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Dynamic balance during running using running-specific prostheses

Lauren A. Sepp [a], Brian S. Baum [b], Erika Nelson-Wong [b], Anne K. Silverman [a,*]

[a] Department of Mechanical Engineering, Colorado School of Mines, Golden, CO 80401, USA
[b] School of Physical Therapy, Regis University, Denver, CO 80221, USA

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ABSTRACT

Running is beneficial for physical, social, and emotional health, and participating in physical activity, including running, is becoming more popular for people with an amputation. However, this population has a greater risk of falling relative to people without an amputation, which may be a barrier to running. Understanding how dynamic balance is maintained during running is important for removing this barrier. To investigate dynamic balance, we quantified whole-body angular momentum in eight people with a unilateral transtibial amputation (TTA) using running-specific prostheses (RSPs) compared to eight people without TTA during running at 2.5, 3.0, and 3.5 m/s. People with TTA had greater ranges of whole-body angular momentum compared to people without TTA in the frontal and sagittal planes ($p < 0.01$). These greater ranges resulted from smaller peak medial, lateral, and braking ground reaction forces from the amputated leg compared to the intact leg and people without TTA. Reduced RSP mass relative to the biological leg also influenced whole-body angular momentum as evidenced by smaller ranges of amputated leg angular momentum compared to the intact leg in the frontal and sagittal planes. Smaller amputated leg angular momentum corresponded with smaller contralateral arm angular momentum in the sagittal plane ($p < 0.01$). People with TTA maintain balance during running with altered muscle coordination and prosthesis characteristics. Restoring mediolateral force generation through prosthetic design advances may help in regulating the frontal plane component of whole-body angular momentum for people with TTA, with potential to improve their ability to maintain balance during running.

1. Introduction

Running is beneficial for physical, mental, emotional, and social health and has been growing in popularity as a recreational activity (Bragaru et al., 2011; Valliant et al., 1985; Lundstrom, 2017). People with an amputation are increasingly interested and participating in physical activity, including running, which can be beneficial for rehabilitation and quality of life (LimbPower, 2016; Bragaru et al., 2011; Lundstrom, 2017). Despite growing interest in participation in physical activity, people with an amputation cite prosthetic limitation and fear of falling among the top reasons they choose not to engage in or return to physical activity (LimbPower, 2016). Thus, understanding how dynamic balance is maintained is important for enabling people with an amputation to engage in physical activity.

Whole-body angular momentum must be generated to facilitate locomotor tasks and must also be regulated to maintain dynamic balance during gait (Herr and Popovic, 2008). The regulation of angular momentum during running has not been widely studied with the exception of the contribution of arm swing in relation to balance, which is known to assist lateral balance during running in combination with alterations in step width (Hinrichs, 1987; Arellano and Kram, 2011). The time rate of change of whole-body angular momentum equals the net external moment about the body center of mass, which results from ground reaction forces (GRFs) and their associated moment arms (Fig. 1). Thus, changes in the net external moment can help explain observed changes in whole-body angular momentum.

Transtibial amputation (TTA) is characterized by the functional loss of ankle musculature. The ankle plantarflexors, compared to all major muscle groups, have the largest contributions to support and propulsion during running (Hamner et al., 2010), and both the dorsiflexors and plantarflexors are important for controlling balance, regulating sagittal plane angular momentum, and responding to perturbations during walking (Winter, 1995; Neptune and McGowan, 2011; Mueller et al., 1995). In addition, running on uneven surfaces results in decreased ankle work in people without TTA relative to level ground, which is suggested to be a mechanism

* Corresponding author at: Department of Mechanical Engineering, Colorado School of Mines, 1500 Illinois St., Golden, CO 80401, USA.
E-mail address: asilverm@mines.edu (A.K. Silverman).

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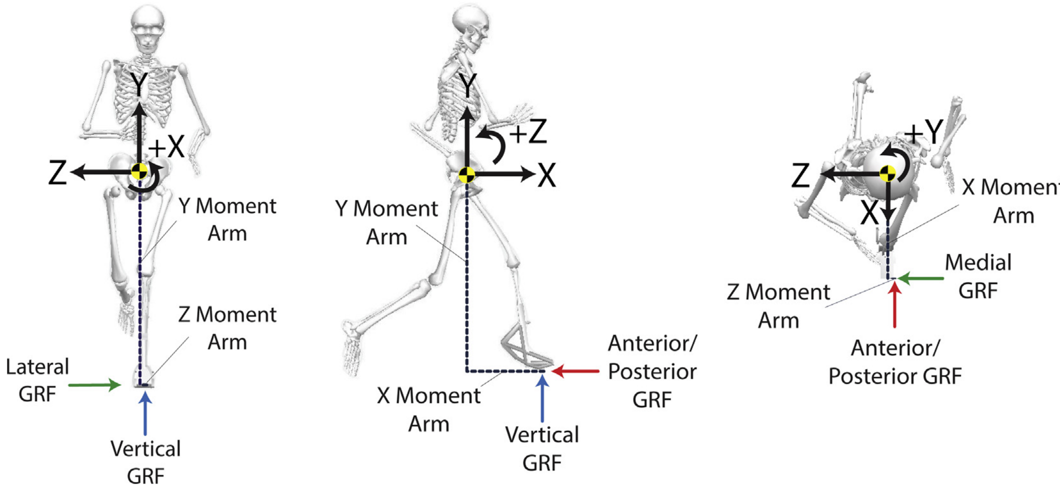

Fig. 1. Dynamic model viewed in the frontal, sagittal and transverse planes. The time rate of change of whole-body angular momentum equals the net external moment about the body center of mass. During running, the external moment results from ground reaction forces and their associated external moment arms (body center of mass to center of pressure distance).

to maintain balance based on proprioceptive feedback and the sensitivity of the ankle joint (Voloshina and Ferris, 2015). The inability to modulate ankle work effectively through muscle action may contribute to the greater risk of falling for people with TTA (Miller et al., 2001). People with TTA have greater ranges of angular momentum during walking at a range of speeds (Silverman and Neptune, 2011), particularly during the prosthetic leg stance phase, when proprioception and muscular control are compromised. In addition, the range of normalized angular momentum decreases with increased walking speed (Silverman and Neptune, 2011; Bennett et al., 2010). However, little is known about how people with TTA modulate whole-body angular momentum during running with a running-specific prosthesis (RSP). Investigating dynamic balance may have implications for prosthetic design improvements and for reducing barriers for people with TTA to participate in physical activity.

