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Effect of age on center of mass motion during human walking

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ABSTRACT

The objective of this study was to investigate the effects of age and speed on body center of mass (COM) motion over a gait cycle. Whole body kinematics and ground reactions were recorded for 21 healthy young (21–32 y) and 20 healthy older adults (66–81 y) walking at 80%, 100% and 120% of preferred speed. The limb-induced COM accelerations and the work done on the COM by the limbs were computed. Despite walking with similar gait speeds, older adults did significantly ($p < 0.05$) less positive work on the COM during push-off but then performed more positive work on the COM during midstance. As a result, older adults induced lower tri-axial COM accelerations via the trailing limb and higher vertical COM acceleration via the leading limb during double support. Older adults also reduced the mediolateral COM acceleration induced by the leading limb during the last third of double support. The forward and vertical components of the limb-induced COM accelerations were highly correlated ($p < 0.005$) but were not correlated to the mediolateral component during double support, at any speed. Together, these results suggest that older adults use the leading limb to compensate for reduced vertical support and work done by the trailing limb. Further, older adults seem to adapt their gait patterns to reduce mediolateral COM accelerations. These findings are relevant for understanding the factors that underlie walking performance and lateral balance in old age.

1. Introduction

Aging induces a shift in joint power production during walking, with older adults exhibiting reduced ankle plantarflexor power during push-off and increased hip flexor power during late stance or hip extensor power during early and midstance [1–5]. These changes have been detected experimentally whether the young and older adults have walked at equal speeds [3,5] or the power measures have been adjusted to account for a slower gait speed in the older adults [4]. Further, as walking speed increases, the power differences become larger [2,5,6]. Although age-related changes in walking coordination have been documented, the factors that underlie these changes are not well understood. Proposed mechanisms include distal muscle weakness [2,3] and a loss of flexibility at the hip [6]. Increased difficulty and/or concerns with lateral balance [7–9] may also be contributing factors. A better understanding of the relative importance of these factors can be obtained by considering how individual limbs influence movement at the whole body level.

At least three simulation studies have investigated the coordination of whole body motion during normal walking [10–12]. Although each study looked at a different combination of segments to represent the bulk of the body mass, their combined results imply that the ankle plantarflexors contribute significantly to the center of mass (COM) forward and vertical accelerations during late stance and pre-swing. In view of the reduced ankle power output of older adults during push-off, this result suggests that the sagittal-plane accelerations of the COM during double support may be reduced by aging. It is possible, then, that increased hip extensor power is compensatory [2,3], providing additional acceleration during single support to maintain walking speed.

Age-related changes in joint kinetics may also influence COM motion in the mediolateral direction. It has previously been established that even healthy older adults experience difficulty in controlling mediolateral stability [8]. Walking includes a substantial single support period such that control of mediolateral balance may be an issue [1,13], particularly when transitioning support from one limb to the other. Interestingly, the age-related decrement in ankle power emerges during the double support period. Thus, it is possible that observed changes in sagittal-plane joint kinetics could alter the control of mediolateral COM motion. This potential coupling of induced forward and mediolateral COM motion could arise mechanically from linked-segment dynamics [14] or neurally, from motor control synergies [15].

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This study was designed to investigate age-related changes in tri-axial (forward, vertical, and medial) COM motion during normal walking. We hypothesized that the older adults would exhibit a decrease in induced COM accelerations and work done via the trailing limb during double support, with compensatory increases in induced COM accelerations and work done by the leading limb. Further, we expected the medial and sagittal-plane accelerations induced by the trailing limb during pre-swing to be coupled to each other. Finally, we expected any observed age-related changes in COM accelerations and work to interact with walking speed, becoming larger as speed increased.

