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Kinematic comparison of walking on uneven ground using powered and unpowered prostheses ☆

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ABSTRACT

Background: Recent research has focused on the design of intelligent prosthetic ankle devices with the goal of adapting behavior of the device to accommodate all walking surfaces that an individual encounters in daily life. To date, no studies have looked at how such devices perform on uneven terrain.
Methods: 11 young adults with unilateral transtibial amputation participated in two data collection sessions spaced approximately 3 weeks apart. In each session they walked across a loose rock surface at three controlled speeds. In the first session, they wore a passive, energy storage and return prosthesis and in the second, they wore a powered prosthesis (BiOM, iWalk, Bedford, MA, USA).
Findings: Subjects had a 10% faster self-selected walking speed when wearing the powered (1.16 m/s) compared to unpowered prosthesis (1.05 m/s; $p = 0.031$). They walked with increased ankle plantarflexion on their prosthetic limb throughout the gait cycle when wearing the powered compared to unpowered prosthesis. This was especially evident in the increased plantarflexion during push-off ($p < 0.001$). There was a small (<3°), but statistically significant decrease in knee flexion during early stance when wearing the powered device ($p = 0.045$). Otherwise, the kinematics of the knee and hip were nearly identical when wearing the different devices. Subjects had decreased medial–lateral motion of their center of mass when wearing the powered prosthesis ($p = 0.020$), but there were no differences in medial–lateral margins of stability between the devices ($p = 0.662$).
Interpretation: Subjects did not significantly alter their proximal joint kinematics on this irregular surface as a result of the addition of power.

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1. Introduction

Individuals with transtibial amputations (TTA) commonly report difficulty walking on uneven surfaces. This may be due to a lack of ankle joint mobility, distal muscles, and sensory feedback on their prosthetic side needed to provide information about where the foot is in space and to make subtle adjustments during stance. The incidence of and fear of falling are greater in individuals with lower limb amputations (Kulkarni et al., 1996; Miller et al., 2001) than able bodied individuals. Challenging surfaces like uneven ground, can further increase the frequency of falling in this population (Ülger et al., 2010).

Previously we studied both individuals with TTA in passive energy storage and return (ESR) prostheses (Gates et al., 2012a) and able bodied adults (Gates et al., 2012b) walking on a loose rock surface. Individuals with TTA took shorter and wider steps on uneven ground compared to level ground and shorter steps on their prosthetic limbs. Both groups altered the position of the foot at initial contact to decrease their foot contact angle. Since individuals with TTA were unable to control the ankle angle, they were not able to achieve this flat foot position on their prosthetic side. Both subject groups also increased hip and knee flexion during swing. Subjects with TTA increased hip and knee flexion more on their prosthetic side to clear the toe. This adaptation may have occurred because they could not dorsiflex the prosthetic foot during swing to achieve toe clearance (Moosabhoy and Gard, 2006), or because they were not able to plantarflex the foot during push-off, limiting their ability to initiate swing (Neptune et al., 2001).

Most commercially available prosthetic devices, like the ones tested previously, are passive mechanical devices (Gates et al., 2012b). However, recent research has focused on the design of intelligent prosthetic ankle devices that are capable of adapting behavior to accommodate the various walking surfaces that an individual encounters in daily life (Au et al., 2007; Sup et al., 2011; Williams et al., 2009). While numerous devices have been proposed, there are currently only two, commercially available, ankle prostheses that claim to provide "intelligent terrain adaptation." The Proprio Foot (Össur, Reyjavik, Iceland) provides terrain adaptation by prepositioning the ankle to better accommodate stairs

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and slopes (Alimusaj et al., 2009; Fradet et al., 2010). However, the controller requires several strides to determine the activity type in order to make the proper adjustments (Williams et al., 2009) and functions identically to a passive ESR foot during the stance phase. In contrast, the BiOM (iWalk, Bedford, MA, USA), uses a series-elastic actuator and reflexive control based on a Hill-type muscle with a positive force feedback scheme to provide power during push-off (Herr and Grabowski, 2012). An onboard microcontroller is used to provide real-time adaptation in ankle position and power based on activity type (Eilenberg et al., 2010).

