

## Compensatory mechanisms of transtibial amputees during circular turning

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### ABSTRACT

Turning plays a prominent role in daily living activities and requires the modulation of the ground reaction forces to accelerate the body's center-of-mass along the path of the turn. With the ankle plantarflexors being prominent contributors to the propulsive ground reaction forces, it is not clear how transtibial amputees perform turning tasks without these important muscles. The purpose of this study was to identify the compensatory mechanisms used by transtibial amputees during a simple turning task by analyzing the radial and anterior–posterior ground reaction impulses and sagittal, transverse and coronal joint work of the residual and intact legs. These quantities were analyzed with the residual leg on both the inside and outside of the turn and compared to non-amputees. The analysis showed that amputees and non-amputees use different joint strategies to turn. Amputees rely primarily on sagittal plane hip joint work to turn while non-amputees rely primarily on ankle work in the sagittal plane and hip joint work in the coronal plane. Differences in strategies are most likely due to the minimal power output provided by the passive prosthetic feet used by amputees and perhaps a desire to minimize the risk of falling. Understanding these differences in turning strategies will aid in developing effective rehabilitation therapies and prosthetic devices that improve amputee mobility.

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### 1. Introduction

An important goal for successful rehabilitation of lower-limb amputees is to return patients to normal daily living activities, in which mobility tasks such as turning play a prominent role [1]. To redirect the center-of-mass, turning requires modulation of the radial ground reaction forces (GRFs) to accelerate the body's center-of-mass towards the center of curvature. Glaister et al. [2] and Strike and Taylor [3] analyzed 90° left and right turns performed by non-amputees and found that modulation of the radial ground reaction force impulse (GRI) is critical to successfully perform the turning tasks. Orendurff et al. [4] investigated joint kinematic and kinetic changes in non-amputees while walking along a 1 m radius circular path and found that relative to straight-line walking, the primary changes occurred in the radial GRI. In addition to redirecting of the center-of-mass, turning requires the rotation of the body towards the new heading [5,6], requiring further modulation of lower-limb joint moments. Non-amputee turning required similar magnitudes of plantarflexor power generation relative to straight-line walking, which suggests that reduced ankle push-off power would also impact turning gait.

Therefore, compensations in joint work are likely needed for amputees to perform a turn, especially since they are only able to minimally modulate the power output of passive foot-ankle prosthetic devices in response to changes in task demands (e.g., increased walking speed [7]).

Although analyses of three-dimensional joint power and work have provided much insight into the ability of non-amputees to both propel and control the lower limbs during straight-line walking [8–11], a limited number of studies have analyzed the three-dimensional joint powers in amputee gait [12]. Sadeghi et al. [13] examined three-dimensional peak powers at the hip, knee and ankle of transtibial amputees ( $n = 5$ ) during straight-line walking at their self-selected speed wearing a SACH foot and found differences in peak power between the intact and residual legs for the hip in all three planes, the knee in the sagittal and coronal planes and the ankle in the sagittal plane. Since turning likely requires even larger modulations in the other planes of motion, analyzing three-dimensional joint work for each leg may provide further insight into the compensatory mechanisms required for amputees to complete a turning task.

Few studies have analyzed amputee turning. Segal et al. [14] compared the transverse plane joint moments and radial GRIs between amputees and non-amputees while walking along a 1 m radius circular path at the same speed. They found that amputees had decreased internal rotation moments at the residual leg hip

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and knee, which may result from a protective response to reduce stress at the prosthesis-skin interface and facilitate the change in orientation. Perhaps as compensation, the amputees increased the external rotation moment for their intact leg hip compared to non-amputees. Amputees also decreased their inside leg radial GRI compared to non-amputees, possibly to minimize the center-of-mass acceleration in the direction of the turn and maximize stability. The prescription of a torsion adapter, a device with an adjustable transverse plane stiffness, further reduced the internal rotation moments at the knee and hip while on the inside of a turn [15]. However, neither study analyzed the anterior–posterior GRIs nor explored how amputees modulate their joint powers and work to generate needed GRIs in the anterior–posterior or radial directions.

