Hip recovery strategy used by below-knee amputees following mediolateral foot perturbations

Sarah E. Miller [a], Ava D. Segal [b], Glenn K. Klute [b,c], Richard R. Neptune [a,*]

[a] Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX, USA
[b] Department of Veterans Affairs, Puget Sound Health Care System, Seattle, WA, USA
[c] Department of Mechanical Engineering, University of Washington, Seattle, WA, USA

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ABSTRACT

Lower-limb amputees have a higher risk of falling compared to non-amputees. Proper regulation of whole-body angular momentum is necessary to prevent falls, particularly in the frontal plane where individuals are most unstable. However, the balance recovery mechanisms used by lower-limb amputees when recovering from a perturbation are not well-understood. This study sought to understand the balance recovery mechanisms used by lower-limb amputees in response to mediolateral foot perturbations by examining changes to frontal plane whole-body angular momentum and hip joint work. These metrics provide a quantitative measure of frontal plane dynamic balance and associated joint contributions required to maintain balance during gait. Nine amputees and 11 non-amputees participated in this study where an unexpected medial or lateral foot placement perturbation occurred immediately prior to heel strike on the residual, sound or non-amputee limbs. Lateral perturbations of all limbs resulted in a reduced range of whole-body angular momentum and increased positive frontal plane hip work in the first half of single limb support. Medial perturbations for all limbs resulted in increased range of whole-body angular momentum and decreased positive frontal plane hip work, also in the first half of single limb support. These results suggest that medial foot placement perturbations are particularly challenging and that hip strategies play an important role in balance recovery. Thus, rehabilitation interventions that focus on hip muscles that regulate mediolateral balance, particularly the hip abductors, and the use of prostheses with active ankle control, may reduce the risk of falls.

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1. Introduction

Approximately 1.6 million Americans were living with limb loss in 2005 and this number is expected to grow to 3.2 million by 2050, with lower-limb amputees making up 39% of this population (Ziegler-Graham et al., 2008). Falls are a common source of injury with more than 50% of amputees falling each year, which is 20% more frequent than non-amputees (Miller et al., 2001). Most of these falls occur while walking (Tinetti et al., 1995) during which individuals are more unstable in the mediolateral direction (Bauby and Kuo, 2000; Kuo, 1999). Research has shown the majority of same level falls occur due to trips or unexpected perturbations (Chang et al., 2016). Thus, understanding the balance recovery mechanisms used by amputees to recover from such perturbations could provide insight into developing rehabilitation strategies aimed at decreasing their risk for falls and injuries.

Whole-body angular momentum, which is the segmental sum of angular momentum about the body's center-of-mass, is a commonly used measure to assess dynamic balance during human locomotion (e.g., Herr and Popovic, 2008; Pijnappels et al., 2004; Simoneau and Krebs, 2000). Previous studies have shown that lower-limb amputees walk with a larger range of frontal plane angular momentum compared to non-amputees (D'Andrea et al., 2014; Silverman and Neptune, 2011) and during pseudo-random mediolateral platform oscillations (Sheehan et al., 2015). A larger range in frontal plane angular momentum has been correlated with reduced second vertical ground reaction force (GRF) peaks (Silverman and Neptune, 2011), greater step widths (Vistamehr et al., 2016) and lower clinical balance scores (Nott et al., 2014).

Whole-body angular momentum is largely dictated by foot placement and the corresponding GRFs. The ankle plantarflexors have been shown to be primary contributors to GRFs (e.g., John et al., 2012; Neptune et al., 2004) and are essential for performing

the biomechanical subtasks of walking such as body support, forward propulsion and balance control (e.g., Anderson and Pandy, 2003; Neptune et al., 2001; Pandy et al., 2010). Thus, the functional loss of the residual limb ankle muscles in lower-limb amputees likely contributes to the increased challenge of regulating their angular momentum compared to non-amputees. Prior study of the effect of medial and lateral step width perturbations on amputee balance control revealed that a sudden decrease in residual limb step width required three additional recovery steps compared to when the disturbance originated with the sound limb or for non-amputees (Segal and Klute, 2014). This delayed recovery was likely influenced by the absence of an immediate mediolateral shift in residual center-of-pressure that was present for the sound limb and non-amputees (Segal et al., 2015). However, the corresponding effect on the range of whole-body angular momentum remains unknown. Since the time rate of change of whole-body angular momentum depends on the moment arm from the center-of-mass to the center-of-pressure (i.e., the step width), we would expect that an increase (decrease) in step width would result in an increase (decrease) in the range of whole-body angular momentum.

