. 2017). As highlighted in the reviewed literature, MC control is anecdotally reported as more advantageous in gait activities other than level walking. A general improvement in stair descent, ramp or hill descent, and walking on uneven terrain are observed by the amputees after switching to using a MC control knee prosthesis (Sawers and Hafner 2013).

The second hypothesis was that limb system prosthesis increased negative mechanical work could influence the work done by the contralateral trailing intact limb when the device's ramp descent mode is activated compared to inactive state. Results for each participant showed that 5 out of 7 participants had a decrease in intact limb negative work when MC was active. Interestingly, in absolute terms, results showed that there was insignificant effect of MC conditions on intact side. Having said that, in percentage terms there was convincing evidence that the intact side had/ illustrated a significant decrease in its contribution to total limb work done by both limbs on CoM during gait termination on declined surface. The intact limb had to do more negative work as a compensation when the MC was inactive (Figure 29). MC inactive condition means there was no control of MC on prosthetic ankle plantarflexion nor shank forward rotation (only the passive hydraulic control) thus, the prosthetic knee perhaps more vulnerable to flexion hence less stable. Accordingly, amputees compensate on the intact side during the penultimate step by increasing intact limb negative work.

6.4.2 Comparison between limbs (limb effect)

Irrespective of surface and MC effect, the present study findings indicate that, the prosthetic limb's contribution to the total negative work done by both limbs to halt gait was much smaller than the intact limb's contribution. However, such a difference in work done by the terminating and trailing limbs was expected from previous studies on amputees' gait termination due to the reduced role in generating deceleration forces that are needed to terminate gait.(Vrieling et al. 2009) and analogous to previous findings in chapter 4

On the level, Prinsen study showed that the intact limb contribution was about 87% of total from both limbs deceleration forces (Prinsen et al. 2017). Such asymmetry/ imbalance in external work performed done by limbs on CoM is also approved in the current study. A prosthetic limb generates less positive trailing work was suggested to be due to ankle weakness (limitation compared to the physiological one). Thus, the contralateral (intact) limb will absorb greater negative terminating work (Houdijk et al. 2009). More negative limb work at the terminating intact limb was suggested to be a compensation for affected limb weakness (Doets et al. 2009; Houdijk et al. 2009). Such weakness involves stiffness of the prosthetic ankle inhibits a smooth anterior displacement of the CoP and can be one of the factors influencing the lack of contribution from the affected limb. Another drawback/ weakness at the prosthetic limb that increases intact limb compensation is the absence of prosthetic knee flexion during the loading response which prevents a posterior positioning of the CoM relative to the CoP (Vrieling et al. 2008a). The duration of single limb support on the prosthetic leg was thought to be decreased because stability is challenged during this phase. Accordingly, persons with an amputation heavily rely on their intact leg for the absorption of energy during gait termination.

This thesis investigated how use of a recently developed microprocessor-controlled Limb System affected termination of ramp descent with the prosthetic limb as the terminating limb. Future work should investigate how use of the Limb System prosthesis affects ramp descent termination when the terminating limb is the intact limb. When the MC is inactive the knee resistance will be reduced to the lowest pre-swing level and simultaneously ankle dorsiflexion resistance would be normal (default level) thus the control on shank forward rotation would be minimal and therefore the prosthetic limb would be collapsing/yielding during the transfer of body CoM towards the terminating intact side, i.e. amputee will ‘fall’ forwards over prosthetic limb. When the MC is active, a slower and more controlled placement of the intact limb on the lower part of ramp surface would occur due to the ‘braking effect’ provided by the intermediate knee resistance compared to the usual pre-swing lower resistance, and the increased dorsiflexion resistance at the ankle. This suggests that stopping (terminating descent) on the intact limb might be easier (more controlled) when the MC is active, because rather than the limb having to do work to control the ‘falling’ body CoM it could instead concentrate on doing work to halt the CoM.

6.4.3 Surface effect

Despite the significance of MC control involvement, prosthetic limb negative mechanical work (Figure 28) was decreased for declined compared to level trials, unlike the intact limb which showed an increased negative work performed on CoM. This suggests a surface effect on gait termination's limb negative work in a manner that the intact limb did more work as a trailing limb similar to that of AB (chapter 4). The decrease in negative work done by the prosthetic side on declined versus level surface also indicated an inability of the prosthetic limb to absorb power required to lower CoM down the ramp. This might be due to limited knee flexion compared to intact side specifically during early stance. Lack in prosthetic limb negative work on declined surface might be related to the fact that the prosthetic limb was the terminating limb in gait termination representing the behaviour of that in AB (chapter 4).

The increased intact limb negative work (9% and 13% more than that on level surface at MCon and MCoff respectively) (Figure 29) was related to the increased power absorption during gait termination on the declined surface. Amputees rely more on their intact side (7% more compared to AB on declined) to primarily halt CoM velocity and secondarily lower it down the ramp. The increased reliance on the intact limb compared to prosthetic limb is a common TFA adaptation, and such adaptation would be exacerbated on declined surface compared to level (Vrieling et al. 2008a).

6.4.4 Amputees versus able-body individuals

During the early part of the braking phase (Figure 27), surface effect on trailing limb negative work was more noticeable for AB compared to TFA group. Negative power done by the trailing limb altered during double support phase (i.e. during the period where transition from one limb to another occurred) (Franz et al. 2012). This considerable increase in trailing limb negative power was for down-slope compared to level-ground walking for AB (Franz et al. 2012). Whereas in the current study, power profile for AB presented same trend (Figure 27b) but not amputees; during double support phase when body CoM being transferred from the prosthetic limb to the intact limb was similar on both surfaces (Figure 27a).

As mentioned previously, intact trailing limb negative work (Figure 28) was greater compared to that of the trailing limb of AB when terminating gait on declined surface suggesting a compensation strategy of using intact limb deceleration force to halt CoM velocity and lower its vertical displacement prior to the last step performed by the prosthetic limb. This is clearly related to the power profile of the intact limb during late stance i.e. higher magnitude and longer duration (Figure 27). In chapter 4 a greater negative power of the trailing limb on declined during late stance was associated with the knee joint which showed the greater influence by the surface.

During level walking, TFA trailing limb increased power compared to that of AB group during late stance was related to the increased negative work at the knee joint during k3 phase and corresponded to eccentric activity in the rectus femoris (Nolan and Lees 2000; Kuster 1995; Chapman and Fraser 2008). During the time period between intact limb foot-strike to prosthetic limb foot-off, TFA did not absorb greater negative work at the intact limb compared to AB group (Mahon et al. 2017). The studies done by Nolan and Lees 2000 and Mohan et al 2017 were related to walking not gait termination however this comparison is still valid since in the current study intact limb went through/performed a full gait cycle during two steps of gait termination.