The purpose of this study was to identify the compensatory mechanisms used by amputees during turning by analyzing the anterior–posterior and radial GRIs and the hip (sagittal, transverse and coronal), knee (sagittal) and ankle (sagittal and coronal) joint power and work of the residual and intact legs. These measures were chosen after considering the structure and functional capabilities of each lower leg joint and analyzed with the residual leg on both the inside and outside of the turn and compared to non-amputee turning. In order to avoid confounding effects associated with transient turns, we had the subjects complete a steady-state turn about a defined circle. We tested the general hypothesis that there would be differences in the GRIs and joint work between the intact, residual and non-amputee legs regardless of whether the leg was on the inside or outside of the turn. Understanding how transtibial amputees compensate to successfully perform the turning task will provide needed insight for designing effective rehabilitation therapies focused on specific muscle groups and prosthetic interventions intended to improve amputee mobility.

## 2. Methods

### 2.1. Participants

Ten unilateral transtibial amputee participants (9 males, 1 female, age =  $56 \pm 12$  yrs, height =  $1.79 \pm 0.08$  m, mass =  $88 \pm 11$  kg) and ten speed-matched non-amputee participants (6 males, 4 females, age =  $44 \pm 14$  yrs, height =  $1.73 \pm 0.10$  m, mass =  $78 \pm 21$  kg) participated in this institutional review board-approved study. All amputees considered themselves moderately active community ambulators, had walked with a prosthesis for at least two years, wore their prosthesis at least 8 h a day, and were free from known neurological or musculoskeletal disorders. All amputee participants wore their clinically prescribed prosthetic components (Table 1) and provided informed consent prior to the study.

**Table 1**

Prosthetic components for transtibial amputee participants. Total contact patellar tendon bearing (PTB) sockets, rigid pylons, and Alpha<sup>®</sup> liners<sup>a</sup> were used by all participants, except subject 8 who used the Iceross<sup>®</sup> liner.<sup>g</sup>

Subject	Cause of amputation	Socket	Liner	Suspension	Foot/Ankle	Socks
1	Traumatic	Carbon laminated	6 mm uniform cushion	sleeve	Endolite Dynamic Response 2/ Multiflex Ankle <sup>b</sup>	Knit-Rite <sup>™</sup> 2 ply <sup>h</sup>
2	Vascular	Carbon laminated	6 mm tapered cushion	sleeve	Freedom Innovation Runway <sup>c</sup>	Royal Knit <sup>™</sup> 5 ply <sup>h</sup>
3	Diabetic infection	Thermal plastic	9 mm uniform locking	pin	FS-3000 <sup>c</sup>	Knit-Rite <sup>™</sup> 5 ply <sup>h</sup>
4	Traumatic	Carbon laminated	6 mm uniform locking	pin	FS-1000 <sup>c</sup>	None
5	Diabetic/Vascular	Carbon laminated	9 mm uniform locking	pin	Dynamic Response/ Multi-axial ankle <sup>d</sup>	None
6	Traumatic	Carbon laminated	6 mm uniform locking	pin	Genesis <sup>e</sup>	Knit-Rite <sup>™</sup> 3 ply <sup>h</sup>
7	Traumatic	Carbon laminated	6 mm contoured locking	pin	Seattle Lite Foot <sup>f</sup>	None
8	Traumatic	Carbon laminated	3 mm Iceross <sup>®</sup>	pin	Vari-Flex Low Profile <sup>g</sup>	Knit-Rite <sup>™</sup> 1 ply <sup>h</sup>
9	Diabetic infection	Carbon laminated	9 mm uniform locking	pin	Low Profile Renegade <sup>c</sup>	Knit-Rite <sup>™</sup> 5 ply, 1 ply <sup>h</sup>
10	Tumor (cancer)	Carbon laminated	6 mm contoured locking	pin	Low Profile FS2000 <sup>c</sup>	Knit-Rite <sup>™</sup> 5 ply <sup>h</sup>

<sup>a</sup> Ohio Willow Wood, Mt. Sterling, OH.

<sup>b</sup> Blatchford, Endolite North America, Centerville, OH.

<sup>c</sup> Freedom Innovation Inc, Irvine, CA.

<sup>d</sup> College Park Industries, Fraser, MI.

<sup>e</sup> Genesis Prosthetic Arts, Howell, MI.

<sup>f</sup> Seattle Systems, Poulso, WA.

<sup>g</sup> Ossur America's, Aliso Viejo, CA.

<sup>h</sup> Knit Rite Inc., Kansas City, KA.

