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Short communication

Kinematic error magnitude in the single-mass inverted pendulum model of human standing posture

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ABSTRACT

Background: Many postural control studies employ a single-mass inverted pendulum model (IPM) to represent the body during standing. However, it is not known to what degree and for what conditions the model's kinematic assumptions are valid.

Research question: Our first objective was to quantify the IPM error, corresponding to a distance change between the ankle joint and center of mass (COM) during unrestricted, natural, unperturbed standing. A second objective was to quantify the error of having the ankle joint angle represent the COM angle.

Methods: Eleven young participants completed five standing conditions: quiet standing with eyes open (EO) and closed (EC), voluntarily swaying forward (VSf) and backward (VSb), and freely moving (FR). The modified Helen-Hayes marker model was used to capture the body kinematics.

Results: The COM distance changed < 0.1% during EO and EC, < 0.25% during VSf and VSb, and < 1.5% during FR. The ankle angle moderately and positively correlated with the COM angle for EO, EC, and VSf, indicating that temporal features of the ankle angle moderately represent those of the COM angle. However, a considerable offset between the two existed, which needs to be considered when estimating the COM angle using the ankle angle. For VSb and FR, the correlation coefficients were low and/or negative, suggesting that a large error would result from using the ankle angle as an estimate of the COM angle.

Significance: Insights from this study will be critical for deciding when to use the IPM in postural control research and for interpreting associated results.

1. Introduction

Since upright standing resembles an inverted pendulum, many postural control studies have employed a single-mass inverted pendulum model (IPM) to represent the body during standing, with the body's center of mass (COM) rotating about the ankle joint [1–5]. The equation of motion of this model, however, is only valid if the pendulum length h, i.e., the distance between the ankle joint and the COM, is constant. Considering that h could vary due to anti-phase actions between the upper and lower body [6–9], it is surprising that no study to date has quantified to what degree and for what conditions this assumption is valid.

When using the IPM to model upright standing, studies have furthermore relied on approximating the angle between vertical and the IPM (θ_{COM}) by the angle between vertical and the shank (θ_{ANK}) [1,3]. Similar to the use of the IPM itself for characterizing postural control, this approximation again assumes, to some degree, a rigid human body during standing. Therefore, its validity is not guaranteed.

Motivated by the need to validate the above assumptions, two study objectives were identified: (1) to quantify to what extent the length h of the IPM is constant during unrestricted, natural, unperturbed standing; and (2) to quantify the error in using θ_{COM} instead of θ_{ANK} to capture the standing kinematics. Five standing conditions were adopted to investigate the robustness of the described assumptions during standing. Similar to previous studies, only the anterior-posterior direction was considered since the ankle joint primarily regulates balance in this direction.