In chapter 4 findings showed that irrespective of surface, AB terminate gait via negative ankle joint work as the foremost contributor to the negative mechanical limb work done. Considering the surface effect, terminating gait on a declined surface was with greater negative mechanical limb work and was associated with the increased negative knee joint work in early stance as the key contributor to the increased mechanical limb work. As above-the-knee prostheses have limited ability to have any early stance knee flexion, TFA cannot have such knee joint contribution during down-slope walking. This highlights that the limb system’s ability to increase the resistance at its ankle mechanism (i.e. absorb more power) was the key factor in why the prosthetic limb was able to do more negative mechanical limb work when it was in the ramp descent mode.

This study was limited by a number of factors such as the sample size of only active TFA (K3 activity level) was presented. However, the number of study's participants was 7 out of 11 TFA participants available in UK using limb system. Determining whether TFA at different activity level can gain functional benefits from using a limb system prosthesis similar to those highlighted in the present study can be investigated through upcoming research. Contrary to study's recommendation that amputees should use the intact side to terminate gait taking advantage of intact limb deceleration GRFs (Vrieling et al. 2008a), the prosthetic side was chosen to lead termination as an attempt to challenge the prosthetic limb and investigate its pure potential/contribution as a terminating limb required to do the foremost contribution in generating braking force and arresting body CoM (Bishop et al. 2002). Terminating with the intact limb was not investigated since including such number of trials increase the likelihood of participants becoming fatigued. Future work is thus required to determine how use of the limb system prosthesis effects gait termination when the terminating limb is the intact limb.

6.5 Conclusion

The use of simultaneous MC control of the hydraulic resistances at the knee and ankle leads to considerably more negative work done by the prosthetic limb. Importantly this increase in prosthetic limb work was only observed on declined surface, i.e. there was no effect of the MC when terminating gait on the level. With this in mind, it can be concluded that 'ramp descent mode' was controlled by the involvement of MC and it is indeed responsive to surface terrains. Which in turn makes difference in terms of improving the mechanical work relative contribution done to arrest CoM during gait termination of TFA. In this chapter, the discussion centres on ramp descent mode efficacy compared to using an above-the-knee prosthesis incorporating knee and ankle devices that had pre-set / unchanging hydraulic resistances. It has been proved that gait termination of TFA on declined surface was influenced by ramp descent mode. The chapter that follows moves on to consider the effect of speed on MC efficacy and ramp descent mode in addition to examine whether these effects will be clinically meaningful using effect size metrics.

Chapter Seven

Gait Termination on a Declined Surface in Trans-Femoral Amputees: Impact of Using Limb System Prosthesis across Different Walking Speeds

7.1 Introduction

Ramp descent mode of limb system prosthesis is designed (according to manufacturer) to accommodate not only different terrains but also users' self-selected speeds. Findings of the previous chapter highlight that when the limb system's ramp-descent mode was active, the prosthetic limb's involvement in terminating gait when descending a ramp at customary speed was enhanced. This thesis investigates the efficacy MC control prosthesis when TFAs terminate gait on declined surface particularly the worthiness of 'ramp descent mode'. It is uncertain if prosthetic limb can have contribution at slow speed and demonstrate MC efficacy. Therefore, as part of this intention the present chapter is dedicated to investigating whether this mode would be also responsive to anecdotally evidenced more challenging slow walking speed. Previous research showed that with decreased walking speed, high differences/imbalance were shown in the contact times of the prosthetic limb and the intact limb (Jaegers et al. 1995; Nolan et al. 2003; Schaarschmidt et al. 2012). Amputees spend more time on the intact limb to transfer CoM towards the prosthetic side and such time was even more increased at slow speed. Accordingly, descending at speed lower than the preferred walking speed, seems to be more difficult to handle for the amputees, likely due to difficulty controlling (slowing) how the CoM progresses (falls forward) over the prosthetic foot during single-support when traditional type prosthetic feet are used. When using traditional prosthetic feet there is a tendency for the heel-keel to recoil following weight acceptance (from its deflected position that resulted in simulated plantarflexion). This recoil means TFAs have a tendency to lurch forwards over the prosthetic foot which can result in the feeling of 'falling' forwards and downwards when descending ramps. To compensate for such CoM forwards 'lurching' the intact trailing limb likely has to perform compensatory negative mechanical work to control CoM forward/downward velocity especially with the lack in prosthetic knee flexion as the chapter 6 highlighted. The purpose of the present chapter was to determine how the use of limb system prosthesis affected the external negative mechanical work done by the prosthetic and intact limbs when TFAs terminated gait during ramp descent at slow compared to customary speed.

7.2 Methods

7.2.1 Participants

Eight TFAs completed planned gait terminations (stopping on prosthesis) on a ramp from slow and customary walking speeds, with the limb's MC being active or inactive. Demographic details of the eight male TFA who participated in the study are presented in Table 3. (Further details can be found in methods chapter section 3.2.2 and 3.2.2.1).

7.2.2 Data collection and processing

Participants were asked to complete repeated trials involving walking at customary and slow speeds down a 4 m long 5-degree declined walkway. Terminations were completed from both customary and slow walking speeds with the terminating limb always being the prosthetic limb. Trials at each walking speed were repeated 10 times with a block of 5 trials completed with the MC active (MCon) and 5 trials with it inactive (MCoff). When inactive (MCoff) the knee and ankle devices provided hydraulic resistances at default levels. Default levels are set to provide the optimal function for level gait at the user's customary walking speed. The order in which walking speed (slow, customary) and MC condition (MCon/MCoff) were completed, were counterbalanced across participants.

7.2.3 Data analysis

Walking speed was determined as the peak CoM forward velocity during intact limb foot contact (i.e. at the start of the two-step gait termination). Time of stopping was determined as the time-period between intact limb foot contact (penultimate step) up to instant of final bipedal standing stance.

In addition to the variables assessed in the previous chapter that confirmed the hydraulic resistances at the ankle changed in the way that was predicted by the limb system's 'ramp descent' mode, in this chapter the time from prosthetic limb foot-contact up to intact limb foot-off (i.e. weight transfer time, WT_{time}), and time of prosthetic limb single-support (SS_{time}) were also determined. The reason that if the plantarflexion resistance was reduced to facilitate attaining foot-flat, then WT_{time} would be reduced. Similarly, if dorsiflexion resistance was increased after attaining foot-flat, then SS_{time} would be increased.

7.2.4 Statistical analysis

Limb work done, walking speed, time of gait termination, time to foot-flat, time from foot-flat to bipedial standing stance and finally CoP velocity in A/P direction after prosthetic intial contact were analysed using repeated measures analysis of variance (ANOVA). Post-hoc analyses were undertaken using Tukey HSD tests. Statistical analyses were performed using Statistica (StatSoft, Inc., Tulsa, OK, USA). Level of significance was set at $p < 0.05$. The following factors, and interaction between these factors, were tested: 1) Speed (i.e. walking speed prior to gait termination): two levels, slow and customary; 2) MC: two levels, inactive (MCoff), active (MCon).

Diferrences in outcome variables between limbs and between MC conditions were additionally evaluated using effect size (ES) to find out whether these diferences would be clinically meaningful. Clinical significance was evaluated using the effect size metric, which is a useful way of quantifying the impact of interventions, since "statistical" significance may not be sufficient to the task of serving as the sole criterion for evaluating result import (Thompson 2002). ES of 0.1-0.29 where considered small; those between 0.3-0.59 where considered moderate; and those 0.6 and above where considered as large (Cohen 1988).