For such devices to be effective, they must function in a way that detects the change in surface and adapts its behavior quickly. The BiOM, in particular, may be beneficial when walking on uneven ground since it allows for ankle plantarflexion which could enable the user to achieve a shallower foot/ground initial contact angle. The supplied power and ankle plantarflexion may also help push the leg into swing such that less hip and knee flexion is required for toe clearance. Unlike a passive device, however, the BiOM provides push-off force that is proportional to the amount that the user loads the foot. On surfaces such as uneven ground, it may be difficult for the user to predict the best way to load the foot as the surface itself is unpredictable. Gait is also far more variable on this surface (Gates et al., 2012a, 2012b). Therefore, the device response may vary to a greater extent than during level ground walking. This could cause the users to adapt a more cautious strategy when utilizing such devices under novel circumstances. A cautious gait pattern is typically characterized by reduced speed, stride length and slight widening of the base of support (Nutt, 2001). To date, no studies have examined the behavior of intelligent prostheses on irregular or unstable surfaces.

The purpose of this study was to determine if a powered device changed gait kinematics in comparison with a passive ESR prosthesis when individuals with TTA walked on a loose rock surface. We proposed two alternative hypotheses: 1) The addition of ankle power and ability to plantarflex the ankle would increase symmetry in lower extremity kinematics and increase self-selected walking velocity when patients walked on this unstable surface. 2) The patients would adopt a cautious strategy when walking on the rock surface using the powered device as they may perceive the device as unpredictable because it may not recognize and adopt the appropriate mode for that surface. Finally, we hypothesized that any affects seen would be amplified by the additional challenge of walking at faster speeds.

2. Methods

2.1. Subjects

11 young adults (10 male, 1 female) with traumatic unilateral transtibial amputation participated (Table 1). Prior to the experiment, all participants were screened to ensure they were free of orthopedic and neurological disorders to the intact side that might have affected their ability to complete the testing protocol. Subjects gave their written informed consent prior to participation in this institutionally approved study.

2.2. Experimental protocol

Subjects came to the lab for two separate data collections. During the first session, they wore their clinically prescribed energy storage and return (ESR) prosthesis (Table 1). After this session, they were given a powered device, the BiOM (iWalk, Bedford, MA, USA). They were given a brief instruction on how to use the BiOM and given the device for at least three weeks to acclimate. They then returned to the lab for a second, identical, data collection session while wearing the BiOM.

During each session, subjects walked over a loose rock surface, previously described in (Gates et al., 2012a, 2012b). The rock surface was a 4.2-m long by 1.2-m wide by 10-cm deep pit filled with loose river rocks from a major hardware store. Kinematic data from 57 reflective markers were used to track full body kinematics at 120 Hz using a 26-camera motion capture system (Motion Analysis, Santa Rosa, CA, USA) (Wilken et al., 2012). All subjects wore their own athletic shoes during data collection. Additionally, data was collected during overground, stair, and slope walking. Those data are reported elsewhere (Aldridge et al., 2012; Ferris et al., 2012).

Walking speeds were normalized to scale speed to each subject's leg length, l, according to Walking Speed $= \sqrt{Fn \cdot g \cdot l}$, where Fn is the Froude number, and g is the gravitational constant (Vaughan and O'Malley, 2005). Subjects walked at three controlled speeds corresponding to Fn of 0.10, 0.16, and 0.23. An audible cuing system was used to control the walking speed (BiofeedTrak, Motion Analysis, Santa Rosa, CA, USA). Additionally, subjects walked at their self-selected (SS) walking speed. The order of testing was always SS pace, followed by controlled speeds 1–3 in order of ascending speed. A minimum of five left and five right strides were collected at each speed. There were several incomplete data sets, due to issues with the collection system, that were not included in these analyses. These include controlled speeds 2 and 3 and SS for P01 in the 'BiOM' condition and SS for P02 in the 'ESR' condition.

2.3. Data analysis

Marker position data were filtered using a 4th order low-pass Butterworth filter with a 6 Hz cut-off frequency. Joint centers were defined relative to segmental markers, found by manual palpation and recorded using a digitizing pointer (C-Motion, Inc., Germantown, MD, USA). Segmental markers and landmarks were used to create a 9 segment whole-body model consisting of feet, shanks, thighs, pelvis, trunk, and head. Local coordinate systems for each segment were defined using the International Society of Biomechanics (ISB) recommendations (Wu et al., 2002, 2005). Whole body center of mass (COM) was calculated as the weighted average of segment COMs using Visual3D (C-Motion, Inc., Germantown, MD, USA). $\dot{\text{COM}}$ velocity ($\dot{\text{COM}}$) was calculated from the COM using a first difference formula. Lateral stability

Table 1
Subject details.

