Does Use of a Powered Ankle-foot Prosthesis Restore Whole-body Angular Momentum During Walking at Different Speeds?

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Abstract

Background Whole-body angular momentum (H) influences fall risk, is tightly regulated during walking, and is primarily controlled by muscle force generation. People

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with transtibial amputations using passive-elastic prostheses typically have greater H compared with nonamputees. **Questions/purposes** (1) Do people with unilateral transtibial amputations using passive-elastic prostheses have greater sagittal and frontal plane H ranges of motion during walking compared with nonamputees and compared with using powered prostheses? (2) Does use of powered ankle-foot prostheses result in equivalent H ranges in all planes of motion compared with nonamputees during walking as a result of normative prosthetic ankle power generation?

Methods Eight patients with a unilateral transtibial amputation and eight nonamputees walked 0.75, 1.00, 1.25, 1.50, and 1.75 m/s while we measured kinematics and ground reaction forces. We calculated H for participants using their passive-elastic prosthesis and a powered ankle-foot prosthesis and for nonamputees at each speed.

Results Patients using passive-elastic prostheses had 32% to 59% greater sagittal H ranges during the affected leg stance phase compared with nonamputees at 1.00 to 1.75 m/s ($p < 0.05$). Patients using passive-elastic prostheses had 5% and 9% greater sagittal H ranges compared with using powered prostheses at 1.25 and 1.50 m/s, respectively ($p < 0.05$). Participants using passive-elastic prostheses had 29% and 17% greater frontal H ranges at 0.75 and 1.50 m/s, respectively, compared with nonamputees ($p < 0.05$). Surprisingly, patients using powered prostheses had 26% to 50% greater sagittal H ranges during the affected leg stance phase compared with nonamputees at 1.00 to 1.75 m/s ($p < 0.05$). Patients using powered prostheses also had 26% greater frontal H range compared with nonamputees at 0.75 m/s ($p < 0.05$).

Conclusions People with a transtibial amputation may more effectively regulate H at two specific walking speeds when using powered compared with passive-elastic prostheses.

Clinical Relevance Our results support the hypothesis that an ankle-foot prosthesis capable of providing net positive work during the stance phase of walking reduces sagittal plane H; future studies are needed to validate our biomechanical findings with larger numbers of patients and should determine whether powered prostheses can decrease the risk of falls in patients with a transtibial amputation.

Introduction

The active generation of whole-body angular momentum (H) is a key strategy for maneuverability and must be regulated to maintain dynamic balance [3, 6, 7]. Humans regulate H by using muscles that contribute to the external moment generated about the center of mass [15]. When confronted with an increased risk of falling such as walking downhill compared with on level ground, nonamputees must tightly control H to maintain dynamic balance [21]. In contrast, people with a unilateral transtibial amputation using passive-elastic prostheses have greater frontal and sagittal plane H ranges during level-ground walking compared with nonamputees [20], which may explain the increased risk of falling in this population [14], particularly during the affected leg stance phase [11]. Approximately 52% of people with a lower extremity amputation incur at least one fall per year and 49% report a fear of falling [14]. Furthermore, it has been shown that when the H range is normalized to speed, height, and weight, H decreases with walking speed in both nonamputees and persons with a transtibial amputation using passive-elastic prostheses [20]. Thus, understanding the regulation of H across walking speeds may give important insight into how biomechanical factors such as ankle power and net positive work influence H.

The regulation of H is influenced by the contributions of the ankle muscles during walking [17, 20]. The rapid development of ankle torque is important in restraining H after a trip to avoid falling [18] and the ankle torque developed during walking is lower in fallers compared with nonfallers [22]. Conventional passive-elastic prosthetic feet cannot provide active ankle torque or power generation, and people with an amputation may therefore have difficulty regulating and restoring H in response to a trip. This functional loss likely contributes to the increased risk of falling in people with an amputation compared with nonamputees.

Newly developed powered ankle-foot prostheses (BiOM, Bedford, MA, USA) can provide biomimetic net positive work during the stance phase of walking [1, 8]. To our knowledge, the BiOM is the only commercially available ankle-foot prosthesis that provides power and net positive work during the stance phase of walking. Use of

this powered prosthesis has enabled people with a transtibial amputation to achieve normative metabolic costs, preferred walking speeds, and step-to-step transition work while walking over level ground across the full range of speeds [8]. However, the effects of the powered prosthesis on H across this range of walking speeds are not yet known. A prosthesis that provides normative ankle torque and power may improve the regulation of H in people with a lower extremity amputation.

