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Impact on the biomechanics of overground gait of using an 'Echelon' hydraulic ankle–foot device in unilateral trans-tibial and trans-femoral amputees

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ABSTRACT

Background: If a prosthetic foot creates resistance to forwards shank rotation as it deforms during loading, it will exert a braking effect on centre of mass progression. The present study determines whether the centre of mass braking effect exerted by an amputee's habitual rigid 'ankle' foot was reduced when they switched to using an 'Echelon' hydraulic ankle–foot device.

Methods: Nineteen lower limb amputees (eight trans-femoral, eleven trans-tibial) walked overground using their habitual dynamic-response foot with rigid 'ankle' or 'Echelon' hydraulic ankle–foot device. Analysis determined changes in how the centre of mass was transferred onto and above the prosthetic-foot, freely chosen walking speed, and spatio-temporal parameters of gait.

Findings: When using the hydraulic device both groups had a smoother/more rapid progression of the centre of pressure beneath the prosthetic hindfoot ($p \leq 0.001$), and a smaller reduction in centre of mass velocity during prosthetic-stance ($p < 0.001$). As a result freely chosen walking speed was higher in both groups when using the device ($p \leq 0.005$). In both groups stance and swing times and cadence were unaffected by foot condition whereas step length tended ($p < 0.07$) to increase bilaterally when using the hydraulic device. Effect size differences between foot types were comparable across groups.

Interpretation: Use of a hydraulic ankle–foot device reduced the foot's braking effect for both amputee groups. Findings suggest that attenuation of the braking effect from the foot in early stance may be more important to prosthetic-foot function than its ability to return energy in late stance.

1. Introduction

Commonly available energy storing and return (ESR) and/or dynamic-response prosthetic-feet (DRF) store and return energy during stance via elastic deformation and recoil. This energy return is intended to replicate the 'power burst' seen at the intact-ankle during terminal stance/pre-swing. However, as there are no active mechanisms involved, this 'power burst' is of a far lower magnitude than that seen in an intact-limb (Sanderson & Martin, 1997; Sadeghi et al., 2001; Ventura et al., 2011), so much so that a recent review article concluded that passive prosthetic devices are "not capable of providing the individual with sufficient ankle power during gait" (Versluys et al., 2009); although the authors did not clarify how 'sufficient' should be defined. It has recently been demonstrated that the use of a prosthetic-foot (Echelon; Chas. A Blatchford & Sons Ltd., Basingstoke, UK) incorporating an 'ankle' device allowing hydraulically damped sagittal plane articulation between a DRF and the shank resulted in significant changes to the biomechanics of overground ambulation in active unilateral trans-tibial amputees (TTs; De Asha et al., 2013a,b). From henceforth we refer to this foot here as a hydraulic ankle–foot device (*hy*A-F). These changes included a reduction in the amount/magnitude of inappropriate disruptions to the forwards progression (i.e. a 'stall' or a negative displacement) of the centre of pressure (CoP) beneath the prosthetic-foot — particularly under the hindfoot during early stance (De Asha et al., 2013a) and an increase in angular velocity of the prosthetic-shank during early stance (De Asha et al., 2013a). These changes resulted in an increase in freely chosen walking speed (De Asha et al., 2013a,b). CoP progression reflects how the whole-body centre of mass (COM) is transferred onto and above the prosthetic-foot (De Asha et al., 2013a). Higher walking speeds decrease the temporal asymmetries in amputee gait (Nolan et al., 2003) and such speed increases are positively correlated with amputees' self-perception of gait quality (Miller et al., 2001). The increase in freely chosen walking speed when using the

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*hy*A-F occurred despite an increase in power absorption and reduction in power return (i.e. increased energy dissipation) by the prosthetic-foot and a concurrent reduction in mechanical work done per metre travelled by the intact-limb (De Asha et al., 2013b). The increased energy dissipation would have been due to the hydraulically damped 'ankle' articulation. This suggests that the speed increases were not a result of an increase in 'propulsion' but were driven by an attenuation of the 'braking effect' (Silverman & Neptune, 2012) that would have otherwise been exerted by the prosthetic-foot during prosthetic-limb stance. This braking effect comes from resistance to forwards shank rotation as the foot deforms during loading, which prolongs the period in which the fore-aft ground reaction force (GRF) is directed posteriorly and/or increases the magnitude of this force. This force determines fore-aft accelerations of the COM, and thus the net (from both limbs) fore-aft GRF impulse will determine the change in COM forwards velocity during prosthetic limb stance. Anecdotally this 'braking effect' is perceived by amputees as a sensation of having to 'climb over' the prosthetic-limb, or a 'stop' or a 'dead-spot' during prosthetic-limb stance. Such 'braking effect' may be greater in unilateral trans-femoral (TF) amputees compared to TTs because of the requirement for the prosthetic-knee to remain stable and/or 'locked' during stance.

