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Biomechanics of ramp descent in unilateral trans-tibial amputees: Comparison of a microprocessor controlled foot with conventional ankle–foot mechanisms

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ABSTRACT

Background: Walking down slopes and/or over uneven terrain is problematic for unilateral trans-tibial amputees. Accordingly, 'ankle' devices have been added to some dynamic-response feet. This study determined whether use of a microprocessor controlled passive-articulating hydraulic ankle–foot device improved the gait biomechanics of ramp descent in comparison to conventional ankle–foot mechanisms.
Methods: Nine active unilateral trans-tibial amputees repeatedly walked down a 5° ramp, using a hydraulic ankle–foot with microprocessor active or inactive or using a comparable foot with rubber ball-joint (elastic) 'ankle' device. When inactive the hydraulic unit's resistances were those deemed to be optimum for level-ground walking, and when active, the plantar- and dorsi-flexion resistances switched to a ramp-descent mode. Residual limb kinematics, joints moments/powers and prosthetic foot power absorption/return were compared across ankle types using ANOVA.
Findings: Foot-flat was attained fastest with the elastic foot and second fastest with the active hydraulic foot ($P < 0.001$). Prosthetic shank single-support mean rotation velocity ($p = 0.006$), and the flexion ($P < 0.001$) and negative work done at the residual knee ($P = 0.08$) were reduced, and negative work done by the ankle–foot increased ($P < 0.001$) when using the active hydraulic compared to the other two ankle types.
Interpretation: The greater negative 'ankle' work done when using the active hydraulic compared to other two ankle types, explains why there was a corresponding reduction in flexion and negative work at the residual knee. These findings suggest that use of a microprocessor controlled hydraulic foot will reduce the biomechanical compensations used to walk down slopes.

1. Introduction

The majority (~85%) of clinically available prosthetic feet are connected to the shank-pylon via a rigid, non-articulating means of attachment. During overground gait such feet facilitate attainment of foot-flat following initial contact via simulated plantar-flexion, which is achieved by deformation of the foot's heel (comprising of cushioning material or of a leaf-spring keel). When walking down a ramp/slope the heel is unable to deform sufficiently to achieve foot-flat, and as a consequence knee flexion is increased (Fradet et al., 2010; Vickers et al., 2008). Although increased knee flexion helps the prosthetic foot attains foot-flat it also potentially reduces knee stability because of the increased load it places on the residuum (Perry et al., 1997; Vickers et al., 2008). This likely explains why individuals with a unilateral trans-tibial amputations (UTAs) find ramp descent problematic (Vickers et al., 2008; Vrieling et al., 2008). The amount of residual-knee flexion needed to achieve foot-flat, can be reduced by using a foot with a more compliant heel. However, while an increase in heel compliance may be beneficial for ramp descent, an increase in compliance has been demonstrated to increase the 'braking effect' exerted by the prosthetic limb during weight acceptance in level-ground gait (Fey et al., 2013; Silverman and Neptune, 2012). This braking effect occurs because deformation of the heel section creates resistance to forwards shank rotation, which in turn slows the forward velocity of the whole-body centre of mass during early- to mid-stance on the prosthetic side (De Asha et al., 2014). Rather than having a more compliant heel, attainment of foot-flat when descending ramps can be achieved using a prosthetic foot that allows articulation at its point of attachment (i.e. has an ankle mechanism). A simple ankle device is achieved by incorporating a rubber-snubber at the point of attachment. Subjective assessment has highlighted that walking down slopes is perceived to be easier when using such a foot (Su et al., 2010). Recently, prosthetic ankle–foot devices allowing hydraulically damped (passive) articulation between the foot and shank-pylon have become clinically available. One such device (*Echelon*; Chas. A Blatchford and Sons, Ltd., Basingstoke, UK), has

