Stability-normalised walking speed: a new approach for human gait perturbation research

perturbation research

Christopher McCrum[1,2*], Paul Willems[1], Kiros Karamanidis[3], Kenneth Meijer[1]

[1]Department of Nutrition and Movement Sciences, NUTRIM School of Nutrition and

Translational Research in Metabolism, Maastricht University Medical Centre+, Maastricht,

The Netherlands

[2]Institute of Movement and Sport Gerontology, German Sport University Cologne, Germany

[3]Sport and Exercise Science Research Centre, School of Applied Sciences, London South

Bank University, London, UK

*Correspondence:

Christopher McCrum

Department of Nutrition and Movement Sciences

Maastricht University,

PO Box 616, Maastricht, 6200 MD, The Netherlands

+31 (0) 43 388 1621

chris.mccrum@maastrichtuniversity.nl

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Abstract

In gait stability research, neither self-selected walking speeds, nor the same prescribed walking speed for all participants, guarantee equivalent gait stability among participants. Furthermore, these options may differentially affect the response to different gait perturbations, which is problematic when comparing groups with different capacities. We present a method for decreasing inter-individual differences in gait stability by adjusting walking speed to equivalent margins of stability (MoS). Eighteen healthy adults walked on a split-belt treadmill for two-minute bouts at 0.4m/s up to 1.8m/s in 0.2m/s intervals. The stability-normalised walking speed (MoS=0.05m) was calculated using the mean MoS at touchdown of the final 10 steps of each speed. Participants then walked for three minutes at this speed and were subsequently exposed to a treadmill belt acceleration perturbation. A further 12 healthy adults were exposed to the same perturbation while walking at 1.3m/s: the average of the previous group. Large ranges in MoS were observed during the prescribed speeds (6-10cm across speeds) and walking speed significantly ($P<0.001$) affected MoS. The stability-normalised walking speeds resulted in MoS equal or very close to the desired 0.05m and reduced between-participant variability in MoS. The second group of participants walking at 1.3m/s had greater inter-individual variation in MoS during both unperturbed and perturbed walking compared to 12 sex, height and leg length-matched participants from the stability-normalised walking speed group. The current method decreases inter-individual differences in gait stability which may benefit gait perturbation and stability research, in particular studies on populations with different locomotor capacities. [Preprint: https://doi.org/10.1101/314757]

Introduction

Mechanical perturbations have been used for decades to investigate the stability of human walking (Berger et al., 1984; Marigold and Patla, 2002; Nashner, 1980; Quintern et al., 1985; Vilensky et al., 1999) and are now frequently <u>applied</u> in falls prevention contexts (Gerards et al., 2017; Mansfield et al., 2015; Pai and Bhatt, 2007). In gait perturbation studies, self-selected walking speeds (for example: Pai et al., 2014) or a prescribed walking speed for all participants (for example: McCrum et al., 2016a) are commonly used, but each comes with drawbacks that complicate the interpretation of results.

A prescribed walking speed <u>(for example, 1.5m/s for all participants)</u> will not result in comparable stability for all participants. This is problematic when comparing groups with different capacities during a gait perturbation task, as the relative challenge of the task will vary. <u>In such a situation, the difficulty in recovering stability following mechanical perturbations will be affected by the relative neuromuscular and biomechanical demands of the task. As well as the demand of recovering from one perturbation, the need for adaptation following repetition of a perturbation may be different. As a result, it is common to use the self-selected or preferred walking speed in gait perturbation research, but this can</u> introduce other problems.

<u>Having participants walk at their own self-selected speeds implies that there will be variation across participants, which is likely to be much greater when multiple groups with different locomotor capacities are involved. There is ample evidence that</u> walking speed affects recovery strategy choice following slips (Bhatt et al., 2005) and trips (Krasovsky et al., 2014), the direction of balance loss following slipping (Smeesters et al., 2001) and differentially affects falls risk following tripping and slipping (Bhatt et al., 2005; Espy et al., 2010; Pavol et al., 1999). Gait stability at perturbation onset may also not be optimised at the self-selected speed and may differ across groups (Bhatt et al., 2005; Hak et al., 2013; Mademli and Arampatzis,

75   2014; Süptitz et al., 2012). For example, older adults walk with a lower safety factor than young

76   adults at self-selected walking speeds (Mademli and Arampatzis, 2014) and reduce stability to

