A pelvic-oriented margin of stability is robust against deviations in walking direction

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Abstract

The Margin of Stability (MOS) is often assessed relative to the intended, linear path of walking progression. When an unanticipated or irregular change in direction occurs, such as during a sudden turn or during activities of daily living, distinguishing the lateral from anteroposterior MOS can be challenging. The purpose of this study was to assess an anatomically orientated method of calculating the MOS using the pelvic orientation to define lateral and anteroposterior directions. We hypothesized that when straight walking was disrupted with a curved path, the pelvic-oriented MOS measure would be less variable compared to the global-oriented MOS. We recruited 16 unimpaired participants to walk at preferred and fast walking speeds along a straight walking path, as well as a path with an exaggerated, curvilinear deviation. We determined the within-subject mean and standard deviation of the anterior MOS at mid-swing and the lateral MOS at ipsilateral foot strike. For straight walking and curved walking separately, repeated measures factorial ANOVAs assessed the effects of model (global or pelvic-oriented), limb (left or right), and speed (preferred or fast) on these MOS values. Based on reduced variability during curved walking, the pelvic-oriented MOS was more robust to walking deviations than the globally defined MOS. In straight walking, the pelvic-oriented MOS was characterized by less lateral and more anterior stability with differences exacerbated by faster walking. These results suggest a pelvic-oriented MOS has utility when the path of progression is unknown or unclear.

Keywords: gait, stability, dynamic balance, stability margins, biomechanics

1 Introduction

The margin of stability (MOS) is a spatial measure that quantifies dynamic mechanical stability from the position and velocity of the whole-body center of mass (COM) relative to the edge of the base of support (BOS) (Hof et al., 2005). This measure has been used to quantify stability margins of individuals with compromised balance, including older adults (Roeles et al., 2018; Süptitz et al., 2013), children with cerebral palsy (Tracy et al., 2019), stroke survivors (Hak et al., 2013), individuals with multiple sclerosis (Peebles et al., 2016), and individuals with Parkinson's Disease (Stegemöller et al., 2012). In all of these cases, the MOS during walking was defined in anteroposterior or lateral directions relative to a prescribed, straight walking direction.

In adult studies, it can be reasonably assumed that participants will adhere to an instructed, straight walking path unless lateral stability is compromised. However, we observed that children (5-12 years) would occasionally deviate from a designated straight path, not necessarily due to instability, but possibly the result of distraction or apprehension about walking near a force plate (Tracy et al., 2019). Such deviations, as well as the corrected trajectory to the target endpoint, alter the lateral MOS relative to the prescribed path. This observation revealed a limitation in defining stability relative to an instructed walking direction, because laterally stable gait could be misclassified as being unstable due to incongruent instructed and performed walking directions. In our published study (Tracy et al., 2019), we ignored trials where a deviation was apparent, but we sought a method of calculating the MOS that was robust against slight, less-perceivable deviations.

An alternative approach to defining anteroposterior and lateral MOS relative to prescribed trajectories is to do so relative to anatomy. An anatomical orientation of stability has relevance when considering balance control, as the orientation of the body influences control strategies and the consequences of a loss of stability (Winter, 1995). For example, during quiet stance with the lower extremities in the anatomical position, stability in the sagittal plane is primarily controlled at the ankle, while the hip muscles play a more prominent role in controlling frontal plane stability (Winter, 1995). Given different control strategies, it is reasonable to assess anteroposterior and lateral stability separately, relative to the orientation of the lower body—specifically the pelvis.

We suggest that calculating the anteroposterior and lateral MOS relative to pelvic orientation would account for these differences and constraints in stability control, as well as the implications on injury risk. Such an approach would not necessarily agree with that of the MOS calculated relative to a path of progression. For example, steady state gait is characterized by long-axis rotation of the pelvis (Lewis et al., 2017) leading to slightly different orientations of the pelvis and the progression path. An alternative example is an individual stepping laterally without reorienting their body to the path of progression (i.e., a "side-shuffle"). In this second case, the MOS in the direction of travel is lateral in the pelvic orientation.