The aim of the present study was to determine whether the COM 'braking effect' exerted by unilateral trans-tibial and trans-femoral amputees' habitual rigid 'ankle' foot was reduced when they switched to using an 'Echelon' hydraulic ankle–foot device. It was hypothesized that the use of a *hy*A-F would attenuate the 'braking effect' during prosthetic-limb stance compared to when participants used their *rigF*, and thus the reduction in COM velocity during prosthetic-limb single support would not be as great, and thus walking speed would be higher, when using the *hy*A-F compared to when using a *rigF*. Furthermore, it was hypothesized that because TFs use a prosthetic-knee device which is required to 'lock' during stance, switching to a *hy*A-F from their habitual rigid 'ankle' DRF (*rigF*) would result in a greater increase in walking speed and a greater reduction in inappropriate CoP trajectory fluctuations than that in TTs. Finally, as an increase in walking speed has been previously found to lead to reductions in spatio-temporal asymmetries in lower-limb amputees (Nolan et al., 2003), it was also hypothesized that spatio-temporal inter-limb symmetry would be improved in both groups as a consequence of the increased freely chosen walking speed when using the *hy*A-F.

2. Methods

2.1. Participants

Nineteen physically active unilateral lower-limb amputees (eight TFs mean (SD); age 42 (14.8) years, mass 86.3 (15.3) kg, height 1.74 (0.06) m and eleven TTs mean (SD); age 47 (10.3) years, mass 84.5 (17.3) kg, height 1.79 (0.07) m) took part, each giving written informed consent prior to their involvement. All had undergone amputation at least two years prior to participation (range 2–45; TF, mean 17.4 (15.8) years; TT, mean 12.5 (13.7) years), were free from neurological, musculoskeletal (other than limb amputation) or cardiovascular disorders, and had used their current prostheses for at least six months (Table 1). All participants were classed as at least K3 on the Medicare scale by their prescribing clinician and all habitually used prosthetic-foot devices that would be classed as a DRF, all of which had rigid, non-articulating 'ankle' attachments. The study was conducted in accordance with the tenets of the Declaration of Helsinki and Institutional bioethics committee approval was obtained.

2.2. Prosthetic intervention

All prosthetic alterations were made by an experienced prosthetist. Everything about the prosthesis was kept as near to constant as possible when one foot type was exchanged for the other. The overall length, socket, suspension, set-up and alignment of knee-device (TF) and shank were unchanged across foot types. Once the *hy*A-F was fitted, participants walked both indoors and outdoors for a minimum of 45 min or until each felt comfortable, whichever was longer, 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 'ankle' articulation (damping) within the *hy*A-F were adjusted until deemed to provide 'optimal function' at self-selected, customary walking speed. The adjustment consisted of systematically altering the levels of damping of both plantar- and dorsi-flexion while each participant walked using the device. The final settings were decided upon using a mixture of participant feed-back regarding perceived comfort and function and the prosthetist's experience.

Table 1
Individual descriptive details for participants.

Age (years)	Time since amputation (years)	Level	Reason for amputation	Foot	Knee
30	12	TF	T	FlexFoot	EUK SAKL
31	28	TF	T	Elite	KX06
61	45	TF	T	Seattle Litefoot	3R 45
42	12	TF	S	1D 10	Total knee
46	32	TF	S	Flexfoot	Total knee
38	2	TF	T	Variflex	KX06
65	3	TF	T	Esprit	Total knee
23	5	TF	S	Esprit	KX06
34	8	TT	T	Esprit	n/a
62	19	TT	T	Esprit	n/a
39	9	TT	T	Esprit	n/a
38	5	TT	T	Esprit	n/a
61	3	TT	T	Esprit	n/a
39	6	TT	T	Esprit	n/a
45	2	TT	T	Esprit	n/a
40	2	TT	T	Esprit	n/a
54	8	TT	T	FlexFreedom	n/a
60	45	TT	T	Seattle Litefoot	n/a
48	30	TT	T	Elite	n/a

TF trans-femoral, TT trans-tibial.
T trauma, S sarcoma.

2.3. Data acquisition and processing

Participants completed two blocks of 10 walking trials; one block was undertaken using their *rigF* 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 (as detailed above). Participants completing trials using their *rigF* in the first block (block 1), completed these on arrival at the laboratory. For those completing trials using their *rigF* 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.

Participants walked in a straight line along a flat and level 8 m walkway at their freely-chosen walking speed, i.e. irrespective of foot condition they were instructed to "walk as they would normally". 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 force platforms (AMTI, Watertown, MA, USA) using the approach described previously (De Asha et al., 2013a). 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. Reflective markers were placed bilaterally on key body landmarks as described previously (De Asha et al., 2013a).