77   benefit from centre of mass velocity when descending stairs; a potential compensation for

78   reduced lower limb neuromuscular capacities (Bosse et al., 2012). Taken together, this

79   evidence means that gait perturbation tasks could have very different effects across participants

80   walking at their self-selected speeds, and it may be difficult to determine if group differences

81   are true differences or artefacts of the above walking speed-related effects. These issues can be

82   further confounded, as walking speed directly affects measures of dynamic gait stability using

83   a centre of mass – base of support relationship model (Bhatt et al., 2005; Hak et al., 2013;

84   Süptitz et al., 2012). Therefore, more sophistication in the choice of walking speed may be

85   necessary for detailed study of reactive gait stability and adaptation processes.

86   Two possible solutions have been applied in previous gait perturbation studies. A Froude

87   number (a dimensionless parameter) for walking speed (Hof, 1996) has been applied to

88   normalise the walking speed based on leg length (Aprigliano et al., 2016; Aprigliano et al.,

89   2017; Martelli et al., 2013; Martelli et al., 2016). Originally developed to analyse the dynamic

90   similarity of differently sized boats (Vaughan and O'Malley, 2005), the Froude number has

91   been applied for the purpose of comparing the gaits of different sizes and species of animals

92   and results in dynamic similarity of the inverted pendulum motion in gait (Alexander, 1989,

93   1991; Vaughan and O'Malley, 2005). However, while the inverted pendulum motion may be

94   dynamically similar between participants, this normalisation based on leg length is not

95   necessarily synonymous with a normalisation of gait stability, because factors such as

96   individual differences in foot placement, posture, leg length to truck length ratio and internal

97   properties of the neuromotor and neuromuscular systems are ignored. Task demand in such

98   gait perturbation protocols (and most locomotor tasks) depends critically on these other factors

99   and not only on the dimensions of the body; an 18-year-old and an 80-year-old with the same

leg length are unlikely to be equally challenged by a gait perturbation while walking at the

same speed. Two studies have used 60% of the walk-to-run velocity to normalise the speed to

participants' walking-related neuromuscular capacities (Bierbaum et al., 2010, 2011).

However, this procedure did not lead to comparable stability during non-perturbed walking,

with the margins of stability and the components of the margins of stability showing differences

between the young and older subjects (Bierbaum et al., 2010, 2011), again probably due to the

fact that gait stability is not determined exclusively by the neuromuscular properties

responsible for gait speed. As both existing normalisation methods are based on a single

parameter, neither of which are the sole determinants of gait stability, one cannot expect

equivalent gait stability among participants. Therefore, further attempts to tackle these issues

are warranted (McCrum et al., 2016b; McCrum et al., 2017).

Here, we present a new method for decreasing inter-individual differences in gait stability by

normalising the walking speed based on gait stability. For this method we use the margins of

stability (MoS) concept (Hof et al., 2005), one of the few well-defined and well-accepted

biomechanical measures of mechanical stability of the body configuration during locomotion

(Bruijn et al., 2013), useful for assessing changes in gait stability due to mechanical

perturbations and balance loss. Additionally, we present results from a gait perturbation

experiment comparing participants walking at their stability-normalised walking speed with

participants walking all at the same prescribed speed.

Methods

Participants

Eighteen healthy adults participated in the first part of this study (eight males, 10 females; age:

24.4±2.5y; height: 174.9±7.4cm; weight: 74.6±15.2kg). Twelve healthy adults participated in

the second part of the study (Table 1). The participants had no self-reported history of walking

125 difficulties, dizziness or balance problems, and had no known neuromuscular condition or

126 injury that could affect balance or walking. Informed consent was obtained and the study was

127 conducted in accordance with the Declaration of Helsinki. The study protocol was approved

128 by the Maastricht University Medical Centre medical ethics committee.

129

130 *Setup and Procedures*

131 The Computer Assisted Rehabilitation Environment Extended (CAREN; Motekforce Link,

132 Amsterdam, The Netherlands), comprised of a dual-belt force plate-instrumented treadmill

133 (Motekforce Link, Amsterdam, The Netherlands; 1000Hz), a 12-camera motion capture system

134 (100Hz; Vicon Motion Systems, Oxford, UK) and a virtual environment that provided optic

135 flow, was used for this study. A safety harness connected to an overhead frame was worn by

136 the participants during all measurements. Five retroreflective markers were attached to

137 anatomical landmarks (C7, left and right trochanter and left and right hallux) and were tracked

138 by the motion capture system.