An anatomically-oriented approach to characterizing stability margins would also be applicable when the path of progression is not prescribed, known *a priori*, or consistent across trials (e.g., free play of children, tai chi, complex athletic maneuvers). Compared to the common approach of determining the MOS relative to a prescribed path (often the "global" reference frame), a MOS relative to pelvic orientation would have the advantages of (1) quantifying stability relative to lower-limb anatomy and anisotropic stability control strategies and (2) not being reliant on a known, non-deviated path of progression. The purpose of this study was to assess an anatomical orientation method of calculating the MOS by using the pelvic orientation to define lateral and anteroposterior directions. We hypothesized that the pelvic-oriented MOS measure (MOS_{pelvis}) would be less variable compared to the global-oriented MOS (MOS_{global}) when the walking path has a curved deviation. We also compared the two measures during the commonly assessed task of straight walking. Given long-axis pelvic rotation during gait, we anticipated less lateral stability using the pelvic-oriented stability margin with between-method differences exacerbated by faster walking.

2. Methods

2.1. Participants

Sixteen healthy young adults (mean (standard deviation); 7M/9F; age = 21.4 (3.6) years; BMI = 21.2 (2.5) kg/m^2) were recruited for this study. Participants were free of any self-reported neuromuscular or skeletal injuries that would affect gait or balance, and they had not experienced a lower extremity fracture or surgery within the year before participation. This study was approved by the University of Delaware institutional review board, and all participants provided written, informed consent before participation.

2.2. Procedure

Participants were recruited to walk along two paths: (1) a straight nine-meter walking path (straight) and (2) a nine-meter walking path with an exaggerated curvilinear deviation with a radius of 1.5 meters (curved) beginning at the three-meter mark (Figure 1). For the curved walking condition, steps prior to the initiation of the turn and steps after the termination were excluded from the analysis. The total path length analyzed for the curved condition was defined as 1.5π meters. For preferred and fast speeds, subjects were instructed to “walk at a usual comfortable speed” and “walk as fast as possible without running”, respectively. Each combination of speed and path consisted of 25 trials, for a total of 100 trials. Subjects alternated between straight and curved conditions within each walking speed and completed the curved condition in the same walking direction for each trial. Participants completed all preferred walking speed trials before the fast-walking speed trials. Walking speed was calculated as path length divided by time to reach the end point.

Forty-one reflective markers were placed throughout the entire body to calculate the whole-body COM from 13 body segments: head, trunk, pelvis, upper arms, forearms, thighs, shanks, and feet (Kadaba et al., 1990). Trials were recorded with a 12-camera motion capture system recording at 120 Hz (Qualysis®, Göteborg, Sweden). Marker data were filtered via a fourth-order low-pass Butterworth filter with a 6-Hz cutoff frequency.

2.3. Anatomically-Oriented Reference of the Margin of Stability

The pelvic coordinate system was determined from the anterior superior iliac spine (ASIS). The mediolateral axis was defined by the vector from the left ASIS to the right ASIS, the vertical axis was defined by a vector aligned with gravity, and the anteroposterior axis was defined by the cross product between the mediolateral and vertical axes (Figure 2). For visualization purposes, the origin of the pelvic coordinate system was located on a sacral shell placed along the center of the spine and in the same horizontal plane as the ASIS markers. Prior to defining the coordinate system, the ASIS and sacral markers were projected onto the floor to remove pelvic tilt and to maintain gravity in the vertical direction.

The MOS was defined by the distance between the extrapolated center of mass (xCOM) and the BOS (Equation 1) (Hof et al., 2005).

$$xCOM = COM + \frac{vCOM}{\sqrt{g/l}}, \tag{1}$$

where $vCOM$ is the COM velocity, g is the acceleration due to gravity (9.81 m/s^2), and l is the pendulum length defined as the distance from the ankle joint center to the COM.

The edge of the BOS is dependent on the direction of the MOS of interest. Laterally, the position of the heel is used, and anteroposteriorly, the position is the 2nd metatarsal. The MOS_{global} was calculated by separating the horizontal vector from the BOS to the xCOM into a lateral component and an anterior component, as defined by the global coordinate system. Prior to determining the MOS_{pelvis}, a rotation matrix defined by the pelvis coordinate system was applied to the xCOM and BOS positions. As defined by this coordinate system, the lateral and anterior components of the vector between these two points was the MOS_{pelvis}. Regardless of the coordinate system, a positive MOS indicated that the xCOM is within the edge of the BOS and a negative MOS indicated that the xCOM is outside the edge of the BOS.

