Individuals with transtibial limb loss use interlimb force asymmetries to maintain multi-directional reactive balance control

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A R T I C L E  I N F O

Article history:
Received 19 June 2014
Accepted 18 August 2014

Keywords:
Posture
Stability
Amputee
Artificial limb
Feedback
Ground reaction force

A B S T R A C T

Background: Deficits in balance control are one of the most common and serious mobility challenges facing individuals with lower limb loss. Yet, dynamic postural balance control among individuals with lower limb loss remains poorly understood. Here we examined the kinematics and kinetics of dynamic balance in individuals with unilateral transtibial limb loss.

Methods: Five individuals with unilateral transtibial limb loss, and five age- and gender-matched controls completed a series of randomly applied multi-directional support surface translations. Whole-body metrics, e.g. peak center-of-mass displacement and net center-of-pressure displacement were compared across cohorts. Stability margin was computed as the difference between peak center-of-pressure and center-of-mass displacement. Additionally, center-of-pressure and ground reaction force magnitude and direction were compared between the prosthetic, intact, and control legs.

Findings: Peak center-of-mass displacement and stability margin did not differ between individuals with transtibial limb loss and controls for all perturbation directions except those loading only the prosthetic leg; in such cases the stability margin was actually larger than controls. Despite similar center-of-mass displacement, greater center-of-pressure displacement was observed in the intact leg during anterior–posterior perturbations, and under the prosthetic leg in medial–lateral perturbations. Further, in the prosthetic leg, ground reaction forces were smaller and spanned fewer directions.

Interpretation: Deficits in balance control among individuals with transtibial limb loss may be due to their inability to use their prosthetic leg to generate forces that are equal in magnitude and direction to those of unimpaired adults. Targeting this force-generating deficit through technological or rehabilitation innovations may improve balance control.

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1. Introduction

One of the most commonly cited functional mobility limitations facing individuals with lower limb loss (LLL) is a deficit in balance control (Jayakaran et al., 2012). Over half of community-living individuals with LLL report having fallen in the past year and 49.2% report a fear of falling (Miller et al., 2001). Ulger et al. (2010) found that 80% of geriatric amputees had fallen within the past year, with 64% falling more than once (Ulger et al., 2010). These impairments in balance control carry an immense cost to an individual's functional mobility at home and in the community (McWhinnie et al., 1994; van Velzen et al., 2006), lowering their health-related quality of life (Pezzin et al., 2000). With LLL affecting more than 1.6 million individuals living in the United States, and forecast to increase to more than 3.6 million individuals by the year 2050 (Ziegler-Graham et al., 2008), the impact of balance impairments will continue to exact an increasing toll on the healthcare system. Although falling is clearly a common and serious problem among individuals with LLL (Pauley et al., 2006), very little is known regarding the challenges and mechanisms underlying deficits in balance control among individuals with LLL (Curtze et al., 2010).

To date, research investigating postural balance control in individuals with LLL has generally been limited to unperturbed standing (Jayakaran et al., 2012; Ku et al., 2014). These posturography studies have revealed increases in measures of postural sway including center of pressure velocity and displacement compared to age-matched controls (Buckley et al., 2002; Fernie and Holliday, 1978; Hermodsson et al., 1994; Isakov et al., 1992; Kolarova et al., 2013; Rougier and Bergeau, 2009). These differences in postural sway are influenced by

Funding: NIH NS053822, NSF EFRI 1137229.

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http://dx.doi.org/10.1016/j.clinbiomech.2014.08.007
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age (Hermodsson et al., 1994; Vittas et al., 1986), cause of limb loss (Hermodsson et al., 1994), residual limb length (Lenka and Tiberwala, 2010), level of limb loss (Fernie and Holliday, 1978; Rougier and Bergeau, 2009), and time since limb loss (Mayer et al., 2011), but not prosthetic alignment (Isakov et al., 1994; Kolara et al., 2013). Additionally, the intact leg may contribute substantially more to maintaining static postural stability than the prosthetic leg (Hlavackova et al., 2011).