139 In the first part of the study (18 participants), the measurement sessions began with 60s

140 familiarisation trials of walking at 0.4m/s up to 1.8m/s in 0.2m/s intervals. After approximately

141 five to ten minutes rest, single two-to-three-minute-long measurements were then conducted at

142 the same speeds. Following these measurements, the stability-normalised walking speed was

143 calculated. To determine the stability-normalised walking speed, the mean anteroposterior

144 MoS (see below) at foot touchdown of the final 10 steps of each walking trial (0.4m/s to 1.8m/s)

145 were taken and fitted with a second order polynomial function. For each participant, the speed

146 resulting in MoS of 0.05m was calculated. Based on our pilot testing, this value would result

147 in walking speeds that would be possible for healthy adults of most ages (Bierbaum et al., 2010,

148 2011; Süptitz et al., 2013). With certain populations, slower walking speeds would be required

149 and then a greater MoS could be used. Participants then walked for three minutes at their

stability-normalised walking speed, at the end of which, a gait perturbation was applied without warning. The perturbation consisted of <u>an 80% increase in the right treadmill belt speed from the stability-normalised walking speed of the participant with a 3m/s^2 acceleration, and</u> thereby, we also normalised the magnitude of the perturbation to the already normalised walking speed. The <u>acceleration began before touchdown of the to-be-perturbed limb to ensure the belt was already at a higher speed when the foot touched down</u> (triggered automatically by the D-Flow software of the CAREN, when the hallux marker of the to-be-perturbed limb became anterior to the stance limb hallux marker in the sagittal plane). The belt decelerated after toe-off of the perturbed limb.

In the second part of the study, 12 participants completed the same familiarisation protocol and then walked for three minutes at 1.3m/s (average stability-normalised walking speed of the 18 participants in the first part of the study). After this, they experienced the same treadmill belt acceleration perturbation. To compare these results with a matched sample, 12 participants from the first group of 18 were selected and matched specifically for sex, height and leg length to the participants in part two of the study (Table 1).

Data Processing

Marker tracks were filtered using a low pass second order Butterworth filter (zero-phase) with a 12Hz cut-off frequency. Foot touchdown was detected using a combination of force plate (50N threshold) and foot marker data (Zeni et al., 2008). The anteroposterior MoS were calculated at foot touchdown as the difference between the anterior boundary of the base of support (anteroposterior component of the hallux marker projection to the ground) and the extrapolated centre of mass as defined by Hof et al. (2005), adapted for our reduced kinematic model based on <u>Süptitz et al. (2013)</u>, as follows:

$$X_{CoM} = \frac{P_{TroL} + P_{TroR}}{2} - P_{HalluxP} + \frac{0.5\left(\frac{V_{TroL} + V_{TroR}}{2} + V_{C7}\right) + |V_{Belt}|}{\sqrt{\frac{g}{L_{Ref}}}}$$

where P_{TroL}, P_{TroR} and $P_{HalluxP}$ are the trochanter and the posterior hallux marker anteroposterior positions respectively; V_{TroL}, V_{TroR} and V_{C7} are the anteroposterior velocities of the trochanter and C7 markers respectively; V_{Belt} is the treadmill belt velocity; g is gravitational acceleration ($9.81 m/s^2$); and L_{Ref} is the reference leg length. This reduced kinematic model was previously shown to be suitable for assessing the MoS and it's components during unperturbed and perturbed treadmill walking in young, middle and older-aged healthy adults, with high correlations and no clear differences compared to a full kinematic model (Süptitz et al., 2013). Note that a large proportion of the CoM velocity is derived from the treadmill belt speed, potentially improving the accuracy compared with overground walking when the entire CoM velocity is derived from the markers. The MoS was calculated for: the final 10 steps of each set walking speed in the first part of the study; the mean MoS of the eleventh to second last step before each perturbation (Base); the final step before each perturbation (Pre); and the first recovery step following each perturbation (Post1).

