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Leveraging Virtual Reality to Assess Walking Gait Performance in Altered Depth Perception Walking Tasks

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Leveraging Virtual Reality to Assess Walking Gait Performance in Altered Depth Perception Walking Tasks

THESIS

submitted in partial satisfaction of the requirements
for the degree of

MASTER OF SCIENCE

in Mechanical Engineering

by

Michael Thorne

Thesis Committee:
Assistant Professor Alexandra S. Voloshina
Associate Professor Natascha Buswell
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2026

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ABSTRACT OF THE THESIS

Leveraging Virtual Reality to Assess Walking Gait Performance in Altered Depth Perception Walking Tasks

by

Michael Thorne

Master of Science in Mechanical Engineering

University of California, Irvine, 2026

Professor Alexandra Voloshina, Chair

Vision contributes to the control of walking, balance regulation, orientation, and anticipatory control, which are important for spatial navigation. How the amount of visual information available, as distinct from its accuracy, shapes balance control during walking remains poorly understood. This study used virtual reality to determine how graded visual occlusion, imposed alongside visual and physical gait speed perturbations, affects gait and balance control during treadmill walking.

Eight healthy adults (Age 20.6 ± 2.8 years) completed nine 5-minute walking trials on a split-belt treadmill in a custom virtual environment using a 3×3 design consisting of three walking conditions crossed with three levels of visual occlusion. During baseline walking, participants walked at 1.0 m/s with the optical flow matched to the treadmill speed. During optical flow trials, participants continued to walk at 1.0 m/s while we sinusoidally varied the relative translation speed of the visual environment between 0.5 to 2.0 times the treadmill speed. Finally, during speed modulation trials, belt speed varied sinusoidally between 0.7 m/s and 1.4 m/s while optical flow remained matched to treadmill speed.

Gait performance was quantified using normalized mean step length and step width, their associated variabilities, margin of stability (MoS), and foot placement control. Mean step lengths and step width changed little across the nine trials while the step width variability increased with occlusion in every trial type by 6.0% during baseline walking, 9.1% during optical flow modulation, and 16.4% during speed modulation from the lowest to highest level of visual occlusion. Mean MoS during speed modulation trials was lower than during baseline walking and optical flow modulation. MoS variability was also substantially higher during speed modulation, exceeding that observed in the other two trial types by approximately 23.4% to 27.3%. Foot placement predictability increased across trials during baseline walking and speed modulation but remained relatively consistent during optical flow modulation trials.

Each manipulation revealed different responses to increasing visual occlusion during regular and speed-perturbed walking. Greater visual occlusion was consistently associated with increased step width variability and stronger coupling between center of mass state and subsequent foot placement, whereas mean spatiotemporal measures remained relatively stable across conditions. These findings support graded visual manipulation in virtual reality as a method for assessing balance control and developing balance training protocols.

INTRODUCTION

Stable walking depends on the integration of visual, proprioceptive, and vestibular information, as well as the nervous system's ability to shift its reliance on these sensory inputs as environmental conditions change. Vision contributes to walking speed regulation, orientation, balance and anticipatory control, all of which are key for navigating the environment [1]. When visual information becomes degraded, the nervous system can reweigh sensory information and rely more heavily on proprioceptive and vestibular information to maintain balance during walking [2]. These adjustments may be reflected during walking in spatiotemporal parameters, postural control, and center-of-mass motion, although different measures may vary in their sensitivity to changes in visual information. Because everyday environments can reduce visual information through conditions such as darkness, fog, or low contrast, understanding how the nervous system maintains walking stability as visual information becomes less available is important for characterizing balance control and developing rehabilitation approaches that account for sensory limitations.

The contribution of vision to walking can be examined by manipulating optical flow, which is the perceived motion of the environment relative to the user as they move through the environment. When the speed of the visual scene does not match the walker's physical speed, visual information about self-motion conflicts with proprioceptive and vestibular feedback. This sensory mismatch can alter gait control as the nervous system adjusts its reliance on the available sensory information. Previous studies have shown that manipulating optical flow relative to treadmill speed can increase gait variability even when average strides, step lengths, and step widths remain relatively unaltered [3]. Experimentally controlling optic flow independently of physical walking speed therefore provides a way to examine how visual information contributes to gait regulation.

Virtual environments make visual and physical speed manipulations possible with precision and repeatability that natural environments do not permit. Virtual environments are increasingly being used to study human locomotor adaptations during treadmill walking, allowing users to walk through simulated scenarios that otherwise would not be feasible [4][5]. Because the visual scene is rendered independently of treadmill motion, optical flow can be decoupled from physical speed, creating a mismatch between what the walker sees and experiences. Scene motion can be oscillated at prescribed amplitudes and frequencies [4], while depth cues can be degraded in graded increments to control the visual information available to the user. Head-Mounted Displays (HMDs) can also influence the visual information available. A restricted field of view and stereoscopic mismatch between the rendered scene and natural vision can cause users to alter gait parameters even in the absence of perturbations [6]. Reduced depth cues and VR scene design can limit the information available to the central nervous system (CNS) when planning foot placement [7], making virtual environments (VEs) a useful means of experimentally controlling depth information.

Visual perturbations can reveal gait and balance deficits that are not apparent during unperturbed walking [8]. For example, individuals with movement deficits, such as chronic ankle instability, show greater postural instability and altered muscle activation when compared with individuals without impairment when visual input is altered during standing and overground walking. This suggests that individuals with ankle instability rely more heavily on vision for balance [9]. During walking with perturbed optical flow, they also exhibit greater variability in step length, step width, and mediolateral sacrum motion, along with shorter average step lengths [10]. People with multiple sclerosis show a similar pattern, exhibiting balance impairments during optic flow perturbations that are not apparent during unperturbed walking [11], [12]. These findings suggest that challenging visual conditions may help identify deficits in balance control that are otherwise difficult to detect.