In contrast to unperturbed standing balance, dynamic postural balance in response to internal or external perturbations has received much less attention among individuals with LLL, despite offering several advantages to other means of balance assessment (Visser et al., 2008). Aruin et al. (1997) studied anticipatory postural adjustments among a cohort of individuals with unilateral transtibial limb loss (TTLL) in response to rapid arm raises. They found that individuals with TTLL exhibited asymmetry in anticipatory changes of background muscle activity. These responses were much larger on the intact side of the body, and frequently absent on the side affected by limb loss (Aruin et al., 1997). In response to external perturbations, individuals with unilateral TTLL exhibit delayed muscle responses in both the intact and prosthetic leg (Rusaw et al., 2013), as well as greater reliance on the intact versus prosthetic leg to generate corrective responses (i.e. CoP displacement, ankle torque) (Curtze et al., 2012; Nederhand et al., 2012; Vrieling et al., 2008).

In these prior studies, perturbations were often limited to the sagittal plane (Curtze et al., 2010; Nederhand et al., 2012; Rusaw et al., 2013; Vrieling et al., 2008). Sagittal plane perturbations load both legs, allowing participants to compensate for sensorimotor impairment in one leg by using the contralateral leg more. Conversely, mediolateral perturbations do not allow contralateral leg compensation, providing a direct assessment of each leg’s capabilities. Medial–lateral balance control is also known to present greater fall risk (Maki et al., 1994; Topper et al., 1993), impose greater processing demands on the central nervous system (Carpenter et al., 1999), and be particularly challenging to individuals with LLL (Aruin et al., 2009), and time since limb loss (Maki et al., 1994; Topper et al., 1993), impose greater processing demands on the central nervous system (Carpenter et al., 1999), and be particularly challenging to individuals with LLL (Aruin et al., 2009), and time since limb loss (Maki et al., 1994; Topper et al., 1993), impose greater processing demands on the central nervous system (Carpenter et al., 1999), and be particularly challenging to individuals with LLL (Aruin et al., 2009), and time since limb loss (Maki et al., 1994; Topper et al., 1993), impose greater processing demands on the central nervous system (Carpenter et al., 1999), and be particularly challenging to individuals with LLL (Aruin et al., 2009), and time since limb loss (Maki et al., 1994; 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A. Coordinate system for support surface translations

B. Right leg response to forward perturbation

Platform Acceleration (g)

Right Tibialis Anterior EMG

Right A-P GRF (N)

Right Vertical GRF (N)

Center of Pressure (cm)

Center of Mass (cm)

Time (ms)
For each trial, whole-body reactive balance responses were quantified by calculating peak CoM and peak net CoP displacement from initial position, as well as the stability margin for each trial. Stability margin is the difference between peak CoP and CoM displacements during a perturbation (Horak et al., 2005; Winter et al., 1996). Peak net CoP, CoM and stability margin were analyzed from 160 to 300 ms following perturbation onset (Fig. 1B). This time period is referred to as the active force production period (Henry et al., 2001; Torres-Oviedo and Ting, 2007). 160 ms post-perturbation reflects the point in time when active force production begins, accounting for the excitation–contraction coupling time period. The ending time of 300 ms, which is prior to peak platform deceleration, was chosen to minimize the effect of the additional unwanted counterforce on the body caused by the deceleration of the platform.

Individual leg contributions to whole-body reactive balance responses were assessed for each perturbation direction by calculating leg-specific measures including: peak CoP displacement, the mean vertical, anterior–posterior (AP) and medial–lateral (ML) GRF magnitude, and the direction of the resultant horizontal GRF vector. The mean background GRFs were calculated during quiet stance, 250–50 ms prior to perturbation onset. The mean active GRF magnitude and the resultant horizontal force directions were computed over the first 60 ms time window of the active force production period (160–220 ms post-perturbation) as the change in force from background levels. All data was analyzed using custom MATLAB™ (MathWorks, Natick, MA) code.