2. Methods

Twenty-one healthy young (age 26 ± 3 y, height 1.73 ± 0.11 m, mass 69 ± 12 kg) and 20 healthy older adults (age 72 ± 5 y, height 1.69 ± 0.09 m, mass 69 ± 11 kg) performed five walking trials at 80%, 100%, and 120% of preferred speed along a 10 m walkway instrumented with three fixed force plates (AMTI, Watertown, MA) and an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). These subjects were a subset of a larger gait study of young and older adults, and details of the experimental setup can be found in a previous communication [5]. Exclusion criteria for this study included major orthopedic diagnoses (bone fractures, joint fusions or replacements, limb amputations) in the lower back, pelvis and lower extremity; joint pain; cardiac, neurologic or balance impairments; and failure to pass cognitive (24 score on mini-mental state exam) and plantar sensation (perception of a 10-g monofilament) tests. Subjects gave informed consent prior to the study. The test protocol was approved by the Health Sciences Institutional Review Board of our institution.

Force plate data were first used to identify the heel strike time for two consecutive foot landings (vertical force > 10 N), from which the start and end times of a gait cycle (GC) were determined. Ground reaction force data were extracted from the middle 50% (25–75%) of this GC, where no limb was in touch with the ground outside of the force plate region. This period involved the end of a step (step 1, 25–50% GC) and the beginning of the next step (step 2, 50–75% GC) on the opposite limb. We set the 0% gait cycle mark at the beginning of step 2. We then shifted the step 1 data forward in time so that its start would merge with the end of step 2 at the 25% GC mark. This manipulation required changing the sign on the mediolateral component of the ground reactions, resulting in the half gait cycles that were analyzed in this study.

Full body kinematics were measured using 42 passive motion capture markers with 23 of them placed on anatomical landmarks of the pelvis, arms, legs and feet, and the other 19 placed on limb segments to facilitate segment tracking [16]. Kinematic data were used in conjunction with heel strike times to determine walking velocity, step length, step width, mediolateral COM excursion and mediolateral stability. Step time was defined as the time between two consecutive heel strikes. Walking velocity was defined as step length divided by step time. Step width was defined as the mediolateral distance between the average positions of the heel markers of the two feet during their respective stance times. Both step length and step width were normalized to body height. Mediolateral COM excursion was defined as the range of mediolateral motion observed over a gait cycle. Because stability is believed to depend on keeping COM motion within the base of support [9], the mediolateral COM excursion was divided by the step width to obtain an indicator of mediolateral stability.

Tri-axial COM accelerations were computed by dividing the directional components of the net ground reaction force (sum of the two limb contributions) by body mass, and subtracting gravity's contribution in the vertical direction. COM velocity and position were then computed by integrating the acceleration curves in each direction [17,18]. Integration constants were assigned by using the average forward velocity as measured by the pelvis markers, and assuming that the average mediolateral and vertical velocities were zero over a full gait cycle. Net acceleration, velocity and position traces were averaged across five trials for each subject to obtain representative COM motion curves at every speed.

The tri-axial COM accelerations induced by each limb were calculated by dividing the directional components of the individual ground reactions by body mass [17,18]. Limb-induced COM acceleration traces were then averaged over three phases of the gait cycle: double support (when two limbs contacted the ground), midstance (when one limb contacted the ground and the forward COM acceleration was negative) and terminal stance (when one limb contacted the ground and the forward COM acceleration was positive). We also computed the dot product of the individual limbs' ground reactions and the COM velocity vector at each point in time to evaluate the instantaneous power delivered to the COM. We integrated these quantities with respect to time in order to obtain the external mechanical work done by each limb on the COM [18].

A two factor analysis of variance (ANOVA) with two levels on age (young, old) and three repeated measure levels on speed (slow, preferred, fast) was then carried out for each limb-induced acceleration and work quantity. Post hoc Tukey comparisons were performed to determine the source of significant age and/or age-by-speed effects by comparing the young and old populations at each of the three speeds. Potential coupling between the directional components of the limb-induced COM acceleration was evaluated by pooling the acceleration data from young and older adults along each direction, and calculating Pearson correlation coefficients for every pair of directional components, both within a limb and between limbs. Significance for all statistical tests (ANOVAs, Tukey comparisons and correlation coefficients) was established at $p < 0.05$.

