

1 **Harmonization of Margin of Stability Calculations and Investigation of the Impact of**
2 **Foot Length, Foot Width, Gait Speed, and Body Mass**

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18 **Conflict of interest statement:**

19 No conflict of interest to declare

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21 **Abstract**

22 The margin of stability (MoS), the minimum distance between the extrapolated center of mass
23 and the edges of the base of support (BoS), is one of the most widely used metric to describe
24 the mechanical stability during gait. In the current literature, the markers used to define the
25 edges of the BoS are variable and the MoS model neglects the influence of anthropometric
26 factors, such as foot length, foot width, and body mass. This study aimed to evaluate differences
27 between anteroposterior (AP) and mediolateral (ML) MoS measures using various BoS edge
28 definitions (AP: n = 3 methods, ML: n = 4 methods) and to investigate the impact of foot length,
29 foot width, gait speed, and body mass on the MoS measures. Results show that the BoS edges
30 definition affects the resulting MoS across the entire stance phase (AP: p<0.001 between the 3
31 methods; ML: p<0.001 between the 4 methods). Moreover, the AP MoS is influenced by foot
32 length (p<0.029), as well as gait speed and body mass on both the AP (gait speed: p<0.001;
33 body mass: p<0.038) and ML (gait speed: p<0.032; body mass: p<0.001) MoS. This study
34 proposes a new approach based on optimal foot markers for defining the edges of the BoS,
35 which may contribute to better assess mechanical stability during gait. Finally, the results
36 suggest that normalizing the MoS (i.e., the AP MoS by foot length, gait speed, and body mass,
37 and the ML MoS by gait speed and body mass) can facilitate comparisons between populations.

38

39 1. Introduction

40 Various methods have been employed to assess stability during human locomotion (Brujin et
41 al., 2013). These methods span from ordinal scale clinical assessments (e.g., the Berg Balance
42 Scale (Miranda-Cantellops and Tiu, 2024) and the Functional Gait Assessment (Leddy et al.,
43 2011)) to biomechanical measures derived from simple mechanical system (e.g., the margin of
44 stability (MoS) (Devetak et al., 2019; Hof et al., 2005)). The MoS is one of the most widely
45 used metric to describe the instant mechanical stability of the body configuration during
46 pathological (Watson et al., 2021) and non-pathological gait (Ohtsu et al., 2019), and is related
47 to the minimal impulse needed for destabilizing the walking person (Curtze et al., 2024). The
48 MoS represents the minimum distance between the extrapolated center of mass (xCoM) and
49 the edges of the base of support (BoS) (*Eq. 1*), and can be calculated in the antero-posterior
50 (AP) and the medio-lateral (ML) directions of the stance phase of walking.

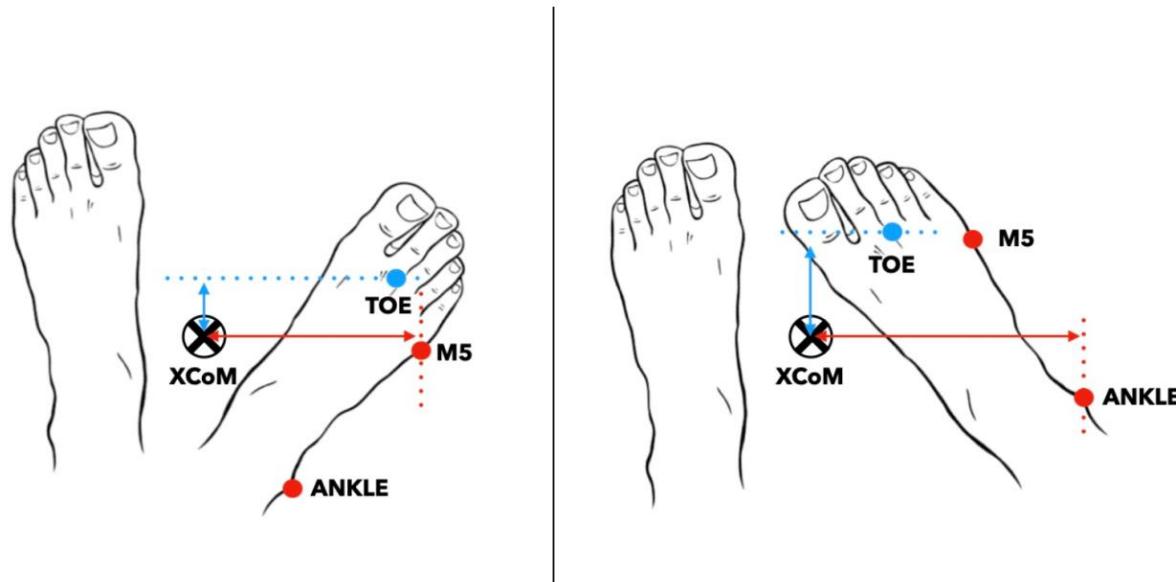
$$51 \quad Eq. \ 1 \qquad MoS = BoS - xCoM$$

52 The xCOM was introduced by Hof and colleagues in 2005, and is based on the traditional
 53 linearized inverted pendulum model (Hof et al., 2005). This latter considers that the CoM
 54 location is continually changing according to even small body position changes by combining
 55 the CoM position and its velocity (\dot{COM}) divided by the pendulum's eigen frequency, i.e., the
 56 square root of gravity (9.81 m/s^2) divided by the pendulum length (l) (Eq. 2).

$$57 \quad Eq.~2 \quad xCoM = COM + \frac{COM}{\sqrt{\frac{g}{l}}}$$

The edges of the BoS can be described using the limits of the center of pressure (CoP) (Brandt et al., 2019; Day et al., 2007; De Jong et al., 2020; Hof et al., 2007; Major et al., 2013; Van Meulen et al., 2016; Vistamehr et al., 2016) or foot markers (Beltran et al., 2014; Gates et al., 2013; Hak et al., 2015, 2014, 2013b, 2013c; Kao et al., 2014; Major et al., 2019; Martelli et al., 2017; Peebles et al., 2017, 2016; Rijken et al., 2015; Simon et al., 2017; Tisserand et al., 2018). Using foot markers, previous studies have described the lateral edge of the BoS with the fifth metatarsal marker (Beltran et al., 2014; Gates et al., 2013; Kao et al., 2014; Major et al., 2019; Martelli et al., 2017; Peebles et al., 2017, 2016), the lateral malleolar marker (Hak et al., 2015, 2014, 2013b, 2013c; Rijken et al., 2015), or the mid-point between the fifth metatarsal and the lateral malleolar markers (Simon et al., 2017; Tisserand et al., 2018). Although the definition of the AP edges seem to be more standardized, based on the toe (anterior edge) (Carty et al.,

69 2011; Hak et al., 2013c; Kao et al., 2014; Ma et al., 2021; McAndrew Young et al., 2012;
70 McAndrew Young and Dingwell, 2012; McCrum et al., 2014; Peebles et al., 2017, 2016;
71 Sangeux et al., 2024; Sivakumaran et al., 2018; Tracy et al., 2019) or heel landmark (posterior
72 edge) (Arora et al., 2019; Hak et al., 2015, 2013a; Herssens et al., 2020; Rijken et al., 2015;
73 Wang et al., 2024; Yamaguchi and Masani, 2022), the heterogeneous methodologies relating
74 the definition of the lateral edge of the BoS is a key factor making interpretation and
75 comparison of MoS results challenging (Watson et al., 2021). Uncertainty persists regarding
76 the optimal marker to define the edges of the BoS at different instants of the stance phase
77 (Curtze et al., 2024). For the ML BoS, it can be suggested that using the midpoint on the virtual
78 line relating the fifth metatarsal and the lateral malleolar markers would have the advantage,
79 compared to using the lateral malleolar or the fifth metatarsal marker, of considering the
80 orientation of the foot (Tisserand et al., 2018, 2016), especially in individuals with consequent
81 internal or external foot rotation (example in **figure 1**). Regardless of the approach chosen, the
82 marker representing the BoS edge is often placed on a foot edge that is not always in contact
83 with the ground (e.g., the lateral malleolar marker during late stance), which limits its
84 effectiveness because only a body part in contact with the ground can contribute to quickly
85 move the CoP. To overcome this problem, the marker defining the edge of the BoS could be
86 chosen according to two conditions: **1)** the marker should be the most anterior (for the AP edge
87 of the BoS) or the most lateral (for the ML edge of the BoS), and **2)** be fixed on a physical edge
88 of the foot that is in contact with the ground at the instant when the MoS is calculated. To date,
89 no study has investigated the effect of employing these various approaches of defining the
90 edges of the BoS, which could potentially induce significant differences in the MoS measure,
91 particularly for the ML MoS.