Subject	Gender	Age	Height (m)	Weight (kg)	LL (cm)	Affected limb	Original prosthesis
1	M	32	1.93	102.0	108.0	R	Re-Flex VSP
2	M	38	1.80	99.0	101.0	L	Re-Flex VSP
3	M	29	1.93	108.9	113.0	R	FlexFoot
4	M	29	1.78	93.2	95.0	R	LP Re-Flex VSP
5	M	22	1.87	96.4	104.0	L	Renegade
6	M	38	1.70	97.7	92.5	R	Renegade
7	M	26	1.83	87.7	98.0	R	Re-Flex VSP
8	F	34	1.65	85.5	86.0	L	Renegade
9	M	29	1.83	93.2	92.5	R	Re-Flex VSP
10	M	25	1.93	97.5	107.0	R	Pathfinder
11	M	26	1.75	84.3	98.5	L	Re-Flex VSP
	Mean (SD)	30 (5)	1.82 (0.09)	95.0 (7.3)	99.6 (8.0)		

was then quantified as the minimum margin of stability (MOS) during stance (as previously described in Gates et al., in press).

$$\mathrm{MOS} = \mathrm{BOS} - \left(\mathrm{COM}_Z + \frac{\mathrm{COM}_Z}{\omega_o}\right). \quad (1)$$

Where the base of support (BOS) was defined as the medial–lateral position of the 5th metatarsal marker (Gates et al., in press), COM_Z and $\dot{\mathrm{COM}}_Z$ were the position and velocity of the COM in the medial–lateral direction, and ω_o was the eigenfrequency of a non-inverted pendulum with a length of 1.34 times the trochanteric height (Hof et al., 2007).

Heel strikes were determined using a velocity-based detection algorithm (Zeni et al., 2008) and then verified by visual inspection. Step width and step length were calculated as the difference between the heel markers in the medial–lateral and anterior–posterior directions, respectively. Foot angle at initial contact was defined as the angle between a line connecting the heel and toe markers and horizontal (Menant et al., 2009). The initial position of the foot was subtracted such that the foot angle was zero during quiet stance. Toe clearance was defined as the minimum height during swing measured relative to the height of the toe when the foot was flat on the rock surface. Since the rock surface was mobile, the height of the rocks may have been different in different parts of the pit or between days. Therefore, the height was estimated as the average position of the toe marker between 20 and 30% of the gait cycle, when the foot was flat and before it began to push into the moveable surface.

2.4. Statistics

Comparisons of walking speeds were made using a series of paired t-tests. Comparison of all other dependent measures were made using 4-way ANOVAs with Condition (BiOM, ESR), Walking Speed (1–3), and Limb (prosthetic, intact) as fixed factors and Subjects as a random factor. Due to the larger inter-subject variation in self-selected walking velocity, data for this condition is presented in figures but not included in the statistical analysis of dependent measures. Statistical analyses were performed using SPSS 16 (SPSS Inc., Chicago, IL, USA), with a level of significance of $p \leq 0.05$ for all comparisons. Significant interactions were explored in SPSS using the estimated marginal means with a Bonferroni correction factor for multiple comparisons.

3. Results

Patients had a 10% greater self-selected walking speed when wearing the powered BiOM than energy-storage and return (ESR) prosthesis (ESR: 1.05 (SD 0.17) m/s, BiOM: 1.16 (SD 0.018) m/s, $p = 0.031$; Fig. 1B). Subjects did not accurately match the tone during controlled speed conditions 2 and 3 (Fig. 1A). Their actual speed was significantly slower than their target speed (~7%) when wearing either prosthesis ($p < 0.05$). Despite walking at a slower speed than targeted during the controlled speed conditions, there were no differences in walking speed between the different prostheses ($p > 0.276$; Fig. 1A).

There were no differences in step width ($p = 0.975$) or step length ($p = 0.163$) between prostheses (Fig. 2). Patients took slightly longer steps with their intact limbs ($p = 0.076$), and longer steps at faster walking speeds ($p < 0.001$).