Walking is regulated through multiple control objectives, including speed, balance, foot placement accuracy, and energetic cost [14]. Perturbations can affect these objectives differently, and the resulting gait responses depend on the specific demands imposed. For example, perturbed walking often increases spatiotemporal variability, particularly step width variability, which can increase energetic cost and reduce gait economy [13]. Visual perturbations alone can produce similar responses by creating a perceived threat to balance without physically destabilizing the walker. Some gait measures also scale with perturbation intensity; for example, step width variability increases with stronger visual perturbations [14], whereas foot placement error responds more directly to explicit foot placement demands. These responses may also vary across individuals, particularly as task difficulty increases. Graded visual occlusion therefore provides a way to examine how different aspects of gait control respond as visual information becomes progressively less available.

What remains less understood is how the amount of visual information available, rather than its accuracy, affects balance control during walking. Existing studies have primarily manipulated the accuracy of visual information through optic flow perturbations or scene oscillations, or removed visual input entirely by comparing eyes-open and eyes-closed conditions. This distinction is important because real-world environments often reduce the amount of visual information available without eliminating it completely. Graded visual occlusion has rarely been combined with controlled perturbations, limiting our understanding of how reduced visual information influences responses to both visual and physical disturbances during walking. Examining these conditions together may help determine whether progressively reducing visual information alters the strategies used to maintain balance.

To understand how reduced visual information affects walking, measures are needed that capture both gait behavior and dynamic balance control. While spatiotemporal parameters describe how a person walks, they do not fully characterize how balance is maintained step to step. Balance control is particularly important in the mediolateral direction, where stability depends strongly on appropriate foot placement. Because stability depends not only on the position of the center of mass (CoM) but also on its velocity, CoM position alone is insufficient for assessing dynamic balance. Measures that account for both where the body is and where it is moving are therefore needed to characterize how walkers respond to visual and physical perturbations. Two measures that can provide complementary information about balance control are margin of stability (MoS) and foot-placement control. The margin of stability is defined as the distance at heel strike between the boundary of the base of support and the extrapolated center of mass, which incorporates both CoM position and velocity [15]. This measure reflects whether the placement of the foot provides a sufficient margin to maintain dynamic stability. In populations with motor impairments, destabilizing walking conditions have been associated with reductions in MoS [11], [12], [16].

MoS describes the outcome of each step but does not explain how that foot placement was selected. Quantifying foot placement control can provide this additional information by quantifying the relationship between the mechanical state of the body during stance and the location of the subsequent foot strike. In unimpaired walking, center-of-mass position and velocity at midstance can predict a substantial portion of the variability in subsequent foot placement [17]. Examining this relationship may therefore reveal changes in how walkers regulate lateral foot placement when visual information is reduced.

The purpose of this study was to determine how graded visual occlusion affects gait variability and dynamic balance control during treadmill walking in virtual reality. Participants completed nine 5-minute trials in a 3×3 design combining three walking conditions - baseline walking, optical flow modulation, and treadmill speed modulation - with three levels of visual occlusion. Each trial was evaluated against the unperturbed baseline to isolate the combined effects of perturbation type and visual occlusion levels, and trials were additionally compared within each condition to isolate the effect of occlusion alone. We hypothesized that mean spatiotemporal parameters, including average step width and step lengths, would remain relatively unchanged, whereas their variability would increase with increasing occlusion in the three trial types. Because visual manipulations were applied in the anterior-posterior direction, we also examined how fore-aft visual and gait speed perturbations influenced mediolateral balance control similar to previous studies examining walker behavior in VR [18], [19]. We further hypothesized that greater visual occlusion would reduce dynamic stability, reflected by lower margin of stability and greater MoS variability. Finally, we hypothesized that visual occlusion would increase the coupling between center of mass state and subsequent foot placement, reflecting greater reliance on nonvisual sensory information for estimating body motion.

METHODS

Participants

We recruited twelve young unimpaired adults for this study. Two participants did not complete all nine experimental trials, and data from two additional participants were excluded due to excessive marker dropout that prevented reliable processing of their trials. Data from the remaining eight participants are included in the analysis (N = 8, Age 20.6 ± 2.8 years, Height: 165.7 ± 10.2 cm, Weight: 66.5 ± 12.5 kg, Leg Length 861.7 ± 71.3 mm). Inclusion criteria included no self-reported visual impairments affecting daily walking function, with exclusion criteria including any neurological conditions, musculoskeletal impairments, pregnancy, or somatosensory deficits. The study was approved by the University of California, Irvine Institutional Review Board, and all participants provided written informed consent.

Experimental Setup:

A.) Visual Conditions

The virtual environment was developed in Unity (Unity Technologies, San Francisco, CA) and presented to users using a head-mounted display (Bertec, Columbus, OH). The headset allowed us to precisely manipulate visual depth perception and anteroposterior optical flow. The environment consisted of a straight walking path containing randomly distributed objects that provided spatial and distance cues. Forward translation of the virtual scene was coupled to the treadmill belt speed. To simulate varying levels of visual occlusion, we implemented a fog-based visual shader in Unity that progressively reduced scene contrast and the visibility of environmental features. This manipulation reduced texture gradients and the visibility of nearby objects, thereby reducing the visual information available for estimating depth.

