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Toe clearance when walking in people with unilateral transtibial amputation: Effects of passive hydraulic ankle

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Abstract—Most clinically available prosthetic feet have a rigid attachment or incorporate an "ankle" device allowing elastic articulation during stance, with the foot returning to a "neutral" position at toe-off. We investigated whether using a foot with a hydraulically controlled articulating ankle that allows the foot to be relatively dorsiflexed at toe-off and throughout swing would increase minimum toe clearance (MTC). Twenty-one people with unilateral transtibial amputation completed overground walking trials using their habitual prosthetic foot with rigid or elastic articulating attachment and a foot with a hydraulic ankle attachment (hyA-F). MTC and other kinematic variables were assessed across multiple trials. When using the hyA-F, mean MTC increased on both limbs (p = 0.03). On the prosthetic limb this was partly due to the device being in its fully dorsiflexed position at toe-off, which reduced the "toes down" foot angle throughout swing (p = 0.01). Walking speed also increased when using the hyA-F (p = 0.001) and was associated with greater swing-limb hip flexion on the prosthetic side (p = 0.04), which may have contributed to the increase in mean MTC. Variability in MTC increased on the prosthetic side when using the hyA-F (p = 0.03), but this did not increase risk of tripping.

Key words: amputation, dorsiflexion, gait, hydraulic ankle, prosthesis, toe clearance, transtibial, tripping, unilateral, walking speed.

INTRODUCTION

In order for people with unilateral transtibial amputation to ambulate safely, they must make substantial motor-control adaptations on both the prosthetic and intact sides in order to compensate for the absent limb [1–2]. Current prosthetic feet typically incorporate flexible heel and forefoot keels that are capable of storing energy during early- and mid-stance and returning a portion of the stored energy to aid forward progression and push-off (so-called energy storage and return feet). Although they represent a notable advancement in technological design, these prosthetic devices only partially compensate for the missing foot and ankle. For example, it has been shown that because of the loss of the ankle joint and associated musculature (resulting in decreased push-off from the prosthetic limb and a shorter intact

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Abbreviations: COM = center of mass, EPSRC = Engineering and Physical Sciences Research Council, habF = habitual prosthetic foot, hyA-F = foot with a hydraulic ankle attachment, MTC = minimum toe clearance, SD = standard deviation.

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limb step length), compensatory work is performed by the intact-limb hip in early stance [2–3].

Given the mechanical constraints imposed by a prosthesis, it is unsurprising that people with amputation tend to be at increased risk of falling compared with age-matched nondisabled individuals [4–5]. Minimum foot clearance/minimum toe clearance (MTC) is defined as the vertical distance between the foot (usually the toe region) and the walking surface at the lowest point of swing. It has been suggested that MTC occurs at the same time as peak forward velocity of the foot [6–8], and we recently determined that this is indeed the case for a group of individuals with transtibial amputation [6]. MTC and variability in MTC are thus critical aspects of gait [9] since failure to consistently achieve adequate foot-ground clearance will increase the risk of tripping and falling. People with amputation have been shown to have a reduced MTC on the prosthetic limb compared with the intact limb when walking over level ground [6,10], with interlimb differences increasing when walking on an uneven surface [10]. The reduced MTC on the prosthetic side is likely to be at least partly due to the prosthetic foot's inability to actively dorsiflex during swing, and this may partly explain the greater falls risk in people with amputation.

Most clinically available prosthetic feet have either a rigid, nonarticulating attachment or an "ankle" device that allows elastically controlled articulation, for example, by incorporation of a rubber snubber at the point of attachment. Such elastically controlled devices have an inherent tendency to return to the neutral position once unloaded. Therefore, once the prosthetic foot leaves the ground the ankle angle returns to neutral and remains so throughout swing. This could partly explain why MTC has been shown to be reduced on the prosthetic compared with intact side [10]. Using a hydraulic ankle device that provides dampened stance-phase passive articulation would allow the foot to passively dorsiflex during stance and thus enable the foot to leave the ground in a relatively dorsiflexed position and remain so throughout swing. This would raise the "toes" during swing and thus increase MTC, which, so long as variability in MTC did not increase, would reduce the likelihood of tripping. Recently, a prosthetic foot with a hydraulic ankle attachment (hyA-F) (Echelon, Chas A Blatchford & Sons Ltd; Basingstoke, United Kingdom) has become clinically available. Relative to its neutral (standing) position, the hydraulic ankle provides 6° plantar flexion and 3° dorsiflexion. The purpose of the present study was to determine how mean and variability in prosthetic foot MTC during overground walking were affected when people with unilateral transtibial amputation switched from using their habitual prosthetic foot (habF) with a rigid or elastically controlled articulating attachment to one with a hyA-F. We hypothesized that mean MTC would increase and variability in MTC would be unchanged when participants switched to using the hyA-F.

