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Attenuation of centre-of-pressure trajectory fluctuations under the prosthetic foot when using an articulating hydraulic ankle attachment compared to fixed attachment

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ABSTRACT

Background: Disruptions to the progress of the centre-of-pressure trajectory beneath prosthetic feet have been reported previously. These disruptions reflect how body weight is transferred over the prosthetic limb and are governed by the compliance of the prosthetic foot device and its ability to simulate ankle function. This study investigated whether using an articulating hydraulic ankle attachment attenuates centre-of-pressure trajectory fluctuations under the prosthetic foot compared to a fixed attachment.
Methods: Twenty active unilateral trans-tibial amputees completed walking trials at their freely-selected, comfortable walking speed using both their habitual foot with either a rigid or elastic articulating attachment and a foot with a hydraulic ankle attachment. Centre-of-pressure displacement and velocity fluctuations beneath the prosthetic foot, prosthetic shank angular velocity during stance, and walking speed were compared between foot conditions.
Findings: Use of the hydraulic device eliminated or reduced the magnitude of posteriorly directed centre-of-pressure displacements, reduced centre-of-pressure velocity variability across single-support, increased mean forward angular velocity of the shank during early stance, and increased freely chosen comfortable walking speed ($P \leq 0.002$).
Interpretation: The attenuation of centre-of-pressure trajectory fluctuations when using the hydraulic device indicated bodyweight was transferred onto the prosthetic limb in a smoother, less faltering manner which allowed the centre of mass to translate more quickly over the foot.

1. Introduction

During normal able-bodied gait the centre-of-pressure (CoP) 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 (Kirtley, 2006; Schmid et al., 2005). In lower-limb amputees the CoP 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', 'stuttering' or experiencing a 'dead spot' during stance on the prosthetic limb are common features of unilateral amputee gait. Such perceptions are likely to be reflected by interruptions in the forward progression of the CoP which in turn reflect how bodyweight is transferred over the prosthetic limb (Winter, 2009). In amputee gait CoP forward progression will be governed by the compliance of the prosthetic foot device (Hafner et al., 2002) and in particular its ability to simulate ankle function to provide 1st and 2nd rocker phases of gait.

The functional performance of one particular prosthetic foot versus another is often evaluated using inverse dynamic 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 current so-called energy-storing and return (ESR) prosthetic feet have no articulating components, and instead deformation of the foot's flexible keels provides 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 and Childress, 2005). To avoid such interpretation problems Hansen et al. (2000) proposed using the trajectory of the CoP, transformed from a

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laboratory-based global coordinate system to the local coordinate system of the shank, to determine the effective 'rocker' or roll-over shape when using a particular prosthetic foot device. In essence the radius and shape of this 'rocker' describe the global functioning of the prosthetic foot-ankle device and remove the necessity of modelling it as a segment and joint. Although this approach has been adopted by others (e.g. Curtze et al., 2009; Major et al., 2011) a limitation of using roll-over shape characterisation is that it determines the radius of a 'best fit' curve onto a limited number of CoP displacement samples and thus overlooks short-duration disruptions in CoP progression. The magnitude of any such disruptions has been hitherto unmeasured, thus an important characteristic of the prosthetic device is disregarded.

Most current prosthetic feet either have a rigid attachment or incorporate an 'ankle' device allowing elastic articulation. The purpose of the present study was to examine whether use of a foot incorporating a device which allowed hydraulically controlled stance-phase articulation would attenuate the disruptions in CoP progression commonly reported in amputee gait. This foot (Echelon™, Chas. A. Blatchford and Sons Ltd., Bassingstoke, UK, *hy*A-F) has recently become clinically available and patients who use it report improved comfort and function. When set-up correctly, a *hy*A-F provides 6° plantarflexion and 3° dorsiflexion relative to its neutral (standing) position. We hypothesised that use of a *hy*A-F would facilitate bodyweight transfer onto the prosthetic limb in a smoother less faltering manner, and as a consequence, CoP forward progression would be less disrupted compared to when using participants' habitual feet (*hab*F) 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 *hy*A-F the shank would rotate forward above the prosthetic foot more 'smoothly' (i.e. with fewer velocity fluctuations) and with greater mean velocity, particularly so during early stance (double-support period) when the *hy*A-F would have greatest influence.