2.5. Statistical analysis

To limit the number of statistical comparisons, four cardinal directions (forward, backward, right and left) of the 12 perturbation directions were subjected to statistical analysis. To determine the effect of group (TTLL versus control) and sway direction (4 directions) on leg-specific measures including: peak CoP displacement, the mean vertical, anterior–posterior (AP) and medial–lateral (ML) GRF magnitude, and the direction of the resultant horizontal GRF vector. The mean background GRFs were calculated during quiet stance, 250–50 ms prior to perturbation onset. The mean active GRF magnitude and the resultant horizontal force directions were computed over the first 60 ms time window of the active force production period (160–220 ms post-perturbation) as the change in force from background levels. All data was analyzed using custom MATLAB™ (MathWorks, Natick, MA) code.

3. Results

3.1. Participant demographics

Five males with unilateral TTLL and five age- and gender-matched unimpaired adults were recruited to participate in the study (Table 1).

3.2. Differences in peak CoM, peak net CoP, and stability margin between groups

Although CoM displacement and net CoP magnitude did not differ between groups, we found directionally specific differences between individuals with TTLL and controls in net CoP displacement and stability margin. Peak CoM displacement did not differ between groups (group effect: P = 0.052), across direction of sway (direction effect: P = 0.16), nor were there significant interaction effects (interaction: P = 0.345) (Fig. 2A–B). The magnitude of net CoP displacement did not differ between groups, while the stability margin did (group effect: P = 0.06); stability margin P = 0.038. Net CoP displacement and stability margin varied across directions (direction effect: net CoP P < 0.001; stability margin P = 0.003). Specifically, individuals with TTLL had significantly larger net CoP displacement (P = 0.002) and stability margin (P = 0.002) than controls when swaying towards and loading their prosthesis (leftward for controls) (Fig. 2A, C–D). Controls and individuals with TTLL had significantly larger peak net CoP displacements during medial–lateral than anterior–posterior perturbations (controls P < 0.029; TTLL P < 0.001) (Fig. 2A, C). These same directionally-sensitive differences in stability margin were only observed in individuals with TTLL (P < 0.003), with controls demonstrating limited differences in stability margin across directions, leftward versus anterior–posterior sway (P < 0.011) (Fig. 2D).

3.3. Individual leg contributions to reactive balance responses (peak CoP, GRFs)

Individual leg force measurements revealed directionally specific differences in CoP displacement in the intact leg compared to the prosthetic and control legs. Peak CoP displacements for the right and left legs of controls were not significantly different (P = 0.491). Therefore the right leg was used for comparison to the peak CoP displacement of the prosthetic and intact legs of individuals with TTLL. Leg-specific peak CoP displacement (Fig. 3) differed between legs (leg effect: P = 0.004), as well as by sway direction (sway direction effect: P < 0.001) with a significant interaction between leg and sway direction (interaction: P < 0.001). During forward sway, the intact leg of individuals with TTLL had a significantly larger peak CoP displacement than the prosthetic leg (P < 0.001) or the control leg (P = 0.002). The control leg also had a significantly larger peak CoP displacement than the prosthetic leg (P < 0.001) (Fig. 3). During lateral sway (towards the prosthesis), the prosthetic leg of individuals with TTLL had larger peak CoP displacement that the control leg (P = 0.006). Additionally, peak CoP displacement under the intact and prosthetic leg of individuals with TTLL was significantly different across all sway directions (intact leg: P < 0.001, prosthetic leg: P < 0.001), while the peak CoP displacement among controls was only found to be significantly different between medial–lateral and anterior–posterior sway directions (P < 0.001).

