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Article *in* Journal of Back and Musculoskeletal Rehabilitation · September 2015

DOI: 10.3233/BMR-150628

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Journal of Back and Musculoskeletal Rehabilitation 29 (2016) 295–300
DOI 10.3233/BMR-150628
IOS Press

Age-related alterations in the activation of trunk and lower limb muscles during walking

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Abstract.
BACKGROUND: Walking is a complex motor task that requires an integrated coordination of the trunk, lower limb, and upper limb movements. Previously, few studies have investigated the activation pattern of trunk muscles during walking. However, the mechanisms by how aging affects the recruitment of trunk muscles during walking remain unclear.
OBJECTIVE: The present study aimed to compare the activation of trunk and lower limb muscles during walking in younger and older women.
METHODS: Fifteen younger women between 18 and 30 yr and 19 older women between 60–82 yr walked on the treadmill at a self-selected speed, while 1-min surface electromyography (EMG) signals were recorded from the multifidus, internal oblique, gluteus maximus, rectus femoris, biceps femoris, tibialis anterior, and gastrocnemius lateralis. EMG signals were processed and a linear envelope was calculated at an initial stance (50 ms after heel contact) and final stance (50 ms before toe-off).
RESULTS: Compared with younger women, older women had 52.32% lower activation of the internal oblique ($p = 0.027$) and 39.95% lower activation of the rectus femoris ($p = 0.003$) at initial stance.
CONCLUSION: Results of this study demonstrated that older women had lower activation of trunk and knee muscles during the initial stance, which may have resulted from weakness and balance impairments caused by aging.

Keywords: Aging, gait, electromyography, trunk stability

1. Introduction

Falls in older adults incur significant costs to the healthcare system and are responsible for approximately 50% of injuries causing death in people older than 65 years [1,2]. Up to 70% of falls in older adults occur during walking [3]. Thus, the differences in gait biomechanics in older patients who experienced a fall and controls have been previously investigated [4–7]. However, a majority of these studies focused on the kinematics of the lower limbs [4–6].

Walking is a complex motor task requiring an integrated coordination of the trunk, lower limb, and upper limb movements [8]. Furthermore, the head, arms, and trunk represent more than 50% of the body mass and greatly influence the dynamic stability of the body [9]. Thus, it is suggested that the neuromuscular control of trunk motion may be prioritized over that of other inferior segments to maintain stability during movements, such as walking [9].

Aging reduces the dynamic stability across all segments, and especially in trunk motion [9]. According to Kang and Dingwell [9], the greater inertia of the trunk

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ISSN 1053-8127/16/$35.00

may attenuate the effect of a given perturbation on trunk motion, making feedback control less effective and reducing the ability to regain stability after a balance perturbation [9]. Further, Menz et al. [3] demonstrated that older people at the risk of falling have difficulty in controlling the rhythmic displacements of the trunk during walking. However, to our knowledge, no study has investigated the impact of aging on the recruitment of the trunk muscles [3].

Few studies have described the activation pattern of trunk and lower limb muscles during walking [10,11]. Anders et al., verified the extent of trunk muscle activation during walking at different speeds in younger healthy adults [10,11]. These studies showed that trunk muscles had two functions during gait: execute intervertebral and pelvic movements, required to maintain balance and reduce the metabolic cost of walking, and stabilize spinal and pelvic joints in the stance phase [10,11]. Schmitz et al. [12] investigated age-related differences in lower limb muscle activation and demonstrated that older adults have greater co-activation of ankle muscles in mid-stance [12].

Considering that trunk and lower limb muscles may play an important role in maintaining postural control and balance, knowledge of how aging affects the recruitment pattern of these muscles could have important clinical application in preventing falls in older adults. Our study aimed to compare the activation of trunk (multifidus [MU] and internal oblique [IO]) and lower limb (gluteus maximus [GM], rectus femoris [RF], biceps femoris [BF], gastrocnemius lateralis [GL], and tibialis anterior [TA]) muscles between younger and older women in the stance phase of gait. We hypothesized that older women would have lower activation of lower limb and trunk muscles, possibly associated with age-related weakness and balance impairments.

