# Whole-body angular momentum during stair ascent and descent

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A R T I C L E   I N F O

A B S T R A C T

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; $21.8 \pm 4.2$ years; $73.3 \pm 14.8$ kg; $1.7 \pm 0.1$ 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:

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

where $\vec{r}_i^{\text{COM}}$, $\vec{v}_i^{\text{COM}}$ and $\vec{\omega}_i$ are the position, velocity and angular velocity vectors of the $i$-th segment's COM, $\vec{r}_{\text{body}}^{\text{COM}}$ and $\vec{v}_{\text{body}}^{\text{COM}}$ 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 ($\alpha = 0.05$).

## 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.
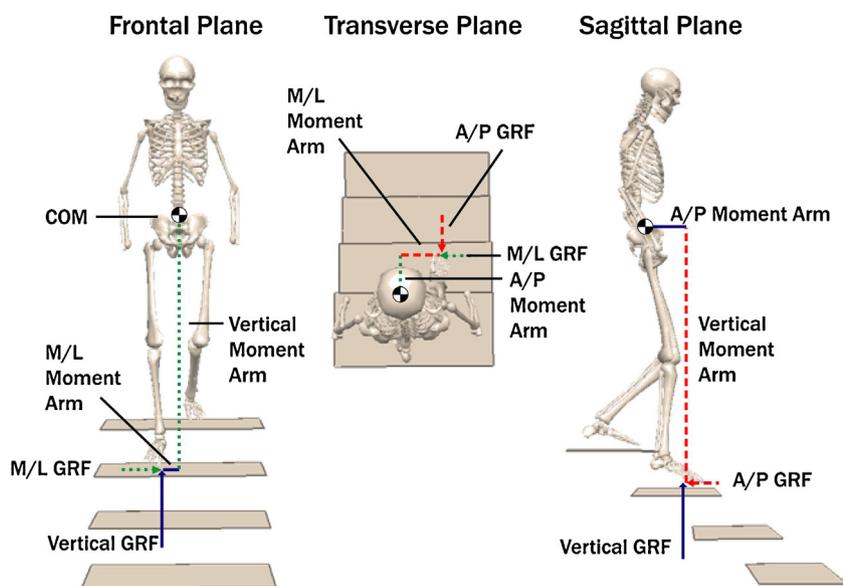


**Fig. 1.** Model used to calculate whole-body angular momentum. The external moment about the center-of-mass (COM) equals the time rate of change of whole-body angular momentum, which is computed as the cross product of the external moment arm and ground reaction force (GRFs) vectors. The right leg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

**Table 1**
Mean values (standard deviation) of the angular momentum ranges, ground reaction force peaks, external moment arm peaks and joint moment peaks. Main effect *p*-values are also shown. Pairwise comparisons were performed between each walking condition, including stair descent (SD), level walking (LW) and stair ascent (SA). Significant differences relative to level walking (*) and stair descent (#) signify that $p \leq 0.05$.

