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Differences in whole-body angular momentum between below-knee amputees and non-amputees across walking speeds

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ABSTRACT

Unilateral, below-knee amputees have an increased risk of falling compared to non-amputees. The regulation of whole-body angular momentum is important for preventing falls, but little is known about how amputees regulate angular momentum during walking. This study analyzed three-dimensional, whole-body angular momentum at four walking speeds in 12 amputees and 10 non-amputees. The range of angular momentum in all planes significantly decreased with increasing walking speed for both groups. However, the range of frontal-plane angular momentum was greater in amputees compared to non-amputees at the first three walking speeds. This range was correlated with a reduced second vertical ground reaction force peak in both the intact and residual legs. In the sagittal plane, the amputee range of angular momentum in the first half of the residual leg gait cycle was significantly larger than in the non-amputees at the three highest speeds. In the second half of the gait cycle, the range of sagittal-plane angular momentum was significantly smaller in amputees compared to the non-amputees at all speeds. Correlation analyses suggested that the greater range of angular momentum in the first half of the amputee gait cycle is associated with reduced residual leg braking and that the smaller range of angular momentum in the second half of the gait cycle is associated with reduced residual leg propulsion. Thus, reducing residual leg braking appears to be a compensatory mechanism to help regulate sagittal-plane angular momentum over the gait cycle, but may lead to an increased risk of falling.

1. Introduction

The regulation of whole-body angular momentum is important for preventing falls and recovering from trips (e.g., Pijnappels et al., 2004). Previous studies have shown that the ankle muscles are important in regulating angular momentum (Pijnappels et al., 2005a), and that individuals with a history of falls have reduced ankle plantar flexor output (LaRoche et al., 2010). For example, when recovering from a trip, fallers have a reduced peak ankle plantar flexor moment relative to non-fallers, which limits their ability to restrain their forward angular momentum (Pijnappels et al., 2005a). Similarly, older adults with a history of falls have reduced ankle dorsiflexor strength (Skelton et al., 2002) and ankle plantar flexor moments during walking (Simoneau and Krebs, 2000).

The importance of the ankle muscles was further highlighted in a simulation study of non-amputee walking that showed the plantar flexors are the only muscles with the ability to regulate sagittal-plane angular momentum throughout the stance phase (Neptune and McGowan, 2010). In addition, the tibialis anterior had significant contributions to backward angular momentum in early stance. Thus, below-knee amputees may have difficulty regulating their angular momentum due to the functional loss of the ankle muscles, which may explain their increased risk of falling relative to non-amputees (Miller et al., 2001).

The net external moment on the body, which is a function of the ground reaction forces (GRFs) and foot placement (Fig. 1), equals the time rate of change of whole-body angular momentum. Unilateral amputees often have asymmetric step lengths (Barth et al., 1992; Underwood et al., 2004; Zmitrewicz et al., 2006), center of pressure (COP) trajectories (Hansen et al., 2004), and GRFs (Nolan et al., 2003; Sanderson and Martin, 1997; Silverman et al., 2008) between the intact and residual legs. Thus, the external moment on the body and corresponding angular momentum during amputee walking would likely differ from non-amputees. However, no study has examined whole-body angular momentum in amputees.

The purpose of this study was to identify differences in three-dimensional (3D), whole-body angular momentum between below-knee amputees and non-amputees over a range of walking speeds. We analyzed angular momentum across walking speeds to further highlight differences between amputees and non-amputees, as angular momentum has been shown to vary with speed (Bennett et al., 2010). Due to amputee GRF and foot placement asymmetry, we expected that the range of angular momentum would be different in amputees compared to non-amputees at all walking speeds, which may provide insight into their increased risk of falling.

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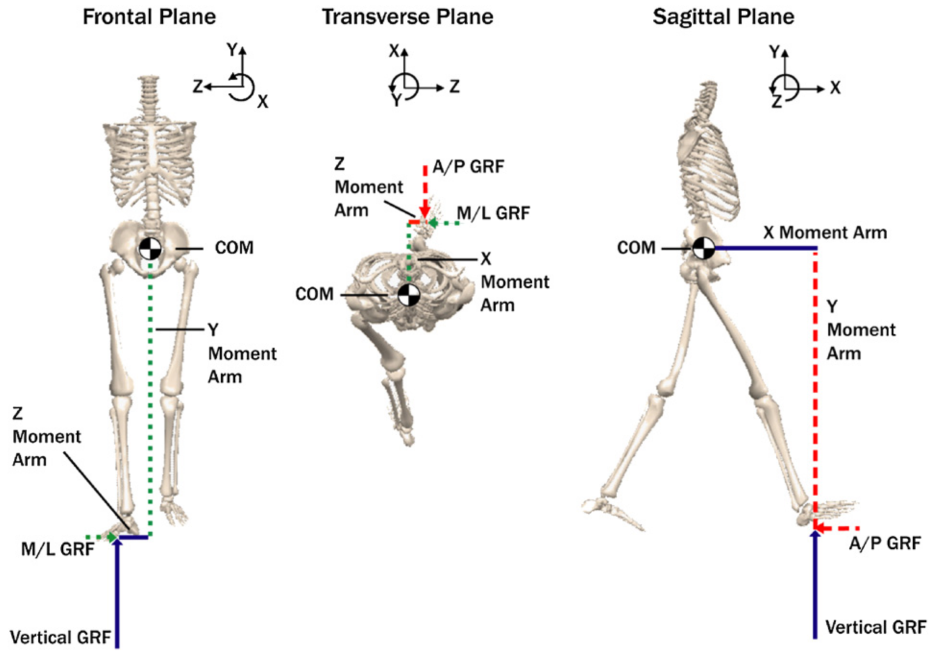

