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**BIOMECHANICAL ADAPTATIONS OF LOWER-LIMB
AMPUTEE GAIT: EFFECTS OF THE ECHELON
HYDRAULICALLY DAMPED FOOT**

**Segmental kinetic and kinematic responses to hydraulically damped
prosthetic ankle-foot components in unilateral, trans-tibial amputees.**

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**A thesis submitted in fulfilment of the degree of
Doctor of Philosophy**

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School of Engineering**

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Abstract

The aim of this thesis was to determine the biomechanical adaptations made by active unilateral trans-tibial amputees when they used a prosthesis incorporating a hydraulically-damped, articulating ankle-foot device compared to non-hydraulically attached devices. Kinematic and kinetic data were recorded while participants ambulated over a flat and level surface at their customary walking speeds and at speeds they perceived to be faster and slower using the hydraulic device and their habitual foot.

Use of the hydraulic device resulted in increases in self-selected walking speeds with a simultaneous reduction in intact-limb work per meter travelled. Use of the device also attenuated inappropriate fluctuations in the centre-of-pressure trajectory beneath the prosthetic foot and facilitated increased residual-knee loading-response flexion and prosthetic-limb load bearing during stance. These changes occurred despite the hydraulic device absorbing more, and returning less, energy than the participants' habitual ankle-foot devices. The changes were present across all walking speeds but were greatest at customary walking speeds.

The findings suggest that a hydraulic ankle-foot device has mechanical benefits, during overground gait, for active unilateral trans-tibial amputees compared to other attachment methods. The findings also highlight that prosthetic ankle-foot device 'performance' can be evaluated using surrogate measures and without modelling an 'ankle joint' on the prosthetic limb.

Keywords: Biomechanics, Kinetics, Kinematics, Gait, Unilateral trans-tibial amputee, Prosthesis, Hydraulic ankle-foot device

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Publications and Presentations

The following is a list of peer-reviewed publications and conference presentations/abstracts completed during my doctoral studies at the University of Bradford.

Publications and presentations/abstracts containing data included within this thesis:

De Asha, AR, Johnson, L, Munjal, R, Kulkarni, J and Buckley, JG (2013). Attenuation of centre-of-pressure trajectory fluctuations under the prosthetic feet when using an articulating hydraulic ankle attachment compared to fixed attachment. *Clinical Biomechanics*, 28, 218-224. This work is detailed in Chapter 4.

De Asha, AR, Munjal, R, Kulkarni, J and Buckley, JG (2013). Walking speed related joint kinetic alterations in trans-tibial amputees: impact of hydraulic 'ankle' damping. *Journal of NeuroEngineering and Rehabilitation*, 10(1): doi:10.1186/1743-0003-10-98. This work is detailed in Chapter 8.

De Asha, AR and Buckley, JG (2014). The effects of walking speed on minimum toe clearance and its temporal relationship with peak swing foot velocity in unilateral trans-tibial amputees. *Prosthetics and Orthotics International*. doi:10.1177/0309364613515493. This work is detailed in Chapter 6.

De Asha, A.R., Johnson, L., Kulkani, J. and Buckley, J.G. (2011). The effects of a hydraulic ankle on gait function and symmetry in unilateral lower limb amputees. Proceedings of International Society of Prosthetics and Orthotics Annual Scientific Meeting, London, England, 7-8 October; p 9. From Chapter 4.

De Asha, A.R., Johnson, L., Kulkani, J., Munjal, R. and Buckley, J.G. (2012). Prosthetic limb 'ankle' kinetics and energy storage and return when using a hydraulic ankle device in unilateral trans-tibial amputees. Proceedings of 1st Joint World Congress of International Society of Posture and Gait Research and Gait & Mental Function, Trondheim, Norway, 24-28 June; p 510-11. From Chapter 8.

De Asha, AR and Buckley, JG (2013). A simple to implement and robust method of identifying 'minimum toe clearance' events during unilateral trans-tibial gait. Institute of Physics and Engineering in Medicine Meeting 'Capability after Limb Loss', Birmingham, England, 26 September. From Chapter 6.

De Asha AR, Johnson, L, Kulkarni, J, Munjal, R and Buckley JG (2013). Minimum toe clearance during unilateral trans-tibial amputee gait; effects of using a hydraulically controlled ankle-foot device. International Society of Prosthetics and Orthotics Annual Scientific Meeting, Blackpool, England, 4-5 October. From Chapter 7.

Publications and presentations/abstracts from other studies I have been involved in the past 4 years:

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Buckey, JG, **De Asha, AR**, Johnson, L and Beggs, CB (2013). Understanding adaptive gait in lower-limb amputees: insights from multivariate analyses. *Journal of NeuroEngineering and Rehabilitation*, 10(1): 98 doi:10.1186/1743-0003-10-98. I was involved in data collection, processing and variable output and the writing of the manuscript.

Johnson, L, **De Asha, AR**, Munjal, R, Kulkarni, J and Buckley, JG (2014). Toe clearance in unilateral transtibial amputee gait: effects of a passive hydraulic ankle. *Journal of Rehabilitation Research and Development*, In Press. This study uses the 'event detection' approach detailed in Chapter 6. I was involved in the data collection, processing and variable output and the writing of the manuscript.

Foster, RF, **De Asha, AR**, Reeves, ND, Elliot, DB, Maganaris, CN and Buckley, JG (2014). Identification of touch-down and foot-off when descending and ascending a non-instrumented staircase. *Gait and Posture*,

39, 816-821. I was involved in the conception of the idea and the processing of data, statistical analyses and writing of the manuscript.

Barton, GJ, **De Asha, AR**, van Loon, ECP, Geitjtenbeek, T and Robinson, MA (2013). Manipulation of visual feedback during gait with a time delayed adaptive Virtual Mirror Box. *Journal of NeuroEngineering and Rehabilitation*, In Review. This work was undertaken as part of my MSc.

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Brown, S, **De Asha, A.R.**, Johnson, L, Munjal, R and Buckley, J.G. (2011). Roll-over characteristics and ankle joint kinetics using low-profile dynamic response foot with fixed versus hydraulic ankle in trans-tibial amputees. Proceedings of International Society of Prosthetics and Orthotics Annual Scientific Meeting, London, England, 7-8 October; p 16. This work is from Steve Brown's MEng project. I assisted in this work by taking part in the data collection and processing and took part in the writing of the abstract.

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Chapter 1. Introduction

1.1 Background

The purpose of any lower limb prosthetic device is to replicate, as best as possible, the function of the absent physiological structures. While such function can be approximated by passive prosthetic devices, such devices are inherently inferior to a physiological limb. The functional requirements of prostheses vary, on a continuum, across the amputee population. The functional requirements range from those of highly active and otherwise healthy individuals to those of users suffering with co-morbidities which render them immobile, inactive and sedentary. For this reason various prosthetic devices have been designed and manufactured in order to address the needs of a wide range of users. These designs vary from a simply cosmetic prosthesis with no functionality, through those providing basic support for tasks such as standing or transfers from bed-to-chair, to those providing unaided ambulation up to those providing optimum function to allow an amputee to perform their sport at 'elite athlete' level.

The majority of prostheses are designed to provide rehabilitation of everyday mobility by facilitating ambulation. Each prosthetic device will have its own 'functional performance' characteristics which may, or may not be appropriate for each individual amputee. The effects of switching from using a habitual prosthesis with non-hydraulic ankle-foot attachment to an Echelon™ prosthetic ankle-foot (Chas. A Blatchford and Sons, Basingstoke, UK) on the biomechanics of overground gait are central to this thesis. This device is designed for, and intended to be used by, active lower limb

amputees. It comprises a foot, constructed from separate heel and fore-foot keels which are made from a carbon fibre composite, and a carrier/attachment. The attachment provides hydraulically damped sagittal plane articulation between the prosthetic foot and shank pylon. In common with most other modern prosthetic foot devices the heel and forefoot keels deform and recoil during stance. The prosthetic foot devices used habitually by the participants recruited had either a rigid attachment to the prosthetic shank pylon, which allowed no articulation or were attached via a rubber snubber device that allowed a small amount of elastically controlled articulation between the prosthetic foot and shank. Irrespective of the amount of articulation at the point of attachment ('ankle') all such devices rely, albeit to differing degrees, on the deformation of the keels to simulate plantar- and dorsi- flexion during stance. During weight acceptance the heel keel will deform allowing the forefoot to lower to the floor in simulated plantarflexion. As the shank subsequently rotates above the foot the forefoot keel is loaded and deforms in simulated dorsiflexion then recoils as the foot is unloaded during pre-swing/terminal stance. This recoil back to the neutral position simulates plantarflexion

This thesis is, in essence, comprised of a series of repeated measures experiments. These are designed to investigate, describe and understand the kinetic and kinematic changes during overground gait brought about through the use of an Echelon (hydraulic 'ankle' damping) ankle-foot device compared to more traditional devices (rigid or elastic attachment). A number of previously published studies have sought to investigate and describe

changes brought about by the use of newly developed prosthetic ankle-foot devices. Many of these reports have modelled the prosthetic as if it were an intact limb. Thus they tend to assume the presence of a definable 'ankle' joint which is capable of articulation. This is anomalous because true articulation between segments may not exist but is mimicked by the deformation of the prosthetic foot. For example, 'plantarflexion' and 'dorsiflexion' angular displacements have previously been reported when there is no mechanical articulation between the prosthetic foot and shank. An underlying theme of the present thesis is, therefore, "how do you compare 'ankle' function when there is no 'ankle'?" For this reason 'ankle' kinematics and kinetics were not reported for the prosthetic limb during the experimental chapters. Instead measures such as self-selected walking speed, power flow at the distal end of the prosthetic shank and centre of pressure progression beneath the prosthetic foot were used to describe and compare the functional performance between devices. The thesis contains a review of the salient literature, a detailed description of the methods used during the experiments, chapters detailing each experiment and finally a general discussion of the findings. The following sections describe firstly the general aims of the thesis followed by the specific objectives of each experiment. It then goes on to provide a brief outline of each of the five separate experimental chapters.

1.1.1 Purpose of the thesis

The aim of this thesis was to determine the effects of using a hydraulically damped, uniaxial articulating prosthetic ankle-foot device on the biomechanics of overground walking in unilateral trans-tibial amputees

compared to use of the participants' habitual ankle-foot devices (none of which had a hydraulic attachment). The majority of participants used a prosthetic foot device which shares the same carbon fibre keels as the Echelon but has a rigid, non-articulating attachment to the shank pylon (Esprit™, Chas. A Blatchford and Sons, Ltd., Basingstoke, UK). The other participants used various different devices. In order to achieve this general aim the specific objectives were to determine how switching to use of a hydraulic ankle-foot device from a non-hydraulic device affected:

1. How body weight was transferred onto, and progressed over, the prosthetic-foot.
2. How centre of pressure progression was affected by sagittal plane misalignments of the prosthetic-foot.
3. The toe-ground separation between the intact- and prosthetic-feet at different walking speeds
4. The stance phase, energy storage and return at the distal end of the pylon and the stance phase joint kinetics of the intact- and residual-limbs at different walking speeds.

1.1.2 Thesis outline

The first experimental chapter (Chapter 4) investigated use of the hydraulically articulating attachment device compared to participants' habitual ankle-foot devices. Prosthetic devices are typically aligned and 'set-up' to provide 'optimal' function at the users' self-selected, customary walking speeds; thus this chapter focussed on differences between devices at each participant's freely chosen speed. A cross-sectional, repeated measures design was used to test the hypothesis that the damped, and consequently time-dependent, articulation provided by the hydraulic attachment would allow the prosthetic shank to rotate more smoothly and quickly above the plantigrade foot thereby reducing resistance to the progression of the whole-body centre of mass and consequently increasing centre of mass velocity (walking speed). It was further hypothesized that this 'easier' translation of the centre of mass would be reflected in a less disrupted progression of the centre of pressure.

The second experimental chapter (Chapter 5) examined how centre of pressure progression beneath the prosthetic foot during customary speed walking was affected by sagittal plane misalignments – forwards and backwards tilts and shifts. The effects of such misalignments were compared across three different prosthetic feet which were identical, save for the nature of their attachments to the shank pylon. In the first experimental chapter (Chapter 4) data indicated that spatial and temporal locations of inappropriate fluctuations to the centre of pressure's progression were consistent within

individuals across ankle-foot attachment devices but varied between individuals, thus a single-subject, repeated measures design was used. A subsidiary aim was to determine if each type of 'ankle' attachment would be more, or less, accommodating of misalignment.

The two subsequent experimental chapters (Chapter 6 & Chapter 7) investigated minimum toe clearance during overground walking. The first investigated the effects of altered walking speeds on toe-ground separation when using the participants' habitual prostheses and also described the temporal relationship between peak swing-foot velocity and minimum toe-ground separation. In the subsequent chapter it was hypothesized that use of the hydraulic device would increase stance phase residual-knee flexion and consequently reduce intact-limb toe-ground separation. This chapter therefore examined how stance phase articulation at the hydraulic ankle-foot device, compared to a rigid attachment, affected inter-limb and inter-segmental control of the intact- and residual-limb and in particular intact-limb minimum toe-ground separation. A cross-sectional, repeated measures design was used to investigate how intact-limb minimum toe clearance was controlled across a range of walking speeds.

The fifth and final experimental chapter (Chapter 8) investigated the effects on lower-limb joint kinetics of using a hydraulic attachment device compared to a rigid, non-articulating attachment across walking speeds. Again a cross-sectional, repeated measures design was used. It was hypothesized that when using the hydraulic attachment device speed-related compensatory

increases in joint kinetics on the intact-side would be reduced. It was further hypothesized that residual-knee involvement during weight bearing would increase. In order to test these hypotheses joint moments and powers on the intact- and residual-limbs were measured and compared in both absolute terms and also when normalised to walking speed.

The final chapter of the thesis (Chapter 9) consists of an overall discussion and conclusions. In this chapter the primary findings from experimental chapters were highlighted. The chapter also contains a summary of limitations of the studies and suggested future research directions building upon the work contained within this thesis.

Chapter 2. Literature review

2.1 Epidemiology

In this review 'Amputation' means the surgical removal of a limb or part of a limb; and the review will concentrate on lower-limb amputations. Limb amputations are performed due to one of any number of reasons such as traumatic injury, infection or disease. Unwin et al. (2000) retrospectively examined the epidemiology of lower-limb amputations across 10 centres in Europe, Eastern Asia and the United States of America (USA) between 1995 and 1997. They found that the incidence of major amputations (those which resulted in the loss of the ankle joint) was approximately double that in males than in females and that across all centres approximately 65% of amputations occurred in patients over the age of 60 years. Diabetes was associated with between 20% (Tochigi, Japan) and 90% (Navajo, USA) of amputations across the different centres. The highest incidence of amputations occurred in the Navajo area of the USA; 43.9 per 100,000 head of population per year, which the authors accredited to the extremely high, and anomalous, prevalence of diabetes among this indigenous population: more than one in five Navajo adults aged over 20 years suffer from either type 1 or type 2 diabetes (Epple et al., 2003). In contrast, the lowest incidence of amputation occurred in Madrid; 2.8 per 100,000 head of population per year. Within the United Kingdom, across four centres, the annual incidence of lower limb amputations was 13.6 per 100,000 head of population per year. Of these, 60% were directly related to diabetes or peripheral vascular disease (PVD).

The National Amputee Statistical Database for the United Kingdom (NASDAB) reported that between 1998 and 2005 there were, on average, 5010 lower-limb amputations conducted within the UK each year (NASDAB, 2007). Between 70% and 75% of amputations that occurred were due to PVD. Their most recent report (2006/7) shows 4574 lower-limb amputations were conducted over the 12 month period with the proportion due to PVD being 72%. The report also stated that during 2006/7 53% of all lower-limb amputations were unilateral trans-tibial, 39% were unilateral trans-femoral, 4% bilateral lower-limb and 1% foot, partial foot or lower digit removal (NASDAB, 2007). It would therefore be reasonable to estimate that at present approximately 5,000 lower limb amputations are carried out each year in the UK, of which approximately 2500 are unilateral trans-tibial. Of the annual ~ 2500 trans-tibial amputations within the UK approximately 600 (24%) are due to trauma, infection or neoplasia. These patients are more active than those suffering from dysvascularity.

As the hydraulically controlled articulation device, the effects of which are the subject of this thesis, is intended for more active amputees the following review of literature concerning amputees' locomotor function will focus on studies in which active amputees (typically K3 or K4 on the Medicare functional level scale) were participants.

2.2 Amputation techniques and factors affecting rehabilitation

2.2.1 Amputation techniques: Trans-tibial amputation

As stated above, the majority of lower-limb amputations performed in the UK are trans-tibial. The current surgical procedures involved will be briefly described. All information, except where stated, is taken from the current guidelines published by the British Association of Plastic and Reconstructive Surgeons (BAPRAS). Issues caused by surgery regarding the condition and health of the residuum will also be highlighted.

Trans-tibial amputations occur at a level below the knee and above the ankle. Amputations where the site is within the distal 1/3 to 1/4 of the shank can be problematic, post operatively. This is due to the relative lack of muscular tissue available to use as padding within the stump and the limited space available below the residuum, into which a prosthetic device may be fitted. Conversely, those procedures which are conducted at the most proximal end of the tibia (near to the knee) result in a residuum with reduced lever arm that can cause difficulties with subsequent fitting of a socket due to the stump having a conical rather than cylindrical shape. Historically, surgeons performed elective trans-tibial amputations 'one hand's breadth' (approximately 10–15 cm) distal to the tibial tubercle. Current guidelines (www.bapras.org.uk) suggest a preference for the long posterior skin flap (Burgess) technique and that, wherever possible, between 1/3 to 1/2 the

length of the tibia should be retained (~ 15-17 cm). This method is preferred due to it facilitating successful and durable healing of the wound which, in turn, allows early use of robust post-operative rehabilitation programmes (Smith & Ferguson, 1999). The exact operation site is dependent on the quality of soft tissue available to provide an 'envelope' (soft tissue which encloses the distal end of the residuum, Figure 1), the height of the individual patient and the shape and size of calf muscles. Ideally the operation site should be at a location which positions the distal end of an appropriate length posterior flap at the junction of the Achilles tendon and soleus (Figure 1).

Once the amputation site is identified and marked the skin and soft tissue are cut. As the name of the 'long posterior skin flap' procedure implies the posterior skin and soft tissue is left longer than the anterior. This allows closure of the amputation site by wrapping the skin and soft tissue from the posterior aspect of the shank over the distal end of the limb and attaching it onto the anterior aspect of the tibia (Figure 1), using myoplasty techniques (Blanc & Borens, 2004). An alternative closure procedure is where the gastrocnemius muscle attaches directly to the posterior aspect of the tibia – a myodesis. A myodesis is recommended in cases where post-surgery activity levels are expected to be high (Blanc & Borens, 2004). The attachment of the gastrocnemius, to either the posterior or anterior aspect of the tibia, allows it to retain some of its function as a knee flexor. Following the incisions to the skin and soft tissue the anterior and posterior tibial arteries and the peroneal artery, which are severed during the procedure, are isolated and ligated. These severed vessels are typically positioned, by the surgeon, within the

amputation site to ensure an adequate blood supply to the whole of the residuum. The major nerves within the shank segment (saphenous, deep peroneal, superficial peroneal, tibial and sural nerves) are identified, isolated, drawn down and severed. They are allowed to retract into the soft tissue away from the amputation site which reduces potentially painful, post-operative irritation of them. This technique ensures the nerves are away from normal areas of pressure caused by subsequent wearing of a prosthesis.

Both the tibia and fibula bones are then severed. Typically the tibia is divided at the level of the anterior skin incision perpendicular to its long axis. It is then shaped with an anterior bevel to facilitate better prosthetic fitting and minimise soft tissue irritation. The fibula is also divided, perpendicular to its long axis, usually approximately 1 – 2 cm proximally to the tibial division. This more proximal division prevents the severed fibula becoming a bony prominence which could be uncomfortable within a socket and in turn may lead to tissue damage through prosthetic use. Prior to closure the fibula is smoothed on the distal, anterior corner, similarly to the tibia bevel, to facilitate prosthetic fitting. When a procedure is required to be performed more proximally than the ideal location, due to a more proximal trauma, the fibula can be removed totally rather than be divided.

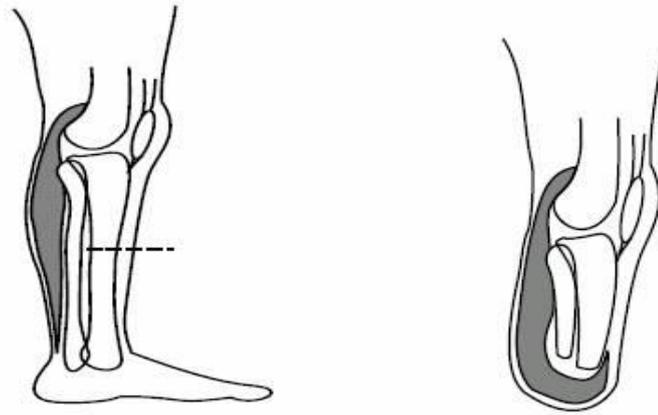


Figure 1. Pre (left) and post (right) trans-tibial amputation showing level of surgery (dotted line) and position of posterior muscle attachment on the anterior aspect of the tibia. From www.rompglobal.org.

Closure is then performed (Figure 2). This comprises three principal parts;

- i. Muscle closure – the fascia of the superficial posterior muscle compartment is advanced anterior and proximal to the distal end of the tibia. It is attached to the periosteum of the tibia and the fascia of the anterior muscle compartment.
- ii. Subcutaneous tissue closure, which assists skin edge approximation.
- iii. Skin closure – usually sutures are used rather than staples as they cause less irritation. These are typically reinforced with skin tapes to minimize tension on the sutures.



Figure 2. A typical soft tissue 'envelope' around the base of the residuum following a trans-tibial amputation. From Tintle et al. (2011).

2.2.2 Factors affecting rehabilitation

Following lower-limb amputation patients are faced with coping with, what is unquestionably, a life changing experience. Some have residual symptoms from the cause of the amputation (e.g. continuing cancers or other injuries from trauma) and almost all are faced with issues resulting directly from the surgery and its outcome. These issues include self efficacy, discomfort, compromised postural stability (Kaufman et al., 2007) and reduced mobility. Gallagher and Maclachlan (2001) used a focus group methodology to investigate factors perceived as important during life adjustment following lower limb amputation and initial prosthetic usage. Three separate groups, each made up of 5 individuals (age range; 20–50 years) who had been amputees for a minimum of 5 years took part. Self-image, acceptance of amputation, and support from, significant others were seen as the factors with the highest impact on adjustment. Other factors which have been

reported to be dominant in the success or otherwise of rehabilitation following an amputation were patient age – older people were less successful at rehabilitation; stump length – shorter stumps reduce the lever arm of the segment, however overly long trans-tibial stumps limit usable prosthetic components; level of amputation – higher amputation levels are associated with reduced rehabilitation (Kelly & Dowling, 2008; Kulkarni, 2008) and the quality and amount of information given to patients both before and soon after the amputation procedure (Mortimer et al. 2002). Another factor which affects rehabilitation is prosthetic usage. The following section therefore focuses on available prosthetic devices.

2.3 Prosthetic development

Prosthetics have been used to replace absent structures since early history. This section however is focused on prosthetic devices which are currently in use. Modern prostheses are comprised of distinct parts which tend to be common across all devices, albeit with slight variations – i.e. a modular system. A trans-tibial prosthesis comprises a socket, suspension system, shank pylon and a foot. Apart from the socket, which is tailor-made to precisely fit the individual's stump, the other components are usually modular in their construction for ease of assembly and service i.e. made up of common component parts which can be independently interchanged - so called modular assembly prostheses (MAPs). Modern prosthetics are constructed from materials such as carbon fibre composites, co-polymer plastics and lightweight metals such as titanium and aluminium. Standard dimensions for components have been adopted by manufacturers. This has

facilitated clinicians being able to individually prescribe selected component parts in order to achieve the best functional outcomes. The following sections briefly discuss the socket and suspension and then, in more depth, feet.

2.3.1 Prosthetic sockets

The socket provides the interface between the residuum and the prosthesis. It is where the bodyweight of the users is transferred from the residuum to the prosthetic limb. It is also where propulsive and supporting moments and forces act on the user and is therefore regarded as the most important component within a prosthesis (Datta et al., 1996). The fit and security of the socket and suspension is vital to enable comfortable and safe use of prostheses (Legro et al., 1999). Discomfort and tissue damage to the residuum caused by in-socket pressures and stresses are a common issue among lower-limb amputees (Chadderton, 1978; Salawu et al., 2006; Kaufman et al., 2007) which can in turn lead to infection, pain and restricted mobility. Sockets are commonly made of a plastic laminate and are typically moulded from a plaster-of-paris cast of the residuum. Each is individually manufactured. The suspension aims to keep the socket securely and comfortably on to residuum, while preventing excess movement between the two.

Until recently trans-tibial amputees most commonly used a patellar tendon bearing (PTB) socket. This socket utilises the tibial condyles and the patellar tendon as the weight bearing areas, avoiding pressure on more sensitive areas of the residuum such as the distal ends of the divided tibia and fibula

and the proximal fibula head. It uses a supracondylar–suprapatellar suspension mechanism, where the femoral condyles and patellar of the residual knee are fully enclosed by the socket. A more modern, and now more common, socket type is the full contact socket. As the name implies the full surface of the residuum is in contact with the socket. All of the participants who took part in the experiments within this thesis used a full contact socket. Suspension of a socket can be achieved by a variety of methods, the choice of which is usually made dependent upon user comfort. Suspension types, which are used in conjunction with full contact sockets, are;

- i. Suction, where a thin silicone liner is worn over the residuum and creates an air tight seal with the socket (this is sometimes combined with ii below). This type of suspension requires a one-way valve within the socket which allows air to be expelled as the socket accepts the residuum, thereby creating a partial vacuum allowing outside air pressure to secure the attachment.
- ii. Liner, pin and lock, where a liner is worn over the residuum which has a ratcheted pin at the distal end. This pin engages with a 'shuttlecock' lock at the distal end of the socket, holding it in place.

In conjunction with the above methods of suspension a tubular sleeve may be worn over the socket which extends up the user's residual thigh. This

sleeve increases the security of the attachment between the residuum and socket. It can be manufactured from neoprene, urethane, latex or other similar materials.

2.3.2 *Prosthetic feet*

Prosthetic feet provide a stable base of support during standing and while ambulating. Globally, one of the most widely used feet is the solid ankle, cushion heel (SACH) foot. This foot was developed during the 1950s (Goh et al., 1984) and is popular, particularly in less economically developed countries, due to low cost and durability (Gordon & Ardizzone, 1960). These feet are of solid construction, having a rigid, normally wooden, keel and a softer, usually rubberised, heel which provides some degree of shock-absorption during early stance while ambulating. More modern feet, termed energy-storing and returning (ESR) or dynamic response feet tend to be constructed from materials with elastic properties such as carbon-fibre. These feet are designed to deflect and deform during weight bearing, thus storing energy elastically, and then recoil, thereby returning a proportion of the stored energy in order to assist propulsion. Unlike ESR feet, SACH feet offer little, if any, energy return during ambulation (Graham et al., 2007). Manufactures of ESR feet claim that during gait, dynamic response feet store energy during stance from initial contact through to late stance when they then return a portion of that stored energy in order to mimic the usual actions of the ankle plantarflexor muscles. Prosthetic 'ankle power' reportedly increased by 300% when using a dynamic response foot compared to a SACH foot for active trans-femoral amputees (Graham et al., 2007) indicating

that such feet do contribute energetically to gait propulsion. In an intact limb the plantarflexors (gastrocnemius; soleus) normally generate a large amount of power during late stance, thus facilitating toe off, forward propulsion and knee flexion during swing (Winter et al., 1995; Neptune et al., 2001; Kirtley, 2006; Winter, 2009). This is far greater than energy return provided by passive prosthetic feet thus while ESR feet provide energy return they cannot replace an intact ankle-foot complex (Gitter et al., 1991; Seroussi et al., 1996; Sanderson & Martin, 1997; Nolan & Lees, 2000; Sadeghi et al., 2001).

The SACH foot was considered 'conventional' prior to the introduction of ESR feet (Goh et al., 1984; McFarlane et al., 1991). A number of studies have compared the effects of using different types of ankle-foot device on the gait of UTAs. Barth et al. (1992) compared the effects of using a SACH foot with non-articulating ESR feet in six UTAs, three of whom had lost their limbs through trauma and three as a consequence of PVD which differentiated the six into being considered high (trauma) and low (PVD) activity levels. This sub-group difference was confirmed by the significantly higher walking speed among the 'trauma' group. Use of ESR feet increased prosthetic-side step length and late stance 'dorsiflexion' at the prosthetic ankle for the 'trauma' group but not for the 'PVD' group. The authors described and discussed their results in terms of gait symmetry i.e. variables were normalised to the intact side rather than in absolute terms. Two implicit assumptions were therefore made. Firstly that kinematic symmetry was the 'ideal' and secondary that the intact limb remained 'constant' between devices. The latter assumption would potentially prevent bilateral changes from being evident and also make the

absolute direction of any changes between devices impossible to interpret. i.e. Any increase or reduction in symmetry could be brought about by a change on either the intact- or prosthetic-side.

Torburn et al. (1990) compared the SACH foot to four ESR feet in a group of five UTAs (three traumatic, two dysvascular) walking at their customary and 'fast' walking speeds. They reported significant differences in kinematics at the prosthetic 'ankle' (larger ranges of motion using ESR compared to SACH feet). Similar differences were also described in other reports (e.g. Barth et al., 1992; James & Stein, 1986) and were no doubt due to the increased deformation of the ESR keels compared to the solid SACH keel. Torburn et al. (1990) went on to report no other kinematic, kinetic or energetic/metabolic differences between prosthetic devices, concluding there to be '*no clinically significant advantage of any one of the feet tested*'. Likewise, other studies which compared UTAs' customary walking speeds (Doane & Holt, 1983, Culham et al., 1986; Mizuno et al., 1992), metabolic cost (Barth et al., 1992) and ground reaction forces (Arya et al., 1995; Postema et al., 1997; Zmitrewicz et al., 2006) when using SACH and ESR feet reported no differences between devices. In contrast, when using an ESR compared to SACH foot, UTAs were reported as having significantly higher walking speeds (Nielson et al., 1989), longer (more symmetrical with the intact-side) prosthetic-limb stance phase (McFarlane et al., 1991; Van Leeuwen et al., 1990) and having lower oxygen uptake (Colborne et al., 1992; Casillas et al., 1995; Nielson et al., 1989), although Nielson et al. (1989) made no inferential statistical comparison. In addition to qualitative comparisons subjective

preferences were recorded for groups of seven (Nielson et *al.*, 1989) and five (Torburn et *al.*, 1990) UTAs who all stated a preference for ESR feet over SACH feet, citing perceived increases in walking speed and stability as the reasons.

These reports focused on differences between the SACH foot, which was considered to be conventional prior to the introduction of ESR feet (Goh et *al.*, 1984; McFarlane et *al.*, 1991), and ESR feet. They report findings which are contradictory. One possible reason for this conflict in findings is the nature of participant groups used in various studies. For example, like Torburn et *al.* (1990), Barth et *al.* (1992) reported no difference in metabolic cost between ESR and SACH. Both studies made these comparisons for groups of UTAs made up of vascular and non-vascular amputees who have significantly different levels of oxygen uptake per meter travelled, respectively (Barth et *al.*, 1992). Differences in the metabolic cost of gait between SACH and ESR feet were reported in otherwise healthy UTAs by Nielson et *al.* (1989) while Colbourne et *al.* (1992) reported metabolic cost differences in children rather than adults. Similar differences existed between the participants of Mizuno et *al.* (1992) who were made up of both younger and elderly, otherwise healthy, adults and Nielson et *al.* (1989) who had participants that were both traumatic and dysvascular amputees. These differences most likely also contributed to the differing findings between ankle-foot devices. In addition any changes due to different devices may not always be large enough across participants of different activity levels or ages to significantly alter outcome measures such as customary walking speeds or

the metabolic cost of walking. Thus these comparative studies highlight that changing one prosthetic foot for another can affect gait function but that the effects of using different prosthetic ankle-foot devices vary across different participant groups.

2.3.3 Hydraulic ankle-foot device

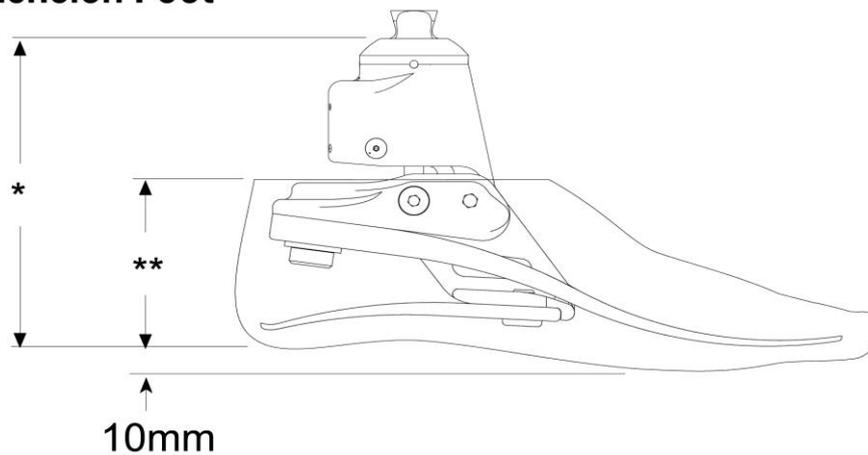
The biomechanical effects of using a hydraulic ankle-foot device are central to this thesis. The Echelon™ device is manufactured by Chas. A Blatchford and Sons (Endolite), Basingstoke, UK and has been clinically available since 2009. It allows nine degrees of sagittal plane articulation between the foot and shank about a fixed axis which is 10 mm anterior to the 'build line' of the prosthetic shank pylon. This plantar- and dorsi- flexion movement is hydraulically dampened/controlled. The device has separate settings which alter the rates of plantar- and dorsi- flexion independently. These arbitrary settings range linearly from 1 (minimum) to 9 (maximum) which equate to damping coefficients of 1.28 to 3.48 Nm.s/deg. The 'foot' is constructed of separate heel and fore foot keels which are manufactured from a carbon fibre composite (Figure 3). The fore-foot and heel keels are fitted as matching pairs which are rated according to the user's weight. There are eight categories which range from 44 kg to 125 kg (44-52 kg, 53-59 kg, 60-68 kg, 69-77 kg, 78-88 kg, 89-100 kg, 101-116 kg, 117-125 kg). The keels are designed to store and return energy during the stance phase of gait. During fitting of this device the settings which control the rates of articulation within the *hyA-F* are adjusted by the prosthetist until deemed to provide "optimal function" at the user's self-selected, customary walking speed. This

adjustment consists of systematically altering the levels of damping of both plantar- and dorsi-flexion while each participant walks using the device. The final settings are decided upon using a mixture of participant feed-back regarding perceived comfort and function and the prosthetist's experience. The manufacturers claim the "*hydraulic yielding function creates a self aligning feature providing stability and security while standing on or traversing varied terrain*" (www.endolite.com). They also claim hydraulic ankle dorsiflexion "*allows greater toe clearance in swing phase*" (www.endolite.com). These claims have not been independently or robustly evaluated.

The device is intended for use by higher functioning amputees who are mobile and active. Amputees who have used the device have reported that it improves comfort and that they feel as though their gait function is enhanced by it (www.endolite.co.uk), although these reports are anecdotal. Prior to this thesis, however, there has been only one published study which investigated any effects on gait function of using the Echelon ankle-foot device. Portnoy et al. (2012), using in-socket sensors, measured pressure at the distal end of the tibia in 10 UTAs during ambulation. Each participant used both their own habitual device and the Echelon device while walking overground and while ascending and descending stairs and slopes. Peak pressure and loading rates were reported to be significantly reduced across all conditions as a result of using the Echelon ankle-foot device, compared to participants' habitual prosthetic foot devices. While the study was limited by having a sensor only at one location on the residuum and the sensors measured only

normal, not shear, forces the authors concluded that use of the device would increase comfort and could potentially reduce pressure related damage caused to the soft tissue of the residuum.

Echelon Foot



* sizes

22 - 24 = 115mm

25 - 26 = 120mm

27 - 30 = 125mm

** sizes

22 - 24 = 70mm

25 - 30 = 75mm



Figure 3. Schematic (top) and photograph (bottom) showing the hydraulic ankle-foot device which is the subject of this thesis– the Endolite Echelon™

2.4 Functional rehabilitation

Balance, posture and gait function are all affected and compromised by a lower limb amputation. The biomechanical difficulties resulting from having to use a prosthetic ankle-foot device, which the amputee has no direct control over, are compounded by the reduction in somatosensory input (Isakov et al., 1992) such as from the plantar surface of the absent foot and other sensory organs, for example the golgi tendon organs within removed muscles.

Balance has been defined as “*a generic term describing the dynamics of body posture to preventing falling. It is related to the inertial forces acting on the body and the inertial characteristics of body segments*” (Winter, 1995) and “*the ability to maintain the body’s centre of mass over its base of support*” (Buckley et al, 2002). It is reasonable to suggest that being able to stand unaided, to move voluntarily and to withstand unexpected perturbations are all actions that one would expect to be achievable for someone with a healthy, functioning balance system. In everyday life able-bodied people take the ability to control balance for granted and are not consciously aware of the complexity of maintaining balance. It tends only to be when a person has difficulty in performing balance-related tasks that the complexities are noticed (Horak, 1997).

Balance strategies, either in quiet stance or during locomotion, are 'learnt' skills (McFadyen et al., 2001). Patients who have lost a lower limb through amputation suffer from a reduction in their ability to maintain balance (Fernie & Holliday, 1978; Isakov et al., 1992; Hermodsson et al., 1994; Kaufman et al., 2007). Therefore these skills have to be re-learnt following amputation. The following section will discuss the issues surrounding this re-education and what strategies may be used to maintain balance by amputees.

2.4.1 Static stability in amputees

Hlavackova et al. (2009) used mirror therapy in an effort to improve upright stance control in a group of unilateral trans-femoral amputees (N = 12). They measured weight bearing symmetry (ratio of vertical components of the ground reaction forces under the intact- and prosthetic-limbs) and the trajectory of the centre of pressure under both the intact- and prosthetic-feet while standing quietly (upright, still and on both limbs) with eyes open during three separate 30 second trials. The protocol was then repeated with mirror feedback provided by a large mirror, placed directly in front of the participants that allowed them to see themselves from the front view. They found that amputees tended to present an asymmetrical weight bearing strategy, with more body weight supported by the intact-limb than by the prosthetic-limb. A weight bearing index was calculated by dividing the measured vertical component of the ground reaction force on the intact side by the corresponding value on the residual side. A mean weight bearing index of 1.51 ± 0.49 without and 1.44 ± 0.36 with mirror feedback was recorded for the group indicating that body weight was distributed approximately 60/40 %

between the intact- and prosthetic-limbs. While the intervention had no significant effect on weight bearing symmetry or in the surface area covered by the centre of pressure beneath the prosthetic-foot it did lead to a significantly reduced area covered by the centre of pressure of approximately 30% under the intact-foot. This, the authors suggested, may have been as a result of the augmented visualisation of their standing position adding to the somatosensory information received via the intact-limb which compensated for the lack of such input from the residual side. The authors concluded that amputees were unable to use this extra visual input when using mirror feedback in order to reduce weight bearing asymmetry. It would appear credible that an increase in visual information in the mirror condition allowed participants to use cues from the surrounding environment to monitor their own 'performance' directly. However the weight bearing asymmetry displayed may well have other, perhaps mechanical or psychological causes which affected whether and how the additional visual information available was used i.e. amputees may simply have 'felt' more secure or more comfortable placing a larger proportion of body weight onto their intact- rather than prosthetic-limb. Hlavackova et al. (2009) offered no potential explanation why there was no change of centre of pressure surface area beneath the prosthetic-foot with additional visual feedback. One potential explanation is that, unlike on the intact-side, participants were unable to make fine postural adjustments which would be possible at the intact-ankle. Unfortunately there was no control group tested so comparisons between the trans-femoral amputees tested and the able bodied cannot be made. Mouchnino et al. (2006) also examined balance during quiet stance however they included a

voluntary movement – a lateral leg raise. Participants were instructed to stand stationary and then raise their leg by abducting the hip. UTAs were significantly slower than controls in completing the movement while standing on both the intact- and prosthetic-sides. This led the authors to conclude that the somatosensory input from the plantar surfaces of both feet prior to the onset of the movement was crucial as feed forward inputs. While their suggestion is plausible other mechanical or psychological reasons may exist. For example the longer time taken by amputees may simply reflect a lack of confidence or be a consequence of residuum discomfort.

Vanicek et al. (2009) recruited UTAs and able-bodied subjects, who were classified as ‘fallers’ and ‘non-fallers’ based on their fall history in the 9 months prior to participation (N = 4 ‘fallers’ + 5 ‘non-fallers’ amputees; 5 ‘fallers’ + 4 ‘non-fallers’ controls) and were subjected to a ‘sensory organisation test’ (Nashner & Peters, 1990) which required that postural sway be measured during quiet stance in 6 different conditions using a Smart Equitest system (NeuroCom, OR, USA):

1. Condition 1: sway measured in a static condition with uncompromised visual, vestibular, and somatosensory feedback (eyes open, fixed surround, and fixed surface). This condition established a baseline level.
2. Condition 2: sway measured in a static condition with eyes closed.
3. Condition 3: sway measured with inaccurate visual cues (sway-referenced moving surround).

4. Condition 4: sway was measured under dynamic conditions with inaccurate somatosensory cues (sway-referenced moving support surface).
5. Condition 5: sway measured with visual cues removed and inaccurate somatosensory information (eyes closed, sway-referenced moving support surface).
6. Condition 6: sway measured with inaccurate visual and somatosensory cues (sway-referenced moving surround and moving support surface).

The only statistically significant difference found was that amputee fallers were significantly less stable than amputee non-fallers when visual and somatosensory cues were inaccurate due to a combination of moving floor surface and swaying surroundings (condition 6). Somewhat surprisingly, given that significant differences between the groups were only observed in one test condition, it was concluded that amputee fallers relied more heavily than amputee non-fallers on visual rather than somatosensory input. However, if that were the case it would be reasonable to expect some differences to have been apparent in other conditions where visual cues were inaccurate or absent. In addition, the sub-dividing of the amputees resulted in groups of four and five individuals which may suggest that extrapolating these findings into a larger population should be done cautiously, if at all. During the same investigation, a motor control test, which involved perturbations of participants' balance by moving the floor surface, was undertaken. All amputees presented significantly less force response under the prosthetic-foot than either the intact- and control-feet in all conditions.

Unfortunately results for amputees, regardless of their fall history, were not reported and compared to the control group so generic strategies used by amputees were not identified or discussed. One interesting finding of their study, particularly in light of the consistent weight-bearing asymmetry reported by Hlavackova et al. (2009), was that amputee fallers were most readily distinguished by the level of asymmetry in body weight borne by each limb. The non-fallers bore more weight on the intact- than on the prosthetic-side whereas in the fallers bodyweight was more evenly distributed between limbs. Vanicek et al. (2009) did not present quantitative kinematic data but did state that the primary adjustment strategy used in both limbs for control participants and in intact-limbs for amputees was an ankle strategy. This strategy may be more effective for those amputees who bear more weight on the intact-limb than on the prosthetic-side. If this were so it would suggest that Hlavackova et al.'s (2009) proposal that unilateral amputee's weight bearing asymmetry during quiet stance should be reduced may be misguided and that it could actually increase fall-risk.

Vrieling, et al. (2008) investigated standing balance in trans-tibial and trans-femoral amputees and controls (N = 5 trans-tibial, 3 trans-femoral + 9 controls). Subjects were exposed to sinusoidal, anterior-posterior perturbations of varying magnitudes for 60 seconds while stood 'quietly' on a moving platform. This was done in three conditions – eyes open, blind-fold and performing a dual task. Weight bearing symmetry and centre of pressure excursion were measured along with the anterior-posterior component of the ground reaction forces. Similarly to the previous studies

(Vanicek et al., 2009; Hlavackova et al., 2009) amputees tended to bear more weight on the intact- rather than prosthetic-limb across all three conditions. Ground reaction forces were found to be significantly larger under the intact-foot than beneath the prosthetic-foot, as was anterior-posterior excursion of the centre of pressure. This larger excursion of the centre of pressure under the intact-foot was similar to that reported by Hlavackova et al. (2009). This is most likely a result of balance being maintained via an ankle strategy on the intact-side as suggested by Vanicek et al. (2009). Another strategy which can be used to maintain balance is a 'hip' strategy (Winter, 1995) which is used during larger perturbations or when moments generated at the ankle are insufficient (Buckley, 2002). Given that amputees appear to rely on the intact-ankle during static balance (Vanicek et al., 2009; Hlavackova et al., 2009) it would perhaps be expected that the participants of Vrieling, et al. (2008) would have done the same until the size of perturbations reached some threshold level at which point they would employ a hip strategy. If such a hip strategy was employed one would expect the larger centre of pressure excursions observed beneath the intact-foot to have been bilateral which it wasn't. This suggests that despite amputees finding sagittal plane movements more challenging to deal with than controls they were able to cope without resorting to a hip strategy which may have become evident if the magnitude of perturbations had been larger. Weight bearing asymmetry was also significantly greater during the perturbations than while the surface was stationary which suggests that as maintenance of balance becomes more challenging amputees place more reliance on their intact-limb.

Buckley et al. (2002) investigated both static and dynamic balance among six highly active unilateral amputees, three trans-tibial and three trans-femoral who they treated as a single group, and able-bodied controls. Dynamic balance was assessed using a uniaxial stabilimeter in both the sagittal and, unlike Vrieling, et al. (2008), the frontal, planes during trials of 20 seconds duration. The stabilimeter was limited to a range of $\pm 5^\circ$ from the horizontal. Participants were deemed to be 'unbalanced' when the stabilimeter was at an angle greater than 4° to the horizontal. Static balance was assessed using a force platform to record centre of pressure excursions. During the dynamic test amputees spent significantly less time 'in balance' than controls and made significantly more downwards movements of the stabilimeter on the prosthetic- compared to intact-side. Both amputees and controls found lateral balance easier to maintain than antero-posterior. Statically, all centre of pressure measures indicated that amputees were less able to maintain a constant pose than the controls and, as with the dynamic test, all participants found lateral (frontal plane) posture more easy to maintain than antero-posterior (sagittal plane). This was likely a manifestation of the centre of pressure having a shorter distance to travel in the antero-posterior direction to be beyond the base of support compared to in the medio-lateral direction. This increased difficulty to maintain sagittal plane pose makes the apparent lack of a hip strategy by the amputees in the Vrieling, et al. (2008) report even more surprising although there were differences in the two methodologies which could potentially account for this. In the Vrieling, et al. (2008) report participants were required to cope with temporally and spatially

constant perturbations rather than maintain balance on an unstable surface thus they would have been able to anticipate perturbations and make adjustments in advance. This could have reduced the need for hip involvement whereas in the Buckley et al. (2002) report participants were required to remain balanced on an unstable surface and may therefore have responded to sudden and unexpected movements with movements at the hip.

