Adaptation and vision change the relationship between muscle activity of the lower limbs and body movement during human balance perturbations

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Running title:
Correlation between EMG activity and body movements.

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Abstract

Objective: Investigate the relationship between changes in lower limb EMG root mean square (RMS) activity and changes in body movement during perturbed standing. Specifically, linear movement variance, torque variance and body posture were correlated against tibialis anterior and gastrocnemius RMS EMG activity during perturbed standing by vibration of the calf muscles.

Methods: Eighteen healthy participants (mean age 29.1 years) stood quietly for 30s before vibration pulses were randomly applied to the calf muscles over a period of 200s with eyes open or closed. Movement variance, torque variance and RMS EMG activity were separated into five periods, thereby allowing us to explore any time-varying changes of the relationships.

Results: Changes of tibialis anterior muscles EMG activity were positively correlated with changes in linear movement variance and torque variance throughout most of the trials, and negatively correlated with some mean angular position changes during the last two minutes of the trials. Moreover, the initial changes in Gastrocnemius EMG activity were associated with initial changes of mean angular position. Additionally, both tibialis anterior and gastrocnemius muscle activities were more involved in the initial control of stability with eyes closed than with eyes open.

Conclusions: Visual information and adaptation change the association between muscle activity and movement when standing is perturbed by calf muscle vibration.

Significance: Access to visual information changes the standing strategy to calf muscle vibrations. Training evoking adaptation could benefit those susceptible to falls by optimizing the association between muscle activities and stabilizing body movement.

Key Words: Motor control; Postural balance; Kinematics; Sensory perception; Learning and memory

1. Introduction

Everyday experience suggests that we are able to improve multiple motor skills through practice and this is more commonly termed adaptation. In current approaches to motor learning, adaptation is viewed as a process in which prediction errors result in proportional changes in parameter estimates (Krakauer et al., 2006). More recently, studies have indicated that adaptation through motor learning can be applied to the human standing posture during balance perturbations (Fransson et al., 2007b; Fujiwara et al., 2007). As such, repeatedly perturbing balance could be one way of decreasing the number of falls in those at risk. The maintenance of the human standing posture depends on the availability and accuracy of somatosensory (muscle, joint, skin and pressure receptors), visual and vestibular sensory inputs and descending commands from the central nervous system (CNS) (Akram et al., 2008). When the information from one or more of the sensory inputs becomes unreliable, a re-weighting occurs as the CNS places an increased demand on the reliable system or systems (Oie et al., 2002). Hence, one way of perturbing balance is by altering the information from the somatosensory receptors through vibration of the calf muscles. By vibrating the calf muscles, body sway and torque at the ankles increases (Fransson et al., 2000; Ivanenko et al., 1999; Kavounoudias et al., 1998) and when the vibrations are repeated, the adaptations are evidenced as a decrease in movement variability and ankle torque (Fransson et al., 2007b).

In patients with suspected balance disorders, it is often useful to assess the muscle activity using electromyography (EMG), along with other recordings of balance such as exerted forces and body movements, to determine the severity of disorder or rehabilitation status. However, additional important information might also be gained by analysing the way EMG activity relates to body movement. Some authors have suggested that since angular acceleration is proportional to joint torque in single joint movements, there should be a clear relationship between kinematics and EMG activity (Gabriel, 2002; Soechting and Flanders, 1991). St-Onge and Feldman have also suggested that EMG activity of multiple muscles should correlate with the direction, magnitude and velocity of movement (St-Onge and Feldman, 2004). However, the relationship between muscle activity and exerted body movements might change through adaptation (Buchanen and Horak, 2003). In addition, the availability of key information from the visual system might also influence this relationship (Buchanen and Horak, 1999).

During balance perturbations, postural muscles in the lower extremities such as the tibialis anterior and gastrocnemius are responsible for initiating counteractive movements (Gollhofer et al., 1989). Two key goals of the present study, then, were to: (1) identify whether adaptation through balance training (i.e. repeated calf muscle vibration) affects the relationship between tibialis anterior and gastrocnemius EMG activity and body movement, recorded as segmental body movements, body posture and exerted torque to the support surface and (2) whether access to vision can affect this relationship. These goals would provide clinicians with information about the importance of the visual system and usefulness of repeated balance perturbations for rehabilitation. We hypothesize that there would be a correlation between tibialis anterior and gastrocnemius muscle activity and body movement and that this relationship might be affected by adaptation due to the online updating of motor performance (Pavol and Pai, 2002; Pavol et al., 2002) but should not be affected by the availability of visual information.

2. Methods and Materials

2.1 Subjects

Posturographic tests were performed on 18 healthy subjects (nine men and nine women; mean age 29.1 years, range 18-49 years; mean height 1.74 m, range 1.50-1.85 m; mean mass 73.4 kg, range 58.1-95.0kg). Subjects had no previous history of balance deficits, neurological disease or injury to the musculoskeletal system of the lower extremities, nor were they taking any medication and were instructed to refrain from alcohol for at least 24 hours prior to testing. Full, informed consent was obtained from all subjects before any experiments were performed in accordance to the Helsinki declaration of 1975.

2.2 Equipment

Vibration of amplitude 1.0mm and frequency of 85Hz was produced using a DC-motor (Escap, Geneva, Switzerland) with a 3.5g mass attached eccentrically to the spindle within a cylindrical plastic coating of dimensions 6cm length and 1cm diameter. The vibrators were attached over the middle of the gastrocnemius muscles on both legs using elastic straps.

A force platform, developed in co-operation with the Department of Solid Mechanics, Lund Institute of Technology, recorded forces actuated at the feet with six degrees of freedom and with an accuracy of 0.5N. A customized computer program controlled the vibratory stimulation, and sampled the force platform data at 50Hz.