Fig. 1. Model viewed from the frontal, transverse, and sagittal planes. Frontal-plane angular momentum was defined about the *X* axis, transverse-plane angular momentum defined about the *Y* axis, and sagittal-plane angular momentum defined about the *Z* axis. The ground reaction forces (GRFs) and foot placement contribute to the external moment about the body center of mass (COM), which equals the time rate of change of angular momentum. Each GRF is shown in the same color as its corresponding external moment arm. Note that the figure only shows external moment contributions from the right leg for clarity. However, the GRFs and foot placement of the left leg will also contribute to the external moment about the COM.

Table 1
Mean (σ) subject characteristics for amputees and non-amputees (time from amputation, prosthetic foot type and etiology are reported for the amputees). The prosthetic foot type is classified as energy storage and return (ESAR) or solid ankle cushioned heel (SACH).

	Age (years)	Body mass (kg)	Body height (m)	Time since amputation (years)	Foot type	Etiology
Amputees	46.3 (9.2)	89.3 (18.5)	1.76 (0.1)	5.4 (3.1)	7 ESAR/5 SACH	9 Traumatic/3 vascular
Non-amputees	34.1 (13.0)	70.9 (13.6)	1.76 (0.1)	–	–	–

2. Methods

Data collection methods have been previously described (Silverman et al., 2008) and are briefly described here. Kinematic and kinetic data were collected from 12 unilateral, below-knee amputees and 10 non-amputees (Table 1). All subjects provided informed consent to an Institutional Review Board approved protocol. Kinematics were collected using a cluster marker set including 46 reflective markers at 120 Hz. GRFs were collected at 1200 Hz using four force plates in the center of a 10-m walkway. Subjects walked overground at four randomly-ordered walking speeds (0.6, 0.9, 1.2 and 1.5 m/s), which were measured using two infrared timing gates separated by 1.8 m.

Data were processed using Visual3D (C-Motion, Inc.). Kinematic and GRF data were low-pass filtered with a fourth-order Butterworth filter with a cutoff frequency of 6 and 20 Hz, respectively. GRFs were normalized by body weight. An eight-segment model was used to find the center of mass (COM) location and velocity of each segment including the torso, pelvis, thighs, shanks, and feet. Each segment mass was determined as a percentage of total body mass (Table 2, Dempster and Aitkens, 1995). The inertial properties of each segment were determined by assuming segment geometry and specifying proximal and distal ends of the segment with kinematic markers (Table 2). For the amputees, the residual shank mass was reduced to 2.325% body weight and the shank COM location was moved proximally such that it was 25% of the total knee-to-ankle distance distal from the knee. The whole-body angular momentum about the COM was determined as

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

where $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$, and $\vec{\omega}_i$ are the position, velocity, and angular velocity vectors of the *i*-th segment's COM, respectively, $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the body COM, respectively, m_i and I_i are the mass and moment of inertia of each segment, respectively, and n is the number of segments. Angular momentum was normalized by body mass (kg), walking speed (m/s), and body height (m) and expressed as a percentage of the residual gait cycle for amputees and the left gait cycle for non-amputees.

The range of each angular momentum component, defined as the peak-to-peak value, was compared between groups and across speeds using a two-factor ANOVA. The first between-subjects factor (group) had two levels: amputee and non-amputee. The second within-subjects factor (speed) had four levels: 0.6, 0.9, 1.2 and 1.5 m/s. The interaction effect between group and speed was also tested with the ANOVA. Mauchly's Test of Sphericity was used to test if the variance was significantly different across conditions. If the sphericity condition was violated, a Huynh–Feldt adjustment ($\varepsilon > 0.75$) or a Greenhouse–Geisser adjustment ($\varepsilon < 0.75$) was applied. When significant group, speed or interaction effects were found, pairwise comparisons using a Bonferroni adjustment were performed to determine which conditions were significantly different ($\alpha = 0.05$). Peak 3D GRFs and external moment arms (i.e., distance from body COM to COP normalized by body height) were similarly compared between amputees and non-amputees, as these quantities directly affect the external moment about the COM and therefore the time rate of change of angular momentum (Fig. 1). To identify quantities that were most strongly correlated with observed differences in angular momentum, Pearson correlation analyses were performed between the range of angular momentum and biomechanical variables that were significantly different between the amputee and non-amputee groups.

