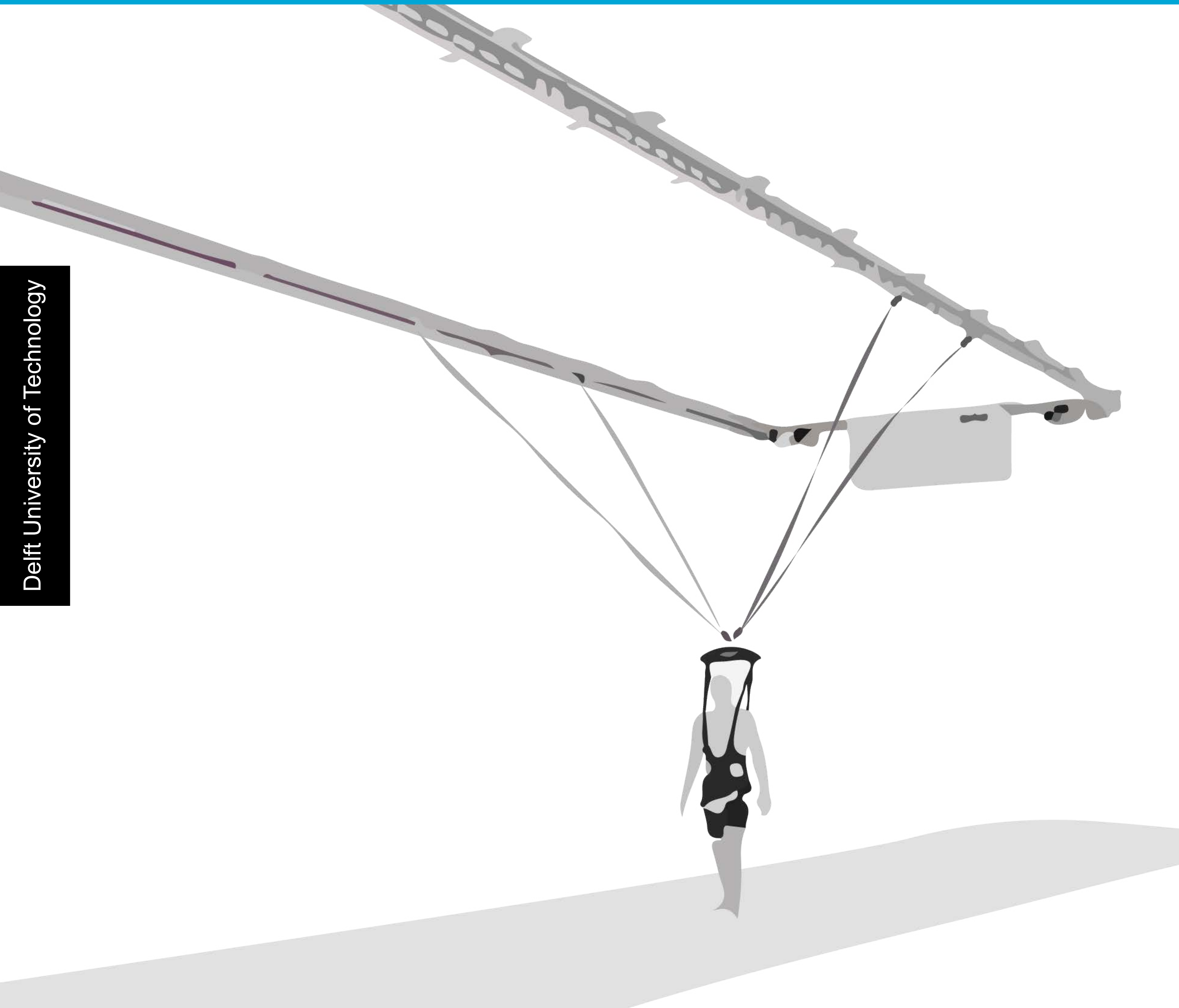
Mediolateral balance during gait in 2D and 3D bodyweight support systems

Delft University of Technology



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TUDelft

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PREFACE

I am happy and proud to finally present this thesis to conclude my master Biomechanical Design. It hasn't been an easy year. Writing your thesis is a challenging project for anyone, and COVID-19 has not made this task any simpler. But I kept myself motivated with the mantra, "If it's hard, that means I'm learning something," and I have sure learned a lot. I would like to sincerely thank all the people who helped me learn so much and made sure I had a great year.

Firstly, I would like to thank my supervisors, who shared so much of their enthusiasm and knowledge with me. Heike, thank you for introducing me to this topic that I still find very interesting. I really appreciated the weekly meetings we had together. Your critical questions motivated me to work hard and have definitely improved all of my work. You also made sure to involve many different people in our meetings, making them very dynamic and fun.

Patrick, thank you for joining this TU Delft project from a whole different department and university. You introduced a lot of new knowledge and enthusiasm to the group. Thank you for being so involved, letting me come over to the lab in Rotterdam, and teaching me how to legally electrocute people, of course. You have helped me on many occasions where I was stuck and always motivated me.

Marco and Shannon, thank you for joining the weekly meetings. Your feedback was always greatly appreciated, and I enjoyed learning about your projects as well. Riding a bicycle on a treadmill at full speed while being pushed off is an experience I certainly will never forget.

Finally, a huge thank you to all my friends and family who have supported me throughout my education. I want to thank my boyfriend, Arden, for listening to my problems, putting them in perspective, making me laugh about them, and overall just having a lot more faith in me than I had in myself sometimes. Thank you to my friends from Proteus-Eretes for all the fun times (even if they had to be online); you made my time as a student unforgettable. I'm very grateful for my family and wonderful friends from Utrecht, who were always a great support to me and know me through and through. And a very special thank you to the two people I spent the most time with the past year: Jitske, thank you for being a great roommate and friend, giving me a pep talk whenever I needed it, and taking care of me; my studies wouldn't have been the same without you the past six years! And Didi, thank you for making your home into our shared office. I really enjoyed all the coffee breaks and lunch walks, and being able to share our problems and motivate each other. And lastly, thank you to everyone who participated in my experiment. Without you, I literally wouldn't have been able to write this thesis and I had a lot of fun during those two weeks of experiments.

Suzanne Rademaker
Delft, February 2022

CONTENTS

Differences in the mediolateral control of balance during gait between 2D and 3D bodyweight support systems

Suzanne L. J. Rademaker

Abstract—**Training with bodyweight support (BWS) systems can improve the likelihood of regaining normal locomotor abilities for neurologically impaired patients. It is known that people alter their gait parameters when walking with BWS. However, it is unclear whether 2D (vertical and lateral support) and 3D (only vertical support) BWS systems affect these gait parameters differently. In this study, participants walked overground in both a 2D and a 3D BWS system to investigate the effects of this lateral support. To compare the contribution of the vestibular system between the different BWS systems, participants received galvanic vestibular stimulation (GVS). Motion capture and force plates were used to find the coupling between the GVS stimulus and the mediolateral ground reaction forces and to calculate the gait parameters. Differences in gait parameters were observed between the 2D and the 3D system. Compared to unsupported gait, participants increased their step width variability by $\sim$10% in the 3D system. Contrarily, participants decreased step width variability by more than 15% in the 2D system. Mean step width decreased slightly in only the 3D system. The margin of stability did not change significantly in any condition. The coupling between the GVS signal and mediolateral ground reaction forces decreased in the 2D and 3D systems compared to unsupported gait, but no significant differences were observed between different BWS conditions. These results suggest that 2D and 3D BWS systems influence gait parameters differently and that they influence the contribution of the vestibular system to balance, but no significant differences between the systems can be observed in this aspect.**

I. Introduction

Sensorimotor disorders that impair our ability to walk directly influence the quality of life [1], [2]. While recovering locomotor function has proven to be difficult, the likelihood of regaining ambulatory function can be improved by task-specific training [3]. For example, providing body weight support (BWS) that is progressively removed through training can help neurologically impaired patients to regain normal locomotor abilities [4], [5]. Some BWS systems (ZeroG [6], Vector, SafeGait) use a ceiling-mounted rail, allowing two degrees of freedom for the patient (2D system): in the vertical (VT) and anteroposterior (AP) direction. The patient experiences ML forces from the system in such a system, pulling them towards the center. These forces may reduce the demands on the patient to stabilize actively during locomotion [7]. More advanced BWS systems (FLOAT [8], RYSEN [9]) allow for mediolateral (ML) movement, creating three degrees of freedom for the patient (3D system). Such systems can come closer to rendering purely vertical forces. While human gait is believed to be passively stable in the AP direction [10], gait has to be controlled actively in the ML direction [11]. If lateral stabilizing forces are present in the system, there is no need to control the ML balance actively [12]. The added degree of freedom in 3D systems makes balance training more challenging and realistic [3]. While the effect of BWS on locomotion characteristics and AP balance of walking have been well-studied [7], [13]–[17], the influence of 2D versus 3D systems and BWS on ML balance, in general, remain largely unknown. To quantify these differences in ML balance, we used mediolateral balance-related gait parameters. We employed step width and the lateral margin of stability (MoS) in this study, as they are most used in ML balance studies. We looked at variabilities as well as the mean values since variability measures are sensitive to changes in stability and have proven to be a good indicator for stability [18].

To maintain a stable gait, humans rely on the integrated input from their sensory systems. The nervous system constantly integrates and modulates information from the visual, somatosensory, and vestibular systems during gait [19]. The vestibular system generates information on the movement of the head [20]. It is thought to contribute to mediolateral stability by influencing the step width [21], [22]. However, in the anteroposterior direction, the vestibular contribution to balance during gait is almost zero [23]. The need for vestibular feedback may thus also be reduced when a person is stabilized laterally [24].