Labelling and gap filling of marker trajectories were undertaken within Workstation software (Vicon, Oxford, UK). The resultant C3D files were then exported to Visual 3D motion analysis software (C-Motion, Germantown, MD, USA), where a nine segment (head, thorax/abdomen, pelvis, thighs, shanks and feet) 6DoF model (Cappozzo et al., 1995) of each participant was constructed. The COM was calculated within Visual 3D as the weighted average of the nine segments. Kinematic and kinetic data were filtered using a fourth order, zero-lag Butterworth filter with a 6 Hz cut-off. 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. Stance was defined as being from IC to ipsilateral limb TO. Swing was from TO to ipsilateral IC. As there were no kinetic data for the intact-limb, IC and TO on the intact-limb were defined using kinematic data: IC was defined as the instant of prosthetic-limb peak hip extension (De Asha et al., 2012) and TO as instant of peak posterior displacement of the intact-toe marker relative to the pelvis (Zeni et al., 2008).

2.4. Data analysis

The following parameters were determined: average walking speed; aggregate negative CoP displacement beneath the prosthetic foot; timing (percentage of stance) of when the CoP passed anterior to the distal end of the prosthetic-shank; prosthetic- and intact-limb stance time, step time and step length (and inter-limb asymmetry in these measures) and cadence. We have previously shown that when TTs used a *hy*A-F compared to *rig*F, CoP fluctuations beneath the prosthetic foot were attenuated and angular velocity of the prosthetic-shank during early stance and freely chosen walking speed were increased (De Asha et al., 2013a). We highlighted that these findings indicated that the COM was transferred onto the limb in a smoother less faltering manner (De Asha et al., 2013a). Therefore in the present study we also determined the reduction in instantaneous walking speed (COM forwards velocity) during prosthetic-limb stance to determine the 'braking effect' exerted by the foot as it passively yielded during loading. COM forwards velocity maximum during prosthetic-limb stance was also determined. Walking speed was defined as the mean forwards (antereo-posterior) velocity of the COM through the central portion of the data collection volume (length ~ 3 m, where/when participants were walking at a steady state). The aggregate negative CoP displacement was determined as the accumulative distance travelled by the CoP in the negative antereo-posterior direction (i.e. opposite direction to the direction of travel) under the prosthetic-foot. Stance and swing time were defined as the absolute duration of stance and swing phases, respectively. Step length was defined as the antereo-posterior distance between the heel of one foot and the heel of the contralateral foot for consecutive foot-falls. Inter-limb asymmetry was assessed by dividing the intact-limb value of a parameter by the corresponding prosthetic-limb value: thus a value of '1' indicated perfect inter-limb symmetry while values higher than '1' indicated degree of asymmetry with the value of the parameter higher on the intact-limb. Cadence was defined as the number of steps per minute. The reduction in instantaneous walking speed was defined as the difference between the COM forwards velocity at intact-limb TO and COM forwards velocity minima during prosthetic-limb single support (see Fig. 1). The above variables were calculated for each individual trial and then averaged across trial repetitions to give a mean value for each foot condition per participant.

2.5. Statistical analyses

Data were analysed using mixed-mode repeated measures ANOVAs with group (TF, TT) as 'between' factor and foot type (*rigF*, *hy*A-F) as 'within' factor. Where comparison between the intact and prosthetic

