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Theoretical and experimental indicators of falls during pregnancy as assessed by postural perturbations

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ABSTRACT

Throughout pregnancy, women experience physical, physiological, and hormonal alterations that are often accompanied by decreased postural control. According to one study, nearly 27% of pregnant women fell while pregnant. This study had two objectives: (1) to characterize the postural responses of pregnant fallers, nonfallers, and controls to surface perturbations, and (2) to develop a mathematical model to gain insights into the postural control strategies of each group. This retrospective analysis used experimental data obtained from 15 women with a fall history during pregnancy, 14 women without a fall history during pregnancy, and 40 nonpregnant controls. Small, medium, and large translational support surface perturbations in the anterior and posterior directions were performed during the pregnant participants' second and third trimesters. A two-segmented mathematical model of bipedal stance was developed and parameterized, and optimization tools were used to identify ankle and hip stiffness, viscosity, and the feedback time delay by searching for the best fits to experimental COP data. The peak differences between the center of pressure and center of gravity (COP-COG) values were significantly smaller for the pregnant fallers compared with the pregnant nonfallers and controls ($p < 0.01$). Perturbation magnitude was a significant factor ($p < 0.01$), but perturbation direction was not ($p = 0.24$). Model fits were obtained with a mean goodness of fit value of $R^2 = 0.92$. Theoretical results indicated that pregnant nonfallers had higher ankle stiffness compared with the pregnant fallers and the controls, which suggests that ankle stiffness itself may be the dominant reason for the different dynamic response characteristics (e.g., peak COP–COG) observed. We conclude that increasing ankle stiffness could be an important strategy to prevent falling by pregnant women.

1. Introduction

As pregnancy advances, women undergo various physical, physiological, and hormonal alterations. For example, they typically gain 11–16 kg in weight [1]. These weight gains are primarily concentrated in the abdominal region and can increase lumbar lordosis [2]. Hormonal fluctuations can increase ligamentous laxity [3,4], and changes in plantar foot pressures are observed [5]. Such alterations can lead to balance problems. According to one study, nearly 27% of pregnant women experienced an accidental fall [6], which is a rate comparable to the 30% rate of falls observed in individuals aged 65 yrs and older [7]. Falls that cause fractures and sprains can contribute to the fear of falling [8], while very serious falls can terminate maternal or fetal life [9,10].

Several researchers have studied the changes in postural control during pregnancy. Butler et al. reported that the center of pressure (COP) excursion in a pregnant group increased in length compared with a control group during quiet stance, and that the amount of weight gained was not significantly associated with the postural sway measures investigated [11]. Nagai et al. showed an increased area of body sway and length of anterior–posterior (A/P) body sway in a pregnant group compared with nonpregnant controls during quiet stance, and that high anxiety correlated with instability [12]. Oliveira et al. reported that pregnant women exhibited larger elliptical fits to COP trajectories as pregnancy progressed and higher COP frequency content along the A/P direction in the absence of visual inputs [13]. However, none of these studies addressed changes of postural control in response to external perturbations.

McCrory et al. investigated pregnant women's responses to A/P support surface translations [14]. Their main finding was that

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pregnant fallers, who reported at least one fall during pregnancy, had a truncated COP displacement immediately in response to the perturbation compared with pregnant nonfallers and controls.

Both the COP and center of gravity (COG) variables have been used individually to quantify postural stability in biomechanics studies. In terms of postural control during quiet standing, the COG and COP can be interpreted as the controlled and controlling variables, respectively [15–17], where the COP is proportional to the ankle torque [18]. COP and COG can also be measured simultaneously and the scalar difference between COP and COG (COP–COG) can be computed as a metric to characterize postural control. The COP–COG has been characterized in both the time and frequency domain. The most common metrics associated with the COP–COG variable are amplitude [18], standard deviation [17], root mean square [19], peak magnitude of displacement [20], latencies of initial and peak displacement [20], and frequency spectra [21]. COP–COG metrics have been applied, for example, to elderly stroke patients [19] and healthy elderly and young subjects [17], but have not yet been used to characterize balance in pregnant women.