92
93 **Figure 1.** Representation of the most lateral marker during foot rotation between the fifth
94 metatarsal (M5) and lateral malleolar (ANKLE) markers. The medio-lateral margin of stability
95 is the difference between the extrapolated center of mass (xCOM) and the most lateral marker.
96 The antero-posterior margin of stability is the difference between xCOM and the most anterior
97 marker, which is always the second metatarsal head marker (TOE). The medio-lateral margin
98 of stability is the difference between the extrapolated center of mass (xCOM) and the most
99 lateral marker.

100
101 In addition to being calculated heterogeneously (mostly in the ML direction), the MoS is
102 typically computed at a single time point, specifically at the initial contact of the ipsilateral foot
103 (Watson et al., 2021). Calculating the MoS at this point is crucial as it reflects the mechanical
104 effects of the contralateral stance phase. However, it has been observed that the minimal MoS
105 is achieved before the contralateral toe-off (Hof, 2007; Hof et al., 2005), and later suggested to
106 be the instant when the MoS should be measured (Curtze et al., 2024).

107 The linear model used to calculate the MoS includes few limitations, which arise from
108 simplifying assumptions made to facilitate biomechanical analysis such as the unchanging
109 pendulum length, the non-deformable pendulum, and the non-consideration of potential
110 additional external forces. Since the MoS accounts for the CoM velocity in its calculation, the
111 AP MoS is strongly influenced by gait speed, with more negative AP MoS when gait speed
112 increases (Curtze et al., 2024; McCrum et al., 2019). Indeed, the AP MoS remains negative
113 during the stance phase of a steady-speed gait because the BoS is consistently positioned
114 posterior to the xCoM (Curtze et al., 2024; Hof, 2008). Mainly, the body is always "falling
115 forward," and stability is maintained by continuous repositioning of the BoS, as the foot
116 contacts the ground in the next step (Kuo and Donelan, 2010). However, the influence of gait

117 speed on ML MoS is less understood and has been investigated by fewer studies (Gates et al.,
118 2013; Guaitolini et al., 2019; Peebles et al., 2016). Moreover, the MoS model neglects the
119 influence of anthropometric factors, such as foot length, foot width, and body mass. Since the
120 MoS is a widely used clinical measure of gait stability and is often compared across populations
121 with different anthropometric characteristics and gait speed, the effect of these factors should
122 be investigated to better personalize/individualize clinical assessments.

123 This study aimed to 1) assess differences between AP and ML MoS measures resulting from
124 the different approaches of defining the edges of the BoS that have been previously used in the
125 literature and the one proposed in this current study, and 2) investigate the effect of foot length,
126 foot width, gait speed, and body mass on the MoS measures. It was hypothesized that 1) the
127 different approaches of defining the BoS used will lead to different AP and ML MoS values,
128 and that 2) AP MoS measure will be negatively related to foot length, gait speed, and body
129 mass, and ML MoS will be positively related to foot width, gait speed, and body mass.

130 **1. Method**

131 *1.1. Participants*

132 This study used an open access dataset from Riglet et al. (2024)' study, including 30 healthy
133 participants (16M/14F) aged between 21 and 41 years old (27.97 ± 5.59 years old) (Riglet et
134 al., 2024). Participants were on average 1.73 ± 0.92 m tall and weighed 68.16 ± 11.06 kg.

135 *1.2. Procedure*

136 Records

137 Each participant was instructed to walk at three different speeds along 10 meters of a flat and
138 regular laboratory surface: slow, comfortable, and fast with walking shoes provided for the
139 experiment (Riglet et al., 2024) (**supplementary figure 1**). All participants were equipped with
140 a set of 63 reflective markers following the Conventional Gait Model (pyCGM, v.2.5) (Baker
141 et al., 2018). Eighteen optoelectronic cameras (Vicon System®, Oxford, UK; 100 Hz sampling
142 rate) and Nexus software were used to collect the marker trajectories.

143 Data processing

144 The c3d files that include the labelling of the marker trajectories were then exported from the
145 Riglet et al. (2024) (Riglet et al., 2024) database and further processed in MATLAB (vR2022b,

146 Mathworks Inc., Natick, USA) using the open-source biomechZoo toolbox (v.1.9.10) (Dixon
147 et al., 2017) and custom codes. The foot strikes and foot-offs were identified using the method
148 of Zeni et al. (2008), based on the positional changes of the heel, foot, and sacrum markers
149 (Zeni et al., 2008). Then, the walking trials were partitioned into individual gait cycles.
150 Considering the natural asymmetry in able-bodied gait (Sadeghi et al., 2000), gait cycles of
151 both legs were included. For each participant, the first 12 gait cycles of each walking speed
152 condition were used for the MoS calculation (i.e., 6 gait cycles on each leg).

153 Calculations

154 For each gait cycle, the continuous MoS was calculated during the stance phase in the AP and
155 ML directions using **Eq.1** and **Eq. 2** (Hof et al., 2005). The stance phase started at the ipsilateral
156 foot contact and ended at the ipsilateral foot-off. The anterior direction of walking was
157 described as the vector of the walking direction in the transverse plane of the laboratory
158 whereas the lateral direction of walking was described as the vector perpendicular to the
159 anterior direction of walking. A table summarizing the different approaches used in the
160 previous literature to calculate the AP and ML MoS is provided in the appendix
161 (**supplementary table 1**).

162 The AP MoS was calculated at each time point of the stance phase following 3 approaches that
163 have been used in the previous literature (HEEL, TOE, MOST POSTERIOR):

- 164 1) **HEEL:** Using the heel marker (**HEEL**) as the anterior limit of the BoS.
- 165 2) **TOE:** Using the second metatarsal marker (**TOE**) as the anterior limit of the BoS.
- 166 3) **MOST ANTERIOR:** Using the most **anterior** marker between the **HEEL** and
167 **TOE** markers. The part of the foot to which the most anterior marker is attached
168 had to be in contact with the ground. For instance, at foot strike, the **HEEL** marker
169 was chosen if the forefoot (**TOE**) was elevated, whereas the **TOE** marker was
170 selected if the heel was elevated (i.e. during late stance).

171 The ML MoS was calculated at each time point of the stance phase following 4 approaches
172 (ANKLE, M5, MIDPOINT, MOST LATERAL):

- 173 1) **ANKLE:** Using the lateral malleolar marker (**ANKLE**) as the lateral limit of the
174 BoS.

- 175 2) **M5:** Using the fifth metatarsal marker (**M5**) as the lateral limit of the BoS. By
176 identifying the most lateral marker as the lateral limit of the BoS.
177 3) **MIDPOINT:** Using the midpoint of the virtual line relating the **ANKLE** and **M5**
178 markers as the lateral limit of the BoS.
179 4) **MOST LATERAL:** Using the most **lateral** marker between the **ANKLE** and **M5**
180 markers. The part of the foot to which the most lateral marker is attached had to be
181 in contact with the ground. For instance, at foot strike, the **ANKLE** marker was
182 chosen if the midfoot/forefoot (**M5**) was elevated, whereas the **M5** marker was
183 selected if the heel was elevated (i.e. during late stance).

184 All calculations were performed using MATLAB (vR2024a, Mathworks Inc., Natick, USA).
185 For each gait speed condition, calculation approach, and participant, the MoS curves were
186 averaged.

187 Gait speed was calculated as the distance covered by the head marker during a gait cycle
188 divided by the gait cycle duration in seconds. The foot length was provided within the Riglet
189 et al. (2024) database (Riglet et al., 2024), and the foot width was calculated as the distance
190 between the first and the fifth metatarsal markers.

191 *1.3. Statistical analysis*

192 All statistical analyses in this study were carried out using the Statistical Parametric Mapping
193 (SPM) toolbox (Pataky, 2010) and using custom-made Matlab scripts.