Thus, the purpose of this study was to analyze dynamic balance in people with and without TTA during running at a range of speeds. We quantified dynamic balance using whole-body angular momentum, and people with TTA ran using RSPs. To interpret the whole-body angular momentum results, we also evaluated body segment angular momenta, GRFs and external moment arms. We hypothesized that the ranges of normalized whole-body angular momentum in the frontal, sagittal, and transverse planes would be greater in people with TTA compared to people without TTA at all running speeds. We also hypothesized that the ranges of each component of normalized whole-body angular momentum would decrease as speed increased, similar to prior studies of walking.

2. Methods

Eight male runners with a unilateral TTA and eight male runners without TTA (Table 1) provided written informed consent to participate in the protocol approved by the Institutional Review Board. Each person with TTA ran with their prescribed RSP. The manufacturer and stiffness category of each RSP was selected by each participant's clinician.

Participants ran continuously along a 100 m track at randomized speeds of 2.5 m/s, 3.0 m/s, and 3.5 m/s. Average running speed was monitored using a set of six laser sensors around a track. A collection was successful if the participant ran within ±0.2 m/s of the prescribed speed. Whole-body kinematics were collected at 200 Hz using a 10-camera motion capture system (Vicon, Centennial, CO, USA) and a set of 59 reflective markers for people without TTA and 62 reflective markers for people with TTA (Hobara et al., 2013). GRFs were collected using 10 in-ground force plates (Kistler, Amherst, MA, USA) at 1000 Hz along a 25 m straightaway. Kinematic and GRF data were low-pass filtered using a 4th-order Butterworth filter with cutoff frequencies of 6 Hz and 50 Hz, respectively (Winter, 2005). Five gait cycles were analyzed at each speed for each participant. GRFs were normalized by body weight for each participant and have been previously analyzed in detail (Baum et al., 2016).

Whole-body angular momentum ($\vec{H}$) was calculated using a 13-segment model (upper arms, forearms, thighs, shanks, feet, pelvis, torso, and head; 18-segment model for people with TTA, including a six-segment RSP) in Visual3D (C-Motion, Inc., Germantown, MD, USA). Each RSP was modeled as six segments, with markers that were evenly spaced on the lateral side of the RSP keel to capture deflection of the prosthesis (Baum, 2012). Whole-body $\vec{H}$ was calculated from the tracked body segment kinematics as:

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

where $\vec{r}_{i,COM}$ and $\vec{v}_{i,COM}$ are the position and velocity of the ith's segment's center of mass, respectively; $\vec{\omega}_i$ is the angular velocity of the

Table 1
Participant characteristics (mean ± standard deviation). Participants with a transtibial amputation (TTA) ran with their clinically prescribed running-specific prosthesis (RSP).

	Age (years)	Body Mass with RSP (kg)	Height (m)	Time Since Amputation (years)	Running Specific Prosthesis Type	Etiology
8 People with TTA	32 ± 10	82.3 ± 13.0	1.80 ± 0.07	11.9 ± 16.1	4 Össur FlexRun 2 Össur Cheetah 2 Freedom Innovations Catapult	7 Traumatic 1 Congenital
8 People without TTA	29 ± 7	79.3 ± 7.9	1.84 ± 0.05	N/A	N/A	N/A