2.4. Data Analysis

Lateral and anteroposterior MOS_{global} and MOS_{pelvis} values were determined throughout the gait cycle. Our analysis focused on lateral MOS values at ipsilateral foot strike and anterior MOS values at contralateral mid-swing. For all curved trials, the left limb was the outer limb and the right limb was the inner limb. Variability of the MOS was measured using standard deviations of MOS values within a given condition.

Differences in curved-walking MOS variability across global and pelvis-oriented coordinate systems (MODEL), left and right limbs (LIMB), and preferred and fast walking speeds (SPEED) were assessed with a full-factorial, repeated-measures ANOVA. Post-hoc comparisons were adjusted with a Sidak correction. An identical analysis was performed for mean MOS values during straight walking. A repeated-measures t-test was used to determine differences in walking speed. All statistical analyses were performed using SPSS software (v25, IBM, Armonk, NY).

3. Results

3.1. Walking Speeds

Based on repeated-measures t-tests, we observed significant differences between preferred and fast walking speeds for both straight walking ($p < 0.001$, mean (SD) preferred: 1.42 (0.21) m/s, fast: 2.09 (0.27) m/s) and curved walking ($p < 0.001$, preferred: 1.55 (0.27) m/s, fast: 2.36 (0.37) m/s).

3.2 Lateral margin of stability variability during curved walking

We observed a significant interaction between MODEL and SPEED for the standard deviation of lateral stability assessed at ipsilateral foot strike ($p < 0.001$, $\eta_p^2 = 0.633$). The pelvic-oriented model reduced variability at both speeds, with larger reductions in variability at fast walking speeds ($p < 0.001$, mean difference (SE) = 6.69 (0.85) cm.), compared to those at preferred walking speeds ($p < 0.001$, 3.43 (0.63) cm.) (Figures 3A & 3B).

3.3 Anterior margin of stability variability during curved walking

We observed a significant interaction of MODEL and LIMB when assessing anterior MOS variability at mid-swing ($p < 0.001$, $\eta_p{}^2 = 0.664$). The pelvic-oriented model reduced variability for each limb with a larger reduction in variability for the outer limb ($p < 0.001$, mean difference (SE) = 4.95 (0.53) cm.), compared to the inner limb ($p < 0.001$, 2.39 (0.55) cm.) (Figures 3C & 3D).

3.4 Mean lateral margin of stability during straight walking

We observed a significant interaction of MODEL and SPEED for mean lateral MOS values assessed at ipsilateral foot strike ($p < 0.001$, $\eta_p{}^2 = 0.629$). the pelvic-oriented model resulted in less stability for each speed with greater reductions at fast walking compared to slow speeds. Post-hoc analysis revealed further reduced stability when walking at fast walking speeds ($p = 0.001$, mean difference (SE) = 4.37 (0.76) cm.) compared to reductions at preferred walking speeds ($p < 0.001$, 1.14 (2.19) cm.) (Figures 4A & 4B).

3.5 Mean anterior margin of stability during straight walking

We observed significant effects of MODEL ($p < 0.001$, $\eta_p{}^2 = 0.677$) and SPEED ($p < 0.001$, $\eta_p{}^2 = 0.888$) for anterior stability values assessed at mid-swing. The pelvic-oriented model resulted in less negative values of the MOS (i.e., a less unstable condition; $p < 0.01$, mean difference (SE) = 0.13 (0.02) cm.). Walking at a fast speed resulted in more negative MOS values (i.e., a more unstable condition; $p < 0.01$,19.86 (1.82) cm; Figure 4C & 4D).

4 Discussion

This study evaluated an anatomical orientation method of calculating the MOS by using the pelvic orientation to define lateral and anteroposterior directions. We hypothesized that, with an exaggerated curvilinear deviation, the pelvic-oriented model would reduce MOS variability. Our hypothesis was supported by a significant decrease in MOS variability when using the pelvic-oriented model regardless of walking speed, reference limb, or MOS direction. These results validate our pelvic-oriented MOS measure as one that is robust against deviations in walking direction.

