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Effects of prosthetic ankle power and foot stiffness category on biomechanical asymmetry and knee moment during walking at different speeds

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Abstract

People with unilateral transtibial amputation (uTTA) using a passive-elastic prosthesis typically walk with contact time (t_c) and first and second peak vertical ground reaction force (F_1 and F_2) asymmetry and greater first peak external knee adduction moment in their unaffected versus affected leg. A previous study found that use of stance-phase powered prosthesis (BiOM) at a recommended power setting compared to a passive-elastic prosthesis can reduce t_c asymmetry at self-selected speed and unaffected leg first peak external knee adduction moment at 1.50 –1.75 m/s. However, the BiOM includes a passive-elastic prosthesis that can have different stiffness categories and can be tuned to different power settings, which may affect t_c and F_1 and F_2 asymmetry and unaffected leg first peak external knee adduction moment. Thirteen people with uTTA used 16 different passive-elastic prosthetic foot stiffness categories and BiOM power settings to walk at 0.75–1.75 m/s. We found that use of the stiffest compared to least stiff category reduced F_2 asymmetry. Use of the BiOM reduced t_c asymmetry compared to a passive-elastic prosthesis and the effects of power setting on F_1 and F_2 asymmetry depended on walking speed. To minimize biomechanical asymmetry during walking at 1.25 m/s, people with uTTA should use the BiOM with power settings up to 20% greater than those that match biological ankle joint biomechanics. Such prosthetic settings could potentially reduce unaffected leg joint pain and/or osteoarthritis risk.

List of Abbreviations

+1 Cat: one category stiffer than recommended

+10%: 10% greater than recommended power setting

+20%: 20% greater than recommended power setting

-1 Cat: one category less stiff than recommended

-2 Cat: two categories less stiff than recommended

AL: affected leg

Cat: prosthetic stiffness category

EKAM: external knee adduction moment

F_1: first peak vertical ground reaction force

F_2: second peak vertical ground reaction force

Fig.: figure

LP: low-profile

Rec: recommended

SEM: standard error of the mean

SI: symmetry index

t_c: contact time

UL: unaffected leguTTA: unilateral transtibial amputation

1. Background

People with unilateral transtibial amputation (uTTA) typically walk using a passive-elastic prosthesis that is comprised of a composite material and allows elastic energy storage and return during the stance phase of level ground walking but cannot generate net positive mechanical work. When people with uTTA walk on level ground at 1.5 m/s, a passive-elastic prosthesis returns only ½ the positive mechanical work typically generated by the soleus and gastrocnemius muscles (1). During the step-to-step transition from the affected to the unaffected leg, people with uTTA compensate for the reduced positive work done by the affected leg with an increase in the magnitude of the work done by the unaffected leg, which has been shown experimentally in people with uTTA to be related to greater loading (ground reaction forces and joint moments) of the unaffected leg compared to the biological legs of non-amputees (2). People with uTTA using a passive-elastic prosthesis walk with asymmetric biomechanics such as ground contact time (t_c) and peak ground reaction forces between their unaffected and affected legs. For example, people with uTTA using a passive-elastic prosthesis have longer t_c and greater first and second peak vertical ground reaction forces (F_1 and F_2) exhibited by their unaffected compared to their affected leg during level walking at a range of speeds from 0.75 to 1.75 m/s (3–7). In addition to walking with biomechanical asymmetry, people with uTTA exhibit a greater peak unaffected leg external knee adduction moment during the stance phase of level ground walking compared to non-amputees (4,5). Biomechanical asymmetry and greater peak external knee adduction moment may be associated with an increased risk of joint pain and osteoarthritis in their unaffected leg and back (2,8–13). Therefore, prosthetic mechanical properties and designs that better replicate biological ankle and foot function may allow people with uTTA to reduce biomechanical asymmetry, peak unaffected leg external knee adduction moment, and the risk and burden of joint pain and/or osteoarthritis.

Prosthetic foot stiffness may affect biomechanical asymmetry and peak unaffected leg external knee adduction moment of people with uTTA during walking. Passive-elastic prosthetic feet are manufactured in different stiffness categories that are recommended based on the user's bodyweight and activity or impact level (14,15). Previous studies have found that the use of a prosthesis with different stiffness can affect t_c, ground reaction forces, and lower limb joint moments of people with uTTA during level-ground walking (16–22). Previous studies suggest that either the prosthetic stiffness category does not affect t_c asymmetry or that use of a less stiff compared to a stiffer prosthesis can increase t_c asymmetry during walking at 1.3 m/s or a self-selected speed, respectively (17,21). In addition, previous studies have shown that use of a less stiff compared to a stiffer prosthesis can decrease unaffected leg F_1 or affected leg F_2 during walking at 0.7-1.5 m/s (17–20). Based on these previous studies and that people with uTTA using passive-elastic prostheses walk with greater F_1 and F_2 for their unaffected compared to their affected leg (7), use of a less stiff prosthetic foot stiffness category may not affect t_c asymmetry, but may reduce F_1 asymmetry and increase F_2 asymmetry compared to use of a stiffer prosthetic foot stiffness category.

The stiffness category of passive-elastic prosthetic feet can affect the lower limb joint moments of people with uTTA during walking (16–18,20,21,23). Stiffer compared to less stiff prosthetic feet have longer effective foot lengths or larger roll-over radii (21,24). Larger roll-over radii can reduce the required change in center of mass velocity during the step-to-step transition and the mechanical work done by the leading unaffected leg (20), which may be related to a reduction in unaffected leg external knee adduction moment (25). Slater et al. found that when people with uTTA walked at 1.3 m/s using three prostheses of varying stiffness categories, use of the less stiff compared to the stiffer prosthetic category resulted in an increase in unaffected leg external knee adduction moment (23) suggesting that decreasing the prosthetic stiffness category may increase peak unaffected leg external knee adduction moment. Ultimately, there may be a prosthetic stiffness category that minimizes peak unaffected leg external knee adduction moment of people with uTTA, which could lead to a reduction in the risk of joint pain and/or osteoarthritis.

Several stance-phase powered ankle-foot prostheses and prosthetic emulators have been designed to increase the mechanical work and power provided by the prosthesis compared to passive-elastic prostheses (26–29). The BiOM (now Ottobock Empower, Duderstadt, Germany) (Fig. 1) is a commercially available, battery-powered ankle-foot prosthesis that can generate net positive mechanical work and power during stance (26,30). Use of the BiOM compared to a passive-elastic prosthesis can affect t_c and ground reaction forces of people with uTTA during level walking (7,31,32). For example, use of the BiOM compared to a passive-elastic prosthesis can result in reduced t_c asymmetry at self-selected speeds (31), reduced unaffected leg F_1 at 1.25-1.75 m/s (7,33), and increased unaffected leg F_2 at 0.75 m/s (7). Use of the BiOM can also affect peak unaffected leg external knee adduction moment. Though Russell-Esposito and Wilken did not find a difference in first peak unaffected leg external knee adduction moment in people with uTTA when walking at ~0.99-1.50 m/s using the BiOM compared to a passive prosthesis (33), Grabowski and D'Andrea found that use of the BiOM reduced first peak unaffected leg external knee adduction moment compared to a passive prosthesis when walking at 1.50 and 1.75 m/s (32). Ultimately, based on previous studies, use of the BiOM likely affects biomechanical asymmetry and first peak unaffected leg external knee adduction moment during walking, which may result in a change in the risk and burden of joint pain and osteoarthritis (2,8,12).