7.3 Results

Group ensemble mean power profiles (W/kg) of the terminating- and trailing-limbs across the repeated gait terminations for the MCon and MCoff conditions at customary and slow speeds are presented in Figure 30. The amount of negative work (W(-ve)) done by the prosthetic (terminating) and intact (trailing) limbs to terminate gait from slow and customary walking speeds for the MCon and MCoff conditions is presented in Figure 31. Group average limb work as a percentage of total negative mechanical work (%) done on CoM during gait terminations from slow and customary speeds for the terminating (prosthetic) limb and the trailing (intact) limbs is presented in Figure 32.

The slow and customary freely chosen walking speeds were significantly different ($p<0.001$), but there was no significant difference in walking speeds ($p>0.31$) between MC conditions. Group average chosen slow walking speed was 0.92 (0.15) and 0.90 (0.13); ($p>0.31$) and the chosen customary speed was 1.20 (0.21) and 1.21 (0.22) m/s; ($p>0.80$), for MCon and MCoff conditions respectivly. Time of stopping was significantly affected by speed ($p<0.001$) and MC condition ($p=0.021$) but there was no interaction between terms ($p=0.21$). Time of stopping was shorter for customary compared to slow speed trials and was shorter when the MP was active compared inactive (slow speed, 1.30 (0.22) and 1.34 (0.23) sec, and customary speed, 1.07 (0.13) and 1.08 (0.12) sec, for MCon versus MCoff condition respectively). Braking phase duration was significantly effected by walking speed ($p<0.001$) but unaffected by MP condition ($p=0.095$) and there was no interaction between terms ($p=0.37$). Braking phase duration was shorter (~12%) for customary compared to slow speed trials.

7.3.1 Prosthetic limb

Negative limb work done was significantly affected by speed ($p=0.016$) and mircoprocessor condition ($p=0.005$) but there was no interaction between terms ($p=0.817$). Negative work done was greater when terminating gait from walking at customary compared to slow speed, and was greater for the MCon compared to MCoff condition (Figure 31). On average 14% and 16% more negative work was done by the prosthetic limb for MCon compared to MCoff condition at slow and customary speeds respectively, with 6 out of 8 and 7 out of 8 participants for slow and customary speeds respectively showing such an increase in negative work when the MC was active.

It is worthy of note that the mechanical limb power profiles of the terminating/prosthetic limb were smilar in shape and size for both speeds

(Figure 30a). WT_{time} was significantly affected by walking speed ($p<0.005$) and by MP condition ($p=0.05$) but there was no interaction between terms ($p=0.51$). WT_{time} was shorter for customary compared to slow speed trials and was shorter when the MC was active compared to inactive (slow speed, 0.235 (0.09) and 0.248 (0.10) sec, and customary speed, 0.184 (0.06) and 0.192 (0.06) sec, for MCon versus MCoff condition respectively). SS_{time} was significantly affected by walking speed ($p<0.001$) but unaffected by MC condition ($p=0.47$) and there was no interaction between terms ($p=0.50$). SS_{time} was shorter for customary (0.304 sec) compared to slow (0.326 sec) speed trials.

When terminating gait from walking at customary speed, there was an increase in negative mechanical work done by the prosthetic limb, for the MCon compared to MCoff condition, and as a result the negative work done by the intact limb decreased slightly (group mean effect size change for MCon compared to MCoff was; 0.60, -0.48 for prosthetic and intact limbs respectively, Table.7). The sizes of effect suggest MCon induced a moderate to high practical difference in the amount of work done by each limb. Congruently, on average 16% more negative work was done by the prosthetic limb, and importantly 7 out of 8 participants showed an increase in negative work done on the prosthetic limb, for MCon compared to MCoff condition (Table 7a). The negative work done by the intact limb decreased in 6 out of 8 participants, however, the decrease in work for when the MC was active compared to inactive was only 2.9% on average (Table 7b). At customary speed, the contribution of negative work done by each limb to the total negative work done by both limbs in halting CoM velocity (terminating gait), showed the prosthetic limb's contribution to the overall work done increased from 17.2% to 19.7% for the MCon compared to MCoff condition. As the intact limb's contribution to the overall work done would have decreased by the same amount (2.5%), this indicates that the contribution of the prosthetic limb relative to the intact limb increased by 5% when the MC was active.

When terminating gait from walking at slow speed, there was a small increase in negative work done by the prosthetic limb for the MCon compared to MCoff condition (effect size; 0.35, Table 7a) but there was not a corresponding reduction in negative work done by the intact limb (Table 7b). Congruently, 14% more negative work was done by the prosthetic limb, and 6 out of 8 participants showed an increase in negative work done on the prosthetic limb, for the MCon compared to MCoff condition (Table 7a).

The contribution of negative work done by each limb to the total negative limb work done by both limbs to terminate gait showed the prosthetic limb's contribution to the overall work done increased from 22.0% to 24.2% for the MCon compared to MCoff condition (an increase of 2.2.%). This meant that the contribution of the prosthetic limb relative to the intact limb increased by 4.4% when the MC was active.

7.3.2 Intact limb

Negative mechanical work done was significantly affected by speed (p=0.004) but was unaffected by the MC condition (p=0.38) and there was no interaction between terms (p=0.30). Negative work done was greater when terminating gait from walking at customary compared to slow speed (Figure 31).

It is noteworthy that the power profile of the trailing/intact limb, highlights the negative power throughout the braking phase was greater (magnitude and period) for customary speed than that at slow speed (Figure 30b). Furthermore, except for a brief period, trailing limb power was negative for almost the entire braking phase: with the positive period being noticeably small (magnitude and duration) for both customary and slow speed.

7.3.3 Comparison to able body individuals

When able bodied individuals halt gait (from a customary walking speed of 1.15 (0.18 m/s), and slow speed of 0.86 (0.21m/s); chapter 5), the terminating limb contributed 25.6% and 30.0%, at customary and slow speeds respectively (Figure 32). In comparison, for the amputee participants in the present study the terminating (prosthetic) limb's contribution to the total negative work done by both limbs when terminating gait from customary speed (1.2 m/s), was 17.2% and 19.7% for when the MC was inactive and active respectively and for slow speed (0.9 m/s) it was 24.0% and 22.2% for when the MC was inactive and active respectively (Figure 32).