Galvanic vestibular stimulation (GVS) can be applied to investigate the contributions of the vestibular system during gait. A small current is sent through electrodes placed on the mastoid processes to evoke vestibular reflexes. Essentially, when GVS is applied to a person while standing up and looking forward, they start to sway [25]. When applying GVS during gait, the stimulation may influence mediolateral balance-related gait parameters such as step width and ML trunk movement [26]–[28].

This study aimed to assess the influence of 2D and 3D BWS systems on the mediolateral control of balance during walking. To answer this question, healthy participants walked in both a simple passive 2D BWS system and a 3D BWS system, the RYSEN (Motek Medical BV, Houten, The Netherlands). The mediolateral control of balance was measured by the contribution of the vestibular system to ML balance. We applied GVS to participants while walking in the different systems and calculated the coupling between the stimulus and their ML ground reaction forces to measure this contribution. We hypothesized that if the 2D system does indeed give more lateral stabilization, then participants will elicit a smaller response to the GVS compared to the 3D system. This stabilization effect would yield a lower coupling between the GVS and the ML ground reaction forces for the 2D system than the 3D system. Secondly, we measured the changes in balance-related gait parameters between the BWS systems. If the 2D BWS system gives more lateral stabilization than the 3D system, the gait parameters should decrease in the 2D system compared to 3D BWS.

II. METHODS

A. Participants

Fifteen healthy participants (7 male, 8 female, age 24 (2) years, height 1.79 (0.08) m, weight 72.9 (10.0) kg; (mean (SD)) took part in the study. The study was approved by the TU Delft Human Research Ethics Committee and conformed to the Declaration of Helsinki. All participants provided informed written consent prior to participation in the experiment. A copy of the consent form can be found in Appendix A.

After the data was analyzed, it was discovered that one participant received a lower amount of BWS than intended in the GVSP25 condition. They received 16.7% BWS, which was decided to be too low compared to the average of 22.7% for the other participants in this condition. Therefore, this participant was excluded, and the data of only 14 participants were used for the results. More information and an overview of the results with this participant included can be found in Appendix B.

B. Instrumentation

The experiments were performed in the BioMechaMotion Lab at the Mechanical, Maritime and Materials Engineering faculty at the Delft University of Technology. Participants walked overground, and ground reaction forces in three directions were recorded by five force plates (Kistler B.V., Eemnes, The Netherlands) at a frequency of 500 Hz. The force plates were embedded in the center of a 6 m walkway and provided a measurable distance of 2.4 m. A visual representation can be found in figure 1a. The kinematics of the participants were recorded with a 12 camera motion capture system (Qualisys, Götenburg, Sweden) at a 100 Hz frame rate. Twenty reflective markers were placed on the lower body according to the IOR (Istituto Ortopedico Rizzoli) lower body model [29], [30]. The exact placement of the markers can be observed in Appendix C, figure A4.

2D and 3D body weight support systems were both used for the experiments in this study. The 2D system was comprised of a cable and spring mounted on a linear rail, which provided support in the vertical and mediolateral direction and near-unimpeded movement in the anteroposterior direction. The system with the spring was selected to have a minimal influence on gait dynamics and be as close as possible to unsupported gait [31]. Participants wore a harness that was attached to the rope with a slingbar (Direct Healthcare Group, Caerphilly, UK). This system cannot actively track the participant and passively follows the participant through rails on the ceiling. The 3D system was the RYSEN, which provides bodyweight support in the vertical direction and near-unimpeded movement in the anteroposterior and mediolateral directions. It was comprised of cables attached to a harness worn by the participants to provide bodyweight support. The system can track the participant's position and follow them through the workspace in three dimensions. In order to quantify the force tracking performance, the reference and rendered forces were recorded using D-Flow (Medical BV, Houten, The Netherlands). A schematic representation of the experimental setup and a representation of some secondary outcome measures can be found in figure 1.

C. Galvanic vestibular stimulation

We used a continuous galvanic vestibular stimulus (GVS) to deliver an isolated vestibular disturbance to subjects during all GVS trials. We calculated the coupling between the GVS stimulus and ML ground reaction forces over each stride to determine the magnitude and timing of the vestibular contribution to the whole-body responses. The GVS stimulus modulates the afferent firing rate of the semicircular canal and otolith afferents [32]–[34]. While looking straight ahead and if the GVS is delivered in binaural-bipolar configuration, it induces a sensation of head roll rotational velocity [35] about an axis directed posteriorly and superiorly by 18° relative to the Reid's plane [25], [36]. This results in swaying mediolaterally to compensate for the induced roll error signal [23], [37].

To apply the GVS stimulus, we used flexible carbon rubber electrodes of $9\,\mathrm{cm}^2$, which were coated with Spectra 360 electrode gel (Parker Laboratories, Fairfield, NJ, USA). Tape and an elastic headband were used to attach the electrodes securely to the mastoid processes. The stimulus was then sent as an analog signal through a data acquisition device (National Instruments Corp., Austin, TX, USA) to an isolated constant current stimulator (STMISOLA, Biopac, Goleta, CA, USA). All participants received the same stimulus, which was a stochastic GVS signal designed with a limited bandwidth of 0 to 25 Hz, zero-mean low-pass filtered white noise, 25 Hz, zero lag, fourth-order Butterworth, peak amplitude of 4.5 mA, root mean square of 1.3 mA, lasting 30 min and created with Matlab software (MathWorks, Natick, MA, USA). This particular bandwidth was selected to evoke responses across the entire frequency bandwidth of vestibular-evoked responses during gait [38], and the amplitude was selected to evoke a measurable response in ML ground reaction forces but not be uncomfortable for the participants.

D. Protocol

The experiment consisted of six conditions (see table I). The first condition was normal walking (NW), where the participants were not attached to a BWS system and received no perturbations. During the second condition, participants walked normally in the RYSEN system with 25% BWS (NR25) without any perturbations.

In the subsequent four conditions, participants were perturbed with GVS while walking without any BWS (GVSW), receiving 25% body weight support in either the RYSEN (GVSR25) or the passive 2D system (GVSP25), or receiving 10% body weight support in the RYSEN system (GVSR10).

TABLE I: The six experimental conditions were performed by participants in a randomized order.

Condition	BWS	Perturbations
NW	None	None
NR25	RYSEN 25%	None
GVSW	None	GVS
GVSR10	RYSEN 10%	GVS
GVSR25	RYSEN 25%	GVS
GVSP25	Passive system 25%	GVS

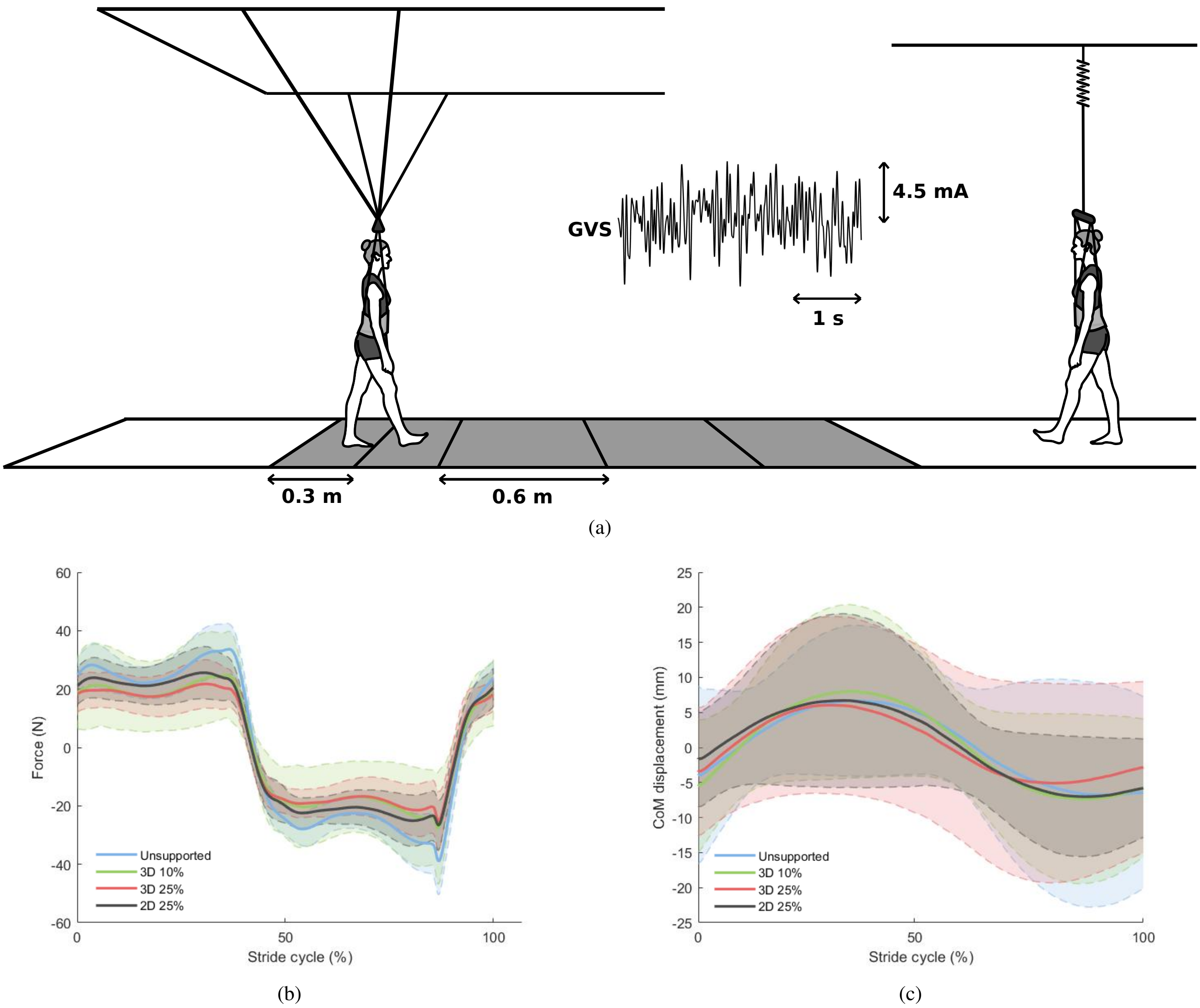


Fig. 1: Overview of the experimental setup and some secondary outcome measures. 1a) Schematic representation of the setup for the experiment, with the RYSEN BWS system and harness on the left, the passive 2D system on the right, the walkway and force plates, and a small fragment of the GVS signal. 1b) The average mediolateral ground reaction forces for all participants in the four conditions where GVS was applied. The standard deviation for each condition is indicated with the shaded area between the dashed lines. 1c) The average mediolateral displacement of the center of mass for all participants in the four conditions were GVS was applied, shifted to oscillate around 0. The standard deviation for each condition is indicated with the shaded area between the dashed lines.

