Effects of vestibular stimulation on control of gait stability

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Abstract

Vestibular information modulates muscle activity during gait, presumably to contribute to stability. If this is the case, stronger effects of perturbing vestibular stimulation on gait stability can be expected for narrow-base walking (which needs more control) than for normal walking and smaller effects for wide-base walking (which needs less control). An important mechanism to stabilize gait in the mediolateral direction is to coordinate foot placement based on a sensory estimate of the center of mass (CoM) state, to which vestibular information appears to contribute. We, therefore, expected that stochastic vestibular stimulation would decrease the correlation between foot placement and CoM state. In fourteen healthy participants, we measured the kinematics of the trunk (as a proxy of CoM), and feet, while they walked on a treadmill in six conditions: control, narrow-base, and wide-base, each with and without stochastic electrical vestibular stimulation (EVS) (peak amplitude of 5 mA; RMS of ~ 1.2 mA; frequency band of 0 to 25 Hz). Stochastic EVS decreased local dynamic stability. Foot placement correlated stronger with trunk motion during walking with EVS in the control and narrow-base conditions. However, residual variance in foot placement was increased with EVS indicating less accurate foot placement, although this effect was not significantly different between step width conditions. Thus, vestibular information contributes to control of gait stability, but this contribution is not modulated by step width.

Introduction

Electrical vestibular stimulation (EVS) has been applied to probe the role of the human vestibular system during standing and walking. The response to this stimulus consists of illusions of body motion, changes in perceived orientation and motor reactions, due to stimulation of the afferents of the semicircular canals and otoliths (Kim and Curthoys 2004; St George and Fitzpatrick 2011). Using a bilateral electrode configuration and a galvanic stimulus, studies showed a shift of the center of pressure (CoP) to the anodal side in standing (Goldberg et al. 1982; Kwan et al. 2019), while during overground walking, the stimulus led to deviations in heading towards the anodal side (Fitzpatrick et al. 1999; Jahn et al. 2000; Bent et al. 2004). Stochastic, zero-mean and rapidly fluctuating EVS is expected not to cause an illusion of movement in one direction, but instead to increase the noise in the afferent signals, thus presenting a perturbation of stability, as was confirmed by its effects on gait stability and variability (van Schooten et al. 2011). Here, we aimed to study how people deal with such increased noise, and how this interacts with the stabilization demands imposed by the task.

Gait stability has been defined as the ability to walk without falling in spite of the intrinsic or environmental perturbations (Bruijn et al. 2013). Gait stability is most challenged in the mediolateral direction (Bauby and Kuo 2000), due to the narrow base of support in this direction, and the movement towards the lateral edge of the base of support during each single-leg stance phase.

Recently, the step-by-step relationship between foot placement and the center of mass (CoM) state has been proposed as a measure of the control of gait stability (Wang and Srinivasan 2014; Bruijn and Van Dieën 2018). The mediolateral CoM state during swing, or closely related pelvis or trunk states, predict > 80% of the variance in mediolateral foot placement at the end of swing, and this measure of predictability of foot placement is gaining popularity as an index of gait stability or the control thereof (Rankin et al. 2014; Wang and Srinivasan 2014; Dean and Kautz 2015; Perry and Srinivasan 2017; Seethapathi and Srinivasan 2019; Mahaki et al. 2019; van Leeuwen et al. 2020). For instance, when the CoM is positioned more medial relative to the stance leg, or when it is moving faster medially than on average during the swing phase, this control will be apparent in a more lateral foot placement than average at the end of this swing phase, and vice versa. This relationship between foot placement and CoM state can be seen as a measure of a discrete control, depending on a

feedback process where the error correction is done when the foot is placed. Vestibular stimulation during gait initiation was shown to affect foot placement, suggesting that it affects the estimate of the CoM movement used for foot placement control (Reimann et al. 2017). Thus, we expected that walking with stochastic EVS, would decrease the degree of foot placement control and consequently reduce local dynamic stability of walking.