Subjects exhibited a decreased foot contact angle when wearing the BiOM ($p = 0.007$) and on their intact limb ($p < 0.001$; Fig. 3). There was also a significant Limb × Prosthesis interaction ($p = 0.016$). Post-hoc analysis determined that the change in foot contact angle only occurred on the intact limb. The difference between the limbs was approximately 8°, while the difference between prostheses on the intact limb was approximately 4° (Supplemental Material).

There were several differences in lower extremity kinematics between devices that reached statistical significance (Fig. 4). Patients

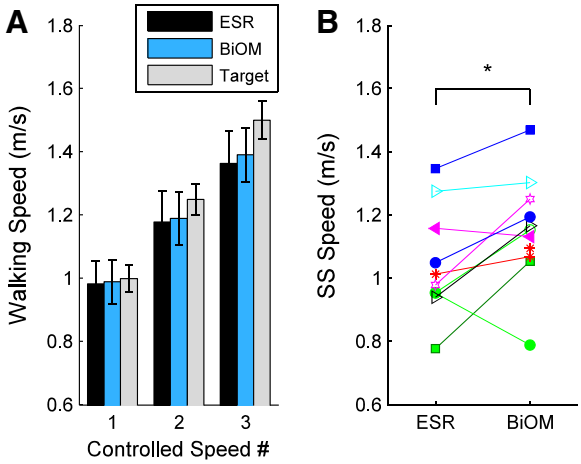

Fig. 1. Walking speed. A) Walking speeds for the controlled speed condition are shown for the passive energy-storage and return (ESR) prosthesis and for powered BiOM prosthesis with the targeted speed. Error bars represent the 95% confidence interval about the mean. B) The self-selected (SS) walking speed is shown for patients when wearing both prostheses. Lines represent each subject. Data from P02 is not shown because there was no data for the ESR prostheses. '*' represents a significant difference ($p < 0.05$).

exhibited greater ankle plantarflexion throughout the gait cycle when wearing the BiOM compared to ESR. Specifically, patients had increased ankle plantarflexion during loading response ($p = 0.002$), and pre-swing ($p < 0.001$), and decreased ankle dorsiflexion during terminal stance ($p = 0.012$) and swing ($p = 0.011$) when wearing the BiOM. There were Limb × Condition interaction effects for ankle plantarflexion during pre-swing ($p < 0.001$) and dorsiflexion during swing ($p < 0.001$). Patients had greater between-limb differences in peak plantarflexion at pre-swing when wearing the ESR prosthesis. There was a significant between-limb difference in the peak dorsiflexion angle during swing when patients were wearing the BiOM ($p < 0.001$), but not the ESR ($p = 0.074$).

At the knee, subjects were approximately 3° more flexed during loading response when wearing the ESR ($p = 0.045$; Fig. 4). They also had greater loading response, and swing knee flexion on their intact limbs ($p = 0.003$ and 0.023, respectively). The between-limb difference in knee flexion during loading response increased with walking speed ($p_{\mathrm{Limb} \times \mathrm{Spd}} = 0.017$). In contrast, the between-limb difference in knee flexion during swing decreased with walking speed ($p_{\mathrm{Limb} \times \mathrm{Spd}} = 0.024$). Late stance extension was not affected by prosthetic type, limb or speed.

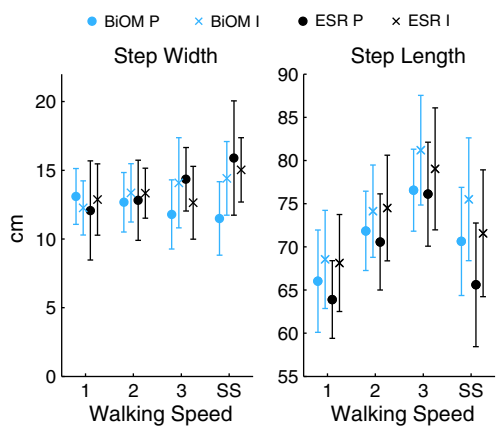

Fig. 2. Temporal–spatial measures. Step width and length are shown for patients walking on a rock surface with an ESR or BiOM prostheses. Data for the intact ('I') and prosthetic ('P') limb are shown separately for each of the three controlled speeds (1–3) and self-selected (SS) walking speed. Error bars represent the 95% confidence interval about the mean.

