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Center of pressure velocity reflects body acceleration rather than body velocity during quiet standing

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ABSTRACT

The purpose of this study was to test the hypothesis that the center of pressure (COP) velocity reflects the center of mass (COM) acceleration due to a large derivative gain in the neural control system during quiet standing. Twenty-seven young (27.2 ± 4.5 years) and twenty-three elderly (66.2 ± 5.0 years) subjects participated in this study. Each subject was requested to stand quietly on a force plate for five trials, each 90 s long. The COP and COM displacements, the COP and COM velocities, and the COM acceleration were acquired via a force plate and a laser displacement sensor. The amount of fluctuation of each variable was quantified using the root mean square. Following the experimental study, a simulation study was executed to investigate the experimental findings. The experimental results revealed that the COP velocity was correlated with the COM velocity, but more highly correlated with the COM acceleration. The equation of motion of the inverted pendulum model, however, accounts only for the correlation between the COP and COM velocities. These experimental results can be meaningfully explained by the simulation study, which indicated that the neural motor command presumably contains a significant portion that is proportional to body velocity. In conclusion, the COP velocity fluctuation reflects the COM acceleration fluctuation rather than the COM velocity fluctuation, implying that the neural motor command controlling quiet standing posture contains a significant portion that is proportional to body velocity.

1. Introduction

Postural sway during quiet standing, also known as static posturography, has been used to assess postural balance abilities [1–8]. The center of pressure (COP) is one of the most popular measurements when quantifying postural sway. Among the postural sway measures yielded from COP, the COP velocity has been suggested to be most sensitive for detecting changes in balance abilities due to aging and/or neurological diseases [1–4,6,7]. Since the COP and center of mass (COM) trajectories agree very well with each other due to the underlying body dynamics [4,9], the COP velocity has been believed to be an approximate representation of the COM velocity. However, there is no study to date that has confirmed this relationship by investigating the neuromechanical meaning of the COP velocity.

The equation of motion of an inverted pendulum is given by:

$$x_{COP} \approx x_{COM} + \ddot{x}_{COM} \frac{I}{mgh}, \tag{1}$$

where x_{COP}, x_{COM}, and $\ddot{x}_{COM}$ denote the COP displacement, the COM displacement, and the COM acceleration, respectively. I, m, h, and g denote the body inertia, mass, height of mass, and standard gravity, respectively [4,9] (see Appendix 1 for details). Note that, in this study, we focused only on the anteroposterior body sway, since body sway is more prominent in this direction compared to the mediolateral direction. Differentiating Eq. (1) yields:

$$\dot{x}_{COP} \approx \dot{x}_{COM} + \dddot{x}_{COM} \frac{I}{mgh}, \tag{2}$$

where $\dot{x}_{COP}$, $\dot{x}_{COM}$, and $\dddot{x}_{COM}$ denote the COP velocity, the COM velocity, and the derivative of the COM acceleration (i.e., jerk), respectively. Based on Eq. (2), $\dot{x}_{COP}$ can be correlated with $\dot{x}_{COM}$,

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which has led to the belief in the field that the COP velocity is an approximate representation of COM velocity. However, if the relative power of the fluctuation of the second term on the right-hand side is large, $\dot{x}_{COP}$ may not be highly correlated with $\dot{x}_{COM}$.

At the same time, since $TQ \approx mgx_{COP}$, where TQ denotes the ankle torque, COP reflects the ankle torque controlling COM during quiet standing. In several studies, the strategy for controlling the ankle torque during quiet standing was modeled using linear controllers [10,11]. For example, Peterka [11] modeled the control strategy using a proportional-integral-derivative (PID) controller and a proportional-derivative (PD) controller in parallel, which correspond to a neural and mechanical controller, respectively (cf. Fig. 1). Based on these suggested models, TQ is given by:

$$TQ = Kp(\theta - \tau) + Kd(\dot{\theta} - \tau) + Ki\int(\theta - \tau)dt + K\theta + B\dot{\theta}, \quad (3)$$

where θ denotes the COM angle; Kp, Kd, and Ki denote proportional, derivative and integral gains for the neural controller, respectively; K and B are proportional and derivative gains for the mechanical controller, respectively; and τ denotes the time delay within the feedback loop of the neural controller. By differentiating Eq. (3), we obtain:

$$\dot{TQ} = Kp(\dot{\theta} - \tau) + Kd(\ddot{\theta} - \tau) + Ki(\theta - \tau) + K\dot{\theta} + B\ddot{\theta}. \quad (4)$$

In the case of quiet standing ($\theta \approx 0$), we can linearly approximate that $x_{COM} \propto \theta, \dot{x}_{COM} \propto \dot{\theta}$ and $\ddot{x}_{COM} \propto \ddot{\theta}$. Since it has been suggested that Kd is relatively large [10,12–14] and B very small [15] in the control system of human quiet standing, it can be hypothesized that the Kd term is dominant in the right-hand side of Eq. (4). In this case, TQ, which is proportional to the COP velocity based on $TQ \approx mgx_{COP}$, can be approximately proportional to the COM acceleration, as the Kd term includes $\ddot{\theta}$. If this is true, our previous findings that the COM acceleration was similarly sensitive as the COP velocity in detecting effects of aging [4] and neurological disease [16] on postural control can be explained well.

