

Effects of Aging on Feedforward Postural Synergies

by

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We investigated the effects of aging on postural muscles covariate patterns prior to voluntary perturbations. Nine healthy young and nine older subjects were instructed to release a load in a self-paced manner. The results of cross-correlation analyses showed that the average time lag corresponding peak correlation coefficient between trunk flexor and extensor muscles in the older group was significantly shorter, compared to that in the young group. The results of principal component analysis showed that the co-contraction Muscle-modes in the older group were observed more frequently than those in the young group. These results indicate that the older group showed changes in the anticipatory postural muscle co-variation, suggesting the transition from reciprocal to co-activation pattern with aging.

Keywords: aging, postural stability, co-contraction

Introduction

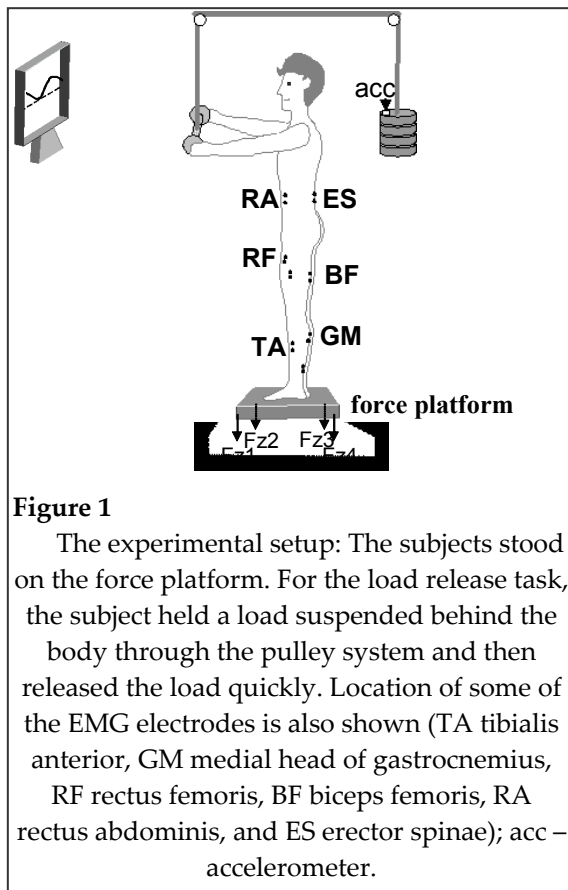
The postural control system has been shown to decline with age (Woollacott et al. 1986). Feed-forward postural control of performance stability correlates with a subsequent reduction in balance loss (Man'kovskii et al. 1980). The anticipatory postural control represents changes in the center of pressure (COP) and the activity of postural muscles prior to the initiation of voluntary motor actions (Aruin & Latash 1995; Belen'kii et al. 1967). With age, COP instability (Bugnariu & Svestrup 2006; Maki 1993), slowing of onset of electromyographic activity (Bleuse et al. 2006; Inglin & Woollacott 1988) and large variability in the sequence of postural

muscle activation (Garland et al. 1997) increase in anticipatory postural adjustments.

Muscle synergies (Ivanenko et al. 2004; Torres-Oviedo & Ting 2007) or muscle-modes (M-modes) (Krishnamoorthy et al. 2004; Wang et al. 2006) have been defined as groups of muscles activated in synchrony or with fixed time delays for use by the central nervous system based on a hierarchical scheme (Ting LH & Macpherson 2005). We have reported recently that in young subjects, M-mode composition changed to co-contraction M-mode in challenging postural tasks, but that recovered to reciprocal M-mode with practice (Asaka et al. 2008). Few investigations on age-related differences of muscle activation in agonist-antagonist pairs have been reported. Benjuya and colleagues (2004) suggested that elderly used co-contraction around

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the ankle in order to deal with changing conditions and sensory inputs. Woollacott and colleagues (1993) suggested that older subjects showed more frequent activation of the quadriceps in addition to hamstrings activation than young subjects prior to quick arm raising. However, the effects of aging on M-mode have not been documented. The present study therefore investigated the effects of aging on postural muscles covariate patterns prior to voluntary perturbations. We hypothesized that older subjects showed more co-activation muscle synergies compared to young subjects.

