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

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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 0.4 m/s up to 1.8 m/s in 0.2 m/s intervals. The stability-normalised walking speed (MoS = 0.05 m) 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 1.3 m/s: the average of the previous group. Large ranges in MoS were observed during the prescribed speeds (6–10 cm 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 0.05 m and reduced between-participant variability in MoS. The second group of participants walking at 1.3 m/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]

1. 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 applied 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 (for example, 1.5 m/s for all participants) 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. 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 introduce other problems.

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 walking speed affects recovery strategy choice following slips

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(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, 2014; Süptitz et al., 2012). For example, older adults walk with a lower safety factor than young adults at self-selected walking speeds (Mademli and Arampatzis, 2014) and reduce stability to benefit from centre of mass velocity when descending stairs; a potential compensation for reduced lower limb neuromuscular capacities (Bosse et al., 2012). Taken together, this evidence means that gait perturbation tasks could have very different effects across participants walking at their self-selected speeds, and it may be difficult to determine if group differences are true differences or artefacts of the above walking speed-related effects. These issues can be further confounded, as walking speed directly affects measures of dynamic gait stability using a centre of mass – base of support relationship model (Bhatt et al., 2005; Hak et al., 2013; Süptitz et al., 2012). Therefore, more sophistication in the choice of walking speed may be necessary for detailed study of reactive gait stability and adaptation processes.

Two possible solutions have been applied in previous gait perturbation studies. A Froude number (a dimensionless parameter) for walking speed (Hof, 1996) has been applied to normalise the walking speed based on leg length (Aprigliano et al., 2016, 2017; Martelli et al., 2013, 2016). Originally developed to analyse the dynamic similarity of differently sized boats (Vaughan and O'Malley, 2005), the Froude number has been applied for the purpose of comparing the gaits of different sizes and species of animals and results in dynamic similarity of the inverted pendulum motion in gait (Alexander, 1989, 1991; Vaughan and O'Malley, 2005). However, while the inverted pendulum motion may be dynamically similar between participants, this normalisation based on leg length is not necessarily synonymous with a normalisation of gait stability, because factors such as individual differences in foot placement, posture, leg length to truck length ratio and internal properties of the neuromotor and neuromuscular systems are ignored. Task demand in such gait perturbation protocols (and most locomotor tasks) depends critically on these other factors 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, 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.

2. Methods

2.1. Participants

Eighteen healthy adults participated in the first part of this study (eight males, 10 females; age: 24.4 ± 2.5y; height: 174.9 ± 7.4 cm; weight: 74.6 ± 15.2 kg). Twelve healthy adults participated in the second part of the study (Table 1). The participants had no self-reported history of walking difficulties, dizziness or balance problems, and had no known neuromuscular condition or injury that could affect balance or walking. Informed consent was obtained and the study was conducted in accordance with the Declaration of Helsinki. The study protocol was approved by the Maastricht University Medical Centre medical ethics committee.

2.2. Setup and procedures

The Computer Assisted Rehabilitation Environment Extended (CAREN; Motekforce Link, Amsterdam, The Netherlands), comprised of a dual-belt force plate-instrumented treadmill (Motekforce Link, Amsterdam, The Netherlands; 1000 Hz), a 12-camera motion capture system (100 Hz; Vicon Motion Systems, Oxford, UK) and a virtual environment that provided optic flow, was used for this study. A safety harness connected to an overhead frame was worn by the participants during all measurements. Five retroreflective markers were attached to anatomical landmarks (C7, left and right trochanter and left and right hallux) and were tracked by the motion capture system.

In the first part of the study (18 participants), the measurement sessions began with 60 s familiarisation trials of walking at 0.4 m/s up to 1.8 m/s in 0.2 m/s intervals. After approximately five to ten minutes rest, single two-to-three-minute-long measurements were then conducted at the same speeds. Following these measurements, the stability-normalised walking speed was calculated. To determine the stability-normalised walking speed, the mean anteroposterior MoS (see below) at foot touchdown of the final 10 steps of each walking trial (0.4 m/s to 1.8 m/s) were taken and fitted with a second order polynomial function. For each participant, the speed resulting in MoS of 0.05 m was calculated. Based on our pilot testing, this value would result in walking speeds that would be possible for healthy adults of most ages (Bierbaum et al., 2010, 2011; Süptitz et al., 2013). With certain populations, slower walking speeds would be required 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 an 80% increase in the right treadmill belt speed from the stability-normalised walking speed of the participant with a 3 m/s^2 acceleration, and thereby, we also normalised the magnitude of the perturbation to the already normalised walking speed. The 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 (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 1.3 m/s (average stability-normalised walking speed of the 18

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

	Sex	Age (y)	Height (cm)	Weight (kg)	Leg length (cm)
1.3 m/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

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).