We chose 25% and 10% BWS to match the range of conditions typically employed during patient training. The order of the conditions was randomized for each participant using balanced Latin squares [39]. The BWS in the RYSEN was automatically applied and monitored after selecting the desired amount in the app. In the passive 2D system, selecting the correct amount of BWS had to be done manually. The participant stood on the force plate while the cable was pulled tight by the experimenter, who was constantly checking the real-time values from the force plate. When a value of 75% of their bodyweight was reached, the cable was locked and the participant walked a few times back and forth to check if the BWS level remained constant.

Participants wore tight clothing with shorts and walked barefoot to allow for secure marker placement. Participants were outfitted with the RYSEN harness, and their weight and height were measured. Participants stood in the middle of the walkway, and their head was positioned in an orientation with the Reid's plane 18° higher than horizontal to maximize the effectiveness of GVS on ML balance [25], [36]. A piece of tape was placed on the wall on both sides of the lab as a visual target to maintain this angle during walking. Participants were instructed to look at the tape when they walked and were reminded of this multiple times during the experiment.

Two participants were outfitted with four extra markers on the face (see Appendix C, figure A5 for the exact placement). These markers were used to assess whether participants could maintain the 18° angle.

Participants walked at a cadence of 78 steps per minute during the trials, guided by a metronome. This cadence was selected to replicate the experiment by Forbes et al. [40]. The cadence was first practiced on a treadmill at a speed of $0.6\,\mathrm{m/s}$ or directly on the walkway. By maintaining this specific cadence and speed, exactly five heel strikes fit the force plates. A constant velocity for all participants was also needed because of the sensitivity of vestibular responses to walking speed [21]. Participants then made a few test walks over the walkway to determine where they needed to start to ensure that the first step on the force plate was positioned correctly.

E. Data analysis

Both the ML gait parameters of the participants and the coupling between the vestibular stimulus and mediolateral forces were measured for each condition.

To determine the ML gait parameters of the participants during gait, step width, step width variability, the margin of stability (MoS), and margin of stability variability were used as balance-related gait parameters. The gait parameters were calculated for each condition and only when participants walked on the force plates.

The vestibular contribution to gait stability was determined by calculating the time-dependent coherence and gain between the vestibular stimulus (i.e., GVS) and the ML ground reaction forces on the force plates [22], [40].

The data were analyzed using custom Matlab scripts. Heel strike events were extracted from the force data based on a threshold in the vertical ground reaction force of $10\,\mathrm{N}$. For each walkway pass, a full stride (i.e., three heel strikes) was extracted from the measured data (i.e., force plate, motion capture and electrical stimulus).

1) Gait parameters: Step width was measured as the distance between ankle joints at heel strike. The location of the ankle joints was estimated from markers placed on the lateral malleolus, calcaneous, and the first and fifth metatarsal head. The midline of the foot was calculated as the vector from the calcaneous marker to the midpoint between the two metatarsal markers. The ankle joint was then estimated as the projection of the lateral malleolus marker on this midline. The step width variability was determined as the standard deviation of all step width values in one condition.

To determine the MoS, the method proposed by Hof, Gazendam, and Sinke [41] was used. They define the concept of the extrapolated center of mass XcoM as

$$\mathrm{XcoM} = x_{\mathrm{CoM}} + \frac{v_{\mathrm{CoM}}}{\omega_0}\,, \tag{1}$$

with x_{CoM} the lateral position of the center of mass (CoM), v_{CoM} the velocity of the CoM, and ω_0 the eigenfrequency of the body as an inverted pendulum. Here $\omega_0 = \sqrt{g/l}$, with $g = 9.81\,\mathrm{m/s^2}$ the gravitational acceleration and l is the estimated height of the CoM during standing, which is calculated as 55% of the total body length [42].

The position of the CoM for the x_{CoM} was estimated as the midpoint of all four pelvis markers (anterior and posterior superior iliac spine, both left and right) [43]. The lateral position x_{CoM} was then determined as the x-coordinate of the CoM position, and v_{CoM} as its derivative.

The dynamic margin of stability was then determined as

$$\mathrm{MoS} = \mathrm{XcoM} - \mathrm{BoS}\,, \tag{2}$$

with BoS the base of support, which was determined as the lateral center of pressure. The MoS was calculated for each step at heel strike [44] [45]. The MoS variability was calculated as the standard deviation of all MoS values for one condition.

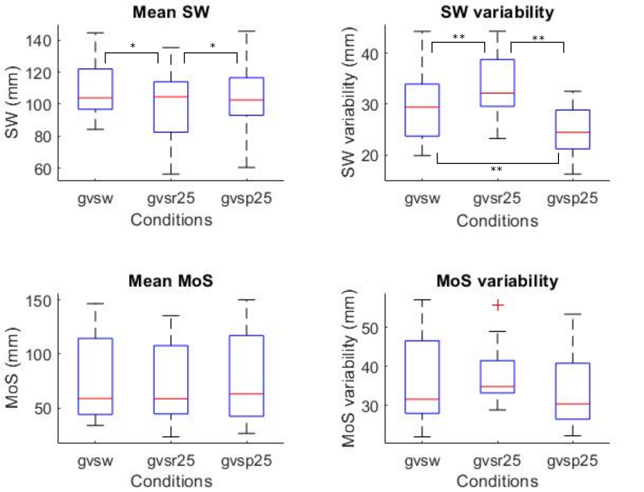
2) Coherence and gain analysis: The time-dependent coherence and gain between the force data and the stimulus was calculated using the continuous Morlet wavelet decomposition used by Forbes et al. [22], [46]. The window from heel strike to heel strike that was created previously was shifted by 15% to create a window approximating the time of toe-off for each limb [47]. This ensured that the entire body weight was on the force plates and that the total mediolateral force produced by the participant onto the ground was being measured. A manual check of each identified step was performed to eliminate artifacts (e.g., tipping the force plate when stepping close to the edge).

Because participants walked back and forth over the force plates, all force data when participants returned to the starting position was inverted to ensure all GVS-evoked responses were in the same direction. The corrected data were then used to perform the Morlet analysis, which resulted in a time-frequency coherence and gain plots for each participant for the four GVS conditions.

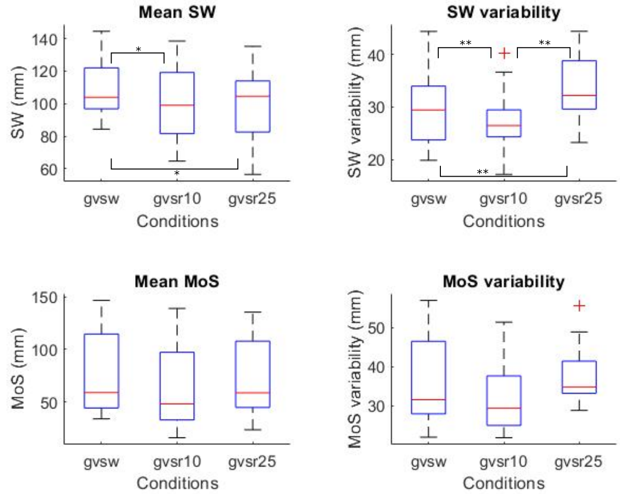
F. Statistical analysis

For the gait parameters, a statistical analysis was performed in SPSS (IBM, Armonk, NY, USA). Data were compared using one-way repeated measure analysis of variance (ANOVA). First, the data were assessed for the validity of using ANOVA by checking for normality (Shapiro-Wilk) and sphericity (Mauchly's test for sphericity). If these criteria were not met, the non-parametric Kruskal-Wallis test or a Greenhouse-Geisser correction were applied, respectively. If a significant main effect was found, posthoc pairwise comparisons with Bonferroni correction were applied to test for significant differences between conditions. To test for significant effects of GVS on gait parameters, only two conditions were compared at a time, and independent samples t-tests were used. A significance level of $p < 0.05$ was used for all statistical tests.

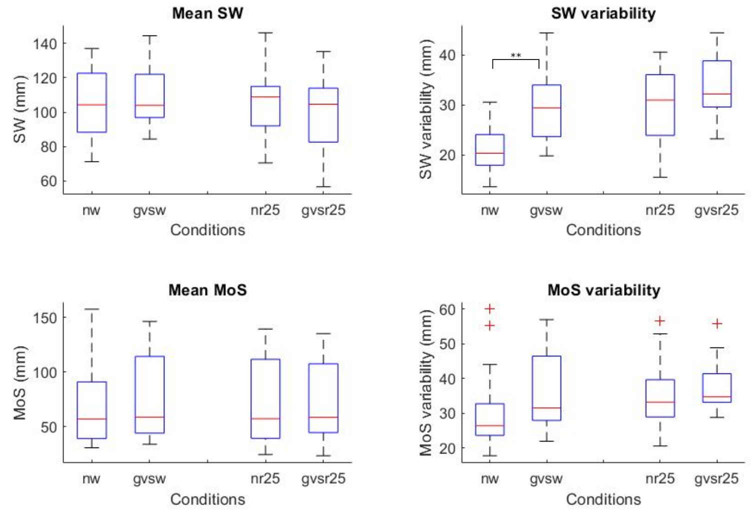
The time-frequency coherence and gain data were analyzed for significant differences with a custom Matlab script that applied a bootstrapping analysis. Differences between conditions were visualized in coherence and gain difference plots. Non-significant areas in these difference plots were masked to distinguish only significant effects.