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been demonstrated to increase walking speed and toe clearance and reduce compensatory intact side kinetics and the braking effect exerted by the prosthetic limb in level-ground gait in active UTAs when compared to using a device with either a non-articulating or rubber-snubber means of attachment (De Asha et al., 2013a,b, 2014; Johnson et al., 2014; Portnoy et al., 2012). More recently, a passive hydraulic ankle-foot device incorporating microprocessor control (MPC) has been introduced (*Elan*; Chas. A Blatchford and Sons, Ltd., Basingstoke, UK). This device has sensors that determine the angle of the terrain being walked on and the cadence of walking. This information is used to automatically alter the hydraulic damping to pre-defined settings so as to facilitate a more optimal foot-ground interaction for the current terrain or cadence: one such setting being a 'ramp descent' mode. According to the manufacturer, when in this mode the resistance to plantar-flexion following initial contact is reduced to facilitate attainment of foot-flat, and dorsi-flexion resistance is increased in order to control the rate of subsequent forwards shank rotation as body weight is transferred onto and over the prosthetic foot. Attaining foot flat without the need for compensatory knee flexion should improve residual knee stability during loading-response, while controlling the rate at which the shank rotates forwards over the limb should prevent UTAs 'falling' forwards during single-support; thus improving dynamic stability (Perry et al., 1997). Use of such a device should therefore improve ramp descent gait biomechanics over using feet with elastically controlled 'ankle' articulation. In previous studies that have investigated whether ramp descent is improved using an adaptive prosthetic foot, UTAs used a *Proprio-Foot* (Ossur, Reykjavik, Iceland) (Agrawal et al., 2015; Darter and Wilken, 2014; Fradet et al., 2010). This device incorporates a MPC actuator at the ankle that pushes the foot into plantar-flexion during swing but the device allows no passive 'ankle' articulation during stance. Findings indicate that having the MPC actuator active had minimal impact on gait biomechanics but it did result in UTAs reporting feeling 'safer' (Fradet et al., 2010). Also, use of a *Proprio-Foot*, in comparison to using a participant's habitual foot, led to increased symmetry in the work done on the whole-body centre of mass by the vertical ground reaction force (GRF) from each limb (Agrawal et al., 2015), and reduced the energy costs of ambulation (Darter and Wilken, 2014). The later finding occurred irrespective of whether the MPC actuator was active or inactive, which suggests it must have been due to factors unrelated to the 'adaptive ankle' function.

The present study determined if use of an *Elan* foot improves the gait biomechanics of ramp descent in UTAs. Active UTAs repeatedly descended a ramp using an *Elan* with the microprocessor active (MPC) or inactive (non-MPC). The foot incorporates a hydraulic ankle device allowing passive articulation during stance of up to 6° of plantarflexion and up to 3° of dorsiflexion. With the microprocessor active the hydraulic resistances automatically switch to settings deemed to be optimum for ramp descent (see methods). With the microprocessor inactive, the hydraulic resistances to plantar- and dorsi-flexion default to the settings determined as being optimum for each individual user during overground customary speed walking. To determine how use of an MPC hydraulic foot compares to using a foot with elastically articulating ankle device, participants also completed trials using an *Epirus* foot (Chas. A Blatchford and Sons, Ltd., Basingstoke, UK). The *Epirus* has an ankle device incorporating a spherical rubber-snubber, which provides multi-axial articulation. The device has the ability to plantarflex up to 15° while dorsiflexion is restricted by a 'hard stop' within the mechanism. Both the *Elan* and *Epirus* use a uniform dynamic-response foot; meaning the only difference across prosthetic conditions was the 'ankle' articulation resistance. It was hypothesised that the attainment of foot-flat would occur sooner using the MPC and elastic device compared to non-MPC device, then following foot-flat the shank's forwards rotation during prosthetic limb single-support would be slower when using the MPC device compared to non-MPC device and slower still in comparison to the elastic device. In addition, because of the MPC device exerting increased control over forwards shank rotation during early- to mid-stance, it was hypothesised that use of this device would also lead to a reduction in residual-limb knee flexion during early stance, and as a consequence reduce the magnitude of the stance-phase joint moment and amount of mechanical power absorbed at the residual-knee during single-support, in comparison to that observed using the non-MPC device, and reduced further still to that observed using the elastic device.

2. Methods

2.1. Participants

Nine male adults with a unilateral trans-tibial amputation (mean (SD), age 41.2 (12.9) years, height 1.76 (0.06) m, mass 74.14 (15.7) kg, time since amputation 7.5 (6.4), range 2.5–22.9 years) participated in the study. All were classed as at least K3 on the Medicare scale. Four habitually used an *Elan*, four an *Echelon VT* (Chas. A Blatchford and Sons, Ltd., Basingstoke, UK; this foot incorporates an 'ankle' device that is equivalent to an *Elan* with MPC inactive) and one a *Re-flex Rotate* (Ossur, Reykjavik, Iceland): thus all were familiarised to using an articulating ankle–foot device. The study was conducted in accordance with the tenets of the Declaration of Helsinki and approval was gained from the Institutional Committee for Ethics in Research.

