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Gait Termination on a Declined Surface in Trans-Femoral Amputees: Impact of Using Microprocessor-Controlled Limb System

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Abstract

Background: Walking down ramps is a demanding task for transfemoral-amputees and terminating gait on ramps is even more challenging because of the requirement to maintain a stable limb so that it can do the necessary negative mechanical work on the centre-of-mass in order to arrest (dissipate) forward/downward velocity. We determined how the use of a microprocessor-controlled limb system (simultaneous control over hydraulic resistances at ankle and knee) affected the negative mechanical work done by each limb when transfemoral-amputees terminated gait during ramp descent.

Methods: Eight transfemoral-amputees completed planned gait terminations (stopping on prosthesis) on a 5-degree ramp from slow and customary walking speeds, with the limb's microprocessor active or inactive. When active the limb operated in its 'ramp-descent' mode and when inactive the knee and ankle devices functioned at constant default levels. Negative limb work, determined as the integral of the negative mechanical (external) limb power during the braking phase, was compared across speeds and microprocessor conditions.

Findings: Negative work done by each limb increased with speed ($p<0.001$), and on the prosthetic limb it was greater when the microprocessor was active compared to inactive ($p=0.004$). There was no change in work done across microprocessor conditions on the intact limb ($p=0.35$).

Interpretation: Greater involvement of the prosthetic limb when the limb system was active indicates its ramp-descent mode effectively altered the hydraulic resistances at the ankle and knee. Findings highlight participants became more assured using their prosthetic limb to arrest centre-of-mass velocity.

Keywords: Gait termination; Ramp descent; Transfemoral-amputee; Microprocessor-controlled; Above-knee prosthesis; Limb mechanical work.

1. Introduction

Walking down ramps can be a demanding task for lower-limb amputees (Vrieling et al., 2008). This is because the deformation/deflection of a prosthetic foot's heel-region/keel that occurs following heel contact (simulated plantarflexion) may be insufficient to attain foot-flat on a declined surface. Not attaining foot-flat will mean the prosthetic shank has the propensity to rotate forward during early stance, and because the whole body centre-of-mass (CoM) is posterior to the foot during this time (Kuster et al 1995; Lay et al., 2006), this can create a flexion moment at the prosthetic knee which can make the knee unstable(Vrieling 2008; Bellman 2010; Highsmith et al., 2014). Knee instability would be particularly undesirable in trans-femoral amputees (TFA) due to the lack of direct control of the prosthetic knee. As a result, TFA have to utilise certain biomechanical adaptations when descending ramps (Vrieling et al., 2008)

To help counter the difficulties lower-limb amputees have in walking down ramps, various prosthetic devices have been developed. Microprocessor-controlled (MC) hydraulic knees have been available for several years. Within these devices electronic sensors send information to a microprocessor, and via complex control-algorithms, the microprocessor changes knee resistance by altering the orifice size within the hydraulic cylinder. Such devices typically have an 'adaptive stance' mode that automatically alters the stance-phase hydraulic resistance to allow a degree of controlled knee flexion in late stance. Use of such prostheses has been shown to improve ramp descent knee kinematics (Burnfield 2012, Bellman 2012, Lura et al., 2015, Bell 2016) and associated gait pattern (Hafner et al 2007, Sawers and Hafner 2013, Highsmith 2013, Bell 2016) as well as increase descent speed (Hafner et al 2007, Burnfield 2012, Highsmith 2013). Their use has also been suggested to improve dynamic stability during descent (Bellmann et al., 2012).