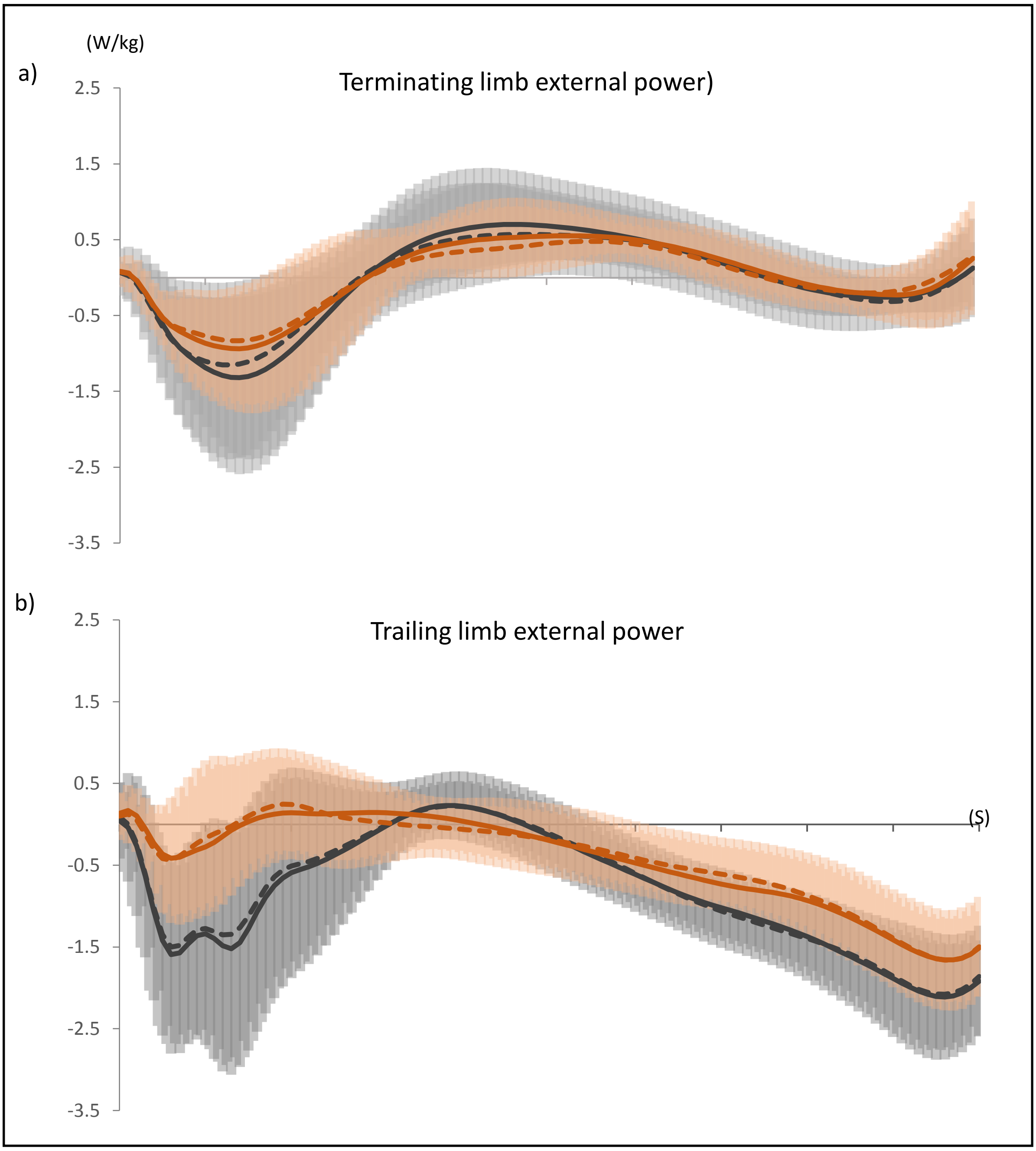


Figure 30. TFA group ensemble mean (+/-SD) limb power profiles (W/kg) during gait terminations at customary and slow speed for a) the terminating (prosthetic) limb and b) the trailing (intact) limb. Bold line= MCon; dashed line= MCoff. Grey band = group SD for customary speed, orange band= group SD for slow speed. Both speeds' data are plotted for the braking phase of each limb.

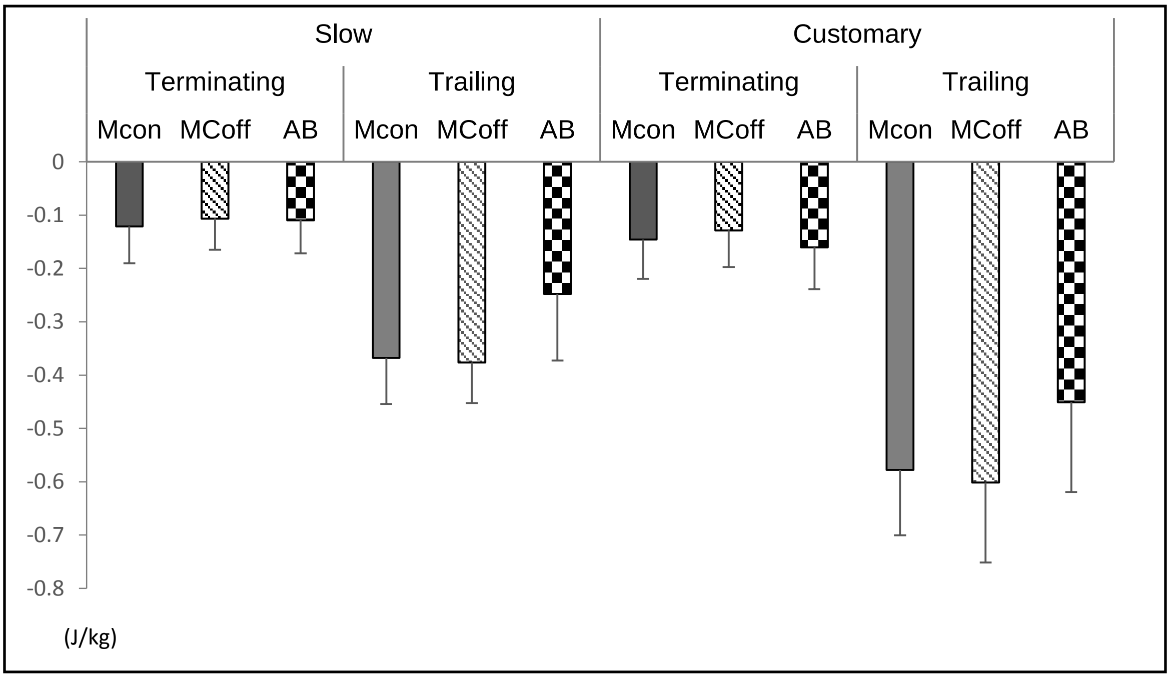


Figure 31. TFA group mean (+SD) limb negative mechanical work done (J/kg) during gait terminations at customary and slow speed for the terminating (prosthetic) limb and the trailing (intact) limbs. AB data, from chapter 5, are shown for comparison.

