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Associations between bipedal stance stability and locomotor stability following a trip in unilateral vestibulopathy

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Running Title: Stance and Locomotor Stability in Vestibulopathy

Abstract

Posturography is used to assess balance in clinical settings, but its relationship to gait stability is unclear. We assessed if dynamic gait stability is associated with standing balance in 12 patients with unilateral vestibulopathy. Participants were unexpectedly tripped during treadmill walking and the change in the margin of stability (MoS_{change}) and base of support (BoS_{change}) relative to non-perturbed walking was calculated for the perturbed and first recovery steps. The centre of pressure (COP) path during 30s stance with eyes open and closed, and the distance between the most anterior point of the COP and the anterior BoS boundary during forward leaning (A_{Dist}) were assessed using a force plate. Pearson correlations were conducted between the static and dynamic variables. The perturbation caused a large decrease in the BoS, leading to a decrease in MoS. One of 12 correlations was significant (MoS_{change} at the perturbed step and A_{Dist}; r = -.595, P = .041; non-significant correlations: .068 ≤ P ≤ .995). The results suggest that different control mechanisms may be involved in stance and gait stability, as a consistent relationship was not found. Therefore, posturography may be of limited use in predicting stability in dynamic situations.

Keywords: vestibular, dynamic gait stability, falls, balance, locomotion

Word Count: 2504

Introduction

Posturography assesses balance and postural sway through centre of pressure (COP) measurements during stance and has been used in groups at an increased falls risk, such as lower limb amputees,[1] elderly[2,3] and vestibulopathy patients.[4,5] Contributions of sensory systems to postural control can be estimated by disturbing vision,[6] changing the support surface[7] or via Achilles tendon vibration.[7-9] However, the majority of falls occur during ambulation,[10-14] not static stance, which may be one reason why posturography appears to be limited as a clinical test, rehabilitation tool and falls risk assessment method.[15]

From a mechanical perspective, the vertical projection of the centre of mass (CoM) is within the base of support (BoS) during bipedal stance and is controlled through anticipatory adjustments of the sensory and neuromuscular systems. However, in dynamic settings, the extrapolated CoM is often situated outside of the BoS and the CoM has a velocity and specific direction, and effective reactive postural adjustments (e.g. by increasing BoS) are required to control stability.[16-19] These differences may explain why posturography could not separate fallers from non-fallers in a slip recovery test during gait.[20] Similarly, the maximum recoverable forward lean angle is not generally predicted by static posturography.[21,22] However, such comparisons between static and dynamic tasks have not, to our knowledge, been conducted in subject groups with balance disorders. Vestibulopathy is associated with imbalance, dizziness and falls[23-26] and decreased motor performance[17,27,28] and therefore, it is important to determine if posturography can provide some insight into gait stability issues seen in these patients.

In a recent study of our group, we reported that patients with unilateral peripheral vestibular disorder have a diminished ability to control and adapt their dynamic gait stability following unexpected trip perturbations while walking compared to healthy participants.[17] Additionally, it has been well documented that various posturography methods can discriminate vestibular patients from healthy subjects.[29-32] Given that both static and dynamic methods reveal differences between healthy and vestibulopathy groups, and that posturography can be easily and cheaply conducted, an assessment of the relationship between such tasks is needed to determine if posturography alone is sufficient to estimate dynamic gait stability. To address this, we collated previously collected data from the dynamic gait stability measurements[17] and from posturography measurements conducted with the same patients.[33] An explorative analysis was conducted to determine correlations between dynamic stability control following a trip and COP parameters during a forward leaning task and during quiet standing with the eyes open and closed. Based on previous results demonstrating a lack of relationship between static and dynamic stability tasks,[20-22] we did not expect to find a consistent relationship between the dynamic stability parameters and the COP parameters during quiet standing, but we suspected that the forward lean task may reveal some correlations with the dynamic task due to the fact that the anterior limit of stability is more challenged in this task than during quiet standing.

