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Gait and Posture 18 (2003) 101–108

GAIT & POSTURE

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Aging, muscle activity, and balance control: physiologic changes associated with balance impairment☆

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Accepted 6 December 2002

Abstract

Older adults demonstrate increased amounts of postural sway, which may ultimately lead to falls. The mechanisms contributing to age-related increases in postural sway and falls in the elderly remain unclear. In an effort to understand age-related changes in posture control, we assessed foot center-of-pressure (COP) displacements and electromyographic data from the tibialis anterior, soleus, vastus lateralis, and biceps femoris collected simultaneously during quiet-standing trials from elderly fallers, elderly non-fallers, and healthy young subjects. Both traditional measures of COP displacements and stabilogram-diffusion analysis were used to characterize the postural sway of each group. Regression analyses were used to assess the relationship between the COP measures and muscle activity. Elderly fallers demonstrated significantly greater amounts of sway in the anteroposterior (AP) direction and greater muscle activity during quiet standing compared with the young subjects, while elderly non-fallers demonstrated significantly greater muscle activation and co-activation compared with the young subjects. No significant differences were found between elderly fallers and elderly non-fallers in measures of postural sway or muscle activity. However, greater postural sway in both the AP and mediolateral (ML) directions and trends of greater muscle activity were found in those older adults who demonstrated lower scores on clinical measures of balance. In addition, short-term postural sway was found to be significantly correlated with muscle activity in each of these groups. This work suggests that high levels of muscle activity are a characteristic of age-related declines in postural stability and that such activity is correlated with short-term postural sway. It is unclear whether increases in muscle activity preclude greater postural instability or if increased muscle activity is a compensatory response to increases in postural sway.

Keywords: Fallers; Postural control; Muscle activity; Stabilogram-diffusion analysis

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1. Introduction

Increased postural sway in older adults is well-documented [1–3], and research has linked greater amounts of postural sway to increased risk of falling, which is a serious problem for older adults [4,5]. For instance, Fernie et al. [4] found significantly greater average speed of sway in older adults who had fallen one or more times in a year compared with those who had not fallen. In a prospective study, Maki et al. [5] found significantly greater anteroposterior (AP) sway in those older adults who had a fall in the 1-year period following the balance measurements. These studies used traditional balance measures, which provide descriptive information on the postural sway exhibited by an individual.

☆ Data were collected at the Hebrew Rehabilitation Center for the Aged, Roslindale, MA and processed at the Applied BioDynamics Laboratory in the Department of Biomedical Engineering, Boston University, Boston, MA.

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0966-6362/02/$ - see front matter
doi:10.1016/S0966-6362(02)00200-X

In 1993, Collins and De Luca [6] developed a statistical-biomechanics method of assessing quiet-standing foot center-of-pressure (COP) time series, called stabilogram-diffusion analysis (SDA). This analysis revealed that over short-term time intervals during quiet stance, the COP tends to drift away from a relative equilibrium point, while over long-term time intervals, the COP tends to return to a relative equilibrium point [6–8]. This finding was interpreted as an indication that the postural control system utilizes open-loop control mechanisms over the short-term and closed-loop control mechanisms over the long-term to maintain upright stance. An open-loop control system is one which operates without sensory feedback, and in the case of the human postural control system may correspond to descending commands which set the steady-state activity levels of the postural muscles. Closed-loop control systems, on the other hand, operate with sensory feedback, and in the case of the human postural control system correspond to the visual, vestibular and somatosensory systems. This interpretation and modeling framework enables one to relate SDA parameters to the steady-state behavior and functional interaction of the neuromuscular mechanisms underlying the maintenance of upright posture [6–8].

In a cross-sectional study of healthy subjects, Collins et al. [8] found that the steady-state behavior of the open-loop postural control mechanisms, which operate over short-term time intervals, is more unstable in older adults compared with young adults, i.e. the output of the overall system has a greater tendency to move or drift away from a relative equilibrium point. It is possible that these age-related changes in the open-loop postural control mechanisms may be due to a postural control strategy adopted by elderly individuals whereby they increase the level of muscle activity across their lower-limb joints. It is important to note that the force output of skeletal muscles contains noise-like fluctuations [9], which increase with muscle activity [10,11]. Increased muscle activity would, therefore, lead to larger noise-like fluctuations across joints, thereby increasing the amount of short-term postural sway. Moreover, compared with healthy young adults, older adults exhibit significantly increased levels of muscle co-activation in response to postural perturbations [12]. Therefore, it is likely that older adults also demonstrate increased muscle activity during quiet standing.

