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Assessing preparative gait adaptations in persons with transtibial amputation in response to repeated medial-lateral perturbations☆

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Abstract

Preventing loss of balance in individuals with transtibial amputation is important, as they are susceptible to a high frequency of fall related injuries. In order to validate fall prevention and balance therapies, methods to assess gait stability must be developed. Kinematic, temporal-spatial, and center of mass data from six healthy young participants with transtibial amputation were collected during treadmill walking during exposure to 10 randomly ordered discrete medial-lateral perturbations. The 20 strides prior to each perturbation were assessed for anticipatory changes. The only consistent postural adjustment made as a result of the perturbations was a significantly lowered center of mass height ($p = 0.016$).

Keywords

Perturbations; Kinematics; Amputees; Variability; Gait

1. Introduction

Falls induced by loss of balance are common in individuals with lower extremity amputations and can lead to serious injuries and decreased balance confidence [1–5]. Improving the ability to respond to a loss of balance is integral in reducing fall related injuries. An equally important but often overlooked consideration is the method selected to assess walking stability. A common strategy is to quantify an individual's response to repeated walking perturbations within a single session. However, repeated perturbations

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may elicit altered gait mechanics compared to unperturbed walking. If gait mechanics are changed, then using the response to these perturbations as a method to analyze stability would be invalid. Several studies have investigated the gait adaptation resulting from inducing anterior-posterior (A/P) slips in healthy individuals. [6–9]. Results show an anterior shift in the COM [6,7] and a reduction in foot contact angles [8,9] during unperturbed walking in response to the induced slips. Moreover, the altered gait patterns were retained as long as one year following the single session [6–8].

Presently, no studies have assessed whether repeated medial-lateral (M/L) perturbations applied to the base of support result in gait adaptations. A lack of data examining M/L stability is particularly relevant in persons with transtibial amputation (TTA). The absence of proprioceptive feedback and musculature below the level of the amputation compromises the normal ankle inversion and eversion strategy used to maintain M/L gait stability [10]. As a result persons with TTA may be more susceptible to M/L instability [11]. Improving the response to M/L perturbations could reduce the frequency and severity of fall related injuries among individuals with TTA. However, a method of validating improved stability and perturbation response is needed. As a precursor to a balance training intervention, this study evaluates a method to analyze balance and stability, and addresses whether individuals with TTA apply anticipatory gait adaptations as a result of repeated M/L perturbations.

2. Methods

2.1. Subjects

Six healthy, young men (age: 29 ± 6 years, height: 1.87 ± 0.04 m and mass: 99.9 ± 10.2 kg) with traumatic TTA participated. Participants were screened to ensure that, for a minimum of two months prior to testing, they were able to independently ambulate without an assistive device for at least five consecutive minutes. Participants provided written informed consent prior to participation in this institutionally approved study.

2.2. Experimental protocol

All participants walked on a treadmill in a Computer Assisted Rehabilitation Environment (CAREN) system (Motek, Amsterdam, Netherlands) consisting of a 7 m diameter dome with a virtual environment projected 300° around the individual, providing optic flow [12]. Participants completed a 3–5 min acclimation period, followed by 15 min of walking including ten (five left, five right) randomly ordered platform perturbations. Participants were asked whether they would like to rest after the acclimation period, and several times during the 15 min of walking. Perturbations were directed medially and were initiated at contralateral toe-off. Total displacement of the platform for each perturbation was 5 cm, and the maximum attained velocity and acceleration during each event was 0.28 m/s and 0.46 m/s^2 respectively. Full body kinematics were collected at 60 Hz during all trials using 57 reflective markers and a 24-camera Vicon motion capture system (Vicon, Oxford, UK) [13].

2.3. Data analysis

Marker position data were filtered using a 4th order low-pass Butterworth filter with a 6 Hz cut-off frequency. Marker positions and joint centers were used to create a 13-segment

whole body model with center of mass (COM) [14]. Kinematics were assessed using previously described methods [13]. Data were time normalized to 0–100% of the gait cycle.

Step length (SL), step width (SW), and step time (ST) were defined as the A/P distance, M/L distance, and time between successive, contralateral heel strikes respectively. Within-subject variability for temporal-spatial measures was defined as the standard deviation across 20 (10 right/10 left) continuous strides prior to each perturbation. Sagittal plane kinematic and COM variability were quantified as mean ± SD: the average width of the standard deviation for each entire gait cycle throughout the 20 strides [15]. Finally, we quantified lateral stability as the minimum margin of stability during stance [16].