Methods

For this explorative analysis, we pooled previously collected data of patients with unilateral peripheral vestibular disorder from two previous studies, the first involving a tripping while walking task[17] and the second involving stance posturography tasks.[33] For each parameter of interest (see below) we included all patients with data from each variable. In total, 12 patients were included with age, height and weight of 50.5(5.4) years, 169.7(6.6) cm and 72.5(9.6) kg respectively (means and SD). All patients were assessed for inclusion by an otolaryngologist to confirm their diagnoses. Further inclusion criteria were that participants did not exercise more than once per week and had no other health issues. The studies were approved by the ethical board of the university, the procedures were explained to the participants, and written informed consent was obtained in accordance with the Declaration of Helsinki.

Previous work has reported the effects of repeated trip perturbations on these subjects.[17] Here, we consider only the impact of the first unexpected trip, to exclude the possibility of adaptation influencing the results and to analyse a more ecologically valid response to the trip perturbation.[34] Full details on the trip perturbation device have been reported previously.[17,35,36] Briefly, the tripping task was conducted during treadmill walking at 1.4 $m \cdot s^{-1}$ (pulsar 4.0, h/p/cosmos, Nussdorf-Traunstein, Germany) using a custom built electronically driven magnet system to provide a trip perturbation. The perturbation consisted of a single unilateral resistance of 2.1 kg, applied and removed unexpectedly to the right leg during the swing phase via a Teflon cable and ankle strap. Participants wore a safety harness connected to an overhead track during all trip recovery and posturography trials. Four to seven days before the measurement session, all participants took part in a treadmill walking familiarisation session lasting approximately 30 minutes, to become accustomed to the treadmill walking conditions. On the day of the measurement, the session began with five minutes of walking to ensure participants were comfortable on the treadmill. The ankle strap was then attached to the right leg and participants walked for another four minutes in order to establish a baseline (about 20 seconds was recorded towards the end of this period to be used as a non-perturbed walking baseline). Directly following the baseline period, the perturbation was applied for the entire duration of the swing phase and was subsequently removed. Participants were not given a warning about the upcoming perturbation. An example of a typical recovery response to the perturbation from one participant can be seen in Fig. 1.

Insert Fig. 1

In order to examine dynamic gait stability, we tracked a twelve-segment, full kinematic model using 26 reflective markers (radius 16 mm) recorded by an eight camera

(120 Hz) Vicon Nexus motion capture system. Segmental masses and locations were calculated based on the data of Dempster et al..[37] The margin of stability (MoS) in the anteroposterior direction was calculated, as defined by Hof et al.[38] (see Fig. 2), as the difference between the BoS anterior boundary (anteroposterior position of the toe marker) and the extrapolated CoM at the instant of foot touchdown (determined using tibia accelerometer data (ADXL250; Analog Devices, Norwood, MA, USA)) during baseline non-perturbed walking, and at touchdown of the perturbed step (PERT) and the first recovery step following the perturbation ($POST_1$). The extrapolated CoM was defined as follows:

$$\text{Extrapolated CoM} = P_{CoM} + \frac{(V_{CoM} + |V_{BoS}|)}{\sqrt{g \cdot L^{-1}}}$$

where P_{CoM} is the horizontal (anteroposterior) component of the projection of the CoM to the ground, V_{CoM} is the horizontal velocity of the CoM, V_{BoS} is the average horizontal velocity of the foot markers during stance (approximately the treadmill belt speed), g is gravitational acceleration and L is the pendulum length (the distance between the CoM and the centre of the ankle joint in the sagittal plane). We focussed here on these two steps as our previous work demonstrated the importance of the perturbed and first recovery step when recovering stability following such perturbations.[17] Baseline values for MoS and BoS (BoS defined as the anteroposterior distance between the left and right toe markers) were calculated at foot touchdown by averaging 12 consecutive steps of non-perturbed walking. In order to account for individual differences in walking stability, the change in the MoS and BoS relative to baseline non-perturbed walking at PERT and $POST_1$ was used for this study (MoS_{change} and BoS_{change} respectively), where negative MoS_{change} and BoS_{change} values represent lower stability and smaller BoS respectively relative to baseline non-perturbed walking.