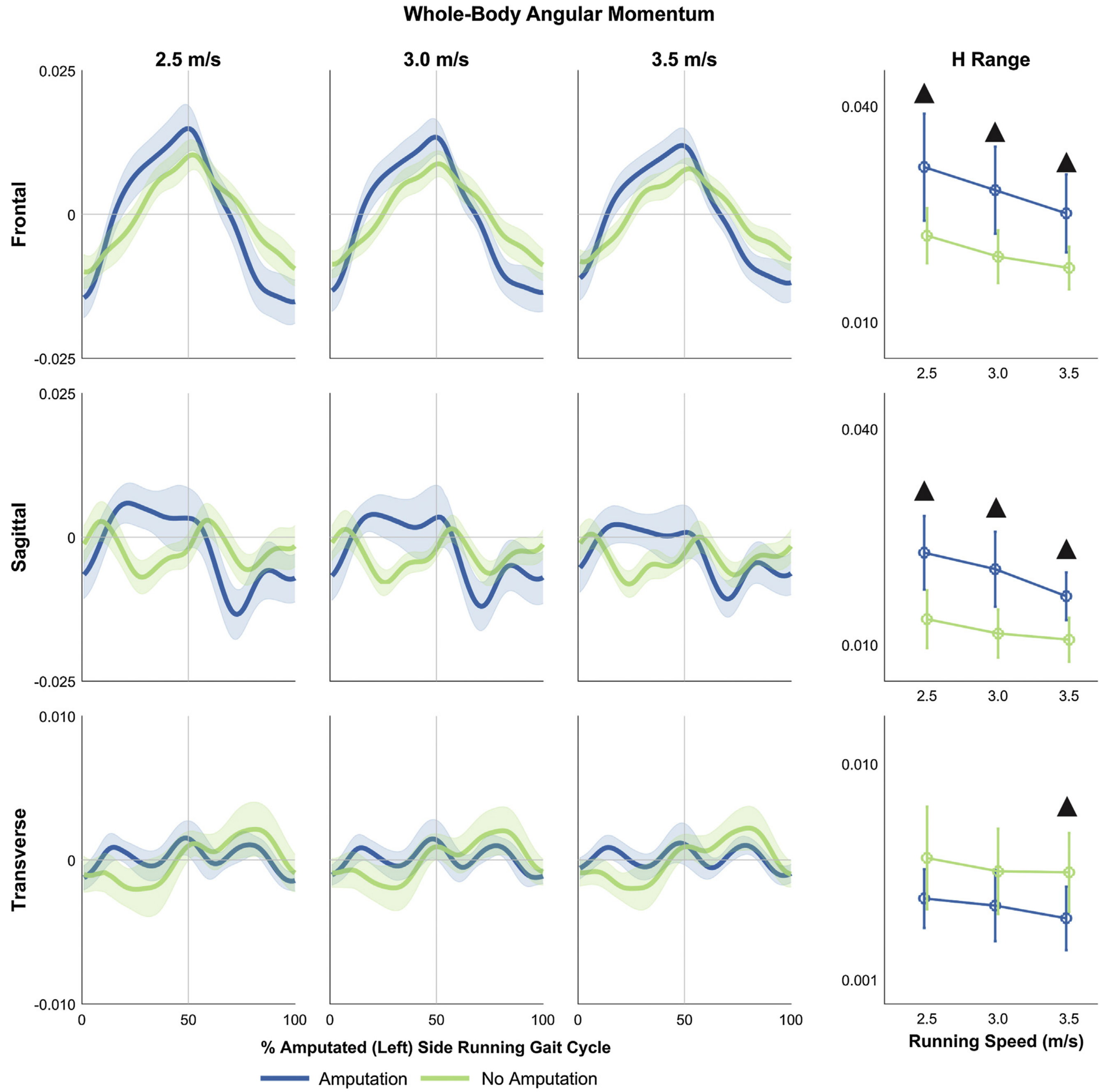

Fig. 2. Mean and one standard deviation (shaded) whole-body angular momentum in frontal, sagittal, and transverse planes during running at 2.5 m/s, 3.0 m/s, and 3.5 m/s for people with and without a transtibial amputation. Significant pairwise comparisons in the range of angular momentum between groups are indicated by "▲". Note the different vertical-axis scale for the transverse plane (bottom row) relative to the frontal and sagittal planes (top and middle rows).

ith segment; $\vec{r}_{body,COM}$ and $\vec{v}_{body,COM}$ are the position and velocity of the center of mass of the whole body, respectively; and m_i and I_i are the mass and inertia matrix of the ith segment, respectively. $\vec{H}$ was normalized by participant height, mass, and average running speed. RSP and socket mass were measured for each participant and inertial properties were estimated from Baum et al. (2013, 2018).

Differences in the range of each component of whole-body and trunk (torso + pelvis + head) $\vec{H}$ (peak-to-peak amplitude in each anatomical plane) were assessed using a two-factor (group × running speed), mixed model ANOVA in R Statistical Computing Software v 1.1.153 ($\alpha = 0.05$). We also compared the range of each component of $\vec{H}$ for the arms (upper arms + forearms) and legs (thighs, shanks, and feet or RSP), peak GRFs, and peak external moment arms using a three-factor (group × side × running speed), mixed-model ANOVA. Post-hoc tests using Tukey's correction for multiple comparisons were performed when significant main or interaction effects were found.

3. Results

The ranges of the frontal and sagittal plane components of whole-body $\vec{H}$ were greater for people with TTA compared to people without TTA at all running speeds ($p \leq 0.011$), which partially supported our hypothesis (Fig. 2). Contrary to our hypothesis, the range of the transverse plane component of whole-body $\vec{H}$ was greater for people without TTA compared to those with TTA at 3.5 m/s ($p = 0.03$).

Table 2
Main and interaction effects of whole-body and segmental angular momentum ranges for frontal, sagittal, and transverse plane across speeds. '–' indicates that a significant effect was not observed.

			Frontal	Sagittal	Transverse
ANOVA Main Effects for Whole-Body $\vec{H}$ Ranges			p-values		
		Group	<0.01	<0.01	–
		Speed	<0.01	<0.01	<0.01
		Interactions	–	Group × Speed (<0.01)	–
ANOVA Main Effects for Segmental $\vec{H}$ Ranges	Trunk	Group	<0.01	<0.01	–
		Speed	<0.01	<0.01	<0.01
		Interactions	Group × Speed (0.02)	–	–
	Arms	Group	–	0.03	–
		Speed	–	<0.01	<0.01
		Side	<0.01	<0.01	<0.01
		Interactions	–	–	Group × Side (<0.01)
	Legs	Group	<0.01	<0.01	–
		Speed	<0.01	<0.01	0.04
		Side	<0.01	<0.01	–
		Interactions	Group × Side (<0.01)	Group × Side (<0.01)	Group × Side (0.03)

Table 3
Mean (standard deviation) ranges of segmental angular momentum and results of post hoc pairwise comparisons when significant ANOVA main or interaction effects were observed. All values are shown in scientific notation (x10^{-2}). Significant differences from people without an amputation are represented by "▲". Significant differences compared to the amputated side are represented by "●".

Speed	Segment	Side	Frontal	Sagittal	Transverse
			Mean (SD) H Range (values shown as 10^{-2})		
2.5 m/s	Trunk	TTA	1.36 (0.46)▲	0.96 (0.21)▲	0.54 (0.12)
		AB	0.79 (0.22)	0.66 (0.17)	0.65 (0.13)
	Arm	Amputated Side	0.31 (0.05)	0.20 (0.05)	0.54 (0.11)
		Intact Side	0.28 (0.08)	0.16 (0.03)●▲	0.45 (0.09)
		AB	0.26 (0.05)	0.24 (0.06)	0.52 (0.07)
	Leg	Amputated Side	0.90 (0.18)	4.10 (0.55)▲	0.64 (0.09)
		Intact Side	1.25 (0.24)●▲	5.85 (0.56)●	0.68 (0.13)●
		AB	0.77 (0.18)	6.00 (0.40)	0.59 (0.14)
3.0 m/s	Trunk	TTA	1.23 (0.36)▲	0.95 (0.24)▲	0.52 (0.10)
		AB	0.70 (0.22)	0.54 (0.13)	0.61 (0.10)
	Arm	Amputated Side	0.30 (0.05)	0.19 (0.03)	0.54 (0.11)
		Intact Side	0.26 (0.07)	0.15 (0.03)●▲	0.43 (0.10)
		AB	0.25 (0.05)	0.22 (0.06)	0.48 (0.07)
	Leg	Amputated Side	0.82 (0.18)	4.10 (0.55)	0.63 (0.10)
		Intact Side	1.13 (0.31)●▲	5.59 (0.34)●	0.67 (0.14)●
		AB	0.65 (0.17)	5.70 (0.23)	0.58 (0.11)
3.5 m/s	Trunk	TTA	1.07 (0.33)▲	0.81 (0.15)▲	0.50 (0.08)▲
		AB	0.65 (0.21)	0.49 (0.15)	0.60 (0.06)
	Arm	Amputated Side	0.29 (0.05)	0.18 (0.04)	0.51 (0.10)
		Intact Side	0.24 (0.07)	0.13 (0.03)●▲	0.41 (0.10)●
		AB	0.25 (0.05)	0.21 (0.06)	0.47 (0.05)
	Leg	Amputated Side	0.72 (0.14)	3.94 (0.53)▲	0.60 (0.09)
		Intact Side	0.98 (0.23)●▲	5.33 (0.44)●	0.65 (0.13)
		AB	0.58 (0.15)	5.32 (0.30)	0.54 (0.09)