3. Results

3.1. Frontal plane

In the frontal plane, the range of angular momentum had significant group ($F = 8.577$, $p = 0.008$), speed ($F = 272.5$, $p < 0.001$)

Table 2
Mass percentage, geometry, and proximal and distal markers for each body segment in the model. For the thighs, the distance from the greater trochanter to the hip joint center was measured rather than using a medial marker. The mass of the residual leg shank was modified for the amputee model, and the percentage of total body mass for the amputee shank is shown in parentheses.

Segment	% of total body mass	Geometry	Proximal lateral maker	Proximal medial marker	Distal lateral marker	Distal medial marker
Torso	35.5	Cylinder	Right acromion process	Left acromion process	Right iliac crest	Left iliac crest
Pelvis	14.2	Cylinder	Right iliac crest	Left iliac crest	Right greater trochanter	Left greater trochanter
Thigh	10	Cone	Greater trochanter	–	Lateral femoral condyle	Medial femoral condyle
Shank	4.65 (2.325)	Cone	Lateral femoral condyle	Medial femoral condyle	Lateral malleolus	Medial malleolus
Foot	1.45	Cone	Lateral malleolus	Medial malleolus	Fifth metatarsal head	First metatarsal head

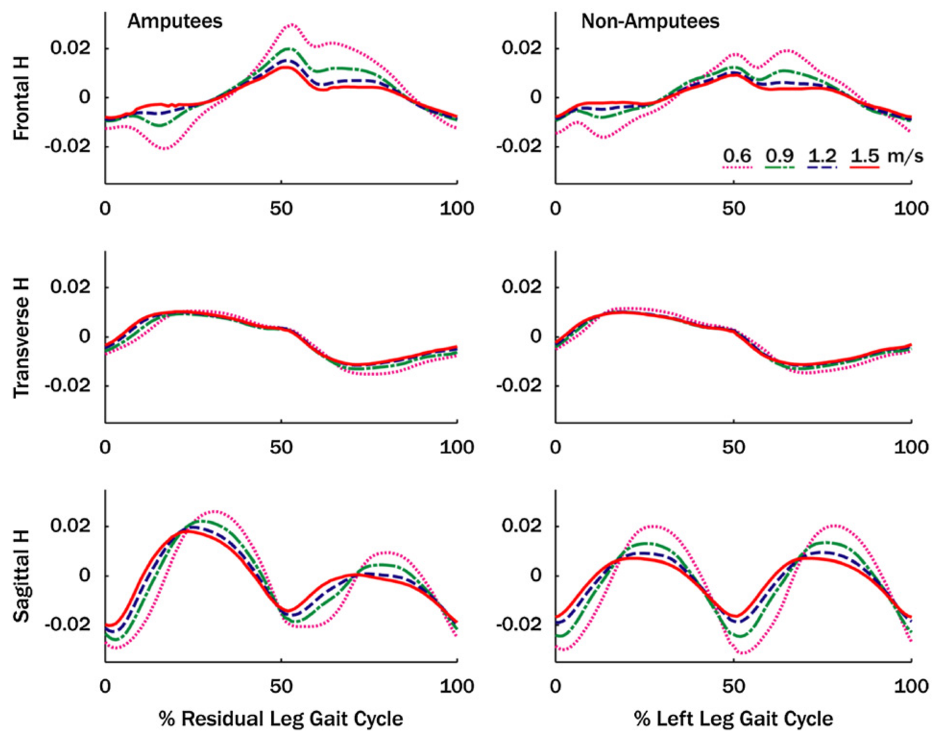

Fig. 2. Mean normalized 3D angular momentum (H) for amputee and non-amputee subjects over the gait cycle. Angular momentum was normalized by body mass, body height and walking speed.

and interaction ($F=7.701$, $p=0.007$) effects (Figs. 2 and 3). The range of angular momentum was larger in amputees compared to non-amputees at 0.6, 0.9 and 1.2 m/s (all $p \leq 0.013$) and decreased with each increase in walking speed for both amputees (all $p < 0.001$) and non-amputees (all $p \leq 0.022$).

