Both static and dynamic postural controls are fundamental components of human activity. Aging has been associated with deterioration in postural control manifesting itself by an increase in postural sway (1,2) and a decrease in the voluntary movement capacity of the body’s center of gravity (3,4). Deterioration in these functions leads to a higher risk of falling, which in turn may increase the number of people with high levels of disability (4,5).

Recently, several studies have reported age-associated increases in muscle coactivation during dynamic movements (ie, walking and stair climbing: 6,7) and postural control (8,9), although evidence on whether coactivation is elevated during maximal effort contraction is inconclusive (10). Increased muscle coactivation in older adults is most commonly described as a compensatory mechanism to increase joint stiffness, which thereby enhances stability (6,7,8,9,11).

However, some researchers have pointed out negative effects of coactivation on postural control and movement, observing that excessive muscular coactivation increases postural rigidity (12,13). A rigid posture induced by strong muscle coactivation reduces the degrees of freedom to be organized by the postural control system (14) and may actually compromise the execution of voluntary or compensatory responses (12,15).

In an effort to prevent deterioration in postural control ability, patients undergo balance training in clinical settings. Multiple studies have reported on the effect of balance training on postural control ability. A systematic review of these studies has documented improvements in the ability to stand on one leg and to reach forward without losing balance in postural control tasks (16).

However, the effect of balance training on muscle coactivation during postural control in older adults remains unclear. Investigation of muscle coactivation changes after balance training thus could be an important step toward clarifying the mechanisms by which balance training might...
improve postural control. In addition, this information could serve as a useful reference for optimizing rehabilitation strategies in older people. The purpose of this study was to clarify the effect of balance training on muscle coactivation during postural control in older adults. In our previous cross-sectional study, muscle coactivation was negatively correlated with postural control ability in older adults (9). This result raises the possibility that the improvement in postural control after balance training correlates with decreased muscle coactivation.

We hypothesized that muscle coactivation during postural control in older adults would decrease after balance training.

**METHODS**

**Participants**

Resident subjects were recruited using advertising literature from three nursing homes. The inclusion criteria were set as follows: aged 65 or older, dwelling in a nursing home, able to walk independently (or with a cane), willing to participate in group exercise classes, and with minimal, if any, auditory or visual impairment. Oral and written explanations of the study were offered to the participants. Subjects were excluded if they had acute neurological impairment (stroke, Parkinson’s disease, paresis of the lower limbs), severe cardiovascular disease, severe cognitive impairment (Rapid Dementia Screening Test score of four points or less; 17), persistent joint pain, or severe musculoskeletal impairment (inability to participate in the training regimen).

Any subjects who were willing to participate and met the entry criteria were accepted into the study. Written informed consent was obtained from each participant in the trial in accordance with the Declaration of Human Rights, Helsinki, 1975. This research was approved by the Ethical Review Board of Kyoto University Graduate School of Medicine, Kyoto, Japan.

**Study Design and Randomization**

Randomization via computer-generated random numbers was performed in blocks of four subjects stratified according to 10 m walking time. The 48 subjects were randomly assigned to the intervention group (n = 24) or the control group (n = 24). The subject assignments were not blinded to the research staff members who performed measurements during the postural control tasks.

**Sample Size**

Our pilot study investigated the immediate effect of repetitive balance training on muscle coactivation during wobble board standing in young adults and demonstrated that muscle coactivation during standing on the wobble board decreased 12% (cocontraction index; CI), which is accompanied by significant improvement in balance ability (18). The longer intervention period would provoke greater reduction of muscle coactivation in older adults. Therefore, we aimed to detect a 15% (CI) reduction in muscle coactivation in the intervention group compared with the control group. Furthermore, based on the data from our pilot study, we estimated the standard deviation of this degree of change in coactivation to be 14% (CI; 18). Based on these assumptions, a total sample size of 34 participants would be required with alpha set at 0.05, beta at 0.2, and a given power of 0.8. We increased the sample size to a minimum of 44 to account for an anticipated dropout rate of 20%.

**Intervention**

Subjects assigned to the intervention group received 40 minutes of group balance training sessions twice a week for 8 weeks focused on improving postural control ability. There were three subgroups in the intervention group. Each subgroup consisted of 4–10 people. Exercise classes, which were supervised by a physical therapist, consisted of 10 minutes of warm-up and stretching exercises followed by 30 minutes of balance training. Subjects with a risk of falling during the exercise session were permitted to hold the back of a chair to ensure their postural stability. The level of exercise difficulty was adjusted according to the ability of each subject by altering reach distance or base of support. Subjects who held a chair during exercise were instructed to decrease the assistive level when they were able to acquire postural ability without losing balance during the tasks.