The BiOM prosthesis is tuned to each user by setting the device to a biomimetic power setting (30,34,35) and this may affect the user's biomechanical asymmetry and peak external knee adduction moment. The BiOM power settings range from 0% (no power) to 100% (maximum power setting) and the manufacturer recommends tuning the power settings so that the BiOM net mechanical ankle work per step at a given walking speed matches non-amputee values within a 95% confidence interval (34,35). In experimental studies, researchers have tuned the BiOM settings based on prosthetist and manufacturer recommendations, user feedback, and/or to match non-amputee values for ankle range of motion, peak power, peak moment, and net mechanical work (7,27,31–33,36–38). However, few studies have examined the effects of tuning the BiOM to different power settings and power settings greater than recommended may offer potential metabolic and biomechanical benefits to people with uTTA during walking (24,34). For example, we found that when people with uTTA walked at 0.75-1.00 m/s using the BiOM at power settings 10% and 20% greater than the recommended setting based on biological ankle values, the magnitude of positive work done by the affected leg increased and the magnitude of negative work done by the unaffected leg decreased (24). Use of the BiOM at power settings greater than recommended likely also affects variables associated with joint pain or osteoarthritis, such as biomechanical asymmetry and unaffected leg external knee adduction moment, but these effects have not been examined previously. Based on the observed effects of the BiOM at recommended power settings (7,31–33), we expect that use of the BiOM at greater than recommended power settings would further reduce t_c asymmetry, reduce F_1 asymmetry, not affect F_2 asymmetry, and decrease first peak unaffected leg external knee adduction moment during walking. In addition, the effect of the power settings of the BiOM may depend on the stiffness category of the prosthesis that is attached to the BiOM (Fig. 1) (15) or the walking speed at which the BiOM is being used. To better inform the choice of prosthetic stiffness category and power setting based on the intended use of the device, it is useful to determine if the effects of stiffness category and power setting interact with each other or with walking speed. Ultimately, there may be a combination of prosthetic stiffness category and power setting that minimizes biomechanical asymmetry and reduces first peak unaffected leg external knee adduction moment of people with uTTA at a range of walking speeds.

To inform the design and utilization of lower limb prostheses, we determined how different prosthetic foot stiffness categories and stance-phase power settings affect variables associated with joint pain and osteoarthritis, t_c symmetry index (SI), F_1 and F_2 SI, and unaffected leg first peak external knee adduction moment, of people with uTTA walking at a range of speeds. We also determined if the effects of prosthetic foot stiffness category and stance-phase power setting interact with each other or with walking speed. First, we hypothesized that a prosthetic foot stiffness category less stiff than manufacturer-recommended would not affect t_c asymmetry but would decrease F_1 asymmetry, increase F_2 asymmetry, and increase first peak unaffected leg external knee adduction moment during walking. Second, we hypothesized that increasing the power setting of the BiOM would reduce t_c asymmetry, reduce F_1 asymmetry, not affect F_2 asymmetry, and decrease first peak unaffected leg external knee adduction moment during walking. Third, we hypothesized the null hypothesis that the effects of prosthetic foot stiffness category and power setting would not interact with each other nor with walking speed.

Figure 1. The BiOM powered ankle-foot prosthesis. The BiOM includes an Össur Low Profile (LP) Vari-flex passive-elastic prosthetic foot and uses battery-power to generate net positive mechanical work about the prosthetic ankle joint during the stance phase of walking. The BiOM includes a series elastic actuator that generates power about the ankle joint that is adjusted by a torque sensor within the prosthesis and uses positive torque feedback so that an increase in the sensed torque about the prosthetic ankle joint results in an increase in the magnitude of power delivered in the second half of the stance phase. Thus, the power provided by the BiOM can change at different walking speeds. This figure is from Tacca et al. 2024 *(24)*.

2. Methods

a. Participants

We asked 13 subjects (10M, 3F, Table 1) with unilateral transtibial amputation (uTTA) to participate. The inclusion criteria for the study were people with uTTA who have at least one year of experience using a prosthesis, are between 18-67 years old, have no known cardiovascular, pulmonary, or neurological disease or disorder other than uTTA, and are at or above a K3 Medicare Functional Classification Level (39). Functional classification levels were reported by each subject and confirmed by a certified prosthetist. All other comorbidities were reported by the subject. Subjects gave written informed consent prior to participation according to the protocol approved by the United States Department of Veteran Affairs' Human Subjects Institutional Review Board (COMIRB #19-1052). The participants reported no harms from participation in the study.

b. Protocol

This study is a subset of a larger study that not only examined the effects of prosthetic foot stiffness category and power setting on biomechanical asymmetry and external knee adduction moment but also examined the effects on step-to-step transition work, roll-over shape, lower limb joint work and power, lower limb muscle activity, metabolic rates, and user satisfaction. The protocol utilized for the larger study is briefly provided here and is also presented in Tacca et al. (24). Subjects completed an acclimation and tuning session and then three experimental sessions. The experimental sessions each occurred on a separate day and were ≥24 hours apart. Within the acclimation and tuning session, subjects were aligned with the BiOM powered prosthesis (BiOM T2, now Ottobock Empower, Duderstadt, Germany), which included a low-profile (LP) Vari-Flex (Össur, Reykjavik, Iceland) prosthetic foot with a manufacturer recommended stiffness category and with the LP Vari-Flex prosthetic foot without the BiOM (15) by a certified prosthetist. After alignment, we placed reflective markers over the joint centers of subjects' lower limbs and on the pelvis and clusters of markers on lower limb segments. For the BiOM prosthesis, we placed markers near the positions of the 1st and 5th metatarsal heads, and posterior heel based on the positions used for the unaffected leg. We placed malleoli markers on the BiOM prosthetic ankle joint (Fig. 1), which represents the sagittal plane center of rotation of the prosthesis. Then, subjects walked at 1.25 m/s on a level dual-belt force measuring treadmill (Bertec, Columbus, OH, USA) while we simultaneously measured marker trajectories at 200 Hz and ground reaction forces at 1000 Hz (Vicon Motion Systems, Oxford, United Kingdom).

We used the manufacturer-supplied computer application and different tuning parameters (stiffness, power at fast cadence, power at slow cadence, power sensitivity, power timing– fast cadence, power timing– slow cadence, stiffness duration, stance damping, cadence range, and hardstop sensitivity) to iteratively tune the BiOM prosthesis for each subject (35) until their prosthetic ankle joint range of motion, peak power, peak moment, and net mechanical work normalized to body mass including the prosthesis matched values from the biological ankle joints of non-amputees within two standard deviations of the mean (40,41). We calculated prosthetic and biological ankle joint range of motion, peak power, peak moment, and net mechanical work with a custom script (MATLAB, Mathworks Inc., Natick, MA, USA), where the foot was defined from metatarsal and malleoli markers and the shank was defined from malleoli and knee markers. Sagittal plane ankle joint range of motion was calculated from the angle between the foot and shank and sagittal plane peak ankle moment and power were calculated using inverse dynamics. We only considered ankle joint angle, moment, and power in the sagittal plane to iteratively tune the BiOM prosthesis. We calculated ankle joint net mechanical work as the integral of the ankle joint power with respect to time during a step. After a series of 30 sec trials where we iteratively tuned the BiOM to each subject, we chose the setting that best matched values from non-amputees and the unaffected leg and this setting was used as the recommended BiOM setting in the subsequent sessions (Supplementary Material: Tuning Procedure). For each participant, we were able to identify BiOM parameters that resulted in prosthetic and biological joint angle range of motion, peak moment, peak power, and net mechanical work that matched our requirements (within two standard deviations of reference data) after 2 to 12 iterations (Supplementary Material: Tuning Procedure, Fig. S1). The first session was approximately 3 hours.