2.2. The ramp

The ramp (5 degree incline) was custom made, with a 2.8 m long and 1 m wide walking surface: a raised surface at its upper end provided a 'lead on' to the decline (Fig. 1). The middle section of the ramp was positioned over a force-platform. This middle section incorporated an inclined solid block (chip-board) that was bolted onto the platform surface. The top surface of the block was flush with the surface of the ramp, and there was a 2 mm gap between the block (all sides) and the rest of the ramp (which was secured to the floor). The ramp and block surfaces were painted with grey anti-slip paint.

2.3. Prosthetic conditions

Ramp descent trials were completed in two blocks; with block order counterbalanced across participants. In one block participants used an *Epirus* foot and in the other they used an *Elan* foot. For the block using the *Elan* foot, trials were undertaken, in random order, with the micro-processor being active (MPC) or inactive (non-MPC). Participants were 'blinded' to which mode the device was in. When active, the device normally requires at least two steps before it switches to its ramp descent mode. However, in the present study, in order to ensure the *Elan* foot was in the ramp descent mode, it was remotely triggered into this mode at the start of a trial via Bluetooth connection with the foot's microprocessor.

All prosthetic alterations were made by the same experienced prosthetist. When swapping between *Elan* and *Epirus* foot types, everything about the prosthesis was kept as near to constant as possible. The overall length, socket, suspension, and shank alignment were unchanged across foot types (as were the size and stiffness of the heel and forefoot keels). The *Elan* foot has nine damping settings for each of the two hydraulic cylinders that control separately the rates of plantar- and dorsi-flexion, and moving between these settings is achieved by altering (via microprocessor control) the position of a valve in each hydraulic cylinder. The settings for plantar- and dorsi-flexion are set to suit each participant's overground gait. When in the ramp descent mode the plantar-flexion resistance goes to its second lowest setting and the dorsi-flexion resistance goes to its second highest. Prior to data collection participants were familiarised to each foot type by walking on the level floor of the laboratory for approximately 20 min. During 'familiarisation' with the *Elan*, the hydraulic damping settings were adjusted, by systematically altering the levels of damping of both plantar- and dorsi-flexion; with the final settings decided upon using a mixture of participant feed-back regarding perceived comfort and function, and

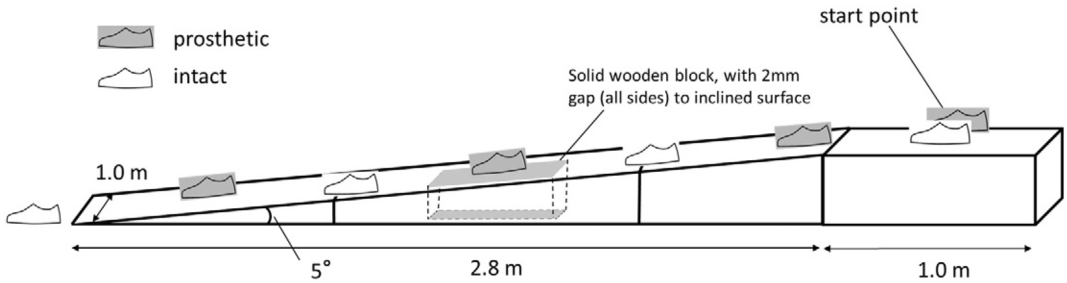

Fig. 1. Schematic of the ramp (modular design). The inclined walkway was 2.8 m long at a 5 degrees inclination; the middle of the walkway had a solid inclined block fixed on top of a force-platform. A raised surface (1 m square) at its upper end provided a 'lead on' to the decline.

the prosthetist's experience. Those participants whose habitual foot was an *Elan* used their device's customary settings.