For those quantities that contribute to the frontal-plane external moment (i.e., peak vertical and medial/lateral (M/L) GRFs and peak Y-moment and Z-moment arms, Fig. 1), the second vertical GRF peak in the residual and non-amputee legs had significant group ($F=29.703$, $p < 0.001$), speed ($F=22.674$, $p < 0.001$) and interaction ($F=8.329$, $p=0.003$) effects (Fig. 4, Table 3). The residual second vertical GRF peak was less than the non-amputees at all walking speeds (all $p \leq 0.002$). The intact and non-amputee second vertical GRF peaks had significant group ($F=25.268$, $p < 0.001$) and speed ($F=78.587$, $p < 0.001$) effects (Fig. 4, Table 3). The intact second vertical GRF peak was less than the non-amputees at all walking speeds (all $p \leq 0.001$).

The residual and non-amputee peak Y-moment arm also had significant group ($F=8.907$, $p=0.007$) and speed ($F=15.330$, $p < 0.001$) effects (Table 4). Similarly, the intact and non-amputee peak Y-moment arm had significant group ($F=10.863$, $p=0.004$) and speed ($F=17.174$, $p < 0.001$) effects (Table 4). The Y-moment arm was significantly greater in amputees compared to non-amputees at all walking speeds for both the residual (all $p \leq 0.013$) and intact (all $p \leq 0.007$) legs (Table 4).

Pearson correlation analyses were performed between the range of frontal-plane angular momentum and the second vertical GRF peaks. There was a correlation with the second residual leg GRF peak at 1.2 and 1.5 m/s ($-0.435 \leq r \leq -0.433$, $p \leq 0.044$, $n=22$) that also approached significance at 0.9 m/s ($r=-0.375$, $p=0.086$, $n=22$). Similarly, the range of frontal-plane angular momentum and the second intact vertical GRF peak were correlated at 0.9 and 1.5 m/s ($-0.520 \leq r \leq -0.429$, $p \leq 0.047$, $n=22$), and approached significance at 0.6 ($r=-0.366$, $p=0.094$, $n=22$) and 1.2 m/s ($r=-0.421$, $p=0.051$, $n=22$).

3.2. Transverse plane

In the transverse plane, there were no significant group or interaction effects. However, the range of angular momentum had a significant speed effect ($F=95.193$, $p < 0.001$, Figs. 2 and 3). The range of transverse-plane angular momentum decreased between 0.6 and 0.9 m/s ($p < 0.001$) and between 0.9 and 1.2 m/s ($p=0.004$).

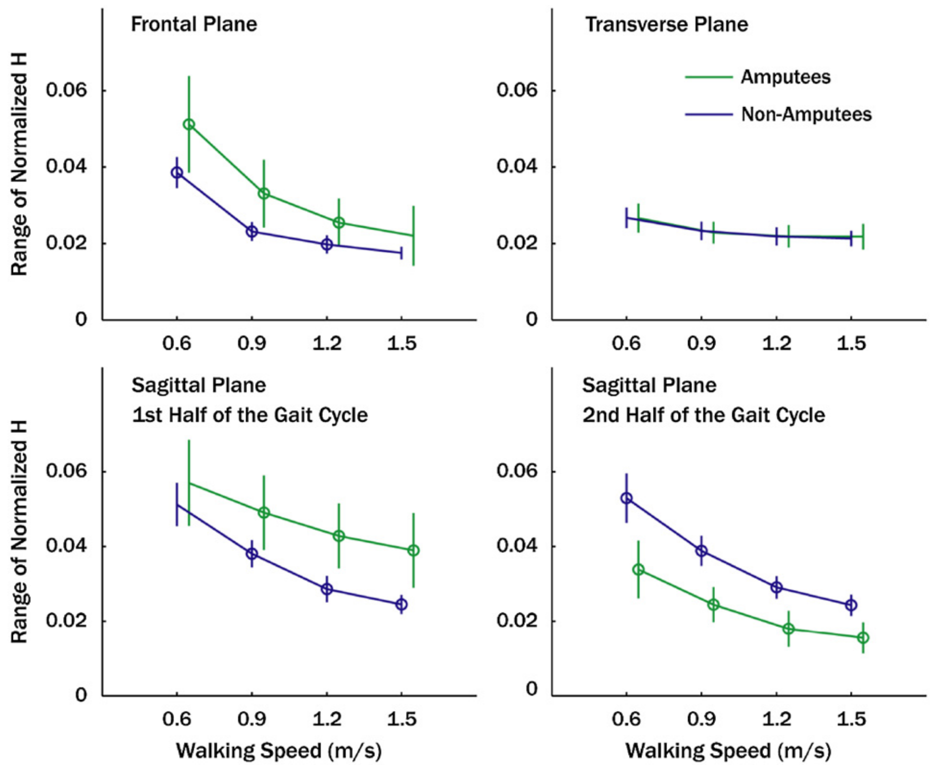

Fig. 3. Mean (σ) range of normalized 3D angular momentum (H). There were significant group, speed and interaction effects in the range of angular momentum in the frontal and sagittal planes. Significant differences between the amputee and non-amputee groups are indicated with an open circle (○).