To look at anticipatory responses as a result of repeated perturbations during gait, we analyzed the data from gait cycles prior to each perturbation. The 20 stride cycles preceding the first perturbation were used as a baseline and compared with the analogous cycles for perturbations 2–10. Kinematic peaks and temporal-spatial parameters were compared using a series of two-factor (Time by Limb), within-subjects, ANOVAs to test for differences between prosthetic and intact limbs during walking prior to perturbations (2–10) (SPSS 16, Chicago, IL). A single-factor, within-subjects ANOVA was used to explore differences in COM variability over time. Estimated marginal means with a Bonferroni correction for multiple comparisons were used for post-hoc analysis of significant interaction effects.

3. Results

A significant main effect of time (from pre-perturbation 1 to pre-perturbation 10) was observed for the average COM height during stance (<0.004 m, approximately 0.2% body height; $p = 0.016$; Fig. 2) as well as a small, but significant decrease in peak knee flexion during swing (<1°; $p = 0.04$; Fig. 3). There was a significant difference between limbs for mean SW ($p = 0.024$; Fig. 1) and hip kinematic variability ($p = 0.025$). There was a significant limb × perturbation interaction effect for ankle dorsiflexion in mid to late stance phase ($p = 0.014$). Post-hoc analyses found no significant changes over time for either limb when assessed independently. There were no other significant differences in kinematics or step measures. The lateral margin of stability was unaffected by limb ($p = 0.461$) or perturbation ($p = 0.599$).

4. Discussion

While the importance of maintaining lateral-stability is widely accepted, the effects of repeated M/L perturbations during gait are unknown. In the present study, a unique environment was used to assess M/L stability in individuals with TTA. Exposure to repeated M/L perturbation in this study elicited only small postural changes in these individuals with TTA. While there were a few between limbs differences (mean SW, hip kinematic variability), which are expected in individuals with TTA [18], there were no differences in the way the limbs anticipated the perturbations.

Participants decreased the height of their center of mass slightly (~0.2% body height), but significantly over time. It has been suggested that lowering COM, and therefore decreasing the moment arm between COM and ground reaction force, increases stability by requiring a

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greater amount of force to induce a fall [17]. The amount of lowering required to provide a clinically meaningful improvement in stability, however, is uncertain. Significant adaptations in step parameters and kinematics might be expected in conjunction with changes in the COM [17], however, we did not observe that here. Such small changes over the course of ten perturbations could also be interpreted as the result of fatigue rather than a strategic adaptation. However, no participant opted to rest at any point, and the investigators did not observe any cardiovascular fitness issue that prompted them to mandate a rest period.

It is possible that we did not have a large enough sample size to detect changes in the joint kinematics. However, this seems unlikely as we did have sufficient statistical power to detect a change in knee kinematics of <1° (Fig. 2). Additionally all observed changes in kinematics were quite small, and thus are unlikely to be physically meaningful despite statistical significance.

The results of this study show that only a slight postural change was made by the individuals with TTA as a result of exposure to repeated discrete perturbations. The lack of substantial adjustments made may be due to an inability of the participants to recognize when a perturbation was about to occur. The CAREN system facilitates a continuous walking environment with no physical cues to indicate an upcoming perturbation. Utilizing a treadmill mounted on a moveable platform to deliver a perturbation greatly reduces the ability of subjects to anticipate during which step a perturbation may occur; whereas, walking in a room with floor mounted force plates constrains the perturbations to a visibly observable range of steps. The nature of the perturbations used in this study was not so large to produce a training effect. Rather, because no preparatory adaptations have been perceived, this study illustrates the feasibility of using similar protocols to compare the post perturbation lateral stability and perturbation recovery strategies before and after various training therapies.