In terms of mathematical modeling, the inverted pendulum, a one-link representation capturing a single degree of freedom, is the simplest mathematical model for describing bipedal postural control [22]. Simple one and two degree of freedom models have been used to study the effects of biofeedback on individuals with vestibular loss [23] and the risk of falling due to obesity [24]. However, to the best of our to our knowledge, these models have not been applied to pregnancy.

The specific goals of this study are (1) to investigate whether COP–COG can differentiate pregnant fallers from nonfallers; and (2) to use mathematical models to gain insights into the differences in postural control strategies between pregnant fallers and nonfallers.

2. Methods

The experimental data were obtained from a prior study of 15 women with a fall history during pregnancy (29.4 ± 4.7 yrs), 14 women without a fall history during pregnancy (30.6 ± 3.8 yrs), and 40 controls (26.5 ± 6.4 yrs) who were not pregnant and had a body mass index that matched the pre-pregnancy indices of the pregnant subjects [14]. The study had originally enrolled 41 pregnant women, however 12 subjects could not complete the study: four delivered pre-term, four had complications (preeclampsia, toxemia, fall with ankle sprain), three did not follow through, and one moved out of area. The average subject height was 165.8 ± 5.6 cm for the controls and 166.1 ± 6.6 cm for the pregnant women. Controls had a mass of 64.7 ± 8.8 kg, whereas the pregnant women had a mass of 73.9 ± 9.9 kg and 81.3 ± 11.1 kg in the second and third trimesters, respectively. Subjects in the pregnant and control groups were not matched based on the number of previous pregnancies. In the pregnant group, 27 women were primigravid; five stated it was their second pregnancy, and nine of the women said it was their third pregnancy. Thirty-three of the control women were nulligravid. Six controls reported that they were pregnant one time and one reported that she had been pregnant twice.

The pregnant subjects were tested twice. The first visit occurred in the middle of the second trimester. The average gestational age during the first data collection session was 20.9 ± 1.2 weeks. The second visit occurred at 35.8 ± 1.5 weeks. The controls participated in a single study visit.

Each participant gave informed consent prior to the start of the experimental procedures, and the study was conducted in accordance with the Helsinki Declaration and approved by the University of Pittsburgh Institutional Review Board. Participants were questioned as to their fall history during this pregnancy. Subjects were retrospectively classified as "pregnant fallers" if they fell at any point during their pregnancy. A fall was defined as a loss of balance such that any part of their body except the sole of the foot touched a support surface. Subject height and weight were obtained using a standard medical scale and stadiometer. Anthropometric data were collected according to the methods of Pavol et al. [25].

Translational surface perturbations in the anterior and posterior directions were generated using the Equitest (NeuroCom International, Inc., Clackamas, OR, USA) Motor Control Test (MCT). Three trials were performed at small, medium, and large perturbation magnitudes. The perturbation magnitude, i.e., the translation magnitude in inches, was determined through the manufacturer's formula $xh/72$, where h is the subject's height in inches, and x is 0.5 inches, 1.25 inches, and 2.25 inches for the small, medium, and large perturbations, respectively. All subjects were fitted with a chest and hip harness. The straps of the harness were only placed around the shoulders and upper thighs, thereby protecting the fetus (no subjects actually lost balance during testing). Subjects were instructed to stand on the platform with their feet hip-width apart and stare straight ahead. COP was directly measured and COG was estimated by the Equitest platform.

The peak COP–COG metric obtained was analyzed using a three-way ANOVA with Tukey's post hoc test ($\alpha = 0.05$), where the subject group (controls (C), pregnant nonfallers (PNF), pregnant fallers (PF)), perturbation direction (backward, forward), and perturbation magnitude (small, medium, large) were designated as fixed factors. The experimental data from the second and third trimesters were averaged based on a one-way ANOVA (with trimester as the fixed factor) that showed that there was no significant difference in peak COP–COG between trimesters for any perturbation condition (p-values ranged from 0.12 to 0.99).

