A comparison of two prosthetic feet on the multi-joint and multi-plane kinetic gait compensations in individuals with a unilateral trans-tibial amputation

Heather A Underwood, MD^{1,3}, Craig D Tokuno, MSc^{2,3}, and Janice J Eng, PhD^{2,3}

¹Division of Physical Medicine and Rehabilitation, University of British Columbia, Vancouver, BC, Canada

²Department of Physical Therapy, University of British Columbia, Vancouver, BC, Canada

³Rehabilitation Research Laboratory, GF Strong Rehab Centre, Vancouver, BC, Canada

Abstract

Objective—To determine the effects of two different prosthetic feet on the three-dimensional kinetic patterns of both the prosthetic and sound limbs during unilateral trans-tibial amputee gait.

Design—Eleven individuals with a unilateral trans-tibial amputation participated in two walking sessions: once while using the conventional SAFE foot, the other while using the dynamic Flex foot.

Background—Despite the wide variation in the design of prosthetic feet, the benefits of these prostheses remain unclear.

Methods—During each test session, peak joint moments and powers in the sagittal, transverse and frontal planes were examined, as subjects walked at a comfortable speed.

Results—The majority of the kinetic differences that occurred due to the changing of prosthetic foot type were limited to ankle joint variables in the sagittal plane with greater peak moments and power during propulsion for the Flex foot compared to the SAFE foot. However, effects were also found at joints proximal to the prosthesis (e.g. knee) and differences were also found in the kinetics of the sound limb.

Conclusion—The dynamic Flex foot allowed subjects to rely more heavily on the prosthetic foot for propulsion and stability during walking with minimal compensations at the remaining joints.

Relevance—Determining the biomechanical differences between the conventional and dynamic prosthetic feet may advocate the use of one prosthetic foot type over another. This information, when used in conjunction with subjective preferences, may contribute to higher functioning and greater satisfaction for individuals with a lower limb amputation.

Keywords

trans-tibial amputation; prosthetic foot; gait

1. Introduction

The task of walking is an important activity of daily life. Though many individuals consider the movement to be a simple and effortless task, for others the task can be much more complex. In particular, individuals who undergo a unilateral trans-tibial amputation must learn to adapt to the loss of musculature and the altered sensation from a lower leg while at the same time, compensating for the properties of their prosthetic foot. Some, such as the Flex foot and Seattle foot, are regarded as “dynamic” prostheses, where energy can be stored within the prosthesis during the early and mid-stance periods of gait and consequently, return a portion of the stored energy back to the individual during the push-off phase of gait. Conversely, cheaper and simpler models, such as the SACH (single axis cushion heel) and SAFE (stable ankle, flexible endoskeleton) feet, do not contain this dynamic component and hence, the individual with an amputation will need to adjust by relying on alternative sources of propulsion during steady state walking.

Despite the potential differences in energy storing capacity, previous research has found few sagittal plane kinetic differences between the ankle joints of dynamic and more conventional prosthetic feet; moments in other planes have not been examined. Some have shown energy storing prosthetic feet to allow the individual with a trans-tibial amputation to increase the plantarflexor moment during the push-off phase of walking (Torburn et al., 1990) and to increase the amount of energy stored and released at the prosthetic ankle (Barr et al., 1992; Ehara et al., 1993; Winter and Sienko, 1988). In contrast, others have found negligible changes in the plantarflexor moment (Lehmann et al., 1993), while Gitter et al. (1991) were not able to detect statistical differences in energy release between energy storing feet and the conventional SACH foot.

The knee and hip joints on the prosthetic side also appear to show little change as a result of using a dynamic versus a non-dynamic prosthetic foot. During the initial phase of stance, the knee has been found to exhibit a flexor moment when using the SACH (conventional model), Seattle (dynamic model) or Carbon Copy (dynamic model) foot (Barr et al., 1992; Gitter et al., 1991; Lehmann et al., 1993; Torburn et al., 1990). Yet, others have found either a slightly flexor (Lehmann et al., 1993) or an extensor moment (Gitter et al., 1991) when subjects were using the energy storing Flex foot. Finally, the magnitude of the flexor moment, as well as the sagittal plane power generation and absorption bursts at the hip joint have not been found to be different when using the energy storing feet, as compared to the more conventional ones (Barr et al., 1992; Torburn et al., 1990). Though the sagittal plane kinetics at the knee and hip have been well documented, the effects of prosthetic foot type on the knee and hip joint kinetics of the frontal and transverse planes have not yet been investigated.