Based on our own previous work, we expected walking speed would increase when using the hyA-F and the center of mass (COM) would translate more quickly and smoothly over the prosthetic foot [11]. Thus, we hypothesized that there would be changes in stance and swing limb kinematics that might also affect MTC. We thus also determined the differences in sagittal plane joint angles at instant of MTC for both the stance and swing limbs, as well as the relationship (correlation) between all outcome variables and walking speed. The latter analysis enabled us to examine to what extent any change in MTC was due to a change in walking speed rather than a change in foot type. Finally, to determine whether changes in MTC when using the hyA-F were a result of participants' altering their gait in the frontal plane, we also assessed differences in the amount of hip-hiking at instant of prosthetic-limb MTC.

METHODS

Participants

Twenty-one active people with unilateral transtibial amputations (18 male, 3 female, mean ± standard deviation [SD] age 48.2 ± 12.8 yr, mass 87.4 ± 13.2 kg, height 1.78 ± 0.07 m) were recruited. All had undergone amputation because of trauma, infection, or carcinoma at least 2 yr prior to participation (median ± interquartile range, 8.8 ± 14 yr; range 2–60 yr). All were free from neurological, musculoskeletal (other than limb amputation), or cardiovascular disorders and were not taking medication that could interfere with balance or coordination. All ambulated independently and had used their current prosthesis for a minimum of 6 mo. Twelve participants' habitual prosthesis incorporated an Esprit foot (Chas A Blatchford & Sons Ltd).This foot is identical to the hyA-F used in the present study, except that it has a nonarticulating point of attachment (ankle) so that passive (simulated) plantar/dorsiflexion is only available via heel and forefoot keel deformation and recoil. The other participants used a range of feet; six used a Multiflex (Chas A Blatchford & Sons Ltd), one a Flex-foot (Össur; Reykjavík, Iceland), one an Elite (Chas A Blatchford & Sons Ltd) and one a Seattle Litefoot (Trulife; Poulsbo, Washington). These feet all had a nonarticulating attachment except the Multiflex, which incorporated an elastically controlled articulating attachment (i.e., rubber snubber).

Walking Protocol and Prosthetic Intervention

Data collection took place within a motion analysis laboratory. Participants were told the study's purpose was to investigate how the hydraulic ankle prosthesis affected walking but were given no further details. Wearing flat-soled shoes, shorts, and T-shirt, participants were instructed to walk straight across the laboratory (approximately 8 m) at a speed they perceived to be comfortable (customary). Segmental kinematic data were recorded (100 Hz) using a multicamera system (Vicon MX, Vicon; Oxford, United Kingdom) using the set-up and approach we have previously described [11]. Participants completed 20 walking trials (in two blocks), 10 using their habF and 10 using the hyA-F. The order in which the two blocks were completed was counterbalanced across participants.

Prosthetic intervention involved exchanging the existing foot (habF) of each participant's prosthesis with a hydraulic foot (hyA-F). Everything else about the prosthesis (length, alignment, socket, etc.) was kept constant. Further details regarding how the intervention was undertaken can be found in our previous report [11].

Once the hyA-F was fitted, participants underwent familiarization for at least 45 min. This included walking over slopes, stairs, and a variety of surfaces (e.g., grass verges, vinyl floors). During this familiarization period, the hyA-F's hydraulic settings (separate setting for dorsi- and plantar flexion) were altered until it was deemed by an experienced prosthetist that the foot was functioning optimally at the participant's customary walking speed. The mean ± SD hydraulic settings for the group were 7.30 ± 0.55 and 6.53 ± 0.67 (equating to damping coefficients of 3.28 ± 0.13 and 3.08 ± 0.18 N ms/°, respectively) for plantar and dorsiflexion, respectively. All participants confirmed that they felt accustomed to the hyA-F before data collection commenced. Participants who used the hyA-F in the first block undertook a similar period of (re)familiarization when their habF was refitted into their prosthesis. Limb length was measured as the height of the greater trochanter above the floor during quiet standing. Group mean ± SD prosthetic limb length was 0.886 ± 0.046 and 0.888 ± 0.044 m for the habF and hyA-F, respectively, and the intact limb length was 0.892 ± 0.038 m; differences in limb length between the intact and prosthetic sides ($p = 0.32$) and between the habF and hyA-F ($p = 0.23$) were nonsignificant.