| | Main effect | SD | LW | SA |
|---|---|---|---|---|
| **Angular momentum ranges (m/s)** | | | | |
| Frontal | 0.000 | 0.034 (0.012) | 0.032 (0.010) | 0.048 (0.014)*# |
| Transverse | 0.000 | 0.009 (0.003)* | 0.014 (0.003) | 0.009 (0.002)* |
| Sagittal | 0.000 | 0.026 (0.004)* | 0.040 (0.006) | 0.036 (0.005)*# |
| **Ground reaction forces (BW)** | | | | |
| Anterior/posterior, 1st peak | 0.000 | −0.112 (0.014)* | −0.171 (0.028) | −0.073 (0.013)*# |
| Anterior/posterior, 2nd peak | 0.000 | 0.096 (0.013)* | 0.188 (0.023) | 0.054 (0.012)*# |
| Vertical, 1st peak | 0.000 | 1.217 (0.102)* | 1.069 (0.058) | 1.001 (0.050)*# |
| Vertical, 2nd peak | 0.000 | 0.915 (0.042)* | 1.050 (0.053) | 1.060 (0.054)# |
| Medial/lateral, 1st peak | 0.000 | 0.072 (0.013)* | 0.059 (0.016) | 0.051 (0.012)*# |
| Medial/lateral, 2nd peak | 0.000 | 0.066 (0.013)* | 0.053 (0.014) | 0.045 (0.011)*# |
| **External moment arms (m)** | | | | |
| Anterior/posterior, 1st peak | 0.000 | 0.213 (0.013)* | 0.286 (0.025) | 0.177 (0.017)*# |
| Anterior/posterior, 2nd peak | 0.000 | −0.102 (0.010)* | −0.287 (0.026) | −0.100 (0.013)* |
| Vertical, 1st peak | 0.000 | −1.140 (0.055)* | −1.049 (0.049) | −0.923 (0.051)*# |
| Vertical, 2nd peak | 0.000 | −0.892 (0.069)* | −1.047 (0.048) | −1.125 (0.048)*# |
| Medial/lateral, 1st peak | 0.000 | −0.127 (0.030)* | −0.069 (0.013) | −0.107 (0.028)*# |
| Medial/lateral, 2nd peak | 0.000 | −0.088 (0.022)* | −0.073 (0.017) | −0.078 (0.022)# |
| **Ankle moment (Nm/kg)** | | | | |
| Ab/adduction, 1st peak | 0.002 | 0.036 (0.149)* | −0.087 (0.064) | −0.024 (0.106)* |
| Abduction, 2nd peak | 0.000 | 0.111 (0.070)* | 0.251 (0.108) | 0.189 (0.098)*# |
| Internal rotation peak | 0.000 | 0.092 (0.039)* | 0.134 (0.032) | 0.056 (0.037)*# |
| Plantarflexion, 1st peak | 0.000 | 0.817 (0.162)* | 0.280 (0.291) | 0.678 (0.201)*# |
| Plantarflexion, 2nd peak | 0.000 | 0.995 (0.111)* | 1.339 (0.094) | 1.009 (0.217)* |
| **Knee moment (Nm/kg)** | | | | |
| Abduction, 1st peak | 0.000 | −0.311 (0.108) | −0.306 (0.077) | −0.197 (0.072)*# |
| Abduction, 2nd peak | 0.000 | −0.211 (0.089)* | −0.300 (0.104) | −0.213 (0.088)* |
| Internal rotation, 1st Peak | 0.000 | 0.044 (0.031)* | 0.088 (0.042) | 0.211 (0.056)*# |
| Internal/external rotation, 2nd peak | 0.000 | 0.142 (0.044)* | −0.133 (0.037) | −0.048 (0.043)*# |
| Extension, 1st peak | 0.000 | 0.497 (0.207)* | 0.405 (0.195) | 1.101 (0.222)*# |
| Flexion/extension, 2nd peak | 0.000 | 1.000 (0.157)* | −0.362 (0.135) | −0.119 (0.270)*# |
| **Hip moment (Nm/kg)** | | | | |
| Abduction, 1st peak | 0.000 | −0.826 (0.148) | −0.796 (0.127) | −0.689 (0.113)*# |
| Abduction, 2nd peak | 0.000 | −0.738 (0.148) | −0.732 (0.115) | −0.522 (0.113)*# |
| External rotation, 1st peak | 0.000 | −0.139 (0.061)* | −0.202 (0.055) | −0.197 (0.076)# |
| External rotation, 2nd peak | 0.000 | −0.061 (0.044)* | −0.100 (0.046) | −0.102 (0.062)# |
| Extension peak | 0.000 | −0.118 (0.123)* | −0.579 (0.130) | −0.315 (0.144)* |
| Flexion peak | 0.000 | 0.317 (0.112)* | 0.615 (0.140) | 0.285 (0.139)*# |

### 3.1. Whole-body angular momentum

Our hypothesis that the range of angular momentum would be greater for stair conditions relative to level walking was only partially supported in the frontal plane (Table 1 and Fig. 2). The range of frontal-plane angular momentum was significantly larger for stair ascent relative to level walking and stair descent (Table 1 and Fig. 2), but stair descent was not significantly different from level walking. The range of transverse- and sagittal-plane angular momentum was significantly smaller for both stair conditions relative to level walking.

### 3.2. Ground reaction forces and external moment arms

The anterior/posterior (A/P) GRF had significantly smaller braking (first) and propulsive (second) peaks during stair walking relative to level walking (Table 1 and Fig. 3). The peak A/P GRFs during stair ascent were also smaller than those during stair descent. The vertical GRFs had significant differences between walking conditions; the initial peak was largest during stair descent, followed by level walking and stair ascent. In late stance, the vertical peak was reduced for stair descent relative to stair ascent and level walking. Both medial/lateral (M/L) GRF peaks were largest during stair descent and smallest in stair ascent.