3. Results

3.1. Spatiotemporal measures

Average gait speed, normalized step length, normalized step width, and mediolateral COM excursions were not significantly different between young and older adults at any speed (Table 1). Further, the COM excursion/step width ratio, an indicator of mediolateral stability, was not significant with age.

3.2. COM accelerations

Older adults walked with different COM acceleration patterns than young adults during double support (Fig. 1a). The older adults showed a tendency for reduced trailing limb-induced tri-axial accelerations and increased leading limb-induced vertical acceleration during double support (Fig. 1b). Post hoc analysis revealed that these double-limb support differences were significant at the preferred and fast speeds (Table 2). Generally, the average limb-induced COM accelerations during midstance and terminal stance were not significantly different between the age groups at any speed. The only exception was the leading limb vertical acceleration during midstance, which reached significance at the fast speed.

The trailing limb accelerated the COM forward and the leading limb decelerated it during double support (Fig. 2). Older adults showed decreased COM forward acceleration during push-off (first

Table 1
Spatiotemporal gait measures of young and older adults.

		Slow speed		Preferred speed		Fast speed	
		Ave (SD)	p-Value	Ave (SD)	p-Value	Ave (SD)	p-Value
Gait speed (m/s)	Young	1.058 (0.100)	0.770	1.326 (0.133)	0.919	1.587 (0.133)	0.533
	Old	1.048 (0.103)		1.322 (0.128)		1.557 (0.154)	
Step length, normalized by height	Young	0.376 (0.028)	0.570	0.415 (0.038)	0.960	0.458 (0.042)	0.830
	Old	0.369 (0.043)		0.415 (0.028)		0.453 (0.039)	
Step width, normalized by height	Young	0.038 (0.016)	0.404	0.041 (0.014)	0.299	0.039 (0.019)	0.255
	Old	0.034 (0.018)		0.035 (0.017)		0.033 (0.017)	
Mediolateral COM excursion (m)	Young	0.030 (0.009)	0.083	0.025 (0.007)	0.101	0.021 (0.005)	0.065
	Old	0.026 (0.008)		0.021 (0.007)		0.018 (0.006)	
COM excursion/step width	Young	0.517 (0.212)	0.734	0.391 (0.162)	0.566	0.360 (0.140)	0.800
	Old	0.544 (0.291)		0.428 (0.239)		0.374 (0.188)	

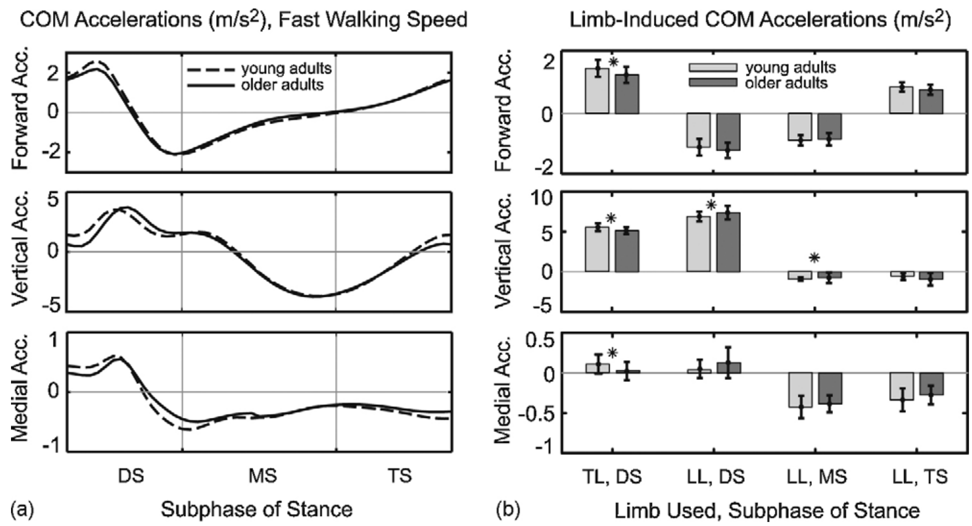

Fig. 1. Center of mass accelerations of young and older adults walking at fast speed during the subphases of stance. (a) Ensemble averaged COM acceleration plots. (b) Average limb-induced COM accelerations. *p-Value < 0.05. Abbreviations: Acc., Acceleration; DS, double support; MS, midstance; TS, terminal stance; TL, trailing limb; LL, leading limb.

