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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, $\dot{TQ}$, 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.

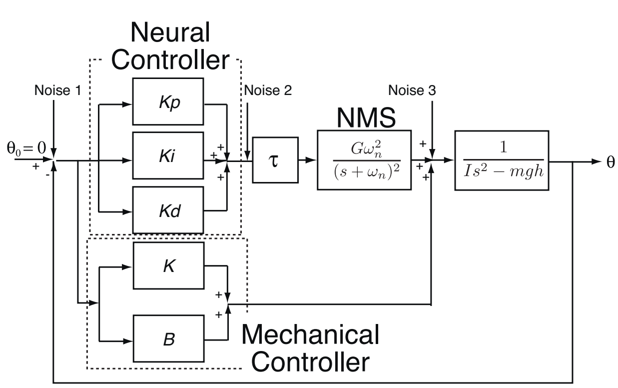

Fig. 1. Computational model used for simulating the control system of quiet standing. The model consisted of a neural controller using a PID controller with gains Kp, Ki, and Kd, and a mechanical controller using a PD controller with gains K and B. A total constant time delay (τ) representing motor and 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 dCOMaccl. They were compared with respective root mean square values of the obtained COP displacement (Eq. (1)) and COP velocity (Eq. (2)). This was done using a Pearson correlation analysis and a linear regression analysis. $p < 0.05$ served as the level of statistical significance.

2.2. Simulation study

2.2.1. Model

The simulation study was performed using Matlab (ver. 7.13, Mathworks, USA) and Simulink (ver. 7.8, Mathworks, USA). A computational model representing the control system of quiet standing was developed (Fig. 1). 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 level of the neural controller. Subsequent to this delay, a critically damped, 2nd order model of the neuromuscular system was inserted to account for the ankle torque generation process [21] producing the neural torque component. Note that the neuromuscular system was not incorporated in [11]. However, as this component improves the model of quiet stance control [21], we decided to include it. As a consequence, Eqs. (3) and (4) do not exactly hold as the amplitude of the output of the neuromuscular system was damped by the neuromuscular system, especially for the fast fluctuations. However, as the slow fluctuations are dominant in TQ, we can still use the following approximations:

$$TQ \approx Kp(\theta - \tau) + Kd(\dot{\theta} - \tau) + Ki\int(\theta - \tau)dt + K\theta + B\dot{\theta} \tag{3'}$$

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

Same as in Eqs. (3) and (4), the neural torque component in addition to the mechanical torque component controlled the body. An inverted pendulum was used as the model of the standing body. Three inverted pendulums were used based on the subjects' body sizes. Three noise inputs corresponding to sensory, motor, and neuro-mechanical noise were injected into the system (Fig. 1) to drive the simulation. All model parameters are listed in Appendix 2.

2.2.2. Simulation and analysis

At first, the stability of the closed-loop system was investigated using the Nyquist stability criterion with gain and phase margins of 2 dB and 5°, respectively. Then, the identified stable control systems were used to simulate postural sway during quiet standing for 150 s. This simulation period, which is longer than the trial period in the experimental study, was adopted to ensure more reliable results for the root mean square measures [22]. However, since the duration of the COP measurement affects the outcome measures [23], this is a potential limitation that may cause differences between the experimental and simulation results. The COP and COM displacements, the COP and COM velocities, the COM acceleration, and the derivative of the COM acceleration were calculated based on the motion of the inverted pendulum.

The fluctuation amount of each variable was quantified using the root mean square. The linear correlations between the fluctuation amounts of the COP and COM displacements, the COP velocity and COM velocity, and the COP velocity and COM acceleration were evaluated using Pearson's correlation coefficient. The contribution of each term on the right-hand side of Eq. (4′) to the torque TQ derivative was evaluated by calculating the ratios between the root mean square of the derivative of each channel output of the model and the TQ derivative. A Friedman test was applied to compare the contribution among the terms, along with a Wilcoxon signed-rank test with Bonferroni correction as a post hoc test. Finally, the correlation between the fluctuation of the TQ derivative and the derivative of each channel output of the model was investigated using Pearson's correlation coefficient. $p < 0.05$ served as the level of statistical significance.

