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Old adult fallers display reduced flexibility of arm and trunk movements when challenged with different walking speeds

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ABSTRACT

Specific patterns of pelvic and thorax motions are required to maintain stability during walking. This cross-sectional study explored older-adults' gait kinematics and their kinematic adaptations to different walking speeds, with the purpose of identifying mechanisms that might be related to increased risk for falls. Fifty-eight older adults from self-care residential facilities walked on a treadmill, whose velocity was systematically increased with increments of 0.1 meters/second (m/s) from 0.5 to 0.9 m/s, and then similarly decreased. Thorax, pelvis, trunk, arms, and legs angular total range of motion (tROM), stride time, stride length, and step width were measured. Twenty-one of the subjects reported falling, and 37 didn't fall. No significant effect of a fall history was found for any of the dependent variables. A marginally significant interaction effect of fall history and walking speed was found for arms' tROM ($p = 0.098$). Speed had an effect on many of the measures for both groups. As the treadmill's velocity increased, the non-fallers increased their arm ($15.9 \pm 8.6^\circ$ to $26.6 \pm 12.7^\circ$) and trunk rotations ($4.7 \pm 1.9^\circ$ to $7.2 \pm 2.8^\circ$) tROM, whereas for the fallers the change of arm ($14.7 \pm 14.8^\circ$ to $20.8 \pm 13^\circ$) and trunk ($5.5 \pm 2.9^\circ$ to $7.3 \pm 2.3^\circ$) rotations tROM were moderate between the different walking speeds. We conclude that walking speed manipulation exposed different flexibility trends. Only non-fallers demonstrated the ability to adapt trunk and arm ROM to treadmill speed i.e., had a more flexible pattern of behavior for arm and trunk motions, supporting the upper-body's importance for stability while walking.

1. Introduction

Approximately 30% of older adults and 50% over the age of 80 experience a fall at least once a year [1]. One out of five falls causes serious injury [2], which constitutes a significant part of the healthcare spending [3] and increases mortality [4]. Falls also have a serious psychological impact that results in fear of falling [5], and decrease in activity [5], participation, and quality of life [4]. The high prevalence and serious consequences of falls highlight the importance in detecting age-related mechanisms responsible for the increased falling rates. Understanding these underlying mechanisms is fundamental for the development of adequate intervention plans to reduce these risks.

One approach for exploring fall-related mechanisms is by comparison of gait characteristics between older adults who experienced falls (i.e., fallers) with those who didn't (i.e., non-fallers). Since the majority of older adults' falls occur while walking [6,7] and because gait balance disorders were identified as one of the main causes for these falls [8], many studies examined fallers vs. non-fallers differences during walking. Compared to non-fallers, fallers show lower knee-flexor muscle strength and swing leg clearance [9], decreased ability to walk at faster velocities, smaller stride lengths and hip extension during push-off, and increased stride frequency [10] and stride-to-stride variability [10,11].

It was suggested that some of the age-related gait pattern changes, such as alteration in stride spatiotemporal characteristics, may be associated with inadequate pelvic and thorax motions, since these motions, and their kinematic adaptations to different walking conditions, were found to deteriorate with age [12].

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However, changes in these parameters between faller and non-faller older adults are still not well understood.

Specific patterns of thorax and pelvis rotations are required for gait stability [13], for efficient energy expenditure, and for control of vertical (i.e., by pelvic-list and pelvic-transverse rotations) displacement of the center of mass (COM) [14]. Moreover, the pelvic transverse rotational momentum is attenuated by the thorax counter rotation (i.e., their combination form the trunk rotation), resulting in a smoother gait [15]. Difficulty in controlling trunk stability is associated with an increased risk of falls [16]. Consequently, altered pelvic and thorax movements may result in impaired and less stable gait pattern, and therefore contribute to a higher susceptibility to falls.

We aimed to compare gait kinematics, especially pelvis-thorax rotations, between older adults with and without a fall history. Altering walking speed conditions requires a different motor control of gait to maintain stability [10,12,17]. We therefore observed the pattern of behavior of these measures in terms of flexibility and stability during changing walking speeds. Gait flexibility was defined as the ability to adapt to a different gait kinematics after a velocity change and to make a transition between different walking speed conditions, operationalized as the measure's mean value during the strides within a specific walking speed. Stability was defined as the low stride-to-stride variability within a specific walking speed, operationalized as the mean standard deviation between strides [10,12,17]. We hypothesized that fallers would display reduced flexibility, i.e., will be less able to make a transition between different walking speeds. We also predicted that, compared to non-fallers, older adult fallers will display: (1) reduced pelvic and thorax total range of motion (tROM) in the transverse and frontal planes, (2) reduced arm and leg tROMs in the sagittal plane, (3) reduced stride lengths and increased step width, and (4) greater variability in kinematic measures (i.e., hypo-stability).

