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Automatic postural responses are generated according to feet orientation and perturbation magnitude

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ABSTRACT

This investigation aimed to assess the effect of feet orientation angle in upright stance on automatic postural responses (APRs) to mechanical perturbations of different magnitudes. Perturbation was produced by releasing suddenly a load attached to the participant's trunk, leading to forward body sway. We evaluated APRs to loads corresponding to 5% (low) and 10% (high) of the participant's body weight, comparing the following feet orientations: parallel, preferred (M = 10.46°), 15° and 30° for each foot regarding the body midline. Results showed that APRs were sensitive to perturbation magnitude, with the high load leading to increased amplitudes of center of pressure displacement and joints rotation, in addition to stronger and earlier muscular responses. Feet orientation at 30° led to a greater amplitude of center of pressure displacement than the other feet orientations. The low perturbation magnitude led to similar responses both at the hip and ankle across feet orientations, whereas the high load induced increased rotation amplitudes in both joints for feet orientation at 30°. Our results suggest that APRs are generated by the nervous system taking into consideration the biomechanical constraints in the response production. Relevant for standardization of feet placement in evaluations of balance recovery, our results indicated that a moderate range of outward feet orientation angles in stance lead to comparable APRs, while increased outward feet orientation angles lead to distinct postural responses.

1. Introduction

Biomechanical constraints are known to affect balance stability. Spatial orientation of feet positioning, in particular, is critical for balance control given that it affects not only the anteroposterior length of the support base but also the direction of the force vector applied on the ground by each foot to stabilize and recover a stable body balance. Analysis of several stabilometric parameters of quiet standing in the preferred foot positioning has shown that feet orientation angle is a relevant biomechanical variable for balance control [1]. Further investigation comparing different feet orientation has shown that increased angles of outward orientation of the feet lead to stabilization of the center of pressure (CoP), with higher balance stability achieved between the angles of 15–45° [2] (see also [3,4]). While large angles of feet orientation seem to induce stabilization of quiet standing, feet positioning other than in parallel might be thought to be maladaptive to recover body balance following a postural perturbation inducing forward body oscillation. In those perturbations, the nonparallel force vectors applied at the ankle of each leg is expected to reduce the compound torque exerted on the ground to recover balance stability in comparison with the parallel orientation. Additionally, orienting both feet outward reduces the length of the stability limit of CoP excursion in the anteroposterior direction, increasing the chance of body disequilibrium following a perturbation leading to increased forward postural sway. Although no information has been gathered thus far about the effect of feet orientation on automatic postural responses[1] (APRs) to extrinsic perturbations, previous investigation has shown that reduced length of the support base leads to modulation of postural responses, with increased participation of hip motion [5]. This finding suggests that the postural control system takes into consideration the geometrical configuration of the feet on the ground in the generation of APRs.

In order to respond appropriately to a postural perturbation, the control system not only must select a good response strategy but also scale the power of the muscular response taking into consideration the magnitude of the perturbation and biomechanical constraints. Park et al. [6] assessed APRs for a range of perturbations, applied by means of a random sequence of different amplitudes of sudden displacement of the support base inducing forward body sway. They found that APRs were tailored in agreement with the perturbation magnitude, with feedback gains being scaled gradually with the perturbation size. Additionally, increasing perturbation magnitude induced a trade-off

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between ankle and hip feedback gains, suggesting that the postural control system uses a representation of biomechanical constraints in the generation of APRs (see also [7,8]). Scaling of APRs to the magnitude of a perturbation is consistent with the observation that power of the initial burst of the long latency muscular activation of the postural response scales linearly with amplitude [9], velocity [10] and maximum acceleration [11] of displacement of the support base. Further research has indicated that muscular responses to a postural perturbation are adjusted not only in magnitude but also in the activation onset latency [12,13]. The aforementioned results suggest that APRs are tailored from both perturbation magnitude and biomechanical constraints. In this regard, it would be of interest to understand how APRs are modulated from the geometry of feet positioning on the ground to different magnitudes of postural perturbations.

In the present investigation, we aimed to assess the effect of feet orientation while standing on APRs to sudden perturbations inducing a fast forward body sway. We compared the feet positioned in parallel against different angles of outward feet orientations in response to distinct magnitudes of perturbation. We hypothesized that APRs are modulated as a function of feet orientation on the ground, and that postural responses are scaled to perturbation magnitude.

2. Methods

2.1. Participants

Twenty-two healthy university students (12 males), age range 18–37 years ($M = 23.27$, $SD = 5.01$), participated in this study. Experimental procedures were carried out with provision of written informed consent by the participants, after approval by the local university ethics committee in accordance with the standards established in the Declaration of Helsinki.