In this study, we examined the effects of age and fall status on measures of muscle activity and short-term postural sway. We also assessed the relationship between short-term postural sway and muscle activity. We hypothesized that: (1) the previously reported age-related increase in short-term postural sway [8] was related to an age-related increase in muscle activity during quiet standing, and (2) older individuals who have had recent unexplained falls have greater amounts of postural sway and greater lower-extremity muscle activity during quiet standing compared with healthy older adults.

2. Methods

Seventy ambulatory, community-dwelling older adults and 15 healthy young adults (seven male, eight female; age range 22–32 years, mean 27 ± 3 years) volunteered for this study. Subjects were recruited through advertisements in a weekly newsletter distributed throughout the local area to community-dwelling individuals over the age of 65, through newspaper advertisements in local newspapers, and through visits to assisted-living communities in the area. Thirty-three of the older adults had a history of two or more unexplained falls (i.e. ones which could not be explained by a trip over an obstacle, a slip on ice, or from a biomechanical perturbation such as a push) in the previous year and were classified as fallers (elderly fallers: six male, 27 female; age range 65–92 years, mean 75 ± 7 years). Thirty-seven of the older adults did not have a history of falls and were classified as non-fallers (elderly non-fallers: 15 male, 22 female; age range 68–89 years, mean 75 ± 5 years). Informed consent was obtained from all participants prior to participation in accordance with the Institutional Review Boards of the Boston University Charles River Campus and the Hebrew Rehabilitation Center for Aged.

All elderly subjects were screened with a medical history and physical examination. Exclusion criteria for fallers were falls secondary to syncope, acute illness, or other specific causes including metabolic disorders, medication side effects, true vertigo, and neurologic or lower-extremity orthopedic diagnoses. Additional exclusion criteria included: (1) focal findings of 1/5 strength difference between sides of the hip flexors, extensors, and abductors, as well as knee and ankle flexors and extensors, as determined by a manual muscle test [13], (2) absent knee extensor reflexes or positive Babinski responses [14], (3) corrected visual acuity worse than 20/100 or presence of a field defect, (4) a Mini-Mental Status Examination [15] score less than or equal to 24/30, (5) orthostatic hypotension, and (6) unilateral sensory deficit in the lower extremities, bilateral sensory loss in stocking distribution, or bilateral proprioception less than 75% correct in identifying great toe position.

A commonly used clinical assessment of balance and gait, the Tinetti Performance Oriented Balance and Mobility Assessment (POMA) [16], was administered to the elderly subjects. This test grades balance performance as normal, adaptive, or abnormal based on maneuvers which include sitting balance, arising from a chair, immediate standing, standing balance, balance with eyes closed while feet are close together, turning balance, response to a nudge, balance while turning the head, standing balance on one leg, back extension, reaching forward, reaching down, and sitting down. The POMA was administered by the same physical therapist to all subjects.

Electromyographic data (Neurodyne Medical Corp., Cambridge, MA) were collected bilaterally from the tibialis anterior (TA), soleus (SO), vastus lateralis (VL), and biceps femoris (BF). The electrodes were placed superficial to the belly of the TA, VL, and BF muscles, and on the medial aspect of the SO belly in an area where the gastrocnemius was not superficial to the SO muscle. Three resting baseline muscle activity trials and three maximal voluntary isometric contractions (MVIC) of 3 s in duration each were collected from each muscle with the participant in either a supine or sitting position. Participants then stood upright and barefoot on a force plate (Kistler Instrument Corp., Amherst, NY) in a comfortable stance with arms at their sides, looking straight ahead. Foot position was self-selected in a stance which was most comfortable for the subject on the force plate, with the lateral borders of the feet no further than 40 cm apart and the big toes of the feet aligned evenly in the AP direction. The participants' feet were then traced onto a sheet of paper so that foot position would be consistent from trial to trial. Ten 30-s quiet-standing trials were obtained from each participant. Rest breaks were provided as needed. Force-plate and electromyographic (EMG) data were sampled at 1500 Hz.