half of double support) via the trailing limb. In the vertical direction, the trailing limb reduced its upward induced acceleration from its maximum value to zero while the leading limb's contribution increased from zero to its maximum value during this phase. Older adults displayed lower vertical acceleration in the first half of double support but greater vertical acceleration in the second half, during leading limb loading. In the mediolateral direction, the trailing limb accelerated the body toward the leading limb during the first half of double support. The leading limb contributed to the COM acceleration toward the new stance side during the first two thirds of double support, and then accelerated the COM back toward the midline during the last third of double support. The net mediolateral acceleration (sum of the two limb contributions) crossed the zero acceleration line at about two thirds of the phase for both age groups, representing a functional transition point between mediolateral acceleration and deceleration subphases. Distinguishing between these functional subphases and evaluating the average acceleration over the deceleration subphase yielded a significant reduction in the COM mediolateral deceleration induced via the leading limb for the older adults at the two fastest speeds (Table 2).

3.3. Intra- and inter-limb correlations

The forward COM accelerations induced by each limb during double support correlated to the vertical accelerations induced by

Table 2
p-Values of age and age-by-speed ANOVA effects on limb-induced center of mass accelerations and work at each subphase of the gait cycle.

Limb, subphase	Age p-value	Age-by-speed p-value	Post hoc Tukey comparisons[a]					
			Slow speed		Preferred speed		Fast speed	
			Difference	p-Value	Difference	p-Value	Difference	p-Value
Forward acceleration (m/s^2)								
TL, DS	0.062	0.019*	−0.08	0.083	−0.12	0.002*	−0.20	<0.001*
LL, DS	0.209	0.508						
LL, MS	0.362	0.771						
LL, TS	0.130	0.097						
Vertical acceleration (m/s^2)								
TL, DS	0.019*	0.051	−0.15	0.273	−0.32	<0.001*	−0.38	<0.001*
LL, DS	0.021*	0.066	+0.29	0.020*	+0.56	<0.001*	+0.53	<0.001*
LL, MS	0.310	0.023*	−0.02	1.000	+0.09	0.745	+0.24	0.009*
LL, TS	0.114	0.062						
Medial acceleration (m/s^2)								
TL, DS	0.087	0.036*	−0.03	0.432	−0.06	0.002*	−0.08	<0.001*
LL, DS	0.188	0.105						
LL, MS	0.287	0.677						
LL, TS	0.078	0.587						
LL, Dec	0.003*	0.031*	+0.09	0.209	+0.13	0.006*	+0.22	<0.001*
Work (J/kg)								
TL, DS	0.013*	0.021*	−0.02	0.318	−0.04	<0.001*	−0.06	<0.001*
LL, DS	0.461	0.695						
LL, MS	<0.001*	0.014*	+0.04	0.041*	+0.09	<0.001*	+0.09	<0.001*
LL, TS	0.120	0.681						

Abbreviations: TL, trailing limb; LL, leading limb; DS, double support; MS, midstance; TS, terminal stance; Dec, medial deceleration phase of double support.
[a] Difference magnitudes (older adults relative to young) and p-values of post hoc comparisons are included only for those limb-phase conditions where significant age or age-by-speed effects exist.
* p-Value < 0.05.