### 2.2. Data collection

Thirty-eight reflective markers were placed bilaterally on the arms, legs, trunk and head of the participants, consistent with Vicon's Plug-In-Gait model (Oxford Metrics, Oxford, England). A 10-camera Vicon 612 system collected kinematic data at a minimum of 120 Hz. GRFs were collected at a minimum of 1200 Hz from two consecutive Bertec force plates (Columbus, OH) embedded in the walkway. Raw marker trajectories were filtered using Vicon's Woltring quintic spline algorithm with MSE value set to 20.

Participants walked at their self-selected speed along a 1 m radius circular path while keeping their body centered over the path marked on the floor. Participants were told that stepping on the line was allowed and practiced the task before data collection. A constant speed 1 m radius circular path was chosen to explore the mechanisms of turning because it represented a typical turn radius found in daily activities (e.g., a 90° hallway turn) but minimized the confounding effects of variable speed observed during non-circular turning gait [16]. Each discrete trial consisted of participants walking clockwise around the 1 m radius circular path starting a few strides from the force plates and ending a few strides after the plates in order to achieve a steady-state pattern throughout force plate contact. The discrete trials were repeated until three force plate hits per leg were measured, and again for the counter-clockwise direction. Subjects were allowed to rest between trials and no subject reported feelings of dizziness. Walking speed was measured for each turning trial by averaging the instantaneous velocity of the outside leg's posterior superior iliac spine marker across the time period when subjects walked across the force plates.

### 2.3. Data analysis

Kinematic and GRF data were low-pass filtered in Visual 3D (C-Motion Inc., Germantown, MD) using a fourth-order Butterworth filter with cut-off frequencies of 6 Hz and 20 Hz, respectively. Intersegmental joint angles and moments were determined from marker trajectories using standard inverse kinematics and dynamics techniques (e.g., [17]). Joint powers were calculated as the product of the three-dimensional joint moments and corresponding angular velocities. Positive (negative) joint work was calculated as the time integral of the positive (negative) joint power over the gait cycle.

A rotating reference frame was defined with the origin at the center-of-mass, the *x*-axis through the center of the 1 m radius circular path and the *y*-axis (forward progression) perpendicular to the *x*-axis and tangent to the circle which the subjects were following [14]. Positive radial GRFs (*x*-axis) were defined in the direction of the circle center. Positive anterior–posterior GRFs (*y*-axis) were defined in the anterior direction. Positive (negative) GRIs were calculated as the time integral of the positive (negative) GRFs over the gait cycle.

Amputee data were grouped according to leg placement with respect to the circular path (condition). Right-leg amputee data from the clockwise trials were grouped with left-leg amputee data from the counter-clockwise trials to obtain *Residual Inside* and *Intact Outside* data. The *Residual Outside* and *Intact Inside* data were similarly obtained. In non-amputee gait, the right leg has been cited as the dominant and mobilizing leg [18,19]; therefore, both amputee legs were compared to the right leg of non-amputees. *Non-amputee Outside* data was obtained from counter-clockwise trials and *Non-amputee Inside* data was obtained from clockwise trials.

2.4. Statistical analysis

Statistical analysis was performed using PAWS Statistics 18 (IBM Corp., Somers, NY). Two-factor (leg and condition), repeated-measures analysis of variances (ANOVAs) were used to compare the GRI and joint work of the residual, intact and non-amputee legs across conditions. The 'leg' factor had two levels and all three combinations of legs were compared (residual with intact, residual with non-amputee, and intact with non-amputee leg). The 'condition' factor had two levels corresponding to the legs' two conditions (inside and outside of the turn). The ANOVAs comparing the residual and intact legs had 'within-subjects' comparisons across all levels (leg and condition), whereas the ANOVAs comparing the non-amputee values to either the residual or intact legs had 'within subjects' comparisons across conditions but 'between-subjects' comparisons across legs. When significant differences were found ( $p < 0.05$ ), pair-wise comparisons with a Bonferroni adjustment for multiple comparisons were used to determine which conditions were significantly different from each other.

3. Results

The average self-selected speed along the 1 m radius circular path was  $0.90 \pm 0.09$  m/s for amputees and  $0.92 \pm 0.06$  m/s for non-amputees.

