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AGE-RELATED CHANGES IN UPPER BODY CONTRIBUTION TO BRAKING FORWARD LOCOMOTION IN WOMEN

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Abstract

INTRODUCTION: Gait termination is a transitory task that requires the lower body to produce braking forces and inhibit forward propulsion. However, it is still unknown whether the upper body plays an active role in braking of gait and whether this mechanism is impaired with ageing.

RESEARCH QUESTION: Do older women exhibit an impaired control of upper body segments during gait termination with respect to young women?

METHODS: Ten young and 10 older women performed three gait termination trials at comfortable speed while fixing the gaze on a visual target. A 3D motion analysis system was used to measure head, trunk and pelvis angular displacement and velocity, and estimate neck, waist and hip moments through Plug-in Gait modeling. Cross-correlation analysis of kinematic waveforms between paired adjacent segments (head-trunk and trunk-pelvis) was performed to investigate upper body coordination. Surface EMG activity of erector spinae (L3), sternocleidomastoid and neck extensor muscles was recorded. Statistics was carried out by MANOVA.

RESULTS: Older participants exhibited delayed peak extensor torques of neck, waist and hip compared to young participants, along with lower progression speed. Both groups showed a slight flexion of the trunk counteracted by a backward tilt of head and pelvis during braking. In addition, older women displayed a peculiar upper body coordination pattern, with the head coupling with trunk motion, as shown by cross-correlation. Older women displayed shorter lumbar erector spinae onset latency relative to last heel contact than young (16±68 ms vs 92±37 ms).

SIGNIFICANCE: The upper body plays an active role in the braking of gait and this mechanism is impaired in older women. Moreover, the age-related coupling of head and trunk motion may produce an unbalancing effect on whole-body stability during the braking mechanism, thus leading to a higher risk of falls.

Keywords: gait termination; head stability; elderly; upper body; locomotion

Introduction

The main role of upper body movement during locomotion is to contribute to whole-body balance and head stabilisation [1,2]. Coordination between the head, trunk and pelvis motion is fundamental to reduce the unbalancing effect generated by the lower limb movement during walking [3,4]. Previous findings showed that older individuals have an altered control of upper body segments, including flexed trunk posture and coupling of upper body segment motion in the same direction, which are less efficient in preserving balance during various locomotor tasks [5,6]. In particular, older adults reported an impaired ability to control angular velocity of the trunk segment, that is known to play a key role in avoiding fall after a trip or slip [7]. Consequently, these age-related changes in upper body behaviour could increase the risk of falling while performing locomotor transitions, such as gait termination, which require more complex interaction between neural and biomechanical factors compared to steady-state walking [8,9]. However, it is still unknown whether similar age-related modification of upper body coordination occurs also during the termination of gait.

Gait termination is the transient period between steady-state gait and a quasi-static standing position that requires to stop the forward momentum of the locomotor pattern [10]. Older individuals have been reported to recruit fewer lower limb muscles, with greater delay, compared to young individuals, thus producing less extensor torque necessary to terminate gait [11,12]. Moreover, older women showed longer response time and distance than older men when terminating gait in response to an unexpected, external stimulus [13]. Although the stopping strategy depends on sensorimotor integration, motor planning and execution in both planned and unplanned gait termination [14,15], previous research has shown that the age-related slowing in walking behaviour is linked to slower information processing [16], thus suggesting that whole-body balance control might be altered during planned gait termination compared to sudden stopping.

Whether or not gait termination is suddenly required or pre-planned, dynamic balance is maintained by keeping the whole-body centre of mass (COM) within the base of support boundaries defined by foot placement [17]. The upper body represents 2/3 of the whole body mass and therefore its motion has a fundamental role in controlling whole-body COM position during gait termination. Noteworthy, older women have been shown to poorly dampen acceleration across upper body segments during planned gait termination at different walking velocities compared to young women [18]. However, the peculiar changes in upper body coordination and its contribution to the braking mechanism during planned gait termination have not been identified yet.

The aim of the study was to investigate upper body mechanics and muscle activity in young and older women during planned gait termination. A more variable and less coordinated motion of upper body segments with lower extensor torque was hypothesised to arise in older compared to young women.

Methods

Participants

Ten healthy young women from university community (age: 23.1 ± 1.1 years; height: 1.66 ± 0.06 m; body mass: 56.9 ± 6.6 kg) and ten healthy and independent community-dwelling older women (age: 73.8 ± 2.4 years; height: 1.60 ± 0.06 m; body mass: 62.1 ± 13.6 kg) took part in the study after being informed and signing an informed consent form approved by the institution ethics committee. Independent t-test showed no differences in both height and body mass between groups (p =0.14). The eligibility of subjects was determined through a health status questionnaire during the recruitment period [19]. Participants with any history of neurological and/or orthopaedic disturb that could impair gait or balance were excluded from the study.