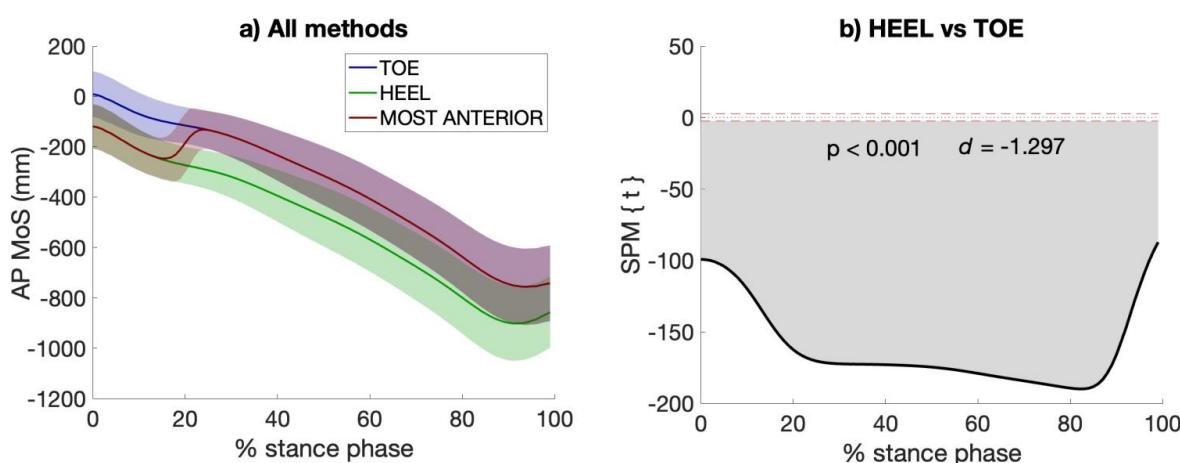
194 To test our first hypothesis, a paired t-test was conducted to assess differences between the of
195 the AP MoS calculation approaches (i.e., HEEL and TOE). To evaluate the differences between
196 the three ML calculation approaches (i.e., ANKLE, M5, and ANKLE-M5), a repeated-
197 measures Analysis of Variance (ANOVA) was performed. In the presence of significant results
198 for the latter, post-hoc analyses using paired t-test were carried out. The level of significance
199 was adjusted following a Bonferroni correction (n = 3) to account for multiple comparisons
200 (i.e. M5 vs MIDPOINT, M5 vs ANKLE, ANKLE vs MIDPOINT) (Altman, 1990). The mean
201 Cohen's d (d) effect size was calculated for each significant cluster (i.e., intervals during the
202 stance phase when the p-value is below the threshold) (Cohen, 1977). Only clusters lasting 5%
203 of the stance phase or more were discussed (Armijo-Olivo et al., 2011). Only the MoS curves
204 at comfortable speed were included (n = 30 curves).

205 To test our second hypothesis, a correlation analysis was conducted to assess the relation of
206 foot length, gait speed, and body mass with the continuous AP MoS curves using the SPM
207 toolbox (spm1d.stats.regress function, spm1d version M.0.4.11). The same analysis was
208 conducted to assess the relation of foot width, gait speed, and body mass with the continuous
209 the ML MoS curves. The AP MoS and the ML MoS were calculated using the MOST
210 ANTERIOR and the MOST LATERAL approaches, respectively. For all the significant
211 clusters, the mean coefficient of correlation (r) was calculated. For the gait speed correlation
212 analysis, the MoS curves of each condition (i.e., slow, comfortable, fast) were included ($n = 90$
213 curves). The relationship was interpreted as small ($r = 0 – 0.290$), moderate ($r = 0.300 – 0.490$),
214 strong ($r = 0.500 – 0.69$), and very strong ($r = 0.700 – 1.000$). Differences in gait speed for
215 each condition are presented in **supplementary figure 1**.

216 **2. Results**

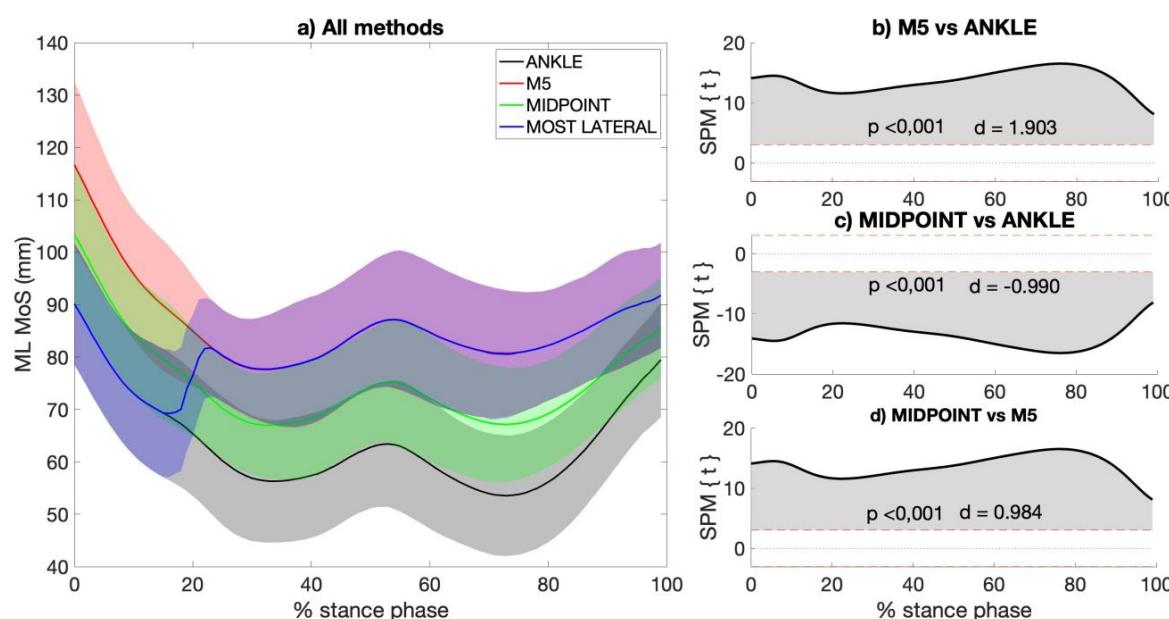
217 *2.1. Differences between the calculation approaches*

218 The calculation approach yielded to different results throughout the entire stance phase: HEEL
219 vs TOE (0 – 100%, $p < 0.001$, $d = -1.297$) (**figure 2a, 2b**). When using the MOST ANTERIOR
220 approach, the HEEL marker was used at the beginning of the stance phase (0 – 20%) after
221 which the TOE marker was selected (**figure 2a**).



222 **Figure 2.** Antero-posterior (AP) margin of stability (MoS) calculated using the two most
223 widely used approaches in the literature (i.e., TOE, HEEL), as well as the approach proposed
224 in this study (i.e., MOST ANTERIOR) to describe the anterior limit of the base of support (a).
225 A negative AP MoS refers to an extrapolated center of mass that is in front of the anterior limit
226 of the BoS. The differences between approaches are presented (statistical parametric mapping
227 (SPM) paired t-test, $p < 0.05$) (b). The red dashed lines indicate the critical thresholds for
228 statistical significance. Values above or under these lines indicate statistically significant
229 differences between the compared approaches at that specific point in the stance phase.
230

231 Concerning the ML MoS, the ANOVA showed a significant difference between all calculation
232 approaches ($p < 0.001$) over the entire stance phase (**supplementary figure 2**). Consequently,
233 paired t-tests were performed. Compared to each other, all calculation approaches lead to
234 different ML MoS values throughout the stance phase: ANKLE vs M5 (0 – 100%, $p < 0.001$,
235 $d = 1.701$), ANKLE vs MIDPOINT (0 – 100%, $p < 0.001$, $d = 0.887$), M5 vs MIDPOINT (0 –
236 100%, $p < 0.001$, $d = -0.866$) (**figure 3b, 2c, 2d**). When using the approach of the most lateral
237 marker, the ankle marker was chosen during the first 20% of stance phase. Else, the fifth
238 metatarsal marker was chosen as the most lateral marker (**figure 3a**).