An ultrasound 3D-Motion Tracking system (Zebris™ CMS-HS Measuring System) recorded linear body movements at five anatomical landmarks. The first marker (‘Head’) was attached to the subject's cheekbone (os zygomaticum), the second marker (‘Shoulder’) to tuberculum majus, the third (‘Hip’) to the crista iliaca, the fourth (‘Knee’) to the lateral epicondyle of femur, and the fifth (‘Ankle’) to the lateral distal fibula head, see figure 1. Each marker registered its position in three directions, i.e., its anteroposterior, lateral and vertical position with a resolution of 0.4mm. The same Zebris™ system simultaneously recorded EMG activity of the tibialis anterior and gastrocnemius medialis muscle of both legs using eight active surface electrodes. The computer sampled the marker position data simultaneously at 50 Hz and EMG activity at 1500Hz.

The recorded data from the force platform and Zebris™ measurement systems were later synchronised by off-line time matching of a reference signal, which had been simultaneously sampled by both measurement systems.

2.3 Procedure

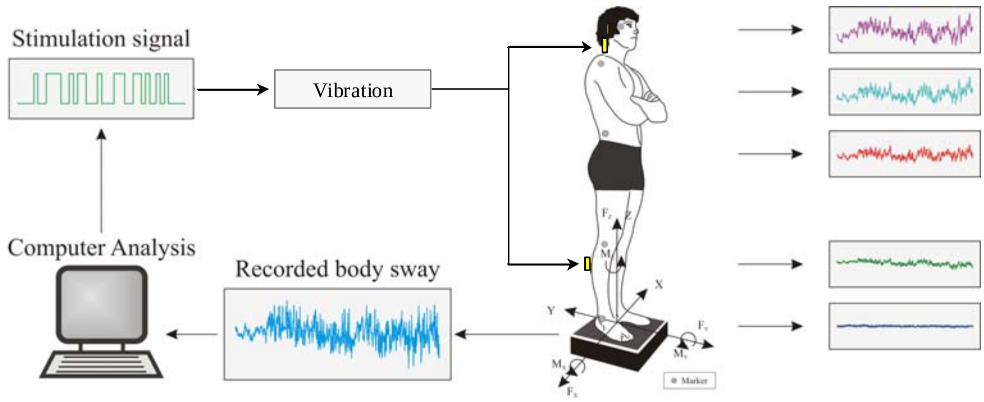

Figure 1. Schematic picture of the measurement setup and placement of the five Zebris markers attached to a subject standing on a force platform. The marker locations are shown as small circles.

The five position markers were attached on the right side of the subject, directly facing the transmitter of the Zebris™ system. EMG surface electrodes were fixed on the skin over the middle and upper end of the tibialis anterior and the gastrocnemius medialis muscles on both legs and the vibrators strapped in place on both legs. Each subject was then instructed to stand barefoot on the force platform in a relaxed posture with arms folded across the chest. The subject’s heels were 3cm apart with feet at an angle of 30° open to the front using guidelines on the force platform. Subjects were instructed to focus on a target 1.5m in front at eye level or stand with their eyes closed depending on the test condition.

The following 2 tests were performed by all subjects in a randomized order using a Latin Square design:

- Vibration of the calf muscles with eyes closed (EC-Calf) and eyes open (EO-Calf).

Before vibration, a 30-second control period of quiet stance was recorded. The vibratory stimulation pulses were applied according to a pseudorandom binary sequence (PRBS) schedule (Tjernstrom et al., 2002) during a period of 200 seconds making the combined test 230 seconds long. Each pulse had random time durations between 0.8 and 6.4 seconds. The PRBS stimulation sequence was selected because the randomised vibratory stimulation is difficult to predict and therefore lessened the likelihood of pre-emptive responses and is known to result in postural adaptation (Fransson et al., 2007b). Additionally, the PRBS stimulation sequence has a broad effective bandwidth in the region of 0.1-2.5 Hz. There was a five-minute rest period between the 4 tests.

2.4. Analysis

Vibratory calf muscle stimulation induces body movement primarily in the anteroposterior direction. Therefore only responses in this direction are considered here. The anteroposterior linear body movements were quantified in terms of movement variance at the head, shoulder, hip and knee, as recorded with the Zebris™ system using the formula below, in this example calculated for the head marker:

$$\bar{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n}$$

$$x(head)_{\mathrm{var}} = \frac{1}{n-1}\sum_{i=1}^{n}\left(x_{Head}(i) - \bar{x}_{Head}\right)^2$$

where $x(head)_{var}$ represents the variance of the linear anteroposterior head movements and x represents the marker's sampled anteroposterior position under the period analyzed. The anteroposterior torque variance was calculated using the same formula as above, where the x = torque exerted in anteroposterior direction to the surface recorded by a force platform (Fransson et al., 2007a). The mean angular position with respect to the ankle joint was calculated for each marker position using the ankle marker as the zero-position reference point and using the vertical and anteroposterior linear perpendicular distances to the marker (Fransson et al., 2007a) according the formula below, in this example calculated for the head marker:

$$\bar{x}_{Head} = \sum_{i=1}^{n} \frac{x_{Head}(i)}{n} \qquad \bar{x}_{Ankle} = \sum_{i=1}^{n} \frac{x_{Ankle}(i)}{n}$$

$$\bar{z}_{Head} = \sum_{i=1}^{n} \frac{z_{Head}(i)}{n} \qquad \bar{z}_{Ankle} = \sum_{i=1}^{n} \frac{z_{Ankle}(i)}{n}$$

$$x(head)_{ang} = \arcsin\left(\frac{\bar{x}_{Head} - \bar{x}_{Ankle}}{\bar{z}_{Head} - \bar{z}_{Ankle}}\right)$$

where $x(head)_{ang}$ represents the mean angular position of the head, x represents the marker's sampled anteroposterior position and z the marker's sampled vertical position under the period analyzed. The Zebris measurement accuracy allowed the marker angular values to be calculated with an error of less than 1.5%. If the mean angular positions of all individual markers are viewed upon together in a simple stick model, a view also of the entire body posture is obtained (Fransson et al., 2007a). Body posture and body movement amplitude commonly change independently of one another, and were therefore analysed separately as mean angular position and linear body movement variance respectively (Fransson et al., 2007b).