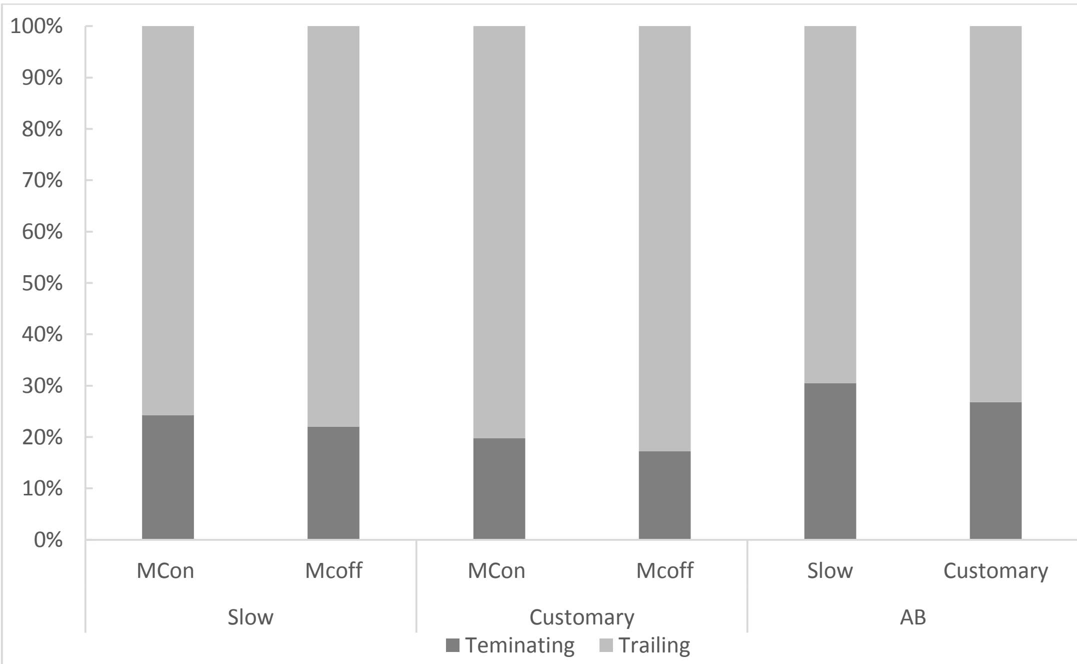


Figure 32. TFA group limb work as a percentage of the total negative mechanical work done (%) during gait terminations on declined surface from slow and customary speed for the terminating (prosthetic) limb (dark grey shading) and the trailing (intact) limb (light grey shading). AB data from chapter 5 are shown for comparison.

Table 7. Negative mechanical limb work done by A) terminating (prosthetic) limb and B) trailing (intact) limb, to terminate gait from walking at slow and customary speeds. ‘on v off’ = percentage difference of ‘on’ condition relative to ‘off’, ES= effect size.

A)	*Slow*				*Customary*			
	Prosthetic side (J/kg)		on v off (%)	ES	Prosthetic side (J/kg)		on v off (%)	ES
Participant	On	Off			On	Off		
TF01	-0.246	-0.182	35.581	0.770	-0.224	-0.219	2.294	0.114
TF02	-0.109	-0.099	10.101	0.437	-0.154	-0.154	-0.242	-0.020
TF03	-0.098	-0.088	11.305	0.332	-0.122	-0.097	25.446	0.793
TF04	-0.080	-0.055	46.044	1.376	-0.115	-0.094	23.317	0.684
TF05	-0.104	-0.088	17.878	0.683	-0.125	-0.098	28.181	1.256
TF06	-0.039	-0.048	-17.571	-0.865	-0.058	-0.048	19.598	1.234
TF07	-0.087	-0.082	6.551	0.226	-0.086	-0.077	11.231	0.163
TF08	-0.207	-0.210	-1.660	-0.157	-0.283	-0.240	17.773	0.624
Mean	-0.121	-0.106	13.529	0.350	-0.146	-0.128	15.950	0.606

B)	*Slow*				*Customary*			
	Intact side (J/kg)		on v off (%)	ES	Intact side (J/kg)		on v off (%)	ES
Participant	On	Off			On	Off		
TF01	-0.369	-0.365	0.950	0.102	-0.672	-0.681	-1.287	-0.184
TF02	-0.303	-0.313	-3.256	-0.290	-0.652	-0.719	-9.314	-1.332
TF03	-0.303	-0.318	-4.912	-0.443	-0.642	-0.707	-9.180	-1.374
TF04	-0.465	-0.515	-9.663	-1.180	-0.725	-0.811	-10.642	-1.204
TF05	-0.326	-0.414	-21.078	-1.806	-0.365	-0.392	-6.848	-0.661
TF06	-0.324	-0.332	-2.290	-0.155	-0.441	-0.433	1.989	0.413
TF07	-0.535	-0.449	19.110	1.179	-0.579	-0.512	13.124	0.671
TF08	-0.320	-0.303	5.793	0.466	-0.549	-0.555	-1.042	-0.194
Mean	-0.368	-0.376	-1.918	-	-0.578	-	-2.900	-

	0.266	0.601	0.483

7.4 Discussion

The aim of the current study was to determine how use of an above knee limb system with simultaneous MC affected the external negative mechanical work done by the prosthetic and intact limbs when TFA terminated gait on a declined surface from slow and customary speed. The study focused on gait terminations where the terminating limb was the prosthetic limb. The major findings of this chapter were that the negative mechanical work done by the prosthetic limb was significantly increased when the limb system's MC was active and when terminated gait from customary speed.

The analysis for the parameters of foot-flat and CoM dynamics in the previous chapter showed that ramp descent mode was indeed affecting the limb work done on CoM on declined. In this chapter, WT_{time} was significantly shorter with MC involvement implying that foot-flat was attained easier/quicker when the MC was active. There was no significant difference across MC conditions in the subsequent single-support period. However, braking time (equivalent to WT_{time} plus SS_{time} combined) did not differ across MC conditions, suggests that single support must have been longer when the MC was active even though it wasn't significantly so (given that WT_{time} was shorter when the MC was active). A prolonged single support means that there was no tendency to lean forwards over the foot. These findings emphasise that when the MC was active the hydraulic resistance at the ankle was adapted (impacting the performed limb negative work) consistently to terrain as well as speed by means of the device's 'ramp descent' mode: plantarflexion resistance reduced at initial contact to facilitate attaining foot-flat; dorsiflexion resistance then increased to control/slow forward progression of shank pylon over foot. The accompanying increased negative work done when MC was active suggests participants became more assured in using their prosthetic limb to arrest CoM velocity.

WT_{time} (which represents the double support phase) was significantly affected by walking speed ($p<0.005$) and by MP condition ($p=0.05$). Unsurprisingly, WT_{time} was shorter for customary compared to slow speed trials and was shorter when the MC was active compared to inactive. At slow speed when MC was inactive, the period of double support (WT_{time}) was extended compared to customary speed perhaps as an attempt to carefully control cautious loading of the prosthetic limb being shifted from the intact limb. Amputees extend the duration of the load transfer onto the prosthetic limb to shorten single support demonstrating the increased effort of the amputees to stay on their intact leg as long as possible (Schaarschmidt et al. 2012). The current study showed that the duration of the load transfer (WT_{time}) was shortened with MCon indicating increased reliance on the prosthetic limb.