(a) The effect of degrees of freedom on the gait parameters. The gvsw condition is the baseline in this case, gvsr25 is the 3D condition, and gvsp25 is 2D.



(b) The effect of the amount of BWS on the gait parameters. The gvsw condition is the baseline in this case, the gvsr10 condition has 10% BWS, and the gvsr25 condition has 25% BWS.



(c) The effect of GVS on gait parameters in normal walking and walking in the RYSEN. The nw and nr25 conditions are the baseline in this case, gvsw and gvsr25 are with GVS.

Fig. 2: Boxplots of the gait parameters for all 14 participants combined. Different comparisons are made between the six conditions: nw = baseline condition; gvsw = no BWS with GVS perturbations; nr25 = 3D BWS at 25%; gvsr25 = 3D BWS at 25% with GVS perturbations; gvsr10 = 3D BWS at 10% with GVS perturbations; gvsp25 = 2D BWS at 25% with GVS perturbations. Figure 2a demonstrates the effect of the 3D vs the 2D system on the gait parameters, figure 2b of the amount of BWS, figure 2c of GVS on walking without BWS, both unsupported and in the 3D system. Significant differences between conditions are indicated with a $*$ for $p < 0.05$ and with $**$ for $p < 0.01$.

III. Results

A. Gait parameters

Participants maintained a constant gait pattern in time with the metronome and did not fall during all conditions, despite the GVS. The average velocity was close to the intended $0.6\,\mathrm{m/s}$, with a slightly higher average during unsupported conditions of $0.62\,\mathrm{m/s}$ compared to the $0.60\,\mathrm{m/s}$ for conditions in a BWS system.

Comparisons between normal walking and BWS supported walking conditions revealed that relative to normal walking, step width variability increased during 3D BWS (normal: $29.8(6.8)\,\mathrm{mm}$ (mean (SD)); 3D: $33.7(6.2)\,\mathrm{mm}$, $p < 0.01$) and decreased during 2D BWS (2D: $25.0(5.0)\,\mathrm{mm}$, $p < 0.01$), see figure 2a. In contrast, mean step width decreased by $\sim$ 10% during 3D BWS (normal: $109.2(18.6)\,\mathrm{mm}$; 3D: $100.1(22.4)\,\mathrm{mm}$, $p = 0.03$), and did not significantly differ from normal for 2D BWS (2D: $105.2(23.0)\,\mathrm{mm}$, $p = 0.57$). The margin of stability did not change significantly for any condition (normal: $74.1(36.8)\,\mathrm{mm}$; 3D: $70.7(38.7)\,\mathrm{mm}$; 2D: $72.9(39.5)\,\mathrm{mm}$, $p = 0.96$), and neither did the MoS variability (normal: $35.7(11.0)\,\mathrm{mm}$; 3D: $37.8(7.8)\,\mathrm{mm}$; 2D:

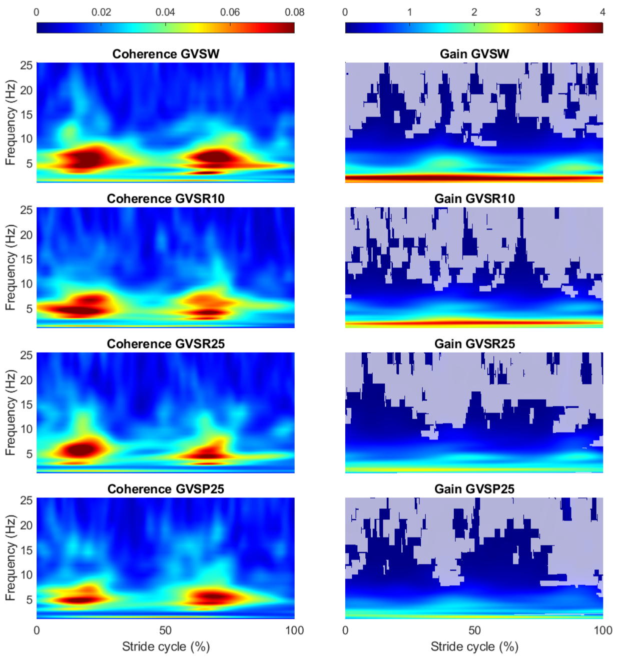


Fig. 3: Surface plots of the coherence and gain. In the left column, the coherence between the GVS and the ML ground reaction forces of the four different GVS conditions are pictured. In the right column, the corresponding gain plots are pictured. Areas are plotted slightly opaque in the gain plots where the coherence was below the significance threshold. The magnitude of the coherence and gain is indicated by the color bars.

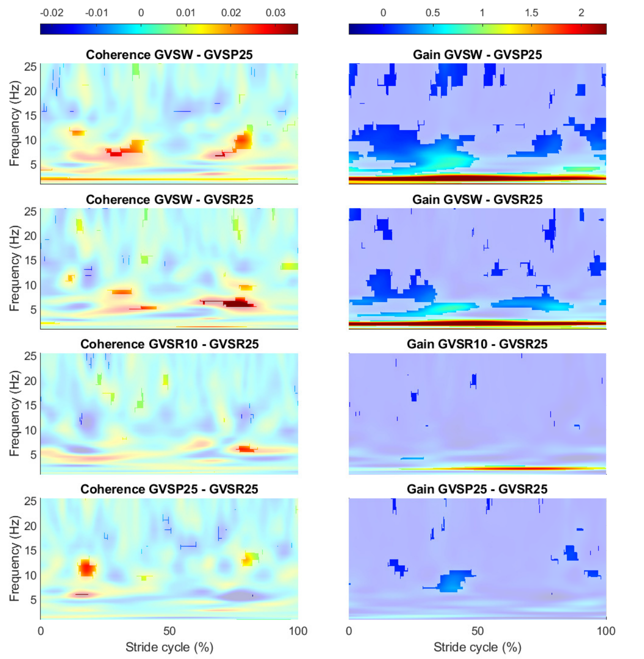


Fig. 4: Surface plots of the differences between conditions in coherence and gain. In the left column, the differences in coherence between four comparisons of GVS conditions are pictured. In the right column, the corresponding differences in gain plots are pictured. Areas that were not found to be significant after bootstrapping analysis are plotted slightly opaque. The magnitude of the difference in coherence and gain is indicated by the color bars.

$33.9(10.3)$ mm, $p = 0.20$).

The amount of BWS also modified the gait parameters when participants walked in the 3D system (see figure 2b). Surprisingly, against the increase in step width variability that we already observed for the 25% condition (normal: $29.8(6.8)$ mm; 25%: $33.7(6.2)$ mm, $p < 0.01$), step width variability decreased at the lower level of BWS compared to normal (10%: $26.8(6.2)$ mm, $p < 0.01$). Both levels of BWS in the 3D system resulted in a reduced step width (normal: $109.2(18.6)$ mm; 10%: $100.9(23.6)$ mm; 25%: $100.1(22.4)$ mm, $p < 0.01$). Again, no significant differences could be observed between any conditions for the mean MoS (normal: $74.1(36.8)$ mm; 10%: $61.6(39.7)$ mm; 25%: $70.7(38.7)$ mm, $p = 0.47$) or the MoS variability (normal: $35.7(11.0)$ mm; 10%: $31.6(8.5)$ mm; 25%: $37.8(7.8)$ mm, $p = 0.07$).

To assess the influence of the stochastic GVS on gait parameters, comparisons were made between walking with and without GVS, both unsupported and in the 3D system (see figure 2c). Step width variability in unsupported gait increased by almost 50% when applying GVS (normal: $20.9(4.9)$ mm; GVS: $29.8(6.8)$ mm, $p < 0.01$). but the increase in step width was not significant (normal: $103.9(20.2)$ mm, GVS: $109.2(18.6)$ mm, $p = 0.48$). Both the mean MoS and the MoS variability increased slightly, but not significantly (normal: $70.4(40.9)$ mm, GVS: $74.1(36.8)$ mm, $p = 0.83$; and normal: $31.5(12.7)$ mm, GVS: $35.7(11.0)$ mm, $p = 0.36$, respectively). In the 3D system, no significant changes were observed due to the GVS in any gait parameter. Step width variability showed a small and not significant increase (normal: $29.5(7.3)$ mm, GVS: $33.7(6.2)$ mm, $p = 0.12$) and step width even showed a slight but not significant tendency to decrease with GVS (normal: $104.3(19.3)$ mm, GVS: $100.1(22.4)$ mm, $p = 0.60$). Both the mean MoS and MoS variability showed no discernible difference (normal: $71.4(40.3)$ mm, GVS: $70.7(38.7)$ mm, $p = 0.99$; and normal: $35.3(10.4)$ mm, GVS: $37.8(7.8)$ mm, $p = 0.48$, respectively).

B. *Coupling GVS and ML ground reaction forces*

In all conditions, a clear phase-dependent group mean response that peaks about 20% of the stride cycle after toe off can be observed (see figure 3). The response in the gain is later and peaks about 40% of the stride cycle after toe off. For these results, GVSW can be considered as the baseline condition. The remaining three conditions are displayed in order of how much the system restricts the participant.