Interestingly, although the aforementioned studies could only find minor adaptations with changes in prosthetic feet, many individuals with a unilateral trans-tibial amputation still

perceive large differences in walking performance. Subjects often feel that the energy storing feet enable them to walk faster and maintain a more stable movement pattern, resulting in a strong preference for the dynamic feet over the more conventional models (Nielsen et al., 1989; Torburn et al., 1990). This lack of relationship between subject perception and biomechanical measures may indicate that other compensations, such as those that have not yet been examined, occur during locomotion with individuals with amputations.

As mentioned previously, a drawback to the previous studies is that the analyses have focused solely on the sagittal plane joint kinetics of the prosthetic limb. However, human walking is considered a bipedal task that requires the co-ordination and movement of both limbs in all three (i.e. sagittal, transverse and frontal) planes. It is of importance to examine the kinetic patterns not only in the transverse and frontal planes, but also from the contralateral limb. Recent studies in the healthy adult population provide evidence of the importance and roles of these parameters. MacKinnon and Winter (1993), for example, have shown that the hip abduction moment can help to maintain body balance while others have found that hip movements in both the frontal and transverse planes can contribute to the total work generated during steady-state walking (Eng and Winter, 1995; Sadeghi et al., 2001). The role of the contralateral limb has also been recently documented by Sadeghi et al. (2001). Using bilateral gait data, they determined that the contralateral limb was relied upon for body control and quite possibly, the correction of body propulsion from the other limb.

Thus, it is possible that by simply focusing on these select variables (i.e. those in the sagittal plane and on the prosthetic limb), previous studies have only explained a small part of the adaptations that occur due to differences in the properties of varying prosthetic feet. Stability and propulsion properties will differ depending on the prosthesis used and so, it would be expected that individuals with lower limb amputations will exhibit differences in these variables in order to be able to walk more effectively.

Therefore, this study was designed to examine the effects of two prosthetic feet (the conventional and semi-rigid SAFE foot versus the dynamic elastic response Flex foot) on the three-dimensional kinetic patterns during steady-state gait in individuals with a unilateral trans-tibial amputation. It was hypothesized that due to the different properties between the two prosthetic feet, individuals with a unilateral trans-tibial amputation would demonstrate kinetic differences in the proximal joints on both the prosthetic and sound limbs when using the SAFE foot.

2. Methods

Eleven individuals with a unilateral trans-tibial amputation (8 males, 3 females; subject characteristics are listed in Table 1) participated in this study. Inclusion criteria were that participants had lost their limb due to traumatic reasons, were able to ambulate independently in the community, did not require assistive devices for ambulation, did not experience stump pain or tenderness, and did not have any other cardiovascular, neurological or musculoskeletal conditions.

For this study, a SAFE II prosthetic foot (Campbell-Childs Inc., White City, USA) was chosen as the conventional prosthetic foot (Figure 1A). It is made of polyurethane foam that approximates the structure and function of the anatomical ankle joint (Wing and Hittenberger, 1989). Although it does not have a built-in spring or leaf system, the keel can compress and cause some motion in all three planes. In contrast, the Flex-Walk prosthetic foot (Flex-Foot Inc., Aliso Viejo, USA) was used as the dynamic elastic response prosthesis (Figure 1B). Unlike the conventional foot, it is designed as two carbon leaves in a soft foam cover (Wing and Hittenberger, 1989). The two leaves flex as a downward force is applied, causing it to bend back during the push-off phase of gait (Wing and Hittenberger, 1989). In addition, the split toe was designed to accommodate inversion/eversion for stability on uneven terrain.

Prior to the study, six subjects were using the Flex foot, while the remaining five were using the SAFE foot. The preference of prosthetic foot did not appear to be related to the individual's prosthetic foot that they used on a regular basis or the sequence in which they tested the prosthetic foot conditions (Fisher Exact test, $P > 0.200$). All subjects gave informed consent in accordance with university and hospital policies.