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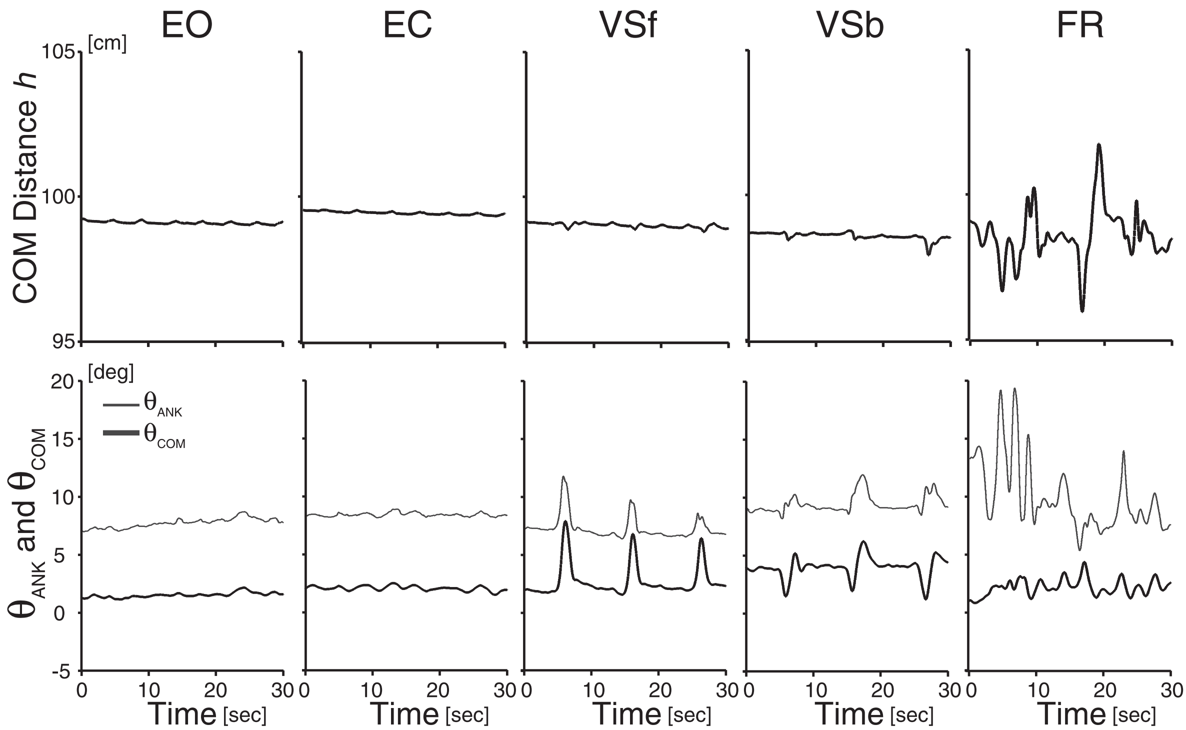

Fig. 1. Example time series for the center of mass (COM) distance (h), the ankle angle (θ_{ANK}) and the COM angle (θ_{COM}) for each condition of eyes open (EO) and closed (EC); voluntarily swaying forward (VSf) and backward (VSb); and free movement (FR). During the VSf and VSb periods presented here, the participant swayed forward and backward three times each. Only temporal features for 30 s are shown.

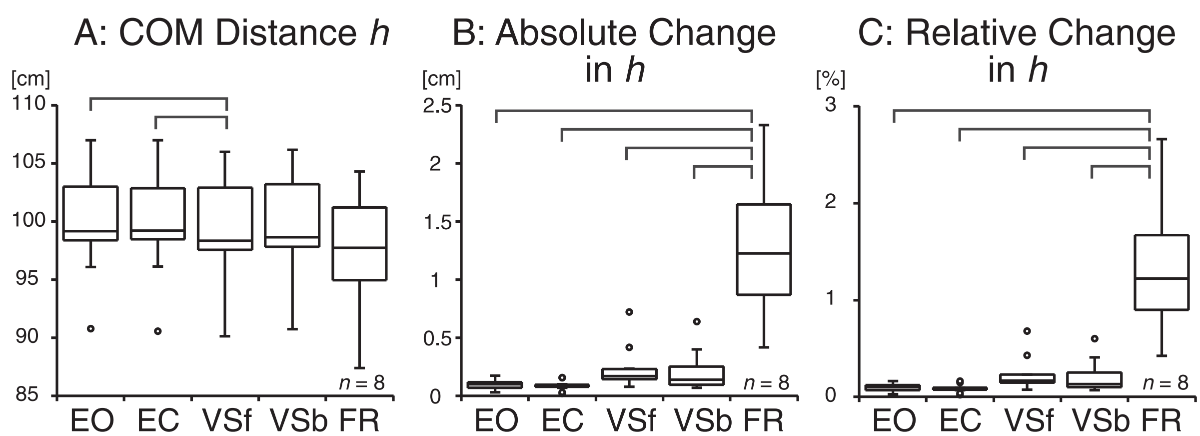

Fig. 2. A: The distribution of the mean value of the center of mass (COM) distance for each condition of eyes open (EO) and closed (EC); voluntarily swaying forward (VSf) and backward (VSb); and free movement (FR). B: The absolute change of the COM distance for each condition. C: The relative change of the COM distance, or coefficient of variation, for each condition. Each horizontal bar indicates a statistically significant difference between the two corresponding conditions. Note that FR included n = 8.