When recording the EMG activity from the tibialis anterior and gastrocnemius muscles, a significant effort was made to determine that the crosstalk from muscles near the muscle of interest did not contaminate the recorded signal. EMG data from the tibialis anterior and gastrocnemius muscles of both legs were band-pass filtered, using 20Hz and 200Hz, respectively, as frequency cut-off limits, and the root mean square (RMS) value was calculated (Fransson et al., 2007a). Gastrocnemius EMG signals were further notch filtered between 100 and 130Hz to remove the distortion effects caused by vibratory stimulations of the calf muscles. The distortion in the EMG recordings due to the vibratory stimulation was at about 115Hz. Notably, the distortion frequency was different from the mechanical vibration, which was at about 85Hz. Hence, the most likely source of the distortions is the electrical device producing the vibration, not the mechanical vibration itself. No notch filtering was required for tibialis anterior EMG signals, as they were not distorted by vibration of the calf muscles. A fifth-order digital Finite duration Impulse Response (FIR) filter, selected to avoid aliasing, was used for spectral separation. Quiet stance EMG activity assessed during calf vibration with eyes open served as the reference (Fransson et al., 2007a). Hence, the EMG results presented for each test were normalised for each subject.

Each test was divided into five periods: Quiet Stance (0–30s), and four 50s stimulation periods (period 1: 30–80s; period 2: 80–130s; period 3: 130–180s; period 4: 180–230s). The selection of 50s analysis periods were based on prior studies on how postural control are gradually affected by prolonged randomised vibratory proprioceptive stimulation (Tjernstrom et al., 2002).

2.5 Data analysis

Torque variance values were normalized to account for anthropometric differences between the subjects, using the subject’s squared height and squared mass (Fransson et al., 2007b;Johansson et al., 1988). Similarly, the anteroposterior linear movement variance values were normalised using the subject’s squared height before the statistical analysis. The averaged RMS values from the right and left tibialis anterior and gastrocnemius muscles were calculated and used in the analysis.

Four quotients were calculated to assess individual changes over time in RMS EMG activity; segmental body movement variance; body posture; and torque variance. The data on which the quotient calculations were done are presented in figures 2-5. The first quotient (dividing quiet stance values by stimulation period 1 values) shows how the assessed parameters were initially affected by the balance perturbations evoked by vibratory proprioceptive stimulation compared to the activity during quiet stance. The three other quotients (dividing stimulation periods 2, 3 and 4 values by stimulation period 1 values) show how the assessed parameters were gradually affected by repeated vibratory stimulation, quantifying possible effects of adaptation to vibratory proprioceptive stimulation.

2.6 Statistical analysis

The Wilcoxon non-parametric matched-pairs signed-rank test (Exact sig. 2-tailed) [25] was used for analysis of variations over time for each test condition. The changes between Quiet Stance and Period 1 in EMG RMS activity, body movement variance, mean angular position and torque variance were evaluated to determine how the assessed parameters were initially affected by vibratory proprioceptive stimulation under the test condition compared to the activity during quiet stance [26]. The changes in these parameters between Period 1 and Period 4 were also evaluated to determine the totally gained improvement under the entire trial, quantifying possible effects of adaptation to vibratory proprioceptive stimulation [26]. The Spearman two-tailed rank correlation coefficient test was used to analyze the correlation between the RMS EMG quotient values and the quotient values of linear movement variance, mean angular position and torque variance. Non-parametric statistics were used for the Spearman's correlations because values were not normally distributed using the Shapiro-Wilk test. In the analysis p<0.01 were considered statistically significant (Altman, 1991). However, we present p-values <0.05 in the correlation figures for consistency.

3. Results

3.1 Recorded RMS EMG activity, linear body movement, mean angular position and torque variance.

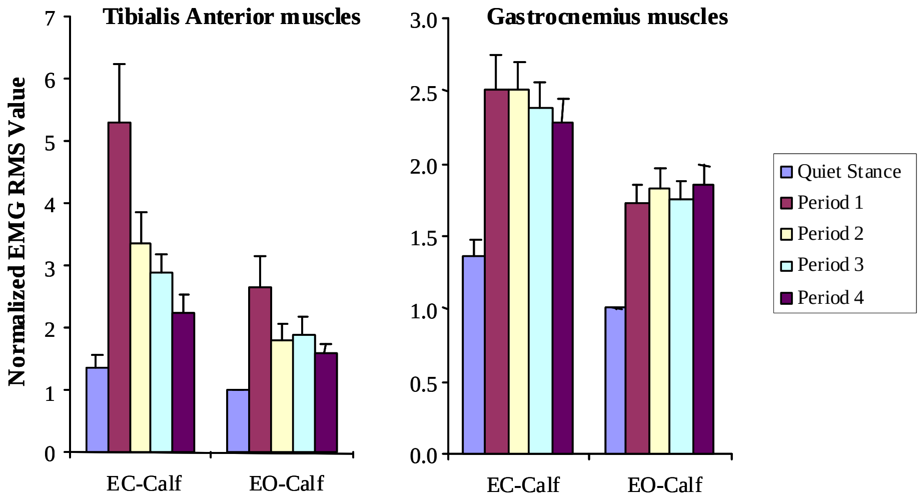

Figure 2. Tibialis anterior and gastrocnemius EMG RMS values (mean and standard error of mean (SEM)).