The study is a repeated-measures design where each subject completed each experimental condition. Subjects completed a series of trials for each of 16 different prosthetic configurations during three experimental sessions. Each subject walked using four different passive-elastic prosthetic foot stiffness categories including the manufacturer-recommended stiffness category (Rec Cat), one category stiffer (+1 Cat), one category less stiff (-1 Cat), and two categories less stiff (-2 Cat) than recommended. While using each LP passive-elastic prosthetic foot stiffness category, subjects walked either without or with the BiOM stance-phase powered prosthesis while using three power settings: the recommended power setting (Rec), which was determined from the tuning day, and a power setting 10% greater (+10%), and 20% greater (+20%) than the recommended power setting. The subjects were blinded to the stiffness category and power setting, but not to whether they were using the BiOM or a passive-elastic prosthesis. We randomized the order of prosthetic configurations each day by selecting a passive-elastic prosthetic foot stiffness category (-2 Cat, -1 Cat, Rec Cat, +1 Cat) and then randomizing the order of the power settings (passive/without the BiOM, Rec, +10%, +20%) for the selected stiffness category. The same series of trials were then performed by each subject using each of the 16 prosthetic configurations. Subjects walked at 1.25 m/s on a level dual-belt force measuring treadmill for 5 minutes while we measured their metabolic rates. During minutes 3 and 4 of each trial, we simultaneously measured 3D reflective marker trajectories at 200 Hz and ground reaction forces at 1000 Hz for 30 sec. Subjects then walked on a level dual-belt force measuring treadmill at four speeds (0.75 m/s, 1.00 m/s, 1.5 m/s, and 1.75 m/s) for at least 30 seconds per speed while we measured 3D reflective marker trajectories and ground reaction forces.

c. Data Collection

Prior to each data collection, we placed reflective markers on the anterior superior iliac spines, posterior superior iliac spines, iliac crests, greater trochanters, lateral and medial epicondyles, lateral and medial malleoli, 1st metatarsal heads, 5th metatarsal heads, and the posterior heels of each leg. For the BiOM and LP Vari-flex prostheses, we placed markers near the positions of the 1st and 5th metatarsal heads, and posterior heel based on the positions used for the unaffected leg. We placed malleoli markers on the BiOM prosthetic ankle joint (Fig. 1), which represents the sagittal plane center of rotation. We placed malleoli markers on the LP Vari-Flex prosthesis based at the approximate location of the unaffected leg ankle joint center of rotation.

d. Data Analysis

We labeled reflective markers (Nexus, Vicon Motion Systems, Centennial CO) and filtered 3D reflective marker positions and ground reaction forces with a fourth-order, low-pass Butterworth filter with a 7 Hz cut-off (Visual3D, C-Motion, Boyds, MD, USA). We exported the ground reaction force data and used a custom script (MATLAB, Mathworks Inc., Natick, MA, USA) to calculate contact time (t_c), first peak vertical ground reaction force (F_1), and second peak vertical ground reaction force (F_2). We determined ground contact using a 20 N vertical ground reaction force threshold for each leg. F_1 was defined as the maximum vertical ground reaction force during the first half of stance phase and F_2 was defined as the maximum vertical ground reaction force during the second half of stance phase. We calculated the Symmetry Index

(SI) between the affected and unaffected legs using the formula defined by Robinson et al. (42) where 'X' refers to a biomechanical variable, 0% indicates no asymmetry, a positive value indicates a greater value for the unaffected than the affected leg, and a negative value indicates a greater value for the affected than the unaffected leg:

$$SI = \frac{X_{unaffected\ leg} - X_{affected\ leg}}{0.5(X_{unaffected\ leg} + X_{affected\ leg})} \times 100\%. \quad (1)$$

We used a rigid segment model and inverse dynamics (Visual 3D, C-Motion, Boyds, MD, USA) to determine knee joint adduction moments calculated from data collected during the experimental sessions. We used reflective marker trajectories to define lower limb segments and joints. We defined lower limb segment masses based on the Dempster regression equations for non-amputees (43) when the subject was using the BiOM because the BiOM has a similar mass and mass distribution to a biological shank and foot. When the subject was using the LP Vari-flex prosthetic foot without the BiOM, we defined segment masses based on Ferris et al. 2017 (44) to account for differences in mass between the passive-elastic prosthesis and a biological ankle and foot. We report the first peak unaffected leg external knee adduction moment during stance. To calculate the first peak unaffected leg external knee adduction moment, we used the findpeaks function in MATLAB to determine the two most prominent peaks in external knee adduction moment during stance. The first peak external knee adduction moment was the peak that occurred during the first half of the stance phase.

e. Statistical Analysis

We constructed linear mixed effects models (45) to test for the effects of prosthetic foot stiffness category, stance-phase power setting, and walking speed on t_c asymmetry, F_1 asymmetry, F_2 asymmetry, and first peak unaffected leg external knee adduction moment. The fixed effects in each linear mixed model were stiffness category (categorical; -2, -1, Rec, +1), stance-phase power setting (categorical; passive-elastic, Rec, +10%, +20%), and walking speed (numerical; speed in m/s). We included the interaction between stiffness category and power setting, and the interactions between each fixed effect and speed. For each comparison we controlled for the remaining fixed effects. We used a significance level of $p < 0.05$. The models were simplified using backward elimination where all non-significant interactions were removed (46). We set the participant as a random effect. We report unstandardized model coefficients (B) for each significant association (dependent variable = B*independent variable + intercept). B represents the change in the dependent variable related to a unit change in the independent variable. All statistical tests were done in RStudio (Boston, MA, USA). To determine an appropriate sample size, we conducted an a prior power analysis based on previous results of people with amputation walking over ground using the BiOM at recommended settings and their own passive prosthesis (32). At 1.5 m/s, Grabowski and D'Andrea found effect sizes (Cohen's d) of 1.13 and 0.86 when seven participants with uTTA walked using the BiOM compared to a passive prosthesis on unaffected leg first peak ground reaction force and external knee adduction moment, respectively (32). For our study design with four prosthetic power conditions, we estimated Cohen's f effect sizes to be 0.57 and 0.42 based on the Cohen's d from Grabowski and D'Andrea (32,47). We set α to 0.05, power to 0.90, used a repeated measures F test, and found that a sample size of 12 was needed to detect an effect size of at least 0.42 (Cohen's f) (48).

3 Results

a. Biomechanical Asymmetry

We did not detect a significant main effect of prosthetic foot stiffness category on contact time symmetry index (t_c SI) ($p > 0.31$; Fig. 2a, Table 2). However, people with uTTA walked with lower t_c SI (less asymmetry) using the BiOM compared to the passive-elastic prosthesis regardless of stiffness category ($p < 0.0001$; Fig. 2a, Table 2). t_c SI averaged across all speeds and stiffness categories was 4.72% when using the passive-elastic prosthesis and 2.66%, 2.51%, and 2.53% when using the BiOM at the recommended, +10%, and +20% power settings, respectively (Fig. 2a). The effect of power settings on t_c SI did not depend on walking speed ($p > 0.06$) nor on prosthetic foot stiffness category ($p > 0.06$) (Fig. 2a). Ultimately, all the prosthetic configurations that minimized t_c asymmetry at each walking speed included the BiOM, whereas walking using the -2 category passive-elastic prosthesis without the BiOM resulted in the greatest t_c asymmetry at 1.00, 1.25, 1.50, and 1.75 m/s (Fig. 2a).