3.1. Ground reaction force impulses

The propulsive impulses had a significant leg effect for both conditions, with the residual leg reduced compared to the intact and non-amputee legs ( $p = 0.017$  and  $p < 0.001$ , Fig. 1). There was a significant condition effect for all legs, with the outside impulse greater than the inside ( $p = 0.017$ ). The braking impulses had a significant condition effect for the intact and non-amputee legs, with the inside impulse greater than the outside ( $p = 0.034$  and  $p = 0.044$ ). The inward radial impulses had a significant condition effect for all legs, with the outside impulse greater than the inside ( $p < 0.001$ ). For the inside condition, the non-amputee leg had greater inward radial impulses than the residual and intact legs ( $p = 0.002$  and  $p = 0.008$ ). The outward radial impulses had a

significant condition effect for the residual and intact legs, with the outside impulse lower than the inside ( $p = 0.036$  and  $p = 0.006$ ). For the inside condition, the non-amputee leg had a lower outward radial impulse than the residual and intact legs ( $p = 0.026$  and  $p = 0.026$ ).

3.2. Hip work

Positive hip work in the sagittal plane had a significant leg effect for the inside condition, with the residual leg greater than the intact and non-amputee legs ( $p = 0.036$  and  $p = 0.001$ , Fig. 2). There was a significant condition effect for the residual and non-amputee legs, with the inside greater than the outside ( $p = 0.010$  and  $p = 0.048$ ). Negative hip work in the sagittal plane had a significant leg effect for the inside condition, with the non-amputee leg greater than the residual leg ( $p = 0.012$ ). Positive hip work in the coronal plane had a significant leg effect for the inside condition, with the non-amputee leg greater than the residual and intact legs ( $p = 0.002$  and  $p = 0.039$ ). The non-amputee leg was also greater than the intact leg for the outside condition ( $p = 0.034$ ). There was a significant condition effect for the intact and non-amputee legs, with the inside greater than the outside condition ( $p = 0.001$  and  $p < 0.001$ ). Negative hip work in the coronal plane also had a significant leg effect for the inside condition, with the residual leg lower than both the intact and non-amputee legs ( $p = 0.042$  and  $p = 0.008$ ), and for the outside condition, with the intact leg greater than both the residual and non-amputee legs ( $p = 0.020$  and  $p = 0.046$ ). There was a significant condition effect for the non-amputee leg, with the inside greater than the outside condition ( $p = 0.001$ ). There were no significant differences in the transverse plane.

3.3. Knee work

There were no significant differences in knee work (Fig. 3).