A 'waist-pull' was used to perturb the balance, during quiet stance, of 15 UTAs by Curtze et al. (2012). Pulls occurred in four directions (two antero-posterior – forwards and backwards and two, lateral – left and right). The perturbations were temporally unexpected but in a known direction. They were created by applying a horizontal force to a cable attached at the waist, which participants were required to resist, and then releasing the cable. Joint moment impulses were compared at the hip and ankle on both the intact- and prosthetic-sides. Antero-posterior perturbations were controlled using predominantly an ankle strategy as seen by Vrieling, et al. (2008) while medio-lateral perturbations were controlled using a hip strategy. Regardless of the direction of the perturbation these adjustments occurred primarily at the intact-limb. Interestingly the authors reported that the passive properties of the prosthetic-foot device played an important part in recovering from the antero-posterior perturbations.

These investigations provide insights into issues faced by lower limb amputees while standing. While each report identifies differences in control

strategies employed they do all identify use of ankle and/or hip strategies similar to those seen in the able-bodied. The differences between reports are most likely due to the differing paradigms and methodologies employed by the authors rather than differences among the amputee populations used. All the studies involved participants who were typically at least two years post-amputation and had undergone rehabilitation. Barnett et al. (2013) investigated postural responses to volitional and perturbed balance tasks in seven UTAs. These participants had all just completed an initial course of in-patient rehabilitation following amputation procedures and were assessed across the following six months. The authors reported that, unsurprisingly, participants moved from an ankle strategy to a hip strategy as external perturbation increased, however they also noted that use of a hip strategy reduced over time. This must have reflected the 'learning effect' (McFadyen et al., 2001) among 'new' amputees as they became more accustomed to prosthetic usage. The authors also noted that even at the end of the six month period the participants were less able to remain in balance when perturbed than 'mature' amputees and therefore, rightly, concluded that full rehabilitation takes in excess of the six month period they observed.

The difficulties in maintaining balance due to loss of a lower limb are easy to understand intuitively, however Hendershot and Nussbaum (2013) investigated postural control during seating by lower-limb amputees (N = 4 trans-tibial, 4 trans-femoral, 8 controls) and reported similar results to previous standing experiments. They found centre of pressure trajectory length, velocity and surface area covered was larger among amputees than

controls. Interestingly there was no difference between the trans-tibial and trans-femoral groups. The authors suggested that differences observed between amputees and controls were most likely due to the effects of compensations being made in the musculature of the trunk following limb loss. Other investigations have been conducted into how gait function is compromised by lower limb amputation and what compensation strategies are used to enable locomotion. These are discussed and contrasted to able-bodied gait in the following sections.

2.4.2 Amputee gait

Typically unilateral lower-limb amputees display asymmetries during ambulation (Van Leeuwen et al., 1990; McFarlane et al., 1991; Arya et al., 1995; Postema et al., 1997; Powers et al., 1998; Nolan et al., 2003; Highsmith et al., 2010). By convention it is usually assumed that 'normal' gait is relatively symmetrical and that asymmetry in gait is symptomatic of some pathology. However, in a number of studies which have focussed on 'normal' gait, asymmetry is evident. Some examples include statistically significant levels of asymmetry observed in recorded ground reaction forces (Herzog et al., 1989; Giakis & Baltzopoulos, 1996; Yiou & Do, 2010), ankle kinematics (Stefanyshyn & Engsberg, 1994), peak swing-phase knee flexion (Maupaset et al., 2002), double support, stance and swing times (Rosenrot et al., 1980), knee and hip power (Sadegi et al., 1997; Sadegi, 2003) and swing-phase toe clearance (Sparrow et al., 2008).

One potential explanation for lower limb asymmetry displayed in able-bodied participants' gait is laterality. Laterality is displayed in human behaviour as 'handedness' or 'footedness' – i.e. a person's preference for using one hand or foot above the other for motor tasks. Devita et al. (1991) reported that the preferred leg of able-bodied participants generated significantly more positive work than the contralateral limb during gait. However Yiou and Do (2010) found no significant differences between the vertical, propulsive and braking components of the measured ground reaction forces generated on the preferred and contralateral sides. These contradictory findings across studies suggest that symmetry, or the lack of symmetry, may not be constant across the able-bodied population or perhaps across tasks. Regardless of the reason for its existence it is apparent that asymmetry is a consistently reported feature of able-bodied gait. This would suggest that perfect symmetry is an unrealistic rehabilitation aim for lower limb amputees. Symmetrical gait suggests identical movements bilaterally, however as human movement is variable a 'looser' definition is perhaps more appropriate. Griffin et al. (1995) suggested gait symmetry occurs when there are no statistically significant differences in measured variables across the two lower limbs.

A number of different methods for quantifying gait symmetry have been reported. The simplest method (e.g. Raggi et al., 2009) is the index of symmetry (IOS). In order to calculate this value a measured value (X) from one side is divided by the corresponding contralateral value as shown below.

$$\text{IOS} = X_{(\text{left})} / X_{(\text{right})}$$

In this calculation an IOS value of 1 indicates perfect symmetry while a higher or lower value indicates asymmetry.

Another measure, used by Becker *et al.* (1995), is the symmetry index (SI). This gives a value as a percentage with a higher value indicating larger asymmetry. It is calculated thus:

$$SI = ((X_{(left)} - X_{(right)}) / 0.5(X_{(left)} + X_{(right)})) \times 100\%$$

This method was criticised by Sadeghi *et al.* (2000) as differences between sides which are small with respect to the magnitude of the measured values will lower the calculated index and “reflect symmetry”. While this criticism is true, asymmetry exists in ‘normal’ gait so perhaps a more sensitive measure would lack specificity and also potentially identify asymmetry as being significant when it may, in fact, be within a “normal” range.

Sadeghi *et al.* (2000) also report the ratio index (Ia) developed by Vagenas and Hoshizaki (1992) which was originally used to quantify asymmetry during running. This is calculated as:

$$Ia = ((X_{(left)} - X_{(right)}) / \max(X_{(left)}, X_{(right)})) \times 100$$

All three of these methods share common weaknesses. Each only detects and quantifies asymmetry in discreet variables chosen by the investigator. They do not indicate whether certain asymmetries are present throughout a gait cycle (time series asymmetries). If the asymmetry is small, in proportion to the signal, they will indicate a low level of asymmetry which could be

perceived as being indicative of a symmetrical movement pattern – this may not necessarily be a bad thing in ‘normal’, non-pathological gait but during a clinical assessment may result in important information being possibly disregarded.

Other investigators have used inferential statistical difference tests (e.g. Powers et al., 1998; Nolan et al., 2003; Highsmith et al., 2010), as promoted by Griffin et al. (1995), and relied on statistically significant differences occurring between sides as an indicator of asymmetry. This approach shares the limitations of the three above indices in that they rely upon discrete variables at a specific point or predefined event. Consequently they are unable to determine symmetry or otherwise throughout an entire gait cycle and in addition will identify asymmetry between groups but not on an individual basis.

Regardless of how asymmetry is measured it is accepted that UTA gait displays a large amount of kinematic and kinetic asymmetry. Typical UTA gait patterns are discussed in the following section.

2.4.3 Typical amputee gait patterns

Amputee gait is asymmetrical (Van Leeuwen et al., 1990; McFarlane et al., 1991; Arya et al., 1995; Postema et al., 1997; Powers et al., 1998; Nolan et al., 2003) and the level of asymmetry increases with higher levels of amputation (Highsmith et al. 2010; Raggi et al., 2009). These asymmetries are due to differences on both the intact- and prosthetic-sides compared to

the able-bodied. Changes on the intact-side tend to be associated with compensation strategies while changes on the prosthetic-side tend to be a result of the mechanical limitations of the prosthesis compared to a physiologically intact limb.

Spatial and temporal gait variables have commonly been reported to be asymmetrical in UTAs. For example, stance phase, as a proportion of the gait cycle, on the intact-limb increases relative to the involved limb. Sanderson & Martin (1997) reported normalised stance phase values (relative to gait cycle) of approximately 66% intact side and 61% involved side as opposed to 58-62% typically seen bilaterally in able-bodied gait (Kirtley, 2006). It has also been reported that the double support phase while the prosthesis accepts weight is extended relative to double support while the intact-limb accepts weight (Nolan et al., 2003). The authors suggested this was due to compromised balance and reduced comfort on the prosthetic-limb. Step and swing time on the prosthetic-side are significantly higher compared to the intact-side in 'typical' trans-tibial and trans-femoral amputee gait (Isakov, et al. 2000) which is as expected given the increased intact-limb stance time displayed simultaneously (Sanderson & Martin, 1997; Nolan et al., 2003). These temporal differences between intact- and prosthetic-limbs tend to reduce as the velocity of walking increases (Nolan et al., 2003). However self-selected walking speeds of amputees are lower than those in the able bodied (Nolan et al., 2003) which lead to temporal asymmetry being ubiquitous in amputee gait. These findings are in agreement with Schmid et al. (2005) who studied the gait of unilateral lower-limb amputees and

reported double support time was longer when loading the prosthetic-limb than the intact-limb and that the intact-limb's stance phase was longer compared to able-bodied controls (N = 12 + 7 controls). Interestingly, as temporal asymmetry appears to reduce with increased walking speed, kinetic asymmetry was reported to increase at higher speed (Silverman et al., 2008) with differences between the intact- and residual-hip power generation becoming greater (Kinetic asymmetry is described and discussed in more detail below).

Schmid et al. (2005) also investigated centre of pressure progression during stance beneath the prosthetic-foot. The centre of pressure describes the origin of the ground reaction force vector (Winter, 2009) and is also the application point of that force on the foot segment. It is a mathematical concept and is the mean location of all forces applied to the plantar surface of the foot during stance (Kirtley, 2006). During able-bodied gait the centre of pressure is at the lateral posterior border of the foot at initial contact and then progresses along the foot until it reaches the medial, anterior border (halux) at toe off (Kirtley, 2006). This progression occurs as the shank rotates over the foot and the body's centre of mass progresses forwards. Schmid et al. (2005) reported that the centre of pressure remained in the hind-foot area of the foot significantly longer during stance beneath the prosthetic-limb than beneath the intact-limb or an able-bodied control-limb. There were no significant differences in centre of pressure progression during stance between the intact-side and control-limbs. Similarly, in a separate report, the centre of pressure progression was found to be interrupted beneath the

prosthetic-limb of a trans-tibial amputee and at times even move backwards (Ranu, 1988). This delay in the centre of pressure travelling along the plantar surface of the foot is consistent with anecdotal feedback from amputees who describe a “flat” or “dead” spot during early or mid stance on the prosthetic-limb and also the feeling of their movement “stalling” or of having to “climb over” the prosthetic-foot. Furthermore, although not statistically significant, differences in centre of pressure velocities were observed between the intact-side and controls during late stance (Schmid et al. 2005) which led the authors to postulate that asymmetry seen in amputees’ was due not only to a different spatio-temporal trajectory of centre of pressure beneath the prosthetic-limb but also to modifications of the spatio-temporal trajectory of centre of pressure under the intact-limb. This, they suggested, indicated adaptation and compensation occurred in the control of the stance phase on the intact-side during gait. Such compensation in amputees is similar to the variations reviewed above in centre of pressure displacement under the intact-foot reported during quiet standing (Vrieling, et al., 2008; Vanicek et al., 2009; Mouchnino et al., 2006 and Hlavackova et al., 2009). These similarities between ambulation and quiet stance highlight that amputees tend to rely more on the intact-limb than the prosthetic- to maintain and control balance (Curtze et al., 2012).

The effects of changes to prosthetic alignment on centre of pressure trajectories were investigated by Geil and Lay (2004) who quantified plantar-foot pressures in six UTAs during clinical, dynamic alignment processes (i.e. fitting of a newly prescribed prosthetic device). They reported that angular

adjustments of the device typically resulted in an increase of plantar pressure on the opposite part of the prosthetic-foot – i.e. an adjustment at the pyramid between the socket and the pylon which moved the foot laterally would cause a medial shift in the centre of pressure. As this protocol was conducted within a clinical setting while participants' prostheses were being adjusted until the 'optimal' alignment was obtained the authors were unable to systematically alter alignments. However what their report does demonstrate is that centre of pressure progression is influenced by (mis)alignment of prosthetic devices.

As well as 'balance' related asymmetry, as discussed above, amputees typically display kinetic and kinematic inter-limb differences which are not related to maintenance of stability. Typically trans-tibial amputees display reduced loading response knee flexion as well as reduced joint moments (peak and impulse) and powers (peak and work) at the residual- compared to the intact-knee (Gitter et al., 1991; Seroussi et al., 1996; Sanderson & Martin, 1997; Powers et al., 1998). These reduced moments and powers were suggested to be a protection mechanism for the residual-knee joint. Knee kinematics and kinetics of the intact-limb were reported to alter in response to an experimentally induced prosthetic mis-alignment (prosthesis internally rotated beneath the socket) in 17 UTAs (Beyaert et al., 2008). When the prosthetic limb was misaligned peak knee flexion and work done at the knee were significantly higher on the intact- than the prosthetic-side and also significantly higher than for control subjects. There were no changes at the residual-knee as a result of the mis-alignment. This lack of change on the prosthetic-side following the intentional mis-alignment of the prosthetic-foot

was explained as a 'protective' response to the increased discomfort while walking which was reported by participants. However if this was a protective response to increased discomfort at the residual-knee one might, perhaps, expect to see some alteration in residual knee kinematics to increase comfort following the application of a mis-alignment to the prosthetic. Another potential explanation is that rather than minimising knee flexion and moments as a method of protecting the residual-knee joint itself, UTAs may well reduce residual-knee involvement during stance to minimise in-socket torques which could generate the sensation of the prosthesis being twisted from the residuum. In other words a strategy to reduce stump-socket torques, primarily in the sagittal plane. Powers et al. (1998) also examined knee function during gait in UTAs (N = 10 + 10 controls). They found that EMG activity of knee extensors and flexors in the residual-limb was of significantly greater intensity and duration than that observed in the control group. Despite this, residual-knee moments and powers were both significantly reduced. They also found peak flexion at the residual-knee during loading response was significantly less and the peak occurred later than controls' knees. They did not report data from the intact-limb. Their explanation for these findings was that limited or no prosthetic 'ankle' mobility during weight bearing/stance meant that participants kept their weight on the rear-foot longer than able-bodied controls. This explanation was supported by Schmid et al. (2005) who reported the centre of pressure 'dwelling' beneath the prosthetic hind-foot. Powers et al. (1998) suggested that muscular co-contraction around the knee was a stabilizing measure which reduced flexion during loading response and, in turn, reduced the knee flexion moment by

preventing the knee moving anterior to the centre of pressure. Given this muscular stabilisation of the knee it may be reasonable to suggest that rather than the lack of change in kinematics of the residual-knee as reported by Beyaert et al. (2008) being a protective mechanism due to the intentional mis-alignment which they applied, it occurred because stability at the residual-knee (and thus minimisation of in-socket torque) is paramount. If this were so the 'protection response' Beyaert et al. (2008) described existed despite, rather than because of, the intentional mis-alignment applied to the prosthesis. Beyaert et al. (2008) did however identify a different intact-side compensatory mechanism used by amputees – increased knee flexion and increased work done at the knee joint. Together these reports tend to indicate that both the intact- and prosthetic-side are able to alter both kinematically and kinetically in unilateral amputees in order to achieve optimal gait function and/or to maintain residual-limb loads and joint moments within certain, comfortable and functional, limits.

It has also been reported that kinetic asymmetry is present at the hip joints in UTAs (Gitter et al., 1991; Czerniecki & Gitter, 1996; Sadeghi et al., 2001). Amputees use the hip flexors on the residual-limb more than able-bodied controls in order to 'pull' the prosthetic foot off the floor into swing. This is a compensation for absent ankle plantarflexors not pushing the foot against the ground during late stance and pre-swing however it does not provide propulsion. In addition Silverman et al. (2008) observed the primary compensation when increasing walking speed in UTAs (N = 14) to be an increase in intact-limb hip joint peak power generation and work done during

early stance (double support). This compensated for absent ankle power which would normally occur in late stance on the contralateral, prosthetic-side. This increased kinetic asymmetry as walking speed increases described by Silverman et al. (2008) occurred simultaneously with the reduced spatial and temporal asymmetry reported by Nolan et al. (2003).

Asymmetry in swing-foot toe clearance has been reported in UTAs, with toe clearance being lower on the prosthetic- than intact-side (Gates et al., 2012; Wuderman et al., 2012). The reduced residual-knee stance flexion typical in amputee gait (Gitter et al., 1991; Seroussi et al., 1996; Sanderson & Martin, 1997; Powers et al., 1998) may well contribute towards this as a straighter (less flexed) stance limb will be functionally longer than if it were more flexed and thus will elevate the pelvis. Interestingly toe clearance asymmetry has also been reported in older able-bodied males (Sparrow et al., 2008) during overground walking. This toe clearance asymmetry was observed to occur in conjunction with step-time asymmetry in that the foot with the shorter step-time had a larger clearance even though step length remained symmetrical. The authors suggested that as there was temporal but not spatial asymmetry the most likely explanation was that 'safety margins' between the swing foot and the floor was increased on the side with the shorter swing time. As step times and step lengths are typically asymmetrical in unilateral amputee gait this 'safety margin' theory may contribute, to some extent, to the toe clearance asymmetries described by Gates et al. (2012) and Wuderman et al. (2012) however as yet no reports have investigated this. If the 'safety margin' was the driver of asymmetry so it would be expected that toe

clearance on both limbs would increase with increased walking speeds which is what occurs during able-bodied gait (Schulz, 2011). Given that the risk of trips and falls is highest at the instant of minimum toe clearance (Mills & Barrett, 2001) and that amputees are at higher risk of tripping than age matched able-bodied controls (Kulkarni et al., 1996; Miller et al., 2001) minimum toe clearance is clinically relevant – especially so in the context of this thesis which investigates the effects of a prosthetic ankle-foot device that is claimed to increase toe clearance.

It has been reported that kinematic asymmetry increases with higher (more proximal) amputation level. Highsmith et al. (2010) reported that trans-femoral amputees displayed significantly shorter and wider steps during overground gait than trans-tibial amputees and controls. Step durations on both the intact- and prosthetic-sides were significantly longer for trans-femoral amputees than trans-tibial amputees and as a result cadence and walking speed were significantly lower in the trans-femoral than trans-tibial subjects. This is unsurprising given that trans-femoral amputees have to cope with the loss of a knee joint in addition to an ankle joint. Hof et al. (2007) examined lateral dynamic balance of trans-femoral amputees during bouts of two minutes steady state treadmill walking. They, like others (e.g. Highsmith et al., 2010, Nolan et al., 2003) found that amputees exhibited a more asymmetrical temporal and spatial gait pattern than able-bodied participants. Amputees spent longer in stance on the intact-side than on the prosthetic- (~68% / ~60% - intact / prosthetic, ~64% controls both sides) and had a greater stride width than the able-bodied. The authors suggested that

this asymmetry was caused by less precise foot placement on the prosthetic-side resulting in a wider stride. They further suggested this increased stride width (and asymmetry) should not be interfered with or altered by medical interventions and may be crucial in enabling amputees to walk safely. Increased stride width would increase the medio-lateral base of support during double support during which weight is being transferred either onto or from the prosthetic-limb. This suggestion that asymmetry may be a functionally appropriate outcome during ambulation (Highsmith et al., 2010) echoes the findings of Vanicek et al. (2009) who demonstrated that higher levels of weight-bearing asymmetry during quiet stance were associated with non-fallers. Raggi et al. (2009) examined step and stance time asymmetry in trans-tibial amputees, trans-femoral amputees and control participants using sample sizes of 25 trans-tibial, 26 trans-femoral and 5 controls. Indices of symmetry were determined by dividing the variable value for the intact-side by the corresponding value for the prosthetic-side. Similar to Highsmith et al. (2010), they reported that asymmetry significantly increased with level of amputation. Unremarkably, they found that step time was significantly correlated with stance time for both limbs in both trans-tibial and trans-femoral participants. They therefore suggested only one of these parameters needs to be monitored to assess temporal asymmetry. Step and stance times were also correlated in the able-bodied participants and a level of asymmetry was present however this was not mentioned during their discussion other than stating levels of symmetry were different across the three groups.

Vrieling, et al. (2008 a & b) examined gait initiation and termination in trans-tibial and trans-femoral amputees and controls (N = 12 trans-tibial, 7 trans-femoral + 10 controls). During gait initiation it was reported that amputees preferred to lead with the prosthetic-limb. When compared to able-bodied subjects, amputees demonstrated a decrease in peak anterior (propulsive) ground reaction force, a smaller or absent posterior centre of pressure shift and a slower increase in walking speed. In addition all amputee participants increased intact-limb loading, which prolonged the period of propulsive force production by the intact-limb. Tokuno et al. (2002) also studied gait initiation in UTAs. Slightly differently, subjects were asked to initiate gait with both intact- and prosthetic-limbs. It was observed that regardless of which limb was used amputee subjects took significantly longer than able-bodied controls to initiate movement. This longer time, regardless of which limb movement was initiated with, was similar to the slower movements and longer movement times reported to accomplish lateral 'leg raises' by amputees during quiet stance by Mouchnino et al. (2006). This tends to suggest the reason for the temporal differences between amputee and control participants is perhaps more mechanical or comfort related than sensory-dependant. In addition, for the amputee participants, the propulsive impulse generated by the prosthetic-limb was significantly smaller regardless of which limb gait was initiated with when compared to able-bodied controls. In a similar way, during gait termination, Vrieling et al. (2008b) reported that amputees demonstrated significantly decreased peak braking ground reaction forces and no anterior shift in the centre of pressure under the prosthetic-limb. The authors therefore concluded that leading with the intact-

limb at termination was preferable for adequate deceleration and for optimal balance control.

2.4.4 Biomechanical effects of different ankle-foot devices during gait

The primary objective of this thesis was to investigate the biomechanical effects on the overground gait of trans-tibial amputees, of using a uniaxial, hydraulically damped, articulating prosthetic ankle-foot device. The previous sections have discussed gait issues prevalent following amputation and also the typical disruptions to, and adaptations of, posture, balance and gait which occur post-amputation. This section focuses on literature which reports how the design and function of the attachment of the prosthetic ankle-foot device to the shank pylon can affect kinetic or kinematic measures of gait function.

The majority of published studies which have compared between ankle-foot devices attempt to quantify differences in their 'performance' when the devices are used during overground walking. Such studies are discussed at section 2.3.2 and typically involved changing one prosthetic foot for another and reporting kinematic and/or kinetic differences between foot conditions. Studies have reported changes in variables such as ground reaction forces (e.g. Goh et al., 1984; Mizuno et al., 1992; Arya et al., 1995; Postema et al., 1997), joint kinetics (e.g. Postema et al., 1997; Schmalz et al., 2002; Underwood et al., 2004), energy expenditure (e.g. Casillas et al., 1995; Schmalz et al., 2002; Nielsen et al., 2006) and spatial-temporal measures such as walking speed, step/stride length or cadence (e.g. Goh et al., 1984;

Mizuno et al., 1992; Underwood et al., 2004; Nielsen et al., 2006). However a prosthetic foot may be attached to the shank pylon via a number of different methods. The majority of prosthetic feet are fixed to the shank pylon using a non-articulated, rigid connection. Other attachment types link the prosthetic-foot to the shank pylon via a non-rigid connection which provides some level of articulation between the two. These devices can be either single or multi axial meaning they are able to move in one plane (usually 'plantar-' and 'dorsi- flexion') or in more than one plane ('plantar-' and 'dorsi- flexion' coupled with inversion and eversion and/or internal/external rotation) respectively. This articulation is typically achieved by inserting a small rubber grommet between the base of the shank pylon and the top of the prosthetic foot. These feet are designed to allow more mobility although they do tend to weigh slightly more than those without articulation and provide only limited movement (e.g. MultiFlex foot has ~ 15° range of motion) compared to an intact ankle.

Zimitrewicz et al. (2006) compared braking and propulsive ground reaction force impulses among 15 older (over 55 years) UTAs walking with both a rigidly attached ESR foot and a SACH foot and with the same feet attached to the pylon via a multiaxial connection. All participants had undergone amputation for vascular reasons but the authors describe them as "healthy and active". Regardless of foot type there was a significant increase in the propulsive ground reaction force impulse when articulation was present compared to when the feet were rigidly attached. Interestingly, given that other reports discussed in section 2.3.2 presented contradictory results

regarding effects of using ESR versus SACH feet, there was no main effect of foot type on ground reaction force impulse. Given that vascular amputees tend generally to be less active than non-vascular that may well have been a manifestation of ESR feet being of no benefit compared to SACH feet for less active users. The report did however suggest that articulation can be of benefit. Ventura *et al.* (2011) also investigated the effects of altering the type of attachment between the foot and shank pylon. The stiffness of the attachment, and hence rate of articulation, was altered by using different thickness of the 'U' shaped 'ankle' attachment (Figure 4).

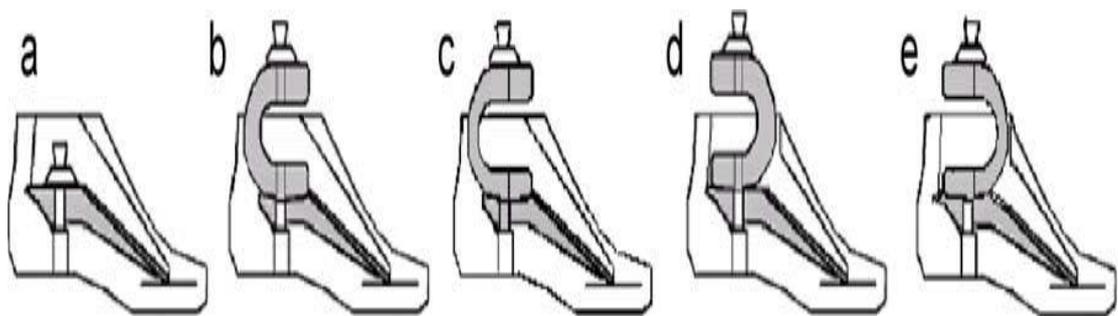


Figure 4. Ankle configurations. a – fixed. b – stiff dorsiflexion. c- compliant dorsiflexion. d- stiff plantarflexion. e – compliant plantarflexion. From Ventura *et al.*, (2011).

Twelve trans-tibial amputees used all the 'ankle' attachments with the same foot and the shank pylon. Compared to the fixed ankle (a) there was increased prosthetic-limb 'ankle' power absorption and generation in mid to late stance with the other four 'ankles' (b – e). Ankles d and e (allowing

predominantly plantarflexion) also showed increased prosthetic-limb 'ankle' power absorption during early stance. These results are unsurprising given that the 'ankles' were made of elastic material and so had intrinsic energy storage and return properties. However the increased power absorption and return may in part, have also been due to increased motion at the 'ankle' that subsequently altered how the dynamic response feet operated. The results also highlight that absorption and return of power by a prosthetic ankle-foot device needs to occur at specific temporal locations within the stance phase in order to contribute to either, comfort, weight acceptance or propulsion.

The impact on trans-tibial amputee gait of using an 'adaptive' ankle-foot device (Proprio-Foot TM, Osser hf, Iceland) has recently been reported (Fradet et al., 2010). The Proprio-Foot weighs 1.4 kg, so is heavier than the Echelon. This device is similar to the Echelon ankle-foot device in that it uses an ESR foot and allows articulation but unlike the Echelon it is microprocessor controlled. This control adjusts the 'ankle' angle during swing but the attachment does not articulate during stance. While the device was 'active' (microprocessor control on) during ramp ascent there was significantly increased 'dorsiflexion' at the prosthetic 'ankle' and increased residual-knee flexion compared to when using the device with the ankle fixed at neutral (Fradet et al., 2010). This was coupled with a reduction in residual-knee flexion moment immediately following initial contact. Similarly, when the device was 'active' during descent, there was a significant increase in prosthetic 'ankle' peak 'plantarflexion' and a reduction in residual-knee flexion at midstance compared to when using the device with the ankle fixed at

neutral. The authors commented that these changes decreased the differences between the amputees and the control group. The same device was also the subject of a study which examined effects of its use during stair ambulation (Alimusaj et al., 2009) in UTAs (N = 16). The study compared the device set with a neutral angle to when it was set in 'dorsiflexion' and reported more 'normal' residual hip and knee kinematics and a 'higher kinetic involvement' of the residual-knee during stair ascent when the device was in 'dorsiflexion' compared to a neutral position. The authors postulated that the increased prosthetic 'ankle' 'dorsiflexion' evident during stair descent may have assisted the shank to rotate above the prosthetic foot throughout stance thereby being "a substantial benefit" to users. However typically during stair descent an intact ankle would be in plantarflexion allowing the landing to be made on the fore-foot (McFadyan & Winter, 1988). An intact ankle would then dorsiflex eccentrically during loading. Thus a 'dorsiflexed' prosthetic 'ankle' may assist in shank rotation during stance but could contribute to higher loading rates immediately after landing.

While introducing a novel analysis variable – Symmetry in External Work (SEW) which describes changes in the whole-body's centre of mass kinetic and potential energy, and compares those changes between the intact- and prosthetic-limb, Agrawal et al. (2009) made comparisons between different prosthetic foot devices, including the Proprio-Foot. These comparisons were made during overground ambulation using a single-subject design thus no inferential statistical analyses were undertaken. Use of the Proprio-Foot and a rigidly attached ESR foot device (Trias+™; Ottobock, Duderstadt,

Germany) resulted in higher levels of inter-limb symmetry than a SACH foot but the authors describe the difference between the Propri-Foot and Trias+ as “not substantially different” (SEW, $94.5 \pm 1.1\%$ Propri-Foot; $92.1 \pm 2.5\%$ Trias+; $35.7 \pm 11.1\%$ SACH). This lack of difference between the micro-processor controlled, adaptive device and rigidly attached foot may have been due to the Propri-Foot simply not functioning differently to the rigidly attached device, each of the devices respective appropriateness for the participant or due to the novel variable being unable to distinguish between them with sufficient resolution. The energetic cost of walking at self-selected, customary walking speed for ten trans-tibial amputees was compared when using a Propri-Foot and their customary ESR feet (Delussu et al., 2013). During normal overground gait use of the Propri-Foot, which was heavier than all participants’ habitual feet, resulted in significantly lower metabolic cost at the same walking speeds compared to the habitual foot devices. This could suggest that a similar benefit would be obtained through use of the Echelon device which is investigated during this thesis. Metabolic cost was also measured while participants walked on inclines (‘up’ and ‘down’) and on the level whilst on a motorised treadmill. There were no significant differences between foot conditions during any of these measurements.

The previous sections have described and discussed the findings of previous published reports which have sought to investigate amputee gait. Regardless of whether the assessment is related to attachment type or to foot design or both, the majority of published studies have typically reported biomechanical measures e.g. prosthetic ‘ankle’ angle and have modelled the prosthetic in an

identical manner to an intact-limb. These measures allow direct comparison between limbs. The following section describes briefly such biomechanical measures and discusses their appropriateness for use in an amputee population.

2.5 Measurement and assessment methodologies

This section focuses on methodologies used and variables chosen to assess prosthetic function. The authors of the majority of studies discussed in previous sections modelled the prosthetic foot-ankle in a similar way to a physiologically intact foot-ankle. That is, an 'ankle' (i.e. articulating joint between shank and foot) was defined on the prosthetic-limb at the same (mirrored) location as on the contralateral intact-limb. When undertaking biomechanical modelling, segments are treated as rigid bodies with joints having pin locations and definable, constant axes (Winter, 2009). This can lead to the anomaly of an 'ankle' angle being reported when there is no articulation between the foot and shank (e.g. Goh et al., 1984). ESR feet, regardless of attachment type are constructed of elastic materials which deform and then recoil during stance as they passively store and return energy. Deformations of a prosthetic foot, which can occur separately for heel and fore-foot keel components, cause 'pseudo' plantar- and dorsi-flexion and occur about indefinable or constantly changing axes. Even when the attachment allows articulation the foot deformation, which occurs during weight-bearing, must contribute to any reported 'ankle' angle (e.g. Ventura et

al., 2011; Fradet *et al.*, 2010; Alimusaj *et al.*, 2009). When the attachment allows articulation it is impossible, within a standard biomechanical model, to differentiate what proportion of any measured 'joint' angle is true articulation between segments and what is due merely to deformation of the foot segment. Even the neutral 'baseline' condition, typically assumed to be that in anatomical pose, is difficult to establish for the prosthetic 'ankle': for an intact limb ankle angles are described with reference to the angle between the foot and shank segments while the participant is standing still in the anatomical position. When this methodology is applied to a lower limb amputee it can be problematic because when standing the prosthetic foot is loaded by the participant's body weight and is therefore deformed. This deformation will be affected by the individual's posture e.g. if more weight is on the heel the foot will be in 'plantarflexion' and conversely if more weight is on the fore-foot the foot will be in 'dorsiflexion' compared to when the foot is unloaded. This can potentially result in the prosthetic 'ankle' appearing to be plantar- or dorsi- flexed in swing, relative to the angle established when standing. Such apparent plantar- or dorsi- flexion is mechanically impossible for a passive, rigidly attached device. During the calculation of joint kinetics (moments and powers) the kinematics and inertial properties of adjacent segment are used. Thus, modelling the prosthetic-limb as if it were an intact-limb, 'ankle power' determined as the dot product of the moment about, and angular velocity of, the 'joint' has been reported for a passive, non-articulating prosthetic ankle-foot device (e.g. Postema *et al.*, 1997; Underwood *et al.*, 2004). Similarly 'ankle power' has been reported for a passive, but articulating via deformation, device (e.g. Ventura *et al.*, 2011).

Even when undertaking kinetic calculations for the residual joints proximal to the amputation, the approach often made is that the foot/ankle complex is modelled in an identical manner as an intact foot-ankle. Such an assumption can result in joint kinetics for the entire prosthetic side (i.e. not just the 'ankle') being at best problematic and sometimes misleading (Geil et al., 2000; Miller & Childress, 2005; Sagawa et al., 2011). Researchers have therefore attempted to provide more ecologically valid tools/approaches when modelling different prosthetic ankle-foot devices to subsequently evaluate its function. The following section describes and discusses some of these techniques.

The whole-body centre of mass can be considered as the single point where the body's weight acts. It is the net algebraic summation of all body segments individual centres of mass (Winter, 2009). The vertical projection of the centre of mass onto the floor surface allows comparison of the horizontal trajectories of centre of mass and centre of pressure. Winter et al. (1998) used these concepts to model a person in quiet stance as an inverted pendulum in which the position of the centre of pressure relative to the centre of mass orientated the direction of any angular accelerations acting upon the pendulum. The authors stated that for stability to be maintained the centre of mass must remain within the pendulum's base of support. During locomotion, however, there are occasions, during single support, when the centre of mass is beyond the base of support yet dynamic stability is still achieved. Hof et al. (2005) introduced the concept of the extrapolated centre of mass to take into account the relative positions of the boundaries of the base of

support and the magnitude and direction of the centre of mass velocity. This allowed the inverted pendulum model to be extended to include ambulation as well as quiet stance. It introduced a measure of dynamic stability which was termed the “margin of stability” defined as the horizontal distance between the border of the base of support and the extrapolated centre of mass.

Basing their assumptions on the body acting as an inverted pendulum during single support, Hansen and Childress (2004) described the trajectory of the centre of pressure during walking transformed from a lab-based global coordinate system to the local coordinate system of the shank segment as an effective ‘rocker’ or ‘roll-over shape’ (Figure 5). They described the radii of these ‘rockers’ or ‘cams’ in terms of either the foot, the ankle-foot, or the knee-ankle-foot co-ordinate systems. This was achieved by altering the ‘distal’ end-point definitions of a ‘virtual shank’ segment in which the ‘proximal’ end-point, and hence the origin, was the ankle. While examining the different radii of curvature produced by different prosthetic feet Hansen *et al.* (2009) concluded that during gait a larger radius was preferable as this provided higher stability during stance on the prosthetic-limb. This may well be true during relatively static tasks such as quiet stance and sit-to-stand movements, or indeed for less mobile or active amputees where stability is paramount. However, for ambulation in more active amputees with greater ranges of motion and higher mobility and activity levels, it may be that a smaller radius of curvature may be beneficial. A small radius would indicate the shank rotated more easily/quickly above

the plantigrade foot as the centre of mass translates forwards. In such instances, it may well be that ‘roll over’ stability, as described by Hansen et al. (2009), is reduced while dynamic ability is increased.

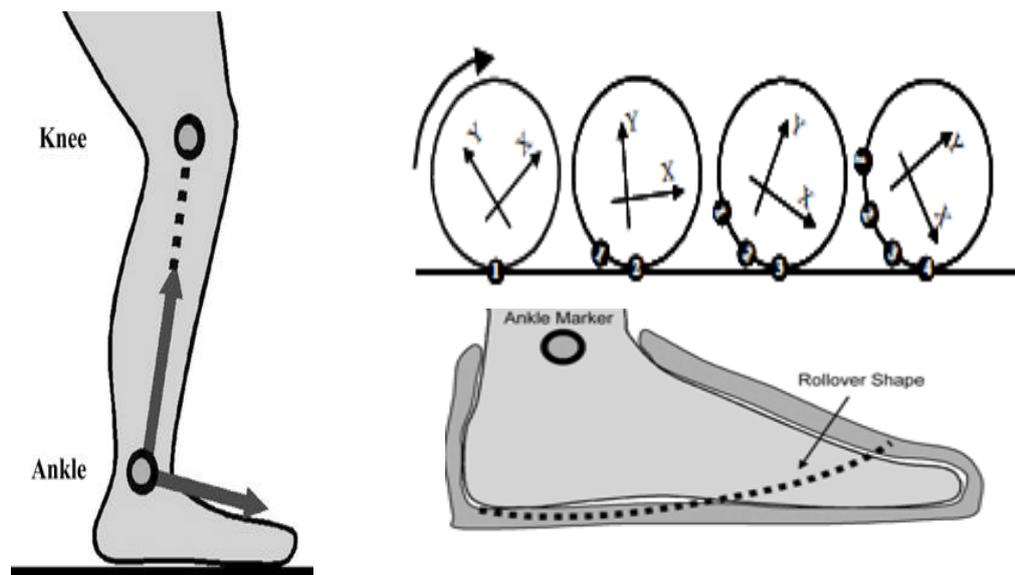


Figure 5. The origin of the segment is defined as the ankle (left). The centre of pressure coordinates are then transferred from a lab based to a segment based system (top right) and the path of the centre of pressure during stance is plotted (bottom right). From: Hansen and Childress (2004).

Roll-over shapes were also investigated using an inverted pendulum analogue by Curtze et al., (2009). Using a weighted pole to simulate the body's centre of mass and lower limb they compared seven different prosthetic feet (all rigidly attached to the shank pylon). They found that each foot had its own distinct roll-over shape and pattern of centre of pressure translation. Importantly the roll-over shapes were modulated by ‘wearing’ shoes which made the roll-over shapes more alike – implying that function of prosthetic ankle-foot devices is altered by foot wear. Major et al. (2011) also

investigated the roll over shape of a prosthetic foot using an analogue. The foot was held stationary in different orientations against a 'floor' surface to simulate initial contact, midstance and toe off as well as two further angles halfway between midstance and the others. A force was then applied to simulate weight-bearing during stance. Major et al.'s (2011) study was different to the other in the literature investigating 'roll over shape' in that rather than fit curves to the shapes and report radii, they presented the shapes as a first order polynomial. This curve fitting likely provides a better description of the centre of pressure trajectory than a single number (i.e. radii) can. Major et al.'s (2011) method could, however, still be compared to other studies by the calculation of instantaneous radii at points along the curve. Given that in essence an inverted pendulum model is the basis for the roll-over concept, it would seem reasonable to use a mechanical analogue to compare prosthetic feet. However, as people are not solid objects moving or rotating at constant velocities, it is a rather large leap to assume these findings would describe how the prosthetic limb/foot is loading during walking. Having said that, these measurements do provide assessments of, and insights into, the function of prosthetic feet which are completely independent of the amputee. As such they may provide a more robust (more repeatable) means of evaluating the function of a prosthetic ankle-foot device.

Another limitation of the roll-over concept is the use of a limited number of data points along the centre of pressure trajectory on which to fit a 'best fit' curve. This may provide a 'global' descriptor of prosthetic foot function,

however it could not reflect small fluctuations or perturbations to the centre of pressure progression such as those reported by Schmid *et al.* (2005) and Ranu (1988). If such fluctuations are ignored an important characteristic of the prosthetic device might be disregarded. Extrapolating roll over characterisation results into the 'real world' should therefore be done with caution.

To get around the problem (as highlighted above) of modelling an 'ankle' joint when often there isn't a definable joint centre researchers have suggested defining the prosthetic 'ankle' without reference to 'anatomy'. Rusaw and Ramstrand (2010) applied a 'functional joint centre' approach to identify the sagittal plane location of an 'ankle' for six different prosthetic feet. The 'functional joint centre' methodology (Schwartz & Rozumalski, 2005) uses angular displacements between adjoining segments to identify the modal point of deflection between the segments, which is then defined as the joint centre. Rusaw and Ramstrand (2010) used motion data from the prosthetic-limb stance phase of overground gait to generate 'displacements' between the prosthetic foot and shank pylon. Their findings indicated that the locations of 'ankle' joint centres varied between the prosthetic feet but generally tended to be anterior and inferior to an 'anatomically' defined 'ankle' (location comparable to the contralateral, intact limb). Five of the six feet tested had a rigid, non-articulating attachment to the shank pylon and one had a limited amount of sagittal plane articulation. While their approach highlighted differences between prosthetic devices it appears anomalous to define an 'ankle' joint that assumes rigid segments and pin-joint articulation, when in

fact the displacements recorded were a result of deformations of the prosthetic foot. In addition, if this methodology were to be used to calculate 'ankle' kinetics for different prosthetic feet, results would be difficult to interpret because of the differently located 'ankle' centres (which may or may not be within the physical confines of the device). Sawers and Hahn (2011) also attempted to identify the 'ankle' joint centres for non-articulating ESR feet. They used a finite helical axis approach to identify the 'centre of rotation' between a prosthetic foot and shank during the prosthetic-limb stance phase. They found, like Rusaw and Ramstrand (2010), that the 'joint centre' tended to be anterior and inferior to an 'anatomically' defined 'ankle'. However, rather than using the modal point (Schwartz & Rozumalski, 2005) as the definition of the 'ankle', they reported that the 'joint centre' was not a fixed point but described a loci throughout the stance phase. They acknowledged that this approach would be impracticable to implement within current modelling and inverse dynamic techniques. Their report did highlight, however, that standard inverse dynamic techniques used in the majority of published research tend to overestimate joint kinetics at the prosthetic 'ankle' (Gitter et al., 1991). This overestimation then continues along/up the kinetic chain and influences moments and powers obtained for more distal joints on the residual limb (Gitter et al., 1991). In order to avoid such measurement issues the following non-standard inverse dynamic techniques have been suggested.

Takahashi et al. (2012) proposed a unified deformable segment model for all structures distal to the knee. This removed the need of modelling a shank

and foot separately and allowed calculation of the total power flow during stance. Their method however only returned the total, scalar power, thus was unable to differentiate sagittal plane power which would contribute to gait propulsion. Prince et al. (1994) suggested a kinetic analysis technique that determined the energy absorbed and returned by the prosthetic-foot by determining the power flow at the distal end of the shank pylon which, regardless of the type of attachment and/or foot, is the physical application point of the forces and moments transferred to and from the shank. As such this modelling approach, as the authors highlighted, can be used for either articulating or non-articulating ankle-foot devices.

The energy entering or leaving the prosthetic foot was assessed by summing the sagittal plane translational and rotational power flows at the.

Translational power (P_{trans}) was defined as:

$$P_{\text{trans}} = (F_z \cdot V_z) + (F_y \cdot V_y)$$

Where; F_y and F_z are the antero-posterior and vertical components of the reaction force (N) acting at distal end of the shank pylon and V_y and V_z are the antero-posterior and vertical velocities of distal end of the shank pylon ($\text{m}\cdot\text{s}^{-1}$).

Rotational power (P_{rot}) was defined as:

$$P_{\text{rot}} = M_x \cdot \omega_s$$

Where; M_x is the sagittal moment acting at the distal end of the shank pylon (Nm) and ω_s is the angular velocity of the shank segment (rads^{-1}).

Total power (P_{dist}) at the distal end of the shank pylon was calculated as:

$$P_{\text{dist}} = P_{\text{trans}} + P_{\text{rot}}$$

Thus the time integral of negative power over the stance phase yielded the energy leaving the shank and flowing to the prosthetic foot ($P_{\text{dist}}^{(\text{neg})}$) while the time integral of the positive power yielded the energy entering the shank and flowing from (returned by) the prosthetic foot ($P_{\text{dist}}^{(\text{pos})}$).

Dumas et al. (2009) directly measured knee moments at the mechanical knee joint of a unilateral trans-femoral amputee's prosthetic-limb. These were compared with those calculated using 'standard' inverse dynamic techniques and those calculated with the assumption that the foot and shank acted as a single rigid segment (i.e. no 'ankle'). The 'standard' technique treated the prosthetic as an intact-limb (i.e. modelled as a shank and foot with articulation between the two at the 'ankle' joint). They reported that the 'standard' technique resulted in an over estimation, similar to that reported by Gitter et al. (1991) and that assuming one rigid foot/shank segment resulted in more ecologically valid values of joint moments at the knee.

Therefore, to avoid modelling the prosthetic as an intact-limb, which could be inappropriate (Sawers & Hahn, 2011; Kent & Franklyn-Miller, 2011) methods advocated by Prince et al. (1994) and Dumas et al. (2009) are those which

will be used for kinetic calculations on the prosthetic side throughout the subsequent chapters of this thesis.

2.6 Summary of literature review

Following a lower limb amputation a person's posture, balance and ability to ambulate are severely compromised. Because the motor coordination strategies involved are adapted compared to able-bodied adults they have to be learnt. Following this 'learning' period, amputees tend to display greater asymmetries in standing balance, and overground gait. These asymmetries are the result of a number of factors such as reduced function on the involved side, mechanical limitations of prosthetic devices, discomfort, altered somatosensory input and compensation strategies on the intact side. These asymmetries increase with higher levels of amputation but importantly reduce with rehabilitation.

While standing, amputees tend to bear more weight through the intact limb than through the prosthetic limb by a ratio of 60/40 compared to 50/50 in the able bodied. In quiet stance amputees tend to find it more difficult to maintain balance during a perturbation than able-bodied individuals.

During overground gait amputees tend to walk slower and have a shorter stride length and higher stride width than the able-bodied. Typically amputees have extended stance time on the intact-limb and reduced stance time on the prosthetic-limb. They generate less propulsive and braking forces on the prosthetic-limb and have altered muscle activation patterns on both

the residual- and intact-sides. Modern passive, prosthetic feet use deformable heel and forefoot keels to simulate plantar- and dorsi- flexion while at the same time storing and returning energy. These devices can improve function compared to SACH feet but they are still far inferior to an intact ankle-foot complex. During able-bodied gait the centre-of-pressure normally progresses in a linear manner along the plantar surface of the foot while in stance. In amputees the centre-of-pressure remains beneath the prosthetic hind-foot area for longer, with anterior progression being delayed. This delay is suggested to be most likely due to the lack of mobility at the ankle and/or the elastic manner in which the prosthetic foot deforms. UTAs display reduced residual-knee flexion during loading response, believed to be the result of muscular co-contractions around the knee in order to stabilize the joint. Changes are also seen at the hips on both the intact- and residual-side where more power is generated during the residual-side pre-swing phase and during early stance on the residual-side in order to compensate for the lack of prosthetic-side ankle power generation.