Irrespective of walking speed, the negative work contribution of the intact trailing limb to halt gait was greater than that of the prosthetic leading limb. Such an imbalance in work done by the leading and trailing limbs is comparable to how AB terminate gait at slow and customary speed (chapter 5). During level walking external negative work must be performed on the CoM at the time of weight transfer from prosthetic to intact side mid-stance (Kuo 2007). Additionally, more negative work is performed than positive at the prosthetic-to-intact transition (Houdijk et al. 2009). During walking down-slope, double support negative work of the trailing limb (during CoM transition from prosthetic towards the intact) proved to be significantly increased with speed (Franz and Kram 2012). Double support phase (contact times) was found to be more adapted to speed than single support of the prosthetic side during walking (Schaarschmidt et al. 2012). In the current study, during the transition from the prosthetic limb towards the intact which represents the former part of the braking phase of the intact limb power profile (Figure 30b) the negative power during braking phase was greater (magnitude and period) at customary speeds than that at slow highlights the importance of this time period in directing and decelerating body CoM.

When the MC was active, the negative work done by the prosthetic limb as a percentage of the total negative work done by both limbs showed a 5% increase relative to the intact limb, when terminating gait from customary speed walking, and for slow speed trials the increase was 4.4%. The corresponding effect sizes (ES=0.606 and 0.350 respectively) indicate these percentage increases were meaningful. These findings highlight that by having a prosthesis that automatically reduces ankle resistance at initial contact to expedite attainment of foot-flat, and then increases ankle resistance to slow how quickly the shank (and thus CoM) rotates forward over the prosthetic foot, MC had clinical benefits enable TFA to do more mechanical limb work with their prosthesis.

Potential limitations of this study include the following. Three of the participants did not habitually use a limb system prosthesis and were only given a relatively short familiarisation period (~20 mins) to become accustomed to it prior to using it within the study. A lack of familiarisation may have affected how these participants used the limb system prosthesis. However, these participants habitually used a prosthesis with a MC knee in combination with a hydraulic ankle-foot device (either with or without MC control): thus, they were experienced users of a MC prosthesis. Furthermore, to guide against any learning and/or fatigue effects the order of which of the MC conditions (MCon, MCoff) was completed first was counterbalanced across participants. Therefore, participants' lack of habituation with the limb system did not affect the results presented.

7.5 Conclusions

In conclusion, compared to using an above-the-knee prosthesis incorporating knee and ankle devices that had pre-set/unchanging hydraulic resistances, the use of limb system with simultaneous MC control, leads to significantly more negative mechanical work being done on CoM not only as a response to terrains as per the previous chapter but also to users freely selected speed. These findings suggest that use of a limb system prosthesis will improve the way TFA descent and stop on ramps, and more generally that use of a limb system prosthesis should provide TFA clinically meaningful benefits to their everyday walking where adaptations to an endlessly changing environment are required. Examining the effect of speed on the external mechanical work during different gait tasks and compared the MC 'behavior' (external power and work pattern) when MC active versus inactive will provide advance knowledge about MC contribution/potentials in TFA ambulation.

Chapter Eight

General Discussion, Research Limitations, Conclusions and Final Remarks

8.1 Introduction

Research conducted in this thesis progressively developed a view that using MC limb system can positively impact gait termination locomotive task in TFAs compared to performing this task with conventional knee and ankle prosthetic devices. The first two chapters presented findings highlighting the gait termination strategies employed by AB, and this provided a normative data base to which the proceeding two chapters on TFAs could be compared. Findings for each study undertaken were discussed in depth at the end of the respective chapters. The current chapter gives a summary of the main findings from each experimental chapter and then provides a general discussion of the key findings of the thesis. This is followed by a summary of limitations in the research undertaken, recommendations for future work and finally the research main remarks are identified.

Summary of key findings from the experimental chapters:

(1) When AB terminate gait on level or declined surfaces the external negative limb work performed is mainly done by the trailing limb, with the associated negative joint muscle work performed mainly at the ankle, knee, and to a lesser extent at the hip. An increase in knee negative work, joint muscle work was the main adaptation to the surface angle changing from the level to declined surface.

(2) The ratio of trailing limb and terminating limb negative work performed on CoM to terminate gait in AB was unaffected by changes in walking speed.

(3) TFAs gait termination demonstrated a similar trend found in AB studies regarding limb work contribution in arresting body CoM at different surfaces. However, a significant increase in prosthetic limb work contribution was observed due to the involvement of MC and ‘ramp descent mode’ on declined surfaces.

(4) In comparison to an above-the-knee prosthesis incorporating knee and ankle devices that had pre-set/unchanging hydraulic resistances, ‘ramp descent mode’ indicated a contribution in the performed limb negative work.

This contribution presented at customary as well as slow speed and was not only statistically significant but also clinically meaningful as the analysis of effect size suggested.

8.2 General discussion

Planned gait termination task (halting of CoM velocity) involved negative work being performed mainly by the trailing limb with lesser contribution from the terminating/leading limb. This trend was observed for both AB and amputees but in slightly deferent ratio; on level surface, 71% and 67% was the trailing limb percentage of total work done by both limbs on CoM for TFA and AB respectively, and 82% and 74% was the trailing limb percentage of total work done by both limb on CoM for TFA and AB respectively on declined surface. Therefore, 4% and 8% more negative work performed by the trailing limb than that of AB on level surface and declined respectively. Thus, slightly more reliance on the trailing (intact side) to perform negative work was observed for TFAs than AB, which highlight TFAs adaptation strategy for stopping irrespective of surface angle. For both groups, trailing (intact) limb negative work was found to increase on declined surface compared to level ground (chapter 4 and 6) and increased with speed of walking (chapter 5 and 7). The most important findings were that use of MC limb system lead to considerably greater negative work being done by the prosthetic limb compared to limb system without the MC, and in turn reduced the negative work done by the intact limb. This increase in prosthetic limb negative work only occurred on declined surface: but this would be expected because the device's ramp-descent mode would have no impact on overground gait (and thus gait termination on a level surface)

Assessment of walking on declined surface has been widely used to evaluate the adaptations amputees use/require to walk down slopes/ hills (Vrieling et al. 2008b; Bellmann et al. 2010; Bellmann et al. 2012; Wolf et al. 2012; Highsmith et al. 2014; Lura et al. 2015; Struchkov and Buckley 2015; Villa et al. 2015; Alexander et al. 2017) but no previous research has investigated the adaptations amputees use to terminating gait when descending slopes. Descending slopes requires greater involvement of the lower-limb joints particularly the knee (i.e. increase in peak moments and muscle power) (Kuster et al. 1995). Transferring the body CoM from one limb to another requires an extended and stable limb, and involves the knee being kept relatively rigid via muscle action (Kuo 2007). For TFAs prosthetic knee contribution to gait termination task is limited because knee resistance is designed/programmed to be stiff during weight acceptance. Reducing the knee resistance to an intermediate level rather than overground pre-swing level, which helps provide braking effect during terminal stance, only occurs in terminal stance (see methods chapter). Given that, the prosthetic limb did more negative work when the limb system's MC was active compared to inactive on declined surface (chapter 6) and at both walking speed (chapter7). This highlights that the ability of the limb system to accommodate to changes in surface angle must have be due to ankle adapted resistance. When AB terminate gait on both surfaces (chapter 4), the increased terminating limb negative work was associated with an increase in negative limb work at the ankle joint. The ankle's role was associated with slowing rotation of the limb (and thus CoM) over the planted foot (i.e. controlling kind of inverted pendulum motion) during the final step of gait termination. In having the prosthetic limb as the terminating limb, there is no terminal stance for the prosthetic limb, stance yielding feature would likely not be activated for sure. Thus, lack of knee flexion shown in the results confirms that stance yielding feature was not involved and thus the increased work done by the prosthetic limb must have been due to the adaptations offered by the ankle mechanism.