Traditionally MC knee devices have been incorporated into a prosthetic limb containing a foot with either a 'rigid' ankle or an ankle device incorporating a rubber snubber providing a small range of motion(Bellmann et al., 2010; Bellmann et al., 2012; Villa et al., 2015; Vrieling et al., 2008). Recently, prosthetic feet that have a MC hydraulic articulating ankle have become available. Such foot-ankle devices also have a ramp-descent mode whereby the ankle device's plantar-flexion resistance is decreased following ground contact to facilitate attaining foot-flat, and then dorsiflexion resistance is increased to slow/control the progression of the shank-pylon over the foot. Use of MC ankle-foot devices have been shown to facilitate/improve how unilateral transtibial amputees walk down slopes in comparison to using the same foot-ankle device but having hydraulic resistances at constant default levels (Agrawal et al., 2015; Fradet et al., 2010; Struchkov and Buckley, 2015). In TFA, having a prosthetic foot that facilitates attainment of foot-flat and then slows/controls the progression of the shank-pylon over the foot would prevent the CoM ‘lurching’ forwards/downwards during ramp descent, which TFA anecdotally indicate can happen when using more traditional type feet. Recently an above-knee prosthesis that has combined and simultaneous MC of the hydraulic resistances at the ankle and knee has become clinically available. This limb system prosthesis – (commercial name Linx, Endolite, Chas A Blatchford & Sons Ltd, Basingstoke, UK) - has several modes in which the hydraulic resistances at the knee and/or ankle are simultaneously altered in response to a change in terrain and or a change in walking speed. One of these modes is aimed at helping descend ramps (‘ramp descent’ mode). According to the manufacturer, this mode alters the resistance at the ankle as per the manner described above (for MC foot-ankle device). In addition, the knee resistance in late stance is reduced to an intermediate level rather than the usual overground gait pre-swing (low resistance) level. The resulting higher limb stiffness means the limb provides a braking effect to help reduce forward /downward momentum and promote better dynamic stability in late stance.

If the use of a limb system prosthesis facilitates walking down ramps, it should also help TFA to terminate gait during ramp descent. Terminating gait (stopping) while descending slopes is likely to be even more challenging than simply walking down a ramp because of the requirement to maintain a stable/rigid limb in order for the limb to do the necessary mechanical work on the CoM to bring it under control. To arrest (dissipate) forward and downward velocity to terminate ramp descent, such work will be predominately negative. The aim of the present study was to determine how the use of a limb system prosthesis affected the external negative mechanical work done by the prosthetic and intact limbs when TFA terminated gait during ramp descent.

2. Methods

2.1 Participants

Participants were recruited on a volunteer basis (convenience sample) from a pool of TFA known, by Chas A Blatchford and Sons Ltd, to be willing to help with field testing of new prosthetic devices and currently using a prosthesis with some type of MC knee. Demographic details of the eight male TFA who participated in the study are presented in Table 1. All self-reported they had been classified by their rehabilitation consultant as K3 activity level (according to Medicare mobility scale) and had no balance, musculoskeletal, or residuum problems. Five participants had habitually used a limb system prosthesis (for technical information visit http://www.blatchford.co.uk/endolite/linx/) for at least two years; two of the other three were using an Orion with Elan foot and the other an Orion with Echelon VT (i.e., limbs with MC knee and MC or passive-hydraulic ankle devices: all Endolite devices). All participants used a full contact socket. Ethical approval was obtained from the institutional ethics committee and all participants provided written informed consent. The tenets of the Declaration of Helsinki were observed.

The three participants who did not habitually use a limb system prosthesis were provided one for the duration of the study. This was achieved by attaching a limb system device to the participant's habitual socket; and altering the length of the shank pylon so that the limb was the same length as their habitual limb and maintaining the same limb alignment. The limb system was then set-up as per manufacturer guidelines (see Clinical Manual, http://linx.endolite.co.uk/downloads). Essentially this involved completing a calibration procedure administered through the device's software, while the participant walked, on a flat level surface, at their self-selected customary walking speed, then at self-selected slow and fast walking speeds. The procedure involves sequential stages so as to determine how the hydraulic (and pneumatic, knee only) resistance levels at the ankle and knee should be altered (and when they should), to optimize features such as: 'stance release' (knee release into free swing), overground walking at multiple speeds, intermediate knee release (which provides a 'brake assist' when descending ramps), and so on. The calibration procedure for each participant was undertaken by the same experience prosthetist. The values of the impedance control parameter (e.g., stiffness and damping) for the different gait phases/tasks were heuristically fine-tuned using observations of the participant's gait and their feedback until their gait ''looked good''. The optimality criterion in this process is similar to the procedure described in (Sup 2009, Liu et al. 2014, Huang et al. 2016).

Once limb set-up was completed an accommodation period of about 20 minutes of walking over level and declined surfaces was provided. For those who habitually used a limb system prosthesis, limb alignment and the hydraulic damping levels were checked and adjusted where necessary (via 'fine tuning' procedure) to ensure they were optimal.

