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Whole-body angular momentum during stair ascent and descent

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ABSTRACT

The generation of whole-body angular momentum is essential in many locomotor tasks and must be regulated in order to maintain dynamic balance. However, angular momentum has not been investigated during stair walking, which is an activity that presents a biomechanical challenge for balance-impaired populations. We investigated three-dimensional whole-body angular momentum during stair ascent and descent and compared it to level walking. Three-dimensional body-segment kinematic and ground reaction force (GRF) data were collected from 30 healthy subjects. Angular momentum was calculated using a 13-segment whole-body model. GRFs, external moment arms and net joint moments were used to interpret the angular momentum results. The range of frontal plane angular momentum was greater for stair ascent relative to level walking. In the transverse and sagittal planes, the range of angular momentum was smaller in stair ascent and descent relative to level walking. Significant differences were also found in the ground reaction forces, external moment arms and net joint moments. The sagittal plane angular momentum results suggest that individuals alter angular momentum to effectively counteract potential trips during stair ascent, and reduce the range of angular momentum to avoid falling forward during stair descent. Further, significant differences in joint moments suggest potential neuromuscular mechanisms that account for the differences in angular momentum between walking conditions. These results provide a baseline for comparison to impaired populations that have difficulty maintaining dynamic balance, particularly during stair ascent and descent.

1. Introduction

Walking on stairs is a common activity of daily living that is important for functional mobility and independence. Stair walking presents a greater biomechanical challenge relative to walking on level ground because the body center-of-mass (COM) must be raised during ascent and lowered during descent during single limb support while maintaining forward progression and proper foot placement. As a result, larger joint moments and joint ranges of motion are required for stair ascent and descent [1–3]. Previous studies have shown that walking on stairs also involves a greater challenge for maintaining dynamic balance, and populations who experience balance deficits, such as the elderly, often have difficulty negotiating stairs [4,5]. Specifically, a review on the causes of falls in the elderly reported that falls most frequently occur on stairs, and falls down the stairs can result in death [4].

The regulation of whole-body angular momentum is important for maintaining dynamic balance during walking to avoid falling [6]. Angular momentum has been shown to vary across walking tasks and to be regulated differently across patient populations [7–11]. A number of studies have investigated angular momentum during trip recovery [12–14] and have highlighted the actions of the support and recovery limbs in arresting angular momentum to prevent falling. Given the greater biomechanical challenge of negotiating stairs and the increased occurrence of falls on stairs, it is reasonable to expect that angular momentum will be different during stair walking relative to level walking.

The external moment about the body COM equals the time rate of change of whole-body angular momentum. Thus, alterations in the external moment arm (i.e., the COM to center-of-pressure distance) or the magnitude of ground reaction forces (GRFs) will affect the net external moment about the COM and therefore the

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angular momentum trajectory. Muscles, as the principal contributors to the GRFs, are the primary mechanism to regulate (i.e., generate and arrest) whole-body angular momentum [15].

A number of studies have investigated the biomechanics of stair climbing including joint kinematics, joint kinetics, GRFs and electromyography (EMG) [1–3,16–19]. These studies have identified important biomechanical differences during stair walking that will likely result in an altered angular momentum trajectory. For example, altered GRF peaks and kinematics will change the net external moment about the COM. Further, through joint kinetic and EMG results, previous studies suggest that the muscles that contribute to support of the body COM and the regulation of angular momentum, such as the gluteus maximus, vastii and ankle plantarflexor muscles [15,20,21] have a critical role during stair ascent and descent. However, how angular momentum is regulated during stair walking is unknown.

Therefore, the purpose of this study was to investigate whole-body angular momentum during stair ascent and descent in healthy subjects. We hypothesized that the overall range of angular momentum would be larger during stair walking relative to level ground because of the greater GRFs and joint kinetics observed during stair walking [1,2,17]. In addition, we investigated GRFs, external moment arms and net joint moments during stair walking to help interpret any observed differences in the angular momentum results. The results of this study will provide insight into how healthy individuals maintain dynamic balance during stair walking and provide a baseline for comparison with balance-impaired populations.