Typically, biomechanical studies investigating gait function in UTAs have tended to model the prosthesis in the same way as a physiologically intact lower limb, including modelling an ankle joint. This produces misleading and difficult to interpret kinematic and kinetic values for a pseudo 'ankle' joint. Various methods have been proposed to provide a more ecologically valid way of modelling prosthetic foot-ankle devices. Such an approach will be used throughout this thesis to evaluate the impact of the Echelon ankle-foot

device during overground ambulation by active unilateral trans-tibial amputees.

Chapter 3. Methods

3.1 Ethical approval

Full ethical approval was obtained via the National Health Service (NHS) National Patient Safety Agency Integrated Research Application System (Project ID: 35078/GM refers) and the University of Bradford's Ethics Committee. All protocols were conducted in accordance with the Declaration of Helsinki. All participants were given verbal and written information regarding their potential involvement in the study at least 24 hours before being asked to provide written, informed consent to take part.

3.2 Participants

Twenty physically active UTAs (Mean \pm SD age 47.4 ± 12.5 years; mass 87.3 ± 13.5 kg; height 1.79 ± 0.06 m) took part in walking protocols. All amputee participants were recruited by the medical staff at the Disablement Services Centre Manchester and the Northern General Hospital, Sheffield. Eligible participants were UTAs who had undergone amputation due to trauma, infection or carcinoma, at least two years prior to participation (mean 11.85 ± 11.83 years; range 2 – 45 years). All were able to walk independently and were free from any problems with their prosthesis and residuum and their intact-limb that could interfere with locomotion. Any with neurological, vascular or musculoskeletal disease or pathology affecting balance were excluded. Also excluded were any with vestibular deficits, pre-existing sensory dysfunction, peripheral neuropathy, cardiac or respiratory disease

which limited basic function or taking medication which interfered with balance, reaction time or coordination. All participants habitually used a prosthetic foot with either a rigid, non-articulating attachment to the shank pylon or elastically controlled articulating attachment. Twelve participants habitually used an Esprit foot (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). This foot is identical in design to the Echelon ankle-foot device, except that it uses a rigid, non-articulating attachment. Of the other eight participants, five used a MultiFlex foot and ankle (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK), one a Flex-freedom (Ossur, hf, Iceland), one an Elite (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) and one a Seattle Litefoot (Trulife, Poulsbo, WA, USA). These 'non-Esprit' feet had rigid, non-articulating attachments to the shank pylon except the MultiFlex foot and ankle which incorporated elastically controlled, multi-axial articulation (~ 15° degrees range in the sagittal plane) between the foot and shank pylon. Full, individual details of all participants are shown in Table 1.

One physically active, male UTA (age 35 years, mass 79 kg, height 1.83 m, 9 years since amputation) took part in the misalignment experiment (chapter 4). He habitually used an Elite VT foot (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) which he had used for two years and also used other prosthetic ankle-foot devices during 'non daily-life' activities such as running and swimming. He had used his current, full contact, socket for nine months and wore a Tech Polyurethane liner.

Table 1. Descriptive details of UTA participants. All used a full-contact socket.

PARTICIPANT I.D. NUMBER	AGE (YEARS)	MASS (KG)	HEIGHT (M)	SEX	SIDE OF AMPUTATION	TIME SINCE AMPUTATION (YEARS)	HABITUAL FOOT DEVICE
M 01	30	96	1.86	M	LEFT	5	MULTIFLEX
M 02	30	79	1.76	M	RIGHT	8	MULTIFLEX
M 03	63	96	1.86	M	RIGHT	21	FLEX-FREEDOM
M 04	54	81	1.80	M	LEFT	36	MULTIFLEX
M 05	30	74	1.72	M	LEFT	8	SEATTLE LITEFOOT
M 06	60	91	1.77	M	RIGHT	7	MULTIFLEX
M 07	45	93	1.73	F	LEFT	30	MULTIFLEX
M 08	49	97	1.93	M	LEFT	5	ELITE
S 01	34	72	1.72	M	RIGHT	8	ESPRIT
S 02	62	82	1.76	M	LEFT	19	ESPRIT
S 03	39	87	1.71	F	LEFT	9	ESPRIT
S 04	38	62	1.80	M	RIGHT	5	ESPRIT
S 05	65	92	1.79	M	RIGHT	8	ESPRIT
S 06	33	86	1.78	M	LEFT	7	ESPRIT
S 07	61	123	1.75	M	RIGHT	3	ESPRIT
S 08	63	93	1.85	M	RIGHT	43	ESPRIT
S 09	59	100	1.78	M	RIGHT	11	ESPRIT
S 10	39	66	1.78	M	RIGHT	6	ESPRIT
S 11	45	84	1.75	M	RIGHT	2	ESPRIT
S 12	40	91	1.89	M	LEFT	2	ESPRIT

3.3 Laboratory set-up

All data were collected in the Biomechanics Laboratory, University of Bradford. The laboratory (5.8 m x 7.0 m x 2.8 m) was illuminated by six fluorescent light fittings housed within the ceiling. All windows within the laboratory were covered using black roller blinds. The temperature was maintained at a comfortable level (~ 20–22°C) for each individual participant. To prevent any disturbances or distractions the laboratory was locked during all sessions. The floor of the laboratory and upper surface of the floor mounted force platforms were covered in a green, non-slip surface, foam backed vinyl.

The origin of the global laboratory co-ordinate system was defined prior to each data collection session such that the origin was at the left hand corner of force platform 1 (Figure 6). Positive Y axis ran along the long edge of the platform towards force platform 2, positive X axis was perpendicular to the Y axis in the horizontal plane and pointed towards the right when moving in the positive Y direction. The positive Z axis ran vertically upwards from the origin.

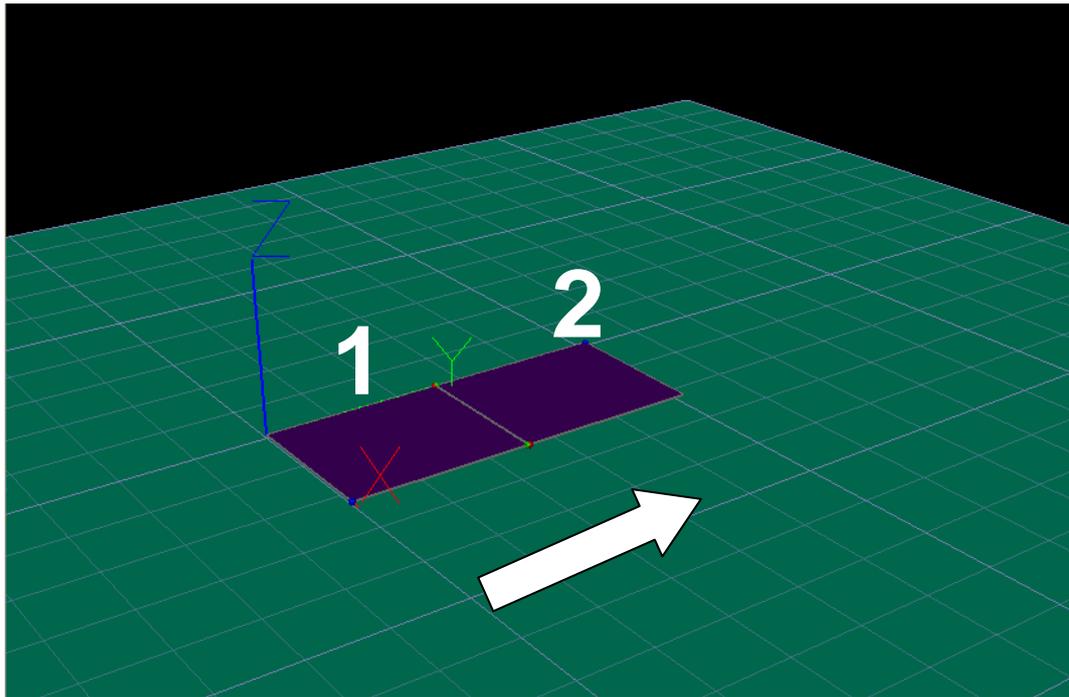


Figure 6. Screen shot from Visual3D motion analysis software (C-Motion, Germantown, MD, USA) showing visualisation of force-plates 1 (left) and 2 (right). The positive direction of each of the three principal axes of movement (X - red, Y - green and Z - blue) are shown at the origin of the global laboratory co-ordinate system. Participants' direction of travel during walking protocols is shown by the arrow.

3.4 Equipment

Kinematic data were recorded using eight infrared cameras (Vicon MX, Vicon, Oxford, U.K.) and passive retro-reflective markers placed on the participants skin or clothing. Marker placement was based on a nine-segment, six degrees of freedom model (see section 3.6). Kinematic data were recorded at 100 Hz. The cameras were operated via Vicon Workstation software (Version 5.2.9, Vicon, Oxford, UK) within a Dell PC. Each camera was fitted with an infrared strobe. Six cameras were wall mounted and two were ceiling mounted, all approximately 2.5 m above the floor. Cameras were positioned such that reconstruction error was minimised within a calibrated

3D volume of approximately 3 m (length) x 2 m (width) x 2.5 m (height). This volume was central to the laboratory and covered the area above the two force platforms. The calibrated volume was defined prior to each data collection session using a 'DynaCal' calibration system (Vicon, Oxford, UK) consisting of a wand and L-frame. The origin of the laboratory co-ordinate system and the three orthogonal axes were firstly defined using the L-frame which was placed so that its arms ran along two sides of force platform 1 and were parallel to, and level with, the floor surface. This procedure also established the position and orientation of each camera relative to the origin of the laboratory based co-ordinate system. The wand was then moved continuously within the 3D capture volume, covering as much area of the volume as possible for approximately 30 seconds in order to define the volume. Calibration reconstruction errors of less than 0.5 mm were deemed to be acceptable during data collection. Participants completed all walking trials in the positive Y direction. All kinetic data were recorded at 400 Hz using two floor mounted strain gauge force platforms (AMTI OR6-7, Advanced Medical Technologies, Boston, USA). The force platforms (508 mm x 464 mm) were mounted in series so that they were flush with the floor of the laboratory with a 3.2 mm gap between them and a 2 mm gap between them and the surrounds. They were in line, along the positive Y direction within the laboratory co-ordinate system running diagonally across the laboratory floor in order to maximise the available walkway space.

The platforms use strain gauge transducers mounted at each corner to measure forces and moments. Each transducer is excited by a constant voltage signal during operation. Whenever a load is applied to the surface of the platform strains occur in each of the elements which alter their electrical resistance. This produces a change in output voltage which is directly proportional to the forces applied. The manufacturers claim less than 2% crosstalk on all channels, $\pm 0.2\%$ of full scale output hysteresis and 0.2% full scale output non-linearity on each of the three orthogonal force measurements (www.amtiweb.com). Each force platform was connected to a six channel amplifier (AMTI MSA-6, Advanced Medical Technologies, Boston, USA). Signals from the amplifiers were sent to the operating PC via a 16 bit analogue to digital (A-to-D) convertor. Each platform had six output channels which each represented one of the forces, along, and moments, about, the three orthogonal axes. Prior to each data collection session the zero levels of both force platforms were calibrated. The amplifiers were also reset before each collection session was begun.

Prior to data collection both force platforms were calibrated using calibrated 'weights' of known mass which were placed onto each platform in turn. The 'weights' were added incrementally from 2 kg to 40 kg and the vertical component of each platform's output compared to the calculated values.

3.5 Participant preparation

During testing all participants wore tight fitting 'lycra' shorts to minimize movement between retro-reflective markers, flesh and clothing. Female

participants wore a t-shirt. Males were bare-chested. All participants wore their own flat-soled, comfortable shoes which they were used to walking in. Data were collected during one single visit to the University of Bradford for each participant (except for the 'misalignment' protocol, Chapter 5, when data collection sessions were conducted on separate days). Frequent breaks during each data collection session were observed to allow time for rest or to become familiar with any alteration to prosthetic equipment, i.e. a change of ankle-foot device. Participants were provided with refreshments as required during their attendance. Each participant's visit to the university lasted for approximately four hours.

Participants' height and weight were measured while wearing their habitual prosthesis and the footwear and clothing they used during data collection. Prior to any data from trials being recorded participants were allowed to familiarize themselves with, and 'practice', data collection protocols.

3.6 Biomechanical model

All modelling was done within Visual3D motion analysis software (Version 4, C-Motion, Germantown, MD, USA). All kinematic data were collected using a nine segment model comprising of head, thorax/trunk, pelvis and both left and right thighs, shanks and feet. This configuration was chosen as it provided an accurate representation of the whole-body centre of mass (Vanrenterghem et al., 2010) without the need for modelling (and tracking markers on) arms and hands.

3.6.1 Model and co-ordinate system

The model and marker set was based on six degrees of freedom (6DoF) segments in which each segment was tracked independently of any other segment (Cappozzo et al., 1995). No constraints were assumed to exist at any of the joints. Thus each segment in the model had, as the name implies, six degrees of freedom; being free to independently translate along and rotate about each of the three principle axes. Each segment was defined anatomically by markers placed on specific landmarks at the segment's proximal and distal endpoints and then tracked during experimental trial using other tracking markers, placed pragmatically on the segment. This modelling technique differs from a 'linked-segment' approach based upon the Newington-Helen Hayes model (Davis et al., 1991). When using a linked-segment model markers are used to define anatomical position as well as to track segments. From the positions of anatomically placed markers the locations of bones and joint centres are calculated using standardised algorithms obtained from cadaveric measurements. Errors made in any marker's placement can affect data obtained across a number of segments and joints (Kirtley, 2002; Charlton et al., 2004). For example a miss-placed ankle marker would impact on where the calculated ankle joint centre was as well as affecting the orientation and tracking of both the foot and shank segments. Consequently this would affect the recorded joint angle at both the knee and ankle.

A 6DoF modelling approach, rather than a linked-segment model, was decided upon in order to avoid using markers placed over joints as tracking markers during trials. This reduced any reconstruction errors caused by soft tissue artefact, which is a major source of error in motion analysis (Leardini, et al., 2005, Cereatti et al., 2006), to a minimum. Soft tissue artefact occurs when marker movement occurs, due to skin deformation/displacement, with respect to the underlying bone. Gao and Zheng (2008) investigated the effect of lower limb soft tissue movement on skin markers during walking and found that markers placed on joints generally exhibited larger movement than other markers placed on the thigh and shank. Karlsson and Tranberg (1999) also reported that the largest soft tissue artefact was found on skin closest to joints. Another practical reason why a linked-segment model was not used was the difficulty of positioning markers accurately on the residual-knee. For all participants the femoral condyles of the residual-knee were enclosed within the socket of the prosthesis and thus impossible to palpate and locate. This was important as variation in knee joint centre location estimation was demonstrated by Holden and Stanhope (1998) to have a significant effect on the magnitude and sign of reported knee moments.

The local coordinate system for each segment was defined at its proximal joint centre; for example the origin of the shank segment was at the knee joint centre. The distal end point was defined as the midpoint between the two distal landmarks; for example the distal end of the shank segment was at the midpoint between markers on 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. A minimum of three points are required for this calculation. When using four points a 'least square' plane was calculated within Visual 3D motion analysis software (C-Motion, Germantown, MD, USA). This was computed such that the sum of the squared distances between the four points and the frontal plane were minimised. The Y axis was the cross product of the Z and X axes. The positive direction of the Z axis was directed from the distal end point towards the proximal end point. For the head segment where there was no proximal joint the origin of the local coordinate system was defined at the midpoint of the two posterior markers.

3.6.1 Marker placement and subject calibration

Markers were placed bilaterally on the anterior and posterior aspects of the head, acromion processes, highest point of the iliac crests, greater trochanters, lateral and medial femoral epicondyles, lateral and medial malleoli and on the shoes over the posterior calcanei, heads of the first and fifth metatarsals, medial and lateral aspect of the feet and on the second toes. All markers placed onto the shoe worn over the prosthetic foot were placed in locations corresponding to the anatomy of the contralateral, intact foot. Markers were also placed on the C7 and T8 vertebrae, sternal notch and xiphoid process. Plate mounted clusters of four markers were attached

to thighs and shanks bilaterally and a skin mounted cluster of four markers was placed around the sacrum.

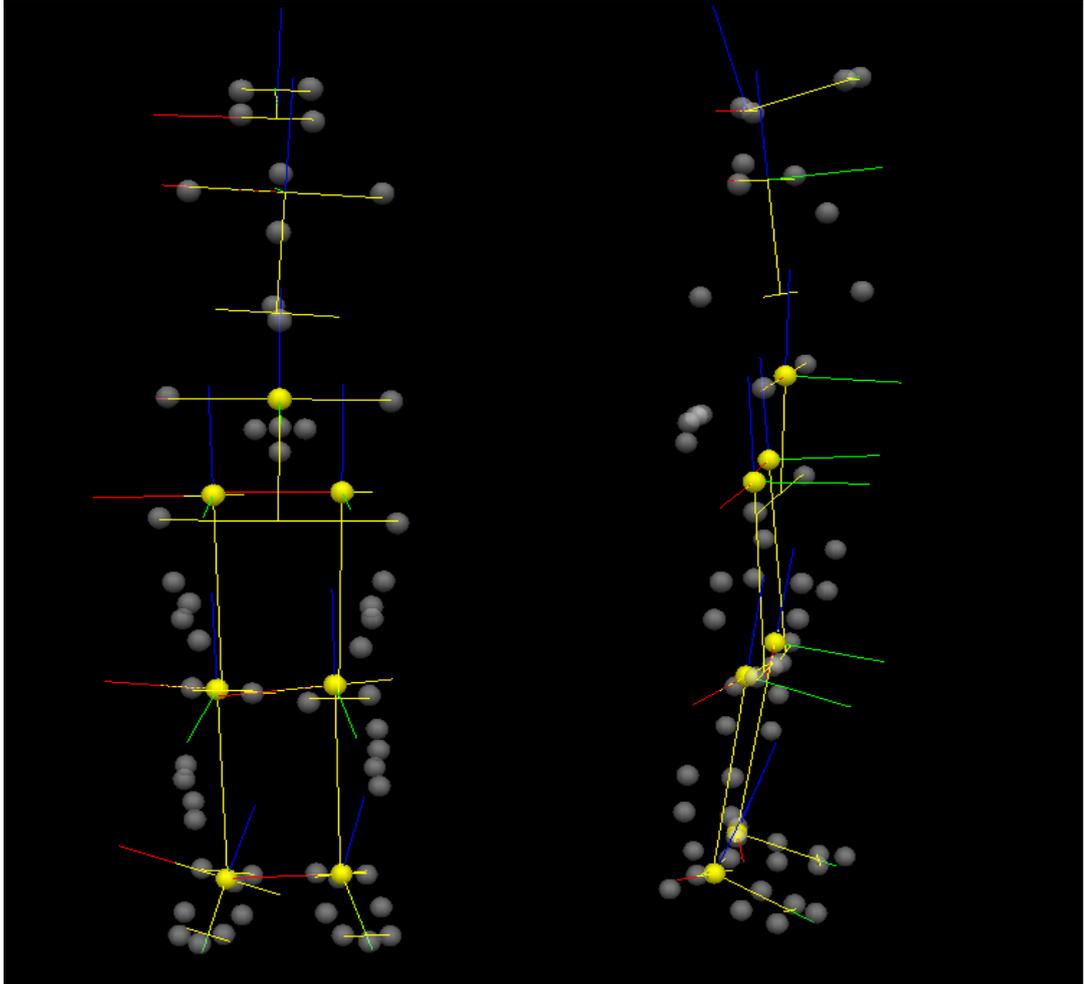


Figure 7. Screen shot from Visual3D motion analysis software (C-Motion, Germantown, MD, USA) showing frontal (left) and side (right) views of the biomechanical model used. Markers (grey) and joint centres (yellow) are shown. Segmental local coordinate systems are shown at the proximal end of each segment.

The marker set for the model was made up of an amalgamation of both 'anatomical' and 'tracking' markers. 'Anatomical' markers were used to define anatomical landmarks so their correct positioning was crucial to ensure as accurate a representation of reality within the model as possible. 'Tracking'

markers were used solely to record the position and orientation (pose) of each segment within the 3D collection volume. Due to this the positioning of these 'tracking' markers was conducted pragmatically to ensure minimum occlusion and movement/soft tissue artefacts during dynamic trials. 'Functional' joint centres (FJCs; Schwartz & Rossumalski, 2005) were created as virtual landmarks throughout. These were used for all intact joints but not for prosthetic 'ankle' joints, which instead were located on the mid-line of the prosthetic pylon at the same height as the FJC on the contralateral, intact ankle. The prosthetic 'ankle' joint was used as the definition of the distal end of the shank segment in all subsequent kinetic calculations. In order to create FJCs movement trials were recorded (as described within the methods section 3.6.3). The foot segments were each defined and tracked using six markers which provided redundancy (i.e. a minimum of three markers were required to estimate the pose of the segment therefore the segment could still be tracked if occlusion of up to three markers occurred during movement trials). Having six markers also facilitated a retrospective conversion from a one segment to a two segment foot model made up of a forefoot and mid/hind-foot configuration, if required. A one segment foot model was used throughout this thesis.

All segments had at least four tracking markers to provide redundancy. These tracking markers needed to be present along with the anatomical markers during the standing calibration trial in order to associate the model template with each individual participant. All participants stood in anatomical pose during the standing trial. All anatomical descriptions of marker locations

relate to this position. The model used a total of 54 markers of which 10 were removed during dynamic trials and 20 were part of plate or headband mounted clusters.

The retro-reflective markers (14 mm except for feet markers which were 9.5 mm) were attached to each participant prior to commencement of data collection. The configuration of the marker set was as per the description above. The vertical and horizontal distances of the toe markers from the anterior distal tip of the shoe and the vertical distance of the heel markers from the posterior distal border of the shoe were recorded. These measurements were used to create virtual landmarks (Heasley et al., 2005; Wuderman et al., 2012) which were embedded within the local coordinate system of the relevant foot segment to provide the best definition and representation of the leading and trailing edges of the shoes respectively.

3.6.2 Static calibration

Following marker placement each participant stood in anatomical pose while a three second capture was recorded as a standing subject calibration file. Participant calibration enabled association between the marker locations for each participant and the model template. Once this was completed the anatomical/calibration markers on left and right acromion processes, lateral and medial femoral epicondyles and lateral and medial malleoli were removed.

3.6.3 Dynamic calibration and definition of joint centre locations

In order to construct FJCs movement trials were recorded. Each movement trial was captured at 100 Hz for 10 seconds. A separate trial was recorded for each of the intact and residual hips and knees and the intact ankle joint. Each trial required the participant to move the joint through a sub-maximal range of motion. Hip joints were flexed and extended, abducted and adducted and circumducted. Knee joints were flexed and extended while ankle joints were plantarflexed and dorsiflexed. All movements were limited to a range of approximately 20° to minimise soft tissue artefact which is known to occur at more maximal movements (Schwartz & Rosumalski, 2005).

3.7 Prosthetic intervention

Prior to completing trials using the hydraulic ankle-foot device each participant's habitual prosthesis was altered by exchanging the existing prosthetic foot for a hydraulic ankle-foot device. All prosthetic adjustments and alterations were carried out by one of two qualified and licensed prosthetists. One made the necessary prosthetic adjustments for all participants from the N.G.H. Sheffield while the other did likewise for all participants from the Manchester D.S.C.

Everything about the prosthesis was kept constant (or near to constant as possible) when one foot type was exchanged for the other. The socket,

suspension and alignment of the shank pylon were unchanged across foot types. When swapping from an Esprit (habitual prosthetic foot device) to an Echelon hydraulic ankle-foot device, or vice versa, only shank length was adjusted (achieved by either shortening the shank pylon or replacing it with a longer one). When swapping one of the other types of habitual prosthetic foot device for the hydraulic ankle-foot device, each habitual foot's alignment was maintained. Functioning of each foot is optimal at its 'ideal alignment', and using such alignment is therefore the fairest way to make comparisons between feet. 'Ideal alignment' for the hydraulic ankle-foot device was achieved during the 'set-up' process while the original alignment of each habitual device was assumed to be 'ideal'.

Once the hydraulic ankle-foot device was fitted, participants walked both indoors and outdoors for a minimum of 45 minutes prior to data collection for accommodation. They negotiated ramps, slopes and stairs and walked over a variety of surfaces including pavements, grass verges and carpeted floors. At the beginning of this period the settings which control the rates of articulation (damping) within the hydraulic ankle-foot device were adjusted by the prosthetist until deemed to provide optimal function at each participant's self-selected, comfortable walking speed as described in section 2.3.3. When participants completed trials using the hydraulic device first and their habitual prosthesis second a similar period of accommodation occurred following their prosthesis being returned to its original, habitual condition.

3.8 Walking protocols

All participants attended at the University of Bradford on one occasion (except for the misalignment protocol, Chapter 5) therefore all overground walking protocols were completed on the same day and in one session. Each session lasted approximately four hours. Refreshments and rest periods were provided during the session as desired. All participants were allowed to familiarise themselves with the walking tasks in the laboratory before kinematic and kinetic data were recorded. Participants completed all walking trials in a straight line along a flat and level 8 m walkway in the positive Y direction of the global, laboratory coordinate system (Figure 1). All participants completed customary walking speed trials.

Typically a prosthetic device is aligned and set-up to provide best function at the user's customary walking speed. Therefore due to the counter-balanced experimental design and because of the methodological limitations associated with speed-controlled studies and the difficulty in generalising findings from such studies to the natural environment (Wilson, 2012) it was decided not to control walking speed. Instead participants were instructed to walk "as they would normally", at their self-selected, customary walking speed across the laboratory. Trials were undertaken in two blocks, one block was undertaken using their habitual prosthetic foot device and the other using the hydraulic ankle-foot device. Block order was counterbalanced. Each then completed a minimum of 10 and a maximum of 20 walking trials until there were 10 successful trials for both the intact and prosthetic side. A successful trial occurred when a 'clean' contact by either of the participants' feet was made with either of the two force platforms without any observable

'targeting' of the platform or alterations in cadence, step length or walking speed during the trial. A 'clean' contact was defined as one in which the foot landed completely within the edges of one of the force platforms and remained thus throughout that limb's stance phase.

Ten of those participants who habitually used an Esprit foot device completed the same customary speed trials as described above and then also completed trials at what each participant perceived to be a slow and a fast walking speed. The participants were instructed to either "walk slowly" or "walk as fast as comfortably possible" prior to these trials respectively. As with the customary walking speed trials, and for the same reasons, it was decided not to control walking speed. Trials at each level of walking speed were completed separately to each other. Participants all completed the customary speed walking trials first and then the slow and fast trials. The order of slow and fast walking trials was counterbalanced across participants as was the order of prosthetic condition. Like while completing customary speed trials (described above), each participant completed a minimum of 10 and a maximum of 20 walking trials in each of the walking speed conditions until there were 10 successful trials for both the intact and prosthetic sides.

3.9 Data processing

Labelling and gap filling of marker trajectories were undertaken within Workstation software (Vicon, Oxford, UK). The C3D files were then exported to Visual 3D motion analysis software (C-Motion, Germantown, MD, USA), where the nine segment 6DoF model of each participant (Cappozzo et al., 1995) was constructed. All signal processing was then undertaken within Visual 3D software.

Kinematic and kinetic data were filtered using a fourth order, zero-lag Butterworth filter with a 6 Hz cut-off. Initial contact and toe-off were defined as the instants the vertical component of the ground reaction force first went above or below 20 N respectively. When there were no kinetic data for the contralateral limb (i.e. not the limb making the clean contact with one of the force platforms), initial contact and toe-off, which were used to determine single and double support for the limb contacting the force platform, were defined using kinematic data. Initial contact was defined as the instant of contralateral limb peak hip extension (De Asha et al., 2013) and toe-off as the instant of peak posterior displacement of the ipsilateral toe marker relative to the pelvis segment (Zeni Jr. et al., 2008). Single support was defined as being from the instant of contralateral toe-off to contralateral initial contact and double support defined as being from initial contact to the instant of contralateral toe-off. Walking speed was defined as the average forwards velocity of the whole-body centre of mass during steady state walking throughout the calibrated collection volume of the laboratory (length ~ 3 m).

The joint kinetics (muscle moments and associated powers) at all joints on the intact limb were calculated using standard inverse dynamics. Kinetics at the distal end of the prosthetic shank were calculated in lieu of prosthetic 'ankle' kinetics using the methods described by Prince et al. (1994) which are described fully within the literature review (Chapter 2.5) and have been used previously (Prince et al., 1998; Morgenroth et al., 2011; Zelik et al., 2011; Segal et al., 2012). The distal end of the prosthetic shank was defined on the segment mid-line at the same height as the contralateral intact ankle. This definition was used in both the habitual and hydraulic foot attachment conditions to enable valid comparisons between them. Plantar- and dorsi-flexion angles on the prosthetic side were not reported. At the residual knee and hip, joint kinetics were determined by assuming the prosthetic foot device and shank to be a single rigid segment with the distal forces acting on this segment being the ground reaction forces (Dumas et al., 2009).

Statistical analyses were made using Statistica (StatSoft, Inc., Tulsa, OK, USA). Prior to any inferential statistical test being applied to data the normality, or otherwise, of their distribution was tested using a Shapiro-Wilk test. Unless stated otherwise within experimental chapters all data were normally distributed ($p > 0.05$). Where data were found to be normally distributed parametric statistical tests (t -tests, ANOVA) were applied.

**Chapter 4. Impact on ‘roll-over’
dynamics of using a hydraulic
ankle-foot device compared to
using participants’ habitual,
non-hydraulic device during
overground customary speed
ambulation**

4.1 Introduction

Passive prosthetic ankle-foot devices are typically, aligned, set-up and adjusted to provide optimal function at the users' self-selected, customary walking speeds thus a fair method of making comparisons between devices is when each are used at what the participant perceives as their 'usual' walking speed. As described previously (Chapter 2) the functional performance of one particular prosthetic foot versus another is often evaluated using inverse dynamics modelling to determine 'ankle' kinetics for the respective feet. A problem with this approach is that it assumes the foot is a rigid segment with definable 'ankle' joint axes (Winter, 2009). Many prosthetic feet have no articulating components, and instead deformation of the foot's flexible keels provide simulated dorsi- and plantar- flexion about an undefined axis. These deformations also occur when an articulated connection device is used. Therefore the interpretation of 'ankle' kinetics is at best problematic and sometimes can be misleading (Geil et al., 2000, Miller & Childress, 2005) thus an alternative method of comparison was used during this experiment.

During normal able-bodied gait the centre of pressure progresses throughout stance along the plantar surface of the foot from the heel forwards to the toes. Such progression reflects how the forward progression of the whole body centre of mass is controlled (Schmid et al., 2005, Kirtley, 2006). In lower-limb amputees the centre of pressure has been found to remain in the

hind-foot area under the prosthetic-foot significantly longer than in both the intact- or control-limbs (Schmid et al., 2005), and at times move backwards towards the heel during early-to-mid stance (Ranu, 1988). Anecdotal perceptions of having to 'climb over the prosthetic foot', 'stopping' or experiencing a 'dead spot' during stance on the prosthetic-limb are common features of UTA gait. Such perceptions are likely to be reflected by interruptions in the forwards progression of the centre of pressure which in turn reflect how bodyweight is transferred over the prosthetic limb. In amputee gait centre of pressure forwards progression will be governed by the compliance of the prosthetic foot device and in particular its ability to simulate ankle function to provide 1st and 2nd rocker phases of gait (Hafner et al., 2002).

The purpose, therefore, of this study was to examine whether use of an Echelon ankle-foot device (*hyA-F*) would attenuate the disruptions in centre of pressure progression commonly reported in UTA gait. It was hypothesised that its use would facilitate bodyweight transfer onto the prosthetic-limb in a smoother less faltering manner, and as a consequence, centre of pressure forward progression would be less disrupted compared to when using the participants' habitual prosthetic feet (*habF*) with traditional attachment; either non-articulating fixed attachment or elastically controlled articulating device. It was further hypothesised that due to the controlled articulation provided by the hydraulic device the shank would rotate forwards above the prosthetic-foot more uniformly and with greater velocity, particularly so during early stance when the hydraulic device has greatest influence.

4.2 Methods

4.2.1 Participants

Twenty physically active UTAs (mean \pm SD age 47.4 ± 12.5 years, mass 87.3 ± 13.5 kg, height 1.79 ± 0.06 m) took part. Full details are given in Chapter 3.2.

4.2.2 Experimental protocol and data acquisition

Details of prosthetic interventions and accommodation procedures are given in Chapter 3. Participants completed two blocks of 10 walking trials; one block was undertaken using their *habF* and the other using a *hyA-F*. Block order was counter-balanced across participants. Prior to completing the block using the *hyA-F* each participant's habitual prosthesis was altered by exchanging the existing foot for a hydraulic ankle-foot device as per Chapter 3.7.

Participants completing trials using their *habF* in the first block (block 1) completed these on arrival at the laboratory. For those completing trials using their *habF* in the second block (block 2), the foot was refitted to their prosthesis following completion of block 1.

Kinematic and kinetic data were recorded as per 'general methods', Chapter 3 while participants walked at their freely-selected comfortable walking

speed. They were instructed to walk “as they would normally”. During data collection, participants wore retro-reflective markers as described in ‘general methods’.

4.2.3 Data processing

All data were reduced and processed within Workstation (Vicon, Oxford, UK) and Visual 3D (C Motion, Germantown, MD, USA) as per ‘general methods’, Chapter 3. When there were no kinetic data for the intact limb, initial contact and toe off on the intact limb (which were used to determine single and double support for the prosthetic limb) were defined using kinematic data: initial contact was defined as the instant of contralateral limb peak hip extension (De Asha et al., 2012) and toe off as instant of peak posterior displacement of the ipsilateral toe marker relative to the pelvis (Zeni Jr. et al., 2008).

4.2.4 Data analysis

The following parameters were determined: average walking speed; mean and peak positive and peak negative (or minimum: if values remained positive) antero-posterior (A-P) centre of pressure velocity; negative (A-P) centre of pressure displacement; variability in A-P centre of pressure velocity across single support; mean sagittal plane angular velocity of the prosthetic shank during the initial double support and single support phases. The first 5 ms of centre of pressure data following initial contact were disregarded to avoid results being affected by any ‘foot-scuff’ during initial contact. To

determine centre of pressure negative displacement, the displacements occurring between each frame were first calculated and any negative displacements were then summed to give the total distance travelled by the centre of pressure in the opposite direction to the direction of travel. Negative displacements in the centre of pressure tended to occur during distinct periods (i.e. over several consecutive frames). Centre of pressure A-P velocity variability was determined as the standard deviation in centre of pressure velocity across single support. Angular velocity of the prosthetic shank was defined as the rate of rotation of the shank segment in the sagittal plane within the global co-ordinate system. These parameters were calculated for each individual trial and then averaged across trials to give a mean value for each foot condition per participant.

4.2.5 Statistical analyses

The normality (or otherwise) of the data was determined using a Shapiro-Wilk test. As the hypotheses were directional comparisons between foot conditions (*habF*, *hyA-F*) were undertaken using 1-tailed, paired *t*-tests and by determining effect size differences. Effect size was calculated as Cohen's 'd' (Cohen, 1977). Statistical analyses were made using Statistica (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

4.3 Results

The magnitude of the peak negative centre of pressure velocity was significantly reduced from $-0.153 \pm 0.110 \text{ ms}^{-1}$ when using the *habF* to -0.043

$\pm 0.057 \text{ ms}^{-1}$ when the *hyA-F* was used ($p < 0.001$, $d = 0.9$, Table 2, also see Figure 8 and Appendix D). The distance travelled posteriorly by the centre of pressure reduced significantly from $-0.022 \pm 0.018 \text{ m}$ using the *habF* to $-0.010 \pm 0.008 \text{ m}$ when using a *hyA-F* ($p = 0.001$, $d = 0.6$, Table 2, also see Figure 9).

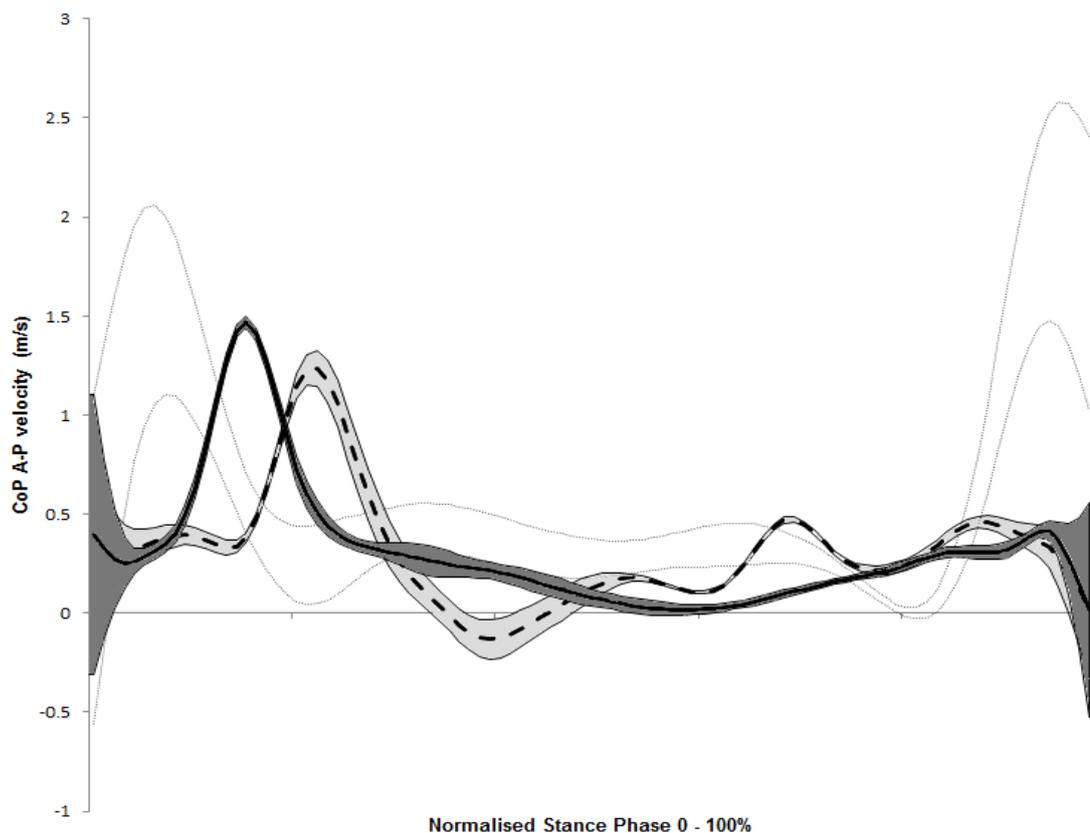


Figure 8. Mean (SD) CoP A-P velocity of the 10 repeat trials, normalised to full stance phase, for one participant when using a *hyA-F* (solid line / dark shading) and *habF* (broken line / light shading). Able-bodied control group CoP velocity $\pm 1\text{SD}$ ribbon is shown (dotted lines) for reference purposes.

There were no significant differences in mean ($p = 0.24$) or peak ($p = 0.28$) anterior centre of pressure velocity between foot conditions (Figure 8).

Centre of pressure velocity variability across single-support was reduced from $0.273 \pm 0.070 \text{ ms}^{-1}$ when using the *habF* to $0.201 \pm 0.063 \text{ ms}^{-1}$ when using the *hyA-F* ($p < 0.001$, $d = 1.0$, Table 2).

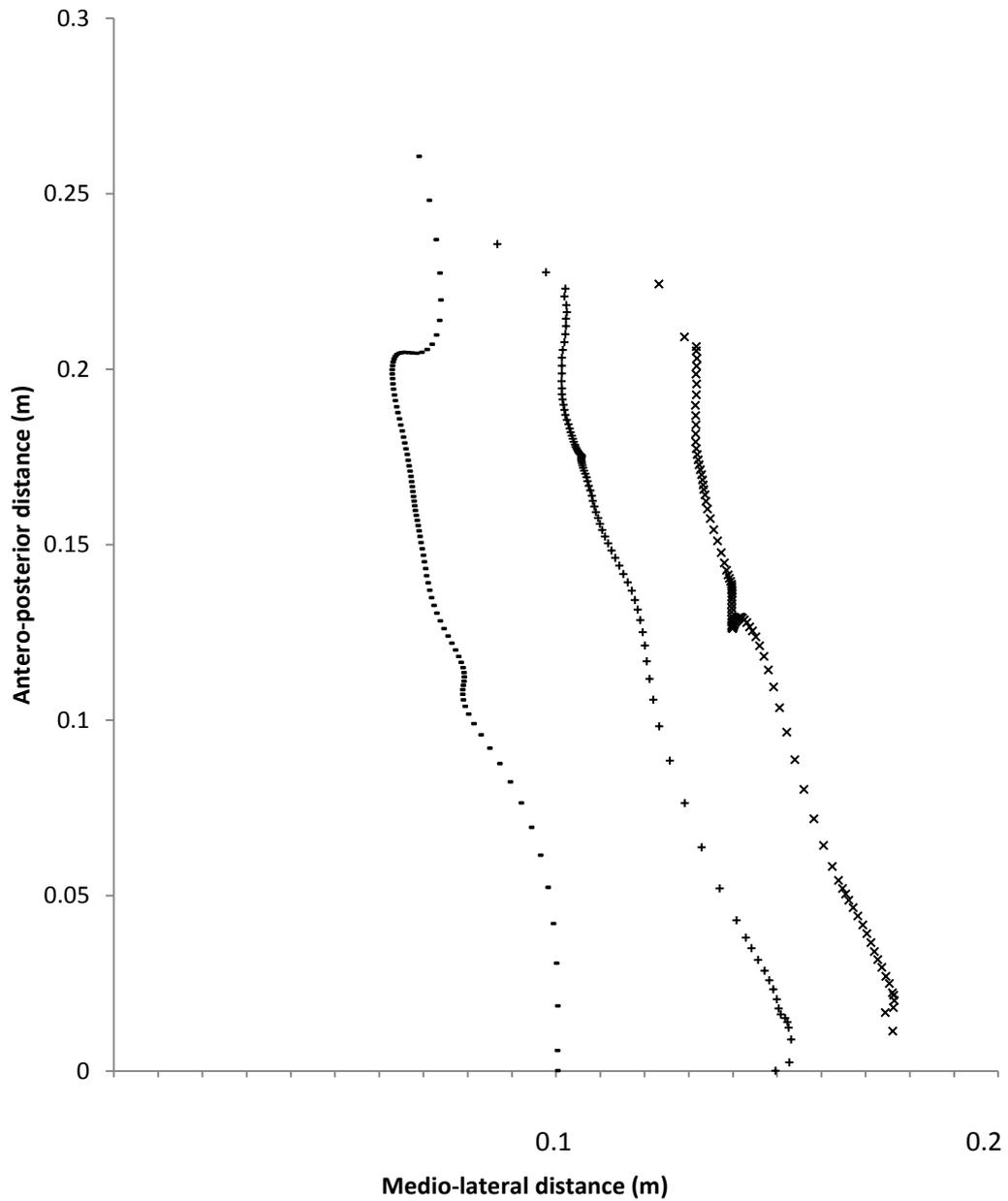


Figure 9. Exemplar CoP displacement traces from one participant while using a *hyA-F* (centre) and *habF* (right). A trace from an able-bodied control is shown for reference purposes (left). The medio-lateral scale has been expanded to allow better view of CoP trajectory fluctuations.

Mean angular velocity of the prosthetic shank during the initial double support phase increased significantly from $94.5 \pm 20.2 \text{ }^\circ\text{s}^{-1}$ when using the *habF* feet to $101.7 \pm 19.2 \text{ }^\circ\text{s}^{-1}$ when using the *hyA-F* ($p < 0.001$, $d = 0.3$, Table 2, also see Figure 10). There were no significant differences in shank angular velocity between foot conditions during single support ($p = 0.37$).

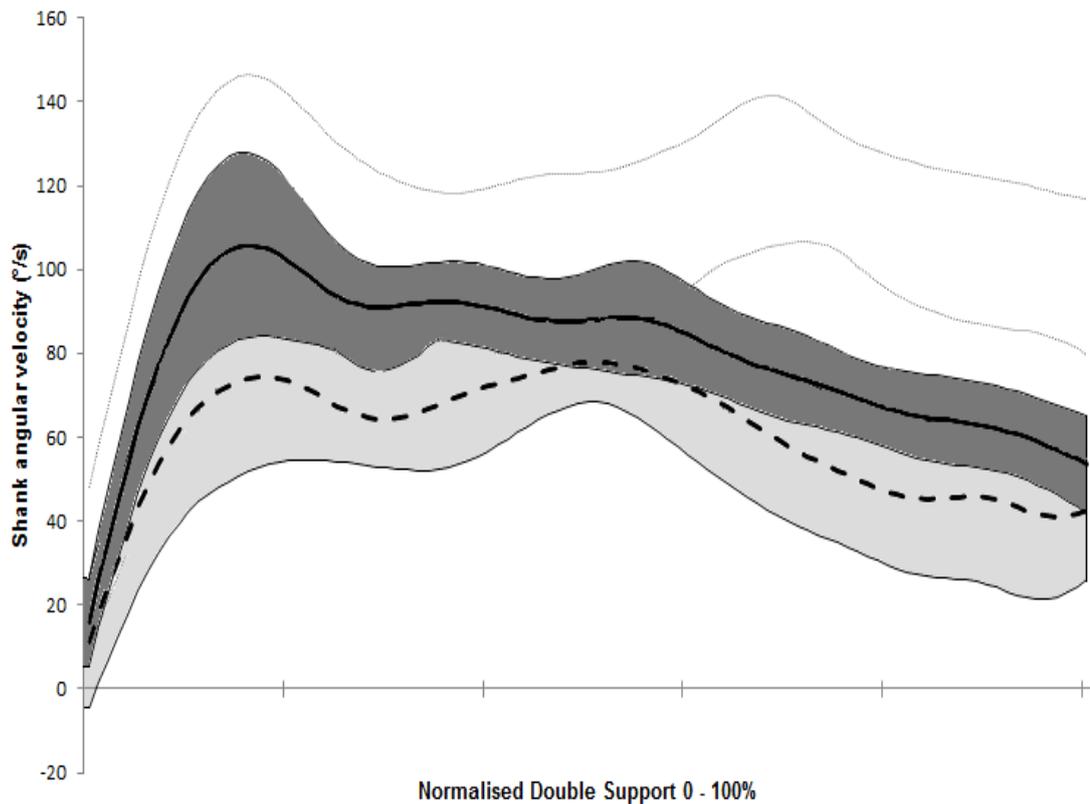


Figure 10. Exemplar mean (SD) shank angular velocity of the 10 repeat trials, normalised to initial double support phase, for one participant when using a *hyA-F* (solid line / dark shading) and *habF* (broken line / light shading). Able-bodied control group shank angular velocity ± 1 SD ribbon is shown (dotted lines) for reference purposes.

