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# Differences in balance control between healthy younger and older adults during steady-state walking

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ABSTRACT

Each year approximately one third of older adults fall and experience extensive musculoskeletal injuries and functional disabilities. An important element in maintaining dynamic balance is the regulation of whole-body angular momentum, which is achieved by proper foot placement with respect to the body center-of-mass as well as generation of appropriate ground reaction forces. Analyzing these quantities in younger and older adults may provide insight into differences in their underlying mechanics for maintaining dynamic balance. This study examined three-dimensional whole-body angular momentum in 13 healthy older (71.8 ± 8.3 years) and 9 younger (23.2 ± 2.8 years) adults walking at their self-selected and fastest-comfortable speeds. The older adults had a significantly higher range of frontal-plane angular momentum compared to the younger adults at both speeds, suggesting poorer mediolateral balance control. This difference was related to the older adults having a wider foot placement with respect to the body center-of-mass, which when combined with the vertical ground reaction force, created a higher destabilizing external moment during single-limb stance that acts to rotate the body towards the contralateral swing leg. To counteract this destabilizing moment, the older adults generated a higher hip abduction moment. There were no differences in the range of sagittal- and transverse-plane angular momentum between age groups at either speed. These results suggest that control of dynamic balance in the frontal-plane is more challenging than in the sagittal-plane for older adults and highlight the importance of proper weight transfer mechanisms and hip abductor force production for maintaining mediolateral balance during walking.

## 1. Introduction

Each year approximately one third of older adults fall, leading to extensive musculoskeletal injuries and functional disabilities (Centers for Disease Control and Prevention CDC, 2008). A major concern and cause of death in the elderly is hip fracture (Deprey, 2009; Kung et al., 2008), which is associated with the lack of balance control in the mediolateral direction (Greenspan et al., 1998; Nevitt and Cummings, 1993). Previous work has shown that dynamic balance may be maintained passively in the sagittal-plane, but active control is needed to stabilize lateral motion (Bauby and Kuo, 2000). Foot placement is a common strategy for controlling mediolateral balance (e.g., Hof et al., 2010). Previous studies have suggested that wider steps are used to increase lateral stability (Bauby and Kuo, 2000; Dean et al., 2007; Donelan et al., 2004; Gabell and Nayak, 1984) and some have observed that elderly fallers take narrower steps (Guimaraes and Isaacs, 1980). However, others have associated wider steps with increased step width variability and increased instability (McAndrew Young and Dingwell, 2012) while some have observed higher rate of falls in individuals with wider steps (Gehlsen and Whaley, 1990; Maki, 1997; Moe-Nilssen and Helbostad, 2005; Nelson et al., 1999). In a comprehensive review, Bruijn and van Dieen (2018) note that older adults are reported to walk with wider steps perhaps as an adaptation strategy to the larger and faster mediolateral center-of-mass (*M/L CoM*) movements compared to younger adults. However, they point out the unclear cause and effect relationship between changes in step width and *M/L CoM* sway, noting that wider steps also contribute to the increase of *M/L CoM* sway. Thus, the influence of foot placement and step width on the control of dynamic balance in older adults is still unclear.

Whole-body angular momentum (*H*), a mechanics-based measure of balance control, has been used to investigate dynamic balance in various populations (Neptune and Vistamehr, 2018) including in older adults

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during step initiation (Begue et al., 2019) and perturbed walking (Pijnappels et al., 2005). In order to maintain dynamic balance during walking, whole-body angular momentum needs to be regulated. Although segmental angular momenta can have high values, the net sum of the angular momentum about the body's *CoM* is kept low (e.g., Herr and Popovic, 2008), which is also manifested in a small peak-to-peak range of *H*. The regulation of whole-body angular momentum can be quantified by analyzing the time rate of change of angular momentum about the body's *CoM*, which is equivalent to the net external moment (i.e., the cross-product of the ground reaction force (*GRF*) vector and moment arm vector from the body *CoM* to the center-of-pressure (*CoP*)). Identifying the changes in *GRFs* and moment arms can provide insight into the underlying mechanisms for maintaining dynamic balance (e.g., Silverman and Neptune, 2011). During perturbed walking, older fallers have shown insufficient reduction of angular momentum, improper placement of the recovery limb and inadequate joint moment generation from the support limb (Pijnappels et al., 2005). However, no study has analyzed the external moment components (i.e., *GRFs* and moment arms) during steady-state walking to identify potential mechanisms underlying dynamic balance deficits in older adults. Steady-state walking has been reported as a key element in maintaining health and physiological activity and promotes independence and social well-being in all adults, especially in the elderly (e.g., Morris and Hardman, 1997). Normative data has reported that healthy older adults on average take 2000–9000 steps per day (Tudor-Locke et al., 2011). Thus, it is important to understand the mechanisms of balance control during steady-state walking in older adults.

