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Whole-body angular momentum during stair walking using passive and powered lower-limb prostheses*

Nathaniel T. Pickle [a], Jason M. Wilken [b], Jennifer M. Aldridge [b], Richard R. Neptune [c], Anne K. Silverman [a,*]

[a] Department of Mechanical Engineering, Colorado School of Mines, Golden, CO 80401, USA
[b] Center for the Intrepid, Department of Orthopaedics and Rehabilitation, Brooke Army Medical Center, JBSA Ft. Sam Houston, TX 78234, USA
[c] Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX 78712, USA

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ABSTRACT

Individuals with a unilateral transtibial amputation have a greater risk of falling compared to able-bodied individuals, and falling on stairs can lead to serious injuries. Individuals with transtibial amputations have lost ankle plantarflexor muscle function, which is critical for regulating whole-body angular momentum to maintain dynamic balance. Recently, powered prostheses have been designed to provide active ankle power generation with the goal of restoring biological ankle function. However, the effects of using a powered prosthesis on the regulation of whole-body angular momentum are unknown. The purpose of this study was to use angular momentum to evaluate dynamic balance in individuals with a transtibial amputation using powered and passive prostheses relative to able-bodied individuals during stair ascent and descent. Ground reaction forces, external moment arms, and joint powers were also investigated to interpret the angular momentum results. A key result was that individuals with an amputation had a larger range of sagittal-plane angular momentum during prosthetic limb stance compared to able-bodied individuals during stair ascent. There were no significant differences in the frontal, transverse, or sagittal-plane ranges of angular momentum or maximum magnitude of the angular momentum vector between the passive and powered prostheses during stair ascent or descent. These results indicate that individuals with an amputation have altered angular momentum trajectories during stair walking compared to able-bodied individuals, which may contribute to an increased fall risk. The results also suggest that a powered prosthesis provides no distinct advantage over a passive prosthesis in maintaining dynamic balance during stair walking.

1. Introduction

Ascending and descending stairs is frequently required during the completion of activities of daily living. Walking on stairs is biomechanically demanding and requires the generation of large joint moments relative to level walking (Andriacchi et al., 1980), which necessitates greater muscle output. The increased physical requirements may explain why populations with muscle weakness or impaired balance, such as the elderly, more frequently experience serious injury due to falling on stairs than able-bodied (AB) individuals (Startzell et al., 2000). Similarly, individuals with transtibial amputation (TTA) are characterized by reduced walking ability and altered muscle function compared to their able-bodied counterparts (Silverman and Neptune, 2012). Further, individuals with TTA have impaired balance (Jayakaran et al., 2012), have a high risk and fear of falling (Miller et al., 2001), and report a decreased ability to walk on stairs (De Laat et al., 2013). The increased biomechanical demand of stair walking combined with greater fall risk in individuals with TTA make walking on stairs a significant safety concern for this patient population. Understanding how balance is maintained during stair walking is critical to ultimately reducing the incidence of falls in individuals with TTA.

Whole-body angular momentum ($\vec{H}$) is tightly regulated by AB individuals during level walking (Herr and Popovic, 2008) to maintain dynamic balance. Regulation of $\vec{H}$ is achieved through the generation of an external moment about the body center-of-mass (COM), which is a function of the ground reaction forces (GRFs) applied to the feet and the external moment arm representing the distance of the center-of-pressure (COP) relative to the body COM (Fig. 1). Muscles are the primary contributors to the net external moment about the body COM (Neptune and McGowan,

* Corresponding author. Department of Mechanical Engineering, Colorado School of Mines, 1500 Illinois Street Golden, CO 80401, USA. Tel.: +303 384 2162; fax: +303 273 3602.
E-mail address: asilverm@mines.edu (A.K. Silverman).

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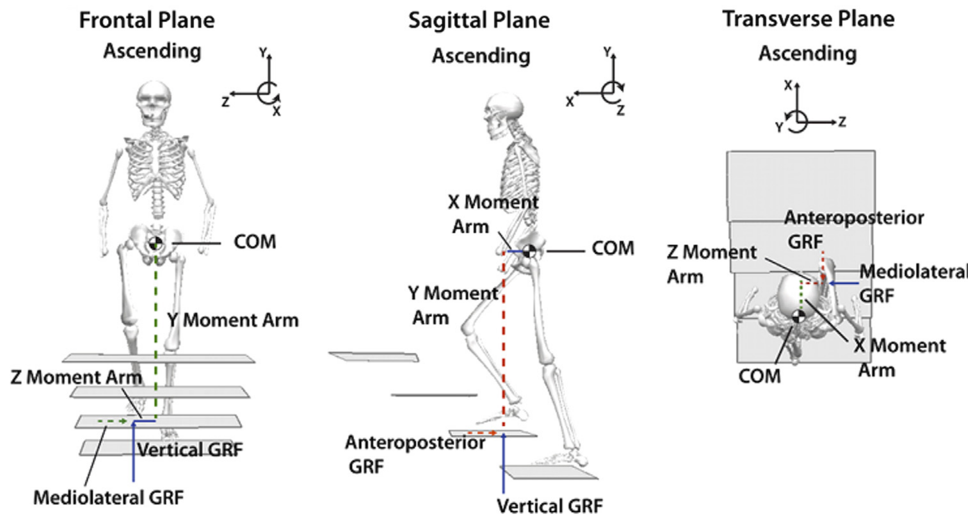

Fig. 1. Diagrams depicting the 13-segment model with external moment arms and ground reaction forces (GRFs) in each of the three anatomical planes for stair ascent. External moment arms are defined by the three dimensional distance from the body center of mass (COM) to the center of pressure of the foot.

2011) and to the net joint moments. Thus, analysis of the joint kinetics can provide insight into potential mechanisms used to regulate $\vec{H}$.

The ankle plantarflexor muscle group, which includes the soleus and gastrocnemius, is the only muscle group that contributes to the regulation of $\vec{H}$ throughout the gait cycle during level walking (Neptune and McGowan, 2011). The ankle muscles are also important for trip recovery (Pijnappels et al., 2005a, 2005b), and fallers can be distinguished from non-fallers by decreased peak ankle plantarflexion moments (Simoneau and Krebs, 2000). The importance of the plantarflexors in the regulation of $\vec{H}$ and fall prevention may partially explain why TTA have a greater risk and fear of falling relative to AB.

The ankle plantarflexors are also critical in stair walking. During stair descent, the plantarflexors are predominantly active during touch-down at the beginning of the stance phase (Spanjaard et al., 2008). However, a conventional passive energy-storage-and-return prosthesis cannot provide active ankle plantarflexion and the associated energy absorption through controlled dorsiflexion that occurs during early stance in stair descent (Sinitski et al., 2012). During stair ascent, the inability of the passive prosthesis to generate plantarflexion power during late stance (i.e., as the trailing limb) is evident in the decreased ankle moment and work output, and the intact limb generates increased plantarflexion power (Sinitski et al., 2012).

Recently, powered prostheses with motorized ankle joints have been developed (Au et al., 2007). These devices aim to restore natural gait by performing positive net work at the ankle joint over the gait cycle and have shown promising results in reducing metabolic costs and increasing preferred walking velocity during level-ground walking (Herr and Grabowski, 2012). However, these prostheses are not explicitly designed for stair walking (Eilenberg et al., 2010), and individuals with TTA have similar kinematics and kinetics when walking with both powered and passive prostheses on stairs (Aldridge et al., 2012). In addition, the influence of powered prostheses relative to passive prostheses on maintaining dynamic balance is unknown.