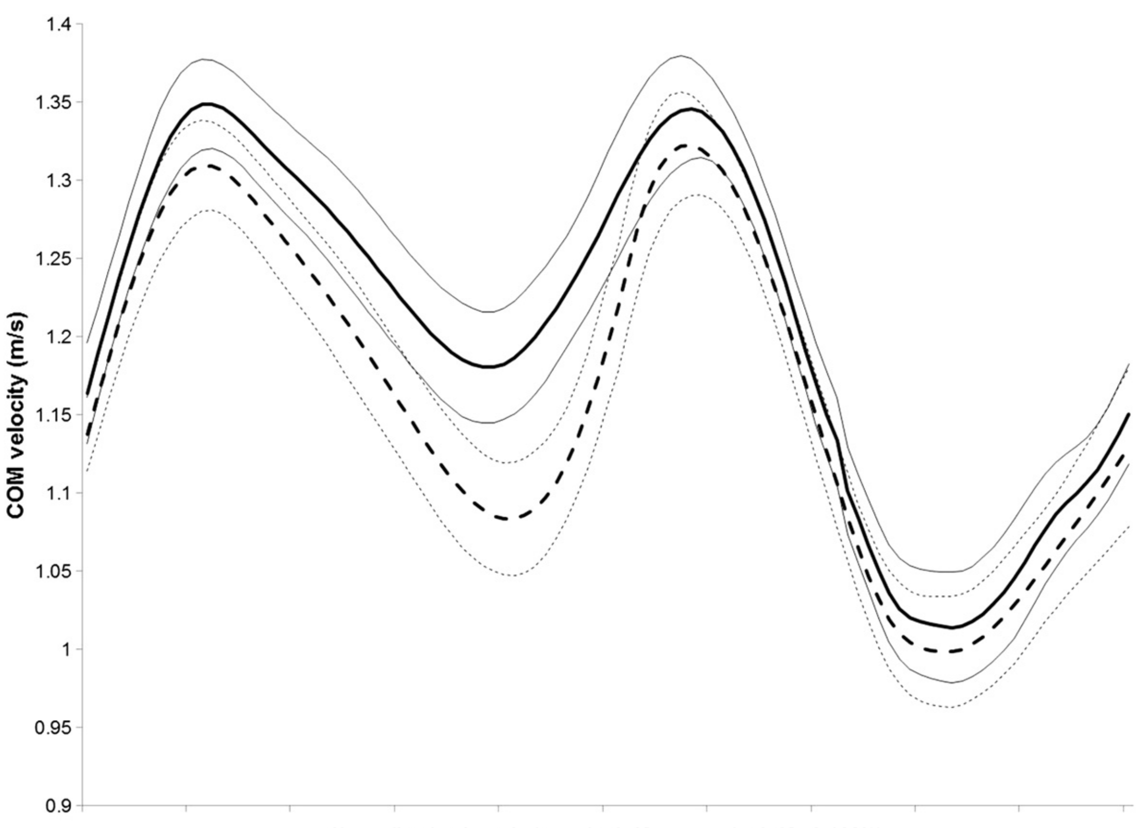

Fig. 1. Centre of mass velocity profiles when using a *hy*A-F (solid line) and *rigF* (broken line) for one exemplar trans-tibial participant. Data are mean and SD of 10 repeat trials.

sides was made (e.g. step length, swing time), ‘limb’ (intact, prosthetic) was also included as an additional ‘within subjects’ factor. Differences between foot type conditions were also determined as effect size differences; calculated as Cohen's ‘d’ (Cohen, 1977), and this was done separately for each group. Statistical analyses were made using Statistica for Windows (StatSoft, Tulsa, OK, USA). The alpha level was set at 0.05.

3. Results

Unless stated otherwise, there were no significant interactions between group and foot condition (see Tables 2 & 3 for details). There were significant main effects of foot condition and group on freely chosen walking speed ($p < 0.005$). Walking speed was significantly faster when the *hy*A-F was used and was significantly faster for TTs than TFs. There was a significant main effect of foot condition ($p = 0.001$) but no main effect of group ($p = 0.59$) on aggregate negative displacement of the CoP trajectory. The aggregate negative CoP displacement was reduced when the *hy*A-F was used. There was a significant main effect of foot condition ($p < 0.001$) but no main effect of group ($p = 0.53$) on the instant the CoP passed anterior to the shank. The CoP passed anterior to the shank earlier in stance when the *hy*A-F was used.

Instantaneous COM velocity at intact-limb TO was unchanged across foot conditions ($p = 0.89$) but instantaneous COM velocity minimum during the subsequent prosthetic-limb single support phase was higher when using the hydraulic device ($p < 0.001$, Fig. 1). As a result there was significantly less slowing of COM velocity (walking speed) during prosthetic-limb single support for both groups when using the *hy*A-F compared to *rigF* ($p < 0.001$). Peak COM velocity during prosthetic-limb stance was unchanged across foot conditions ($p = 0.1$). All instantaneous COM velocity values (Table 2) were significantly higher for TTs than TFs ($p \leq 0.045$). Effect size differences (d) between foot conditions tended to be smaller in TFs (mean; 0.48 (0.22)) compared to TTs (mean; 0.65 (0.31), Table 2).

There was no effect of foot condition ($p = 0.50$) on swing time, but a significant main effect of limb ($p < 0.001$) and a group-by-limb interaction ($p < 0.001$) indicated that swing time was longer for the prosthetic-limb compared to the intact-limb and the difference between limbs was greater for the TFs compared to TTs. There was no effect of foot condition ($p = 0.57$) on stance time, but a significant main effect of limb ($p < 0.001$) and a group-by-limb interaction ($p < 0.001$) indicated that stance time was longer on the intact-limb compared to the prosthetic-limb and the difference between limbs was greater for the TFs compared to TTs. There was a significant main effect of limb ($p < 0.001$) but no effect of foot condition ($p = 0.062$) or group ($p = 0.069$) on step length. Step length was higher on the prosthetic-limb compared to the intact-limb (Table 3). There were no significant main effects of foot condition ($p = 0.84$) or group ($p = 0.063$) on cadence. There were no significant effects of foot condition on inter-limb asymmetry in swing time ($p = 0.22$), stance time ($p = 0.77$) or step length ($p = 0.70$, Table 3). Swing and stance time inter-limb asymmetry were significantly higher for TFs than for TTs ($p < 0.001$) but there was no group effect on step length inter-limb symmetry ($p = 0.17$).