The purpose of this study was to assess dynamic balance in healthy younger and older adults during steady-state walking by comparing the angular momentum profiles and external moment components (i.e., *GRFs* and moment arms) in the frontal, sagittal, and transverse planes. To help interpret any observed differences, net intersegmental joint moments were analyzed. We hypothesized that the regulation of whole-body angular momentum would be more challenging in older adults (manifested by a higher peak-to-peak range of *H*). In addition, we expected that older adults would use different mechanisms for regulating *H* due to age-related differences in *GRF* generation and foot placement relative to the body *CoM*.

## 2. Methods

### 2.1. Participants

Nine young and 13 older adults were enrolled (Table 1). The participants did not report any neurological or orthopedic impairment that affected walking. All participants reported to be physically active with at least once a week engaging in physical exercise or yard work. The study protocol and consent form were approved by an Institutional Review Board and all participants provided written, informed consent prior to study participation.

### 2.2. Experimental data collection

Participants walked overground on a 10-meter walkway at their self-selected (SS) and fastest-comfortable (FC) speeds (Table 1). At least 3 trials were collected for each walking speed. A modified Helen Hayes full-body marker set was used to define 15 body segments (head, trunk, pelvis, and each upper arm, lower arm, hand, thigh, shank, and foot). A 12-camera motion capture system (VICON, Los Angeles, USA) was used to record three-dimensional body-segment kinematics at 100 Hz. During the SS walking condition, three-dimensional *GRF* data were collected at 1200 Hz using four overground force plates (AMTI, Inc., Watertown, MA, USA). Participants walked as many trials as needed with their natural gait pattern across the room (i.e., they did not modify their spatiotemporal parameters or target stepping on the force plates) until five good force plate strikes per leg were recorded. Due to the challenge of successfully acquiring overground force plate measurements and fatigue-related concerns in older adults, the protocol only included collection of *GRF* data during the SS walking condition.

**Table 1**
Participant characteristics: gender, age, leg length, height, weight, overground self-selected (SS) and fastest-comfortable (FC) walking speeds.