The tibialis anterior EMG RMS activity increased significantly during all test conditions in period 1 compared with quiet stance (p<0.001), see figure 2. The EMG increases were about 300% for EC-Calf and 165% for EO-Calf. The tibialis anterior EMG RMS activity was significantly smaller in period 4 compared with period 1 during both test conditions (p<0.01). The EMG RMS activity decreases were about 60% for EC-Calf and 40% for EO-Calf.

The gastrocnemius EMG RMS activity increased significantly during both test conditions in period 1 compared with quiet stance (p<0.001). The EMG RMS activity increases were about 80% for EC-Calf and 70% for EO-Calf. In contrast to the tibialis anterior EMG RMS activity, the gastrocnemius EMG RMS activity was not significantly different in period 4 compared with period 1 during any of the test conditions.

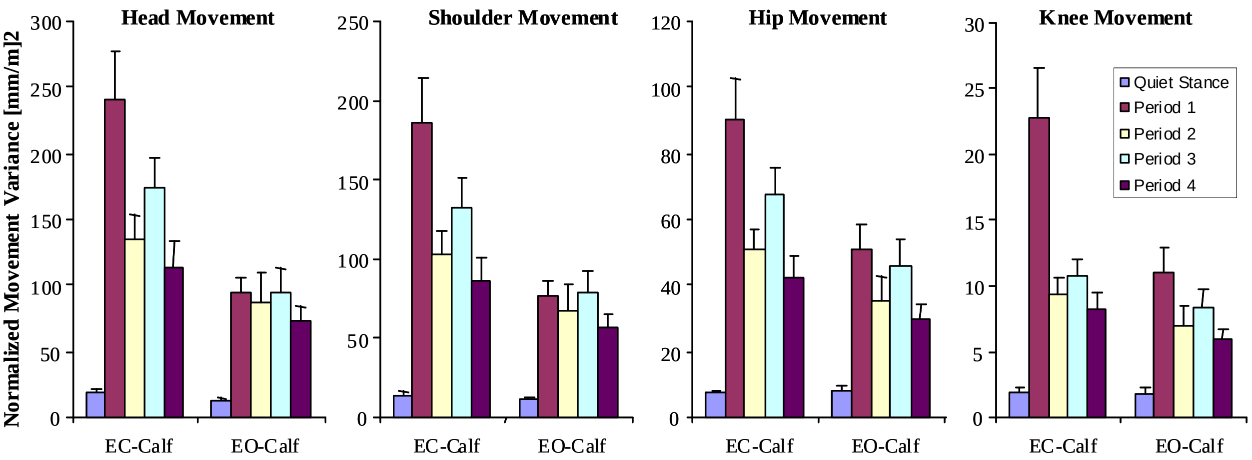

Figure 3. Normalised movement variance values for anteroposterior linear head, shoulder, hip and knee movements (mean and standard error of mean (SEM)).

The stimulus onset significantly increased body movement variances at all measured sites ($p<0.001$), see figure 3. The significant movement variance increases for the tests was approximately 1180% with EC-Calf and 570% with EO-Calf at all positions.

Analysis of the variance values showed that with EC-Calf, there was an equal reduction of the movement variances at all measured sites by about 55% ($P<0.001$) in period 4 compared with period 1. However, with EO-Calf another pattern was observed. With EO-Calf only the movement variances at the lower segments decreased significantly in period 4 compared with period 1, the knee movement variance by about 45% and the hip by about 40% ($p<0.01$).

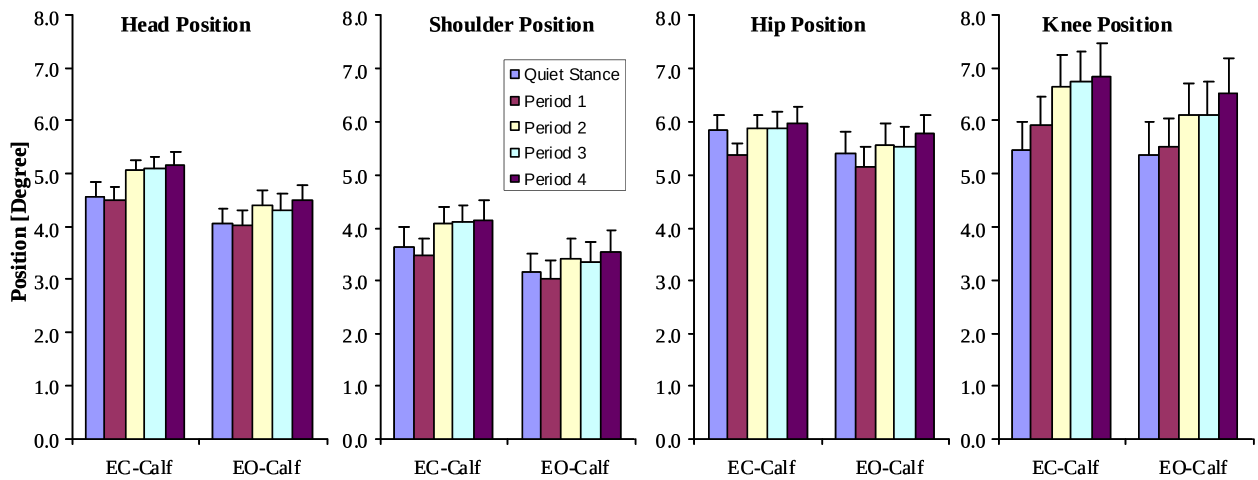

Figure 4. Angular values for anteroposterior head, shoulder, hip and knee positions (mean and standard error of mean (SEM)).