During straight-line walking the hip abductors, particularly the gluteus medius, act to rotate the body towards the ipsilateral limb and counteract the effects of gravity (Neptune and McGowan, 2016). Wider (narrower) steps have been correlated with an increase (decrease) in gluteus medius muscle activity (Hof and Duysens, 2013; Kubinski et al., 2015). Without the use of the ankle muscles in amputee walking, studies have shown that hip muscle work is an effective compensatory mechanism during straight-line walking (Silverman et al., 2008) and turning (Ventura et al., 2011). Segal and Klute (2014) found that during medial perturbations of the residual limb, the center-of-mass surpassed the lateral edge of the foot which decreased the peak base of support and reduced their stability. In contrast, the base of support increased during lateral perturbations (Segal and Klute, 2014). Thus, to return the base of support to undisturbed levels with the lateral perturbations, we expect the primary compensatory mechanism will be muscle work from the perturbed limb hip abductors that act to rotate the body towards the perturbed limb. In contrast, during the medial perturbations we expect a decrease in muscle work from the perturbed limb hip abductors.

With these expectations, the purpose of this study was to understand the balance recovery mechanisms used by lower-limb amputees in response to mediolateral foot perturbations. Specifically, we examined the relationships between changes in dynamic balance, quantified using whole-body angular momentum, and corresponding hip joint work.

2. Methods

The data collection methods were previously described in detail (Segal and Klute, 2014) and will be briefly presented. Nine unilateral transtibial amputees (all male; height: 1.84 ± 0.07 m; body mass: 86.0 ± 16.0 kg; age: 47 ± 16 years; leg length: 0.94 ± 0.04 m) and eleven non-amputees (9 males; height: 1.77 ± 0.07 m; body mass: 80.5 ± 14.4 kg; age: 40 ± 13 years; leg length: 0.92 ± 0.04 m) free of neurological deficits and musculoskeletal disorders gave informed consent to participate in this IRB-approved study. All amputees were fit and aligned with the same prosthetic foot (Highlander, FS3000, Freedom Innovation Inc., Irvine, CA) appropriate for their body weight and activity level by a certified prosthetist. Subjects walked at their self-selected walking speed (amputees: 1.20 ± 0.1 m/s, non-amputees: 1.23 ± 0.1 m/s) on a split-belt instrumented treadmill (Bertec, Columbus, OH) while wearing a perturbation device that was attached to the ankle of interest. The

perturbation device was a pneumatic system designed to impose a repeatable mediolateral shift in foot placement of approximately 50 mm just prior to heel strike. Data collected from a force sensor on the bottom of the participant's shoe was used to calculate stride time. The stride time was then used to determine the timing delay required for a solenoid valve to release a medial or lateral airburst ~135 ms prior to heel strike and provide the perturbation. For more information on the perturbation device and protocol, please see Segal and Klute (2014). Six conditions (i.e., a medial or lateral perturbation on the residual, sound, or non-amputee limb) were collected. Approximately five random repeated trials per condition were collected from all participants. Subjects were aware of the perturbation direction but not the timing.

Kinematic and GRF data were collected at 120 Hz and 1200 Hz, respectively, using a 12-camera motion capture system (Vicon Motion Systems, Centennial, CO) and a split-belt instrumented treadmill (Bertec, Columbus, OH). Thirty-five 14 mm reflective markers were placed according to Vicon's Plug-in-Gait full-body model on each participant. Prosthetic foot marker placements were symmetric with the sound foot placement.

Marker and GRF data were filtered with Vicon's Woltring quintic spline algorithm with a mean-square-error value of 20. The data were then filtered with a 3rd–order, low-pass Butterworth filter with cutoff frequencies of 25 and 20 Hz, respectively. To determine the biomechanical quantities of interest, a 13-segment model including head, upper arms, forearms, hands, torso, pelvis, thighs, shanks, and feet was created in Visual 3D (Visual 3D, C-Motion, Inc., Germantown, MD). Residual limb shank inertial properties were adjusted by reducing the segment mass by 39% and altering the COM distance 24% closer to the knee (Smith et al., 2014). Intersegmental joint angles and moments were determined from the GRFs and body segment kinematics using standard inverse kinematic and dynamics techniques (e.g., Winter, 1991). Joint powers were calculated as the dot-product of the three-dimensional joint moment and angular velocity vectors. Positive (negative) joint work was calculated as the time integral of the positive (negative) joint power over the gait cycle.