The local divergence exponent is an often used measure of local dynamic stability of gait (Bruijn et al. 2013). It represents the rate of divergence between neighboring trajectories in state space over a short time interval, and indicates resistance to small perturbations (Dingwell and Cusumano 2000; Van Emmerik et al. 2016). Local dynamic stability has been shown to be associated with fall risk in several studies (Toebes et al. 2012; Rispens et al. 2014; van Schooten et al. 2016). Stochastic EVS provides erroneous sensory information, and, as such, has been shown to decrease gait stability (i.e. increase the local divergence exponent) (Sloot et al. 2011; van Schooten et al. 2011; Magnani et al. 2021) and to increase trunk movement variability (van Schooten et al. 2011).

Step width adjustments are commonly needed in daily life, to deal with environmental constraints, such as avoiding a puddle or a hole. Moreover, a wider step width is used as a response to impaired lateral stability during locomotion (Kubinski et al. 2015; Aboutorabi et al. 2016). At the same time, previous studies have also shown that local dynamic stability is reduced when walking with wider steps (Young and Dingwell 2012; Magnani et al. 2021). Narrow-base walking, due to the challenge it poses to gait stability, is associated with more tightly controlled CoM motion and foot placement (Arvin et al. 2016; Perry and Srinivasan 2017) and this may suggest increased reliance on sensory, including vestibular information. On the other hand, walking with wide steps requires less tight control and may thus be less affected when vestibular information is disturbed. In a recent study, we showed that the coupling of EVS to muscle activity and to the resulting ground reaction forces was modulated as a function of stabilization demands. In this earlier study, we measured the same step widths as studied here (narrow, normal, and wide-base) and we also measured walking with a device that stabilized walking by means of elastic cords providing conservative forces to maintain the mediolateral position of the body CoM (Magnani et al. 2021). We found that the mediolateral ground reaction force was less coupled to vestibular stimulation during externally stabilized walking than during normal walking. Moreover, when walking with a narrower step width, a stronger coupling between ground reaction force and EVS was

present. Overall, these findings suggests that the importance of vestibular input for control of human walking is dependent on stability demands.

If indeed vestibular contributions to gait depend on stability demands, we expect the destabilizing effects of EVS (Sloot et al. 2011; van Schooten et al. 2011) to also depend on stability demands. For instance, as decreased coupling between EVS and ground reaction force has been shown in wide-base walking, we hypothesized the decrease in stability due to EVS during wide-base walking to be less than during narrow-base walking. Conversely, since narrow-base walking has been shown to increase coupling between EVS and ground reaction forces, we hypothesized a larger decrease in gait stability due to EVS during narrow-base walking.

In the current study, we tested how stochastic EVS affects gait stability and how it affects the regulation of gait stability in normal walking compared to waking with imposed narrow and wide steps. We assessed foot placement, which has been suggested to be the main mechanism to regulate mediolateral stability (Wang and Srinivasan 2014; Arvin et al. 2016) and we measured the effect of EVS on gait stability by means of the local divergence exponent, as done in previous studies (Sloot et al. 2011; van Schooten et al. 2011). In addition, we calculated CoM position variability, since trunk movement variability has previously been shown to increase due to EVS (van Schooten et al. 2011). We hypothesized that stochastic EVS would decrease the correlation between foot placement and CoM state, increase the residual variance in foot placement and increase the local divergence exponent and variability of trunk movement. Moreover, we hypothesized that, given the dependence of the role of vestibular information on stabilization demands, these effects would be stronger in narrow-base walking than in control walking and less strong in wide-base walking.

Methods

Ethics Statement and Participants

We measured 23 healthy young adults between 24 and 33 years recruited from the university campus. Nine participants were excluded during data analysis due to technical problems in the collection of kinematic data in any of the trials recorded, as a consequence of construction work in the lab building. Data of fourteen healthy participants (five female/nine male; mean age 28.42±2.87 years) were included in the analysis. None of the participants reported any auditory, vestibular, neurologic, cardiovascular, or orthopedic disorders, and all had a normal

or corrected-to-normal vision. The participants agreed to participate in the study by signing the informed consent form, and the study was approved by the Ethics Committee of the Faculty of Behavioural and Movement Sciences of the VU Amsterdam (VCWE-2017-158).