Insert Fig. 2

Our previous study of stance stability assessed many variables from different sensory conditions in these patents.[33] In the current study, we include three variables and two tasks that are conducted in clinical settings and provide information on general stance stability with and without visual sensory information, and stability control near the anterior limit of stability. Participants stood on a custom made strain gauge force plate which was used to measure (at 1000 Hz) the position of the COP during forward leaning and upright standing tasks. Participants stood barefoot with their feet at pelvic width and with their heels on a marked line on the platform. The positions of both feet were marked with a line on the force plate in order to transform the coordinates of the anterior and posterior boundaries of the BoS into the coordinate system of the force plate. In this way, the position of the COP could be calculated in relation to the boundaries of the BoS. For the leaning task, participants were instructed to lean as far forward as possible without moving joints other than the ankles. The task was repeated three times, with the trial showing the least difference between the most anterior position of the COP under the feet and the anterior boundary of the BoS (the line connecting left and right metatarsal five) being used for each subject (A_{Dist}). Participants were then asked to stand as still as possible on the platform for three trials, under both eyes open and eyes closed conditions each with a time frame of 30 seconds. For the eyes closed condition, participants wore blackout glasses (custom made) to ensure that there was no visual sensory input during this condition. A Hamming low-pass filter with a cut off frequency of 5 Hz was used to remove high frequency noise and eliminate sampling error. Postural stability was assessed by the total excursion distance of the COP (COP_{Path}) over the 30 seconds analysis window. The average values of the COP parameters from the three trials for each participant were used in the analysis.

Pearson correlations were used to analyse the relationships between the posturography measures (A_{Dist}, eyes open and eyes closed COP_{Path}) and MoS and BoS values of the trip recovery task. 12 and eight participants' data were included for the A_{Dist} and COP_{Path} correlation analyses respectively. The level of significance for all tests was set at $\alpha = .05$. The distribution normality of the results was checked prior to applying statistical analysis using the Shapiro-Wilk test, which revealed normal distributions for all parameters (P > .05). GraphPad Prism version 7.00 software (GraphPad Software Inc., La Jolla, California, USA) was used for the statistical analysis. All results are presented as mean and standard deviation.

Results

The perturbation resulted in large changes in both the BoS and MoS. Changes in BoS and MoS relative to baseline at touchdown of the perturbed step and first recovery step are presented in Fig. 3. The perturbation caused a large decrease in the BoS at touchdown of the perturbed step, leading to a decrease in MoS (Fig. 3). A larger step was then taken in an attempt to control stability (see BoS at $POST_1$ in Fig. 3) but due to the forward velocity induced by the trip, the MoS did not return to baseline level (Fig. 3).

Consistent correlations between the posturography and dynamic stability parameters were not found. The three posturography tasks yielded results of 5.96(1.6) cm, 21.17(5.87) cm and 30.98(9.54) cm for A_{Dist}, eyes open and eyes closed COP_{Path} respectively. The correlation analyses revealed a significant negative correlation between A_{Dist} and $MoS_{change}PERT$ (r = -.595, P = .041; Fig. 4). The other 11 correlation coefficients were not significant (see all r and P values in Fig. 4).

Insert Fig. 3 and Fig. 4

Discussion

The current study aimed to determine if balance maintenance during quiet stance and dynamic gait stability recovery performance were related in patients with unilateral peripheral vestibular disorder. Only one significant correlation was found out of 12 (Fig. 4), suggesting that performance during static stability tasks is not closely related with stability in dynamic situations. This may be particularly relevant for clinical settings where assessment of static stance, but not dynamic gait stability is conducted in patient groups at an increased falls risk.

The significant negative correlation between A_{Dist} and MoS_{change}PERT means that, in these participants, the ability to bring the COP closer to the anterior boundary of the BoS during forward leaning was associated with a less negative MoS at touchdown of the tripped step during walking. It could be speculated that a more anterior limit of stability was the underlying mechanism for this finding, as this could facilitate both a more anterior COP position during leaning, as well as the ability to apply force to the ground more anteriorly following the perturbation. Similarly, this could also be related to the ability to control CoM velocity in the anterior direction, although this is perhaps less likely, due to the large difference in movement speed of the tasks. In either case, this result suggests that such an anterior leaning task may have some value in assessing the ability to control stability in the anterior direction. However, given the lack of significant correlations in general, the use of posturography tasks for the purpose of estimating stability in dynamic settings is not well supported by our results. This agrees with previous studies in other populations using forward lean and release or slip perturbations.[20-22] That being said, one limitation of this study was that it was not powered to test a null hypothesis such as that no correlations between the tasks would be found and therefore, it may be that with a larger sample size, more significant correlations could have been detected. We do not think that this is likely, however, as when we included age, height and weight matched healthy subjects in the analysis (data not shown), thereby artificially increasing the variation of performance, this did not greatly change the results.

