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Whole-body angular momentum in incline and decline walking ☆

Anne K. Silverman [a,*], Jason M. Wilken [b], Emily H. Sinitski [c], Richard R. Neptune [c]

[a] Department of Mechanical Engineering, Colorado School of Mines, Golden, CO 80401, USA
[b] Center for the Intrepid, Department of Orthopedics and Rehabilitation, Brooke Army Medical Center, Ft. Sam Houston, TX 78234, USA
[c] Department of Mechanical Engineering, The University of Texas at Austin, Austin, TX 78712, USA

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ABSTRACT

Angular momentum is highly regulated over the gait cycle and is important for maintaining dynamic stability and control of movement. However, little is known regarding how angular momentum is regulated on irregular surfaces, such as slopes, when the risk of falling is higher. This study examined the three-dimensional whole-body angular momentum patterns of 30 healthy subjects walking over a range of incline and decline angles. The range of angular momentum was either similar or reduced on decline surfaces and increased on incline surfaces relative to level ground, with the greatest differences occurring in the frontal and sagittal planes. These results suggest that angular momentum is more tightly controlled during decline walking when the risk of falling is greater. In the frontal plane, the range of angular momentum was strongly correlated with the peak hip and knee abduction moments in early stance. In the transverse plane, the strongest correlation occurred with the knee external rotation peak in late stance. In the sagittal plane, all external moment peaks were correlated with the range of angular momentum. The peak ankle plantarflexion, knee flexion and hip extension moments were also strongly correlated with the sagittal-plane angular momentum. These results highlight how able-bodied subjects control angular momentum differently on sloped surfaces relative to level walking and provide a baseline for comparison with pathological populations that are more susceptible to falling.

1. Introduction

Walking on sloped surfaces is an important aspect of daily mobility, and the risk of slip-related falls is greater on slopes compared to level ground (Redfern et al., 2001). The increased risk is due to the shear ground reaction force (GRF) generated during sloped walking being greater than in level walking. To prevent a fall, the shear GRF must not exceed the frictional force developed between the foot and ground (Redfern et al., 2001). GRFs are largely developed by muscle forces (e.g., Anderson and Pandy, 2003; Neptune et al., 2004). As a result, many impaired populations, such as the elderly or individuals with amputation, have a higher risk of falling relative to the able-bodied population (e.g., Kannus et al., 1999; Miller et al., 2001). This increased risk is partly due to the inability of these populations to rapidly modulate their GRFs in response to changing GRF requirements.

An important measure of GRF modulation is whole-body angular momentum, which is highly regulated in level walking (Herr and Popovic, 2008). Whole-body angular momentum must be quickly modulated to recover from trips and prevent falls (Pijnappels et al., 2004; Pijnappels et al., 2005a) and is affected by pathology, such as lower-limb amputations (Silverman and Neptune, 2011). The range of angular momentum may be of particular importance, as large angular momentum deviations from zero may result in a greater control challenge to maintain dynamic stability. However, little is known about how whole-body angular momentum is affected by walking on a slope. The time rate of change of angular momentum equals the net external moment about the body's center-of-mass (COM). Two GRF components from each foot produce the net external moment in each plane. For example, in the sagittal plane, the moments produced by the anterior/posterior (A/P) and vertical GRFs sum to produce the net external moment (Fig. 1). Therefore, the changes in foot placement, joint kinematics, body COM position and GRFs observed during incline and decline walking (Lay et al., 2006; Leroux et al., 2002; McIntosh et al., 2006; Redfern and DiPasquale, 1997) will result in differences in the external moment, and therefore the whole-body angular momentum relative to level walking. Changes in net joint moments, which are primarily the result of muscle forces (e.g., Pandy et al., 2010), have also been observed in sloped walking (Lay et al., 2006; McIntosh et al., 2006). Thus, understanding how the joint moments are related to angular momentum will provide insight into how angular momentum is controlled on sloped surfaces. Quantifying differences in angular momentum and identifying the

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* Corresponding author. Tel.: +303 384 2162; fax: 303 273 3602.
E-mail address: asilverm@mines.edu (A.K. Silverman).