B.) Data Collection

Participants walked on a split-belt instrumented treadmill while we recorded kinematic and ground reaction force data. We captured whole-body kinematics using an 8-camera motion capture system. Twenty-one reflective markers were placed bilaterally on the shoulders, anterior and posterior superior iliac spines, thighs, knees, tibias, ankles, heels, and toes, with an additional marker over the C7 vertebra. We used Kinamoto software (Bertec, Columbus, OH) to control treadmill speed and virtual reality settings and Vicon Nexus (Vicon Motion Systems Ltd., Oxford, UK) to record marker trajectories and ground reaction forces. We then exported the data to MATLAB for further processing and calculation of kinematic variables, including joint positions and joint angles.

Experimental Protocol

Each participant completed nine 5-minute walking trials comprising three visual occlusion levels (None, Moderate, High) and three different trial types (baseline walking, optical flow manipulation, and speed modulation), for a total of 45 minutes of virtual-reality walking.

During baseline walking trials, participants walked at 1.0 m/s with the optical flow matched to the treadmill speed. During optical flow trials, participants continued to walk at 1.0 m/s while we sinusoidally varied the relative translation speed of the visual environment between 0.5x to 2.0x the physical treadmill speed. The perturbation acted in the anteroposterior direction and created a mismatch between visually perceived information and actual walking speed. The sinusoidal pattern of visual flow speed repeated every 2.5 minutes for two full cycles in the 5-minute trial. While other studies have employed frequencies up to 0.25 Hz [18], the perturbation rate used here allows participants to experience multiple types of flow within each trial without too much variation. Fore-aft perturbations induced by the optical flow changes were applied as a translational motion parallel to the walking motion, creating a sensory mismatch between perceived speed and actual speed.

Finally, during the speed modulation trials, the optical flow remained matched to treadmill speed while we varied treadmill speed between 0.7 m/s and 1.4 m/s according to a prescribed sinusoidal waveform. This was done to observe how visual occlusion affected participant ability to match changes in physical speed.

Before each trial, we informed participants of the upcoming condition and instructed them to walk in a straight line for the full 5 minutes of the trial. Participants wore a body-weight support harness and standardized running shoes throughout testing and were given rest breaks between trials. Participants could discontinue participation at any time. Before and after the walking trials, participants completed a Motion Sickness Questionnaire to assess symptoms associated with exposure to virtual-reality.

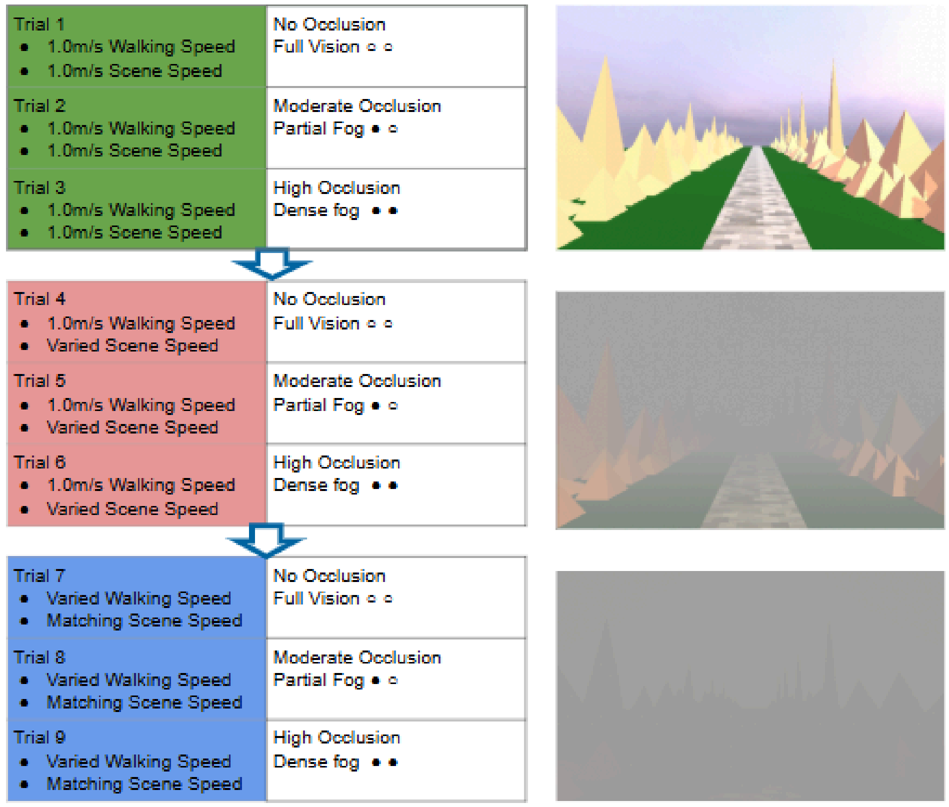

Figure 1: Experimental paradigm and virtual environments

Participants completed a 3×3 experimental design consisting of three trial types (baseline walking, optical flow manipulation, and treadmill speed modulation) under three levels of visual occlusion (none, moderate, and high). Seventy-two trials from eight participants were included in the analysis.

Data Processing:

We processed biomechanical data in MATLAB 2024b (MathWorks, Natick, MA) to quantify gait responses across walking conditions. We gap-filled missing marker trajectories in Vicon Nexus and used only the last 2.5 minutes of each trial for analysis. All marker trajectories were low-pass filtered using a fourth-order Butterworth filter with a 6 Hz cutoff frequency.

We relied on gait events to separate gait data into strides. We identified heel strikes using a custom MATLAB algorithm based on estimated sacrum and heel marker positions. Since our marker set did not include a sacrum marker, we estimated sacrum position as the midpoint between the left and right posterior superior iliac spine markers. For each limb, we computed the effective leg angle between the estimated sacrum and each heel marker in the sagittal plane and identified heel strike from the local minimum of this angle. This minimum corresponded to the point at which the heel reached its maximal anterior position relative to the pelvis.