Data Analysis

Initial processing was undertaken using Workstation software (Vicon). Data files (C3D) were then exported into Visual 3D software (version 4, C-Motion; Germantown, Maryland). For each participant, a kinematic nine-segment model [12] was constructed. All intact lower-limb joints were created as functional joints [13], and on the prosthetic limb a virtual ankle was defined at the midpoint of the prosthetic shank at the same level (height from the ground) as the intact ankle. Virtual landmarks were also created at the anteroinferior end point of each shoe (virtual shoe tip). This was done by determining the vector displacement between the second toe marker and the inferior distal tip of the shoe during a static calibration trial [14]. Data were filtered with a zero-lag fourth-order Butterworth filter cut-off of 6 Hz. The following variables were analyzed for the intact and prosthetic limbs:

- MTC: vertical distance between virtual shoe tip and ground at instant of peak forward foot velocity during swing phase [7–8].
- Joint angles at MTC: stance and swing limb sagittal plane angles at the hip (thigh relative to pelvis), knee (shank relative to thigh), and foot (foot relative to the floor). All angles were relativized to those determined during a standing calibration file.

We also determined the following for the prosthetic side only:

- Prosthetic limb hip-hiking: determined as height of prosthetic (swing) limb hip relative to intact (stance) limb hip at instant of prosthetic limb MTC.

Peak forward foot velocity was used as a temporal marker to determine MTC. We have recently demonstrated that, although there is a slight temporal offset between peak forward foot velocity and MTC (MTC occurs around 0.014 s earlier), there is no significant difference in the toe clearance values at each point [6]. We used this approach as our automatic event detection algorithm because of its ease of use and because it ensures the local minima in toe-ground clearance that occurs at or just after toe-off would not be identified in error as the point of MTC.

We measured foot angle rather than ankle angle because of the problems of assuming a prosthetic foot to be a rigid segment with definable ankle joint axes [15–16]. Even if we had made the necessary assumptions, it would still have been difficult to compare ankle angles between the hyA-F and habF because of problems determining the neutral (zero) angle of the hyA-F, in that during stationary standing there was ongoing (nonstationary) articulation at the hydraulic ankle device because of anteroposterior postural sway. Thus, instead we assessed foot segment angle relative to the laboratory floor and used the foot segment angle during standing as the zero reference. A negative foot angle indicated the foot was in a “toes down” angle relative to that determined during standing.

For each participant, the mean and variability (SD) across the 10 repeat walking trials were determined for each of the listed variables. To avoid acceleration and deceleration periods, only data from the middle part of each trial (typically 3 to 5 steps) were included in the analysis. Hence, the mean and variability for each participant were determined across approximately 40 steps. The mean of all participants’ SD was used to evaluate group variability. Although the main focus of our analysis was to determine how use of a hyA-F affected prosthetic limb MTC, we also compared variables for the prosthetic limb with those for the intact limb. This secondary analysis provided more general insights into how walking with a unilateral prosthesis affects MTC.

Statistical Analysis

Except for hip-hiking, variables were analyzed using repeated measures analysis of variance with limb (×2; intact, prosthetic) and prosthetic foot-type (×2; hyA-F, habF) as repeated factors. Post hoc analyses were undertaken using Sidak correction for multiple comparisons. Hip-hiking was analyzed using a paired *t*-test (1-tailed). Alpha level for significance was $p < 0.05$. All analyses were undertaken with SPSS statistical software (version 19, SPSS Inc; Chicago, Illinois).