2.2. Task and apparatus

The experimental task consisted of recovering stable upright stance following a perturbation caused by suddenly releasing a load attached to the participant's trunk. The initial position was sustaining upright stance, keeping the arms crossed over the chest, while resisting to a load pulling the participant's trunk backward. Participants worn a harness (20-cm wide) at the lumbar-sacral region connected to the pulling load while standing on a tri-axial force plate (AMTI, OR6-WP). An electromagnetic system embedded into the backside of the harness was connected to the load by means of a steel cable (Fig. 1). The load was released by means of a soundless remote switch unanticipatedly by the participant, leading to a fast forward body sway. In addition to ground reaction forces, we measured activation of the muscle gastrocnemius medialis (GM) of the right leg. Muscular activation was measured by means of wireless surface electrodes (Delsys Inc., Boston, MA, model Trigno). Measurement of activation of the muscle GM was made on the right leg, with the EMG electrode positioned in agreement with the SENIAM project recommendations (http://www.seniam.org/). For kinematic analysis, body motion was measured by tracking round reflective markers (15 mm diameter) attached to the following anatomic points: fifth metatarsophalangeal joint, lateral malleolus, lateral knee joint center, greater trochanter and acromion. The markers were tracked through four optoelectronic cameras (Vicon, Model MX3+).

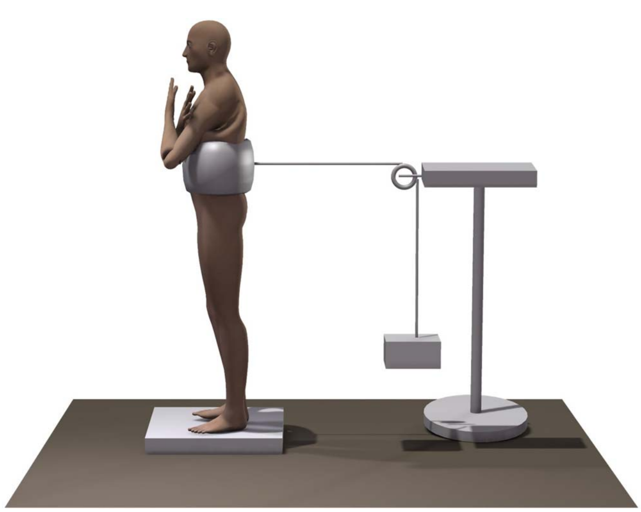

Fig. 1. Schematic representation of the experimental setup, with the participant standing on a forceplate during load application before its sudden release.

2.3. Experimental design and procedures

We employed a single group design, testing participants in eight conditions resulting from the combination of feet orientation and load magnitude. The feet orientations were the following: parallel (0°), preferred, 15°, and 30° for each foot regarding the body midline. The inner border of the feet was used as reference for measuring the angle of feet orientation, keeping the heels 5 cm apart across feet orientations. We tested APRs to perturbations provoked by two loads: 5% (low) versus 10% (high) of participant's body weight. Following preparation, participants stayed barefoot onto the platform and oriented their feet in a personally comfortable way. This orientation was marked with adhesive tape on the force plate and adopted as the preferred feet orientation. Participants were asked to gaze at a 10-cm diameter spot, presented on a monitor screen 2 m away, at eyes' height. Immediately before the probing trials, participants were provided with one familiarization trial for each probing condition. Evaluation was made through three trials for each condition. Sequence of loads was counterbalanced and sequence of feet orientation was randomized across participants. Intertrial intervals within a condition were 30-s long, while intervals between conditions endured 1 min. After half the trials, a 2-min. sitting rest interval was provided. Trials in which participants presented a stepping response were canceled out and immediately repeated.

2.4. Analysis

Data were extracted and processed through MATLAB (Mathworks, Natick, MA) routines. Data sampling frequency was set at 2000 Hz for EMG, and at 200 Hz for kinematics and ground reaction forces. EMG signals were amplified with a gain of 1000, and digitally band-pass filtered between 20 and 400 Hz. Kinematic and ground reaction forces data were digitally low-pass filtered with a cut-off frequency of 10 Hz. Signals filtering were made through a dual-pass fourth-order Butterworth filter. Estimation of center of mass (CoM) displacement was based on the anthropometric model proposed by Winter [14].

Analysis was made for the period immediately ensuing load release. Dependent variables were the following: maximum displacement in the anteroposterior axis of (a) center of pressure (CoP) and (b) center of mass (CoM); CoP displacement was normalized by the anteroposterior length of the base of support (L_{CoP}/L_{bs}, Fig. 2); (c) maximum CoP velocity; (d) latency of activation onset of the muscle GM, having as criterion the time of onset of sustained growing EMG values two standard deviations above the average in the interval of 200 ms preceding load release; (e) magnitude of GM muscle activation, estimated by means of the root mean square (RMS) of the EMG envelope in the interval of 75 ms following muscular activation onset, with raw values being normalized to the respective individual maximum value in the interval of interest following load release across experimental conditions; maximal rotation amplitude at the (f) hip (peak to peak) and (g) ankle (plantar flexion).