2.3. Data processing

Marker tracks were filtered using a low pass second order Butterworth filter (zero-phase) with a 12 Hz cut-off frequency. Foot touchdown was detected using a combination of force plate (50 N 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 Süptitz et al. (2013), 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).

2.4. 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 1.3 m/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 intervals were used to confirm the similarity of the groups' demographics. Significance was set at $\alpha = 0.05$. When sphericity was violated, a Greenhouse-Geisser correction was applied. Normality of the distributions was assessed with Q-Q plots. Analyses were performed using Prism version 8 for Windows (GraphPad Software Inc., La Jolla, California, USA).

3. Results and discussion

3.1. Stability during unperturbed walking

Walking speed significantly affected the MoS ($F_{[2.547,\ 42.93]} = 1485$, $P < 0.0001$, $\hat{\varepsilon} = 0.3638$; Fig. 1) and Tukey's multiple comparisons tests revealed significant differences for each speed compared to all other speeds ($P < 0.0001$; Fig. 1). These results agree with previous work (Bhatt et al., 2005; Hak et al., 2013; Süptitz et al., 2012). A range of MoS values were observed for each speed (approximately 6–10 cm), even among these healthy participants, confirming some of the issues related to prescribed walking speeds in gait stability research discussed above. The strong relationship between walking speed and MoS also has relevance for clinical studies conducting self-paced gait measurements with an assessment of gait stability. Patients who improve in walking speed may demonstrate a reduction in MoS, which may not be reductions in the stability of the patients' gait *per se*, but simply an artefact of the improved walking speed.

The stability-normalised walking speeds (range from 1.22 m/s to 1.51 m/s with a mean ± SD of 1.3 ± 0.1 m/s) resulted in MoS very close to the desired outcome of 0.05 m (within one SD of the mean MoS for 15 of the 18 participants; Fig. 2A). The stability-normalised walking speed also reduced between-participant variability in MoS (as shown by the group level standard deviations; Fig. 2B). These combined results indicate that the stability-normalisation was successful in reducing between-participant differences in MoS during walking, even in a homogenous group of healthy young adults.

Small, non-significant correlations between the determined stability-normalised walking speeds and the participants' height and leg length were found (Fig. 3). The outcomes of our correlation analysis suggest that height and leg length did not significantly

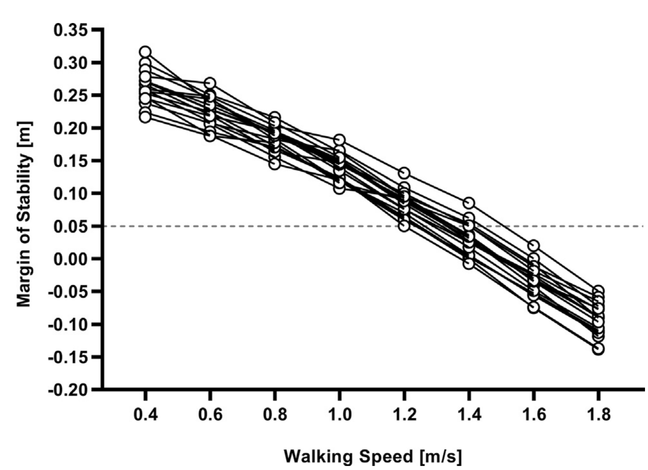

Fig. 1. Individual margins of stability at foot touchdown over the different walking speeds. The dashed line represents the margin of stability used to determine the stability-normalised walking speed.