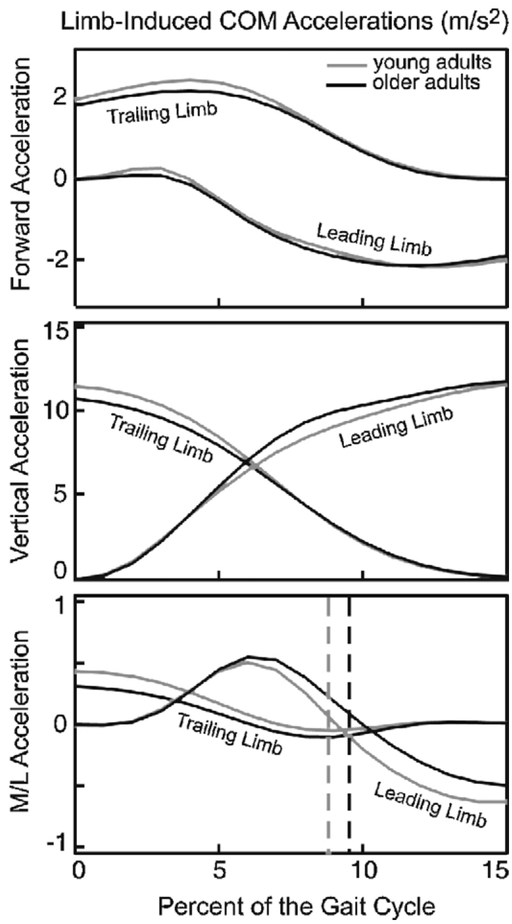

Fig. 2. COM accelerations induced by each limb (solid lines) during the double limb support phase for young and older adults walking at fast speed. The vertical dashed lines show the approximate locations of the mediolateral (M/L) acceleration's zero crossing points for each group.

the same limb (Table 3). The mediolateral accelerations induced by each limb during double support did not correlate to either the forward or the vertical accelerations induced by the same limb. The accelerations induced by the trailing limb negatively correlated with the accelerations induced by the leading limb along each direction. These correlations were present at all speeds (Table 3).

3.4. External COM power and work

The external power and work done by the limbs on the COM exhibited differences between the age groups (Fig. 3). Specifically, the mechanical work done by the trailing limb on the COM during double support was lower in the older adults than in the young, reaching significance at preferred and fast speeds (Fig. 3b). The mechanical work done by the leading limb on the COM during midstance was significantly different between the age groups at every speed, with midstance work being positive in the older adults but slightly negative in the young adults (Fig. 3b). Both the work done by the trailing limb during double support and the work done by the leading limb during midstance showed discernible age-by-speed effects (Table 2).

4. Discussion

The older adults exhibited similar preferred walking speed and step length as the young adults at all speeds. However, older adults generated less forward acceleration and performed less work via the trailing limb during double support. Similar to this study, reduced trailing limb work during double support has previously been shown in older adults [19]. These results lead to the question of how a similar walking speed was maintained. Our analyses indicate that compensation was not achieved by the leading limb during double support, since forward accelerations were not significantly different between the groups in this phase (Fig. 1b, Table 2). Instead, the older adults performed more net positive work, relative to the young adults, during the subsequent midstance portion of single support. This result is consistent with previous observations of older adults performing more work than young adults via the hip extensor power burst [3,5], which extends into midstance. Although we did not find significant age-related differences in the forward acceleration when averaged over midstance, this was the only subphase of the gait cycle when there was a tendency for slightly greater forward acceleration (i.e., less deceleration) in the older adults (Fig. 1a). This suggests that the older adults likely coordinated the leading limb during midstance to functionally compensate for reduced trailing limb push-off during double support, thereby allowing them to maintain similar walking speeds as the young adults.

Table 3
Pearson correlations between the COM accelerations induced during double support by the trailing and leading limbs.

