WHOLE-BODY ANGULAR MOMENTUM IN INDIVIDUALS WITH UNILATERAL TRANSTIBIAL AMPUTATION DURING STAIR WALKING

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INTRODUCTION

Whole-body angular momentum (H) must be carefully controlled during walking to maintain dynamic balance and avoid falling. Humans regulate H during walking by altering the net external moment about the body center-of-mass (COM), which equals the time rate of change of H and is a function of ground reaction forces (GRFs) and external moment arms. Muscles are the primary contributors to the net external moment [1], and are therefore the primary regulators of H. Individuals with transtibial amputation (TTA) have altered H compared to able-bodied (AB) individuals during walking on level ground [2], likely due to the functional loss of the ankle plantarflexors. On stairs, TTA also have altered kinematics and kinetics relative to AB [3], but the differences in H between TTA and AB during stair walking remain unclear.

Recently, prostheses with active ankle joints have been developed [4]. These devices generate power at the ankle and reduce the metabolic cost of overground walking in TTA [5]. However, the effects of powered prostheses on H are unknown. Therefore, the goal of this study was to investigate H in TTA using both passive and powered prostheses relative to AB during stair ascent, level walking and stair descent. GRFs, external moment arms and net joint powers were also investigated to help interpret the H results.

METHODS

Nine TTA and nine AB participants completed a biomechanical walking assessment. TTA were assessed while using their clinically prescribed passive energy storing and return (ESR) prosthesis and with the BiOM (iWalk, Bedford, MA) powered prosthesis (PWR). Whole-body kinematics and GRF data were collected at 120 Hz and 1200 Hz, respectively, while participants ascended and descended a 16-step staircase with four instrumented steps using a step-over-step pattern at a fixed cadence of 80 steps/min. The participants also walked on level ground at a fixed walking speed based on leg length.

Kinematic and kinetic data were filtered at 6 Hz and 50 Hz, respectively, using a 4th order low pass Butterworth filter. A 13-segment inverse dynamics model was used to compute net joint powers and H as

$$\mathbf{H} = \sum_{i=1}^{n} [(r_{i}^{COM} - r_{body}^{COM}) \times m_{i} (v_{i}^{COM} - \mathbf{\omega}_{body}^{COM}) + \mathbf{I}_{i} \mathbf{\omega}_{i}]$$

where n is the number of segments, \(r_{i}^{COM}, v_{i}^{COM},\) and \(\mathbf{\omega}_{i}\) are, respectively, the position, velocity, and angular velocity of the \(i^{th}\) segment, \(r_{body}^{COM}\) and \(v_{body}^{COM}\) are, respectively, the position and velocity of the whole-body COM, and \(m_{i}\) and \(I_{i}\) are the mass and inertia matrix of the \(i^{th}\) segment. H was normalized by subject height and body weight and expressed from 0 to 100% of the gait cycle. The range of H in all three anatomical planes, peak values of the GRFs, external moment arms, and net joint powers were statistically compared across walking condition (stair ascent, level walking and stair descent) and subject group (ESR, PWR, AB) using a mixed model, repeated measures ANOVA with post hoc pairwise comparisons (\(\alpha=0.05\)).

RESULTS AND DISCUSSION

No statistically significant differences in the range of H were observed between groups (ESR, PWR, AB) in the frontal or transverse planes for any walking condition.
During stair ascent, the range of $H$ in the sagittal plane during the first half of the gait cycle (during prosthetic limb stance) was greater for both ESR ($p=0.008$) and PWR ($p=0.004$) relative to AB (Fig. 1). The range of sagittal $H$ during the second half of the gait cycle was similar between AB and both ESR and PWR. The range of $H$ was not significantly different between ESR and PWR. Similarly, in level walking, there was no difference in the range of $H$ between ESR and PWR. The range of $H$ was larger for ESR and PWR relative to AB in the first half of the gait cycle, but not significantly different in the second half, contrary to previous results [2].

The peak vertical GRF in the intact (trailing) limb was increased in ESR ($p=0.015$) and PWR ($p<0.001$) relative to AB during the push-up phase (0-10% prosthetic limb gait cycle) of stair ascent, contributing to a negative (forward) net external moment and negative slope of the $H$ trajectory. In addition, the prosthetic limb anterior external moment arm was increased during pull-up in ESR ($p<0.001$) and PWR ($p=0.017$) relative to AB, contributing to a greater positive (backward) external moment from the prosthetic (leading) limb vertical GRF (10-30% of the prosthetic limb gait cycle). Further, peak prosthetic limb hip extension power generation during pull-up was increased in ESR ($p<0.001$) and PWR ($p<0.001$) relative to AB during stair ascent, suggesting a greater contribution from the hip extensors to positive $H$ [1].

During stair descent, all groups had a significantly reduced range of sagittal $H$ relative to level walking, but there were no significant differences between groups. This reduced range of sagittal $H$ during stair descent is similar to results for AB during decline slope walking [6], and may be a protective mechanism to reduce fall risk.

**CONCLUSIONS**

Significant differences in net joint powers between TTA and AB suggest TTA use compensations (e.g., prosthetic limb hip power and intact limb ankle power) during stair walking, regardless of prosthesis type [3]. The lack of significant differences in $H$ between the passive and powered prosthesis suggests that, although the powered prosthesis provides increased ankle power generation in the prosthetic limb, it does not provide a distinct advantage over a passive prosthesis in the regulation of $H$.

**REFERENCES**


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