2. Methods

2.1. Subjects

Fifty-eight community-dwelling older adults were recruited for a laboratory explorative cross-sectional study. Our sample size estimation was based on Barak et al. [10] that showed a 3.6° difference in hip extension with a standard deviation of 3.3° at 0.63 meters/second (m/s) walking speed condition. Using the above numbers, for a two-sided estimate at a significance level of 0.05 and 80% power, a minimum of 14 subjects were required for each group (fallers vs. non-fallers). Subjects were included in the study if they were: over the age of 70; independently ambulatory without assistive devices; and received approval for participation by a physician. Subjects were excluded if they had: a history of total hip or knee replacement; Mini-Mental Score ≤24; severe visual impairment, cardiovascular, respiratory, or neurological diseases; active cancer. The study was approved by the Helsinki Committee of the Barzilai University Medical Center, Ashkelon, Israel (ClinicalTrials.gov Registration number #NCT01439451). Subjects signed an informed consent prior to participation.

2.2. Study protocol

The experimental set up was performed as previously described [12]. Subjects walked on a treadmill without handrails, wearing their own walking shoes, with their hands free to swing. Prior to data collection, a familiarization period on the treadmill of four to seven minutes was performed for each subject. The subjects wore a loose safety harness that could prevent a fall, but allowed them to walk comfortably without suspension. The instructions were: "Walk as naturally as possible at your preferred stride frequency". The treadmill's velocity was systematically increased, with increments of 0.1 m/s, from 0.5 m/s to 0.9 m/s, and then similarly decreased. Each walking speed condition was maintained for 35–40 s, composed of 5–10 s for acclimation and 30 s of motion data collection. If the subject felt unsafe during one of the walking conditions, the treadmill's velocity was decreased, and data for the walking speed where the subject felt unsafe wasn't included in data analysis.

2.3. Instrumentation and outcome measures

Three-dimensional (3D) kinematic data was collected using the Ariel Performance Analysis System (APAS, Ariel Dynamics Inc., CA, USA). The APAS was shown to be valid and reliable, with a system mean point estimate error less than 3.5 mm, 1.4 mm mean linear error, and 0.26° mean angular error [18]. Two video cameras that were placed approximately 7 m in front of the treadmill, at an angle of 45° to one another, recorded the motion of 8 reflective markers attached bilaterally to the subjects' midline of the anterior aspect of ankle joints, the Anterior Superior Iliac Spines (ASIS), the shoulder acromion processes, and the radial styloid processes. The marker locations were sampled simultaneously by the two cameras at a frequency of 60 Hz, and the videos from both were mapped onto a 3D coordinate system using an internal direct linear transformation algorithm. This data was grabbed, digitized, transformed, and low-pass filter smoothed (Butterworth second-order forward and backward passes) with a cut-off frequency of 5 Hz. The coordinate system for the purpose of data collection and analysis was defined by three coordinates: the X-axis represented the anterior-posterior direction, the Y-axis was vertical, and the Z-axis medio-lateral (Fig. 1(1)).

Gait kinematic parameters were calculated using Matlab (Math Works Inc. Cambridge, MA, USA). Algorithms identified the toe-off and heel strike events within the gait cycle by using the ankle markers, temporal data from the X, Y, and Z coordinates, and by visual inspection of the video. The toe-off event was identified by the peak vertical velocity of the ankle marker, and the heel strike when the ankle reached its most forward value. Identification of these events enabled the calculation of: stride time and length, defined respectively as the time in seconds and distance in meters from the toe-off until the subsequent toe-off of the same leg and step width, measured during the double support by the distance in centimeters between the ankles in the Z-axis.

For calculation of arm and leg angular tROM in the sagittal plane, maximal (i.e., when the arm/leg reached its most flexed position) and minimal (i.e., most extended position) angles were calculated between a frontal reference vector and the vector between the shoulder acromion processes and radial styloid process markers for the arms, and the vector between the ASIS and ankle markers for the legs (Fig. 1(2)). A tROM was calculated by computation of maximal and minimal angles (peal-to-peak movement) for each gait cycle. Similarly, pelvic and thorax tROM in the transverse and frontal planes were calculated as peak-to-peak movement between the vector between the ASIS and shoulder acromion processes markers, respectively, and the body's central axis (for the upward-downward movement) (Fig. 1(3)), and the frontal reference vector (for the transverse rotation) (Fig. 1(4)). After adjustment of thorax and pelvic angles in the transverse plane to the same quarter, trunk rotation tROM was obtained by subtraction of the adjusted pelvic and thorax angles (Fig. 1(5)).