We evaluated the effects of using a BiOM prototype on H during level-ground walking across a full range of speeds. We hypothesized that people with unilateral transtibial amputation using passive-elastic prostheses would have greater sagittal and frontal plane H ranges of motion (ROMs) during walking at each speed compared with nonamputees and compared with using powered prostheses. We also hypothesized that use of powered prostheses would result in equivalent H ranges in all planes of motion compared with nonamputees during walking at each speed as a result of normative prosthetic ankle power generation.

Materials and Methods

Study Participants

Before participation, eight men with a traumatic unilateral transtibial amputation and eight age-, sex-, height-, and weight-matched nonamputees gave informed written consent according to the Department of Veterans Affairs Research Service institutional review board. Subjects with an amputation were at least 2 years postamputation and were at or above a K3 Medicare Functional Classification Level, which requires that a person has the ability or potential for ambulation with variable cadence, has the ability to traverse most environmental barriers, and may have vocational, therapeutic, or exercise activity that demands prosthetic use beyond simple locomotion. All subjects had no known cardiovascular, pulmonary, or neurological disease or disorder and no additional musculoskeletal issues (Table 1). Before participation, subjects with an amputation were evaluated by a certified prosthetist who quantified and confirmed their level of amputation and disability. All subjects with an amputation used conventional passive-elastic prostheses to walk during their normal daily activities.

Powered Ankle-foot Prosthesis

During experimental trials, subjects with an amputation used a prototype of the BiOM powered ankle-foot prosthesis (Fig. 1) that has been previously described [1, 2, 8], which generates net positive prosthetic ankle work during

Table 1. Individual and average (SD) anthropometrics of male subjects with a transtibial amputation (sTTA) and of nonamputee (NA) subjects

Subject number	Age (years)	Height (m)	Mass (kg)	Years since amputation	Affected leg	Prosthesis
1	37	1.89	90.0	17	R	Flex-Foot VSP (Össur, Reykjavik, Iceland)
2	45	1.74	92.7	19	R	Venture (College Park, Warren, MI, USA)
3	50	1.74	90.7	39	L	Renegade (Freedom Innovations, Irvine, CA, USA)
4	50	1.80	106.7	31	L	Flex-Foot Re-Flex VSP (Össur)
5	60	1.84	91.0	7	R	Silhouette (Freedom Innovations)
6	39	1.94	111.0	20	R	Flex-Foot Vari-Flex EVO (Össur)
7	42	1.82	112.7	20	L	Axtion (Otto Bock, Duderstadt, Germany)
8	51	1.73	92.6	2	L	Pathfinder (WillowWood, Mt Sterling, OH, USA)
sTTA Average	47 (8)	1.81 (0.07)	98.6 (9.7)	19.4 (11.8)		
NA Average	50 (8)	1.85 (0.06)	96.5 (11.5)			

R = right; L = left.

Fig. 1 Powered ankle-foot prosthesis. The powered prosthesis performs nonconservative positive work about the prosthetic ankle by using a series-elastic actuator comprised of a brushless 200-W DC motor, ball screw transmission, and carbon-composite series leaf spring. The motor, transmission, and electronics are housed above

the prosthetic ankle, and a lithium-polymer battery provides power to the motor. The base of the prosthesis consists of a carbon-composite leaf spring, which adds compliance at the heel and forefoot. (Image printed with permission of *Proceedings of the Royal Society B*.)

the stance phase of walking and provides normative ankle mechanical power during late stance. The prosthesis uses both passive and motorized elements to emulate biological ankle and foot function (Fig. 1). There are prosthetic ankle torque sensors within the prosthesis that provide feedback to ensure that the prosthetic ankle achieves normative ankle motion, torque, and power production throughout the walking gait cycle. The powered prosthesis uses the acquired feedback in combination with biologically inspired control schemes to govern the behavior of the device, which enables the proper timing and magnitude of ankle power indicative of nonamputee level-ground walking across a wide range of speeds [5, 13].