We calculated step length and step width from the anteroposterior and mediolateral distances, respectively, between the left and right heel markers at each heel strike. We normalized both measures to leg length, which we estimated from the maximum vertical distance between the ipsilateral anterior superior iliac spine and ankle markers averaged across each of their nine trials. Step length and step width variability were quantified by taking the standard deviations of the step length and step widths respectively within each trial.

We calculated mediolateral margin of stability (MoS) following Hof et al. (2005) [15]. Center of mass (CoM) position was estimated as the midpoint of the posterior superior iliac spine markers. CoM velocity was calculated as the first derivative of CoM position and scaled by the sampling rate. We then calculated extrapolated center of mass as

$$X_{CoM} = x_{CoM} + \frac{v_{CoM}}{\omega_0}$$

where $\omega_0 = (g/L)^{0.5}$, $g = 9810\ mm/s^2$, and L represents participant leg length. MoS was calculated at each heel strike as the mediolateral distance between the XCoM and the lateral boundary of the base of support, with positive values indicating stable foot placement within the base of support.

We assessed foot-placement control using a linear regression model adapted from Wang and Srinivasan (2014) [17]. At mid-stance of each step, we extracted mediolateral CoM position relative to the stance foot position (p), as well as mediolateral CoM velocity (v). We then defined subsequent foot placement (z) as the mediolateral distance between the swing foot and stance feet at the following heel strike and fit the model:

$$z = b0 + (b_{pos} \times p) + (b_{vel} \times v)$$

The coefficient of determination (R^2) quantifies the proportion of variation in subsequent foot placement that is explained by the CoM state at midstance. The position gain (b_{pos}) reflects the degree to which the CoM position relative to the stance foot drives foot placement on the subsequent step, and the velocity gain (b_{vel}) reflects the degree to which mediolateral CoM velocity drives foot placement on the subsequent step.

For stride-level analyses of CoM motion, we estimated CoM position from the pelvis markers. We defined each stride from one right heel strike to the subsequent right heel strike and time-normalized each stride to 100 data points. We then evaluated mediolateral and anteroposterior CoM trajectories across strides to characterize within-trial gait consistency and variability. This allowed us to monitor gait stability over time and to evaluate how participants varied over the duration of the walking trial.

Statistical Analysis

All statistical analyses were performed in MATLAB 2024b (MathWorks, Natick, MA). Nine outcome measures were analyzed including mean step length, mean step width, step length variability, step width variability, mean margin of stability, margin of stability variability, foot placement R^2, and foot placement position and velocity gains. All spatiotemporal measures were normalized to leg length before analysis.

Two sets of comparisons were performed using two-tailed paired samples t-tests, treating each participant as their own control. The first compared every trial against the first baseline trial with no occlusion (BL1), isolating the combined effects of the trial type and level of visual occlusion. The second compared trials within each condition, isolating the effect of visual occlusion alone. Only participants with valid data in all trials were included in the comparison. Statistical significance was set at $\alpha = 0.05$, and a Bonferroni correction was applied within each measure by multiplying each p value by the number of comparisons performed in that set. For each comparison, we computed the group mean and standard deviation of the trials, the uncorrected and corrected p values, and Cohen's d for paired samples, calculated as the mean of paired differences divided by the standard deviation of those differences, where values of 0.2, 0.5, and 0.8 denote small, moderate, and large effects respectively. Unless otherwise stated, p values reported in the Results are uncorrected. With eight participants, these comparisons had approximately 80% power to detect effects of $d \geq 1.13$, which was calculated for the two-tailed test at the specified α with seven degrees of freedom. Effect sizes are reported as d alongside p values throughout the Results, including for comparisons that did not reach significance.

RESULTS

A.) Spatiotemporal Gait Parameters

During the baseline walking trial with no visual occlusion (BL1), participants walked with a normalized step length of 0.637 ± 0.047, step width of 0.152 ± 0.039, step length variability of 0.0297 ± 0.0078, and step width variability of 0.0334 ± 0.0047. As visual occlusion increased across BL2 and BL3 (Moderate and High occlusion levels, respectively), mean step length and step width remained relatively consistent. Step length variability also changed little across occlusion levels, whereas step width variability increased by about 6.0% from BL1 to BL3 ($d = 0.33$, $p = 0.38$). No spatiotemporal measure differed significantly from BL1 and no comparison between baseline trials reached significance (all $p \geq .13$).

During optical flow modulation, mean step length and step width again remained relatively consistent across occlusion levels. Step length variability decreased by approximately 12.6% ($d = -0.97$, $p = 0.029$) while step width variability increased by approximately 9.1% from OFM1 to OFM3 ($d = 0.62$, $p = 0.12$). The pattern observed during optical flow modulation was similar to that observed during baseline walking, with the clearest change occurring in step width variability as visual occlusion increased.

During speed modulation, step width variability showed the largest increase with visual occlusion, increasing by approximately 16.4% from SM1 to SM3 ($d = 1.19$, $p = .012$). Step width variability at SM1 did not differ from BL1 ($d = 0.11$, $p = 0.77$), indicating that this increase was associated with the visual occlusion level rather than the speed of the treadmill itself. Step length variability was also much higher during speed modulation than during the baseline and optical flow modulation conditions; however, this is a mechanical consequence of the imposed treadmill oscillation rather than a change in motor variability. Mean step width remained relatively unchanged across the three occlusion levels, whereas mean step length decreased by 1.5% from SM1 to SM3 (d = -1.00, p = 0.025).