Table 2. Group mean (SD) CoP trajectory measures, shank velocities and walking speeds when walking with *habF* and *hyA-F*. Participants in Sub-G1 habitually used an Esprit foot and participants in Sub-G2 habitually used a range of other types of feet (see text for detail). Measures that differed significantly when switching to a *hyA-F* are shown in **bold**. Where differences are significant effect sizes (*d*) are provided (in *italics*).

		Negative CoP displacement	Negative CoP velocity	Maximum CoP velocity	Mean CoP velocity	Mean CoP velocity variability (Single Support)	Shank mean angular velocity (Double Support)	Shank mean angular velocity (Single Support)	Walking speed
		(m)	(ms ⁻¹)	(ms ⁻¹)	(ms ⁻¹)	(ms ⁻¹)	(°s ⁻¹)	(°s ⁻¹)	(ms ⁻¹)
<i>habF</i>	ALL	-0.022 (0.018)	-0.153 (0.110)	2.392 (0.892)	0.365 (0.041)	0.273 (0.070)	94.5 (20.2)	66.5 (9.9)	1.12 (0.14)
	SubG1	-0.026 (0.019)	-0.210 (0.092)	2.607 (1.043)	0.361 (0.036)	0.283 (0.060)	91.8 (23.2)	66.6 (10.4)	1.11 (0.15)
	SubG2	-0.016 (0.013)	-0.066 (0.073)	2.072 (0.504)	0.371 (0.051)	0.267 (0.080)	98.7 (15.3)	66.5 (10.0)	1.14 (0.14)
<i>hyA-F</i>	ALL	-0.010 (0.008) <i>0.6</i>	-0.043 (0.057) <i>0.9</i>	2.305 (0.890)	0.370 (0.043)	0.210 (0.063) <i>1.0</i>	101.7 (19.2) <i>0.3</i>	66.2 (8.8)	1.17 (0.15) <i>0.3</i>
	SubG1	-0.010 (0.00) <i>0.8</i>	-0.062 (0.066) <i>1.4</i>	2.535 (1.006)	0.378 (0.036)	0.212 (0.073) <i>1.5</i>	100.6 (21.6) <i>0.3</i>	66.7 (9.3)	1.17 (0.15) <i>0.3</i>
	SubG2	-0.009 (0.008) <i>0.5</i>	-0.014 (0.021) <i>0.7</i>	1.960 (0.577)	0.358 (0.055)	0.204 (0.050) <i>0.8</i>	103.3 (16.0) <i>0.2</i>	65.3 (8.4)	1.18 (0.14) <i>0.2</i>

4.4 Discussion

When walking with a non-articulated prosthetic foot the fore-foot is lowered to the floor following initial contact via a combination of heel deformation (creating simulated plantarflexion) and forward limb rotation (caused by bodyweight translating over the foot). When using a hydraulic ankle-foot device, the lowering of the foot is also facilitated by the passive mechanical

plantarflexion at the hydraulic unit. The key findings of the present study were that use of a hydraulic ankle-foot device significantly reduced or eliminated centre of pressure posterior displacement, reduced the peak negative centre of pressure velocity, reduced centre of pressure velocity variability across single support, increased the mean forwards shank rotational velocity during weight transfer onto the prosthesis and thus increased overall walking speed. These findings support the hypotheses and indicate that use of the device led to bodyweight being transferred onto the prosthetic limb in a smoother, less faltering manner which allowed the centre of mass to translate more quickly over the foot. This is likely why freely chosen walking speed was found to increase. In addition, participants reported the perception of having to 'climb over' their prosthesis was no longer present, which presumably was a consequence of the above findings.

Schmid *et al.* (2005) and Ranu (1988) have previously reported that 'stalling' of the centre of pressure tended to occur under the hind-foot during early or mid-stance in trans-femoral and trans-tibial amputees respectively. In the present study the exact locations and timings of the disruptions to the anterior progression of the centre of pressure were not consistent between participants with most participants displaying disruptions in centre of pressure progression beneath the mid-foot in addition to the hind-foot. However, the location and timing of any disruptions to the centre of pressure trajectory tended to be consistent within participants across foot conditions. This suggests that when and where centre of pressure disruptions occur is not solely a function of the prosthetic foot used; rather it is an individual's

response to the foot and/or their style of walking. Low variability across the 10 repeated trials (see Figure 8) suggests such responses were consistent for each participant. In able-bodied gait the centre of pressure A-P velocity pattern can be associated with the notion of the three 'rockers' of gait; with relatively high positive velocities during the first and third rockers (during which the foot rotates about the heel and toe regions respectively) and a slower, near constant velocity during the second rocker (during which the foot is relatively plantigrade and the ankle becomes the rocker, see Figure 8). In general both the habitual feet and *hyA-F* devices were able to mimic, albeit to differing degrees, the first two rockers with regard to centre of pressure velocity. However, the first rocker (reflected by a short duration rapid increase in centre of pressure velocity (Figure 8), was temporally delayed compared to that in able-bodied gait, particularly so when using the habitual feet. This delay was likely a consequence of the compression/deformation of the prosthetic heel keel needed to allow the foot to become plantigrade. Such 'early stance' centre of pressure disruption corroborates previous findings that have indicated the centre of pressure becomes 'stalled' under the prosthetic hindfoot (Schmid *et al.* 2005). This delay was reduced, but not removed when using the *hyA-F*. There were also fewer centre of pressure velocity fluctuations during the second rocker period (single support) when using the *hyA-F* compared to *habF* (as evidenced by the reduced variability in centre of pressure velocity). As single-support represents the period when there are no propulsive or braking forces applied by the contralateral (intact) limb, fewer fluctuations in centre of pressure velocity during this period reflect a more uniform transfer of bodyweight over the prosthesis. Finally, it is

apparent that participants were unable to facilitate generating a short duration rapid increase in centre of pressure A-P velocity during the third rocker as seen in able-bodied controls irrespective of which prosthetic foot device was being used. This is due to the lack of active plantarflexion via concentric muscle action prior to toe off and highlights the major limitation of all passive prosthetic feet.

Due to the counter-balanced experimental design and because of the methodological limitations associated with speed-controlled studies and the difficulty in generalising findings from such studies to the natural environment (Wilson, 2012) it was decided not to control walking speed. Instead participants were asked to walk at what they perceived to be their own customary walking speed. As highlighted above, this freely chosen walking speed was found to be significantly greater when participants used the *hyA-F*. While this increase was statistically significant the mean absolute increase was 0.05 ms^{-1} and the effect size ($d = 0.3$) was small to medium. In order to establish whether the centre of pressure trajectory changes found when participants used the *hyA-F* were simply due to an increase in walking speed rather than the functioning of the device, a retrospective investigation of the relationship between walking speed and centre of pressure progression was undertaken. This analysis highlighted that there was no significant relationship between walking speed and the magnitude of peak negative centre of pressure displacement, mean centre of pressure velocity or peak positive centre of pressure velocity irrespective of foot type ($R^2 \leq 0.015$, $p \geq 0.1$). However, peak negative centre of pressure velocity was significantly

related to walking speed when using the *habF* ($r = -0.1672$, $R^2 = 0.0280$, $p = 0.025$), but not when using the *hyA-F* ($p = 0.35$). This indicates that when using the *habF* the velocity of the negatively directed centre of pressure excursion was greater (i.e. increased in negative direction) in trials completed at higher walking speeds. One can only speculate about the cause of this relationship. It is possible that at higher walking speeds the *habF* had a tendency to 'bottom out' during loading of the heel-heel i.e. period of weight acceptance, and this then affected (delayed) how weight was transferred onto the fore-foot heel as the centre of mass progressed forwards over the foot. Given that walking speed was found to be unrelated to any of the centre of pressure measures when using the hydraulic device, this suggests the findings, indicating use of the *hyA-F* attenuated centre of pressure trajectory fluctuations, resulted from the functioning of the foot itself rather than simply being due to an increase in customary walking speed when using it.

Irrespective of prosthetic ankle-foot type, the mean forwards angular velocity of the prosthetic shank was significantly higher during double support (weight transfer onto the prosthetic limb) compared to single support, and velocities during double support became significantly increased when using the *hyA-F*. This may have contributed to the significant increase in overall walking speed when using the *hyA-F*. A systematic review of the variables used in amputee gait research highlighted that self-selected comfortable walking speed was the most often reported parameter (Sagawa Jr. et al., 2011), which reflects the importance of walking speed as a measure of overall gait quality. Increasing walking speed has been found to decrease temporal asymmetries

in amputee gait (Nolan et al., 2003) and is positively correlated with amputees' self-perception of gait quality (Miller et al., 2001). A previous study that compared use of ESR and non ESR feet with and without an elastic ankle articulation device (Zmitrewicz et al., 2006) found no difference in walking speed across foot and ankle device conditions. This suggests that ankle articulation alone does not result in an increase in walking speed. It has recently been demonstrated that use of the same type of hydraulic ankle-foot device as used in the present study led to a reduction in in-socket pressure in trans-tibial amputees during ambulation at self-selected speeds (Portnoy et al., 2012). This suggests that comfort might be increased when using the *hyA-F* and it may also be this increased comfort which facilitates higher walking speeds. Although the absolute mean increase of walking speed may appear small ($\sim 0.05 \text{ ms}^{-1}$) it was an increase of 4.5 % and of a medium effect size. Also, walking speed increased consistently (increases were observed in all but one participant) thus these increases can be described as being clinically meaningful.

Five of the 20 participants used a Multiflex (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) device which allows elastically controlled articulation at the 'ankle' attachment. Indeed, the Multiflex foot-ankle provides up to 15° of sagittal plane articulation which is more than the *hyA-F* does. This suggests that the observed effects of using the *hyA-F* were due to the hydraulically dampened nature of the articulation rather than solely the magnitude of the articulation provided. The *hyA-F* provides passive damped, and therefore time-dependent, resistance which slows and thus temporally

extends the period during which articulation occurs compared to elastically controlled devices. With elastically controlled devices, such as a Multiflex, articulation is permitted via deformation at the point of attachment (e.g. by use of a rubber snubber). The rate of articulation is governed by the stiffness of the snubber and is, by and large, time-independent. These devices will reach their limit of articulation very quickly when loaded at which point they will act more like a rigid device. This would explain why there were no significant differences between Esprit users and non-Esprit (predominantly Multiflex) users.

4.5 Conclusions

Use of the hydraulic ankle-foot device reduced or eliminated the backwards directed centre of pressure displacement, reduced centre of pressure velocity fluctuations beneath the prosthetic foot and allowed the prosthetic shank to rotate over the foot quicker during double support. These changes were associated with an increase in self-selected customary walking speed. This suggests that such a device may be functionally beneficial for active amputees. In addition, this study has highlighted that among other measures, which aim to quantify comparative performance of prosthetic feet, the examination of the centre of pressure progression beneath the prosthetic foot is a useful tool.

Chapter 5. How are centre-of-pressure trajectory fluctuations beneath the prosthetic foot (i.e. 'roll-over' dynamics) affected by sagittal plane misalignments, and are the effects of misalignment reduced by using a hydraulic ankle-foot device?

5.1 Introduction

In the previous experimental chapter (Chapter 4) the results revealed that, as previously reported (Ranu, 1988; Schmidt et *al.*, 2005) the centre-of-pressure trajectory beneath the prosthetic foot dwells beneath the hind-foot and is also subject to inappropriate, negative displacements. The spatial and temporal locations of these inappropriate trajectory fluctuations were varied across participants, however they were consistent within each participant regardless of which prosthetic ankle-foot device they used. In view of this, this second study was conducted as a single-subject analysis.

Beneath the foot, the centre-of-pressure represents the origin of ground reaction forces which act to provide support and propulsion (De Cock et *al.*, 2008). Throughout stance, the centre-of-pressure typically progresses along the plantar surface of the foot from the heel forwards to the toes. Its velocity is higher during early and late stance as the toe and then heel region is respectively lowered to, and raised from, the floor, and slower during mid-stance when the foot is plantigrade (De Asha et *al.*, 2013, Chapter 4). Such movements of the centre-of-pressure reflect how the forward progression of the whole-body centre of mass is controlled (Schmid et *al.*, 2005; Kirtley, 2006) with centre of mass velocity higher during the transfers of weight onto and off the stance limb (double support) than during the period of single limb support.

In able-bodied gait the ankle, and associated musculature, facilitates the lowering of the foot to the floor, the shank rotating as the centre of mass translates above the foot, and the foot 'pushing-off' (Schmid et al., 2005; Winter, 2009). The principle limitation of passive lower-limb prostheses is their inability to replicate, fully, intact-ankle function. Thus, in unilateral lower limb amputees, the centre-of-pressure remaining beneath the prosthetic hind-foot significantly longer than beneath the hind-foot of the intact foot is suggested to be attributable to a delay, compared to an intact limb, between the heel contacting the floor and the foot becoming plantigrade (Ranu, 1988; Schmid et al., 2005). Anecdotal perceptions of a 'dead spot' during prosthetic limb stance are a common feature of UTA gait and reflect interruptions in the forwards progression of the centre-of-pressure which in turn reflect how bodyweight is transferred over the prosthetic limb (Ranu, 1988; Schmid et al., 2005; De Asha et al., 2013, Chapter 4).

The function of a prosthetic device is affected by its alignment (Hannah et al., 1984). Previous studies have investigated how alignment changes alter a number of gait variables such as temporal and spatial gait features (Fridman et al., 2003), proximal joint kinetics (Blumentritt et al., 1999; Schmalz et al., 2002), gait symmetry (Chow et al., 2006), moments at the distal end of the socket (Boone et al., 2012) and in-socket pressures (Parker et al., 1999; Seelen et al., 2003). One previous study investigated plantar pressure changes in UTAs during alignment optimization (Geil & Lay, 2004), however due to the study's clinical setting the changes in alignment could not be

systematically controlled but were simply those observed during the set-up and alignment process conducted by a prosthetist.

The purpose of this experiment, therefore, was to investigate whether systematically altering alignment of the prosthetic foot in the sagittal plane affected the progression of the centre-of-pressure beneath the foot during UTA overground gait and, if so, whether the use of different types of articulating or non-articulating foot attachments ameliorate such effects.

5.2 Methods

5.2.1 Participant

One physically active, male UTA (age 35 years, mass 79 kg, height 1.83 m, 9 years since amputation) took part. He habitually used an Elite VT foot (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) which he had used for two years and also frequently used other prosthetic devices during activities such as running and swimming. This meant he had regular experience of accommodation between different prosthetic devices. He had used his current, full contact, socket for nine months and wore a Tech Polyurethane liner.

5.2.2 Experimental protocol and data acquisition

During the study the participant used a rigidly attached foot device (*rigF*, Esprit) and two articulating ankle-foot devices, the articulation of which was either multiaxial and controlled elastically (*e/A-F*, Epirus) or hydraulically controlled uniaxial (*hyA-F*, Echelon). These devices all share the same manufacturer (Chas. A. Blatchford and Sons Ltd., Basingstoke, UK) and were chosen as they are identical in design other than their attachments to the shank pylon. The Esprit device was attached rigidly to the shank pylon, allowing no mechanical articulation. The Epirus device articulated around a rubber snubber at the attachment between foot and pylon and therefore provided multi-axial articulation. The Echelon device, as described previously (Chapter 2.3.3), provided hydraulically damped sagittal plane articulation.

Kinematic and kinetic data were recorded as per 'general methods, Chapter 3, while walking trials were completed at the participant's customary walking speed in three sessions; each on a separate day. In each session data were collected for one of the three attachment types (i.e. separate session for each attachment condition). The participant was instructed to walk "as he would normally". In each session trials were conducted in five blocks. One block was conducted with the prosthesis in its 'optimal' alignment which was decided upon using a mixture of participant feed-back regarding perceived comfort and function and the prosthetist's experience. For each of the other four blocks the prosthesis was intentionally misaligned in one of the following ways; tilting the shank pylon forwards (toe-up) or backwards (toe-down) 6° in

the sagittal plane or shifting the foot forwards or backwards 6 mm. Block order was randomised. All alterations were made using a bespoke adaptor beneath the pyramid at the distal end of the socket. Following each alignment change the participant used the device, for a minimum of five minutes, until he felt comfortable prior to data being recorded. While the participant was aware of adjustments being made to the prosthesis he was blinded to the nature of each change.

During data collection, the participant wore retro-reflective markers as described in 'general methods', Chapter 3.

5.2.3 Data processing

All data were reduced and processed within Workstation (Vicon, Oxford, UK) and Visual 3D (C Motion, Germantown, MD, USA) as per 'general methods', Chapter 3. The area beneath the prosthetic foot was divided in to hind-foot (posterior from the markers at the midline of the shank pylon), mid-foot (shank markers to the metatarsal markers) and forefoot (anterior from the metatarsal markers).

5.2.4 Data analysis

In the previous chapter (Chapter 4) the outcome variable were chosen *a priori*. Following that experiment parameters which were different between foot types were known and therefore they were selected as the outcome variable for this experiment. Therefore the following parameters were

determined for each of the three areas of the foot: the time (normalised to stance phase) the centre-of-pressure was beneath that area of the foot (the instant of centre-of-pressure moving from the hind-foot to the mid-foot coincided with it moving anterior to the base of the shank pylon); peak negative (or minimum, if values remained positive) antero-posterior (A-P) centre-of-pressure velocity; negative A-P centre-of-pressure displacement; variability in A-P centre-of-pressure velocity. To determine negative centre-of-pressure displacement, the displacements occurring between each frame were first calculated and then negative values summed to give the total distance travelled by the centre-of-pressure in the opposite direction to the direction of travel. Centre-of-pressure velocity variability was determined as the standard deviation in centre-of-pressure velocity. These parameters were calculated for each individual trial and then averaged across trials to give a mean value for each alignment condition for each of the three attachment types used.

5.2.5 Statistical analyses

Due to the single-subject design of the study no statistical comparisons were made. Effect size differences were calculated as Cohen's 'd' (Cohen, 1977).

5.3 Results

For each attachment type there were four misalignment conditions and four variables measured which totalled 16 result values beneath each area of the prosthetic ankle-foot devices. The 'in text' results focus on differences in the effect sizes of changes due to alignment between attachment types and also on changes to inappropriate, negative centre-of-pressure displacements due to alignments. All results beneath the hind-foot, mid-foot and forefoot for each of the attachment types are listed separately, and in full, in the results tables.

5.3.1 Effect size

The mean effect sizes for changes due to mis-alignment for each attachment type were 1.44 ± 2.25 , *rigF*; 0.85 ± 1.91 , *e/A-F* and 0.88 ± 0.87 , *hyA-F*. The mean effect sizes for changes due to mis-alignment for each area beneath the prosthetic foot were 1.33 ± 2.59 , hind-foot; 1.36 ± 0.98 , mid-foot and 0.38 ± 0.44 , forefoot.

5.3.2 Hind-foot

When the *rigF* was used the mean effect size for changes due to mis-alignment was 1.99 ± 3.26 (range; 13.9 – 0.4). When the *e/A-F* was used the mean effect size for changes due to mis-alignment was 1.20 ± 2.99 (range; 12.3 – 0.1). When the *hyA-F* was used the mean effect size for changes due to mis-alignment was 0.78 ± 1.6 (range; 6.0 – 0.3).

5.3.3 Mid-foot

Full results are shown in Table 4. When the *rigF* was used the mean effect size for changes due to mis-alignment was 1.81 ± 1.19 (range; 2.8 – 0.3). When the *e/A-F* was used the mean effect size for changes due to mis-alignment was 0.86 ± 0.42 (range; 1.3 – 0.2). When the *hyA-F* was used the mean effect size for changes due to mis-alignment was 1.43 ± 0.99 (range; 3.2 – 0.2).

5.3.4 Forefoot

Full results are shown in Table 5. When the *rigF* was used the mean effect size for changes due to mis-alignment was 0.33 ± 0.35 (range; 1.0 – 0.0). When the *e/A-F* was used the mean effect size for changes due to mis-alignment was 0.36 ± 0.39 (range; 0.9 – 0.0). When the *hyA-F* was used the mean effect size for changes due to mis-alignment was 0.46 ± 1.6 (range; 1.8 – 0.0).

5.3.5 Negative centre-of-pressure displacement

Regardless of attachment device or alignment condition the majority of negative centre-of-pressure displacement occurred beneath the prosthetic hind-foot with a small amount occurring beneath the mid-foot and none beneath the forefoot. In all but one condition (*rigF*, posterior tilt) misalignment caused an increase in negative centre-of-pressure displacement compared to

the optimally aligned condition beneath the hind-foot. Conversely all misalignment conditions caused a significant reduction in negative centre-of-pressure displacement beneath the mid-foot across all attachment types. There was no negative displacement of the centre of pressure beneath the fore-foot across all devices and alignment conditions. Centre-of-pressure velocity in each condition is shown at Figure 11, 14 & 15)

Table 3. Mean (SD) differences in centre-of-pressure variables due to misalignment beneath the hind-foot area of the prosthetic ankle-foot devices. Effect sizes 'd' are shown in parentheses in superscript.

	Attachment type	Negative CoP displacement (cm)	Minimum CoP velocity (m/s)	CoP velocity variability (m/s)	% stance phase
Optimal	<i>rigF</i>	-1.7 (0.4)	-2.06 (0.56)	0.44 (0.09)	25.9 (0.9)
	<i>eIA-F</i>	-2.4 (0.5)	-3.04 (0.59)	0.74 (0.16)	24.3 (1.2)
	<i>hyA-F</i>	-1.2 (0.6)	-1.14 (0.24)	0.56 (0.11)	24.0 (1.1)
Anterior Shift	<i>rigF</i>	-2.8 (0.4) ^(2.0)	-2.99 (0.52) ^(1.3)	0.56 (0.08) ^(1.0)	37.2 (4.0) ^(2.9)
	<i>eIA-F</i>	-2.5 (0.5) ^(0.1)	-2.98 (0.62) ^(0.1)	0.69 (0.13) ^(0.3)	23.8 (1.2) ^(0.3)
	<i>hyA-F</i>	-1.8 (0.3) ^(0.9)	-1.42 (0.24) ^(0.9)	0.53 (0.10) ^(0.2)	27.9 (2.3) ^(0.5)
Posterior Shift	<i>rigF</i>	-2.6 (0.3) ^(1.9)	-2.77 (0.61) ^(0.9)	0.56 (0.08) ^(1.1)	30.8 (2.5) ^(1.9)
	<i>eIA-F</i>	-2.6 (0.6) ^(0.3)	-2.99 (0.82) ⁽⁰⁾	0.58 (0.10) ^(0.9)	31.4 (6.3) ^(0.5)
	<i>hyA-F</i>	-2.0 (0.6) ^(0.9)	-1.54 (0.63) ^(0.6)	0.51 (0.15) ^(0.3)	30.0 (2.8) ^(0.8)
Anterior Tilt	<i>rigF</i>	-2.7 (0.4) ^(1.8)	-2.90 (0.36) ^(1.3)	0.43 (0.04) ^(0.1)	56.5 (2.1) ^(13.9)
	<i>eIA-F</i>	-3.0 (0.4) ^(1.0)	-3.38 (0.76) ^(0.4)	0.47 (0.07) ^(1.6)	60.8 (2.8) ^(12.3)
	<i>hyA-F</i>	-2.2 (0.4) ^(1.4)	-1.77 (0.48) ^(0.4)	0.41 (0.08) ^(1.1)	48.3 (2.3) ^(3.4)
Posterior Tilt	<i>rigF</i>	-1.4 (0.7) ^(0.4)	-1.52 (0.88) ^(0.5)	0.41 (0.15) ^(0.2)	24.5 (2.0) ^(0.7)
	<i>eIA-F</i>	-2.9 (0.9) ^(0.5)	-3.16 (1.08) ^(0.1)	0.75 (0.29) ⁽⁰⁾	26.2 (2.8) ^(0.7)
	<i>hyA-F</i>	-2.3 (0.5) ^(1.4)	-1.91 (0.54) ^(0.2)	0.66 (0.19) ^(0.5)	24.9 (1.9) ^(0.4)

Table 4. Mean (SD) differences in centre-of-pressure variables due to misalignment beneath the mid-foot area of the prosthetic ankle-foot devices. Effect sizes 'd' are shown in parentheses in superscript.

	Attachment type	Negative CoP displacement (cm)	Minimum CoP velocity (m/s)	CoP velocity variability (m/s)	% stance phase
Optimal	<i>rigF</i>	-0.4 (0.1)	-0.15 (0.03)	0.28 (0.02)	65.2 (1.7)
	<i>eIA-F</i>	-0.3 (0.2)	-0.12 (0.19)	0.19 (0.04)	67.7 (3.4)
	<i>hyA-F</i>	-0.2 (0.3)	-0.09 (0.04)	0.26 (0.04)	64.3 (7.3)
Anterior Shift	<i>rigF</i>	-0.02 (0.03) ^(2.8)	-0.05 (0.08) ^(1.3)	0.22 (0.03) ^(1.8)	53.8 (5.7)
	<i>eIA-F</i>	-0.01 (0.01) ^(1.2)	-0.03 (0.04) ^(0.5)	0.14 (0.02) ^(1.3)	67.4 (1.0)
	<i>hyA-F</i>	-0.02 (0.03) ^(0.6)	-0.02 (0.05) ^(1.0)	0.18 (0.03) ^(1.6)	62.6 (2.2)
Posterior Shift	<i>rigF</i>	-0.1 (0.1) ^(1.6)	-0.13 (0.13) ^(0.2)	0.26 (0.03) ^(0.6)	62.6 (5.3)
	<i>eIA-F</i>	-0.02 (0.02) ^(1.1)	-0.07 (0.09) ^(0.2)	0.18 (0.03) ^(0.2)	62.7 (4.7)
	<i>hyA-F</i>	-0.003 (0.005) ^(0.6)	0.01 (0.02) ^{** (1.6)}	0.16 (0.03) ^(2.5)	59.1 (2.2)
Anterior Tilt	<i>rigF</i>	-0.007 (0.0051) ^(3.1)	0.08 (0.08) ^(2.9)	0.33 (0.07) ^(0.7)	34.1 (2.4)
	<i>eIA-F</i>	0.0 ^(1.3)	0.14 (0.08) ^(1.3)	0.24 (0.06) ^(0.7)	31.5 (3.1)
	<i>hyA-F</i>	0.0 ^(0.6)	0.10 (0.04) ^(3.2)	0.15 (0.03) ^(2.7)	42.9 (2.3)
Posterior Tilt	<i>rigF</i>	-0.01 (0.02) ^(3.0)	-0.02 (0.03) ^(3.4)	0.29 (0.03) ^(0.3)	49.3 (7.4)
	<i>eIA-F</i>	-0.02 (0.04) ^(1.1)	0.00 (0.08) ^(0.6)	0.23 (0.04) ^(0.8)	49.2 (9.9)
	<i>hyA-F</i>	-0.007 (0.001) ^(0.6)	0.03 (0.04) ^(2.0)	0.25 (0.03) ^(0.2)	52.3 (3.1)

Table 5. Mean (SD) differences in centre-of-pressure variables due to misalignment beneath the forefoot area of the prosthetic ankle-foot devices. Effect sizes 'd' are shown in parentheses in superscript.

	Attachment type	Negative CoP displacement (cm)	Minimum CoP velocity (m/s)	CoP velocity variability (m/s)	% stance phase
Optimal	<i>rigF</i>	0.0	0.34 (0.21)	0.56 (0.06)	8.9 (1.3)
	<i>e/A-F</i>	0.0	0.28 (0.21)	0.70 (0.16)	8.0 (2.9)
	<i>hyA-F</i>	0.0	0.15 (0.09)	0.57 (0.08)	11.6 (1.4)
Anterior Shift	<i>rigF</i>	0.0	0.31 (0.19) ^(0.1)	0.63 (0.14) ^(0.5)	9.0 (2.7)
	<i>e/A-F</i>	0.0	0.37 (0.28) ^(0.3)	0.49 (0.09) ^(1.2)	8.7 (1.2)
	<i>hyA-F</i>	0.0	0.13 (0.10) ^(0.1)	0.59 (0.16) ^(0.1)	9.4 (0.9)
Posterior Shift	<i>rigF</i>	0.0	0.55 (0.70) ^(0.3)	1.05 (0.73) ^(0.7)	6.7 (3.4)
	<i>e/A-F</i>	0.0	0.36 (0.25) ^(0.3)	0.71 (0.03) ^(0.1)	5.9 (2.8)
	<i>hyA-F</i>	0.0	0.23 (0.12) ^(0.5)	0.43 (0.12) ^(1.0)	10.9 (2.4)
Anterior Tilt	<i>rigF</i>	0.0	0.12 (0.09) ^(1.0)	0.58 (0.18) ^(0.1)	9.3 (1.2)
	<i>e/A-F</i>	0.0	0.29 (0.23) ⁽⁰⁾	0.63 (0.13) ^(0.4)	7.7 (0.9)
	<i>hyA-F</i>	0.0	0.41 (0.22) ^(1.1)	0.51 (0.07) ^(0.5)	8.7 (1.1)
Posterior Tilt	<i>rigF</i>	0.0	0.16 (0.12) ^(0.7)	0.50 (0.09) ^(0.6)	26.2 (7.0)
	<i>e/A-F</i>	0.0	0.09 (0.07) ^(0.9)	0.57 (0.09) ^(0.7)	24.6 (9.3)
	<i>hyA-F</i>	0.0	0.10 (0.06) ^(0.4)	0.42 (0.03) ^(1.8)	22.7 (4.3)

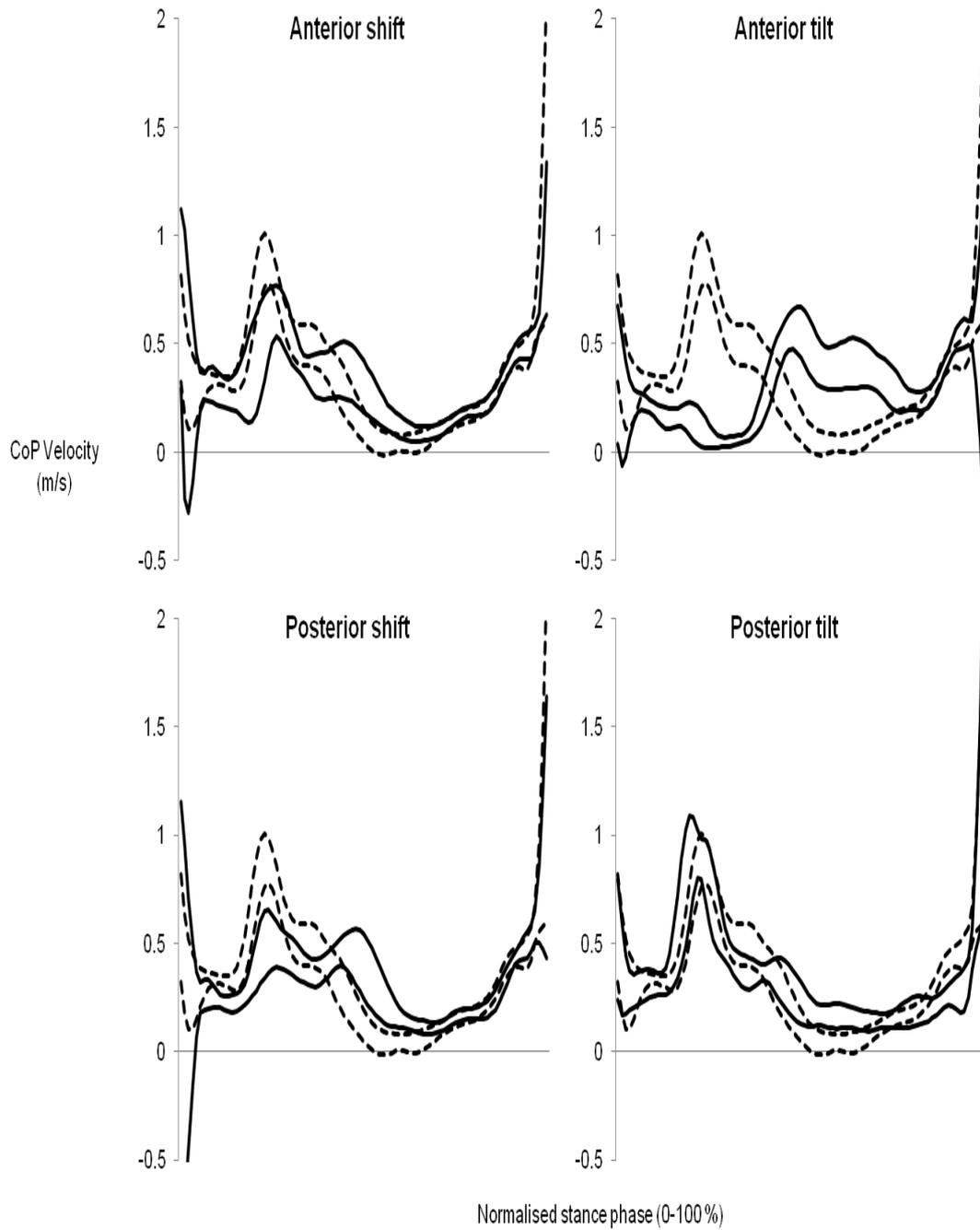


Figure 11. ± 1 SD ribbons of CoP A/P velocity normalised to stance phase in each alignment condition (solid lines) and optimally aligned (dashed lines) when using a *hyA-F*.

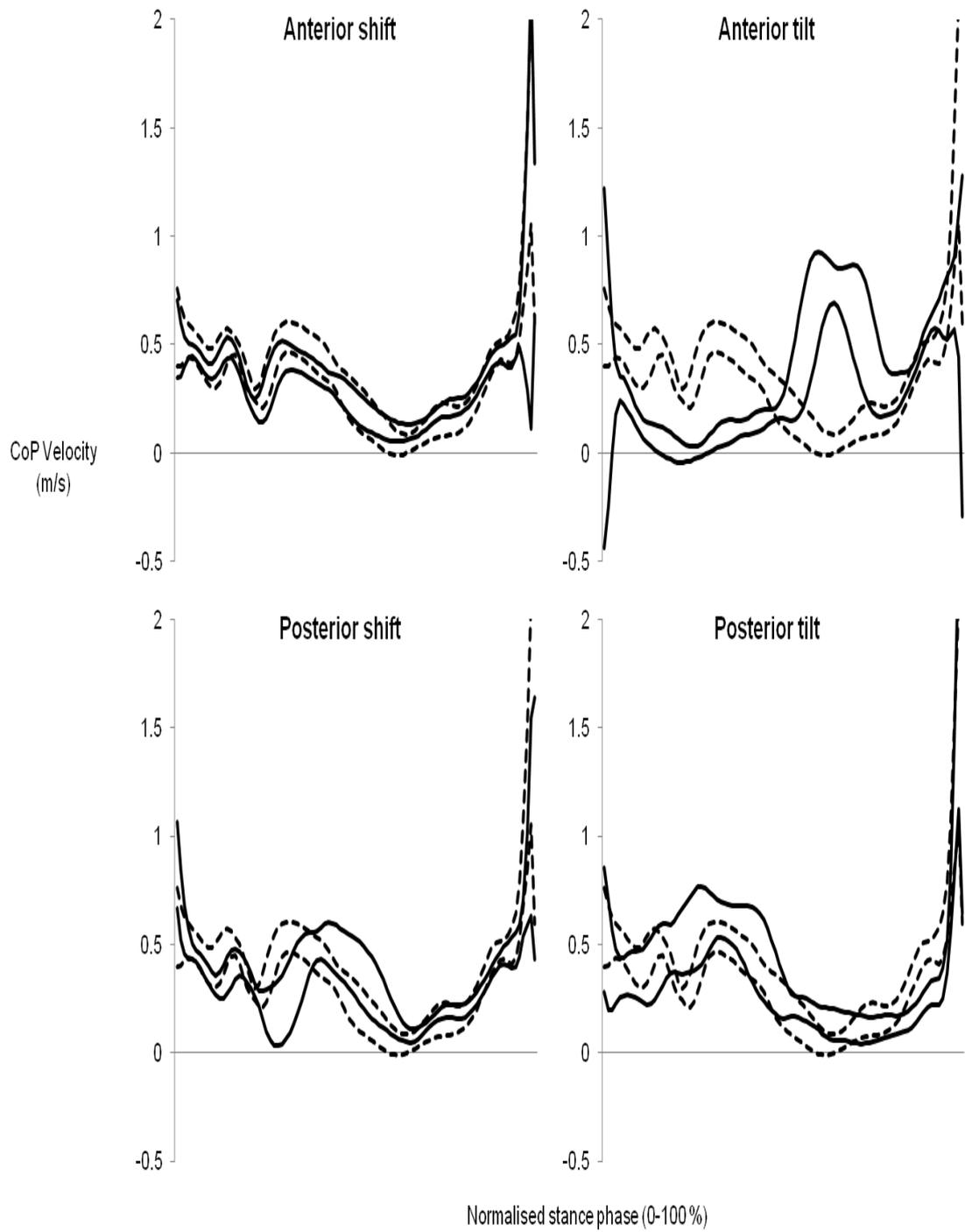


Figure 12. ± 1 SD ribbons of CoP A/P velocity normalised to stance phase in each alignment condition (solid lines) and optimally aligned (dashed lines) when using a eIA-F.

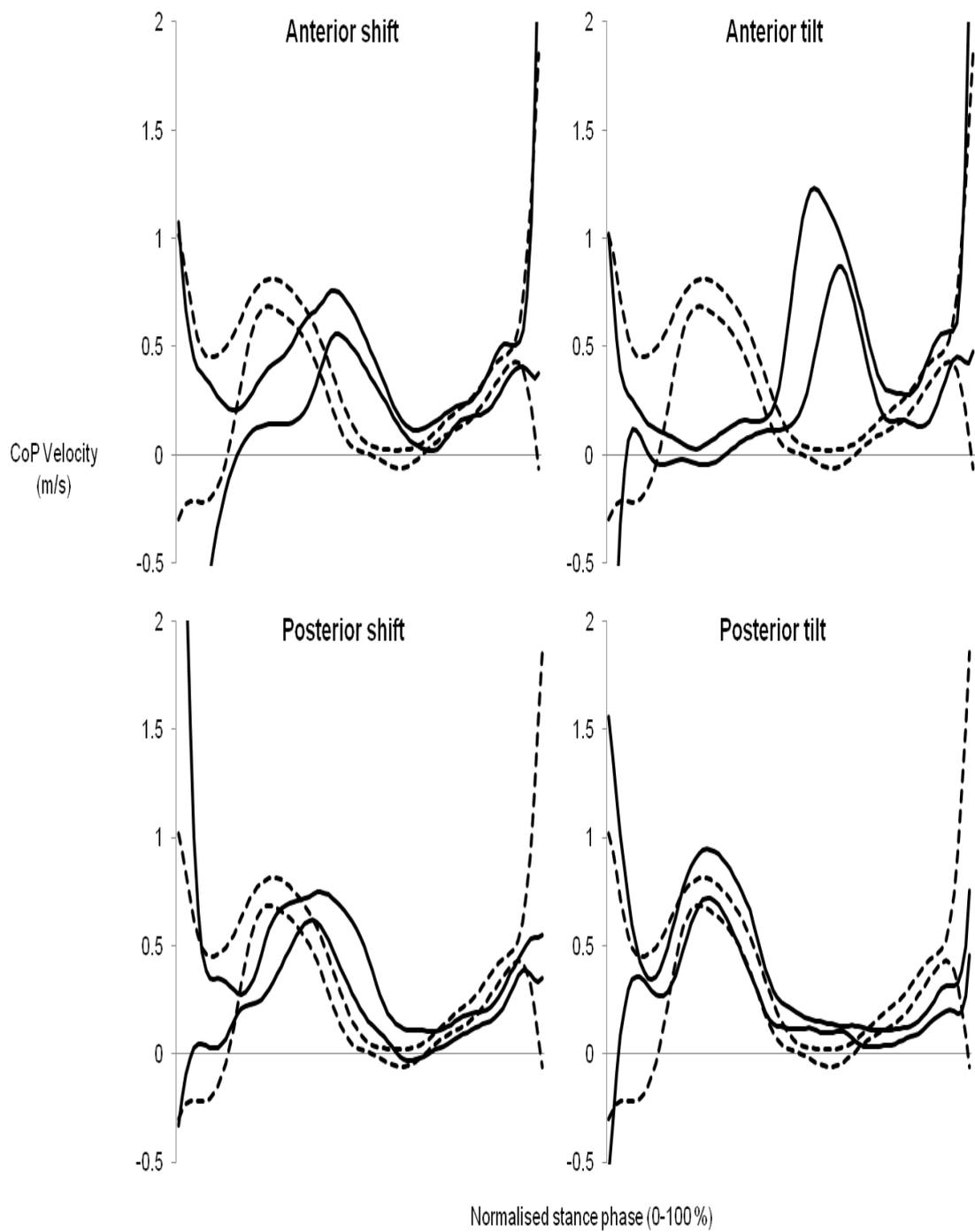


Figure 13. ± 1 SD ribbons of CoP A/P velocity normalised to stance phase in each alignment condition (solid lines) and optimally aligned (dashed lines) when using a *rigF*.

5.4 Discussion

The purpose of the current study was to investigate if and how systematically altering alignment of the prosthetic foot in the sagittal plane affected the progression of the centre-of-pressure beneath the foot during overground gait and whether the use of different types of foot attachments could ameliorate such effects. Unsurprisingly, given the reports of numerous more proximal effects of prosthetic mis-alignment, the misalignments during the current study did have profound effects on centre-of-pressure progression. The effect sizes suggested that when the foot device was attached rigidly it was more sensitive to mis-alignment than when the attachment device allowed only hydraulically controlled sagittal plane articulation or elastically controlled multi-axial articulation.

It was demonstrated previously (De Asha et al., 2013, Chapter 4) that inappropriate fluctuations of, and disruptions to, centre-of-pressure trajectory vary, both temporally and spatially, between amputees but tend to be consistent within individuals across prosthetic devices thus a single-subject analysis was considered appropriate. The participant in this study was a highly active (K4) individual. While one should draw no statistical inferences from the results of this study into a general amputee population regarding the location and magnitudes of inappropriate centre-of-pressure trajectory fluctuations it is reasonable to state that the current study demonstrates sagittal mis-alignments do influence centre-of-pressure progression and perhaps also to suggest that less active individuals could potentially be more

affected, albeit perhaps differently depending on individual gait characteristics, by such mis-alignments.

When the prosthetic devices were optimally aligned the participant primarily experienced disruption to the centre-of-pressure progression beneath the hind-foot which manifest as a period of negatively directed travel of the centre-of-pressure and as a delay in the centre-of-pressure passing anterior to the 'ankle' when compared to an intact limb similar to that observed previously (Ranu, 1988; Schmid et al., 2005; De Asha et al., 2013, Chapter 4). This passing of the centre-of-pressure anterior to the 'ankle' has been reported to occur at approximately 30% of stance on the prosthetic limb for unilateral trans-tibial amputees (Schmalz et al., 2002; Ventura et al., 2011; De Asha et al., 2013, Chapter 4) compared to approximately 9% of stance for an intact limb (Kirtley, 2006). For the participant in the current study this occurred at approximately 25% of stance across all attachment types. This most likely reflects the higher activity level of this particular individual but also demonstrates that despite this higher activity level the hind-foot stalling and negative displacement of the centre-of-pressure is an issue due to the function of passive prosthetic devices *per se*. All the mis-alignment conditions which resulted in changes to the timing of this event made this delay longer. The previously suggested reason for this delay is the longer time between the heel contacting the floor and the foot becoming plantigrade for a prosthetic foot compared to an intact foot (Ranu, 1988; Schmid et al., 2005). The posterior tilt mis-alignment, which would be expected to result in the foot becoming plantigrade sooner, had no effect on the timing of the centre-of-

pressure passing anterior of the 'ankle' for the *rigF* or *hyA-F* and delayed the event for the *e/A-F*. The cause of this is unclear but these findings suggest the reason behind this phenomenon of delayed hind-foot centre-of-pressure progression is more complicated than simply the time taken for the prosthetic foot to lower to the floor. It may, perhaps, be significant that the device with the multiaxial component to articulation between the shank and foot was the one adversely affected by this particular mis-alignment and which possibly suggests that this device was more sensitive to any non-ideal alignment than the other two.

Similarly, when the devices were optimally aligned, negative centre-of-pressure displacement occurred primarily beneath the prosthetic hind-foot. As reported previously this was reduced when using the *hyA-F* compared to the other attachment types (De Asha et al., 2013, Chapter 4). Interestingly this negative displacement was highest when using the *e/A-F* which could suggest that multiaxial articulation may exacerbate negative centre-of-pressure displacements beneath the hind-foot. This inappropriate, negative displacement was increased by misalignment across all attachment types except in the posterior tilt condition when there was a small reduction when using the *rigF*.

Although no statistical analysis of the centre-of-pressure A-P velocity across the full stance phase was made a qualitative assessment of FIG N supports the effect size comparison in that sagittal mis-alignment predominantly affected centre-of-pressure velocity and progression beneath the hind- and

mid-foot. It also corroborates the previous finding that, while the magnitudes may vary, the temporal and spatial locations of disruptions to centre-of-pressure progression remain constant within individuals across prosthetic devices (De Asha et al., 2013, Chapter 4). In addition these data also suggest that this intra-subject consistency extends across alignment conditions.

This study is possibly limited due to its single-subject design and consequent lack of inferential statistical power however it does succeed in demonstrating that centre-of-pressure progression is profoundly affected by sagittal plane misalignment of a prosthetic device. The single-subject design was, however, chosen to prevent aggregation errors which may have occurred due to disruptions to centre-of-pressure trajectories occurring at different temporal and spatial locations between participants. While the majority of previous investigations into the effects of prosthetic (mis)alignment have tended to focus on more proximal outcome measures it would be appropriate, given the relationship between centre-of-pressure and whole body centre of mass progression, for future work to investigate more fully this distal manifestation of changes in prosthetic device function due to alignment. Simple measures such as negative centre-of-pressure displacements or the time, as a percentage of stance, the centre-of-pressure is beneath each part of the prosthetic foot could potentially be used as tools to assist a prosthetist when conducting an initial alignment of a device or even, given the differences between devices reaction to misalignment for one participant, inform any decision into which device may be appropriate for the individual user.

5.5 Conclusions

This experiment demonstrated that centre-of-pressure progression was adversely affected by sagittal mis-alignment regardless of attachment type, although the magnitude of effects did appear to differ between attachment devices. The temporal and spatial locations of inappropriate fluctuations of, or disruptions to, the centre-of-pressure trajectory were consistent across prosthetic attachment types and alignment conditions supporting the suggestion that while centre-of-pressure disruption is associated with most, if not all, prosthetics the location of such disruptions is influenced by individual walking styles.