Procedure

Before experimental sessions, each subject with an amputation completed an acclimation session with the powered

ankle-foot prosthesis. Then they completed two experimental sessions, one session using their own passive-elastic prosthesis and one session using the powered prosthesis. Each nonamputee completed one experimental session. All data were collected at the Gait and Motion Analysis Laboratory of the Providence, RI, USA, VA Medical Center.

The acclimation session with the powered ankle-foot prosthesis was at least 2 hours long. During this session, a certified prosthetist fit and aligned the powered prosthesis. Then, the stiffness and power delivery of the prosthesis were tuned so that the prosthetic ankle angle at toe-off and net positive mechanical work matched average biological ankle data [16, 23] within 2 SDs of the mean over a full range of walking speeds [8] (0.75, 1.00, 1.25, 1.50, and 1.75 m/s). The tuning parameters we saved and subsequently used during the powered prosthesis experimental session.

Before each experimental session, we placed 72 reflective markers on the legs, trunk, and arms using a modified

Cleveland Clinic marker set to measure whole body kinematics based on standard techniques [7, 9, 10, 24]. Clusters of at least four markers were placed on the feet, shanks, thighs, pelvis, trunk, upper arms, and forearms, and individual markers were placed on the medial and lateral joint centers between each segment. We matched the marker positions of the amputated leg using the prosthesis, or affected leg, to those of the unaffected leg.

During all experimental sessions, subjects walked 0.75, 1.00, 1.25, 1.50, and 1.75 m/s across a 10-m level walkway. We used an eight-camera three-dimensional motion analysis system (Qualisys Oqus, Gothenburg, Sweden) and two force platforms (Advanced Medical Technology Incorporated, Watertown, MA, USA) embedded in the walkway to simultaneously measure kinematics at 100 Hz and ground reaction forces at 1000 Hz during each trial. We analyzed three trials from each subject at each walking speed and only considered walking trials in which the participant's speed, measured as the horizontal distance per unit time of the marker placed over the seventh cervical vertebrae, was within 0.10 m/s of the target speed and where each foot made full contact with each force plate. We asked subjects to repeat the walking trials until they met these criteria.

Data Analysis

We digitized the reflective marker positions using motion tracking software (Qualisys Track Manager, Gothenburg, Sweden) and filtered the marker data with a 6-Hz Butterworth low-pass filter. We used a 12-segment model to determine segmental kinematics from marker data and to calculate H (Visual3D™; C-Motion, Inc, Germantown, MD, USA). Our model included the feet, shanks, thighs, pelvis, trunk/abdomen, lower arms, and upper arms (Fig. 2). The masses and inertial properties of the body segments were based on Dempster and Aitkens [4]. Using a custom code (MathWorks, Waltham, MA, USA), we calculated H from the model as described by Herr and Popovic [7], where the vector of whole-body angular momentum ($\vec{H}$) equals the sum of the orbital and spin components, as expressed in equation 1:

$$\vec{H} = \sum_{i=1}^{12} [(\vec{r}^i - \vec{r}_{COM}) \times m_i(\vec{v}^i - \vec{v}_{COM}) + I^i \vec{\omega}^i] \quad (1)$$

where i = segment, $\vec{r}$ = position vector in the global frame, COM = model center of mass; m_i = mass of segment i , $\vec{v}$ = linear velocity vector in the global frame, I^i = segment moment of inertia matrix, and $\vec{\omega}^i$ = angular velocity vector of segment i .

We time-normalized H to a percentage of a stride beginning with the affected leg heel strike of subjects with

an amputation and the right leg heel strike of nonamputees through the subsequent heel strike with the same leg. Then we divided H by velocity (m/s), body mass (kg), and subject height (m) to calculate H in dimensionless units. We calculated the average H range, defined as the peak-to-peak value of H in each plane of motion for all trials of each subject. We calculated sagittal H ranges for the first and second portions of a stride because there is a clear biphasic pattern in sagittal H that corresponds to the contact phases of the affected or right leg and unaffected or left leg, respectively.