SDA was performed on the COP trajectories. SDA, as described by Collins and DeLuca, provides a number of physiologically meaningful parameters [6–8]. Among these are AP diffusion coefficients obtained by plotting the mean squared COP displacement in the AP direction ($\langle \Delta y^2 \rangle$) as a function of the time interval (Δt). This plot demonstrates two distinct regions: a short-term region and a long-term region. The short-term (D_{ys}) and long-term (D_{yl}) AP diffusion coefficients, which characterize the effective stochastic activity of open-loop and closed-loop postural control mechanisms in the AP direction, respectively, are derived from the slopes of the short-term and long-term regions of this plot (e.g. see Fig. 1A).

Traditional measures of postural sway were also calculated. These parameters included the standard deviation (S.D.) and range of the COP trajectory in the AP direction (AP S.D. and Range AP, respectively), and the S.D. and range of the COP trajectory in the ML direction (ML S.D. and Range ML, respectively).

EMG data were band-pass filtered between 20 and 450 Hz using an inverse Chebyshev filter and then processed using a 40-ms running-window root mean square (RMS). The mean RMS value for each muscle for each quiet-standing trial was then normalized to the average baseline RMS value and the peak RMS of the MVIC (TA, SO, VL, BF). 'On' levels for each muscle were determined as three S.D. above baseline level. This level was used to determine the percentage of a trial that each muscle was active (%TA, %SO, %VL, %BF). Muscle activity parameters were averaged across trials and then over right and left sides. The percent of a trial for which the antagonist muscle pairs, tibialis anterior and soleus (%TA–SO) and vastus lateralis and biceps femoris (%VL–BF), respectively, were active simultaneously was also determined and averaged across trials and sides.

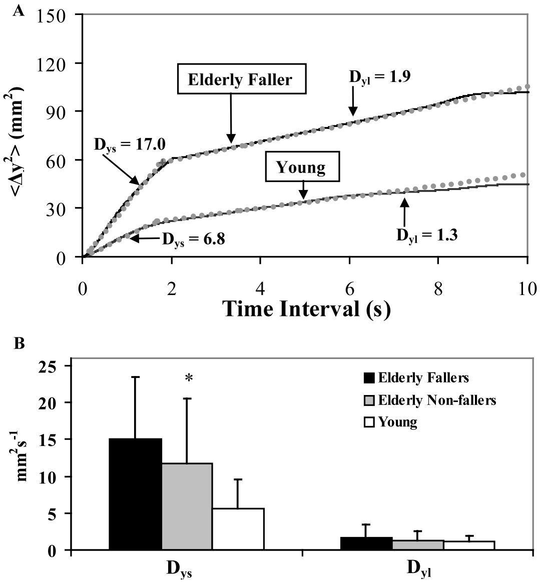

Fig. 1. Stabilogram-diffusion analysis. (A) Experimental anteroposterior linear stabilogram-diffusion plots (solid line) and fitted regressions (dotted lines) for one representative young subject (age 30 years) and one representative elderly faller subject (age 86 years) (solid lines). The computed short-term (D_{ys}) and long-term (D_{yl}) diffusion coefficients (in units of $mm^2\ s^{-1}$) are shown for each subject. (B) Short-term (D_{ys}) and long-term (D_{yl}) diffusion coefficients in the anteroposterior direction. Group means and standard deviations are shown for the elderly fallers ($N = 33$), elderly non-fallers ($N = 37$), and young subjects ($N = 15$). Significant differences ($P < 0.008$) between elderly fallers and young subjects are indicated by an asterisk *.

Six one-way analyses of variance (ANOVA) with Tukey post-hoc tests were performed on the postural sway parameters (D_{ys}, D_{yl}, AP S.D., Range AP, ML S.D., and Range ML) and 10 one-way ANOVAs were performed on the muscle activity parameters (TA, SO, VL, BF, %TA, %SO, %VL, %BF, %TA–SO, and %VL–BF) to determine differences between groups. Significant α levels were corrected for multiple comparisons with a Bonferroni adjustment for multiple comparisons ($P = 0.05/6 = 0.008$ for the postural sway parameters and $P = 0.05/10 = 0.005$ for the muscle activity parameters). To determine the relationship between muscle activity and postural sway, backwards-stepwise regression analyses were conducted for each subject group (elderly faller, elderly non-faller, young)

Table 1
Traditional center-of-pressure measures for elderly fallers ($N = 33$), elderly non-fallers ($N = 37$), and young ($N = 15$)