For walking along a straight prescribed path, we observed a decrease in stability in the lateral direction and a small increase in stability in the anterior direction when using the pelvic-oriented MOS. Long-axis rotation of the pelvis is a characteristic of walking and is proportional to walking speed (Crosbie and Vachalathiti, 1997). With such pelvic rotation, the COM forward velocity vector “bleeds” into the lateral stability calculation using the pelvic-oriented method. The magnitude of the changes we observed also reflect this concept. We observed very small changes in anterior stability at mid-swing, a time point in which the pelvis is in a similar orientation to the path of progression, while there were larger changes in lateral stability at foot strike, a time point in the gait cycle characterized by a rotated pelvis (Crosbie and Vachalathiti, 1997). Additionally, at faster walking speeds, changes in lateral MOS were exacerbated. While the influence of COM forward velocity on lateral stability may be considered erroneous when considering only unperturbed gait, our approach may reflect an increased risk of hip impact following a loss of balance while the pelvis is rotated as lateral stability is decreased (Yang et al., 2020, 2016). Other studies defined stability relative to the path of progression defined by the stepping vector (Conradsson et al., 2018a, 2018b; He et al., 2018), the velocity of the center of mass (Fino et al., 2020; Mellone et al., 2016), or an unspecified method (Havens et al., 2018; Mehdizadeh et al., 2020). In these methods, the MOS was defined as vectors parallel or orthogonal to the path of progression, effectively addressing situations where the prescribed walking direction is unclear or changing (Huxham et al., 2006). We anticipate that these path-of-progression-based MOS definitions would also show reduced variability compared to the global-oriented MOS for our curved walking condition. A limitation of this approach is that variability in motor control could influence both the defined progression path *and* stability control about that path. It may be difficult to determine if group differences arise because the intended path, accuracy in following the intended path, or the control of stability about the intended path is different. Our anatomically-oriented approach is not dependent upon a defined progression path. Instead, it may be driven by differences in pelvic orientation or long-axis rotation.

Because the MOS is proportional to the impulse needed to change stability states (Hof et al., 2005), it is indicative of the size of the perturbation needed to become unstable (when the MOS is positive) or, alternatively, the size of the corrective response needed to regain stability (when the MOS is negative). We suggest that the response to a perturbing impulse or loss of stability is constrained by anatomy, as the moment-producing capabilities and joint range of motions of the lower extremity are greatly affected by the plane of motion (Winter, 1995). Accordingly, the capabilities in responding to a perturbation are different between anatomical frontal and sagittal planes (Crenshaw and Kaufman, 2014). This constraint further justifies an anatomical reference to calculating the MOS. Indeed, if someone is laterally unstable (i.e., a negative MOS), they can reduce that lateral instability by rotating the pelvis, so that it is defined as an anterior instability—a direction in which they are better suited to execute recovery steps. Furthermore, a fall to the side and/or impact to the lateral aspect of the pelvis elevates the risk of a resulting hip fracture (Hayes et al., 1993). This direction-specific influence of injury risk also warrants consideration of stability relative to pelvic orientation.

A potential limitation of our study may be the narrow scope of time points we selected to determine the MOS. Initially, we planned to assess MOS at mid-swing, foot strike, and the greatest point of lateral instability. Each of these events represent a point in which stability is likely to be lost. However, when analyzing the minimum lateral MOS, we encountered model-specific differences in timing within the gait cycle. Therefore, we decided not to compare MOS values between methods as these values no longer represented comparable points of the gait cycle. Another potential limitation of our pelvic-oriented MOS is the reliance on secure and proper marker placement for the markers defining the pelvic coordinate system. While all MOS values are dependent on proper marker placement, any shift in the markers that define the pelvis can greatly influence the pelvic-oriented MOS. In Figures 3 and 4, we noted a subject with abnormal stability values that may have come from deviations in marker placement. During data analysis, we noted downward trending MOS values for this participant over time. We reviewed the pelvic rotation angle across trials and determined that the pelvic angle gradually shifted towards the right side. This was likely due to markers defining our pelvic coordinate system slowly displacing or becoming loose without being corrected. This artificial pelvic rotation greatly influenced our MOS values as the anterior COM velocity had a greater “bleed” into the lateral values. While odd, these data points were not excluded from our analysis as they were not objectively defined as an outlier (MOS values and standard deviations were within a 95% Confidence Interval of all subjects). Interestingly, excluding this participant from the analysis did not change our conclusions which may suggest at least some resilience of the pelvic-oriented method to marker placement.