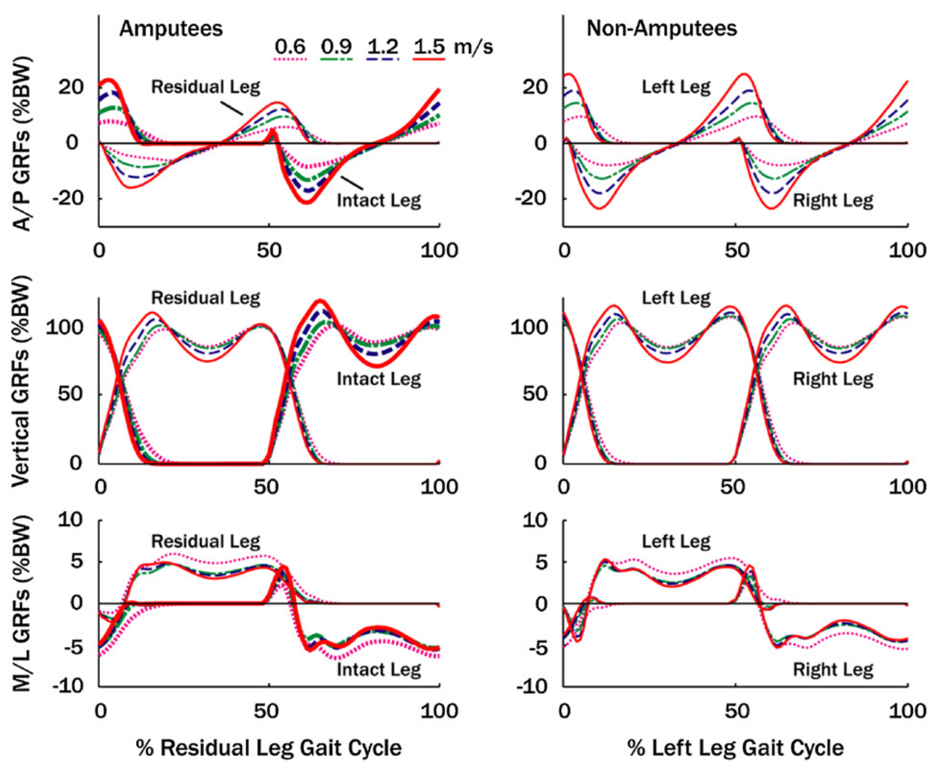

Fig. 4. Mean anterior/posterior (A/P), vertical and medial/lateral (M/L) ground reaction forces (GRFs) for amputees and non-amputees at all walking speeds, normalized to the residual and left leg gait cycles, respectively. The GRF quantities are normalized by body weight (BW). The heavier lines indicate the intact leg.

3.3. *Sagittal plane*

In the first half of the residual gait cycle, the range of angular momentum had significant group ($F=11.999$, $p=0.002$), speed ($F=341.093$, $p<0.001$) and interaction ($F=14.446$, $p<0.001$) effects (Figs. 2 and 3). The range was larger in the amputees compared to non-amputees at 0.9, 1.2 and 1.5 m/s (all $p \leq 0.004$). In the second half of the gait cycle, the range of angular momentum had significant

Table 3
Mean (σ) peak 3D ground reaction forces (GRFs) for the residual, intact, and non-amputee legs in percent body weight.

Speed (m/s)	0.6	0.9	1.2	1.5
Residual leg				
Peak propulsive GRF	6.478 (1.833)*	10.060 (1.736)*	12.661 (2.374)*	15.368 (3.364)*
Peak braking GRF	−7.122 (1.143)*	−10.102 (1.541)*	−13.839 (2.550)*	−17.534 (3.199)*
First peak vertical GRF	99.652 (4.028)	102.772 (4.546)	106.265 (4.267)	110.883 (6.194)
Second peak vertical GRF	100.160 (3.074)*	99.255 (3.637)*	100.762 (3.517)*	101.847 (4.247)*
First peak M/L GRF	6.077 (1.495)	4.829 (1.071)	5.146 (1.210)	5.587 (1.127)
Second peak ML GRF	5.859 (1.148)	4.737 (0.737)	4.670 (0.937)	4.462 (0.742)
Intact leg				
Peak propulsive GRF	8.144 (1.150)*	13.158 (1.368)*	18.601 (1.299)	23.452 (1.983)
Peak braking GRF	−8.723 (1.463)	−13.485 (1.893)	−17.274 (2.361)	−21.672 (2.641)
First peak vertical GRF	103.312 (3.003)	104.062 (4.855)	110.856 (5.249)	118.366 (5.392)
Second peak vertical GRF	98.595 (1.583)*	99.495 (2.325)*	103.003 (2.514)*	106.772 (3.819)*
First peak M/L GRF	6.702 (1.345)	5.390 (1.332)	5.745 (1.192)	5.949 (1.206)
Second peak ML GRF	6.392 (1.188)	5.394 (1.298)	5.556 (1.158)	5.558 (1.449)
Non-amputee average leg				
Peak propulsive GRF	9.981 (0.946)	14.948 (1.400)	19.527 (2.417)	25.722 (2.712)
Peak braking GRF	−8.441 (0.804)	−13.105 (1.374)	−18.504 (1.689)	−23.993 (1.978)
First peak vertical GRF	102.810 (3.674)	105.045 (4.036)	108.592 (6.300)	114.679 (6.041)
Second peak vertical GRF	106.007 (4.796)	106.841 (5.188)	109.293 (4.975)	113.950 (3.894)
First peak M/L GRF	5.542 (0.767)	4.668 (0.741)	5.127 (0.617)	5.453 (0.600)
Second peak ML GRF	5.535 (0.700)	4.655 (0.550)	4.676 (0.856)	4.558 (0.826)