2.4. Data acquisition and processing

Participants completed trials at two speeds: a speed they perceived to be customary and at a speed they perceived to be comfortable slow. We included a slow speed because UTAs anecdotally report that walking down slopes is more difficult at a slow speed; possibly because of not being able to prevent 'falling' forwards over the prosthetic limb during single-support. The customary speed trials were always completed first. Trials at each speed, for each foot condition (elastic, MPC, non-MPC) were completed six times (random order). Participants wore their own flat-soled shoes and 'lycra' shorts. Kinematic and kinetic data were recorded at 200 Hz using a ten-camera motion capture system (Vicon MX, Oxford, UK) and a floor-mounted force-platform (AMTI, Watertown, MA, USA). Retro-reflective markers were placed as per previous work (De Asha et al., 2013a). Labelling and gap filling of marker trajectories were completed within Nexus software (Vicon, Oxford, UK). Gap filling was completed either using 'pattern fill' of the trajectory from another marker attached to the same segment as the missing marker, or using the spline fill option if 'pattern fill' was not possible. The C3D files were then exported to Visual 3D motion analysis software (C-Motion, Germantown, MD, USA), where all further processing was undertaken. Data were filtered using a fourth order, zero-lag Butterworth filter with a 6 Hz cut-off. A 'force structure' was created above the force-platform with the exact dimensions of the inclined solid block, allowing the centre of pressure (CoP) coordinates to be transformed within Visual 3D from the platform to the top surface of the 'force structure'. A CalTester (C-Motion, Germantown, MD, USA) was used to confirm no error in the transformed determination of CoP coordinates and force vector orientation within the global reference system. A nine segment 6DoF model (Cappozzo et al., 1995), incorporating head, thorax/abdomen, pelvis, thighs, shanks and feet, of each participant was constructed. All anatomical joint centres (both hips, both knees, and the intact ankle) were determined using a functional joint centre approach (Schwartz and Rozumalski, 2005). The distal end-point of the prosthetic shank ('ankle') was created on the shank's segmental mid-line, at the same height (consistent across foot types) as the contralateral intact ankle. The deformations of the foot's flexible keels, which simulate plantar- and dorsi-flexion, break the assumptions of rigid segments and pin-joint articulation that are used in a standard inverse dynamic approach (Geil et al., 2000). Therefore we used the 'unified deformable segment' methodology (Takahashi and Stanhope, 2013; Takahashi et al., 2012), which determines the total (scalar) power absorbed, returned and dissipated by the prosthetic ankle-foot device. Joint kinetics at the residual-knee were determined by assuming the foot and shank to be a single segment with the distal forces acting on the segment being the ground reaction forces (De Asha et al., 2013b; Dumas et al., 2009). Initial contact (IC) and toe off (TO) were defined as the instants the vertical component of the GRF first went above or below 20 N, respectively. Double support (transfer onto the prosthetic limb) was from prosthetic limb IC to intact limb TO. Single-support was from intact limb TO to IC.

2.5. Data analysis

The following parameters were determined: residual-knee loading response flexion, and single-support minimum flexion, time to foot-flat, prosthetic-limb shank mean angular velocity during single-support, single-support residual-knee moment impulse, and the single-support negative mechanical work at the residual hip and knee joints and unified deformable segment (UDS). Residual-knee loading response flexion and single-support minimum flexion were defined as the local maximum and minimum flexion during early stance and single-support respectively. Time to foot-flat was defined as the time between IC and the instant the downward velocity of the prosthetic foot's toe marker became less than 0.1 ms^{-1}. The angular velocity of the prosthetic shank was defined as the rate of rotation of the shank segment in the sagittal plane. The following integrals were determined for the prosthetic-limb stance phase: extension moment impulse at the residual-knee; and the negative work (power integral) at the hip and knee joints and UDS. Hip and knee work were the integrals of their sagittal plane negative joint powers, and work at the UDS was the integral of its negative scalar power. All kinetic variables were normalized to participant's body-weight. All parameters were calculated for each individual trial and then averaged across trials to give a mean value at each speed and foot condition for each participant.

2.6. Statistical analysis

The mean values (which Kolmogorov–Smirnov tests indicated where normally distributed) were compared using repeated measures analysis of variance (ANOVA) with ankle type (elastic, MPC hydraulic, non-MPC hydraulic) and speed-level (slow and customary) as repeated factors. Where main effects were significant post hoc analyses were conducted using Tukey HSD tests. Statistical analyses were made using Statistica 6.0 (StatSoft, Inc., Tulsa, OK, USA). The alpha level was set at 0.05.

3. Results

To investigate if results were affected by a participant's familiarity with foot design, we repeated our statistical analyses but included participant's habitual foot (*Elan*, *Echelon VT*) as a 'between factor' in a mixed-design ANOVA (again with ankle type and speed level as repeated factors: note the participant whose habitual foot was a *Re-flex Rotate* was included in the *Echelon VT* sub-group). This analysis indicated that for all but one of the parameters investigated there were no significant main or interaction effects of habitual foot type sub-group (all $P > 0.13$). The one exception was a significant sub-group by ankle type interaction effect ($P = 0.008$) for residual hip negative work; indicating no difference between sub-groups for the MPC and non-MPC foot conditions but an increase in negative hip work for the habitual *Elan* users and a