During calf vibration, the mean angular position did not significantly change between quiet stance and vibration period 1, see figure 4. Instead, with EC-Calf, the angular positions increased by approximately 15% at all measured sites in period 4 compared with period 1 (head ($p=0.002$); shoulder ($p=0.002$); hip ($p=0.001$); knee ($p=0.008$)), i.e., the subjects increased their leaning forward. Similarly, with EO-Calf, the angular positions increased in period 4 compared with period 1, (head ($p=0.014$); shoulder ($p=0.014$); hip ($p<0.001$); knee ($p<0.001$)).

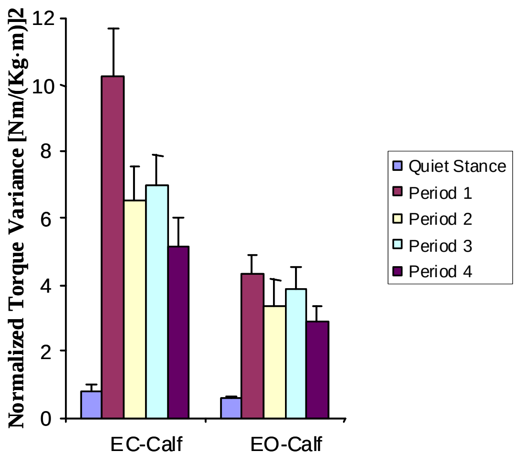

Figure 5. Normalised torque variance values for anteroposterior linear head, shoulder, hip and knee movements (mean and standard error of mean (SEM)).

The anteroposterior torque variance increased significantly during all tests in period 1 compared with quiet stance (p<0.001), see figure 5. The increase in torque variance was about 1180% for EC-Calf and 665% EO-Calf. Moreover, the torque variance values were significantly smaller in period 4 compared with period 1 in both test conditions, EC-Calf (p<0.001) and EO-Calf (p<0.01). The decrease in torque variance for the tests was about 50% for EC-Calf and 35% for EO-Calf.

3.2 Correlations between alteration of RMS EMG activity and alterations of linear body movement, mean angular position and torque variance.

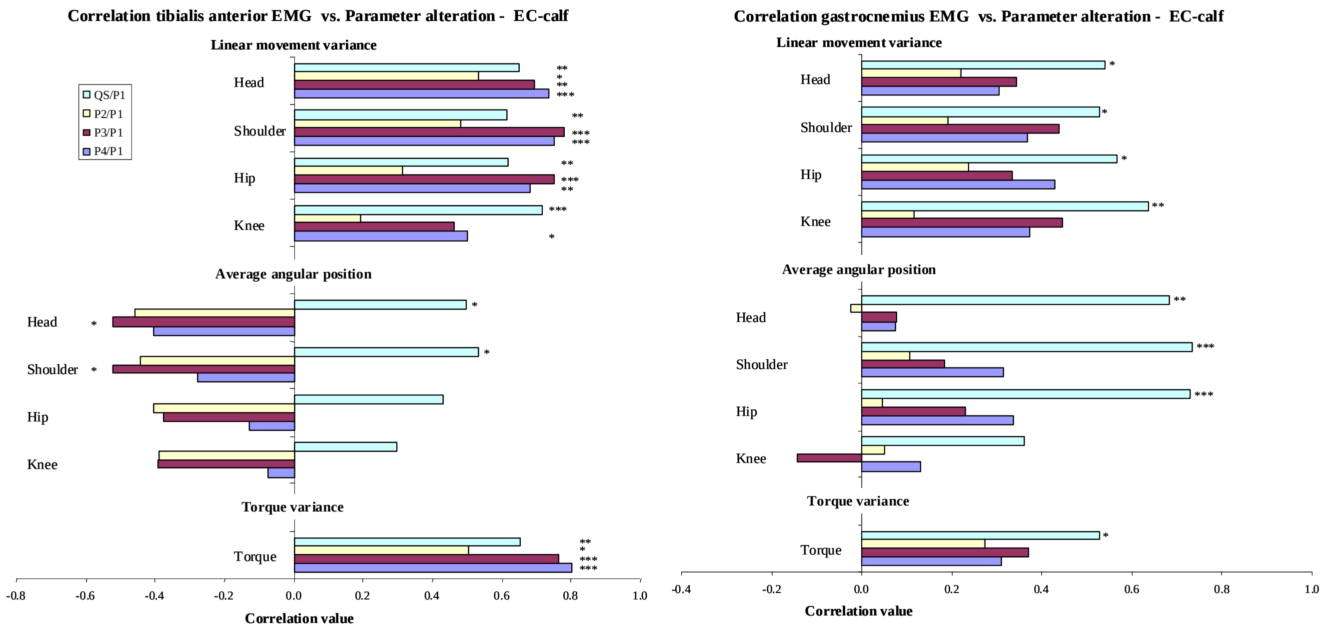

Figure 6. Correlation values between alterations in tibialis anterior and gastrocnemius RMS EMG activity and alterations of linear body movement, mean angular position and torque variance with EC Calf. The statistical differences found are marked with asterisk, where *= p<0.05, ** = p<0.01 and *** = p<0.001.