Statistical Analysis

We performed Shapiro-Wilk tests to confirm that our data were normally distributed. Then we implemented paired t-tests at each walking speed to compare the H ranges in each plane of motion for subjects with an amputation using a passive-elastic prosthesis compared with using a powered prosthesis. We used unpaired t-tests at each walking speed to compare the H ranges in each plane of motion for subjects with an amputation using each prosthesis compared with nonamputees. Significant differences were further analyzed with a Tukey's honestly significant difference followup procedure and detected significance as $p \leq 0.05$. We confirmed that all significant comparisons were normally distributed for $p \leq 0.05$. We calculated Cohen's effect size (ES) and 95% confidence intervals (CIs) for statistically significant results. Post hoc, we calculated the statistical power to detect differences in each H range using an alpha of 0.05 and eight subjects and then averaged these powers over all velocities [12]. We had statistical powers of 0.89 in the sagittal H range, 0.71 in the frontal H range, and 0.57 in the transverse H range for detecting 15% differences.

Results

Effects of Passive-elastic Prostheses

Subjects with a unilateral transtibial amputation using passive-elastic prostheses had greater sagittal plane H ROMs at most speeds compared with nonamputees and compared with using a powered prosthesis and greater frontal plane H ROMs at two speeds compared with nonamputees. Specifically subjects with an amputation using passive-elastic prostheses had greater sagittal H ranges during the affected leg or right leg stance phase compared with nonamputees (Fig. 3) at 1.00 m/s (ES = 0.87; CI = 0.047–0.053, 0.035–0.041; $p < 0.001$), 1.25 m/s (ES = 0.83; CI = 0.039–0.047, 0.028–0.034; $p < 0.001$), 1.50 m/s (ES = 0.79; CI =

Fig. 2 Model of whole-body angular momentum in sagittal, frontal, and transverse planes. The anterior-posterior (AP) (F_y), vertical (F_z), and mediolateral (F_x) ground reaction forces contribute to the external moment about the center of mass (COM). External moment

arms (R_x , R_y , and R_z) indicate the distance from the COM to the center of pressure. Only the right leg contributions to the external moment are shown; however, both legs contribute in all three planes of motion.

0.034–0.042, 0.023–0.029; $p < 0.001$), and 1.75 m/s (ES = 0.89; CI = 0.032–0.038, 0.019–0.025; $p < 0.001$) (Table 2) and compared with using a powered prosthesis at 1.25 m/s (ES = 0.25; CI = 0.039–0.047, 0.037–0.045; $p = 0.032$) and 1.50 m/s (ES = 0.22; CI = 0.034–0.042, 0.031–0.039; $p = 0.032$). During the unaffected leg stance phase, subjects with an amputation using passive prostheses had greater sagittal H ranges at 1.25 m/s compared with nonamputees (ES = 0.46; CI = 0.029–0.039, 0.026–0.032; $p = 0.030$) and at 0.75 m/s (ES = 0.33; CI = 0.046–0.060, 0.042–0.054; $p = 0.031$) and 1.75 m/s (ES = 0.33; CI = 0.023–0.031, 0.019–0.027; $p = 0.017$) compared with using powered prostheses. Subjects with an amputation using passive-elastic prostheses had greater frontal H ranges (Fig. 3) at 0.75 m/s (ES = 0.57; CI = 0.034–0.046, 0.026–0.036; $p = 0.043$) and 1.50 m/s (ES = 0.37; CI = 0.017–

0.023, 0.015–0.021; $p = 0.042$) compared with nonamputees (Table 3). There were no differences in frontal H ranges between prostheses.

Effects of Powered Prostheses

Use of the powered prosthesis did not result in equivalent H ranges in all planes of motion compared with nonamputees during walking at each speed. Subjects with an amputation using powered prostheses had greater sagittal H ranges during the affected leg or right leg stance phase compared with nonamputees at 1.00 m/s (ES = 0.75; CI = 0.044–0.052, 0.035–0.041; $p = 0.001$), 1.25 m/s (ES = 0.76; CI = 0.037–0.045, 0.028–0.034; $p = 0.002$), 1.50 m/s (ES = 0.76; CI = 0.031–0.039, 0.023–0.029; $p = 0.002$),

Fig. 3 Normalized whole-body sagittal, frontal, and transverse angular momentum at each walking speed. We time-normalized angular momentum (H) to a stride beginning with the affected leg heel strike of subjects with an amputation and the right leg heel strike of nonamputees. Red curves represent the average H from subjects with an amputation using passive-elastic prostheses (Passive), blue

curves represent the average H from subjects with an amputation using powered prostheses (Powered), and black curves represent the average H from nonamputee subjects (NonAmp). Dashed lines below each graph indicate the stance phase of the affected and left legs of nonamputees, and solid lines indicate the stance phase of the unaffected and left legs from subjects with an amputation and nonamputees, respectively.