Thus, using an experimental and simulation study, the purpose of this study was to test the hypothesis that the COP velocity fluctuation reflects the COM acceleration fluctuation due to a large derivative gain in the neural control system during quiet standing.

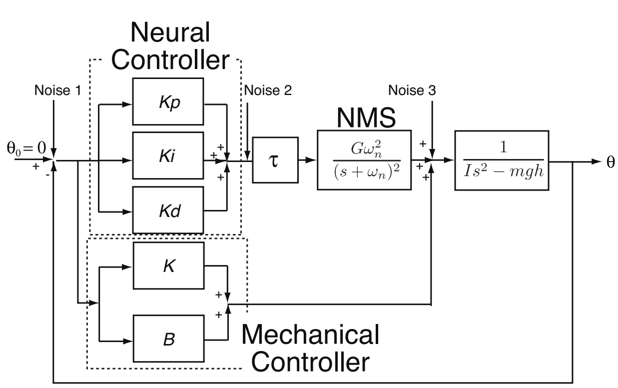

Fig. 1. Computational model used for simulating the control system of quiet standing. The model consisted of a neural controller using a PID controller with gains Kp, Ki, and Kd, and a mechanical controller using a PD controller with gains K and B. A total constant time delay (τ) representing motor and sensory transmission delays was inserted at the output of the neural controller. Subsequent to the delay, a critically damped, 2nd order model of the neuromuscular system (labeled as NMS) was included to replicate the ankle torque generation process producing the neural torque component. The sum of the neural and mechanical torque components controlled the inverted pendulum model of the standing body. Three noise inputs corresponding to sensory, motor, and neuro-mechanical noise were injected to drive the simulation. All model parameters are listed in Appendix 2.

2. Methods

2.1. Experimental study

2.1.1. Subjects

Experimental data were acquired in the context of a previous study [17]. Twenty-seven healthy young adults (14 female; age 27.2 ± 4.5 years; height 168 ± 9 cm; weight 62.3 ± 10.9 kg) and twenty-three healthy elderly adults (12 female; age 66.2 ± 5.0 years; height 157 ± 7 cm; weight 59.3 ± 8.4 kg) participated in this study. They had no medical history or signs of neurological disorders. All subjects gave written informed consent to participate in the study, and the experimental procedures were approved by the local ethics committee.

2.1.2. Procedure

Each subject stood quietly with bare feet, eyes open, and the arms hanging along the sides of the body for the duration of 90 s. The subject was instructed to stand relaxed and quietly and to refrain from any voluntary limb or head movements. Each subject completed five trials with sufficient resting time in between the trials. The horizontal position around the third lumbar vertebra (L3) was measured with a high-accuracy laser displacement sensor (LK-500, Keyence, Japan). A force platform (Type 9281B, Kistler, Switzerland) was used to measure the subjects' COP displacement and the horizontal ground reaction force during quiet standing. All data were sampled at 1 kHz and stored on a personal computer for subsequent analysis.

2.1.3. Analysis

Based on the subject's body mass (M) and height (H), the moving mass (m), the COM height (h), and the moment of inertia with respect to the ankle joint (I) were estimated according to: $m = 0.971M$ [18]; $h = 0.520H$ [19]; and $I = 0.347MH^2$ [20] (using the shape factor $k = 1.32$ in $I = kmh^2$). The COM displacement was estimated using the output of the laser displacement sensor with a height correction, i.e., $d' = d \times (h/h_{laser})$, where d denotes the output of the laser displacement sensor; d' denotes its corrected value; and h_{laser} denotes the height of the laser displacement sensor.

The COM velocity and COP velocity time series were obtained by differentiating the COM and COP displacements. The COM acceleration was obtained in two ways: (1) using the measured horizontal force according to $\ddot{x}_{COM} = f_{AP}/m$, where f_{AP} denotes the horizontal force in the anteroposterior direction (COMaccf) and (2) by differentiating the COM velocity calculated above (COMaccl). Thus, the former is based on the force plate output and the latter on the laser sensor output. The derivatives of COMaccf (dCOMaccf) and COMaccl (dCOMaccl) were also calculated via differentiation.

The amount of fluctuation was summarized using the root mean square for each variable, i.e., the COP displacement, COM displacement, COP velocity, COM velocity, COMaccf, and COMaccl, after removing the mean value from each time series. The comparison between the young and the elderly was made for each variable using a t-test. $p < 0.05$ served as the level of statistical significance.

The linear correlation of the root mean square fluctuation was evaluated using Pearson's correlation coefficient for each pair of: (1) COP displacement and COM displacement; (2) COP velocity and COM velocity; (3) COP velocity and COMaccf; (4) COP velocity and COMaccl; (5) COP velocity and dCOMaccf; and (6) COP velocity and dCOMaccl.

To evaluate the accuracy of the measured data, we also tested if Eqs. (1) and (2) hold true for the measured data. For that purpose, the fluctuations of the right-hand sides of Eqs. (1) and (2) were quantified with root mean square values using the obtained COM displacement, COMaccf, COMaccl, COM velocity, dCOMaccf, and