In the literature, use of MC knees have been shown to lead to improvements in gait biomechanics which in turn increase the ability to perform a greater range of daily activities (Hafner et al. 2007). The most recent Clinical Commissioning Policy by NHS regarding the efficacy of MC prosthetic knees highlighted that there is strong evidence in relation to using a MC compared to a non MC knees (Clinical Commissioning Policy 2016). Similarly, the improved ankle function when using a prosthetic foot with hydraulically articulating versus rigid ankle has proved to enhance the biomechanics of gait. For instance, use of a hydraulic foot-ankle device has been found to reduce the joint moments at the hip of the prosthetic side in an individual with unilateral TFA (Alexander et al. 2017). An ankle mechanism combined with the lack of stance phase knee flexion proved to force the TFAs to spend relatively more time rotating their prosthetic legs forward until foot flat (prolonged double support) was achieved compared to the control group (McNealy and Gard 2008). In the current research, findings that highlight the decrease in double support phase (WT_{time}) with a prolonged single support when MC was involved. This proposed an easier/quicker foot flat achievement and a limited or controlled leaning forwards over the foot owing to slowed down shank rotation over foot thus improved CoM dynamic stability. The use of the MC hydraulic ankle-foot device can improve knee stability used to walk down-slopes because of a greater negative 'ankle' work done with the MC hydraulic compared to other conventional ankle types resulted in a corresponding reduction in flexion and negative work at the residual knee in TTA (Struchkov and Buckley 2015). In the present study foot flat and CoM dynamic control via shank rotation revealed that 'ramp descent mode' thus MC involvement via 'ramp descent mode' was the likely reason why more negative limb work was done when the MC was active. Consequently, the current findings are suggesting that limb system 'ramp descent' mode' enhanced TFA's ability to halt CoM velocity during gait termination on decline.

8.3 Research limitation

Amputee Participants were traumatic active healthy TFAs with K3 Medicare activity scores. Such amputees represent the high end, functionally, of lower limb amputees and make up approximately 3% of the total UK TFA population (i.e. 512 out of 2055; (Limbless Statistics 2011-2012). It is known that TFA who have lost their limb due to vascular insufficiency have a higher energy cost when walking compared to those TFA who have lost their limb due to traumatic causes (Detrembleur et al. 2005). Future studies should therefore investigate whether the benefits as a result of using a limb system prosthesis that were found in this thesis will also be seen in less active/vascular TFAs. Another limitation was the Limited time of accommodation given to some of the amputees (~20 mins) to get used to the limb system prosthesis, and the fact that trials with the limb system active or inactive were conducted within the same lab session. Five of the eight participants habitually used a limb system prosthesis and the other three were using a prosthesis with MC knee in combination with a hydraulic ankle-foot device (either with or without MC control): thus, all study participants were experienced users of MC prosthesis. Thus, it is unlikely that a lack of accommodation affected results. However, having amputees complete trials with the limb system active and inactive within the same lab session could have influenced the adaptations used by amputees because of learning effect. Although the order in which the ‘active’ and ‘inactive’ trails were conducted was block counterbalanced across participants, it cannot be ruled that such adaptations may have affected limb work findings thus future work should investigate MC conditions during separate sessions after a period of accommodation to each MC condition. It is worth bringing up that limb system has several other modes (e.g. ‘stair descent’, ‘stop and lock’), and thus future research is also required to determine whether it enhances the execution of other adaptive gait tasks.

This study did not assess TFAs’ gait termination from fast walking speed as per AB since it was difficult for them to stop on the prosthetic side from fast walking speed and bring the intact side next to it without needing an extra step outside the force platform which was irrelevant to the protocol used. Plus, adding more conditions to the protocol would likely influence TFAs fatigue. The key research question was evaluating the effectiveness of ‘ramp descent mode’, therefore it was important to have overground gait termination as normative data to be investigated rather than a third speed condition that TFA do not really use/perform down ramps. It is also important to note that when using a limb system prosthesis for everyday locomotion, the ‘ramp descent mode’ is automatically initiated after the initial two walking steps along a declined surface. However, in the present study, switching the ‘ramp descent’ mode on (and off) was done via Bluetooth connection. This allowed certainty that the device was in the correct mode (MCon, MCoff). Future work should thus assess when and if the ‘ramp descent’ mode becomes initiated when walking down-slope. Furthermore, gait termination on ramp angle (5 deg) was only evaluated. This angle was chosen since it was recommended as maximum angle for wheel-chair access ways (British Standards Institution 8300: 2009). Future research should examine the effect of different declined surfaces on this task along with the impact of ramp descent mode.

Thesis studies (and particularly chapter 4 and 5) only investigated gait terminations that were predictable in terms of where and when they occurred. The focus was on such gait terminations because most daily locomotor activity involves volition over where and when to stop, and thus it is important to understand how such terminations are achieved. Future work could determine the mechanical limb work involved in abrupt/unplanned gait terminations and determine whether the contributions from each limb are different to that for predictable/planned gait terminations. A related point to consider regarding the general protocol applied to assess planned gait termination in this research, is that one would expect more contribution from the terminating/prosthetic limb and perhaps increased difference between MC conditions for the abrupt stop task. Since during abrupt stop, greater GRFs would be generated as a response to increased forward momentum (Hase and Stein 1998; Bishop et al. 2002) in which TFAs would rely more on the leading prosthetic limb to stop such increased momentum.

The first experimental chapter examined how gait termination was performed by AB on declined surfaces versus level. This was done by means of the external limb work approach (Donelan et al. 2002a; Franz et al. 2012) along with traditional joint muscle power muscle (Quanbury et al. 1975) approach to examine how these biomechanical variables are altered as a result of a change in surface angle. Analysis of mechanical work performed on the body also involves the work performed to accelerate and decelerate the limbs (particularly the intact side and the residuum) relative to the CoM i.e. internal mechanical work was not assessed in this study. However, the external limb work approach quantified by the Individual Limb Method (ILM) (Cavagna and Margaria 1966) used in the current study was previously shown to be equivalent to the internal joint muscle work (Willems et al. 1995). The principal aim of this research was to evaluate the efficacy of MC involvement to gait termination task. This aim was achieved using the ILM (Donelan et al. 2002b; Franz et al. 2012) as a proposed ‘clinically friendly tool’ (Agrawal et al. 2013) for practicing evidence-based evaluation of TFA amputees gait kinetic and prosthetic functional differences (Detrembleur et al. 2005; Adamczyk 2008; Agrawal et al. 2009; Houdijk et al. 2009; Agrawal et al. 2013; Bonnet et al. 2014; Agrawal et al. 2015) as previously highlighted in the reviewed literature. Although external mechanical work can be considered as a global measure of muscle work for the limb's three joints (Cavagna et al. 1976; Willems et al. 1995; Kuo 2007; Houdijk et al. 2009), further research could be conducted regarding residual hip joint power and moments when using the limb system prosthesis to investigate any adaptations (e.g. in/decrease hip moments and powers) as a response to ramp descent mode and MC control.