2. Methods

2.1. Participants

Data from 34 women were considered in this prospective study (Table 1). Fifteen younger women (age, 18–30 years) were recruited from a university setting and 19 older women (60–82 years) were recruited from community-based physical activity groups. To determine the sample size, we considered the mean and standard deviation of IO activation during the stance phase obtained in the pilot study ($n = 14$, in each group; effect size = 1.3, power = 0.95, α-error = 0.05).

The subjects were invited to participate in the study by personal contact and provided written informed consent as approved by the Institutional Ethics Committee. People who had uncontrolled cardiovascular disease (2 patients with heart failure), dementia or cognitive impairment (defined as a Mini-Mental State Examination score < 20), balance disturbance (defined as a BERG balance score < 36), hemiparesis, pain in the lower limbs or trunk, or a progressive motor disorder were excluded.

2.2. Procedure

The volunteers visited the laboratory on two separate occasions within 24–72 h. On Day 1 of data collection, the preferred overground and treadmill walking speeds (PTWS) were determined. On Day 2, volunteers were familiarized with treadmill walking at the PTWS (10 min), which was followed by recording of surface electromyography (EMG) signals.

Electromyographic activity was assessed using an 8-channel, telemetered electromyogram (Noraxon®, Phoenix, AR, USA), with signals recorded at a sample frequency of 2000 Hz using silver/silver chloride (Ag/AgCl) disc electrodes (Miotec®, Porto Alegre, Brazil) with an active area of 1 cm^2 and inter-electrode distance of 2 cm arranged in a bipolar configuration. The electrodes were positioned over muscles on the participants' right side as follows: IO, 2 cm medial and inferior to the anterior superior iliac spine; MU, 2 cm lateral to the space between the L4–L5 spinous processes; GM, midpoint of the line between the sacral vertebrae and the greater trochanter; BF, midpoint of the line between the ischial tuberosity and the lateral epicondyle of the tibia; RF, midpoint of the line from the anterior superior iliac spine to the superior border of the patella; TA, at one third of the line between the proximal head of the fibula and the tip of the medial malleolus; and GL, at of the line between the proximal head of the fibula and the calcaneus [13,14]. A reference electrode was placed on the right medial malleolus. Before placing the electrodes, the subject's skin was shaved and cleaned with alcohol to reduce impedance [14]. Heel contact and toe-off were detected by a footswitch (Noraxon®, Phoenix, USA), composed prising four sensors attached to right foot at the heel, first and fifth metatarsals, and toe, which was synchronized with the EMG system. The lower limb dominant side was assessed by the Hoffman and Payne protocol [15].

Table 1
Subject characteristics

Variables	Young women group ($n = 15$)	Older women group ($n = 19$)	P-value
Age (years)	22.13 (2.58)	68.21 (7.44)	$< 0.001^{\dagger}$
Mass (kg)	59.62 (3.6)	64.94 (7.75)	0.016*
Height (m)	1.63 (0.06)	1.53 (0.04)	$< 0.001^{\dagger}$
BMI (kg/m^2)	22.44 (1.39)	27.69 (3.73)	$< 0.001^{\dagger}$
Walking speed (m/s)	1.42 (0.07)	1.08 (0.19)	$< 0.001^{\dagger}$

†= difference between groups, $P < 0.01$; * = between-group differences, $P < 0.05$; BMI = body mass index.

2.3. Gait analysis

Preferred overground walking speed, calculated by dividing the distance walked (10 m) with the time to cover this distance (s), was determined on Day 1 of data collection. Volunteers were instructed to walk at their self-selected speed at a natural pace over 20 m and the duration for each trial was determined with 2 infrared timing gates placed at 5 and 15 m.

The PTWS was determined by starting treadmill walking at 50% of the preferred overground speed followed by a gradual increase in speed until the subject reported that the current speed was faster than the preferred gait speed; the speed was then slowly decreased until the subject reported that the current speed was slower than the preferred gait speed. This procedure was repeated thrice, and the average of the 3 "faster" and 3 "slower" than preferred speeds was taken as the subject's PTWS. Then, the volunteers performed underwent a habituation trial involving 5 min of walking on the treadmill at the self-selected speed.

On Day 2 of data collection, volunteers were familiarized with the PTWS for 10 min. After that, while still walking at the PTWS, 1 min of EMG data was recorded.