Experimental procedure

Subjects walked at their self-selected comfortable speed and were asked to stop and stand still for at least 3 s with both feet parallel on a pre-set target area (length 60 cm; width 40 cm), which was located in the floor in the middle of a 10 m walkway. Participants were asked to maintain the gaze fixed on a target positioned 1 meter after the stopping area at eye level. The dominant limb was determined by asking the participants which leg they use to kick a ball and all participants in this study were right-leg dominant. Therefore, three familiarisation trials were performed before the data collection to ensure the subject approached the target area with the right foot. During the data collection, three valid trials were obtained and used for the analysis [20].

Thirty-five retro-reflective markers were located on the subject skin and their 3D position was reconstructed through a seven-camera motion capture system (Vicon MX3, Oxford, UK) at a sampling rate of 100 samples/s. The whole-body COM and kinematics of head, upper trunk and pelvis segments were calculated based on a 15-body segments 3D model (Plug-in Gait, Vicon). The sagittal neck, waist and right hip torques that acted on head, trunk and about the right hip, respectively, were obtained through inverse dynamic solution from the same biomechanical model [21]. The relevant measured and estimated data were filtered using a second order low-pass Butterworth filter with a cut-off frequency of 5 Hz.

A surface EMG device (PocketEMG, BTS Bioengineering, Italy) was used to record upper body muscle activity at a sampling frequency of 1000 Hz. EMG activity was recorded on left side, that is contralateral to the braking leg, using bipolar disposable electrodes (Ag/AgCl, 1 cm disc-electrodes, 2 cm inter-electrode distance) from the sternocleidomastoid (SCM), neck extensor (NE) and erector spinae muscles at L3 level (ESL). Electrodes were positioned 2 cm lateral of the spinal processes of C4 and L3 vertebrae for NE and ESL, respectively. For the SCM, electrodes were placed half the distance between the mastoid process and the sternal notch. To reduce the electrode-skin impedance and to increase signal quality, the skin was gently abraded, cleansed and rubbed with a conductive paste before electrode placement.

Data Analysis

Since participants were asked to terminate gait onto a known ground location (thus defining the planned nature of the task), each gait termination was divided into three phases: Approaching, Braking and Stabilisation phases [18]. The Approaching phase consisted of the last complete stride of the right leg, with the last right heel contact (RHC2) hitting the target area. The Braking phase consisted of the last step that brought both feet parallel on the target area (from RHC2 to the last left heel contact). The Stabilisation phase lasted until the COM progression speed was less than 0.05 m/s and a full stopping position was reached. The progression speed was estimated from the AP COM speed, which was calculated over the entire task and time-normalised to yield an ensemble average for each group (100 points). The duration of each phase as well as the total duration were calculated and expressed in absolute terms (ms). Then, the duration of each phase was expressed as percentage of the total task duration to investigate the stopping strategy of each group.

To evaluate postural stabilisation at the end of the planned gait termination, mean AP COM velocity and position in the Stabilisation phase were normalised by body height and foot length, respectively, and compared with the boundaries of stability region as modelled by Pai and Patton [22].

Maximum angular excursion, mean flexion/extension angle and angular velocity of head, trunk and pelvis segments along the sagittal plane were calculated in each phase. To investigate the net extensor/flexion torque acting on the upper body segments during the braking, the peak torques of neck (head-trunk), waist (trunk-pelvis) and hip of the supporting leg were calculated about the first heel contact with the target area (RHC2, beginning of the Braking phase) and normalised by the total body mass. The latency of the peak with respect to the RHC2 was also obtained. A cross-correlation analysis between the angular displacement waveforms of paired segments (head-trunk and trunk-pelvis) was used to evaluate the coordination between upper body segments during Braking and Stabilisation phases. Specifically, the cross-correlation function provides the time lag between the paired segment motion as well as a negative or positive coefficient ($-1 < r < +1$) that refers to paired segments movement in the opposite or same direction, respectively.

The raw EMG signal was firstly bandpass-filtered with a 4th order Butterworth filter (30-450 Hz) to retain only relevant physiological information, filtered with a 2nd order high-pass Butterworth filter with a cut-off frequency of 30 Hz in order to remove possible ECG artefacts [23]. A linear envelope of the EMG signals was then obtained by full-wave rectification and low-pass filtering with a 2nd order Butterworth filter (cut-off frequency of 30 Hz). The visual muscle onset determination method was used by the same experienced experimenter to estimate the latency of the muscle activation with respect to the mechanical event of RHC2 [24]. EMG waveforms were amplitude-normalised by the mean value of the entire trial [25], time-normalised (100 points) and ensemble averaged over each group. These amplitude- and time-normalised EMG waveforms were then plotted against joint extensor torques to visually examine differences between groups in EMG activation patterns across the entire gait termination task.