* Significant differences with the non-amputee average leg.

Table 4
Mean (σ) peak 3D external moment arms normalized by body height at all walking speeds.

Speed (m/s)	0.6	0.9	1.2	1.5
Residual leg				
Anterior *X*-moment arm	0.134 (0.022)	0.162 (0.030)	0.177 (0.014)	0.196 (0.019)
Posterior *X*-moment arm	0.107 (0.029)*	0.136 (0.017)*	0.170 (0.017)*	0.200 (0.020)*
Y-moment arm	0.544 (0.016)*	0.544 (0.015)*	0.545 (0.016)*	0.545 (0.016)*
Z-moment arm	0.084 (0.020)	0.072 (0.019)	0.063 (0.012)	0.064 (0.024)
Intact leg				
Anterior *X*-moment arm	0.117 (0.026)	0.143 (0.022)	0.170 (0.033)	0.172 (0.032)
Posterior *X*-moment arm	0.110 (0.020)*	0.152 (0.021)	0.178 (0.020)	0.214 (0.018)*
Y-moment arm	0.544 (0.014)*	0.545 (0.013)*	0.545 (0.015)*	0.546 (0.015)*
Z-moment arm	0.089 (0.021)	0.072 (0.013)	0.064 (0.016)	0.061 (0.014)
Non-amputee average leg				
Anterior *X*-moment arm	0.120 (0.013)	0.141 (0.015)	0.172 (0.026)	0.188 (0.026)
Posterior *X*-moment arm	0.137 (0.029)	0.162 (0.024)	0.196 (0.022)	0.232 (0.016)
Y-moment arm	0.527 (0.009)	0.527 (0.009)	0.528 (0.009)	0.530 (0.010)
Z-moment arm	0.073 (0.021)	0.068 (0.023)	0.057 (0.012)	0.052 (0.014)

* Significant differences with the non-amputee average leg.

group ($F=61.396$, $p<0.001$), speed ($F=183.585$, $p<0.001$) and interaction ($F=8.492$, $p=0.003$) effects (Figs. 2 and 3). The range was smaller in the amputees compared to non-amputees at all four walking speeds (all $p<0.001$), resulting in a more negative (forward) angular momentum in this region. In both halves of the gait cycle, the range of angular momentum decreased with each increase in walking speed for both amputees (all $p\leq 0.014$) and non-amputees (all $p<0.001$).

For those quantities that contribute to the sagittal-plane external moment (i.e., anterior/posterior (A/P) and vertical GRFs and *X*-moment and *Y*-moment arms, Fig. 1), the residual and non-amputee peak braking GRFs had significant group ($F=30.889$, $p<0.001$), speed ($F=325.838$, $p<0.001$) and interaction ($F=12.285$, $p<0.001$) effects in addition to the differences found above in the vertical GRF peaks between groups (Fig. 4, Table 3). The residual peak braking GRF was less than the non-amputees at all walking speeds (all $p\leq 0.011$). The residual and non-amputee peak propulsive GRF also had significant group ($F=58.577$, $p<0.001$), speed ($F=295.275$, $p<0.001$) and interaction ($F=27.616$, $p<0.001$) effects (Fig. 4, Table 3). The residual peak propulsive GRF was less than the non-amputees at all walking speeds ($p<0.001$). The intact and non-amputee peak propulsive GRFs had significant group ($F=8.988$, $p=0.007$) and speed ($F=688.563$, $p<0.001$) effects (Fig. 4, Table 3), with a smaller intact peak propulsive GRF at 0.6, 0.9 and 1.5 m/s compared to the non-amputees (all $p\leq 0.023$).