The lack of association between the posturography and trip recovery outcome measures may be due to differences in the governing control strategies and mechanisms of stability associated with the tasks. Posturography during quiet stance assesses the ability to keep the vertical projection of the body's CoM within the BoS, principally by using anticipatory adjustments. In contrast, the ability to regain dynamic stability after tripping or slipping where the extrapolated CoM is located outside of the BoS is governed principally by reactive postural adjustments.[18,19] Here, a key factor in preventing a fall is the ability to take a large recovery step to lengthen the BoS and increase the MoS.[16-18]

It is important to note, that while the patients all had unilateral vestibulopathy, the degree of vestibular function remaining varied, and this information was not available for all patients. That being said, there were no significant outliers among the patients in our results, suggesting that while the vestibular function may have varied between patients, the overall impact on stability control was reasonably consistent. This was not a concern for our results, as we treated these subjects as a generalised group with balance disorders that should be distinct from healthy subjects in terms of stability and balance control.

In conclusion, no consistently significant relationship between posturography and the trip recovery task measures was found, indicating that different mechanisms of postural control appear to be involved in our static and dynamic stability tasks. Balance maintenance during quiet stance alone may be of limited use in predicting dynamic stability during perturbed walking. We therefore recommend that task specificity should be considered in clinical and research settings regarding stability and falls risk assessment. Future research should aim to relate laboratory-induced gait perturbation outcomes with real life falls.

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Figure Captions

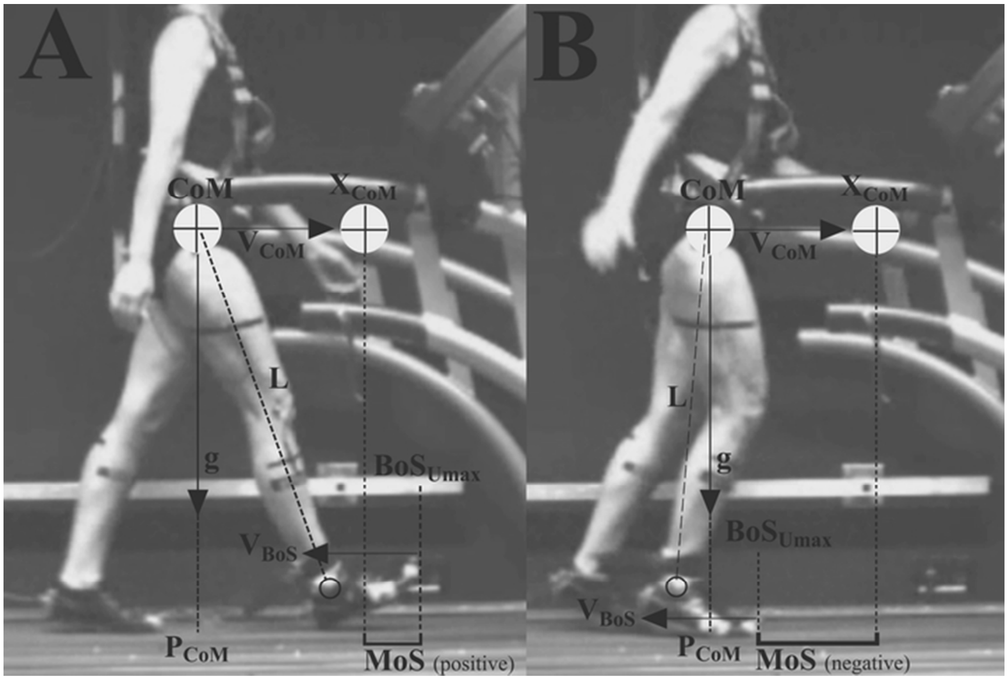
Figure 1 - Example of a typical recovery response to the trip perturbation in one participant. The perturbation adds resistance to the swing phase of the right leg, leading to a reduction in the base of support at foot touchdown. This causes a more anterior position and higher velocity of the centre of mass at touchdown, relative to non-perturbed walking. In response, an increased base of support is required in the following recovery step to maintain gait stability.