The A/P external moment arm was significantly greater in level walking relative to stair conditions in both early and late stance. The vertical moment arm was smallest for stair ascent and greatest for stair descent in early stance. This relationship was reversed for the peak in late stance, where stair descent was the smallest and stair ascent was the largest. In the M/L direction, the external moment arm was largest for stair descent in early and late stance. The M/L moment arm for stair ascent was larger than level walking in early stance.

### 3.3. Joint moments

The net joint moments had a number of significant differences between walking conditions in all three planes (Table 1 and Fig. 4). In the frontal plane, there was an adductor moment at the ankle in early stance, which transitioned to an abductor moment in late stance during level walking. During stair walking, the ankle moment remained in abduction throughout stance. The peak abduction moment in late stance was much smaller during stair descent relative to level walking and stair ascent. At the knee, the frontal plane abduction moment was reduced in both early and late stance during stair walking relative to level walking. There were fewer significant differences at the hip, with the peak hip abduction moment being smaller in early and late stance for stair ascent relative to level walking and stair descent.

In the transverse plane, the peak ankle internal rotation moment was smaller for the stair conditions relative to level walking. At the knee, the internal rotation moment for stair ascent was largest in early stance. In late stance, the internal knee rotation moment was greatest for stair descent, whereas the level walking condition was characterized by an external rotation moment. At the hip, all three conditions had an external rotation moment throughout stance, which was significantly smaller during stair descent.

In the sagittal plane, both stair conditions had an early ankle plantarflexion moment peak that was not seen in level walking. The peak plantarflexion moment in late stance was largest for level walking. At the knee, the peak extension moment in early stance was greatest during stair ascent. In late stance, the knee moment transitioned to a flexor moment for both stair ascent and level walking, but the stair descent moment remained flexor. At the hip, the peak extension moment was reduced in early stance for stair walking relative to level walking. Similarly, the hip flexion moment was also reduced for both stair walking conditions.

## 4. Discussion

The ranges of whole-body angular momentum were similar to those reported previously for healthy subjects (e.g., [6–9]). Our hypothesis that the range of angular momentum would be greater during stair walking was only partially supported, with the range
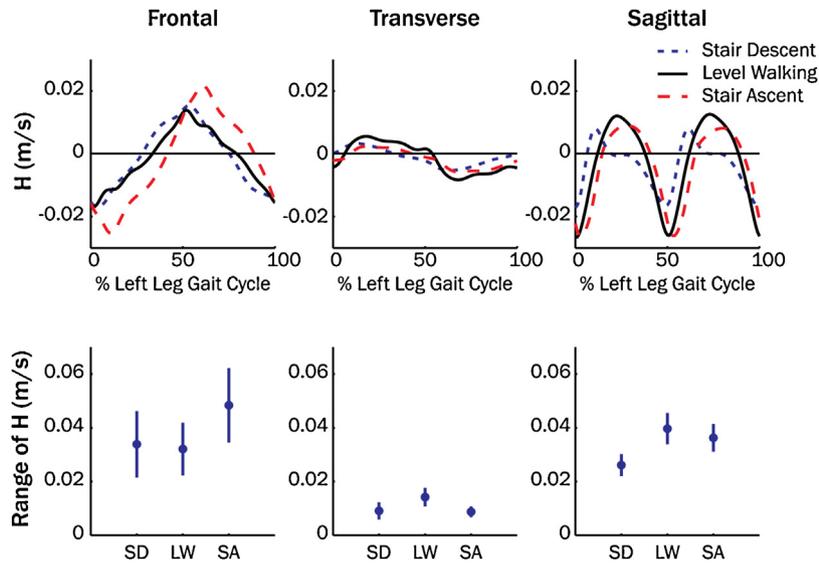
**Fig. 2.** Whole-body angular momentum (H) trajectories and ranges in the frontal, transverse and sagittal planes over the left leg gait cycle. Angular momentum was normalized by body height and body weight and has units of m/s. Vertical lines indicate the standard deviation of the range for stair descent (SD), level walking (LW) and stair ascent (SA).

of angular momentum during stair walking differing from level walking in all conditions except in the frontal plane during descent. Differences in the GRFs, external moment arms and net joint moments helped explain the altered angular momentum trajectories and the mechanisms used to maintain dynamic balance during stair walking.