Instrumentation

Kinematic data were recorded using a 3D motion analysis system (Optotrak Northern Digital Inc., Waterloo, Ontario, Canada) operating at 100 samples per second. Light emitting diodes (LED) clusters were positioned at the occipital lobe, the spinous process of the sixth thoracic vertebra (T6), the posterior superior iliac spines and bilaterally at the calcaneus. Forces were collected from force plates embedded in the treadmill (Motekforce Link, The Netherlands) sampling at 200 samples per second.

Continuous percutaneous bipolar electrical stimulation was used to modulate the firing rate of the vestibular nerves, using an isolated linear stimulator (STMISOLA, BIOPAC System Inc., USA). When the head is facing forward, galvanic binaural stimulation evokes an illusion of head roll rotational velocity (Peters et al. 2015) about an axis directed posteriorly and superiorly by 18 degrees relative to Reid's plane (Fitzpatrick and Day 2004; St George and Fitzpatrick 2011), and results in a postural response in the frontal plane to compensate for the induced roll error signal (Forbes et al. 2016; Tisserand et al. 2018). Here, we used bandwidth-limited stochastic EVS with a frequency content of 0 to 25 Hz. This stimulus was created using MATLAB software (MathWorks, Natick, MA, USA) (Figure 1) from zero-mean low-pass filtered (25 Hz cutoff, zero lag, fourth-order Butterworth) white noise, and had a peak amplitude of 5 mA and root mean square (RMS) of ~ 1.2 mA. This stimulus does not produce a movement illusion but induces noise in the vestibular afference. All volunteers were exposed to the same stimulus during eight minutes while walking. The stimulation was applied by flexible carbon surface electrodes (about 9 cm^2). The electrodes were coated with Spectra 360 electrode gel (Parker Laboratories, Fairfield, NJ, USA) and affixed with adhesive tape to the clean and dry skin of the participants' mastoid processes, and further stabilized by an elastic headband.

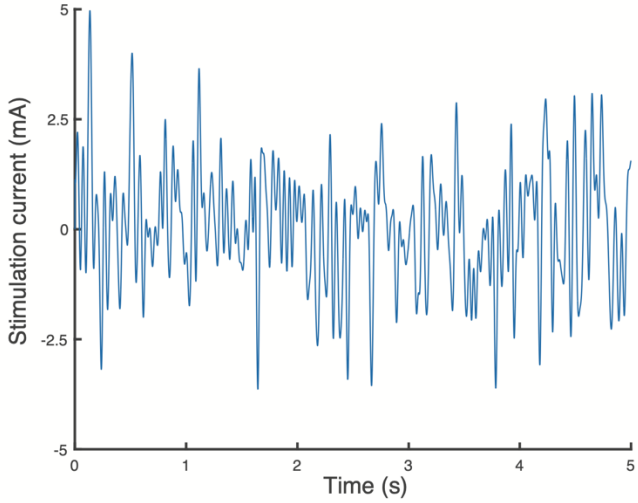

Figure 1: Vestibular stimulation signal.

Protocol

Participants walked on a level treadmill (walking surface width of 102cm and length of 157cm) in six different conditions while wearing their own shoes. All conditions were performed at a fixed speed of 0.8 m/s, and cadence was imposed by the beat of a metronome, which was set at 78 steps/min. We controlled walking speed and cadence since both of these have been shown to influence vestibular contributions to locomotion (Dakin et al. 2013; Forbes et al. 2017). The walking speed of 0.8 m/s and cadence 78 steps/min were chosen to replicate the conditions used by Dakin et al. 2013. The conditions were characterized by step width, here called Condition (control, narrow-base, wide-base) and by the stimulus presence (EVS and no-EVS). For the control condition, participants walked with their usual step width; for narrow-base walking, participants were asked to adopt a smaller base of support (smaller than hip width); for wide-base walking participants were asked to increase the base of support adopting step width greater than hip width. The duration of the EVS conditions was eight minutes, and the no-EVS conditions took one minute. The order of the conditions was randomized for each participant, and participants were given a 2-minute break between conditions.