All trials were examined and included in the analysis if the change in perturbed step width was greater than one standard deviation of the normal step width for each subject and the perturbed limb GRF data was clearly defined (i.e., no cross-over steps). The average perturbation size for the amputee and non-amputee subjects was 48 ± 18 mm and 39 ± 15 mm, respectively. If a GRF was undefined during unperturbed walking (e.g., there was a centerline cross-over) then the stride was removed from the analysis. This resulted in fewer subjects included in the medially-directed prosthetic (n = 5) and sound (n = 8) limb perturbation analyses. Also, any trials where a subject used the safety handrail for support were not included.

Whole-body angular momentum (H) about the body center-of-mass (CoM) was calculated as (Fig. 1):

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

where $\vec{r}_i^{COM}$ and $\vec{v}_i^{COM}$ are the position and 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 whole-body CoM. $\vec{\omega_i}$, m_i and I_i are the angular velocity vector, and mass and moment of inertia of the i-th segment, respectively, and n is the number of segments.

Whole-body angular momentum was time normalized to the entire gait cycle and normalized by subject mass (kg), speed (m/s) and height (m). The range of H (H_R) was defined as the peak-to-peak difference between the maximum and minimum values of H over the stance phase of the perturbed limb.
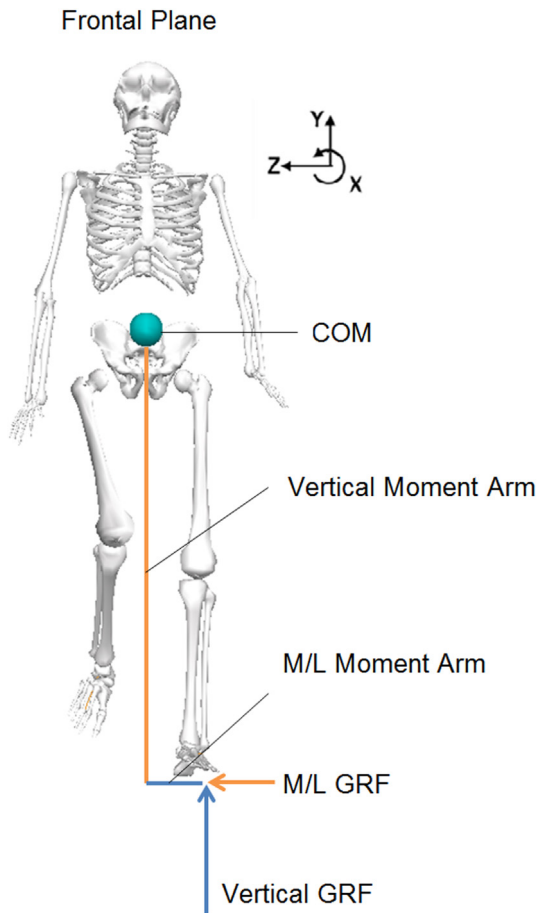
Frontal Plane

Fig. 1. Frontal plane whole-body angular momentum was defined about the x-axis.

Positive and negative hip joint work were calculated for six phases of the gait cycle by time-integrating the positive and negative joint powers, respectively, using custom MATLAB code (The MathWorks, Inc., Natick, MA). Specific emphasis was placed on the second and third phases of the gait cycle, which were identified as the first half of single limb support (phase 2) and the second half of single limb support (phase 3), since these time periods have been associated with higher instability (Nott et al., 2014).

Statistical analyses were completed in SAS (SAS Institute Inc., Cary, NC) using a mixed-effects model to determine if mean changes in outcome metrics between steps (normal versus perturbed) differed by disturbed limb (Amputee Residual, Amputee Sound, and Non-Amputee Right). Unperturbed steps were defined as the average of the first three consecutive steps prior to the perturbation and then averaged across trials for each subject. The perturbed step analyzed was the first step immediately following the perturbation. Medial and lateral perturbation conditions were analyzed with separate models for all metrics. Differences between normal and perturbed steps per condition were separated into six individual models depending on the perturbation direction and limb perturbed: Lateral Residual, Lateral Sound, Lateral Non-Amputee Right, Medial Residual, Medial Sound, and Medial Non-Amputee Right. The mixed model contained one fixed effect (Step – undisturbed step versus perturbed step) and two random effects (Subject and Trial). The statistical models used to analyze the joint work were also separated by the six phases of the gait cycle. Significance was defined as p < 0.05. Emphasis was placed on the joint work in the first and second halves of the single support (phases two and three) to identify the balance recovery mechanisms used by the perturbed limb.