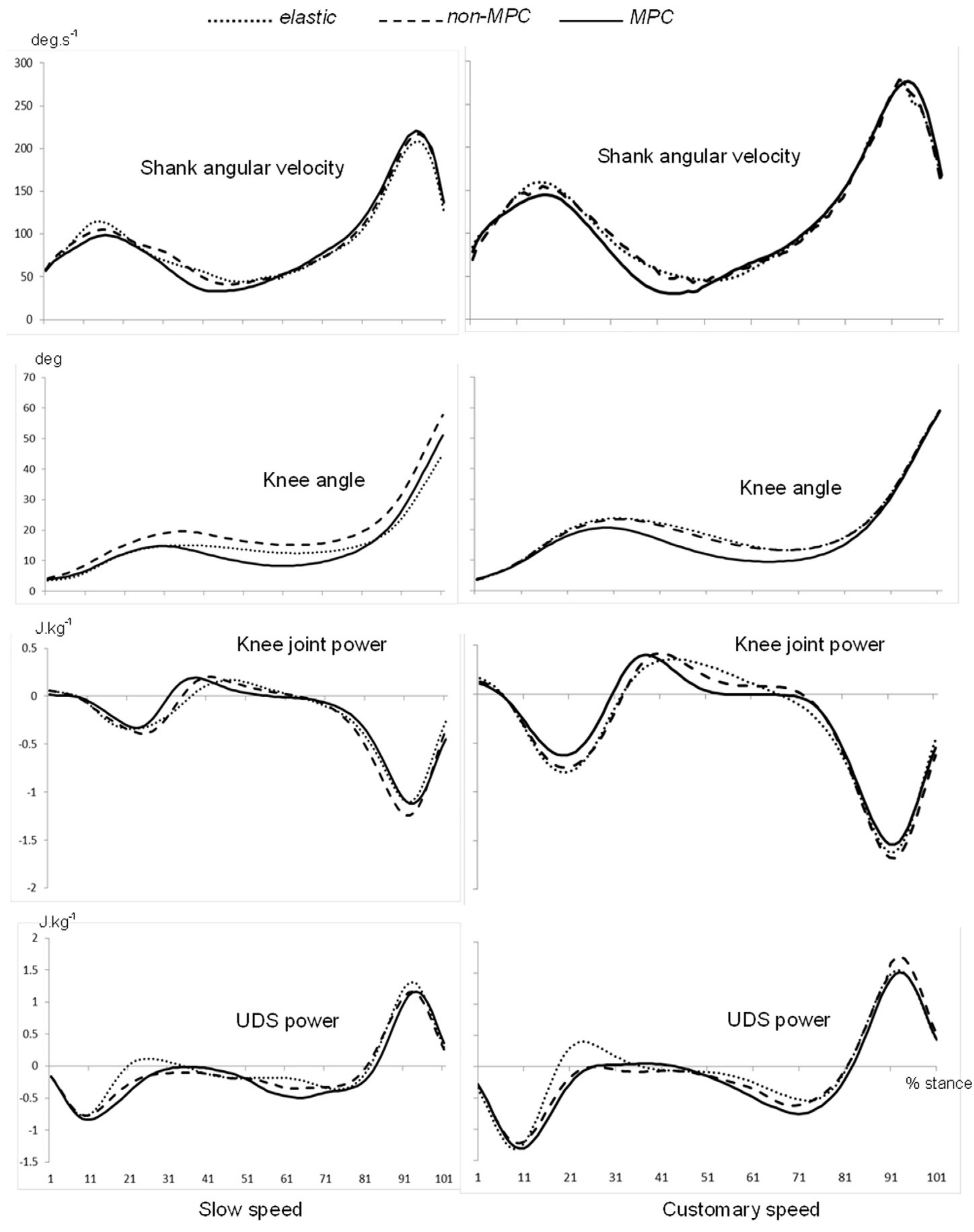

Fig. 2. Ensemble group mean stance phase shank angular velocity, residual-knee angular displacement, knee joint power, and UDS power when using the elastic (dotted line), non-MPC (dashed line), and MPC (solid line) ankle–foot.

decrease in negative hip work for the habitual *Echelon VT* users in the elastic ankle condition. This highlights that familiarity with foot design had minimal affect on the results presented. Henceforth, therefore, the results presented are for the participant cohort considered as a single group.

There were main effects of speed-level ($P < 0.001$) and ankle type ($P < 0.001$) on time to foot-flat. Time to foot-flat was longer at the slow walking speed and, irrespective of speed level, was shortest for the elastic foot and longest for the non-MPC foot with differences between each ankle type being significant. There were main effects of speed-level ($P < 0.001$) and ankle type ($P = 0.006$) on shank angular velocity (Fig. 2) during single-support (Table 1). Angular velocity was lower at the slow walking speed and, irrespective of speed level, was significantly lower for the MPC foot than either the non-MPC or elastic foot, with no significant differences between the non-MPC and elastic feet.