We are aware that MTC has been shown to increase with increasing walking speed [6,17], and based on our own previous work we expected walking speed would be increased when using the hyA-F [11]. However, we chose not to control walking speed because of the methodological difficulties of such an approach and its possible lack of generalizability to the real world [18]. Thus, to determine to what extent any change in MTC was due to a change in walking speed rather than a change in foot type, we examined, using regression analyses, the relationship between each outcome variable and walking speed. These regression analyses were conducted using the combined data across both foot types. If we found a significant relationship between an outcome variable and trial walking speed, then this would suggest that any differences between foot conditions found for this variable was at least partly related to the higher walking speed when using the hyA-F and thus not solely because of the function of the hyA-F. Alternatively, if we found no relationship between an outcome variable and trial walking speed, then this would highlight that any difference found when using the hyA-F would be related to the function of the device.

RESULTS

Minimum Toe Clearance

Mean MTC was significantly affected by foot type ($p = 0.03$) and by limb ($p = 0.04$), while variability in MTC was significantly affected by foot type ($p = 0.03$), and there was a significant foot type by limb interaction ($p = 0.04$). Mean MTC increased for both limbs when using the hyA-F compared with habF (2.17 vs 1.90 cm) and was also greater on the intact compared with prosthetic limb (2.20 vs 1.91 cm). Variability in MTC increased when using the hyA-F compared with the habF (0.62 vs 0.54 cm), and the significant foot type by limb interaction indicated that variability increased on the prosthetic but not the intact limb. MTC results are presented in **Table 1** and the **Figure**.

Joint Angles at Minimum Toe Clearance

Swing Limb

Mean foot angle was significantly affected by foot type ($p = 0.01$) and by limb ($p = 0.02$), and there was a significant foot type by limb interaction ($p = 0.049$). The foot angle was reduced (indicating a slightly less toes-down foot angle) on the prosthetic compared with intact side (–17.7° vs –20.8°), and this angle was reduced for both limbs when using the hyA-F compared with habF; the interaction indicated that the reduction in foot angle was only significant on the prosthetic side (reduction: intact limb 1.0°; prosthetic limb 4.8°). Variability in foot

Table 1.
Group mean ± standard deviation and within-participant variability in minimum toe clearance (MTC) for intact and prosthetic limbs and freely chosen walking speed when using habitual prosthetic foot (habF) and hydraulic ankle attachment (hyA-F).

Measure	habF		hyA-F	
	Intact	Prosthetic	Intact	Prosthetic
MTC Mean (cm)*†	2.12 ± 0.91	1.76 ± 0.85	2.27 ± 0.63	2.07 ± 0.63
MTC Variability (cm)*‡	0.58 ± 0.25	0.50 ± 0.20	0.58 ± 0.17	0.65 ± 0.24
Walking Speed (m/s)*	1.12 ± 0.14		1.17 ± 0.14	
Walking Speed Variability (m/s)*	0.01 ± 0.01		0.03 ± 0.01	

*Significant main effect for foot-type.
†Significant main effect for limb.
‡Significant interaction.