The control group did not receive any intervention but were simply instructed to spend their time as usual during the intervention phase. In order to avoid contamination during the exercise period, the training location was set at a place where the subjects in the control group were not usually visiting when the training session was under way. Subjects assigned to the control group were offered the same exercise program after the conclusion of the study period.

**Warm-up and Stretching**

Before training, subjects participated in warm-up and stretching exercises consisting of finger joint movement, bending fingers backward, shoulder rotation, waist rotation, upward stretching, lateral bending of the trunk, forward bending, and lower leg stretching. Lower leg stretching was targeted on the hamstrings (bending trunk forward with extended knee) and gluteus maximus (hip flexion using arms) in a sitting position.

**Balance Training**

Balance training consisted of standing on one leg, tandem standing, shifting weight laterally from one foot to another (19), anterior–posterior or lateral weight shifting, and reaching forward and laterally (20). Subjects were instructed to...
maintain their position for 5 or more seconds during each task and performed three sets of these exercises during each training session. The training sessions also included stepping forward and sideways in which the instructor called out each stepping direction. Subjects were provided a wall or chair for safe support as needed.

Testing Procedures and Protocol

The postural control tasks selected for testing consisted of postural sway during quiet standing, functional reach (4,21), and functional stability boundary (forward and backward; 3) because similar movements are performed frequently during activities of daily living.

Postural Sway

Postural sway during quiet standing was measured by a force plate (Kistler 9286 force platform, Kistler Instruments Inc., Amherst, NY). Signals were sampled at 20 Hz and processed by a low-pass filter (6 Hz cut-off frequency). The participants stood on the force plate with their feet together and then were asked to gaze at a mark at eye level while maintaining a stationary posture as symmetrically as possible. Quiet standing balance was registered for a period of 10 seconds, from which the root mean square area was calculated. Electromyography (EMG) activity was also recorded for the first 3 seconds of quiet standing. The intra-class correlation coefficient (ICC1,1) for the root mean square area was .72 in this study, which indicates “substantial” reliability (22).

Functional Reach

Functional reach was defined as the difference between arm’s length and maximal forward reach (4). The position of the fingertip was determined with the shoulder flexed at 90° along a wall. Subjects then were instructed to reach as far forward as possible without moving their feet, thus moving the center of gravity forward over a fixed base, and to maintain this maximal forward reach position for 3 seconds for EMG measurements.

Functional Stability Boundary

Functional stability boundary tasks were performed on the force plate (3). Standing with their heels on a line 10 cm anterior to the posterior edge of the plate, subjects were instructed to standstill for 5 seconds and then to shift their body weight first toward their toes and then toward their heels over the largest possible amplitude while maintaining full contact between their feet and the plate (avoiding toes off or heels off). For each direction (forward and backward), the subject maintained their posture for 3 seconds for EMG measurements, from which the averaged peak center of pressure displacement from the initial position was calculated. The center of pressure displacement for each subject was normalized by the length of that subject’s foot.

Additional Physical Function Characteristics

To calculate the 10 m walking time, which we utilized as the stratification variable, we had subjects perform walking trials at their preferred speed over a 12-m walkway, during which we measured the walking time for the middle 10 m (23). The timed up and go test (24) and the timed one-leg standing test for the dominant leg with eyes open were performed without EMG monitoring. The maximum duration of the one-leg standing test was set at 30 seconds.

EMG Recording

EMG data were collected with the Telemyo 2400 (Noraxon USA Inc., Scottsdale, AZ). The skin of the dominant leg was shaved over the fibula head, tibialis anterior, and soleus (25) and then washed with alcohol. Bipolar surface electrodes (Ambu, Blue sensor M, Denmark) with a 2,0-cm interelectrode distance were placed on the skin around the probable motor point of the muscles (26). The ground electrode was affixed to the skin over the fibula head of the dominant leg. The EMG data were sampled at 1500 Hz.

EMG activity was recorded from the soleus and tibialis anterior while the subjects were performing maximal voluntary contractions (MVC; 27). The MVC of the soleus was obtained during maximal isometric plantar flexion, and maximal tibialis anterior activation was recorded during maximal isometric dorsiflexion of the ankle at 90° (anatomically neutral position). Strong verbal encouragement was given during every contraction to promote maximal effort. The EMG data from the MVCs were used to normalize the EMG amplitude (percent MVC) during the postural tasks. The MVCs were recalculated for the posttraining measurement.