In addition to the differences found in the *Y*-moment arm between groups, the posterior *X*-moment arm (i.e., *X*-moment arm from the trailing leg) also showed significant differences (Table 4). The residual and non-amputee posterior *X*-moment arms had significant group ($F=13.185$, $p=0.002$) and speed ($F=170.614$, $p<0.001$) effects. The non-amputee posterior *X*-moment arm was significantly greater than that of the residual leg at all walking speeds (all $p\leq 0.026$). The intact and non-amputee posterior *X*-moment arms also had significant group ($F=6.606$, $p=0.018$) and speed ($F=172.133$, $p<0.001$) effects, with the non-amputee moment arm significantly larger than the intact moment arm at 0.6 ($p=0.015$) and 1.5 m/s ($p=0.026$).

Pearson correlation analyses were performed in the first half of the residual (left) gait cycle with the peak residual braking GRF and the range of angular momentum. A correlation was found for 0.9, 1.2, and 1.5 m/s ($0.632 \le r \le 0.708$, $p \le 0.002$, $n=22$). The range of angular momentum and the second intact vertical GRF peak were also correlated at 0.9, 1.2, and 1.5 m/s ($-0.687 \le r \le -0.436$, $p \le 0.043$, $n=22$). The range of angular momentum and the intact posterior X-moment arm were correlated at 0.9 m/s only ($r=-0.475$, $p=0.026$, $n=22$).

In the second half of the residual gait cycle, the range of angular momentum and the peak residual propulsive GRF were correlated at all walking speeds ($0.571 \le r \le 0.775$, $p \le 0.006$, $n=22$). Similarly, the intact peak Y-moment arm, which occurred at approximately 80% of the residual gait cycle, was negatively correlated with the range of angular momentum at all walking speeds ($-0.530 \le r \le -0.466$, $p \le 0.029$, $n=22$).

4. Discussion

The purpose of this study was to investigate differences in whole-body angular momentum between amputees and non-amputees across walking speeds. The range of all three angular momentum components decreased for both amputees and non-amputees as walking speed increased (Figs. 2 and 3), which was consistent with previous non-amputee results (Bennett et al., 2010). However, there were significant interaction effects in the range of frontal- and sagittal-plane angular momentum. There were also interaction effects in the GRFs between amputees and non-amputees across walking speeds (Fig. 4, Table 3), which was consistent with previous amputee walking studies (Nolan et al., 2003; Sanderson and Martin, 1997; Silverman et al., 2008). The time rate of change of angular momentum is dependent on the GRFs; therefore, the interaction effect on the range of angular momentum across walking speed is consistent with the GRF results.

In the frontal plane, amputees had a greater range of angular momentum compared to non-amputees at 0.6, 0.9 and 1.2 m/s (Figs. 2 and 3). In the first half of the residual (left) gait cycle, the time rate of change in frontal-plane angular momentum, and therefore the frontal-plane external moment, was generally positive, and in the second half was generally negative (slope of the frontal-plane angular momentum, Fig. 2 upper left). A significantly smaller second intact vertical GRF peak, which would increase the positive time rate of change of the angular momentum, was found compared to non-amputees (Fig. 4, Table 3). In addition, this quantity had a negative correlation with the range of frontal-plane angular momentum at 0.9 and 1.5 m/s. Similarly, in the second half of the residual gait cycle, a more negative time rate of change of angular momentum would result from a reduced residual vertical GRF, which was smaller compared to non-amputees at all walking speeds and had a negative correlation with the range of angular momentum at 1.2 and 1.5 m/s. Thus, the greater range of amputee frontal-plane angular momentum may be a result of the reduced intact vertical GRF early in the residual gait cycle, and the reduced residual vertical GRF during the second half of the residual gait cycle.

In the first half of the residual gait cycle, the range of sagittal-plane angular momentum was higher in amputees compared to non-amputees at the three highest walking speeds (Figs. 2 and 3). At the beginning of the residual gait cycle, the residual leg is anterior to the body COM and provides braking, while the intact leg is posterior to the body COM and provides propulsion. Reduced residual braking would act to decrease the negative external moment on the body, and therefore increase the positive time rate of change of angular momentum (Fig. 1). The amputees had reduced residual peak braking compared to non-amputees at all walking speeds (Fig. 4, Table 3), and residual peak braking was correlated with the range of sagittal-plane angular momentum at the three highest speeds. In addition, a reduced intact vertical GRF and posterior X-moment arm would decrease the negative external moment on the COM. The intact second vertical GRF peak was reduced compared to non-amputees, and had a negative correlation with the range of angular momentum at the three highest walking speeds. The intact posterior X-moment arm was less than non-amputees at 0.6 and 1.5 m/s (Table 4), and was correlated with the range of angular momentum at 0.9 m/s only. Thus, of the GRF and moment arm components, reduced residual braking appears to be the most likely mechanism for the greater range of sagittal-plane angular momentum in amputees in the first half of the gait cycle. The reduced intact second vertical GRF peak and posterior X-moment arm may also be contributing to the greater range of angular momentum, but the correlations were not as strong compared to residual peak braking.