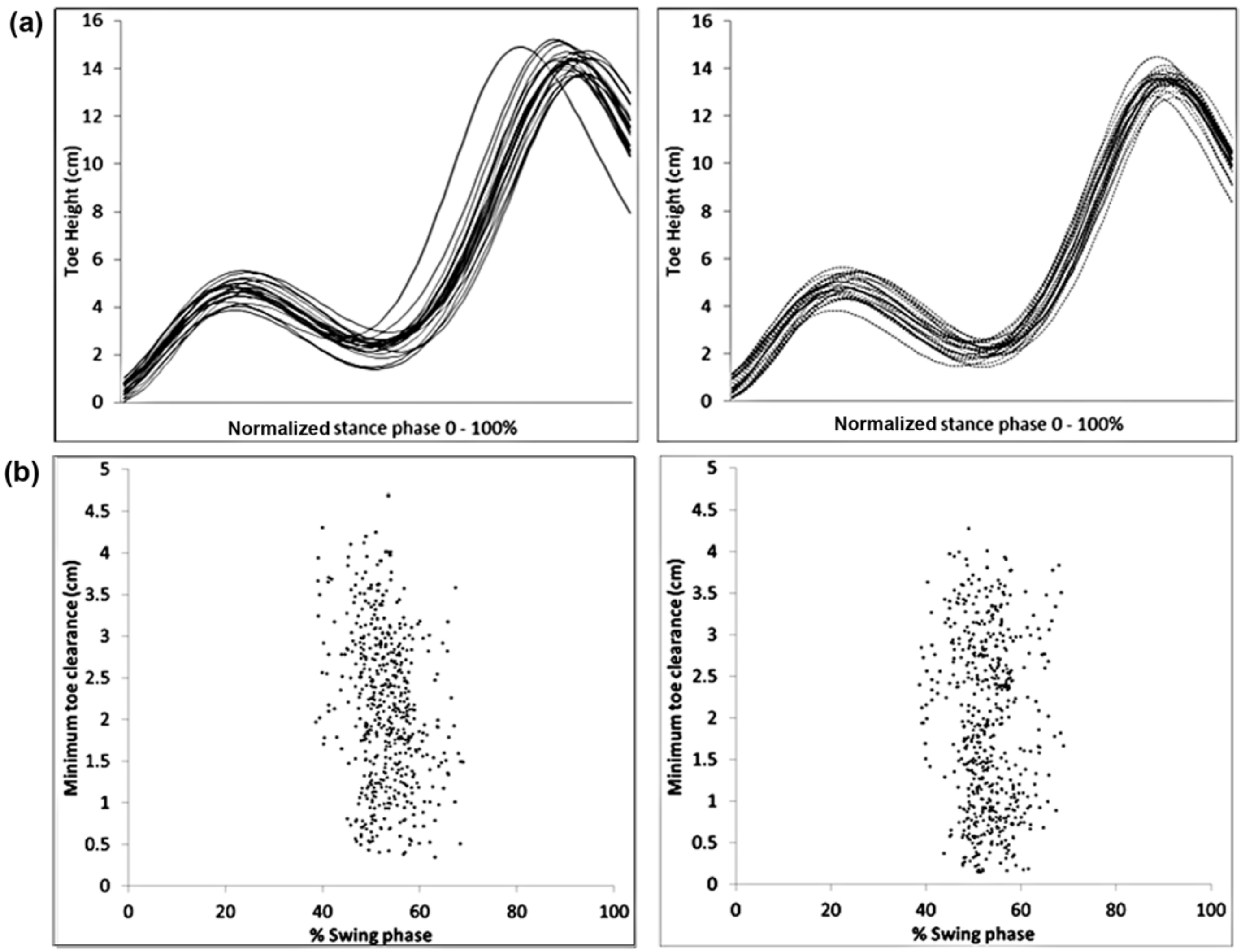

Figure.
(a) Swing-limb vertical displacement toe trajectory profiles for all trials for one participant, and **(b)** minimum foot clearance (MFC) values for all trials of all participants (values plotted at instant in swing when MFC event occurred). Data shown are for prosthetic limb only. Panels on left indicate data for foot with hydraulic ankle attachment and panels on right indicate data for habitual prosthetic foot.

angle was significantly reduced on the prosthetic compared with intact limb (5.4° vs 8.4°; $p = 0.01$) but was not significantly different when using the hyA-F compared with habF (5.6° vs 5.1°; $p = 0.15$). On both limbs, the hip was significantly more flexed when using the hyA-F compared with habF (25.8° vs 22.8°, $p = 0.001$) but was unaffected by limb, and there was no interaction between terms (**Table 2**). There were no differences in mean knee angle across foot types or limbs ($p > 0.13$), but variability in knee angle was significantly affected by limb ($p = 0.001$) and there was a significant foot-type by limb interaction ($p = 0.03$). Variability in knee angle was reduced on the prosthetic compared with intact limb (4.2° vs 6.5°), and the interaction indicated that variability decreased on the prosthetic limb but increased on the intact limb when using the hyA-F compared with the habF.

Stance Limb

Mean hip ($p = 0.01$) and knee ($p < 0.001$) angles at contralateral MTC were significantly affected by limb but were unaffected by foot type (hip $p = 0.10$; knee $p = 0.09$), and there were no interactions between terms (hip $p = 0.32$; knee $p = 0.05$). The hip was more extended/less flexed (–0.5° vs +3.3°) and the knee was less flexed (7.2° vs 13.6°) during stance on the prosthetic compared with intact side (**Table 2**). There was a trend for the knee to be more flexed at contralateral MTC when using the hyA-F compared with habF (11.0° and 9.7°; $p = 0.09$). Variability in the hip angle of each limb at contralateral MTC was significantly decreased when using the hyA-F compared with habF (2.17° vs 2.83°; $p = 0.04$).