2.2 Data collection and processing

Participants were asked to complete repeated trials involving walking at customary and slow speeds down a 4 m long 5-degree declined walkway. Trials were repeated with the limb system's microprocessor active ('ramp descent' mode: see introduction) or inactive. Terminations were completed from both customary and slow walking speeds with the terminating limb always being the prosthetic limb. A slow speed condition was included because there is anecdotal evidence that TFA find walking down ramps more challenging at slow speeds, possibly because of difficulty controlling the prosthetic limb's forward progression over the foot during prosthetic-limb single-support. Note the limb system also incorporates a MC pneumatic cylinder which acts to provide swing-phase control at the knee. As the focus of the current study was on the mechanical limb work done to halt ramp-descent, findings only provide insight into the device's stance-phase functions.

Starting location was adjusted for each participant so that each foot landed 'cleanly' within the bounds of two adjacent sloped blocks integrated within the ramp that were positioned approximately two-thirds of the way down the ramp. The sloped blocks (solid wood) were bolted onto two floor-mounted force-platforms, and there was a gap of 2 mm (all sides) between them and the surrounding ramp (which was secured to the floor). Trials at each walking speed were repeated 10 times with a block of 5 trials completed with the MC active (*MC*on) and 5 trials with it inactive (*MC*off). When inactive (*MC*off) the knee and ankle devices provided hydraulic resistances at default levels. Default levels are set to provide the optimal function for level gait at the user's customary walking speed. Switching between MC conditions was done via a Bluetooth connection. The order in which walking speed (slow, customary) and MC condition (*MC*on/*MC*off) were completed, were counterbalanced across participants.

Ground Reaction Forces (GRF) and kinematic data were collected at 200 Hz using the two-floor mounted force-platforms (508*464mm, AMTI, Watertown, MA, USA) and a 10-camera motion capture system (Vicon MX, Oxford, UK). Retro-reflective markers (all 12.5mm diameter except on 'cluster plates' and head band which were 14 mm diameter) were placed bilaterally on the following body landmarks (or equivalent locations on the prosthesis): superior aspects of first and fifth metatarsal heads, distal end of second toe, posterior calcaneus, medial and lateral aspects of the mid-foot, medial and lateral malleoli, medial and lateral femoral condyles, greater trochanter, iliac crest directly above greater trochanter, and acromion process. Markers were also placed on vertebrae C7 and T8, sternal notch, and xiphoid process. Plate-mounted 4-marker clusters were worn on lateral aspects of thighs and shank, a skin-mounted 4-cluster was attached around the sacrum, and a sweatband with 4 markers was worn on the head. Following 'participant' calibration the markers on ankles, knees and acromions were removed.

Labelling and gap filling were done using Vicon Nexus 1.8.5 software. Data were filtered using a fourth order, zero-lag Butterworth filter with force data filtered with 20 Hz cut-off and marker trajectory data with 6 Hz cut-off. Data were subsequently exported in C3D format to Visual3D software (Version 5.02.27 C-Motion, Germantown, MD, USA) where all further processing took place. Force structures were created to represent the sloped blocks above the force-platforms, which allowed GRF and Centre of Pressure (CoP) data to be transferred to the ramp surface (for further details see: https://www.c-motion.com/v3dwiki/index.php?title=Force_Structures).

A six degrees of freedom, nine segments model (head, thorax/abdomen, pelvis, thighs, shanks and feet; Cappozzo et al., 1995) was constructed for each participant. A functional joint centre approach was used to determine intact limb, and the residual hip, joint centres using data collected in limb 'waggling' trials(Schwartz and

Rozumalski, 2005). The prosthetic ankle was defined midway between the markers placed on each side of the pylon at the same height as the corresponding markers on the intact ankle (De Asha et al., 2013). The prosthetic knee was defined by the makers placed bilaterally at the axis of the knee hinge. The location of the CoM was determined within Visual 3D as the weighted average of the nine tracked segments. Mass and CoM location for segments of the prosthetic limb were determined in the same way as the intact side.

For the two steps of gait termination, foot contact and toe-off events were determined as the instants the vertical (Z) GRF rose above or dropped below a threshold of 50 N respectively. The 50 N threshold was based on published work (Franz et al 2012). The instant of trailing/intact limb contact that indicated final bipedal standing stance was defined as the instant the CoP's velocity under the terminating/prosthetic limb went above 0.2 m/s in the medial direction following toe-off of the intact foot (Van Keeken et al. 2012).

