WHOLE-BODY ANGULAR MOMENTUM DURING SLOPED WALKING WITH POWERED AND PASSIVE ANKLE-FOOT PROSTHESES

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INTRODUCTION
Sloped walking places increased demand on the ankle joint [1], and results in a greater risk of slipping than level walking due to an increased shear force between the foot and ground [2]. Sloped walking may thus be especially challenging for individuals with transtibial amputation (TTA), who have a greater risk of falling compared to able-bodied individuals (AB) [3]. This increased fall risk is partially attributed to the functional loss of the ankle muscles, which are critical for controlling dynamic balance during walking [4].

Whole-body angular momentum ($H$) has been previously used to investigate dynamic balance [5, 6], but has not been quantified for individuals with TTA during sloped walking. In addition, powered prostheses (PWR) actively generate torque at the ankle joint, which may affect dynamic balance differently than passive energy storage and return prostheses (ESR). The influence of PWR on dynamic balance during sloped walking is unclear, and thus the purpose of this study was to analyze $H$ during sloped walking while using both powered and passive prostheses.

METHODS
Six adult males with (n=3) and without (n=3) TTA walked on slopes of ±10°, ±5°, and 0°. Individuals with TTA were assessed using their clinically prescribed ESR prosthesis and also with the BiOM (BiOM, Bedford, MA) powered prosthesis. Whole-body kinematics were collected at 120 Hz while participants walked at a fixed walking speed based on leg length.

Kinematic data were filtered at 6 Hz using a 4th order low pass Butterworth filter. A 13-segment inverse dynamics model was used to compute $H$ as:

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

where $n$ is the number of segments, $\vec{r}_i^{COM}$, $\vec{v}_i^{COM}$, and $\vec{\omega}_i$ are, respectively, the position, velocity, and angular velocity of the $i$th segment, $\vec{r}_i^{COM}_{body}$ and $\vec{v}_i^{COM}_{body}$ are, respectively, the position and velocity of the body center of mass (COM), and $m_i$ and $I_i$ are the mass and inertia matrix of the $i$th segment. For each of the three anatomical planes as well as the overall vector magnitude, $H$ was normalized by body height and weight and expressed from 0 to 100% of the left or prosthetic limb gait cycle. Characteristic features (range, mean) of $H$ during the first (0-50%) and second (50-100%) halves of the gait cycle were compared across groups and slopes using a linear mixed effects ANOVA with slope angle and group (AB, ESR, PWR) as fixed effects and subject as a random effect nested within type (AB, TTA). Post-hoc pairwise comparisons were performed when significant main or interaction effects were found, and $p$-values were adjusted using Tukey’s method. All tests used a significance level $\alpha=0.05$.

RESULTS AND DISCUSSION
Although $H$ was analyzed in each of the three anatomical planes, the primary findings were in the sagittal plane and in the vector magnitude of $H$ during the first half (0-50%) of the prosthetic (left) limb gait cycle (Fig. 2). For all slope conditions, both ESR and PWR resulted in increased range of sagittal-plane $H$ relative to AB (Figs. 1 & 2). In addition, the mean magnitude of $H$ was increased in ESR relative to AB on declines, and was increased in both ESR and PWR relative to AB on inclines. The increased range and magnitude of $H$ in individuals with TTA suggest that it may be more difficult for them to maintain their balance on slopes, since the affected limb lacks ankle muscles.
that are important for regulating $H$ [7]. This potential increase in fall risk for individuals with TTA is in addition to the already increased risk of falls on slopes relative to level ground [2].

Figure 1: Normalized whole-body angular momentum ($H$) trajectories in the sagittal plane (top row) and total vector magnitude (bottom row) for 3 sloped walking conditions.

Figure 2: Mean ($\pm$SD) of the range (peak-to-peak value) of sagittal $H$ (top) and magnitude of the $H$ vector over the first 50% of the gait cycle. Significant differences between groups are indicated by ‘$*$’.

Few differences were observed in $H$ when using ESR relative to PWR, similar to previous results on stairs [6]. However, one notable difference was that individuals with TTA had increased sagittal-plane range and mean magnitude of $H$ with ESR compared to PWR when walking on a 10° decline. This result suggests that PWR performs more similarly to AB than ESR, potentially because of controlled plantarflexion in early stance and active plantarflexion in late stance [8]. These functions may help in regulating $H$ during decline walking for individuals with TTA.

All groups had increased sagittal-plane range and mean vector magnitude of $H$ at $+10^\circ$ relative to $0^\circ$ in the first half of the gait cycle. However, only AB had a decreased mean magnitude of $H$ at $-10^\circ$ relative to $0^\circ$, suggesting that individuals with TTA do not regulate $H$ as tightly as AB during decline walking. The risk of falling during decline walking is greater than during incline walking because of differences in the timing of the peak shear GRF [2]. AB may be more tightly controlling $H$ in response to this greater risk of falling [5], which may be difficult for individuals with TTA.

CONCLUSIONS

Differences in $H$ suggest altered control of dynamic balance across different sloped walking conditions, between individuals with TTA and AB, and when using different prostheses. Future work will incorporate additional subjects and investigate the external moment to further understand dynamic balance control.

REFERENCES


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