Statistical Analysis

The normal distribution of data was tested using a Shapiro-Wilk test and the presence of outliers was checked through the Z-score method. To investigate the effect of age as between-subject factor, six separate MANOVAs were performed on kinematic, kinetic and EMG measures due to the large number of dependent variables. Specifically, the first MANOVA was performed on AP COM speed and duration of each phase of the task; the second, third and fourth MANOVAs were performed on sagittal angular displacement, velocity and maximum excursion; the fifth and sixth MANOVAs were performed on timing and amplitude of sagittal peak torques. If MANOVA showed statistical significance follow-up testing was performed using separate ANOVAs with Holm-Bonferroni correction. A one-way ANOVA was performed on the latency of muscle activation with respect to RHC2 mechanical event. The significance level α was set to 0.05 and all statistical tests were performed using the SPSS 23.0 software (Chicago, IL, USA). Partial eta squared (η^2_p) was used as a measure of effect size and Cohen's η^2_p benchmarks of 0.0099, 0.0588 and 0.1379 were used for small, medium and large effect size, respectively [26].

Results

MANOVA showed a significant effect of age on phase time and AP COM speed during the three phases (Wilk's $\Lambda = 0.409$, $F(6,13) = 3.134$, $p < 0.05$, $\eta^2_p = 0.591$). Specifically, older women spent more time in the Approaching phase and less time in the Stabilisation phase than young women ($F(1,18) > 7.029$, $p < 0.05$, $\eta^2_p > 0.281$; Table 1), with no difference in the total duration. The ensemble average of AP COM speed for both groups are reported in Fig.1. Both groups showed the characteristic decrease in AP COM speed during the Braking phase, with older women displaying a lower mean AP COM speed in each phase compared to young ($F(1,18) > 5.358$, $p < 0.05$, $\eta^2_p > 0.229$; Table 1).

TABLE 1 AND FIGURE 1 HERE

Figure 2 displays the AP COM velocity-position dispersion plot of both young and older women. Visual inspection indicated that 10% of trials in older women fell above the upper boundary of the stability region, whereas all trials in young women were within the stability region. In particular, 30% of trials in older women were above the limit indicated by Pai and Patton [22], with the adjusted upper boundary of the stability region accounting for the 59% reduction in strength in elderly.

FIGURE 2 HERE

Upper body kinematics

Mean angular position, velocity and maximum excursion of head, trunk and pelvis in the sagittal plane are reported in Table 2. Older and young women had similar mean angular position of the three segments across each phase, with the head and pelvis being backward tilted, and the trunk slightly flexed. MANOVA showed a significant effect of age on sagittal angular velocity of head, trunk and pelvis across the three phases (Wilk's $\Lambda = 0.142$, $F(9,10) = 6.703$, $p < 0.01$, $\eta^2_p = 0.858$). Particularly, the two groups were flexing (negative mean angular velocity) both pelvis and trunk and were extending (positive mean angular velocity) the head during the Approaching phase, with young flexing the trunk with a greater magnitude ($F(1,18) = 22.481$, $p < 0.001$, $\eta^2_p = 0.555$). No differences were found in mean angular velocity of pelvis and trunk during the Braking phase, whereas the head was extending in older women and was flexing in young ($F(1,18) = 8.491$, $p < 0.01$, $\eta^2_p = 0.321$). During the Stabilisation phase, older women were flexing again both pelvis and trunk while young were extending them ($F(1,18) > 16.404$, $p < 0.01$, $\eta^2_p > 0.477$; mean angular velocity values in Table 2), with no differences at head level. Young mostly displayed greater maximum angular excursion of the three segments than older women across each phase (MANOVA: Wilk's $\Lambda = 0.182$, $F(9,10) = 5.008$, $p < 0.05$, $\eta^2_p = 0.818$; results of post-hoc analysis are shown in Table 2).

TABLE 2 HERE

Cross-correlation analysis showed more consistency in the coordination patterns between paired segments in the young than older. Specifically, most of the young subjects adopted an opposite motion of the head with respect to both trunk (73% of cases) and an in-phase movement of trunk with respect to the pelvis (83% of cases) (Fig.2a). In contrast, older women presented both opposite and in-phase motion of the head with respect to trunk (46% and 54% of cases, respectively), but a trunk-pelvis coordination pattern similar to young (Fig.2b, 2c).

FIGURE 3 HERE

Upper body kinetics

Figure 3 shows group ensemble averages of the observed torques as well as EMG activity. Despite MANOVA indicated no significant effect of age on peak extensor torque magnitude about the RHC2 ($p > 0.05$), thus showing similar values between groups at neck and waist, there was a tendency for the right hip extensor torque peak to be lower in older than in young women (0.39 ± 0.19 $Nm \cdot kg^{-1}$ and 0.59 ± 0.20 $Nm \cdot kg^{-1}$, respectively). In contrast, the latencies of neck, waist and right hip torque peaks with respect to RHC2 were significantly greater in older compared to young (173±17, 182±31 and 138±25 ms in older and 145±13, 151±11 and 117±11 ms in young, respectively; MANOVA: Wilk's $\Lambda = 0.499$, $F(3,16) = 5.357$, $p < 0.05$, $\eta^2_p = 0.501$).