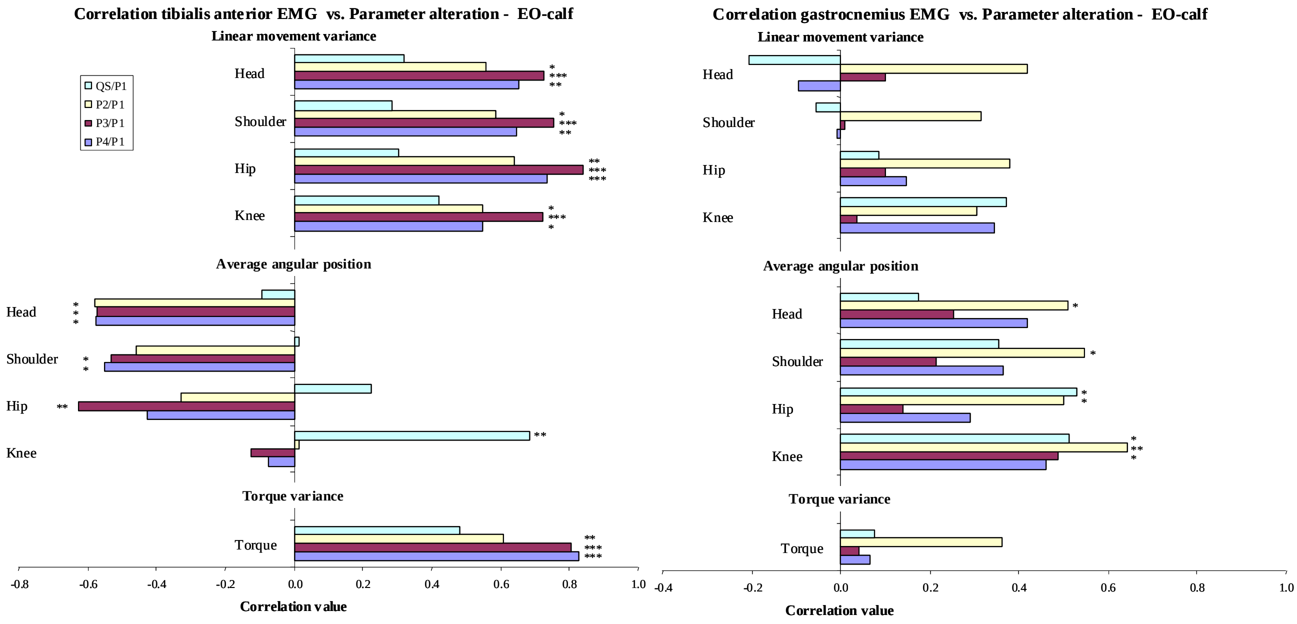

Figure 7. Correlation values between alterations in tibialis anterior and gastrocnemius RMS EMG activity and alterations of linear body movement, mean angular position and torque variance with EO Calf.

3.2.1 Linear body movement variance and RMS EMG activity

When studying the initial effects of balance perturbations (see QS/P1 quotient correlation) we found that the increase in muscle activity of the tibialis anterior muscles correlated to the increases in linear body movement variance at all positions (head, shoulder and hip, $P<0.01$; knee $P<0.001$) during the EC-Calf test (Figures 3 and 6). In the following period, we observed a sharp decrease in linear body movement variance in all tests, which levelled off in Periods 3 and 4 (Figure 3). However, the initial decrease in the tibialis anterior muscle activity did not reflect the adaptive decrease in linear movement variance at any position during EC-Calf, see P2/P1 quotient correlation. Additionally, when reaching Period 3 of EC-Calf test (see P3/P1 quotient), there was a significant correlation between the decrease in tibialis anterior muscle activity and the decrease in linear body movement variance at all positions except at the knee (head, $P<0.01$; shoulder and hip, $P<0.001$), and these correlations were almost the same in Period 4 (head and shoulder, $P<0.001$; hip, $P<0.01$), see P4/P1 quotient.

In the EO-Calf test (see Figure 7), there was no significant relationship between the initial changes (QS/P1 and P2/P1 quotient correlations) in tibialis anterior muscle activity and recorded changes in body movement variance (Figure 3). However, during the second half of the test (P3/P1 quotient; at all positions, $P<0.001$) and (P4/P1 quotient; head and shoulder, $P<0.01$; hip, $P<0.001$), the correlation values shows that the decrease in tibialis anterior muscle activity reflected the decrease in body movement variance.

At the onset of vibration (QS/P1 quotient), the increase in gastrocnemius muscle activity correlated with the increase in linear body movement variance at the knee (knee $P<0.01$) in the EC-Calf test (see Figures 3 and 6). In the EO-Calf test, there was no significant relationship between the changes in gastrocnemius muscle activity and changes linear body movement variance, i.e., whereas movement variances decreased markedly over time in these tests, the gastrocnemius muscle activity changes were not associated with these changes (see Figure 7).

3.2.2 Mean angular position and RMS EMG activity

At the onset of vibration (QS/P1 quotient), the initial increase in tibialis anterior muscle activity correlated with a small increase in mean angular position at the knee (*P*<0.01) in the EO-Calf test, see figure 6. Although there were some clear trends between decreased tibialis anterior activity and increased mean angular position in a number of comparisons in both EC-Calf and EO-Calf tests, the adaptive decrease in tibialis anterior muscle activity only significantly correlated negatively with the increase in the mean angular position at the hip between Period 1 and Period 3 (*P*<0.01) during the EO-Calf test, see P3/P1 quotient. Hence, decreased tibialis anterior muscle activity was associated with increased leaning forward of the hip.

At the onset of vibration, the increase in gastrocnemius muscle activity correlated with the increase in mean angular position at the head (*P*<0.01), shoulder (*P*<0.001) and hip (*P*<0.001) in EC-Calf, see QS/P1 quotients figures 4 and 6. In contrast, there was no indication of a relationship between gastrocnemius muscle activity and mean angular position with EC-Calf at any position during the reminder of the test. During the remaining test periods, the small change in gastrocnemius muscle activity only correlated with a slight change in mean angular position forward at the knee in EO-Calf (*P*<0.01), see P2/P1 quotient figures 4 and 7.