We did not detect a significant effect of prosthetic foot stiffness category on first peak vertical ground reaction force symmetry index (F_1 SI) ($p > 0.14$; Fig. 2b, Table 2). However, we found a significant effect of prosthetic power setting on F_1 SI that depended on walking speed ($p < 0.002$; Fig. 2b, Table 2) but not on prosthetic foot stiffness category ($p > 0.10$). At 0.75 m/s, F_1 SI averaged across prosthetic foot stiffness categories was 3.32% with the passive-elastic prosthesis and 1.61%, 2.66%, and 4.99% with the BiOM at the recommended, +10%, and +20% power settings, respectively (Fig. 2b). Thus, when walking at the slowest speed, using the BiOM at recommended and +10% power settings had less F_1 asymmetry but the BiOM at +20% power setting had greater F_1 asymmetry than the passive-elastic prosthesis. At 0.75 m/s, F_1 SI was positive for all power settings, meaning that F_1 was greater for the unaffected than the affected leg. However, at 1.75 m/s, F_1 SI averaged across prosthetic foot stiffness categories was 4.59% using the passive-elastic prosthesis and -7.28%, -8.86%, and -8.94% using the BiOM at the recommended, +10%, and +20% power settings, respectively (Fig. 2b). Therefore, when walking at 1.75 m/s and using the passive-elastic prosthesis, F_1 was greater for the unaffected compared to affected leg, but when using the BiOM, F_1 was lower for the unaffected compared to affected leg, and the magnitude of F_1 asymmetry was greater compared to the passive-elastic prosthesis.

Walking using the +1 stiffness category prosthesis resulted in a lower (less asymmetry on average) second peak vertical ground reaction force symmetry index (F_2 SI) compared to the -2 stiffness category prosthesis ($p = 0.005$; Fig. 2c, Table 2); however, we did not detect a significant effect of the other stiffness categories on F_2 SI ($p > 0.24$; Fig. 2c, Table 2). Walking using the BiOM at the recommended power setting did not result in a significantly different F_2 SI compared to the passive-elastic prosthesis across all walking speeds ($p > 0.09$; Fig. 2c, Table 2). However, walking using the BiOM at the +10% and +20% power settings resulted in different F_2 SI compared to the passive-elastic prosthesis and these effects depended on walking speed ($p < 0.01$; Fig. 2c, Table 2). At 0.75 m/s, F_2 SI averaged across stiffness categories was 0.98% using the passive-elastic prosthesis and -0.33% and -3.25% using the BiOM at the +10% and +20% power settings, respectively (Fig. 2c). Thus, when walking at 0.75 m/s, increasing power settings to +20% resulted in greater F_2 asymmetry so that F_2 was lower for the unaffected compared to affected leg. However, at 1.75 m/s, F_2 SI averaged across stiffness categories was 17.19% using the passive-elastic prosthesis and 19.33% and 19.81% using the BiOM at the +10% and +20% power settings, respectively (Fig. 2c). Therefore, when walking at 1.75 m/s, increasing the power setting resulted in greater F_2 asymmetry so that F_2 was greater for the unaffected compared to affected leg.

b. First Peak Unaffected Leg External Knee Adduction Moment

When participants walked using a passive-elastic prosthesis, we did not detect a significant effect of prosthetic stiffness category ($p > 0.09$) nor an interaction between prosthetic stiffness category and walking speed on unaffected leg first peak external knee adduction moment ($p > 0.62$; Fig. 3). When participants walked using the recommended prosthetic foot stiffness category without and with the BiOM, we did not detect a significant effect of power setting on unaffected leg first peak external knee adduction moment ($p > 0.19$; Fig. 3, Table 3) nor an interaction between power setting and walking speed ($p > 0.42$; Fig. 3). Based on statistical analyses from trials with all combinations of prosthetic foot stiffness categories and power settings, we found some significant interaction effects between prosthetic foot stiffness category and power setting on unaffected leg first peak external knee adduction moment. We found that use of the BiOM at the +10% power setting resulted in a 0.06 Nm/kg lower unaffected leg first peak external knee adduction moment when the BiOM was attached to the -1 and +1 prosthetic foot stiffness categories compared to when it was attached to the -2 prosthetic foot stiffness category ($p < 0.04$; Fig. S2, Table S1).