Therefore, the purpose of this study was to investigate the influence of passive versus powered prostheses on maintaining dynamic balance, quantified by $\vec{H}$, during stair walking. GRFs, external moment arms, and joint powers were used to help interpret differences in the $\vec{H}$ trajectories. Level walking was also analyzed as a baseline for comparison with the stair walking conditions. We hypothesized that individuals with TTA using both passive and powered prostheses would have a decreased range and maximum vector magnitude of $\vec{H}$ relative to AB during stair descent. This result was expected due to the increased fear and risk of falling in participants with TTA, possibly leading to a more conservative strategy. Passive and powered prostheses were expected to have similar performance during stair descent as neither device provides the ankle power absorption that occurs during stair descent in AB (Sinitski et al., 2012). We also hypothesized that individuals with TTA using passive and powered prostheses would have an increased range and maximum vector magnitude of $\vec{H}$ relative to AB during stair ascent, similar to previous results on level ground (Silverman and Neptune, 2011). Due to the ability of the powered prosthesis to generate net positive power at the ankle joint and more effectively emulate a biological ankle, we expected that $\vec{H}$ would be similar between individuals with TTA using a powered prosthesis and AB during stair ascent. Quantifying differences in $\vec{H}$ between individuals with TTA using different prostheses relative to AB will provide insight into the influence of powered and passive prostheses on gait mechanics and strategies used to maintain dynamic balance during stair walking.

2. Methods

Data were collected from nine participants with TTA (one female) with an average age of 30 (SD=6) years, height of 1.80 (SD=0.10) m, and mass of 94.5 (SD=7.8) kg. All participants were capable of walking independently for at least 15 consecutive minutes and were independent walkers for an average of 18.4 (SD=11.1) months prior to the study. Trials were conducted first with the participant's original passive energy-storage-and-return prosthesis and then with the BiOM (iWalk, Bedford, MA) powered prosthesis. The passive and powered trials were separated by an average of 43.4 (SD=18.1) days to allow the user to acclimate to the BiOM. Upon prosthetic fitting of the BiOM, participants were instructed in proper stair climbing technique in order to utilize the functionality of the device. In addition, the device power and timing of power were tuned to user preference using normative values. Nine AB participants (three females) were selected based on similar height and weight to participants with TTA. AB individuals had an average age of 23.2 (SD=4.6) years, height of 1.77 (SD=0.07) m, and mass of 89.6 (SD=8.5) kg. All participants provided their informed consent for the protocol approved by the institutional review board at Brooke Army Medical Center.

Whole-body kinematics were captured using a 26-camera motion capture system (Motion Analysis Corp., Santa Rosa, CA) operating at 120 Hz. A set of 57 reflective markers was used to define 13 body segments (Wilken et al., 2012), and a digitizing process was used to identify anatomical landmarks. GRF data were captured at 1200 Hz using an instrumented 16-step staircase containing two force plates (AMTI, Watertown, MA) configured in an interlaced stairway design (Della Croce and Bonato, 2007). Each step within the staircase had a rise of 18 cm and a

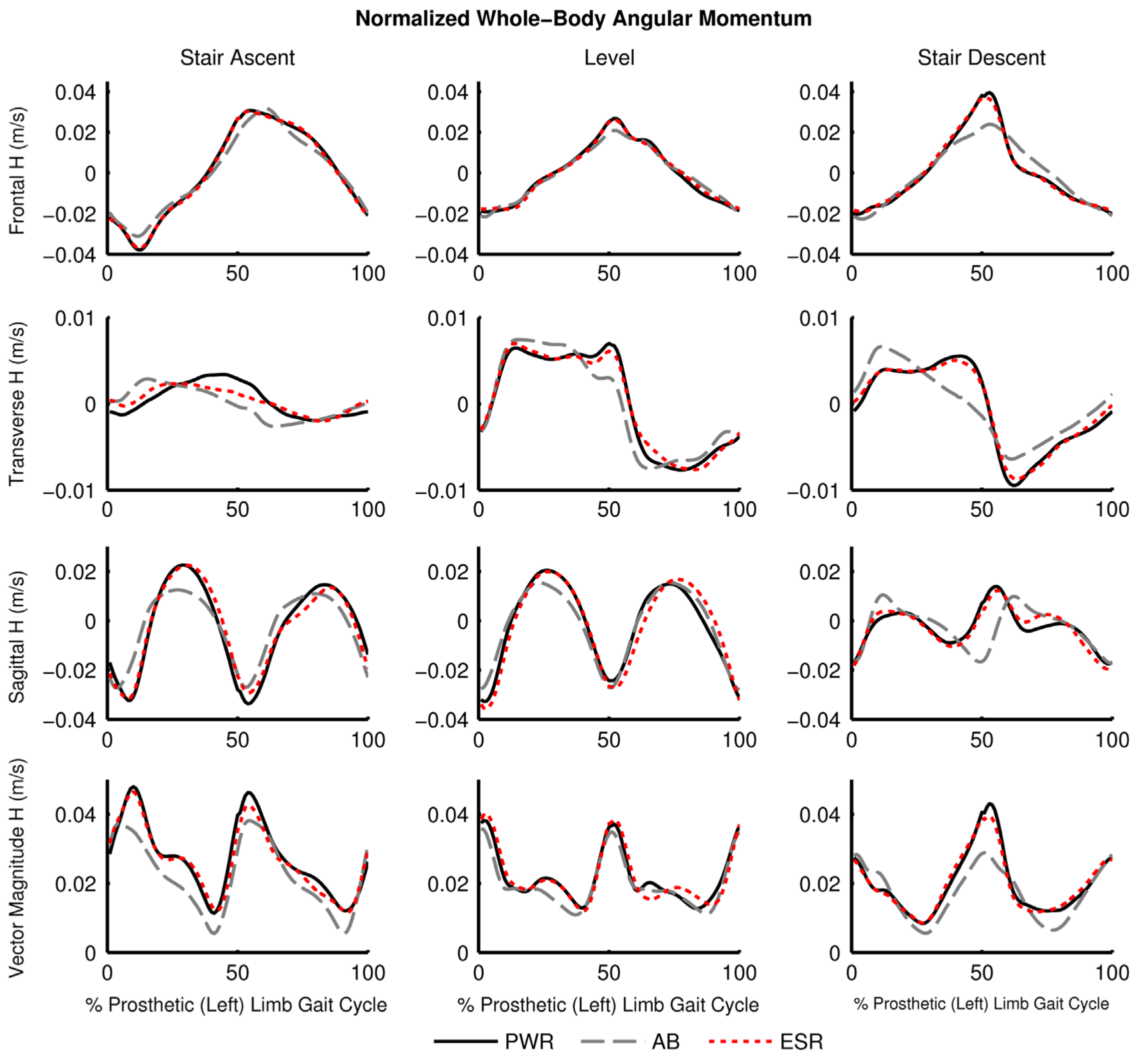

Fig. 2. Whole-body angular momentum ($\vec{H}$) for stair ascent, level walking, and stair descent in each of the three anatomical planes and total vector magnitude, normalized by subject height and mass and expressed relative to percentage of the gait cycle. Results are shown for individuals with an amputation (TTA) using the powered prosthesis (PWR, solid black line), able-bodied participants (AB, dashed gray line), and individuals with TTA using the passive prosthesis (ESR, dotted red line).

run of 26.5 cm. Stair dimensions were consistent with the stairwell in the facility where the participants received their rehabilitation. Participants navigated the staircase in a step-over-step gait, and an auditory cue was used to control cadence at 80 steps/min to eliminate confounding effects due to variation in self-selected climbing cadence. For level walking, participants walked at a fixed speed based on leg length (McAndrew et al., 2010), also controlled with an auditory cue.