There were main effects of speed-level ($P < 0.001$) and ankle type ($P = 0.039$) on residual-knee loading response flexion (Table 1). Loading response flexion was increased at the faster walking speed, and irrespective of speed-level, was significantly reduced when using the MPC foot than either the non-MPC or elastic foot (Fig. 2). There were main effects of speed-level ($P = 0.024$) and ankle type ($P < 0.001$) on single-support minimum flexion (Table 1). Single-support minimum flexion was reduced at the faster walking speed and, irrespective of speed-level was significantly reduced when using the MPC foot than either the non-MPC or elastic foot (Fig. 2). There was no significant difference in residual-knee loading response flexion or single-support minimum flexion between the non-MPC and elastic feet ($P > 0.77$).

There were no significant effects of speed-level or ankle type on residual-knee extension moment impulse ($P > 0.75$, Table 2). There was a significant effect of ankle type ($P < 0.001$), but not speed-level ($P = 0.84$), on negative work at the UDS (Table 2). There was more negative work done at the UDS when using the MPC foot than either the non-MPC or elastic foot, and more when using the non-MPC compared to elastic foot (Fig. 2). There were no significant main effects of speed-level ($P = 0.45$) on negative work done by the residual-knee, but negative work tended ($P = 0.083$) to be reduced at both speeds when using the MPC foot compared to the non-MPC or elastic foot (Fig. 2). There was a main effect of speed-level ($P = 0.036$) but no effect of ankle type ($P = 0.73$) on negative work done by the residual-hip. Negative work by the residual-hip was increased at the faster walking speed (Table 2).

4. Discussion

The present study determined whether use of a microprocessor controlled passive-articulating hydraulic ankle–foot device (MPC foot) improved the gait biomechanics of ramp descent in UTAs in comparison to using feet with non-adaptive 'ankle' articulation mechanisms (non-MPC, elastic). Results indicate that foot-flat was attained most quickly when participants used the elastic foot, and second quickest when participants used the MPC foot. The subsequent forwards shank rotation velocity during single-support was slowest with the MPC foot with similar forwards shank velocities when using the other two feet. An increase in negative work done at the UDS when using the MPC foot compared to either the non-MPC or elastic foot, was likely the cause of the reduced shank velocity. These findings support our first hypothesis. In addition, results indicate that residual-knee flexion during early stance was reduced and there was a trend towards reduced negative mechanical work at the residual-knee with no difference in residual-knee moment impulse when using the MPC foot, in comparison to that using either the non-MPC or elastic foot. These findings partially support our second hypothesis.

UTAs often anecdotally report the sensation of the residual knee being 'thrown' or 'pushed' forwards when descending ramps using a non-articulating ankle–foot device. This sensation, of the residual knee being forced into increased flexion, is no doubt a result of difficulty experienced, and/or the compensations required, in achieving foot-flat when using a non-articulating ankle–foot device (Vickers et al., 2008; Vrieling et al., 2008). Previous research has shown that UTAs report perceiving ramp descent becoming 'easier' when using an elastically articulating (*MultiFlex* type), compared to non-articulating device (Su et al., 2010), or report feeling 'safer' descending a ramp using a foot that is 'pushed' into a plantar-flexed position during swing (Fradet et al., 2010). Together, these findings would suggest that attainment of prosthetic-limb foot-flat is a crucial event in ramp descent for UTAs. In the present study, time to foot-flat was shortest when using the elastic foot and longest when using the non-MPC foot, but importantly, the time was reduced when using the MPC foot compared to non-MPC foot. This highlights that the hydraulic plantar-flexion resistance following initial contact must have been reduced when using the MPC compared to non-MPC foot (as it is designed to be when in its 'ramp descent' mode). Once foot-flat is attained, the MPC foot is designed (when in 'ramp descent' mode) to be at the second highest dorsiflexion resistance setting in order to control forwards shank rotation as body weight is transferred onto and over the prosthetic foot. In contrast, when using a non-MPC foot 'ankle' dorsi-flexion control is the same as that for level ground gait. When using an elastic foot, the plantar-flexed position of the foot at foot-flat means the rubber ball-joint would exert a forwards 'pull', via elastic recoil, to bring the shank back to its neutral position relative to the foot (at which point it hits the 'hard stop' within the mechanism). Subsequent resistance to dorsi-flexion would occur solely due to the stiffness of the forefoot keel. In the present study, residual-knee flexion during early stance was reduced when using the MPC foot compared to the other two feet. This reduced flexion at the residual-knee when using the MPC foot, was likely driven by a combination of reduced plantar-flexion resistance, allowing the foot to reach foot-flat sooner, followed

Table 1
Group mean (SD) time to 'foot-flat', mean angular velocity of shank, residual-knee loading response flexion and single-support minimum flexion when using MPC, non-MPC and elastic feet at slow and customary speeds. Statistically significant differences are highlighted in **bold**.