and 1.75 m/s (ES = 0.77; CI = 0.029–0.037, 0.019–0.025; $p = 0.003$). During the unaffected leg stance phase, subjects with an amputation using powered prostheses had greater sagittal H ranges compared with nonamputees at 1.25 m/s (ES = 0.31; CI = 0.027–0.037, 0.026–0.032; $p = 0.033$). Subjects with an amputation using powered prostheses had greater frontal H ranges at 0.75 m/s compared with nonamputees (ES = 0.57; CI = 0.034–0.044, 0.026–0.036; $p = 0.048$) (Table 3). There were no differences in transverse H ranges between any conditions at any speed, except

for 0.75 m/s (Fig. 3), in which subjects with an amputation had a greater transverse H range using passive-elastic compared with powered prostheses (ES = 0.11; CI = 0.016–0.026, 0.015–0.025; $p = 0.040$) (Table 4).

Discussion

Angular momentum (H) quantifies the rotational characteristics of a body in motion and is tightly regulated during

Table 2. Average (SD) range of normalized whole body sagittal (x) plane angular momentum for male subjects with a transtibial amputation using a passive and powered prosthesis and nonamputees (NonAmp)

Condition	Walking speed				
	0.75 m/s	1.00 m/s	1.25 m/s	1.50 m/s	1.75 m/s
Passive AL	0.060 (0.006)	0.050 (0.004)*	0.043 (0.005)*,†	0.038 (0.005)*,†	0.035 (0.003)*
Powered AL	0.056 (0.007)	0.048 (0.005)*	0.041 (0.005)*	0.035 (0.005)*	0.033 (0.005)*
NonAmp R	0.050 (0.009)	0.038 (0.003)	0.031 (0.003)	0.026 (0.003)	0.022 (0.003)
Passive UL	0.053 (0.008)†	0.041 (0.007)	0.034 (0.006)*	0.030 (0.007)	0.027 (0.005)†
Powered UL	0.048 (0.007)	0.039 (0.004)	0.032 (0.006)*	0.027 (0.004)	0.023 (0.005)
NonAmp L	0.050 (0.006)	0.038 (0.003)	0.029 (0.003)	0.026 (0.005)	0.024 (0.003)

* Significantly different ($p < 0.05$) from nonamputee; †significantly different ($p < 0.05$) from powered prosthesis; AL = affected leg; UL = unaffected leg stance; R = right leg; L = left leg stance.

Table 3. Average (SD) range of normalized whole body frontal (y) plane angular momentum for male subjects with a transtibial amputation using a passive and powered prosthesis and nonamputees (NonAmp)

Condition	Walking speed				
	0.75 m/s	1.00 m/s	1.25 m/s	1.50 m/s	1.75 m/s
Passive	0.040 (0.007)*	0.030 (0.004)	0.024 (0.003)	0.021 (0.004)*	0.019 (0.004)
Powered	0.039 (0.006)*	0.030 (0.004)	0.024 (0.004)	0.021 (0.004)	0.018 (0.003)
NonAmp	0.031 (0.006)	0.025 (0.004)	0.021 (0.003)	0.018 (0.004)	0.016 (0.003)

* Significantly different ($p < 0.05$) from nonamputee.

Table 4. Average (SD) range of normalized whole body transverse (z) plane angular momentum for male subjects with a transtibial amputation using a passive and powered prosthesis and nonamputees (NonAmp)

Condition	Walking speed				
	0.75 m/s	1.00 m/s	1.25 m/s	1.50 m/s	1.75 m/s
Passive	0.021 (0.006)*	0.017 (0.004)	0.016 (0.004)	0.016 (0.004)	0.014 (0.002)
Powered	0.020 (0.006)	0.017 (0.003)	0.014 (0.003)	0.015 (0.004)	0.013 (0.003)
NonAmp	0.018 (0.003)	0.016 (0.003)	0.014 (0.003)	0.013 (0.003)	0.013 (0.003)

* Significantly different ($p < 0.05$) from powered prosthesis.