Differences in force magnitude and direction were also observed between the prosthetic, intact and control legs. There were no significant differences in the mean background forces or the peak active forces between the left and right legs of control subjects (P > 0.05). The mean ML, AP, and vertical background forces were not significantly different
Fig. 2. Whole-body reactive balance responses to multi-directional support surface translations. A: Peak center of mass displacement (blue squares), peak net center of pressure excursion (black circles), and stability margin (red diamonds) averaged across subjects (transtibial amputees: unfilled; controls: filled) for all 12 sway directions. B: Peak center of mass displacement (mean ± 1SD) across the four cardinal directions for transtibial amputees (unfilled) and controls (filled). There was no significant difference in the magnitude of whole-body displacement between individuals with limb loss and controls, or across directions within either cohort. C: Peak net center of pressure excursion (mean ± 1SD) across the four cardinal directions for individuals with transtibial limb loss (unfilled) and controls (filled). Individuals with transtibial limb loss had significantly larger net center of pressure excursion in response to lateral sway that loaded the prosthetic leg \((P = 0.002)\) than controls. D: Stability margin (mean ± 1SD) across the four cardinal directions for individuals with transtibial limb loss (unfilled) and controls (filled). Individuals with transtibial limb loss had a significantly larger stability margin during lateral sway that loaded the prosthetic leg \((P = 0.002)\) than controls.
between the prosthetic, intact or control legs (P > 0.05) (Fig. 4). Neither the ML (P = 0.891), nor the vertical (P = 0.635) peak active forces were significantly different between the prosthetic, intact or control legs; only the AP peak active force was significantly different between leg conditions (leg effect: P = 0.002). Specifically, the intact leg of individuals with TTLL produced a significantly larger peak AP force than the prosthetic leg (P = 0.002) and the control leg (P = 0.032). While the prosthetic leg produced a smaller peak AP force than the control leg, this difference was not statistically significant (P = 0.416) (Fig. 4). Beyond the magnitude of the horizontal forces, inspection of Fig. 5 reveals that the direction of the resultant horizontal forces also differed between cohorts, specifically when comparing the prosthetic leg with that of the control leg (Fig. 5B).

4. Discussion

Individuals with TTLL employ different interlimb kinetic strategies to achieve surprisingly similar dynamic balance performance compared to controls in most directions. Specifically, during lateral perturbations that loaded the prosthetic leg, similar CoM but greater net CoP displacement (Fig. 2A) was found, resulting in a higher, not lower stability margin in the direction considered the least stable. Differences in interlimb coordination of individual leg CoP and ground reaction forces were also observed across directions suggesting more reliance on the intact leg for balance. Limited directions of force were identified in the prosthetic leg compared to the intact and control legs, however it is unclear whether this reflects a compensatory strategy or a functional limitation related to the design of prostheses.

The exaggerated net CoP response that was observed among individuals with TTLL may impose additional unintended balance challenges and result from a fear of falling and/or sensorimotor limitations of the prosthesis. The exaggerated net CoP response may have developed as a result of a history of falls. Repeated falls could create the desire to maintain a greater margin for error to reduce the likelihood of a loss of balance, particularly when using the prosthetic leg (Hof et al., 2007). Four of the five individuals with TTLL in this study reported at least one fall in the past 12 months, suggesting that their control strategy for maintaining balance could have been influenced by their falls’ history. Another explanation could be the inability to accurately perceive the perturbation, or to generate appropriately scaled forces with the prosthetic leg. It is possible that the prosthetic leg was used passively as a strut rather than actively to generate forces. Regardless of its cause, this exaggerated response may impose unforeseen challenges to maintaining balance control. If individuals with unilateral TTLL use motor responses that are larger than necessary, and that lie at the limits of their capabilities, this may constrain their ability to respond to continued or larger perturbations, ultimately resulting in a loss of balance and potential for serious injury. Testing this hypothesis would require identifying the maximal capacity of subjects during reactive balance, comparing it to responses across other perturbation magnitudes, and assessing whether the use of an exaggerated response across perturbation magnitudes does in fact limit their ability to respond effectively to continued perturbations. This would likely require an examination of stepping (Curtze et al., 2010, 2012; Mansfield et al., 2013; Mille et al., 2003; Rogers et al., 2003) and non-stepping responses.