| | Gender | Age (yrs) | Leg Length (m) | Height (m) | Weight (kg) | SS Speed (m/s) | FC Speed (m/s) |
|---|---|---|---|---|---|---|---|
| **Younger Adults** | | | | | | | |
| 1 | M | 24 | 0.88 | 1.80 | 66.2 | 1.26 | 1.74 |
| 2 | M | 21 | 0.91 | 1.78 | 64.0 | 1.44 | 1.98 |
| 3 | M | 21 | 0.90 | 1.74 | 60.0 | 1.34 | 2.05 |
| 4 | F | 22 | 0.98 | 1.85 | 82.1 | 1.50 | 2.00 |
| 5 | M | 23 | 0.94 | 1.83 | 82.6 | 1.27 | 1.75 |
| 6 | M | 22 | 0.81 | 1.73 | 80.7 | 1.14 | 1.81 |
| 7 | F | 24 | 0.74 | 1.66 | 47.6 | 1.37 | 1.86 |
| 8 | M | 22 | 0.96 | 1.89 | 74.4 | 1.28 | 1.78 |
| 9 | M | 30 | 0.96 | 1.89 | 75.3 | 1.40 | 1.90 |
| Mean | | 23.2 | 0.90 | 1.79 | 70.3 | 1.33 | 1.87 |
| SD | | 2.8 | 0.08 | 0.08 | 11.8 | 0.11 | 0.12 |
| **Older Adults** | | | | | | | |
| 1 | F | 65 | 0.84 | 1.61 | 57.6 | 1.38 | 1.65 |
| 2 | M | 66 | 0.97 | 1.83 | 75.7 | 1.61 | 1.85 |
| 3 | F | 64 | 0.77 | 1.57 | 45.4 | 1.22 | 1.43 |
| 4 | F | 64 | 0.84 | 1.67 | 69.2 | 1.58 | 1.83 |
| 5 | F | 59 | 0.85 | 1.73 | 82.1 | 1.46 | 1.80 |
| 6 | M | 72 | 0.79 | 1.64 | 70.3 | 1.25 | 1.36 |
| 7 | M | 72 | 0.97 | 1.83 | 96.6 | 1.46 | 1.86 |
| 8 | M | 82 | 0.86 | 1.71 | 78.0 | 1.30 | 1.49 |
| 9 | M | 83 | 0.95 | 1.85 | 98.4 | 1.41 | 1.61 |
| 10 | F | 77 | 0.77 | 1.58 | 60.3 | 1.02 | 1.49 |
| 11 | F | 70 | 0.77 | 1.65 | 63.5 | 1.59 | 1.72 |
| 12 | F | 74 | 0.86 | 1.62 | 56.7 | 1.20 | 1.56 |
| 13 | F | 86 | 0.71 | 1.53 | 64.9 | 0.96 | 1.11 |
| Mean | | 71.8 | 0.84 | 1.68 | 70.7 | 1.34 | 1.60 |
| SD | | 8.3 | 0.08 | 0.10 | 15.4 | 0.21 | 0.22 |

### 2.3. Data processing

The kinematic and *GRF* data were low-pass filtered using a fourth-order Butterworth filter with cutoff frequency of 7 Hz and 20 Hz, respectively (e.g., Shell et al., 2018; Sinclair et al., 2013; Vistamehr et al., 2018; Winter et al., 1974). A 15-segment model (C-Motion, Inc., Germantown, USA) was used to calculate intersegmental joint moments, body center-of-mass (CoM) position and velocity as well as angular momentum for each segment. At each time step, whole-body angular momentum (*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 $\vec{r}_i^{\,COM}$ and $\vec{v}_i^{\,COM}$ are the position and velocity vectors of the $i^{th}$ segment's CoM, respectively. $\vec{r}_{body}^{\,COM}$ and $\vec{v}_{body}^{\,COM}$ are the position and velocity vectors of the whole-body CoM. $\vec{\omega}_i$, $m_i$ and $I_i$ are the angular velocity vector, and mass and moment of inertia of the $i^{th}$ segment, respectively, and $n$ is the number of segments. Angular momentum was normalized by the product of subject mass (kg), height (m) and $\sqrt{g \cdot l}$, where $g$ is the gravitational acceleration and $l$ is the subject height. The term $\sqrt{g \cdot l}$ has units of m/s and provides a normalization technique similar in concept to the Froude number (e.g., Vaughan and O'Malley, 2005). Dynamic balance control was assessed using the peak-to-peak range of $H$ ($H_R$), which was calculated as the difference between the minimum and maximum values of $H$ in each plane over the entire gait cycle. To further understand the regulation of $H$, the individual components of the net external moment were examined during the self-selected walking condition as:

$$\vec{M}_{ext} = \vec{r} \times \overrightarrow{GRF}$$

where $\vec{r}$ is the moment arm vector from body *CoM* to *CoP* and $\overrightarrow{GRF}$ is the vector of *GRFs* (Fig. 1). Peak moment arm values were calculated during the stance phase. The peak moment arm in the mediolateral direction was also calculated during single-limb stance. Step width was calculated as the mediolateral distance between the proximal end position of the foot at ipsilateral heel strike to the proximal end position of the foot at the next contralateral heel strike. Step width and moment arms were normalized by leg length. In addition, variability in moment arm and step width were calculated as the standard deviation of the normalized peak moment arms and step width, respectively. Both the *GRF* peaks and impulses were calculated during early and late stance to capture the effects of instantaneous and average changes in the *GRFs* between age groups. *GRF* impulses were calculated as the time integral of *GRFs* during early (0–50%) and late (51–100%) stance. Peak hip, knee and ankle intersegmental joint moments were analyzed to further understand observed differences between age groups. *GRFs* and joint moments were normalized by body weight and mass, respectively.