	Slow			Customary			*P* value
	MPC hydraulic	Non-MPC hydraulic	Elastic	MPC hydraulic	Non-MPC hydraulic	Elastic	
Time to 'foot-flat' (s)	0.195 (0.016)	0.212 (0.026)	0.170 (0.019)	0.172 (0.016)	0.180 (0.020)	0.150 (0.021)	**Speed < 0.001** **Foot < 0.001** Int. 0.058
Mean shank angular velocity ($^\circ$ s^{-1})	63.8 (12.5)	70.4 (14.5)	70.8 (11.4)	85.1 (15.6)	91.2 (14.6)	92.0 (19.9)	**Speed < 0.001** **Foot 0.006** Int. 0.96
Loading response flexion (°)	18.0 (5.4)	21.3 (6.1)	21.7 (5.1)	21.8 (4.7)	24.7 (5.3)	24.4 (5.3)	**Speed < 0.001** **Foot 0.039** Int. 0.57
Single support minimum flexion (°)	10.8 (7.7)	16.7 (7.6)	17.4 (6.7)	7.1 (7.4)	12.1 (9.3)	13.4 (6.2)	**Speed 0.024** **Foot < 0.001** Int. 0.94

Table 2
Group mean (SD) residual-knee extension moment impulse, and the negative work done at the unified deformable segment (UDS) and at the residual knee and hip when using MPC, non-MPC and elastic feet at slow and customary speeds. Statistically significant differences are highlighted in **bold**.

	Slow			Customary			*P* value
	MPC hydraulic	Non-MPC hydraulic	Elastic	MPC hydraulic	Non-MPC hydraulic	Elastic	
Residual-knee ext moment impulse (Nm s kg^{-1})	0.010 (0.007)	0.010 (0.016)	0.001 (0.009)	0.011 (0.018)	0.012 (0.021)	0.009 (0.007)	Speed 0.75 Foot 0.94 Int. 0.83
UDS negative work (J kg^{-1})	−0.158 (0.085)	−0.114 (0.041)	−0.088 (0.039)	−0.143 (0.052)	−0.125 (0.055)	−0.088 (0.017)	Speed 0.84 **Foot < 0.001** Int. 0.29
Residual-knee negative work (J kg^{-1})	−0.004 (0.004)	−0.010 (0.012)	−0.019 (0.028)	−0.011 (0.012)	−0.019 (0.018)	−0.017 (0.011)	Speed 0.45 Foot 0.083 Int. 0.33
Residual-hip negative work (J kg^{-1})	−0 .010 (0.005)	−0.015 (0.011)	−0.018 (0.009)	−0.026 (0.022)	−0.022 (0.021)	−0.026 (0.014)	**Speed 0.036** Foot 0.73 Int. 0.37

immediately by increased dorsi-flexion resistance which ‘slowed’ (passively controlled) the forwards shank rotation occurring during prosthetic limb single-support. This was evidenced by the significantly reduced forwards shank angular velocity when using the MPC foot compared to other two feet. There was also significantly more negative work done by the ankle-foot device (i.e. at the UDS) during single-support when using the MPC foot compared to the other two ankle types, and greater for the non-MPC compared to elastic foot. This increase in negative work suggests the ‘plantar-flexion knee extension couple’ effect was greater when using the MPC compared to non-MPC or elastic foot, and greater for both hydraulic ankle-foot conditions compared to the elastic foot condition. This likely explains why there was a trend ($P = 0.083$) towards a reduction in negative mechanical work done at the residual-knee when using the MPC foot: i.e. because the device (i.e. the UDS) absorbed more mechanical power, the residual-knee did not have to absorb as much. The above results suggest that stance-phase dynamic stability was better using the MPC foot compared to the other two foot types. To determine further how dynamic stability was affected by foot type, we retrospectively determined the mean forwards (A–P) centre of pressure (CoP) velocity during single-support. This analysis found that CoP velocity was lower for the slow compared to customary speed level ($P < 0.001$), but there was no main effect of foot type ($P = 0.078$). However, there was a significant speed-level by ankle type interaction ($P = 0.006$). This interaction indicated that mean CoP velocity was significantly lower for the MPC compared to non-MPC and elastic feet at the slow speed level, while at the customary speed level, mean CoP velocity was again lower for the MPC foot but only compared to the non-MPC foot.