Secondary outcome measures included the: 1) Self-reported Fall Efficacy Scale-International (FES-I) [19] in which subjects rated their fear of falling on a scale from 1 to 4 for 16 tasks, with 1 being confident and 4 being very concerned; 2) Performance-Oriented

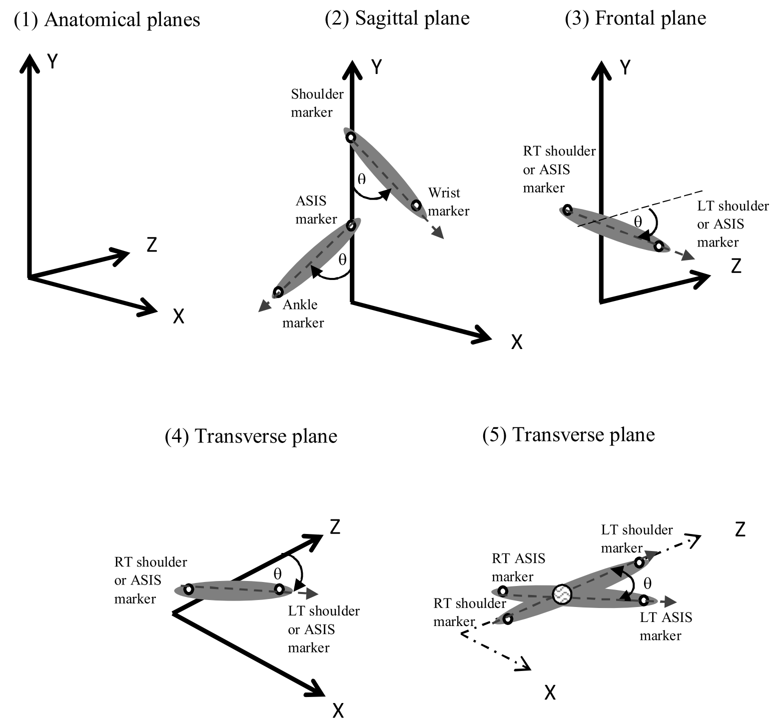

Fig. 1. (1) Definition of a three-dimensional coordinate system, i.e., the X-axis represents the anterior-posterior direction, the Y-axis represents the vertical direction, and the Z-axis represents the medio-lateral direction; (2) Definition of the arm and leg angles in the sagittal plane (Y, X). i.e., 'θ' between the shoulder and wrist markers represents the arm angle when the arm is flexed, and 'θ' between the ASIS and ankle markers represents the leg angle when the leg is extended; (3) Definition of the pelvic and thorax angles 'θ' in the frontal plane (Y, Z), the dashed line is parallel to the Z-axis and represents the center axis of the body; (4) Definition of the pelvic and thorax angles 'θ' in the transverse plane (Z, X); (5) Definition of the pelvic minus shoulder (i.e., trunk rotation) angle 'θ' in the transverse plane (Z, X), the central circle represents the central vertical axis of the body (Y).

Mobility Assessment (POMA) [20], a task-oriented test that measures an older adult's gait and balance abilities. Higher score indicates higher levels of balance function; 3) Mini-Mental State examination (MMSE) [21], a tool for assessment of mental status. Score lower than 24 indicate mild cognitive disorder; 4) Subjects' retrospective recall of fall events during the past year; 5) "Preferred treadmill speed" defined as the midpoint between the speed reported as "fast" and "slow" during increasing and decreasing treadmill's velocity respectively.

2.4. Statistical analysis

PASW statistics version 18.0 was used for statistical analysis (Somers, NY, USA). For every dependent variable, group mean values and SDs were averaged from each subject's sum of 25–30 strides at each walking speed. A 2×5 way ANOVA with repeated measures was performed to examine the effects of fall history (fallers vs. non-fallers), five increasing walking speeds (0.5-0.6-0.7-0.8-0.9 m/s), and the interaction effect between them for each dependent variable. Only subjects who completed all five walking speed conditions were included for the purpose of this analysis. In case of a significant effect of walking speed, two additional one-way ANOVA analyses for speed conditions as the between group factor were implemented for each dependent variable, one for fallers and one for non-fallers separately. These analyses included all successful trials of all subjects. The purpose of these analyses was to examine the flexibility behavior of each group's separately. In case of significance, post hoc analysis (LSD) was carried out to determine the within-groups significant differences between the walking speeds. Finally, clinical and gait characteristics of subjects who completed the examination in all walking speeds vs. subjects who didn't in each group (fallers and non-fallers), using Mann-Whitney and independent *t*-test analyses. The level of statistical significance was set at $p \leq 0.05$.