### 2.4. Statistical analysis

To compare dynamic balance between the older and younger adults, $H_R$ was compared within each anatomical plane and walking speed using an independent *t*-test with equal variances. To further understand the regulation of dynamic balance during the SS walking condition, net external moment components (*GRFs* and peak moment arms) and intersegmental joint moments, averaged across legs, were compared between age groups using a two-sample *t*-test with equal variances. The equality of variances was confirmed using an F-test. An independent *t*-test was suitable for this study since the goal was to compare differences in the biomechanical quantities between age groups within each walking condition. Also, when significant differences were found in $H_R$, Pearson correlation analyses were performed between $H_R$ and the individual biomechanical quantities. Data for older and younger subjects were combined for the correlation analyses. To minimize the incidence of false negative findings, multiple comparisons were corrected with a Benjamini-Hochberg post-hoc adjustment (Benjamini and Hochberg, 1995).

## 3. Results

### 3.1. Frontal plane

In the frontal plane, $H_R$ was significantly higher in the older adults during both the self-selected and fastest-comfortable walking conditions (Fig. 2). Of the components that contribute to the frontal-plane net external moment (i.e., vertical and *M/L* moment arms and *GRFs* (Fig. 1)), the normalized peak *M/L* moment arm was significantly wider in the older adults (Fig. 3). This was true both during double-limb stance (older: 0.097 ± 0.021; younger: 0.078 ± 0.011; p = 0.033) and single-limb stance (older: 0.073 ± 0.015; younger: 0.060 ± 0.008; p = 0.036) phases. However, there were no differences in the normalized step width between the older (0.13 ± 0.02) and younger (0.14 ± 0.02) adults. Also, there were no differences in the *M/L* moment arm variability or step width variability between the older (*M/L* moment arm variability: 0.0169 ± 0.0049; step width variability: 0.024 ± 0.0044) and younger (*M/L* moment arm variability: 0.0173 ± 0.0062; step width variability: 0.020 ± 0.0068) adults. The $H_R$ in the frontal plane was correlated with the peak *M/L* moment arm during double-limb stance (r = 0.46, p = 0.029) and single-limb stance (r = 0.53, p = 0.011) phases. There were no differences in the *GRF* peaks and impulses, except the vertical *GRF* impulse during early stance was significantly lower in the older adults (Fig. 3).

For the frontal-plane joint moments, the peak hip abduction moment was significantly higher in older adults during both early and late stance (Table 2). Although there were no significant correlations with the frontal-plane $H_R$, the peak hip abduction moment was correlated to the peak *M/L* moment arm (r = 0.50, p = 0.016).

### 3.2. Sagittal plane

In the sagittal plane, there were no significant differences in the $H_R$ between age groups (Fig. 2). Of the components that contribute to the sagittal-plane net external moment (i.e., *A/P* and vertical moment arms and *GRFs*), there were no differences between age groups, except for lower vertical *GRF* impulse in the older adults during early stance (Fig. 3). For the sagittal-plane joint moments, the peak ankle plantarflexor moment was significantly lower in older adults (Table 2). The vertical *GRF* impulse was correlated to the peak ankle plantarflexor moment (r = 0.47, p = 0.027).