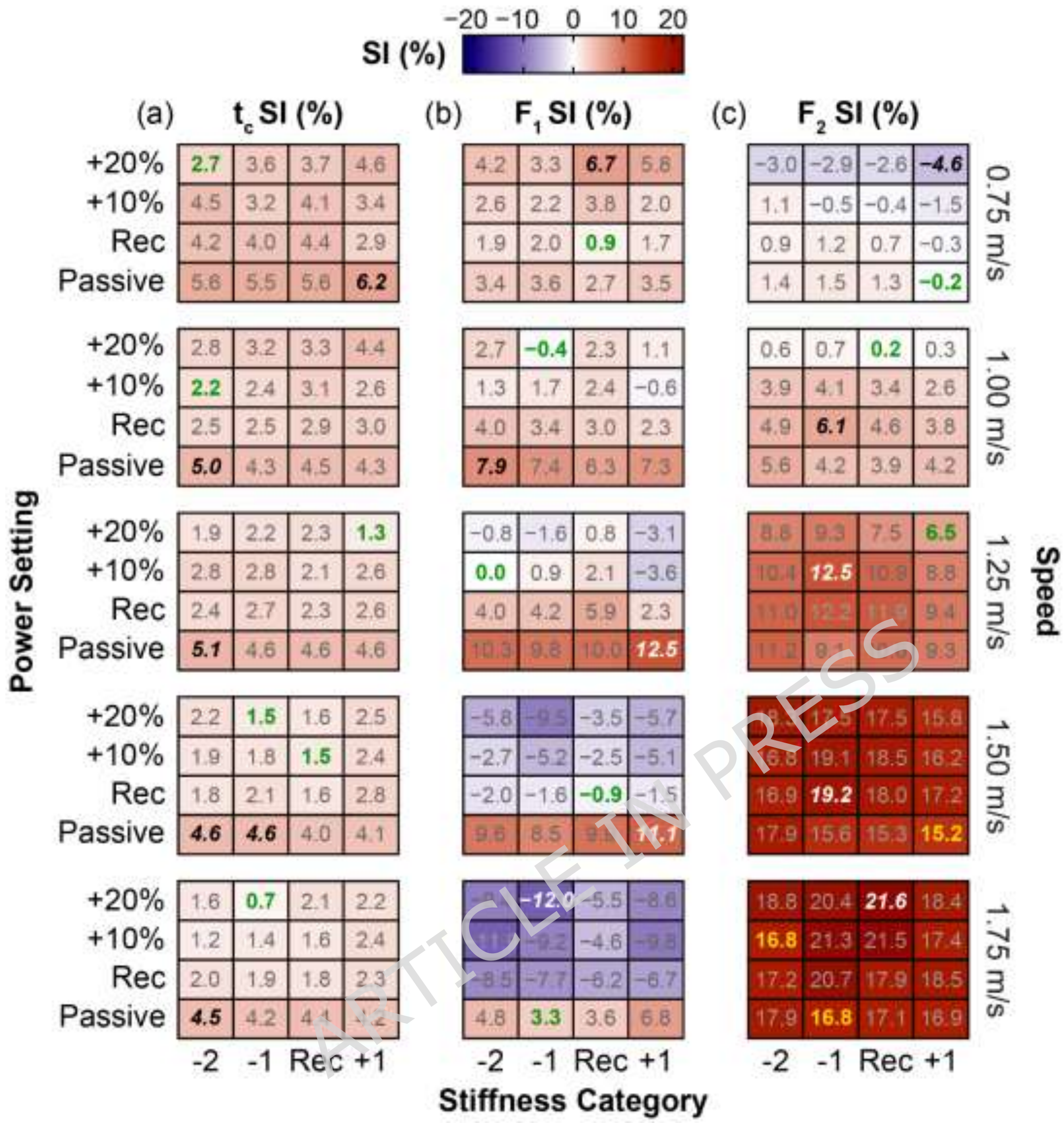

Figure 2. **Contact time and vertical ground reaction force asymmetry for stiffness category and power setting configurations.** (a) Contact time (t_c), (b) first peak vertical ground reaction force (F_1), and (c) second peak vertical ground reaction force (F_2) symmetry index (SI). The intensity and shade of the color refers to the SI (%) value. Red colors indicate that the unaffected leg value is greater than the affected leg value (positive SI), blue colors indicate that the unaffected leg value is less than the affected leg value (negative SI), and white indicates no asymmetry (SI = 0%). Darker colors refer to greater asymmetry. Each row of four-by-four grids indicates a different walking speed (m/s). Each row of small rectangles within each grid indicates a different power setting (passive/no BiOM, Rec, +10%, and +20%). Each column of small rectangles within each grid indicates a different prosthetic foot stiffness category (-2, -1, Rec, and +1). Gray numbers in each box indicate the SI values (%) for a given configuration and condition. Green (yellow in some squares to increase contrast with background) and bolded numbers are the SI values (%) for the prosthetic configuration with the least asymmetry within each grid. Black (white in some squares to increase contrast with background) and italicized

numbers are the SI values (%) for the prosthetic configuration with the greatest asymmetry within each grid.

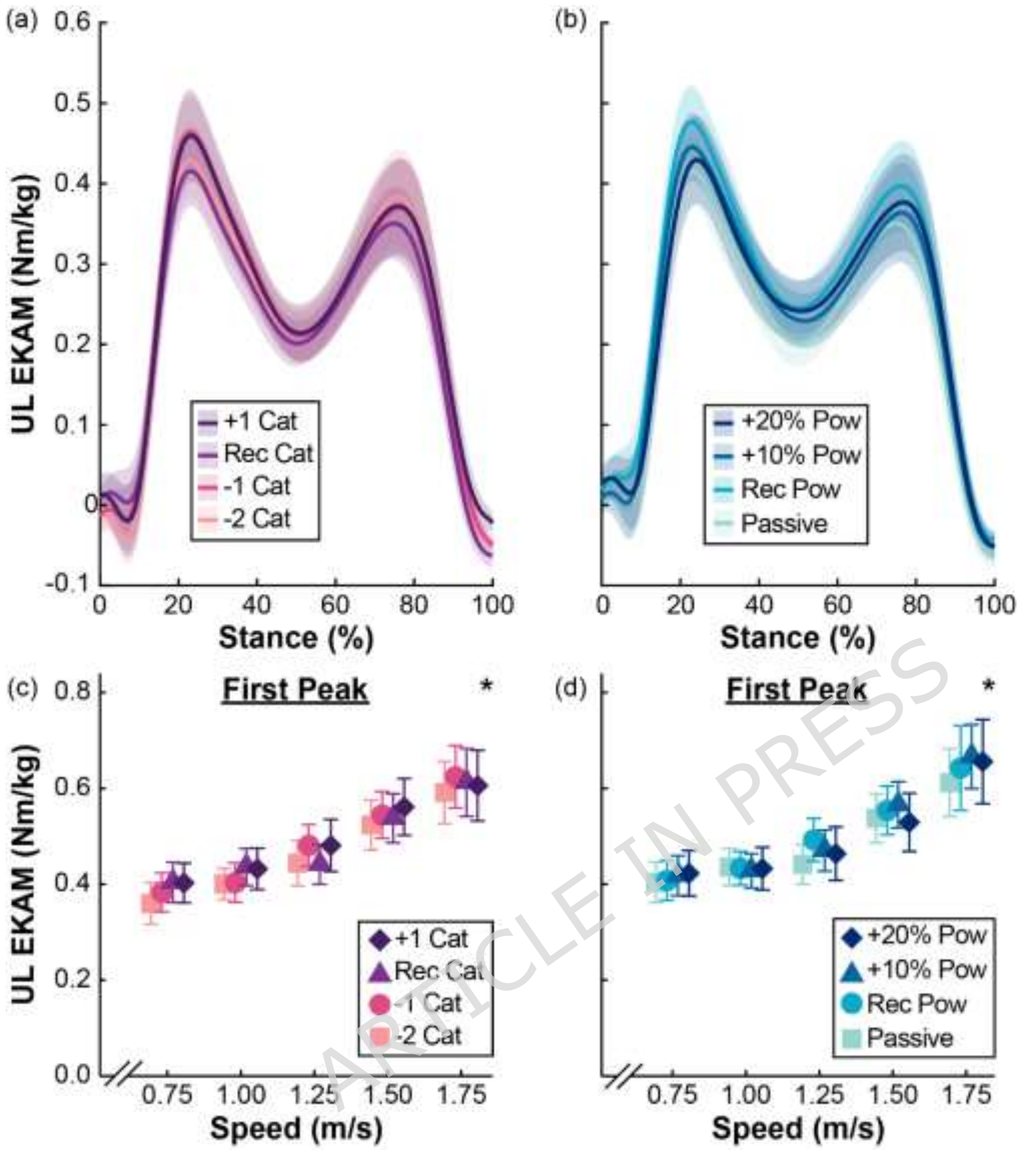

Figure 3. First peak unaffected leg external knee adduction moment for stiffness category and power setting configurations. Average first peak unaffected leg external knee adduction moment (UL EKAM) curves during the stance phase from heel strike to toe-off for all 13 participants walking at 1.25 m/s using a (a) passive-elastic prosthesis with the +1, recommended (Rec), -1, and -2 stiffness categories (Cat) and (b) the recommended prosthetic foot stiffness category without and with the BiOM at Rec, +10%, and +20% power settings. Shading is SEM. Average first peak UL EKAM for 13 participants with uTTA using (c) a passive-elastic prosthesis with the +1, recommended (Rec), -1, and -2 stiffness categories (Cat) and (d) the recommended prosthetic foot stiffness category without and with the BiOM at Rec, +10%, and +20% power settings. Colors and symbols represent different prosthetic foot stiffness categories (+1, Rec, -1, and -2) and power settings (Passive, Rec, +10%, and +20%). Error bars are SEM. * indicates a significant effect of speed.

4 Discussion

In support of our first hypothesis, we found that when subjects with uTTA used different prosthetic foot stiffness categories, there was no significant effect on t_c asymmetry and use of the stiffest passive-elastic prosthesis reduced F_2 asymmetry compared to use of the least stiff prosthesis. Our results suggest that prosthetists may not need to consider prosthetic foot stiffness category to minimize t_c asymmetry, but they should choose stiffer than recommended prosthetic foot stiffness categories to reduce F_2 asymmetry in people with uTTA. However, use of the stiffest prosthesis reduced F_2 asymmetry by only 1.22 percentage points compared to use of the least stiff category, and we did not detect significant effects of the other prosthetic stiffness categories on F_2 asymmetry. It is unclear if a 1.22 percentage point change in second peak vertical ground reaction force asymmetry is clinically meaningful to affect overall function, joint pain, or osteoarthritis. A post-hoc, pairwise comparison between the +1 and recommended prosthetic foot stiffness category showed no difference in F_2 asymmetry ($p = 0.34$). This suggests that prosthetists should only consider the effect of prosthetic foot stiffness category on F_2 asymmetry when making large changes in stiffness (e.g., a three-category change).