2.3 Data analysis

Using the approach described by Donelan and colleagues (Donelan et al., 2002), global limb mechanical power was determined as the sum of limb mechanical powers in each orthogonal direction. Limb power in each direction was calculated as the dot product of the respective GRF component and the corresponding component of the CoM velocity. Limb negative work, was determined as the time integral of negative global limb mechanical power during the braking phase of each limb.

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

To determine if the hydraulic resistances at the ankle changed in the way that would be predicted by the limb system's 'ramp descent' mode, we also determined time from prosthetic limb foot-contact up to intact limb foot-off (i.e. weight transfer time, WT_{time}), and time of prosthetic limb single-support (SS_{time}). We reasoned that if the plantarflexion resistance was reduced to facilitate attaining foot flat, then WT_{time} would be reduced. Similarly, if dorsiflexion resistance was increased after attaining foot flat, then SS_{time} would be increased.

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

2.4 Statistical analysis

Data were analysed using a random-effects regression model with maximum likelihood estimator, using the Stats version 13.0 statistical program (StataCorp, College Station, TX, USA). Factors of interest were incorporated sequentially and their statistical significance was determined. Level of significance was set at $p < 0.05$, and factors and interaction between factors, reaching this level were retained in the final model. The following factors and interaction between these factors, were tested: 1) Walking speed: two levels, slow and customary; 2) Microprocessor: two levels, inactive (*MC*off), active (*MC*on).

In addition, in order to understand how the mechanical work done by the terminating (prosthetic) and trailing (intact) limbs compares to that in able-body individuals, group ensemble-average mechanical power profiles(±SD band) for each limb were plotted alongside ensemble average mechanical power profiles(±SD band) for a group of able-body individuals (customary speed trials only, Figure 1). The data for able-body individuals are from our earlier study (Abdulhasan and Buckley, unpublished). A comparison of the current data to these previous data allowed us to subjectively evaluate how TFA produce the negaitive mechanical limb work to terminate gait on a ramp in comparison to how able-body individuals do.

3. Results

The group mean mechanical limb power profiles (W/kg) of the terminating/prosthetic- and trailing/intact- limbs for ramp-descent terminations from customary walking speed for the *MC*on and *MC*off conditions are presented in Figure 1. Group mean negative mechanical limb work ($W_{(-ve)}$) done by the prosthetic and intact limbs to terminate ramp-descent from slow and customary walking speeds for the *MC*on and *MC*off conditions is presented in figure 2.

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

3.1 Prosthetic limb

Negative mechanical limb work done was significantly affected by speed (*P* <0.001) and MC condition (*P* =0.004), but there was no interaction between terms (*P* =0.582*).* Negative work done was greater when terminating ramp-descent from walking at customary compared to slow speed, and was greater for the *MC*on compared to *MC*off condition (Figure 2). On average 14% and 16% more negative work was done by the prosthetic limb for *MC*on compared to *MC*off condition at slow and customary speeds respectively; with 6 out of 8 and 7 out of 8 participants for slow and customary speeds respectively showing an increase in negative work when the microprocessor was active.

It is worthy of note that the mechanical limb power profile of the prosthetic/terminating limb was within the normal range (SD band) of that for the able-bodied group (Figure 1a).

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

3.2 Intact limb

Negative mechanical work done was significantly affected by speed ($P<0.001$) but was unaffected by the MC condition ($P=0.35$) and there was no interaction between terms ($P=0.317$). Negative work done was greater when terminating ramp-descent from walking at customary compared to slow speed (Figure 2).

It is noteworthy that the mechanical limb power profile of the intact/trailing limb, highlights the negative power during late stance was close to the ‘upper edge’ (higher negative value) of the SD band of the able-bodied group (Figure 1b). Furthermore, except for a brief period, trailing limb power was negative for almost the entire braking phase: with the positive period being noticeably smaller (shorter, smaller magnitude) and outside the SD band of the able-bodied group.

3.3 Comparison to able body individuals

We have previously shown when able-bodied individuals terminate ramp-descent (from a customary speed of 1.14 (0.16 m/s), the negative mechanical limb work done by the terminating limb contributes around 26% of the total negative work done by both limbs (Abdulhasan and Buckley, unpublished). In comparison, for the amputee participants in the present study the terminating (prosthetic) limb's contribution to the total negative work done by both limbs when terminating ramp-descent from customary speed, was 17.2% and 19.7% for when the microprocessor was inactive and active respectively.