human walking [7, 15, 19–21]. People with transtibial amputations using passive-elastic prostheses do not regulate H in a manner similar to nonamputees, which may explain their increased risk of falling [14]. H is primarily controlled through muscular force generation [15]. Passive-elastic prostheses store and return energy but cannot generate force nor produce net positive work [25]. Use of powered ankle-foot prostheses that mimic biological ankle work [8] may allow better regulation of H compared with use of passive-elastic prostheses. We measured and compared H over a wide range of walking speeds on level ground for people with an amputation using passive-elastic and powered prostheses and for nonamputees. Similar to previous studies, subjects with an amputation using passive-elastic prostheses had larger sagittal H ranges at most speeds and larger frontal H ranges at two speeds compared with nonamputees [20, 21]. However, subjects with an

amputation using powered prostheses also had greater sagittal H ranges at most speeds and a larger frontal H range at one speed compared with nonamputees. Use of powered prostheses allowed subjects with an amputation to better regulate sagittal H at two speeds compared with using passive prostheses.

Our study had a number of limitations. We had a relatively small sample size of high-functioning men with a traumatic amputation. A larger sample size would likely increase our ES and confirm or refute differences in H between subject groups. Because our subjects were high-functioning, it may have been difficult to detect significant differences in H between prostheses. The effects of using powered prostheses on H may be more pronounced in lower functioning individuals with greater fall risk. A broader subject pool would also improve the generalizability and applicability of our results. We tuned the

Fig. 4 Average ranges of whole-body sagittal, frontal, and transverse angular momentum at each walking speed. Average angular momentum ranges for subjects with an amputation using passive-elastic prostheses (Passive, red diamonds), subjects with an amputation using

powered prostheses (Powered, blue diamonds), and nonamputees (NonAmp, black squares) across five speeds over level ground. Error bars show $\pm$ SD.

powered prosthesis for each subject to match biological net positive ankle work within 2 SDs of the mean [8]. At 1.25 and 1.50 m/s, speeds closest to preferred, the average powered prosthetic net positive ankle work was within 1 SD of the average biological net ankle work [8]. However, if prosthetic tuning was inadequate, this may have affected H. Future research will determine how prosthetic tuning, specifically the timing and magnitude of power delivery, affects H in people with transtibial amputations. Subjects had at least 2 hours to acclimate to the powered prosthesis before experimental testing. Future research should determine how acclimation time affects adaptation and H. To our knowledge, there is only one commercially available powered ankle-foot prosthesis. New designs and further research will elucidate how prosthetic ankle power affects walking biomechanics. All of our experiments were done over level ground. Future research is planned to address how use of powered prostheses affects H on uphill and downhill slopes.

Subjects with an amputation using passive-elastic prostheses had larger sagittal H ranges during the affected leg stance phase compared with nonamputees (Fig. 4) and

consistent with previous results [20], which can be attributed to a large positive slope of H between approximately 10% and 30% of the stride (Fig. 3). H equals the time rate of change of the net external moment, a product of the ground reaction force and perpendicular distance of that force from the center of mass, or external moment arm (Fig. 2). The sagittal H trajectory results from AP and vertical ground reaction forces (Fig. 5) and external moment arms. The AP braking force acting on the leading leg contributes to a negative (forward) external moment about the center of mass. Thus, if the braking force is reduced in early stance, there will be a more positive external moment and greater positive slope of H. Subjects with an amputation using passive-elastic prostheses produced lower peak braking forces at 1.00 to 1.50 m/s compared with nonamputees (Table 5), which may partially explain the greater sagittal H ranges. At 1.00 to 1.50 m/s, the biological ankle produces net positive work [8] and likely regulates sagittal plane H [17, 20]. At 1.25 and 1.50 m/s, subjects with an amputation reduced sagittal H ranges using powered compared with passive-elastic prostheses during affected leg stance (Fig. 4), suggesting

Fig. 5 Mediolateral, AP, and vertical ground reaction forces at each walking speed. Mediolateral (Fx), AP (Fy), and vertical (Fz) ground reaction forces are normalized to a stride beginning with the affected leg heel strike of subjects with an amputation and the right leg heel strike of nonamputees. Red dashed lines represent the average forces from the affected leg and red solid lines represent the average forces

from the unaffected leg of subjects with an amputation using passive-elastic prostheses (Passive). Blue dashed lines represent the average forces from the affected leg and blue solid lines represent the average forces from the unaffected leg of subjects with an amputation using powered prostheses (Powered). Black solid lines represent the average forces from nonamputee subjects (NonAmp).