Kinematic marker trajectories and analog force data were filtered using a 4th-order low pass Butterworth filter with cutoff frequencies at 6 Hz and 50 Hz, respectively. Kinematic and GRF data were combined in Visual3D (C-Motion, Inc., Rockville, MD) to compute joint powers using an inverse dynamics approach. A 13-segment model was used in the inverse dynamics and $\vec{H}$ calculations. The masses of individual segments were determined as a percentage of total body mass (Dempster and Aitkens, 1995) for all participants. $\vec{H}$ was calculated as

$$\vec{H} = \sum_{i=1}^{n} \left[\left(\vec{r}_i^{\,COM} - \vec{r}_{body}^{\,COM} \right) \times m_i \left(\vec{v}_i^{\,COM} - \vec{v}_{body}^{\,COM} \right) + I_i \vec{\omega}_i \right]$$

where n is the number of segments; $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$, and $\vec{\omega}_i$ are, respectively, the position, velocity, and angular velocity of the ith segment; $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are, respectively, the position and velocity of the whole-body COM; and m_i and I_i are the mass and inertia matrix of the ith segment. $\vec{H}$ was normalized by body height and mass and expressed as a percentage of the left or prosthetic limb gait cycle for the AB and TTA groups, respectively.

The ranges (peak-to-peak values) of $\vec{H}$ in all three anatomical planes were calculated and compared statistically using the R Statistical Computing Software, v. 2.15.1 (R Core Team, 2012). The peak values of the maximum magnitude of the $\vec{H}$ vector (i.e., maximum $|\vec{H}|$), defined as

$$|\vec{H}| = \sqrt{(H_{frontal})^2 + (H_{transverse})^2 + (H_{sagittal})^2},$$

along with 3D GRFs, external moment arms and joint powers were similarly compared. Two-factor, mixed model ANOVAs (Lawrence, 2012) were performed on these data to determine significant subject group (first factor) and walking condition (second factor) main effects. The ANOVAs compared individuals with TTA using the powered versus passive prosthesis, individuals with TTA using the powered prosthesis versus AB, and individuals with TTA using the passive prosthesis versus AB across the three walking conditions (ascent, descent, level) as repeated trials with the same participants. If data were found to violate the assumption of a normal distribution, a non-parametric version of the repeated measures ANOVA was used (Noguchi et al., 2012). When significant main effects were found, pairwise comparisons were performed. For normally distributed data, the unpaired t-test was used for parametric comparisons between groups and the paired t-test was used for parametric comparisons within groups, accounting for unequal variances if necessary. For non-normally distributed data, the Wilcoxon rank sum test was performed for non-parametric comparisons between groups, and the Wilcoxon signed rank test was used for non-parametric comparisons within groups. Pairwise comparisons included a Holm adjustment for multiple comparisons ($\alpha = 0.05$).

3. Results

3.1. Frontal plane

Significant condition main effects were observed for the range of frontal-plane $\vec{H}$ for all three ANOVAs (powered prosthesis vs. AB, passive prosthesis vs. AB, powered prosthesis vs. passive prosthesis). Group effects were only significant for the powered

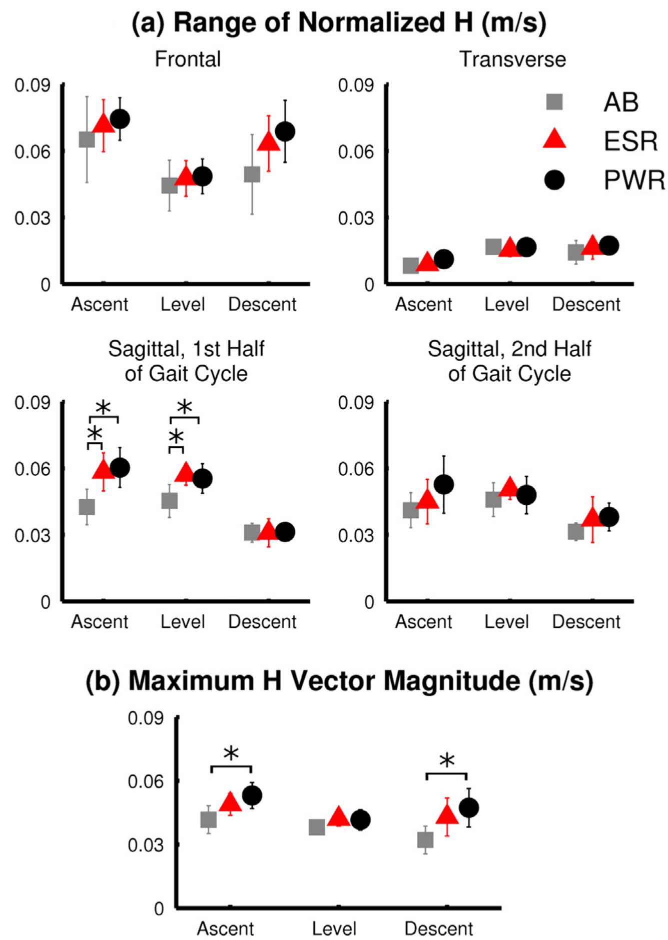

Fig. 3. Average and standard deviation of the range of whole-body angular momentum ($\vec{H}$) in each of the three anatomical planes (*a*) and vector magnitude (*b*), normalized by subject height and mass. In the sagittal plane, the range of $\vec{H}$ was calculated separately for the first half (0–50%) and second half (51–100%) of the gait cycle. Results are shown for individuals with an amputation (TTA) using the powered prosthesis (PWR, black circles), able-bodied participants (AB, gray squares) and individuals with TTA using the passive prosthesis (ESR, red triangles). Significant differences between subject groups are indicated by "*".

vs. passive prosthesis ANOVA (Figs. 2 and 3, Table 1). However, pairwise comparisons resulted in no significant differences between groups following the Holm adjustment. All subject groups had an increased range of frontal-plane $\vec{H}$ during stair ascent relative to level walking, and only individuals with TTA had an increased range of frontal-plane $\vec{H}$ during stair descent relative to level.

There were several significant differences between groups in the moment arms and GRFs in both the vertical and M/L directions (Tables 2 and 3, Figs. 4 and 5), which contribute to the external moment in the frontal plane (Fig. 1). There were also significant differences in nearly all of the peak external moment arms and GRFs in the vertical and M/L directions during stair ascent and descent relative to level walking (Tables 2 and 3).