2. Methods

Thirty healthy subjects (13 male, 17 female; 21.8 ± 4.2 years; 73.3 ± 14.8 kg; 1.7 ± 0.1 m) provided written, informed consent to participate in this study approved by the Institutional Review Board at Brooke Army Medical Center, Ft. Sam Houston, TX. Subjects walked at a fixed cadence (80 steps per minute) up and down an instrumented staircase with 16 stairs as well as over a level walkway. A 26-camera motion capture system tracked 55 markers at 120 Hz to quantify full-body motion [22]. Three-dimensional GRFs were measured at 1200 Hz using two force plates embedded in an interlaced staircase design [23].

Biomechanical data were processed in Visual3D (C-Motion, Inc., Germantown, MD, USA). A low-pass, fourth-order Butterworth filter was applied to the kinematic and GRF data, with cut-off frequencies of 6 Hz and 50 Hz, respectively. A 13-segment model was used to estimate the COM location and velocity of each segment including the head, torso, pelvis, upper arms, lower arms, thighs, shanks and feet (Fig. 1, [8]). Each segment mass was calculated as a percentage of total body mass [24] and segment inertial properties were determined from kinematic marker placement and estimates of segment geometry. Whole-body angular momentum ($\vec{H}$) about the COM was calculated as:

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

where $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$ and $\vec{\omega}_i$ are the position, velocity and angular velocity vectors of the i-th segment's COM, $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the whole-body COM, m is the segment mass, I is the segment moment of inertia, and n is the number of segments. Whole-body angular momentum was normalized in magnitude by body mass (kg) and body height (m), and normalized in time to the left leg gait cycle.

The ranges of the frontal, transverse and sagittal angular momentum components, defined as the peak-to-peak values over the gait cycle, were compared across the three conditions (stair descent, level walking and stair ascent). To help interpret the angular momentum results, the magnitudes of the peak GRFs, external moment arms and joint moments, averaged between the right and legs, were also compared across condition. Significant main effects were assessed using a one-factor, repeated-measures ANOVA for normally distributed data and Friedman's test for non-normally distributed data. Pairwise comparisons were performed using paired t-tests with a Bonferroni adjustment for multiple comparisons for normally distributed data and Wilcoxen Signed Rank tests for non-normally distributed data (α = 0.05).

3. Results

Significant main effects were observed for the range of angular momentum in all three planes (Table 1). Similarly, the peak GRFs, external moment arms and joint moments had significant main effects across walking conditions (Table 1), with significant differences between walking conditions.

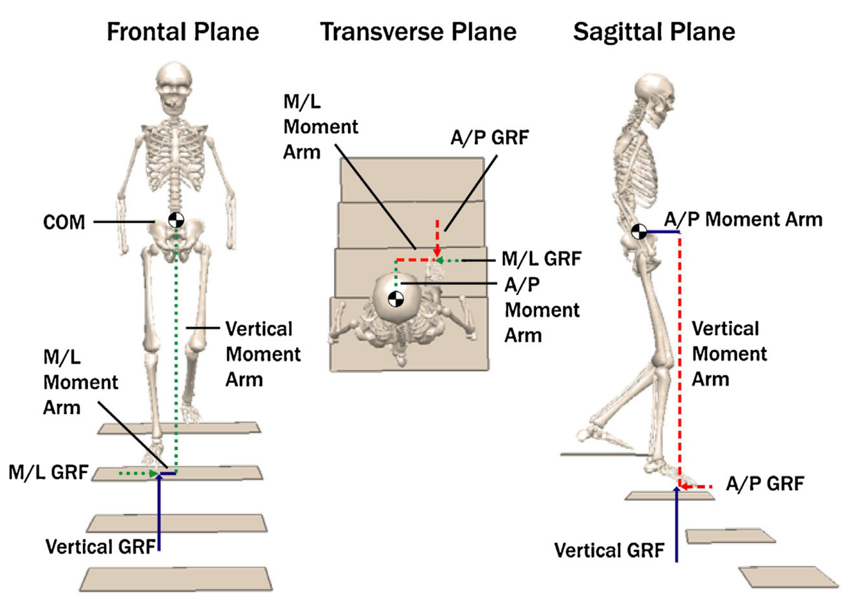

Fig. 1. Model used to calculate whole-body angular momentum. The external moment about the center-of-mass (COM) equals the time rate of change of whole-body angular momentum, which is computed as the cross product of the external moment arm and ground reaction force (GRFs) vectors. The right leg contributions to the external moment are shown during stair descent.