5 Conclusion

Based on our findings, calculating the margin stability relative to pelvic orientation is a strong alternative approach to using the instructed path of progression. A pelvic-oriented margin of stability is best suited for use when the path of progression is unclear or is not reflective of the body's orientation. For gait with a clear path of progression, such as straight walking, a pelvic-oriented margin of stability may underestimate stability compared to the traditional margin of stability.

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Conflict of Interest Statement

The authors have no conflict of interest to disclose.

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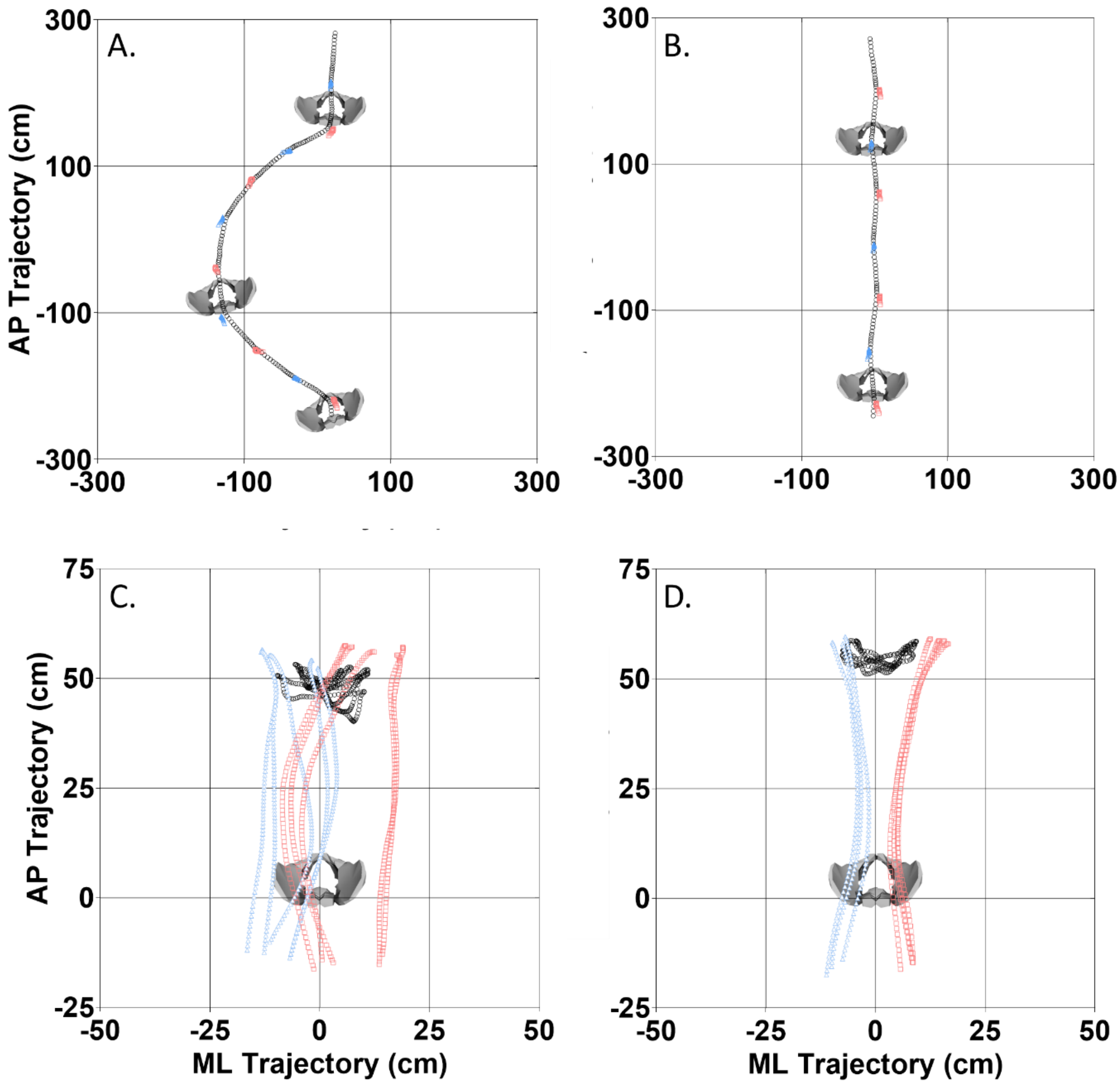