**Chapter 6. Minimum toe
ground separation during
unilateral trans-tibial gait:
event detection and
comparison of clearance
values between limbs**

6.1 Introduction

The first experimental chapter (Chapter 4) demonstrated that use of the *hyA-F* device resulted in increased customary walking speeds compared to use of the participants' *habF*. It describes increased angular velocity of the prosthetic shank and reduced disruption of the centre-of-pressure trajectory during prosthetic limb stance using the *hyA-F* compared to the participants' *habF*. This chapter describes a simple to implement method of temporally indentifying the instant of minimum toe ground separation (MTGS) of the swing foot during ambulation. This technique was developed initially to implement on data not included as part of this thesis but can be found in Johnson et *al.*, 2013. The technique described below is also used in the following chapter (Chapter 7). For the purpose of this and the following chapter MTGS is defined as the local minimum in separation between the ground and the toe of the forwards moving foot during the swing phase. The risk of tripping, the predominant cause of falls during ambulation (Blake et *al.*, 1988), is highest at the point of MTGS (Mills & Barrett, 2001). In practice MTGS is a difficult event to determine as part of an automated event detection process within motion analysis software. This difficulty is because at the beginning of swing, toe-off, the toe-ground separation (TGS) is minimal and thus, when searching for MTGS, this point, at or just after toe off tends to be incorrectly identified. To ensure MTGS is correctly identified when using an automatic event detection process one could narrow the 'search window' so that it identifies an MTGS event after a temporal delay from the instant of toe-off. However the magnitude of such an off-set would be affected by

factors such as walking speed. It would, therefore, involve subjective decisions being made for each individual participant and/or each individual trial. An alternative approach is to define the local minima event manually, however both approaches are time consuming and laborious.

In able-bodied gait the instant of peak forwards velocity of the swing-foot (PFV) has been reported to coincide with MTGS (Winter, 1992) although empirical data to support this assertion were not presented. Numerous published studies allude to this previous study (e.g. Mills & Barrett, 2001; Begg et al., 2007; Nagano et al., 2011) but as with the original study, they do not present supporting data. No previous studies have investigated the relationship between PFV and MTGS in UTAs.

UTAs have been shown to have a higher risk of falls than age-matched, able-bodied controls (Kulkarni et al., 1996; Miller et al., 2001), have lower MTGS on the prosthetic compared to intact side (Gates et al., 2012; Wuderman et al., 2012) and exhibit increased MTGS variability, which is likely to increase the risk of trips, on both the intact- and prosthetic-limb (Wuderman et al., 2012). UTAs have altered gait kinematics and kinetics due to the mechanical constraints imposed on them by their prostheses (Sanderson & Martin, 1997; Nolan & Lees, 2000; Prinsen et al., 2011). These constraints result in reduced walking speeds and increased asymmetry compared to able-bodied individuals (Nolan et al., 2003) therefore the assumed synchronicity between PFV and MTGS, reported in able-bodied gait, may not occur in UTAs.

The aim of the present study was to determine whether PFV was synchronous with MTGS in UTAs during overground ambulation across a range of walking speeds for both the intact and prosthetic limbs and to compare the values of TGS recorded at each event in order to assess whether PFV was appropriate to use as a temporal marker of MTGS.

6.2 Methods

6.2.1 Participants

Ten physically active UTAs (mean \pm SD age; 48 ± 11.7 years, mass; 86 ± 17.7 kg, height; 1.78 ± 0.06 m) took part. All participants habitually used the same prosthetic ankle-foot device - an Esprit foot (*rigF*, Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). Full details are given in Table 6.

6.2.2 Experimental protocol and data acquisition

Kinematic and kinetic data were recorded as per 'general methods', Chapter 3 while participants walked in a straight line along a flat and level 8 m walkway at three different speed levels: customary, 'slow' and 'fast'. Participants were instructed to walk "at their normal walking speed", "slowly" and "as fast as comfortably possible". Participants completed trials at each speed until 20 clean contacts had been made by both the intact and prosthetic feet with either of the force platforms (20 trials x 3 speeds x 2 limbs = 120 PFV events).

During data collection, participants wore retro-reflective markers as described in 'general methods', Chapter 3.

Table 6. Descriptive details of UTA participants.

PARTICIPANT I.D. NUMBER	AGE (YEARS)	MASS (KG)	HEIGHT (M)	SEX	SIDE OF AMPUTATION	TIME SINCE AMPUTATION (YEARS)	HABITUAL FOOT DEVICE
S 01	34	72	1.72	M	RIGHT	8	ESPRIT
S 02	62	82	1.76	M	LEFT	19	ESPRIT
S 03	39	87	1.71	F	LEFT	9	ESPRIT
S 04	38	62	1.80	M	RIGHT	5	ESPRIT
S 07	61	123	1.75	M	RIGHT	3	ESPRIT
S 08	63	93	1.85	M	RIGHT	43	ESPRIT
S 09	59	100	1.78	M	RIGHT	11	ESPRIT
S 10	39	66	1.78	M	RIGHT	6	ESPRIT
S 11	45	84	1.75	M	RIGHT	2	ESPRIT
S 12	40	91	1.89	M	LEFT	2	ESPRIT

6.2.3 Data processing

All data were reduced and processed within Workstation (Vicon, Oxford, UK) and Visual 3D (C Motion, Germantown, MD, USA) as per 'general methods', Chapter 3. Due to equipment failure there were no kinetic data recorded for two participants therefore for these initial contact and toe off were defined

using kinematic data: initial contact was defined as the instant of contralateral limb peak hip extension (De Asha et al., 2012) and toe off as the instant of peak posterior displacement of the ipsilateral toe marker relative to the pelvis (Zeni Jr. et al., 2008).

6.2.4 Data analysis

The following parameters were determined: The instants of intact- and prosthetic-foot PFV; the instants of intact- and prosthetic-foot MTGS; TGS at the instants of PFV and MTGS. Swing was defined as being from the instant of toe off until ipsilateral initial contact. PFV was defined as the maximal forwards velocity, in the direction of travel, of the foot-segment centre of mass during swing; and was determined automatically within Visual 3D. TGS was defined as being the vertical component of the shoe-tip trajectory. MTGS was defined as the local minimum of the vertical component of the shoe-tip trajectory during mid swing; and was determined manually by examining the shoe-tip trajectory of each trial (see Figure 14). This 'manual' approach was used to ensure the local minima in toe-ground clearance occurring at or just after toe off would not be identified in error; which would have been the case if MTGS had been determined automatically. Toe-ground clearance values at PFV and MTC were determined as the height of shoe-tip above the ground at each event. These parameters were calculated for each individual trial and then averaged across trials to give a mean value for each foot condition per participant.

6.2.5 Statistical analyses

A “Limits of Agreement” (LOA) analysis (Bland & Altman, 1999) and 95 % confidence intervals established agreement (or otherwise) between the instants of when PFV and MTGS events occurred. The normality (or otherwise) of the data was determined using a Shapiro-Wilk test. Two-tailed paired *t*-tests were undertaken to determine whether there was a statistical bias between the timings of PFV and MTGS. Values of TGS at PFV and MTGS were compared using repeated measures ANOVA with limb (prosthetic, intact), event (PFV, MTGS) and speed level (slow, customary, fast) as between factors. Post hoc analyses were conducted using Tukey HSD tests. The alpha level was set at 0.05.

6.3 Results

In total 1200 PFV and 1200 MTGS events (600 of each, for intact- and prosthetic-foot) were analysed. Data were normally distributed ($p > 0.05$). Mean walking speeds for the slow, customary and fast levels were $0.93 \pm 0.12 \text{ ms}^{-1}$, $1.13 \pm 0.17 \text{ ms}^{-1}$, and $1.36 \pm 0.27 \text{ ms}^{-1}$ respectively (range $0.73 - 1.77 \text{ ms}^{-1}$). The synchronicity between the timing of PFV and MTGS at each walking speed level, and the average across all speeds, are shown in **Error! Reference source not found.** Foot Velocity and toe trajectory profiles across speeds are shown in Figure 14.

On the intact-limb, PFV occurred $0.015 \pm 0.011 \text{ s}$ after MTGS. This bias was significant ($p < 0.001$). The 95 % LOA (repeatability range) between PFV and

MTGS was between + 0.006 s and - 0.037 s. On the prosthetic-limb, PFV occurred 0.012 ± 0.010 s after MTGS. This bias was significant ($p < 0.001$). The 95 % LOA (repeatability range) between PFV and MTGS was between + 0.008 s and – 0.033 s.

There were no significant differences between the TGS values at MTGS and PFV ($p = 0.38$). MTGS was significantly lower on the prosthetic-limb (slow; 1.11 ± 0.69 cm, customary; 1.09 ± 0.68 cm, fast; 1.10 ± 0.64 cm) compared to the intact-limb (slow; 2.28 ± 0.87 cm, customary; 2.52 ± 0.90 cm, fast; 2.57 ± 0.85 cm, $p < 0.001$). MTGS increased significantly with speed ($p = 0.011$) with clearance at the fast and slow speed levels being significantly different ($p = 0.010$); though a speed-by-limb interaction ($p = 0.004$) indicated that only the speed-related increases on the intact-limb were significant.

Table 7. Group mean (SD) walking speeds, temporal difference between PFV and MTC events, and toe-ground clearance at PFV and MTC.

Walking speed (ms ⁻¹)		Temporal difference* (s)	Range (s)	95% Levels of Agreement (s)	Toe clearance @ PFV (cm)	Toe clearance @ MTC (cm)
Overall	Intact	-0.015 (0.011)	-0.04 / +0.01	-0.037 / +0.006	2.65 (0.76)	2.46 (0.87)
	Pros	-0.012 (0.010)	-0.04 / +0.05	-0.033 / +0.008	1.21 (0.71)	1.10 (0.66)
Slow 0.93 (0.12)	Intact	-0.015 (0.011)	-0.05 / 0	-0.037 / +0.006	2.49 (0.78)	2.28 (0.87)
	Pros	-0.012 (0.010)	-0.04 / +0.03	-0.031 / +0.007	1.22 (0.73)	1.11 (0.69)
Customary 1.13 (0.17)	Intact	-0.015 (0.011)	-0.04 / 0	-0.038 / +0.007	2.70 (0.79)	2.52 (0.90)
	Pros	-0.011 (0.011)	-0.04 / +0.05	-0.033 / +0.010	1.20 (0.71)	1.09 (0.68)
Fast 1.36 (0.27)	Intact	-0.016 (0.010)	-0.04 / +0.01	-0.036 / +0.005	2.77 (0.74)	2.57 (0.85)
	Pros	-0.014 (0.011)	-0.04 / +0.02	-0.035 / +0.007	1.22 (0.74)	1.10 (0.64)

*All temporal differences were significant ($p < 0.001$).

A negative temporal difference indicates PFV occurred after MTC

6.4 Discussion

The mean temporal difference between when PFV and when MTGS occurred was small - approximately 0.014 ± 0.01 s across both limbs. The bias was significant indicating that PFV occurred consistently after MTGS; indeed only 7 of 1200 PFV events occurred prior to the corresponding MTGS event. This temporal relationship between PFV and MTGS was invariant, across the prosthetic- and intact-sides and across all walking speeds. This invariance suggests that swing phase inter-segmental coordination is the same for both limbs. It also suggests that during swing the lower limbs act as simple mechanical pendulums, and thus toe-ground clearance is at least partially a result of how the entire limb swings about the hip rather than being solely controlled by swing-limb knee and/or ankle flexion. In the study by Winter (1992) it was highlighted that PFV and MTGS were synchronous. However, no empirical data were presented to support this contention, and in addition the sampling rate of the kinematic analysis was not detailed. It is reasonable to infer that the video-based methodology used to collect the kinematic data in Winter's (1992) study would have been sampled at a lower rate (likely ~ 30 Hz) than that used in the present study. The lower temporal resolution may have given the appearance of absolute synchronicity (no temporal difference) between PFV and MTGS. The present study, which used a sampling rate of 100 Hz, demonstrated that, MTGS occurs, on average, slightly (i.e. just over one sampling frame @ 100 Hz) before PFV. The small but consistent difference between the timing of when each event occurs explains the

slightly, but non-significantly, higher values of TGS at PFV than MTGS .

The reduced MTGS on the prosthetic-side compared to the intact-side corroborates previous findings (Gates et al., 2012; Wuderman et al., 2012). However the normal distribution of MTC was unlike the typically skewed distribution reported by Begg et al. (2007). The increase in intact-side clearance with increasing walking speed is similar to the speed related increases reported in able-bodied participants (Schulz, 2011). Having greater clearance on the intact-side is likely to be, at least to some extent, a result of UTAs typically presenting reduced prosthetic-limb knee flexion during the loading-response of early stance (Gitter et al., 1991; Bateni & Olney, 2002), which would raise the height of the swing-limb hip.

It has been reported previously that older able-bodied males have some degree of inter-limb asymmetry in toe clearance (Sparrow et al., 2008). The authors noted that the inter-limb asymmetry in toe clearance was associated with step time asymmetry, i.e. the limb with the shorter step time and higher swing velocity had higher toe clearance. They suggested that increased safety margins required at faster swing-foot speeds, may be driving the asymmetry. Such speed-accuracy considerations cannot explain toe clearance inter-limb asymmetries in UTAs who typically present spatially longer steps on the prosthetic-limb than the intact-limb as well as higher swing-foot velocities on the prosthetic-side (as highlighted in Figure 14). If speed-accuracy considerations were the primary driver of such differences it

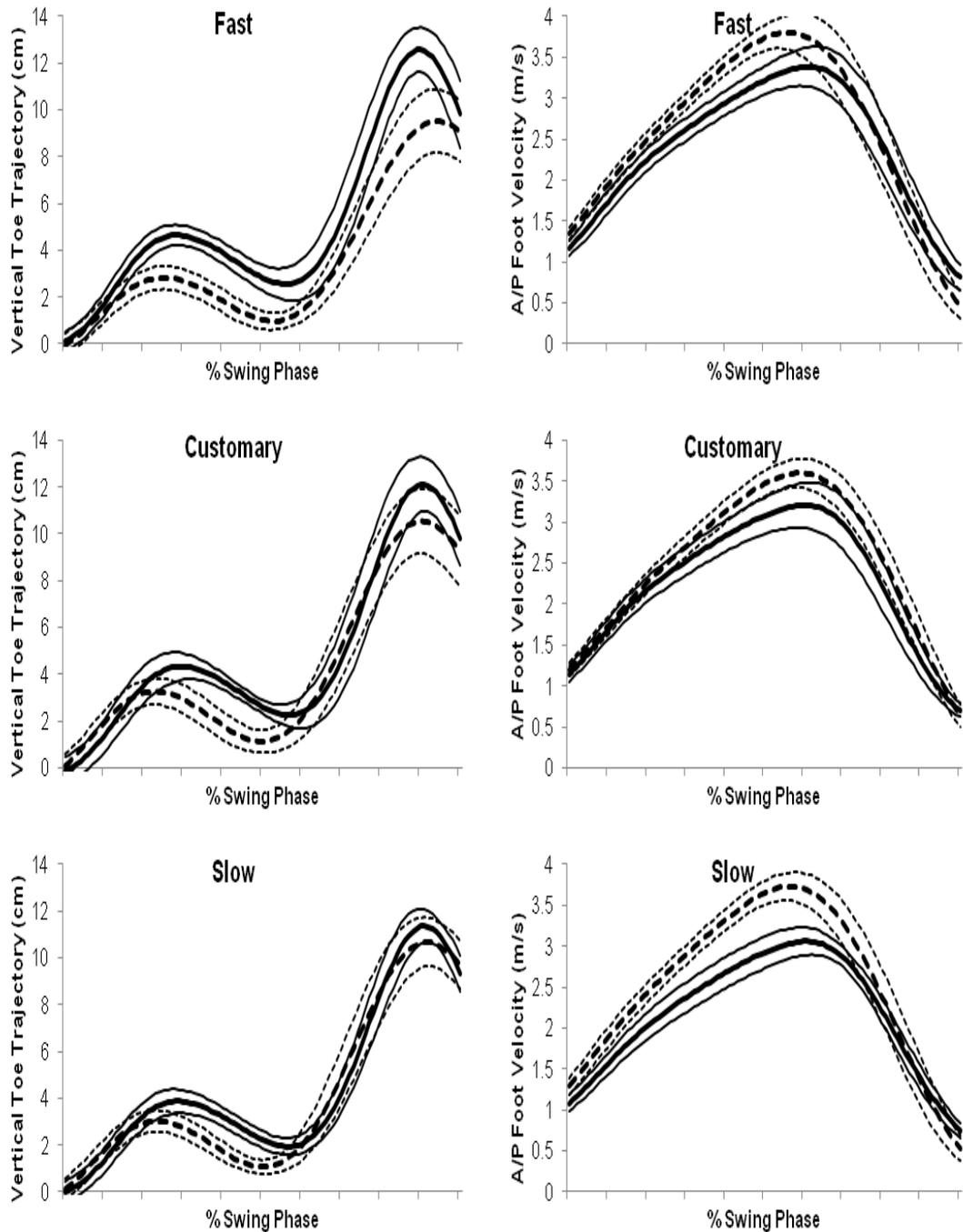


Figure 14. Ensemble Mean (SD) swing phase vertical toe trajectory (left-hand column) and A/P foot velocity (right-hand column) for the intact- (solid lines) and prosthetic- (dashed lines) limbs for one participant at slow, customary and fast walking speeds.

would be expected that higher clearances would occur on the prosthetic-side at all walking speeds. The finding that MTGS on the prosthetic-side did not

increase with speed but did on the intact-side indicates that step time/length asymmetry is not the driver of amputees' toe clearance asymmetries. The fact that MTGS increased with speed on the intact-side but not on the prosthetic-side suggests some level of active, 'online' central motor control of the swinging foot was present on the intact-limb and absent on the prosthetic-limb. It would seem apparent that this control on the intact-limb occurred at the ankle; which would explain why such control was not evident on the prosthetic-side. Given the small magnitudes in speed-related changes of MTGS (2–3 mm), this 'online' control would require only minimal dorsiflexion (~ 1 degree) to affect such changes. It is reasonable to argue therefore that walking speed-related modulation of MTGS in an intact-limb is dependent upon a combination of both passive (limb acting as a simple pendulum) and active (modulation of dorsiflexion) control mechanisms. Hence, as well as its relevance to trips and falls, analysis of MTGS metrics may also provide insights into underlying neural control strategies and coordination patterns. Interestingly MTGS data were normally distributed for both the intact- and prosthetic-limbs. This was unlike a positively skewed distribution reported in the able-bodied by Begg et al. (2007). Begg et al. (2007) described how modulation of this skew could be used to reduce trip-risk thus its absence among UTAs may well be another contributory factor towards their heightened trip risk compared to the able-bodied.

In undertaking the present study, it was difficult to implement an automatic event detection process that could consistently identify the local minima of TGS during swing. This event was difficult to identify because at the

beginning of swing TGS was minimal and thus, when automatically searching for the instant of MTGS (using a sub-routine within the software), the point at or just after toe off was often incorrectly identified. To ensure MTGS is correctly identified when using an automatic event detection process, one could narrow the 'search window' so that it looks for an MTGS event from a number of frames after the instant of toe off onwards. However the magnitude of the off-set from toe off would be affected by factors such as walking speed and would, therefore involve subjective decisions being made for each individual participant and/or each individual trial. This would negate it being an 'automatic' event detection process. An alternative approach is to identify the local minima events manually, as was done in the present study. Both of the above approaches are time consuming and laborious. The findings of the present study indicate that determining the instant of peak foot velocity will also identify when the instant of minimum toe clearance occurs to within 0.014 ± 0.010 s for both intact- and prosthetic-sides. Thus the suggestion is that PFV could be used as a robust and easy to implement kinematic marker for the instant of MTGS. While magnitudes of toe-ground clearance at PFV and MTGS were not significantly different the value of toe-ground clearance was 1 – 2 mm higher at PFV than MTGS. Given the significantly consistent temporal relationship between the events the accuracy and precision of using PFV as a marker for MTGS could be increased, by applying an off-set of + 0.014 s from the instant of PFV. Such an offset could be applied across the range of walking speeds reported in the present study, which encompass the range of speeds used by the majority of ambulatory UTAs.

6.5 Conclusions

The timing of when PFV occurred was virtually synchronous with MTGS. The minimal temporal difference between the two events was invariant across limbs and speed levels. This temporal consistency suggests both lower-limbs act as simple mechanical pendulums during swing. The lack of walking speed related toe-ground clearance changes on the prosthetic-side may potentially increase UTAs' risk of trips at faster walking speeds. Finally, the consistent and minimal temporal differences between events, regardless of speed and limb, indicate that determining the instant of peak swing-foot velocity will also identify the instant of minimum toe clearance to within 0.014 s. This approach could therefore be implemented in automated processing procedures when evaluating minimum toe clearance metrics.

Chapter 7. Impact of using a hydraulic ankle-foot device on inter-segmental coordination and intact-limb minimum toe ground separation

7.1 Introduction

In order to make progression during ambulation each leg swings forward, in turn, while the contralateral leg is in stance. During the swing phase the instant of PFV of the swinging foot coincides with MTGS for the able-bodied (Winter, 1992) and, as demonstrated in the previous chapter, is virtually synchronous with MTGS for both the intact- and prosthetic-limbs in UTAs. Consequently this is when the risk of trips, a predominant cause of falls during ambulation (Blake *et al.*, 1988), is highest. This is the result of a combination of the proximity of the swing-foot to the ground, the high velocity of the swinging foot and the forward-travelling centre of mass being in front of, and moving away from, the base of support (Winter, 1992). Providing adequate TGS is therefore essential for safe ambulation, especially for UTAs who tend to be at higher risk of falls than age-matched able-bodied controls (Kulkarni *et al.*, 1996, Miller *et al.*, 2001). TGS can be affected by changes in joint angle at any individual joint in either the stance or swing limb or by any combination of such. For example, Winter (1992) demonstrated that changes, in isolation, of swing-knee or stance-knee flexion of 1.35° and 3.30° respectively would alter TGS by ± 0.45 cm. Thus successful positioning of the swing-foot, in order to provide sufficient TGS, is a complex and coordinated multi-segment and multi-limb movement strategy involving both the stance and swing limbs. This coordinated movement strategy has been shown to be sensitive to changes such as occlusions of the peripheral visual field (Graci *et al.*, 2009) and increases in walking speed (Schulz, 2011) which both resulted in increased MTGS during overground gait.

In UTAs MTGS is less on the prosthetic than intact side (Gates et al., 2012; Wuderman et al., 2012) therefore it could be reasonable to concentrate on prosthetic-foot MTGS. However the intact-limb also displays increased variability in MTGS which is likely to increase risk of tripping (Wuderman et al., 2012). When an amputee's prosthesis is altered kinematic changes occur during stance on the prosthetic side (De Asha et al., 2013, Chapter 4) which could potentially affect inter-segmental coordination across both limbs.

The purpose of this study was to investigate whether inter-segmental coordination, and consequently MTGS of the intact foot, was affected when UTAs who habitually used a *rigF* used a *hyA-F* when walking at their customary, slow and fast speeds. Kinematic changes in one of, or both, the swing and stance limb can be associated with altered MTGS. Therefore it was hypothesized that increased angular velocity of the prosthetic shank during early stance (De Asha et al., 2013, Chapter 4) would increase early stance residual-knee flexion when using the *hyA-F* which would result in reduced intact-limb MTGS when using a *hyA-F*. It was further hypothesized that this higher residual-knee flexion would increase with higher walking speed and thus, unlike able-bodied participants (Schulz, 2011) and UTAs using a non-articulating foot device (Chapter 6), intact-limb MTGS would decrease with increased walking speed when using a *hyA-F*.

7.2 Methods

7.2.1 Participants

The same ten physically active UTAs (mean \pm SD age; 48 ± 11.7 years, mass; 86 ± 17.7 kg, height; 1.78 ± 0.06 m) as completed the experimental protocol in the previous chapter (Chapter 6) took part. Details are given in Table 6.

7.2.2 Experimental protocol and data acquisition

Participants completed overground walking trials at three different speed levels: customary, comfortable 'slow' and comfortable 'fast'. Kinematic and kinetic data were recorded as per 'general methods', Chapter 3. Each participant completed 10 successful trials at each speed level using both their habitual *rigF* and a *hyA-F*.

7.2.3 Data processing and analysis

The following parameters were determined within Visual 3D software: The instant of intact-limb PFV during swing. Intact-limb TGS at PFV; variability in TGS at PFV; antero-posterior foot separation at PFV; sagittal plane joint angles of the intact- and residual- hips and knees and intact-ankle at PFV and the functional limb length of the intact- and prosthetic- legs at PFV.

Intact-limb PFV was defined as the maximal velocity, in the participant's direction of travel, of the intact-foot segment centre of mass during swing. The timing of PFV was normalised relative to both intact-limb swing and prosthetic-limb stance phases thus two timings are reported. Intact-limb TGS was defined as the vertical distance between the antero-inferior end point of the shoe and the floor. Variability in TGS was defined as the standard deviation of TGS at PFV across trials in each condition. All sagittal plane joint angles were defined within the local coordinate system of the distal segment. Functional limb length was defined as the scalar distance between the origin of the pelvis segment and the antero-inferior end point of the shoe. PFV, TGS, foot separation and all joint angles at PFV were calculated for each trial and then averaged to give a value per prosthetic condition at each walking speed for each participant.

7.2.4 Statistical analysis

The normality (or otherwise) of the data was determined using a Shapiro-Wilk test. Comparisons of parameters were undertaken using repeated measures ANOVA with prosthetic condition (*hyA-F*, *rigF*) and speed level (slow, customary and fast) as repeated factors. Where main effects were significant post hoc analyses were conducted using Tukey HSD tests. Statistical analyses were made using Statistica (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

7.3 Results

Intact-limb TGS at PFV was unaffected by prosthetic condition ($p = 0.11$) but was significantly affected by speed level ($p = 0.030$) in that it increased with increasing speed. Post hoc analysis revealed that the slow and fast speed levels were significantly different ($p = 0.029$). Variability in TGS was unaffected by prosthetic condition or speed level ($p \geq 0.43$). The timing of PFV relative to both prosthetic-limb stance and intact-limb swing phases was unaffected by prosthetic condition or speed level ($p \geq 0.31$). Temporally, intact-limb PFV occurred consistently at 57.1 ± 0.5 % intact swing phase and 58.8 ± 0.6 % prosthetic stance phase. Foot separation at PFV was unaffected by prosthetic condition ($p = 0.21$) but was significantly affected by speed level ($p = 0.004$) in that the swing foot distance ahead of the stance foot increased with increased speed (slow; 0.15 ± 0.06 m, customary; 0.17 ± 0.07 m, fast; 0.20 ± 0.07 m). Post hoc analysis revealed a significant difference between the slow and fast speed levels ($p = 0.003$).

A full list of kinematic variables at PFV is shown at

Table 7. Intact-limb knee and ankle angles at PFV were unaffected by prosthetic condition or speed level ($p \geq 0.29$). Intact-limb hip angle at PFV was significantly affected by prosthetic condition ($p = 0.022$) and by speed level ($p < 0.001$). The hip was more flexed when using the *hyA-F* and flexion increased with increased walking speed. Post hoc analysis revealed no significant difference in hip flexion between the slow and customary speed levels ($p = 0.30$) but significant differences between fast and the other two

speed levels ($p \leq 0.003$). The residual-limb hip was extended at PFV across all conditions. This was significantly affected by prosthetic condition ($p = 0.042$) in that it was less extended when using the *hyA-F*. Although not significant, there was also a trend towards increased residual-hip extension with increased speed level ($p = 0.055$). Residual-limb knee angle at PFV was significantly affected by prosthetic condition ($p = 0.003$) in that it was more flexed when using the *hyA-F*. There was also a significant prosthetic-by-speed interaction ($p < 0.001$) in that residual-knee flexion at PFV increased with increased speed when using the *hyA-F* but decreased with increased speed when using the *rigF*. Functional intact-limb length at PFV was unaffected by prosthetic condition or by speed level ($p \geq 0.19$). Functional prosthetic-limb length was affected by prosthetic condition ($p = 0.049$) but not by speed level ($p = 0.12$). The prosthetic-limb was functionally longer at PFV when the *rigF* was used compared to when the *hyA-F* was used.

Table 7. Group mean (SD) kinematic variables at peak foot velocity when walking at the slow, customary and fast speed levels using a *hyA-F* and *rigF*. Statistically different differences are in **bold**.

	SLOW <i>hyA-F</i> / <i>rigF</i>	CUSTOMARY <i>hyA-F</i> / <i>rigF</i>	FAST <i>hyA-F</i> / <i>rigF</i>	P value
Walking speed (ms ⁻¹)	0.94 (0.13) / 0.92 (0.10)	1.18 (0.16) / 1.09 (0.17)	1.37 (0.22) / 1.34 (0.24)	Pros 0.003 Speed < 0.001 Interact. 0.043
TGS @ PFV (mm)	26.1 (27.3) / 23.8 (25.2)	28.4 (29.6) / 25.5 (26.7)	28.4 (29.3) / 27.1 (28.6)	Pros 0.11 Speed 0.030 Interact. 0.63
TGS @ PFV variability (mm)	3.8 (1.7) / 3.9 (1.1)	3.6 (0.7) / 4.7 (2.2)	4.6 (1.7) / 4.3 (1.4)	Pros 0.48 Speed 0.43 Interact. 0.36
Timing of PFV (% prosthetic limb stance)	56.2(2.5) / 57.0 (2.7)	57.0 (2.0) / 57.5 (3.1)	57.3 (1.5) / 57.3 (1.5)	Pros 0.31 Speed 0.43 Interact. 0.37
Timing of PFV (% intact limb swing)	57.8 (2.3) / 58.4 (2.2)	58.7 (4.6) / 59.1 (2.5)	59.6 (3.8) / 59.2 (3.6)	Pros 0.63 Speed 0.47 Interact. 0.76
Foot separation @ PFV (m)	0.14 (0.06) / 0.15 (0.06)	0.17 (0.06) / 0.16 (0.07)	0.20 (0.07) / 0.19 (0.08)	Pros 0.21 Speed 0.004 Interact. 0.57
Residual limb hip angle @ PFV (°)	- 1.5 (9.0) / - 4.2 (8.1)	- 1.5 (8.5) / - 4.6 (8.2)	- 1.7 (7.6) / - 5.7 (8.5)	Pros 0.042 Speed 0.055 Interact. 0.54
Residual limb knee angle @ PFV (°)	6.3 (3.6) / 3.9 (4.7)	6.6 (4.3) / 3.3 (4.8)	8.7 (4.2) / 3.2 (5.0)	Pros 0.003 Speed 0.55 Interact. < 0.001
Intact limb hip angle @ PFV (°)	26.4 (7.7) / 25.1 (6.9)	27.8 (7.7) / 25.3 (7.3)	30.0 (7.7) / 27.2 (7.5)	Pros 0.022 Speed < 0.001 Interact. 0.29
Intact limb knee angle @ PFV (°)	43.1 (5.1) / 42.1 (4.0)	43.0 (4.6) / 41.7 (3.3)	41.4 (3.3) / 42.6 (3.0)	Pros 0.64 Speed 0.70 Interact. 0.15
Intact limb ankle angle @ PFV (°)	1.2 (2.3) / - 0.5 (4.2)	1.5 (2.2) / - 0.1 (4.7)	1.0 (3.0) / 0.0 (4.8)	Pros 0.29 Speed 0.69 Interact. 0.23

7.1 Discussion

Intact-limb hip angle and residual-limb hip and knee angles at PFV were all significantly affected by prosthetic condition. Thus the first part of the hypothesis, that residual-knee angle would be higher at the instant of PFV

when participants used the *hyA-F* than when they used the *rigF* and that increased knee angle would also further increase as speed level increased, was supported. Despite this intact-limb TGS at PFV was unaffected by prosthetic condition and, as in able-bodied controls (Schulz, 2011), increased as speed level increased while using both prosthetic devices. Thus the second part of the original hypothesis that TGS at PFV would decrease with increased speed level when using the *hyA-F*, was not supported. In addition there were no significant changes to the timing of PFV or to TGS variability. The participants were therefore able to spatially and temporally maintain consistent TGS at PFV while joint kinematics of both limbs altered due to altered prosthetic components. In addition, and perhaps surprising given that it has been suggested previously that swing-limb knee and ankle angle (Winter, 1992; Mills et al., 2008) play a large role in modulating TGS, there were no changes in either due to prosthetic condition or speed. The functional length of the intact-limb at PFV was unaffected by either speed or by prosthetic condition which indicates that the 'gross changes' in TGS were not being modulated at the swing-limb knee or ankle. Interestingly, given that the TGS at PFV increased with increased speed level, there was a simultaneous increase in swing-hip flexion and advanced swing-foot position in relation to the stance-foot coupled with a trend towards increased stance-hip extension at PFV. This suggests that modulation of TGS may occur at both hips, but primarily at the swing-hip. Together these findings support the suggestion made in the previous chapter that the swing-limb may act as a simple mechanical pendulum beneath the hip.

The absolute temporal consistency of PFV across speed levels reported in Chapter 6 was reflected in the temporally normalised results above. This was the case when the timing of PFV was normalised to both intact-limb swing phase and prosthetic-limb stance phase and was unaffected by prosthetic condition. This suggests that participants were able to maintain temporal coordination both within and between limbs. This invariance of temporal coordination across speeds and prosthetic conditions was achieved despite changes in TGS at PFV and in cadence and/or step length across walking speeds. This tends to support the suggestion of mechanical-like pendulum behaviour of the swing leg being the primary (temporal) modulator of TGS but some other mechanism also being involved in the 'fine-tuning' control of the foot position which is responsible for the speed related increases in TGS at PFV. Given that, as described in the previous chapter, such on-line 'fine-tuning' is only observed on the intact side the logical assumption is that this adjustment is made at the ankle. While no significant differences between speed levels were observed in the intact ankle angle at PFV it could still be involved in on-line 'fine-tuning' however as the angular displacements required to do so would, in practice, be imperceptible. These results therefore reinforce the notion that TGS is controlled by a complex and sophisticated inter- and intra-limb coordination strategy where changes at a joint, such as the increased residual-knee angle when using a *hyA-F*, can be compensated for across one or several other joints in both the stance and swing limbs.

7.2 Conclusions

The findings, and in particular the changes in hip angle at the instant of PFV however do, at least partially, support the suggestions made in the previous chapter (Chapter 6) that TGS is primarily modulated at the swing-limb hip and that the swing-limb tends to act as a simple mechanical pendulum beneath the hip joint. There was no evidence in altered joint kinematics across either the stance or swing limbs as to which, if any, specific joint(s) were the primary controller(s) of speed related TGS modulation.

**Chapter 8. Walking speed
related joint kinetic alterations
in trans-tibial amputees: the
effects of hydraulic 'ankle'
damping**

8.1 Introduction

The previous experimental chapters have tended to focus on changes which occur due to use of a *hyA-F* during the prosthetic-limb stance phase of overground gait. Amputees, however, are known to make kinetic compensations on the intact-side. This chapter therefore investigates stance phase kinetics on both the intact- and residual-limbs and also power flow to and from the prosthetic-foot device.

The determination of muscle moments and associated powers at the joints of the lower limbs provides key insights in to what, mechanically, is driving locomotion. When walking at their customary speed over level ground, able-bodied individuals typically display a period of low-magnitude power absorption at the ankle joint, for the first three quarters of stance, at which point a period of larger magnitude power generation occurs (Winter et al., 1995; Neptune et al., 2001; Kirtley, 2006). At the hip, moderate magnitude power is generated during early and late stance with a short period of power absorption at mid-stance (Paul, 1970; Silverman et al., 2008). In contrast, the knee tends to predominantly absorb power throughout stance with very little power generation occurring (Paul, 1970; Siegel et al., 2004; Kirtley, 2006). When walking at their customary speed UTAs compensate for the absent foot and ankle by increasing early and late stance power generation at both hips and increasing late stance power generation at the intact ankle (Gitter et al., 1991; Seroussi et al., 1996; Sanderson & Martin, 1997; Nolan & Lees, 200; Sadeghi et al., 2001). Despite such compensations, UTAs tend to have

a slower freely chosen walking speed than able-bodied persons (Nolan et al., 2003). In addition, the moments and powers (peaks and integrals) at the residual knee are reduced compared to the intact side (Gitter et al., 1991; Czerniecki & Gitter, 1996, Powers et al., 1997), most likely due to a desire to minimise loads on the residuum.

To increase walking speed, able-bodied individuals predominantly increase stance-phase power generation at the hip (and moderately increase power generation at the ankle) (Silverman et al., 2008; Chen et al., 1997; Riley et al., 2001). They also display increases in both power absorption and generation at the knee (Silverman et al., 2008; Chen et al., 1997). Thus as walking speed increases, the proportional contribution of the ankle to gait propulsion reduces while the contributions of the hips and knees increase (Silverman et al., 2008). In UTAs walking speed is likewise increased by increasing stance-phase power generation at the hip on both the intact- and prosthetic-sides (Silverman et al., 2008).

The purpose of a prosthetic ankle-foot device is to approximate the function provided by the absent physiological structures. Modern, passive prosthetic devices typically incorporate flexible keels that are able to absorb and return power during stance through elastic deformation and recoil. Such deformation occurs irrespective of whether the foot is fixed to the prosthetic shank rigidly or via a device allowing articulation. Not surprisingly, use of a dynamic response (sometimes referred to as energy-storing and return) foot compared to semi-rigid foot (e.g. SACH) has been shown to increase the

amount of late stance power returned at the prosthetic 'ankle' as well as power absorbed at the residual knee (Underwood et al., 2001).

Use of a *hyA-F* by active, UTAs, has been shown to result in reduced in-socket pressures (Portnoy et al., 2012), a less disrupted centre-of-pressure progression and an increased freely chosen walking speed (De Asha et al., 2013, Chapter 4) and increased toe clearance (Johnson et al., 2013). The *hyA-F* can passively realign in the sagittal plane when walking. This ability to change alignment may also accommodate changes in walking speed because of the requirement of the foot/ankle to go through a greater range of motion when walking at faster speeds. Thus use of the device may offer speed-related advantages over more traditional attachment types. The purpose of the present study was to investigate whether a *hyA-F* device reduced speed-related changes in intact-limb joint kinetics compared to when the foot was attached via a rigid, non-articulating attachment (*rigF*). Specifically, joint moments and powers were compared between attachment categories when UTA participants walked at their freely chosen slow, customary and fast speeds. It was hypothesized that due to having fewer and/or smaller disruptions in centre of pressure progression under the foot during prosthetic-limb stance when using a *hyA-F* (De Asha et al., 2013, Chapter 4) the speed-related increases in compensatory intact side stance-phase power generation at the hip and ankle would be reduced. It was further hypothesized that due to a reduction in in-socket pressures while using the *hyA-F* (Portnoy et al., 2012) and increased residual-knee flexion

during stance (Chapter 7) there would be increased loading on the residual knee at all speed levels.

8.2 Methods

8.2.1 Participants

Eight male, physically active UTAs (mean \pm SD age; 44.8 ± 10.7 years, mass; 83.3 ± 19.0 kg, height; 1.77 ± 0.05 m) took part. Full details are given in Table 8.

Table 8. Descriptive details of UTA participants.

PARTICIPANT I.D. NUMBER	AGE (YEARS)	MASS (KG)	HEIGHT (M)	SEX	SIDE OF AMPUTATION	TIME SINCE AMPUTATION (YEARS)	HABITUAL FOOT DEVICE
S 01	34	72	1.72	M	RIGHT	8	ESPRIT
S 02	62	82	1.76	M	LEFT	19	ESPRIT
S 03	39	87	1.71	F	LEFT	9	ESPRIT
S 04	38	62	1.80	M	RIGHT	5	ESPRIT
S 07	61	123	1.75	M	RIGHT	3	ESPRIT
S 08	63	93	1.85	M	RIGHT	43	ESPRIT
S 09	59	100	1.78	M	RIGHT	11	ESPRIT
S 10	39	66	1.78	M	RIGHT	6	ESPRIT

8.2.2 Experimental Protocol and data acquisition

Kinematic and kinetic data were recorded as per 'general methods', Chapter 3, while participants completed overground walking trials at three different speed levels: customary, comfortable 'slow' and comfortable 'fast'. Participants were instructed to walk "at their normal walking speed", "slowly" and "as fast as comfortably possible". Trials were undertaken in two blocks, each made up of sets of trials at each of the three walking speed levels. One block was undertaken using the *rigF* and the other using a *hyA-F*. Attachment type order was counterbalanced across participants as were the order of 'fast' or 'slow' sets, following the customary speed set which was always performed first. Trials were repeated until 10 'clean' contacts of each foot were made at each speed in each attachment condition. Due to the counter-balanced experimental design and because of the methodological limitations associated with speed-controlled studies and the difficulty in generalising findings from such studies to the natural environment (Wilson, 2012) walking speed was not controlled.

During data collection, the participant wore retro-reflective markers as described in 'general methods', Chapter 3.

8.2.3 Data processing

All data were reduced and processed within Workstation (Vicon, Oxford, UK) and Visual 3D (C Motion, Germantown, MD, USA) as per 'general methods', Chapter 3.

8.2.4 Data analysis

The dynamic-response foot, which is integral to both the *hyA-F* and *rigF*, has flexible heel and forefoot keels which, when loaded, deform simulating plantar- and dorsi-flexion about non-defined axes. Thus, as with all such feet, the assumptions of a rigid segment and pin joint articulation (Winter, 2009) are violated. Consequently, the assessment and interpretation of 'ankle' kinetics can be problematic and sometimes misleading (Geil et al., 2000; Miller & Childress, 2005; Sagawa Jr. et al., 2011). Prince et al. (1994) suggested a kinetic analysis technique that determined the energy absorbed and returned by the prosthetic-foot by determining the power flow at the distal end of the prosthetic-shank pylon which, regardless of the type of attachment and/or foot, is the physical application point of the forces and moments transferred to and from the shank. As such this modelling approach, as the authors highlighted, can be used for either articulating or non-articulating ankle-foot devices. Therefore the energy entering or leaving the prosthetic-foot was assessed by summing the sagittal plane translational and rotational power flows at the distal end of the prosthetic-shank (Figure 15), as per the method described by Prince et al. (1994) which is described in full in Chapter 2.

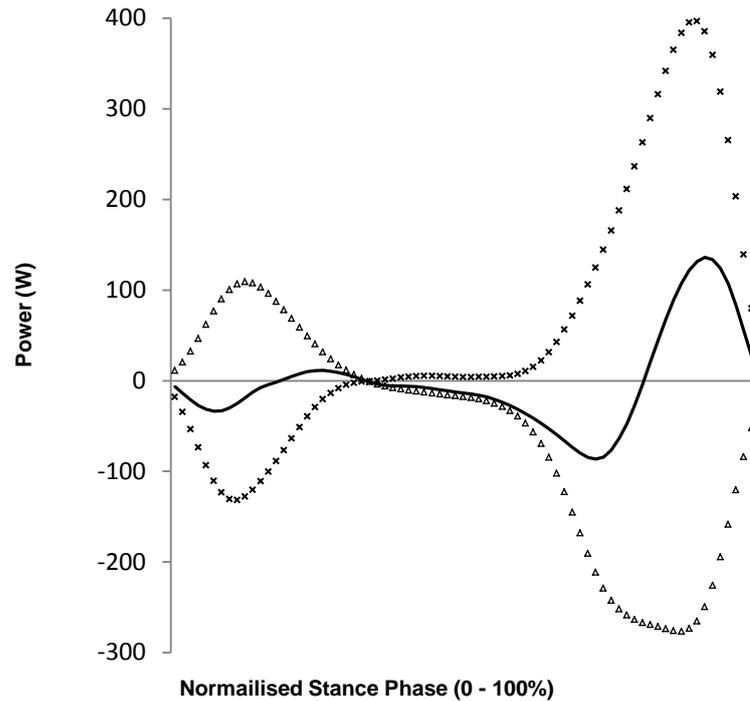


Figure 15. Exemplar trial showing stance phase total power (P_{dist} , solid line) along with the rotational (P_{rot} , triangles) and translational powers (P_{trans} , crosses) at the distal end of the prosthetic shank. Negative values indicate the power leaving the shank (flow to the foot) and positive values indicate power entering the shank (flow from the foot).

The joint kinetics (muscle moments and associated powers) at all physiological joints were calculated using standard inverse dynamics. At the residual-knee, joint kinetics were determined by assuming the foot and shank to be a single rigid segment with the distal forces acting on the segment being the GRF (Dumas et al., 2009).

All moment and power data were normalised to body weight. Based on previous findings (De Asha et al., 2013, Chapter 4) it was expected that the slow, customary and fast walking speed levels would all be higher when using the *hyA-F* than when using the *rigF*. As such, all moment and power

measures were normalised to walking speed (Riley et al., 2001; Stansfield et al., 2006). Thus, in addition to absolute differences, the results presented indicate differences per metre travelled. The effects of attachment category and speed level on the absolute and speed-normalised moments and powers at each joint were determined by comparing joint moment peaks, and joint power peaks and power integrals (work) across conditions. Negative and positive power integrals were assessed separately to provide an insight, respectively, of the eccentric and concentric work (intact joints) or energy absorbed and returned by the prosthetic-foot. The scalar magnitudes of these integrals were also then summed to yield the total work done at each physiological joint for both the intact- and residual-limbs. The difference between negative and positive power integrals at the distal end of the prosthetic-shank was calculated to yield the net energy dissipated by the ankle-foot device.

All parameters of interest were calculated for each individual trial and then averaged across trials to give a mean value for each participant, at each speed, in each attachment condition.

8.2.5 Statistical analyses

Comparisons between foot attachment conditions were undertaken using repeated measures ANOVA with attachment type (*hyA-F*, *rigF*) and speed level (slow, customary and fast) as repeated factors. Where main effect of speed was significant post hoc analyses were conducted using Tukey HSD

tests. Statistical analyses were made using Statistica (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

8.3 Results

As the object of the study was not to investigate speed-effects *per se* the 'in text' results only detail the effects of attachment category and/or attachment category-by-speed interactions. In addition, to avoid repetition only speed-normalised results are detailed. Speed main effects are indicated in the results tables (Tables 10-15); which present data both as speed normalised and in absolute terms. Ensemble average, speed-normalised joint moments and powers for physiological joints of both limbs are shown in Figure 16 (moment) and Figure 17 (power), and for distal end of the prosthetic shank in Figure 18. These figures are reproduced in larger scale at Appendix E.