Subjects participated in two test sessions – one while using the dynamic Flex foot, and the other while using the conventional SAFE foot. In a random assignment, half the subjects were evaluated with the Flex foot first while the other half were evaluated with the SAFE foot first. The time period between the two test sessions was sufficient (a minimum of 30 minutes) such that the study participants could become familiar and comfortable with the properties of each prosthetic foot. During the period of adjustment, subjects were encouraged to take part in a variety of tasks, such as walking on even and uneven (i.e. dense foam) terrain, turning, stair climbing, and if possible, running. Proper alignment and fitting of the prosthesis was ensured by a certified prosthetist who had more than ten years experience with amputee clients. The prosthetist adhered to the detailed guidelines provided by the manufacturer.

For each test session, bilateral gait data were collected by means of an optoelectric sensor (Optotrak, Northern Digital, Waterloo, Canada) synchronized to two force platforms (Bertec model K71003, Bertec Corporation, Columbus, USA) that were flush mounted to the ground in the middle of the 10-meter walkway. Eighteen infrared emitting diodes (0.2 mm accuracy, 0.01 mm resolution) were used to identify the three-dimensional kinematics of the two lower limbs and anatomical landmarks were identified relative to the diodes using a digitizing probe during a standing calibration trial. Marker locations and the segment model used were similar to that described previously by Eng and Winter (1995). For the prosthetic foot, marker locations were approximated using the contralateral non-amputated foot as a guide.

Participants were asked to walk back and forth along the walkway at a comfortable pace. Shoes were worn during testing. Each individual's walking speed was matched for the two test sessions in order to eliminate potential kinematic and kinetic differences that may arise due to changes in gait speed (Murray et al., 1984; Winter, 1983). This was done with the use of a metronome. After the first test session, the participant's step duration was determined and was used to set the beat frequency of the metronome. Consequently, when subjects were

tested for the second session, they were again asked to walk back and forth along the walkway at a comfortable pace. While they were not specifically instructed to walk symmetrically, it is possible that subjects walked more symmetrically due to the use of a metronome to match the speeds, however, the beat was simply used as a guide for each foot contact so that gait speed could be kept relatively constant between test sessions and not to induce symmetry of the steps. From Table 2, it is evident that we were successful in getting similar gait speeds across the two different prosthetic conditions, but the steps were still asymmetrical within each condition. Lastly, given the randomization of the two prosthetic feet, one-half of the subjects would have had their gait speed set with the Flex foot and the other half the SAFE foot.

Coordinate data were sampled at 60 Hz and the force data were sampled at 600 Hz; the noise from both signals was reduced with a second order zero-phase lag Butterworth filter with a cutoff frequency of 6 Hz and 50 Hz respectively. Trials were rejected if the foot did not land on the force platform or if targeting of the platform appeared to occur. Targeting was considered to occur if we observed that the subject was visually focusing on the force platform, if there was an obvious shortening or lengthening of the stride to ensure force plate contact or if the subject appeared to glance downwards. To minimize the effects of fatigue, rest periods were given between trials and between the two prosthetic feet test sessions. Five trials were collected from each leg.

Joint moment and power calculations were determined from synchronized co-ordinate and force data using an inverse dynamics routine (Eng and Winter, 1995). The major power bursts were labeled according to those used by Eng and Winter (1995), where the first letter designates the joint, the number refers to the sequence of the power burst, and the second letter indicates the plane of motion. For example, A1S refers to the first peak power burst of the ankle joint in the sagittal plane. Thirteen peak joint moments and 16 peak joint power variables were identified as consistent variables across subjects and were extracted from individual (not averaged) trials and then the mean was calculated over the five trials.

At the conclusion of each test session, subjects were asked to rate on a scale of 1 to 10 (with 10 being the best possible score), six questions regarding the stability and mobility of the prosthetic foot. The questions were asked in terms of 1) stability while standing on level ground, 2) stability while standing on a slope, 3) stability while walking on level ground, 4) stability while walking on uneven ground, 5) ability to walk quickly and 6) stability while standing on a foam surface.