3.1. Frontal plane

As running speed increased, the range of the frontal plane component of whole-body $\vec{H}$ for people with and without TTA decreased (significant main effect of speed ($p < 0.01$), Table 2). There was also a significant group effect, which points to the greater range for people with TTA compared to people without TTA ($p < 0.01$) (Table 2). In addition to the whole-body results, significant main and interaction effects for the ranges of the frontal plane components of segmental $\vec{H}$ were observed (Table 3). For example, the range of the frontal plane component of trunk $\vec{H}$ was greater for people with TTA compared to people without TTA at all speeds ($p \leq 0.02$). In addition, the intact leg had a greater range of frontal plane $\vec{H}$ compared to the amputated leg and leg of people without TTA at all speeds ($p < 0.01$) (Fig. 3). These segmental results help explain the observed differences in whole-body $\vec{H}$ due to altered coordination of the trunk and legs.

3.2. Sagittal plane

People with TTA had a greater range of the sagittal plane component of whole-body $\vec{H}$ compared to people without TTA (group main effect, $p < 0.01$), and this range decreased more with speed for people with TTA compared to people without TTA (speed × group interaction effect, $p < 0.01$, Table 2, Fig. 2). This range was greater for people with TTA compared to people without TTA at all speeds ($p \leq 0.01$) (Fig. 2). Similar to the whole-body, people with TTA also had greater ranges of the sagittal component of

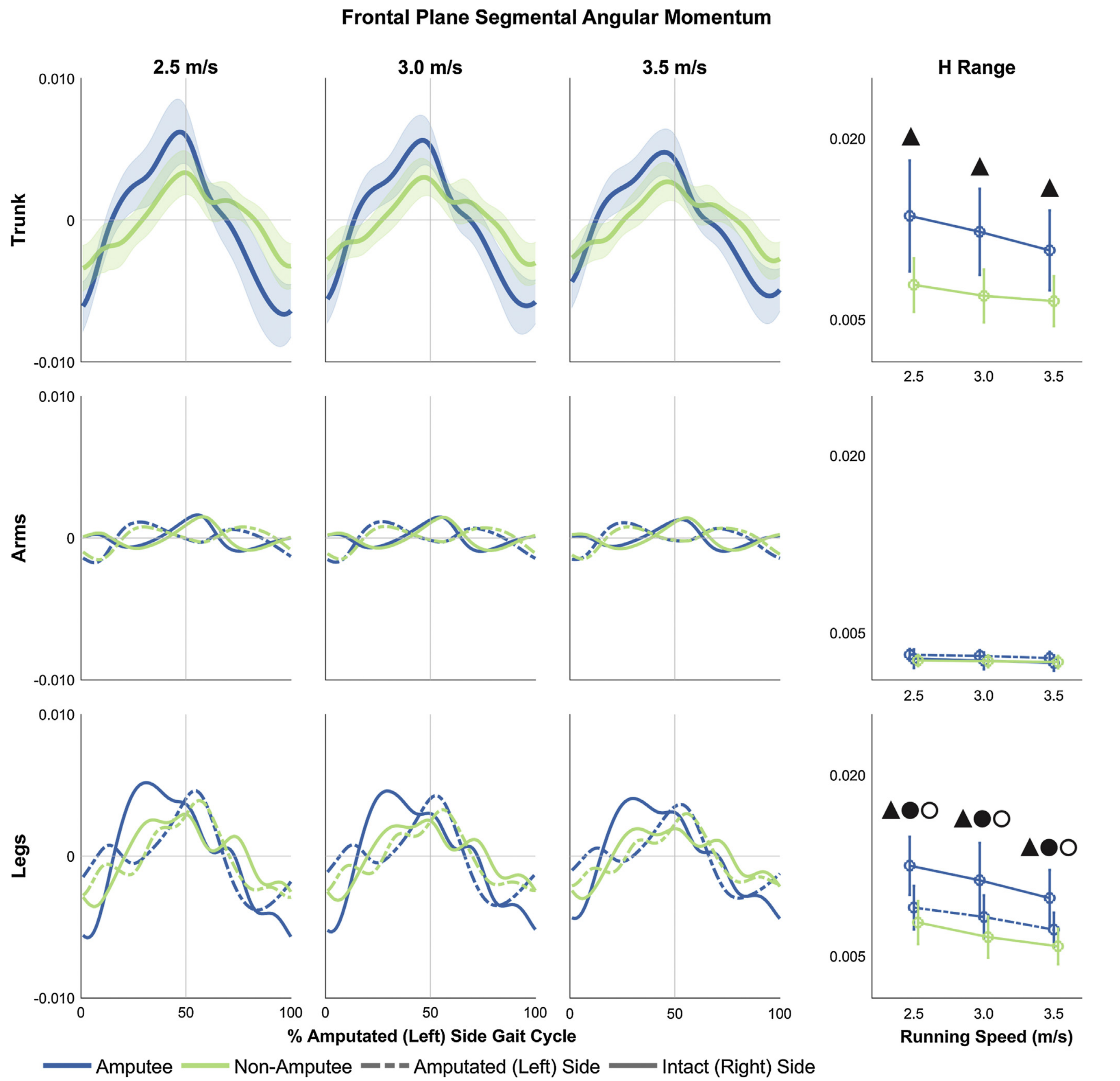

Fig. 3. Mean and one standard deviation (shaded) of the frontal plane component of trunk, arm, and leg angular momentum during running at 2.5 m/s, 3.0 m/s, and 3.5 m/s. Mean and one standard deviation (vertical lines) of trunk, arm, and leg angular momentum range are shown in the right column. Significant differences between people with and without and a transtibial amputation are indicated by "▲". Significant differences between the amputated and intact sides are indicated by "●". Significant differences between intact side and people without an amputation are indicated by "○".