3.2. Transverse plane

A significant condition main effect was found for the range of transverse-plane $\vec{H}$ in all ANOVAs, but the group main effect was only significant in the powered vs. passive prosthesis ANOVA (Figs. 2 and 3, Table 1). However, similar to $\vec{H}$ in the frontal plane, the pairwise comparisons did not result in significant differences between the prostheses. All groups had a reduced range of transverse-plane $\vec{H}$ during stair ascent relative to level. There were significant differences between groups and conditions in the A/P and M/L GRFs and external moment arms (Figs. 4 and 5), which contribute to the external moment in the transverse plane (Fig. 1).

3.3. Sagittal plane

Significant group and condition main effects were found for the range of sagittal-plane $\vec{H}$ (Figs. 2 and 3, Table 1). During stair ascent, individuals with TTA using either type of prosthesis had a significantly increased range of sagittal-plane $\vec{H}$ relative to AB during the first half of the gait cycle (prosthetic limb stance), but no significant difference was observed in the range during the second half of the gait cycle (intact limb stance).

Also during stair ascent, individuals with TTA using the passive prosthesis had reduced peak vertical GRFs in the prosthetic limb relative to AB. Individuals with TTA using either prosthesis had increased peak vertical GRFs in the intact limb relative to AB (Fig. 5). Peak sagittal-plane (flexion/extension) joint powers had several significant group and condition main effects. Pairwise comparisons showed that individuals with TTA had reduced maximum knee power generation and increased maximum hip power generation in the prosthetic limb relative to AB, as well as greater ankle power generation in the intact limb (Table 2, Fig. 6) regardless of the type of prosthesis used. Peak ankle power generation in individuals with TTA using the passive prosthesis was significantly less than AB, but for the powered prosthesis, there was a significant increase relative to the passive prosthesis and no significant difference relative to AB.

During stair descent, all groups had a significantly reduced range of sagittal-plane $\vec{H}$ relative to level walking, and there were no significant differences in the range of sagittal-plane $\vec{H}$ between subject groups. In the prosthetic limb, individuals with TTA using passive and powered prostheses had a reduced anterior moment arm, increased second peak lateral moment arm (Table 2, Fig. 4), reduced first peak vertical GRF (Table 3, Fig. 5), and reduced ankle plantarflexion power absorption relative to AB (Table 4, Fig. 6). In the intact limb, individuals with TTA using both types of prostheses had an increased peak posterior GRF, increased first and second peak vertical GRF, and increased ankle power absorption relative to AB.

3.4. Vector magnitude $|\vec{H}|$

Significant group and condition main effects were found for the overall maximum magnitude of the $\vec{H}$ vector (Figs. 2 and 3, Table 1). Individuals with TTA using the powered prosthesis had increased maximum $|\vec{H}|$ during both ascent and descent relative to AB individuals, but when using the passive prosthesis there were no significant differences relative to AB. Individuals with TTA using either prosthesis had significantly greater maximum $|\vec{H}|$ during stair ascent than level walking, but AB individuals only showed significantly greater maximum $|\vec{H}|$ during ascent relative to descent.

4. Discussion

The purpose of this study was to investigate dynamic balance in individuals with TTA using both passive and powered prostheses relative to AB during stair walking. We expected that individuals with TTA would have an increased range and magnitude of $\vec{H}$ during stair ascent relative to AB but a decreased range and magnitude of $\vec{H}$ during stair descent relative to AB, with

Table 1
Significant differences in the range of whole-body angular momentum ($\vec{H}$) in the three anatomical planes and maximum magnitude of the $\vec{H}$ vector (i.e., maximum $|\vec{H}|$). Significance values are shown for the main group and condition effects. Main effects < 0.0005 are listed as 0.000 and "-" denotes main effects that were not significant. Mean (SD) values of the range of $\vec{H}$ and maximum magnitude of the $\vec{H}$ vector are also given for the different groups and conditions, and significant pairwise results are indicated. PWR refers to individuals with an amputation using the powered prosthesis and ESR refers to use of the passive energy-storage-and-return prosthesis. Pairwise comparisons were performed after significant main effects were found.

Range of $\vec{H}$							**Maximum $\vec{H}$**
			Frontal	**Transverse**	**Sagittal**		**Vector magnitude**
					1st half	2nd half	
			p-values				
ANOVA Main effects for Range of $\vec{H}$	**PWR vs ESR**	Group	0.020	0.001	–	–	–
		Condition	0.000	0.002	0.000	0.006	0.000
	PWR vs AB	Group	–	–	0.002	0.023	0.0007
		Condition	0.000	0.000	0.000	0.000	0.000
	ESR vs AB	Group	–	–	0.002	–	0.000
		Condition	0.000	0.000	0.000	0.000	0.000
			Mean (SD) Range of $\vec{H}$				**Mean (SD)**
Pairwise comparisons	**Stair Ascent**	PWR	0.074 (0.010)*	0.011 (0.003)*	0.060 (0.009)$^{\dagger o}$	0.053 (0.013)	0.053 (0.006)^{*o}
		ESR	0.071 (0.012)*	0.009 (s0.002)*†	0.058 (0.009)$^{\dagger o}$	0.045 (0.010)	0.049 (0.005)*
		AB	0.065 (0.019)*†	0.008 (0.003)*†	0.043 (0.008)†	0.041 (0.008)†	0.042 (0.007)†
	Level	PWR	0.049 (0.008)	0.017 (0.003)	0.055 (0.007)^{o}	0.048 (0.008)	0.042 (0.005)
		ESR	0.048 (0.008)	0.015 (0.003)	0.057 (0.005)^{o}	0.051 (0.005)	0.042 (0.003)
		AB	0.044 (0.012)	0.017 (0.003)	0.045 (0.007)	0.046 (0.008)	0.038 (0.003)
	Stair Descent	PWR	0.069 (0.014)*	0.017 (0.004)	0.031 (0.004)*†	0.038 (0.006)	0.047 (0.009)^{o}
		ESR	0.063 (0.013)*	0.016 (0.005)†	0.031 (0.006)*†	0.037 (0.010)	0.043 (0.009)
		AB	0.049 (0.018)†	0.014 (0.005)†	0.031 (0.004)*†	0.031 (0.004)*†	0.032 (0.007)†

* Significant differences relative to level walking.
[a] Significant differences relative to the passive prosthesis.
[o] Significant differences relative to able-bodied (AB) individuals for the same walking condition.
† Significant differences between stair ascent and descent.

Table 2
Mean (SD) values for peak external moment arms in the three anatomical planes, normalized to percentage of body height (%BH). PWR refers to individuals with an amputation using the powered prosthesis, and ESR refers to use of the passive energy-storage-and-return prosthesis. Pairwise comparisons were performed after significant main effects were found.