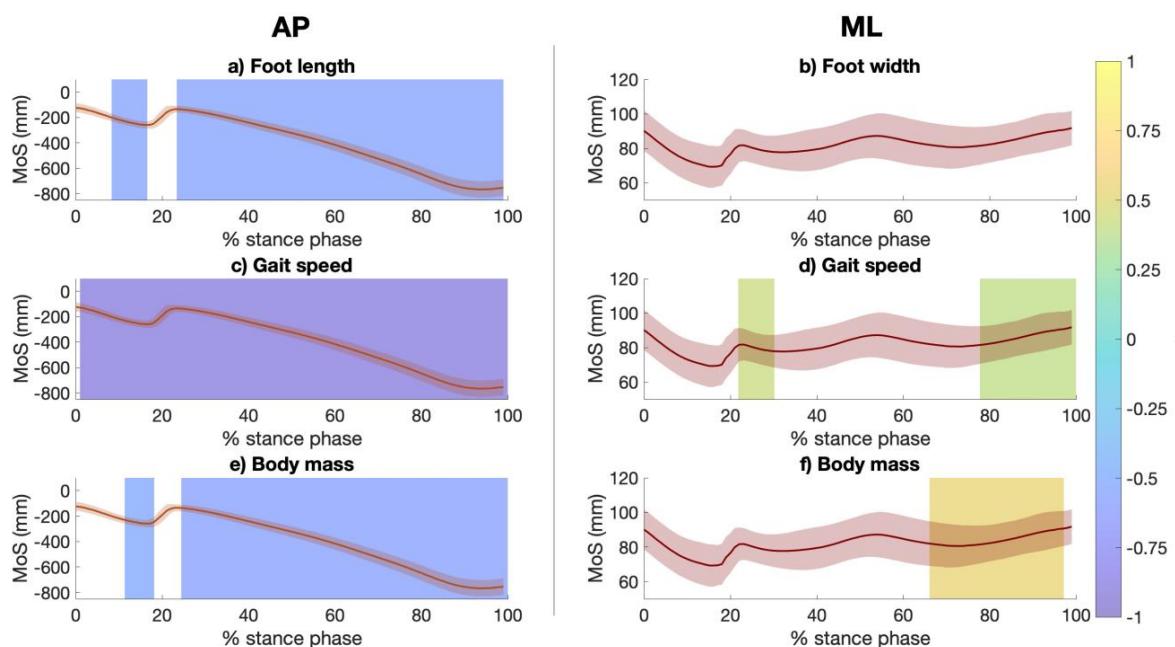


239
240 **Figure 3.** Mediolateral (ML) margin of stability (MoS) calculated using the three most widely
241 used approaches in the literature (i.e., ANKLE, M5, MIDPOINT), as well as the approach
242 proposed in this study, (i.e., MOST LATERAL) to describe the lateral limit of the base of
243 support (a). A negative ML MoS refers to an extrapolated center of mass that is more lateral
244 than the lateral limit of the BoS. The differences between approaches are presented (statistical
245 parametric mapping (SPM) paired t-test, $p < 0.05$) (b, c, d). The red dashed lines indicate the
246 critical thresholds for statistical significance. Values above or under these lines indicate
247 statistically significant differences between the compared approaches at that specific point in
248 the stance phase.

249 2.2. Factors influencing the margin of stability

250 The AP MoS was negatively correlated with foot length (8 – 18% of stance phase, $p = 0.029$,
251 $r = -0.559$; 23 – 100%, $p < 0.001$, $r = 0.640$), gait speed (0 – 100%, $p < 0.001$, $r = -0.905$), and
252 body mass (11 – 18%, $p = 0.038$, $r = -0.524$; 24 – 100%, $p < 0.001$, $r = -0.598$) (**figure 4a, 4c,**
253 **4e**). The ML MoS was positively correlated with gait speed (22 – 30%, $p = 0.033$, $r = 0.449$;
254 78 – 100%, $p = 0.003$, $r = 0.411$), body mass (66 – 97%, $p < 0.001$, $r = 0.563$), but not foot

255 width (**figure 4b, 4d, 4f**). The detailed results of the correlation analysis are presented in the
256 **supplementary figure 3 and 4**.



257
258 **Figure 4.** Correlation analysis results between foot length (a) or width (b), gait speed (c, d),
259 and body mass (e, f) and the margin of stability (MoS) in the anteroposterior (AP) (a, c, e) and
260 mediolateral (ML) (b, d, f) directions during the stance phase of the gait cycle. The AP MoS
261 was calculated using the MOST ANTERIOR approach whereas the ML MoS was calculated
262 using the MOST LATERAL approach. A negative AP MoS refers to an extrapolated center of
263 mass that is in front of the anterior limit of the BoS, whereas a negative ML MoS refers to an
264 extrapolated center of mass that is more lateral than the lateral limit of the BoS. The colored
265 patches represent regions where significant correlation were found (statistical parametric
266 mapping, $p < 0.05$), and are colored by its mean r value.

267

268 **3. Discussion**

269 *3.1. Summary*

270 The aims of the present study were to 1) assess differences between the approaches most widely
271 used in the literature for calculating AP and ML MoS and 2) investigate the effect of foot
272 length, foot width, gait speed, and body mass on the MoS measures. Two main findings
273 emerged. First, significant differences were observed between the MoS calculation
274 approaches (using the HEEL and TOE markers for AP MoS, and using the M5, ANKLE, and
275 MIDPOINT for ML MoS). Second, gait speed, foot length, and body mass are negatively
276 correlated with the AP MoS during almost the entire stance phase, whereas only gait speed and
277 body mass are positively correlated with the ML MoS and only during late stance.

278

3.2. The effect of the calculation approach

279 A key component for calculating the MoS is the BoS. The latter can be defined as the area
280 between the feet during walking outlined by the points of contact with the ground. Placing a
281 marker on specific points such as medial or lateral malleolus, M5, or calcaneus might position
282 the BoS edge too far toward the center, sides, front, or back. Consequently, this can result in
283 misestimations of the ML and AP MoS (Curtze et al., 2024). The results of the present study
284 showed that the marker chosen to describe the anterior or lateral BoS boundaries significantly
285 affects the resulting values of AP and ML MoS, respectively (**figure 2** and **figure 3**). Our
286 results complement those of Havens et al. (2018), who have reported that biases in the MoS
287 value can be introduced by the approach used to estimate the CoM dynamics (Havens et al.,
288 2018). The current study used the pelvis average model (i.e., average position of the four pelvis
289 markers) to minimize this bias, as suggested by the latter study (Havens et al., 2018) when
290 simplified models of CoM dynamics are used. Together, these results highlight that the MoS
291 can be significantly affected by the choice of calculation approach, both in terms of the
292 definition of the BoS and the method used to estimate the CoM kinematics. This underscores
293 the importance of avoiding comparisons between studies that use different calculation
294 approaches and the future adoption of a standardized approach across different studies.

295 The current literature reveals an important heterogeneity in MoS calculation, which makes
296 comparisons between studies and populations difficult (Watson et al., 2021). For example, in
297 children with cerebral palsy, some studies opted for the lateral malleolus to describe the lateral
298 boundary of the BoS (Delabastita et al., 2016; Ma et al., 2021; Rethwilm et al., 2021), while
299 other used the MIDPOINT approach (Sangeux et al., 2024). In populations with in-toeing or
300 out-toeing gait such as those with cerebral palsy (Cao et al., 2020; Rethlefsen, 2006), both
301 approaches may lead to misestimates the BoS (Puszczalowska-Lizis and Ciosek, 2017).
302 Similarly, using the M5 marker to describe the lateral boundary of the BoS may not be relevant
303 if the individual presents a rotation of the medial foot. Thus, these approaches to calculate the
304 MoS are less suited to pathological populations, especially when the MoS value is used to be
305 compared with their healthy peers. Our study reported a significant difference in the AP and
306 ML MoS values based on the choice of the markers used to define the BoS in non-pathological
307 gait. Finally, another potential limitation of the current literature is the calculation of the MoS
308 at a single point in the gait cycle, often at initial contact (Hak et al., 2013b; Rijken et al., 2015;
309 Sangeux et al., 2024), although it has been largely suggested focus on more relevant instant,

310 such as close to the contralateral toe-off (Curtze et al., 2024), which is when the MoS is minimal
311 and stability is mechanically critical (Hof, 2007; Hof et al., 2005).