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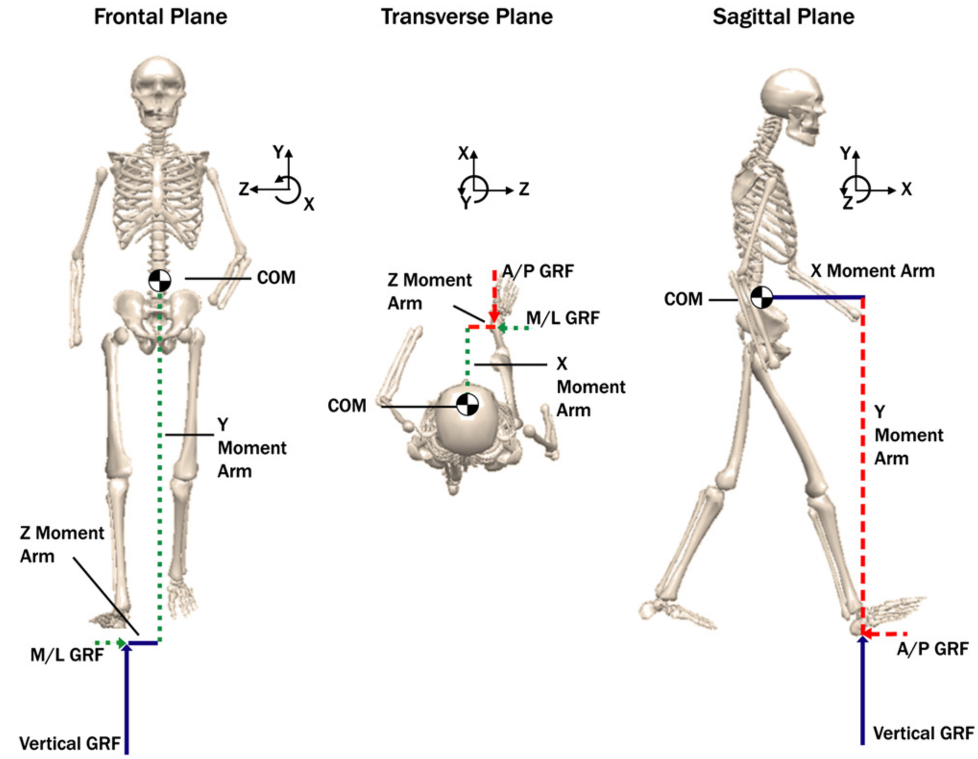

Fig. 1. The inverse dynamics model as viewed from the frontal, transverse and sagittal planes. The ground reaction forces (GRFs) contribute to the external moment about the center of mass (COM). Only the contributions from the right leg are shown for clarity. Both legs contribute to the external moment in all three planes.

biomechanical measures most strongly correlated with those differences will help establish a basis for analyzing pathological populations who are susceptible to falling during sloped walking.

Therefore, the purpose of this study was to analyze three-dimensional (3D) whole-body angular momentum in healthy subjects walking on a range of sloped surfaces. We tested the hypothesis that the range of angular momentum is different in incline and decline walking relative to level walking, and that these differences are directly related to changes in the external moments and joint moments as the incline and decline slope angle changes.

2. Methods

Thirty healthy subjects (13 male, 17 female; 21.8 ± 4.2 years; 73.3 ± 14.8 kg; 1.7 ± 0.1 m) underwent a single gait assessment while walking up and down a 16-ft platform at each of four slopes (0, 5°, 10° and 15°) presented in random order. Slopes of 5°, 10° and 15° were selected to allow comparison to previously published work (e.g., Leroux et al., 2002; McIntosh et al., 2006; Redfern et al., 2001; Redfern and DiPasquale, 1997) and effectively characterize the effect of changing incline angle on angular momentum. In addition, the 5° condition was selected to match the maximum incline allowed under the 1994 Americans with Disabilities Act (ADA) standards for accessible design (28 CFR Part 36:518–521). This study was approved by the Institutional Review Board at Brooke Army Medical Center, Ft. Sam Houston, TX, and all participants provided written, informed consent prior to participation.

A 26-camera motion capture system operating at 120 Hz and a six degree-of-freedom marker set with 55 markers were used to quantify full-body motion (Wilken et al. (in press)). GRFs were measured at 1200 Hz using two force plates embedded in the platform. An automated auditory cue based on trunk marker velocity guided subjects to walk at a pre-determined, scaled walking speed based on the Froude number to reduce kinematic and kinetic variability between subjects (McAndrew et al., 2010; Wilken et al. (in press)). The auditory cue ensured that walking speed remained constant and that gait pattern changes were due to the slope condition only.