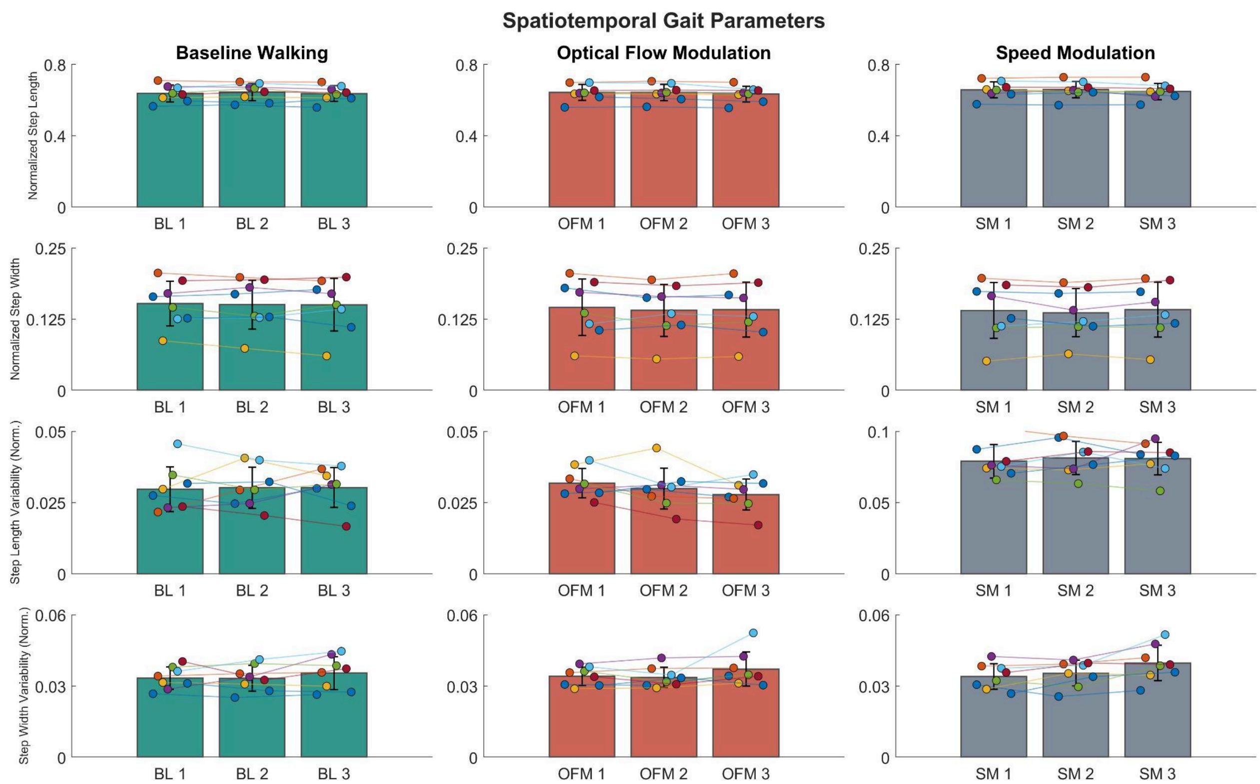

Figure 2: Spatiotemporal gait parameters across walking conditions

Group Average (Standard Deviation) normalized step lengths (Row 1), normalized step width mean (Row 2), step length variability (Row 3), and step width variability (Row 4) for baseline walking, optical flow modulation, and speed modulation conditions. Step lengths and step widths are normalized to leg length. Colored dots represent individual participants across trials to illustrate within subject trends.

Table 1: Spatiotemporal gait parameters across nine trials

Trial	Step Length	Step Width	SL Variability	SW Variability
BL 1	0.637 ± 0.047	0.152 ± 0.039	0.0297 ± 0.0078	0.0334 ± 0.0047
BL 2	0.645 ± 0.049	0.150 ± 0.043	0.0302 ± 0.0072	0.0333 ± 0.0054
BL 3	0.637 ± 0.044	0.150 ± 0.046	0.0303 ± 0.0070	0.0354 ± 0.0069
OFM 1	0.643 ± 0.046	0.146 ± 0.049	0.0318 ± 0.0051	0.0341 ± 0.0039
OFM 2	0.643 ± 0.046	0.140 ± 0.045	0.0299 ± 0.0071	0.0337 ± 0.0042
OFM 3	0.634 ± 0.044	0.142 ± 0.048	0.0278 ± 0.0054	0.0372 ± 0.0072
SM 1	0.658 ± 0.046	0.140 ± 0.049	0.0790 ± 0.0117	0.0341 ± 0.0053
SM 2	0.659 ± 0.046	0.136 ± 0.042	0.0813 ± 0.0117	0.0354 ± 0.0054
SM 3	0.648 ± 0.046	0.142 ± 0.048	0.0809 ± 0.0114	0.0397 ± 0.0074

Values reported as group mean ± SD (N = 8), normalized to leg length. BL = Baseline Walking, OFM = Optical Flow Modulation, SM = Speed Modulation. Trials 1,2 and 3 correspond to no occlusion, moderate occlusion, and high occlusion respectively.

B.) Margin of Stability

Across all conditions, mean mediolateral margin of stability (MoS) generally ranged from approximately 41.6 to 48.6 mm, indicating that the foot typically landed several centimeters lateral to the extrapolated CoM (Figure 3). Individual responses varied considerably, with participant-level mean MoS values ranging from approximately 23.7 to 78.7 mm, reflecting differences in stance width and walking strategy. Within each trial type, mean MoS remained relatively stable across occlusion levels, with trial-to-trial changes no greater than approximately 2.1%.