3. Results

Out of the 58 subjects, 21 reported at least one fall within a year prior to the study (i.e., fallers) and 37 reported that they didn't fall (i.e., non-fallers). Fallers were significantly younger (77.88 ± 5.16) than non-fallers (80.75 ± 5.25). No other significant differences were found for the baseline characteristics between the groups (Table 1A). Fourteen of the fallers and 27 of the non-fallers were able to walk at all walking speeds (i.e. completers). Five fallers and 8 non-fallers that were unable to walk one of highest treadmills walking speeds (i.e., 0.8–0.9 m/s) were defined as non-completers. Two fallers and two non-fallers completed all walking speeds but lacked only the slowest walking speed (0.5m/s) due to technical problem, were defined as completers for group comparisons. Table 1B shows that within the fallers, completers had significantly lower FES-I and higher POMA score compare with non-completers (23.69 ± 10.94 vs. 25.52 ± 11.15, $p = 0.028$ and 27.13 ± 0.99 vs. 24.4 ± 2.79, $p = 0.023$, respectively). There were no differences between completers and non-completers in gait kinematics at all walking speeds (For example, the preferred walking speed in Table 1B). Within the non-fallers, completers reported significantly higher preferred walking speed (0.6 ± 0.03 vs. 0.53 ± 0.03, $p < 0.001$).

A 2×5 way ANOVA didn't reveal a significant main effect of a fall history or an interaction effect of a fall history and walking speed for any of the dependent variables. However, trend to significant interaction effect was found for arm tROM ($p = 0.098$). Non-fallers displayed a larger gradual increase for their arm tROM over the five increasing walking speeds (17.1°–25.6°), compared

Table 1
(A) Characteristics of subjects who participated in the study and group comparisons (mean ± SD). (B) Group comparisons between subjects who completed all walking speed conditions (i.e. completers) and subjects who did not (i.e. non-completers) by fall status. Comparisons of kinematic measures between the groups were performed at the subjects' preferred walking speed condition.

A. Characteristic[a]	Fallers (n = 21)	Non-fallers (n = 37)	p-value
Age (years)	77.88 ± 5.16	80.75 ± 5.25	p = 0.049*
Gender (F%/M)	17 (81%)/4	26 (70.3%)/11	p = 0.372
Height (cm)	158.14 ± 8.38	158.91 ± 9.87	p = 0.765
Weight (kg)	68.71 ± 12.80	69.41 ± 18.89	p = 0.882
BMI(cm/m^2)	27.35 ± 3.66	26.47 ± 4.32	p = 0.442
FES-I	25.52 ± 11.15	20.59 ± 5.16	p = 0.067
POMA	26.45 ± 1.96	26.68 ± 1.37	p = 0.614
MMSE	29 ± 1.45	28.54 ± 1.62	p = 0.305
Number of falls	1.52 ± 0.87	0.00	
Preferred walking speed (m/s)	0.58 ± 0.03	0.58 ± 0.04	P = 0.759

B. Characteristic[b]	Completers (n = 16)	Non-completers (n = 5)	p-value	Completers (n = 29)	Non-completers (n = 8)	p-value
Age (years)	79.1 ± 5.3	74.1 ± 1.8	p = 0.005**	79.5 ± 4.9	85.2 ± 3.9	p = 0.005**
Height (cm)	157.6 ± 7.3	159.8 ± 12.1	p = 0.625	158.6 ± 10.1	160 ± 9.5	p = 0.731
Weight (kg)	68.6 ± 12.4	69 ± 15.6	p = 0.934	68.9 ± 20.7	71.2 ± 9.7	p = 0.277
BMI(cm/m^2)	27.5 ± 3.9	26.7 ± 2.6	p = 0.68	25.9 ± 4.2	28.6 ± 4.1	p = 0.077
FES-I	23.7 ± 10.9	25.5 ± 11.1	p = 0.028*	20.7 ± 5.6	18.9 ± 3.3	p = 0.752
POMA	27.1 ± 1	24.4 ± 2.7	p = 0.023*	26.6 ± 1.4	26.9 ± 1	p = 0.832
MMSE	29.1 ± 1.4	31.4 ± 10.7	p = 0.298	28.7 ± 1.5	27.9 ± 1.8	p = 0.186
Number of falls	1.6 ± 0.9	1.2 ± 0.4	p = 0.401	0.00	0.00	
Preferred walking speed (m/s)	0.59 ± 0.03	0.56 ± 0.03	p = 0.14	0.6 ± 0.03	0.53 ± 0.03	p < 0.001***

Abbreviations: cm = centimeters; Kg = kilogram; cm/m^2 = centimeters/meter^2; m/s = meters/seconds; F = female, M = male; BMI – Body Mass Index; FES-I = Falls Efficacy Scale-International; POMA = Performance Oriented Mobility Assessment; MMSE = Mini Mental State Examination.