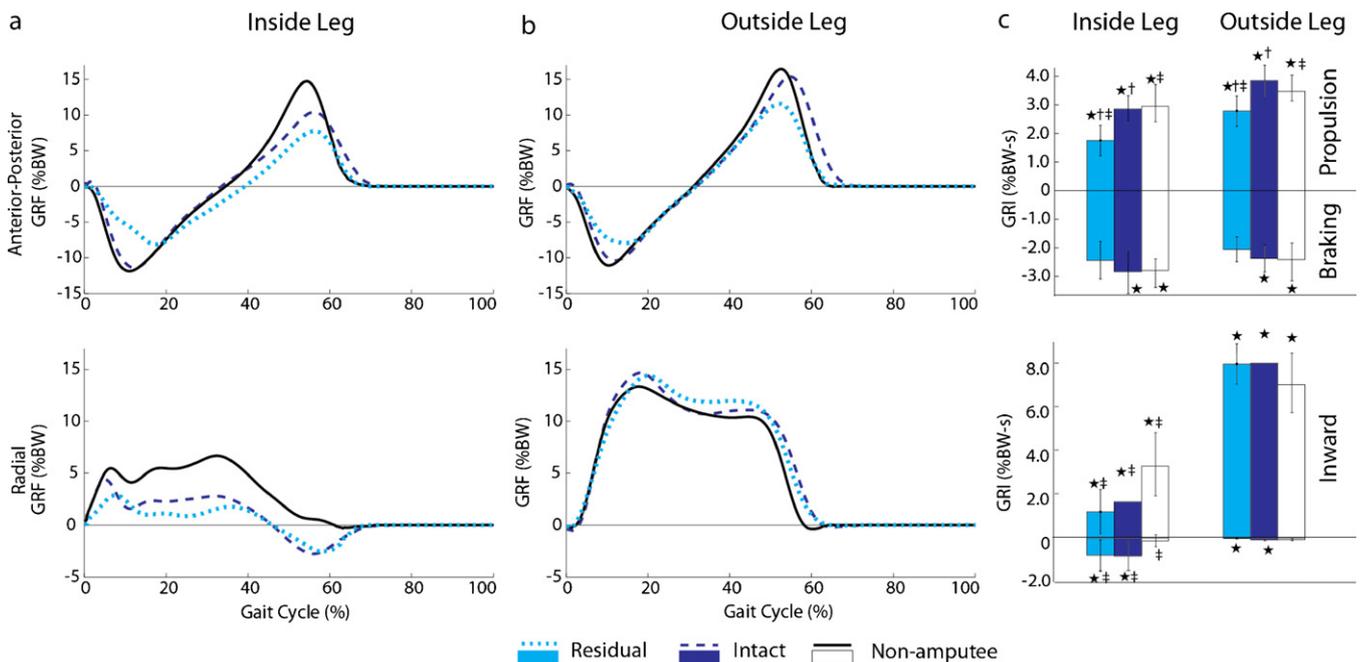
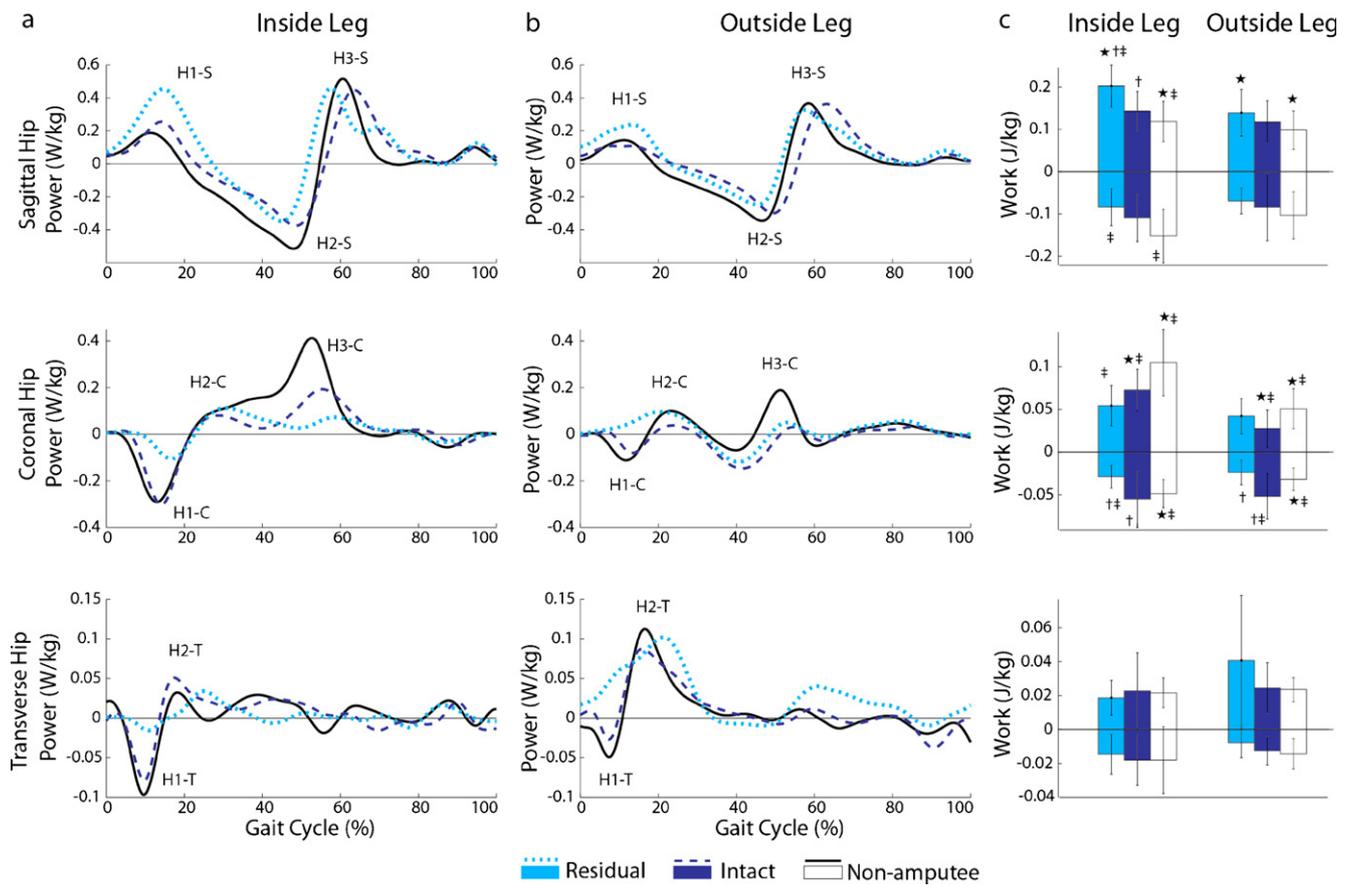
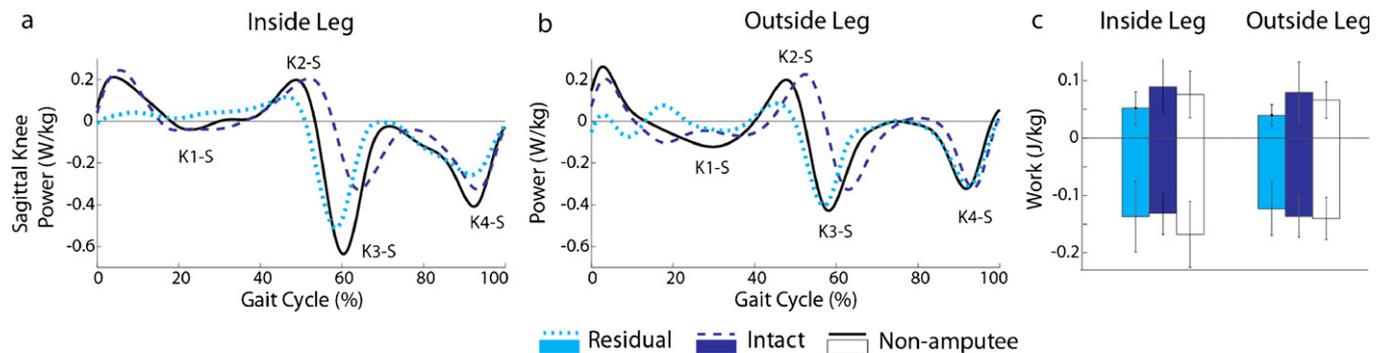