Two factor (Flex versus SAFE prosthetic type and sound versus prosthetic limb) repeated measures ANOVAs were used to determine statistical differences between the mean peak moments and power bursts during the two prosthetic conditions. Alpha was set a priori at $P < 0.05$.

3. Results

Subjects walked at a similar velocity, with a similar cadence, step length, swing time and stance time between the dynamic Flex foot and conventional SAFE foot (Table 2). Since

subjects were asked to walk to the beat of a metronome in order to control for gait speed, none of the above measures were found to be statistically different between the two prosthetic type conditions.

The mean peak moments and powers are listed in Tables 3 and 4 and the results describe the differences between the SAFE and Flex foot, rather than differences between the prosthetic and sound limb for each of the different prosthetic types. Of the 13 peak moments that were examined, significant findings were limited to variables at the ankle and knee joints. At the ankle, subjects were able to apply a 15% greater ankle plantarflexor moment on their prosthetic limb when wearing the dynamic Flex foot as compared to the SAFE foot ($P=0.018$ for prosthetic type $\times$ limb interaction). When subjects were tested with the Flex foot, a 45% greater inverter moment on the prosthetic limb was found compared to the SAFE foot but this did not reach statistical significance ($P=0.107$ for the prosthetic type $\times$ limb interaction).

Differences were also found in both the sagittal and frontal planes of the knee joint. The use of the Flex foot resulted in a larger knee flexor moment on the prosthetic limb but a slightly smaller knee flexor moment on the sound limb, in comparison to the SAFE foot condition. ($P=0.007$ for the prosthetic type $\times$ limb interaction). As push-off began to occur, the second peak knee extensor moment was found to differ between prosthetic types, where a greater knee extensor moment in both the prosthetic (34% increase) and sound (14% increase) limbs was found when subjects walked with the dynamic Flex foot ($P=0.034$ for the prosthetic type main effect). In the frontal plane of the knee, the magnitude of the first peak knee abductor moment during early stance had opposing effects on the sound and prosthetic limbs when using each of the two prosthetic feet. The knee abductor moment was 16% greater on the prosthetic side, when using the Flex foot. On the sound limb, a 13% smaller knee abductor moment was found when subjects wore the Flex foot compared to the SAFE foot; ($P=0.011$ for the prosthetic type $\times$ limb interaction). None of the hip moment variables were found to be significantly different between prosthetic foot type conditions.

Of the 15 peak powers of interest, differences were again limited to those occurring at the ankle and knee. In specific, a larger A1S power absorption burst on both limbs ($P=0.014$ for the prosthetic type main effect) was found with the Flex compared to the SAFE foot. A 27% increase in the A2S power burst on the prosthetic ankle was also found when using the Flex foot and this concurred with the observed increase in the ankle plantarflexor moment. However, it did not quite reach statistical significance ($P=0.082$ for the prosthetic type $\times$ limb interaction). At the knee, a greater K3S power absorption was found on both knees when using the Flex foot ($P=0.007$ for the prosthetic type main effect). Finally, a prosthetic type main effect ($P=0.020$) was found for the K2F power absorption burst, where the use of the Flex foot resulted in less absorption at the knee than the SAFE foot.

While there were again no statistical differences found at the hip joint, several trends were observed in the three sagittal plane power bursts at this joint. The H1S power generation tended to be smaller on both the prosthetic (a 66% decrease) and sound (a 75% decrease) limbs when subjects were wearing the conventional Flex foot ($P=0.059$ for the prosthetic type main effect). Second, the H2S power absorption burst appeared to be the smaller on the

prosthetic limb when using the Flex foot $P=0.060$ for the prosthetic type $\times$ limb interaction). Finally, the magnitude of the H3S power generation burst was found to be greater by 57% on the prosthetic limb during the Flex foot condition ($P=0.062$ for the prosthetic type $\times$ limb interaction).

The results from the ratings of perceived stability and mobility on the six-question questionnaire are listed in Table 5. Although the dynamic Flex foot was, on average, preferred over the SAFE foot for all six questions, only two measures (ability to walk quickly and stability on foam surface) were found to be different by a score of greater than 1.