4. Discussion

The results indicated that, in both groups, the reduction in instantaneous COM velocity during prosthetic-limb single support was not as great when the *hy*A-F was used. Thus our hypothesis that the use of a *hy*A-F would reduce the ‘braking effect’ exerted by the foot during stance was supported. The differences in reduction in instantaneous COM velocity during prosthetic-limb stance, in freely chosen walking speed, and in CoP measures between foot conditions were comparable across TTs and TFs. Thus our hypothesis, that TFs would be affected to a greater extent by switching to using a hyA-F than TTs, was not supported.

Table 2
Group mean (SD) freely chosen walking speed, CoP trajectory measures and instantaneous COM velocities when using a *hy*A-F and *rigF* in trans-tibial (TT) and trans-femoral (TF) amputee participants.

	TT		TF		p value	Foot effect size d	
	*hy*A-F	*rigF*	*hy*A-F	*rigF*		TT	TF
Walking speed (m/s)	1.22 (0.11)	1.14 (0.14)	0.99 (0.10)	0.94 (0.11)	**Foot 0.005** **Group <0.001** Int 0.12	0.5	0.4
Time CoP anterior to shank (% stance)	31.8 (4.1)	34.9 (3.5)	26.9 (6.4)	33.7 (7.7)	**Foot <0.001** Group 0.53 Int 0.76	0.6	0.7
Negative CoP displacement (cm)	0.82 (0.64)	2.11 (0.97)	0.35 (0.39)	1.21 (1.51)	**Foot 0.001** Group 0.59 Int 0.46	1.1	0.6
COM velocity at intact-limb TO (m/s)	1.30 (0.15)	1.29 (0.13)	1.09 (0.17)	1.09 (0.14)	Foot 0.89 **Group 0.019** Int 0.86	n/a	n/a
COM velocity maxima during single support (m/s)	1.35 (0.19)	1.32 (0.19)	1.16 (0.20)	1.14 (0.21)	Foot 0.10 **Group 0.045** Int 0.82	n/a	n/a
COM velocity minima during single support (m/s)	1.09 (0.15)	1.01 (0.15)	0.89 (0.20)	0.83 (0.17)	**Foot <0.001** **Group 0.026** Int 0.39	0.4	0.2

Bold text indicates significant effects and the p-value of such effects.

Table 3
Group mean (SD) spatio-temporal variables for each limb and resulting inter-limb symmetry index (S.I.) when using a *hy*A-F and *rigF* in trans-tibial (TT) and trans-femoral (TF) amputee participants.

	TT						TF						*p* value
	*hy*A-F			*rigF*			*hy*A-F			*rigF*			
	Int	Pros	S.I.	Int	Pros	S.I.	Int	Pros	S.I.	Int	Pros	S.I.	
Swing time (s)	0.54 (0.11)	0.55 (0.09)	0.98 (0.06)	0.54 (0.10)	0.54 (0.09)	1.0 (0.08)	0.42 (0.05)	0.57 (0.02)	0.74 (0.12)	0.42 (0.06)	0.57 (0.04)	0.74 (0.09)	Foot 0.50 **Group 0.19 **Limb <0.001**
Stance time (s)	0.79 (0.08)	0.77 (0.07)	1.03 (002)	0.79 (0.09)	0.76 (0.08)	1.04 (0.05)	0.90 (0.09)	0.76 (0.09)	1.19 (0.11)	0.92 (0.11)	0.78 (0.10)	1.19 (0.09)	Foot 0.57 **Group 0.13 **Limb <0.001**
Step length (m)	0.68 (0.06)	0.74 (0.08)	0.91 (0.07)	0.67 (0.07)	0.71 (0.07)	0.95 (0.14)	0.60 (0.09)	0.69 (0.09)	0.88 (0.12)	0.58 (0.10)	0.68 (0.09)	0.85 (0.12)	Foot 0.062 Group 0.069 **Limb <0.001**
Cadence (steps per minute)	96.8 (6.6)		n/a	97.2 (6.4)		n/a	91.0 (6.6)		n/a	90.2 (8.6)		n/a	Foot 0.84 Group 0.063 Int 0.53

** Significant group-by-limb interaction effect ($p < 0.001$). Bold text indicates significant effects and the p-value of such effects.

Finally, there was no change in either group between foot conditions in spatio-temporal inter-limb asymmetry, therefore our hypothesis that symmetry would be improved when using a *hy*A-F, as a consequence of an increased freely chosen walking speed, was not supported.