FIGURE 4 HERE

Upper body EMG

Both groups displayed a constant activity of SCM during the Approaching phase with occasional low amplitude burst not clearly related to heel contact. Young but not older women showed a decrease in SCM activity during the Braking and Stabilisation phases, as shown by the normalised EMG profiles in Fig.3. In NE, a clearer footfall-related activity was observed in young compared to older women, although the amplitude of bursts was still low. On the other hand, both groups showed bursts of ESL clearly related to ground contact, with a greater activation during the Braking phase that was prolonged to the Stabilisation phase in the older group only. Consequently, a clear visual identification of the burst onset in both groups was exclusively achievable in ESL, with older women showing lower muscle activation latency with respect to RHC2 compared to young (29±94 ms in older and 72±37 ms in young; $F(1,18) = 6.282$, $p < 0.05$, $\eta^2_p = 0.260$).

Discussion

The main result of the study was that older women showed a more variable coordination of the head with respect to both trunk and pelvis compared to young, coupled with a reduced maximal excursion of upper body segments. In addition, delayed extensor torque peak of neck, waist and supporting hip were exerted by older women during the Braking phase.

During gait termination, older women showed lower maximal angular excursion and no differences in sagittal angular position of upper body segments compared to young, with the head and pelvis being backward tilted and trunk slightly flexed in both age groups. Although the trunk had a flexed position in each phase, young showed angular velocity towards extension during the Stabilisation phase, whereas older women were still flexing the segment. The forward flexed position of the trunk is a feature of walking and its role has been suggested to assist (not impede) forward locomotion [3]. Conversely, in this study, participants were asked to terminate locomotion onto a known target area on the ground, thus probably adopting a pre-planned anticipatory behaviour which did not require a forward flexion of the trunk. In line with that, the AP COM velocity-position plot showed that persisting forward flexion in older women would produce an unbalancing effect by moving the COM closer to the age-adapted anterior boundary of the stability region as modelled by Pai and Patton [22]. This anticipatory motor behaviour of the upper body, therefore, may play an important role in balance control when a rapid stopping is required, in which fast visual processing, strategy planning and motor output are involved [13]. Indeed, elderly have been reported to adopt more frequently two-step stopping strategy when they are asked to rapidly stop in response to an unexpected external cue, wherein no anticipation can facilitate the stopping motor behaviour [11].

Stabilisation of the head is maintained through head-on-trunk and trunk-on-pelvis movement during locomotion to improve dynamic balance control and gaze stabilisation [27–29]. In this study, a visual target constraint was present and, therefore, a stable position of the head was required to facilitate gaze fixation. EMG results showed an evident muscle activity of lumbar erector spinae related to heel contacts in both groups and an occasional muscle activation with low amplitude bursts of neck muscles, especially in the SCM. Indeed, young people managed to stabilise the head through a ground impact-related muscle activity of NE and a constant SCM activity until the Stabilisation phase. Although a similar activity of erector spinae muscles has been reported in the literature [3,9], the results about neck muscle activity found in the young group were partially in contrast with those obtained by Cromwell et al. [3], showing a less random and occasional muscle activity. This different neck muscle activity behaviour might be related to the need of an active and task-specific control of head stability during gait termination that is less evident in older women. Kinematic results showed that the flexed trunk position was compensated by an extended position of the head in both groups and an oppositional movement between the two segments aimed at recovering vertical alignment (i.e. head was forward tilting while trunk was extending) was performed during the Braking and Stabilisation phases. However, cross-correlation analysis revealed that the head-trunk coordination pattern was more variable in the older than in young women as an in-phase movement of the two segments (i.e. both head and trunk were flexing) was also adopted. Such a coordination pattern suggests a rigid motion of the upper body segments and it has been indicated as a compensation for age-related decrement in lower limb muscle activity during walking and gait initiation [6,30,31]. Although it may have a functional role by facilitating forward locomotion, the head-trunk coupled motion can decrease head stability, as demonstrated by the synchronisation of head and trunk acceleration profiles during walking [30]. From a biomechanical point of view, gait termination requires to stop and not facilitate forward locomotion, thus the adoption of a single unit motion of the upper body does not comply with the task requirements, mainly resulting in a reduction of head and whole-body stability. Accordingly, older women may increase the odds of balance loss by adopting the aforementioned head-trunk coordination pattern in more challenging condition, such as at fast walking speed, resulting in an eventual fall.