Dependent values were frontal plane H_R over the stance phase of the perturbed limb and the positive and negative hip joint work in specific regions of the gait cycle. Group averages were calculated as the averages across trials and subjects within the amputee and non-amputee groups.

3. Results

3.1. Range of whole-body angular momentum

Frontal plane H_R of the perturbed limb was lower for lateral perturbations for all limbs ($p < 0.05$ for the residual limb, $p < 0.0001$ for the sound and non-amputee limbs) compared to unperturbed walking. In contrast, frontal plane H_R was greater for medial perturbations for all limbs ($p < 0.05$ for the residual limb, $p < 0.0001$ for the sound and non-amputee limbs) (Figs. 2 and 3).

3.2. Hip joint work

3.2.1. Lateral perturbations

In the first half of single support, the lateral perturbations of all limbs ($p < 0.05$) for residual, $p < 0.0001$ for sound and non-amputee) resulted in greater positive frontal plane hip joint work. Lateral perturbations of the residual ($p < 0.05$) and non-amputee ($p < 0.0001$) limbs resulted in reduced negative hip work. In the second half of single support, the frontal plane positive hip work increased for lateral perturbations of the sound limb ($p < 0.05$). In addition, the frontal plane negative hip work was reduced for lateral perturbations of the non-amputee limb ($p < 0.05$) (Figs. 4 and 5).

3.2.2. Medial perturbations

All limbs responded with reduced positive frontal plane hip work for medial perturbations ($p < 0.0001$) in the first half of single support. Medial perturbations of the residual limb also resulted in greater negative frontal plane hip work ($p < 0.05$) during the first half of single support. During the second half of single support, medial perturbations of the sound ($p < 0.05$) and non-amputee ($p < 0.05$) limbs resulted in reduced positive and increased negative frontal plane hip work of the non-amputee limb ($p < 0.0001$) (Figs. 4 and 5).

4. Discussion

The purpose of this study was to understand the balance recovery mechanisms used by lower-limb amputees in response to mediolateral foot perturbations by examining frontal plane H_R and hip work. Contrary to our hypotheses, lateral perturbations decreased frontal plane H_R while medial perturbations increased H_R during stance for all limbs (Figs. 2 and 3). Amputees and non-amputees used a similar immediate hip strategy to maintain balance, with increased positive frontal plane hip work for lateral perturbations and decreased positive frontal plane hip work for medial perturbations (Figs. 4 and 5). These results were consistent with previous studies that correlated wider (narrower) step width with increased (decreased) gluteus medius activity (Hof and Duysens, 2013; Kubinski et al., 2015).