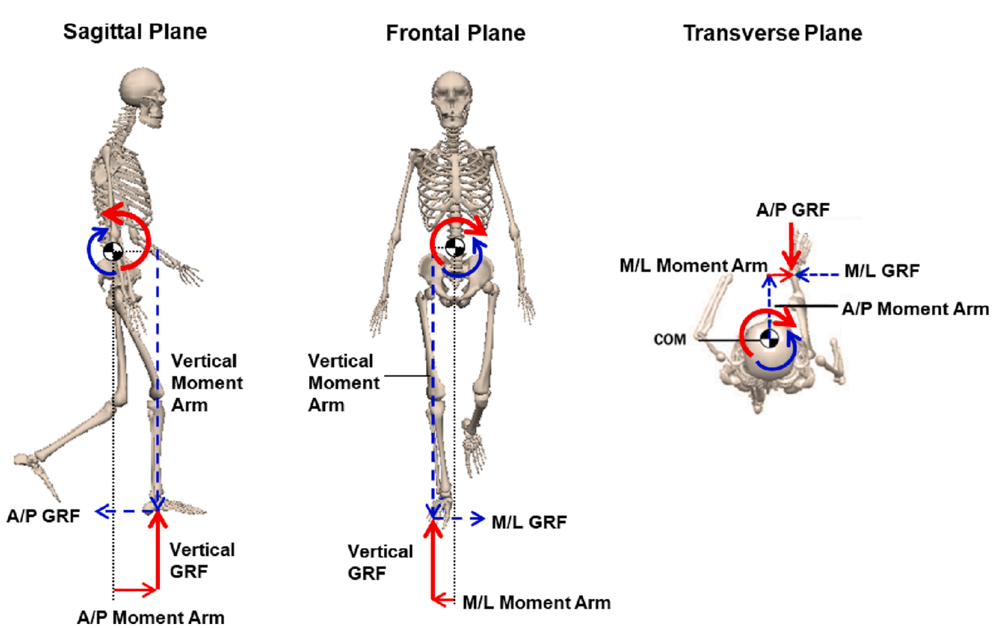


**Fig. 1.** The net external moment components in the frontal, sagittal, and transverse planes during single-limb stance. Whole-body *CoM* is shown with '◐'. The *GRF* vectors, their corresponding moment arms, and the resulting external moment component appear in the same color. For instance, in the sagittal plane, the A/P moment arm and vertical GRF create a backward external moment about the *CoM* (all shown in red). The higher magnitude of the vertical *GRF* compared to other components (after normalizing moment arms and *GRFs* by leg length and body weight, respectively) is highlighted by the line thickness. Also, the dominant moment components are shown with thicker and larger circular arrows.

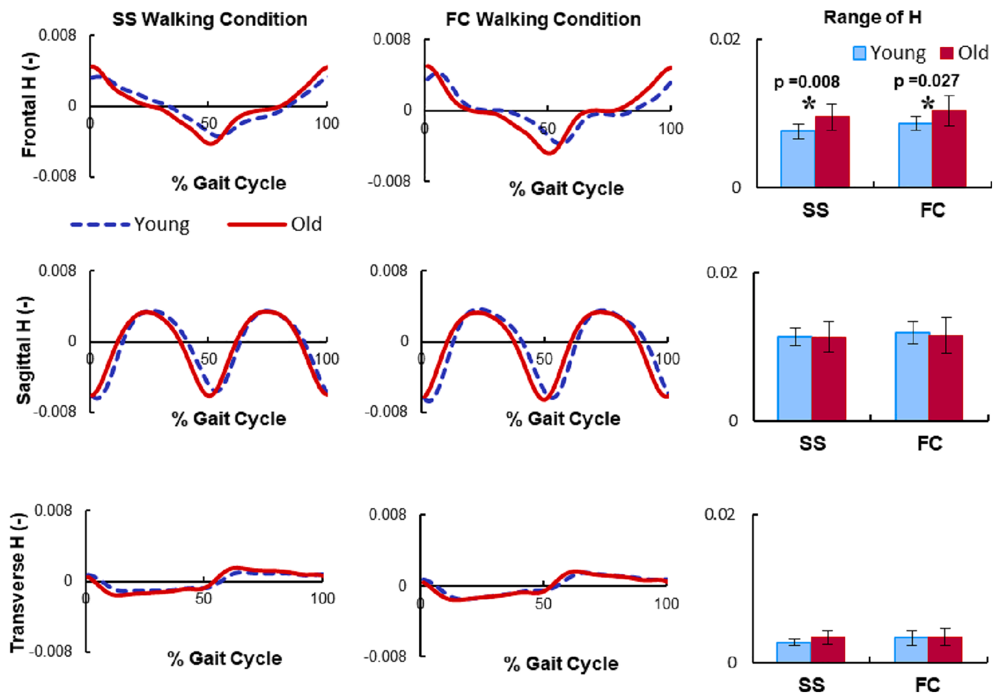


**Fig. 2.** Normalized, mean whole-body angular momentum ($H$) in the frontal, sagittal, and transverse planes during the self-selected (SS) and fastest-comfortable (FC) walking speed conditions. The mean (±SD) range of whole-body angular momentum ($H_R$) is shown on the right. Statistically significant differences between the younger and older subjects are indicated with '*' ($p < 0.05$).