2.4. Data analysis

The EMG signal was processed according to specific routines developed in Matlab (Mathworks®, Natick, USA) using a band-pass filter with a cut-off frequency of 20–500 Hz, full-wave rectification and a low-pass, fourth-order filter with a cut-off frequency of 10 Hz. Then, the mean of the linear envelope of the EMG signal was obtained 50 ms after heel contact (initial stance phase) and before toe-off (final stance phase) of in the first 10 strides. All linear envelope values were normalized to the mean activation obtained during the gait.

The PASW version 18.0 (SPSS inc.) was used for all statistical analyses. We used appropriate descriptive statistics (mean and standard deviation) to summarize participant characteristics. The Shapiro-Wilk test was

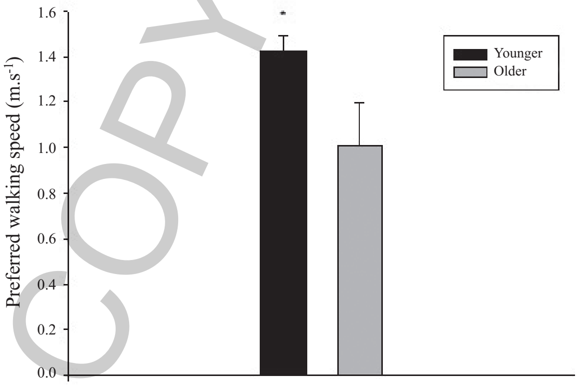

Fig. 1. Comparisons of preferred walking speed between younger and older women. $^*P < 0.05$.

used to test the normality of the data. A covariance multivariate analysis (MANCOVA) was used to compare dependent variables (muscles activation) between groups, using the gait speed as covariate. Moreover, in order to evaluate the influence of age on muscle activation, we computed the partial correlation coefficients to quantify the association between age and EMG activity, and gait velocity was factored in as a covariate. The significance level was set to $p < 0.05$.

3. Results

Older women were heavier ($p = 0.016$), shorter ($p < 0.001$), and had a higher body mass index ($p < 0.001$).

The MANCOVA analysis showed a significant main group effect ($F = 2.623$ and $p = 0.026$). Younger women had 23.96% faster PTWS ($p < 0.001$; Fig. 1), 52.32% higher activation of the IO ($p = 0.001$; Fig. 2), and 39.95% higher activation of the RF ($p = 0.004$; Fig. 2) at initial stance. Further, at final stance, younger women had a 57.01% higher activation of BF ($p = 0.008$; Fig. 2) and older women had 39.82% higher activation of TA ($p = 0.34$; Fig. 2). In addition, partial correlations showed that RF activation at initial stance and BF activation at final stance were negatively as-

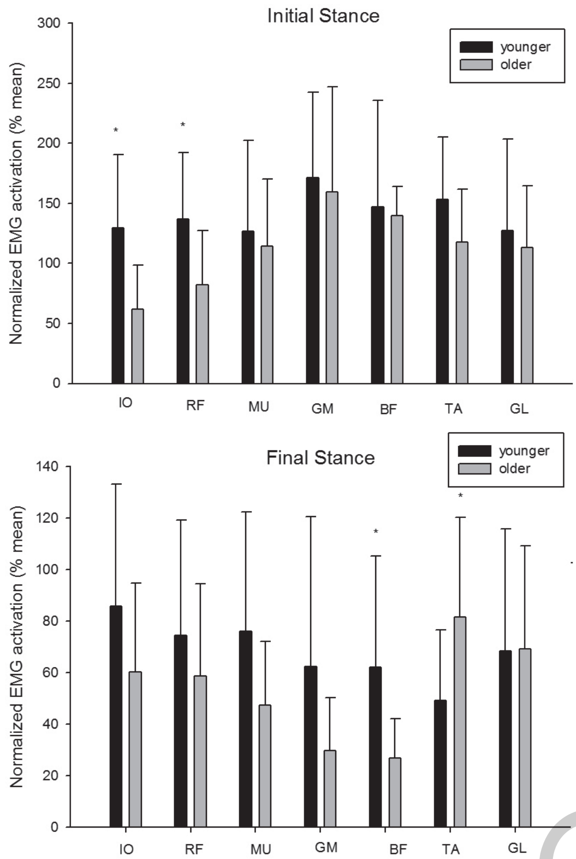

Fig. 2. Comparisons of trunk and lower limb muscle activation between younger and older women. $^{*}P < 0.05$.

sociated with age ($r = -0.422$ and $p = 0.013$; and $r = -0.341$ and $p = 0.041$).