8.4 Conclusion

This thesis contributes to an important body of knowledge and scientific research concerned with prosthesis selection and rehabilitation programs. This thesis has demonstrated that amputee and AB exhibit similar biomechanical characteristics (mechanical work ratios between trailing and terminating limb) to halt gait regardless of the surface or walking speed prior to stopping. Findings indicate clinically important biomechanical benefits that TFAs obtain from using the MC limb system prosthesis when terminating gait down-slope. The increased prosthetic side contribution in negative work done to halt CoM, reduced the time required to transfer CoM onto prosthetic limb during the two steps of gait termination. Controlling how quickly attaining foot flat was achieved, then carefully bringing body CoM onto the prosthetic limb via the controlled/pre-set/coordinated shank rotation was likely the key factor in increasing negative work contribution. Prostheses' functional performance during real world activities such as negotiating stairs/ramps and adapting to variable walking speeds and direction is essential to evaluate the relative value of prosthetic components. MC involvement seems to contribute more towards completing such tasks compared to mechanical conventional prostheses.

8.5 Final Remark

The goal of current prosthetic engineering technology is to develop user-adaptive artificial limbs that mimic the function of a physiologic limb and hence increase prosthetic limb usage which in turn should potentially reduce intact limb compensatory effort/loading. The current study's findings suggest that a limb system prosthesis represents another step towards such technological advancement. Undoubtedly, there was satisfactory evidence suggesting the increased efficacy of the MC limb system with ramp descent mode impacted amputees' performance to execute the gait termination task.

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Appendices

Appendix A. Participant Information Sheet

Study title:

Gait termination when walking down slopes in unilateral trans-femoral amputees; the effects of microprocessor-controlled compared to conventional prostheses.

You are being invited to take part in a research study. Before you decide it is important for you to understand why the research is being undertaken and what it will involve. Please take time to read the following information carefully. Ask if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

<u>Purpose of the Study?</u>

To determine the effects of a microprocessor-controlled prosthetic-knee/ankle device's 'ramp descent mode' on the biomechanics of gait termination.

<u>Why have I been chosen?</u>

You have been chosen because you are healthy trans-femoral amputee and do not have any gait impairments.

<u>Do I have to take part?</u>

It is up to you to decide whether or not to take part. If you do decide to take part you will be given this information sheet to keep and be asked to sign a consent form. If you decide to take part you are still free to withdraw at any time and without giving a reason, and without this affecting your healthcare at any way in the future.

<u>What will happen to me if I take part?</u>

You will be asked to visit the Biomechanics Laboratory at the University of Bradford; a visit could last up to 3 hours. You will be asked to undergo or complete the following:

- Have your height and weight measured.

- Perform the following walking tasks:

i. Walk along the walkway towards the second square, and then with the final step being on to your prosthetic foot, come to a halt looking at picture of a smiley face. Then after 10 seconds of stationary standing start to walk again leading with the intact foot.

ii. Walk down a ramp (5 deg decline) and then, with the final step being on to your prosthetic foot, come to a halt looking at a picture of a smiley face. Then after 10 seconds of stationary standing start to walk again leading with the intact foot.

Repeat (ii) at two self-selected walking speeds: customary, slow.

During these tests you will have a number of small spherical markers placed on your clothing and shoes, and infra-red cameras will track your body movements.

The results of this study will be used for research purposes. Any personal information will not be retained.

What do I have to do in preparation?

You will be asked to maintain your usual diet and activity level and refrain from drinking alcohol for 24 hours before your visit and to bring with you Lycra shorts, t-shirt, comfortable flat-soled shoes, and your normal corrective spectacles (if worn) that you use for walking. When you arrive you will be asked to change into shorts, and t-shirt. If necessary Lycra shorts and t-shirt will be provided.

Is there any risk of harm to myself?

There is a hypothetical risk of you losing your balance when performing the tasks, but we have never had anybody do so after many of these studies.

Further information: if you would like more information about the study and what is being asked of you please contact:
Mrs Zahraa Abdulhasan (email: z.m.a.abdulhasan@bradford.ac.uk.ac.uk, (mob. 07405681677) at the University of Bradford.
or
Dr John G Buckley (email: J.Buckley@bradford.ac.uk, (tel. 01274 234641) at the University of Bradford.

Appendix B. Participant Consent Form

UNIVERSITY OF BRADFORD

PARTICIPANT CONSENT FORM

Gait termination when walking down slopes in unilateral trans-femoral amputees; the effects of microprocessor-controlled compared to conventional prostheses.

Researcher: Zahraa Abdulhasan, UoB School of Engineering and Informatics, Biomedical Engineering Department

1. I confirm that I have read and understand the information provided for the above study. ☐
2. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily ☐
3. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving a reason and that this will not affect my medical care or legal rights. ☐
4. I understand that any personal information collected during the study will be anonymised and remain confidential. ☐
5. I agree to take part in the above study. ☐

Name of Participant	Date	Signature
Name of Researcher	Date	Signature
Name of Person taking consent *(if different from researcher)*	Date	Signature

Note: When completed, 1 copy for participant and 1 copy for researcher

Appendix C. Participant Base-line Data Collection Form

Participant name..

Contact telephone number..

Gender: M / F DOB........................

Height (m)............... Weight (kg).................. BMI................ Shoe size.............

Date of amputation.................. Side & level of amputation.........................

Cause of amputation...

Details of current prosthesis

Prosthesis Wt (kg) ……………

Type of knee device: ……………………………….. Settings:

Type of foot device: …………………………………..Settings:

Type of socket...Type of liner...

How long had current prosthesis..................

Others prostheses used in last 3 years………………………………..

Phantom pain / sensation..

Residuum areas of reduced sensation………………………………………………

Residuum areas of sensitivity………………………………………………………………..

Residuum length (cm)............. Intact limb length; ASIS - med malleolus (cm).....................

Stump/socket distal circumference at ~20 mm from socket end, (cm)……………

Stump/socket proximal circumference at the level of greater trochanter (cm)……………

The distance between the two circumferences (cm)……………..

Residuum problems in last 3 months Y/N: details if ‘Y’...

Other musculoskeletal problems (either limb):

Activity level (K-levels)...................

Visual problems, wears specs (walking), contact lenses...

Vertigo / balance problems Y/N: details if ‘Y’..

Current medications:…………………………….