Methods

Subjects

Subjects comprised 9 young adults (4 men, 5 women; mean age, 22.3±1.2 years old) and 9 older adults (5 men, 4 women; 66.4±3.5 years). Mean height and body mass of subjects were 161.5±7.6 cm and 56.3±9.3 kg in young subjects and 160.8±7.5 cm and 65.3±8.6 kg in older subjects, respectively. All subjects were healthy, with no known neurological or motor disorders that were established by questioning by a medi-

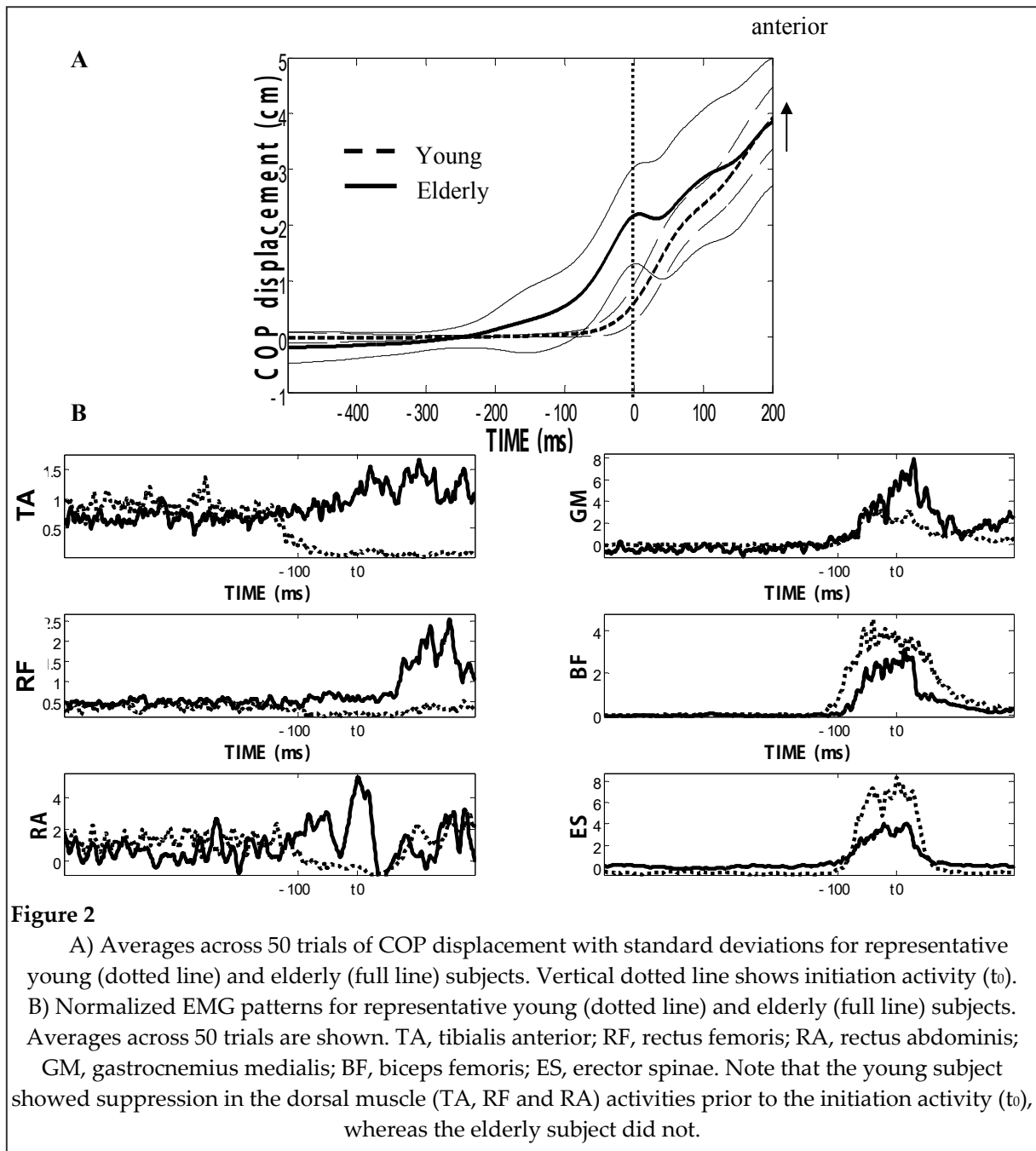
cal doctor. All of the older subjects were independent and capable of self-care and normal household tasks. All subjects provided informed consent based on the procedures approved by the ethics committee at Hokkaido University School of Medicine.