Kinematic and kinetic data were low-pass filtered using fourth-order Butterworth filters with cut-off frequencies of 6 Hz and 50 Hz, respectively. A 13-segment model including the head, torso, pelvis, upper arms, lower arms, thighs, shanks and feet was used to determine the COM location and velocity of each segment. Segment masses were determined as a percentage of total body mass (Dempster and Aitkens, 1995). Segment inertial properties were determined from marker placement and by assuming segment geometry. Whole-body angular momentum about the COM was determined as

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

where $\vec{r}_i^{\,COM}$, $\vec{v}_i^{\,COM}$ and $\vec{\omega}_i$ are the position, velocity and angular velocity vectors of the ith segment's COM, $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the whole-body COM, m_i and I_i are the mass and moment of inertia of the ith segment and n is the number of segments. Angular momentum was normalized by body mass (kg), walking speed (m/s) and body height (m) and expressed as a percentage of the left leg gait cycle.

The range of each 3D angular momentum component, defined as the peak-to-peak value, was compared across slope conditions. External moment components were calculated for each direction as the product of the GRF and the COM to center-of-pressure distance. The magnitude of the peak external moments and joint moments, averaged between the right and left legs, were also compared across slope conditions. Five trials were used for each subject during each condition. External moment and joint moment peaks were compared during early stance (first peak) and late stance (second peak). Normally distributed data were compared using a one-factor (condition), repeated measures ANOVA ($\alpha = 0.05$). Significant main effects were explored using paired t-tests with a Bonferroni adjustment, in which the significance level was adjusted by dividing by the number of pairwise comparisons ($\alpha^* = \alpha/n$, where n is the number of comparisons and $\alpha = 0.05$). Incline and decline conditions were compared to level walking. Non-normally distributed data were compared across conditions using Friedman's test and pairwise comparisons were performed using Wilcoxen Signed Rank tests, also with a Bonferroni adjustment.

To identify quantities that were correlated with differences in angular momentum, correlation analyses were performed between the range of angular momentum and the peak external and joint moment magnitudes that were significantly different across conditions. Correlations for timing of the peaks were not considered. A Pearson correlation was performed on normally distributed data and a Spearman correlation was performed on non-normally distributed data.

3. Results

All significant main effects and pairwise differences for the sloped conditions relative to level walking are reported in Table 1.

Table 1
The *p*-values from the statistical analysis are shown, including significant main effects as well as the pairwise comparisons. Pairwise comparisons were performed between the magnitude at an individual sloped condition and level walking only. If a sloped condition had a magnitude significantly less than level walking, then the *p*-value is in bold. Otherwise, the magnitude was greater than level walking. If no *p*-value is given, then the comparison was not significant ($p > 0.05$). A value of 0.000 indicates that the *p*-value was less than 0.0005.