3. Results

3.1. Experimental study

Fig. 2A shows the comparison of postural sway amount quantified with the root mean square value for each variable, between the young and the elderly. Postural sway amounts were not different between the two age groups for COP displacement ($p = 0.079$), COM displacement ($p = 0.927$), and COM velocity ($p = 0.104$). Postural sway amounts were significantly different between the age groups for COP velocity ($p < 0.0001$), COMaccf ($p < 0.0001$), and COMaccl ($p = 0.002$).

The subsequent analyses were executed with the young and elderly samples grouped together as they appeared in the same distribution based on the correlation plots (cf. Fig. 2B), even though the average values were different. Fig. 2B shows the correlation between the selected variables. While all correlations were statistically significant ($p < 0.001$), the correlations between the COM and COP displacements, between the COP velocity and COMaccf, and between the COP velocity and the derivative of COMaccf exhibited very high correlation coefficients ($r > 0.9$).

Fig. 3 shows the results for the validation of Eqs. (1) and (2). All examined correlations were statistically significant ($p < 0.001$). The right and left sides of Eq. (1) were highly correlated and on the identity line ($r > 0.9$; Fig. 3A and B). In case of Eq. (2), the relationship between the right and left sides showed a very high correlation when using COMaccf ($r = 0.945$), but deviated slightly from the identity line (Fig. 3C). When the right hand side of Eq. (2) was calculated using COMaccl, the correlation became much worse ($r = 0.501$), and the relationship deviated more from the identity line (Fig. 3D).

3.2. Simulation study

1.782 simulations were executed in total. Fig. 4 shows the correlations between the variables obtained via the computational simulations. All correlations were statistically significant ($p < 0.001$). The COM and COP displacements ($r = 0952$; Fig. 4A) as well as the COP velocity and COM acceleration ($r = 0.985$; Fig. 4C) were highly correlated. However, the COP velocity and COM velocity exhibited a much lower correlation ($r = 0.799$; Fig. 4B).

Fig. 5A shows example time series of the derivatives of each channel output corresponding to each term in Eq. (4′). We can see that the fluctuation of the Kd-channel and K-channel outputs dominate the fluctuation in the TQ derivative. This observation is quantified in Fig. 5B. The Friedman test revealed that there was a significant difference in the contributions across the terms ($p < 0.0001$), and the post hoc analysis revealed that the contributions were Kd-term > K-term > Ki-term > Kp-term > B-term ($p < 0.0001$ for all comparisons; Fig. 5B). Fig. 5C shows the relationships between the TQ derivative and each term. Among the channels, the Kd- and B-terms were more correlated with the TQ derivative than the other terms. Considering that the contribution of the Kd-term was much higher than of the B-term (Fig. 5B) and that the K-term did not highly correlate with the TQ derivative (Fig. 5C), we can conclude that the Kd-term dominated the fluctuation in the TQ derivative.