Statistics

A mixed effects model for repeated measures with walking speed as a fixed effect and Tukey post hoc comparisons was used to confirm a walking speed effect on the MoS. To determine whether a normalisation of walking speed based on body dimensions would assume equivalent gait stability, Pearson correlations between the stability-normalised walking speeds and participants' height and leg length were conducted. A two-way repeated measures ANOVA with participant group (Stability-normalised walking speed [Norm] and 1.3m/s) and step (Base, Pre, Post1) as factors with post hoc Sidak's tests for multiple comparisons were used to determine between group differences in the MoS. Equivalence tests using 90% confidence

199 intervals were used to confirm the similarity of the groups' demographics. Significance was

200 set at α=0.05. When sphericity was violated, a Greenhouse-Geisser correction was applied.

201 Normality of the distributions was assessed with Q-Q plots. Analyses were performed using

202 Prism version 8 for Windows (GraphPad Software Inc., La Jolla, California, USA).

203

204 **Results and Discussion**

205 *Stability during unperturbed walking*

206 Walking speed significantly affected the MoS ($F_{[2.547, 42.93]}=1485$, $P<0.0001$, $\hat{\varepsilon}=0.3638$; Fig. 1)

207 and Tukey's multiple comparisons tests revealed significant differences for each speed

208 compared to all other speeds ($P<0.0001$; Fig. 1). These results agree with previous work (Bhatt

209 et al., 2005; Hak et al., 2013; Süptitz et al., 2012). A range of MoS values were observed for

210 each speed (approximately 6-10cm), even among these healthy participants, confirming some

211 of the issues related to prescribed walking speeds in gait stability research discussed above.

212 The strong relationship between walking speed and MoS also has relevance for clinical studies

213 conducting self-paced gait measurements with an assessment of gait stability. Patients who

214 improve in walking speed may demonstrate a reduction in MoS, which may not be reductions

215 in the stability of the patients' gait *per se*, but simply an artefact of the improved walking speed.

216 *Insert Fig. 1*

217 The stability-normalised walking speeds (range from 1.22m/s to 1.51m/s with a mean±SD of

218 1.3±0.1m/s) resulted in MoS very close to the desired outcome of 0.05m (within one SD of the

219 mean MoS for 15 of the 18 participants; Fig. 2A). The stability-normalised walking speed also

220 reduced between-participant variability in MoS (as shown by the group level standard

221 deviations; Fig. 2B). These combined results indicate that the stability-normalisation was

222 successful in reducing between-participant differences in MoS during walking, even in a

223 homogenous group of healthy young adults.

9

224   *Insert Fig. 2*

225   Small, non-significant correlations between the determined stability-normalised walking

226   speeds and the participants' height and leg length were found (Fig. 3). The outcomes of our

227   correlation analysis suggest that height and leg length did not significantly affect the calculation

228   of stability-normalised walking speed, suggesting that a normalisation of walking speed based

229   on body dimensions does not assume equivalent gait stability, at least not when assessed by the

230   MoS concept.

231   *Insert Fig. 3*

232

233   <u>*Stability during perturbed walking*</u>

234   For the second part of the study, the 12 participants were successfully matched to the 12 of the

235   18 participants from part one of the study (Table 1). During the perturbations, the 1.3m/s group

236   had a greater range in MoS values during Base, Pre and Post1 (Fig. 4). A two-way repeated

237   measures ANOVA revealed a significant effect of group ($F_{[1, 22]}=6.409$, $P=0.019$), step ($F_{[1.097,}$

238    $_{24.14]}=8.34$, $P=0.0068$<u>, $\hat{\varepsilon}=0.5486$</u>) and a significant group <u>(Norm and 1.3m/s)</u> by step <u>(Base, Pre,</u>

239   <u>Post1)</u> interaction ($F_{[2, 44]}=15.4$, $P<0.0001$) on MoS. Sidak post hoc tests revealed a significant

240   difference between Norm and 1.3m/s groups at Post1 ($P=0.0049$). While part of the differences

241   found may be due to chance, the current comparison suggests that the stability-normalised

242   walking speed and the normalised perturbation (acceleration to a peak speed 180% of the

243   walking speed) reduce the inter-individual differences in MoS during both unperturbed and

244   perturbed walking, at least with the current protocol. The significant difference found at Post1

245   between the groups also aligns with the previous studies reporting different responses to

246   perturbations experienced while walking at different speeds (Bhatt et al., 2005; Krasovsky et

247   al., 2014).