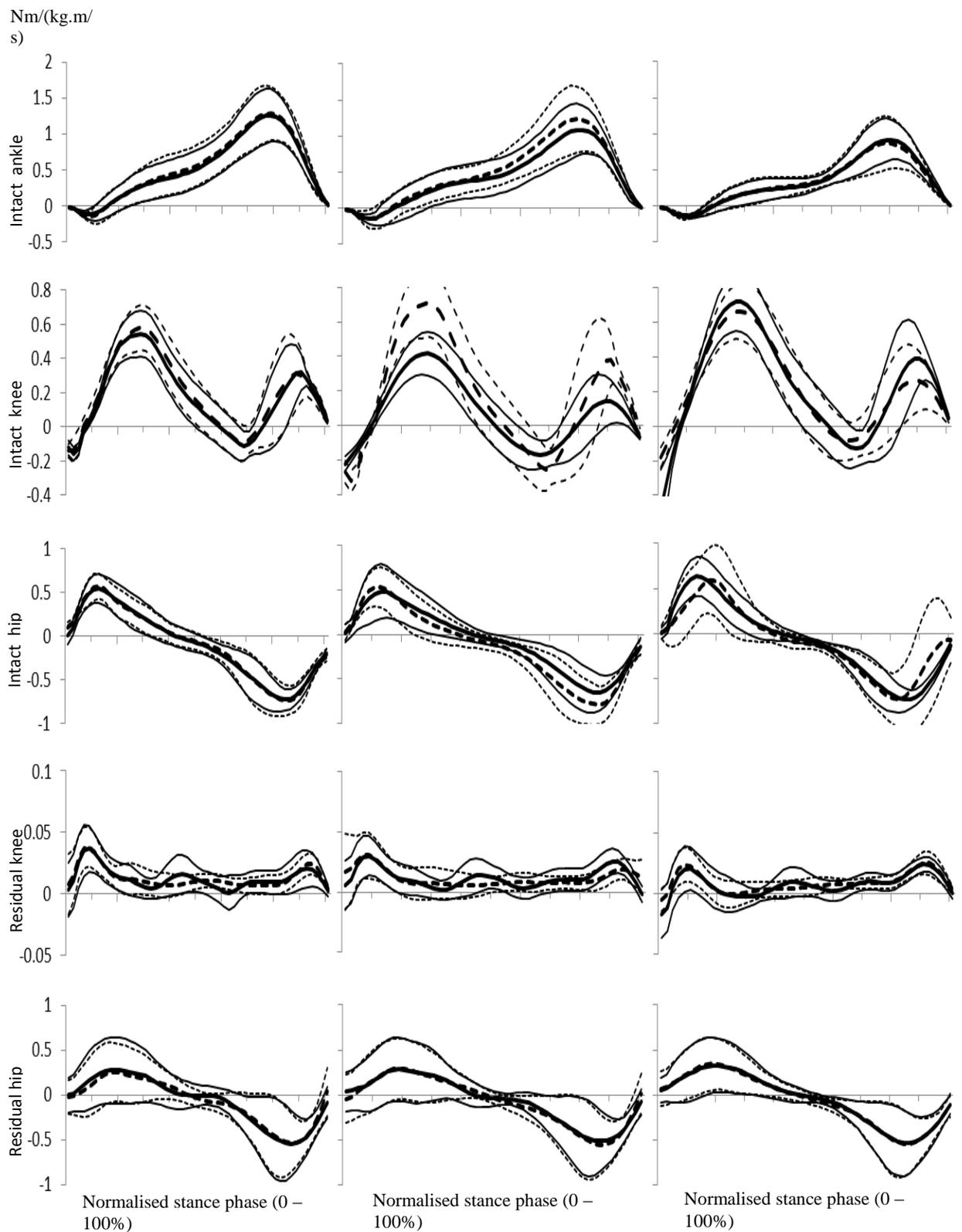


Figure 16. Ensemble mean (SD) speed normalised stance phase moments at hips, knees and ankle (intact limb only) joints when using rigid (*rigF*, dotted lines) and hydraulic (*hyA-F*, solid lines) ankle-foot devices at slow (left column), customary (centre column) and fast (right column) speed levels.

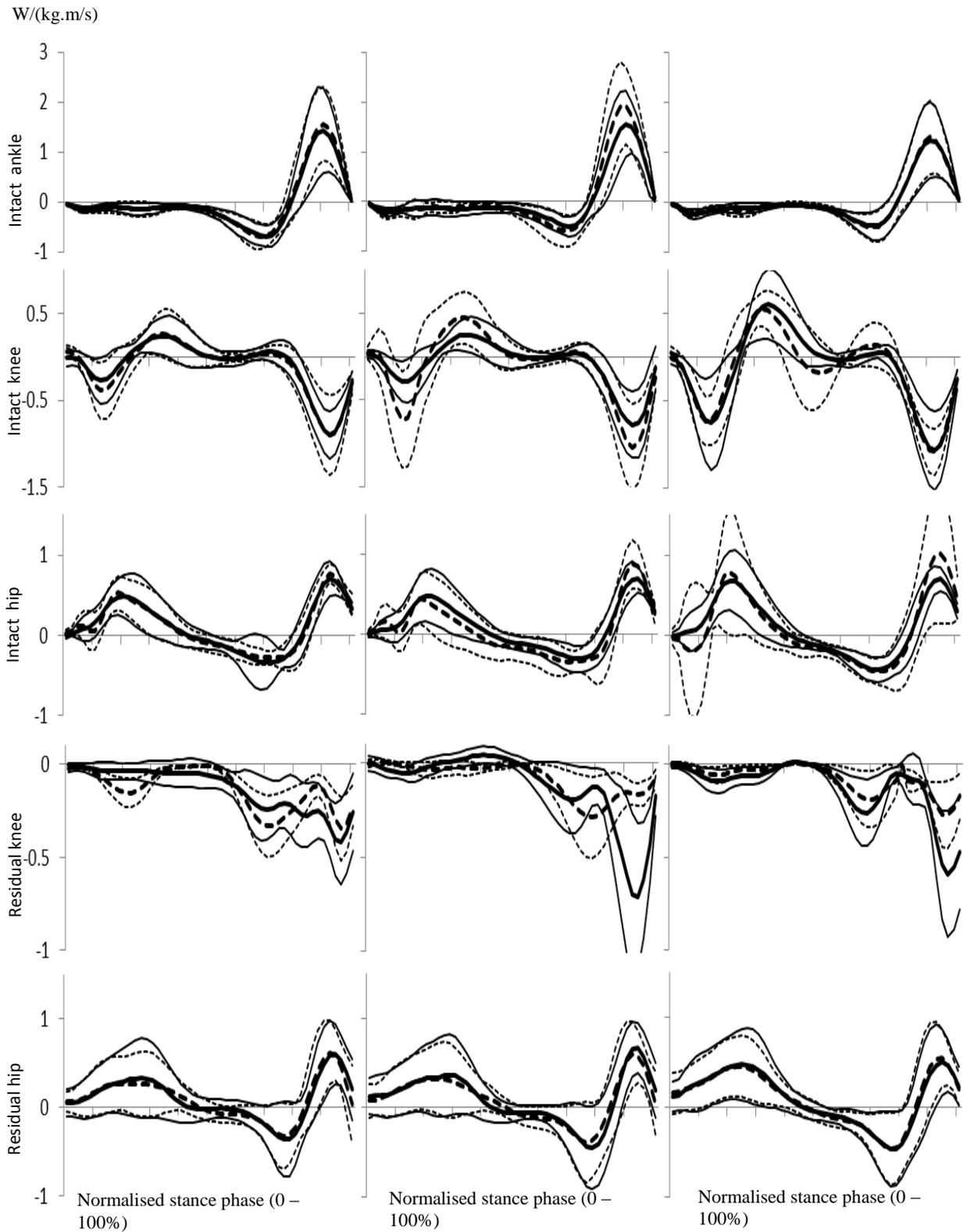


Figure 17. Ensemble mean (SD) speed normalised stance phase power generation (positive) and absorption (negative) at hip, knee and ankle (intact limb only) joints when using rigid (*rigF*, dotted lines) and hydraulic (*hyA-F*, solid lines) ankle-foot devices at slow (left column), customary (centre column) and fast (right column) speed levels.

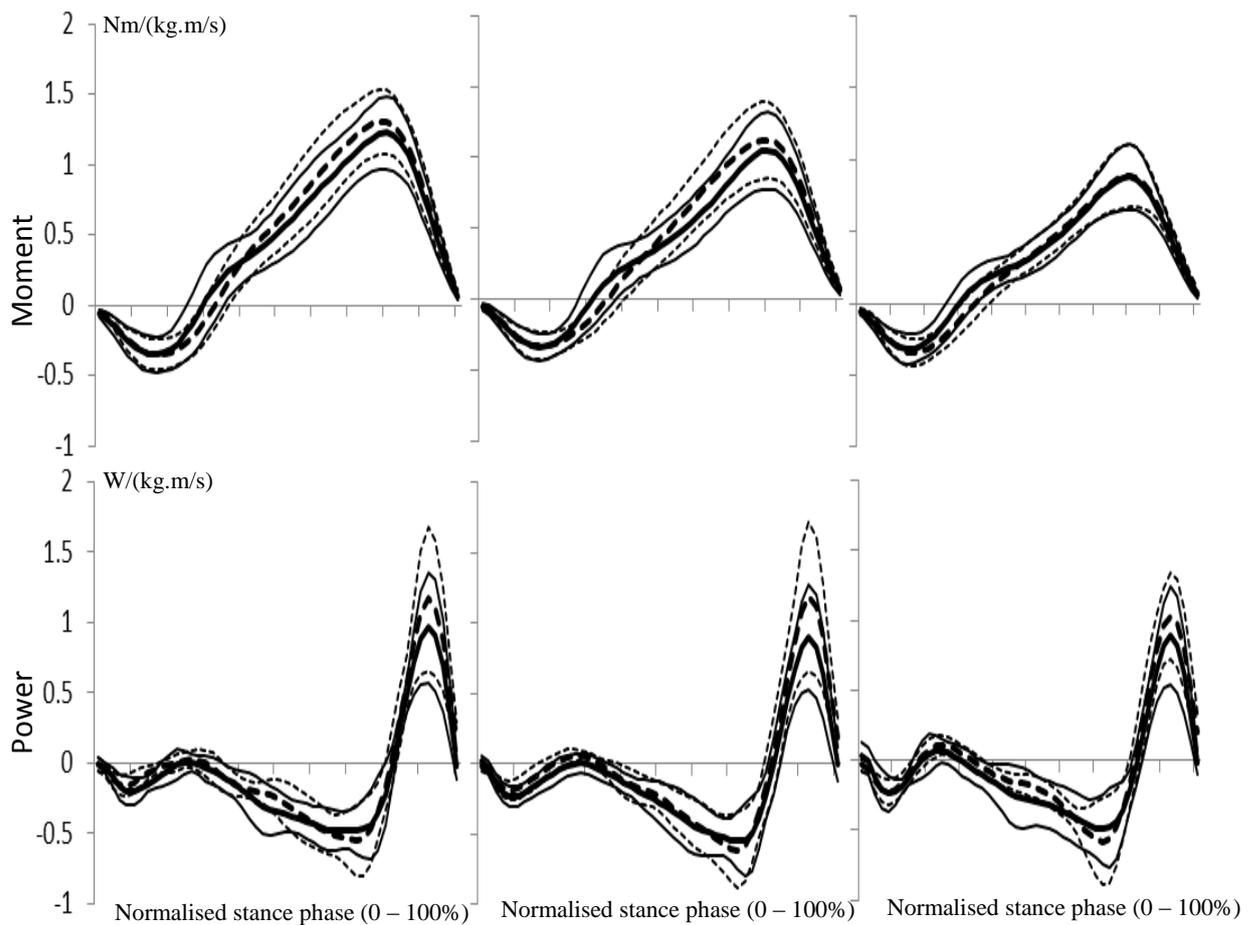


Figure 18. Ensemble mean (SD) speed normalised stance phase external moment (plantarflexion – negative, dorsiflexion – positive, top) and power profiles at *pros-end* when using rigid (*rigF*, dotted lines) and hydraulic (*hyA-F*, solid lines) ankle-foot devices at slow (left column), customary (centre column) and fast (right column) speed levels. Negative power is power leaving the shank into the foot and positive power is power returning to the shank from the foot.

8.3.1 Intact limb hip, knee and ankle

The intact-limb hip peak flexion moment was significantly affected by attachment category ($p = 0.038$), and was reduced (all speeds) when using the *hyA-F*. There was no significant main effect of attachment category or

interaction between attachment category and speed level on the joint kinetics at the intact-knee. The intact-limb peak dorsiflexion moment was significantly affected by attachment category ($p = 0.044$), and was lower when using the *hyA-F* (all speeds). The intact-ankle negative work ($p = 0.032$) and total work ($p = 0.003$) done were significantly affected by attachment category; less work was done when using the *hyA-F*. There was also a significant effect of attachment category on the total joint work done by the intact-limb (summation of work done across all joints, $p = 0.047$); indicating reduced work when using the *hyA-F* device.

8.3.2 Residual limb hip and knee

There was no significant effect of attachment category or interaction between attachment category and speed level on the joint kinetics at the residual-hip. The negative power peak at the residual-knee was affected by a speed level-by-attachment category interaction ($p \leq 0.034$), and was higher when using the *hyA-F* at customary speed than all other conditions ($p \leq 0.016$). Residual-knee negative work was affected by attachment category ($p = 0.047$); indicating there was increased eccentric work when using the *hyA-F*.

Table 9. Group mean (SD) peak extension (positive) and flexion (negative) muscle moments for intact hip, knee and ankle joints and residual knee and hip joints when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Results are listed downwards - slow, customary speed and fast. Statistically significant differences are in **bold**.

	Speed Normalised Nm/kg.ms			Absolute Nmkg ⁻¹		
	<i>hyA-F</i>	<i>rig-F</i>	<i>p</i> value	<i>hyA-F</i>	<i>rig-F</i>	<i>p</i> value
Intact Hip Moment	+ 0.56 (0.17) - 0.77 (0.16) + 0.53 (0.25) - 0.69 (0.21) + 0.64 (0.22) - 0.77 (0.16)	+ 0.61 (0.13) - 0.82 (0.21) + 0.64 (0.19) - 0.88 (0.23) + 0.83 (0.73) - 1.00 (0.45)	0.18 / 0.038 (foot) 0.28 / 0.30 (speed) 0.71 / 0.35 (interact)	+ 0.53 (0.49) - 0.73 (0.70) + 0.61 (0.55) - 0.82 (0.77) + 0.88 (0.76) - 1.05 (0.99)	+ 0.56 (0.52) - 0.76 (0.73) + 0.69 (0.63) - 0.94 (0.90) + 1.20 (0.73) - 1.38 (1.10)	0.30 / 0.11 (foot) 0.06 / 0.005 (speed) 0.59 / 0.33 (interact)
Intact Knee Moment	+ 0.55 (0.22) - 0.23 (0.19) + 0.47 (0.23) - 0.16 (0.17) + 0.69 (0.25) - 0.15 (0.17)	+ 0.60 (0.31) - 0.21 (0.18) + 0.76 (0.14) - 0.17 (0.17) + 0.69 (0.26) - 0.13 (0.19)	0.17 / 0.29 (foot) 0.21 / < 0.001 (speed) 0.09 / 0.28 (interact)	+ 0.54 (0.24) - 0.20 (0.15) + 0.56 (0.28) - 0.17 (0.14) + 0.96 (0.43) - 0.18 (0.16)	+ 0.56 (0.31) - 0.18 (0.15) + 0.79 (0.35) - 0.18 (0.15) + 0.96 (0.46) - 0.15 (0.17)	0.27 / 0.13 (foot) 0.002 / 0.12 (speed) 0.11 / 0.40 (interact)
Intact Ankle Moment	+ 1.29 (0.38) - 0.14 (0.05) + 1.13 (0.37) - 0.19 (0.06) + 0.95 (0.30) - 0.15 (0.03)	+ 1.32 (0.39) - 0.18 (0.04) + 1.27 (0.47) - 0.18 (0.09) + 0.92 (0.37) - 0.16 (0.04)	0.11 / 0.044 (foot) < 0.001 / 0.28 (speed) 0.10 / 0.07 (interact)	+ 1.20 (0.30) - 0.13 (0.05) + 1.30 (0.28) - 0.22 (0.07) + 1.27 ±(.35) - 0.20 (0.04)	+ 1.19 (0.26) - 0.16 (0.04) + 1.31 (0.31) - 0.20 (0.09) + 1.21 (0.37) - 0.22 (0.41)	0.54 / 0.28 (foot) 0.35 / 0.004 (speed) 0.52 / 0.06 (interact)
Residual Hip Moment	+ 0.37 (0.32) - 0.62 (0.37) + 0.36 (0.32) - 0.60 (0.36) + 0.37 (0.28) - 0.58 (0.35)	+ 0.41 (0.25) - 0.64 (0.32) + 0.43 (0.27) - 0.66 (0.34) + 0.39 (0.27) - 0.63 (0.32)	0.21 / 0.14 (foot) 0.76 / 0.65 (speed) 0.48 / 0.27 (interact)	+ 0.33 (0.25) - 0.57 (0.32) + 0.40 (0.31) - 0.68 (0.35) + 0.47 (0.31) - 0.75 (0.39)	+ 0.37 (0.20) - 0.59 (0.29) + 0.44 (0.24) - 0.69 (0.30) + 0.48 (0.27) - 0.81 (0.30)	0.26 / 0.43 (foot) 0.016 / < 0.001 (speed) 0.59 / 0.44 (interact)
Residual Knee Moment	+ 0.04 (0.02) - 0.02 (0.01) + 0.04 (0.02) - 0.01 (0.01) + 0.03 (0.02) - 0.02 (0.01)	+ 0.04 (0.02) - 0.01 (0.02) + 0.04 (0.02) - 0.01 (0.01) + 0.03 (0.01) - 0.01 (0.00)	0.47 / 0.09 (foot) 0.021 / 0.049 (speed) 0.20 / 0.18 (interact)	+ 0.03 (0.01) - 0.02 (0.01) + 0.04 (0.01) - 0.01 (0.01) + 0.06 (0.02) - 0.02 (0.01)	+ 0.03 (0.01) - 0.01 (0.01) + 0.05 (0.02) - 0.01 (0.01) + 0.05 (0.01) - 0.02 (0.01)	0.76 / 0.78 (foot) < 0.001 / 0.84 (speed) 0.51 / 0.35 (interact)

Table 10. Group mean (SD) peak concentric (positive) and eccentric (negative) muscle powers for intact hip, knee and ankle joints and residual knee and hip joints when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Results are listed downwards - slow, customary speed and fast. Statistically significant differences are in **bold**.

	Speed Normalised W/kg.ms			Absolute Wkg ⁻¹		
	<i>hyA-F</i>	<i>rig-F</i>	<i>p</i> value	<i>hyA-F</i>	<i>rig-F</i>	<i>p</i> value
Intact Hip Power	+ 1.51 (0.89) - 0.43 (0.32) + 1.68 (0.67) - 0.36 (0.17) + 1.28 (0.75) - 0.49 (0.16)	+ 1.68 (0.98) - 0.39 (0.13) + 1.99 (1.00) - 0.47 (0.25) + 1.26 (0.75) - 0.77 (0.65)	0.13 / 0.24 (foot) 0.53 / 0.11 (speed) 0.53 / 0.24 (interact)	+ 0.73 (0.22) - 0.39 (0.23) + 0.91 (0.26) - 0.42 (0.21) + 1.15 (0.51) - 0.67 (0.24)	+ 0.78 (0.18) - 0.37 (0.14) + 1.04 (0.34) - 0.51 (0.30) + 1.66 (1.83) - 1.12 (1.20)	0.25 / 0.24 (foot) 0.12 / 0.036 (speed) 0.46 / 0.27 (interact)
Intact Knee Power	+ 0.38 (0.17) - 0.92 (0.24) + 0.37 (0.16) - 0.84 (0.38) + 0.62 (0.31) - 1.03 (0.32)	+ 0.38 (0.23) - 0.98 (0.39) + 0.60 (0.38) - 1.31 (0.56) + 0.69 (0.25) - 1.12 (0.37)	0.09 / 0.08 (foot) < 0.001 / 0.25 (speed) 0.11 / 0.06 (interact)	+ 0.36 (0.18) - 0.91 (0.35) + 0.44 (0.21) - 1.02 (0.51) + 0.87 (0.55) - 1.46 (0.61)	+ 0.35 (0.23) - 0.93 (0.43) + 0.63 (0.35) - 1.41 (0.58) + 0.96 (0.46) - 1.55 (0.63)	0.13 / 0.13 (foot) < 0.001 / < 0.001 (speed) 0.16 / 0.09 (interact)
Intact Ankle Power	+ 1.51 (0.89) - 0.74 (0.27) + 1.68 (0.67) - 0.77 (0.48) + 1.28 (0.75) - 0.54 (0.27)	+ 1.68 (0.98) - 0.79 (0.29) + 1.99 (1.00) - 0.87 (0.53) + 1.26 (0.75) - 0.56 (0.29)	0.30 / 0.10 (foot) 0.048 / 0.21 (speed) 0.47 / 0.41 (interact)	+ 1.47 (0.90) - 0.69 (0.24) + 1.93 (0.80) - 0.92 (0.68) + 1.85 (1.27) - 0.72 (0.35)	+ 1.55 (0.91) - 0.72 (0.24) + 2.09 (0.91) - 0.95 (0.69) + 1.80 (1.29) - 0.74 (0.38)	0.74 / 0.44 (foot) 0.14 / 0.49(speed) 0.82 / 0.95 (interact)
Residual Hip Power	+ 0.70 (0.38) - 0.49 (0.35) + 0.72 (0.44) - 0.57 (0.47) + 0.69 (0.40) - 0.54 (0.41)	+ 0.72 (0.37) - 0.57 (0.25) + 0.73 (0.36) - 0.58 (0.35) + 0.74 (0.41) - 0.53 (0.38)	0.63 / 0.58 (foot) 0.89 / 0.66 (speed) 0.74 / 0.41 (interact)	+ 0.63 (0.26) - 0.45 (0.28) + 0.81 (0.40) - 0.64 (0.48) + 0.89 (0.39) - 0.69 (0.49)	+ 0.65 (0.30) - 0.53 (0.24) + 0.76 (0.28) - 0.62 (0.34) + 0.94 (0.41) - 0.67 (0.43)	0.93 / 0.82 (foot) < 0.001 / 0.08 (speed) 0.34 / 0.30 (interact)
Residual Knee Power	+ 0.103 (0.154) - 0.459 (0.501) + 0.102 (0.121) - 1.048 (1.115) + 0.089 (0.150) - 0.564 (0.692)	+ 0.06 (0.07) - 0.41 (0.44) + 0.04 (0.03) - 0.42 (0.45) + 0.03 (0.02) - 0.42 (0.46)	0.11 / 0.07 (foot) 0.76 / 0.017 (speed) 0.90 / 0.034 (interact)	+ 0.10 (0.15) - 0.43 (0.48) + 0.11 (0.15) - 1.20 (1.34) + 0.12 (0.21) - 0.79 (1.02)	+ 0.05 (0.06) - 0.37 (0.40) + 0.04 (0.02) - 0.42 (0.41) + 0.04 (0.02) - 0.58 (0.67)	0.10 / 0.08 (foot) 0.99 / 0.025 (speed) 0.83 / 0.036 (interact)

Table 11. Group mean (SD) concentric, eccentric and total stance-phase work done at the intact hip, knee and ankle joints when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Results are listed downwards - slow, customary speed and fast. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Statistically significant differences are in **bold**.

	Speed Normalised J/kg.ms			Absolute Jkg ⁻¹		
	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value
Intact Hip Total work	0.21 (0.12) 0.24 (0.11) 0.29 (0.15)	0.25 (0.09) 0.26 (0.11) 0.36 (0.18)	0.19 (foot) 0.014 (speed) 0.57 (interact)	0.24 (0.11) 0.25 (0.11) 0.36 (0.13)	0.24 (0.08) 0.26 (0.12) 0.45 (0.36)	0.36 (foot) 0.027 (speed) 0.44 (interact)
Intact Hip Concentric work	0.18 (0.11) 0.16 (0.10) 0.17 (0.11)	0.18 (0.08) 0.16 (0.09) 0.20 (0.15)	0.74 (foot) 0.42 (speed) 0.41 (interact)	0.17 (0.08) 0.18 (0.09) 0.22 (0.11)	0.16 (0.07) 0.17 (0.08) 0.27 (0.24)	0.73 (foot) 0.09 (speed) 0.41 (interact)
Intact Hip Eccentric work	0.08 (0.04) 0.06 (0.03) 0.10 (0.05)	0.08 (0.04) 0.105 (0.06) 0.13 (0.08)	0.21 (foot) 0.12 (speed) 0.41 (interact)	0.07 (0.04) 0.07 (0.05) 0.14 (0.06)	0.08 (0.05) 0.10 (0.08) 0.18 (0.14)	0.20 (foot) 0.010 (speed) 0.48 (interact)
Intact Knee Total work	0.21 (0.06) 0.18 (0.08) 0.26 (0.09)	0.22 (0.11) 0.27 (0.12) 0.28 (0.09)	0.18 (foot) 0.030 (speed) 0.09 (interact)	0.21 (0.08) 0.21 (0.11) 0.36 (0.16)	0.17 (0.09) 0.29 (0.12) 0.38 (0.16)	0.46 (foot) < 0.001 (speed) 0.09 (interact)
Intact Knee Concentric work	0.06 (0.03) 0.06 (0.03) 0.08 (0.04)	0.06 (0.03) 0.09 (0.05) 0.09 (0.04)	0.19 (foot) 0.030 (speed) 0.06 (interact)	0.06 (0.03) 0.06 (0.04) 0.17 (0.06)	0.06 (0.03) 0.09 (0.05) 0.12 (0.06)	0.30 (foot) 0.001 (speed) 0.07 (interact)
Intact Knee Eccentric work	0.15 (0.04) 0.12 (0.06) 0.18 (0.06)	0.16 (0.09) 0.18 (0.08) 0.19 (0.06)	0.21 (foot) 0.07 (speed) 0.21 (interact)	0.14 (0.06) 0.15 (0.08) 0.25 (0.10)	0.15 (0.09) 0.20 (0.08) 0.27 (0.11)	0.27 (foot) < 0.001 (speed) 0.25 (interact)
Intact Ankle Total work	0.34 (0.18) 0.35 (0.18) 0.24 (0.11)	0.37 (0.20) 0.39 (0.21) 0.26 (0.13)	0.032 (foot) 0.011 (speed) 0.42 (interact)	0.32 (0.12) 0.40 (0.20) 0.33 (0.17)	0.33 (0.17) 0.40 (0.19) 0.36 (0.22)	0.25 (foot) 0.08 (speed) 0.64 (interact)

Intact Ankle Concentric work	0.17 (0.12) 0.20 (0.10) 0.13 (0.08)	0.19 (0.14) 0.22 (0.13) 0.14 (0.11)	0.17 (foot) 0.036 (speed) 0.89 (interact)	0.17 (0.13) 0.23 (0.12) 0.19 (0.13)	0.17 (0.12) 0.23 (0.13) 0.21 (0.19)	0.54 (foot) 0.10 (speed) 0.67 (interact)
Intact Ankle Eccentric work	0.16 (0.07) 0.15 (0.08) 0.11 (0.05)	0.18 (0.08) 0.17 (0.09) 0.12 (0.05)	0.003 (foot) 0.024 (speed) 0.24 (interact)	0.15 (0.06) 0.17 (0.09) 0.15 (0.06)	0.16 (0.06) 0.17 (0.08) 0.15 (0.07)	0.15 (foot) 0.63 (speed) 0.98 (interact)
Intact limb (all joints) total work	0.76 (0.21) 0.77 (0.27) 0.80 (0.25)	0.84 (0.28) 0.92 (0.24) 0.90 (0.26)	0.047 (foot) 0.36 (speed) 0.50 (interact)	0.73 (0.24) 0.88 (0.29) 1.08 (0.38)	0.77 (0.27) 0.97 (0.33) 1.24 (0.67)	0.15 (foot) < 0.001 (speed) 0.51 (interact)

Table 12. Group mean (SD) concentric, eccentric and total stance-phase work done at the residual hip and knee joints when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Results are listed downwards - slow, customary speed and fast. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Statistically significant differences are in **bold**.

	Speed Normalised J/kg.ms			Absolute Jkg ⁻¹		
	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value
Residual Hip Total work	0.22 (0.19) 0.22 (0.19) 0.24 (0.17)	0.21 (0.17) 0.22 (0.18) 0.23 (0.16)	0.44 (foot) 0.68 (speed) 0.82 (interact)	0.20 (0.15) 0.24 (0.16) 0.31 (0.16)	0.19 (0.14) 0.22 (0.16) 0.27 (0.16)	0.004 (foot) 0.003 (speed) 0.28 (interact)
Residual Hip Concentric work	0.15 (0.17) 0.15 (0.16) 0.15 (0.14)	0.15 (0.14) 0.15 (0.14) 0.16 (0.13)	0.75 (foot) 0.92 (speed) 0.51 (interact)	0.14 (0.12) 0.17 (0.14) 0.19 (0.14)	0.13 (0.12) 0.15 (0.12) 0.19 (0.12)	0.57 (foot) 0.009 (speed) 0.046 (interact)
Residual Hip Eccentric work	0.07 (0.04) 0.07 (0.05) 0.08 (0.05)	0.06 (0.04) 0.07 (0.06) 0.07 (0.05)	0.26 (foot) 0.26 (speed) 0.43 (interact)	0.06 (0.04) 0.08 (0.06) 0.11 (0.07)	0.066 (0.04) 0.07 (0.06) 0.09 (0.06)	0.14 (foot) 0.002 (speed) 0.45(interact)
Residual Knee Total work	0.08 (0.09) 0.10 (0.09) 0.08 (0.09)	0.08 (0.09) 0.07 (0.08) 0.06 (0.06)	0.052 (foot) 0.62 (speed) 0.32 (interact)	0.13 (0.14) 0.13 (0.15) 0.21 (0.24)	0.09 (0.09) 0.09 (0.10) 0.13 (0.14)	0.044 (foot) < 0.001 (speed) 0.32 (interact)
Residual Knee Concentric work	0.01 (0.01) 0.01 (0.01) 0.01 (0.01)	0.01 (0.01) 0.00 (0.00) 0.01 (0.01)	0.91 (foot) 0.90 (speed) 0.35 (interact)	0.01 (0.01) 0.01 (0.01) 0.01 (0.01)	0.01 (0.01) 0.00 (0.00) 0.01 (0.01)	0.99 (foot) 0.47 (speed) 0.36(interact)
Residual Knee Eccentric work	0.08 (0.09) 0.09 (0.09) 0.07 (0.09)	0.07 (0.09) 0.06 (0.07) 0.05 (0.06)	0.036 (foot) 0.48 (speed) 0.29 (interact)	0.07 (0.09) 0.10 (0.09) 0.10 (0.13)	0.07 (0.08) 0.06 (0.07) 0.07 (0.08)	0.025 (foot) 0.45 (speed) 0.19 (interact)
Residual limb (all joints) Total work	0.30 (0.20) 0.32 (0.24) 0.32 (0.18)	0.29 (0.17) 0.28 (0.18) 0.29 (0.17)	0.12 (foot) 0.96 (speed) 0.48 (interact)	0.27 (0.16) 0.35 (0.22) 0.41 (0.19)	0.26 (0.15) 0.29 (0.16) 0.36 (0.18)	0.024 (foot) < 0.001 (speed) 0.11 (interact)

8.3.3 Prosthetic 'ankle'

The timing of when the external moment at the distal end of the prosthetic-shank changed sign was affected by attachment category ($p = 0.029$). The moment changed from one tending to plantarflex to one tending to dorsiflex earlier, at 29% compared to 34% stance phase, when using the *hyA-F* device. The negative power integral in early stance was significantly affected by attachment category ($p = 0.009$); indicating more energy flowed from the shank to the foot (energy absorption) at all speed levels when using the *hyA-F*. The positive power integral during early stance (energy return) was also significantly affected by attachment category ($p < 0.001$); indicating less energy flowed into the shank from the foot when using the *hyA-F*. Despite there being no difference between foot conditions in the negative power integral during mid-stance ($p = 0.08$), the positive power integral during late stance was affected by attachment category ($p < 0.001$) and by a speed level-by-attachment category interaction ($p = 0.024$). There was less energy flow into the shank from the foot when using the *hyA-F* device, at all speeds ($p \leq 0.019$), with greater reduction at the customary speed level compared to both slow and fast speed levels.

Table 13. Group mean (SD) stance phase moment integrals at distal end of prosthetic shank (*pros-end*) when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Results are listed downwards - slow, customary speed and fast. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Statistically significant differences are in **bold**.

	Speed Normalised Nm/Kg.ms			Absolute Nms.kg ⁻¹		
	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value
External early-stance 'plantarflexion' moment impulse	-0.06 (0.03) -0.05 (0.02) -0.05 (0.02)	-0.06 (0.03) -0.06 (0.02) -0.05 (0.02)	0.36 (foot) 0.13 (speed) 0.81 (interact)	-0.05 (0.03) -0.07 (0.03) -0.06 (0.03)	-0.06 (0.02) -0.07 (0.02) -0.07 (0.02)	0.54 (foot) 0.028 (speed) 0.058 (interact)
External mid-late stance 'dorsiflexion' moment impulse	+0.43 (0.14) +0.31 (0.12) +0.27 (0.11)	+0.46 (0.13) +0.34 (0.10) +0.27 (0.08)	0.30 (foot) < 0.001 (speed) 0.12 (interact)	+0.34 (0.10) +0.36 (0.09) +0.35 (0.09)	+0.43 (0.07) +0.36 (0.06) +0.35 (0.05)	0.79 (foot) < 0.001 (speed) 0.12 (interact)

8.1 Discussion

The objective of this study was to compare speed-related joint kinetic alterations when using a hydraulically damped articulating attachment device compared to when the foot was attached rigidly without articulation. The speed of walking at all three freely chosen speed levels was higher when using the *hyA-F* (although only significantly so for customary speed walking), and thus to make the comparison of attachment conditions more balanced joint kinetic data were evaluated both normalised to walking speed and in absolute terms.

Table 14. Group mean (SD) stance phase power integrals at distal end of prosthetic shank (*pros-end*) when using rigid (*rigF*) and hydraulic (*hyA-F*) ankle-foot devices. Results are listed downwards - slow, customary speed and fast. Values normalised to walking speed are in the left-hand columns and absolute values in the right-hand columns. Statistically significant differences are in **bold**.

	Speed Normalised J/kg.ms			Absolute Jkg		
	Speed Normalised			Absolute		
	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value	<i>hyA-F</i>	<i>rigF</i>	<i>p</i> value
Negative work – early stance	- 0.017 (0.007) - 0.027 (0.009) - 0.024 (0.007)	- 0.012 (0.004) - 0.018 (0.007) - 0.019 (0.005)	0.009 (foot) 0.017 (speed) 0.55 (interact)	- 0.023 (0.009) - 0.031 (0.010) - 0.031 (0.012)	- 0.017 (0.007) - 0.020 (0.010) - 0.025 (0.010)	0.011 (foot) 0.005 (speed) 0.21 (interact)
Positive work - early stance	+0.003 (0.002) +0.003 (0.003) +0.008 (0.004)	+0.003 (0.001) +0.006 (0.004) +0.010 (0.004)	< 0.001 (foot) 0.007 (speed) 0.09 (interact)	+0.004 (0.003) +0.003 (0.004) +0.012 (0.004)	+0.004 (0.003) +0.007 (0.004) +0.014 (0.007)	0.011 (foot) < 0.001 (speed) 0.09 (interact)
Negative work – mid-stance	- 0.114 (0.053) - 0.140 (0.066) - 0.110 (0.042)	- 0.097 (0.046) - 0.129 (0.038) - 0.105 (0.035)	0.08 (foot) 0.36 (speed) 0.48 (interact)	- 0.148 (0.043) - 0.157 (0.049) - 0.143 (0.037)	- 0.124 (0.038) - 0.134 (0.029) - 0.137 (0.032)	0.043 (foot) 0.39 (speed) 0.13 (interact)
Positive work – late stance	+0.067 (0.025) +0.081 (0.031) +0.062 (0.027)	+0.081 (0.030) +0.109 (0.032) +0.087 (0.024)	< 0.001 (foot) 0.47 (speed) 0.024 (interact)	+0.090 (0.031) +0.093 (0.032) +0.087 (0.027)	+0.106 (0.032) +0.115 (0.032) +0.118 (0.038)	< 0.001 (foot) 0.47 (speed) 0.027 (interact)
Total work	0.201 (0.076) 0.251 (0.092) 0.204 (0.059)	0.193 (0.073) 0.263 (0.078) 0.222 (0.056)	0.22 (foot) 0.20 (speed) 0.09 (interact)	0.187 (0.060) 0.285 (0.071) 0.287 (0.116)	0.174 (0.050) 0.277 (0.060) 0.305 (0.110)	0.84 (foot) 0.023 (speed) 0.10 (interact)
Energy dissipated (negative + positive work)	0.061 (0.044) 0.083 (0.061) 0.065 (0.041)	0.026 (0.032) 0.031 (0.024) 0.027 (0.024)	0.001 (foot) 0.71 (speed) 0.43 (interact)	0.084 (0.034) 0.101 (0.047) 0.082 (0.037)	0.033 (0.036) 0.034 (0.025) 0.032 (0.025)	< 0.001 (foot) 0.046 (speed) 0.25 (interact)
Time of moment 'flip' (% stance)	28.5 (4.9) 30.8 (5.3) 28.4 (4.8)	32.9 (3.8) 36.1 (5.9) 34.0 (4.1)	0.029 (foot) 0.25 (speed) 0.31 (interact)			

While there were no differences between attachment categories in the amount of compensatory joint work at the intact-limb hip (absolute or normalised), the normalised peak power generation and total joint work at the

intact-limb ankle were reduced at all speed levels when using a *hyA-F*. Therefore the hypothesis that the compensatory joint power generation at the intact-limb hip and ankle would be reduced through use of the *hyA-F* was only partially supported.

It is noteworthy that in absolute terms there was no difference between attachment categories in total work at the intact-ankle across all speed levels (*hyA-F* $0.353 \pm 0.173 \text{ Jkg}^{-1}$, *rigF* $0.365 \pm 0.189 \text{ Jkg}^{-1}$, $p = 0.25$) but as walking speeds were higher when using the *hyA-F*, this resulted in there being a reduction in total ankle work done per meter travelled. The reduction in peak power generation during late stance at the intact-ankle when using the hydraulic attachment could possibly be a result of there being less resistance to forwards progression of the whole body centre of mass in early stance on the prosthetic-limb because the prosthetic-shank was able to rotate forwards more easily during this period (De Asha et al., 2013, Chapter 4). Thus, for the same amount of mechanical effort at the intact-limb ankle (and hip) there was greater centre of mass progression.

At the residual-knee the increased power absorption peak at customary speed when using the *hyA-F* and significant increase in (normalised and absolute) negative, eccentric work at all speed levels, indicate the residual-knee was more active/involved during weight bearing when using the hydraulic device, particularly so at the customary speed. To gain further insight into the increased residual knee loading/involvement when using a *hyA-F* knee flexion during loading response, which is typically reported to be

reduced on the residual side in trans-tibial amputees (Powers et al., 1998; Postema et al., 1997), and the ground reaction forces and residual-knee joint reaction force (both normalised to body weight, BW) during prosthetic-limb stance were retrospectively examined. There was significantly more angular displacement during loading response at the intact-knee compared to the residual-knee (intact $\sim 22^\circ$, residual $\sim 6^\circ$, $p = 0.001$), but there was no difference in residual-knee angular displacement during loading response across attachment categories or speed levels ($p \geq 0.49$). However, the residual-knee was in a more flexed position at initial contact (and thus throughout loading response) when using the *hyA-F* compared to *rigF*, and differences in residual-knee flexion angle between attachment categories increased with speed level (slow *hyA-F* $8.2^\circ \pm 3.2^\circ$, *rigF* $5.1^\circ \pm 6.2^\circ$; customary *hyA-F* $6.5^\circ \pm 3.9^\circ$, *rigF* $3.7^\circ \pm 6.9^\circ$; fast *hyA-F* $10.0^\circ \pm 5.0^\circ$, *rigF* $4.3^\circ \pm 6.5^\circ$, $p = 0.019$). This change in limb posture at initial contact could have been due to an increased 'dorsiflexion' angle at toe-off due to the articulation provided by the hydraulic device which brought the prosthetic-shank forwards, or due to a drive to allow the residual-knee to be loaded more because of perceived increased comfort - previous research has shown that in-socket pressures are reduced when using the *hyA-F* (Portnoy et al., 2012) which suggests increased comfort levels. When the *hyA-F* was used there was a significant increase in both the peak vertical ground reaction force (*hyA-F*, 1.05 ± 0.14 BW, *rigF*, 1.02 ± 0.14 BW, $p = 0.007$) and the vertical ground reaction force impulse (*hyA-F*, 0.60 ± 0.09 BW, *rigF*, 0.57 ± 0.09 BW.s, $p = 0.001$), accompanied by a significant increase in the axial (relative to the shank) residual-knee peak joint reaction force (*hyA-F*, $1.15 \pm$

0.18 BW, $rigF$, 1.11 ± 0.017 BW, $p = 0.044$), indicative of increased weight bearing. Consequently, the hypothesis that the residual-knee would be loaded more during weight bearing was supported. Previous research has reported the power absorption peak to be significantly reduced at the residual-knee compared to that at the intact-knee (Powers et al., 1998; Sadeghi et al., 2001). Thus the findings of the present study suggest that the hydraulic device had an important and clinically meaningful effect on how the residual-knee was loaded. It is worth emphasising however, that no statistical comparisons were made between the joint kinetics of the intact- and residual-knees due to the differing modelling approaches used for each limb. However, it is important to highlight that the magnitude of the moments and powers at the residual-knee were very small in comparison to values at the intact-knee.

At the distal end of the prosthetic-shank, net power flow presented a double bi-phasic pattern of power absorption and return irrespective of attachment category and across speed levels (Figure 18). Such a double bi-phasic pattern was reported by Prince et al. (1994). These two periods must have corresponded respectively with the heel and then the forefoot keel elastically deforming and recoiling. While compression of the heel keel following initial contact plays a role in increasing comfort and allowing the prosthetic forefoot to lower to the ground (Postema et al., 1997) the energy return, during early-to-mid stance, associated with its subsequent recoil is an unusual phenomenon when compared to intact-ankle joint kinetics. Such energy return could not have directly contributed to gait propulsion as it occurred at

approximately 20% of stance, and indeed may have affected gait inappropriately/negatively; potentially causing an early heel rise and/or a 'bouncing' or unstable sensation. This inappropriate energy return from the foot, which increased with speed level for both attachment categories, was significantly reduced at all speed levels when participants used a *hyA-F* (normalised and absolute). The change from inappropriate energy return (recoil) to absorption was synchronous with the external moment switching from one tending to plantarflex to one tending to dorsiflex. This moment 'flip' occurred approximately 5% earlier in stance at all speed levels when using the *hyA-F* (i.e. at 29% of stance compared to 34% of stance when using the *rigF*), and occurred at a similar time to that reported previously for overground gait in trans-tibial amputees (~ 30% of stance, Schmalz et al., 2002; Ventura et al., 2011) but later than that seen in able-bodied gait (~ 9% of stance, Kirtley, 2006). The earlier moment 'flip' when using the *hyA-F* may have been due to the device preventing the centre of pressure trajectory 'stalling' under the prosthetic hindfoot during its progression from the heel to the toe region (Schmid, 2005; De Asha et al., 2013, Chapter 4). In turn this resulted in the centre of pressure passing anterior to the distal end of the prosthetic shank sooner. With the increase in power absorption and reduction in power return more energy was dissipated by the *hyA-F* than the *rigF* across all speed levels. Such energy dissipation is a feature of hydraulic damping. Likely as a result of such damping, many of the participants commented that when using the *hyA-F* they no longer felt they had to "climb over" the prosthetic-foot or commented they felt less resistance to forward progression. A feeling of improved ability to 'get onto and over' the prosthetic-

foot suggests that the hydraulic attachment attenuates/reduces the 'braking' effect the foot conveys to centre of mass progression. This would further explain why walking speed increased and there was a reduction in intact-limb joint work done per meter travelled when using the *hyA-F*. A reduction in work per meter travelled also suggests use of an *hyA-F* may potentially result in a reduction in metabolic energy costs, and this should be investigated in future work.

Regardless of speed level or attachment category, during early stance the magnitude of P_{rot} (positive) was smaller than that of P_{trans} (negative) yielding a net negative power flow as the heel deformed. Once the foot became plantigrade, P_{rot} increased relative to P_{trans} (Figure 15) as the distal end of the shank became the fulcrum of shank rotation above the foot. At this time there was a net positive power flow which must have been due to the recoil of the heel keel. However, when the *hyA-F* was used the earlier moment 'flip', and relative increase in P_{rot} , tended to reduce this period of positive power flow at all speed levels. When using the *hyA-F* the heel keel would have still recoiled but some of the energy returned was likely dissipated within the hydraulic unit rather than being transferred to the shank segment. As a result there was almost continuous negative power flow for the first three-quarters of stance. The earlier moment 'flip' when using the *hyA-F* signified that the passive control exerted on the forward rotating shank (external moment tending to dorsiflex) occurred sooner when using the device which is consistent with the increased time of negative power flow. This indicates the *hyA-F* provided

'ankle' function that was more akin to that typical of able-bodied gait (Winter, 1995; Neptune et al., 2001; Kirtley, 2006).

There were no significant differences in the normalised negative power associated with forefoot compression (i.e. during mid-to-late stance period) across speed levels or between attachment categories. This is, perhaps, unsurprising given that the *hyA-F* reaches its limit of articulation during mid-stance and thus the behaviour of the foot in late stance should be similar for both attachment categories. Having said this, the positive power flow as the forefoot keel recoiled, which mimics the A2 power burst seen in able-bodied gait, was significantly reduced at all walking speed levels when using the hydraulic attachment (normalised and absolute). There is no obvious explanation for this, but it is possible that the increased negative work at the residual-knee facilitated power flow into the shank from the thigh, which in turn may have contributed to the net power flow observed at the distal end of the prosthetic-shank.

The rate at which the hydraulic unit articulates during stance is a function of walking speed and level of hydraulic damping. This means the 'setting up' process conducted by the prosthetist is paramount for optimal functioning of the device. The late stance power flow into the shank was higher at customary walking speed than the other speed levels when using the *hyA-F* which suggests its time-dependent nature contributed indirectly to the energy returned by the prosthetic-foot. The differences observed in the energy absorbed, and dissipated or returned, by the *hyA-F* and *rigF* were of a much

smaller magnitude than the kinetic changes observed on the intact-limb. This highlights how small alterations of a prosthetic device can have profound effects further along the kinetic chain.

Irrespective of attachment category, the normalised peak power generation and total joint work at the intact-ankle became reduced as speed level increased, but there were simultaneous increases in (normalised and absolute) total joint work at the intact-limb knee and hip. These speed effects are similar to those from previous studies investigating speed-related joint kinetics changes in able-bodied gait (Chen et al., 1997; Riley et al., 2001). As knee kinetics play a limited role in gait propulsion (Paul, 1970), these findings suggest that amputees, like able-bodied individuals, predominantly alter hip kinetics in order to increase walking speed. In the present study there was no significant increase with speed level in the normalised residual-hip peak power absorption, generation or total joint work, indicating that amputees relied predominantly on the intact-limb hip for increased propulsion, regardless of prosthetic foot attachment used. These findings are consistent with the findings of Powers et al. (1998) who reported residual-limb hip peak power generation to be lower than that for the intact-limb hip at customary speed walking.

8.2 Conclusions

Findings indicate that a trans-tibial prosthesis incorporating a dynamic-response foot attached via an articulating hydraulic device reduced the speed-related increases in compensatory intact-limb joint kinetics compared

to when the foot was attached rigidly. In addition, residual-knee loading/involvement and weight bearing on the prosthetic-side were increased when using the hydraulic device. Differences between attachment types were highest at the customary speed level which indicates the hydraulic ankle-foot device, like other passive prosthetic devices, is most effectual at the walking speed it is set up for. Finally, the kinetic changes observed at the prosthetic ankle-foot were of a much smaller magnitude than those observed in the intact limb, which highlights how small alterations at the prosthetic device can cause relatively large kinetic effects in the intact-limb.