		Main effect	Decline conditions			Incline conditions		
			15°	10°	5°	5°	10°	15°
Frontal								
Angular momentum range		0.000	-	-	-	0.000	0.000	0.000
External moments	Vertical (Y) moment arm and M/L GRF, 1st peak	0.000	0.000	0.000	0.000	**0.004**	**0.002**	-
	Vertical (Y) moment arm and M/L GRF, 2nd peak	0.000	-	-	-	**0.004**	**0.001**	**0.000**
	M/L (Z) moment arm and Vertical GRF, 1st peak	0.000	0.000	0.000	0.000	-	-	-
	M/L (Z) moment arm and Vertical GRF, 2nd peak	0.000	**0.008**	-	-	-	-	-
Joint moments	Ankle adduction peak	0.000	0.000	0.000	0.000	**0.000**	**0.000**	**0.000**
	Ankle abduction peak	0.000	**0.000**	**0.000**	**0.000**	0.000	0.000	0.000
	Knee abduction, 1st peak	0.000	**0.004**	-	-	**0.000**	**0.000**	**0.000**
	Knee abduction, 2nd peak	0.000	**0.000**	**0.000**	**0.001**	**0.002**	**0.000**	**0.000**
	Hip abduction, 1st peak	0.000	0.000	0.000	0.000	**0.000**	**0.000**	**0.000**
	Hip abduction, 2nd peak	0.000	**0.002**	**0.001**	-	**0.003**	**0.000**	**0.000**
Transverse								
Angular momentum range		0.000	**0.000**	**0.001**	**0.000**	-	-	0.003
External moments	A/P (X) moment arm and M/L GRF, 1st peak	0.000	-	-	0.007	**0.000**	**0.000**	**0.000**
	A/P (X) moment arm and M/L GRF, 2nd peak	0.000	-	-	0.002	**0.000**	**0.000**	**0.000**
	M/L (Z) moment arm and A/P GRF, 1st peak	0.000	0.000	0.000	0.000	**0.014**	**0.000**	**0.002**
	M/L (Z) moment arm and A/P GRF, 2nd peak	0.000	0.046	0.003	0.002	**0.003**	**0.000**	**0.001**
Joint moments	Ankle internal rotation peak	0.003	**0.018**	**0.002**	**0.000**	-	-	**0.000**
	Knee internal rotation peak	0.000	0.000	0.000	0.000	0.000	0.000	0.000
	Knee external rotation peak	0.000	**0.000**	**0.000**	**0.000**	-	-	-
	Hip external rotation, 1st peak	0.021	-	-	-	-	-	-
	Hip external rotation, 2nd peak	0.000	**0.000**	**0.000**	**0.000**	-	0.032	0.007
Sagittal								
Angular momentum range		0.000	**0.001**	**0.000**	**0.000**	0.000	0.000	0.000
External moments	A/P (X) moment arm and vertical GRF, 1st peak	0.000	-	-	-	-	**0.018**	**0.013**
	A/P (X) Moment arm and vertical GRF, 2nd peak	0.000	**0.000**	**0.000**	**0.000**	0.000	0.000	0.001
	Vertical (Y) moment arm and A/P GRF, 1st peak	0.000	0.000	0.000	0.000	**0.000**	**0.000**	**0.000**
	Vertical (Y) moment arm and A/P GRF, 2nd peak	0.000	**0.002**	-	-	**0.001**	**0.000**	**0.000**
Joint moments	Ankle plantarflexion peak	0.000	**0.000**	**0.000**	**0.000**	0.000	0.000	0.000
	Knee extension peak	0.000	0.000	0.000	0.000	0.004	0.000	0.000
	Knee flexion peak	0.000	**0.000**	**0.000**	**0.000**	0.000	0.000	0.000
	Hip extension peak	0.000	-	-	**0.001**	0.000	0.000	0.000
	Hip flexion peak	0.000	-	-	**0.034**	-	-	**0.001**

3.1. Frontal plane

The range of frontal–plane angular momentum was greater for all incline conditions (Table 1; Fig. 2, top row). There were significant main effects for both external moments during early and late stance (Table 1; Fig. 3, top row). The peak external moment resulting from the medial/lateral (M/L) GRF (Fig. 3, upper-left plot) was greater during early stance for all decline conditions and was smaller for 5° and 10° incline. The peak external moment resulting from the vertical GRF (Fig. 3, upper-middle plot) was greater for all decline conditions during early stance. Frontal-plane ankle, knee and hip joint kinetic quantities also had significant differences (Table 1; Fig. 4, top row).

The range of angular momentum was significantly correlated with several of the kinetic quantities ($n=210$, Table 2). Both peaks of the external moment resulting from the vertical GRF were significantly correlated. All peak joint moments, except for the peak ankle adduction moment, were also significantly correlated with the range of angular momentum. The strongest joint moment correlations were with the knee and hip abduction peaks in early stance.

3.2. Transverse plane

The range of transverse–plane angular momentum was smaller for all decline conditions and greater for 15° incline only (Table 1; Fig. 2, middle row). Both peaks of the external moment component resulting from the M/L GRF were smaller for incline conditions and greater for 5° decline (Table 1; Fig. 3, middle row). Both peaks of the A/P GRF external moment component were greater for all decline conditions and smaller for all incline conditions. Significant joint moment differences were again found at the ankle, knee and hip (Table 1; Fig. 4, middle row). The range of the transverse angular momentum was correlated with the first peak of the external moment arising from the M/L GRF ($n=210$, Table 2) in early stance. The range was also correlated with both knee rotation moment peaks and the second hip rotation moment peak. The external rotation moment at the knee in late stance had the strongest correlation with the range of angular momentum.

3.3. Sagittal plane

In the sagittal plane, the range of angular momentum was smaller for decline walking and greater for incline walking. All peaks of the external moment components had significant main effects (Table 1; Fig. 3, bottom row). The peak of the vertical GRF external moment component was smaller for the decline conditions and greater for the incline conditions in late stance. Both peaks of the external moment resulting from the A/P GRF (Fig. 3, bottom-middle plot) had smaller magnitudes for all incline conditions. Significant differences in the peak joint moments