Termination of gait requires the generation of an extensor torque, mainly exerted by lower limb muscles during the Braking phase, in order to stop forward locomotion [10,17]. Older adults have been reported to fail in muscle recruitment, especially gluteus medius as hip extensor, and to delay the onset of muscle activity during stopping, thus leading to a lower extensor torque production compared to young adults [12]. In line with that, the present study reported delayed extensor torque peak of supporting hip as well as neck and waist in older than young women during the Braking phase. In addition, although the multivariate approach did not show statistical significance, there was a tendency for a lower extensor torque peak of the supporting hip in older women compared to young, which agrees with the impaired gluteus medius activation during gait termination in elderly [12]. However, the delayed exertion of extensor torque at upper body level may indicate a reduced control of the braking and stabilising mechanisms in older women, as confirmed by the flexing movement of the trunk during the Stabilisation phase. Therefore, these results point out an ineffective upper body extensor torque production that, in synergy with the previously reported deficiencies in generation of lower body extensor torque, suggests an overall weakened capacity of terminating gait.

There are some limitations in the present study that are worth to acknowledge. First, the upper limb movement was not taken into account, except for the whole-body COM position computation.

It has been demonstrated that arm movement influences the COM position during stopping in response to a slip [32], therefore it may also play a role in the upper body motion during planned gait termination. Second, only female participants were included in this study, thus limiting the applicability of the present results to the overall older population. Therefore, further investigation including both upper limbs analysis and male participants are needed for a better understanding of the upper body role in gait termination.

In conclusion, this study demonstrated that older women adopted a more variable and less stable coordination of head and trunk movement compared to young women. The delay in the extensor torque peak of neck, waist and supporting hip suggested an impaired mechanism of extensor torque generation in older women compared to young while braking forward locomotion. Overall, the upper body exhibited a lower contribution to the braking mechanism and even an unbalancing effect in older women, thus indicating the importance of a major upper body movement control for safeguarding elderly from the occurrence of balance loss and falling.

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TABLES

Table 1.
Mean and standard deviation of gait termination spatio-temporal parameters during Approaching, Braking and Stabilisation phases. * = significantly different from young (ANOVA: *p* <0.05)

	Approaching Phase		Braking Phase		Stabilisation phase	
	Older	**Young**	**Older**	**Young**	**Older**	**Young**
Mean AP COM speed (m/s)	0.91 ± 0.23*	1.34 ± 0.18	0.60 ± 0.12*	0.79 ± 0.09	0.10 ± 0.02*	0.13 ± 0.02
Phase duration (s)	1.32 ± 0.24*	1.10 ± 0.09	0.60 ± 0.09	0.54 ± 0.07	0.39 ± 0.12*	0.52 ± 0.08
Phase duration (%)	57 ± 4*	51 ± 3	26 ± 2	25 ± 2	17 ± 6*	24 ± 4

Table 2.
Mean and standard deviation of sagittal kinematic parameters during Approaching, Braking and Stabilisation phases in both groups. Values of mean angular position are given with respect to the horizontal (i.e. 90° vertical angular position of the segment). * = significantly different from young (ANOVA: *p* <0.05)

		Approaching Phase		Braking Phase		Stabilisation phase	
		Older	**Young**	**Older**	**Young**	**Older**	**Young**
Mean angular position (°)	Head	104.5 ± 8.2	111.4 ± 6.8	107.4 ± 9.3	113.1 ± 7.5	107.0 ± 8.9	110.0 ± 6.8
	Trunk	88.2 ± 5.2	84.3 ± 3.0	86.2 ± 5.6	80.5 ± 3.1	85.2 ± 5.9	82.2 ± 2.9
	Pelvis	95.7 ± 6.6	98.2 ± 1.7	95.3 ± 6.1	98 ± 4.9	95.4 ± 5.9	102.4 ± 5.2
Mean angular velocity (°/s)	Head	1.1 ± 2.1	0.3 ± 2.1	2.6 ± 3.1*	-1.3 ± 2.9	0.2 ± 3.0	-0.9 ± 6.2
	Trunk	-2.5 ± 1.5*	-5.7 ± 1.4	-0.8 ± 2.4	0.6 ± 3.1	-4.4 ± 3.6*	2.6 ± 4.1
	Pelvis	-1.6 ± 1.2	-3.2 ± 2.8	1.6 ± 3.7	0.5 ± 5.5	-1.2 ± 4.2*	9.6 ± 2.4
Maximum Excursion (°)	Head	5.1 ± 2.3	5.3 ± 1.2	3.4 ± 1.5 *	5.2 ± 2.1	1.8 ± 0.9 *	3.4 ± 1.7
	Trunk	4.5 ± 1.4*	7.1 ± 0.9	2.7 ± 0.5 *	4.5 ± 1.5	2.0 ± 0.6	3.0 ± 1.3
	Pelvis	3.9 ± 0.9*	5.2 ± 1.8	2.9 ± 1.1 *	4.3 ± 1.8	1.9 ± 1.0 *	5.2 ± 1.7