Apparatus

A force platform (Type 9286A; Kistler, Switzerland) was used to record shear force (F_y) in the anteroposterior (AP) direction and the vertical component of the reaction force (F_z). Disposable self-adhesive electrodes (Ambu, Denmark) were used to record surface electromyography (EMG) of the following muscles: tibialis anterior (TA); medial head of gastrocnemius (GM); rectus femoris (RF); biceps femoris (BF); rectus abdominis (RA); and erector spinae (ES). The electrodes were placed on the left side of the subject's body over the muscle bellies, with their centers 3 cm apart. In addition, a reference electrode was attached to the lateral aspect of the fibula. EMG signals were transmitted wirelessly, then amplified ($\times 2000$) and band-pass filtered (0.5-250 Hz). All signals were digitized at a sampling frequency of 1000 Hz with 12-bit resolution. A uni-directional accelerometer (TA-512G; Nihonkoden, Japan) was taped to the load and used for trial alignment (time zero: t_0).

Procedures

The task was used based on a previous study (Krishnamoorthy et al. 2004; Wang et al. 2006). Subjects stood barefoot on the force platform with feet parallel, about hip-width apart. Subjects were asked to hold a load of approximately 5% body weight per-1Kg suspended behind the body through a pulley system by pressing with the hands of the extended arms on the two round ends of a light horizontal handle. The pulley system redirected load action upwards (Figure1). Subjects were instructed to release the load in a self-paced manner with a quick, large amplitude and bilateral shoulder abduction movement after beep generated by a personal computer. Note that the composition of the modes was similar across all the cited studies and methods of their identification. In total, each subject performed 50 trials to principal component analysis (see later). Prior to each trial, the subject was required to adjust their



posture to make sure that the cursor corresponding to the instantaneous position of the difference (F_z) between the vertical force components anterior to the subject ($F_{z1} + F_{z2}$) and the posterior components of the vertical force ($F_{z3} + F_{z4}$) was on the horizontal target line corresponded to $\Delta F_z = 0$, shown on a monitor placed about 2 m from the subject, at eye level (Asaka et al. 2008). In addition, subjects were reminded to keep the body upright. A rest period of about 5 min was given after 25 trials within the task. Subjects reported no fatigue over the duration of the experiment. In addition, 3 control trials were performed. In the first control trial, sub-

jects were asked to stand as quietly as possible for 10 s. In the second and third control trials, subjects were instructed to stand upright and hold the load in front of the body and behind the body through the pulley system for 10 s by pressing with the hands of the extended arms on the light horizontal handle. These data were used for EMG normalization.

Data processing

All signals were processed off-line using LabView-8 (National Instruments, USA) and MatLab 7.3 software (MathWorks, USA). EMG signals were rectified and filtered with a 50-Hz

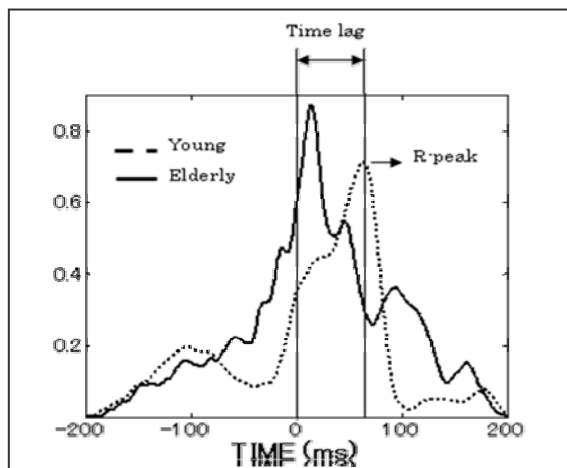


Figure 3 Correlation coefficients computed from cross-correlations between rectus abdominis and erector spinae muscles from -200 ms to t_0 for representative young (dotted line) and elderly (full line) subjects. Time lag is defined at the moment of peak correlation coefficient (R-peak). Positive time lag indicates earlier activation in dorsal muscle compared to ventral muscle activation.

low-pass, second-order, zero-lag Butterworth filter. Rectified EMG signals were integrated from 100 ms prior to t_0 up to t_0 ($JEMG$). These values were corrected by EMG integrals quantified over 100-ms intervals from 400 ms prior to t_0 ($\Delta IEMG$). To allow comparison across muscles and subjects, $\Delta IEMG$ indices for the dorsal muscles (GM, BF, and ES) were normalized by the EMG integrals ($IEMG_c$) in the control trial where the load was acting downward, while $\Delta IEMG$ indices for the ventral muscles (TA, RF, and RA), were normalized by the EMG integrals ($IEMG_c$) in the control trial where the load was

acting upward.