During baseline walking, mean MoS changed little as visual occlusion increased from BL1 to BL3 ($d = 0.12$, $p = 0.75$). MoS variability, however, increased by approximately 5.9% from BL1 to BL3 ($d = 0.47$, $p = 0.23$). Thus, although the average margin of stability remained relatively constant, step-to-step fluctuations in MoS increased as visual information was reduced.

During optical flow modulation, mean MoS also remained relatively stable across occlusion levels and was similar to that observed during baseline walking. MoS variability increased by approximately 4.3% from OFM1 to OFM3 ($d = 0.42$, $p = .27$). This pattern was similar to baseline walking, with increasing visual occlusion associated primarily with greater variability rather than a change in mean MoS. While none of the comparisons made were significant, MoS variability was higher than BL1 for all other baseline and optical flow modulation trials, and every condition had the highest MoS variability at higher occlusion levels.

During speed modulation, mean MoS was lower than in any baseline walking or optical flow modulation trial. MoS variability was also substantially higher, exceeding that observed in the other two trial types by approximately 23.4% to 27.3% and also increased from SM1 to SM3 ($d = 0.86$, $p = .046$). Both the perturbation and the occlusion contributed, with the perturbation effect the larger of the two, such that the difference between SM1 and BL1, which isolates the speed perturbation from visual occlusion ($d = 1.33$), was larger than the difference between SM3 and SM1, which isolated occlusion within the perturbation ($d = 0.86$).

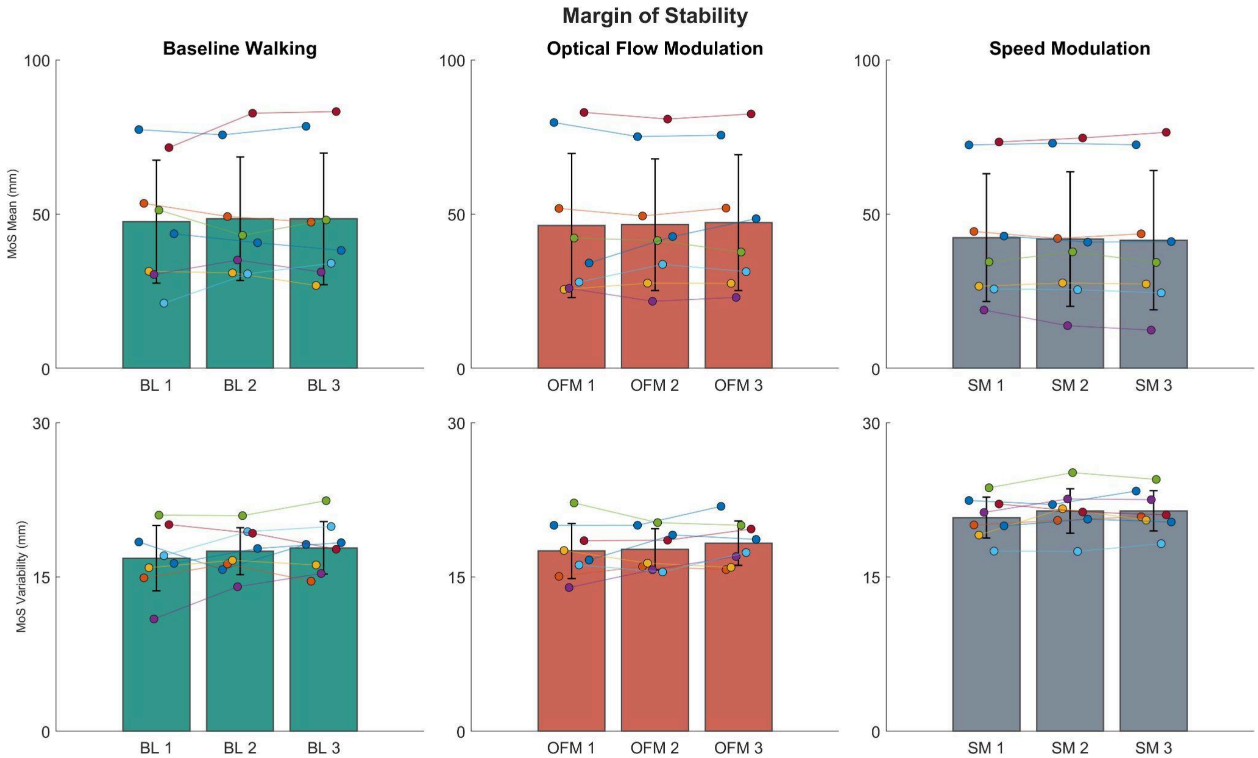

Figure 3: Mediolateral margin of stability across nine trials

Mediolateral margin of stability across nine trials (3 conditions x 3 occlusion levels) Top Row: MoS Mean (mm). Bottom Row: MoS Variability (mm). Bars represent group mean ± SD. Individual participant data are shown as connected dots. MoS was computed at each heel strike as the distance between foot position and extrapolated center of mass.

Table 2: Mediolateral margin of stability across nine trials

Trial	MoS Mean (mm)	MoS Variability (mm)
BL 1	47.60 ± 19.93	16.83 ± 3.17
BL 2	48.56 ± 20.01	17.50 ± 2.27
BL 3	48.50 ± 21.32	17.83 ± 2.54
OFM 1	46.34 ± 23.39	17.52 ± 2.68
OFM 2	46.62 ± 21.29	17.69 ± 2.01
OFM 3	47.31 ± 21.99	18.28 ± 2.15
SM 1	42.40 ± 20.71	20.76 ± 1.99
SM 2	41.99 ± 21.76	21.42 ± 2.16
SM 3	41.58 ± 22.60	21.42 ± 1.96

Values reported as mean ± SD in mm. MoS computed at heel strike following Hof [15].