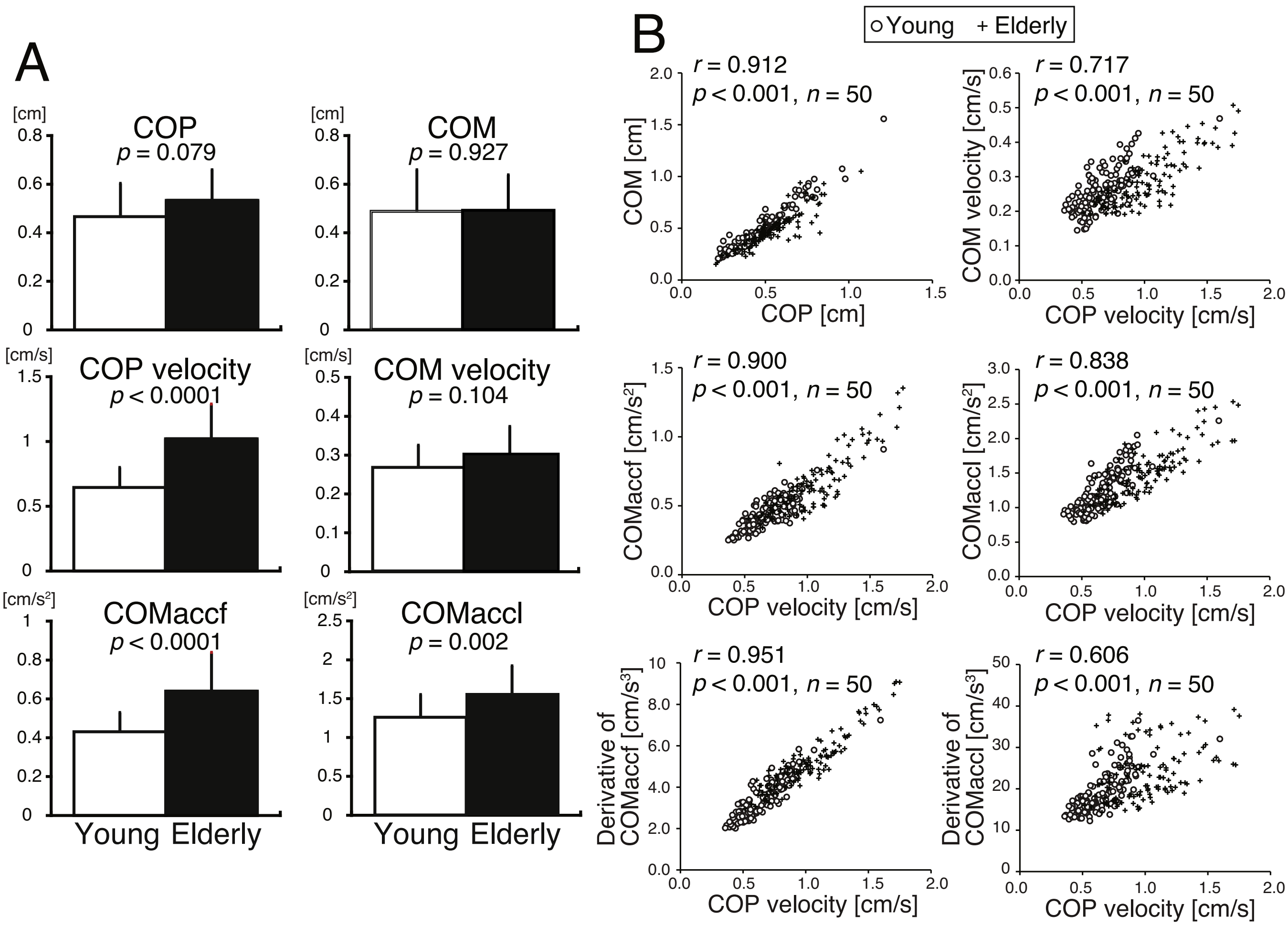

Fig. 2. Comparison of postural sway quantified with the root mean square value for each variable between the young and the elderly (A), and the correlation between the selected variables (B) in the experimental study. The p-value of the t-test for the comparison between the young and the elderly is shown in (A) for each bar graph. The Pearson's correlation coefficient (r) is shown in (B) for each plot combining the young and the elderly samples.

4. Discussion

In the experimental study, we first demonstrated that the COP velocity as well as the COM acceleration detected the age difference of postural sway, whereas the COM velocity did not (Fig. 2A). In addition, the COP velocity was more highly correlated with COMaccf and the derivative of COMaccf than with the COM velocity (Fig. 2B). In the simulation study, we also demonstrated that the correlation between the COP velocity and the COM acceleration was much higher than the one between the COP velocity and the COM velocity (Fig. 4). The contribution of the *Kd*-term dominated the fluctuation in the TQ derivative (Fig. 5).