2. Methods

2.1. Participants

Twenty physically active, unilateral trans-tibial amputees (mean (SD) age 47.4 (12.5) years, mass 87.3 (13.5) kg, height 1.79 (0.06) m) took part, each giving written informed consent prior to their involvement. All had undergone amputation at least two years prior to participation (mean 11.85 (SD 11.83) years, range 2–45 years) and all had used their current foot for at least six months. All participants habitually used a prosthetic foot with a fixed or elastically controlled articulating attachment (*hab*F). Twelve participants habitually used an Esprit foot (Esprit™, Chas. A. Blatchford and Sons Ltd., Basingstoke, UK). This foot is identical in design to the *hy*A-F, except that it uses a fixed attachment (Fig. 1). Of the other eight participants, five used a Multiflex, one a Flex-freedom, one an Elite and one a Seattle Litefoot. The study was conducted in accordance with the tenets of the Declaration of Helsinki and local bioethics committee approval was obtained.

2.2. Protocol and prosthetic intervention

Participants completed two blocks of 10 walking trials; one block was undertaken using their *hab*F and the other using a *hy*A-F. Block order was counter-balanced across participants and both blocks were conducted on the same day. Prior to completing the block using the *hy*A-F each participant's habitual prosthesis was altered by exchanging the existing foot for a *hy*A-F. All alterations were made by an experienced prosthetist, who was careful to ensure that the two types of feet used had as close to the same alignment as possible. That is the socket, suspension and alignment of the shank pylon were unchanged across foot types and each type of foot was attached to the distal end of the shank pylon with as close to the same alignment and set-up as possible. Thus before exchanging one foot for another, foot orientation and alignment of the attachment at the shank were noted and wherever possible maintained between foot conditions. When swapping from an Esprit (*hab*F) to an Echelon (*hy*A-F), or vice versa, the foot would naturally fall into the existing location and 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 *hab*F for a *hy*A-F, each foot's ideal alignment was used as the guiding criteria. Functioning (i.e. roll over characteristics) of each foot is optimal at its own ideal alignment, and using such alignment is therefore the fairest way to make comparisons between feet. Ideal alignment instructions were readily available from the respective manufacturers, and the experienced prosthetist making the adjustments was familiar with these instructions.

Once the *hy*A-F was fitted, participants walked both indoors and outdoors for a minimum of 45 min 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. During this period the settings which control the rates of articulation within the *hy*A-F (damping) were adjusted by the prosthetist until deemed to provide optimal function at self-selected, comfortable walking speed. The device has separate settings for plantar- and dorsi-flexion ranging from 1 [minimum] to 9 [maximum], equating to damping coefficients of 1.28 to 3.48 Nms/deg respectively. Participants completing trials using their *hab*F in the first block (block 1) completed these on arrival at the laboratory. For those completing trials using their *hab*F in the second block (block 2), the foot was refitted to their prosthesis following completion of block 1 (undertaken using the *hy*A-F), and the original length, set-up, and alignment of the prosthesis were restored. Participants were again given a familiarisation period, similar to that described above, in order to reacquaint themselves with their habitual prosthesis prior to data collection.

2.3. Data acquisition and processing

Participants walked in a straight line along a flat and level 8 m walkway at their freely-selected comfortable walking speed. Kinematic and kinetic data were recorded at 100 Hz and 400 Hz respectively using an eight camera motion capture system (Vicon MX, Oxford, UK) and two floor-mounted force platforms (AMTI, MA, USA) mounted within the floor of the walkway. A successful trial occurred when a 'clean' contact by the prosthetic foot was made with either of the two force platforms without any observable targeting or changes in stride pattern. During data collection, participants wore their own flat-soled shoes and lycra shorts. Spherical, retro-reflective markers (all 14 mm diameter except markers placed onto the feet which were 9 mm diameter) were placed bilaterally on the following body landmarks (or equivalent locations on the prosthesis): acromion process, iliac crest directly above the greater trochanter, greater trochanter, medial and lateral femoral condyles, medial and lateral malleoli, posterior calcaneous, superior aspects of first and fifth metatarsal heads, distal end of second toe and pragmatically on the medial and lateral aspects of the mid-foot. Markers were also placed on the sternal notch, xiphoid process, and vertebrae C7 and T8. A head band was used to mount 4 head markers, and plate-mounted 4-marker clusters were worn on the thighs and shanks, whilst a skin-mounted 4-marker cluster was attached about the sacrum. Following 'subject'-calibration the markers on the acromions, knees and ankles were removed.

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 (Version 4, C-Motion, Germantown, MD, USA), where a nine segment 6DoF model of each participant (Vanrenterghem et al., 2010) was constructed. Functional joint centres were created (as per Schwartz and Rosumalski, 2005) for both hips and knees and for the intact ankle. For the prosthetic limb a virtual 'ankle' centre was defined on the mid-line of the prosthetic shank at the same height as the contralateral intact ankle. This ensured