In the frontal plane, there was a greater range of angular momentum during stair ascent relative to the other walking conditions, partially supporting our hypothesis (Table 1 and Fig. 2). The vertical and M/L GRFs and moment arms contribute to the net external moment in the frontal plane (Fig. 1). As the external moment equals the time rate of change of angular momentum, the greater angular momentum range results from a large positive slope (i.e., external moment) for the first half of the left leg gait

cycle (rotation toward the right leg) and a large negative slope during the second half of the gait cycle (rotation toward the left leg). During the first half of the gait cycle, the smaller M/L GRF and vertical moment arm from the left (leading) leg resulted in a smaller negative external moment, which resulted in an overall increase in the net positive external moment and whole body angular momentum trajectory (toward the right/trailing leg) during early stance. Similarly, in the second half of the gait cycle, the smaller M/L GRF and vertical moment arm from the right leg resulted in a smaller positive contribution, increasing the net negative external moment and angular momentum trajectory (toward the left leg). During stair descent, there were several significant differences in the vertical and M/L moment arms and GRFs relative to level walking, including a larger initial vertical GRF
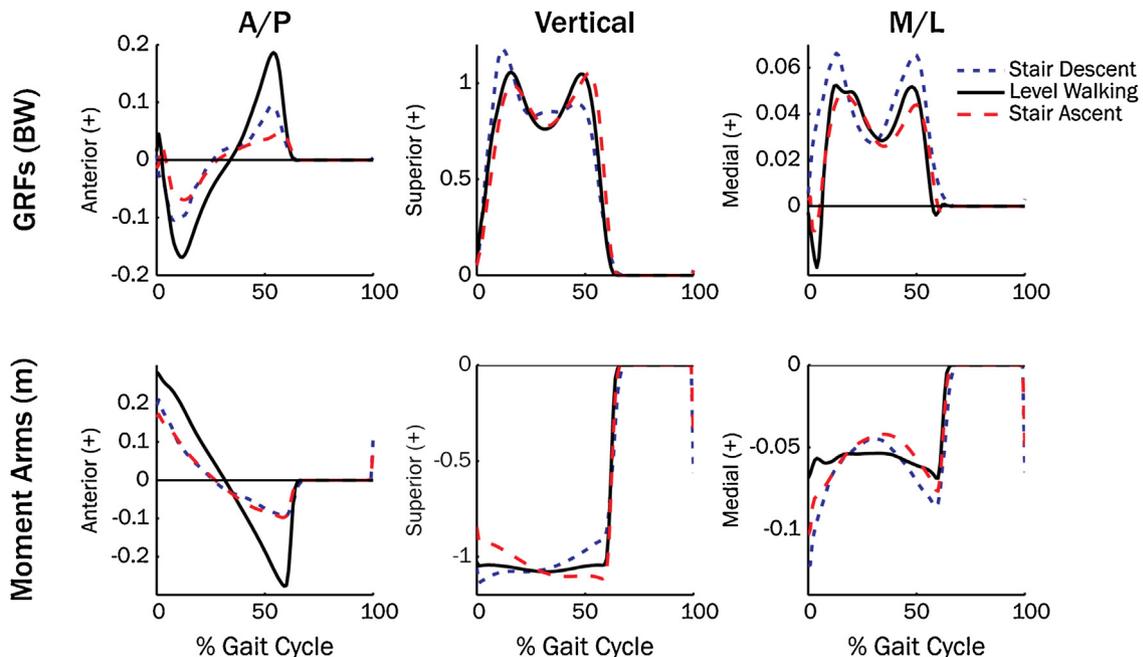