C.) Foot Placement Control Predictability

Foot-placement control, quantified by the coefficient of determination R^2, position gain (b_{pos}), and velocity coefficient (b_{vel}), differed across trial types (Figure 4; Table 3).

During baseline walking, R^2 progressively increased across trials, indicating that midstance CoM state explained a greater proportion of subsequent foot-placement variability as visual occlusion increased. The b_{vel} coefficient showed similar increases, indicating a stronger relationship between mediolateral CoM velocity and subsequent foot placement across baseline trials.

During optical flow modulation, R^2 remained relatively consistent across trials and did not show the progressive increase observed during baseline walking ($d = 0.08$, $p = 0.84$). The b_{vel} coefficient did not increase monotonically, but was 16.4% higher at the highest occlusion level than the lowest. Thus, optical flow modulation altered the progression of foot-placement control observed during baseline walking, even though mean foot-placement measures remained relatively unchanged. The effect size for foot placement predictability under optical flow was negligible compared to the other trial types (Table 4).

During speed modulation, R^2 was initially lower than in the other trial types but increased most strongly across trials. The b_{vel} coefficient followed a similar pattern, increasing across the speed-modulation trials. These results indicate that the relationship between CoM state and subsequent foot placement strengthened across repeated speed-modulation trials despite weaker coupling in the initial condition. These increases were significant from SM1 to SM3 after correction for R^2 and b_{vel} ($d = 1.42$ and 1.89 respectively).

Comparing the highest to the lowest occlusion level within each condition isolates the effect of reduced visual information from the effect of the perturbation itself. Across all nine outcome measures, these effects were non-significant during baseline walking and optical flow modulation while the largest effects occurred during speed modulation (Table 4).

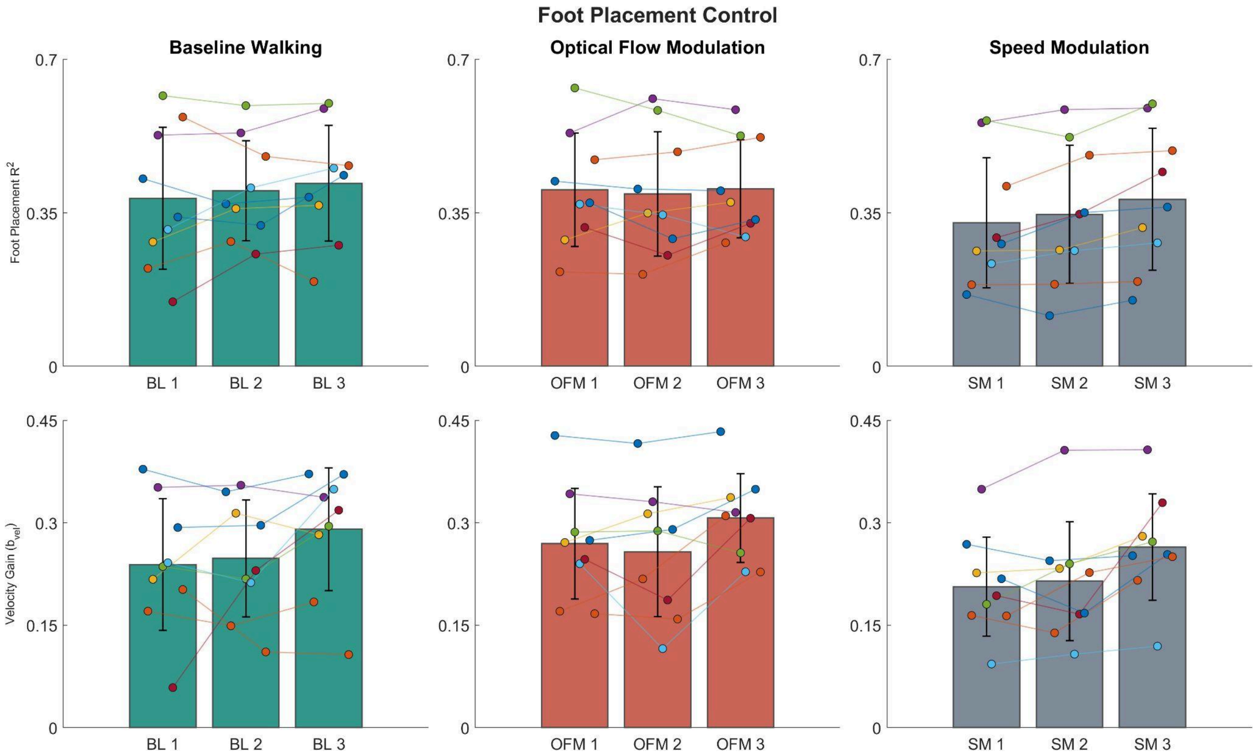

Figure 4: Foot placement control across nine trials

Top Row: Coefficient of Determination (R^2) from Linear Regression Model. Bottom Row: Velocity Gain. Values reported as mean ± SD, individual participant data shown as connected dots.

Table 3: Foot placement control across nine trials

Trial	FP R^2	Velocity Gain (b_vel)
BL 1	0.377 ± 0.172	0.221 ± 0.086
BL 2	0.404 ± 0.121	0.236 ± 0.083
BL 3	0.421 ± 0.140	0.281 ± 0.090
OFM 1	0.400 ± 0.138	0.250 ± 0.059
OFM 2	0.391 ± 0.151	0.238 ± 0.079
OFM 3	0.406 ± 0.119	0.291 ± 0.048
SM 1	0.348 ± 0.144	0.199 ± 0.073
SM 2	0.375 ± 0.140	0.211 ± 0.092
SM 3	0.410 ± 0.146	0.266 ± 0.083

Values reported as mean ± SD. R^2 reflects the proportion of foot placement variance explained by CoM at midstance. Velocity gain reflects the degree to which the CoM velocity at midstance drives subsequent foot placement.