4. Discussion

The purpose of this study was to compare the effects of a conventional versus dynamic prosthetic foot on the three-dimensional kinetic walking patterns of both the prosthetic and sound limbs in individuals with a unilateral trans-tibial amputation. Despite a perceived preference for the more expensive dynamic Flex foot, previous research has found few biomechanical differences between the gait profiles of these two prostheses on the sagittal kinematics and kinetics of the prosthetic ankle. Our study presents several new findings.

Large effects were found at the ankle of the prosthetic limb, and were comparable to those found in previous studies (Barr et al., 1993; Ehara et al., 1993; Torburn et al., 1990; Winter and Sienko, 1988). The deformation properties of the dynamic foot allowed for a greater power absorption (A1S) during weight acceptance and consequently, a trend towards a greater plantarflexor moment and power generation (A2S) took place at push-off. Although the use of the dynamic Flex foot increased the plantarflexor moment and push-off power for the prosthetic ankle, these values were still found to be well below that of the sound ankle.

The present study also found that differences occurred at joints proximal to the ankle joint itself. This is in contrast to previous studies which have reported no differences in the sagittal plane joint moments at the prosthetic hip and knee (Barr et al., 1992; Torburn et al., 1990). A change occurred at the knee joint of the prosthetic leg, where the greater plantarflexor moment with the Flex foot necessitated a subsequent increase of the K3S power absorption burst and late stance knee extensor moment to control for the knee collapse. Furthermore, due to the diminished ankle push-off capabilities with the conventional SAFE foot, several trends (i.e. in the range of $P=0.059$ to 0.062) were observed at the hip of the prosthetic leg. First, there was a slightly larger amount of power being generated on both limbs during the early stance H1S power burst when using the SAFE foot. It is possible that this finding is related to the work by Winter and Sienko (1988), who have found that individuals with a unilateral trans-tibial amputation do walk with greater amounts of hip extensor activity. In fact, they termed this adaptation as a “push-from-behind” because a large hip extensor action early in stance can compensate for the later weaker ankle push-off. Second, the H3S power generation burst, whose role is to add energy to the swinging limb (Eng and Winter, 1995), was found to be smaller on the prosthetic limb during the SAFE foot condition. One may have hypothesized that the opposite result would occur, whereby an increased H3S burst when using the non-dynamic foot could have offset the lack

of push-off with the prosthetic ankle. Instead, it may be possible that other factors had a strong effect in the decreased involvement of the H3S burst.

Work in the non-sagittal planes of the hip joint has also been found to contribute to the total work in the frontal and transverse planes in the healthy population (Eng and Winter, 1995; Sadeghi et al., 2001). However, in the present study, differences at the hip in the non-sagittal planes did not accompany changes in prosthetic feet. This is surprising since the diminished actions of the SAFE foot could have been supplemented by an increase in the frontal and transverse plane hip parameters. It is possible that the lack of significance found in our study is a reflection of the wide range of gait strategies exhibited by individuals with a trans-tibial amputation especially in non-sagittal planes.

While the original intention of the dynamic component in prosthetic feet was to help with one's ability to propel forward, steady state walking requires both the control of dynamic stability and at the same time, the application of the appropriate forces in order to move forward. As such, the choice of prosthesis also appears to have effects during other phases of the gait cycle. For example, when using the SAFE foot condition, there was an increased need for passive stability on the knee of the sound limb (as demonstrated by the increased knee abductor moment) during weight acceptance in order to counteract the instability during the propulsion phase on the contralateral leg. In addition, the slightly larger H1S power burst on the prosthetic limb shortly after heel strike may have assisted to control the trunk and the collapse of the stance limb. These two findings suggest that the dynamic Flex foot requires less passive and active stability to be controlled by the individual as compared to the conventional SAFE foot.