Figure 2 - Schematic diagram of the inverted pendulum model during locomotion.[38] P_{CoM} represents the horizontal (anterior-posterior) component of the projection of the center of mass (CoM) to the ground, V_{CoM} is the anterior-posterior velocity of the CoM, V_{BoS} is the average horizontal velocity of the foot markers during stance (approximately the treadmill belt speed), g is acceleration due to gravity and L is the pendulum length (i.e., distance between the CoM and the centre of the ankle joint in the sagittal plane). Margin of stability (MoS) in the anterior direction is calculated at foot touchdown as the difference between the anterior boundary of the base of support (BoS_{Umax}) and the extrapolated centre of mass (X_{CoM}). A stable body configuration is indicated by positive MoS values (A), whereas an unstable body configuration is indicated by negative margin of stability values (B), where additional motor actions, such as stepping, are required to preserve stability and to avoid a fall.

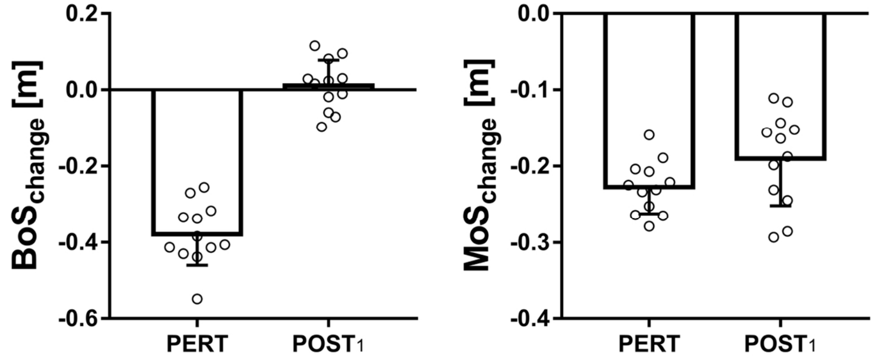
Figure 3 - Change relative to baseline non-perturbed walking in base of support (BoS) and margin of stability (MoS) at touchdown of the perturbed step (PERT) and the first recovery step ($POST_1$) for 12 patients with unilateral vestibulopathy (mean, SD and individual data points).

Figure 4 - Pearson correlation coefficients (*r*) between the static and dynamic stability tasks. 12 patients with unilateral vestibulopathy were included for the A_{Dist} correlations and eight patients were included for the eyes open (EO) and eyes closed (EC) COP_{Path} correlations. $BoS_{change}PERT$ and $BoS_{change}POST_1$: Change in the base of support relative to baseline non-perturbed walking at touchdown of the perturbed and first recovery steps respectively. $MoS_{change}PERT$ and $MoS_{change}POST_1$: Change in the margin of stability relative to baseline non-perturbed walking at touchdown of the perturbed and first recovery steps respectively. A_{Dist}: Distance between the most anterior point of the COP during the forward leaning task and the anterior boundary of the base of support (the line connecting left and right metatarsal five). EO and EC COP_{Path}: total path length of the centre of pressure trajectory during 30s of quiet stance with eyes open and closed respectively.