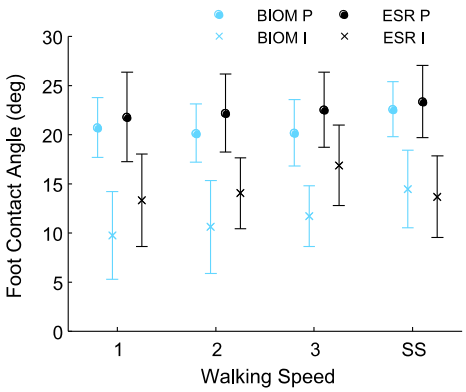

Fig. 3. Foot contact angle. A) The average foot contact angle is shown for patients walking on a rock surface with the ESR and BiOM prostheses. Data for the intact ('I') and prosthetic ('P') limb are shown separately for each of the three controlled speeds (1–3) and self-selected (SS) walking speed. Error bars represent the 95% confidence interval about the mean.

There were no differences in hip kinematics between the two prostheses (Fig. 4). The amount of hip flexion increased with increasing walking speed ($p \leq 0.039$). Peak hip flexion during early stance and swing were greater on the prosthetic limb than the intact ($p \leq 0.004$).

Subjectively, patients tended to stay on top of the surface of the rocks when wearing the ESR prosthesis, but pushed into them when wearing the BiOM. This can be seen in the increased minimum toe position when wearing the BiOM (Fig. 5A and B) and significant Limb × Condition interaction effect ($p = 0.002$). Despite pushing further into the rocks, patients had increased minimum toe clearance during swing when wearing the BiOM compared to ESR ($p = 0.022$; Fig. 5A and C). Toe clearance was also greater on the intact limb compared to prosthetic limb for both prostheses ($p < 0.001$), and increased with walking speed ($p = 0.004$).

Participants walked with decreased medial–lateral COM motion when wearing the BiOM prosthesis ($p = 0.020$). The reduced range of motion did not affect the MOS between the different devices ($p = 0.662$). Medial–lateral motion of the COM and MOS decreased with increasing walking speeds ($p < 0.01$). There were no between-limb differences in COM motion or MOS ($p > 0.195$).

4. Discussion

This study looked at the effect of a powered ankle prosthesis in subjects with unilateral transtibial amputation walking on a loose rock surface. We proposed two alternative hypotheses. First, that the BiOM would increase gait symmetry and increase walking speed in comparison to an ESR prosthesis, or alternatively that subjects would adopt a cautious gait when wearing the BiOM. Subjects were able to cross this surface at all speeds with both devices tested without experiencing any loss of balance. There was no difference in step width or step length between the two devices (Fig. 2), and subjects had a faster self-selected walking speed when wearing the BiOM compared to ESR prosthesis (Fig. 1). Therefore, subjects did not appear to adopt a cautious strategy when wearing the BiOM and hypothesis #2 was rejected.

One proposed benefit of the BiOM device is that the added power may increase symmetry between the limbs. A small change in early stance knee flexion (Fig. 4) was observed with the BiOM, however, the difference was less than previously published minimal detectable change values (Wilken et al., 2012) and not likely of clinical significance. No other kinematic differences were observed between the BiOM and ESR at the hip or knee. Thus the between-limb differences that existed when wearing the ESR prosthesis persisted in the BiOM. Similar to previously published results (Aldridge et al.,