Fig. 1. (a) Average anterior–posterior and radial ground reaction forces (GRF) for the inside leg; (b) average anterior–posterior and radial GRFs for the outside leg; and (c) average ground reaction force impulses (GRI) for the inside and outside legs. Averages are presented for the residual and intact legs of 10 amputees and the right (non-amputee) legs of 10 non-amputees when the leg is inside and outside of a 1 m turn. Significant differences between inside and outside conditions for a given leg are denoted with  $\star$ ; significant differences between residual and intact legs for a given condition are denoted with  $\dagger$  and differences between the non-amputee leg and either residual or intact legs with a  $\ddagger$ .



**Fig. 2.** (a) Average hip joint powers in the sagittal, coronal and transverse planes for the inside leg; (b) average hip joint powers for the outside leg; and (c) average hip joint work for the inside and outside legs. Averages are presented for the residual and intact legs of 10 amputees and the right (non-amputee) legs of 10 non-amputees when the leg is inside and outside of a 1 m turn. Significant differences between inside and outside conditions for a given leg are denoted with ★; significant differences between residual and intact legs for a given condition are denoted with † and differences between the non-amputee leg and either residual or intact legs with a ‡.



**Fig. 3.** (a) Average knee joint powers in the sagittal plane for the inside leg; (b) average knee joint powers for the outside leg; and (c) average knee joint work for the inside and outside legs. Averages are presented for the residual and intact legs of 10 amputees and the right (non-amputee) legs of 10 non-amputees when the leg is inside and outside of a 1 m turn. Significant differences between inside and outside conditions for a given leg are denoted with ★; significant differences between residual and intact legs for a given condition are denoted with † and differences between the non-amputee leg and either residual or intact legs with a ‡.