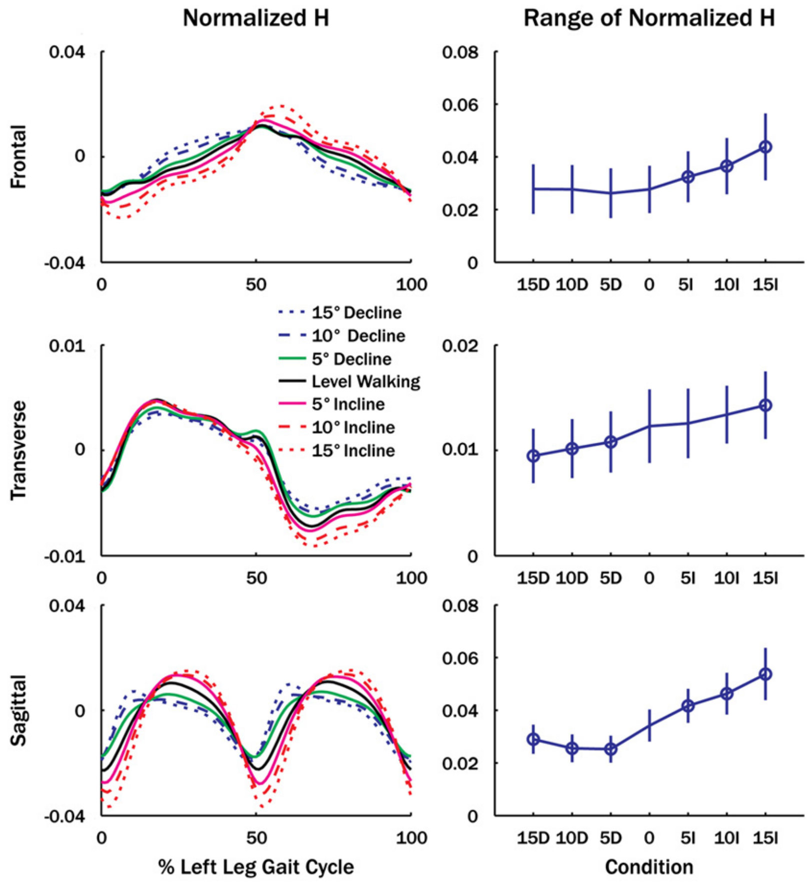

Fig. 2. The mean, normalized, 3D whole-body angular momentum over the left leg gait cycle is shown for all slope conditions on the left. Angular momentum was normalized by body mass (kg), body height (m) and walking speed (m/s). The mean (σ) range of the angular momentum (peak-to-peak value) for each slope condition is shown on the right. Conditions that were significantly different from level walking (0 condition) are indicated with an open circle.

were found at the ankle, knee and hip (Table 1; Fig. 4, bottom row).

The range of sagittal-plane angular momentum was correlated with all external moment peaks and all joint moment peaks except for the peak hip flexion moment ($n=210$, Table 2). The joint moments that were the most strongly correlated with the range of angular momentum were the peak ankle plantarflexion, knee flexion and hip extension moments.

4. Discussion

There were significant differences in angular momentum, external moment components and joint moments in all three planes. The external and joint moment results aided in interpreting the differences in angular momentum across slope condition.

In the frontal plane, the greater range of angular momentum for all incline conditions (Fig. 2, top row) resulted from a greater positive rate of change of angular momentum for the first half of the gait cycle and a more negative rate of change for the second half of the gait cycle. The gluteus medius is a large contributor to both the hip abduction moment and to the frontal-plane external moment (Neptune et al., 2011). Thus, reducing the hip abduction moment for incline conditions may increase the range of angular momentum in the frontal plane, consistent with our correlation results. Conversely, increasing the hip abduction moment may be an effective strategy to reduce the range of angular momentum to prevent falling.

In the transverse plane, the range of angular momentum was correlated with the first external moment peak resulting from the M/L GRF. At approximately 50% of the left gait cycle, the positive contribution from the right leg contributed to the net negative external moment, resulting in the altered range of the angular momentum for the different slope conditions.

In the sagittal plane, the range of angular momentum was smaller for the decline conditions and greater for the incline conditions compared to level walking. For the decline conditions, the net external moment was reduced early in the gait cycle because of a greater braking force and vertical (Y) moment arm from the leading (left) leg, which is consistent with previous studies showing a greater posterior shear force during decline walking (Lay et al., 2006; McIntosh et al., 2006; Redfern and DiPasquale, 1997). Conversely, for the incline conditions, the external moment was more positive early in the gait cycle from a reduced braking force and vertical (Y) moment arm, which is consistent with previous work reporting a smaller posterior shear force (Lay et al., 2006; McIntosh et al., 2006).