It is perhaps this improved stance stability, in addition to the additional push-off power that underlies the overwhelming preference for the dynamic Flex foot. Murray et al. (1998) found that compared to the conventional SACH foot, subjects perceive the dynamic feet (i.e. the Seattle foot in their study) to reduce the magnitude of the stress at the hip and knee, as well as improve their sense of balance. Likewise, Postema et al. (1997) postulated that the flexible feet allow for better balance possibilities in subjects who have good proprioceptive control in their prosthetic limb. In our study, the preference for the dynamic Flex foot became more pronounced as the task became more difficult (i.e. standing on level ground to standing on a foam surface) and may be attributed to the greater power absorption (A1S) during weight-acceptance which can assist in the absorption of loading forces and in the adjustment to uneven terrain.

In summary, it is evident that the kinetic effects when using a dynamic compared to a conventional prosthetic foot were small, especially those not involving the sagittal plane or the prosthetic ankle joint. However, the lesser need for proximal compensations at the hip may be key factors underlying the preference of the dynamic Flex foot over the conventional SAFE foot, in addition to the added ankle push-off power.

It would be useful in the future to evaluate these subjects while they are engaged in more demanding tasks and during longer wear times (e.g. over weeks or months). It is known, for example, that increases in walking and/or running speed are associated with changes in the

magnitude of joint moments and powers in the able-bodied population (Winter, 1983). By altering task difficulty, it would in turn create greater body disequilibrium and may accentuate the discrepancies between the dynamic and conventional feet. A practice period, involving several weeks or even months, may be of further benefit as it would enable the user to be able to practice and form impressions with each of the prosthetic feet with respect to endurance or fatigue (e.g. all day use), varying terrain, or different weather conditions (e.g. wet or icy sidewalks). Lastly, having a larger sample size would certainly aid in detecting statistical differences, as large variability exists in the non-sagittal planes of gait profiles (Eng and Winter, 1995). Numerous trends were observed, especially at the hip joint, and testing several more individuals could help to determine whether those parameters are clinically relevant.

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Figure 1.
Photo of *A*, the SAFE II prosthetic foot and *B*, the Flex-Walk prosthetic foot.

Table 1

Subject characteristics (n = 11). M refers to male; F refers to female. L refers to the left leg, R refers to the right leg.

	Mean (or n)	S.D.	Range
Gender (M/F)	8 M, 3 F	-	-
Amputated Side (L/R)	5 L, 6 R	-	-
Age (years)	42.50	13.11	22 – 59
Mass (kg)	80.33	14.32	56.8 – 104.5
Height (cm)	171.20	9.44	150 – 185
Time since amputation (years)	11.08	13.26	1 – 31

Table 2

Mean (SD) spatio-temporal parameters during the SAFE and Flex foot conditions.

	SAFE foot	Flex foot
Walking speed (m/s)	1.44 (0.18)	1.43 (0.20)
Cadence (steps/min)	108.31 (8.38)	106.26 (7.80)
Step length (Prosthetic) (m)	0.82 (0.12)	0.86 (0.08)
Step length (Sound) (m)	0.79 (0.12)	0.77 (0.11)
Stance time (Prosthetic) (m)	0.66 (0.06)	0.65 (0.05)
Stance time (Sound) (m)	0.69 (0.06)	0.68 (0.06)
Swing time (Prosthetic) (m)	0.44 (0.02)	0.44 (0.02)
Swing time (Sound) (m)	0.42 (0.02)	0.43 (0.02)

Table 3

Mean (SD) peak moments (in Nm/kg) for both the prosthetic and sound limbs during the SAFE and Flex foot conditions (^a indicates $P < 0.05$ for the prosthetic type main effect, ^b indicates $P < 0.05$ for the prosthetic type x limb interaction).