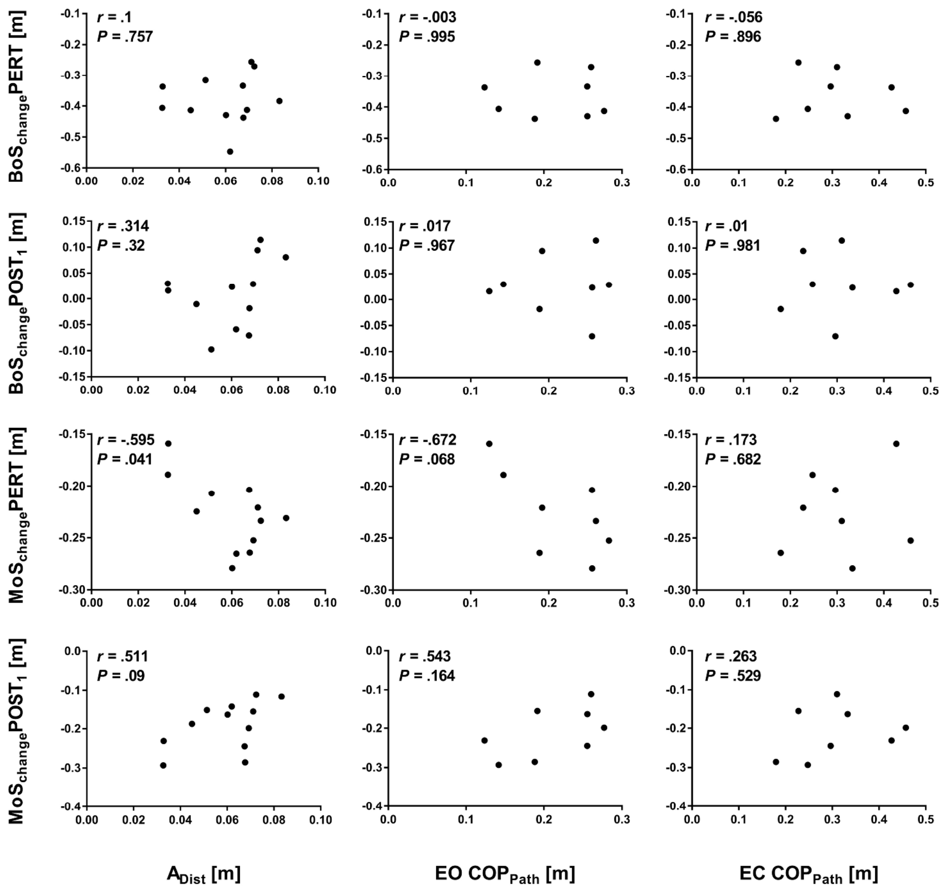
Example of a typical recovery response to the trip perturbation in one participant. The perturbation adds resistance to the swing phase of the right leg, leading to a reduction in the base of support at foot touchdown. This causes a more anterior position and higher velocity of the centre of mass at touchdown, relative to non-perturbed walking. In response, an increased base of support is required in the following recovery step to maintain gait stability.
Fig. 1
28x5mm (300 x 300 DPI)



Schematic diagram of the inverted pendulum model during locomotion.38 PCoM represents the horizontal (anterior-posterior) component of the projection of the center of mass (CoM) to the ground, VCoM is the anterior-posterior velocity of the CoM, VBoS is the average horizontal velocity of the foot markers during stance (approximately the treadmill belt speed), g is acceleration due to gravity and L is the pendulum length (i.e., distance between the CoM and the centre of the ankle joint in the sagittal plane). Margin of stability (MoS) in the anterior direction is calculated at foot touchdown as the difference between the anterior boundary of the base of support (BoSUmax) and the extrapolated centre of mass (XCoM). A stable body configuration is indicated by positive MoS values (A), whereas an unstable body configuration is indicated by negative margin of stability values (B), where additional motor actions, such as stepping, are required to preserve stability and to avoid a fall.
Fig. 2
60x40mm (300 x 300 DPI)



Change relative to baseline non-perturbed walking in base of support (BoS) and margin of stability (MoS) at touchdown of the perturbed step (PERT) and the first recovery step (POST1) for 12 patients with unilateral vestibulopathy (mean, SD and individual data points).
Fig. 3
68x30mm (600 x 600 DPI)



Pearson correlation coefficients (r) between the static and dynamic stability tasks. 12 patients with unilateral vestibulopathy were included for the ADist correlations and eight patients were included for the EO and EC COPPath correlations. BoSchangePERT and BoSchangePOST1: Change in the base of support relative to baseline non-perturbed walking at touchdown of the perturbed and first recovery steps respectively. MoSchangePERT and MoSchangePOST1: Change in the margin of stability relative to baseline non-perturbed walking at touchdown of the perturbed and first recovery steps respectively. ADist: Distance between the most anterior point of the COP during the forward leaning task and the anterior boundary of the base of support (the line connecting left and right metatarsal five). EO and EC COPPath: total path length of the centre of pressure trajectory during 30s of quiet stance with eyes open and closed respectively.

Fig. 4

142x132mm (600 x 600 DPI)