The time rate-of-change of H is equivalent to the net external moment and defined as the cross product of the GRF vector and moment arm vector from the body center-of-mass to center-of-pressure (Fig. 1). Thus, we anticipated that lateral perturbations would lead to an increase in H_R due to the increased mediolateral moment arm. Lateral perturbations increased the step width (Segal and Klute, 2014) and thus increased the mediolateral moment arm (Fig. 1). However, the increased moment arm was likely offset by
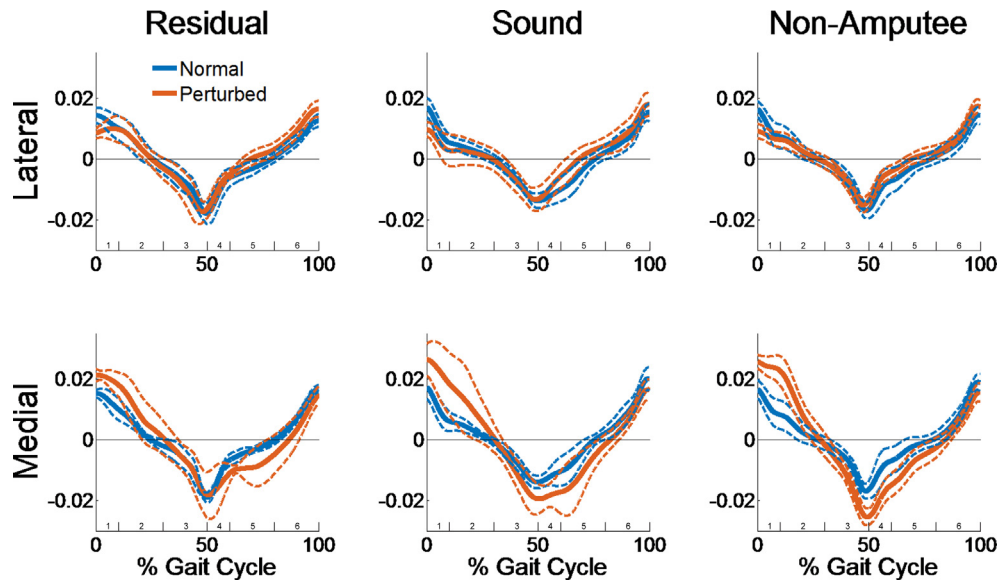
Fig. 2. Normalized frontal plane whole-body angular momentum (solid lines) ± one standard deviation (dashed lines) of the residual, sound, and non-amputee limbs during normal walking (blue line) and the perturbed step (orange line). The tick marks on the x-axis correspond to the beginning and end of the six phases of the gait cycle: (1) 1st double support, (2) 1st half of single support, (3) 2nd half of single support, (4) 2nd double support, (5) 1st half of swing, and (6) 2nd half of swing. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)
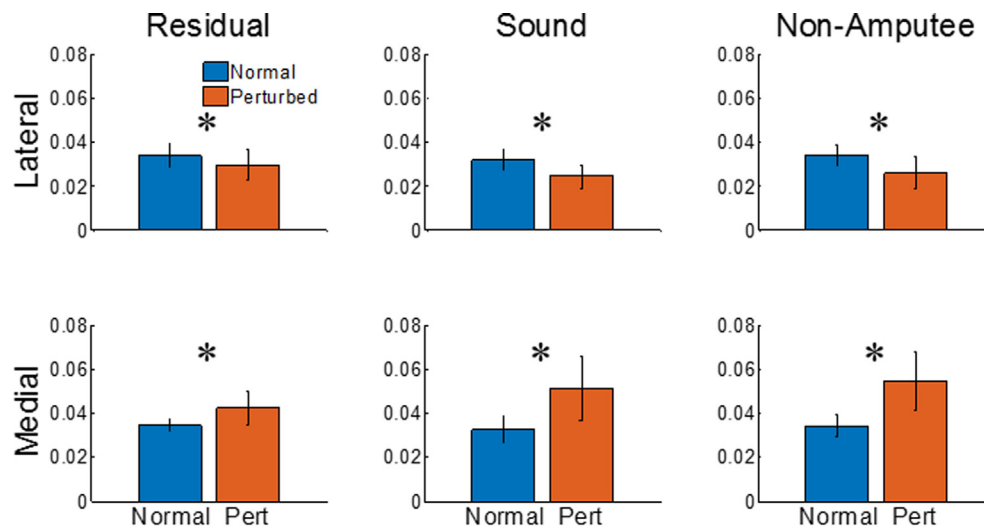

Fig. 3. Normalized range of frontal plane whole-body angular momentum ± one standard deviation of the residual, sound, and non-amputee limbs during normal walking (left blue bar) and the perturbed step (right orange bar). Significant differences between unperturbed and perturbed steps are indicated with a '*' ($p < 0.05$). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

changes in the GRFs. To determine if these changes did indeed occur, we performed a post-hoc analysis of the GRFs.

For lateral perturbations, there was little change in the vertical GRFs (Fig. 6), but there was an increase in the medial GRFs for all limbs (Fig. 7). This was likely due to both the change in step width and observed increases in hip work (Fig. 5) following the perturbation. Previous modeling work has shown that the hip abductor muscles are the primary contributors to the medial GRF during this region of the gait cycle (John et al., 2012). Since the medial GRF contributes to the negative net external moment, an increase in the medial GRF would produce a decrease in the net external moment, and thus a decrease in H_R.

For the medial perturbations, we expected the H_R would decrease due to the smaller mediolateral moment arm. In contrast, the range increased for all limbs (Figs. 2 and 3). Examination of the

vertical GRFs showed little change while the mediolateral GRFs showed a reduction in the medially-directed GRF for all limbs (Fig. 7). A reduction in the medial GRF would have led to an increase in the external moment, and thus an increase in H_R.