4.1. COP velocity reflects COM acceleration rather than COM velocity

In agreement with previous studies [1,3,4,6], the COP velocity sensitively detected changes in postural sway due to aging, whereas the COP and COM displacements did not (Fig. 2A). Based on the high correlation between the COP and COM displacements ($r = 0.951$; Fig. 2B) and their theoretical relationship (Eq. (1); see also [4,9]), the COP velocity is likely to correlate with the COM velocity, which was true ($r = 0.717$). However, the correlation between the COP velocity and the COM acceleration was much higher ($r = 0.900$ for COMaccf and $r = 0.838$ for COMaccl) when compared to the one between the COP velocity and the COM velocity. As a consequence, the COM acceleration (COMaccf and COMaccl) detected postural sway changes due to aging similarly well as the COP velocity (Fig. 2A; see also [4,16]), whereas the COM velocity did not (Fig. 2A). Taken together, these results suggest that the COP velocity reflects COM acceleration rather than COM velocity.

While Eqs. (1) and (2) yielded from the equation of motion do not explain the close relationship between the COP velocity and the COM acceleration, it does explain the high correlation between the COP velocity and the derivative of the COM acceleration. However, if the contribution of the COM velocity is high in controlling *TQ*, the close relationship between the COP velocity and the COM acceleration can be explained as follows: If we model the control system using linear controllers (Fig. 1), the *TQ* derivative can reflect the effect of the COM acceleration (Eq. (4′)). In the stable systems, the contribution of the *Kd*-term was very high in comparison to the other right side terms in Eq. (4′) (Fig. 5). As a result, the COP velocity was more highly correlated with the COM acceleration (Fig. 4C) than with the COM velocity (Fig. 4B). Therefore, we can derive from the simulation results that the high correlation between the COP velocity and COM acceleration is indicative of a postural control strategy during quiet standing that relies notably on velocity information [10,12–14].

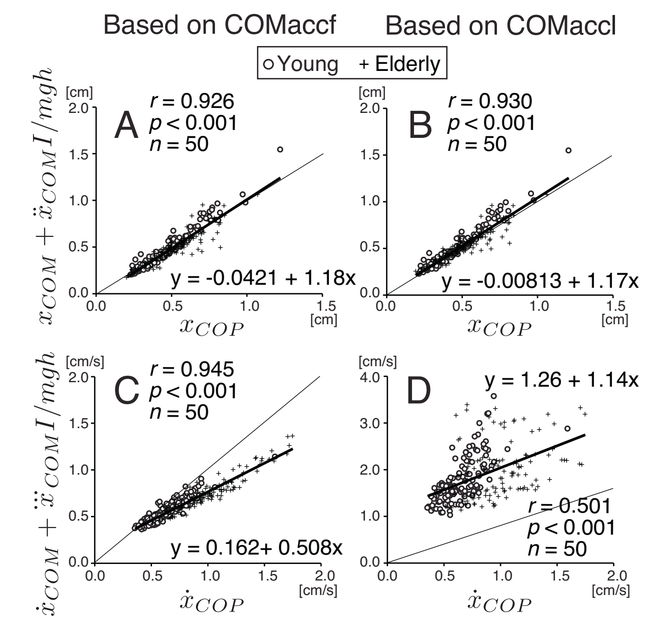

Fig. 3. Validation of Eqs. (1) and (2) in the experimental study. The horizontal axes are the left sides of Eq. (1) (A and B) and Eq. (2) (C and D). The vertical axes are the right sides of the corresponding equations. (A and C) are depicting COMaccf and (B and D) COMaccl. Pearson's correlation coefficient (r) is shown for each plot including the young and elderly samples. The thick lines indicate the linear regression lines for the entire plots including the young and the elderly samples. The thin lines indicate the identity lines.

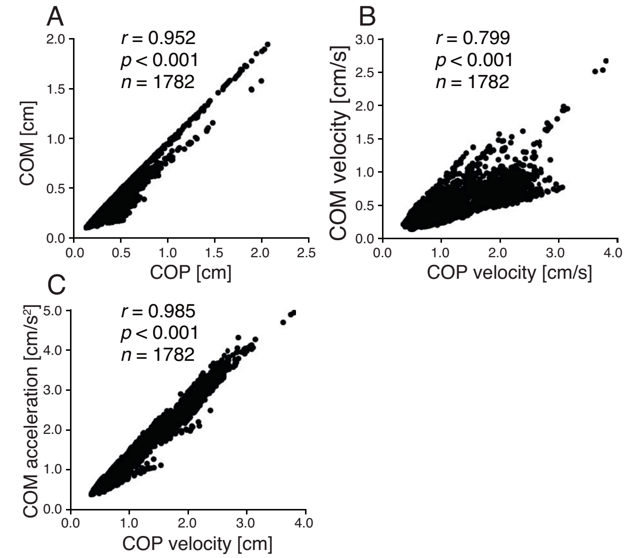

Fig. 4. Correlation of postural sway quantified with the root mean square value between the selected variables in the simulation study. Pearson's correlation coefficient (r) is shown for each plot.