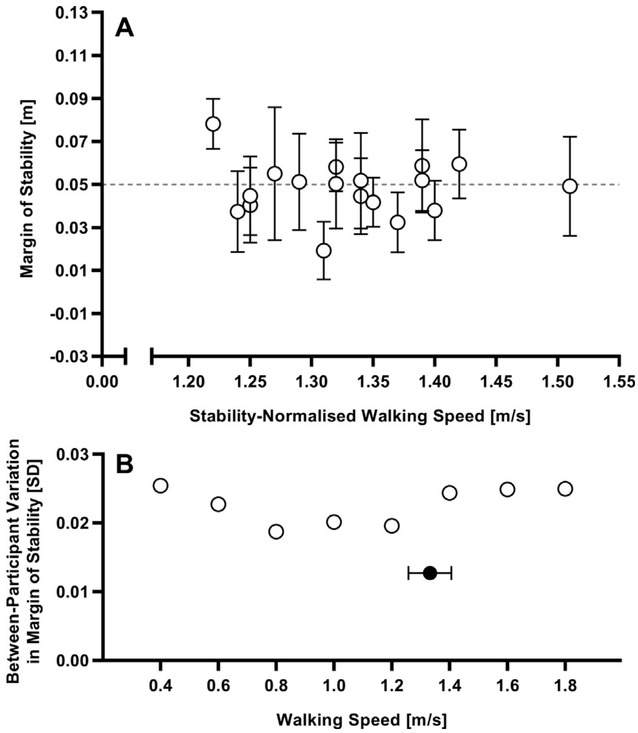

Fig. 2. (A) Means and standard deviations of the margins of stability at touchdown of the final 10 steps at the stability-normalised walking speed for each individual participant. The desired MoS of 0.05 m at foot touchdown is indicated by the dashed line. (B) The between-participant variation in the 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).

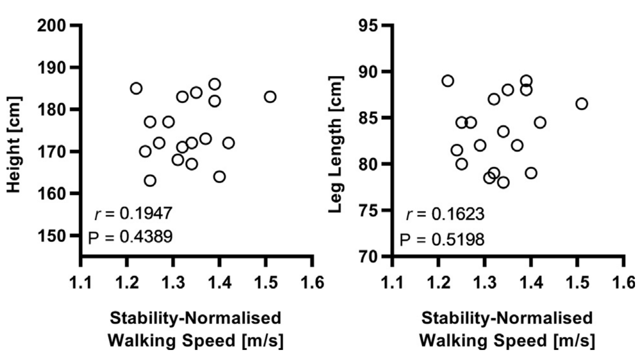

Fig. 3. Pearson correlations between the participants' stability-normalised walking speeds and their height and leg length.

affect the calculation of stability-normalised walking speed, suggesting that a normalisation of walking speed based on body dimensions does not assume equivalent gait stability, at least not when assessed by the MoS concept.

3.2. Stability during perturbed walking

For the second part of the study, the 12 participants were successfully matched to the 12 of the 18 participants from part one of the study (Table 1). During the perturbations, the 1.3 m/s group had a greater range in MoS values during Base, Pre and Post1 (Fig. 4). A two-way repeated measures ANOVA revealed a significant effect of group ($F_{[1,\ 22]}$ = 6.409, P = 0.019), step ($F_{[1.097,\ 24.14]}$ = 8.34, P = 0.0068, $\hat{\varepsilon}$=0.5486) and a significant group (Norm and 1.3 m/s) by step (Base, Pre, Post1) interaction ($F_{[2,\ 44]}$ = 15.4, P < 0.0001) on

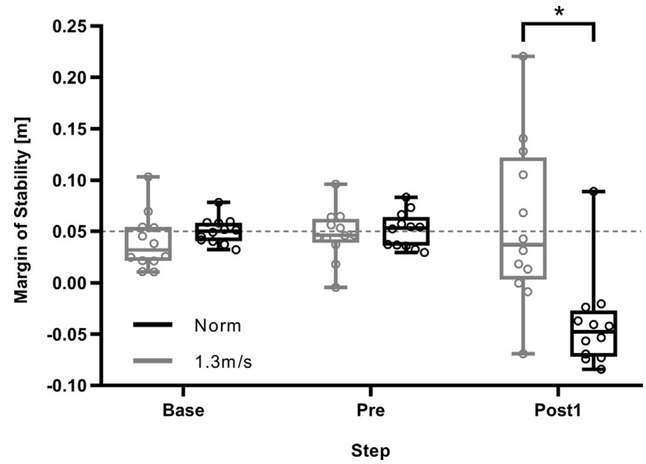

Fig. 4. Margins of stability during unperturbed and perturbed walking of participants walking at their stability-normalised walking speed (Norm) and participants walking at 1.3 m/s. Base: the mean MoS of the eleventh to second last step before each perturbation; Pre: the final step before each perturbation;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

MoS. Sidak post hoc tests revealed a significant difference between Norm and 1.3 m/s groups at Post1 (P = 0.0049). While part of the differences found may be due to chance, the current comparison suggests that the stability-normalised walking speed and the normalised perturbation (acceleration to a peak speed 180% of the walking speed) reduce the inter-individual differences in MoS during both unperturbed and perturbed walking, at least with the current protocol. The significant difference found at Post1 between the groups also aligns with the previous studies reporting different responses to perturbations experienced while walking at different speeds (Bhatt et al., 2005; Krasovsky et al., 2014).