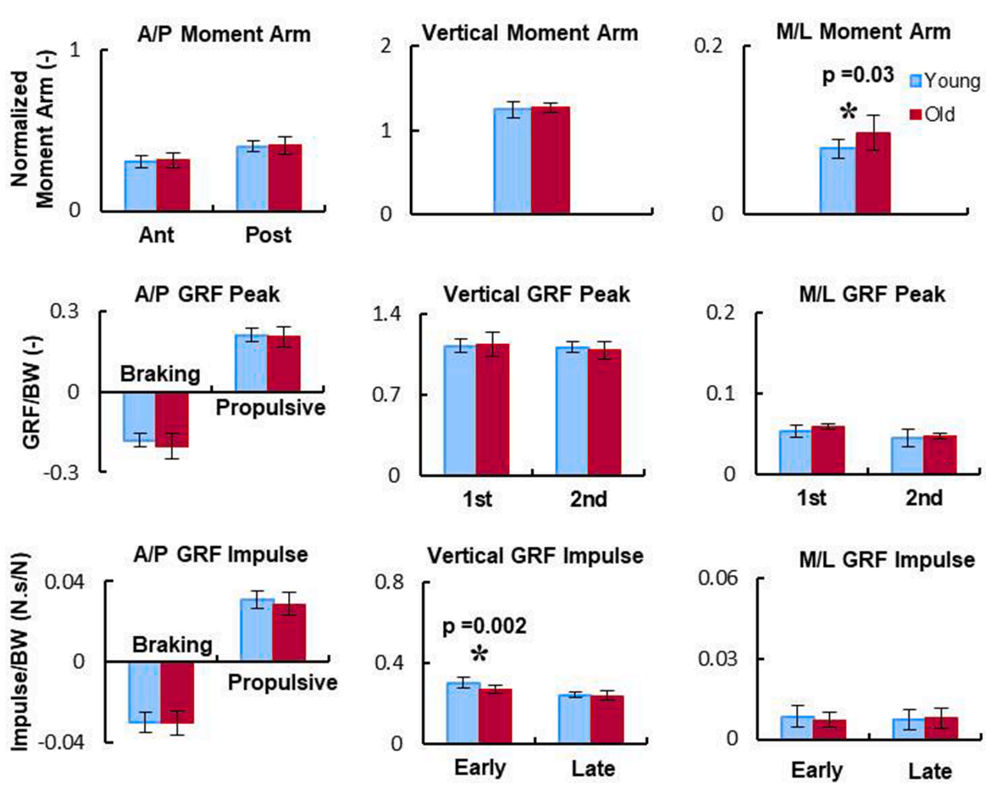


**Fig. 3.** Biomechanical quantities during the self-selected walking speed condition. Top row shows the mean (±SD) peak moment arms normalized by leg length. The A/P moment arm is presented in the anterior (Ant) and posterior (Post) directions. Middle row shows the body weight (BW) normalized, mean (±SD) first and second peak *GRFs*. Bottom row shows the BW-normalized mean (±SD) *GRF* impulses during early and late stance. Significant differences between the two age groups are indicated with '*' ($p < 0.05$).

### 3.3. *Transverse plane*

In the transverse plane, there were no differences in the $H_R$ between age groups (Fig. 2). Of the components that contribute to the transverse-plane net external moment (i.e., *A/P* and *M/L* moment arms and *GRFs*), the peak *M/L* moment arm was significantly wider in the older adults (Fig. 3). However, there were no correlations between the Transverse-plane $H_R$ and peak *M/L* moment arm.