3.2.3 Torque Variance and RMS EMG activity

When studying the initial effects of balance perturbations (see QS/P1 quotient, Figure 6) we found that the large increase in tibialis anterior muscle activity correlated with the initial increase in torque variance with EC-Calf (*P*<0.01). However, the decrease in the tibialis anterior muscle activity from Period 1 to Period 2 (see P2/P1 quotient) did not significantly correlate with the decrease in torque variance. Though, when reaching Period 3 and Period 4 of EC-Calf test (see P3/P1 and P4/P1 quotients), there was a significant correlation between the decrease in tibialis anterior muscle activity and the decrease in torque variance (*P*<0.001).

In the EO-Calf test, there was no significant relationship between the initial changes (QS/P1 quotient) in tibialis anterior muscle activity and recorded toque variance, see figure 7. However, in the last three periods of the test (P2/P1 quotient, P<0.01; P3/P1 and P4/P1 quotients, *P*<0.001), the correlation values shows that the decrease in tibialis anterior muscle activity reflected the adaptive decrease in torque variance.

The initial increase in gastrocnemius muscle activity did not significantly reflect any of the changes in torque variance during any of the tests. Additionally, we found no evidence in any of the tests that the changes in gastrocnemius muscle activity were significantly related to the large decrease in torque variance.

4. Discussion

4.1 Relationship between EMG activity and movement

Both adaptation and somewhat surprisingly the availability of visual information affected the relationship between tibialis anterior and gastrocnemius muscle activity and body movement. During the continuous perturbations using a pseudorandom sequence of vibration pulses the postural challenge, although still threatening, became more controllable, evidenced by a reduction in the body movement variance and ankle torque variance, see figures 3 and 5. On the whole, the postural stability as assessed by body movement and ankle torques towards the surface improved rapidly until vibration Period 3, followed by no real change between Periods 3 and 4. These patterns are similar to the ones we have found previously, and we have deemed that at this plateau phase the adaptation has subsided (Fransson et al., 2002). Body posture and tibialis anterior muscle activity also showed adaptive responses; body posture leaned further forward (figure 4) and tibialis anterior muscle activity decreased, though no adaptive behaviour was evidenced in the gastrocnemius muscle activity (figure 2). These findings are consistent with the findings by Corna et al (2000), showing that within the first few cycles of a balance perturbation, participants predict the characteristics of perturbations and their destabilizing effects, and set their balance control system to minimize these effects (Akram et al., 2008).

When considering the initial changes in response to the balance perturbations, we found that tibialis anterior muscle activity changes correlated well with torque variance and body movement variance changes during eyes closed only, and gastrocnemius muscle activity changes correlated with head, shoulder and hip angular position changes also during eyes closed only. This finding suggests that vision has a significant influence on the relationship between muscle activity and recorded body movements in the initial phase of exposure to balance perturbations. Therefore, the presence or absence of vision dramatically changes the strategy employed for the maintenance of postural stability (Hafström et al., 2002; De Nunzio et al., 2005). This finding is similar to Buchanan and Horak (1999) that without visual information, EMG activity of muscles including the tibialis anterior and gastrocnemius were associated with slow drift of the head and Centre of Mass (CoM) motion, suggesting that either otolith or somatosensory information trigger the muscle responses. Additionally, the initial control of linear movement correlated to both tibialis anterior and gastrocnemius muscle activity. Another implication from the presented findings is therefore that without visual information, initial postural stability might be enhanced through co-contraction of the tibialis anterior and gastrocnemius muscles. However, this initial relationship changed over time, as tibialis anterior activity decreased between periods 1 and 4, which is consistent with reports suggesting that co-contraction can only be maintained for a short period of time before muscular fatigue occurs (Hogan, 1984). One may question whether the EMG activity in tibialis anterior and gastrocnemius was changed due to adaptation or merely as an effect of increased body leaning forward. Some findings in this study suggest that the body leaning forward was of importance for the tibialis anterior and gastrocnemius EMG activity but findings also suggest that body leaning might not be the only factor influencing the EMG activity. Most of the tibialis anterior EMG activity reduction occurred during period 1 to period 3 under the same periods the body leaning was most notably changed forward. Additionally, we found several correlations at $p<0.05$ between tibialis anterior, gastrocnemius EMG activity changes and mean angular position changes, see figure 7, so a relationship between increased body leaning forward and muscle activity reduction can not be excluded. However, the body leaning changes in degrees were about the same with eyes open and eyes closed. Nonetheless, the size and changes of the tibialis anterior and gastrocnemius EMG activity were clearly larger with eyes closed than with eyes open which suggests that vision influenced the muscle EMG activity and the adaptation of the EMG activity independently of body leaning.

Vibratory stimulus of the gastrocnemius muscle can directly influence the fusimotor activity in the muscle and thereby the EMG activity recorded. However, the gastrocnemius EMG activity had almost identical properties when the subjects were exposed to neck stimulation as when exposed to calf stimulation (unpublished observations), which suggests that the vibratory stimulation of the gastrocnemius muscle do not have a major detectable influence on the recorded EMG activity after the precautious filtering procedures used in the study.