For the theoretical part of the study, a single-segmented mathematical model was considered first, but the goodness of fit for the COP data was worse when compared with the goodness of fit for a two-segmented representation; i.e., the single-segmented model was found to be inadequate for this application [26]. Thus, a two-segmented model was implemented to represent the dynamics of the body as shown in Fig. 1 along with the postural

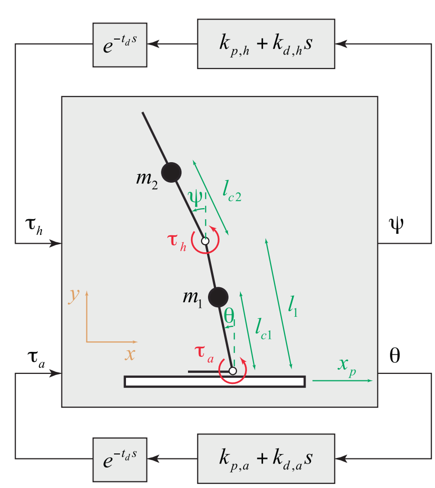

Fig. 1. The mathematical model employed in this study: a two-link inverted pendulum with lumped mass representation describes the dynamics of the body. x_p represents the platform position; l_1 is the first segment length, l_{c1}, and l_{c2} are the center of mass heights for segments 1 and 2, m_1 and m_2 are the segment masses, θ and ψ are the absolute ankle and hip joint ankles, and τ_a and τ_h are the ankle and hip joint torques. The feedback control loops around the ankle and hip joints represent the postural control model.

control model. The system equations, linearized around the upright equilibrium, are given by

$$\begin{bmatrix}\ddot{\theta}\\ \ddot{\psi}\end{bmatrix} = \begin{bmatrix}\frac{g(l_{c1}m_1 + l_1 m_2)}{l_{c1}^2 m_1} & -\frac{g l_1 m_2}{l_{c1}^2 m_1}\\ -\frac{g l_1 (l_{c1}m_1 + l_1 m_2)}{l_{c1}^2 l_{c2} m_1} & \frac{g}{l_{c2}}\left(1 + \frac{l_1^2 m_2}{l_{c1}^2 m_1}\right)\end{bmatrix}\begin{bmatrix}\theta\\ \psi\end{bmatrix} + \begin{bmatrix}-\frac{1}{l_{c1}^2 m_1} & \frac{(l_1 + l_{c2})}{l_{c1}^2 l_{c2} m_1} & \frac{1}{l_{c1}}\\ \frac{l_1}{l_{c1}^2 l_{c2} m_1} & -\frac{l_{c1}^2 m_1 + l_1(l_1 + l_{c2})m_2}{l_{c1}^2 l_{c2}^2 m_1 m_2} & \frac{-l_1 + l_{c1}}{l_{c1} l_{c2}}\end{bmatrix}\begin{bmatrix}\tau_a\\ \tau_h\\ \ddot{x}_p\end{bmatrix} \tag{1}$$

where θ and ψ are the absolute ankle and hip joint ankles, τ_a and τ_h are the ankle and hip joint torques, x_p represents the platform position, g is the gravitational constant, l_1 is the first segment length, l_{c1}, and l_{c2} are the center of mass heights for segments 1 and 2, and m_1 and m_2 are the segment masses, respectively. The ankle torque τ_a and hip torque τ_h are expressed in the Laplace domain as

$$\begin{aligned}\tau_a(s) &= (k_{p,a} + k_{d,a}s)e^{-t_d s}\theta(s)\\ \tau_h(s) &= (k_{p,h} + k_{d,h}s)e^{-t_d s}\psi(s)\end{aligned} \tag{2}$$