Data analysis

Gait events were identified from center of pressure data (Roerdink et al. 2008), and a complete gait cycle was defined by the time between heel contact and subsequent heel contact of the same limb. Since the minimum number of strides over conditions and participants was 32, we calculated all gait parameters for 32 strides of each condition. We chose to analyze the last 32 strides, because in case of any habituation or reweighting as a

consequence of EVS, we expected this to be the most stable part. We also performed the same analysis on the first 32 strides of each condition with similar results (results not reported here). The mean and standard deviation of step time (ST) (calculated as the duration between two consecutive heel contacts) and step width (SW) (determined as the mediolateral distance between the heels during heelstrikes) were calculated. Gait stability was evaluated by the local divergence exponent calculated from the T6 mediolateral velocity time series (Dingwell and Cusumano 2000). To this end, the time series was first normalized to 3200 samples, after which a 5-dimensional state space was constructed from this time normalized signal and 4 time-delayed copies. We used a fixed delay of 10 samples, such that each dimension was shifted by 10 samples with respect to the previous one. We used a fixed number of dimensions and a fixed delay, as previous research has suggested that this yields the most sensitive and reliable results (van Schooten et al. 2013). Next, for each point in the state space, the divergence between each datapoint and its nearest neighbor (defined as the point with the smallest Euclidean distance, while having at least half a cycle temporal separation to the original point), was tracked over time. The mean of the log of these curves was then taken as the logarithmic rate of divergence and the local divergence exponent was calculated as the slope to this curve from 0-0.5 strides (Rosenstein et al. 1993; Stenum et al. 2014; Bruijn 2017). Foot placement control was quantified using a regression equation relating mediolateral foot placement to the position and velocity of mediolateral trunk CoM at heelstrike (Wang and Srinivasan 2014; van Leeuwen et al. 2020). The T6 marker was used to approximate the ML CoM position. The ML trunk velocity (VCoM) was calculated as the first derivative of the ML trunk position time-series. We used the following regression equation:

$$\mathbf{FP} = \boldsymbol{\beta 1} \cdot \mathbf{CoM} + \boldsymbol{\beta 2} \cdot \mathbf{VCoM} + \boldsymbol{\varepsilon} \qquad \text{(Equation 1)}$$

where β1 and β2 are the regression coefficients, ε the error (Wang and Srinivasan 2014; Mahaki et al. 2019; Bruijn and Van Leeuwen 2020; van Leeuwen et al. 2020), with all variables detected at heel strike to focus on the outcome of the feedback process, which likely involves information obtained during the complete swing phase of gait. The ratio of predicted foot placement variance to actual foot placement variance at heelstrike (R^2) was used as outcome measure. A higher R^2 represents a stronger coupling between mediolateral trunk state and the subsequent mediolateral foot placement (Wang and Srinivasan 2014). However, as this R^2 is a ratio of the variance in foot placement explained by the model and the total variance in foot placement, R^2 will reflect coupling between trunk state and foot placement but also

the magnitude of the variance in CoM state. This is especially problematic, as stochastic EVS has been shown to increase trunk movement variability. Thus, we also calculated trunk position variability at heelstrike, and the residual variance in foot placement with respect to the predicted foot placement (i.e. mean(|$\boldsymbol{\varepsilon}$|).

Statistical Analysis

The assumption of normality was checked by the Shapiro-Wilk test ($p > 0.05$). The gait parameters, SW, ST and their standard deviations, local divergence exponent and the R^2 and the residual variance of foot placement, and trunk position variability at heelstrike were compared by repeated measures analysis of variance with two factors, Stimulation (EVS – electrical vestibular stimulation and no-EVS) x Condition (control, narrow, wide). A Bonferroni correction was applied for the post hoc tests, which were only performed for narrow vs. control and wide vs. control. Whenever a significant interaction effect was found, we tested whether the change due to EVS was significant between conditions (i.e. we tested the paired difference of EVS-no EVS between conditions). All statistical analyses were performed in Matlab, using $p < 0.05$ as a threshold for significance.

Results

The results of all repeated measures ANOVAs are presented in Table 1 and the results of all post hoc tests are described in Table 2. As expected, step width showed a significant Condition effect, where step width was significantly wider during wide-base than control walking and significantly narrower in narrow-base compared to the control walking ($p<0.01$). Step width showed no significant EVS effect or Condition x EVS interaction. Step width variability showed a significant Condition x EVS interaction, as well as a significant EVS and Condition effects. These effects indicated that step width variability increased when walking with EVS, but less so for the wide-base condition (Figure 2). Moreover, step width variability was smaller in the control condition than in narrow-base walking.