Muscle Coactivation Analysis

The original raw EMG signal was band-pass filtered at 20–500 Hz. We computed the root mean square amplitude of the signal using a 50-ms window. The EMG of each muscle was then expressed as a percentage of the EMG value during the MVC.

To calculate the relative level of cocontraction of the tibialis anterior and soleus muscles, the CI was calculated using the method of Falconer and Winter (28). Specifically, the following equation was used:

\[ CI(\%) = 2I_{ant} / I_{total} \times 100. \]

\( I_{ant} \) is the area of the total antagonistic activity, calculated in accord with the following equation:

\[ I_{ant} = \int_{t_1}^{t_2} \text{EMG}_{\text{TA}}(t) dt + \int_{t_1}^{t_2} \text{EMG}_{\text{SOLE}}(t) dt, \]
where \( t_1 \) to \( t_2 \) denotes the period during which the tibialis anterior EMG is less than the soleus EMG, and \( t_2 \) to \( t_3 \) denotes the period during which the soleus EMG is less than the tibialis anterior EMG.

\[
I_{\text{total}} = \int_{t_1}^{t_2} \left[ \text{EMG}_{\text{tib}} + \text{EMG}_{\text{sole}} \right] (t) \, dt.
\]

The CI calculation was done by a staff member who was blinded to the subject assignments.

Testing Reliability

The test–retest interday reliability of EMG measurements related to electrode positioning was estimated by calculating ICC\(_{1,1}\). The ICC\(_{1,1}\) for the CI were as follows: 0.68 (95% CI: 0.36–0.86) for quiet standing, 0.75 (95% CI: 0.47–0.89) for functional reach, 0.86 (95% CI: 0.66–0.94) for functional stability boundary (forward), and 0.91 (95% CI: 0.78–0.96) for functional stability boundary (backward). These ICC values indicated substantial to “almost perfect” reliability for all measurements of CI (22). The ICC\(_{1,1}\) for CI without electrode repositioning during quiet standing was 0.93 (95% CI: 0.72–0.98), which indicates almost perfect reliability (22).

Statistics

The results were analyzed using an intention to treat analysis. Baseline characteristics of the intervention and control groups were compared to examine comparability of the two. The Kolmogorov–Smirnov Test was used to test the normality of distributions. Differences between groups were analyzed using the chi-square test for categorical variables, Student’s \( t \) test for continuous variables with normal distribution, and the Mann–Whitney \( U \) test for nonnormally distributed variables.

The effect of exercise on outcome measurements was analyzed using mixed design 2 × 2 group (intervention and control groups) × time (pretraining and posttraining) analysis of covariance. Baseline values were used as covariates in the analysis of covariance. Statistical significances in CI and muscle activation (percent MVC) were set at 0.0125 (0.05/4) because four tasks measured by EMG were included to assess the muscle coactivation. Post hoc Bonferroni tests were used to assess which group or time periods showed significant differences. Statistical significance in post hoc Bonferroni tests was also set at 0.0125 (0.05/4). \( P < .05 \) was considered statistically significant in analysis of covariance for physical functions. Data were entered and analyzed using SPSS (Windows version 12.0, SPSS, Inc., Chicago, IL). For all outcome measures, missing data values were imputed using mean values for each corresponding group.

Results

Study Population

Our initial pool of study subjects comprised 108 residents from three nursing homes, of whom 51 refused to participate in the study and nine did not meet the inclusion criteria. The remaining 48 individuals (mean age 83.0 ± 6.8 years) agreed to participate in the study and provided written consent. Of the 48 subjects who enrolled in the study, 40 (83%) completed the 8-week intervention phase and postintervention assessment: 20 in the intervention group (83%) and 20 in the control group (83%). The other eight subjects dropped out because of hospitalization due to chronic illness (two subjects) or absence on the day of the assessment for personal reasons (six subjects). After the imputation of missing data, we performed an intention to treat analysis on the 48 subjects.

Adherence to the Study Protocol

During the 8-week intervention phase, 16 exercise sessions were scheduled and all took place. Excluding the four who dropped out, the intervention group subjects attended an average of 14 sessions and had an overall attendance rate of 82% over the 8 weeks. No health problems, including cardiovascular or musculoskeletal complications, occurred during training sessions or testing.

Baseline Characteristics

Table 1 summarizes baseline data for the 48 subjects who completed the study. No significant differences between the intervention and control groups were observed in any of the characteristics examined, including age, height, and weight.