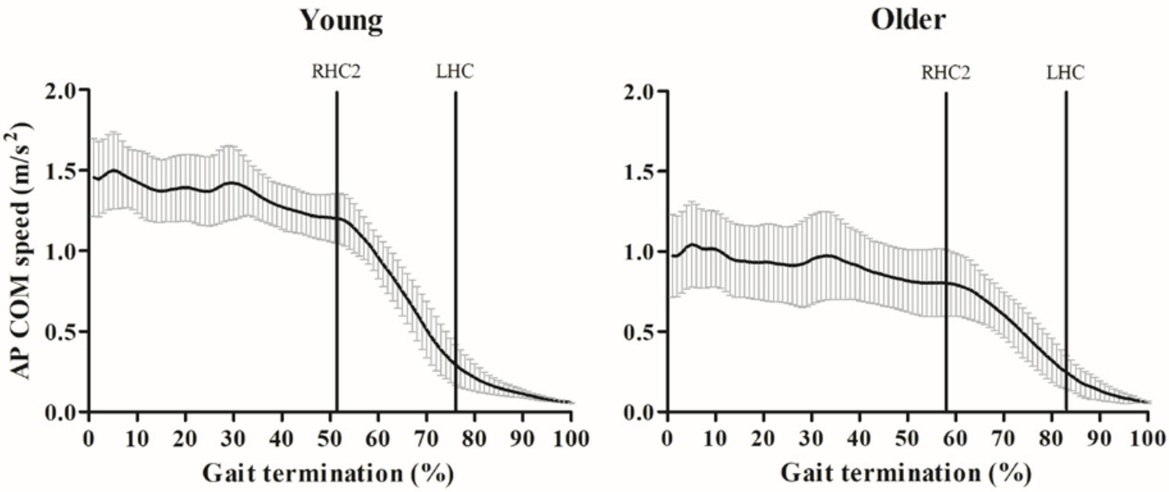
FIGURE CAPTIONS

Fig. 1
Group ensemble average (mean ± SD) of AP COM speed during gait termination. Vertical lines represent the last right heel contact (RHC2) and subsequent left heel contact with the ground (LHC).

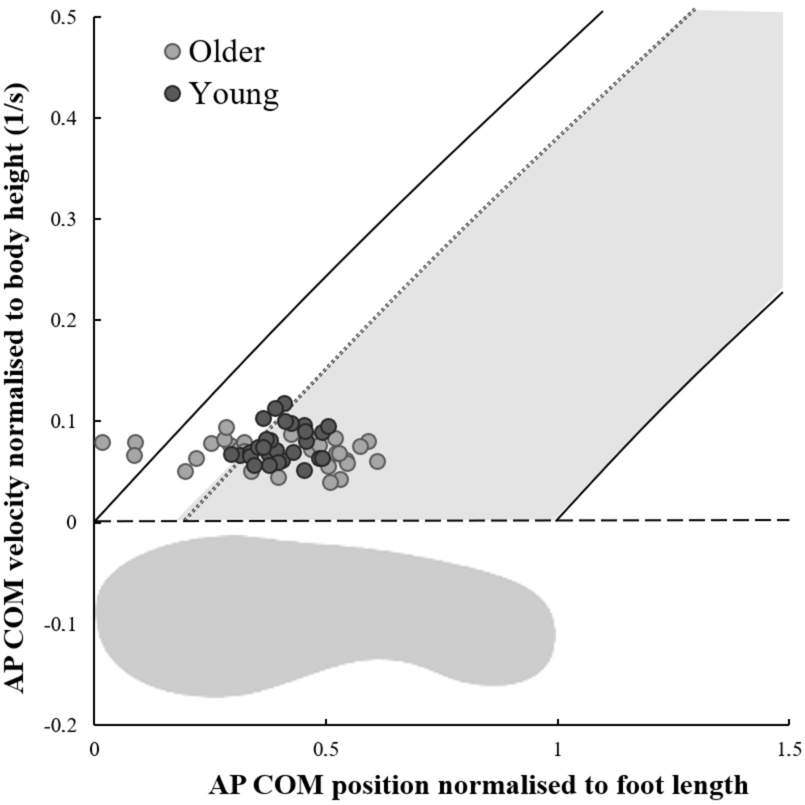
Fig. 2
Mean AP COM velocity-position plot for the two groups during the Stabilisation phase. The upper and lower boundaries (black solid line) define the stability region for young adults as modelled by Pai and Patton. The secondary upper boundary (black dotted line) define the stability region for older adults (grey area) to account for the 59% reduction in strength in elderly as indicated by the same model. If values exceeded the upper boundary of the stability region, a forward fall would be initiated. A backward fall would be initiated if the values were below the lower boundary.