that prosthetic ankle work also affected sagittal plane H. Subjects with an amputation using powered prostheses were able to produce greater affected leg peak braking forces at 1.25 m/s and peak propulsive forces at 1.25 m/s and 1.50 m/s compared with using passive-elastic prostheses (Table 5). Subjects with an amputation using passive-elastic prostheses had larger frontal H ranges at 0.75 and 1.50 m/s compared with nonamputees and consistent with previous results [20]. Frontal H range is influenced by mediolateral and vertical ground reaction forces and external moment arms (Fig. 2). The affected leg of subjects with an amputation using passive-elastic prostheses produced lower second vertical force peaks at 0.75 to 1.25 and 1.75 m/s compared with nonamputees (Table 5), which may partially explain frontal H range differences at 0.75 m/s. At 1.50 m/s, subjects with an

amputation using passive-elastic prostheses produced greater vertical force from their unaffected leg (Table 6), which contributes to a positive slope of H after unaffected leg heel strike and a greater frontal H range relative to nonamputees. The magnitude of transverse H range was similar to previous results [3, 7, 20, 21] and was not affected by amputation or prosthesis (Table 4).

Because H equals the time rate of change of the net external moment, a prosthesis that could restore bilateral ground reaction forces, center of pressure, and center of mass position would normalize H during level-ground walking. Use of the powered prosthesis reduced differences in H ranges of subjects with an amputation at specific speeds but did not normalize H ranges compared with nonamputees (Fig. 4). Subjects with an amputation using powered prostheses had greater sagittal H ranges at 1.00 to 1.75 m/s during

Table 5. Average (SD) affected leg and nonamputee (NonAmp) peak AP (Fy) and vertical (Fz) ground reaction forces in units of body weight (BW)

	Condition	Walking speed				
		0.75 m/s	1.00 m/s	1.25 m/s	1.50 m/s	1.75 m/s
First Fy (BW)	Passive	−0.085 (0.014) [†]	−0.101 (0.012)*	−0.127 (0.013)* [†]	−0.167 (0.039)*	−0.199 (0.071)
	Powered	−0.100 (0.014)	−0.122 (0.028)	−0.161 (0.039)	−0.192 (0.032)	−0.227 (0.062)
	NonAmp	−0.112 (0.025)	−0.139 (0.019)	−0.164 (0.022)	−0.203 (0.033)	−0.242 (0.052)
Second Fy (BW)	Passive	0.098 (0.023)*	0.115 (0.020)*	0.139 (0.019)* [†]	0.155 (0.017)* [†]	0.166 (0.023)* [†]
	Powered	0.109 (0.021)	0.123 (0.014)*	0.151 (0.013)*	0.176 (0.012)*	0.187 (0.020)*
	NonAmp	0.128 (0.018)	0.153 (0.021)	0.188 (0.021)	0.228 (0.028)	0.260 (0.029)
First Fz (BW)	Passive	1.03 (0.05)*	1.03 (0.07)	1.07 (0.07)	1.13 (0.09)	1.28 (0.15)
	Powered	1.04 (0.05)*	1.03 (0.04)*	1.10 (0.09)	1.19 (0.09)	1.30 (0.17)
	NonAmp	0.99 (0.02)	0.98 (0.01)	1.05 (0.06)	1.17 (0.08)	1.25 (0.09)
Second Fz (BW)	Passive	0.97 (0.03)*	0.96 (0.04)*	0.95 (0.05)*	0.96 (0.08)	0.97 (0.08)*
	Powered	0.97 (0.02)*	0.97 (0.02)*	0.96 (0.04)*	0.97 (0.06)	0.93 (0.09)*
	NonAmp	0.99 (0.02)	1.01 (0.03)	1.03 (0.03)	1.06 (0.09)	1.09 (0.06)

* Significantly different ($p < 0.05$) from nonamputee; [†]significantly different ($p < 0.05$) from powered prosthesis.