Prosthetic Limb Hip-Hiking

Hip-hiking was unaffected by foot type (hyA-F: 7.1 mm, habF: 5.5 mm; $p = 0.37$), as was hip-hiking variability (hyA-F: 23.2 mm, habF: 20.2 mm; $p = 0.39$).

Effects of Increases in Walking Speed

As expected, mean walking speed was greater when using the hyA-F compared with when using the habF ($p < 0.001$) (**Table 1**), which is why we investigated a priori the relationship between walking speed and each outcome measure.

Irrespective of foot type, there was no significant correlation between walking speed and MTC for either limb ($p > 0.09$). There were significant correlations between walking speed and the following variables. The amount of swing-phase hip flexion at prosthetic limb MTC increased with walking speed ($r = 0.09$, $p = 0.04$), while the amount of swing-phase knee flexion at MTC of each limb ($r = 0.15$, $p < 0.001$), the amount of prosthetic limb stance-phase knee flexion at instant of intact limb MTC ($r = 0.10$, $p = 0.03$), and the amount of hip-hiking at prosthetic limb MTC ($r = 0.22$, $p < 0.001$) decreased with walking speed.

Table 2.
Group mean ± standard deviation and within-participant variability in swing and stance limb joint angles (degrees) and prosthetic limb hip-hiking at prosthetic limb minimum toe clearance when using habitual prosthetic foot (habF) and hydraulic ankle attachment (hyA-F).

Measure	habF		hyA-F	
	Intact	Prosthetic	Intact	Prosthetic
Swing Limb				
Hip Angle*	24.3 ± 7.9	21.4 ± 8.1	26.9 ± 7.7	24.7 ± 8.4
Hip Angle Variability	2.0	2.1	2.2	1.8
Knee Angle	49.7 ± 5.1	45.5 ± 4.7	49.6 ± 5.6	47.4 ± 7.5
Knee Angle Variability†‡	6.1	4.7	6.9	3.7
Foot Angle*†‡	–21.3 ± 2.6	–20.1 ± 5.3	–20.3 ± 7.8	–15.3 ± 3.1
Foot Angle Variability†	7.9 ± 4.0	5.1 ± 2.6	9.0 ± 3.9	5.6 ± 2.2
Hip-Hiking (mm)	—	5.5 ± 20.2	—	7.1 ± 23.3
Stance Limb				
Hip Angle†	–1.7 ± 6.4	–3.9 ± 6.1	0.7 ± 5.9	–2.7 ± 5.8
Hip Angle Variability*	2.5 ± 1.0	3.2 ± 3.0	2.0 ± 1.1	2.3 ± 0.7
Knee Angle†	12.8 ± 5.3	6.6 ± 7.3	14.3 ± 4.8	7.7 ± 5.1
Knee Angle Variability	2.4 ± 2.1	1.8 ± 0.8	1.9 ± 0.9	1.8 ± 1.1

Note: Negative foot angle indicates “toe down.” Positive hip and knee angle represents flexion; negative hip and knee angle represents extension. Positive hip-hiking indicates height of prosthetic/swing limb is higher than intact/stance limb.
*Significant main effect for foot-type.
†Significant main effect for limb.
‡Significant interaction.

DISCUSSION

Mean MTC significantly increased for both the prosthetic and intact limbs when using the hyA-F compared with habF. This suggests that people with amputation might have a reduced risk of tripping when walking with a hyA-F. However, within-participant variability in MTC, which has been reported to be a risk factor for falling [9], also increased on the prosthetic side when using the hyA-F compared with habF (**Figure (a)**; vertical toe clearance trajectory profiles for a typical participant). Nonetheless, the potential increased risk of tripping associated with this increase in variability was off-set by the increased mean value. **Figure (b)** illustrates that when using the hyA-F compared with the habF, the spread in MTC values across all participants and all trials shifted in an upward direction, indicating that the increased variability in MTC when using the hyA-F occurred with a greater mean toe clearance value. **Figure (b)** also illustrates that there were a number of MTC values less than 2 mm when using the habF, while the lowest MTC values when using the hyA-F were around 4 mm. Thus, although MTC variability was found to be increased when using the hyA-F, such an increase did not cause an increase in risk of tripping. The factors found to contribute to the changes in MTC are discussed here.