4.2. Accuracy of the measurements

As postural sway during quiet standing is very small while a derivative operation (i.e., differentiation) tends to amplify the magnitude of fast components of the time series, we had concerns

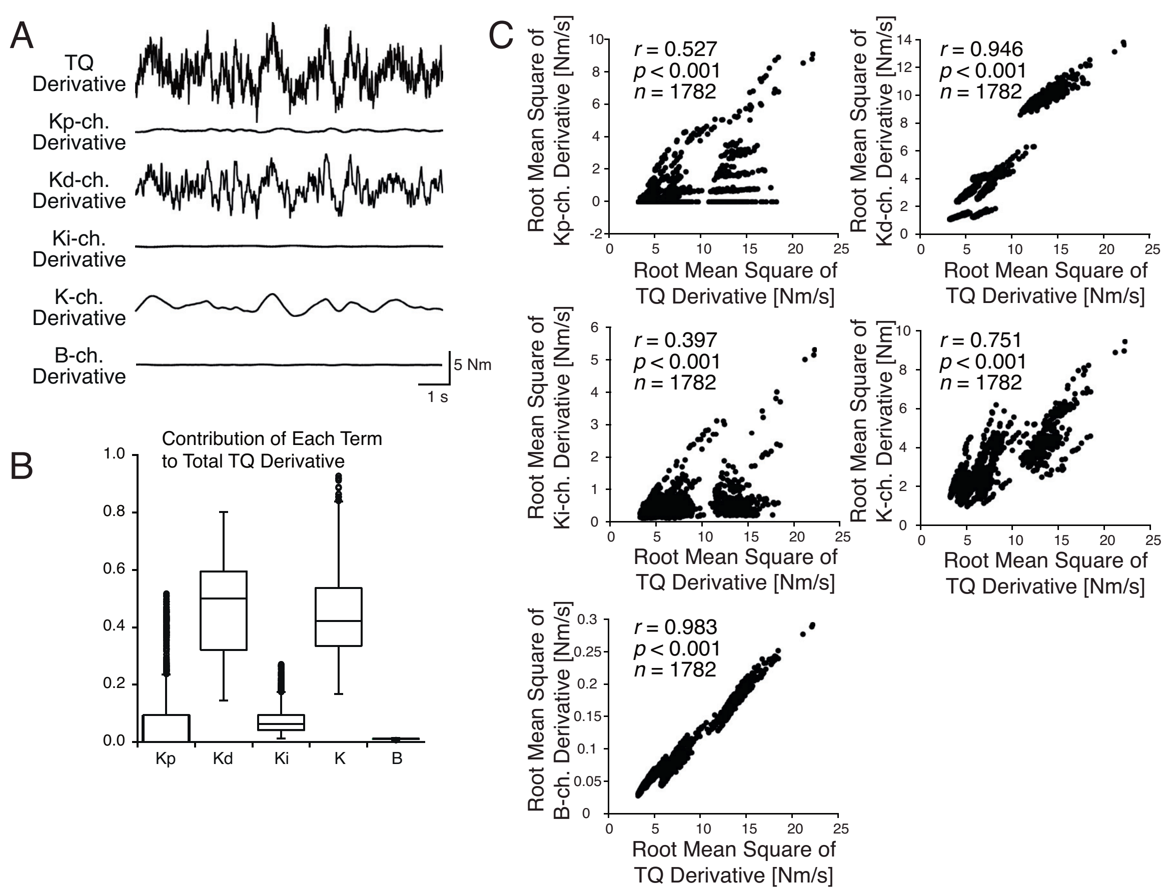

Fig. 5. (A) The example time series of the derivative of each channel output corresponding to each term on the right side in Eq. (4′). (B) The contribution of each term to the total TQ derivative. The ratios between each channel (quantified with the root mean square of the derivative of each channel output of the model) and the root mean square of the TQ derivative are presented as box plots. (C) The correlation of postural sway quantified with the root mean square value between the TQ derivative and each term. Pearson's correlation coefficient (r) is shown for each plot.