trunk $\vec{H}$ compared to people without TTA at all speeds ($p < 0.01$) (Fig. 4). Within participants with TTA, the sagittal plane range of arm $\vec{H}$ was greater for the amputated side arm compared to the intact side arm at all speeds ($p < 0.01$) (Table 3, Fig. 4). In contrast, the amputated leg had a smaller sagittal plane range of $\vec{H}$ compared to the intact leg ($p < 0.01$) and compared to the leg of people without TTA ($p < 0.01$) at all speeds (Table 3, Fig. 4).

3.3. Transverse plane

The ranges of the transverse plane components of whole-body and trunk $\vec{H}$ decreased with increases in speed (speed main effect, $p < 0.01$, Tables 2 and 3). In addition, the amputated side arm had a greater range of the transverse plane component of $\vec{H}$ compared to the intact side arm at all speeds ($p < 0.01$) (Table 3).

3.4. External moment arms

The maximum anterior moment arm was significantly different between sides (side main effect, $p < 0.01$), indicating differences in foot placement relative to the body center of mass (Table 4). In addition, a group × side interaction effect ($p < 0.01$) suggested different levels of asymmetry in foot placement for people with TTA relative to those without TTA. The amputated leg had a greater anterior moment arm than the intact leg at all speeds ($p < 0.01$)

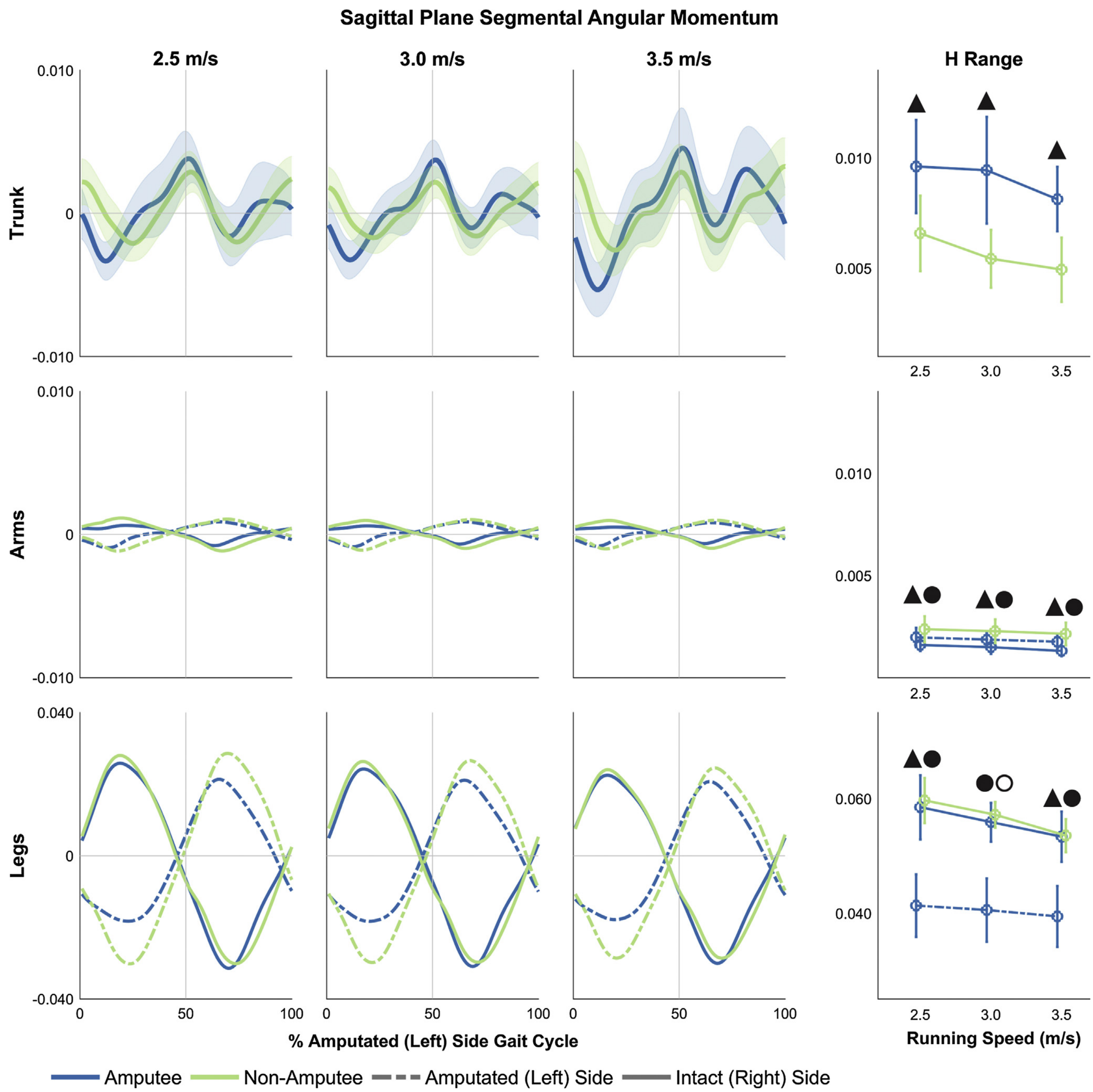

Fig. 4. Mean and one standard deviation (shaded) of the sagittal plane component of trunk, arm, and leg angular momentum during running at 2.5 m/s, 3.0 m/s, and 3.5 m/s. Mean and one standard deviation (vertical lines) of trunk, arm, and leg angular momentum range are shown in the right column. Significant differences between people with and without and a transtibial amputation are indicated by "▲". Significant differences between the amputated and intact sides are indicated by "●". Significant differences between intact side and people without an amputation are indicated by "○". Note the different vertical-axis scale for the legs (bottom row) relative to the trunk and arms (top and middle rows).