The increased walking speed, in both groups, when participants switched from using their habitual DRF with rigid 'ankle' to using a *hy*A-F was similar to that previously reported for a larger group of TTs who, similarly to the present study, used a variety of different *rig*Fs (De Asha et al., 2013a) as well as for a group of TTs who all habitually used the same type of *rig*F (De Asha et al., 2013b). Absolute increases were larger for TTs (~+0.08 ms^{-1}) than for TFs (~+0.05 ms^{-1}) but as TTs tended to walk faster than TFs effect size differences were comparable. While these absolute increases may seem small, they amount to medium effect-sized increases of approximately 7% for TTs ($d = 0.5$) and 5% for TFs ($d = 0.4$): suggesting that the increases to freely chosen walking speed are likely to prove to be clinically meaningful. Walking speed is a function of step length and cadence. The increased walking speed when using a *hy*A-F in both groups occurred without any significant increases to these component parameters; although there was a trend across groups ($p = 0.069$) for step length to increase bilaterally. It would appear that this increase in step length was sufficient, in combination with a maintained cadence, to result in a significantly increased freely chosen walking speed. The increase in walking speed, in both groups, was coincident with a less disrupted progression of the CoP (i.e. reduced aggregate negative CoP displacement) during prosthetic-limb stance. This reduction in inappropriate negative displacement resulted in the CoP moving anterior to the distal end of the prosthetic-shank sooner when using the *hy*A-F. It has previously been shown in TTs (De Asha et al., 2013a; Ranu, 1988) and TFs (Schmid et al., 2005), that the CoP forwards progression beneath the prosthetic-foot can become delayed in the hindfoot area during early stance. While this delay became reduced, in the present study, in TFs (~27% stance, *hy*A-F; ~34% stance *rigF*) and TTs (~32% stance, *hy*A-F; ~35% stance *rigF*) when they switched to using a *hy*A-F, the time periods the CoP spent beneath the hindfoot area were still considerably longer than those seen in able-bodied gait (~9% stance; Kirtley, 2006). In the present study the CoP passed anterior to the shank slightly, but consistently, earlier during stance in TFs compared to TTs regardless of foot type; although there was no statistical difference between groups. This may have been because of TFs' shorter prosthetic-limb step length resulting in the shank being at a more vertical angle at initial contact, which would have helped keep the knee extended and 'locked' during stance. In turn, this resulted in the COM transferring onto and over the limb sooner and hence the CoP progressing beneath the prosthetic- foot more rapidly. Disruptions to how the CoP progresses beneath the prosthetic-foot reflect how the COM is transferred onto and over the prosthetic-limb, thus improvements leading to a smoother, less disrupted, progression of the CoP beneath the prosthetic hindfoot reflect a smoother COM transfer onto the prosthetic-limb. For the TFs in the present study this earlier progression of the CoP beyond the horizontal position of the bottom of the shank when using the *hy*A-F compared to *rigF* occurred despite a concurrent (non-significant) increase in prosthetic-limb step length. Once the CoP passed anterior to the shank, the forefoot would have become loaded and the external moment at the 'ankle' would have changed from one tending to 'plantarflex' to one tending to 'dorsiflex'. Consequently onset of passive control of forwards shank rotation (and thus transfer of the COM over the limb) would have occurred sooner during stance when using the *hy*A-F.

The use of a *hy*A-F in TT participants has been shown to lead to higher angular velocities of the prosthetic-shank during stance (De Asha et al., 2013a); improving 'roll-over' above the prosthetic-foot. In the present study, we found that the reduction in instantaneous COM forwards velocity (walking speed) during prosthetic-limb single support was not as great in either amputee group when they used the *hy*A-F. It should be highlighted that the instantaneous COM velocity at instant of intact-limb TO and the COM velocity maxima during prosthetic-limb stance were no different between foot conditions. It was only the instantaneous COM velocity minimum that was affected (Fig. 1). This provides direct support to the notion that the COM was transferred onto and above the prosthetic limb in a smoother less faltering manner when using a *hy*A-F, and that an accompanying reduction in the 'braking effect' was the main reason for an increased freely chosen walking speed. To gain further insight/evidence that the increased walking speed was due to a reduced 'braking effect' (rather than to an increase in forwards propulsion), we retrospectively investigated the antereo-posterior component of the GRF (GRF_{A-P}) for the period of prosthetic-limb single support. This was done by comparing peak force and impulse of the negatively (braking) and positively (propulsive) directed GRF_{A-P}, normalised to body-weight (BW) during this period. Given that larger GRFs are associated with higher walking speeds we expected that all GRF measures would be increased when using the *hy*A-F as a reflection of the increased walking speed. Our analysis indicated that all GRF_{A-P} measures were significantly higher for TTs than TFs ($p \leq 0.032$, Fig. 2); likely as a result of their higher walking speed. The peak propulsive force, propulsive force impulse and peak braking force during prosthetic-limb single support were higher when the *hy*A-F was used; although not significantly so ($p \geq 0.072$). These higher GRF_{A-P} values were likely due to the increased freely chosen walking speed when using the *hy*A-F. However, braking force impulse was lower when the *hy*A-F was used, although again not significantly so ($p = 0.061$, Fig. 2). This reduced braking impulse when using the *hy*A-F (despite an increase in walking speed) provides further support to the notion that the *hy*A-F exerted a reduced 'braking effect' on COM forwards velocity compared to when using the *rigF*. This reduced 'braking effect' helps explain why it has previously been reported that there is a reduction in intact-limb mechanical work done when using a *hy*A-F compared to *rigF* (De Asha et al., 2013b), however any future study should include measurement of metabolic costs to confirm or refute the suggestion of a physiological energetic saving. In general, such 'braking effect' may be a result of the deformation of the prosthetic-foot 'bottoming', i.e. deflection of the heel keel reaches its maximum. Indeed, decreasing foot keel stiffness has been linked with increased 'braking' (Fey et al., 2013). This 'braking effect' may also result from difficulties in how loading of the foot transfers from the heel keel to the forefoot keel. In our earlier report we showed that the use of a *hy*A-F in TTs resulted in less intact-limb mechanical work done per metre travelled despite more power being absorbed by, and less returned from, the prosthetic-foot (De Asha et al., 2013b). This, together with