The sagittal joint moment results were consistent with previous studies of incline walking (Lay et al., 2006; McIntosh et al., 2006) and provided insight into potential mechanisms for the observed angular momentum changes. For example, a recent simulation study quantified individual muscle contributions to sagittal-plane angular momentum during level walking and found that the hip extensors contributed positively to the external moment in early stance (Neptune and McGowan, 2011). These results support the greater (smaller) range of angular momentum in the sagittal plane for incline (decline) conditions, which was strongly correlated with increases (decreases) in the hip extension moment. In addition, an increased magnitude of muscle activity from the gluteus maximus, biceps femoris and semimembranosus shown during incline walking

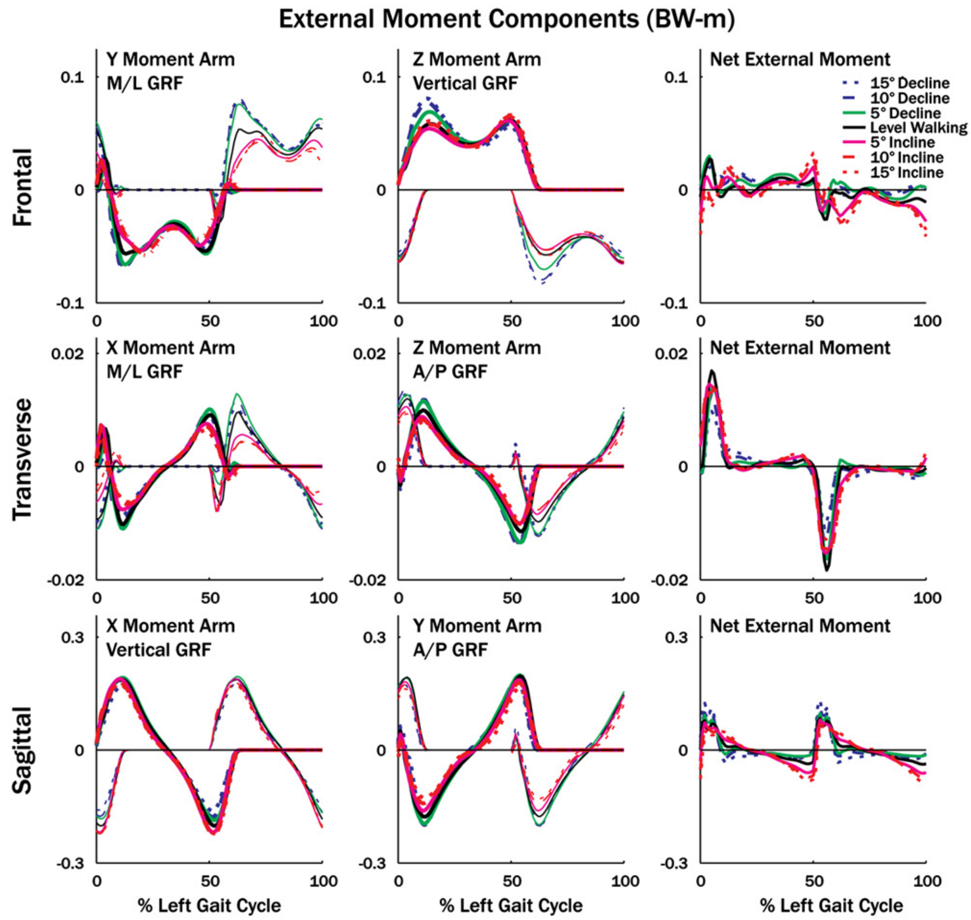

Fig. 3. The components that contribute to the external moment on the body center-of-mass in each plane are shown over the left leg gait cycle. There are four components for each plane (i.e., two components from each leg, see Fig. 1). Contributions from both the left and right legs are shown. The thicker lines indicate the left leg. The peak values of these components were averaged across legs and compared in the statistical analysis. The net external moment, which is the sum of all four components and equals the time rate of change of whole-body angular momentum, is shown in the right column.

(Lay et al., 2007) corresponds to the observed increases in the hip extension moment and range of sagittal-plane angular momentum.