Fig. 3
Sagittal angular displacement of head, trunk and pelvis during three representative gait termination trials of one young (a) and two older women (b and c). Vertical lines represent the last right heel contact (RHC2) and subsequent left heel contact with the ground (LHC). Cross-correlation coefficients of the corresponding head-trunk and trunk-pelvis paired segment motion are presented below and plotted against 25% phase lag (positive and negative) with respect to the entire trial. Positive or negative peak coefficient identifies a motion of the paired segments in the same or opposite direction, respectively. Positive or negative lag indicates that the upper segment leads or follows the lower segment, respectively.

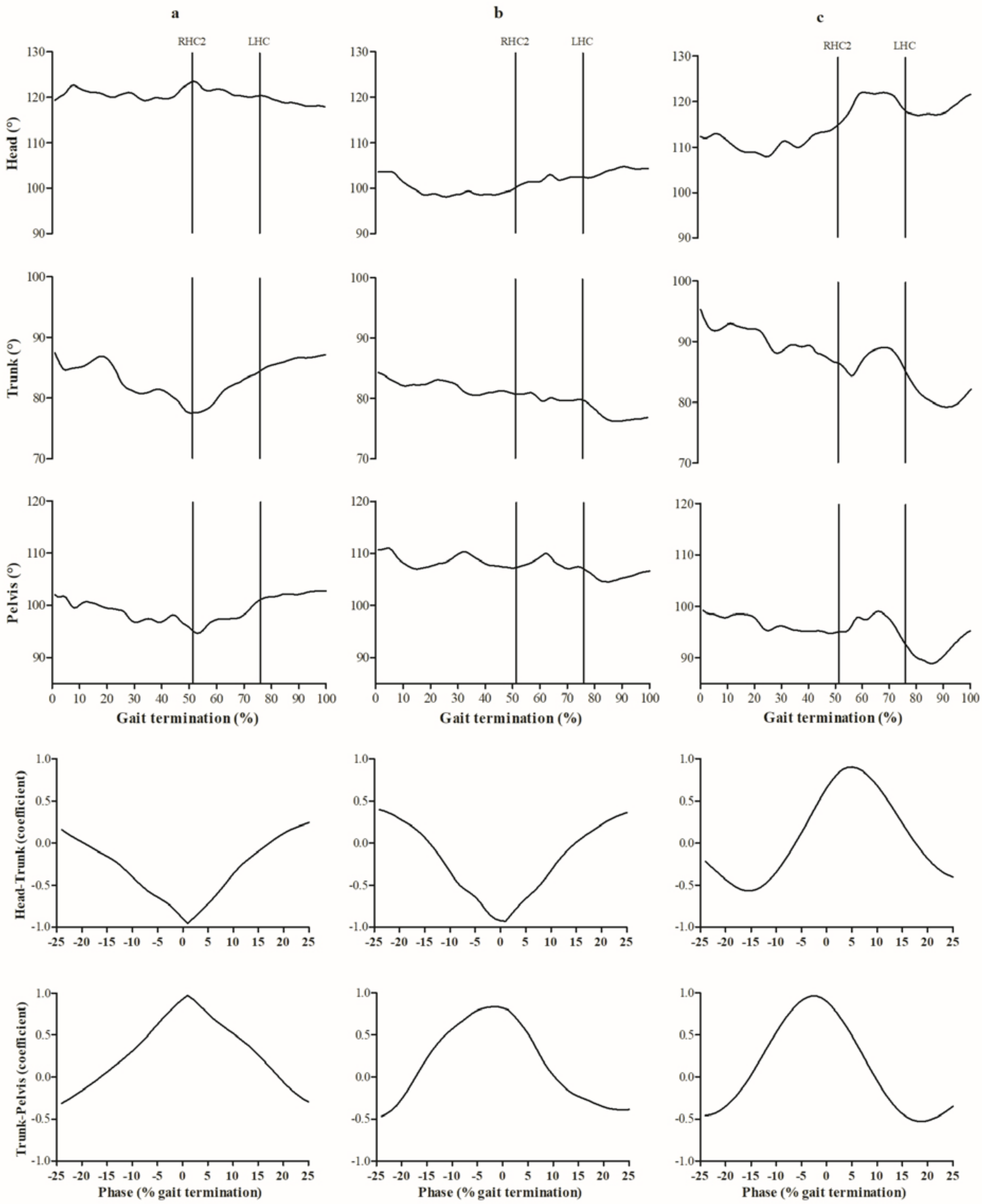
Fig. 4
Group ensemble average of neck, waist and right hip moments (mean ± SD) along with normalised EMG profiles (mean) of SCM, NE and ESL in young and older women during gait termination. Horizontal dotted line in EMG graphs display the mean EMG activity of the muscle over the entire trial.