The differences in balance between individuals with TTLL was likely not due to their initial posture, but rather the execution of the motor response following a perturbation. While visually distinct, we found that the background forces were not significantly different between controls and individuals with TTLL (Fig. 4), consistent with the results of Nederhand et al. (2012). This does not exclude the possibility that adjustment(s) to background posture or force patterns would not improve reactive balance among individuals with LLL (Anker et al., 2008; Genthon and Rouquier, 2005).

Consistent with prior studies in perturbed (Curtze et al., 2012; Vrieling et al., 2008) and unperturbed standing (Nadollek et al., 2002; Quai et al., 2005), compensations in the intact leg were required to preserve anterior–posterior balance. If the individuals with TTLL in our study had not increased the CoP displacement under their intact leg during forward sway (Fig. 3), the peak CoM displacement would have exceeded the net CoP response, resulting in a loss of balance. This asymmetric strategy was effective in maintaining a similar stability margin as controls (Fig. 2A, D), and was achievable due to the relatively high mobility level of the amputee subjects in this study. However, older adults with TTLL, or those with additional co-morbidities such as neuropathy, may lack adequate contralateral sensorimotor function to effectively compensate for the reduced biomechanical contributions of the prosthetic leg to anterior–posterior balance. Additional research is required to determine how these compensations are affected by aging and additional impairment. The differences in interlimb contributions to balance also highlight the need for interventions and innovations in prosthetic technology to reduce the need for contralateral compensations.

Consistent with previous work (Nederhand et al., 2012; Vrieling et al., 2008), force-generating asymmetries between legs were observed in the anterior–posterior direction, with the intact leg exerting significantly greater peak AP forces than the prosthetic or control legs. Deficits in force magnitude were limited to the AP direction, as individuals with unilateral TTLL are able to use their prosthetic leg to produce ML and vertical forces of equivalent magnitude to those of controls and their own intact leg (Fig. 4A, C, E). The large difference in the AP force magnitude may reflect limitations in the ability to actively generate plantarflexion torque with the prosthetic leg. Further, this exaggerated force response may characterize a preferred strategy used by individuals with TTLL to simplify the control of reactive balance. If a single exaggerated yet safe response is consistently used, it may remove the need to scale the response according to the perturbation and/or need for contralateral compensation, thereby simplifying control.

Individuals with TTLL demonstrated limited directions of force with their prosthetic leg, altering interlimb force coordination for directional control of the CoM. Individuals with unilateral TTLL exerted active resultant horizontal GRFs with their prosthetic leg that clustered around the medial–lateral directions, while forces generated by control and intact legs were distributed across all directions in the horizontal plane.
By restricting forces to a set of two direction-invariant vectors, the challenge of maintaining balance in response to horizontal plane perturbations may be simplified (Macpherson, 1988). Such a restriction of force direction has been observed previously for non-preferred stances in both humans (i.e. narrow stance) and animals (Henry et al., 2001; Macpherson, 1994). It is possible that the force constraint strategy simplifies the control of balance under novel, or challenging balance conditions. We only observed the constraint in the prosthetic leg suggesting that the intact leg regulates multi-directional control of the CoM among individuals with unilateral TTLL. However, the medial–lateral force constraint may also be due to limitations in sagittal plane ankle range of motion or plantarflexion torque generation in the prosthetic ankle.