At approximately 50% of the gait cycle during incline (decline) walking there was a greater (smaller) negative external moment from the trailing leg vertical GRF (Fig. 3 bottom left plot), which resulted from a greater (smaller) vertical GRF in late stance and was consistent with the work of others (McIntosh et al., 2006). Thus, there was a greater (smaller) negative rate of change of sagittal-plane angular momentum for incline (decline) walking relative to level walking. In late stance, the gastrocnemius contributes positively to the external moment whereas the soleus contributes negatively to a greater extent (Neptune and McGowan, 2011). Thus, a greater (smaller) ankle plantarflexion moment during incline (decline) walking likely contributes to a greater (smaller) negative external moment, and therefore the greater (smaller) range of angular momentum. In addition, increased muscle activity from the gastrocnemius and soleus during incline walking (Lay et al., 2007) is consistent with the ankle moment and angular momentum results.

The results showed that the range of whole-body angular momentum is, in general, greater for walking up an incline and similar or smaller for walking down a decline in all three planes. We examined the range of angular momentum because it captured the deviation of the angular momentum from zero and quantified the overall differences in the angular momentum trajectory over the gait cycle. It may be interesting in future studies to include detailed comparisons of the shape of the momentum patterns over the gait cycle.

The results also showed that the changes in the range of angular momentum were directly related to changes in the external moment and individual joint moments. We chose to examine peak values of the external joint moments as there were clear, systematic changes in the peaks across conditions (Figs. 2 and 3). Small changes in the GRFs, resulting from muscle forces, can significantly alter the angular momentum trajectory. Previously, angular momentum has been shown to change with walking speed and be different between amputees and non-amputees (Bennett et al., 2010; Silverman and Neptune, 2011), and these changes have also been correlated with specific kinetic measures (Silverman and Neptune, 2011).

Changes in step length can cause changes in the GRF as well as the external moment arm. Both of these quantities will largely affect the external moment and therefore whole-body angular momentum. Thus, we analyzed step lengths to further inform the angular momentum and external moment results (Table 3). The step length was increased for the 5° incline condition and decreased for all decline conditions relative to level walking. This result highlights how individuals change their step length as a way of controlling their angular momentum for different slope conditions.

The regulation of angular momentum has previously been shown to be important in fall prevention (Pijnappels et al., 2004; Pijnappels et al., 2005b). The likelihood of falling on a decline surface is greater than on level or an incline surface because of differences in the magnitude and peak timing of the shear forces developed on the bottom of the foot (Redfern et al., 2001). Thus, it

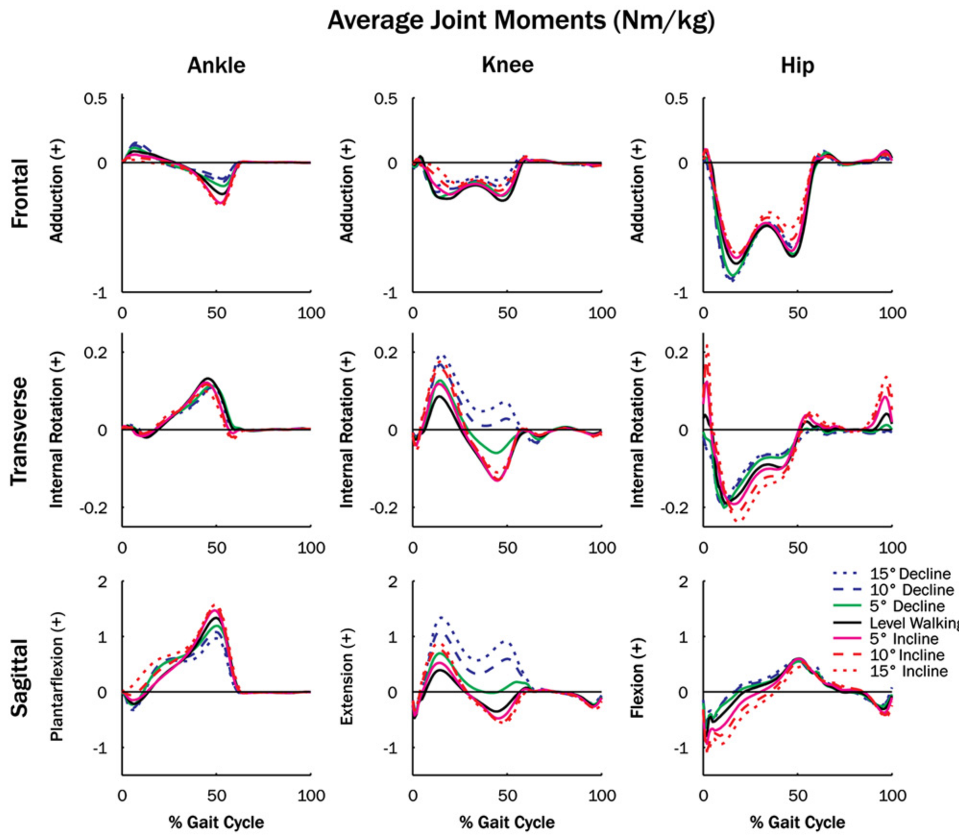

Fig. 4. Average 3D joint moments across slope conditions. The joint moments were normalized by body mass. Peak joint moments were compared in the statistical analysis.