Mean (SD) Values for Peak External Moment Arms (%BH)							
			Max A/P	Min A/P	Min. Vertical	1st Peak M/L	2nd Peak M/L
Stair Ascent	**Prosthetic**	ESR	14.0 (1.4)$^{*\dagger o}$	−6.2 (1.0)*	−71.8 (0.5)†	9.0 (2.0)*	7.1 (1.8)*†
		PWR	13.0 (1.3)$^{*\dagger o}$	−6.7 (1.2)*	−72.6 (1.3)*†	8.5 (1.7)*	6.9 (1.5)*†
	Intact	ESR	11.1 (1.5)*	−7.0 (0.9)*	−74.8 (0.8)$^{*\dagger o}$	8.5 (2.2)*	8.0 (1.7)
		PWR	11.0 (1.5)*	−7.0 (1.3)*	−74.1 (0.8)*†	8.6 (2.4)*	7.2 (1.1)
	Average	AB	10.6 (1.1)*†	−6.5 (0.9)*	−73.2 (0.7)*†	7.7 (1.7)*†	6.3 (1.2)*
Level	**Prosthetic**	ESR	18.4 (2.0)	−16.7 (1.7)^{o}	−70.3 (6.5)	4.5 (1.4)	5.7 (1.1)
		PWR	17.2 (1.0)	−19.0 (1.3)^{a}	−68.6 (1.4)	4.4 (0.8)	5.2 (0.7)
	Intact	ESR	16.5 (1.1)	−19.7 (1.2)	−68.4 (0.7)	5.4 (1.0)	6.4 (0.6)
		PWR	16.1 (1.3)	−20.3 (1.4)	−68.4 (1.3)	5.8 (0.6)	6.5 (0.8)
	Average	AB	16.6 (1.3)	−19.4 (1.4)	−67.5 (0.8)	4.6 (0.7)	5.3 (1.1)
Stair Descent	**Prosthetic**	ESR	4.6 (1.0)$^{*\dagger o}$	−7.2 (1.4)*	−65.1 (0.8)†	6.7 (1.3)*	10.1 (0.9)$^{*\dagger o}$
		PWR	5.4 (1.2)$^{*\dagger o}$	−8.0 (1.4)*	−64.7 (1.1)*†	6.5 (0.8)*	9.7 (1.2)$^{*\dagger o}$
	Intact	ESR	12.6 (1.5)*	−7.2 (1.0)*	−68.2 (0.9)$^{\dagger o}$	9.8 (1.5)*	7.4 (1.4)
		PWR	12.0 (0.7)*	−7.0 (0.9)*	−68.1 (1.1)$^{\dagger o}$	7.5 (1.4)*	7.8 (2.0)
	Average	AB	13.0 (1.1)*†	−7.8 (0.6)*	−66.7 (0.9)*†	9.3 (1.2)*†	6.6 (1.4)*

* Significant differences relative to level walking.
[a] Significant differences relative to the passive prosthesis.
[o] Significant differences relative to able-bodied individuals (AB) for the same walking condition.
† Significant differences between stair ascent and descent.

differences between passive and powered prostheses during stair ascent. Our hypotheses were not supported in the frontal and transverse planes as there were no differences between groups, although the range of $\vec{H}$ was different between stair ascent, level walking, and stair descent for all groups (Table 1, Fig. 3).

Our hypotheses were partially supported in the sagittal plane and total vector magnitude of $\vec{H}$. As expected, individuals with TTA using the passive prosthesis had a larger range of sagittal-plane $\vec{H}$ during the first half of the gait cycle and maximum $|\vec{H}|$ relative to AB for stair ascent. These results are similar to the

Table 3
Mean (SD) values for peak ground reaction forces in the three anatomical planes, normalized to percentage of body weight (%BW). PWR refers to individuals with an amputation using the powered prosthesis and ESR refers to use of the passive energy-storage-and-return prosthesis. Pairwise comparisons were performed after significant main effects were found.

Mean (SD) Values for Peak Ground Reaction Forces (%BW)

			Max A/P	Min A/P	First Peak Vertical	Second Peak Vertical	First Peak M/L	Second Peak M/L
Stair Ascent	**Prosthetic**	ESR	5.2 (1.3)[*†]	−7.6 (1.5)[*]	95.7 (3.6)[*o†]	95.5 (3.8)[*†o]	7.5 (2.5)	6.0 (1.8)[†]
		PWR	5.2 (0.8)[*†]	−6.9 (1.1)[*]	97.2 (4.5)[*]	96.9 (8.0)	7.4 (2.3)	6.3 (1.6)
	Intact	ESR	7.3 (1.8)[*]	−9.1 (0.6)[*†o]	118.0 (6.3)[*†o]	118.4 (9.0)[*†o]	9.2 (1.6)[o]	8.1 (2.3)[o]
		PWR	7.4 (2.0)[*†]	−9.9 (1.9)[*†]	112.3 (6.8)[*†o]	125.8 (9.8)[*†oa]	8.0 (2.4)	7.0 (1.6)
	Average	AB	6.3 (1.2)[*†]	−7.4 (1.8)[*†]	100.7 (3.1)[*†]	105.0 (4.1)[†]	6.1 (1.1)[†]	5.4 (0.9)[†]
Level	**Prosthetic**	ESR	13.5 (1.1)[o]	−13.7 (1.7)[o]	107.8 (8.1)	105.4 (6.9)	5.8 (1.3)	6.2 (0.9)
		PWR	18.1 (2.2)[a]	−17.1 (1.7)[a]	106.4 (6.0)	102.3 (3.7)	6.5 (1.0)	5.9 (1.4)
	Intact	ESR	20.1 (2.7)	−16.2 (2.8)	107.7 (5.0)	103.2 (6.0)	7.5 (1.1)	7.1 (0.8)
		PWR	20.5 (2.7)	−17.1 (2.7)	102.5 (5.0)[o]	104.1 (5.2)	7.0 (1.1)	7.2 (1.2)
	Average	AB	19.8 (1.7)	−18.4 (3.6)	109.2 (4.7)	106.6 (3.2)	6.7 (1.3)	6.4 (1.7)
Stair Descent	**Prosthetic**	ESR	9.4 (0.6)[*†]	−7.5 (2.2)[*o]	109.3 (7.1)[†o]	94.8 (6.9)[*]	7.4 (1.5)[*]	7.9 (1.7)[†]
		PWR	10.9 (1.6)[*†]	−8.4 (2.2)[*]	102.6 (11.3)[o]	89.9 (8.2)[*]	8.7 (1.4)[*]	7.2 (1.2)
	Intact	ESR	9.9 (2.4)[*]	−17.5 (2.4)[†o]	150.7 (20.7)[*†o]	107.6 (6.0)[†o]	10.3 (1.5)[*]	8.8 (1.8)
		PWR	10.6 (1.5)[*†]	−18.0 (2.5)[†o]	164.0 (29.6)[*†o]	109.4 (6.5)[†o]	10.8 (3.7)	9.2 (1.7)
	Average	AB	9.6 (1.6)[*†]	−11.6 (1.4)[*†]	123.3 (8.2)[*†]	93.4 (3.5)[*†]	8.6 (1.2)[*†]	7.6 (1.3)[†]

[*] Significant differences relative to level walking.
[a] Significant differences relative to the passive prosthesis.
[o] Significant differences relative to able-bodied individuals (AB) for the same walking condition.
[†] Significant differences between stair ascent and descent.