where s is the Laplace variable. In other words, a proportional-derivative feedback control scheme was employed for each joint independently, i.e., ankle torque did not explicitly depend on hip joint states and vice versa. The parameters $k_{p,a}$ and $k_{p,h}$ were the proportional control gains, i.e., they led to a torque component proportional to the ankle and hip angles, respectively, and were therefore referred to as ankle stiffness and hip stiffness, respectively. The parameters $k_{d,a}$ and $k_{d,h}$ were the derivative control gains, i.e., they led to a torque component proportional to the ankle and hip angular velocities, respectively, and were therefore referred to as the ankle viscosities and hip viscosities, respectively. Note, however, that even though the terms "stiffness" and "viscosity" are usually associated with passive springs and dampers, Eq. (2) aimed to capture the active torques generated by the postural control mechanism and there were no passive elements in the model. A time delay t_d was also included in each feedback loop.

The input to the model was the platform acceleration $\ddot{x}_p$. The output was the COP as given by

$$\mathrm{COP} = -\frac{\tau_a}{g(m_1 + m_2)} \tag{3}$$

To parameterize the model, averaged anthropometric measurements of subjects were used to find the values for the parameters l_1, l_{c1}, and l_{c2}. The total mass was distributed between m_1 and m_2 according to previously reported formulae for segment weights [25]. Finally, the tuning of the postural control parameters $k_{p,a}$, $k_{d,a}$, $k_{p,h}$, $k_{d,a}$, and t_d was formulated as an optimization problem. The objective of the optimization was defined to match the averaged experimental COP trajectories as well as possible. The cost function was defined as

$$J = \|\mathbf{COP}_{\mathrm{exp}} - \mathbf{COP}_{\mathrm{sim}}\|_2 + \|\Delta\mathbf{COP}_{\mathrm{exp}} - \Delta\mathbf{COP}_{\mathrm{sim}}\|_2 \tag{4}$$

where $\mathbf{COP}_{\mathrm{exp}}$ and $\mathbf{COP}_{\mathrm{sim}}$ were the experimental and simulated COP trajectory vectors, and Δ notation represented the vector of differences between the adjacent elements of the vector to which the Δ operator was applied and could be interpreted as a scaled derivative operator in the discrete domain. The first and second terms of the cost function helped to decrease the difference in magnitude and shape between the COP trajectories, respectively. A combination of genetic and gradient-based algorithms in the MATLAB (MathWorks, Inc., Natick, MA, USA) Optimization and Global Optimization toolboxes was used to globally minimize the cost function. Only COP was used during model fitting, because only COP was directly measured by the Equitest platform.

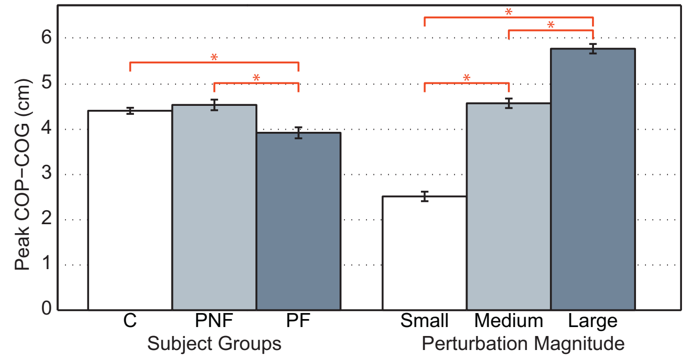

Fig. 2. The population marginal means of subject groups and perturbation magnitudes for peak COP–COG. Asterisks indicate statistically significant differences ($p < 0.05$). Bars denote standard error. C: controls, PNF: pregnant nonfallers, PF: pregnant fallers.

The experimental data for each subject group were ensemble-averaged over all subjects and all trials per each perturbation condition. Only forward perturbations were considered based on the finding that perturbation direction was not a significant factor.