Similar coherence and gain plots can be observed for the BWS conditions. Even though the timing and shape of the coherence and gain peaks remain the same, the size of the peaks appears to decrease for each condition. The 2D condition (GVSP25) shows the smallest peaks. The peaks in the 3D are slightly (GVSR25) larger, and in the 3D condition at 10% BWS (GVSR10), the peaks are almost the same size as in the baseline condition.

In figure 4, the differences between the coherence and gain in different conditions are displayed. A bootstrapping analysis was applied to determine which areas significantly differed between conditions. Comparisons were made between the baseline and 2D condition, baseline and 3D, 3D and 3D at 10% BWS, and between the 2D and 3D condition. The differences in coherence were small and mostly not significant for all comparisons. However, the comparisons in gain between baseline and 2D and between baseline and 3D showed significant differences. Especially at low frequencies, these differences were large. Between the 3D system and the other conditions, both 10% BWS and the 2D condition, differences in gain were small and mostly not significant.

IV. Discussion

The primary aim of this study was to determine the differences in influence on the mediolateral balance between 2D and 3D BWS systems. The influence on ML balance was determined by measuring balance-related gait parameters and the coupling between the GVS and ML ground reaction forces. Participants walked overground on force plates in both a passive 2D BWS system and the 3D BWS system the RYSEN. Comparisons in gait parameters showed significant differences in step width and step width variability between the two systems. The contribution of the vestibular system to ML balance was determined by applying GVS and measuring the ML ground reaction forces. Our data show that BWS reduces the coupling between GVS and the ML ground reaction forces.

A. 2D BWS decreases while 3D BWS increases step width variability

Our results show that the tested 2D and 3D BWS systems have different effects on gait stability measures. Similar to previous studies, participants decreased step width variability when walking in the 2D system [7], [14]. However, participants increased their step width variability when walking in the 3D system at an equivalent level of BWS. This is in contrast with most studies, where BWS was found to decrease ML gait parameters. Dragunas and Gordon [7] found that BWS tends to reduce the requirements for lateral stability. This is supported by studies on healthy [48] and post-stroke [49] participants, which showed that ML trunk accelerations decrease in BWS systems. Similar effects are found in studies where participants are stabilized laterally on the pelvis, where a decrease in step width variability [50] is found. That this decrease in step width variability is not found in the RYSEN system could indicate that the system does not stabilize participants laterally. Moreover, participants walking in the RYSEN would be challenged more than in a 2D BWS system. However, while greater step width variability could indicate a greater challenge to stability [18], it is a complex measure to interpret.

In this study, no significant change in mean step width was found for the 2D system, while step width decreased when participants walked in the 3D system. It has been suggested that healthy individuals increase their step width for challenging conditions that require a high level of stability [51], [52]. A decreased step width is a common effect in BWS gait [13]–[15], [53]. The changes observed for step width and step width variability in our study are in opposite directions. We found only one study describing an increase in step width with additional BWS, though this was accompanied by a decrease in step width variability [7].

Our 2D BWS system used a spring, which allows for a gait pattern closer to normal gait than a constant force [31]. It is often unclear what type of BWS was used for previous BWS studies. A different BWS type could create a difference in study outcomes. Furthermore, some studies used a treadmill while others were overground, which may also result in slightly different outcomes [54], [55]. These potential differences between studies complicate drawing conclusions on the effects of BWS on gait parameters, and more research is needed to understand the underlying causes of these effects.

B. Low and high amounts of 3D BWS show contradictory results

While 3D BWS at 25% increased step width variability, at a level of 10%, step width variability appears to decrease. A decrease in both step width and step width variability was found in this study. The differences in ground reaction forces could explain this difference between the 25% and 10% conditions. At 25% BWS, only a reduced amount of ground reaction forces can be used to control balance and correct for the GVS perturbations. This makes the active control that is needed in ML direction more complex [11]. At 10% BWS, this effect is also present, but in a much smaller amount.

Another possible explanation arises from the fact that spatio-temporal gait parameters have been shown to change significantly after a certain level of BWS by Apte, Plooij, and Vallery [56]. In their review study, they conclude that this "threshold" lies at about 30% BWS, after which gait parameters change drastically. They did not look at stability in the ML direction. Nevertheless, their findings are in accordance with the changes we observe between 10% and 25%, which comes close to the 30% threshold.

C. GVS only affects gait parameters in unsupported gait

For both unsupported gait and walking in the RYSEN at 25% BWS, a comparison was made between an unperturbed and GVS condition. For most parameters, only a small and not significant increase was found when adding GVS, but SW var in unsupported gait showed a large increase with GVS.

Not many studies have been performed on GVS and ML gait parameters to support these findings. Nevertheless, a recent study by Magnani et al. [28] did look at step width and its variability during gait. They found no significant effect for step width but an increase in step width variability when applying GVS. This is in accordance with our findings for unsupported gait. For step width variability in the RYSEN, only a small and not significant increase was found. To our knowledge, no research has been done on the effect of GVS combined with BWS on stability. Further research is needed to determine why the effect of GVS on gait parameters is different in BWS systems from unsupported gait.

D. Step width is more sensitive to balance perturbations than MoS

Something that stands out from our results is that none of the MoS measures are significantly different for different conditions (see figure 2). This contrasts with the four significant differences observed in mean step width and seven in step width variability.

We were able to find more studies where step width or step width variability increased, but MoS showed no significant difference [14], [57], [58]. It may be that the MoS is generally less sensitive to changes in ML stability than step width. Additionally, we found that variability measures generally yield significant effects more often than their mean counterparts, which is also apparent in the results from this study. It might thus be more beneficial to employ variability measures than means when looking for differences between conditions.

Another explanation could lie in the definition of the MoS. The MoS is linked to step width because it uses the base of support in its calculation. What sets it apart from step width is that it also takes the movement of the CoM into account by calculating the distance between the extrapolated CoM and the base of support. So although the two concepts are linked (step width and base of support are closely related), step width and MoS can show different results. Participants in this study appear to have kept their MoS constant, which has been shown in other studies before [18]. Increasing the step width could be seen as a strategy to keep the MoS at the same level.

E. Coupling between GVS and ML ground reaction forces is lower with BWS than in unsupported gait

Compared to the baseline of unsupported gait with GVS, both the 2D and the 3D systems significantly decreased the gain between the GVS stimulus and the ML ground reaction forces. For both comparisons, these differences are largest at the low frequencies. In contrast, neither comparison with the baseline shows much significant change in coherence. The comparisons between 3D at 25% and at 10% and between the 2D and 3D systems show almost no significant differences in both coherence and gain.

The drop in gain from the unsupported gait to the BWS conditions shows a reduction of vestibular influence on the ML ground reaction forces. It is unclear why this change is only present in the gain and not in the coherence. Possibly, more data from either more participants or a larger number of strides per condition would have resulted in significant changes in coherence. Differences between the 3D and 2D systems or a lower level of BWS were small, indicating that coupling between the stimulus and the ML ground reaction forces was not altered significantly.

F. Limitations

Three main limitations were identified in this study. First, the study was limited to only testing a few participants (14), and the number of strides collected in each condition was small. It could be possible that significant differences between conditions would have emerged with more strides or participants. The amount of 100 strides for the GVS conditions was selected based on pilot studies. In the pilots, significant differences in coherence were visible for 100 strides. However, the participants in the pilots were people highly involved and practiced in the experiment. They might have been more accustomed to the setup and thus shown less variability in walking within one condition. Magnani et al. [24], who did find significant differences in coherence, used a much larger number of 256 strides. They used a treadmill, drastically reducing the time needed to collect such a large number of steps. Using a treadmill in the present study was undesired, as the goal here was to study overground BWS systems. Other studies examining GVS and gait used about 115 steps to find significant results [26], [27], but they did not look at coherence.

The second limitation of this study is that there were unintended differences between the 2D and 3D BWS systems. Ideally, only ML forces would differ between the two systems. In section III, we described how one participant was excluded because the unloading force in the 2D system was lower than intended (see Appendix B for more details). The average BWS percentage in the 2D system was 22.7%, versus 25% in the 3D system. However, this difference in VT force is small compared to the AP forces. The RYSEN is controlled to follow the participant in ML and AP direction. We know that the force tracking error is low, with an RMS error of $3.0\,\mathrm{N}$ in ML direction, and $1.4\,\mathrm{N}$ in AP direction (see Appendix E). The 2D system, contrarily, is completely passive and has to be pulled forward by the participant. So in reality, the effect of not only a difference in ML but also slightly AP force is examined in this study.

Finally, after analysis on the two participants who wore face markers during the whole experiment, it was observed that participants did not maintain the intended 18° head angle during the experiment. The observed average head angle for the two participants was 3.5°, see Appendix D. The effectiveness of GVS is known to be highest at an 18° head angle [25], [36]. The effectiveness of the GVS perturbations was thus lower than intended in this study. This could be improved by measuring and adjusting the head angle in real-time during the experiment. Magnani et al. [24] accomplish this by making participants wear a headgear-mounted laser which has to point at a target placed at such a height that the 18° angle is achieved. This maximizes the effectiveness of the GVS and also yields less variability in head angle between participants. However, a different solution has to be created for overground walking where the head angle changes as people approach the target.

G. Recommendations

Three promising research possibilities were identified in this study. First, there appear to be differences between 2D and 3D systems in the effects on ML gait parameters. However, the differences observed in this study were not straightforward to interpret. For example, the increase we found for step width variability in the RYSEN contradicts the decreased mean step width. These results do not comply with the current BWS literature, which is mainly on 2D BWS systems. More research on 3D BWS systems is needed to find the underlying causes for the difference in ML gait parameters and determine the implications on clinical practice. Based on this study alone, it is not possible to conclude whether 3D BWS systems are more challenging and realistic for participants.