248   *Insert Fig. 4*

Further methodological considerations

250 As the MoS – walking speed relationship from 1.0-1.6m/s appeared to be linear in part one of

251 the study (Fig. 1), a simple linear regression was calculated for 1.0-1.6m/s. A significant

252 regression equation was found (Fig. 5). Future research could use this (or similar) as a simple,

253 efficient method for increasing the dynamic similarity in gait stability across participants, by

254 measuring participants walking at a single speed from 1.0-1.6m/s and using this equation to

255 prescribe speeds that would result in similar MoS values. As it is common practice in gait

256 experiments to familiarise participants to the setup and conditions, including some practice

257 walking trials, we would suggest that this may be the ideal opportunity to incorporate our

258 method, without having to conduct any additional trials. It is, however, worth highlighting that

259 the current participants were young healthy adults; the walking speed – MoS relationship may

260 be altered in other populations. Future implementations of this method should consider the

261 capacities of the population of interest and the desired or expected impact on gait stability of

262 the perturbations when selecting an MoS value for normalisation.

263 **Insert Fig. 5**

264

265 *Limitations*

266 Individual responses in the MoS to the perturbation varied (Fig. 4), although the variation was

267 lower in the stability-normalised walking speed group. Part of the reasons for this variation

268 could be the result of uncontrolled factors such as individual physiological, biomechanical or

269 psychological differences affecting the individual response at the onset of the perturbation. It

270 could be argued that using a single trial as opposed to averaging multiple trials is less reliable,

271 however, due to the significant and rapid learning effects following even single perturbations

272 of this kind, the responses seen after averaging trials would no longer accurately represent

273 natural responses to unexpected perturbations. In this sense, our approach is ecologically valid,

274 as the variation is representative of daily life responses to truly unexpected gait perturbations.

275 Another potential limitation relates to a validity constraint of the MoS calculation detailed by

276 Hof et al. (2005), in that the pendulum length (distance from the centre of mass to the axis of

277 rotation) should remain constant. This may not always be the case during dynamic walking and

278 perturbed walking tasks if the knee is slightly flexed at foot contact. However, we have not

279 observed large changes in the pendulum length and small changes are not systematic, as within

280 and between individual variability in responses is large. We therefore believe that this is an

281 acceptable limitation of using the model in this context, but one that should be kept in mind

282 when interpreting the results.

283

284 *Conclusions*

285 In conclusion, large ranges in MoS were observed and walking speed significantly affected

286 MoS even within these young healthy participants, confirming some issues related to walking

287 speed choice in gait stability research. The current methods reduced between-participant

288 variability in MoS during both unperturbed and perturbed walking, meaning that the method

289 could be beneficial for gait stability studies comparing groups with different locomotor

290 capacities. An equation has been provided that can be used following a single gait trial to

291 increase the dynamic similarity of gait stability between participants.

292

293 **Conflict of Interest Statement**

294 The authors declare no conflict of interest.

295

301

302   **References**

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390

391 **Tables**

392 Table 1: Demographic characteristics of the participant groups in part two of the study.

	Sex	Age (y)	Height (cm)	Weight (kg)	Leg Length (cm)
1.3m/s Group	8 males, 4 females	25.1±3.8	178.2±5.2	72.5±9.7	84.2±2.1
Norm Group	8 males, 4 females	24.3±2.9	178.7±5.8	79±15.3	85.5±2.8
Equivalent based on 90% Confidence Intervals?	-	Yes	Yes	Yes	Yes

393

394

Figure Legends

Fig. 1: Individual margins of stability at foot touchdown over the different walking speeds. The dashed line represents the margin of stability used to determine the stability-normalised walking speed.

Fig. 2: **A**: Means and standard deviations of the margins of stability at touchdown of the final 10 steps at the stability-normalised walking speed for each individual participant. The desired MoS of 0.05m at foot touchdown is indicated by the dashed line. **B**: The between-participant variation in the margins of stability (standard deviation at group level) for the final 10 steps at each walking speed (the stability-normalised walking speed trials are indicated with the black circle; mean and standard deviation).

Fig. 3: Pearson correlations between the participants' stability-normalised walking speeds and their height and leg length.

Fig. 4: Margins of stability during unperturbed and perturbed walking of participants walking at their stability-normalised walking speed (Norm) and participants walking at 1.3m/s. Base: the mean MoS of the eleventh to second last step before each perturbation; Pre: the final step before each perturbation; Post1: the first recovery step following perturbation. *: Significant difference (Sidak post hoc test: P=0.0049).

Fig. 5: Margins of stability as a function of walking speed between 1.0 and 1.6m/s. The shaded area represents the 95% confidence intervals of the regression line.