Table 1: ANOVA results. Bold values represent statistically significant results ($p<0.05$). SW: step width; ST: step time; var: variability; LDE: local divergence exponent; R^2: percentage of explained variance of foot placement; EVS: electrical vestibular stimulation.

	EVS	Condition	EVS x Condition
SW (m)	$F(1,13)=1.74$; $p=0.21$; $\eta_p^2=0.12$	**$F(2,26)=49.28$; $p=0.00$; $\eta_p^2=0.79$**	$F(2,26)=2.68$; $p=0.09$; $\eta_p^2=0.17$

SW var (m)	**F(1,13)=15.27; p=0.00; η_p^2=0.54**	**F(2,26)=7.21; p=0.00; η_p^2=0.36**	**F(2,26)=6.00; p=0.01; η_p^2=0.32**
ST (s)	**F(1,13)=6.12; p=0.03; η_p^2=0.32**	**F(2,26)=4.51; p=0.02; η_p^2=0.26**	**F(2,26)=5.06; p=0.01; η_p^2=0.28**
ST var (s)	**F(1,13)=11.24; p=0.01; η_p^2=0.46**	F(2,26)=2.17; p=0.13; η_p^2=0.14	F(2,26)=0.11; p=0.90; η_p^2=0.01
LDE	**F(1,13)=11.34; p=0.01; η_p^2=0.47**	**F(2,26)=19.99; p=0.00; η_p^2=0.61**	F(2,26)=0.02; p=0.98; η_p^2=0.00
R^2	F(1,13)=2.61; p=0.13; η_p^2=0.17	**F(2,26)=12.77; p=0.00; η_p^2=0.50**	**F(2,26)=5.57; p=0.01; η_p^2=0.30**
Residual variance	**F(1,13)=25.76; p=0.00; η_p^2=0.66**	**F(2,26)=5.02; p=0.01; η_p^2=0.28**	**F(2,26)=5.86; p=0.01; η_p^2=0.31**
CoM var	**F(1,13)=33.17; p=0.00; η_p^2=0.72**	**F(2,26)=9.16; p=0.00; η_p^2=0.41**	F(2,26)=2.86; p=0.08; η_p^2=0.18

Table 2: P-values of the post hoc comparisons. Bold values represent statistically significant results (p<0.05). SW: step width; ST: step time; var: variability; LDE: local divergence exponent; R^2: percentage of explained variance of foot placement; EVS: electrical vestibular stimulation. Note: whenever a main effect of the ANOVA was not significant, corresponding post-hoc tests were not performed, hence the empty cells.

Variable	Control-Narrow	Control-Wide	Control*EVS-Narrow*EVS	Control*EVS-Wide*EVS
SW (m)	**0.000**	**0.000**		
SW var (m)	**0.000**	0.307	0.667	**0.040**
ST (s)	0.060	0.667	0.215	0.177
ST var (s)				
LDE	**0.004**	**0.034**		
R^2	**0.001**	0.667	0.105	**0.010**
Residual variance	**0.030**	**0.043**	0.150	0.148
CoM var	**0.001**	0.667		