The goodness of fit of the model's response to the experimental data was evaluated using the coefficient of determination $R^2 = 1 - SS_{\mathrm{err}}/SS_{\mathrm{tot}}$, where $SS_{\mathrm{err}} = \sum_i (\mathrm{COP}_{\mathrm{exp},i} - \mathrm{COP}_{\mathrm{sim},i})^2$ is the sum of squares of the residuals, and $SS_{\mathrm{tot}} = \sum_i (\mathrm{COP}_{\mathrm{exp},i} - \overline{\mathrm{COP}}_{\mathrm{exp}})^2$ is the total sum of squares with $\overline{\mathrm{COP}}_{\mathrm{exp}}$ representing the mean experimental COP.

3. Results

For both the forward and backward perturbations, the COP and COG were initially displaced in the opposite direction of the perturbation followed by an overshoot in the direction of the perturbation before a return to an upright position. For a detailed analysis of the COP data, see [14].

Fig. 2 illustrates that the pregnant fallers showed significantly smaller peak COP–COG values compared with the pregnant nonfallers and controls ($p < 0.01$, $F(2,389) = 7.83$). Perturbation magnitude was a significant factor ($p < 0.01$, $F(2,389) = 247.99$), but perturbation direction was not ($p = 0.24$, $F(1,389) = 1.37$). Peak COP–COG increased significantly from the small to medium perturbations, as well as from the medium to large perturbations.

Model fits were obtained with a mean goodness of fit value of $R^2 = 0.92$. An example model fit is shown in Fig. 3. Table 1 summarizes the postural control parameters obtained for all fits, and Fig. 4 illustrates the normalized and averaged parameter values. The model shows that pregnant nonfallers had on average

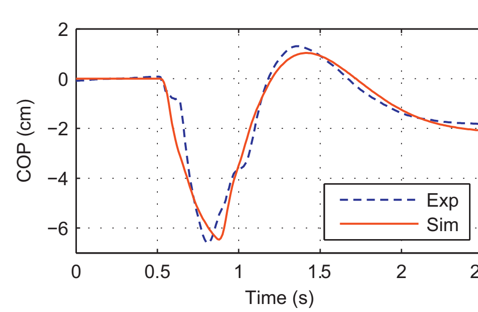

Fig. 3. Example model fitting result for the forward large perturbation and pregnant nonfaller subject group ($R^2 = 0.94$).

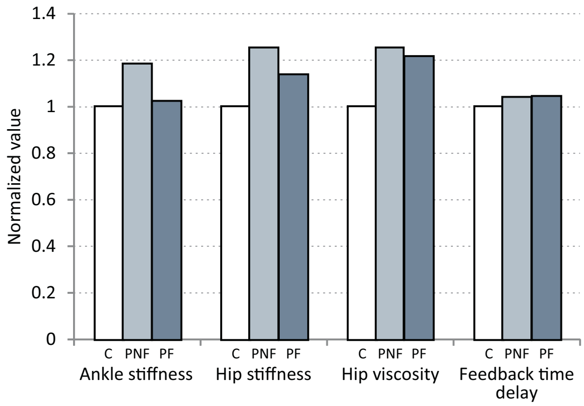

Fig. 4. Postural control parameters for the subject groups averaged over the three perturbation magnitudes. The values are normalized with respect to the corresponding values for the controls. The figure highlights that the main difference between the pregnant fallers and nonfallers is in the ankle stiffness. Ankle viscosity converged to zero in all conditions and is hence not shown in the figure.

about 18% higher ankle stiffness and about 25% higher hip stiffness and viscosity compared with the controls, whereas the feedback delay increased by about 4%. Note that the ankle stiffness values for the pregnant fallers were closer to those of the controls, showing only a 2% difference, whereas the remaining values generally were closer to those of the pregnant nonfallers. In particular, hip stiffness increased by about 14%, hip viscosity by about 22%, and feedback delay by about 4% in the pregnant fallers compared with the controls.

