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Effects of balance training with visual feedback during mechanically unperturbed standing on postural corrective responses

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ABSTRACT

Evidence of a non-specific effect of balance training on postural control mechanisms suggests that balance training during mechanically unperturbed standing may improve postural corrective responses following external perturbations. The purpose of the present study was to examine kinematics of the trunk as well as muscular activity of the lower leg and paraspinal muscles during postural responses to support–surface rotations after short-term balance training. Experiments were performed in control ($n = 10$) and experimental ($n = 11$) groups. The experimental group participated in the 3-day balance training program. During the training, participants stood on a force platform and were instructed to voluntarily shift their center of pressure in indicated directions as represented by a cursor on a monitor. Postural perturbation tests were executed before and after the training period: the slow and fast 10° dorsiflexions were induced at angular velocities of approximately 50° s^{-1} and 200° s^{-1}, respectively. In the experimental group, the amplitude of the trunk displacements during slow and fast perturbations was up to 33.4% and 26.7% lower, respectively, following the training. The magnitude of the muscular activity was reduced in both the early and late components of the response. The kinematic parameters and muscular responses did not change in the control group. The results suggest that balance training during unperturbed standing has the potential to improve postural corrective responses to unexpected balance perturbation through (1) improved neuromuscular coordination of the involved muscles and (2) adaptive neural modifications on the spinal and cortical levels facilitated by voluntary activity.

1. Introduction

Recent advances in technology have resulted in the availability of visual feedback for the retraining of balance function in disabled and at-risk populations [1–5] through improvement of sensorimotor integration [6,7]. Generally, those balance improvements were revealed in a *task-specific* manner [8,9]. That is, balance training in mechanically unperturbed standing was shown to improve the stability of mechanically unperturbed posture [10,11]. However, since the ability to recover balance after external perturbations may be crucial for fall prevention, it is not clear whether the unperturbed balance training would foster that ability. There is growing evidence of *non-specific* effects of balance training on postural control mechanisms [12–15]. These findings raise the question of whether balance training during unperturbed standing may improve the ability to recover balance after external perturbations.

Postural corrective responses to unexpected external perturbations [16–23] are of considerable importance in maintaining stability in the unpredictable circumstances of daily life [11,24,25]. Quantification of postural responses has practical implications for predicting falls [26,27]. For instance, the magnitude of the kinematic and muscular responses, as well as the ability to scale the responses proportionally to the perturbation intensity, can be affected by specific neuromuscular impairments [23,28] or training adaptation [29,30]. It has been long believed that postural recovery strategies are not associated with cortical control [18]. However, growing evidence exists that the cerebral cortex and

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high-level ''cognitive'' processing may be involved in controlling specific aspects of balance [11,27,31]. For example, it has been demonstrated that primarily supraspinal adaptations contribute to improved balance performance following balance training [29]. Although the earliest part of the postural response may not be due to cortical activity, studies suggest that the cerebral cortex may become involved in shaping the postural response as the response progresses [27]. Therefore, as cortical control seems to contribute to postural stabilization in general, the improvement obtained during balance training in unperturbed standing may affect the postural recovery ability against the perturbation.

Based on these considerations, we suggest that postural corrective responses can be improved after balance training during mechanically unperturbed standing. The purpose of the present study was to examine trunk kinematics as well as muscular activity of the lower leg and paraspinal muscles during postural responses to support–surface rotations of different velocities after short-term balance training. In particular, we hypothesized that a reduced magnitude of trunk deflection following the perturbations of different velocities, as well as the increased ability to scale responses, will reveal an improvement of stability, whereas changes in early and late components of the muscular response will indicate spinal and supraspinal adaptation due to the training.

2. Methods

2.1. Participants

Experiments were conducted in 21 male and female volunteers (mean ± SD: age 26.5 ± 3.6 yrs, height 169.2 ± 10.9 cm, body mass 64.7 ± 8.8 kg). Of the 21 participants, 10 (seven males, three females; age 27.4 ± 4.0 yrs; height 169.4 ± 10.1 cm; body mass 65.6 ± 9.4 kg) were randomly assigned to the control group and the other 11 (seven males, four females; age 26.5 ± 3.6 yrs; height 169.1 ± 11.9 cm; body mass 64.7 ± 8.8 kg) to the experimental group. Each participant gave written informed consent to the experimental procedure, which was approved by the local ethics committee in accordance with the declaration of Helsinki on the use of human subjects in experiments.

2.2. Experimental setup and procedure

Postural responses to external perturbations have been tested on the first (first measurement) and on the fifth (second measurement) days of the study. Between these tests, the experimental group performed the balance training, whereas the control group was instructed to maintain a regular level of daily activity.

During the test, the participants stood in an upright position with their feet lightly strapped across the instep to the foot plates. Postural perturbations were evoked by rotating the foot plates in the dorsiflexion direction by a custom-made servo-controlled torque motor. The axis of rotation of the ankle joint was aligned with the axis of rotation of the foot plate. The perturbations were applied at angular velocities of approximately $50^\circ\,s^{-1}$ and $200^\circ\,s^{-1}$, which resulted in joint rotations of 10° dorsiflexion. The time between two successive perturbations was randomized between 8 and 12 s.