4. Discussion

The present study determined how the use of an above-the-knee limb system prosthesis, which has simultaneous MC control of the hydraulic resistances at the knee and ankle, affected the negative mechanical work done by each limb when TFA terminated gait on a declined surface. The study focused on ramp-descent terminations where the terminating limb was the prosthetic limb. The key finding was that the negative mechanical work done by the prosthetic limb was significantly increased when the limb system's microprocessor was active (ramp descent mode). The increase in mechanical work done occurred despite there being no difference between *MC* conditions in freely chosen walking speed.

There is anecdotal evidence that when using traditional type prosthetic feet TFA have a tendency to 'lurch' forwards/downwards over the foot when descending ramps. In the present study, when the limb system was active (*MC*on) WT_{time} was significantly shorter, which suggests foot flat was attained easier/quicker when the microprocessor was active. Although there was no significant difference across MC conditions in the subsequent single-support period, the fact that braking time (equivalent to WT_{time} plus SS_{time} combined) did not differ across MC conditions, suggests that single support must have been longer when the MC was active even though it wasn't significantly so (given that WT_{time} was shorter when the MC was active): and thus there was no tendency to 'lurch' forwards over the foot. These findings highlight that when the microprocessor was active the hydraulic resistance at the ankle was altered in the manner congruent with the device's 'ramp descent' mode: plantarflexion resistance reduced at initial contact to facilitate attaining foot flat; dorsiflexion resistance then increased to control/slow forward progression of shank pylon over foot. The accompanying increased negative work done when the limb system was active suggests participants became more assured in using their prosthetic limb to arrest CoM velocity.

One might expect that because there was greater negative mechanical limb work done on the prosthetic side when the *MC* was active, then the intact limb's contribution would reduce by a comparable amount: however, we found no evidence of this. This null finding maybe related to participants' lack of familiarisation with the limb system. Note, trials for both the active (*MC*on) and inactive (*MC*off) conditions were collected within the same data collection session and no indication was given regarding whether the limb system's microprocessor was active or not. Without familiarisation or knowledge that the MC would be active, participants may have continued to use their 'learnt' intact-limb adaptations. It is also worth noting that the prosthetic limb's contribution to the total negative limb work done by both limbs to terminate ramp-descent was much smaller than the intact limb's contribution irrespective of MC condition. However, such an imbalance in work done by the leading and trailing limbs is comparable to how able-bodied individuals terminate ramp-descent (Abdulhasan and Buckley, unpublished).

The external mechanical limb work done during locomotion represents the work done on the CoM to transfer it from one limb to the other whilst also progressing it forwards as well as sometimes moving it upwards or downwards (i.e. when negotiating stairs or ramps) (Donelan et al., 2002). Following ground contact during ongoing gait, the limb dissipates energy and during the same double-support period the contralateral limb generates mechanical power to restore and redirect the CoM upward and forward (Donelan et al., 2002). This means that for successive double support periods each limb alternates between energy dissipation (negative work) and generation (positive work). When terminating gait, the CoM velocity needs to be arrested, therefore the mechanical work done by the terminating limb will predominantly involve energy dissipation. We chose to analysis limb negative mechanical work as our main outcome variable because in essence it summarises the contribution each limb makes in arresting CoM velocity. The mechanical limb work assessed was that performed by the external forces, and thus it was determined as the product of the GRF and CoM velocity vectors. As such the approach avoided having to make any assumptions about joint centre locations; as would be required if internal work estimates were determined. This was important because the heel and forefoot keels of the ‘dynamic response’ prosthetic foot within the limb system would deflect about non-defined axes (simulated plantar/dorsi-flexion) and at different locations to the ankle device’s articulation axis. It has previously been highlighted that due to the issues regarding where to define an ‘ankle’ in a prosthetic foot, the assessment and interpretation of ‘ankle’ kinetics can be problematic (Geil et al., 2000).