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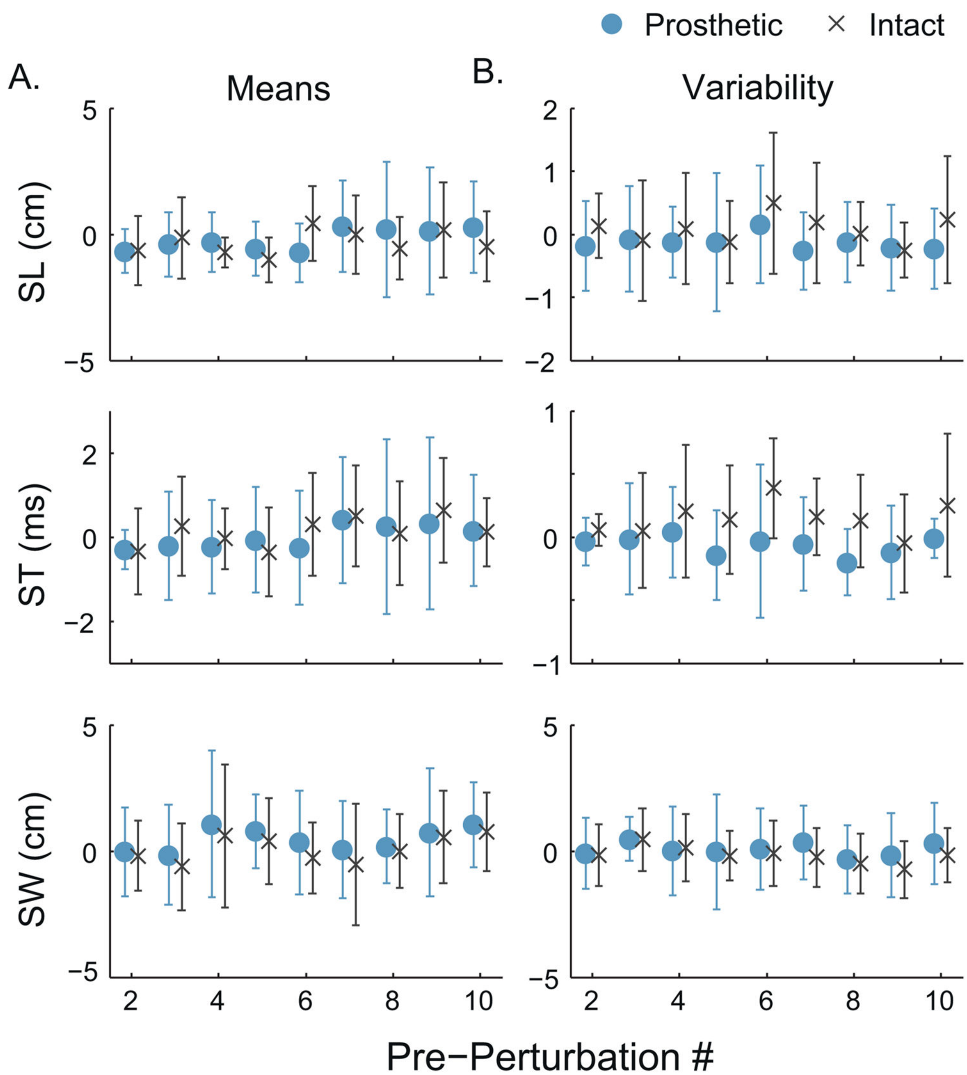

Fig. 1.
(A) Mean and (B) variability of temporal-spatial measures for all subjects are shown as the difference in cm between the 10 strides prior to each perturbation over time (Pert 2–10) compared to the 10 strides prior to the first perturbation, with 0 indicating no difference. Error bars represent the 95% confidence interval about the mean. Significant limb effect is seen in for SW (p = 0.024).

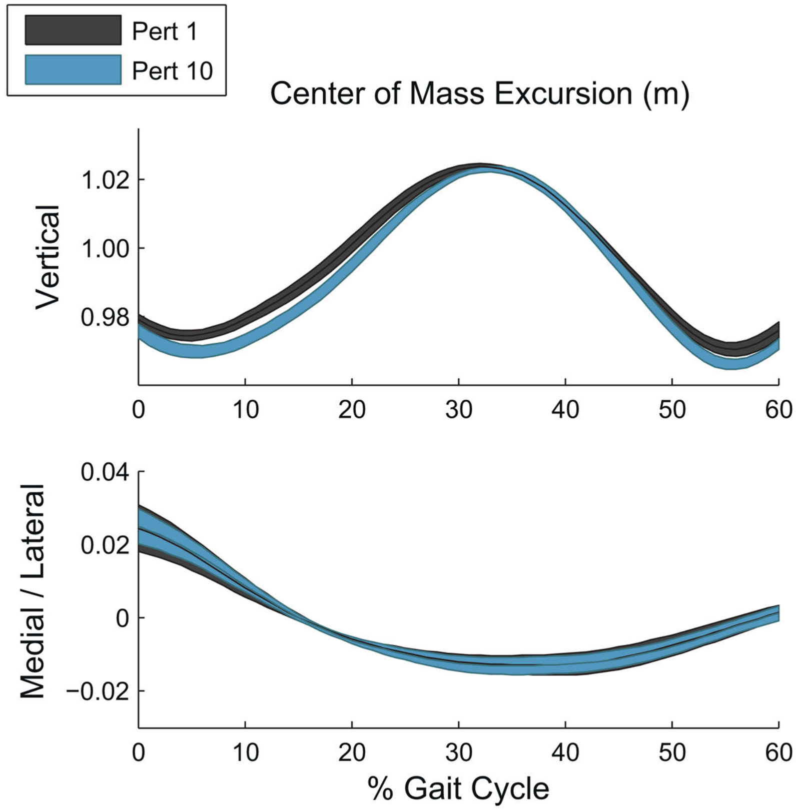

Fig. 2.