Previous studies have shown there are three primary mechanisms for controlling mediolateral balance: a stepping strategy, hip strategy and ankle strategy (Hof, 2007). Our results have shown that while there was a stepping strategy used for recovery from these perturbations (Segal and Klute, 2014), frontal plane hip work also played an important role in balance recovery. We hypothesized increased positive frontal plane hip work for lateral perturbations due to the increased step width. Our results supported this hypothesis across all limbs in the first half of single support. The positive frontal plane hip work also remained increased for lateral perturbations of the sound limb during the second half of single
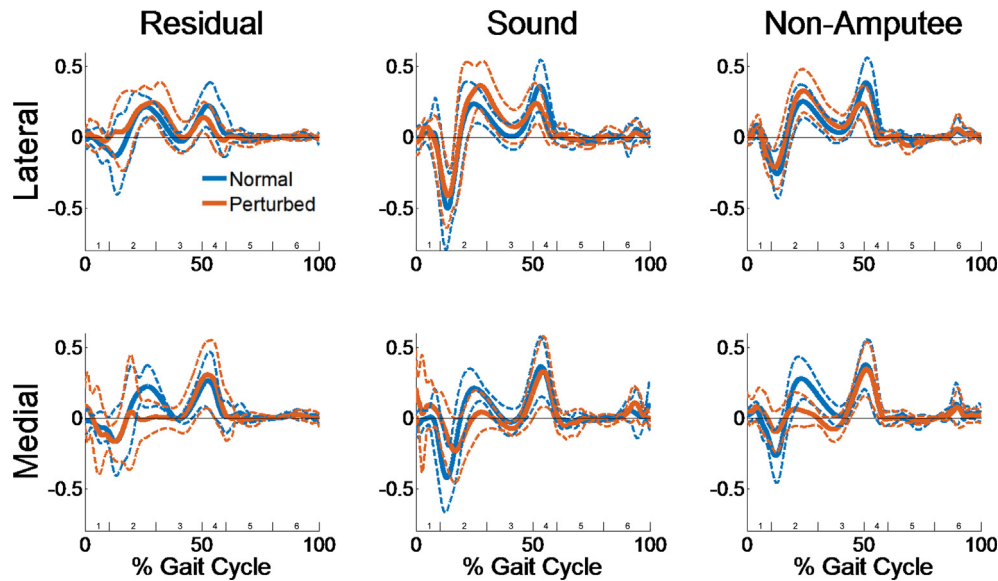
Fig. 4. Normalized frontal plane hip joint power (J/kg) ± one standard deviation of the residual, sound, and non-amputee limbs over the gait cycle. Positive values correspond to power generation. The tick marks on the x-axis correspond to the beginning and end of the six phases of the gait cycle (see Fig. 2 caption).
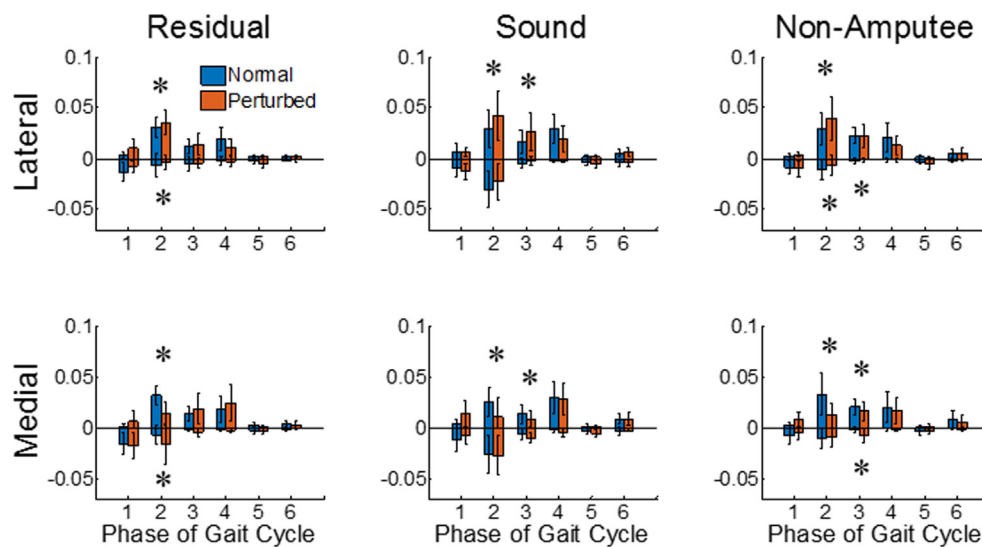

Fig. 5. Normalized frontal plane hip joint work (W/kg) ± one standard deviation of the residual, sound, and non-amputee limbs over the six phases of the gait cycle. Significant differences between unperturbed and perturbed steps are indicated with a '*' ($p < 0.05$).