## 4. Discussion

Our hypothesis that the regulation of whole-body angular

**Table 2**
Mean (±SD) peak intersegmental joint moments, normalized by body mass for old and young participants. P-values from an independent samples *t*-test between the younger and older subjects are shown within each speed. Significant differences ($p < 0.05$) between age groups are highlighted in bold.

| Joint moment | Self-selected Walking Condition | | |
|---|---|---|---|
| | Young | Old | p-value |
| Hip abduction, 1st peak | **0.83 (±0.17)** | **0.99 (±0.16)** | **0.035** |
| Hip abduction, 2nd peak | **0.68 (±0.13)** | **0.85 (±0.11)** | **0.004** |
| Hip extension peak | 0.95 (±0.13) | 0.83 (±0.27) | 0.230 |
| Hip flexion peak | 1.07 (±0.23) | 0.98 (±0.29) | 0.459 |
| Knee extension peak | 0.49 (±0.23) | 0.53 (±0.25) | 0.713 |
| Knee flexion peak | 0.44 (±0.14) | 0.34 (±0.09) | 0.056 |
| Ankle plantarflexion peak | **1.52 (±0.15)** | **1.39 (±0.13)** | **0.033** |

momentum would be more challenging in older adults was supported in the frontal plane. The older adults had a significantly higher $H_R$ in the frontal plane compared to the younger adults during both the self-selected and fastest-comfortable walking conditions (Fig. 2). However, in the sagittal and transverse planes, there were no differences in $H_R$ between age groups at either walking speed (Fig. 2). The lack of difference in the regulation of sagittal-plane $H$ despite the differences in the frontal-plane $H$ suggests that maintaining mediolateral balance during walking is more challenging in older adults. This is consistent with previous studies showing that sagittal-plane balance control can be maintained passively while mediolateral balance requires active control (Bauby and Kuo, 2000; Kuo, 1999). In addition, the lack of difference in transverse-plane $H_R$ between age groups is consistent with previous studies reporting lower angular momentum values in the transverse plane which require little regulation (Bennett et al., 2010; Herr and Popovic, 2008; Silverman et al., 2012).

The group difference in frontal-plane balance control was primarily due to the differences in the mediolateral moment arm (i.e., mediolateral foot placement with respect to the body center-of-mass). The older adults had a significantly wider peak mediolateral moment arm (Fig. 3), which was correlated with a higher $H_R$. The wider mediolateral moment arm combined with the dominant vertical *GRF* creates a greater destabilizing moment about the body *CoM* that acts to rotate the body towards the contralateral swing leg. During single-limb stance when there is no counteracting moment from the contralateral leg, the older adults generated higher hip abduction moment (Table 2) to counteract the destabilizing moment and improve balance control, which is consistent with others (e.g., Brough et al., 2021; MacKinnon and Winter, 1993). However, older adults who cannot generate greater hip abduction moments due to muscle weakness (Johnson et al., 2004) may be subject to decreased balance control and increased risk of falling.

The older adults also generated lower peak ankle plantarflexor moments (Table 2), which is consistent with prior studies (e.g., Thelen et al., 1996; Vandervoort and McComas, 1986) and the distal-to-proximal redistribution phenomenon in older adults (DeVita and Hortobagyi, 2000). Others have reported that fallers generate lower plantarflexor moments compared to non-fallers (e.g., Barak et al., 2006) and that older adults with a lower rate of support limb joint moment generation and lower peak ankle moments have a higher $H_R$ following a trip and are more likely to fall (Pijnappels et al., 2005). The decreased ankle plantarflexor moment has important implications for balance control, as previous studies have shown the plantarflexors play a critical role in maintaining dynamic balance by regulating whole-body angular momentum in both the sagittal (Neptune and McGowan, 2011) and frontal (Neptune and McGowan, 2016) planes during walking. During late stance, ankle plantarflexors contribute to the vertical *GRFs* (Anderson and Pandy, 2003; Liu et al., 2008; Neptune et al., 2001) which along with the mediolateral moment arm creates an external moment towards the contralateral leg. In contrast, the gluteus medius contributes to the medial *GRF* (Neptune and McGowan, 2016), which combined with the vertical moment arm creates an external moment towards the ipsilateral leg. In the present study, the higher mediolateral moment arm in the older adults was identified as the primary contributor to the higher frontal-plane $H_R$ in this age group. Further, the peak mediolateral moment arm was inversely correlated ($r = -0.46$, $p = 0.029$) with the peak plantarflexor moment and directly correlated with the peak hip abduction moment ($r = 0.50$, $p = 0.016$). Thus, one explanation could be that the older adults compensate for their decreased peak ankle plantarflexor moment by increasing their mediolateral moment arm, which increases the external moment towards the contralateral leg. However, the increased mediolateral moment arm imposes higher demand on the stance leg hip abductors to counteract any excessive destabilizing moment and regulate the frontal plane $H$, which is consistent with the observed increases in hip abduction moment.