Figure 1. A top-down view of the MOS components during stance for curved walking (A & C) and straight walking (B & D). Trajectories are displayed in the global coordinate system (A & B) and the pelvic coordinate system (C & D). Left toe marker positions (blue triangle), right toe marker positions (red square), and xCOM positions (black circles) are shown. For curved walking, steps leading to the initiation of the turn and after the termination of the turn were excluded from analysis.

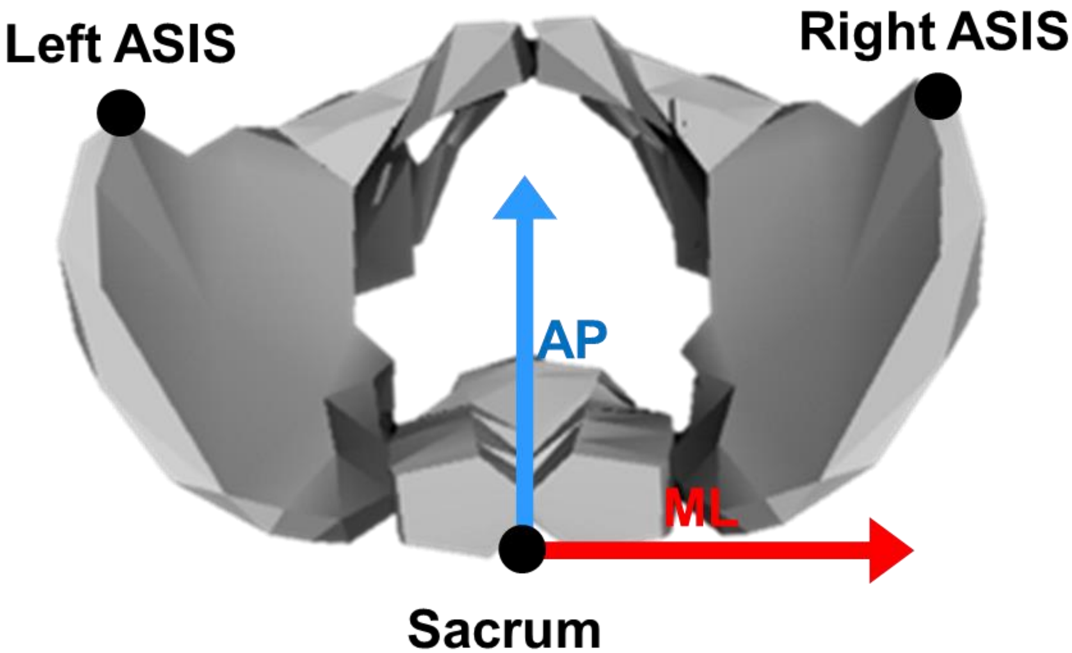

Figure 2. A top-down view of the pelvis coordinate system. The mediolateral axis (red) is defined as the vector from the left ASIS to the right ASIS, the vertical axis is defined as a vector aligned with gravity, and the anteroposterior axis (blue) is defined as the cross product of these two vectors. The origin of this coordinate system is the sacrum.