support (Figs. 4 and 5). The lateral perturbations resulted in increased step width (Segal and Klute, 2014) and increased medial GRF (Fig. 7). The hip abductors are the primary muscles that counteract the influence of gravity in the frontal plane (MacKinnon and Winter, 1993) and contribute to the medial GRF during early stance (Jansen et al., 2014; John et al., 2012). Specifically, the gluteus medius has been shown to rotate the body towards the ipsilateral limb to counteract the net external moment created by gravity (Neptune and McGowan, 2016). Therefore, the increase in medial GRF and frontal plane hip work following a lateral perturbation is likely due to an increase in the hip abductor activity. These results are consistent with previous studies that found increased hip abductor activity, specifically the gluteus medius, with increased step widths (Hof and Duysens, 2013; Kubinski et al., 2015).

In contrast to our hypothesis for lateral perturbations, we hypothesized reduced positive frontal plane hip work for medial perturbations. The results agreed with our hypothesis across all leg types in the first half of single support (Figs. 4 and 5). Narrow step widths have been correlated with decreased gluteus medius activity (Kubinski et al., 2015). Since the gluteus medius counteracts the influence of gravity on the GRFs (Neptune and McGowan, 2016), the gluteus medius would also decrease its contribution to the GRFs following a medial perturbation. Reduced positive hip abduction moment has previously been correlated with increased H_R (Silverman et al., 2012). Thus, the reduced positive hip work for medial perturbations likely indicates decreased gluteus medius activity. These results suggest that subjects respond to medial perturbations with reduced hip abductor activity in response to a decreased destabilizing moment by gravity. However, this response was insufficient to regulate the frontal plane H as the range was higher with the medial perturbations and required additional steps to regain their balance (Segal and Klute, 2014).
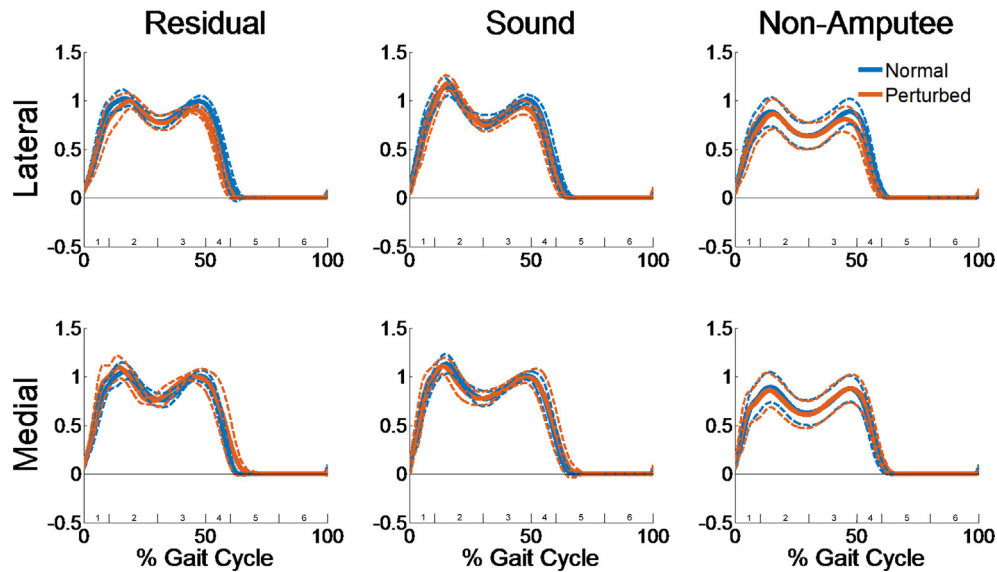
Fig. 6. Normalized vertical GRFs (solid lines) ± one standard deviation (dashed lines) of the residual, sound, and non-amputee limbs during normal walking and the perturbed step. The tick marks on the x-axis correspond to the beginning and end of the six phases of the gait cycle (see Fig. 2 caption).
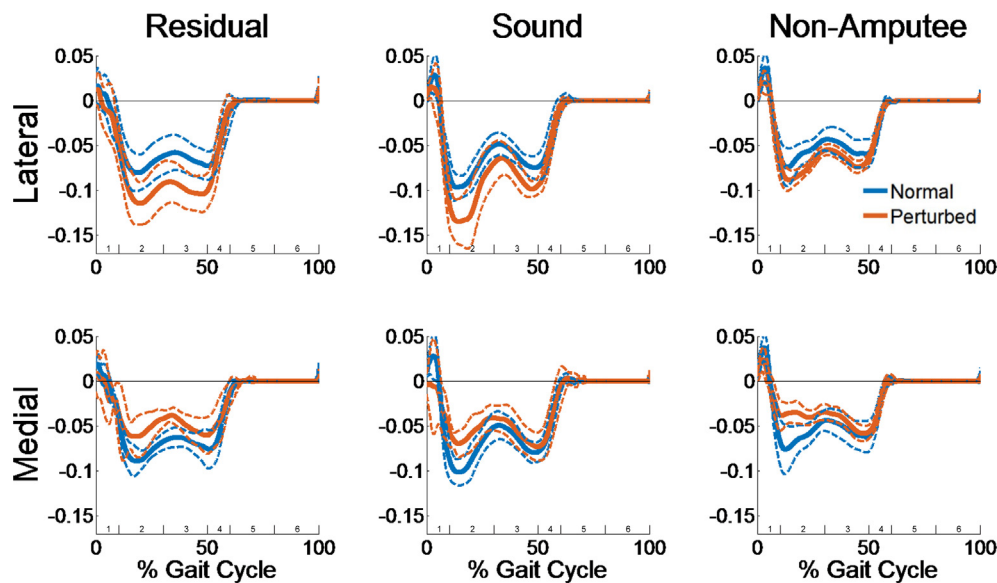