**Fig. 3.** Average ground reaction forces (GRFs) and external moment arms during each walking condition in the anterior/posterior (A/P), vertical and medial/lateral (M/L) directions.
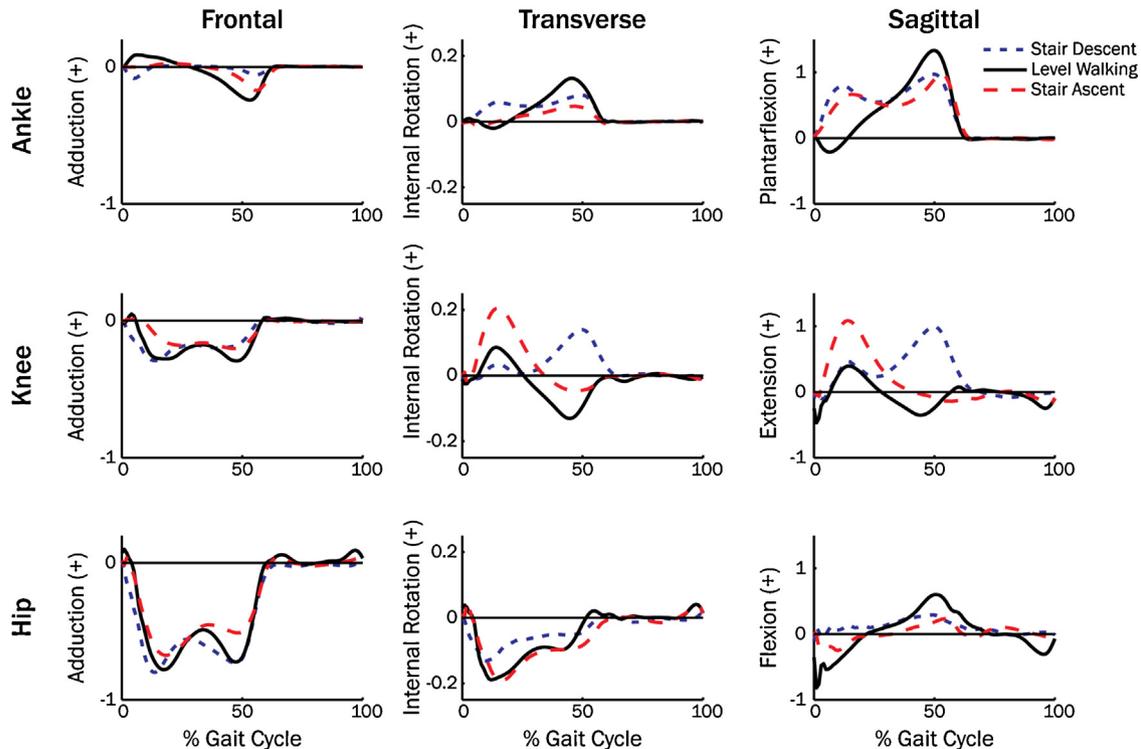
## Net Joint Moments (Nm/kg)



**Fig. 4.** Average three-dimensional net joint moments (Nm/kg) at the hip, knee and ankle for each walking condition.

peak, larger M/L GRF peaks, and a larger M/L moment arm. However, the larger contributions from the vertical GRFs opposed the contributions from larger M/L GRFs, resulting in a similar angular momentum trajectory to level walking.

The joint moment results helped explain how the angular momentum trajectory may be regulated in the frontal plane (Table 1 and Fig. 4). During stair ascent, the subjects had a reduced hip abduction moment in early and late stance, which is consistent with previous results [1] and reduced action of the gluteus medius. The gluteus medius is a major contributor to the frontal plane external moment, and acts to rotate the body toward the ipsilateral leg [25], maintaining the angular momentum close to zero. Reduced action of the gluteus medius may therefore result in greater deviation of angular momentum from zero, which was seen in the frontal plane for stair ascent. Previous work [1] has also reported altered hip abduction angle trajectories in stair ascent relative to level walking, and has hypothesized that changes in the hip abduction angle assist in rotating the pelvis to ensure the swing leg clears the intermediate stair. Thus, greater frontal plane angular momentum may be a necessary strategy to raise the body center-of-mass while avoiding a trip during stair ascent. The vastii muscles have also been shown to be major contributors to frontal-plane angular momentum as they act to rotate the body toward the contralateral leg [25]. The vastii are also large contributors to the knee extension moment, which was much larger in early stance during stair ascent. Therefore, greater contributions from the left leg vastii muscles may contribute to a greater positive (toward the right leg) angular momentum during left leg stance, whereas the right leg vastii muscles contribute to a greater negative (toward the left leg) angular momentum during right leg stance.