4.1. Study limitations

We studied only five individuals with TTLL, limited to those of traumatic etiology. Given the increasing age of the general population

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**Fig. 4.** Background and active period ground reaction forces. A, C, E: Polar plots for each leg depicting the mean background (thin lines) and active period (thick lines) vertical (A), medial–lateral (B), and anterior–posterior (C) ground reaction force for controls (solid black lines) and individuals with transtibial limb loss (dashed red lines) across each direction of sway. B, D, F: The mean background force (±1SD) and peak active force (±1SD) along the vertical (B), medial–lateral (D), and anterior–posterior (F) axes for the control (gray), intact (white) and prosthetic (red) legs. Polar plots reveal a non-significant difference in vertical (prosthetic vs. control leg), medial–lateral (prosthetic vs. control; intact vs. control), and anterior–posterior (prosthetic vs. control; intact vs. control) background forces. The patterns of active forces were largely comparable between controls and individuals with transtibial limb loss along the vertical and medial–lateral axes, with no significant differences in the peak active forces. The patterns of active forces were vastly different between cohorts along the anterior–posterior axis, with significant differences between the control and intact leg ($p = 0.032$), as well as the intact and prosthetic leg ($p = 0.002$). Peak active forces were generally exerted in the same direction by controls and individuals with transtibial limb loss (denoted by star).
(Wiener and Tilly, 2002), and the rising incidence of diabetes (Boyle et al., 2010), there is an unmistakable need to study individuals with dysvascular limb loss as well as older adults with lower limb loss. However, for this initial study the traumatic population was specifically selected to avoid additional confounds that may influence reactive balance control. Despite a limited sample size we were able to identify significant differences between cohorts. Nevertheless, a study including a larger cohort of subjects is certainly warranted to confirm these results. The perturbation parameters selected in this study were selected based upon previous work (Henry et al., 2001; Horak et al., 2005) and sufficient to examine the initial active response period. However, the short time interval between peak platform acceleration and deceleration precluded us from looking at a longer time window. Here we examined whole-body kinematics and endpoint forces to describe differences in reactive postural balance control between individuals with unilateral TTLL and unimpaired adults. An analysis of the muscle coordination patterns that underlie the differences in control strategies used to stabilize the same CoM displacement (Chvatal and Ting, 2013; Chvatal et al., 2011; Torres-Oviedo and Ting, 2007), may provide additional insight into their production and the design of interventions to improve balance control among individuals with LLL.

5. Conclusion

Surprisingly, we found that individuals with non-dysvascular unilateral TTLL were equally capable of maintaining dynamic standing balance in response to external perturbations as age-matched unimpaired controls. Individuals with TTLL used significantly different interlimb contributions to stabilize the same CoM displacement, revealing functional asymmetries in CoP and ground reaction forces between the prosthetic, intact and control legs. These functional asymmetries between legs suggest that individuals with unilateral TTLL may be limited in their ability to exert forces with their prosthetic leg that are equal in magnitude and direction to those of unimpaired adults. These deficits in force production may underlie many of the postural and walking balance impairments attributed to individuals with lower limb loss.

Acknowledgments

The authors would like to thank Stacie Chvatal, PhD and Yi Ying Tsai for their contributions to the design of the experiment and her help with data collection.

Fig. 5. Direction of resultant horizontal force vectors. A: Direction (mean ± 1SD) of the net resultant horizontal force vectors for each of the 12 sway directions. The direction of the net resultant horizontal force vectors was similar between controls (black squares) and individuals with transtibial limb loss (red circles) across sway directions, the main exception being sway in the direction of the prosthesis (leftward). B: Direction (mean ± 1SD) of the resultant horizontal force vector exerted by the prosthetic (red circles) and left leg of controls (black squares). Controls exerted forces across a wider range of directions than individuals with transtibial limb loss who tended to constrain the forces they exerted with their prosthetic leg to two principle directions, approximately 180° and 0°. Note that individuals with transtibial limb loss did not exert forces with their prosthetic leg across a range of directions, namely 45° to 160° (shaded area). C: Direction (mean ± 1SD) of the resultant horizontal force vector exerted by the intact leg of individuals with transtibial limb loss (red circles) and right leg of controls (black squares). Note that the direction of the resultant horizontal force vectors was virtually equivalent across all sway directions.

References