2. Methods

Eleven healthy, young males (age 20.7 ± 3.6years; height 173.6 ± 7.1 cm; body mass 68.0 ± 8.6 kg; mean ± standard deviation (SD)) participated. They gave written informed consent, and the experimental procedures were approved by the local ethics committee.

The conditions to investigate the variability of h were: (1) quiet standing with eyes open (EO) and (2) eyes closed (EC); (3) voluntarily swaying forward (VSf) and (4) backward (VSb), with eyes open; and (5) freely moving (FR) with eyes open. EO, EC, VSf, and VSb lasted for 120 s each, and FR for 30 s. Since FR was added after participant #3, only 8 participants completed it. During EO, participants were instructed to look at a mark placed at eye level two meters ahead and, for EC, to close their eyes. During VSf and VSb, participants had to maximally sway toward and then away from a target in front of (VSf) or behind (VSb) them at a speed of 1 cycle per 12 s (~4 s toward target, ~4 s away from target, ~4 s for resetting body). In one 120 s trial, only one direction (VSf or VSb) was performed ten times, and a metronome helped the participant complete each cycle. During FR, participants had to sway freely in any direction while being allowed to bend their knees. In EO, EC, VSf, and VSb, participants had to cross their arms above the chest, whereas, in FR, to move them dynamically. For all conditions, participants were asked to not lift their feet; no instructions were given regarding the use of ankle or hip strategy.

Body kinematics were recorded at 200 Hz using a motion capture system (Rapter-E, Motion Analysis Corp., USA) with six cameras. Twenty-nine reflective markers were affixed to the body following the modified Helen-Hayes kinematic model with 13 segments [10]. Similar to previous work [11], the kinematic data were smoothed using a 2nd-order Butterworth low-pass filter with a cut-off frequency of 4 Hz.

Each participant's COM location and the representative ankle joint location (the average of both ankle joints) were calculated [12]. h was obtained as the two-dimensional distance between the ankle joint and COM location in the sagittal plane. For all conditions, we calculated the: (1) average COM distance; (2) absolute COM distance change (i.e., SD of the time series); and (3) relative COM distance (i.e., coefficient of variation). To compare θ_{COM} with θ_{ANK}, we also assessed: (4) the absolute change in θ_{ANK} and θ_{COM} (i.e., SD of respective time series); (5) the mean difference between θ_{COM} with θ_{ANK}; and (6) Pearson's correlation coefficient between θ_{COM} with θ_{ANK}.

For each of (1)–(3), and (5), the statistical difference across conditions was examined using a paired t-test with Bonferroni correction. The two variables in (4) were compared, for each condition, using a paired t-test with Bonferroni correction ($\alpha = 0.05$ for all tests).

3. Results

Fig. 1 (top) shows example time series for the COM distance h across conditions. The distribution of the mean of the h time series is shown in Fig. 2A, and of its absolute change in Fig. 2B. The relative changes of the COM distance, or coefficient of variation (Fig. 2C), were 0.092 ± 0.040% (EO), 0.089 ± 0.040% (EC), 0.232 ± 0.175% (VSf), 0.211 ± 0.165% (VSb), and 1.33 ± 0.69% (FR).

Fig. 1 (bottom) shows that the temporal features of θ_{ANK} and θ_{COM} were similar for EO, EC and VSf. Except for FR, the changes of θ_{ANK} and θ_{COM} were not statistically different across conditions (Fig. 3A). A considerable offset existed between the two angular displacements (Fig. 1) as reflected in the mean difference between θ_{ANK} and θ_{COM} (Fig. 3B). θ_{ANK} and θ_{COM} showed moderate to high correlations for EO, EC and VSf (Fig. 3C). For VSb and FR, the correlation coefficient between θ_{ANK} and θ_{COM} varied significantly across participants (Fig. 3C).