Effects of Intervention on Muscle Coactivation and Muscle Activity

After the 8-week intervention phase, CI values in the intervention group showed a significant decrease compared with preintervention values for functional reach (pre: 33.8% ± 21.4%, post: 24.1% ± 15.9%, group × time interaction: \( F[1,46] = 10.311, p = .002, \eta^2 = 0.186 \)). Although there was a significant group-by-time interaction in CI during functional stability boundary (forward; pre: 41.3% ± 25.7%, post: 33.2% ± 20.4%, group × time interaction: \( F[1,46] = 7.226, p = .010, \eta^2 = 0.138 \)), no significant decrease was found in CI during this task by post hoc test (\( p \geq .0125; \text{Figure 1} \)).

CI did not significantly change in standing (pre: 51.3% ± 24.4%, post: 40.1% ± 19.9%, interaction: \( F[1,46] = 4.975, p = .031, \eta^2 = 0.100 \)) or functional stability boundary
Table 1. Baseline Characteristics (mean ± SD) of Subjects

<table>
<thead>
<tr>
<th>Intervention Group (n = 24)</th>
<th>Control Group (n = 24)</th>
<th>p Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age, y</td>
<td>81.0 ± 6.9</td>
<td>81.6 ± 6.4</td>
</tr>
<tr>
<td>Body weight, kg</td>
<td>51.3 ± 8.4</td>
<td>52.4 ± 7.6</td>
</tr>
<tr>
<td>Height, cm</td>
<td>150.7 ± 6.8</td>
<td>153.1 ± 6.5</td>
</tr>
<tr>
<td>Female, %</td>
<td>83.0</td>
<td>92.0</td>
</tr>
<tr>
<td>Medication, n (%)</td>
<td>20 (83)</td>
<td>20 (83)</td>
</tr>
<tr>
<td>Number of comorbidities, n (%)</td>
<td>.604</td>
<td></td>
</tr>
<tr>
<td>≧3</td>
<td>20 (83)</td>
<td>18 (75)</td>
</tr>
<tr>
<td>4–6</td>
<td>4 (17)</td>
<td>5 (21)</td>
</tr>
<tr>
<td>≧7</td>
<td>0 (0)</td>
<td>1 (4)</td>
</tr>
<tr>
<td>Falls in the last year, n (%)</td>
<td>4 (17)</td>
<td>6 (25)</td>
</tr>
<tr>
<td>Postural sway area, cm²</td>
<td>21.3 ± 4.6</td>
<td>19.8 ± 6.3</td>
</tr>
<tr>
<td>Functional reach, cm</td>
<td>19.5 ± 6.8</td>
<td>18.8 ± 7.1</td>
</tr>
<tr>
<td>Functional stability boundary for forward, %</td>
<td>13.1 ± 5.4</td>
<td>11.7 ±5.8</td>
</tr>
<tr>
<td>Functional stability boundary for backward, %</td>
<td>1.7 ±0.9</td>
<td>1.8 ±1.0</td>
</tr>
<tr>
<td>10 m walking time, s</td>
<td>11.3 ± 2.5</td>
<td>11.5 ± 2.9</td>
</tr>
<tr>
<td>TUG, s</td>
<td>8.3 ± 2.2</td>
<td>9.2 ± 2.2</td>
</tr>
<tr>
<td>One-leg stance, s</td>
<td>14.4 ± 11.5</td>
<td>10.1 ± 9.9</td>
</tr>
</tbody>
</table>

Notes: Significance was tested using the $^2$ test for categorical variables, Student’s $t$ test for continuous variables, and the Mann–Whitney $U$ test for nonnormally distributed variables. TUG = timed up and go.

* $p < .05$.

Effects of Intervention on Physical Function

Functional reach, functional stability boundary (forward and backward), timed up and go, and one-leg stance significantly improved after the 8-week intervention (group × time interaction: $p < .05$; Table 3). No significant improvements in other postural control abilities or physical functions were observed.

Discussion

Our study results provided evidence partially supportive of our hypothesis that muscle coactivation during postural control would decrease after balance training. Specifically, we found that balance training decreased muscle coactivation in dynamic postural control tasks (ie, functional reach and functional stability boundary for forward), although no significant group-by-time interactions in tibialis anterior activity were observed during functional stability boundary (backward), but not during functional reach ($p = .055$; Table 2). Although there was no significance across the board, tibialis anterior activity remained constant or decreased, and soleus activity increased during the tasks such as standing, functional reach, and functional stability boundary for forward in the intervention group (Table 2). Tibialis anterior activity increased during functional stability boundary (backward) in the intervention group ($p < .0125$).