312 Treating MoS as a continuous measure has been emphasized in the literature, such as
313 McAndrew Young et al. (2012) who highlighted the importance of studying the MoS
314 throughout the entire stance phase instead of an average value across the gait cycle, to better
315 highlight the instant when the stability is critical (McAndrew Young et al., 2012). Similarly,
316 Kazanski et al. (2022) proposed a step-to-step approach to solve the MoS averaging paradox
317 (Kazanski et al., 2022). These findings support our goal of clarifying MoS interpretation.

318 *3.3. The effect of foot length and width*

319 The findings of this study indicate that the MoS is more forward in individuals with longer feet
320 (**figure 4a**), which may reflect an adaptation to the individual's longer limb structure. This
321 correlation has been noted even before the foot achieves full contact (i.e., between 8 – 18 % of
322 stance phase), suggesting that foot morphology such as foot length may play a role in dynamic
323 stability even when the foot is not completely in contact with the ground. Indeed, with a longer
324 foot, the CoP may extend further from the ankle compared to individuals with smaller feet,
325 potentially allowing for greater ankle moment to help regulate the stability, and in cases of
326 significant instability, to decelerate. To our knowledge, this is the first study that have assessed
327 the effect of foot length on AP MoS during gait (**supplementary table 1**), which limits our
328 understanding on how a forward MoS may promotes a more stable gait in individuals with
329 longer feet. However, as the MoS is a measure of an individual gait stability and is commonly
330 compared between populations, normalizing this value by foot length should be considered to
331 ensure accurate comparisons.

332 Regarding the ML direction, it was expected that larger foot will lead to wider BoS, which
333 could have directly increased the ML MoS. However, no significant relation has been noted
334 between foot width and the MoS (**figure 4b**). During postural balance task (i.e., eyes open and
335 closed), Qiu et al. (2013) observed that wider foot width is related to greater balance
336 performance (measured by the Composite Equilibrium Score), especially in younger
337 individuals (Qiu and Xiong, 2013). Together, these results may suggest that the mechanics of
338 gait stability, in contrast to postural balance tasks, involve more complex factors beyond just
339 foot width especially in the ML direction (Bauby and Kuo, 2000; Kuo and Donelan, 2010).

340 *3.4. The effect of gait speed*

341 It has been also observed that individuals used a more forward MoS when gait speed is
342 increased, which was characterized by a very strong correlation coefficient ($r = -0.887$) (**figure**
343 **4c**). It is important to recognize that this strong correlation is likely influenced by the fact that
344 the calculation of xCoM incorporates a velocity component. Nevertheless, other previous
345 studies have also indicated that gait speed plays a key role in an individual's AP stability
346 (Guaitolini et al., 2019; Hak et al., 2013a; McCrum et al., 2019). Indeed, consistent with our
347 findings, Guaitolini et al. (2018) reported that an increase in the gait speed is related to a
348 forward MoS (i.e., the xCoM is more forward relative to the BoS) in healthy young adults
349 (Guaitolini et al., 2019). Thus, normalization of the MoS by gait speed should be considered to
350 reduce between-participant variability, which is an approach that has been advocated
351 previously to enhance the precision of stability assessments (McCrumb et al., 2019).

352 In the ML direction, an increased gait speed was moderately related to a larger MoS (i.e., the
353 xCoM is more medial than the lateral limit of the BoS), during mid-stance (22 – 30%, $r =$
354 0.449) and late stance (78 – 100%, $r = 0.411$) (**figure 4d**). Especially during late stance, this
355 result can be explained by the reduced ML oscillation amplitude of the CoM when walking
356 faster, as compared to slower walking speeds. This reduced oscillation would result in a lower
357 ML velocity, which in turn increases the ML MoS. However, this finding was not supported
358 by Lencioni et al. (2019), who reported a negative correlation ($r = -0.320$) between gait speed
359 normalized by body height and the ML MoS at mid-stance in healthy adults (Lencioni et al.,
360 2020). The moderate relationship suggests that other factors such as stance time and step width
361 may play a more significant role in influencing the ML MoS during mid-stance and late stance
362 (Buurke et al., 2023; Hof, 2008). Also, the weaker relationship (i.e., lower correlation
363 coefficient) in the ML compared to the AP direction, aligns with the principle that the xCoM
364 incorporates the velocity of the CoM. Consequently, the AP MoS is more strongly related to
365 gait speed than the ML MoS since more motion occurs in that direction (McCrumb et al., 2019).
366 A previous study has shown that the ML MoS may be, among other factors, related to age (i.e.,
367 older individuals will have a larger MoS) (Lencioni et al., 2020). However, the present study
368 did not assess the relationship between age and ML MoS, due to the narrow range of age (21–
369 41 years old) of the included population.

370 *3.5. The effect of body mass*

371 A more forward MoS was observed when body mass is increased during the end of the initial
372 contact phase (11 – 18%, $r = -0.523$), and during foot flat and push-off phases (25 – 100%, $r =$

373 -0.598) (**figure 4e**). This observation may align with previous finding that obese individuals
374 tend to adopt a more forward posture during balance-challenging tasks, which has been
375 attributed to the increased difficulty they encounter in controlling anterior-posterior body
376 movements (Berrigan et al., 2006; Menegoni et al., 2009). In the ML direction, a larger MoS
377 (i.e., the xCoM is more medial than the lateral limit of the BoS) was observed in individuals
378 with increased body mass during late stance (66 – 97%, $r = 0.563$) (**figure 4f**). This may reflect
379 a strategy for conserving energy, possibly driven by their larger mass, which reduces the need
380 for substantial mechanical effort to shift to the other side. This adaptation aligns with the
381 findings of previous studies who also reported a larger MoS during gait in healthy individuals
382 with higher body mass (Herssens et al., 2020; Lencioni et al., 2020).

383 *3.6. Limitations*

384 This study has some limitations. First, the MoS is an instant measure of gait stability that does
385 not allow for a comprehensive understanding of the factors contributing to gait stability such
386 as segmental/joint kinematics and muscle activity. Future research should consider integrating
387 these parameters for a more refined interpretation of the MoS. Second, expanding the study to
388 include pathological populations and a wider age range would help generalize the findings and
389 enhance their clinical relevance. Our approach is intended to be inclusive of pathological
390 populations, but we have not yet validated it with these groups. Third, the age factor is also
391 underrepresented, as the participant age range is too narrow, limiting the applicability of the
392 results to other age groups.

393 **4. Conclusion**

394 This study stands out as one of the few to have continuously evaluated the MoS across the gait
395 cycle. It introduces a novel method using optimal foot markers to estimate BoS in the absence
396 of CoP data. By consistently normalizing MoS (AP by foot length, gait speed, and body mass;
397 ML by gait speed and body mass), this approach enhances gait stability assessment, improves
398 population comparisons, and better identifies stability deviations in pathological gait patterns.

399

400 **Author contributions**

401 Conceptualization; CDP, YC - Formal analysis; CDP, CR, YC - Methodology; CDP, CR, YC
402 - Supervision; YC - Visualization; CDP, CR, YC - Roles/Writing - original draft; CDP, YC -
403 Writing - review & editing; YC, RT