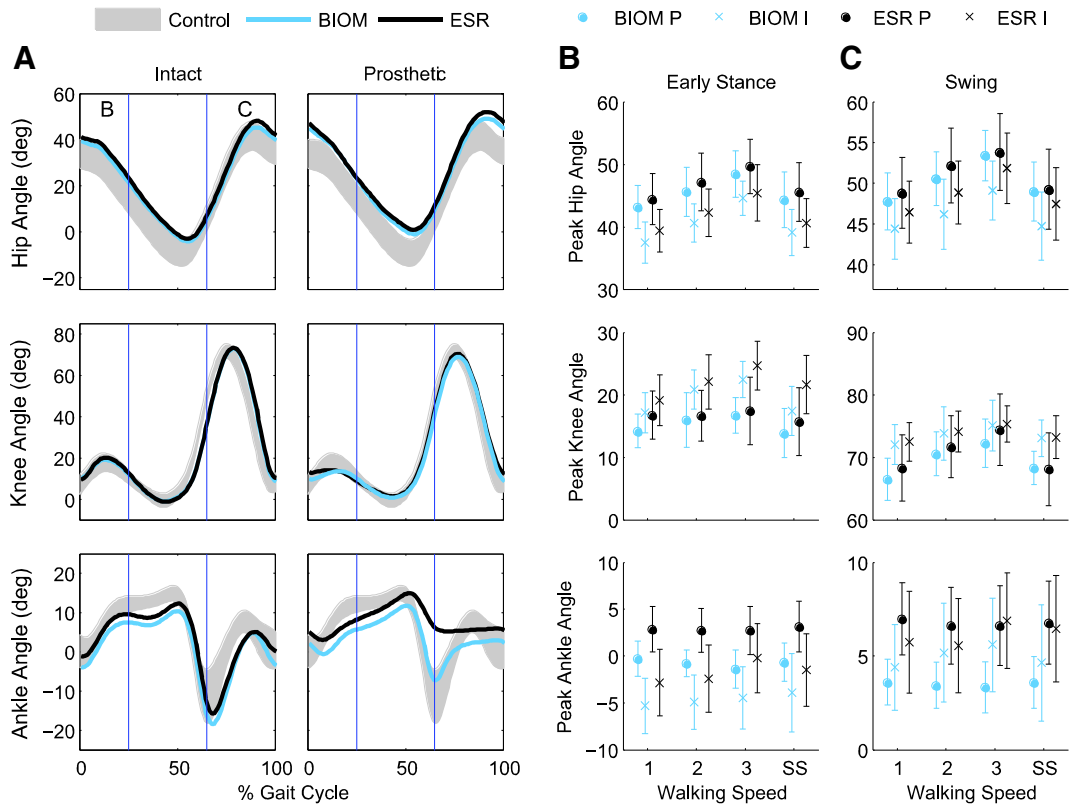

Fig. 4. Lower extremity kinematics. A) Average hip, knee and ankle angles across subjects are shown for the intact ('I') and prosthetic ('P') limbs while subjects walked at controlled speed 2. Normative data is taken from 15 healthy young adults walking at the same controlled speed (Gates et al., 2012b). B) Peak hip and knee flexion and ankle plantarflexion between 0 and 25% of the gait cycle are shown for the BiOM and ESR conditions. C) Peak hip and knee flexion and ankle dorsiflexion during swing (65–100% gait cycle) are shown for controlled speeds 1–3 as well as self-selected (SS) walking speed. Error bars represent the 95% confidence interval about the mean.

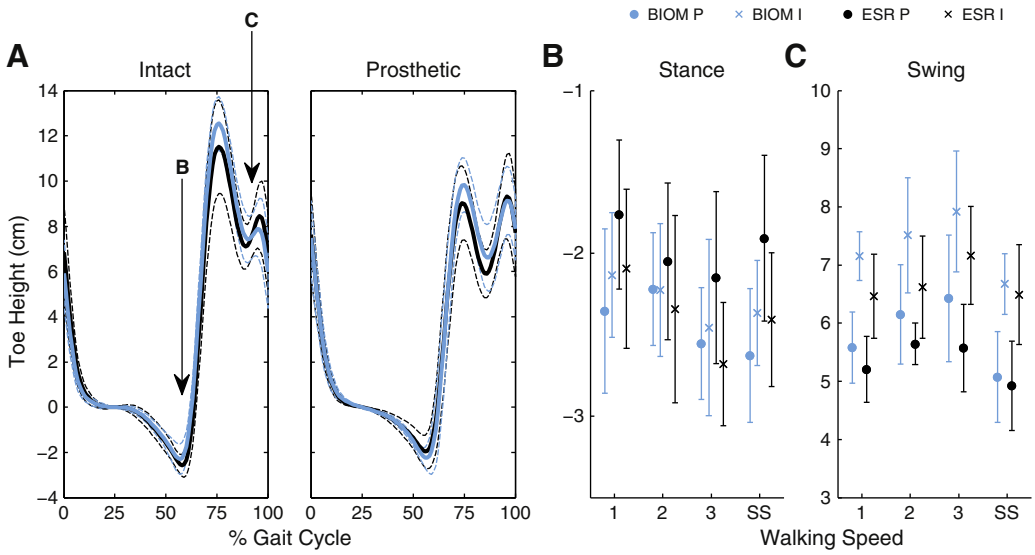

Fig. 5. Toe clearance. A) The average position of the toe marker across subjects is shown for the intact ('I') and prosthetic ('P') sides when wearing the BiOM (blue) and ESR (black) prostheses. The height of the toe marker was normalized to its position during foot flat. The minimum position of the toe marker during pre-swing (B) and swing (C) are shown at each of the three controlled speeds as well as self-selected ('SS'). Error bars represent the 95% confidence interval about the mean.