	Mean and S.D.			Significance		
	Elderly fallers	Elderly non-fallers	Young	EF vs. EN	EF vs. Y	EN vs. Y
AP S.D. (mm)	4.97 (1.55)	4.15 (1.08)	3.54 (1.01)	–	*	–
Range AP (mm)	24.22 (7.21)	20.66 (5.43)	16.69 (4.97)	–	*	–
ML S.D. (mm)	3.17 (1.28)	2.87 (1.33)	2.26 (0.79)	–	–	–
Range ML (mm)	15.27 (6.39)	14.18 (6.44)	10.87 (3.21)	–	–	–

Four traditional parameters were considered: S.D. and range of the COP trajectory in the anteroposterior direction (AP S.D. and Range AP, respectively), and the S.D. and range of the COP trajectory in the mediolateral direction (ML S.D. and Range ML, respectively). The group means and S.D. for each parameter are shown. Significant differences ($P < 0.008$) between the elderly fallers and nonfallers (EF vs. EN), elderly fallers and young (EF vs. Y), and elderly non-fallers and young (EN vs. Y), are indicated by an asterisk *; non-significant findings are indicated by a double dash –.

with D_{ys} as the dependent variable and all of the muscle activity parameters as the independent variables. Only the independent variables that significantly contributed to the variance in the dependent variable were included in the final regression. Diffusion coefficients in the ML direction were not reported as the muscles for which EMG data were collected effect movement primarily in the AP direction. Differences between fallers and non-fallers in the POMA were assessed using the non-parametric Wald–Wolfowitz runs test [17].

We also examined the postural sway of elderly participants with and without balance performance deficits, as measured by the POMA clinical balance assessment. The traditional sway parameters, short-term and long-term AP diffusion coefficients, and muscle activity parameters for participants without clinical balance deficits (those demonstrating a perfect POMA score: 16 points) were compared with those for participants with balance deficits (those demonstrating a less than perfect POMA score: <16 points). These comparisons were made using Student's t-tests. A Bonferroni correction factor was used to adjust the α level for multiple comparisons for the six postural sway parameters (D_{ys}, D_{yl}, AP S.D., Range AP, ML S.D., and Range ML; $P = 0.05/6 = 0.008$) and the ten muscle activity parameters (TA, SO, VL, BF, %TA, %SO, %VL, %BF, %TA–SO, %VL–BF; $P = 0.05/10 = 0.005$).

3. Results

Elderly fallers demonstrated significantly greater short-term AP diffusion coefficients (D_{ys}) compared with the young subjects (Fig. 1B). Significant differences between the elderly fallers and the young subjects were also found in measures of the AP COP standard deviation (AP S.D.) and range (Range AP) (Table 1). There were no significant differences in the postural sway measures between the elderly non-fallers and young subjects, and the elderly fallers and elderly non-fallers (Fig. 1B and Table 1).

On average, elderly fallers scored similarly to elderly non-fallers on clinical tests of balance (POMA). Interestingly, when the elderly non-fallers and fallers were classified according to performance on the POMA, significantly greater short-term AP diffusion coefficients (D_{ys}, $P = 0.004$) were found in the group with lower POMA scores (i.e. those who had difficulty completing one or more of the balance tasks on the test; Table 2). In addition, significantly greater measures of ML COP standard deviation (ML S.D., $P = 0.007$) and range (Range ML, $P = 0.004$) were found in the group with lower POMA scores (Table 2).

In the elderly non-fallers and elderly fallers, the vastus lateralis (%VL) was significantly more active for a greater percent of the quiet-standing trials compared with the young subjects ($P < 0.005$; Fig. 2B). In addition, elderly non-fallers demonstrated significantly more co-activation in antagonistic muscle groups (%VL–BF) compared with the young subjects ($P < 0.005$; Fig. 2C). The normalized tibialis anterior activity (TA), vastus lateralis activity (VL), and biceps femoris (BF) activity in the older adults were in the order of three to 100 times

Table 2
Sway parameters for elderly subjects separated according to the performance oriented mobility assessment (POMA) scores

Parameter	Low POMA, $N = 22$	High POMA, $N = 48$
$D_{ys\,*}$	17.37 (10.71)	11.41 (6.9)
D_{yl}	2.01 (2.01)	1.05 (1.05)
AP S.D.	5.10 (1.75)	4.28 (1.10)
Range AP	25.25 (8.30)	21.00 (5.11)
ML S.D.*	3.65 (1.61)	2.70 (1.02)
Range ML*	17.69 (7.63)	13.32 (5.28)

Comparisons were made between groups that performed well on the POMA test vs. those that demonstrated a lower score. Variables included in this assessment were the short-term and long-term anteroposterior diffusion coefficients (D_{ys} and $D_{yl,}$ respectively), the anteroposterior S.D. (AP S.D.) and range (Range AP) of the COP trajectory, and the mediolateral S.D. (ML S.D.) and range (Range ML) of the COP trajectory. The group means and standard deviations for each parameter are shown. Significant differences ($P < 0.008$) are indicated by an asterisk *.