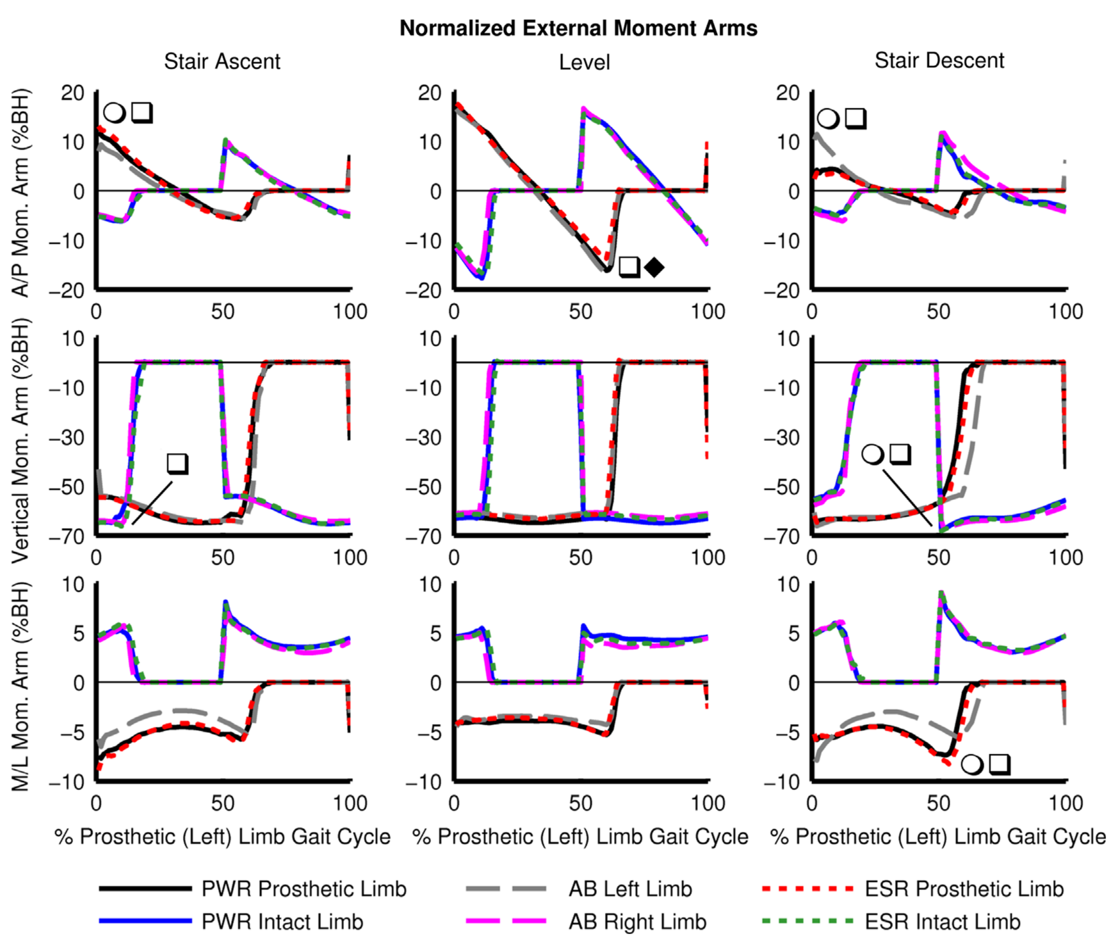

Fig. 4. External moment arms in each of the three anatomical planes during stair ascent, level walking, and stair descent, normalized to percent of body height (%BH) and expressed relative to percentage of the gait cycle. Results are shown for individuals with transtibial amputation (TTA) using a powered prosthesis (PWR, solid lines), able-bodied participants (AB, dashed lines), and individuals with TTA using a passive prosthesis (ESR, dotted lines). Significant differences in peak values are indicated between PWR and AB (❍), ESR and AB (❑), and PWR and ESR (♦). Additional pairwise comparison results are given in Table 2.

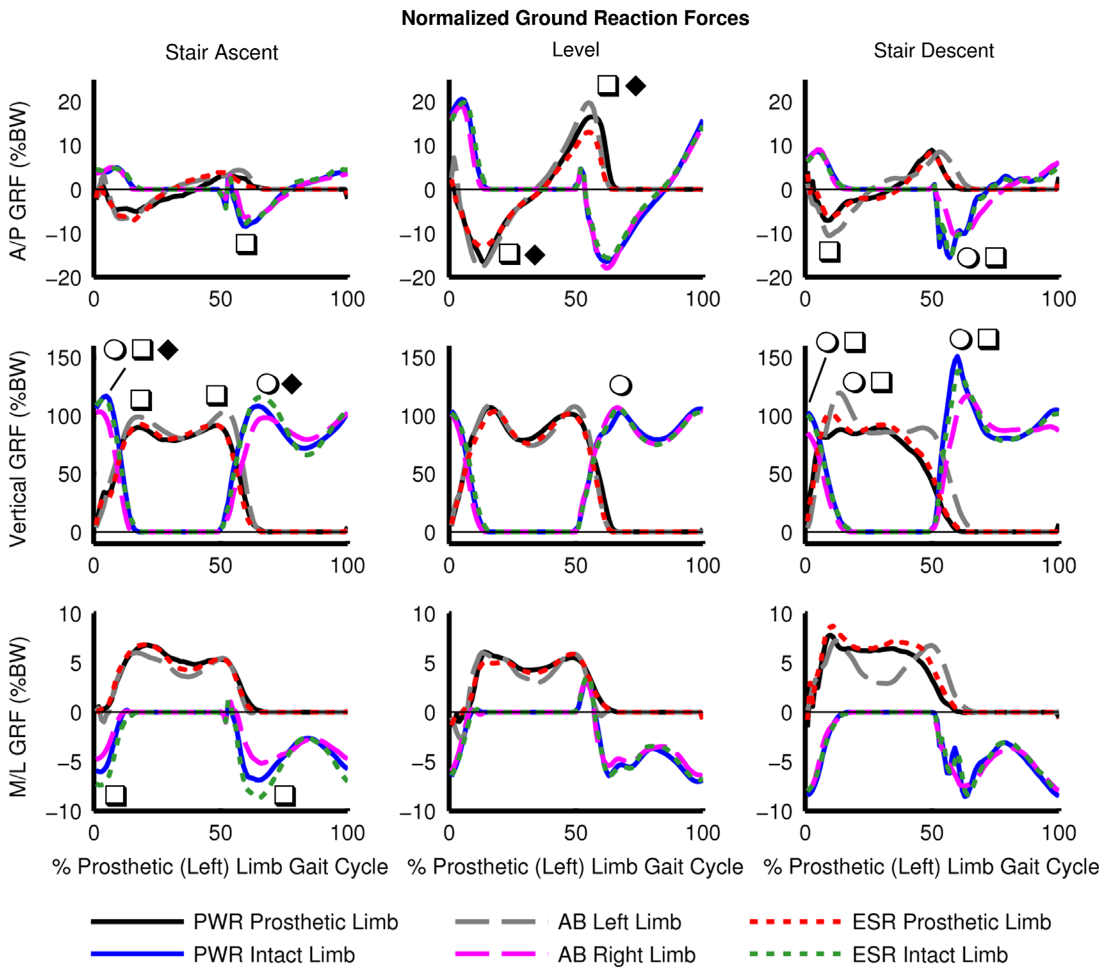

Fig. 5. Ground reaction forces (GRFs) in each of the three anatomical planes during stair ascent, level walking, and stair descent, normalized to percent of body weight (%BW) and expressed relative to percentage of the gait cycle. Results are shown for individuals with transtibial amputation (TTA) using a powered prosthesis (PWR, solid lines), able-bodied participants (AB, dashed lines), and individuals with TTA using a passive prosthesis (ESR, dotted lines). Significant differences in peak values are indicated between PWR and AB (❍), ESR and AB (❑), and PWR and ESR (♦). Additional pairwise comparison results are given in Table 3.

increased range of $\vec{H}$ during prosthetic limb stance seen in previous level walking studies (Silverman and Neptune, 2011). Contrary to our expectations, individuals with TTA using the powered prosthesis also had an increased range of sagittal-plane $\vec{H}$ relative to AB for stair ascent and level walking and, notably, there were no significant differences in the range of $\vec{H}$ between prostheses.