Secondly, we observed significant changes between the conditions where the RYSEN delivered 10% and 25% BWS. A surprising result was that the step width variability decreased for the 10% condition compared to baseline, whereas step width variability increased for the 25% condition. More research is needed to determine the underlying causes of this effect. It is also valuable to know whether this difference in step width variability is a sliding scale or if there is some threshold as was found for AP gait parameters [56]. There could be an "optimal point" where step width variability is not changed from unsupported walking.

Finally, very little is known about the effects of GVS perturbations on ML stability during gait. To our best knowledge, no research has been published before on the combination of GVS and BWS. More research is needed to determine whether GVS is a valuable tool to realistically mimic the behavior of balance-impaired individuals walking in a BWS system.

V. Conclusion

In this study, we investigated the differences between a 2D and a 3D BWS system in how they influence ML balance-related gait parameters and the contribution of the vestibular system to balance.

Regarding the gait parameters, we found that participants increased their step width variability in the 3D system compared to unsupported gait, but in contrast, decreased step width variability in the 2D system. Surprisingly, step width decreased in the 3D system. We conclude that the 2D and 3D BWS systems influence the gait parameters differently, but it remains unclear whether one of the systems was more stabilizing. We also investigated the effects of different levels of BWS. When 3D BWS was applied at 10% of the bodyweight instead of 25%, step width variability and mean step width decreased compared to unsupported gait. We conclude that the level of BWS can influence the gait parameters and that a high level of BWS may be more challenging to ML balance than a lower level. When evaluating the effects of GVS on gait parameters, we found that differences were only present in step width variability during unsupported gait. We conclude that GVS can affect gait parameters, but BWS may overshadow this effect.

Finally, we determined the differences in the contribution of the vestibular system between different BWS conditions. Significant decreases in coupling between the GVS stimulus and the ML ground reaction forces were observed in comparisons between unsupported gait and both the 2D and the 3D BWS system. However, no other significant differences were observed between the 3D system and the 2D system or a lower level of BWS. We conclude that the vestibular contribution to balance decreases when walking with BWS, but not differently for a 2D or 3D system.

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APPENDIX A
INFORMED CONSENT

Informed Consent Form

Research Study:
Influence of 2D versus 3D body-weight support systems on mediolateral balance during gait

Delft University of Technology

Informed Consent Form

This informed consent form is for individuals who are invited to participate in this TU Delft study about the effects of 2D versus 3D body weight support systems on mediolateral balance during gait.

Researchers:	Suzanne Rademaker (contact), Patrick Forbes
Supervisor:	Heike Vallery
Organization Name:	Delft University of Technology (TU Delft)
Faculty:	Biomechanical Engineering Department, Mechanical, Materials, and Maritime Engineering (3ME) Faculty, TU Delft

Below is a brief introduction to the study and your role in it. If you agree to participate after reading this information, please sign the certificate of consent at the end of this form. You will receive a full copy of your signed Informed Consent Form, upon request.

Information Sheet

Introduction:

During walking a human has to actively control their balance to stay upright and not fall sideways. Bodyweight support is commonly used to train individuals with gait impairments who lack the muscle strength to support their body weight. Bodyweight support systems usually consist of a harness that is connected to the ceiling with cables. The cables are pulled tight so that they carry part of your bodyweight. This makes walking much lighter and can be used to learn to walk again after an accident for example because you cannot fall and you need less strength in your muscles.

Traditional bodyweight support systems enable movement in 2 dimensions: up/down and forward/backward. A 3D system however allows movement in 3 dimensions, enabling you to also walk left/right.

This study aims to investigate the differences between a traditional 2D system and a 3D system. The results of this study will help to understand the effect of these systems and how they help with training a person to walk again, improving the current knowledge of rehabilitation. The research will also aid further development of the used 3D bodyweight system (RYSEN – Motek Medical B.V.).

This study is in collaboration with Motek Medical B.V.

Who can participate in this study:

Healthy subjects between the ages of 19 and 50 who are proficient in English and have a body mass between 50 and 90 kg.

Who should not participate in this study:

If you meet any of the following criteria, you should not participate in this study:

- History of neck or back pain
- A known disease affecting your muscles or nerves
- History of balance problems
- History of migraines or severe headaches
- History of severe motion sickness
- Pregnant women
- A prior neuromuscular injury

- Hearing deficits
- Incompetence to give informed consent
- Unable to travel to the lab in Delft without using public transportation[1]

What does the study involve:
If you decide to participate, you will be requested to come to the Department of BioMechanical Engineering, Mekelweg 2 Delft, at the Delft University of Technology.

What is expected of me?
Throughout these experiments, you will receive a stimulus that will activate your vestibular system. This galvanic (i.e. electrical) vestibular stimulation (GVS) will be delivered through two rubber electrodes placed behind your ears and will evoke a sensation of head movement even when remaining completely motionless. This can be compared to the feeling of standing on a swaying boat. You will have some time to get accustomed to the sensation so that you feel comfortable walking with the GVS.
You will then be weighed to initialize the experimental setup according to your weight. After recording your weight, you will be asked to wear a full-body harness. A full-body reflective-marker set will then be placed on your body to facilitate the motion capture system. After this, a calibration procedure for the measurement set-up is performed, in which you will be asked to stand in the measurement space and mimic the movements of the experimenter. After the calibration procedure, you will be asked to walk multiple steps over a walkway to record measurements of unsupported gait. The full-body harness is then supported from the RYSEN device or the passive bodyweight support device, and some time will be spent in different unloading conditions to familiarize yourself with the device. After the familiarization, your walking will be recorded for 6 different conditions of the device. In each condition, you will have to walk over the walkway 30 times back and forth. This means a total of 7 trials (1 free-walking and 6 with a bodyweight support device).
Passive markers for the 3D motion acquisition system will be attached to your body on your feet, legs, and hips. You are advised to wear tight-fitting clothes to ensure we can obtain accurate measurements from our motion capture system. Additionally, you are advised to wear shorts so that the motion capture markers can be placed directly on your skin. The markers can easily be removed afterward. We can also provide the appropriate clothing at the lab.
The experiment is expected to take around 120 minutes.

If you decide to take part in this study, you will be required to perform the tasks described above.

COVID measures:
Due to COVID-19, we take multiple precautionary measures to minimalize your and our own risk. We ask you (and the researchers) to stay home if you or one of your housemates has any COVID symptoms. During the experiment, no more than 4 people will be present in the room, which is large and well-ventilated. Everyone will be asked to wear face masks and disinfect their hands regularly. Finally, participants are asked to participate in pairs of people from the same household so that any procedures that require physical contact (such as placing electrodes) can be done by your roommate. If you have any questions or doubts regarding these safety measures, please contact Suzanne Rademaker.

[1] Due to COVID regulations, we want to minimize travel as much as possible.

Possible harms or side effects of participating:
Wearing a full-body harness might make walking slightly uncomfortable initially because the straps are pulled very tightly, but this will not pose a problem once you are accustomed to it.
There are no known physical or physiological risks associated with the non-invasive electrical vestibular stimulation technique. The electrical stimulation will produce a skin sensation of mild tingling at the site of the stimulating electrodes and may generate dizziness on rare occasions. Most subjects report a moderate sensation of motion and slight flashing in the visual field. Some subjects who are highly susceptible to car motion sickness or with a history of headaches (exclusion criteria) may possibly experience mild nausea, light-headedness, or mild headaches for a brief period (up to 1 hour) following the experiment (in about 20% of subjects we have tested in the past).
At your request, the experiment can be stopped immediately if you feel uncomfortable.

Data Policy:
Personal information such as your weight and height will be measured, and your age will be asked before the experiments. During the experiments, identifiable (full-body) video recordings will be made of your walking gait. All the recorded data will be anonymized and stored safely without access to external parties. Personal data, which links your anonymized data to yourself, will be stored separately and only the researchers may have access to it. The video recordings will not be kept for longer than 12 months. If any video recordings have to be stored for a longer period or used for any type of publication (such as a presentation or open data article) this will only happen with your consent. Any other identifiable data (such as name, email address, telephone number) are stored separately from the recorded data and will not be kept for longer than 6 months. All information will be archived so that no one except the researchers and supervisors as listed above will have access to the data. On request, you will have access to your data. You may discuss with other participants after the study period, but please respect the confidentiality of others' participation in the study. All data is made anonymous for publication purposes. The anonymized data will be processed and uploaded to an online repository in the advent of a possible publication.

Participant's rights:
Participation in this research study is completely voluntary. Even after you agree to participate and begin the study, you are still free to withdraw at any time and for any reason. You have the right to ask that any data you have supplied to that point be withdrawn/destroyed, without penalty. You have the right to omit or refuse to answer or respond to any question that is asked, without penalty. You have the right to have your questions about the procedures answered (unless answering these questions would interfere with the study's outcome). If any questions arise as a result of reading this information sheet, you need to ask the investigators before the start of the experiment.

Cost, reimbursement, and compensation:
No cost, reimbursement, or compensation are applicable for this study.

For further information:
The investigators and supervisors listed above will gladly answer your questions about this study at any time. If you are interested in the final results of this study, you can contact one of the investigators or supervisors. For questions, please contact Suzanne Rademaker at s.l.j.rademaker@student.tudelft.nl, +316 29898719.