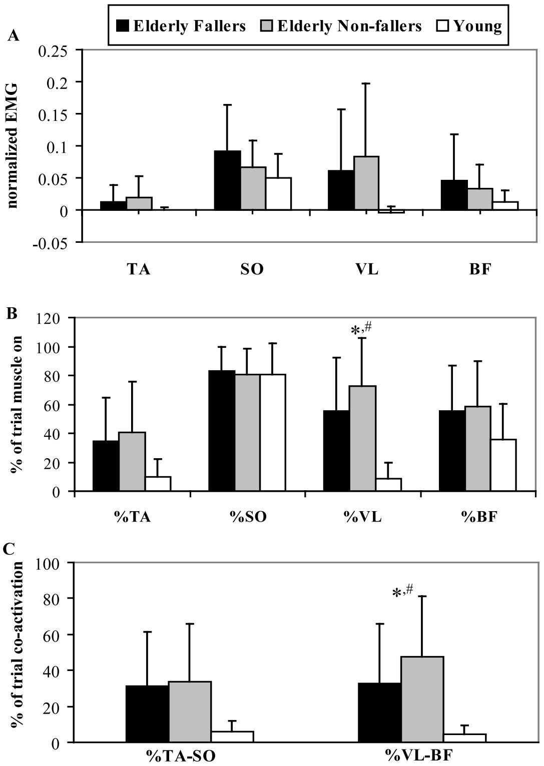

Fig. 2. Quiet-standing muscle activity: group means and standard deviations for the elderly fallers ($N = 33$), elderly non-fallers ($N = 37$), and young ($N = 15$) subjects. Electromyographic (EMG) data were processed with a 40-ms moving root-mean-square window. (A) Normalized muscle activity averaged over the ten 30-s quiet-standing trials (TA, tibialis anterior; SO, soleus; VL, vastus lateralis; BF, biceps femoris). Note that the young group had, on average, negative muscle activity for vastus lateralis during quiet standing. This result occurred because most young participants demonstrated vastus lateralis muscle activity levels during quiet standing that were below the average baseline value (see body of the text). (B) Percent of stance a given muscle was active (%TA, tibialis anterior; %SO, soleus; %VL, vastus lateralis; %BF, biceps femoris). (C) Percent of stance antagonist muscle pairs, tibialis anterior–soleus (%TA–SO) and vastus lateralis–biceps femoris (%VL–BF), respectively, were co-activated. Significant differences ($P < 0.005$) between elderly fallers and young subjects are indicated by an asterisk *. Significant differences ($P < 0.005$) between elderly non-fallers and young subjects are indicated by $^{\#}$.

greater than that of the young subjects. In addition, the percent of the standing trials for which the biceps femoris (%BF) and the tibialis anterior (%TA) were active and for which the tibialis anterior and soleus (%TA–SO) were co-activated were 1.5 to five times greater in the older subjects compared with the young subjects (Fig. 2B and C). When the elderly fallers and elderly non-fallers were classified according to POMA, those who received lower scores, indicating a balance deficit, demonstrated substantially greater muscle activity compared with the group of older adults who received perfect scores on the POMA (Table 3). Due to the high levels of variability in the EMG data, these increases were not statistically significant.

Table 3
Muscle activity parameters for elderly subjects separated according to the performance oriented mobility assessment (POMA) scores

Parameter	Low POMA, $N = 22$	High POMA, $N = 48$
TA	0.029 (0.043)	0.010 (0.021)
SO	0.084 (0.078)	0.076 (0.049)
VL	0.117 (0.161)	0.053 (0.058)
BF	0.038 (0.041)	0.040 (0.063)
%TA	49.63 (31.33)	32.29 (32.11)
%SO	80.36 (17.05)	82.85 (17.88)
%VL	68.36 (38.29)	63.02 (34.76)
%BF	63.77 (26.36)	53.87 (32.81)
%TA–SO	40.00 (28.78)	29.12 (31.93)
%BF–VL	46.54 (35.19)	37.79 (33.36)