[a] p-value compares baselines means between completers and non-completers in the two groups based on the *t*-test and chi-square.

[b] p-value compares characteristic means between completers and non-completers in the two groups based on the *T*-Test for the age and height variables and on the Mann-Whitney for the remaining variables due to non-normal distribution of the data as examined by the shapiro-wilk statistics.

* Significance at $p \leq 0.05$.

** Significance at $p \leq 0.005$.

*** Significance at $p \leq 0.001$.

with fallers (14.5°–19.8°). A significant main effect of walking speed was found for stride time and stride length ($p < 0.001$), for SDs of stride time ($p = 0.003$), length ($p = 0.03$), and width ($p = 0.005$), for leg and arm sagittal trunk rotations, and for thorax frontal ($p < 0.001$) and transverse planes tROM ($p = 0.03$).

3.1. Stride time, stride length, and leg tROM

One-way ANOVA for speed conditions as the between-group factor showed significant main effects for stride time, length, and leg tROM for both fallers ($p = 0.004$, $p < 0.001$) and non-fallers ($p < 0.001$) (Fig. 2). Visual inspection revealed an almost linear decrease in stride time and increase in stride length and leg tROM from 0.5^{i} to 0.9 m/s (increasing walking speed: superscripted 'i') and then a gradual increase in stride time (Fig. 2(1)) and decrease in stride length (Fig. 2(2)) and legs' tROM (Fig. 2(3)) from 0.9 to 0.5^{d} m/s (decreasing walking speed: superscripted 'd'). Post hoc analysis revealed significantly shorter stride time at 0.9 m/s compared with walking speed conditions of 0.5^{i,d} m/s among the non-fallers ($p < 0.05$). Stride length was significantly longer at 0.9 m/s than walking speed conditions 0.7^{i,d} m/s and slower for fallers ($p < 0.01$) and non-fallers ($p < 0.001$). Leg tROM was significantly lower for 0.5 m/s compared to 0.7^{i} m/s and 0.6^{d} m/s and higher, and for 0.6 m/s compared to 0.8^{i}m/s and 0.7^{d} m/s and higher respectively for both fallers ($p < 0.01$, $p < 0.05$) and non-fallers ($p < 0.001$, $p < 0.005$).

3.2. Arm sagittal plane tROM

One-way ANOVA showed a significant main effect of walking speeds only for the non-fallers ($p = 0.002$) (Fig. 3(1)), with a significantly higher tROM at 0.9 m/s compared with 0.5^{i}m/s speed condition at post hoc analysis ($p = 0.011$). Visual inspection revealed an almost linear increase from 0.5^{i} to 0.9 m/s and then a decrease from 0.9 to 0.5^{d}m/s for the non-fallers (Fig. 3(1A)), while for the fallers the increase and decrease were minimal (Fig. 3(1B)).

3.3. Trunk rotation tROM

One-way ANOVA revealed a significant main effect of walking speed for the non-fallers group ($p = 0.001$), but not for the fallers (Fig. 3(2)). Visual inspection revealed an almost linear increase in tROM from 0.5^{i} to 0.9 m/s and then a decrease from 0.9 to 0.5^{d} m/s for the non-fallers group (Fig. 3(2A)), while for the fallers the increase and decrease were moderate (Fig. 3(2B)). "Post hoc analysis for the non-fallers group showed significantly higher trunk rotations at 0.8^{i}, 0.8^{d}, and 0.9 m/s compared with 0.5^{i} ($p = 0.034$, $p = 0.05$, and $p = 0.001$, respectively) and between 0.9 and 0.7^{i} m/s ($p = 0.033$).

3.4. Pelvic and thorax transverse and frontal plane tROM

One-way ANOVA didn't reveal significant main effects of speed for both groups (Fig. 3(3–6)).

4. Discussion

Walking speed had an effect on gait parameters, supporting reports of velocity being a control parameter of gait [10,12,17]. By manipulating the walking speed, we identified different patterns of behavior for arms and trunk motions for fallers and non-fallers, suggesting changes in the control of gait.