The range of sagittal-plane angular momentum was smaller in amputees compared to non-amputees in the second half of the residual gait cycle at all walking speeds (Figs. 2 and 3). During this region, the residual leg provides propulsion and is posterior to the body COM, while the intact leg provides braking and is anterior to the body COM. The amputees had a reduced residual peak propulsive GRF (Fig. 4, Table 3) compared to the non-amputees, which is consistent with previous studies (e.g., Sanderson and Martin, 1997; Zmitrewicz et al., 2006) and would decrease the positive external moment and corresponding positive rate of change in angular momentum. The residual posterior X-moment arm was smaller compared to non-amputees, and would oppose the effect of reduced residual propulsion on the angular momentum. The intact Y-moment arm was larger than non-amputees and negatively correlated with the range of angular momentum at all walking speeds. However, the correlation was stronger with the residual peak propulsive GRF. In addition, the peak Y-moment arm occurred at approximately 80% of the residual gait cycle, when the intact A/P GRF is transitioning from braking to propulsion, and the external moment is small. Thus, it appears that reduced residual propulsion is the most likely mechanism for the smaller range of angular momentum in the amputees in the second half of the gait cycle.

In amputees, the prosthetic foot type can influence the COP, and therefore the magnitude of the external moment arms (e.g, Hansen et al., 2004). In addition, the magnitude of the GRFs on both the intact and residual legs can be affected by prosthetic foot type (e.g., Hafner et al., 2002). Thus, different types of prosthetic feet may alter the overall whole-body angular momentum trajectories. However, the amputees in this study wore different prosthetic feet (Table 1), and the overall trends of the angular momentum, GRFs and external moment arms were similar across subjects. Thus, the differences in angular momentum between amputees and non-amputees are likely greater than differences resulting from different prosthetic feet.

Whole-body angular momentum has been shown to be small and highly regulated in non-amputee walking (Bennett et al., 2010; Herr and Popovic, 2008; Popovic et al., 2004; Robert et al., 2009). In amputees, the range of angular momentum was greater, and therefore not as tightly regulated, compared to non-amputees in the frontal and sagittal planes. Thus, larger angular momentum may result in a less-stable gait pattern, which has been suggested previously (Kaya et al., 1998; Rietdyk et al., 2005; Simoneau and Krebs, 2000). In addition, restraining angular momentum is important for preventing falls (e.g., Pijnappels et al., 2004, 2005b) and the ankle muscles are important in regulating angular momentum (Neptune and McGowan, 2010). Thus, reduced control at the residual ankle combined with greater frontal- and sagittal-plane angular momentum may explain why amputees have an increased risk of falling.

The results of this study suggest that reduced residual braking is an important mechanism to regulate sagittal-plane angular momentum over the gait cycle. Reduced residual propulsion was associated with more negative angular momentum compared to non-amputees in the second half of the gait cycle. Therefore, the increased range of angular momentum at the beginning of the gait cycle, which was associated with a reduced residual braking GRF, was necessary to conserve angular momentum over the gait cycle. However, the greater range of angular momentum suggests reduced stability in amputee walking. Therefore, reduced residual braking, while compensating for reduced residual propulsion to regulate whole-body angular momentum, may also contribute to a less-stable walking pattern in amputees.

An important limitation in this study is that arms were not included in the model, and therefore not included in the angular momentum calculation. However, previous studies have shown that swinging arms contribute mostly to angular momentum in the transverse plane (Bennett et al., 2010; Collins et al., 2009; Herr and Popovic, 2008), and that contributions of the arms to angular momentum in the sagittal and frontal planes during walking are small compared to other body segments (Bennett et al., 2010; Herr and Popovic, 2008). Thus, we do not expect this limitation would significantly affect the results in the sagittal and frontal planes. This limitation may more significantly affect the results in the transverse plane, although we expect arm movement between amputees and non-amputees to be similar, and therefore differences between groups to be minimal.

5. Conclusion

The results of this study highlight the differences in 3D angular momentum in amputees compared to non-amputees over the gait cycle and across walking speeds. In the frontal plane, a greater range of angular momentum was shown in amputees, which was related to a reduced intact vertical GRF at the beginning of the residual gait cycle and a reduced residual vertical GRF in the second half of the residual gait cycle. Reduced residual braking was associated with a greater range of sagittal-plane angular momentum in the first half of the gait cycle while reduced residual propulsion was associated with a smaller range of sagittal-plane angular momentum in the second half of the gait cycle. Thus, decreased residual braking appears to be an important mechanism to regulate sagittal-plane angular momentum in amputee walking, but was also associated with a greater range of angular momentum that may contribute to reduced stability in amputees. Future work is needed to understand individual muscle contributions to angular momentum so that rehabilitation programs can target specific muscles to improve amputee stability.

Conflict of interest statement

There is no conflict of interest regarding the publication of this manuscript.

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