Comparisons were made between groups that performed well on the POMA test vs. those that demonstrated a lower score. Variables included in this assessment were: percent of stance a given muscle was active (%TA = tibialis anterior, %SO = soleus, %VL = vastus lateralis, %BF = biceps femoris) and percent of stance antagonist muscle pairs, tibialis anterior–soleus (%TA–SO) and vastus lateralis–biceps femoris (%VL-BF), respectively, were co-activated. The group means and standard deviations for each parameter are shown. Significant differences between the groups were not detected; however, trends of increased muscle activity in the low POMA score group (i.e. those who clinically demonstrated difficulty in completing one or more tasks of balance) were noted.

The results of the backward stepwise regression analyses are summarized in Table 4. Short-term postural sway, as characterized by D_{ys}, was positively correlated with soleus activity (SO) in the elderly fallers, biceps femoris activity (%BF) in the elderly fallers and non-fallers, and tibialis anterior activity (TA) in the young subjects. Short-term postural sway was negatively correlated with co-activation of the vastus lateralis and biceps femoris (%VL–BF) in both the elderly non-fallers and young subjects.

4. Discussion

We hypothesized that age-related increases in postural sway during short intervals of time was due, in part, to an age-related increase in muscle activity during quiet standing. In support of this hypothesis, we found significant increases in muscle activation and co-activation of antagonistic muscle groups in older adults compared with young adults during quiet standing. We also hypothesized that older adults who have had a history of falling would demonstrate increased short-term postural sway and increased muscle activity during quiet standing compared with older adults who did not have a history of falls. We were unable, however, to find differences in either quiet-standing sway or postural muscle activity when older adults were classified according to their self-reported history of unexplained falls.

The greater values for D_{ys} for the elderly fallers compared with the young suggests an age-related

Table 4
Results of the regression analysis relating muscle activity to the short-term AP diffusion coefficient (D_{ys}) for the elderly fallers, elderly non-fallers, and healthy young subjects

Groups and regression coefficients	Independent variables included in regression	Partial correlation to D_{ys}	P-value for partial correlation
Elderly fallers	SO	0.48	0.007
$R = 0.61$ ($R^2 = 0.38$), $P < 0.001$	%BF	0.46	0.011
Elderly non-fallers	%BF	0.44	0.014
$R = 0.44$ ($R^2 = 0.19$), $P = 0.043$	%VL–BF	−0.41	0.021
Young	TA	0.57	0.031
$R = 0.65$ ($R^2 = 0.42$), $P = 0.005$	%VL–BF	−0.57	0.032

Partial correlations and P-values are shown for the independent variables included in the final predictive regression equation, as determined by a backwards stepwise regression. SO, soleus muscle activity normalized to its maximum voluntary isometric contraction (MVIC), %BF, percent of trial that the biceps femoris was on, %VL–BF, percent of trial that the vastus lateralis and biceps femoris were co-activated; TA, tibialis anterior muscle activity normalized to its MVIC.

increase in postural sway over short-term intervals of time. This result is consistent with the study of Collins et al. [8], who hypothesized that this age-related change in the postural control system involves a strategy by which older individuals increase the level of muscle activity across their lower-limb joints. Age-related increases in muscle activity may be maladaptive with respect to short-term posture control as the force output of skeletal muscle contains noise-like fluctuations [9], which increase with increased muscle activity [10,11]. Thus, the increase in muscle activity observed in the elderly subjects may be responsible for the increase in short-term postural sway and consequently may compromise an individual's ability to maintain upright stability. In support of this explanation, significant positive correlations were found between soleus, biceps femoris, and tibialis anterior muscle activity and short-term postural sway in the elderly fallers, elderly non-fallers, and young subjects. We observed a negative relationship between postural sway and muscle co-activation of the biceps femoris and vastus lateralis for both the elderly non-fallers and young subjects. This indicates that co-activation of these muscles had a stabilizing effect on short-term postural sway.

The elderly fallers and non-fallers in the present study had more than twice as much biceps femoris activity during quiet standing compared with the young subjects. This may be due, in part, to the flexed and rigid stance demonstrated by older adults [18]. This flexed posture places the body's center of gravity anterior to the base of support. By activating the hamstrings, older subjects may be attempting to prevent the body's center of gravity from moving further forward. The positive correlation between the biceps femoris and short-term postural sway in both the elderly fallers and non-fallers indicates that this strategy is ineffective at maintaining short-term postural stability.