During stair ascent, individuals with TTA using both passive and powered prostheses had an increased second peak vertical GRF in the intact (trailing) limb during push-off (Fig. 5, 0–10% prosthetic limb gait cycle), possibly contributing to a more negative slope in the sagittal-plane $\vec{H}$ trajectory that would correspond to movement of the trunk toward the staircase (Fig. 1). In early to mid prosthetic limb stance, individuals with TTA had an increased prosthetic limb anterior moment arm (Fig. 4). This increased anterior moment arm contributed to a greater positive time rate of change of $\vec{H}$ during prosthetic limb stance, approximately 10–30% of the gait cycle (Fig. 4), which would correspond to trunk movement away from the staircase. Both of these mechanisms could contribute to the increased range of $\vec{H}$ during the first half of the gait cycle.

Hip extension power generation in the prosthetic limb for both prostheses was increased in early to mid stance relative to AB (Fig. 6), which is consistent with previous studies of level walking and stair ascent (Aldridge et al., 2012; Winter and Sienko, 1988; Yack et al., 1999). Greater hip extensor power generation raises the trunk and rotates the body away from the staircase and may result in an increased positive time rate of change and greater overall range of $\vec{H}$. While this may be important for lifting the body COM to the next step in individuals with TTA, it may also adversely affect balance during stair ascent by resulting in a greater range of $\vec{H}$.

The greater range of sagittal-plane $\vec{H}$ was also reflected in the larger negative slope of the $\vec{H}$ trajectory in individuals with TTA from approximately 30–50% of the prosthetic limb gait cycle compared to AB individuals. This may be due to impaired function of the ankle plantarflexor muscles, which work cooperatively to regulate $\vec{H}$ during level walking. During early stance, both the gastrocnemius and soleus contribute to a negative (forward) external moment. In late stance, the two muscle groups work in opposition: the uniarticular soleus primarily contributes to negative (forward) sagittal-plane $\vec{H}$ which propels the body COM while providing support, and the biarticular gastrocnemius generates energy to the leg for swing and contributes to positive (backward) sagittal-plane $\vec{H}$ (Neptune and McGowan, 2011). Recent modeling and simulation work of individuals with TTA during level walking has shown that a passive prosthesis provides much of the function of the soleus muscle during level walking (Silverman and Neptune, 2012). The results of the current study suggest that passive and powered prostheses produce similar effects on $\vec{H}$ during stair walking, despite increased ankle power generation from the powered prosthesis. This may be because the powered prosthesis does not replace the function of the gastrocnemius, which would counteract the increased slopes in the $\vec{H}$

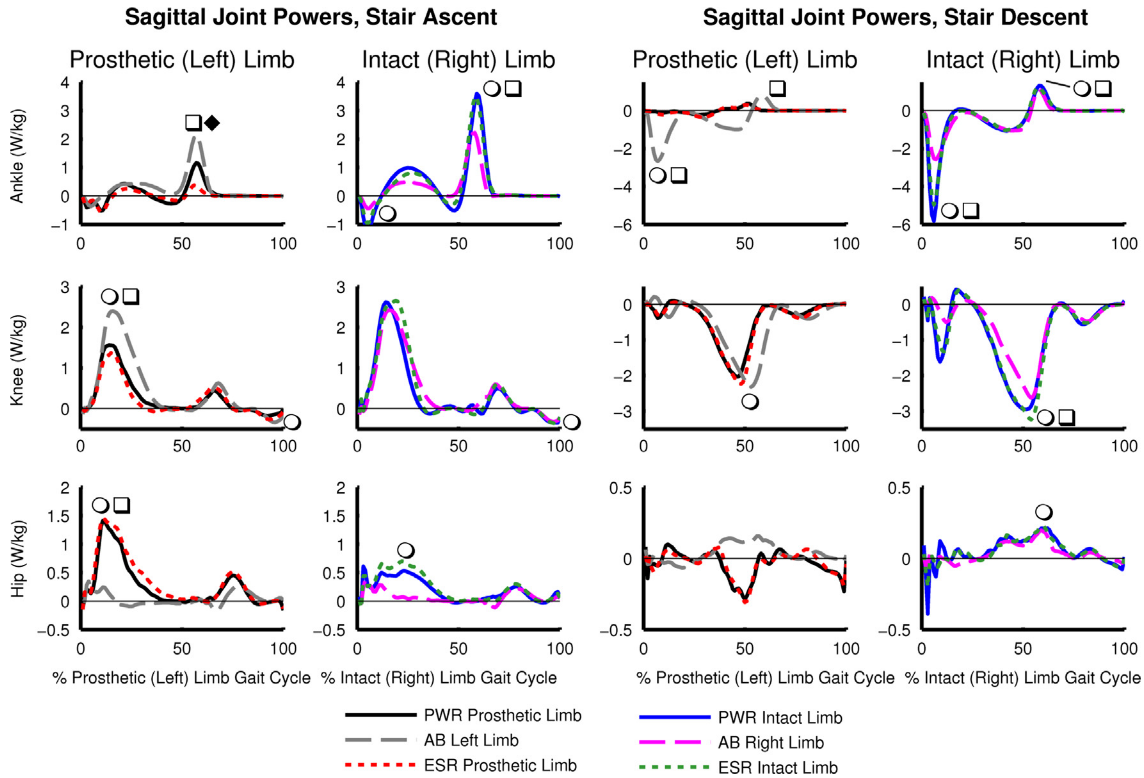

Fig. 6. Sagittal (flexion/extension) joint powers at the ankle, knee, and hip during stair ascent and descent, normalized by subject mass and expressed relative to percentage of the gait cycle. Results are shown for individuals with transtibial amputation (TTA) using a powered prosthesis (PWR, solid lines), able-bodied participants (AB, dashed lines), and individuals with TTA using a passive prosthesis (ESR, dotted lines). Negative values correspond to power absorption and positive values to power generation. Significant differences are indicated between PWR and AB (❍), ESR and AB (❑), and PWR and ESR (♦). Additional pairwise comparison results are given in Table 4.

Table 4
Mean (SD) values for peak joint flexion/extension power generation and absorption in the ankle, knee, and hip, normalized by subject body mass. PWR refers to individuals with an amputation using the powered prosthesis, and ESR refers to use of the passive energy-storage-and-return prosthesis. Pairwise comparisons were performed after significant main effects were found.