Center of mass (COM) excursion over an intact side gait cycle in the vertical and medial-lateral directions are shown for a single representative subject (there were no differences between sides). Bands represent the 95% confidence interval of the mean COM motion across the 10 strides. A significant main effect of time was seen as a change in COM height ($p = 0.016$).

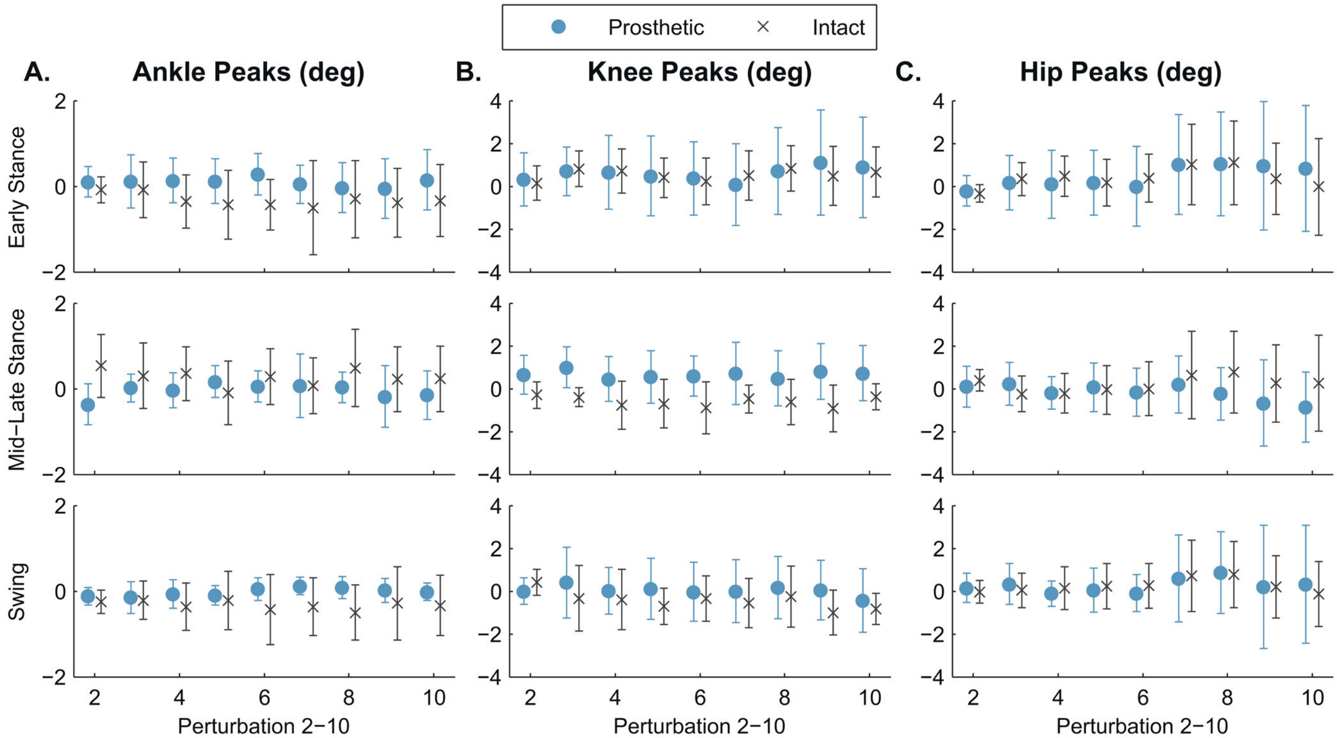

Fig. 3.

Mean kinematic peaks for all subjects during Early Stance (0–25% gait cycle), Mid/Late Stance (26–65% gait cycle), and Swing (66–100% gait cycle) are shown as the difference in degrees between the 10 strides prior to each perturbation over time (Pert 2–10) compared to the 10 strides prior to the first perturbation, with 0 indicating no difference. Error bars represent the 95% confidence interval about the mean. A small (<1°) but significant main effect of time was seen in knee kinematics during swing ($p = 0.04$).