To explore time-dependent covariates in flexor and extensor muscle activity in the major joints of the lower extremities and trunk, cross-correlation analyses were run separately for EMGs in agonist-antagonist pairs over a time period from -200 ms to t_0 . For each trial, peak magnitude of the correlation coefficient (R-peak) and time lag (Δt) of R-peak were computed. In addition, for each subject, $\Delta IEMG_N$ data matrices were obtained with a size of 50x6 (50 rows corresponding to trials; 6 columns corresponding to muscles). The correlation matrix among $\Delta IEMG_N$ was subjected to principal component analysis, using procedures from SPSS 14.0 software (SPSS, USA). The factor analysis module with principal component extraction was employed. For each subject, obtained 6 factor loadings per principal component (PC) were then considered. The first three PCs were selected for further analysis. This was based on two criteria: examination of scree plots; and having at least one muscle significantly loaded per PC with the absolute value of the loading factor over 0.5 (Hair et al. 1995). We addressed these PCs as M-modes and assumed that magnitudes of (coefficients at) M-modes were manipulated by the controller to produce anterior-posterior shifts in the COP (COP_{AP}). In other words, M-modes represent unitary vectors in the muscle activation space that can be recruited by the controller with different magnitudes.

We used unpaired t-tests to analyze possible differences in values of average time lags and peak correlation coefficients between groups. In

Table 1 Cross correlation R-peak and corresponding time lags between flexor and extensor muscle activities in the ankle, knee and hip joints from -200 ms to t_0 .

	Young		Elderly	
	\bar{x}	SD	\bar{x}	SDF
Ankle				
R-peak	0.71	± 0.06	0.70	± 0.02
time lag (ms)	27.2	± 37.5	9.6	± 10.6
Knee				
R-peak	0.65	± 0.03	0.66	± 0.03
time lag (ms)	56.3	± 23.4	41.9	± 17.7
Hip				
R-peak	0.70	± 0.04	0.70	± 0.03
time lag (ms)	57.3	± 39.4	30.3	$\pm 15.0^*$

Values are average \pm S.D. * Significantly shorter than young group at $P < 0.05$.

addition, the Mann-Whitney test was used on the number of the two types of M-modes (detail see Results) across groups. Statistical significance was set at the $P < 0.05$ level.

Results

Older subjects showed greater average and variability in COP_{AP} displacement than young subjects, both prior to and after perturbations (Figure 1A). Figure 1B shows normalized EMG signals averaged across 50 trials for a representative subject in each group. The young subject showed suppression in the dorsal muscle (TA, RF and RA) activities prior to the initiation activity (t_0), whereas the older subject did not.

An example of correlation coefficients computed from cross-correlations between rectus abdominis and erector spinae muscles from -200 ms to t_0 for a representative young and a representative older subject is shown in Figure 2. Time lag is defined at the moment of peak correlation coefficient (R-peak). Table 1 shows average R-peak and time lag between flexor and extensor muscle activities in the ankle, knee and hip joints. Time lag is defined at the moment of peak correlation coefficient (R-peak). Average time lag at the moment of peak correlation coefficient between hip flexor and extensor muscles was significantly shorter in the older group than in the young group ($P < 0.05$). Average time lags between ankle and knee flexor and extensor muscles were tend to be shorter in the older group than in the young group (P values were 0.06 and 0.07, respectively). Peak correlation coefficients between ankle, knee and hip flexor and extensor muscles did not differ significantly

between groups ($P > 0.05$).

On average, the first three PCs accounted for 71.1% of total variance (30.4%, 21.6% and 19.1%, respectively) in the young group, and 71.9% of total variance (31.5%, 22.2% and 18.2%, respectively) in the older group. Unpaired t -test using the factor *Group* (young and elderly) performed on the amounts of variance explained by individual PCs and on total amount of variance explained by principal component analysis showed no significant effects. Across all subjects, the first three principal components accounted for more than 16.7% of variance and were able to classify the data into the two basic types: "reciprocal" M-modes; and "co-contraction" M-modes (Krishnamoorthy et al. 2004). There were two types of reciprocal M-modes, "push-back" (GM, BF, ES) and "push-forward" (TA, RF, RA). There were three types of co-contraction M-modes, "co-contraction at the hip" (ES and RA), "co-contraction at the knee" (BF and RF), and "co-contraction at the ankle" (GM and TA). When only one muscle loaded significantly on one of the first three PCs, this mode was not classified as any of those mentioned, but addressed as a "Singular" mode. Table 2 shows the number of occurrences of different M-modes. Co-contraction M-modes were observed more frequently in the elderly group than in the young group ($P < 0.05$), while reciprocal M-modes occurred less frequently in the older group than in the young group ($P < 0.05$).