The participants were asked to maintain their normal standing posture, with straight knees and arms hanging comfortably at their sides. The position of their hip (at the level of the greater trochanter) was then reset to zero on an oscilloscope to establish a reference value for each participant's preferred stance position (see below). After each perturbation, the participants received feedback on their standing posture by visualizing this position on the oscilloscope placed at eye level, approximately 1.5 m in front of the participant. The participants were instructed to keep the same position during the test.

2.3. Data collection

All biomechanical (kinematics and foot plate rotation) and electromyographic (EMG) recordings were initiated 200 ms prior to the onset of the perturbation and had a sampling duration of 5 s.

The anterior–posterior displacements of points corresponding to the location of the cervical vertebra 7 (C7) and the greater trochanter (HIP) were measured by two charge-coupled laser displacement sensors (LK-2500, Keyence, Japan). During the experimental trials, the laser beams were aimed at the plastic plates (corresponding to the C7 and HIP), and the distance between the plate and the laser sensor was measured. Surface EMG signals from the right soleus muscle (SOL), the medial head of the gastrocnemius muscle (GM), the tibialis anterior muscle (TA), and the lumbar paraspinal (erector spinae) muscles (ES) at the level of the iliac crest were recorded via bipolar surface Ag–AgCl electrodes with a diameter of 7 mm (Vitrode F, Nihon Kohden, Japan). The electrodes were applied at an inter-electrode distance of 20 mm after cleansing of the skin. For the reference electrode, a belt-type surface Ag–AgCl electrode with a width of 20 mm (45400, Shimizu Electronic Ind., Niigata, Japan) was wrapped around the right shin at the tibial tuberosity level. The EMG signals were amplified with a gain of 1k and band-pass filtered (15 Hz to 3 kHz) with a bioamplifier (AB-651J, Nihon Kohden, Tokyo, Japan). Both the biomechanical data and EMG were sampled at a rate of 5 kHz.

2.4. Balance training

The training in the experimental group was performed with the force plate analysis system ''Stabilan-01'' (Rhythm, Russia). During the training, participants stood on the force plate and were instructed to look at the monitor that was placed at eye level approximately 1.5 m in front of the force plate. The center of pressure (COP) position signal was utilized as a visual input to game-based exercises.

During the first day of the experiment, the participants of both groups were requested to attend a familiarization session after the first measurement. During this 10–15-min session, a single trial of each game-based exercise was introduced to the participants. Then, the training was performed in the experimental group during three successive days. On the fifth day, the participants of both groups attended the final 7–10-min session, and performed a single trial (''exam'') of each game-based exercise trying to reach the maximal score for the entire training program. After this exam, the second measurement was performed.

2.5. Training protocols

Details of the training intervention have been reported elsewhere [2,12]. In brief, during the balance training, two types of supervised learning conditions were implemented in random order. For the first type, a given stereotyped pattern of movement had to be generated, requiring a high precision of movement performance. For the second type, the participants applied a general strategy of voluntary postural control that included attention, decision-making, and performance of the task with different movement patterns. The duration of each exercise varied from 1 to 2 min. In order to motivate participants to improve their performance, a score representing different exercise parameters was displayed. The participants were instructed to maximize their score during each exercise. Once a consistent score in each exercise was attained by the participants, the difficulty level of the exercise was increased. During each 60-min training session, usually five rounds of each exercise were presented to the participant.

2.6. Data processing

Participants were familiarized with the perturbation test for each perturbation velocity to reduce the effects of adaptation [19,23]. For each angular velocity, twelve perturbations were used for the analysis. The peak-to-peak amplitude of the posterior displacement of the C7 and HIP points following the perturbation was normalized with respect to the mean HIP displacement during the first measurement. The EMG time series were full-wave rectified and ensemble averaged. Muscle activity was quantified by calculating the average EMG signal amplitude across each of the following six epochs prior to and following the onset of the foot plates rotation: −100 to 0 ms (background activity), 30–80 ms (short latency stretch response (SLSR)), 80–120 ms (medium latency stretch response (MLSR)), 120–220 ms (primary balance correcting response (PBSR)), 240–350 ms (secondary balance correcting response (SBSR)), and 350–700 ms (stabilizing phase (SP)). These time periods are consistent with previous works [19,22]. The response amplitude was reported as a ratio with respect to the background level of activity preceding the perturbation (100 ms) [3].

The dependent variables were submitted to 2 (groups: control, experimental) by 2 (measurements: first, second) by 2 (perturbation velocities: slow, fast) mixed analysis of variance (ANOVA). In this factorial design group was used as a between-subject factor, whereas both perturbation velocity and measurement were within-subject factors. It should be noted that we chose to use this factorial arrangement for each dependent variable. EMG data were analyzed for each epoch independently. A simple *t*-test comparisons were made to decompose significant effects ($\alpha = 0.05$). The results for the pooled data are presented as mean values and standard errors of the mean (SEM).

3. Results

Fig. 1 demonstrates the muscular responses with kinematics of the trunk following fast ($200^\circ\,s^{-1}$) support–surface rotations during the first and second measurements in one participant from the control group (Fig. 1A) and one from the experimental group (Fig. 1B). There was no difference in the level of background muscle activity in the control and experimental groups during the first and second measurements. It can be seen that the rotation of