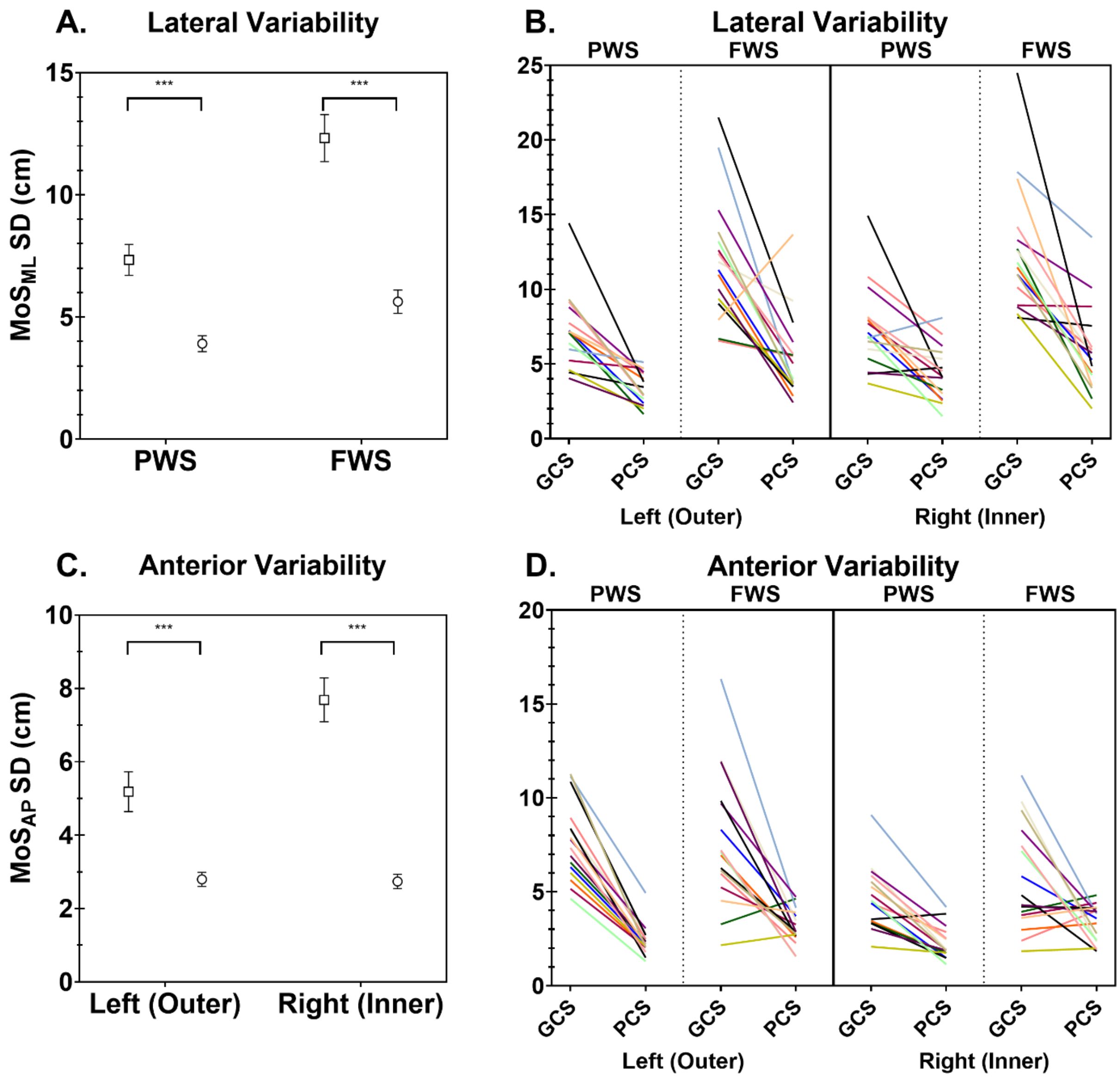

Figure 3. Lateral MOS standard deviations at ipsilateral heelstrike (A & B) and anterior MOS standard deviations at mid-swing (B & C). Panels A and C show summary statistics (marginal mean ± one standard deviation) along with significant interactions ($p < 0.001$). Panels B and D show individual participant data segmented into the lowest level interaction. Note that Y-Axis scales are different across plots. The subject with abnormal variability is represented by the black line (see discussion).