MOMENTS	PROSTHETIC		SOUND	
	SAFE	FLEX	SAFE	FLEX
Ankle Invertor	0.11 (0.08)	0.16 (0.07)	0.17 (0.09)	0.15 (0.12)
Ankle Dorsiflexor	0.33 (0.14)	0.43 (0.15)	0.30 (0.15)	0.30 (0.10)
Ankle Plantarflexor ^{a,b}	1.21 (0.16)	1.39 (0.21)	1.51 (0.30)	1.48 (0.27)
Knee Abductor 1 ^b	0.48 (0.24)	0.57 (0.25)	0.69 (0.35)	0.60 (0.28)
Knee Abductor 2	0.30 (0.19)	0.28 (0.15)	0.48 (0.24)	0.52 (0.25)
Knee Extensor 1	0.31 (0.24)	0.47 (0.19)	0.92 (0.58)	0.85 (0.34)
Knee Flexor ^b	0.11 (0.24)	0.18 (0.14)	0.08 (0.27)	0.03 (0.12)
Knee Extensor 2 ^a	0.44 (0.18)	0.59 (0.20)	0.69 (0.47)	0.79 (0.39)
Hip Abductor 1	0.91 (0.22)	0.85 (0.28)	1.18 (0.33)	1.10 (0.36)
Hip Abductor 2	0.70 (0.28)	0.63 (0.27)	1.03 (0.16)	1.00 (0.26)
Hip Internal Rotator	0.12 (0.05)	0.10 (0.04)	0.23 (0.10)	0.25 (0.10)
Hip Extensor	0.75 (0.31)	0.67 (0.28)	0.95 (0.40)	0.89 (0.44)
Hip Flexor	1.40 (0.50)	1.68 (0.64)	1.60 (0.58)	1.79 (0.71)

Table 4

Mean (SD) peak powers (W/kg) for both the prosthetic and sound limbs during the SAFE and Flex foot conditions (^a indicates $P < 0.05$ for the prosthetic type main effect, ^b indicates $P < 0.05$ for the prosthetic type $\times$ limb interaction). Positive values indicate power generation, negative values indicate power absorption.

POWERS	PROSTHETIC		SOUND	
	SAFE	FLEX	SAFE	FLEX
A1S ^a	-0.56 (0.18)	-0.72 (0.21)	-0.71 (0.32)	-0.78 (0.23)
A2S	1.19 (0.51)	1.52 (0.38)	4.28 (1.34)	4.13 (1.36)
K1S	-0.37 (0.38)	-0.45 (0.47)	-1.06 (0.82)	-1.10 (0.70)
K2S	0.26 (0.28)	0.21 (0.13)	0.91 (0.66)	0.72 (0.41)
K3S ^a	-2.42 (0.89)	-3.38 (1.50)	-2.72 (1.92)	-2.70 (0.98)
K4S	-1.00 (0.28)	-1.03 (0.31)	-0.96 (0.26)	-0.93 (0.30)
H1S	1.23 (0.53)	0.81 (0.38)	2.12 (1.14)	1.59 (0.89)
H2S	-1.66 (1.14)	-1.17 (0.64)	-1.27 (0.66)	-1.34 (0.67)
H3S	1.22 (0.58)	1.92 (1.10)	1.89 (0.76)	1.74 (0.67)
A1F	-0.06 (0.05)	-0.03 (0.01)	-0.03 (0.02)	-0.03 (0.02)
K1F	0.17 (0.13)	0.23 (0.16)	0.46 (0.45)	0.29 (0.11)
K2F ^a	-0.21 (0.16)	-0.11 (0.07)	-0.39 (0.31)	-0.28 (0.26)
H1F	-0.47 (0.22)	-0.46 (0.41)	-0.89 (0.53)	-0.73 (0.38)
H2F	0.28 (0.17)	0.26 (0.19)	0.48 (0.24)	0.49 (0.32)
H3F	0.94 (0.74)	0.63 (0.31)	1.44 (0.72)	1.34 (0.68)

Table 5

Ratings of perceived stability and mobility. Answers were based on a scale of 1 to 10, with 10 being the best.

Question	SAFE Foot		Flex Foot		Preference (%)		
	Mean	SD	Mean	SD	SAFE	Flex	None
Stability while standing on level ground	8.7	1.4	9.2	0.8	18	36	45
Stability while standing on a slope	6.8	2.2	7.7	1.5	18	55	27
Stability while walking on level ground	8.7	1.5	8.8	1.0	36	18	45
Stability while standing on uneven ground	7.3	2.4	7.9	1.2	36	36	27
Ability to walk quickly	7.7	2.1	9.1	1.0	0	64	36
Stability while standing on a foam surface	5.9	1.9	7.7	1.7	0	73	27