In contrast to our first hypothesis, we found that when subjects with uTTA used different prosthetic foot stiffness categories, there was no significant effect on F_1 asymmetry or unaffected leg first peak external knee adduction moment. Unlike our findings, previous studies of walking while people with uTTA used experimental prosthetic feet found that use of less stiff prostheses lowered first peak vertical ground reaction force and increased first peak external knee adduction moment of the unaffected leg (18–20,23). It is possible that over a wider range of prosthetic foot stiffness categories, there would be an effect of stiffness on F_1 asymmetry and first peak external knee adduction moment. For example, the difference in forefoot stiffness values for the stiffest and least stiff experimental prosthetic feet tested in a previous study was 24.6 kN/m (average forefoot stiffness for the least stiff was 35.7 kN/m and stiffest was 60.3 kN/m) (20), whereas the difference in forefoot stiffness values between the stiffest and least stiff prosthesis that we tested was on average 12.0 kN/m (average forefoot stiffness for the least stiff was 31.9 kN/m and stiffest was 43.9 kN/m) (49) for each participant. Even so, our results are consistent with a study of people with uTTA walking using a variable stiffness prosthesis at 0.75-1.50 m/s that did not find a significant effect of prosthetic stiffness on unaffected leg first peak external knee adduction moment (22). Ultimately, we found that the use of different prosthetic stiffness categories had little to no effect on t_c, F_1, and F_2 asymmetry and unaffected leg first peak external knee adduction moment, suggesting that adjusting the stiffness category of the passive-elastic prosthetic feet we tested may not affect the biomechanical variables related to joint pain and osteoarthritis risk in people with uTTA. To minimize the risk of joint pain and osteoarthritis, perhaps people with uTTA should consider use of semi-powered (2) or powered prostheses (32).

In partial support of our second hypothesis, we found that when subjects with uTTA used the BiOM with 10% and 20% greater than recommended power settings t_c asymmetry was reduced, but the effects on F_1 and F_2 asymmetry depended on walking speed. People with uTTA walked with less t_c asymmetry using the BiOM compared to a passive-elastic prosthesis, but there was no difference in t_c asymmetry between different power settings of the BiOM. Therefore, use of the BiOM may be beneficial for minimizing t_c asymmetry, but prosthetists may not need to consider the power settings of the BiOM when addressing t_c asymmetry for people with uTTA. The effects of using different BiOM power settings on F_1 and F_2 asymmetry depended on walking speed. When people with uTTA walked at 1.25 m/s the prosthetic configurations that minimized F_1 and F_2 asymmetry included the BiOM at greater than recommended power settings (+10% and +20%), which suggests that increasing BiOM power settings to greater than recommended may be beneficial for minimizing F_1 and F_2 asymmetry. However, when subjects walked at 0.75 m/s and 1.75 m/s, increasing the BiOM power settings to +10% and +20% increased F_1 and F_2 asymmetry. This result suggests that prosthetists may want to consider different power settings for different walking speeds or that future stance-phase powered prosthesis designs should include control systems that better accommodate changing walking speeds.

In contrast to our second hypothesis, we found that when subjects with uTTA used the BiOM with different power settings, there was no significant effect on unaffected leg first peak external knee adduction moment. Unlike Grabowski and D'Andrea who found that use of the BiOM reduced unaffected leg first peak external knee adduction moment compared to use of a passive-elastic prosthesis when subjects with uTTA walked at 1.50 and 1.75 m/s (32), we found no difference between use of the BiOM at any of the power settings and use of the passive-elastic prosthesis. The average unaffected leg first peak external knee adduction moment when subjects walked at 0.75–1.75 m/s using the BiOM at recommended, +10%, and +20% power settings well-matched the results from Grabowski and D'Andrea at the same walking speeds. However, the average unaffected leg first peak external knee adduction moment in the present study for trials when subjects used the passive-elastic prosthesis was lower than that of Grabowski and D'Andrea (32) so that there was no difference between using the BiOM and a passive-elastic prosthesis in the present study. Perhaps the difference in results between studies was due to the use of different passive-elastic prosthetic foot models or to potential differences between overground walking and treadmill walking (32). Overall, the present results suggest that use of the BiOM at a range of power settings does not reduce first peak unaffected leg external knee adduction moment compared to use of a passive-elastic prosthesis when people with uTTA walk at a range of speeds.

Our results can be used to inform the selection of lower limb prosthetic configurations. For example, to minimize biomechanical asymmetry for people with uTTA walking at 1.25 m/s, prosthetists should consider tuning the BiOM at power settings up to 20% greater than recommended based on biological ankle joint values. While this configuration can minimize biomechanical asymmetry, we did not detect any effects on first peak unaffected leg external knee adduction moment, a metric that has been associated with knee osteoarthritis risk and pain in the unaffected leg (8,12). So, it is unclear if use of the BiOM at power settings up to 20% greater than recommended reduces unaffected leg joint pain and osteoarthritis risk in people with uTTA. Future studies should examine the effects of prosthetic stiffness and power on other biomechanical metrics that may also be related to joint pain and osteoarthritis such as sagittal plane knee moments and sagittal and frontal plane hip moments (50,51). Our results can also be used to inform the design of controllers for powered prostheses. The BiOM utilizes positive torque feedback so that an increase in the sensed torque about the prosthetic ankle increases the magnitude of power delivered by the prosthesis and the relationship between the sensed torque and power delivered depends on the chosen power setting. At 0.75 and 1.75 m/s, increasing power setting increased F_1 and F_2 asymmetry of the participants, but at the speed for which the BiOM was tuned (1.25 m/s) increasing power setting decreased the participants' F_1 and F_2 asymmetry. To minimize biomechanical asymmetry, perhaps future prosthetic controller designs can utilize power settings that adapt to the walking speed so that the relationship between the sensed torque and power delivered is lower at speeds slower and faster than the speed at which the BiOM was tuned (i.e. 0.75 and 1.75 m/s).

While there have been many studies that have examined the effects of using a stance-phase powered prosthesis such as the BiOM compared to a passive-elastic prosthesis, their adoption in daily life has been limited as people with amputation and prosthetists consider if any potential benefits of the BiOM outweigh the added costs and complexity. Potential benefits of the BiOM outlined in previous studies have been mixed, with some studies indicating metabolic benefits (26,27,40) and improvements in biomechanical asymmetry or parameters related to joint pain and osteoarthritis (7,32,33), while other studies have indicated no benefits of BiOM use (37,38). Yet, the effect of the tuning parameters of the BiOM has rarely been studied (24,34). Our findings demonstrate that the power setting of the BiOM can have considerable effects on biomechanical asymmetry. Perhaps tuning parameters of the BiOM need to be optimized for the potential benefits to be realized.

One potential limitation to our study is the acclimation time that subjects had for each prosthetic configuration during experimental trials. Subjects completed a 5-minute walking trial at 1.25 m/s with each prosthetic configuration before completing the 30 second trials at different speeds. Therefore, participants had at least 3 minutes of acclimation with each prosthetic configuration before data were collected during each experimental session. We pseudo-randomized the trial order to account for potential training effects, but a longer acclimation period with each prosthetic configuration may have allowed subjects to better utilize the power provided by the BiOM prosthesis, which could affect asymmetry. Tuning the BiOM and determining recommended power settings after a longer period of acclimation or for each prosthetic stiffness category may also help people with uTTA to better utilize the BiOM prosthesis (40). Future studies should examine the potential effects of acclimation time on use of the BiOM in people with uTTA. While we pseudo-randomized the trial order of different prosthetic conditions, we did not randomize the order of speed conditions, which is another potential limitation of the study. Participants completed the speed conditions in order of increasing speed or in an order of 1.25, 0.75, 1.00, 1.50, and 1.75 m/s. Therefore, there may have been confounding factors related to fatigue or unequal acclimation time that could affect our interpretation of the significant interactions between power setting and walking speed on F_1 and F_2 SI.