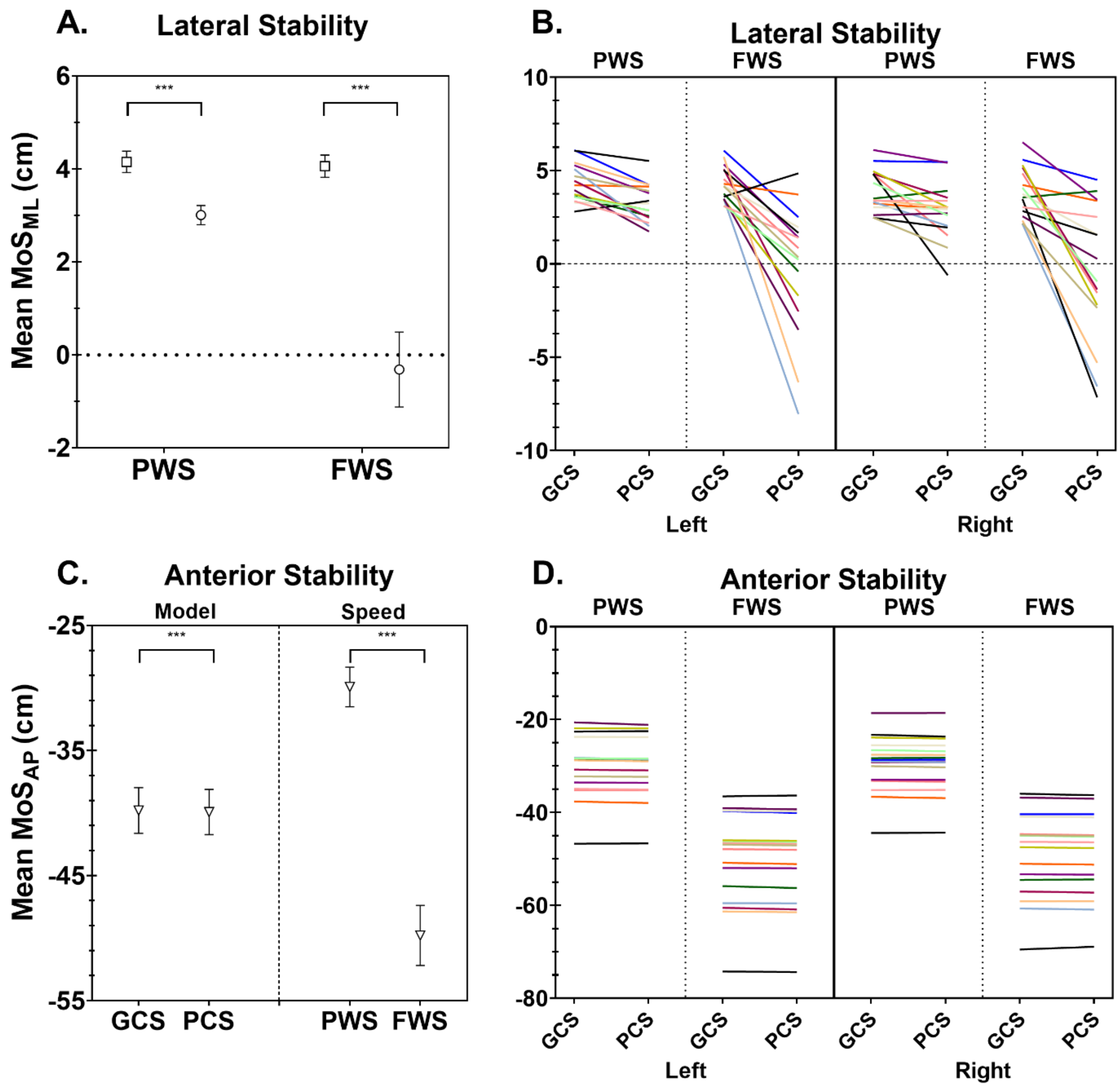

Figure 4. Mean lateral MOS at ipsilateral heelstrike (A & B) and mean anterior MOS at mid-swing (B & C) during straight-line walking. Individual subject data segmented into the lowest level interaction (B & D) and summary statistics (marginal mean ± one standard deviation) along with significant interactions ($p < 0.001$) (A & C) are shown above. Note that Y-Axis scales are different across plots. The subject showing abnormal stability values is represented by the black line (see discussion).