In our previous study we showed that when able-bodied individuals terminate gait on a level or declined surface, negative ankle joint (muscle) work is the foremost contributor to the negative mechanical limb work done(Abdulhasan and Buckley, unpublished). In addition, when terminating gait on a declined surface, greater negative mechanical limb work is done in comparison to terminations on a level surface, with increased negative knee joint work in early stance being the key contributor to the increased mechanical limb work (Abdulhasan and Buckley, unpublished material). Other studies have also found that the knee joint is the primary joint contributing to the gait adjustments needed to walk on ramps (Kuster 1995, Redfern and DiPasquale 1997, Lay 2006, Franz 2012). Although above-the-knee MC prostheses (including a limb system prosthesis) typically provide stance-yielding knee flexion to facilitate walking down ramps, such flexion is designed to occur in late stance. In the present study, the focus was on terminating gait on the prosthesis, and thus there was no ‘late stance’ period on the prosthetic limb and hence there was no stance-yielding knee flexion initiated during the period assessed. Indeed, the knee remained fully extended throughout irrespective of MC condition. This highlights that the limb system’s ability to alter the resistance at its ankle mechanism must have been the key factor in why the prosthetic limb was able to do more negative mechanical limb work when the limb system’s microprocessor was active.

Potential limitations of this study include the following. The study included only active adult TFA (K3 activity level) and the age range was relatively large. At the time of data collection there were only 9 TFA in the UK using a limb system prosthesis. Thus, although the results presented are applicable to the population group at the time, we cannot say they are generalizable to all TFA. Future work is required to determine whether TFA categorised at K2 activity level are able to gain functional benefits from using a limb system prosthesis similar to those highlighted in the present study. When using a limb system prosthesis for everyday locomotion, the ‘ramp descent’ mode is automatically initiated after the initial two walking steps along a declined surface. However, in the present study, we switched the ‘ramp descent’ mode on (and off) via Bluetooth connection. Accordingly, this enabled us to be confident that the device was in the correct mode (*MC*on, *MC*off). Thus, a limitation of the study is that it did not assess when and if the ‘ramp descent’ mode becomes initiated when walking down a ramp. We chose to only investigate ramp-descent terminations in which the prosthetic limb was the terminating limb. We thought that including trials in which ramp-descent was also terminated on the intact limb would increase the likelihood of participants becoming fatigued. Future work is thus required to determine how use of a limb system prosthesis affects termination of ramp-descent when the terminating limb is the intact limb. Additionally, the stopping task investigated in this study was planned/expected ramp-descent terminations. Future work could assess how a limb system prosthesis facilities completing abrupt/unexpected ramp-descent terminations. The ramp used in the present study had an angle of 5 deg. We chose this angle as it is the recommended maximum angle for wheel-chair access ways. (BS 8300: 2009). The negative limb work done to terminate ramp-descent would increase, on both limbs, on a 10 deg ramp (Redfern 1997, Lay 2006, Cham and Redfern, 2002, Franz et al. 2012). However, future work is required to determine whether a limb system prosthesis can provide as much benefit on a 10-degree ramp as it does of a 5-degree ramp. Finally, the limb system has several other modes (e.g. ‘stair descent’, ‘stop and lock’), and thus future research is required to determine whether it enhances the execution of other adaptive gait tasks.

5. Conclusions

In conclusion, when TFA terminated gait on a 5-degree declined walkway using a limb system prosthesis which has simultaneous microprocessor control over the hydraulic resistances at the ankle and knee, significantly more negative mechanical limb work was done when the device's microprocessor was active ('ramp descent' mode) compared to inactive (hydraulic resistances at constant default levels). This indicates that when the microprocessor was active it effectively altered the hydraulic resistances at the ankle and knee, and as a result there was greater involvement of the prosthetic limb. These findings suggest that use of a limb system prosthesis will improve the way TFA descend and stop on ramps, and more generally that use of such a prosthesis should provide TFA clinically meaningful benefits to their everyday walking where adaptations to an endlessly changing environment are required. The goal of current prosthetic engineering technology is to develop artificial limbs that mimic the function of a physiologic limb and hence increase prosthetic limb usage which in turn should potentially reduce intact limb compensatory effort/loading. The current study's findings suggest that a limb system prosthesis represents another step towards such technological advancement.

Conflicts of interest: None

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Figure 1. TFA group ensemble mean limb mechanical power (W/kg) during ramp-descent terminations from customary walking speed for a) the terminating/prosthetic limb and b) the trailing/intact limb. Black bold line= MCon; dashed line= MCoff. For comparison, Grey band = group ± SD-band for able-bodied individuals (Data from (Abdulhasan and Buckley, unpublished). Data are plotted for the braking phase of each limb.

Figure 2. TFA group mean (+SD) total negative mechanical limb work done (J/kg) during termination of ramp-descent from customary and slow walking speeds for the terminating/prosthetic and the trailing/intact limbs. Data are shown for when the limb system's microprocessor was active (MCon; solid bars) or inactive (MCoff; hashed

bars). Able body data (checkered bars), from previous report (Abdulhasan and Buckley, unpublished) are shown for comparison. * indicates significantly different to MCoff condition (p<0.01).