In the transverse plane, the range of angular momentum was greatest for level walking, and therefore did not support our hypothesis (Table 1 and Fig. 2). The A/P and M/L GRFs and external

moment arms contribute to the external moment about the COM in this plane (Fig. 1). Thus, the reduced range of angular momentum is the result of the reduced magnitudes of the A/P GRFs, consistent with previous work [16], and moment arms. The A/P moment arms are much smaller for stair conditions because A/P foot placement is largely constrained by the depth of the stair, and foot placement is not constrained during level walking. While the M/L moment arms were larger during stair walking relative to level walking, the A/P GRFs were much smaller, resulting in a smaller external moment relative to level walking, particularly from 0 to 10% and from 50 to 60% of the gait cycle. This smaller external moment reduced the time rate of change of angular momentum in the transverse plane at this time, reducing the overall range of angular momentum.

In the sagittal plane, our hypothesis was not supported. The range of angular momentum for both stair ascent and descent was smaller than level walking and the range of angular momentum for stair descent was smaller relative to stair ascent (Table 1 and Fig. 2). In the sagittal plane, the vertical and A/P GRFs and moment arms contribute to the net external moment (Fig. 1). The reduced ranges of angular momentum during stair walking are largely a result of the reduced magnitude of the A/P GRFs and moment arms. The reductions in the A/P GRFs and moment arms reduced the magnitude of the net external moment about the COM during stair walking, resulting in smaller overall ranges of angular momentum.

The sagittal plane angular momentum results likely have the greatest application to fall prevention during stair walking, as trips during ascent and slips during descent will have the greatest effect on external forces in the anterior/posterior direction, potentially evoking forward and backward falls. During stair ascent, greater joint ranges of motion (e.g., [1]) are thought to provide toe clearance to avoid tripping during swing [26] as the leg is lifting to the next stair. In our results, the time of maximum positive angular

momentum (30% and 82% of the gait cycle) during stair ascent occurs near the time of mid-swing (34% for the right leg and 85% for the left leg). The toe catching on the stair would result in a negative (forward) external moment about the body COM. Thus, it is advantageous to have a large positive angular momentum at this point in time to counteract a potential negative external moment from a trip and a potential forward fall toward the stairs.

During stair descent, falling backward (positive angular momentum toward the stairs) is far preferable to falling forward (negative angular momentum down the stairs), which can result in serious injury. Our angular momentum results support a strategy to avoid negative angular momentum during stair descent. Early in the gait cycle, the angular momentum rapidly transitions from negative to positive. This positive slope of angular momentum results from the large vertical GRF of the leading limb in early stance. After the peak positive angular momentum, the trajectory is gradually reduced, and then rapidly transitions again at 50% of the gait cycle at right heel strike. Angular momentum is regulated more tightly during descent in that the range is smaller and closer to zero. The risk of falling during descent is higher than during ascent, and the risk of serious injury from a fall is greater during descent relative to ascent [27]. Thus, this tighter regulation of angular momentum during descent may be a strategy to reduce fall risk; similar to what has previously been shown during decline walking [8].

The differences in the hip extension moment in early stance and plantarflexion moment in late stance may partially explain the differences in sagittal angular momentum during stair walking. In early stance, both the gluteus maximus and biarticular hamstrings contribute to positive (backward) angular momentum [15] and the hip extension moment. A reduced hip extension moment in stair walking suggests reduced force output from the hip extensors, and therefore a reduced contribution to positive angular momentum in early stance (0–30% of the gait cycle). Similarly, in late stance (30–50% of the gait cycle), the soleus contributes to the ankle plantarflexor moment and negative angular momentum [15]. A reduced ankle plantarflexor moment was shown for the stair conditions and may also explain the overall reduced range of angular momentum (Fig. 2).

## 5. Conclusions

This study investigated whole-body angular momentum in healthy subjects while walking on stairs. The results help explain how healthy individuals maintain dynamic balance during stair walking and also provide a baseline for comparison with balance-impaired populations. Our hypothesis that the range of angular momentum would be greater during stair walking relative to level walking was only partially supported in the frontal plane, with only stair ascent showing differences. In the transverse and sagittal planes, the range of angular momentum was reduced in stair walking relative to level walking. Differences were seen in the range of angular momentum, ground reaction forces, external moment arms and joint moments between walking conditions, suggesting that angular momentum is regulated differently in stair ascent and descent relative to level walking in order to maintain dynamic balance. An important area of future work is to assess the thresholds in the range of angular momentum in specific movement tasks that will lead to a fall. Knowing these thresholds will help identify individuals who are susceptible to falling and prescribe appropriate locomotor interventions to address fall risk.

## Conflict of interest statement

The authors have no conflict of interest in the preparation or publication of this work.

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