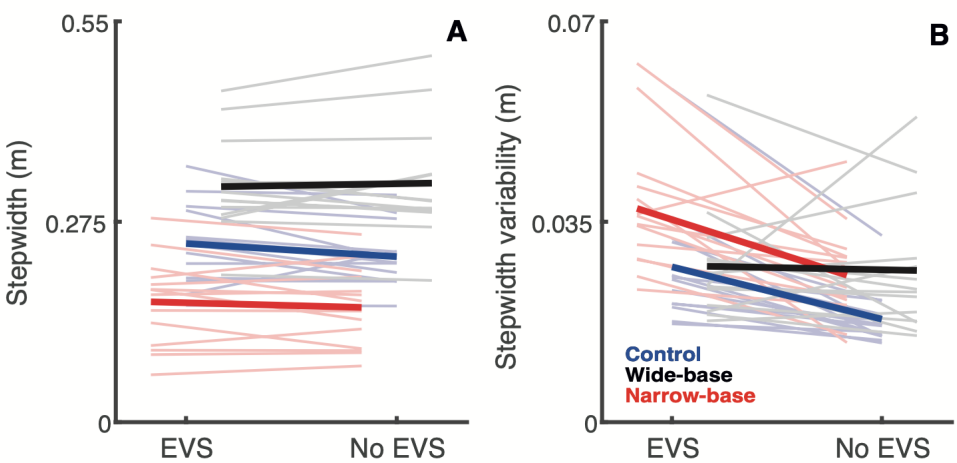

Figure 2: Effects of Condition and EVS on step width (A) and step width variability (B). Thick lines represent mean values and thin lines show individual data for each condition according to the following colors - blue lines: control condition; black lines: wide-base condition; red lines: narrow-base condition. Note: the conditions are jittered on the x-axis.

Step time showed significant EVS and Condition effects, as well as a significant EVS X Condition interaction, but none of the post hoc test was significant (Figure 3A). Step time variability was significantly higher during EVS conditions than during no-EVS conditions, presenting no significant effects of Condition nor an interaction.

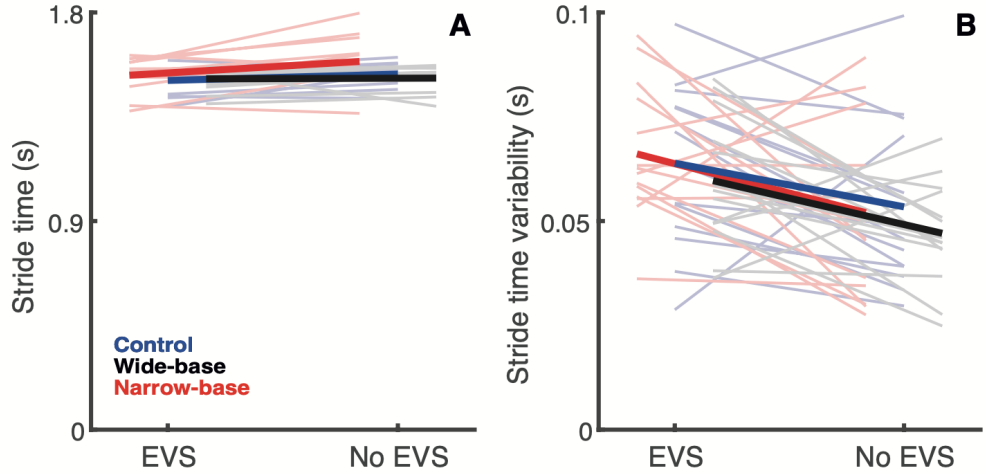

Figure 3: Effects of Condition and EVS on step time (A) and step time variability (B). Thick lines represent mean values and thin lines show individual data for each condition according to the following colors - blue lines: control condition; black lines: wide-base condition; red lines: narrow-base condition. Note: the conditions are jittered on the x-axis.

Local dynamic stability was significantly decreased (i.e. increased LDE values) during EVS conditions. Moreover, there was a significant effect of Condition on LDE, where the post hoc tests indicated that the control condition was less stable than the narrow-base condition and more stable than the wide-base condition. In contrast with our hypothesis, there was no significant interaction effect on LDE (Figure 4).