Correlated variables	Slow speed		Preferred speed		Fast speed	
	Correlation coefficient (r)	p-Value (p)	Correlation coefficient (r)	p-Value (p)	Correlation coefficient (r)	p-Value (p)
Intra-limb correlations						
TL Fwd vs TL Ver	0.471	0.002*	0.509	<0.001*	0.542	<0.001*
TL Ver vs TL M/L	−0.180	0.272	−0.164	0.318	−0.259	0.112
TL Fwd vs TL M/L	0.077	0.642	−0.055	0.739	−0.215	0.189
LL Fwd vs LL Ver	−0.491	0.002*	−0.710	<0.001*	−0.641	<0.001*
LL Ver vs LL M/L	0.037	0.823	−0.018	0.912	−0.011	0.950
LL Fwd vs LL M/L	0.085	0.609	0.138	0.401	0.063	0.703
Inter-limb correlations						
TL Fwd vs LL Fwd	−0.663	<0.001*	−0.635	<0.001*	−0.606	<0.001*
TL Fwd vs LL Ver	0.035	0.833	0.197	0.232	0.136	0.410
TL Fwd vs LL M/L	−0.134	0.416	−0.107	0.517	0.160	0.330
TL Ver vs LL Fwd	0.063	0.702	+0.075	0.649	−0.011	0.949
TL Ver vs LL Ver	−0.656	<0.001*	−0.576	<0.001*	−0.498	0.001*
TL Ver vs LL M/L	0.010	0.951	−0.030	0.855	0.168	0.308
TL M/L vs LL Fwd	−0.153	0.350	−0.139	0.399	−0.092	0.578
TL M/L vs LL Ver	0.145	0.377	0.160	0.330	0.147	0.371
TL M/L vs LL M/L	−0.810	<0.001*	−0.798	<0.001*	−0.816	<0.001*

Abbreviations: TL, trailing limb; LL, leading limb; Fwd, forward; Ver, vertical; M/L, mediolateral.
* p-Value < 0.05.

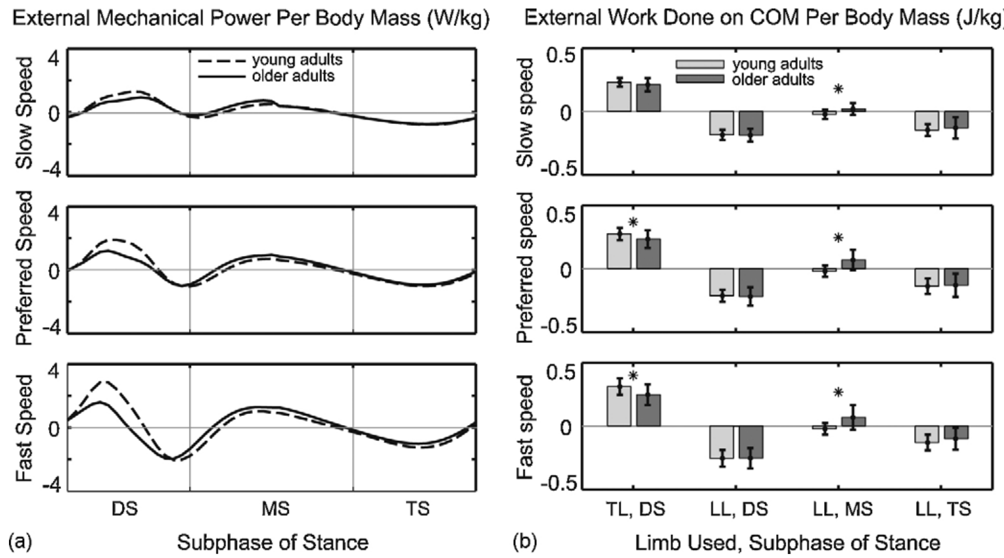

Fig. 3. External mechanical power and work done on the body COM during the subphases of stance in young and older adults. (a) Ensemble averaged power plots. (b) External work done by the limbs. At faster speeds, older adults did less work with the trailing limb during double support. In order to maintain walking speed, the leading limb may compensate by doing net positive work during mid-stance. *p-Value < 0.05. Abbreviations: Acc., Acceleration; DS, double support; MS, midstance; TS, terminal stance; TL, trailing limb; LL, leading limb.