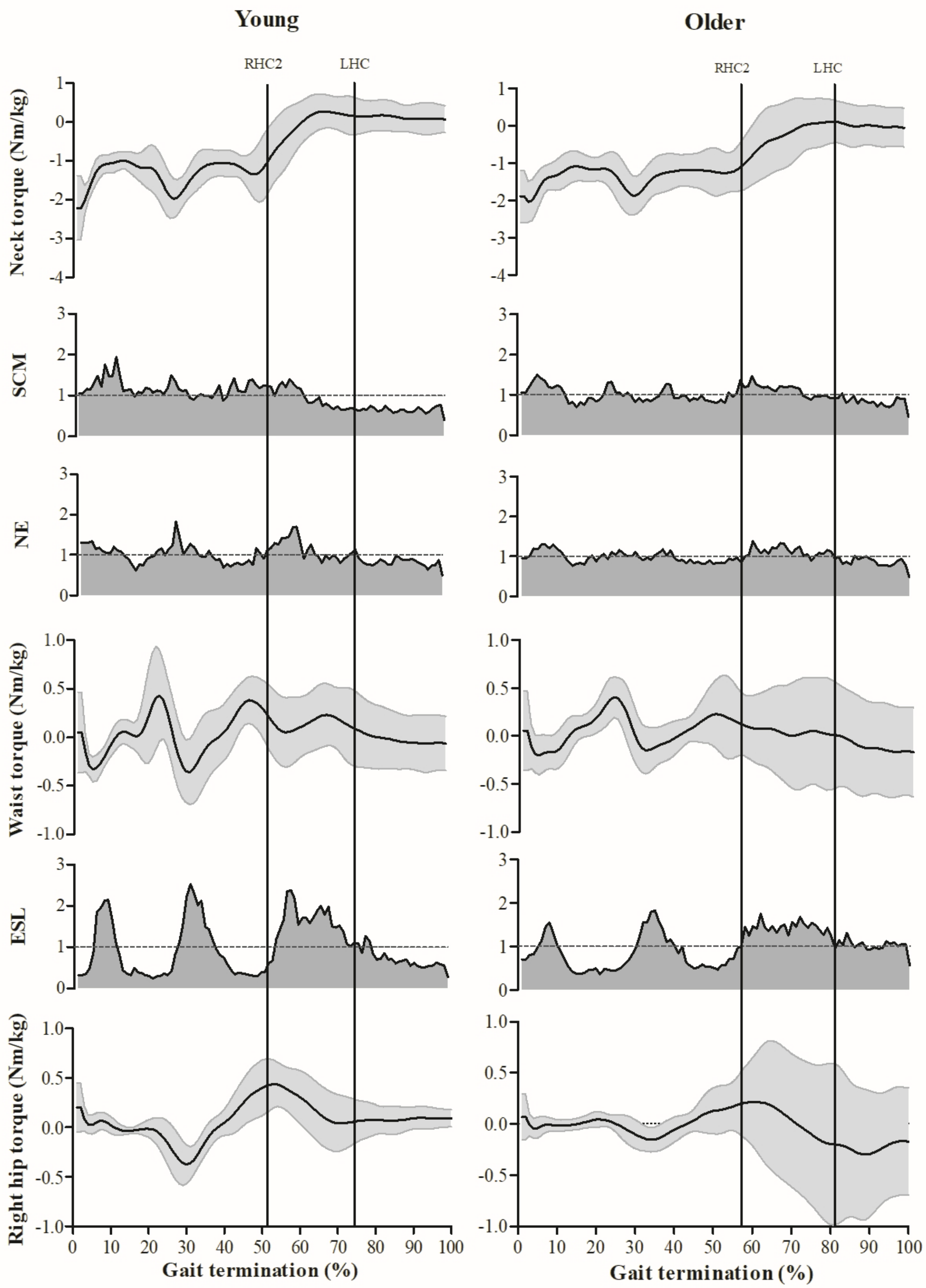
Young
Older
RHC2
LHC
AP COM speed (m/s^2)
2.0
1.5
1.0
0.5
0.0
0
10
20
30
40
50
60
70
80
90
100
Gait termination (%)



0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
Older
Young
AP COM velocity normalised to body height (1/s)
0
0.5
1
1.5
AP COM position normalised to foot length



a
b
c
RHC2
LHC
Head (°)
Trunk (°)
Pelvis (°)
Gait termination (%)
Head-Trunk (coefficient)
Trunk-Pelvis (coefficient)
Phase (% gait termination)



Young
Older
RHC2
LHC
Neck torque (Nm/kg)
SCM
NE
Waist torque (Nm/kg)
ESL
Right hip torque (Nm/kg)
Gait termination (%)