Mean (SD) Values for Peak Flexion/Extension Joint Powers (W/kg)								
			Ankle		**Knee**		**Hip**	
			Generated	Absorbed	Generated	Absorbed	Generated	Absorbed
Stair Ascent	**Prosthetic**	ESR	0.62 (0.21)[*o]	−0.74 (0.27)	1.52 (0.53)[*o†]	−0.36 (0.14)[*†]	1.71 (0.46)[*o†]	−0.25 (0.1)[*]
		PWR	1.59 (0.47)[*a†]	−0.78 (0.25)[†]	1.79 (0.36)[*o†]	−0.28 (0.03)[*o†]	1.69 (0.42)[*o†]	−0.31 (0.17)
	Intact	ESR	4.12 (0.93)[*o†]	−1.07 (0.70)[†]	2.88 (0.69)[*†]	−0.51 (0.14)[*†]	0.88 (0.30)[†]	−0.26 (0.10)[*]
		PWR	4.56 (0.90)[*o†]	−1.46 (0.51)[o†]	2.84 (0.53)[*†]	−0.58 (0.15)[*o†]	0.95 (0.25)[o]	−0.26 (0.15)
	Average	AB	2.41 (0.44)[†]	−0.48 (0.35)[*†]	2.56 (0.73)[*†]	−0.38 (0.08)[*†]	0.54 (0.21)	−0.31 (0.17)[*]
Level	**Prosthetic**	ESR	1.48 (0.24)[o]	−1.03 (0.25)	0.44 (0.20)[o]	−0.94 (0.20)[o]	0.97 (0.20)	−0.46 (0.16)
		PWR	3.29 (0.89)[a]	−0.66 (0.14)[oa]	0.46 (0.14)[o]	−1.07 (0.54)	1.04 (0.43)	−0.49 (0.21)
	Intact	ESR	2.67 (0.53)	−1.00 (0.44)	1.43 (0.45)	−1.43 (0.15)	0.71 (0.15)	−0.59 (0.22)
		PWR	2.74 (0.52)	−0.86 (0.37)	1.26 (0.41)	−1.38 (0.36)	0.70 (0.14)	−0.45 (0.11)
	Average	AB	2.58 (0.42)	−1.07 (0.29)	1.51 (0.54)	−1.52 (0.34)	0.78 (0.14)	−0.57 (0.24)
Stair Descent	**Prosthetic**	ESR	0.48 (0.24)[*o]	−0.53 (0.11)[*o]	0.25 (0.12)[†]	−2.46 (0.56)[*†]	0.35 (0.17)[*†]	−0.52 (0.26)
		PWR	0.56 (0.40)[*†]	−0.37 (0.13)[*o†]	0.24 (0.12)[†]	−2.35 (0.29)[*o†]	0.39 (0.16)[*†]	−0.53 (0.28)
	Intact	ESR	1.78 (0.41)[*o†]	−5.10 (1.81)[*o†]	0.64 (0.25)[*†]	−3.63 (0.89)[*o†]	0.41 (0.10)[*†]	−0.30 (0.15)
		PWR	1.87 (0.37)[*o†]	−6.05 (1.44)[*o†]	0.74 (0.37)[†]	−3.59 (1.44)[*o†]	0.56 (0.36)[o]	−0.48 (0.24)
	Average	AB	1.28 (0.35)[*†]	−2.77 (0.52)[*†]	0.37 (0.17)[*†]	−2.68 (0.20)[*†]	0.32 (0.08)[*]	−0.17 (0.07)[*]

[*] Significant differences relative to level walking.
[a] Significant differences relative to the passive prosthesis.
[o] Significant differences relative to able-bodied individuals (AB) for the same walking condition.
[†] Significant differences between stair ascent and descent.

trajectory that were shown for individuals with TTA during prosthetic limb stance (Fig. 2).

Contrary to previous level walking studies (Silverman and Neptune, 2011), the range of sagittal-plane $\vec{H}$ during the second half of the gait cycle was not significantly reduced in individuals with TTA compared to AB (Table 1, Fig. 3), regardless of prosthesis type. Silverman and Neptune (2011) attributed the reduction in $\vec{H}$ to a reduced anterior GRF from the passive prosthesis in individuals with TTA. In the current study, individuals with TTA using the passive prosthesis did have a significantly reduced A/P GRF during level walking as observed in the prior study, but the A/P GRFs between individuals with TTA using the powered prosthesis and AB were not significantly different (Table 3, Fig. 5). Thus, the powered prosthesis provided an increased anterior GRF during level walking compared to a passive prosthesis. During stair walking, however, all subject groups had a reduced peak anterior GRF relative to level walking, suggesting that this component of the GRF is less critical for stair walking.

The observed decrease in sagittal-plane $\vec{H}$ during stair descent in both individuals with TTA and AB relative to level walking and stair ascent is consistent with previous findings regarding regulation of $\vec{H}$ on declined surfaces (Silverman et al., 2012). Regulating sagittal-plane $\vec{H}$ more tightly during stair descent may be the result of a more conservative walking strategy to decrease fall risk. However, individuals with TTA using the powered prosthesis had significantly increased maximum $|\vec{H}|$ during descent (Table 1, Fig. 3), which suggests the possibility that the powered prosthesis may adversely affect regulation of $\vec{H}$ during stair descent. Power absorption in the prosthetic ankle was significantly reduced in individuals with TTA (Fig. 6), and several significant differences were found between individuals with TTA and AB during stair descent in the GRFs and external moment arms (Figs. 4–6). These findings are consistent with data from previous studies in demonstrating altered strategies in individuals with TTA compared to AB during stair descent (Powers et al., 1997; Ramstrand and Nilsson, 2009; Sinitski et al., 2012).

Our results suggest that the powered and passive prostheses are functionally similar in the regulation of $\vec{H}$ during stair walking. Neither of the prostheses studied replicates the function of the biarticular gastrocnemius, which may explain some of the observed differences between individuals with TTA and AB. Musculoskeletal modeling and simulation of the powered prosthesis during level and stair gait is an important area of future work to better understand the functional capabilities of this device relative to biological ankle function. A potential limitation to our study is that we did not directly evaluate fall risk in the study participants or their ability to recover from perturbations. An important area of future work is to determine the extent to which angular momentum (and other relevant factors) directly contribute to the occurrence or prevention of a fall. In addition, future studies investigating the ability of individuals with TTA to respond to perturbations or trips during stair walking will provide further insight into the relationship between the increased range of $\vec{H}$ and fall risk.

5. Conclusions

In summary, individuals with a transtibial amputation had an increased range of sagittal-plane $\vec{H}$ during the first half of the gait cycle relative to AB during stair ascent; however, no significant differences were observed in range of $\vec{H}$ when using powered versus passive prostheses. While the greater range of $\vec{H}$ during the first half of the gait cycle may be necessary for individuals with TTA to ascend stairs, it may adversely affect the ability to maintain dynamic balance and lead to increased fall risk regardless of the type of prosthesis used. Compensations at other joints, such as increased prosthetic limb hip extensor power generation early in the gait cycle, were observed in individuals with TTA using both powered and passive prostheses similar to previous results (Aldridge et al., 2012), and $\vec{H}$ was regulated similarly in all three anatomical planes regardless of prosthesis type. These results suggest that both prostheses are functionally similar in regulating $\vec{H}$ and maintaining dynamic balance during stair walking. All subject groups had a decreased range of sagittal-plane $\vec{H}$ during stair descent relative to stair ascent and level walking, which may be a protective mechanism to reduce the risk of falling and sustaining serious injuries.

Conflict of interest

There is no conflict of interest.

Acknowledgments

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