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407

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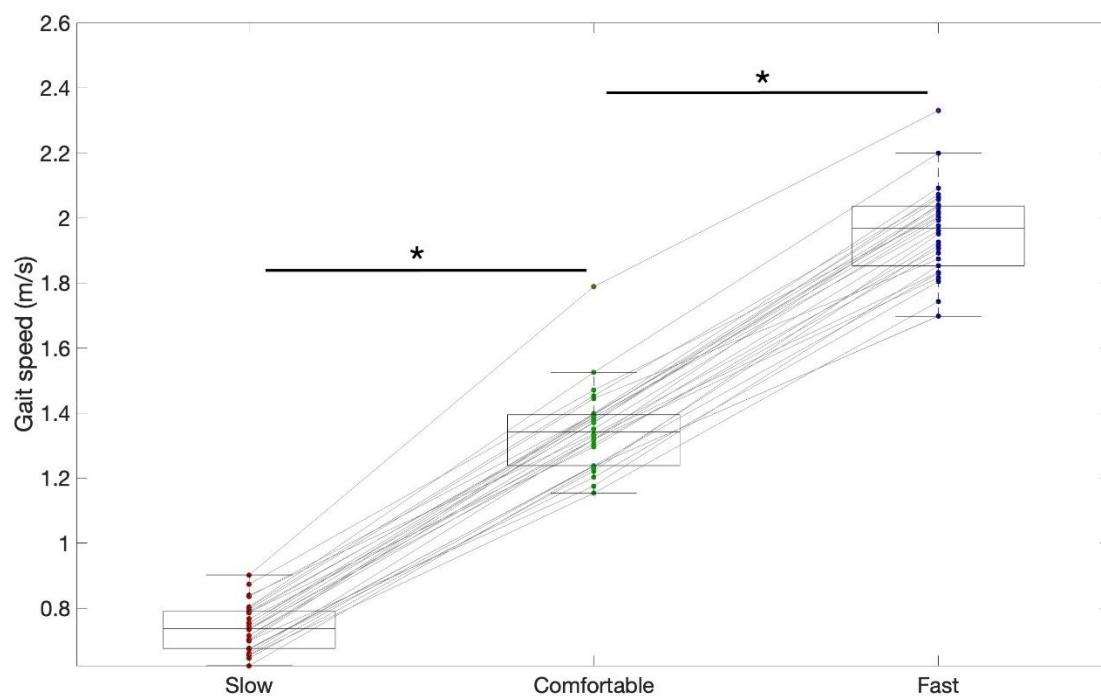
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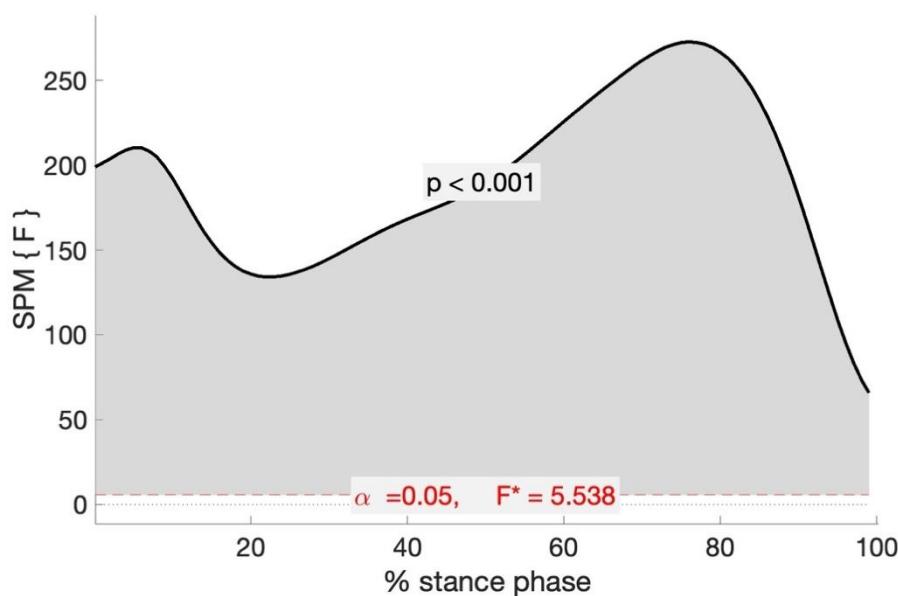
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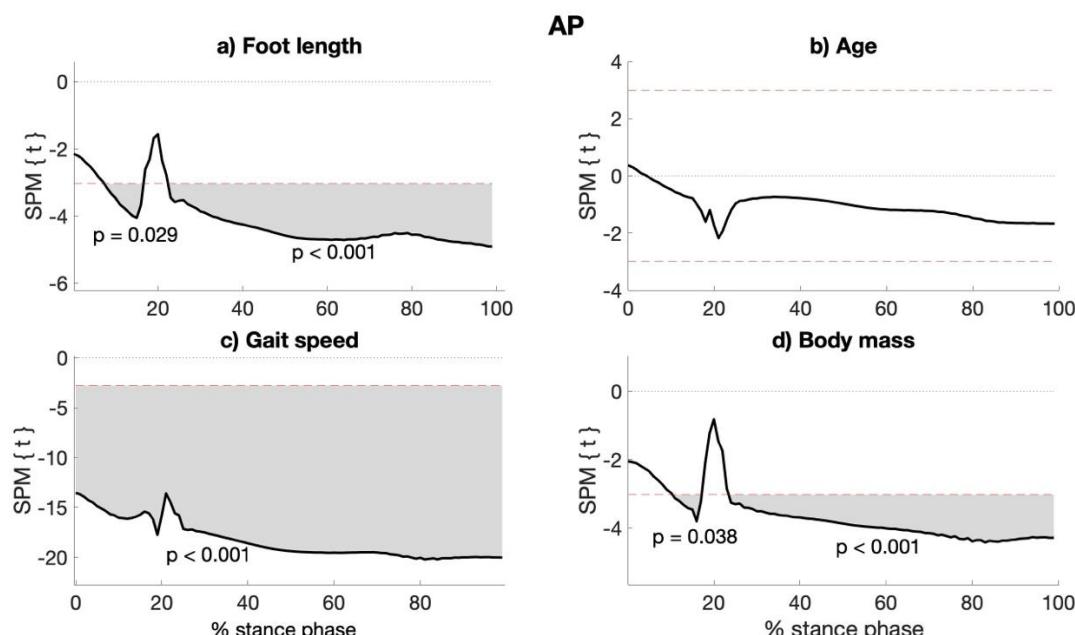
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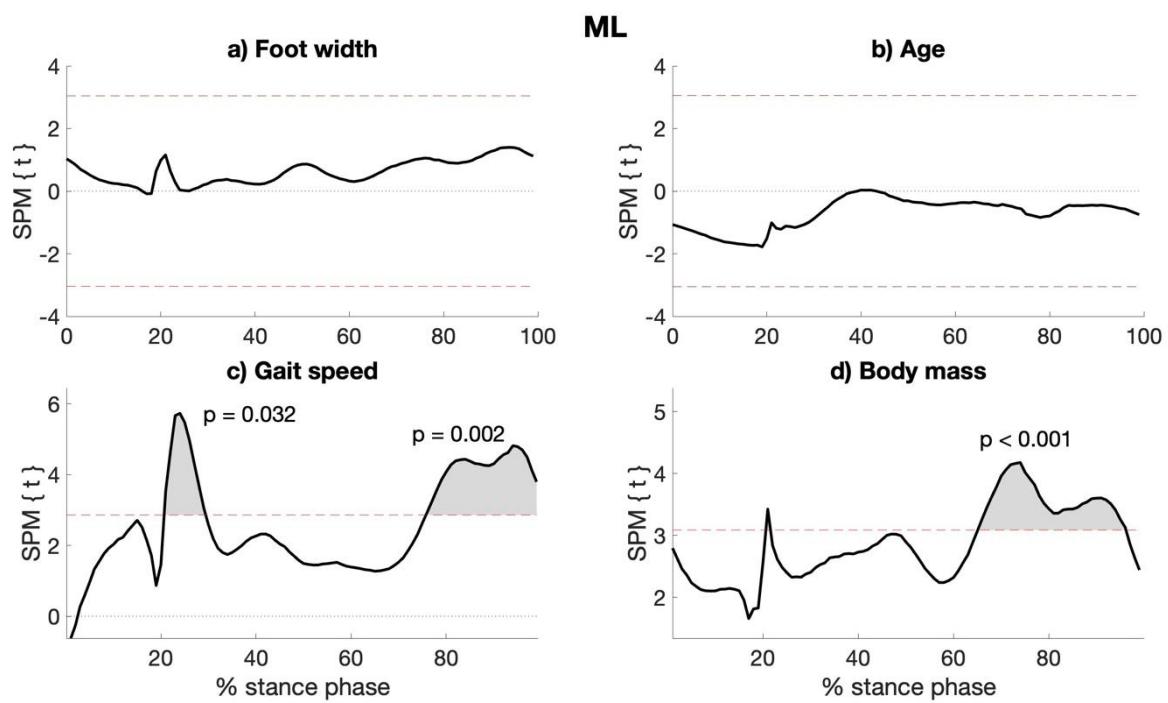
Supplementary figure 1. Boxplots representing the participant mean gait speed (m/s) across 12 gait cycles for each condition (slow, comfortable, fast), with the red line indicating the group median gait speed. Individual changes across the three speed conditions are presented with a grey line that connects the mean gait speed. The top and bottom edges of the box represent the first quartile (Q1) and third quartile (Q3), respectively. The whiskers extend to the largest and smallest values within 1.5 times the interquartile range from Q3 and Q1, respectively. The black horizontal lines with asterisks (*) denote statistically significant differences ($p < 0.05$, Wilcoxon Signed-Rank test) between the speed conditions. Participants gait speed were significantly different between speed conditions: slow vs comfortable (median difference [95% confidence interval (CI)]: 0.60 [0.56, 0.64] m/s, $p < 0.001$, Glass delta effect size (Δ): 4.930); comfortable vs fast (median difference [95% CI]: -0.61 [-0.66, -0.59] m/s, $p < 0.001$, Δ : 4.934) (see figure 2).