and compared to people without TTA at 2.5 m/s and 3.0 m/s ($p < 0.04$) (Table 4). There was a speed main effect ($p < 0.01$) for maximum posterior moment arm, which decreased with increases in speed for both groups (Table 4). The maximum lateral moment arm differed between sides ($p < 0.01$) and groups ($p < 0.01$). The intact leg also had a greater maximum lateral arm than amputated leg at 3.0 m/s ($p = 0.02$) (Table 4), indicating that the amputated leg foot placement was closer to the body center of mass. The maximum vertical moment arm also differed between groups ($p < 0.01$) and across speed ($p < 0.01$). People without TTA ran with their body center of mass closer to the ground compared to people with TTA, as both the amputated ($p < 0.03$) and intact legs ($p < 0.04$) had larger peak vertical moment arms compared to people without TTA at all speeds (Table 4).

3.5. Ground reaction forces

There were significant main effects of speed and side, and a significant group × side interaction effect for maximum anterior, posterior, medial, and vertical GRFs ($p < 0.01$), and all increased with increases in speed (Fig. 5). Peak lateral GRFs had a significant side main effect and group × side interaction effect ($p < 0.01$).

Table 4
Mean (standard deviation) peak external moment arms (m). Moment arms for people without a transtibial amputation (TTA) are the average of the left and right legs. Significant differences compared to the amputated side are denoted with "●". Significant differences compared to the leg of people without an amputation are denoted with "▲".

Speed	2.5 m/s	3.0 m/s	3.5 m/s
Intact Side			
Anterior (X) Moment Arm	0.106 (0.044)●	0.109 (0.037)●	0.116 (0.040)●
Posterior (X) Moment Arm	−0.184 (0.032)	−0.195 (0.030)	−0.217 (0.023)
Mediolateral (Y) Moment Arm	0.027 (0.009)	0.027 (0.006)●	0.025 (0.007)
Vertical (Z) Moment Arm	0.640 (0.027)▲	0.640 (0.034)▲	0.634 (0.033)▲
Amputated Side			
Anterior (X) Moment Arm	0.166 (0.024)▲	0.166 (0.023)▲	0.156 (0.028)
Posterior (X) Moment Arm	−0.194 (0.020)	−0.204 (0.022)	−0.215 (0.027)
Mediolateral (Y) Moment Arm	0.018 (0.007)	0.016 (0.005)	0.017 (0.007)
Vertical (Z) Moment Arm	0.640 (0.024)▲	0.640 (0.028)▲	0.638 (0.029)▲
People without TTA			
Anterior (X) Moment Arm	0.111 (0.027)	0.119 (0.026)	0.127 (0.031)
Posterior (X) Moment Arm	−0.199 (0.040)	−0.219 (0.026)	−0.229 (0.025)
Mediolateral (Y) Moment Arm	0.024 (0.007)	0.021 (0.007)	0.028 (0.009)
Vertical (Z) Moment Arm	0.606 (0.010)	0.603 (0.010)	0.603 (0.010)