Discussion

The goal for anticipatory postural control is to reduce balance loss with respect to perturbations of posture. The ability to maintain stability appears to depend on very much on the postural muscle activation. Our results clearly demonstrate that the short time lag in the elderly group was partly due to the fact that they showed an approximately simultaneous (co-activation) pattern. Co-contraction M-modes were observed more frequently in the older group than in the young group. These results indicate that the older group showed changes in anticipatory postural muscle co-variation, which suggesting a transition from reciprocal to more co-activation pattern.

Co-contraction patterns were defined as those involving high loading factors with the

Table 2
The number of occurrences of different M-modes

M-modes Pattern	Young	Elderly
Reciprocal		
Push-back	6	4
Push-forward	7	3
Co-contraction		
Hip	1	4
Knee	1	4
Ankle	2	1
Singular	10	11

Data across all subjects are presented. Singular: one muscle loaded significantly on a mode

same sign within the same M-mode for two muscles with opposing actions at one of the major joints. Such patterns were mostly observed in individual data in ES & RA and BF & RF. The total number of co-contractions observed in individual results summed over the older group was greater than that of the young group. The co-contraction pattern may be viewed as the means of increasing stiffness of the postural joints and stabilizing the COP displacement to the upcoming perturbation to compensate for changing neuromuscular functions with aging (Wolfson 1997). For elderly, the more efficient 'push-back' and 'push-forward' M-modes are regarded less safe to negotiate perturbation. The observation of co-contraction M-modes fits well Carpenter's study that has reported the stiffness strategy was adopted in a situation accompanied by increased postural threat (Carpenter et al. 2001). The co-contraction M-modes have been reported for challenging postural conditions (Krishnamoorthy et al. 2004; Asaka et al. 2008) and pre-programmed reactions (Robert & Latash 2008) in young subjects. Our study demonstrates that it also shows changes with age. This change in postural control leads to irreproducibility of important performance variables, such as COP, which will then violate the balance and initiate a fall of the body.

In the current study, the time lag at the moment of peak correlation coefficient between the hip flexor and extensor muscles was significantly shorter in the older group than in the young group (Table 1). In addition, the co-contraction M-modes at the hip and knee joints were observed more frequently than those at the ankle joint in the elderly group (Table 2). These results, therefore, suggest that the proximal groups of muscles in older might tend to change into co-activation pattern under the disturbed postural condition comparing to those of the young group. Over-reliance on hip joint action has been found when the elderly subjects respond to unexpected postural disturbances (Manchester et al. 1989). Amiridis and colleagues' study (2003) also showed that increasing difficulty postural demands during quiet standing increased hip muscle activities in the elderly, whereas this finding was not noted in the young adults. Note that elderly persons

typically show increased postural sway (Melzer et al. 2004), a reduced perception of stability limits and an atypical selection of adequate sources of sensory information used for postural stabilization (Hay et al 1996; Redfern et al. 2001). They also show anticipatory postural adjustments that occur later, that is closer in time to the action initiation, and are of a smaller magnitude (Inglin & Woollacott 1988; Rogers et al. 1992). Therefore, daily tasks with self-induced perturbation that are easy to the young people would be perceived by the central nervous system of elderly persons as challenging. Taken together, the possible explanations for the greater hip dependence shown in the elderly could be the greater motor unit loss in distal compared to proximal muscles (Doherty et al. 1993) and insufficient somatosensory contribution from the distal lower limb and foot (Horak et al. 1990).

Dates back to Gelfand and Tsetlin's (Gelfand & Tsetlin 1966) seminal work, the concept of M-modes is based on the idea of a hierarchical control of complex, multi-muscle action and has been assumed to be important for postural tasks (Wang et al. 2006). Within this view, M-modes are at the lower level of the hierarchy to stabilize and ensure proportional involvement of muscles within a certain group. The central neural controller is able to rearrange M-mode composition with regard to the perturbation based on a safety-efficacy trade-off. The composition of M-modes changes in different time intervals following a perturbation (Robert & Latash 2008) and with practice (Asaka et al. 2008). Elderly who commonly have movement constraint of the joints and muscles showed significant co-contraction M-modes, which suggests that there was a central adjustment for a safety-efficacy trade-off. Our study reveals that there are co-contraction M-modes, namely M-modes uniting muscle pairs with opposing actions at major leg joints-co-contraction modes, for the elderly in postural perturbation. We conclude that different sets of M-modes are applied to ensure stable COP trajectories.

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