Our results are directly relevant to prosthetists, prosthetic users, and researchers by informing the choices of prosthetic configuration and tuning parameters, but future work may be needed to increase generalizability. For this study, we chose to consider prosthetic stiffness as a categorical variable to reflect the decisions that a prosthetist, prosthetic user, or researcher has when choosing which prosthesis to use. Considering prosthetic stiffness as a categorical variable may limit the generalizability of our results because the numerical stiffness of a prosthesis for a given category varies between manufacturers and foot sizes (49). Estimating the numerical stiffness of the prosthesis heel and forefoot could allow for more direct comparisons with different prosthetic models and be of use to engineers designing new prosthetic devices. To improve generalizability and inform comparisons with other prosthetic devices, we estimated the numerical stiffness of the heel and forefoot of each prosthesis used in the study based on Tacca et al. and provide these values in the Supplementary Data Table (49). Similarly, we chose to consider power setting as a categorical variable to inform the BiOM tuning process. The power delivered by the prosthesis depends not only on the power setting but also on how the user interacts with the device. To better understand this human-device interaction, for future work we plan to calculate the mechanical power generated by the prosthesis at a given power setting. Furthermore, we chose to conduct the study using an instrumented treadmill because it allowed us to analyze the effects of walking speed and prosthetic configuration independently and collect more observations per condition. Since people with an amputation most often walk overground in daily life, use of a treadmill may limit the generalizability of our results. A previous study found that ground reaction forces and joint kinetics were different between overground and treadmill walking at the same speed, but that the differences were within the normal variability of gait parameters (52). Given that the effect of a treadmill is consistent across all experimental conditions, we expect that the use of a treadmill should not affect the interpretation of our results.

5 Conclusions

We determined the effects of different prosthetic foot stiffness categories and stance-phase power settings on contact time asymmetry, first and second peak vertical ground reaction force asymmetry, and unaffected leg first peak external knee adduction moment, of people with transtibial amputation walking at 0.75–1.75 m/s. We found that use of the stiffest compared to least stiff prosthesis reduced second peak vertical ground reaction force asymmetry independent of walking speed, but we did not detect an effect of stiffness category on the other variables. Moreover, we found that when walking at 1.25 m/s, use of the BiOM at greater than recommended power settings minimized contact time and peak vertical ground reaction force asymmetry compared to use of a passive-elastic prosthesis for people with unilateral transtibial amputation. However, at 0.75 m/s and 1.75 m/s, increasing the power setting of the BiOM can increase peak vertical ground reaction force asymmetry compared to use of a passive-elastic prosthesis for people with unilateral transtibial amputation. To minimize peak vertical ground reaction force asymmetry, prosthetists should consider different power settings for different walking speeds and/or future stance-phase powered ankle-foot prosthesis designs should include control systems that better accommodate to changing walking speeds. Furthermore, we found that use of the BiOM at recommended and up to 20% greater than recommended power settings did not affect unaffected leg first peak external knee adduction moment compared to a passive-elastic prosthesis for people with unilateral transtibial amputation. This suggests that unlike previous findings some people with unilateral transtibial amputation may not reduce peak external knee adduction moment in the unaffected leg and thus the potential associated risk of knee osteoarthritis when using the BiOM prosthesis. Our results can be used to inform the design and configuration of prostheses by demonstrating the relationships between prosthetic stiffness categories and power settings with biomechanical asymmetry and external knee adduction moment.

Declarations

a. Ethics approval and consent to participate

The participants in this study gave written informed consent to participate in the protocol. The study protocol was approved by the United States Department of Veteran Affairs' Human Subjects Institutional Review Board (COMIRB #19-1052) and all studies were conducted in accordance with local legislation and institutional requirements.

b. Consent for publication

The participants gave informed consent for publication of the data included in this article.

c. Availability of data and materials

Data is available in the text and supplementary materials. Additional requests can be directed to the corresponding author.

d. Competing interests

The authors declare that they have no competing interests.

e. Funding

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f. Authors' contributions

All authors reviewed and edited the manuscript. J.T. collected the data, developed the methodology, did the formal data analysis, prepared figures, and wrote the main manuscript text. Z.C. collected the data and developed the methodology. A.G. acquired the funding, conceptualized the experiment, provided resources and supervision, and developed the methodology.

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Figure 1. The BiOM powered ankle-foot prosthesis. The BiOM includes an Össur Low Profile (LP) Vari-flex passive-elastic prosthetic foot and uses battery-power to generate net positive mechanical work about the prosthetic ankle joint during the stance phase of walking. The BiOM includes a series elastic actuator that generates power about the ankle joint that is adjusted by a torque sensor within the prosthesis and uses positive torque feedback so that an increase in the sensed torque about the prosthetic ankle joint results in an increase in the magnitude of power delivered in the second half of the stance phase. Thus, the power provided by the BiOM can change at different walking speeds. This figure is from Tacca et al. 2024 *(24)*.

Figure 2. **Contact time and vertical ground reaction force asymmetry for stiffness category and power setting configurations.** (a) Contact time (t_c), (b) first peak vertical ground reaction force (F_1), and (c) second peak vertical ground reaction force (F_2) symmetry index (SI). The intensity and shade of the color refers to the SI (%) value. Red colors indicate that the unaffected leg value is greater than the affected leg value (positive SI), blue colors indicate that the unaffected leg value is less than the affected leg value (negative SI), and white indicates no asymmetry (SI = 0%). Darker colors refer to greater asymmetry. Each row of four-by-four grids indicates a different walking speed (m/s). Each row of small rectangles within each grid indicates a different power setting (passive/no BiOM, Rec, +10%, and +20%). Each column of small rectangles within each grid indicates a different prosthetic foot stiffness category (-2, -1, Rec, and +1). Gray numbers in each box indicate the SI values (%) for a given configuration and condition. Green (yellow in some squares to increase contrast with background) and bolded numbers are the SI values (%) for the prosthetic configuration with the least asymmetry within each grid. Black (white in some squares to increase contrast with background) and italicized numbers are the SI values (%) for the prosthetic configuration with the greatest asymmetry within each grid.```````

Figure 4. First peak unaffected leg external knee adduction moment for stiffness category and power setting configurations. Average first peak unaffected leg external knee adduction moment (UL EKAM) curves during the stance phase from heel strike to toe-off for all 13 participants walking at 1.25 m/s using a (a) passive-elastic prosthesis with the +1, recommended (Rec), -1, and -2 stiffness categories (Cat) and (b) the recommended prosthetic foot stiffness category without and with the BiOM at Rec, +10%, and +20% power settings. Shading is SEM. Average first peak UL EKAM for 13 participants with uTTA using (c) a passive-elastic prosthesis with the +1, recommended (Rec), -1, and -2 stiffness categories (Cat) and (d) the recommended prosthetic foot stiffness category without and with the BiOM at Rec, +10%, and +20% power settings. Colors and symbols represent different prosthetic foot stiffness categories (+1, Rec, -1, and -2) and power settings (Passive, Rec, +10%, and +20%). Error bars are SEM. * indicates a significant effect of speed.

Table 1. Participant characteristics: sex, age, body mass including the passive-elastic prosthetic foot, recommended prosthetic stiffness category of the LP Vari-flex prosthetic foot, the prosthetic foot size, average axial stiffness of the heel of the recommended stiffness category prosthetic foot without a shoe measured in Tacca et al. *(49)*, and average axial stiffness of the forefoot of the recommended stiffness category prosthetic foot without a shoe measured in Tacca et al. *(49)*.