Fig. 7. Normalized mediolateral GRFs (solid lines) ± one standard deviation (dashed lines) of the residual, sound, and non-amputee limbs during normal walking and the perturbed step. Negative values correspond to a medially-directed GRF. The tick marks on the x-axis correspond to the beginning and end of the six phases of the gait cycle (see Fig. 2 caption).

There was also an increase in the negative frontal plane hip work for medial perturbations of the residual limb. Segal et al. (2015) found that while the sound and non-amputee limbs responded with changes in their frontal plane hip moment, the residual limb frontal plane hip moment did not change and remained reduced compared to unperturbed levels of the sound and non-amputee limbs. This suggests that hip adductor activity may have increased to counteract hip abductor activity that may have still been too high, which would increase the negative frontal plane hip work. Co-excitation of the hip adductors and abductors would also act to stiffen the hip joint as a possible balance control mechanism. This result supports that balance perturbations of the residual limb are particularly challenging and require additional balance recovery mechanisms that are not needed for the sound or non-amputee limbs. Through this additional hip strategy, ampu-

tees were able to maintain a similar H_R on the perturbed step compared to non-amputees. Since amputees required three additional steps to return to their unperturbed step width (Segal and Klute, 2014), we predict that H_R would likely remain increased for the amputee's subsequent steps and a topic for future exploration with a larger sample population.

The amputees also exhibited a similar frontal plane H_R as the non-amputees during unperturbed walking, implying similar stability across populations. These findings were consistent with prior research (Pickle et al., 2014), yet contrary to several others who reported relatively small (<15%) differences between populations when walking at a similar self-selected pace of approximately 1.2 m/s (D'Andrea et al., 2014; Sheehan et al., 2015; Silverman and Neptune, 2011). An interesting finding was that our non-amputees had a higher H_R during the medial perturbation (0.055

N-m) compared to the amputees (0.035 N-m), which may be attributed to differences in walking mechanics. Examination of the GRFs shows the non-amputees responded to the medial perturbation with a lower medial GRF in early stance (Fig. 5), which increased the external moment and generated the greater H_R. Despite this greater range, the non-amputees were able to recover their balance in less steps compared to the amputees (Segal and Klute, 2014). This is likely due to non-amputees having active ankle muscle control, which has been shown to be essential to regulating frontal plane H (Neptune and McGowan, 2016).

This study showed a reduced H_R for lateral perturbations and increased H_R for medial perturbations for both amputees and non-amputees. We also showed increased frontal plane hip work for lateral perturbations and reduced frontal plane hip work for medial perturbations in the first half of single limb stance. These results suggest that frontal plane hip work contributes to dynamic balance recovery following mediolateral foot perturbations. Thus, future work should focus on strengthening the hip abductors to improve the control of dynamic balance following mediolateral perturbations. In addition, developing prostheses better able to replicate the active ankle control strategy used by non-amputee ankles may reduce the need for compensatory hip joint responses.

Conflict of interest

The authors declared that there is no conflict of interest.

Acknowledgements

Conflict of Interest Statement

The authors have no conflict of interest to declare.

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