2012; Ferris et al., 2012) the BiOM did provide increased ankle plantarflexion which was particularly evident in the more than 10° increase from the ESR at push-off. The smaller differences seen at other points in the gait cycle were likely due to the alignment chosen by the prosthetist and not a function of the device itself since the unloaded (i.e. swing) position was also more plantarflexed. Therefore, the device did not result in between limb symmetry on an uneven surface, and hypothesis #1 was only partially supported.

One way that able bodied individuals adapted their gait when walking on this surface was to alter their foot orientation to decrease the angle between the foot and surface (Gates et al., 2012b). This enables them to rapidly achieve a foot flat position. This may be beneficial since having more of the plantar surface in contact with the walking surface, especially on uneven ground, enhances stability (Su et al., 2010). Here, there were no differences in foot contact angle between the two devices for the prosthetic limb. There were also no differences in medial–lateral margins of stability between the prostheses. Although the BiOM is capable of plantarflexing, it does not plantarflex to achieve a foot flat position at initial contact; rather, it rotates only with loading. This feature could be added to future designs of intelligent devices to help improve stability. In practice, this may be difficult to achieve as it would require the device to have some a priori knowledge of the surface prior to stepping.

One of the primary differences between the ESR prosthesis and BiOM is that the BiOM adjusts the ankle into plantarflexion at push-off and supplies power. Here we were not able to measure the ground reaction force, or subsequently joint kinetics. This same subject group exhibited a 125% increase in ankle power when using the BiOM compared to ESR when walking on level ground (Ferris et al., 2012). Therefore it is likely that they also had increased ankle power when walking on the rocks. Subjectively, this additive power contributed not only to accelerating the limb forward and an increased self-selected walking speed, but also caused more rocks to be displaced backward as the forefoot sunk into the loose rock surface and was pushed backward (Fig. 5).

Individuals with transtibial amputation may also have difficulty clearing the toe during swing since they are unable to dorsiflex the foot. The ESR prosthesis will remain in position while the BiOM moves to a set point and locks at a constant angle during swing. To compensate for this, individuals with transtibial amputation can increase hip and knee flexion (Moosabhoy and Gard, 2006). Here, despite the BiOM being set in a more plantarflexed position, which would be expected to decrease toe clearance, there was a mean increase with the BiOM. The mean differences in toe clearance were small, and it is possible that measures lacked sufficient resolution to detect subtle changes in hip and knee kinematics that could possibly account for this increase. It is also possible that the small change in knee stance flexion on the intact limb when wearing the BiOM (Fig. 4) was sufficient to raise the hip joint center, thereby increasing toe clearance. While the toe clearance increased with the BiOM, it was still significantly less than the intact limb. Future designs could include active dorsiflexion during swing, similar to the Proprio foot, to ensure adequate toe clearance on an uneven surface.

Part of the reason for the lack of differences between devices may be that the patients tested here were young, highly active, and already operating at a high functional level with their ESR prosthesis. While there were kinematic differences between individuals with TTA and controls (Gates et al., 2012b), these differences were quite small (Fig. 3). Greater changes in kinematics may be expected in more impaired populations such as the elderly or patients who have had an amputation as a result of vascular disease. We also might expect larger difference in a longer duration walking trial or walking with additional load.

5. Conclusion

Kinematic analysis of persons with unilateral transtibial amputation walking on a loose rock surface revealed that the powered BiOM prosthesis increased self-selected speed, ankle plantarflexion at push-off, and toe clearance in comparison to a passive ESR prosthesis. The addition of power did not normalize joint kinematics at the knee or hip. Future devices designed for navigating irregular surfaces should focus on altering the foot orientation at initial contact and actively dorsiflexing the foot during swing to achieve additional increases in toe clearance.

Conflict of interest

The authors have no conflicts of interest to report.

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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.2013.03.005.

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