Significant group-by-time interactions in tibialis anterior activity were observed during functional stability boundary (backward), but not during functional reach ($p = .055$; Table 2). Although there was no significance across the board, tibialis anterior activity remained constant or decreased, and soleus activity increased during the tasks such as standing, functional reach, and functional stability boundary for forward in the intervention group (Table 2). Tibialis anterior activity increased during functional stability boundary (backward) in the intervention group ($p < .0125$).
significance was seen in functional stability boundary for forward in post hoc test. These findings suggest that exercise may be able to modify redundant muscle coactivation during dynamic postural control in older adults.

Previous studies have reported that older adults showed greater muscle coactivation during postural control in static or dynamic conditions compared with young adults (9,25,29). A previous study of ours showed that muscle coactivation was significantly higher in older adults with low postural control ability than in older adults with high postural control ability (9). All these studies described coactivation with resultant ankle joint stiffness as a compensatory strategy to maintain postural stability. The result of our current study suggests that trained subjects could maintain dynamic postural stability without the stiffening of their ankle joint associated with higher muscle coactivation.

Functional reach, functional stability boundary (forward), timed up and go, and one-leg stance time significantly improved in the intervention group. The improvements we observed in functional reach and functional stability boundary were associated with decreased muscle coactivation after the intervention. Our previous cross-sectional study showed that balance ability is negatively related to muscle coactivation during postural control in older adults (9). The result of the present study suggested that changes in muscular coactivation are related with changes in postural control ability. However, the postural sway area and CI during standing did not improve in the intervention group. The lack of the effect on postural sway might be attributable to training frequency and duration. The training program in this study was 40-minute balance sessions conducted twice a week for 8 weeks. A previous study proposed an effective

<table>
<thead>
<tr>
<th>Tasks</th>
<th>Muscle</th>
<th>Pre (n = 24)</th>
<th>Post (n = 24)</th>
<th>Group x Time</th>
<th>Effect Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Functional stability boundary for forward, %</td>
<td>TA</td>
<td>Intervention group</td>
<td>16.0 ± 15.0</td>
<td>16.4 ± 14.5</td>
<td>0.513</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>13.2 ± 12.0</td>
<td>17.3 ± 12.1</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>SOL</td>
<td>Intervention group</td>
<td>48.6 ± 19.2</td>
<td>61.1 ± 13.2</td>
<td>2.706</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>45.3 ± 19.0</td>
<td>52.9 ± 19.7</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Notes: Significance was tested using mixed design analysis of covariance. SOL = soleus; TA = tibialis anterior.

*p < .0125 F value; F value is a test statistic to decide whether the sample means are within the sampling variability of each other. The null hypothesis is rejected when the F value is large. η², effect size (η²) is a measure of the strength of the relationship between the two variables.

Table 3. Pre- and Posttraining Comparison of Postural Control Abilities and Physical Function

<table>
<thead>
<tr>
<th>Tasks</th>
<th>Muscle</th>
<th>Preintervention (n = 24)</th>
<th>Postintervention (n = 24)</th>
<th>Group x Time</th>
<th>Effect Size</th>
<th>η²</th>
</tr>
</thead>
<tbody>
<tr>
<td>Postural sway area, cm²</td>
<td>TA</td>
<td>Intervention group</td>
<td>1.7 ± 0.9</td>
<td>1.3 ± 1.0</td>
<td>0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>1.8 ± 1.0</td>
<td>1.43 ± 0.87</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Functional reach, cm</td>
<td>TA</td>
<td>Intervention group</td>
<td>21.3 ± 4.6</td>
<td>25.2 ± 4.0</td>
<td>19.808*</td>
<td>0.306</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>19.8 ± 6.3</td>
<td>19.6 ± 6.0</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Functional stability boundary for forward, %</td>
<td>TA</td>
<td>Intervention group</td>
<td>19.5 ± 6.8</td>
<td>25.2 ± 6.8</td>
<td>28.777*</td>
<td>0.390</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>18.8 ± 7.1</td>
<td>17.9 ± 7.4</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Functional stability boundary for backward, %</td>
<td>TA</td>
<td>Intervention group</td>
<td>12.7 ± 5.2</td>
<td>15.7 ± 6.2</td>
<td>3.886</td>
<td>0.079</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>11.7 ± 5.8</td>
<td>12.3 ± 4.6</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>10 m walking time, s</td>
<td>TA</td>
<td>Intervention group</td>
<td>11.3 ± 2.5</td>
<td>10.5 ± 2.1</td>
<td>3.926</td>
<td>0.080</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>11.5 ± 2.9</td>
<td>11.7 ± 3.4</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>One-leg stance, s</td>
<td>TA</td>
<td>Intervention group</td>
<td>14.4 ± 11.5</td>
<td>16.1 ± 10.0</td>
<td>9.737*</td>
<td>0.178</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>10.1 ± 9.9</td>
<td>8.3 ± 6.8</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TUG, s</td>
<td>TA</td>
<td>Intervention group</td>
<td>8.3 ± 2.2</td>
<td>7.5 ± 1.7</td>
<td>7.420*</td>
<td>0.142</td>
</tr>
<tr>
<td></td>
<td>Control group</td>
<td>9.2 ± 2.2</td>
<td>9.2 ± 2.3</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Notes: TUG = timed up and go. Significance was tested using mixed design analysis of covariance.