During double support (~0–15% GC), both limbs are in contact with the ground as the trailing limb passes responsibility to the leading limb to support and accelerate the body. We had hypothesized that tri-axial COM accelerations induced by the trailing limb would be reduced in the older adults during this phase. Our results (Fig. 1b) support this hypothesis. In the forward and vertical directions, this result is likely due to decreased ankle plantarflexor power [5]. We also expected that the older adults would then compensate by increasing vertical support and reducing mediolateral deceleration via the leading limb during this phase. Indeed, we observed a significant increase in the vertical acceleration in the older adults, but no significant change in mediolateral deceleration. Upon inspection of the individual limb contributions to mediolateral acceleration, it can be seen that the leading limb has a dual role during double support, first assisting the trailing limb to accelerate the body toward the new stance limb, and then accelerating the body back toward the midline (Fig. 2). Dividing the double support phase into mediolateral acceleration and deceleration subphases does indeed result in a significant reduction in deceleration by the leading limb during the deceleration subphase. Based on this finding, we believe that it may be beneficial to define gait phases differently for the mediolateral direction based on the zero-crossings of the COM acceleration curves.

The fact that older adults had reduced mediolateral COM accelerations (relative to the young) during double support suggests that joint kinetic changes in the frontal plane may also exist. In addition, it invites the question whether the older adults may have improved their mediolateral stability by reducing COM excursions while keeping a similar step width [9]. However, the COM excursion/step width ratio was not significantly reduced in the older adults with respect to the young (Table 1). Instead, it seems that control of mediolateral accelerations (i.e., the rate at which velocity changes) during the transition from one limb to the other becomes more important as adults age than reducing the excursion/step width ratio. This result suggests that stability measures based simply on COM position relative to foot placement are insufficient to describe the mediolateral balance challenge imposed on the motor control system by walking. Additional dynamic measures (accounting for mediolateral velocity [20] and/or acceleration) may be needed to fully characterize mediolateral stabilization.

Our investigation of two-way correlations during double support also yielded interesting results. First, we found that the leading limb- and trailing limb-induced tri-axial accelerations were negatively correlated. For example, increased forward acceleration by the trailing limb was associated with increased deceleration via the leading limb. This result may be indicative of mechanical constraints due to step-to-step transitions [18], whereby the lower body configuration with the limbs as two sides of a triangle would tend to generate larger impact forces on the leading limb in response to higher trailing limb accelerations. Hence, the leading limb assists the trailing limb to achieve the forward progression, vertical support and mediolateral shift of the COM. Secondly, we saw that during double support, the forward and vertical accelerations induced by the trailing limb correlated to each other but not to the mediolateral accelerations, consistent with independent lateral control [13]. Therefore, age-related reductions in mediolateral COM acceleration during push-off are likely not attributable to mechanical or neural coupling with sagittal-plane motions. Instead, muscles with the potential to directly induce frontal plane body motions (such as the hip adductors/abductors on either limb) may be involved in producing this age-related difference [21].

The approach used in this study was able to determine the contribution of individual limbs to COM accelerations and work, but did not directly identify the joints or muscles responsible for those changes. However, it was previously shown in this same older adult population that the work done by the plantarflexors was diminished and the work done about the hips was enhanced, relative to the young adults [5]. Thus, these underlying joint kinetic changes, which are similar to results found by others [1–4], likely contribute directly to the changes in COM kinetics and kinematics observed in this study. It is worth noting that COM accelerations and limb work can be directly computed using only forceplate data [17], which would represent a simpler way than full gait analysis to identify potential age-related changes in gait mechanics.

5. Conclusion

In this study, we have shown that healthy older adults control COM motion differently than young adults when walking at preferred and fast speeds. In particular, older adults rely less on the trailing limb to induce forward and vertical accelerations during double support, and compensate by using the leading limb to increase support and do additional work during midstance. In addition, a significant reduction in the mediolateral COM acceleration occurs that is not coupled to changes in sagittal COM motion. These findings are relevant for understanding the factors that underlie walking performance and the causes of mediolateral balance difficulties in older adults.

Acknowledgments

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Conflict of interest statement

None of the authors have any conflicts of interest.

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