Table 2
Significant correlation coefficients and associated *p*-values for the range of angular momentum with the magnitude of peak kinetic quantities ($n=210$). If no coefficient is shown, then the variable of interest did not have a significant main effect across conditions or was not significantly correlated with the range of angular momentum.

		r	p
Frontal			
External moment	Vertical (Y) moment arm and M/L GRF, 1st peak	-	-
	Vertical (Y) moment arm and M/L GRF, 2nd peak	-	-
	M/L (Z) moment arm and vertical GRF, 1st peak	0.188	0.007
	M/L (Z) moment arm and vertical GRF, 2nd peak	0.524	0.000
Joint moments	Ankle adduction peak	-	-
	Ankle abduction peak	0.186	0.007
	Knee abduction, 1st peak	−0.521	0.000
	Knee abduction, 2nd peak	−0.252	0.000
	Hip abduction, 1st peak	−0.533	0.000
	Hip abduction, 2nd peak	−0.361	0.000
Transverse			
External moment	A/P (X) moment arm and M/L GRF, 1st peak	−0.227	0.001
	A/P (X) moment arm and M/L GRF, 2nd peak	-	-
	M/L (Z) moment arm and A/P GRF, 1st peak	-	-
	M/L (Z) moment arm and A/P GRF, 2nd peak	-	-
Joint moments	Ankle internal rotation peak	-	-
	Knee internal rotation peak	−0.221	0.001
	Knee external rotation peak	0.461	0.000
	Hip external rotation, 1st peak	-	-
	Hip external rotation, 2nd peak	0.258	0.000
Sagittal			
External moment	A/P (X) moment arm and Vertical GRF, 1st peak	−0.243	0.000
	A/P (X) moment arm and Vertical GRF, 2nd peak	0.511	0.000
	Vertical (Y) moment arm and A/P GRF, 1st peak	−0.730	0.000
	Vertical (Y) moment arm and A/P GRF, 2nd peak	−0.366	0.000
Joint moments	Ankle plantarflexion peak	0.694	0.000
	Knee extension peak	−0.284	0.000
	Knee flexion peak	0.656	0.000
	Hip extension peak	0.620	0.000
	Hip flexion peak	-	-

Table 3
Average step length values and standard deviation (m) across subjects for all incline and decline conditions. Conditions that were significantly different from level walking are indicated with '*'.

	Decline conditions			0°	Incline conditions		
	15°	10°	5°		5°	10°	15°
Average step length (m)	0.629*	0.620*	0.592*	0.670	0.707*	0.683	0.640*
Standard deviation (m)	0.031	0.046	0.053	0.046	0.044	0.063	0.053

appears that actively controlling angular momentum to a greater extent (i.e., reducing the range of angular momentum) while walking down a decline is a protective mechanism to prevent slipping or falling. Such tight control may not be necessary for walking up an incline, when the risk of falling is not as high. Thus, these results support the increased risk of falling shown on decline surfaces and have important implications for those who have a greater risk of falling. Previous studies have shown that fallers have reduced muscle strength (Pijnappels et al., 2008) and reduced peak joint moments during walking and trip recovery (Pijnappels et al., 2005b; Simoneau and Krebs, 2000). Thus, these populations may have difficulty controlling their angular momentum because of muscle weakness. Understanding how muscles contribute to angular momentum and how the angular momentum changes under different conditions can provide insight into why certain populations are at a higher risk for falling and how to design effective rehabilitation therapies that target specific muscle groups.

5. Conclusion

This study identified significant differences in the external and joint moments that were directly related to changes in whole-body angular momentum while walking on sloped surfaces. The results suggest that able-bodied subjects actively regulate angular momentum differently on sloped surfaces relative to level walking. The greatest differences occurred in the frontal and sagittal planes. In general, the range of angular momentum was larger for incline walking compared to level walking, and similar or smaller for decline walking compared to level walking. These results suggest that individuals more tightly control angular momentum while walking down a slope to help prevent a slip or fall.

Conflict of interest statement

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

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