Utilization of flexibility measures was suggested for detection of an increased risk for falling, as the need to adapt to task changes may be in conflict with the ability of a person to execute these changes, resulting in a less stable walking pattern, and a higher

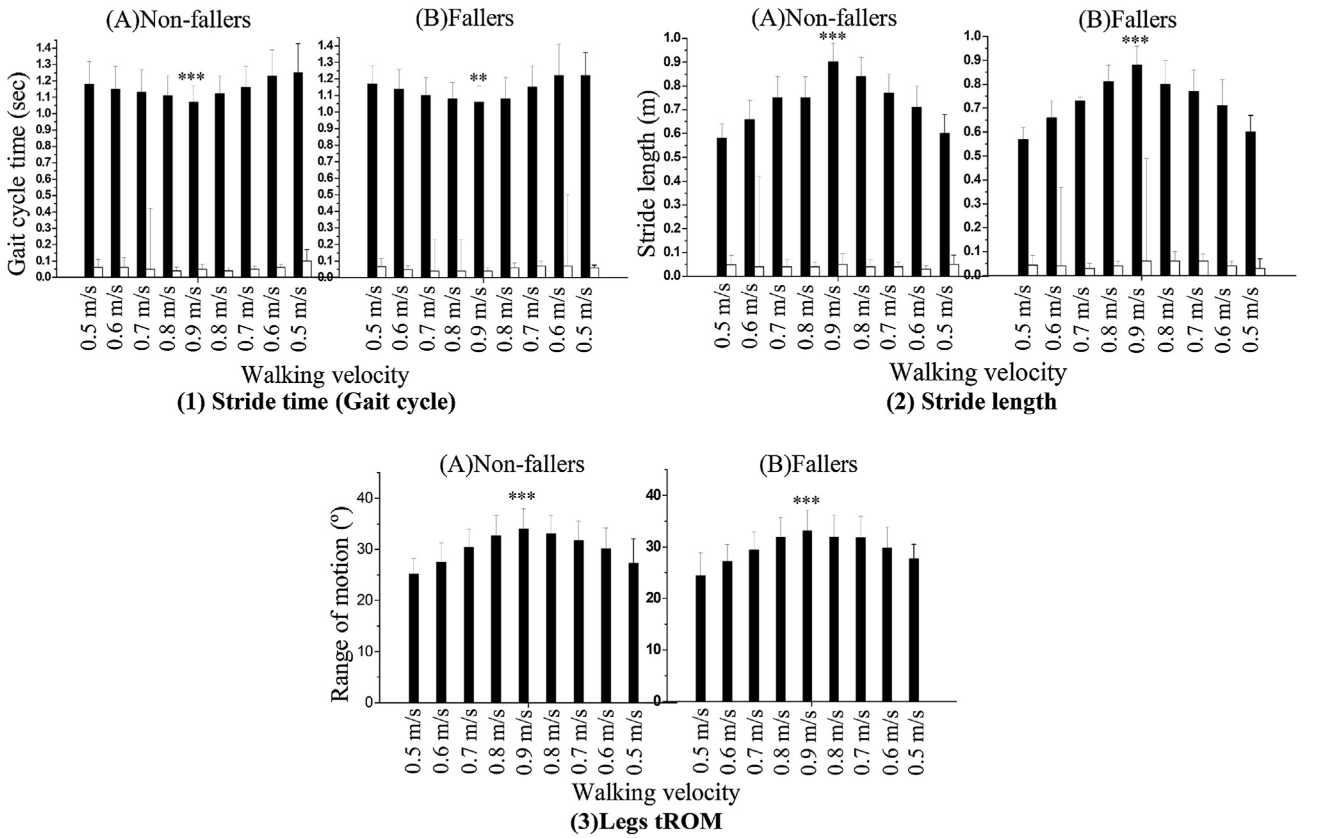

Fig. 2. (1) Stride time in seconds, (2) stride length in meters, and (3) leg sagittal tROM in degrees and its standard deviations for non-fallers (panels A) and fallers (panels B) walking on a treadmill with increasing (i.e., 0.5–0.9 m/s) and decreasing (i.e., 0.9–0.5 m/s) walking speed conditions. Both groups displayed a gradual transition between velocities. When walking speed was increased both groups decreased their stride time (panel 1) and increased stride length (panel 2) and leg tROM (panel 3), and then increased their stride time (panel 1) and decreased stride length (panel 2) and leg tROM (panel 3) when velocity was increased. P-values (*$p < 0.05$, **$p < 0.005$, ***$p < 0.001$) represent significance in the one-way ANOVA model analyzing the effect of walking speed on the measures for each group in each panel.