4. Discussion

The present study compared the activation of trunk (IO and MU) and lower limb (GM, RF, TA, GL, and BF) muscles between younger and older women during the stance phase of the gait. Significant between-group differences were observed only during the initial stance phase for IO and RF muscles. Thus, our results partially agree with the previous hypothesis that older women would have lower trunk and lower limb muscles during the stance phase of gait.

During gait, the IO and RF have similar activation patterns, characterized by 2 peaks of activation – the first at initial stance (heel strike) and the second at final stance (push-off phase). At the initial stance, the IO is recruited to provide stability to the trunk segment, whereas the eccentric contraction of RF contributes to decelerate the anterior displacement of the center of mass [15]. Thus, according to our findings, older women may have impaired trunk and lower limb stability during the initial stance phase that could possibly increase the risk of falling.

In view of the fact that the proportion of active motor units is indirectly related to the force generated, we infer that older adults are more inclined to develop lower limb weakness and aging, particularly influenced by quadriceps muscles strength [16–18]. According to Ikezoe et al., the quadriceps muscles constitute the lower limb group muscle with the most lost muscle mass during aging [19]. Moreover, these authors showed that the muscle mass loss in the quadriceps has an impact on the mobility status [19].

For IO and RF muscles, the walking speed has an important contribution on the amount of EMG activation during this phase. According to Anders et al. [10], the IO activation pattern is mixed, with a continuous activation at low walking speeds (0.55–0.83 m/s) and phasic activation at high speeds (1.11–1.66 m/s) [10]. In addition, there is a positive association between walking speed and weight acceptance force at heel strike, which leads to an increased amount of the RF activation [20].

Our findings demonstrated that there is a negative correlation between age and walking speed, which is in concordance with the majority of previous studies [21–23]. This difference in the walking speeds between younger and older women is the main limitation of this study. Thus, to attenuate the effect of this limitation on the comparison of muscle activation between younger and older women, we considered the walking speed as covariate in the statistical analysis. In addition, from our results, we found that younger women had greater activation of BF older women had greater activation of TA at the final stance, which may contribute to the between-group difference in gait speeds.

During the final stance, the plantar flexors are highly activated to push the foot against the ground while the knee flexors and hip extensors are also activated to promote anterior acceleration of the body [15]. Thus, the higher activation of the BF in younger women may contribute to their increased walking speeds. On the other hand, the greater co-activation between the TA and GL in older women during the final stance may compromise the capacity to generate torque and accelerate the body to execute the next step [12].

Surface EMG is frequently used to describe the muscle activation pattern during human gait [10]. Recently, studies had used surface EMG to measure and describe trunk muscle activation during walking [10,11]. Despite the amount of fat tissues above the electrodes and the cross-talk between the muscles (IO and transversus abdominis; MU and erectors spinae), surface EMG is a noninvasive and safe method to determine spinal stability [10].

The body mass index has a strong correlation with the amount of fat tissue; therefore, in our sample, older women should have had a higher concentration of fat tissue, specifically above the areas where the electrodes were placed. Thus, the amount of EMG activation may have been reduced in this group. However, in an attempt avoid this limitation for the use of surface EMG, we amplified and normalized the signal. In addition, the history of falls in older subjects was not considered in this study; therefore, extrapolations of our results for fall prevention be made.

5. Conclusion

This is an initial study that investigated how age affects the EMG activation pattern of trunk and lower limb muscles during walking. The main findings of this study demonstrated that older women had lower activation of trunk and knee muscles during initial stance. Thus, according to our findings, we suggest that future studies must be conducted in order identify the effects of aging on lower limb joint and trunk stability and loading distribution during walking.

Acknowledgments

This study was funded by grants from the Fundação de Amparo à Pesquisa do Estado de São Paulo (FAPESP, 2011/11639-7), Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq), Fundação para o Desenvolvimento da UNESP (FUNDUNESP), and Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES).

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