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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.

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

* Corresponding author at: Department of Mechanical Engineering, The University of Texas at Austin, 204 E. Dean Keeton Street, Stop C2200, Austin, TX 78712, USA.

E-mail address: rneptune@mail.utexas.edu (R.R. Neptune).

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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.