Table 6. Average (SD) unaffected leg and nonamputee (NonAmp) peak AP (Fy) and vertical (Fz) ground reaction forces in units of body weight (BW)

	Condition	Walking speed				
		0.75 m/s	1.00 m/s	1.25 m/s	1.50 m/s	1.75 m/s
First Fy (BW)	Passive	−0.116 (0.027)	−0.145 (0.025)	−0.189 (0.037)	−0.235 (0.049)*	−0.278 (0.051)
	Powered	−0.116 (0.034)	−0.135 (0.025)	−0.180 (0.043)	−0.220 (0.045)	−0.280 (0.065)
	NonAMP	−0.110 (0.020)	−0.135 (0.023)	−0.169 (0.028)	−0.203 (0.038)	−0.240 (0.024)
Second Fy (BW)	Passive	0.136 (0.019)	0.179 (0.023)*	0.215 (0.032)*	0.272 (0.053)	0.313 (0.060)
	Powered	0.133 (0.018)	0.165 (0.018)	0.222 (0.045)	0.269 (0.066)	0.318 (0.079)
	NonAMP	0.129 (0.017)	0.157 (0.011)	0.186 (0.009)	0.232 (0.024)	0.263 (0.045)
First Fz (BW)	Passive	1.00 (0.03)	1.04 (0.05) [†]	1.13 (0.08)*	1.28 (0.10)* [†]	1.38 (0.11)*
	Powered	0.99 (0.01)	0.98 (0.03)	1.06 (0.07)	1.14 (0.12)	1.35 (0.15)
	NonAMP	0.99 (0.03)	1.00 (0.03)	1.06 (0.05)	1.16 (0.07)	1.23 (0.04)
Second Fz (BW)	Passive	0.97 (0.02)* [†]	1.00 (0.05)	1.05 (0.08)	1.14 (0.13)	1.18 (0.15)
	Powered	0.99 (0.02)	0.98 (0.03)*	1.07 (0.08)	1.16 (0.15)	1.22 (0.20)
	NonAMP	1.00 (0.02)	1.01 (0.02)	1.05 (0.04)	1.08 (0.07)	1.11 (0.06)

* Significantly different ($p < 0.05$) from nonamputee; [†]significantly different ($p < 0.05$) from powered prosthesis; UL = unaffected leg.

the affected leg stance phase compared with nonamputees (Fig. 4) and similar to previous results of passive-elastic prostheses [20], possibly as a result of inadequate tuning or accommodation time. Peak affected leg braking forces from subjects with an amputation using powered prostheses were not different from those of nonamputees (Table 5). Furthermore, subjects with an amputation using powered prostheses had greater first affected leg vertical force peaks at 0.75 and 1.00 m/s compared with nonamputees (Fig. 5). Subjects with an amputation using powered prostheses had larger frontal H ranges at 0.75 m/s compared with nonamputees. The lower second affected leg vertical force peaks at 0.75 to 1.25 and 1.75 m/s in subjects with an amputation

using powered prostheses compared with nonamputees (Table 5) may partially explain the differences in frontal H range at 0.75 m/s.

Greater H ranges in subjects with an amputation may suggest a greater risk of falling compared with nonamputees. In patients with impaired neuromuscular function, greater H ranges are more difficult to regulate given that muscles restore H during cyclic activities [15] and in response to perturbations [18]. Thus, people with a transibial amputation who lack biological ankle function are likely to have difficulty restoring H if exposed to external perturbations such as when they trip or stumble. The powered ankle-foot prosthesis provides net positive ankle work

during the stance phase of walking but does not cross the knee. The gastrocnemius spans the knee and ankle of nonamputees and contributes to the positive sagittal external moment during late stance [15]. Thus, the lack of gastrocnemius function may explain why subjects with an amputation using either prosthesis had greater sagittal H ranges compared with nonamputees. Our results suggest that people with a transtibial amputation using powered compared with passive-elastic prostheses can more effectively regulate H during level-ground walking at speeds closest to preferred. However, subjects with an amputation using the powered ankle-foot prosthesis were not able to restore H trajectories to those of nonamputees. Improvements in the H trajectory are closely linked to changes in ground reaction forces through the net external moment. Thus, use of a powered ankle-foot prosthesis that produces net positive ankle work, ground reaction forces, and external moment arms equivalent to those of nonamputees may reduce fall risk in people with transtibial amputations. However, fall risk was not specifically studied here and should be evaluated in clinical trials that include patient falls and their relationship to H. Future work will implement rehabilitation strategies for people with transtibial amputations that target the restoration of angular momentum.

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