Walking is a dynamic task and proper weight transfer between the legs is an essential element of balance control (Woollacott and Tang, 1997). A number of studies analyzing mediolateral balance control have reported step width as a critical component (e.g., Bauby and Kuo, 2000; Dean et al., 2007; Donelan et al., 2004; Gabell and Nayak, 1984; Guimaraes and Isaacs, 1980; Maki, 1997). Although step width provides insight regarding the base of support boundary during double-limb stance, it does not capture the weight transfer element and control of body *CoM* with respect to the base of support. Understanding the *CoM* movement with respect to the base of support is particularly important during single-limb stance when a greater mediolateral separation between the body *CoM* and stance foot can create a destabilizing moment arm and result in a higher whole-body angular momentum that requires greater control. The present study found that even though step width remained unchanged across age groups or in a prior study across walking conditions (Frame et al., 2020), there were still significant differences in the frontal-plane $H_R$. Thus, step width alone is not a sufficient measure of mediolateral balance control.

A potential limitation of this study is that the older adult participants may have been more physically active than the average older adults in the community. Thus, some of the elderly gait characteristics (e.g., step width and step width variability) may have not been captured in this cohort. However, the age related neuromechanic differences (e.g., body *CoM* control and joint moment generation) were still evident in the dataset. Another potential limitation is that the FC walking condition did not include kinetic data to quantify the external moment components. However, we performed a post-hoc analysis and quantified the *M/L* moment arm as the maximum distance between the body *CoM* and foot *CoM* in the mediolateral direction (normalized by leg length), and the findings were consistent with those during the SS walking condition obtained using the *CoP* data. That is, during FC walking, the normalized peak *M/L* moment arm during single-limb stance was significantly wider in the older adults (older: $0.079 \pm 0.013$; younger: $0.066 \pm 0.011$; $p = 0.021$) and the frontal-plane $H_R$ was correlated with the peak *M/L* moment arm ($r = 0.65$, $p = 0.002$). Thus, it is highly likely that the observed group differences in frontal-plane $H_R$ during FC walking were primarily related to a wider mediolateral foot placement relative to the body *CoM*, which is consistent with our findings during the SS walking. Lastly, a limitation of $H_R$ is that it hasn't been benchmarked against fall rates, and therefore the threshold values predicting a high fall risk are unknown.

## 5. Conclusion

The successful regulation of whole-body angular momentum is critical in maintaining dynamic balance and preventing falls. The range of frontal-plane angular momentum in older adults was higher compared to younger adults, which was related to a wider foot placement with respect to the body *CoM* in the older adults. That is, the older adults did not shift their body *CoM* as closely to their stance foot. During single-limb stance, the wider foot placement (i.e., M/L moment arm) combined with the dominant vertical *GRF* creates a greater destabilizing external moment that acts to rotate the body towards the contralateral leg and creates a greater range of angular momentum in the frontal plane. Thus, in response to a perturbation, it may be more challenging to regulate the already increased angular momentum in older adults with muscle weakness and slow reaction times. Future perturbation studies are needed to test this premise. An interesting finding was that there were no differences in the range of sagittal-plane angular momentum between age groups. These results suggest that maintaining mediolateral balance may be more challenging than maintaining sagittal-plane balance. Further, exercise programs that target appropriate weight transfer mechanisms and increasing lower extremity muscle strength, particularly of the hip abductors and ankle plantarflexors, may improve frontal-plane balance control in older adults.

## Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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## Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2021.110717.

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