Chapter 9. Final Discussion and Conclusions

9.1 Final Discussion

The experiments within this thesis were conducted in order to determine the effects of using a hydraulically damped, uniaxial articulating prosthetic ankle-foot device on the biomechanics of overground walking in UTAs compared to use of non-hydraulic devices. The principle findings were that use of the hydraulic device compared to non-hydraulic devices resulted in increases in participants' walking speeds which occurred despite a reduction in mechanical work done by the intact-limb per meter travelled. Other findings demonstrated that centre of pressure progression beneath the prosthetic-foot was less disrupted and that eccentric loading of the residual-knee and prosthetic-limb weight-bearing increased during stance when using the hydraulic device.

Use of the hydraulic ankle-foot device led to a number of significant changes in the lower-limb kinetics and kinematics of UTAs. These changes were observed across both limbs. In the first experimental chapter (Chapter 4) the principal finding was that as a result of the device's use there was an increase in self-selected customary walking speed. Walking speed is a widely used measure of overall gait quality (Holden et al., 1986; Guralnik et al., 1999; Cesari et al., 2005; Schmid et al., 2007). Typically, amputees walk more slowly than the able-bodied (Nolan et al., 2003) and at their most energetically efficient speed (Barth et al., 1992; Colborne et al., 1992). This would suggest that the significantly higher walking speeds due to use of a hydraulic device can be considered as an 'improvement' in gait quality and

were due to some energetic, as well as mechanical, reason. Although measures of energy consumption were beyond the scope of this thesis the findings of the last experimental chapter (Chapter 8), that there was less mechanical work done at the joints of the intact-limb per meter travelled when using the hydraulic device, suggest that there may well be some energetic benefit from its use. Metabolic cost was assessed previously for a similar adaptive ankle-foot device, the Proprio-Foot (Delussu et al., 2013), and it was found that there was a benefit compared to a rigidly attached foot when participants walked overground at their customary speeds. Like the Echelon, the Proprio-Foot is heavier than a rigidly attached foot device. During the final experimental chapter the findings demonstrated that the increase in walking speed and reduction in intact-limb joint work per meter travelled occurred despite the extra mass of the Echelon and the device absorbing more, and returning less, energy than a rigidly attached foot. One possible explanation, therefore, is that rather than providing additional propulsion the hydraulic control of articulation allowed the whole body centre of mass to progress forwards more easily. As the progression of the centre of pressure reflects that of the centre of mass, this suggestion is supported by the reduction in inappropriate fluctuations to centre of pressure progression observed beneath the prosthetic-limb in the first experimental chapter. As stated above, measurement of metabolic cost during ambulation was beyond the scope of this thesis however it should be a direction of any future research regarding the effects of hydraulic attachment of a prosthetic-foot device.

UTAs typically display reduced loading response knee flexion on the residual-limb (Gitter et al., 1991; Powers et al., 1998; Beyaert et al., 2008) which is associated with lower moments and powers compared to an intact-limb (Seroussi et al., 1996; Sanderson & Martin, 1997; Powers et al., 1998). They also bear less weight on the prosthetic-side (Nolan et al., 2003). This kinetic asymmetry, 'favouring' the intact-limb, is reportedly a contributory factor in amputees developing osteoarthritis in the intact-knee (Norvell et al., 2005; Royer & Konig, 2005; Royer & Wasilewski, 2006) and lower-back pain (Khodadadeh et al., 1988; Gaunard et al., 2011). Use of the hydraulic device was demonstrated in the final experimental chapter (Chapter 8) to result in increased loading response residual-knee flexion, which increased with higher walking speed. This was in conjunction with increased eccentric work, peak power absorption and peak longitudinal joint reaction forces at the residual-knee and increased peak vertical ground reaction force and vertical ground reaction force impulse during prosthetic-limb stance. It was shown during the fourth experimental chapter (Chapter 7) that this increased stance-phase knee flexion continued into mid-stance, at the point of intact-limb minimum toe clearance. This higher level of flexion may well be due, partly at least, to the higher rotational velocity of the prosthetic-shank during early stance reported in the first experimental chapter (Chapter 4). If so it would appear that the time-dependent nature of the hydraulic device's articulation is the ultimate driver of the changes at the residual-knee. It may well be that this time-dependent articulation is also driving the reductions in in-socket pressure reported by Portnoy et al. (2012). While it is speculation, this increased involvement in weight bearing on the prosthetic-side, seen as a

result of using a hydraulic device, could potentially play some part in reducing the incidence or severity of intact-limb knee osteoarthritis and/or lower back pain among UTAs.

9.2 Limitations and Future Research Directions

One potential limitation of this thesis was the participant population. Participants were active and otherwise healthy UTAs. This group is at the high end, functionally, of lower limb amputees and makes up approximately 10% of the UK amputees (NASDAB). The Echelon, hydraulic ankle-foot device which was at the core of the experiments is designed and intended for use by those who have “*the ability or potential for ambulation with variable cadence.... the ability to traverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic utilization beyond simple locomotion or the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels.*” (www.endolite.co.uk). This description does not exclude trans-femoral amputees who were not included as participants in the experiments conducted as part of the thesis. Future studies should, perhaps, therefore investigate whether the effects seen as a result of using the Echelon hydraulic ankle-foot device are also present for active trans-femoral amputees. While this ‘narrowness’ of participant group may be perceived as a limitation it was done to avoid potential null/confounding findings due to groups of mixed functional levels such as those described in Chapter 2 (e.g.

Torburn et al., 1990; Barth et al., 1992; Mizuno et al., 1992; Nielson et al., 1989)

Another potential limitation was the limited accommodation time afforded to UTA participants after the hydraulic device was attached to their prostheses. Each used the device for a minimum of 45 minutes prior to data collection and while differences were observed due to the device these differences could possibly be magnified or reduced through a longer period of habituation. However all overground walking data were collected from each participant during a single session. Thus there were no other factors which may have influenced the kinetic and kinematic differences observed between prosthetic conditions other than the function of the ankle-foot devices used. During the experimental protocols throughout this thesis participants undertook all walking trials on a flat and level surface. Another direction for future research should be to investigate ambulation using a hydraulic ankle-foot device over other surfaces such as ramps and stairs.

Like all biomechanical investigations into amputee gait the findings of this thesis must be viewed within the context of the modelling processes undertaken. Current models with assumptions of rigid segments and pin-joint articulation are inherently flawed with respect to the physical reality of prosthetic device function. It may be that in future the development of 'prosthetic-specific' modelling should be undertaken. While the techniques of Prince et al. (1994) and Dumas et al. (2009), used throughout this thesis, appear to be more ecologically valid than a more 'traditional' inverse dynamic

approach they are still limited by rigid-segment assumptions. Takahashi et al. (2012) propose a deformable-segment approach to measuring below-knee kinetics which, intuitively, appears to be the direction in which future modelling techniques should be accomplished. This, together with some method of accounting for movements at the stump-socket interface, should be a realistic and achievable aim of those involved in lower-limb amputee gait research.

9.3 Final Conclusions

Findings indicate that use of a trans-tibial prosthesis incorporating a dynamic-response foot attached via an articulating hydraulic device has biomechanical benefits for active, unilateral trans-tibial amputees. Its use resulted in increased self-selected walking speeds, a primary measure of gait quality, while simultaneously reducing speed-related increases in compensatory intact-limb joint kinetics. This reduction in mechanical work done per meter travelled suggests there may be a reduction in the metabolic cost of gait associated with the device's use compared to more traditional methods of attachment. In addition, during prosthetic-limb stance, inappropriate centre of pressure fluctuations were reduced while residual-knee loading/involvement and weight bearing were increased when using the hydraulic device. Differences between attachment types were highest at the customary speed level which indicates the hydraulic ankle-foot device, like other passive prosthetic devices, is most effectual at the walking speed it is set up for. Finally, this thesis has highlighted that it is possible to quantify prosthetic

ankle-foot 'performance' using surrogate measures and thus without modelling an 'ankle' joint on the prosthetic-limb.

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Appendices

Study title

The effects on gait of using a hydraulic versus fixed ankle device in unilateral trans-tibial amputees.

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.

What is purpose of the Study?

To determine, how gait function is affected by alignment alterations and changes in walking speed in unilateral amputees when using hydraulic and fixed ankle devices.

Why have I been chosen?

You have been chosen because you are healthy and are a unilateral trans-tibial amputee, capable of using a hydraulic foot-ankle device

Do I have to take part?

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 in anyway in the future.

What will happen to me if I take part?

You will be asked to visit the Biomechanics Laboratory at the University of Bradford for a single visit lasting up to 4 hours. During this visit you will use both a hydraulic and fixed ankle device. These will be fitted to you current prosthesis by an experienced prosthetist

During the session you will be asked to undergo or complete the following:

- *Have your height and weight measured*
- *Walk normally up and down the laboratory while using both a hydraulic and fixed ankle device.*

During these tests a number of small spherical, reflective markers will be attached to you skin or clothing using medical grade tape.

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

After the visit your original prosthesis set-up will be reinstated before you leave.

What do I have to do?

You will be asked to maintain your usual diet and activity level and refrain from drinking alcohol immediately before your visit and to bring with you shorts, a T-shirt, and comfortable flat soled shoes. When you arrive you will be asked to change into your shorts, T-shirt and comfortable shoes. You will be asked to complete the above tasks using the two types of prosthesis set-up. Each time an alteration is made you will be allowed time to become familiar and comfortable with it.

Is there any risk of harm to myself?

There is a hypothetical risk of you losing your balance when performing the walking tasks, but no more than during your normal daily activities. There is a possibility that you may experience some small discomfort when you first use your prosthesis with an unfamiliar set-up, but this will be no different to what

you have experienced in the past when you have been given a new prosthesis.

Will my taking part in the study be kept confidential?

Your confidentiality will be safeguarded during and after the study. All personal information will be held securely and destroyed at the study's conclusion.

Contact Details of Researcher

Alan De Asha
University of Bradford
Phone: 01274 385926
E-Mail: A.R.DeAsha@student.bradford.ac.uk

If you have any issues which you would wish to discuss or would like like *more information regarding the study please feel free to contact the study supervisors ;*

Dr. John Buckley	Dr. Louise Johnson
University of Bradford	University of Bradford
Phone: 01274 234641	Phone: 01274 236587
E-mail: J.Buckley@bradford.ac.uk	E-mail: L.Johnson4@bradford.ac.uk

Appendix C. Participant base-line data collection form.

Participant code..... Participant initials..... Date.....

**The effects on gait of using a hydraulic versus fixed ankle device in unilateral trans-tibial amputees
BASELINE DATA PROFORMA**

Are you feeling well today Y / N
 Consent form signed Y / N
 Meets inclusion criteria Y / N

dob Age Gender M/ F

side L / R time since amp cause

Medication Y / N
 Details

Issues with residuum Y / N
 Details

Phantom symptoms Y / N Pain Y / N Frequency

How long had current prosthesis
 Hours per day typically worn
 Socket
 Foot
 Liner

Weight (kg) Height (m)

Toe	Vertical	L (cm)	R (cm)
	Horizontal	L (cm)	R (cm)
Heel	Vertical	L (cm)	R (cm)

Residuum (cm) Length Circumference Prox Dist

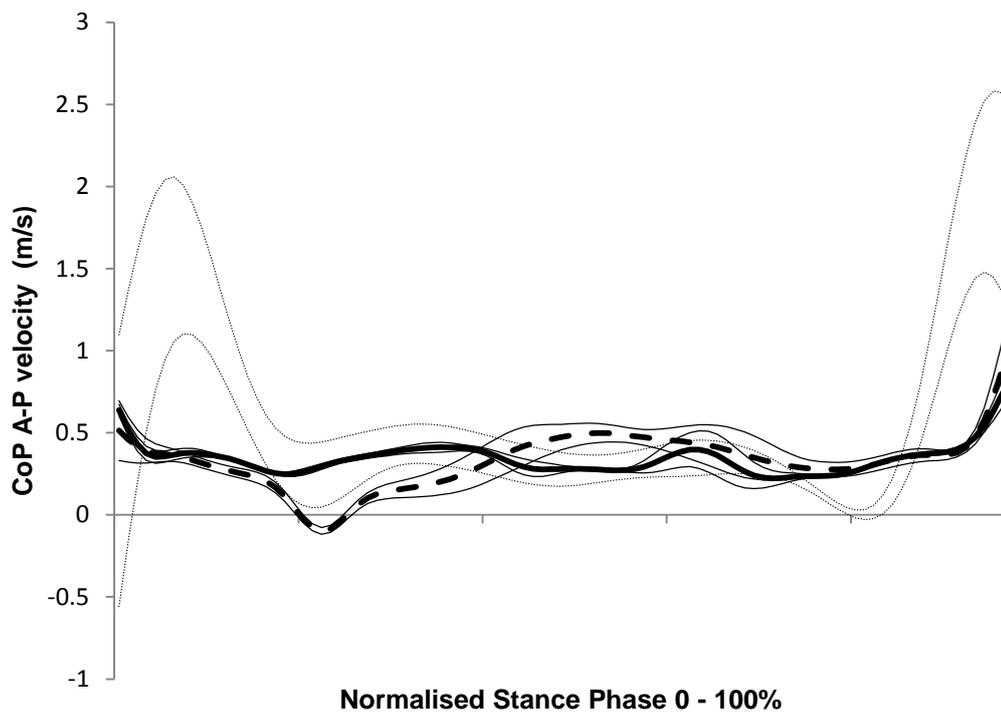
Hydraulic settings PF / DF/.....

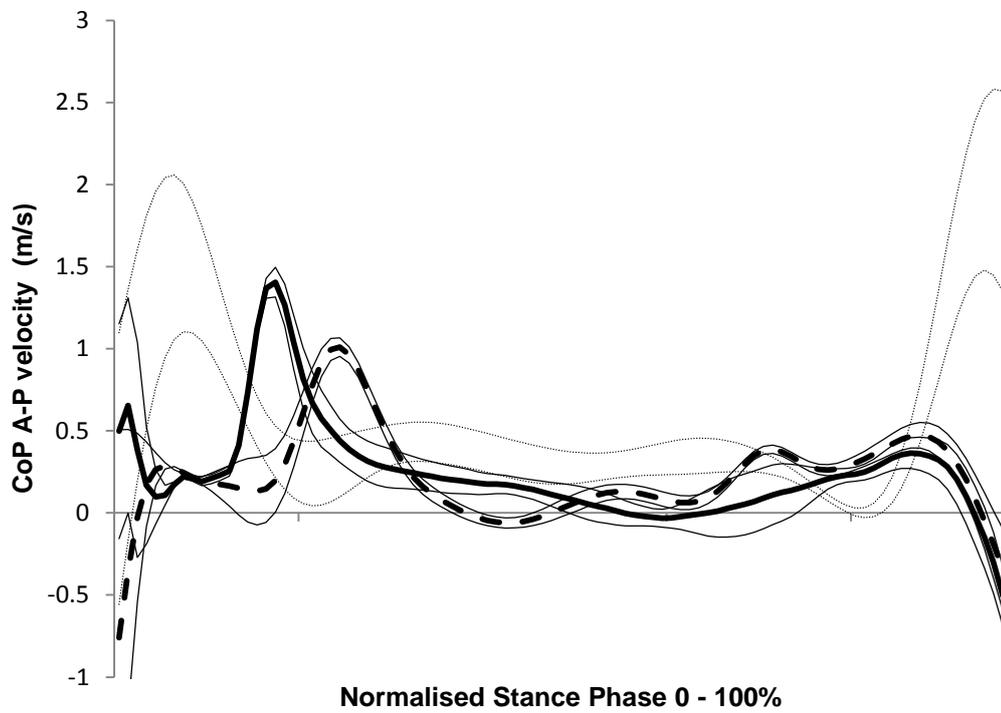
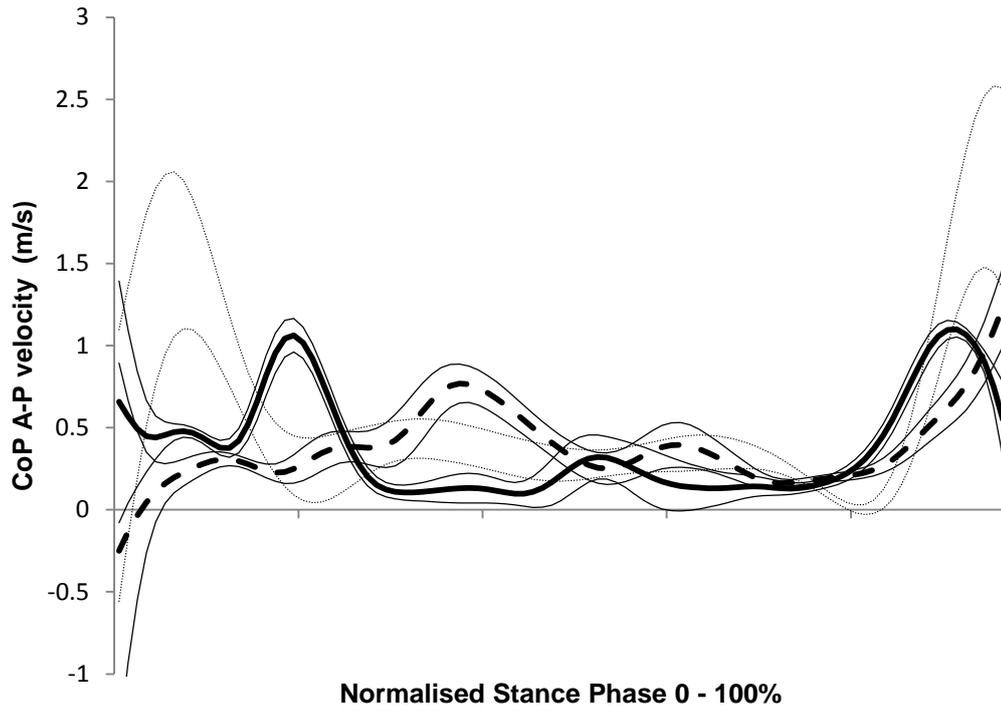
Vision correction worn during walking None / Spectacles / Con Lenses

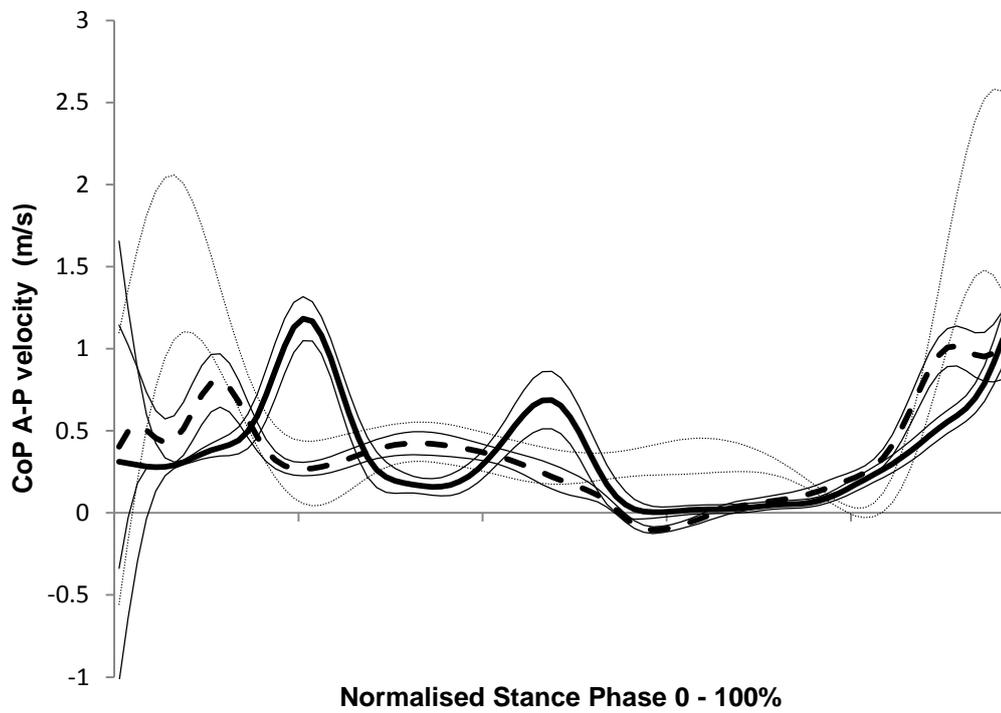
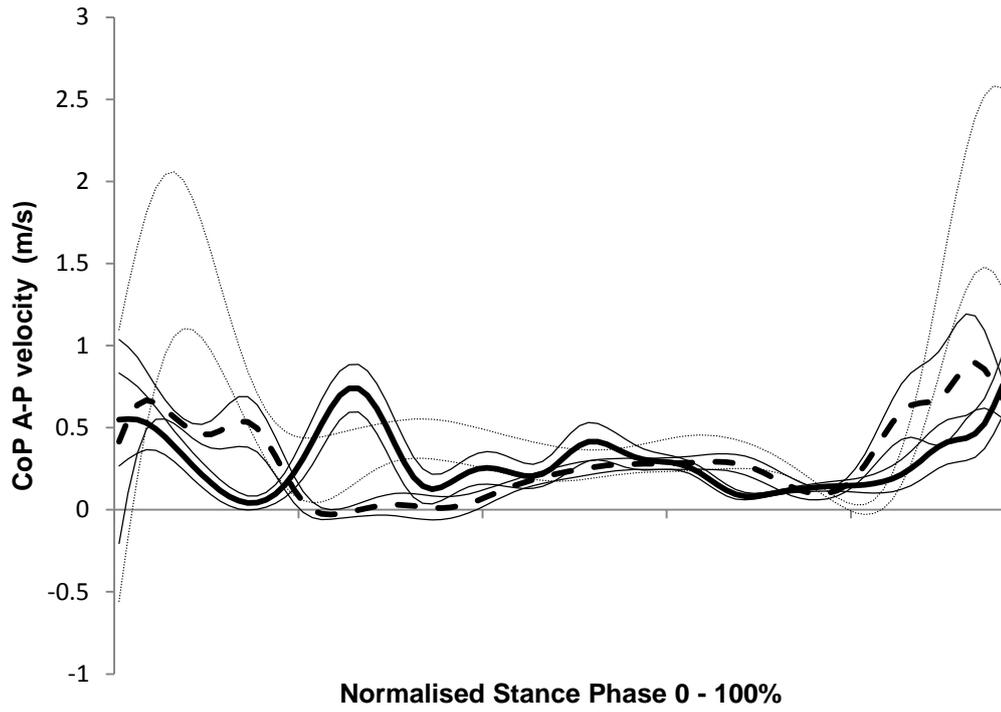
LogMAR (Snellen) Lighthouse binocular visual acuity

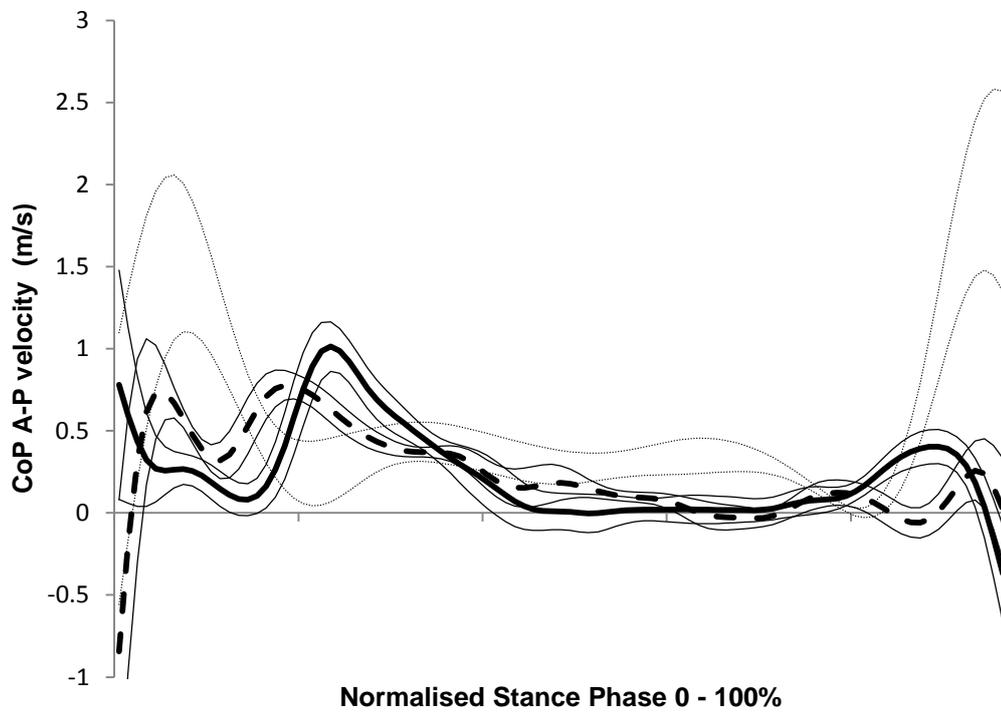
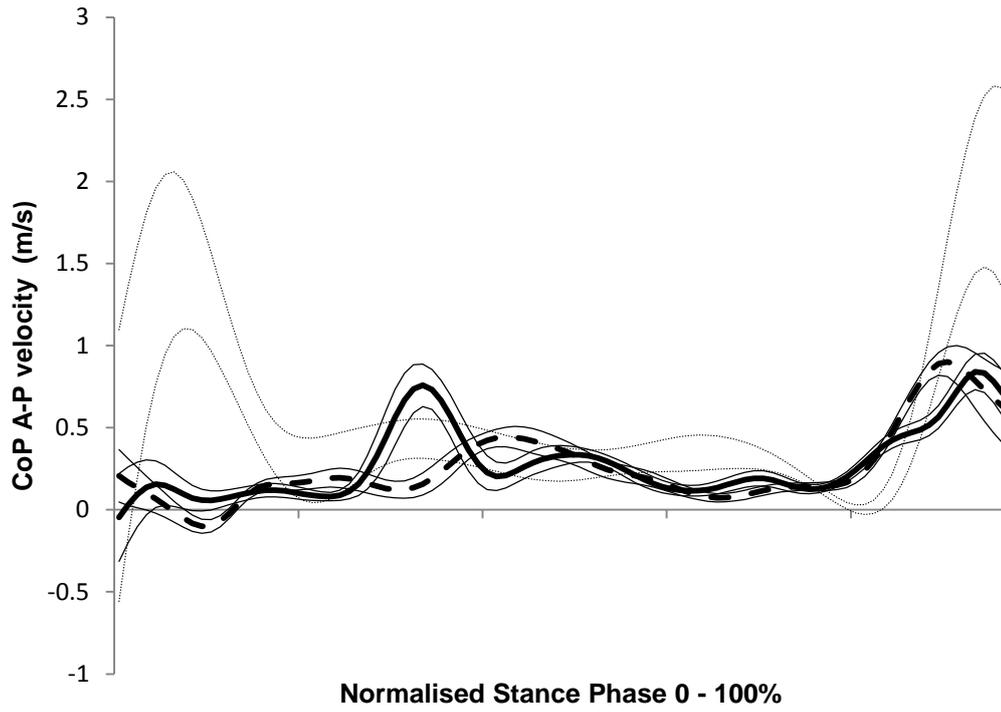
*Appendix D. Individual Participant centre of pressure velocity figures
(Chapter 4)*

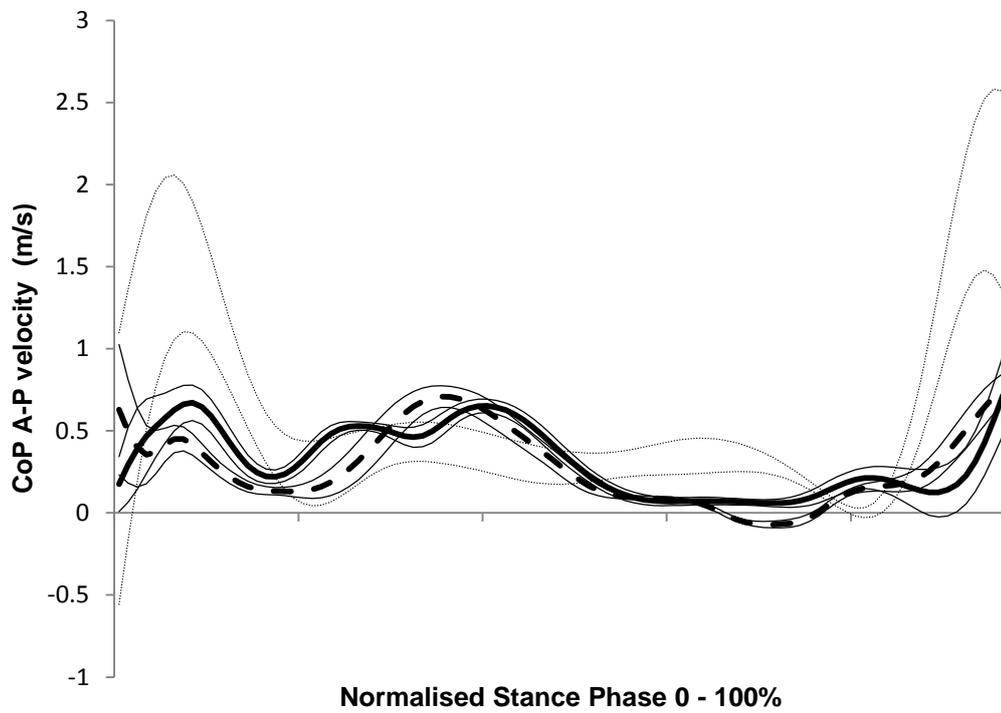
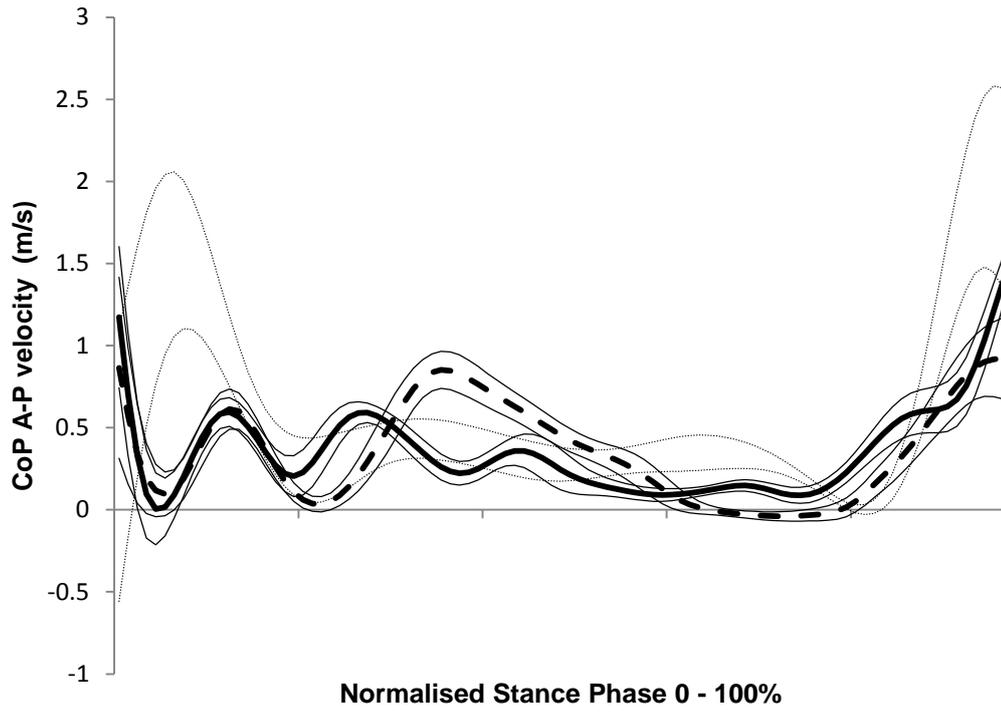
Mean (SD) CoP A-P velocity of the 10 repeat trials, normalised to full stance phase, for each participant when using a *hyA-F* (solid line) and *habF* (broken line). Able-bodied control group CoP velocity ± 1 SD ribbon is shown (dotted lines) for reference purposes.

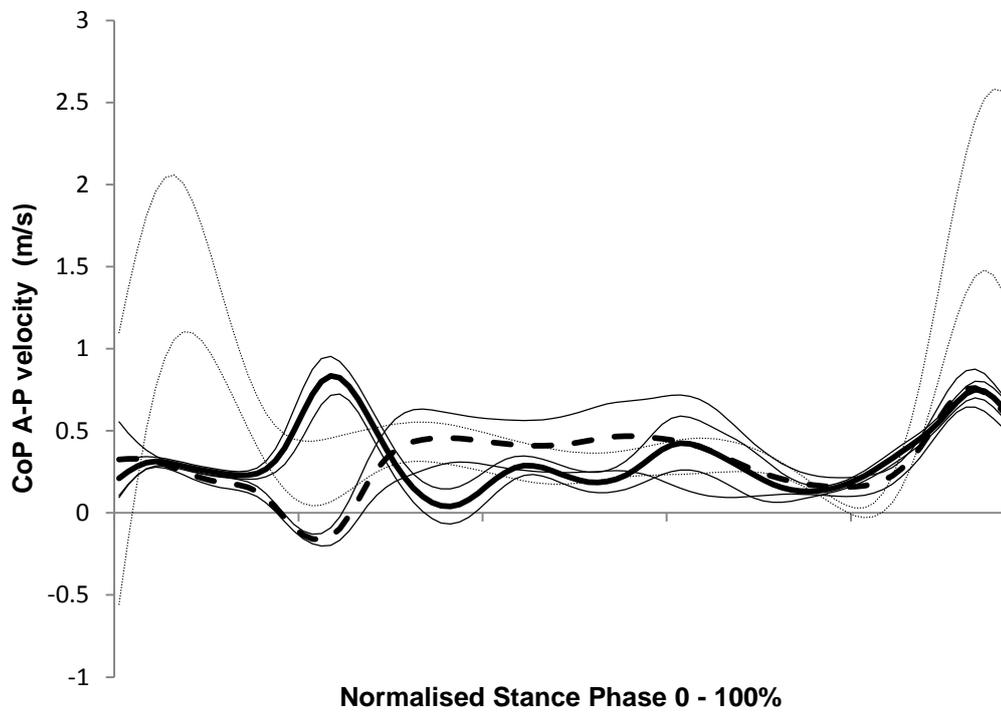
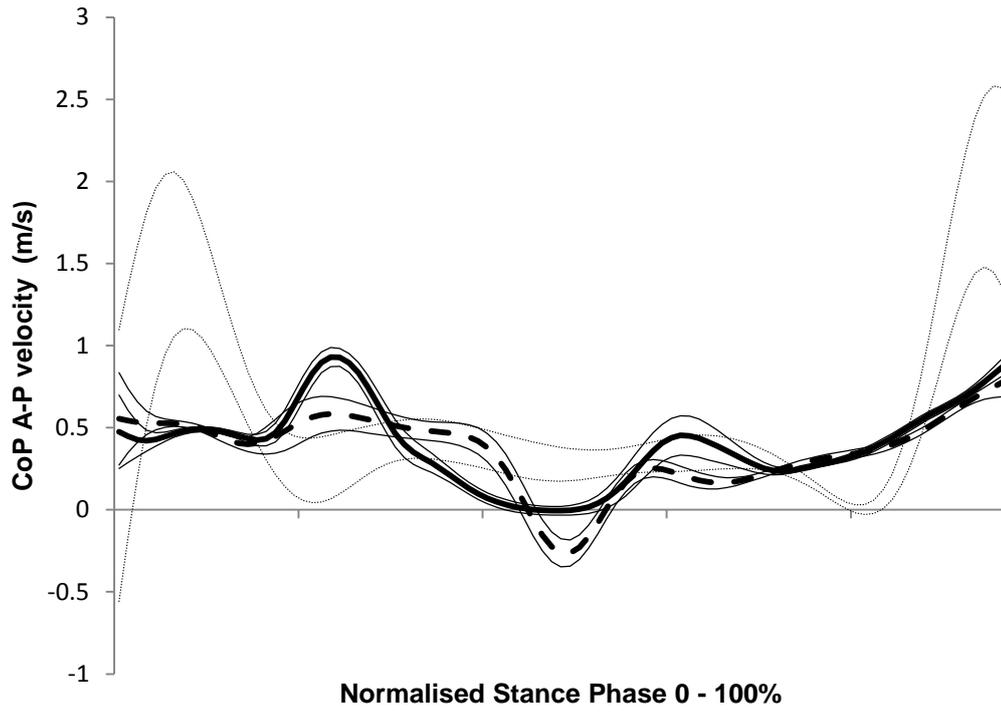


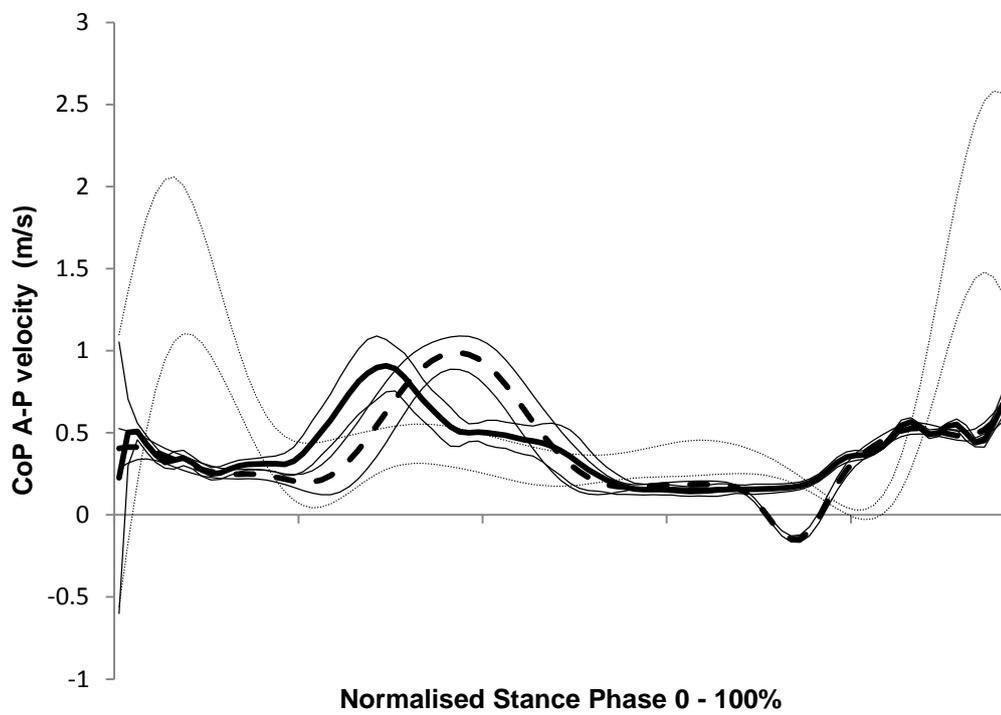
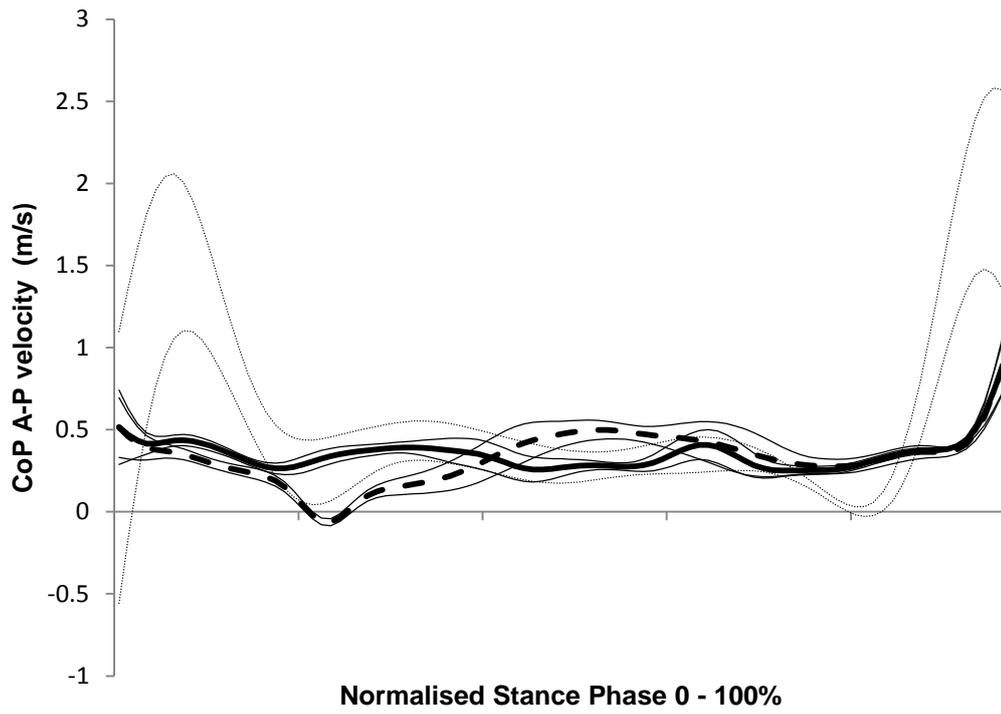


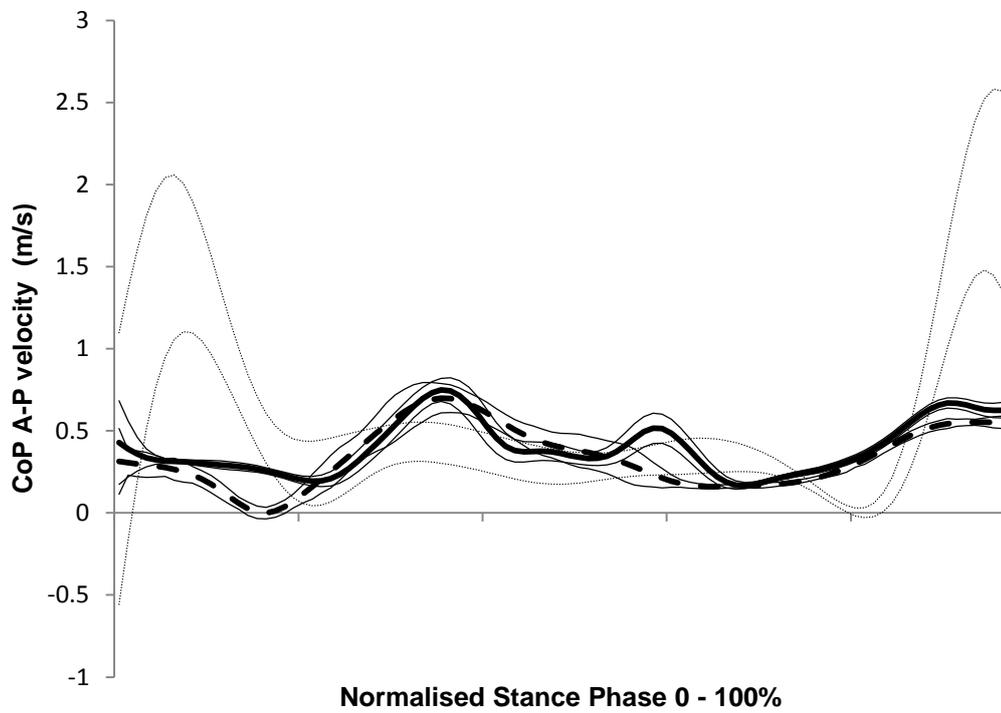
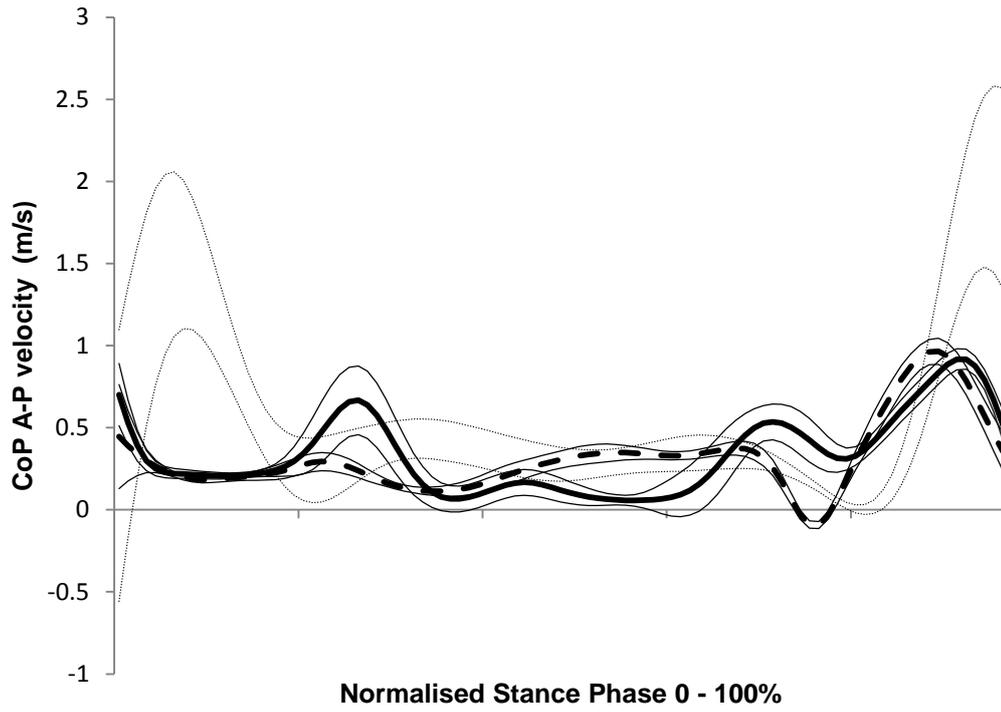