During overground gait, when using a rigid or elastically articulating ankle–foot device, there is typically a small burst of positive ‘ankle’ power during early stance, which is likely due to elastic recoil of the heel-keel (De Asha et al., 2013b). This energy return cannot aid propulsion as it occurs early in the stance phase, and may in fact exacerbate early heel rise—a common issue for UTAs. Such ‘inappropriately’ timed energy return has been shown to become reduced when participants switch to using a hydraulic ankle–foot device (De Asha et al., 2013b). During ramp descent such energy return from the heel could potentially contribute to increasing forwards shank rotation and hence knee flexion. In the present study, there was a burst of UDS positive power in early stance when using the elastic foot, which was not evident when using either the MPC or non-MPC feet (Fig. 2). This positive power burst would be due to energy being returned by recoil of the heel-keel, which must have been dissipated within the hydraulic ‘ankle’ mechanism when using the MPC or non-MPC feet. This highlights, “in a nutshell”, one of the advantages of viscoelastic ankle–foot devices over elastic devices.

When walking down ramps, knee flexion during early stance has been shown to be increased in UTAs compared to that when walking overground (Fradet et al., 2010; Vickers et al., 2008; Vrieling et al., 2008). This increased knee flexion helps the prosthetic foot attain a foot-flat position but potentially also reduces residual-knee stability during prosthetic limb stance because of the increased load it places on the residuum (Vickers et al., 2008). During overground walking, UTAs have been found to employ co-contraction of the residual-limb hamstrings and quadriceps, in order to stabilise the residual-knee (Isakov et al., 1996, 2000), and the amplitude of the associated muscle contraction becomes increased when walking down slopes (Vickers et al., 2008). Such co-contractions contribute, in overground gait, to UTAs having higher metabolic costs compared to the able-bodied (Barth et al., 1992), and thus the metabolic costs are presumably increased further for ramp descent. In the present study, the reduced residual-knee flexion during early stance and trend towards reduced negative mechanical work at the residual-knee when using the MPC foot compared to the other two feet, suggests there could potentially be a metabolic saving when descending ramps using a MPC foot compared to non-MPC or elastically articulating feet. However, the MPC foot weighs approximately 0.3 kg more than the non-MPC foot which is 0.5 kg more than the elastic foot. Thus any stance-phase savings in metabolic costs may be negated because of higher swing-phase metabolic costs associated with a change in mass distribution (Lehmann et al., 1998) but future work is required to confirm this.

There were a number of limitations to the present study. With only one force platform within the ramp system, only trials where the prosthetic limb landed on the force platform were undertaken. Expanding the protocol to include trials in which the intact limb landed on the platform would have increased the likelihood of participants becoming fatigued. We chose to remotely ‘trigger’ the MPC foot into its ‘ramp descent’ mode using Bluetooth connection to the device. Although this is not what would happen when such a device is used in the ‘real world’, we felt that it was more important to assess the effects of the hydraulic resistance alterations associated with the device's ‘ramp descent’ mode, rather than to test if and when switching to this mode occurred. All participants were familiarised to using either an *Elan* or *Echelon VT* foot, both of which are hydraulic ankle–foot devices. Although we found that the type of habitual foot participants used had minimal effect on results, having only 20 min to become familiar with the non-hydraulic *Epirus* foot, may mean the issue of familiarity is a confounding factor. The relatively short familiarisation period may explain why certain parameters investigated returned relatively high group standard deviation values (Tables 1 and 2). Finally, there was no comparison of ramp-descent with a non-articulating (rigid ankle) prosthetic ankle–foot. This was because it was felt that a fourth foot condition would be problematic due to potential fatigue issues for participants, and/or that the biomechanical compensation required when using such feet might have ‘carry-over’ effects to the other foot conditions, particularly as all participants habitually used an articulating ankle–foot device. It would be expected that the time to foot-flat would be longer when using a non-articulating ankle–foot than for any of the three attachments used in the present study, due to the requirement of having to increase residual-knee flexion to achieve foot-flat when using such feet (Fradet et al., 2010; Vickers et al., 2008); although future work would need to confirm this.

Conflicts of interest

None.

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