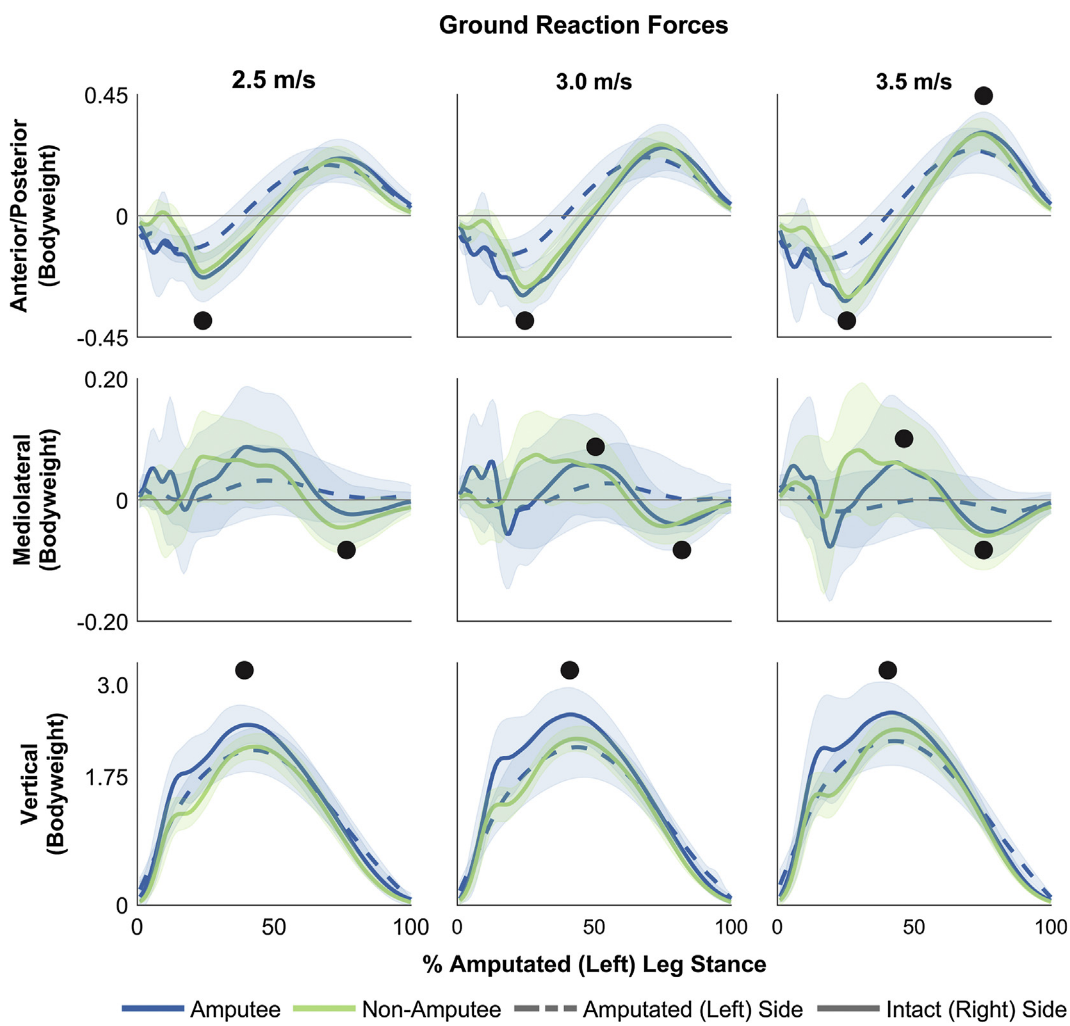

Fig. 5. Mean and one standard deviation (shaded) anterior/posterior, mediolateral, and vertical ground reaction forces (in bodyweights) during running at 2.5 m/s, 3.0 m/s, and 3.5 m/s for people with and without a transtibial amputation. Significant differences between amputated and intact sides are indicated by "●".

The intact leg had a greater propulsive force compared to the amputated leg at 3.5 m/s ($p < 0.01$). The amputated leg had a smaller braking force compared to the intact leg at all speeds ($p < 0.01$) and compared to the leg of people without TTA at 3.0 m/s and 3.5 m/s ($p \leq 0.04$). The intact leg had a greater maximum lateral GRF compared to the amputated leg at all speeds ($p < 0.01$) and greater maximum medial GRF at 3.0 m/s and 3.5 m/s ($p < 0.01$). The intact leg had a greater maximum vertical GRF compared to the amputated leg at all speeds ($p < 0.01$) (Fig. 5).

4. Discussion

The purpose of this study was to analyze dynamic balance in people with and without TTA during running at a range of speeds. The ranges of the frontal and sagittal plane components of whole-body $\vec{H}$ were greater for people with TTA compared to people without TTA, which was expected and is consistent with previous observations of people with TTA during walking at a range of speeds (Silverman and Neptune, 2011). The range of the transverse plane component of whole-body $\vec{H}$ was greater for people without TTA at 3.5 m/s, which did not support our hypothesis.

Changes in $\vec{H}$ of individual body segments alter the range of whole-body $\vec{H}$, and $\vec{H}$ of individual body segments is often counteracted by other body segments, resulting in small values of whole-body $\vec{H}$ (Herr and Popovic, 2008). For example, the movement of the arms and legs offset the movement of one another and also reduce trunk excursions in the transverse plane (Hinrichs, 1987). In our study, differences between groups in ranges of trunk $\vec{H}$ were similar to those of whole-body $\vec{H}$ in all planes (Table 3), which suggests that overall, differences in whole-body $\vec{H}$ are likely closely associated to trunk $\vec{H}$.

Reduced RSP mass and altered inertial properties relative to a biological leg contributed to the observed differences in the frontal and sagittal plane components of $\vec{H}$ between legs (Figs. 3, 4). Smaller mass (m_i) and moment of inertia (I_i) of the amputated leg (residual leg plus prosthesis) decreased the magnitude of the orbital and spin components of $\vec{H}$ (Eq. (1)). A proximal shift in the center of mass of the residual leg (thigh, residual shank, and prosthesis) also decreased the orbital term ($\vec{r}_{i,COM}$, Eq. (1)). This reduction resulted in a greater ranges of the frontal and sagittal plane components of whole-body $\vec{H}$, as the opposing leg contributions did not cancel each other to as great of an extent as in people without TTA. The intact leg had a greater negative value of the frontal plane component of $\vec{H}$ as the hip flexed through mid-swing (~0–10% amputated leg gait cycle) (Fig. 3). This greater magnitude coincided with a greater negative value of trunk frontal plane $\vec{H}$. Both the intact leg and trunk dominated the frontal plane component of whole-body $\vec{H}$. In the sagittal plane, the amputated leg had a reduced contribution during stance and swing phase (Fig. 4), which contributed to more positive whole-body $\vec{H}$ in stance and more negative whole-body $\vec{H}$ in swing.

Although legs are the largest contributor to the sagittal plane component of $\vec{H}$, the accompanying changes in the arms highlight the effect of using a prosthesis on $\vec{H}$ of the whole body. For example, during amputated leg stance in the sagittal plane, the amputated leg and intact side arm both have a reduced $\vec{H}$ range compared to the intact leg and amputated side arm (Fig. 4). The less negative sagittal plane component of $\vec{H}$ of the amputated leg corresponds with a less positive contribution from the intact side arm (~0–40% gait cycle) (Fig. 4). In addition, arm swing is known to be a mechanism for maintaining lateral balance during running for people without TTA (Arellano and Kram, 2011) and during walking over rough surfaces for people with TTA (Curtze et al., 2011). Therefore, changes in arm $\vec{H}$ may help regulate whole-body $\vec{H}$ to maintain balance in response to asymmetry in the legs.

Our results suggest that the regulation of the frontal plane component of whole-body $\vec{H}$ is an important metric for assessing balance in people with TTA. A greater frontal plane component of whole-body $\vec{H}$ in people with TTA during running (Fig. 2) is consistent with prior studies in this population during walking (e.g., Silverman and Neptune, 2011; Gates et al., 2013; Sheehan et al., 2015). We found that smaller medial GRFs from the prosthetic leg reduced the negative external moment in the frontal plane despite a greater vertical moment arm (Fig. 5, Table 3), resulting in a greater positive time rate of change of the frontal plane component of whole-body $\vec{H}$ during amputated leg stance (~0–40% gait cycle, Fig. 2). In addition, during intact leg stance, greater vertical GRFs increased the negative external moment (~60–100% gait cycle, Fig. 5).