4. Discussion

Our experimental data indicate that the pregnant fallers utilize smaller peak COP–COG values compared with the pregnant nonfallers and controls, while our theoretical data assert that the pregnant nonfallers demonstrate increased ankle stiffness compared with the pregnant fallers and controls.

If the COP–COG variable is considered as an error signal that the balance system senses [18], then the pregnant fallers' smaller COP–COG values would indicate smaller deviations requiring correction, resulting in smaller corrective torques, and therefore smaller COP excursions, because COP is proportional to ankle torque. Hence, the pregnant fallers' smaller peak COP–COG values may reflect the inability to generate adequate corrective torque in response to surface perturbations. This interpretation is consistent with the model's prediction that the ankle stiffness does not increase in this population as it does in pregnant nonfallers. Additionally, the peak COP–COG metric can be used to characterize the margin of stability during dynamic posturography. Our results suggest that pregnant nonfallers have a larger margin of dynamic stability than the pregnant fallers, due in part to greater ankle stiffness. Fear of falling could be an additional potential factor for some of the subjects; however, due to the fact that some of the falls were reported before the testing and some after, fear due to a prior fall is unlikely to be a general factor. Furthermore, the perturbations used in the tests were too small to induce a fall. Hence, fear of falling is unlikely to be the dominant cause for the difference observed between pregnant fallers and nonfallers.

The model parameter values for pregnant fallers were generally close to those of the pregnant nonfallers except for ankle stiffness. Our theoretical findings suggest that pregnant nonfallers have stiffer ankles than both the controls and pregnant fallers. Despite the increase in mass during pregnancy, pregnant fallers do not compensate in terms of ankle stiffness and appear to continue behaving like the controls group. These findings suggest that ankle stiffness itself may be the dominant reason for the different dynamic response characteristics (e.g., peak COP–COG) observed between pregnant fallers and nonfallers. The increase in ankle stiffness in pregnant nonfallers may be a compensatory mechanism to some of the changes, such as increased mass and ligament laxity [3,4], or decreased nerve conduction velocity [27] and neuromuscular coordination [11,12] during gestation. We did not observe an increase in ankle stiffness in pregnant fallers, which could explain why this group is more susceptible to falling.

Physical explanations for the higher ankle stiffness in the pregnant nonfallers are not readily apparent. Since all of the sedentary pregnant women fell, exercise participation may play a role, although the pregnant women who were active were equally likely to fall or not fall [14]. Since we did not measure serum hormone levels, we cannot state if the pregnant nonfallers had different levels of pregnancy-associated hormones. There were no anthropometric differences between the pregnant fallers and nonfallers [14]. Therefore, the differences in ankle stiffness between the pregnant fallers and nonfallers could be sensorimotor in nature; some of the pregnant fallers may not have been as familiar with the musculoskeletal, neuromuscular, and weight distribution related changes in their bodies because they were more sedentary. Inactivity may have led to the lack of development of adequate compensatory responses, which caused increased incidences of balance loss throughout the second and third trimesters.

Our findings are also consistent with previously reported COP–COG findings in the elderly [17], elderly with stroke [19], and individuals with stroke [21]. In all three populations studied, a change in the COP–COG metric used reflected a change in the postural control strategy. In the elderly subjects tested during quiet stance, larger COP–COG values were observed compared with healthy young controls. In the elderly stroke subjects tested during quiet stance, the COP–COG values were larger compared with the healthy elderly controls. In the stroke patients who served as their own controls, there was a significant reduction in the COP–COG values following in-patient rehabilitation. During quiet standing, postural control is maintained by minimizing the COP–COG "error signal". In the case of perturbed stance, however, a larger COP–COG "error signal" is required to generate adequate corrective torques to maintain balance. The smaller peak COP–COG values in pregnant fallers can thus explain why this group was prone to fall.