that the measurements, particularly the derivatives, might not be sufficiently accurate. Fig. 3A and B successfully proved that our measurements of the COP and COM displacements as well as of the COM acceleration were sufficiently accurate. The high correlation ($r = 0.945$) in Fig. 3C also proved that the COP and COM velocities as well as the COM acceleration measured via the force plate were sufficiently accurate. The fact that the regression line in Fig. 3C was not very close to the identity line is believed to be caused by the estimation error of the anthropometric variables as well as errors caused by the differentiation. On the contrary, the relatively low correlation ($r = 0.501$) in Fig. 3D, which was still statistically significant, suggests that the measurements used for Eq. (2) were not very accurate. The only difference to Fig. 3C was that the laser output was used for calculating the derivative of the COM acceleration. Therefore, this result suggests that COMaccl is not sufficiently accurate for differentiation, causing the relatively low correlation between the COP velocity and the derivative of COMaccl ($r = 0.606$; Fig. 2B) in comparison to the one between the COP velocity and the derivative of COMaccf ($r = 0.951$). Thus, the results related to the derivative of COMaccl are not accurate enough, while the results related to the derivative of COMaccf are, in fact, reliable.

5. 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.

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Conflict of interest statement

We wish to confirm that there are no known conflicts of interest associated with this publication and there has been no significant financial support for this work that could have influenced its outcome.

Appendix 1

When human posture during quiet standing is approximated by an inverted pendulum, the equation of motion is described as

$$I\ddot{\theta} = mgh\sin\theta - TQ. \tag{A.1}$$

Assuming that the body sway amplitude is small, $\ddot{\theta} \approx \ddot{x}_{COM}/h$, $h\sin\theta \approx x_{COM}$, and $TQ \approx mgx_{COP}$. Therefore,

$$x_{COP} \approx x_{COM} + \ddot{x}_{COM}\frac{I}{mgh}. \tag{A.2}$$

Appendix 2

Three inverted pendulums were used with the parameters of:

- Pendulum 1: $h = 0.846$ m, $m = 59.2$ kg, $I = 55.2$ kg m^2
- Pendulum 2: $h = 0.896$ m, $m = 68.8$ kg, $I = 69.0$ kg m^2
- Pendulum 3: $h = 0.797$ m, $m = 49.6$ kg, $I = 41.3$ kg m^2,

which were the average, largest and smallest subjects in the entire subject group, respectively.

The noise, which was injected at three locations in the feedback loop (Noise 1, 2 and 3 in Fig. 1), represented white noise filtered with three different time constants. The used time constants were:

- Time constant of Noise 1: 5, 100, 200 s
- Time constant of Noise 2: 5, 100, 200 s
- Time constant of Noise 3: 0.025, 0.25, 0.5 s.

Other constants used in the model were based on [15,21]:

- $B = 5$ Nm s/rad
- $\tau = 80$ ms

The model parameters tested in the simulation study were:

- Neural controller: $0 \le Kp \le 500$ Nm/rad, $0 \le Kd \le 300$ Nm s/rad, $0 \le Ki \le 300$ Nm/(rad s), with a step size of 100 for each
- Mechanical controller: $0 \le K \le 1000$ Nm/rad, with a step size of 100
- NMS: $0.1 \le 1/\omega_n \le 0.2$ s, with a step size of 0.05.

The identified gains that stabilized the system had the following ranges:

- Neural controller: $0 \le Kp \le 500$ Nm/rad, $100 \le Kd \le 300$ Nm s/rad, $100 \le Ki \le 300$ Nm/(rad s)
- Mechanical controller: $100 \le K \le 1000$ Nm/rad
- NMS: $0.1 \le 1/\omega_n \le 0.2$ s.

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