probability for falling [22]. Our study revealed that with increasing walking speeds, non-fallers increased their arm swing movements more substantially than the fallers. Moreover, for the arm and trunk rotation tROM the non-fallers were able to modify their movement patterns to the changing walking speeds (i.e., transition behavior). They increased their arms' swing movements and trunk rotations the more the velocity increased and then gradually decreased them with the decreasing walking speed. In contrast, the fallers had a more rigid behavior (i.e., decreased flexibility). Fallers showed a decreased ability to adapt to a new kinematic pattern and a decreased transition ability, as their change in motions with changing walking speeds was minimal. Arm swing movements are known to reduce energy consumption, balance the body's angular momentum, and contribute to a more stable walking pattern by reducing the lateral displacement of the COM [23]. The pelvis transverse rotations during walking are attenuated by the counter-rotation of the thorax, resulting in a smoother gait and in a decreased angular momentum [15]. The decreased axial momentum might improve frontal plane balance due to a decreased necessity for transverse corrective torques around the hip [24]. Decline in medio-lateral stability was associated with falls in older adults [25], and significantly improved medio-lateral trunk stability was found associated with emphasized arm swings for community dwelling older adults [26]. In summary, manipulation of walking velocity exposed fallers who have a rigid trend of behavior for arms and trunk motions, which may result in unstable walking [23]. This might be related to the fact that fallers who were unable to walk at all velocities scored lower at the POMA and higher in FES-I tests than fallers who able to walk all walking speeds. This might indicate that faller older adults with worse balance measures have difficulties walking at faster speeds.

For thorax and pelvis kinematics, we found that both groups displayed a relatively similar movement pattern across walking speeds, different from the more flexible pattern observed for young adults [12,27]. The gait changes that occur with aging were suggested to be associated with changes in the central nervous system (CNS) [28] and with deterioration of the musculoskeletal function [29]. The spatial characteristics of gait were described to be less stable at higher walking speeds, in which the segmental momentum is increased, resulting in a limited time and decreased ability of the CNS to efficiently attenuate kinematic disturbances and control errors [30]. Consequently, we speculate that the reduced ability to change the arm-trunk movement pattern in response to altering walking speeds, may serve the fallers as an attempt to compensate for the impaired ability of the CNS to control the segmental kinematics by decreasing the degrees of freedom. Nevertheless, as daily life is very dynamic, requiring constant change of behavior in response to changes in task requirement, impaired flexibility may contribute to an increased risk for falling due to an inability to efficiently respond and recover from unexpected situations.

In contrast to our hypotheses, our results showed no differences between the fallers and non-fallers for strides' characteristics, i.e., arm, leg, pelvis, and thorax angular motions, and for the stability of measures. This might be attributed to the lack of actual differences between the groups for the evaluated variables or to the study's methodology and limitations. We performed a kinematic evaluation using a protocol that identified differences between