Table 1
Model fitting results for forward perturbations. Ankle viscosity converged to zero in all conditions.

Parameter	Small perturbation			Medium perturbation			Large perturbation		
	C	PNF	PF	C	PNF	PF	C	PNF	PF
Ankle stiffness (N m/rad)	1063.0	1293.4	1069.4	1060.6	1218.3	1043.3	929.6	1101.5	1009.9
Hip stiffness (N m/rad)	166.3	223.3	170.47	130.6	156.5	156.4	127.8	151.9	156.1
Hip viscosity (N m s/rad)	33.0	40.70	39.7	28.9	37.1	32.9	21.3	26.4	28.5
Feedback time delay (ms)	24.1	23.7	24.1	28.0	28.3	27.4	21.2	24.2	25

Corbeil et al. developed a mathematical model of postural control to evaluate the effects of obesity on the stabilizing torque needed at the ankle joints during perturbed stance [24]. In nonpregnant, nonobese persons, the COG is aligned slightly anterior to the ankle joints [28]. No initial resultant ankle torque is required when the COP and COG are in alignment with the ankle joint [24]. Like the model employed in their study, the pregnant women in this study have a proportion of body mass distributed anterior to the projection of the COM, resulting in an anterior displacement of the COM, and a subsequent non-zero ankle torque required to stabilize the system. Corbeil et al. observed that increased muscular ankle torque (defined as the summation of an initial ankle torque and physiological ankle torque) was needed for the obese subjects to maintain postural stability during perturbed stance compared with the nonobese subjects. Our results are consistent with these findings in the sense that pregnant fallers do not increase their ankle stiffness like the pregnant nonfallers do and hence generate less ankle torque.

The pregnant subjects had on average up to 16.6 kg higher body weight than the controls. However, this difference in body weight did not affect the COP–COG values between controls and pregnant nonfallers, despite prior findings that suggest body weight is a strong predictor of postural stability [29,30]. This could indicate that the weight increase during pregnancy is too small to create a significant change in postural stability as measured by the COP–COG metric. Furthermore, despite their similar weights, some pregnant subjects were fallers and others nonfallers, indicating that body weight alone was not a strong indicator of postural control.

This study relied on several assumptions. The COG data used during the COP–COG analysis were derived from the Equitest-measured COP data. We suggest that more accurate results can be obtained by directly measuring COG. Corriveau et al. have demonstrated that the use of four repetitions of the COP–COM variable provides a higher reliability coefficient (0.94 in the A/P direction) measurement of postural stability in healthy elderly [19] compared with the three measurements used per subject per condition in this study. Moreover, our biomechanical linear model is only valid within close proximity of the upright equilibrium position. It does not consider the knees and arms, or motion in the coronal or transverse planes. The model also neglects the potential coupling between the ankle and hip feedback loops. These assumptions were found suitable for the purposes of this study; however, they may need to be revisited when studying more severe or multidirectional perturbations.

It remains to be investigated why ankle stiffness would increase in pregnant nonfallers, but not in pregnant fallers. If ankle stiffness is indeed the key differentiator between pregnant fallers and nonfallers, our results suggest that increasing ankle stiffness could be an important factor for fall prevention strategies targeted at pregnant women. Ankle stiffness could potentially be modified through targeted exercises or the use of braces, supports, or biofeedback. Given that the absolute peak COP–COG values were capable of distinguishing between pregnant fallers and nonfallers, we suggest further exploration of this metric for its potential predictive ability to identify the women at risk of experiencing loss of balance throughout gestation. Further research is necessary to determine if the differences in the peak COP–COG and ankle stiffness are due to altered muscle activation or increased co-contraction in response to the perturbation.

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Conflict of interest statement

We certify that none of the authors have a financial or personal relationship with other people or organizations that could inappropriately influence (bias) this work. The authors have no conflict of interest to disclose.

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