Informed Consent Form

Please tick the appropriate boxes

	Study participation and recorded data	YES	NO
1	I consent voluntarily to be a participant in this study. I understand that I can refuse to answer questions and that I can withdraw from the study at any time, without giving any reason.	☐	☐
2	I understand that taking part in the study involves video recordings being made that are identifiable. I agree that those video recordings are made during the experiments.	☐	☐
3	I understand that during the experiments sensor data is recorded by motion capture equipment, force plates, and by the RYSEN device itself.	☐	☐
4	I understand that I will be asked questions regarding my age and that my height and weight will be measured.	☐	☐
	Data use		
5	I understand that information I provide will be used for the master thesis and a possible research article of Suzanne Rademaker.	☐	☐
6	I understand that personal information that can identify me (such as my name, email address, and telephone number) will not be shared beyond the research team.	☐	☐
7	I understand that personal information and recorded data will be stored separately.	☐	☐
8	I understand that any identifiable data (such as the video recordings) will be either removed or anonymized a maximum of 12 months after the experiments.	☐	☐
9	I agree that the recorded data in the experiments can be used (anonymized) in research outputs and can be published as open data.	☐	☐
10	I consent that non-anonymized photos or videos that were taken in this experiment can be shown in public presentations.	☐	☐
11	I understand that I may request my data at any time and that I can make corrections to any inaccurate data that I provided. I also understand that I have the "right to be forgotten", and can request the deletion of my data.	☐	☐

Consent Certificate

I have read and understood the information above and have had the opportunity to ask questions and my questions have been answered satisfactorily. By signing this form, I voluntarily consent to participate as a research participant in this study.

____

Name of Participant (BLOCK CAPITALS)

____ _______________

Signature of Participant Date

____

Name of Researcher (BLOCK CAPITALS)

____ _______________

Signature of Researcher Date

If you would like a copy of this consent form to keep, please ask the researcher.

APPENDIX B
EXCLUDED PARTICIPANT

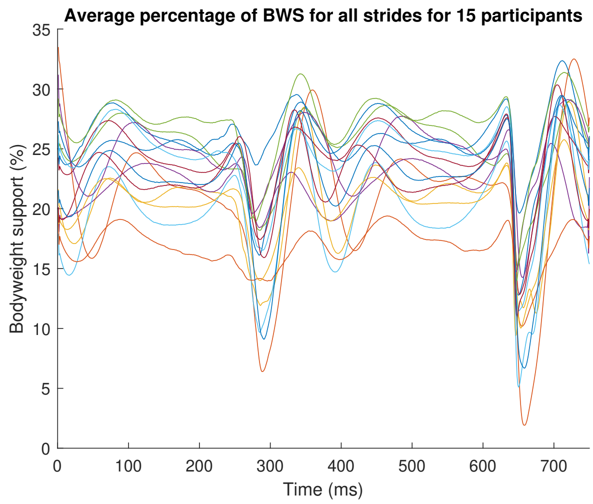


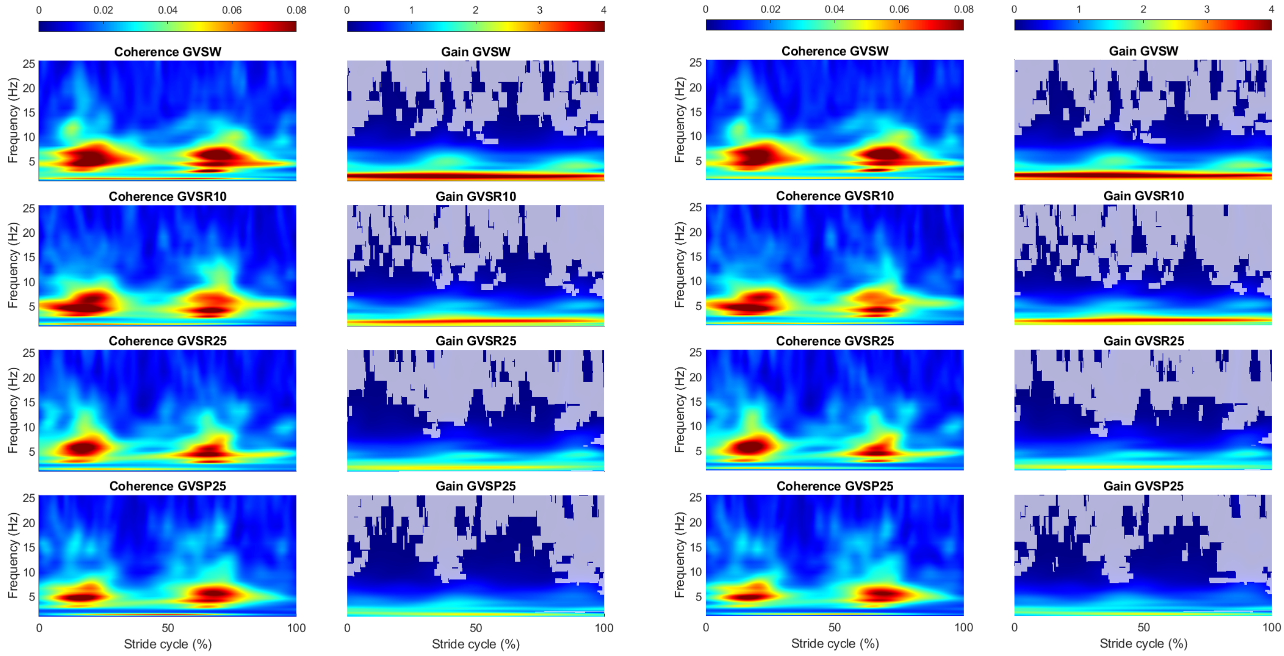
Fig. A1: The average percentage of BWS over all steps in the GVSP25 condition is calculated for all participants by extracting the force in the VT direction on the force plates. The BWS percentage is then calculated using the weight of the participant, which was measured before the experiment.

In the RYSEN system, BWS is automatically monitored and adjusted. However, in the 2D system, the percentage of BWS was set at the beginning of the condition using a force plate. The average percentage of weight taken off by the system is plotted in figure A1 to verify the percentage of BWS participants received during the 2D condition.

It was found that the average amount of BWS for all participants in the GVSP25 condition was 22.7%. This is lower than the intended 25% that is used in the RYSEN condition. For one participant, the average percentage of BWS was 16.7%, whereas all other participants had average percentages higher than 20%. This deviation was decided to be too large since 16.7% is closer to the 10% BWS condition than the intended 25%. Therefore, this participant was excluded from all analyses, and the data of only 14 participants were analyzed for the results.

In figure A2, plots of the coherence and gain results of both 15 and 14 participants can be observed. In figure A2a the outlier participant who received a smaller amount of BWS in the passive condition is included in the plots. Figure A2b was already present in this article (section III-B) and shows the results with this participant excluded. It can be observed that the differences between these two figures are very small. Thus, excluding the outlier participant in this study appears to have a minimal effect.

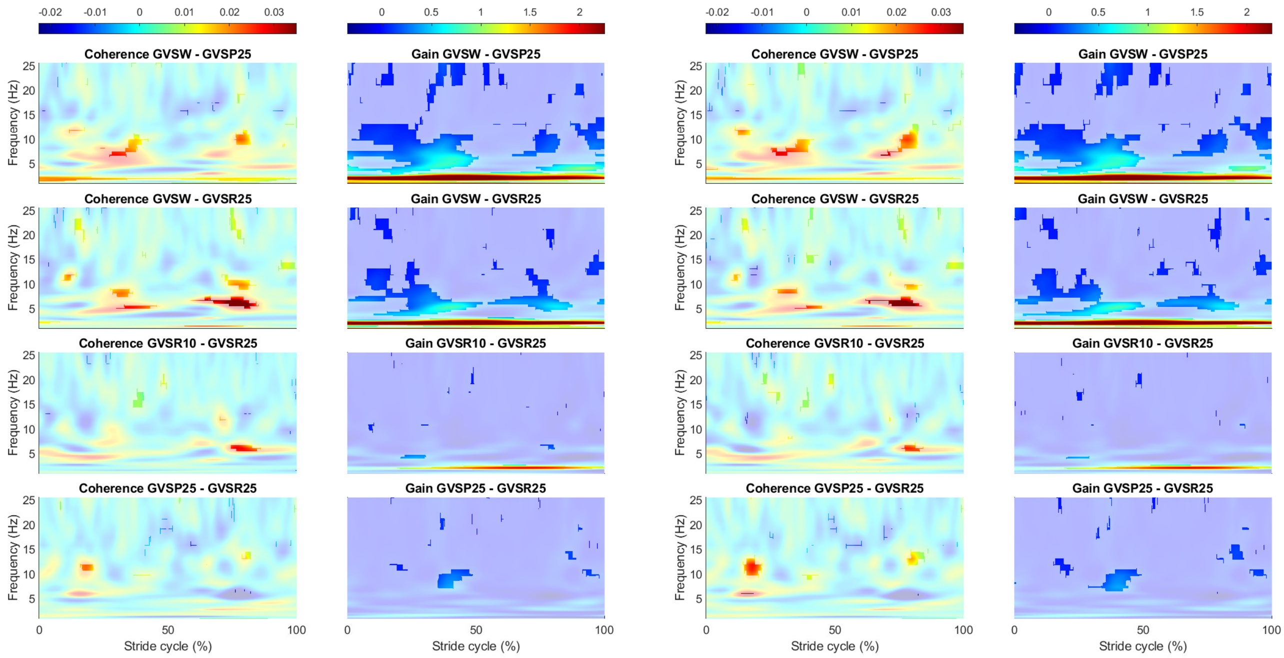
The differences between four different comparisons of conditions for both 15 and 14 conditions are shown in figure A3. Figure A3a shows plots with the outlier participant included, while figure A3b is only of 14 participants (see section III-B). Here as well, differences between the two figures are small, further solidifying the conclusion that excluding the outlier participants has minimal consequences.



(a) Coherence and gain results with all 15 participants included. This figure was not pictured before in this article.

(b) Coherence and gain results with the outlier participant excluded. This figure can also be found in section III-B.

Fig. A2: Comparison between the coherence and gain results with 15 and 14 participants.



(a) Differences in coherence (left column) and gain (right column) for three conditions with all 15 participants included. Differences that are not significant are plotted slightly opaque.