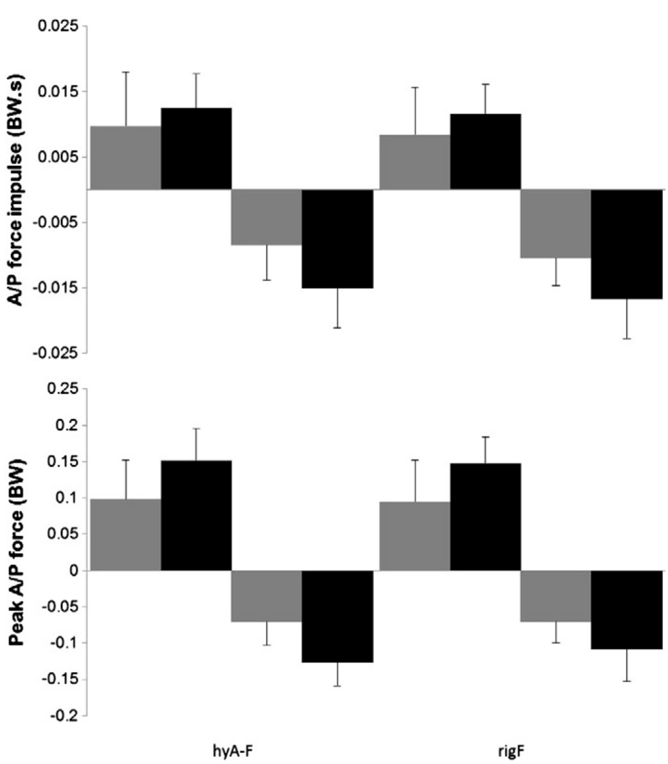

Fig. 2. Group mean (SD) peak force and impulse in propulsive (positive) and braking (negative) GRF_{A-P}, for period of prosthetic-limb single support when using a *hy*A-F and *rig*F in trans-tibial (black bars) and trans-femoral (grey bars) amputee participants (no significant differences between foot conditions $p \geq 0.061$).

the results from the present study, suggests that, contrary to 'common wisdom', improvements in passive prosthetic-foot 'function' may be less dependent on energy return during late stance than on reducing resistance to COM progression ('braking effect') during early-to-mid stance.

During able-bodied gait peak COM forwards velocity typically coincides with minimum COM height during the middle portion of double support (Alexander, 1984). In the present study, peak COM forwards velocity was found to occur just after TO. As this finding appeared to be anomalous with the literature, we retrospectively analysed the COM forwards velocity and COM vertical displacement profiles along with net GRF_{A-P} for the initial prosthetic double support period for five of our participants where we happened to have GRF data from both limbs for consecutive foot falls (along with an exemplar able-bodied individual who completed an identical protocol; see Supplementary material). This analysis allowed us to better understand the relationship between the timings of peak COM velocity, minimum COM height and the instant the net GRF_{A-P} changed from propulsion to braking. These data consistently showed that minimum COM height and peak COM forwards velocity did not occur synchronously for amputees. In able-bodied gait these events are synchronous, which is what we found for our exemplar able-body individual. Furthermore, it was evident that amputees maintained a net propulsive (positive) GRF_{A-P} impulse throughout double support, or at least until much later in double support than is seen in able-bodied gait. This is likely reflective of having increased compensatory propulsion at the intact-limb coupled with relatively (compared to the intact-limb) reduced braking at the prosthetic-limb. This meant that (unlike able-bodied gait) peak COM forwards velocity occurred while the COM was being elevated. Given that up to 70% of overall energy expenditure during gait occurs during step-to-step transition (Donelan et al., 2002) this divergence from a standard 'inverted pendulum' model (Kuo, 2007) may contribute to the higher metabolic cost of gait typically reported for amputees in comparison to able-bodied individuals. It is also worth emphasising that in the current study initial contact and toe off events for the intact-limb were created using kinematic algorithms (due to the absence of kinetic data), and their use may have introduced a small systematic bias. This is likely because the algorithms used were developed using data from able-bodied gait. This may be why peak COM forwards velocity occurred just after TO, rather than at or just before TO. We do not believe, however that this limitation detracts from the finding of reduced 'braking effect' during prosthetic-limb stance.