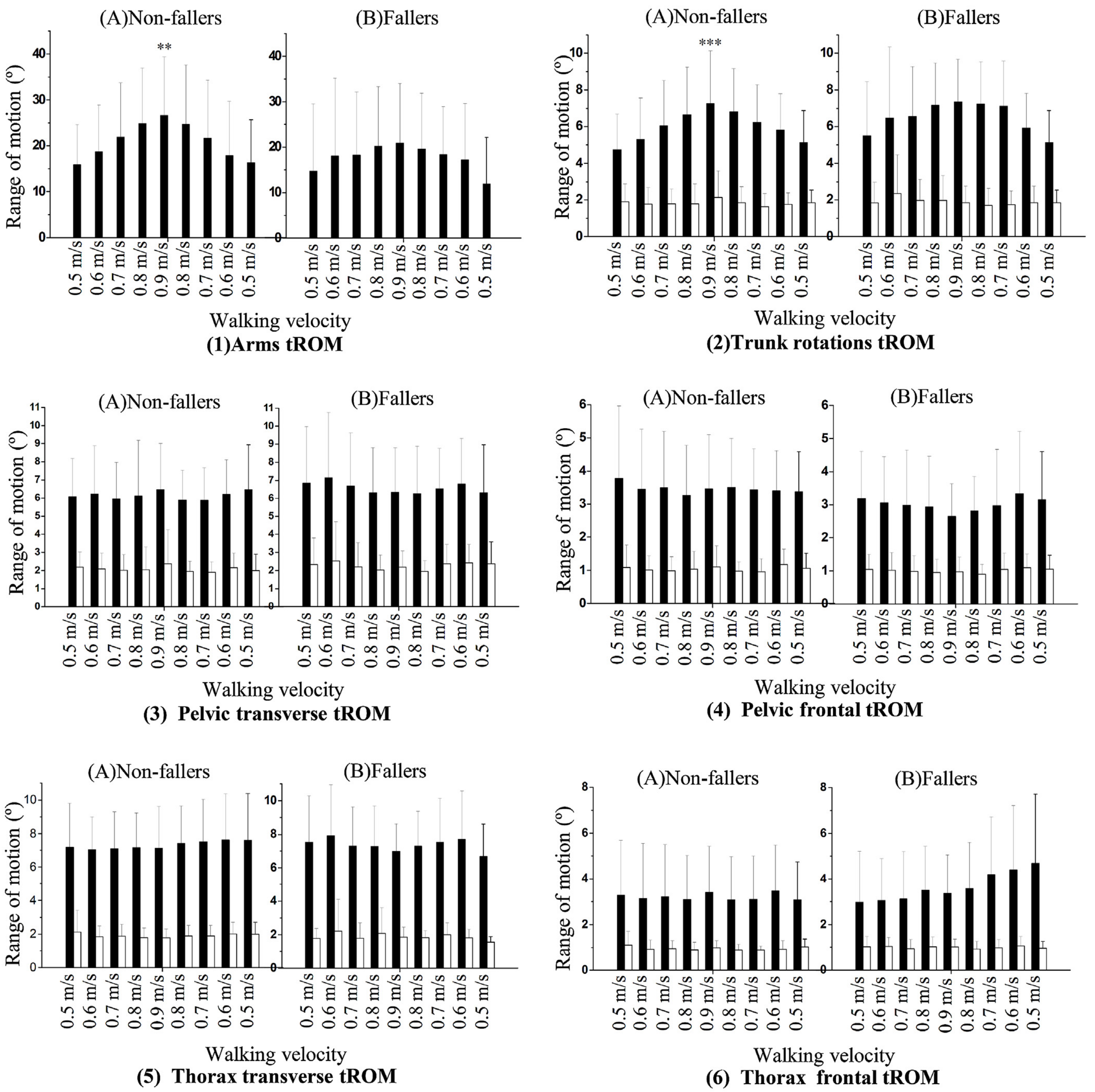

Fig. 3. (1) Arm tROM in the sagittal plane, (2) trunk rotations tROM in the transverse plane, (3) pelvic and (4) thorax tROM in the transverse plane, and (5) pelvic and (6) thorax tROM in the frontal plane in degrees at increasing (i.e., 0.5–0.9 m/s) and decreasing (i.e., 0.9–0.5 m/s) walking speeds, and their standard deviations. For arm and trunk motions, non-fallers (A) displayed a gradual transition of their arm and trunk movements, i.e., flexible behavior (panels 1A and 2A), whereas fallers (B) had a more rigid pattern of behavior, i.e., no transition behavior, as their arm and trunk movements' change was minimal with the changing velocities (panels 1B and 2B). For pelvic and thorax tROM in the transverse (panels 3 and 4) and frontal (panels 5 and 6) planes, both groups displayed a relatively similar pattern of behavior across all walking speeds. P-values ($^{*}p < 0.05$, $^{**}p < 0.005$, $^{***}p < 0.001$) represent significance in the one-way ANOVA model analyzing the effect of walking speed on the measures for each group in each panel.

young and old adults [12]. However, it is possible that the measures used in this protocol are not optimal for differentiating fallers from non-fallers. For instance, as all gait cycles within each walking speed were averaged, long-term correlations between the cycles were neglected. These correlations represent the feedback loops in the motor control of gait [31,32], and were found to appear in a more random manner, associated with an unstable walking pattern among faller compared to non-faller older adults [33]. On the other hand, not recalling minor falls due to the retrospective nature of the study, or a single coincidental fall or exposure of the fallers to more balance-challenging situations that might result in a fall due to their younger age, all might have interfered with our division to groups, and therefore affected our results. Most importantly, subjects from both groups had relatively good dynamic balance, as most subject at both groups were able to complete a challenging study protocol, had low self-reported fear of falling, and both received a score higher than the cut-off point indicative for increased risk for falls in the Performance Oriented Mobility Assessment, with no significant difference between the groups. This restricts the generalization of our results to the heterogenic older adult population. The high variability for the measures' values between the subjects, together with the relatively small sample size, might have decreased the power of the study, and therefore our ability to identify smaller, yet existing differences. Results should be treated in caution since multiple statistical comparisons were performed and false positive results might appear.

In conclusion, our findings suggest that only non-fallers demonstrated the ability to adapt trunk and arm ROM to treadmill speed i.e., had a more flexible pattern of behavior, and supporting reports of the upper body's role in maintaining balance during gait. Future research should address better understanding the mechanisms resulting in the impaired flexibility pattern, and its effect on dynamic stability between faller and non-faller older adults.

Conflict of interest statement

The authors have no conflict of interest to declare.

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