(b) Differences in coherence (left column) and gain (right column) for three conditions with the outlier participant excluded. Differences that are not significant are plotted slightly opaque. This figure can also be found in section III-B.

Fig. A3: Plots of the differences in coherence and gain between four comparisons of GVS conditions, with and without the outlier participant.

APPENDIX C
MARKER PLACEMENT

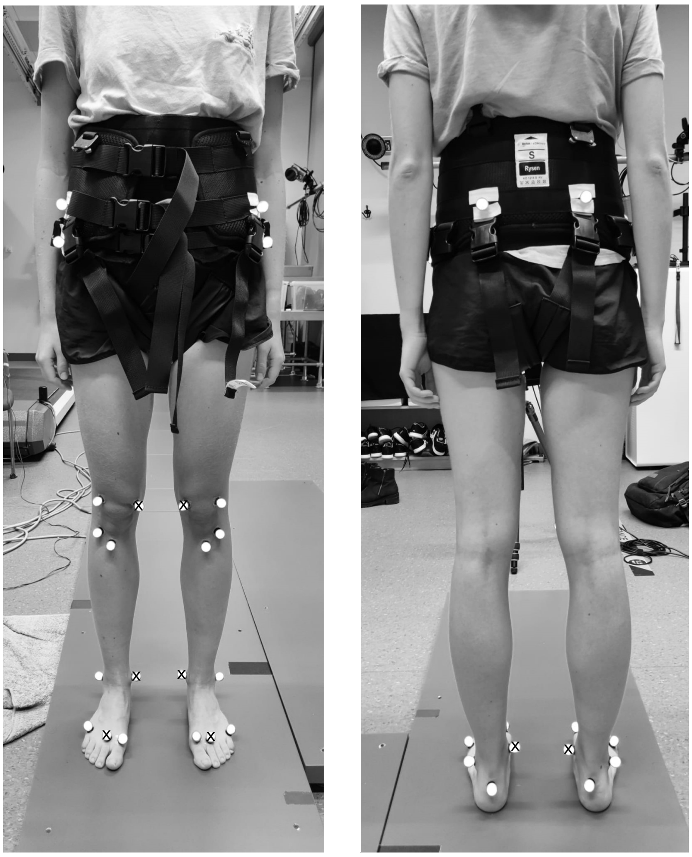


Fig. A4: IOR lower body marker placement on participant, front and backside. The markers with an “x” over them were only used for static analysis and were removed before the experiment started.

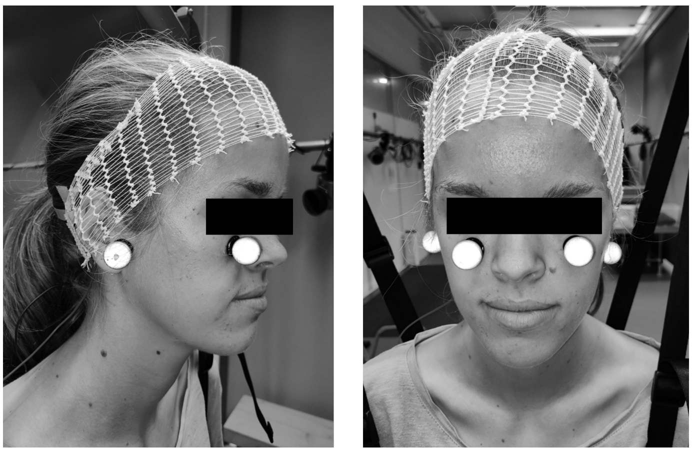


Fig. A5: Front and side view of markers placed on the face to measure head pitch during GVS trials. This markerset was used for only two participants.

APPENDIX D
CONSTANT HEAD ANGLE

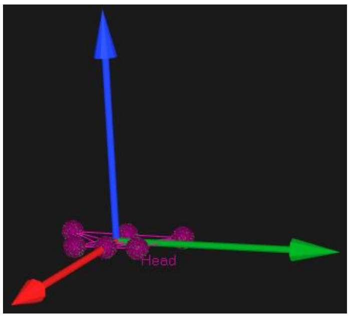


Fig. A6: Representation of the rigid body created in QTM. The two center markers are added virtual markers to help create the local coordinate frame. The origin is the midpoint between all four real markers. The X-axis (red) is created from the midpoint of the ear markers to the midpoint of the eye markers. The Y-axis (green) is parallel to the two ear markers. The Z-axis (blue) is orthogonal to the plane created by the X and Y-axis.

It was important for the accuracy of the experiment that all participants maintained the same head angle while walking on the force plates. As described in section II-D, the effectiveness of GVS is highest when the Reid's plane is 18° higher than horizontal. Therefore, a piece of tape was placed on both walls on the far ends of the room. It was placed at the height that participants had the Reid's plane at 18° when they were in the middle of the force plates. To maintain this angle, it was important that participants did not just look up with their eyes but with their whole head. Participants were told multiple times during the experiment to "keep your head up" when the experimenter noticed them looking down.

To measure what angle was actually achieved during the experiments, two participants performed the whole experiment with four markers on their face as pictured in figure A5. The plane created by the four markers was an approximation to the Reid's plane. In QTM, the Qualisys user software, a rigid body was created from the markers as pictured in figure A6. The Euler angles of this rigid body with respect to the global coordinate system were extracted from QTM. The rotation order for the Euler angles was yaw-pitch-roll (ZYX).

The angle representing the angle of the Reid's plane is the pitch. The pitch angle was extracted and plotted with a custom Matlab script. In figure A7, the average head pitch in all four GVS trials from both participants can be observed. Participant 2 maintained a slightly higher angle of $4.1 \pm 1.6°$ (mean $\pm$ SD), versus $2.8 \pm 1.1°$ for participant 1. This result is far from the desired 18°.

The fact that the average head pitch angles for these participants were about 15° lower than intended means that the applied GVS was likely not as effective as it could have been. Nevertheless, these results show that participants could maintain a more or less constant head angle. The maximum differences between conditions were about 3°, and the difference between these participants is even smaller. This suggests that the GVS perturbations are fairly consistent for all conditions and participants.

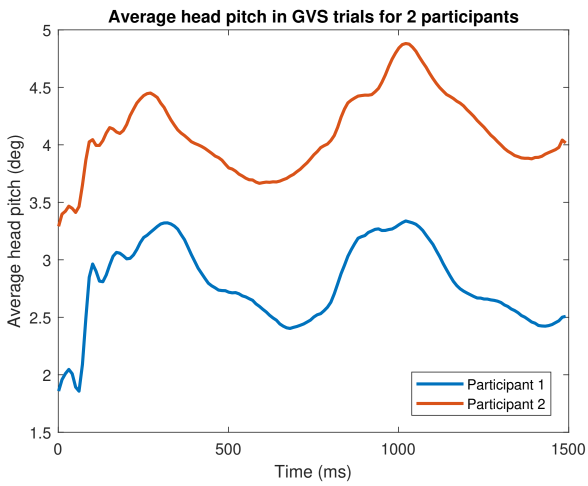


Fig. A7: Head pitch angle averaged over all steps and all four GVS conditions for two participants. Time windows were extracted from first to third heel strikes as in the main Results section. The angle appears to be smaller at heel strike and larger during the swing phase for both participants.

APPENDIX E
TRANSPARENCY RYSEN SYSTEM

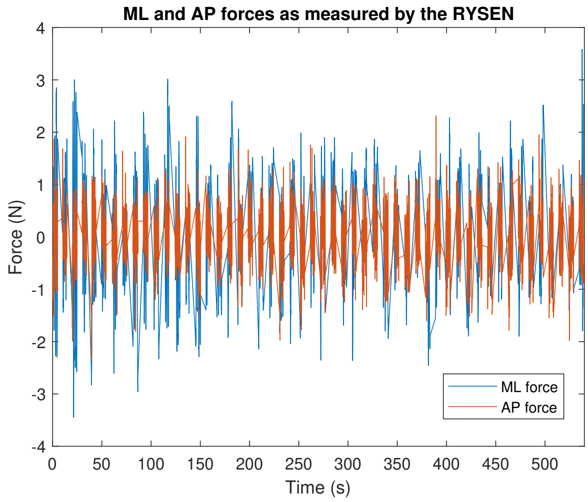


Fig. A8: Typical representation of the ML and AP forces as recorded by the RYSEN during a trial. The straight lines connecting the oscillating parts indicate when the participant was not walking on the force plates but the walkway before or after them. These data are not used in calculating the RMS of the forces.

Balance training for gait rehabilitation relies on high transparency of the BWS system in ML direction [59]. In this study, we assume that the RYSEN system is fully transparent in the ML direction and can thus be classified as a 3D system. As mentioned in section II-B, for two participants, the data from the RYSEN were recorded with D-Flow. The RYSEN system is equipped with a slingbar with sensors that calculate the forces in three directions. The root mean square (RMS) of the ML and AP forces function as a measure of the transparency of the system.

The data captured by D-Flow was analyzed with a custom Matlab script. Only data from when a participant was on the force plates were used, on account of high AP forces arising in the system when a participant stopped and turned at the end of the walkway. Because D-Flow does not use a set sample rate, the data was first resampled to be equidistant. A typical trial from one participant in one condition is shown in figure A8.

The results for the RMS of the AP and the ML RMS forces averaged over two participants are given in table A1. As expected, almost no difference can be observed between the RYSEN 25% conditions with and without GVS. Both the ML and AP RMS force appear to increase with the increasing amount of BWS, the ML force more so than the AP force. Still, these forces are relatively low, indicating high transparency of the RYSEN system.

TABLE A1: Root mean square of the forces in ML and AP direction, detected by the RYSEN in the three different RYSEN conditions. RMS forces are averaged over two participants.

Condition	ML RMS force (N)	AP RMS force (N)
GVSR10	1.5	1.0
GVSR25	3.0	1.3
NR25	3.0	1.4