Participant	*Sex*	*Age (years)*	*Mass (kg)*	*Recommended Prosthetic Stiffness Category*	*Prosthetic Size (cm)*	*Average Heel Stiffness (kN/m)*	*Average Forefoot Stiffness (kN/m)*
1	M	32	59	3	25	45.02	37.24
2	F	35	60	3	26	43.35	35.61
3	F	49	64	3	25	45.02	37.24
4	F	24	67	3	25	45.02	37.24
5	M	38	67	4	25	49.66	41.08
6	M	50	73	4	28	44.65	36.19
7	M	46	74	4	26	47.99	39.45
8	M	34	79	5	28	49.29	40.03
9	M	47	80	5	25	54.30	44.92
11	M	50	86	5	29	47.62	38.40
10	M	38	88	6	27	55.60	45.50
12	M	49	98	6	27	55.60	45.50
13	M	47	110	7	28	58.57	47.71
Average		41.5	77.4				
S.D.		8.5	15.2				

Table 2. Linear mixed model parameters for fixed effects of prosthetic foot stiffness categories, power settings, interactions of prosthetic foot stiffness categories and power settings with speed, and interactions of prosthetic foot stiffness categories with power settings on contact time symmetry index (t_c SI), first peak vertical ground reaction force symmetry index (F_1 SI), and second peak vertical ground reaction force symmetry index (F_2 SI). Linear mixed models were simplified using backward elimination where only non-significant ($p > 0.05$) interaction terms were removed *(46)*. Coefficient estimates, 95% confidence intervals for coefficient estimates (CI), coefficient standard errors (SE), t values (t), and p values (p) are listed. For the prosthetic foot stiffness categories (categorical; -2, -1, Rec, +1), the model coefficients are in reference to the -2 prosthetic foot stiffness category. For the power settings (categorical; passive, Rec, +10%, and +20%), the model coefficients are in reference to the passive-elastic prosthesis. The model coefficients for speed represent the change in the dependent variable for a 1 m/s increase in speed.

t_c SI (%)	***Estimate (B)***	***CI***	***SE***	***t***	***p***
Intercept	6.99	[5.86, 8.12]	0.57	12.32	< **0.0001**
Power Setting [Rec]	-2.04	[-2.40, -1.68]	0.19	-11.04	< **0.0001**
Power Setting [+10%]	-2.19	[-2.55, -1.83]	0.19	-11.83	< **0.0001**
Power Setting [+20%]	-2.16	[-2.53, -1.80]	0.19	-11.69	< **0.0001**
Stiffness Category [-1]	-0.15	[-0.51, 0.21]	0.19	-0.80	0.42
Stiffness Category [Rec]	-0.04	[-0.40, 0.33]	0.19	-0.19	0.85
Stiffness Category [+1]	0.19	[-0.18, 0.56]	0.19	1.01	0.31
Speed [m/s]	-1.83	[-2.20, -1.47]	0.19	-9.74	< **0.0001**

F_1 SI (%)	***Estimate (B)***	***CI***	***SE***	***t***	***p***
Intercept	3.64	[-0.89, 8.16]	2.31	1.57	0.123
Power Setting [Rec]	6.71	[2.41, 11.02]	2.21	3.04	**0.002**
Power Setting [+10%]	7.85	[3.54, 12.16]	2.21	3.55	**0.0004**
Power Setting [+20%]	11.44	[7.13, 15.75]	2.21	5.18	< **0.0001**
Stiffness Category [-1]	-0.78	[-1.96, 0.40]	0.60	-1.30	0.195
Stiffness Category [Rec]	0.90	[-0.29, 2.08]	0.61	1.48	0.139
Stiffness Category [+1]	-0.20	[-1.39, 1.00]	0.61	-0.32	0.748
Speed [m/s]	2.98	[0.64, 5.32]	1.20	2.49	**0.013**
Power Setting [Rec] * Speed [m/s]	-10.94	[-14.28, -7.60]	1.71	-6.39	< **0.0001**
Power Setting [+10%] * Speed [m/s]	-13.34	[-16.68, -9.99]	1.71	-7.79	< **0.0001**
Power Setting [+20%] * Speed [m/s]	-16.38	[-19.72, -13.03]	1.71	-9.57	< **0.0001**

F_2 SI (%)	***Estimate (B)***	***CI***	***SE***	***t***	***p***
Intercept	-12.25	[-16.02, -8.47]	1.91	-6.40	< **0.0001**
Power Setting [Rec]	-1.37	[-4.42, 1.68]	1.56	-0.88	0.381
Power Setting [+10%]	-4.05	[-7.10, -0.99]	1.56	-2.59	**0.010**
Power Setting [+20%]	-10.51	[-13.57, -7.46]	1.56	-6.72	< **0.0001**
Stiffness Category [-1]	-0.06	[-0.90, 0.78]	0.43	-0.14	0.886
Stiffness Category [Rec]	-0.50	[-1.34, 0.34]	0.43	-1.17	0.244
Stiffness Category [+1]	-1.22	[-2.06, -0.37]	0.43	-2.81	**0.005**
Speed [m/s]	18.03	[16.38, 19.69]	0.85	21.28	< **0.0001**
Power Setting [Rec] * Speed [m/s]	2.09	[-0.28, 4.45]	1.21	1.72	0.085
Power Setting [+10%] * Speed [m/s]	3.87	[1.50, 6.23]	1.21	3.19	**0.001**
Power Setting [+20%] * Speed [m/s]	7.66	[5.29, 10.03]	1.21	6.32	< **0.0001**

Table 3. (top) Linear mixed model parameters for fixed effects of prosthetic foot stiffness categories, speed, and the interaction of prosthetic foot stiffness categories with speed on first peak unaffected leg external knee adduction moment (UL EKAM) for trials without the BiOM. (bottom) Linear mixed model parameters for fixed effects of power settings, speed, and the interaction of power settings with speed on unaffected leg UL EKAM for trials without and with the BiOM attached to the recommended prosthetic foot stiffness category. Linear mixed models were simplified using backward elimination where only non-significant (p > 0.05) interaction terms were removed. Coefficient estimates, 95% confidence intervals for coefficient estimates (CI), coefficient standard errors (SE), t values (t), and p values (p) are listed. For the prosthetic foot stiffness categories (categorical; -2, -1, Rec, +1), the model coefficients are in reference to the -2 prosthetic foot stiffness category. For the power settings (categorical; passive, Rec, +10%, and +20%), the model coefficients are in reference to the passive-elastic prosthesis. The model coefficients for speed represent the change in dependent variable for a 1 m/s increase in speed.

First Peak UL EKAM (Nm/kg)	***Estimate (B)***	***CI***	***SE***	***t***	***p***
Intercept	0.18	[0.09, 0.28]	0.05	3.79	**0.001**
Stiffness Category [-1]	0.02	[-0.01, 0.06]	0.02	1.20	0.233
Stiffness Category [Rec]	0.02	[-0.02, 0.06]	0.02	1.16	0.249
Stiffness Category [+1]	0.03	[-0.00, 0.07]	0.02	1.70	0.091
Speed [m/s]	0.23	[0.19, 0.26]	0.02	11.53	< **0.0001**

First Peak UL EKAM (Nm/kg)	***Estimate (B)***	***CI***	***SE***	***t***	***p***
Intercept	0.21	[0.11, 0.31]	0.05	4.19	**0.001**
Power Setting [Rec]	0.02	[-0.01, 0.05]	0.02	1.04	0.299
Power Setting [+10%]	0.02	[-0.01, 0.05]	0.02	1.30	0.194
Power Setting [+20%]	0.01	[-0.02, 0.04]	0.02	0.73	0.468
Speed [m/s]	0.22	[0.19, 0.25]	0.02	13.19	< **0.0001**