Mean swing-limb hip flexion and foot angle at MTC were found to significantly change on both the intact and prosthetic limbs when using the hyA-F, which suggests changes in one or both of these variables were contributing to the changes in MTC. Importantly, post hoc analysis indicated that the reduction in foot angle at MTC (signifying a reduced toes-down foot angle) when using the hyA-F was only significant on the prosthetic side. Hence, the increased MTC observed on the prosthetic limb must at least have been partly driven by the hydraulic device being in its fully dorsiflexed position at toe-off, which reduced the toes down foot angle throughout swing. Hip flexion at MTC was increased on both limbs when using the hyA-F. Thus, since foot angle at MTC did not significantly alter on the intact limb when using the hyA-F, the higher MTC values for the intact limb must have been driven by the increased swing-limb hip flexion.

The use of a hyA-F resulted in increased walking speed, which in turn was associated with an increase in swing limb hip flexion on the prosthetic side. This suggests that an increase in swing-limb hip flexion contributed to the increase in prosthetic limb MTC. However, a reduction in swing-limb knee flexion on the prosthetic limb was also found to be associated with increased walking speed, and this would have likely countered any increase in MTC as a result of the speed-related increase in hip flexion.

Irrespective of foot type, MTC was consistently greater on the intact compared with prosthetic limb. One might expect that because of reduced sensorimotor control of the prosthetic shank and foot, and the inability to dorsiflex the foot (to lift the "toes"), people with amputation would compensate and lift their prosthetic limb higher during swing through increasing hip and/or knee flexion. However, hip or knee flexion did not significantly differ between limbs, and in addition, there was no evidence of prosthetic limb hip-hiking. These findings indicate that participants did not make any compensatory effort to increase prosthetic-limb height during swing. Furthermore, findings indicate that the mean MTC value for the intact limb (2.20 cm) is greater than that previously reported for young nondisabled adults (1.9 cm [19], 1.56 cm [20], 1.85 cm [14], 1.49 cm [21], and 1.29 cm [8]), while the mean MTC value for the prosthetic limb (1.91 cm) is closer to these previously reported values. Although some of these differences may be due to the different methodologies used to determine MTC, collectively they indicate that the intact limb had increased MTC (compared with values reported in the literature) rather than the prosthetic limb having reduced MTC. This between-limb difference in MTC is consistent with previous reports indicating that people with unilateral transtibial amputations have greater MTC on the intact compared with prosthetic limb [6,10,22]. The greater MTC on the intact side must have partly resulted from the significantly decreased stance limb knee flexion on the prosthetic side (**Table 2**), which would have raised the COM and swing (intact) limb. The reduced stance limb knee flexion observed on the prosthetic side likely reflects a strategy to reduce the knee flexor moment when supporting body weight during single-limb stance [23–26]. It has been reported previously that a change of 3.3° stance limb knee angle would alter MTC by ±0.45 cm [8]. This suggests that in people with transtibial amputation, stance phase prosthetic limb knee flexion is an important driver of contralateral limb MTC.

Why variability in MTC was increased on the prosthetic limb when using the hyA-F compared with habF is unclear. One possible explanation for this is the short familiarization period participants were given in which to adjust to the change in ankle articulation afforded by the hydraulic device and/or the device's increased mass (the hydraulic device weighs 0.45 kg more than a typical prosthesis). Note the increased intertrial variability in toe trajectory profiles when using the hyA-F compared with habF in **Figure (a)**. Future work should assess fully adapted subjects. Variability in walking speed was the only other parameter whose variability also increased when using the hyA-F, but this increase in walking speed variability does not explain why MTC variability was only increased for the prosthetic limb.

CONCLUSIONS

In summary, when people with transtibial amputation switched from using their habitual ankle-foot device to one with hydraulically controlled articulation, MTC during overground walking increased on both the prosthetic and intact limbs. Variability in MTC also increased, but only on the prosthetic side. On the prosthetic limb, the mean MTC increase was partly driven by a less toes-down foot angle during swing, which occurred because of the hydraulic device enabling the foot to leave the ground and remain in a more dorsiflexed position throughout swing. Although variability in MTC also increased on the prosthetic limb when using a hyA-F, any associated increased risk of tripping was off-set by a relatively larger concurrent increase in mean MTC. Irrespective of foot type, MTC was greater on the intact limb. This resulted from reduced stance phase knee flexion on the prosthetic side, which raised the swinging limb.

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