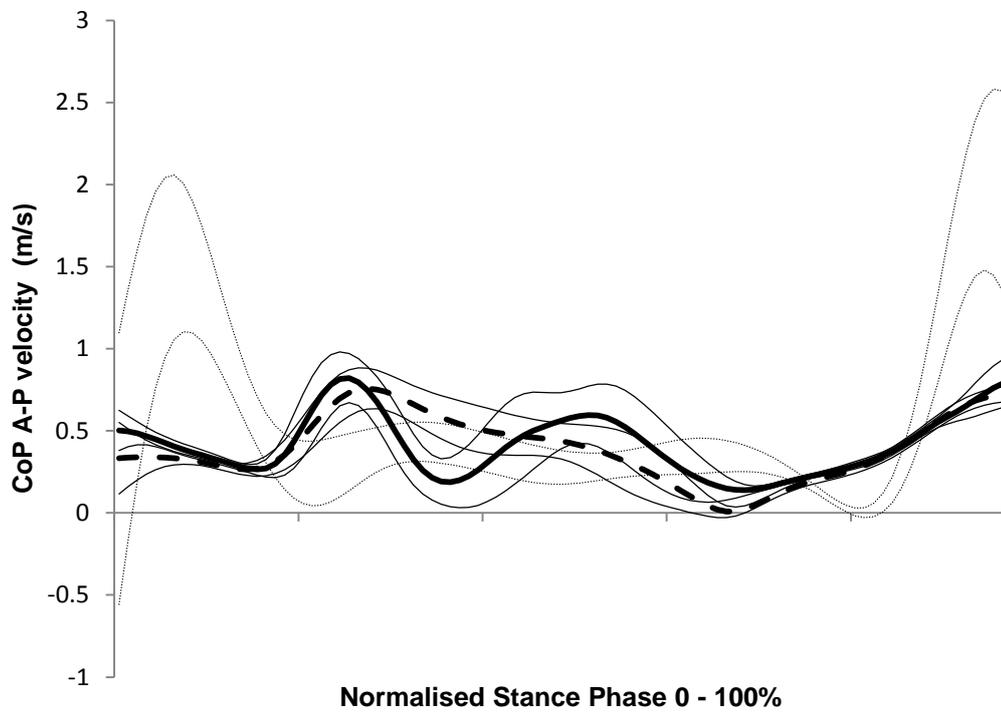
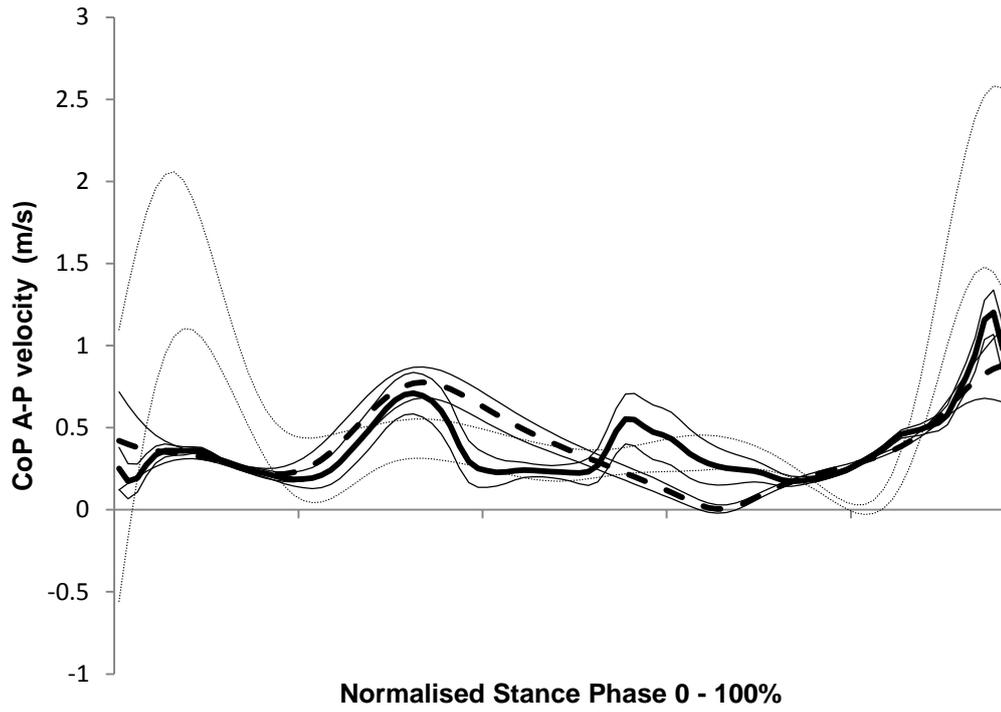


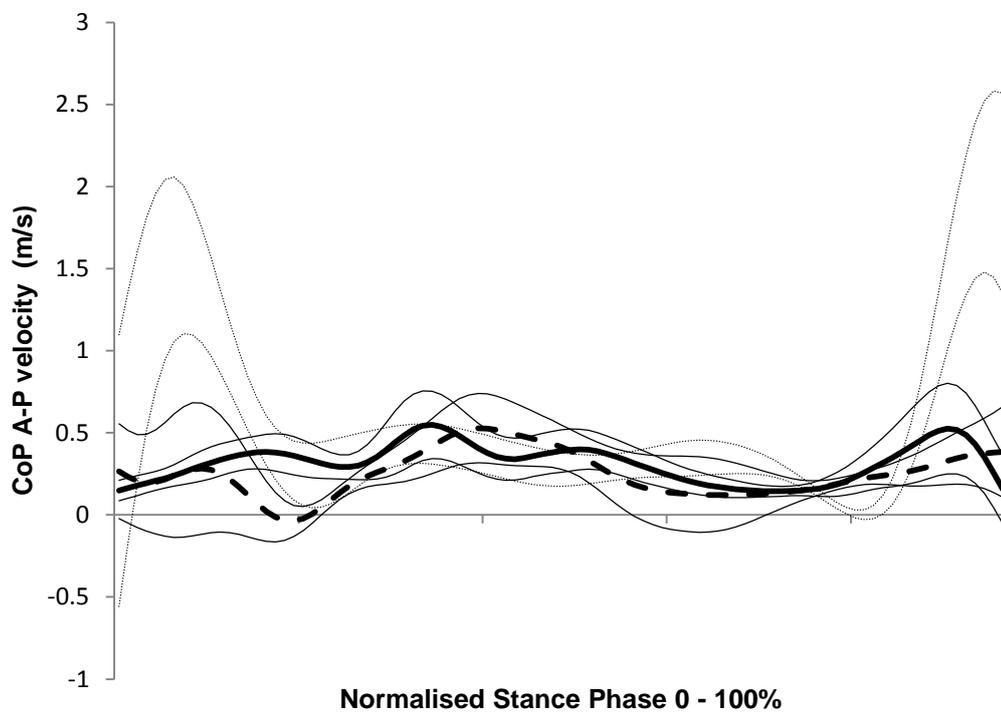
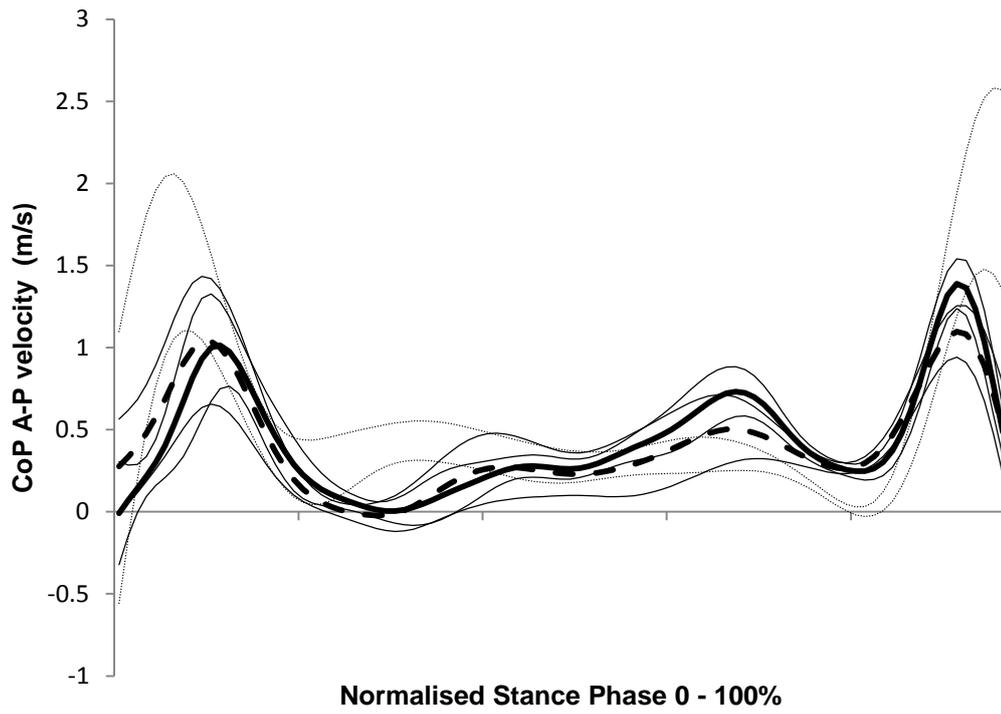








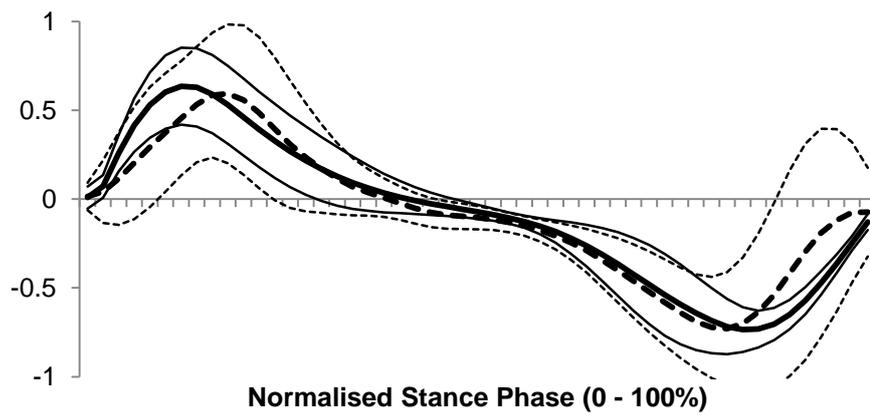
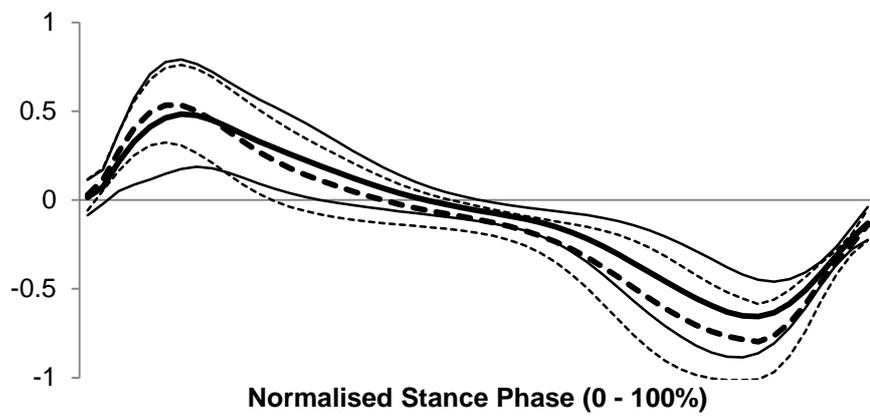
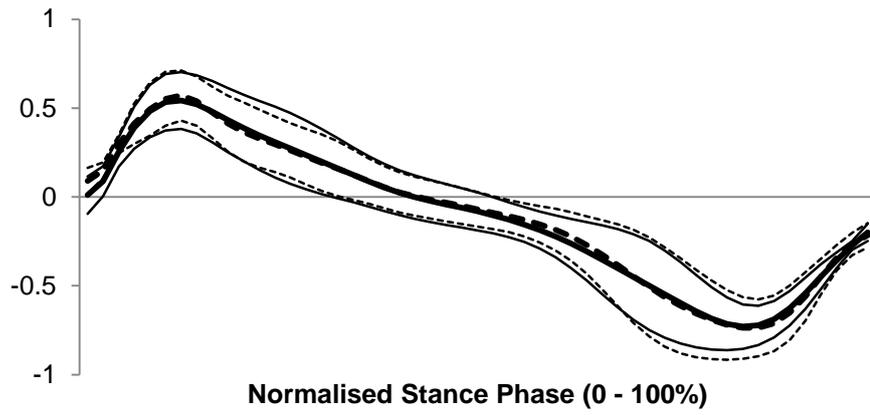




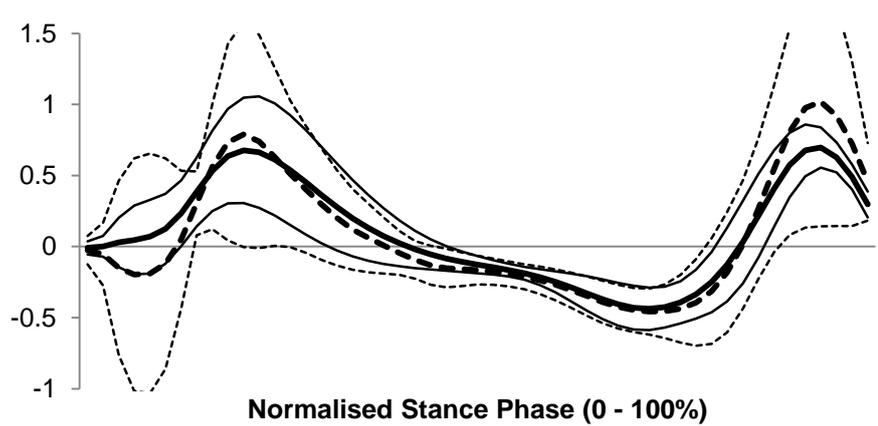
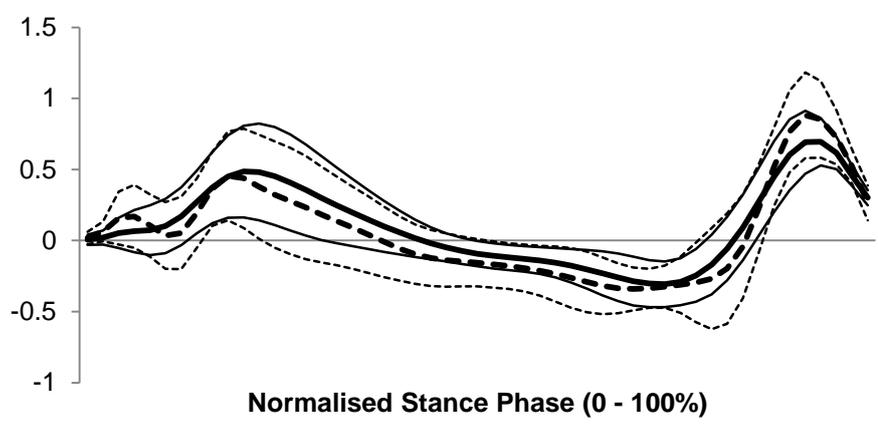
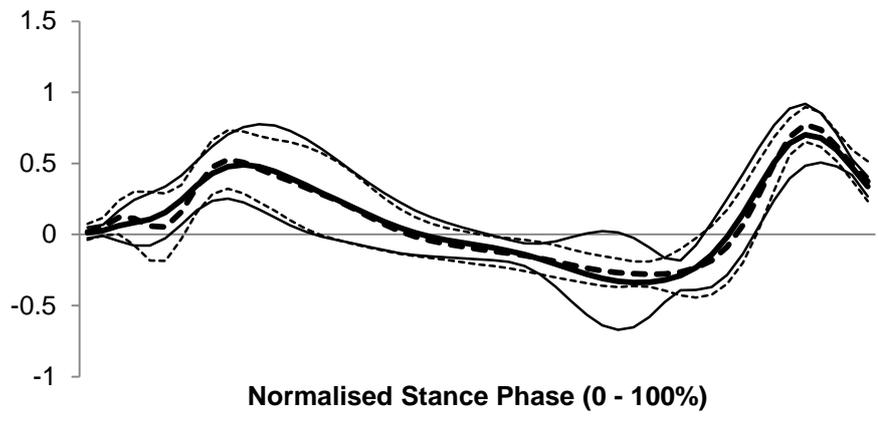
Appendix E. Larger scale plots of moment and power figures at all intact and residual joints and at pros end (Chapter 8).

All figures display ensemble Mean (SD) speed-normalised joint kinetics across a normalised stance phase for participants using a *hyA-F* (solid line) and a *rigF* (dotted line). For each individual joint, kinetics are presented as slow walking speed (top), customary walking speed (centre) and fast walking speed (bottom). Units are listed within the title of each figure.

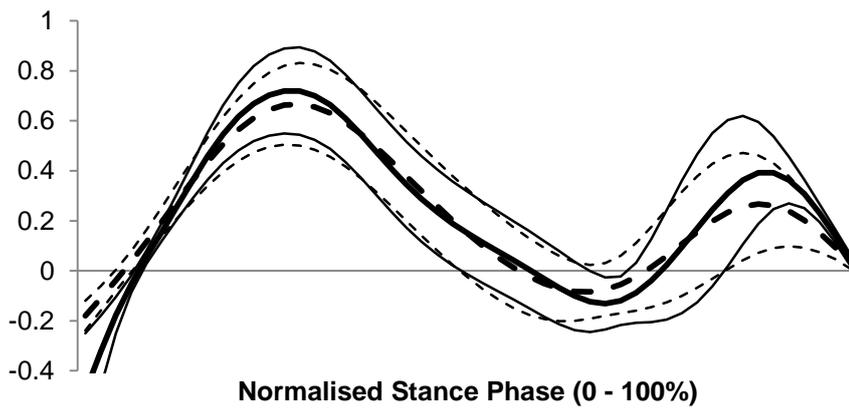
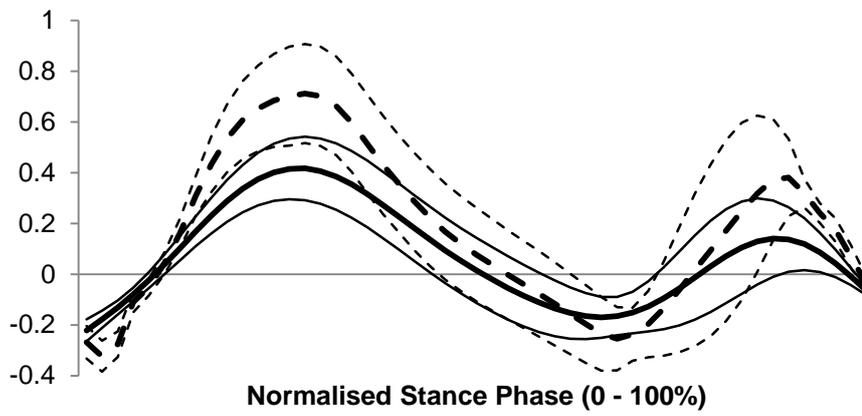
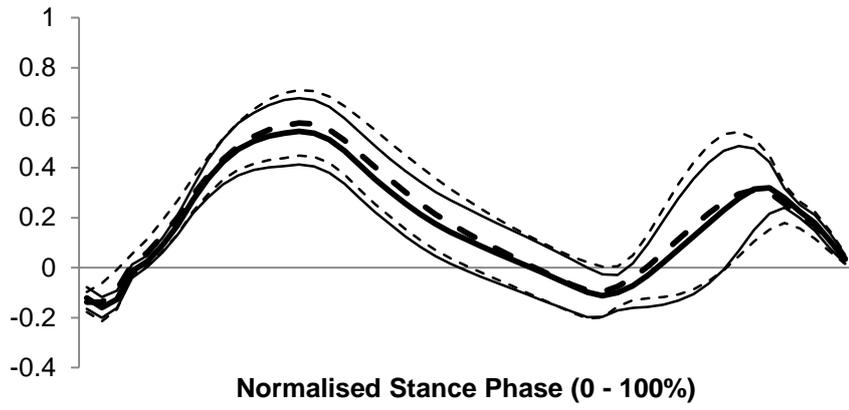
Intact hip moment ($\text{Nm}/(\text{kg}\cdot\text{ms}^{-1})$).



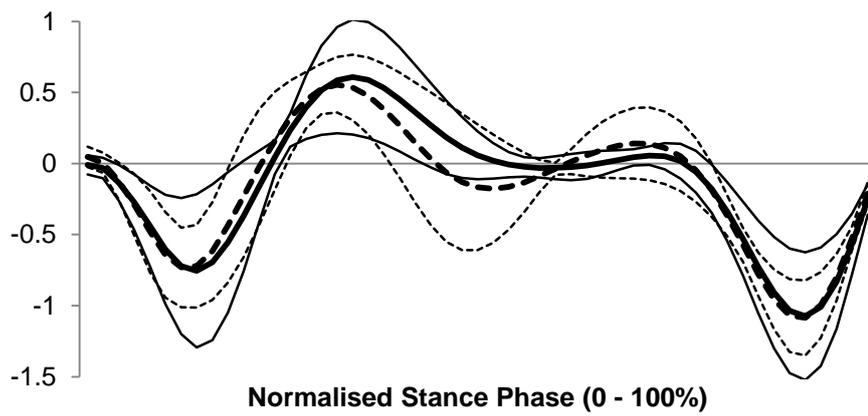
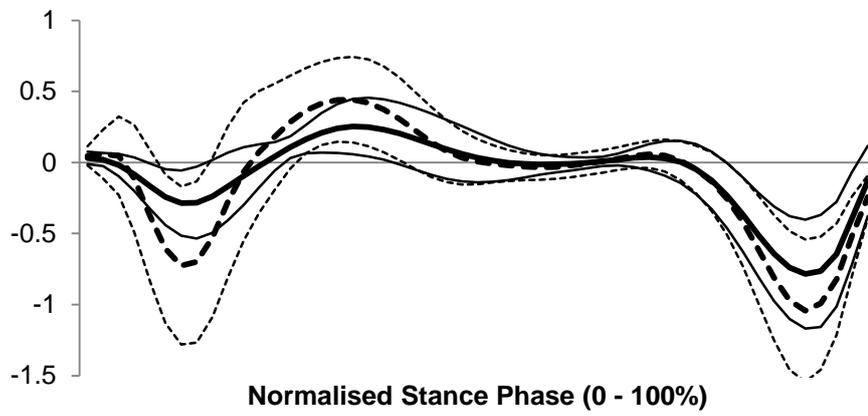
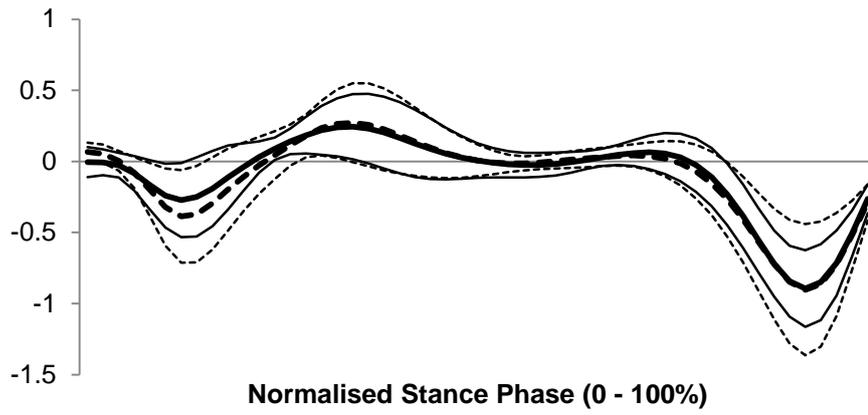
Intact hip power ($W/(kg.ms^{-1})$).



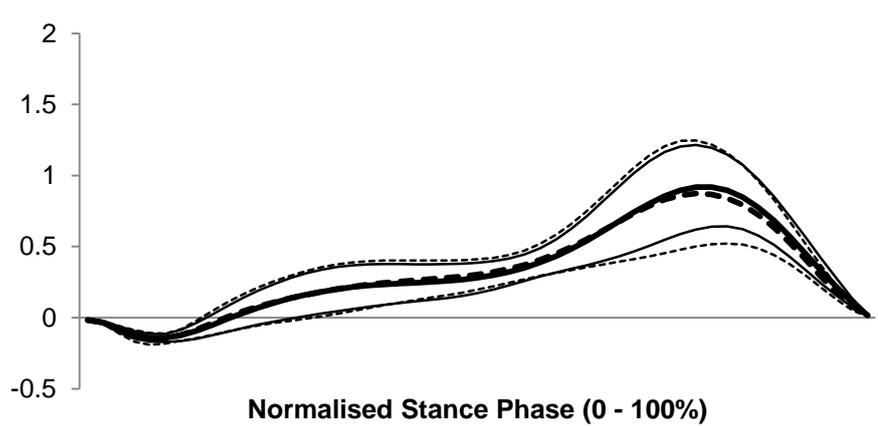
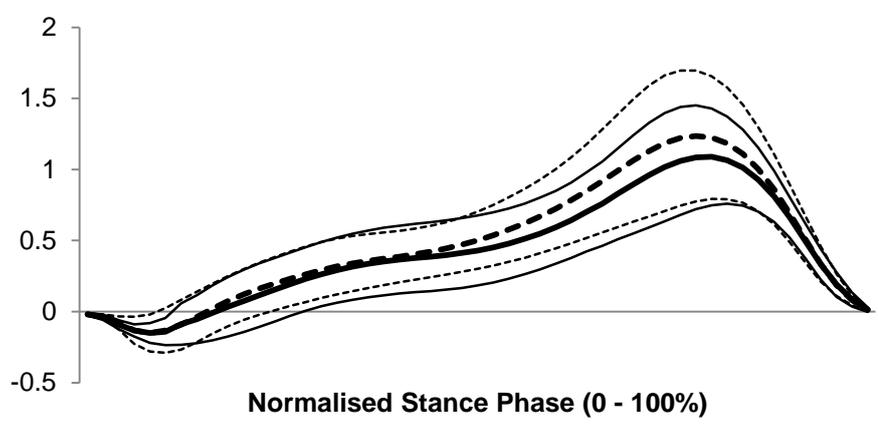
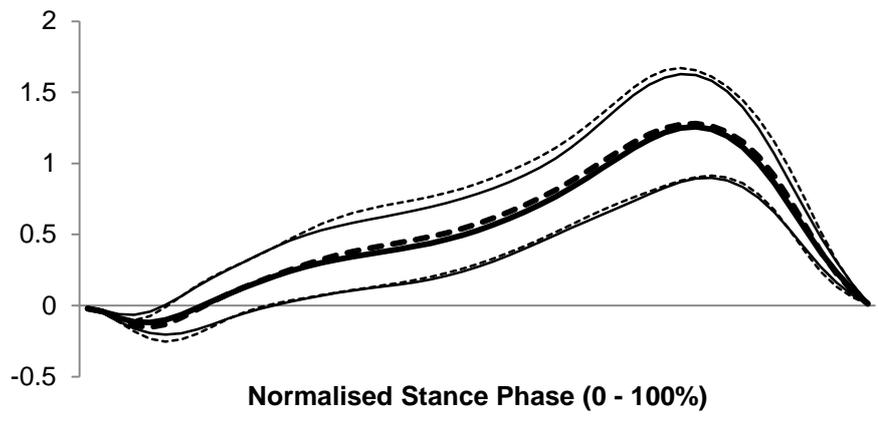
Intact knee moment (Nm/(kg.ms⁻¹)).



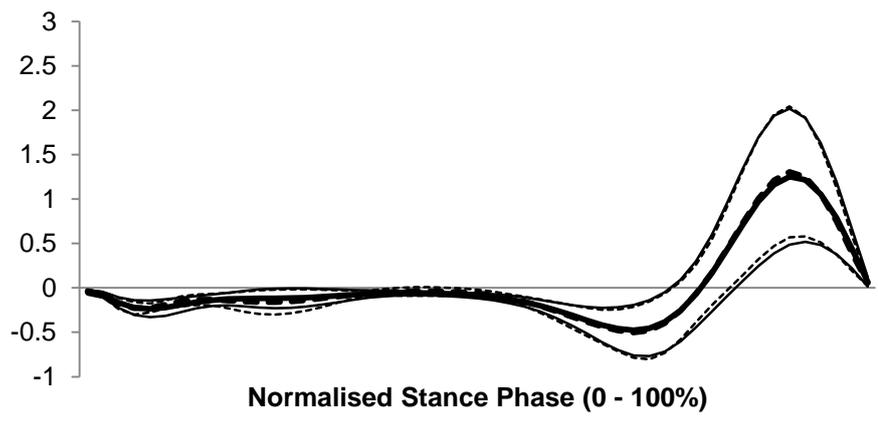
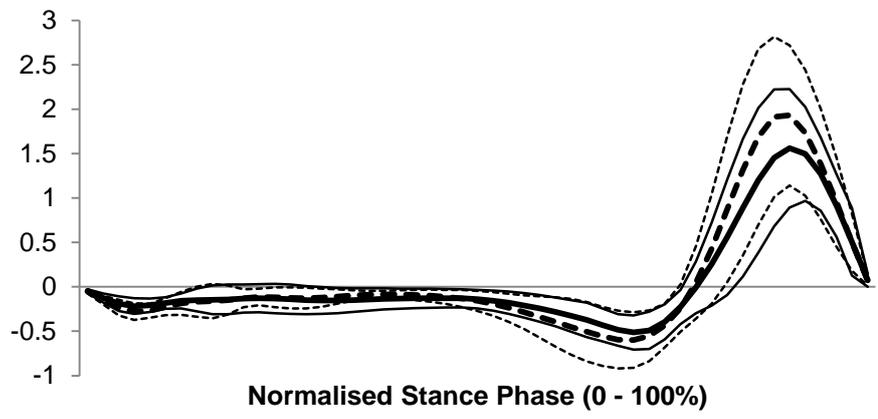
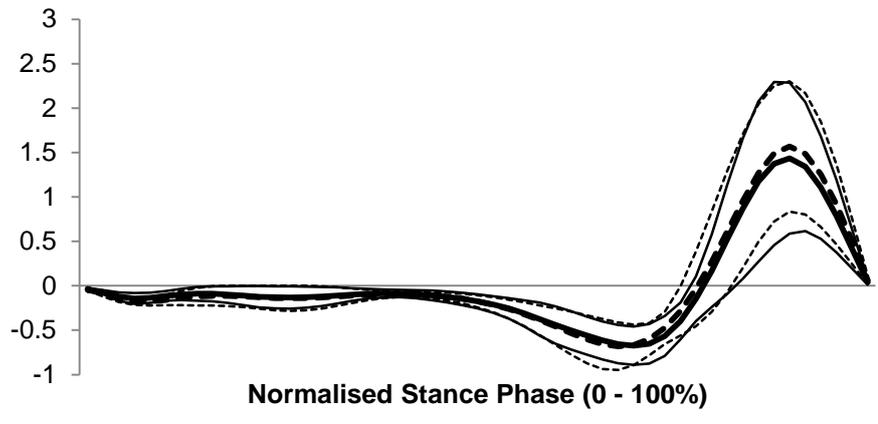
Intact knee power ($W/(kg \cdot ms^{-1})$).



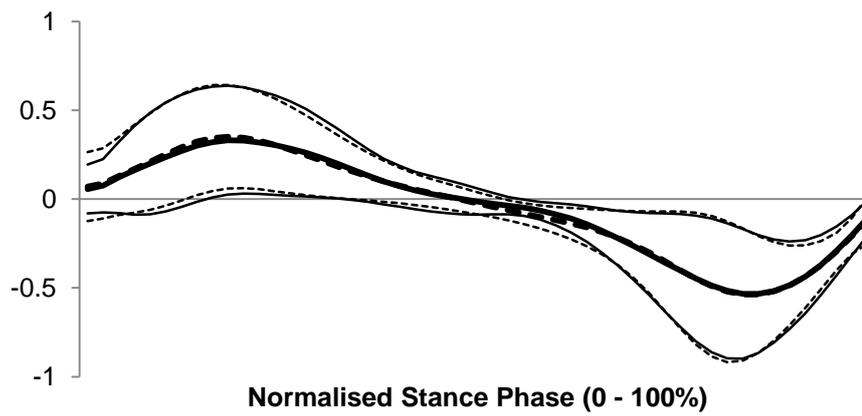
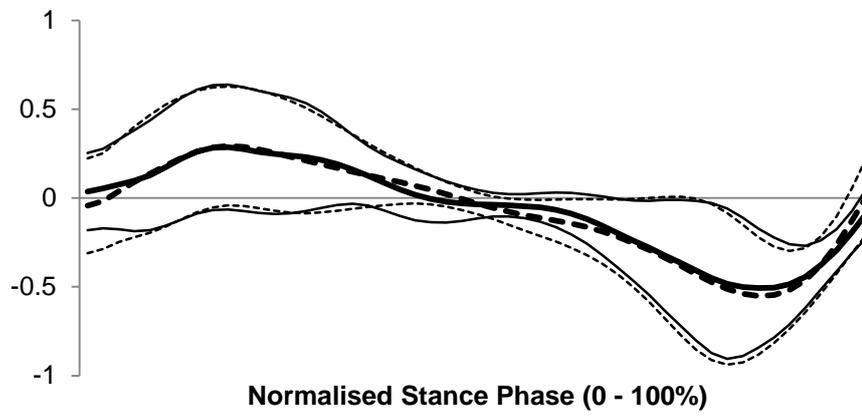
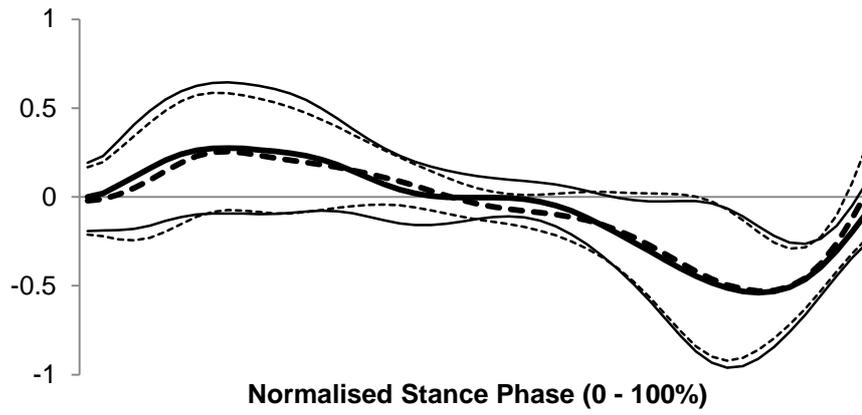
Intact ankle moment ($\text{Nm}/(\text{kg}\cdot\text{ms}^{-1})$).



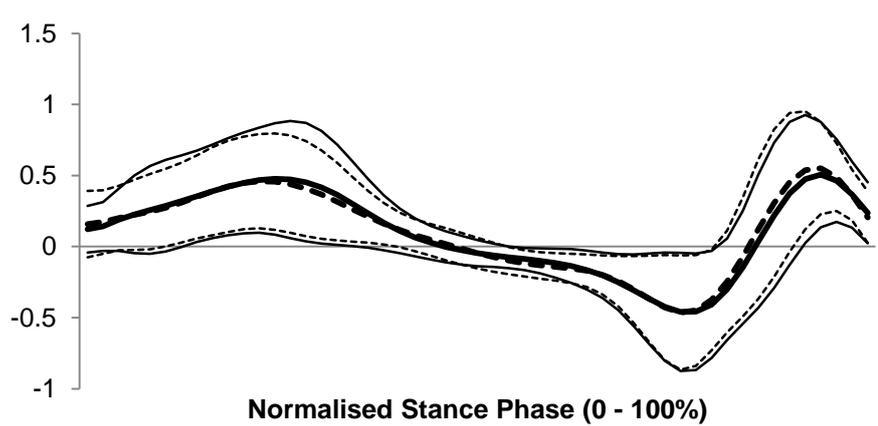
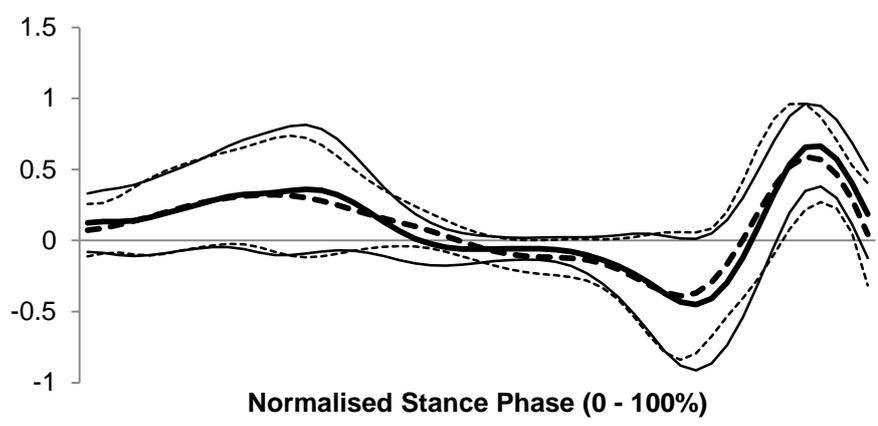
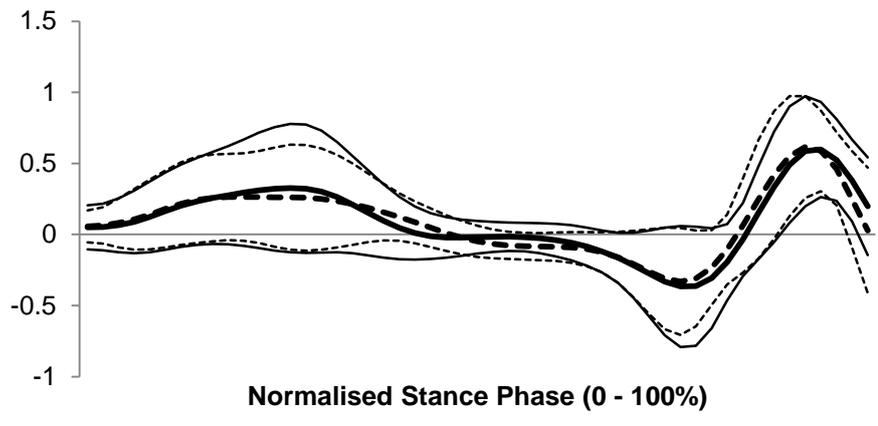
Intact ankle power ($W/(kg.ms^{-1})$).



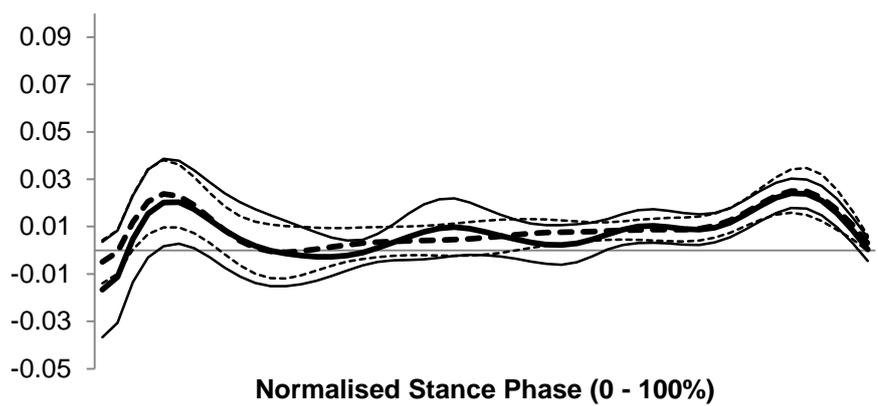
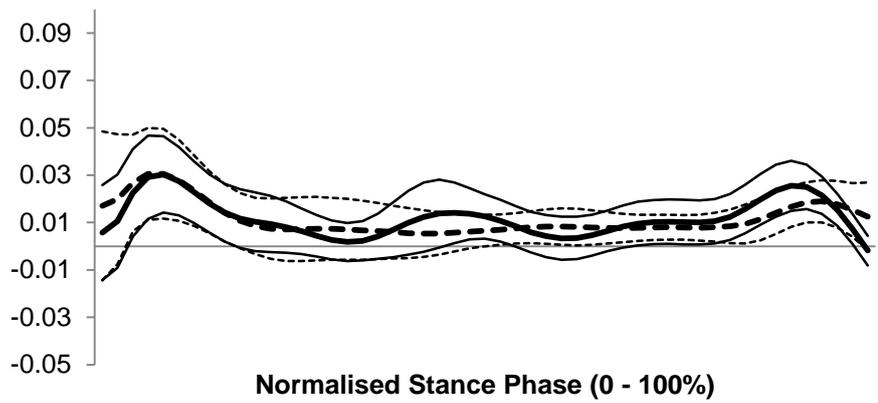
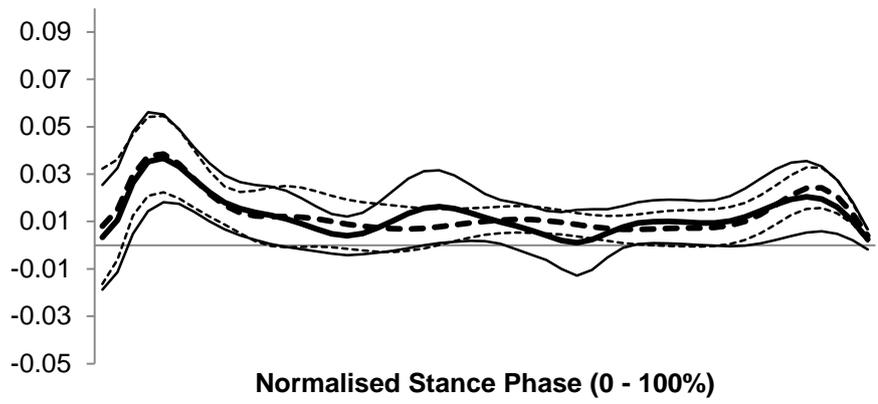
Residual hip moment ($\text{Nm}/(\text{kg}\cdot\text{ms}^{-1})$).



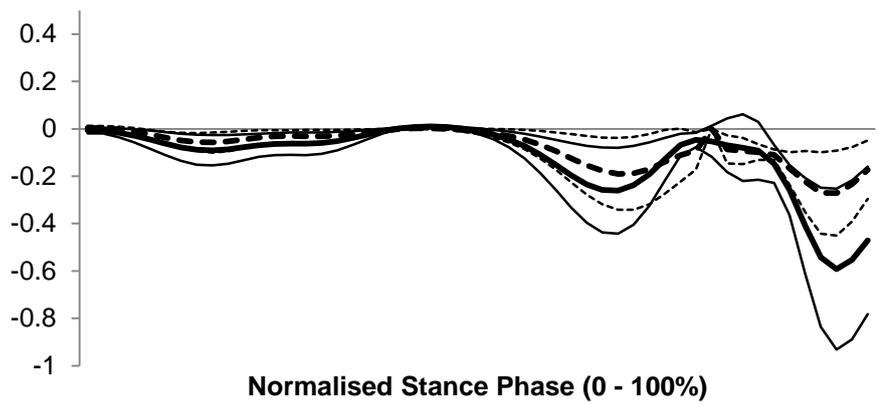
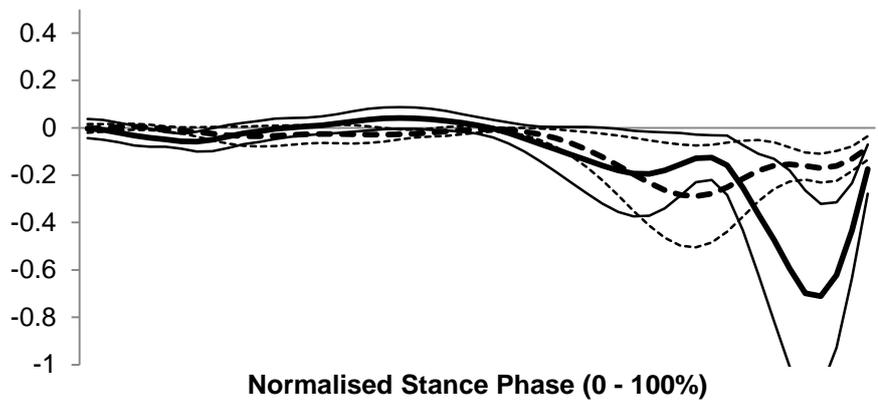
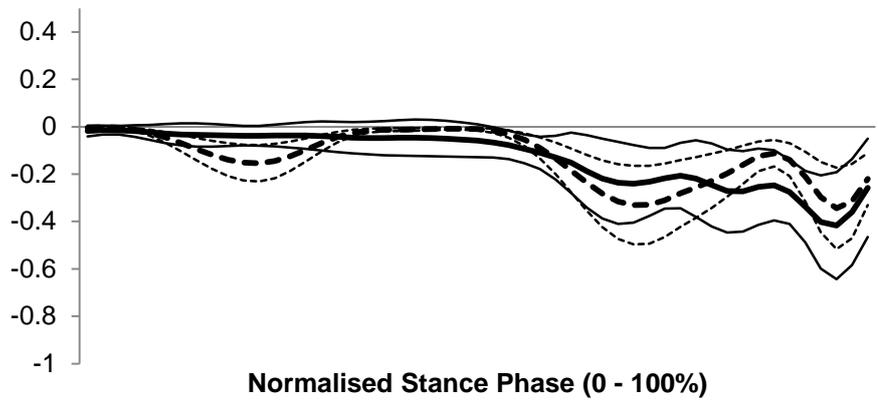
Residual hip power ($W/(kg \cdot ms^{-1})$).



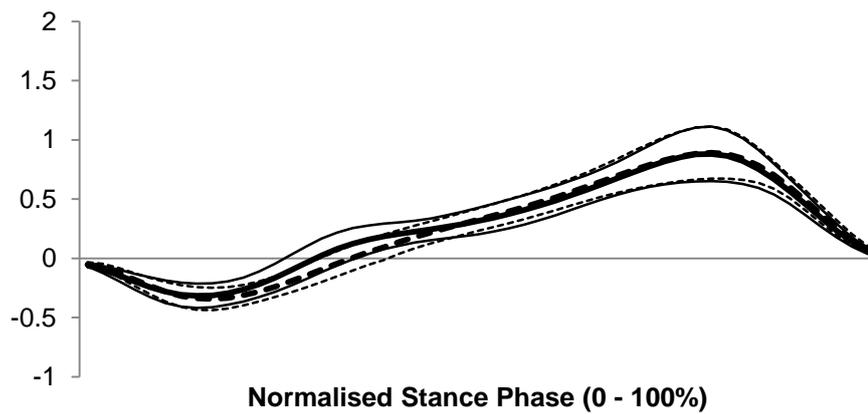
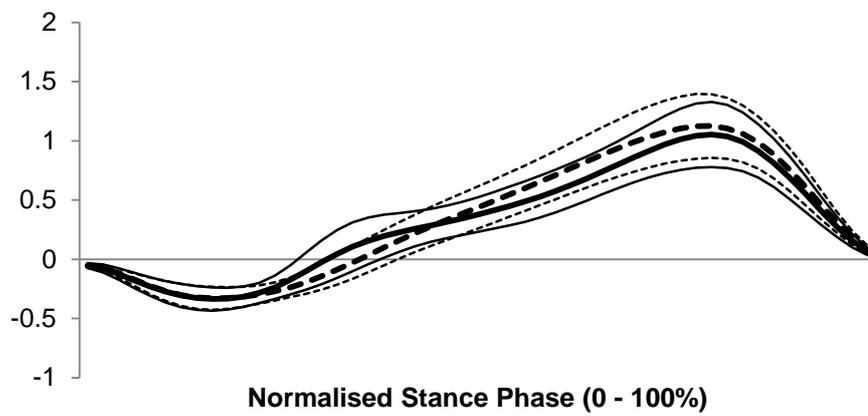
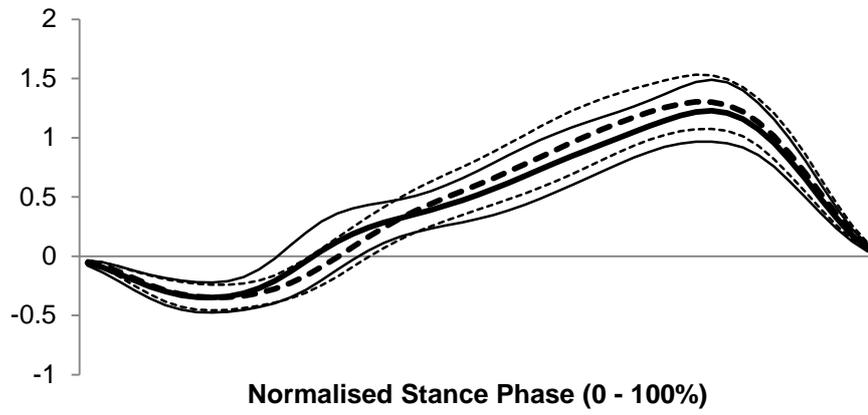
Residual knee moment ($\text{Nm}/(\text{kg}\cdot\text{ms}^{-1})$).



Residual knee power ($W/(kg \cdot ms^{-1})$).



Pros end moment (Nm/(kg.ms⁻¹)).



Proximal 'power' ($W/(kg \cdot ms^{-1})$).