3.3. Further methodological considerations

As the MoS – walking speed relationship from 1.0 to 1.6 m/s appeared to be linear in part one of the study (Fig. 1), a simple linear regression was calculated for 1.0 to 1.6 m/s. A significant regression equation was found (Fig. 5). Future research could use this (or similar) as a simple, efficient method for increasing the dynamic similarity in gait stability across participants, by measuring participants walking at a single speed from 1.0 to 1.6 m/s and

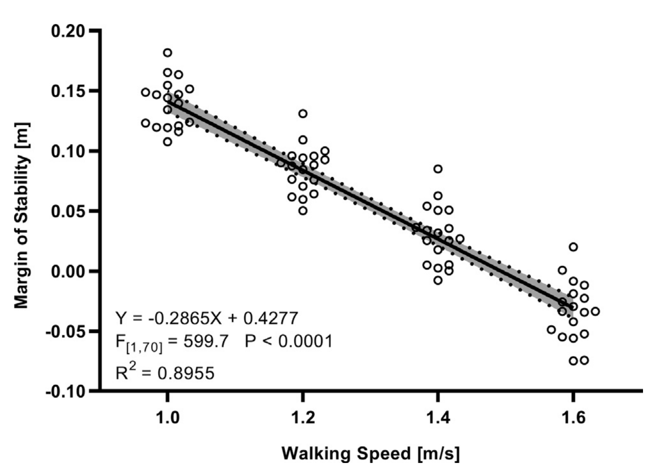

Fig. 5. Margins of stability as a function of walking speed between 1.0 and 1.6 m/s. The shaded area represents the 95% confidence intervals of the regression line.

using this equation to prescribe speeds that would result in similar MoS values. As it is common practice in gait experiments to familiarise participants to the setup and conditions, including some practice walking trials, we would suggest that this may be the ideal opportunity to incorporate our method, without having to conduct any additional trials. It is, however, worth highlighting that the current participants were young healthy adults; the walking speed – MoS relationship may be altered in other populations. Future implementations of this method should consider the capacities of the population of interest and the desired or expected impact on gait stability of the perturbations when selecting an MoS value for normalisation.

3.4. Limitations

Individual responses in the MoS to the perturbation varied (Fig. 4), although the variation was lower in the stability-normalised walking speed group. Part of the reasons for this variation could be the result of uncontrolled factors such as individual physiological, biomechanical or psychological differences affecting the individual response at the onset of the perturbation. It could be argued that using a single trial as opposed to averaging multiple trials is less reliable, however, due to the significant and rapid learning effects following even single perturbations of this kind, the responses seen after averaging trials would no longer accurately represent natural responses to unexpected perturbations. In this sense, our approach is ecologically valid, as the variation is representative of daily life responses to truly unexpected gait perturbations. Another potential limitation relates to a validity constraint of the MoS calculation detailed by Hof et al. (2005), in that the pendulum length (distance from the centre of mass to the axis of rotation) should remain constant. This may not always be the case during dynamic walking and perturbed walking tasks if the knee is slightly flexed at foot contact. However, we have not observed large changes in the pendulum length and small changes are not systematic, as within and between individual variability in responses is large. We therefore believe that this is an acceptable limitation of using the model in this context, but one that should be kept in mind when interpreting the results.

3.5. Conclusions

In conclusion, large ranges in MoS were observed and walking speed significantly affected MoS even within these young healthy participants, confirming some issues related to walking speed choice in gait stability research. The current methods reduced between-participant variability in MoS during both unperturbed and perturbed walking, meaning that the method could be beneficial for gait stability studies comparing groups with different locomotor capacities. An equation has been provided that can be used following a single gait trial to increase the dynamic similarity of gait stability between participants.

Conflict of interest statement

The authors declare no conflict of interest.

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