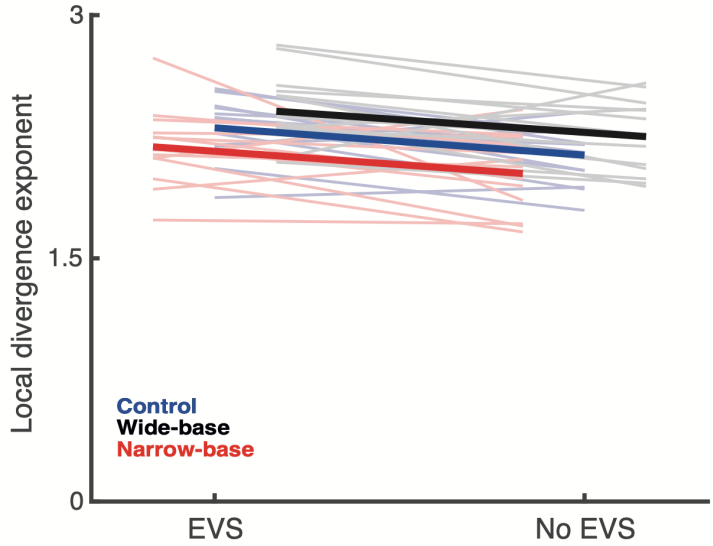

Figure 4: Effects of Condition and EVS on mediolateral local divergence exponent. Thick lines represent mean values and thin lines show individual data for each condition according to the following colors - blue lines: control condition; black lines: wide-base condition; red lines: narrow-base condition. Note: the conditions are jittered on the x-axis.

Contrary to our expectations, R^2 between trunk motion and foot placement was higher during EVS walking, but this effect was not significant. As hypothesized, a significant EVS X Condition interaction indicated that the effect of EVS was smaller for wide-base walking. Finally, the R^2 showed a significant Condition effect, and post hoc tests indicated that it was lower in the control condition compared to narrow-base walking.

The CoM position variability showed significant EVS and Condition effects. As expected, CoM variability increased during the EVS conditions. The post hoc test showed that the control condition was less variable than the narrow-base condition (Figure 6A). The residual variance in foot placement showed a significant Condition x EVS interaction, as well as significant EVS and Condition effects. EVS increased the residual variance (Figure 6B). Moreover, the residual variance in foot placement was higher in the narrow and wide conditions than during control walking. None of the interactions survived post-hoc corrections, but the figure suggested that the residual variance increased more due to EVS in the narrow-base condition (compared to control) and increased less due to EVS in the wide-base condition (again, compared to control).

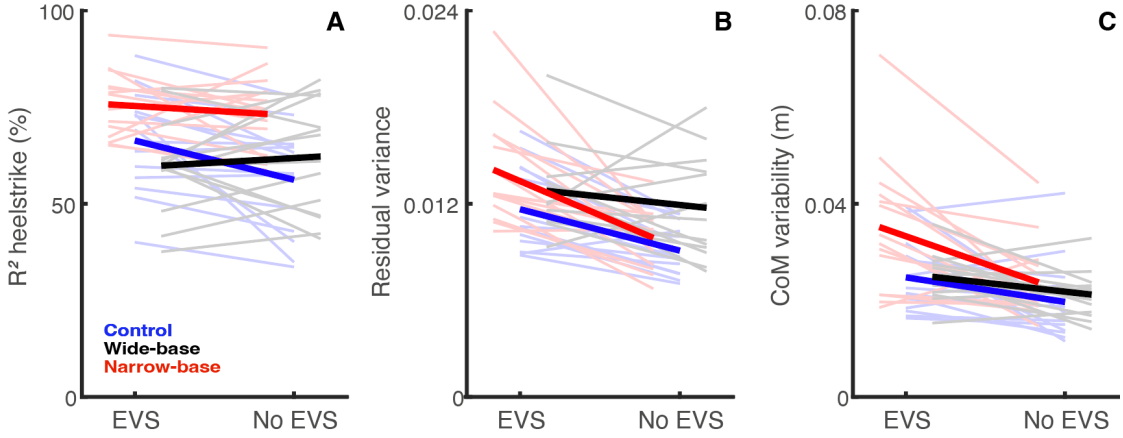

Figure 5: Effects of Condition and EVS on R^2 (percentage of explained variance in foot placement) (A), the residual variance in foot placement (B) and center of mass position variability (C). Thick lines represent mean values and thin lines show individual data for each condition according to the following colors - blue lines: control condition; black lines: wide-base condition; red lines: narrow-base condition. Note: the conditions are jittered on the x-axis.

Discussion

We tested the effects of stochastic EVS on gait stability and its regulation through foot placement in walking with different step widths. As we had previously shown a decreased coupling between EVS and ground reaction forces in wide-base walking compared to control walking, we expected that EVS during wide-base walking would have less effect on gait stability and its regulation than during control walking. Likewise, as we have shown increased coupling between EVS and ground reaction forces in narrow-base walking, we hypothesized a larger effect of EVS on gait stability and its regulation during narrow-base walking than during control walking.