Supplementary figure 2. Results of the statistical parametric mapping analyses investigating the analysis of variance (ANOVA) between the mediolateral margin of stability calculation approaches. F-statistic from the ANOVA, used to determine whether there are significant differences between the three groups is represented by the red dashed lines as the critical thresholds for statistical significance. Values above or under these lines indicate statistically significant difference (ANOVA, $p < 0.05$) between the compared approaches at that specific point in the stance phase.



Supplementary figure 3. Results of the statistical parametric mapping (SPM) analyses investigating the regression between foot length (a), age (b), gait speed (c), and weight (d), and the antero-posterior (AP) margin of stability throughout the stance phase of the gait cycle.



Supplementary figure 4. Results of the statistical parametric mapping (SPM) analyses investigating the regression between foot length (a), age (b), gait speed (c), and weight (d), and the medio-lateral margin (ML) of stability throughout the stance phase of the gait cycle.

Supplementary table 1. Literature review on the different approaches used to calculate the margin of stability

Study information			BoS definition		Population		Analysis
Year	Authors	Title	AP marker	ML marker	Groups (n)	Age (years old)	Relationship assessed
2011	Curtze, C., Hof, A. L., Postema, K., & Otten, B.	Over rough and smooth: Amputee gait on an irregular surface	TOE	Midpoint: Calcaneus-M2	A: 18	A: 55.6 ± 9.5	
2011	Carty, C. P., Mills, P., & Barrett, R.	Recovery from forward loss of balance in young and older adults using the stepping strategy	TOE		E: 31 HYA: 16	E: [65.0 - 80.0] HYA: [20.0 - 35.0]	<u>AP MoS:</u> Age
2012	Young, P. M. M., & Dingwell, J. B.	Voluntary changes in step width and step length during human walking affect dynamic margins of stability	TOE	Lateral heel marker	HYA: 13	HYA: [18.0 - 35.0]	<u>AP MoS:</u> Step length, step width <u>ML MoS:</u> Step width
2012	Young, P. M. M., Wilken, J. M., & Dingwell, J. B.	Dynamic margins of stability during human walking in destabilizing environments	TOE	Lateral heel marker	HYA: 12	n/s	
2012	Süptitz, F., Karamanidis, K., Catalá, M. M., & Brüggemann, G.	Symmetry and reproducibility of the components of dynamic stability in young adults at different walking velocities on the treadmill	HALLUX		HYA: 11	HYA: 25.5 ± 2.1	
2013	Hak, L., Van Dieën, J. H., Van Der Wurff, P., Prins, M. R., Mert, A., Beek, P. J., & Houdijk, H.	Walking in an unstable environment: Strategies used by transtibial amputees to prevent falling during Gait	ANKLE	ANKLE	HA: 9 TA: 10	HA: 37.0 ± 11.4 TA: 38.8 ± 14.6	
2013	Hak, L., Houdijk, H., Van Der Wurff, P., Prins, M. R., Mert, A., Beek, P. J., & Van Dieën, J. H.	Stepping strategies used by post-stroke individuals to maintain margins of stability during walking	ANKLE	ANKLE	HA: 9 PS: 10	HA: 57.3 ± 7.2 PS: 60.8 ± 8.4	
2013	Hak, L., Houdijk, H., Beek, P. J., & Van Dieën, J. H.	Steps to take to enhance gait stability: The stride frequency, stride length, and walking speed on local dynamic stability and margins of stability	HEEL	ANKLE	HYA: 9	HYA: 21.9 ± 1.8	<u>AP MoS:</u> Gait speed, stride length <u>ML MoS:</u> Stride frequency
2013	Gates, D. H., Scott, S. J., Wilken, J. M., & Dingwell, J. B.	Frontal plane dynamic margins of stability in individuals with and without transtibial amputation walking on a loose rock surface		M5	HYA: 15 TA: 13	HYA: 22.0 ± 5.0 TA: 28.0 ± 4.0	<u>ML MoS:</u> Gait speed
2014	Beltran, E. J., Dingwell, J. B., & Wilken, J. M.	Margins of stability in young adults with traumatic transtibial amputation walking in destabilizing environments		M5	HYA: 13 TA: 9	HYA: 24.8 ± 6.9 TA: 30.7 ± 6.8	<u>ML MoS:</u> Platform and visual oscillations
2014	Kao, P., Dingwell, J. B., Higginson, J. S., & Binder-Macleod, S.	Dynamic instability during post-stroke hemiparetic walking	TOE	M5	HA: 9 PS: 9	HA: 61.7 ± 10.0 PS: 60.8 ± 9.0	
2014	McCrumb, C., Eysel-Gosepath, K., Epro, G., Meijer, K., Savelberg, H. H. C. M., Brüggemann, G., & Karamanidis, K.	Deficient recovery response and adaptive feedback potential in dynamic gait stability in unilateral peripheral vestibular disorder patients	TOE	n/s	HA: 17 UPVD: 17	HA: 51.0 ± 8.0 UPVD: 49.0 ± 9.0	

2015	Rijken, N., Van Engelen, B., Geurts, A., & Weerdesteyn, V.	Dynamic stability during level walking and obstacle crossing in persons with facioscapulohumeral muscular dystrophy	HEEL	ANKLE	FSHD: 10	FSHD: [43.0 - 68.0]	
2015	Hak, L., Houdijk, H., Wurff, P., Prins, M., Beek, P., & Van Dieën, J.	Stride frequency and length adjustment in post-stroke individuals: Influence on the margins of stability	HEEL	ANKLE	PS: 10	PS: [26 - 74]	<u>AP MoS:</u> Stride length, frequency, gait speed
2015	Hoogkamer, W., Bruijn, S. M., Sunaert, S., Swinnen, S. P., Van Calenbergh, F., & Duysens, J.	Toward new sensitive measures to evaluate gait stability in focal cerebellar lesion patients	Posterior boundary of the feet	Lateral boundary of the feet	HYA: 14 CL: 18	HYA: 24.4 ± 3.5 CL: 24.4 ± 7.3	
2016	Peebles, A. T., Reinholdt, A., Bruetsch, A. P., Lynch, S. G., & Huisenga, J. M.	Dynamic margin of stability during gait is altered in persons with multiple sclerosis	TOE	M5	HA: 20 MS1: 20 MS2: 20	HA: 47.5 ± 7.8 MS1: 45.8 ± 8.6 MS2: 45.9 ± 8.7	<u>AP and ML MoS:</u> Gait speed
2016	Delabastita, T., Desloovere, K., & Meyns, P.	Restricted Arm Swing Affects Gait Stability and Increased Walking Speed Alters Trunk Movements in Children with Cerebral Palsy		ANKLE	TD: 24 CP: 26	TD: [5.0 - 12.0] CP: [4.0 - 12.0]	
2016	van Meulen, F. B., Weenk, D., van Asseldonk, E. H., Schepers, H. M., Veltink, P. H., & Buurke, J. H.	Analysis of balance during functional walking in stroke survivors	Midpoint: front of each foot	Lateral shoe position	PS: 10	PS: 63.2 ± 8.9	
2017	Simon, A., Lugade, V., Bernhardt, K., Larson, N. A., & Kaufman, K.	Assessment of stability during gait in patients with spinal deformity—A preliminary analysis using the dynamic stability margin	M5	Midpoint: M5- ANKLE	HYA: 12 SD: 17	HYA: [23.2 - 27.1] SD: [23.8 - 50.4]	
2017	Ghomian, B., Mehdizadeh, S., Aghili, R., Naemi, R., Jafari, H., Machado, J., Silva, L. F., Lobarinhas, P., & Saeedi, H.	Rocker outsole shoes and margin of stability during walking: A preliminary study	TOE	M5	DB: 1	DB: 50.0	<u>AP MoS:</u> Using rocker outsole shoes
2017	Acasio, J., Wu, M., Fey, N. P., & Gordon, K. E.	Stability-maneuverability trade-offs during lateral steps		M5	HYA: 10	HYA: 25.6 ± 3.4	
2017	Martelli, D., Luo, L., Kang, J., Kang, U. J., Fahn, S., & Agrawal, S. K.	Adaptation of Stability during Perturbed Walking in Parkinson's Disease	HALLUX	M5	HA: 9 PD: 9	HA: 64.7 ± 7.3 PD: 64.3 ± 7.4	
2017	Peebles, A. T., Bruetsch, A. P., Lynch, S. G., & Huisenga, J. M.	Dynamic balance in persons with multiple sclerosis who have a falls history is altered compared to non-fallers and to healthy controls	TOE	TOE	HA: 27 MS: 55	HA: 44.9 ± 9.9 MS: 45.9 ± 9.4	<u>AP and ML MoS:</u> Fall history
2018	Guaitolini, M., Aprigliano, F., Mannini, A., Sabatini, A. M., & Monaco, V.	Effects of gait speed on the margin of stability in healthy young adults	M1	M5	HYA: 8	HYA: [22.0 - 32.0]	<u>AP and ML MoS:</u> Gait speed
2018	Havens, K. L., Mukherjee, T., & Finley, J. M.	Analysis of biases in dynamic margins of stability introduced by the use of	TOE	Lateral heel marker	HYA: 12	HYA: 26.0 ± 3.0	