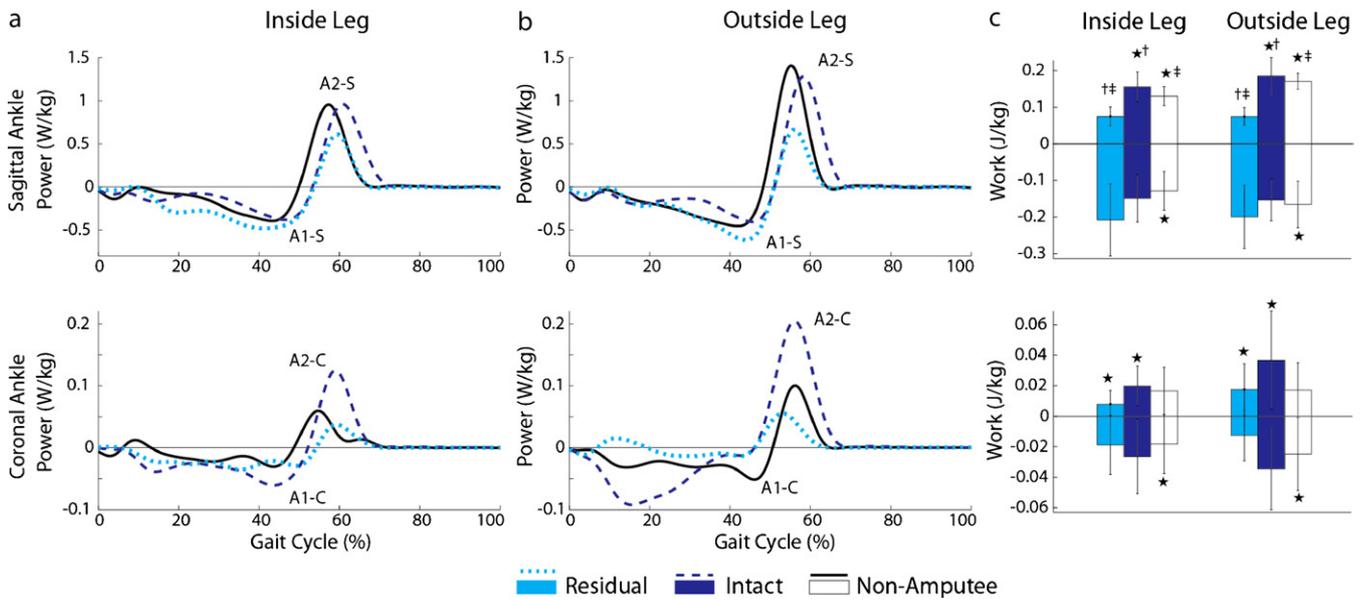
### 3.4. Ankle work

Positive ankle work in the sagittal plane had a significant leg effect for both conditions, with the residual leg lower than the intact and non-amputee legs ( $p = 0.005$  and  $p = 0.005$ , Fig. 4). There was a significant condition effect for the intact and non-amputee legs, with the inside lower than the outside ( $p = 0.005$  and  $p = 0.012$ ). Negative ankle work in the sagittal plane had a significant condition effect for the non-amputee leg, with the inside lower than the outside ( $p = 0.019$ ). Positive ankle work in the coronal plane had a significant condition effect for the residual and intact legs,

with the inside lower than the outside ( $p = 0.009$  and  $p = 0.034$ ). Negative ankle work in the coronal plane had a significant condition effect for the non-amputee leg, with the inside lower than the outside ( $p = 0.038$ ).

## 4. Discussion

The analysis of the GRIs and joint work indicates that amputees and non-amputees use different strategies to perform a turning task. Non-amputees primarily performed the turning task by increasing the propulsive impulse and decreasing the braking



**Fig. 4.** (a) Average ankle joint powers in the sagittal and coronal planes for the inside leg; (b) average ankle joint powers for the outside leg; and (c) average ankle joint work for the inside and outside legs. Averages are presented for the residual and intact legs of 10 amputees and the right (non-amputee) legs of 10 non-amputees when the leg is inside and outside of a 1 m turn. Significant differences between inside and outside conditions for a given leg are denoted with ★; significant differences between residual and intact legs for a given condition are denoted with † and differences between the non-amputee leg and either residual or intact legs with a ‡.

impulse of the outside leg relative to the inside leg while generating an inward directed radial GRI with both legs (Fig. 1). However, amputees generated GRIs with their inside and outside legs differently. Regardless of whether the residual leg was on the inside or outside, the propulsive impulses were lower in the residual leg relative to both the intact and non-amputee legs, likely due to the lost ankle muscles. With regard to the inward radial GRI, the amputees generated a lower inward radial GRI with their inside leg relative to non-amputees, primarily from mid to late stance (Fig. 1). This reduction in inward radial impulse for both the intact and residual legs may be due to amputees not leaning into the turn as much as the non-amputees to minimize the excursion of the center-of-mass outside of the base of support and maximize stability during single-leg support and push-off [4].