*p < .05 F value; F value is a test statistic to decide whether the sample means are within the sampling variability of each other. The null hypothesis is rejected when the F value is large. η², effect size (η²) is a measure of the strength of the relationship between the two variables.
balance training program conducted three times per week for 1 hour for 3 months (16). Longer programs may produce greater effects on physical function and muscle coactivation. If postural control ability in functional stability boundary for backward had improved in the present study, coactivation during this task might also have changed after training.

Our study found that tibialis anterior activity during postural sway, functional reach, and functional stability boundary for forward had a tendency to remain constant or decrease modestly in the intervention group. On the other hand, soleus activity was subject to increase especially in the intervention group. Functional reach and functional stability boundary (forward) involve a forward movement of the center of pressure, which requires increased plantar flexor torque at the ankle joint to control posture, while the tibialis anterior plays a role of antagonist. The results of the study suggest that the balance training led to an increase in agonist (soleus) activity and a decrease or maintenance in antagonist (tibialis anterior) activity. Carolan and Cafarelli (30) have measured muscle activation in the biceps femoris during knee extension and found that as knee extensor strength increased, biceps femoris activity also increased. This finding indicates that greater effort to recruit agonist induces increased muscle coactivation in the antagonist. In the present study, after balance training, subjects could recruit the agonist (soleus) without enhancement of the antagonist (tibialis anterior). This decrease or conservation in the antagonist activity, in turn, could help optimize the work of the agonist (soleus) during postural control. Postural control exercise therefore could potentially lead older adults to more efficient postural control strategies without increasing muscle coactivation.

Not much has been done to define the contribution of muscle coactivation during physical activity or in falls in older adults, although the change of muscle coactivation strategy has been reported during postural control or joint movements (6,8,31,32). The present study showed decreased muscle coactivation during postural control after training. However, it remains unclear from our study how neural adaptation (ie, decrease of muscle coactivation) contributes to fall avoidance or changes with increased physical activity. Further studies should investigate the effect of decreased muscle coactivation after training on functional outcomes other than balance ability.

This study has several limitations. First, our study examined only the ankle joint and thus provides no information on postural strategies for the knee or hip joint. Second, our data did not measure changes in kinematics, which therefore did not enable us to evaluate any possible relationship between muscle coactivation and kinematics during postural tasks in older adults. It is a matter of further study to clarify the effect of kinematic change on muscle coactivation with information from multiple joints. Third, the information on subject assignment was not blinded to the research staff members during the postural control tasks. Possible bias cannot be ruled out in this study design. However, the effect of bias on coactivation, which is the main outcome in this study, is not considered significant because it would be difficult for the research staff members to manipulate the coactivation intentionally. Also, the CI calculation process was done in a blinded way. Finally, the ICC for postural control ability (sway area) during quiet standing was not very high (0.72) because of the short measurement time (10 seconds). This might have had an effect on the results of the postural sway area. However, ICC for CI without electrode repositioning during quiet standing was very high (0.93). We conclude that the results for muscle coactivation were not influenced very much by the measurement time in quiet standing.

**Conclusions**

Our study found that coactivation during postural control decreases after balance training in older adults, which can be associated with improvement of postural control ability. Postural control exercise could potentially lead older adults to develop more efficient postural control strategies without increasing muscle coactivation. Further research is needed to clarify the contribution decreased muscle coactivation makes to balance ability and to other functional outcomes such as fall prevention or increased physical activity.

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**References**


