DIFFERENCES IN SAGITTAL PLANE ANGULAR MOMENTUM BETWEEN BELOW-KNEE AMPUTEES AND NON-AMPUTEES ACROSS A RANGE OF WALKING SPEEDS

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INTRODUCTION

The regulation of whole-body angular momentum is important to prevent falls and recover from trips [1, 2]. Previous studies have shown that the ankle muscles are critical in the regulation of angular momentum. For example, when recovering from a trip, fallers have a reduced peak ankle plantar flexor moment relative to non-fallers [2], which limited their ability to restrain their forward angular momentum. Below-knee amputees have an increased risk and fear of falling relative to non-amputees [3], which may be due to their lack of ankle muscles to help regulate their angular momentum. The purpose of this study was to identify differences in sagittal plane, whole-body angular momentum between below-knee amputees and non-amputees over a wide range of steady-state walking speeds. We analyzed angular momentum across different walking speeds to further highlight differences between amputees and non-amputees, as the regulation of angular momentum has been shown to vary with walking speed [4].

METHODS

Kinematic and kinetic data were collected from twelve amputee and ten control subjects walking at 0.6, 0.9, 1.2, and 1.5 m/s [5]. Kinematic data were collected using an 8-camera motion capture system (Vicon) and a cluster marker set at 120 Hz. Ground reaction force (GRF) data were collected using four AMTI force plates embedded in a 10-m walkway at 1200 Hz. Data were processed in Visual3D (C-Motion). Kinematic and GRF data were low-pass filtered with a cutoff frequency of 6 Hz and 20 Hz, respectively. An inverse dynamics model was used to find the center of mass location and velocity of eight body segments including the left and right thighs, shanks, feet, pelvis and torso. The calculation of whole body angular momentum about the body center of mass was determined as:

$$ H = \sum_{i=1}^{n} \left[ (\vec{r}_i^C - \vec{r}_CM) \times (\vec{v}_i^CM - \vec{v}_CM) + I_i \vec{\omega}_i \right] $$

where $\vec{r}_i^C$, $\vec{v}_i^CM$ and $\vec{\omega}_i$ are the position, velocity and angular velocity vectors of the $i$-th segment’s center of mass in the laboratory coordinate system, $\vec{r}_CM$ and $\vec{v}_CM$ are the position and velocity vectors of the total body center of mass in the laboratory coordinate system, and $m_i$ and $I_i$ are the mass and moment of inertia of each body segment.

Angular momentum was normalized by body mass (kg), walking speed (m/s) and body height (m) and then normalized to the intact leg gait cycle for amputees and the right leg gait cycle for non-amputees. Trials were then averaged at each speed for each subject. Sagittal plane angular momentum was defined as positive when directed counterclockwise (torso tilting backward) and negative when directed clockwise (torso tilting forward).

The range of angular momentum, defined as the peak to peak value, in the first and second halves of the gait cycle was compared between groups and across speeds using two, two-factor ANOVAs. The first factor (group) had two levels: amputee and non-amputee. The second factor (speed) had four levels: 0.6, 0.9, 1.2 and 1.5 m/s. When significant differences were found, pair-wise comparisons using a Bonferroni adjustment for multiple comparisons were performed to determine which conditions were significantly different ($\alpha = 0.05$).

RESULTS AND DISCUSSION

There were significant differences in the range of angular momentum between the amputee and non-amputee subjects across speeds (Fig. 1).
Figure 1: Average sagittal plane angular momentum across walking speeds for amputees and non-amputees.

In the first half of the gait cycle, there were significant group (p<0.001), speed (p<0.001), and interaction (p=0.001) effects on the range of angular momentum. The range was significantly smaller in the amputees compared to non-amputees at all four walking speeds (p≤0.039), resulting in a more negative (forward) angular momentum in amputees (Fig. 1). The range of angular momentum significantly decreased with walking speed for both amputees and non-amputees (p<0.001), except with the amputees between 1.2 and 1.5 m/s.

In the second half of the gait cycle, there were also significant group (p=0.003), speed (p<0.001), and interaction (p<0.001) effects on the range of angular momentum. However, in the second half of the gait cycle, the range was significantly larger in the amputees compared to non-amputees at 0.9, 1.2 and 1.5 m/s (p≤0.004). The range significantly decreased with each increase in speed for both groups (p<0.001).

Previous studies have shown that there is a smaller propulsive GRF from the residual leg during early intact leg stance (late residual leg stance) compared to non-amputees [5]. A reduced propulsive GRF decreases the positive external moment on the body and corresponding positive rate of change in angular momentum. Similarly, there is a reduced residual leg braking GRF in the second half of the intact leg gait cycle [5] that acts to decrease the negative external moment, which increases the positive rate of change in angular momentum. To assess if the reduced residual leg propulsion is associated with the smaller range in amputee angular momentum in early stance (Fig. 1), a post-hoc Pearson correlation analysis was performed between the peak residual leg propulsive GRF and the range of angular momentum in the first half of the gait cycle at each walking speed. A significant correlation was found at each walking speed (0.44 ≤ r ≤ 0.7, p≤0.04). The analysis was also performed with the peak residual leg braking GRF and the range of angular momentum in the second half of the gait cycle, and a significant correlation was found for 0.9, 1.2 and 1.5 m/s (0.63 ≤ r ≤ 0.7, p≤0.002).

CONCLUSION

These results indicate there are differences in how amputees regulate their angular momentum compared to non-amputees. The lack of residual leg propulsion results in more negative (forward) angular momentum in the first half of the gait cycle. To compensate, reduced residual leg braking in the second half of the gait cycle increases the positive (backward) angular momentum to help restore the momentum balance. Thus, decreased residual leg braking appears to be an important mechanism to regulate angular momentum in amputee walking.

REFERENCES


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