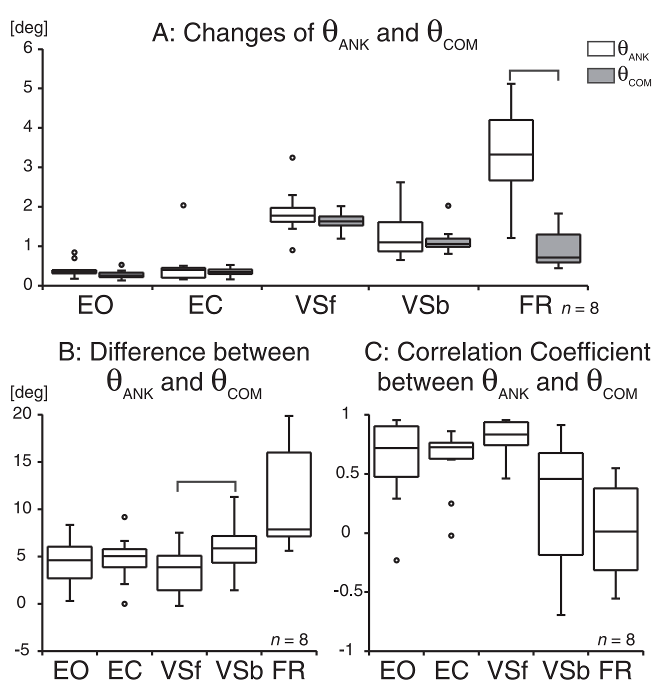

Fig. 3. A: The change of the ankle (θ_{ANK}) and the center of mass (COM) (θ_{COM}) angle for each condition of eyes open (EO) and closed (EC); voluntarily swaying forward (VSf) and backward (VSb); and free movement (FR). B: The mean difference between the ankle and the COM angle for each condition. C: Correlation coefficients between the ankle and the COM angle for each condition. Note that FR included n = 8.

4. Discussion

A key condition for the equation of motion of the IPM to be accurate is that the COM distance is constant. Our findings demonstrate that, during EO and EC, the change in the COM distance was very small, i.e., less than 1 mm, which is close to the motion capture manufacturer's declared accuracy. Therefore, the kinematic error of the IPM can be assumed to be acceptable during *quiet standing*, as long as identifying the contribution of interaction torques from hip or knee motion to the ankle joint torque [6–9] is not of relevance. If it is [13,14], more differentiated, multi-segment models of standing should be used (e.g., [14]).

In VSf and VSb, the COM distance change tended to be larger than for EO and EC (< 2.5 mm or 0.25%), but without statistical significance. This implies that the kinematic error of the IPM would still be acceptable for these well-defined, dynamic postural sway conditions. During FR, however, the COM distance change was significantly larger than for the other conditions (> 1.0 cm or 1.0%), most likely due to an increase in anti-phase actions between the upper and lower body. As the arms were moving, also the body's moment of inertia must dynamically change and not adhere to the IPM's assumption of constancy. Due to these reasons – and as expected – the IPM should not be used to study FR or comparable standing conditions.

While the fluctuations of θ_{ANK} and θ_{COM} were not different and moderately correlated for EO, EC, and VSf, an offset in the angles existed. The correlation results, which agree with Gage et al. [2], imply that the ankle joint angle may be used to capture general trends in body sway fluctuation during EO, EC, and VSf. However, the offset in the angles suggests that, if it is critical to quantify the body sway magnitude over time, θ_{ANK} should not be used as a θ_{COM} surrogate. Finally, the fact that, for FR, the two angles exhibited fluctuations that were significantly different, had the largest offset, and were not substantially correlated, corroborates the previous conclusion that the IPM cannot be used for unconstrained conditions of voluntary body sway.

4.1. Limitations

One limitation was the unbalanced design, with FR being performed by only 8 participants. Another limitation was that we only investigated upright standing in the young; since the elderly generally make greater use of the hip strategy during standing [15], future work should examine whether our results also hold true for this population.

4.2. Conclusions

Using the IPM and/or the ankle angle as an approximation of the model's COM angle are common choices in the field. By quantifying the kinematic error associated with these choices for different standing conditions, the present work allows postural control researchers to consider the model's error in their interpretations or to decide for more complex models when appropriate.

Conflict of interest

None.

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