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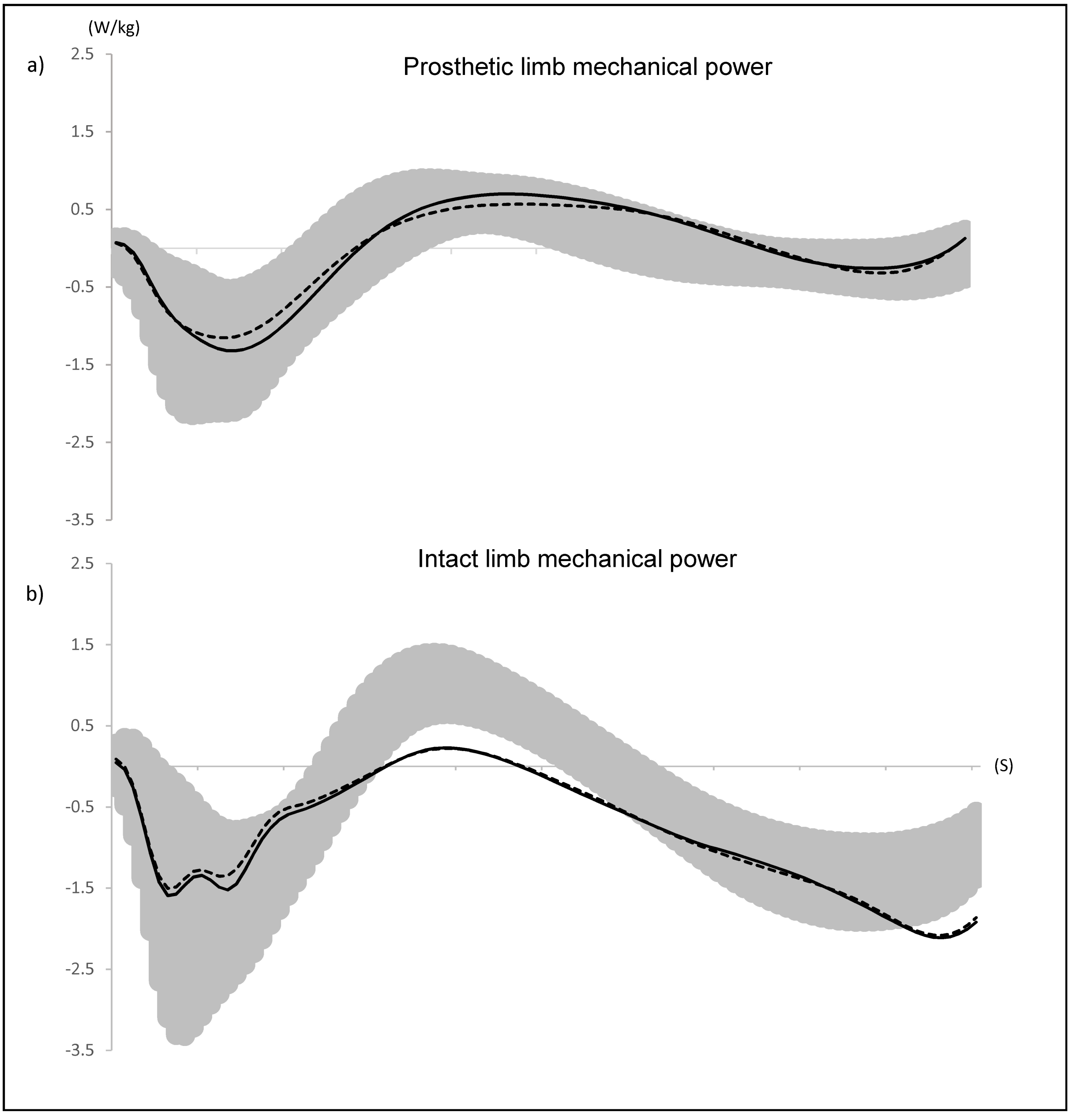
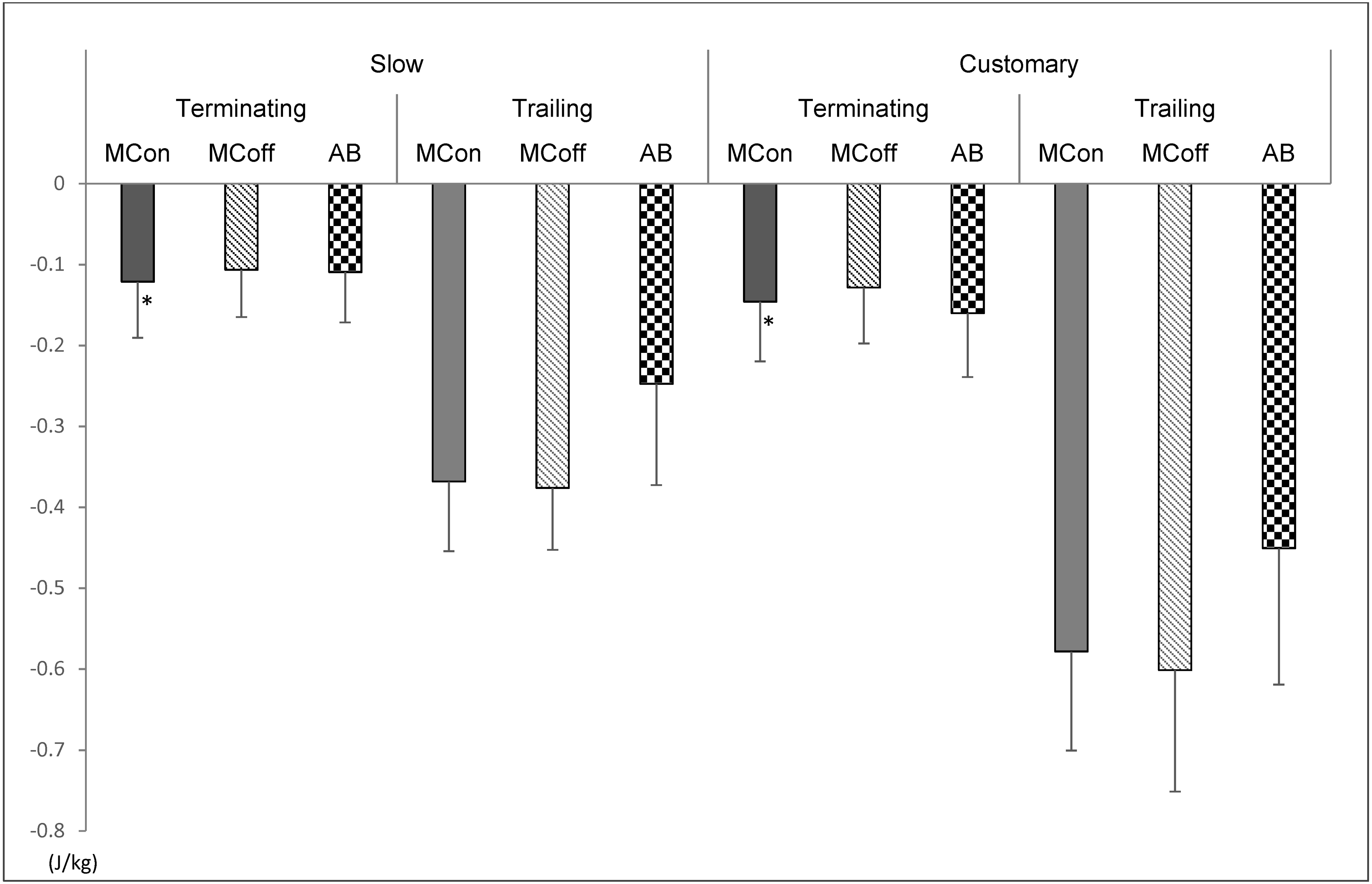


(W/kg)
a)
Prosthetic limb mechanical power
2.5
1.5
0.5
-0.5
-1.5
-2.5
-3.5
b)
Intact limb mechanical power
2.5
1.5
0.5
-0.5
-1.5
-2.5
-3.5
(S)



Slow
Customary
Terminating
Trailing
Terminating
Trailing
MCon
MCoff
AB
MCon
MCoff
AB
MCon
MCoff
AB
MCon
MCoff
AB
0
-0.1
-0.2
-0.3
-0.4
-0.5
-0.6
-0.7
-0.8
(J/kg)
*
*

Table .1 Demographic characteristics of study participants

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