Despite walking to the beat of a metronome in all conditions, step time showed a slight decrease when walking with EVS, while variability of step time and step width increased with EVS for all conditions. Previous studies have also found small and significant changes in stride frequency when participants walked in challenging conditions in spite of the use of a metronome (van Leeuwen et al. 2020; Hoogstad et al. 2022). Walking with EVS led to reduced gait stability during walking at all step widths (as evidenced by a higher LDE), as found previously (Sloot et al. 2011; van Schooten et al. 2011; Magnani et al. 2021). Wide-base

walking was less locally stable, while narrow-base walking was more stable than control walking, as reported before (Young and Dingwell 2012; Magnani et al. 2021).

For the control and narrow-base conditions, R^2 was larger when walking with EVS. This falsifies our hypothesis that there would be a decrease in foot placement control when walking with EVS. However, as in previous studies, we also found that when walking with EVS, the variability of trunk movement increased (van Schooten et al. 2011), which could potentially explain the higher R^2 values. Indeed, the residual variance in foot placement increased when walking with EVS, thus the remaining errors in foot placement were still larger. Interestingly, the results here were more in line with our hypothesis, with larger effects for narrow-base walking, and smaller effects for wide-base walking, although these differences were not statistically significant. Thus, it seems that indeed foot placement control is more affected by EVS during narrow-base walking, but maybe R^2 is not the right metric to show this.

We previously showed an increased coupling of EVS to ground reaction forces during narrow-base walking (as compared to control walking), which is in line with the suggested finding here that the contribution of vestibular input to control of gait stability is larger during narrow-base walking. A tighter foot placement control may be used to maintain stability during narrow-base walking (Perry and Srinivasan 2017). Such a larger R^2 at smaller step width (and lower, albeit non significantly so R^2 at wider step width) might be due to the fact that an increased base of support increases margins of safety, so it would need less foot placement modulation than a narrower base (Perry and Srinivasan 2017; van Leeuwen et al. 2020). Still, we found that residual variance in foot placement during narrow-base walking was larger than during control walking, while local dynamic stability was highest for narrow-base walking. While especially these two latter findings (larger residual variance in foot placement and increased local dynamic stability) may seem contradicting, it should be kept in mind that foot placement is only one mechanism by which stability can be controlled. Other mechanisms include the shifting the center of pressure under the stance foot (Reimann et al. 2018; van Leeuwen et al. 2021) or changing angular momentum around the CoM (Hof 2008; van den Bogaart et al. 2020). The increase in EVS-erector spinae muscle coupling during narrow-base walking that we found previously (Magnani et al. 2021), suggests that indeed angular momentum plays a larger role during narrow-base walking, as do behavioral studies (Bogaart et al. 2021).

This study has some limitations. First of all, our results were obtained during treadmill walking. Mechanically there are no differences between overground and treadmill walking, but step width has been shown to be wider and step width variability was reduced on a treadmill compared to overground (van Ingen Schenau 1980; Rosenblatt and Grabiner 2010). Another limitation is that we used T6 as a proxy of the CoM, and this is a rather high location. Nevertheless, the R^2 between CoM and foot placement was similar as in previous work using a more accurate estimate of the CoM or the pelvis as a proxy (Mahaki et al. 2019; van Leeuwen et al. 2021).

In conclusion, stochastic EVS decreases local dynamic stability of gait, but this effect is not different when walking at different step widths. Foot placement was more coordinated with trunk motion during walking with EVS. However, residual variance in foot placement was also increased with EVS, indicating less accurate foot placement Still, this effect was not significantly different between different step widths. Thus, vestibular information contributes to control of gait stability, but this contribution is not significantly modulated by step width.

Declarations

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Conflicts of interest

The authors declare that they have no competing interests.

Availability of data and material (data transparency)

The data and code for this manuscript can be found at https://surfdrive.surf.nl/files/index.php/s/dUu3uRQyVHlN2g3 and will be published to Zenodo upon acceptance.

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