		simplified center of mass estimates during walking and turning					
2018	Tisserand, R., Armand, S., Allali, G., Schnider, A., & Baillieul, S.	Cognitive-motor dual-task interference modulates mediolateral dynamic stability during gait in post-stroke individuals.		Midpoint: HEEL-M2	HA: 10 PS: 12	HA: 68.5 ± 4 PS: 58.0 ± 12.8	<u>ML MoS:</u> Cognitive tasks of various attention load
2018	Sivakumaran, S., Schinkel-Ivy, A., Masani, K., & Mansfield, A.	Relationship between margin of stability and deviations in spatiotemporal gait features in healthy young adults	TOE	M5	HYA: 11	HYA: 24.0 ± 4.4	<u>AP and ML MoS:</u> Step length, step width
2018	McCrum, C., Willems, P., Karamanidis, K., & Meijer, K.	Stability-normalised walking speed: a new approach for human gait perturbation research	HALLUX		HYA: 18	HYA: 24.4 ± 2.5	<u>AP MoS:</u> Gait speed
2019	AminiAghdam, S., Griessbach, E., Vielemeyer, J., & Müller, R.	Dynamic postural control during (in)visible curb descent at fast versus comfortable walking velocity	HALLUX		HYA: 12	HYA: 25.5 ± 4.7	
2019	Tracy, J. B., Petersen, D. A., Pigman, J., Conner, B. C., Wright, H. G., Modlesky, C. M., Miller, F., Johnson, C. L., & Crenshaw, J. R.	Dynamic stability during walking in children with and without cerebral palsy	TOE	n/s	CP: 15 TD: 14	CP: 8.7 ± 2.4 TD: 9.1 ± 2.5	
2019	Lencioni, T., Carpinella, I., Rabuffetti, M., Cattaneo, D., & Ferrarin, M.	Measures of dynamic balance during level walking in healthy adult subjects: Relationship with age, anthropometry and spatio-temporal gait parameters	M5	M5	HA: 34	HA: [21.0 - 71.0]	<u>AP MoS:</u> Age, body height, body mass, gait speed, stride length and cadence <u>ML MoS:</u> Age, body height, body mass, step width
2019	Van Vugt, Y., Stinear, J., Davies, T. C., & Zhang, Y.	Postural stability during gait for adults with hereditary spastic paraparesis	Metatarsal marker of the stance foot	2cm lateral to M2	HA: 10 HSP: 10	HA: 56.4 ± 16.0 HSP: 53.5 ± 11.5	
2019	Ohtsu, H., Yoshida, S., Minamisawa, T., Takahashi, T., Yomogida, S., & Kanzaki, H.	Investigation of balance strategy over gait cycle based on margin of stability	M5	Medial HEEL	HYA: 30	HYA: 21.2 ± 0.8	
2019	Arora, T., Musselman, K. E., Lanovaz, J. L., Linassi, G., Arnold, C., Milosavljevic, S., & Oates, A.	Walking Stability During Normal Walking and Its Association with Slip Intensity Among Individuals with Incomplete Spinal Cord Injury	n/s		HA: 16 ISCI: 20	HA: 58.9 ± 17.1 ISCI: 60.0 ± 17.8	
2019	Major, M. J., McConn, S. M., Zavaleta, J. L., Stine, R., & Gard, S. A.	Effects of upper limb loss and prosthesis use on proactive mechanisms of locomotor stability		M5	ULL: 10	ULL: 50.0 ± 19.0	
2020	Herssens, N., Van Criekinge, T., Saeys, W., Truijen, S., Vereeck, L., Van Rompaey, V., & Hallemans, A.	An investigation of the spatio-temporal parameters of gait and margins of stability throughout adulthood	HEEL	M5	HA: 105	HA: [20.0 - 89.0]	<u>AP MoS:</u> Body mass index <u>ML MoS:</u> Age, body mass index
2021	Ma, Y., Mithraratne, K., Wilson, N. C., Zhang, Y., & Wang, X.	Kinect V2-Based Gait Analysis for Children with Cerebral Palsy: Validity	TOE	ANKLE	CP: 10	CP: 6.4 ± 2.2	

		and Reliability of Spatial Margin of Stability and Spatiotemporal Variables					
2021	Rethwilm, R., Böhm, H., Haase, M., Perchthaler, D., Dussa, C. U., & Federolf, P.	Dynamic stability in cerebral palsy during walking and running: Predictors and regulation strategies		ANKLE	CP: 117 TD: 25	CP: 11.0 ± 3.2 TD: 10.4 ± 2.5	
2022	Yamaguchi, T., & Masani, K.	Effects of age on dynamic balance measures and their correlation during walking across the adult lifespan	HEEL	M5	HA: 151	HA: [20.0 - 77.0]	<u>ML MoS:</u> Age
2024	Sangeux, M., Viehweger, E., Romkes, J., & Bracht-Schweizer, K.	On the clinical interpretation of overground gait stability indices in children with cerebral palsy	TOE	M5	TD: 20 CP: 20	TD: [7.7 - 16.7] CP: [8.3 - 17.8]	
2024	Wang, Z., Xie, H., & Chien, J. H.	The margin of stability is affected differently when walking under quasi-random treadmill perturbations with or without full visual support	HEEL	M5	HYA: 20	HYA: 22.6 ± 2.8	

Summary of studies ($n = 41$) that have investigated either the antero-posterior (AP) or the mediolateral (ML) margin of stability (MoS). For each MoS calculation, the markers used to define AP and/or ML base of support were retrieved. The population (n, age, and pathology) included, and the relationship assessed are also reported. The age is reported by the range [min - max], or by the mean \pm SD, according to how it was reported in the study. Empty boxes indicate that the element was not assessed by the study. Population abbreviations are as follows: Amputees, AM; Cerebellar lesions, CL; Cerebral palsy, CP; Diabetes, DB; Elderly, E; Facioscapulohumeral muscular dystrophy, FSHD; Healthy adults, HA; Hereditary spastic paraparesis, HSP; Healthy young adults, HYA; Incomplete spinal cord injury, ISCI; Multiple sclerosis, MS; Multiple sclerosis without gait impairments, MS1; Multiple sclerosis with gait impairments, MS2; Parkinson disease, PD; Post-stroke, PS; Spinal deformity, SD; TA, Transtibial Amputees; Typically developing, TD; Upper limb loss, ULL; Unilateral peripheral vestibular disorder, UPVD. Other abbreviations: ANKLE, Lateral malleoli; HALLUX, Hallux; HEEL, Calcaneum; M5, 5th metatarsal; TOE, 2nd metatarsal.