Walking speed is a primary measure of gait function in amputees (Waters et al., 1976). Speed increases are also positively correlated with lower limb amputees' self-perception of gait quality (Miller et al., 2001) and increases in speed not only reflect improved function but also coincide with a decrease in spatio-temporal inter-limb asymmetries (Nolan et al., 2003). As in previous studies (Highsmith et al., 2010; Nolan et al., 2003; Raggi et al., 2009) the results of the present study indicated that TFs had a more temporally asymmetrical gait than TTs. Step length asymmetry was higher for TFs than TTs but this difference was not significant which is most likely reflective of all participants being active with 'good' gait function. Contrary to our hypothesis, the increase in walking speed when using the *hy*A-F was not reflected by improvements in inter-limb symmetry. In the present study, the lack of change in inter-limb asymmetry between foot conditions, despite a significant increase in freely chosen walking speed, suggests that an improvement in gait function does not necessarily result in an increase in symmetry, and conversely, suggests that an increase in symmetry may not necessarily improve function. The goal of improving inter-limb symmetry in lower-limb amputees has been previously questioned because 'fallers' were distinguished from 'non-fallers' by higher levels of load-bearing symmetry during quiet stance (Hlavackova et al., 2009). Interestingly the present study highlights that there was considerable inter-limb asymmetry in the COM velocity profiles for the stance-phase of each limb with such asymmetry becoming increased when using the *hy* A-F with a larger instantaneous COM velocity reduction during the intact-limb compared to prosthetic-limb stance phase (Fig. 1). This corroborates Tesio et al. (1998) who reported similar inter-limb asymmetry in the COM mechanical energy profile for the stance phase of each limb in both unilateral TT and TF amputee gaits.

Another potential limitation of the present study is the different types of habitual prosthetic-feet used by participants; even though all used a type of DRF. This may explain the high intra-group variability in aggregate negative CoP displacement. Although all participants experienced a reduction in aggregate negative CoP displacement, the size of the reduction was, obviously, limited by the magnitude of negative displacement present when using their *rigF*; which varied from as much as 4 cm to ~1 mm. It may also be related to differing levels of heel keel stiffness across participants (keel stiffness is selected using bodyweight categories), even in those with the same type of foot (10 of the 19 participants habitually used an Esprit). Similarly, TF participants used differing knee devices, which may have effected how the foot was loaded. Therefore knee function/type could potentially have been a confounding factor when comparing ankle–foot devices. However, the effects of switching to a *hy*A-F were consistent i.e. when a *hy*A-F was used every participant had reduced negative CoP displacement and all, except one TT, increased walking speed. This suggests that any confounding effect due to type of foot or knee joint was minimal. As highlighted above, the majority of participants' *rigF* was an Esprit. Thus the findings we present suggesting that the 'braking effect' the foot exerted on the COM was reduced when using a *hy*A-F may be due to an Esprit foot exerting an 'untypically' high 'braking effect' when compared to other types of DRF. However, an Esprit foot forms the base of a *hy*A-F: the only difference between these two feet is, in an Esprit the foot is attached rigidly to the shank, and in a *hy*A-F it is attached via a hydraulic 'ankle' device. Thus we don't believe that any bias regarding the number of participants habitually using an Esprit was a factor.

5. Conclusion

After switching from using their habitual DRF with non-articulating (rigid 'ankle') attachment to one with a hydraulic 'ankle' both amputee sub-groups had a significantly smoother/more rapid progression of the CoP beneath the prosthetic-foot, less COM forwards velocity reduction ('braking effect') during prosthetic-limb stance and an increased freely chosen walking speed, but spatio-temporal inter-limb asymmetries were unaffected. The effects of using the *hy*A-F were in the same direction and of a similar magnitude in trans-tibial and trans-femoral amputees: suggesting that it would be appropriate for use by active unilateral lower-limb amputees regardless of amputation level. Finally, findings suggest that attenuation of the 'braking effect' from the foot in early-to-mid stance may be more important to a prosthetic-foot's function than its ability to return energy in late stance.

Conflict of interests

None.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at http://dx.doi.org/10.1016/j.clinbiomech.2014.06.009.

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