The elderly subjects demonstrated more than three times greater tibialis anterior muscle activity and significantly greater vastus lateralis muscle activity compared with the young subjects. This may be due, in part, to a compensation for age-related decreases in muscle strength. While muscle strength was not quantitatively measured in this study, it has been documented that many older individuals have relatively weaker tibialis anterior and vastus lateralis muscles compared with healthy young individuals [19,20]. Weakness in these muscle groups could potentially impair an individual's ability to correct a shift in the body's center of gravity and effectively prevent a fall. It may be that maintaining these muscles in an activated state, thereby increasing muscle co-activation, is an attempt to provide additional stability under conditions of increased muscle weakness.

Age-related deterioration of sensory and neuromuscular control mechanisms may also be responsible for the greater muscle activity of the elderly subjects. Increasing levels of muscle activity may assist in enhancing joint proprioception by increasing the firing rate and recruitment of primary afferents, thereby enhancing the functional behavior of the associated closed-loop postural control mechanisms [21]. The observed age-related increase in muscle activation may also be due, in part, to changes in muscle inhibitory mechanisms. For instance, a loss of giant pyramidal cells (Betz cells) in the motor cortex occurs with aging, and these cells typically inhibit extensor muscles and diminish anti-gravity tone [22]. Research has also demonstrated that in young subjects, motor output is maintained by regulating presynaptic inhibition, while, in older adults, presynaptic inhibition is less effective and motor output is maintained through direct activation [23].

The experimental modeling work of Peterka [24] and Newman et al. [25] may also provide some explanation for these findings. Both studies found that a closed-loop control system with continuous feedback control was consistent with the earlier SDA findings of Collins and colleagues [6–8]. In Peterka's model [24], increases in short-term sway occurred when the damping factor (the corrective torque generated in proportion to body sway or velocity) was decreased, or when the time delay due to sensing, transmission, processing and muscle activation was increased [24]. Such changes could occur in older adults under conditions of decreased muscle strength or with a decline in nerve conduction speed, both of which have been shown to occur with age [19,20,26]. Newman et al. [25] found that postural stability was optimal within a range of muscle activity, and very large amounts or very small amounts of muscle activity created postural instability [25]. Thus, damping, or a decrease in muscle gain, may result in a larger amount of sway, which would then be associated with greater amounts of corrective muscle activation and muscle co-activation. Similarly, greater amounts of muscle activity may also result in larger amounts of stochastic activity, as noted above, resulting in increased postural sway.

There were no differences between the elderly fallers and elderly non-fallers in clinical balance performance (POMA). This finding indicates that, for this sample of community-dwelling elderly, balance performance of the elderly non-fallers was not significantly better than that of the elderly fallers. In this sample, other factors such as risk-taking or activity level may have played a role in the incidence of falls, and thus a fall event did not distinguish participants with and without balance deficits. Therefore, it is not surprising that the measures of postural sway and muscle activity were not significantly different between the elderly fallers and elderly non-fallers. However, when older adults were classified according to clinical measures of balance (POMA), measures of postural sway were significantly greater for those demonstrating lower scores. Of notable importance was the finding that there were no significant differences in ML sway between the elderly fallers and non-fallers. However, sway in the ML direction was significantly greater in those older adults who demonstrated poorer POMA scores. This finding is consistent with Maki et al. [5] and Mitchell et al. [27], who both found increased sway in the ML direction to be indicative for risk of falling and age-related disease.

As noted above, significantly greater amounts of postural sway and muscle activity were present in the elderly compared with the younger subjects. It is unknown, however, whether increased muscle activity is a contributing factor to increased postural sway or whether increased muscle activation is a compensation for increased postural sway. Previous work by our group has shown increased postural sway in young adults when muscle activity levels were increased using biofeedback techniques (unpublished data). Future studies in which muscle activity in the elderly is modified while standing, using similar biofeedback techniques, may assist in defining the nature of this relationship.

Acknowledgements

This research was funded by National Institutes of Aging Grants # PO1 AGO4390 and #AG08812, the National Institutes of Health Grant #K24 HD01351, and the National Institute on Disability and Rehabilitation Research (Award #H133P990003). Dr Lipsitz holds the Irving and Edyth S. Usen and Family Chair in Geriatric Medicine at the Hebrew Rehabilitation Center for Aged.

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