Amputees also modulated their joint work differently than non-amputees. In the sagittal plane, residual leg positive hip work was greater than the intact and non-amputee legs for the inside condition (Fig. 2). This compensatory mechanism occurred primarily in early stance (H1-S), where greater concentric hip extensor work was likely compensating for decreased residual leg ankle push-off later in stance, similar to previous analyses of amputee straight-line walking [7,20–23].

In contrast, amputees generally had lower hip joint work in the coronal plane relative to non-amputees as a result of reduced power absorption in early stance (H1-C) and generation in late stance (H3-C). The negative residual leg coronal work was reduced compared to the intact leg for both the inside and outside conditions. This decrease in negative coronal hip work occurred during early stance (H1-C), likely due to the power being redirected to the sagittal plane (H1-S). The residual leg positive coronal hip work was also reduced compared to the intact leg for the inside condition. The residual leg hip appears to absorb less energy in the coronal plane in early stance relative to non-amputees and the intact leg during turning. The coronal work was also less in late stance for the inside condition.

There were no differences in hip joint work between amputees and non-amputees in the transverse plane. However, the residual leg did not demonstrate the initial H1-T absorption in early stance that controlled rotation in the intact and non-amputee legs (Fig. 2).

The absence of this power burst from the residual leg would allow the pelvis to continue to rotate about the inside leg while being propelled by the outside (intact) leg. The lack of statistical significance in the transverse plane may be due to larger variability, especially for the residual outside leg, which is consistent with other gait studies [10,24].

Thus, it appears non-amputees perform the turning task by primarily modulating hip power in the coronal plane, while amputees modulate their residual leg hip power in the sagittal plane. Differences in hip joint work between the inside and outside legs in the sagittal plane would turn the body much like a vehicle where the outside wheel rotates more than the inside wheel, rather than by tilting the body as would be accomplished by coronal plane hip joint work. It is interesting to note the range (peak-to-peak value) of coronal plane angular momentum is greater in amputees relative to non-amputees during straight-line walking [25], which may indicate a less stable gait pattern. Amputees already have an increased fear of falling relative to non-amputees [26], and thus may adopt a turning strategy that helps minimize the risk of falling. Future work is needed to assess how the different strategies used by amputees and non-amputees influence whole-body angular momentum and overall dynamic stability.

Consistent with studies of straight-line walking, amputees generated lower sagittal plane ankle joint work in the residual leg relative to non-amputees and their intact leg (Fig. 4). This was consistent with the lower propulsive GRIs generated by the residual leg (Fig. 1) since the ankle plantarflexors are the primary contributors to the propulsive GRI (e.g., [27,28]). Non-amputees and the amputees' intact leg had higher ankle joint work for the outside condition relative to the inside condition, but the residual leg ankle joint work did not vary between conditions. These results suggest that amputees are unable to modulate sagittal plane power of their foot-ankle prosthesis during turning, which is consistent with previous studies of straight-line walking [7,29–31].

An area for future work is to analyze the compensatory mechanisms associated with transient turns. In daily life, transient turns with initiation and termination are more common than steady-state turning steps [1]. Strategies used by subjects with respect to foot placement and the timing of body orientation

during transient turns can provide insight into needed stability and control mechanisms [5,6]. Thus, to better understand the mechanisms used by amputees, analyzing joint power and work during transient turns would be a fruitful area for future research.

In summary, the analysis showed that amputees and non-amputees use different strategies to perform a simple turning task. Amputees rely primarily on sagittal plane hip joint work to turn while non-amputees rely primarily on ankle work in the sagittal plane and hip joint work in the coronal plane. Differences in strategies are most likely due to the minimal power output provided by the passive prosthetic feet used in this study and perhaps the desire to minimize the risk of falling. Future work is needed to identify which muscles contribute to the joint work modulation and are directly responsible for accelerating the body center-of-mass throughout the turn. Such analyses will further aid in developing effective rehabilitation therapies and prosthetic devices that improve amputee mobility and reduce the risk of falling.

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### Conflict of interest

None

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