4.2 Muscle activity and adaptation

The correlations between tibialis anterior and gastrocnemius muscle activities and the recorded body movement parameters were prone to adaptation, see figures 6 and 7. The correlation coefficients between gastrocnemius muscle activity and body posture were larger during the initial increase in Period 1 and the initial adaptation in Period 2, whereas there was a significant relationship between changes in tibialis anterior muscle activity and movement variance in Periods 1, 3 and 4 of vibration. Hence, the relationship between muscle activity and body movement is complex, and cannot simply be through a parallel change of EMG activity and movement variance. The control of postural stability is regulated by postural muscles that form an uninterrupted muscular chain that extends from the head to the feet (Roll et al., 1989) as confirmed by the induced balance perturbations caused by proprioceptive vibration at various locations (Courtine et al., 2007). Thus, one possibility is that during some periods of time some other postural muscles along the muscular chain may influence the body movements more than the gastrocnemius and tibialis anterior muscles. Alternatively, the reduced correlation between gastrocnemius muscle activity changes and body posture changes after 100 seconds of vibration suggests that once the level of movement variance had decreased sufficiently through adaptive mechanisms, gastrocnemius may not have the same role in postural control. The muscles shifting role in postural control is also illustrated by the lack of correlation between linear body movement and tibialis anterior EMG activity between Period 1 and Period 2, possibly due to that the muscle activity could not be suppressed to the same extent as body movement, particularly with eyes closed, due to that the muscles during this phase also had an important role for postural stability.

A surprising finding was that the positive correlations between tibialis anterior muscle activity and linear movement variance and the negative correlation between tibialis anterior muscle activity and an increased mean angular position (i.e. forward leaning) generally increased in the P3/P1 and P4/P1 quotients. This latter period of the test represents a settling period where subjects have adapted to use less energy to maintain postural stability (Fransson et al., 2002), and our findings suggest that this is shown by an increased control of movement through forward leaning, and evidenced also by decreased movement variance (figure 3) and lower tibialis anterior muscular activity (figure 2). Benefits of this adaptation is decreased risk for muscular fatigue and an enhanced standing postural strategy (Mihelj et al., 2000), since by forward leaning the reliance on sensory feedback is reduced (Madigan et al., 2006) and the muscle spindles in the plantar flexors gain improved ability to sense changes in muscle length and velocity due to the increased gamma motor neurone drive (Madigan et al., 2006).

As somewhat expected by the negligible change in gastrocnemius muscle activity compared with the changes in body movement and mean angular position, there was almost no relationship between these variables at our Bonferroni-corrected level of significance ($p<0.01$). This furthers and corroborates the findings by Loram et al. (Loram et al., 2005) showing little relationship between gastrocnemius EMG and CoM movements under quiet stance. Furthermore, this implies that, although the gastrocnemius muscles are important in the regulation of the upright standing posture, particularly with sudden balance perturbations, the gastrocnemius muscles might not be fully associated with the tonic maintenance of postural control.

4.3 Muscle activity and movement

Several previous investigations of the relationship between muscle activity and body movement are largely based on theoretical presumptions (Riccio and Stoffregen, 1988; Soechting and Flanders, 1991; St-Onge and Feldman, 2004) or based on studies of arm movements (Darling and Cooke, 1987a, b; Gabriel, 2002). Furthermore, although a strong link between a single muscle and a single joint may be established for some tasks, this is most probably not the case for multi-joint tasks. For example, work employing a two joint arm system (Kelso, Buchanan, and Wallace 1991) demonstrated that prime movers drop out when inertia can accomplish the same action. These models might not therefore be adequate enough to illustrate the complex relationship between local muscle activity and recorded body movements in upright standing posture. Additionally, several findings in this study suggests that several partly independent factors such as vision, body leaning and adaptation may change the relationship between muscle activity and recorded movements. Moreover, complex relationships and adaptive changes might be more obvious when studying the effects over a long period of time, such as the 50s periods used in the present study, rather than studying short periods of EMG activity directly associated with a particular movement.

4.4 Clinical Significance of findings

It is well-known that to maintain upright stance, the central nervous system (CNS) must coordinate motion across many joints and muscles using sensory information provided by the visual, somatosensory and vestibular systems (Akram et al., 2008). The multiple segments of the body are inter-connected (Ivanenko et al., 2000), and as evidenced in this study, a local change in proprioceptive information led to a widespread alteration in posture remote from the vibration site, thus complimenting the findings by Ivanenko (Ivanenko et al., 2000) and Thompson et al. (Thompson et al., 2007). In other words, the CNS must use different strategies for appropriate balance control when the information from one of the sensory receptors is unreliable. However, in some commonly used posturography tests there is sometimes no detectable change in balance, even in patients with sensory disorders, as sensory re-weighting shifts the reliance of afferent information from unreliable sources to other more reliable receptors (Oie et al., 2002). Therefore, the finding in the present study that the change in strategy is detectable when assessing the correlation between muscle activity and body movements during balance perturbation, might warrant a new balance testing approach to asses rehabilitation effects. For example, the used approach might be beneficial assessing patients recovering from surgical procedures performed on the neuromuscular or musculoskeletal system affecting postural control, or to check whether the appropriate control of posture and balance control is gained from vision. Furthermore, postural control’s remarkable ability to learn how to handle vibration-induced balance perturbations as illustrated in this study supports the idea that proprioceptive vibration training could be used as a rehabilitation technique. This is particularly true for the elderly because while the elderly fall frequently when surface somatosensory information is altered, they become capable of maintaining normal steadiness after repetitive experience (Woollacott et al., 1986).

5. Conclusions

Both adaptation and the availability of visual information affected the relationship between tibialis anterior and gastrocnemius muscle activity and body movement. Without visual information, initial postural support might be enhanced through the use of co-contraction of the tibialis anterior and gastrocnemius muscles. However, these initial relationships changed over time as an effect of adaptation. Thus, adaptation training using vibratory proprioceptive stimulation could benefit those susceptible to falls by changing the association between muscle activity and movement.

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