Mediolateral GRFs are primarily modulated by ankle musculature (i.e., soleus, gastrocnemius, ankle everters) in combination with hip adductors and abductors (Pandy et al., 2010), and thus differences in mediolateral force are likely due to the lack of ankle muscles in people with TTA and the inability of the RSP to provide their function. Further, modulation of mediolateral GRFs is important for maintaining balance during single-limb stance (Tropp and Odenrick, 1988; Chou et al., 2003; Pandy et al., 2010) and the ankle plantarflexors are critical for the regulation of the frontal plane component of $\vec{H}$ during walking (Neptune and McGowan, 2016). In combination with a greater range of the frontal plane component of $\vec{H}$, the ability to safely recover from mediolateral perturbation may be difficult for people with TTA, especially during running on uneven surfaces, such as outdoor trails. Thus, the ability for RSPs to better regulate mediolateral force generation, thereby reducing the range of the frontal plane component of $\vec{H}$, may be important to consider in prosthetic design. For example, inclusion of a split-toe design in RSPs may allow for better mediolateral force modulation, by allowing for inversion/eversion in the absence of an articulating subtalar joint.

Braking, propulsive, and vertical GRFs act about the body center of mass to create net external moments to control the sagittal plane component of $\vec{H}$ (Fig. 1). Participants with TTA had a smaller peak amputated leg braking force compared to the intact leg (Fig. 5), which resulted in a more positive (backward) sagittal plane external moment as evidenced by the positive slope of the sagittal plane component of $\vec{H}$ in early prosthetic leg stance (~0–20% gait cycle, Fig. 4). During running, the quadriceps are the main contributor to braking the body center of mass in the first half of stance (Hamner et al., 2010). In the second half of stance, the soleus and gastrocnemius propel the body forward (Hamner et al., 2010). People with TTA have reduced quadriceps strength (Renström et al., 1983; Lloyd et al., 2010) in the amputated leg compared to the intact leg and are also missing the function of the plantarflexors. Therefore, the greater range of the sagittal plane component of $\vec{H}$ is likely a result of altered muscle coordination and function to maintain balance during running with a TTA.

In addition, the amputated leg had an increased anterior moment arm in combination with a decreased vertical GRF (Table 4, Fig. 5). The effect of a greater moment arm was larger than that of the reduced GRF, and thus the positive net external moment in the sagittal plane was greater, also contributing to a greater positive slope of the sagittal plane component of $\vec{H}$ compared to people without TTA (Fig. 2). A more anterior foot placement from the prosthetic leg may be a method to compensate for reduced vertical force generation to regulate whole-body $\vec{H}$.

Greater ranges of $\vec{H}$ suggest that individuals may have a greater risk of falling, because a greater external moment is required to restore $\vec{H}$ to maintain dynamic balance. However, increases in ranges of $\vec{H}$ components do not necessarily indicate a fall is imminent, as people with TTA are able to run successfully at a range of speeds. Sagittal plane movement tasks, such as walking and running, may be facilitated by greater ranges of $\vec{H}$ in this plane and reduced braking forces in people with TTA, where forward movement is a clear objective and ankle plantarflexor muscle function to provide propulsion is impaired.

The results of this study are subject to potential limitations. RSPs are carbon fiber devices that undergo large deformations during dynamic activities, which can be difficult to track and model accurately. Inertial properties for the RSP were estimated from previous research, which included three specific models of RSPs with inertial values calculated from a period of oscillation using a trifilar pendulum (Baum et al., 2013, 2018). Center of mass approximation of a segment, and in this case the RSP, affects its contribution to whole-body $\vec{H}$. However, as RSPs are of a similar shape, the small variants in center of mass location between RSP models would likely not have a large effect on our results. Different RSP models also have varying stiffness, which are categorized into a small range of discrete, pre-determined values set by the manufacturer (Beck et al., 2016). We did not control for the specific stiffness value of each device, which is generally prescribed based on body mass of the participant. We instead included participants running with their own, clinically-prescribed prosthesis. Certainly, different device stiffness values would influence the compression of the device, and also the ability for the participant to generate appropriate GRFs, likely altering the $\vec{H}$ trajectory. We accounted for the varying kinematics and force generation profiles of different devices across participants in our study; however, how different prosthesis stiffness values affect how dynamic balance is controlled is an important area of future work. In addition, while the ranges of the 3D components of $\vec{H}$ are useful metrics to describe the deviation of $\vec{H}$ from zero, changes in $\vec{H}$ trajectory are not fully captured by analyzing ranges alone. Changes in both timing and magnitude of the $\vec{H}$ trajectory may provide more detailed information regarding differences in neuromuscular strategies to maintain balance throughout a movement (Pickle et al., 2017). Finally, we examined running speeds common to recreational runners (10:44 min/mile to 7:40 min/mile). Testing a wider range of speeds in future work has potential to provide additional insight into maintaining dynamic balance during running with an amputation, as some runners may choose to run faster or slower than the specific speeds we tested.

5. Conclusion

The regulation of $\vec{H}$ is important to evaluate in people with TTA as they have an increased risk and fear of falling, which may be magnified when they engage in high velocity activities such as running. Our results indicate that people with TTA have greater ranges of the components of $\vec{H}$ at both the whole-body and segmental level, which are driven by prosthesis characteristics and muscular action. Residual leg and RSP mass and center of mass characteristics in people with TTA reduce amputated leg $\vec{H}$ and therefore increase the magnitude of whole-body $\vec{H}$. Opposing asymmetry in arm $\vec{H}$ mitigated the effects of leg asymmetry on whole-body $\vec{H}$. Smaller peak mediolateral GRF production while using RSPs increased the range of the frontal plane component of whole-body $\vec{H}$, which may have negative implications for fall risk during running. In addition, a smaller peak braking force and a greater anterior moment arm in the prosthetic leg increased the positive net external moment in the sagittal plane, which resulted in a more positive sagittal plane component of whole-body $\vec{H}$ during prosthetic leg stance. This result may have negative implications for fall risk, but also facilitate forward movement during running with a TTA.

To enable people with TTA to engage in physical activity and regulate dynamic balance effectively, RSP design may be improved by incorporating better mediolateral GRF modulation. Future investigation of prosthesis mass, shape, and stiffness effects on whole-body $\vec{H}$ can further influence prosthetic design. Analysis of $\vec{H}$ in runners with TTA using different prosthesis designs during running may also reveal design features that facilitate better regulation whole-body $\vec{H}$ and reduce fall risk.

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Conflict of interest statement

There was no conflict of interest in the preparation or publication of this work.

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