Table 4: Effect sizes for within-condition occlusion comparisons across nine trials

Measure	Baseline Walking	Optical Flow Modulation	Speed Modulation
Step Length Mean	-0.03	-0.61	-1.00
Step Length Variability	0.07	-0.97	0.18
Step Width Mean	-0.14	-0.43	0.14
Step Width Variability	0.33	0.62	1.19
MoS Mean	0.12	0.16	-0.30
MoS Variability	0.47	0.42	0.86
Foot Placement R^2	0.50	0.08	1.42
Foot Placement b_{pos}	0.32	-0.02	1.02
Foot Placement b_{vel}	0.57	0.70	1.89

Values are Cohen's d for paired samples (N = 8) comparing the highest occlusion trial (BL3 vs BL1, OFM3 vs OFM1, SM3 vs SM1) against the no-occlusion trial within each condition. Positive values indicate an increase with greater occlusion. Values of 0.2, 0.5, and 0.8 correspond to small, moderate, and large effects respectively. At this sample size, the study could reliably detect effects of $d \geq 1.13$.

DISCUSSION

This pilot study investigated how visual occlusion, optical flow modulation, and treadmill speed modulation influenced spatiotemporal parameters, margin of stability (MoS), and foot-placement control during treadmill walking in virtual reality. Across conditions, mean spatiotemporal measures changed relatively little, whereas measures of gait variability and foot-placement control were more sensitive to the experimental manipulations. Step width variability increased with greater visual occlusion across all three trial types, and MoS variability increased modestly during baseline walking and optical flow modulation. Speed modulation produced the lowest mean MoS and the greatest MoS variability overall. Foot-placement control also differed across trial types: the relationship between CoM state and subsequent foot placement strengthened across baseline and speed-modulation trials but remained comparatively stable during optical flow modulation. The strengthening observed was strongest for speed modulation and produced the only foot placement comparisons to survive correction for multiple comparisons. Together, these findings suggest that visual and gait speed perturbations influenced the variability and control of walking more strongly than average stepping behavior.

Changes in visual occlusion during treadmill walking had little effect on mean step length or step width, with step width variability increasing from the lowest to highest occlusion level by 6.0% during baseline walking, 9.1% during optical flow modulation, and 16.4% during speed modulation. The effect sizes for these increases were ordered similarly ($d = 0.33$, 0.62, and 1.19 for BL1 → BL3, OFM1 → OFM3, and SM1 → SM3 respectively), consistent with the hypothesis that reducing visual information increases mediolateral stepping variability and that the effect is amplified when physical perturbations are imposed concurrently with the occlusion.

This pattern is consistent with the previously reported contribution of vision to mediolateral balance control [18]. Vision plays a central role in motor control and gait regulation, providing continuous information about limb position, and the environment that enables people to evaluate their own priorities [7]. When visual information becomes less reliable, participants may depend more heavily on vestibular and proprioceptive information and adjust foot placement from step to step to maintain balance [2]. The relatively stable mean step width alongside increasing step width variability suggests that participants did not systematically adopt a wider or narrower gait but instead exhibited greater fluctuations in lateral foot placement. These changes may reflect greater uncertainty in lateral balance control when visual information is degraded rather than a consistent shift in the average stepping strategy [20].

Margin of stability showed a different pattern across perturbation types. Speed modulation produced lower mean MoS and greater variability than BL and OFM trials. Because the extrapolated CoM incorporates CoM velocity, changes in treadmill speed can directly alter the relationship between the CoM and the base of support. The lower MoS observed during speed modulation may therefore reflect the increased challenge imposed by continuously changing walking speed, while the greater MoS variability suggests larger step-to-step fluctuations consistent with ongoing demand on balance control and necessary corrective adjustments during the trial. In contrast, mean MoS remained relatively stable during baseline walking and optical flow modulation, even as MoS variability increased modestly with greater visual occlusion. Thus, reduced visual information appeared to influence the consistency of dynamic stability more than its average magnitude in the absence of a physical speed perturbation. The reduction in mean MoS during speed modulation was consistent across all three trials but was not found to be significant with the given sample size.

Foot placement control also differed across trial types. During baseline walking and speed modulation, R^2 progressively increased across trials, indicating that midstance CoM state explained a greater proportion of subsequent foot-placement variability. The corresponding increases in b_{vel} indicate that mediolateral CoM velocity also became more strongly associated with subsequent foot placement. In contrast, R^2 showed no progressive increase during optical flow modulation, with an effect size of $d = 0.08$ against $d = 1.42$ during speed modulation. One possible interpretation is that altering optical flow interfered with the development of a more consistent relationship between CoM state and lateral foot placement. However, this interpretation should be treated cautiously because the experimental design does not allow changes across trials to be attributed specifically to adaptation or visual occlusion.

A limitation of this study was statistical power. With only eight participants, these comparisons had approximately 80% power to detect effects of $d \geq 1.13$. Many of the effect sizes fall below this threshold, so non-significant results should be understood as undetected rather than absent. Due to this being a pilot study, pairwise comparisons were used rather than a factorial analysis of variance. A second limitation is that trial order was not randomized and participants completed the three baseline walking trials first, then optical flow modulation, then speed modulation. Occlusion level was also not randomized within each condition, always progressing from none to moderate to high. Fatigue and adaptation therefore accumulated differently across the protocol and could affect exposure to later trials differently, meaning that progressive increases in foot placement R^2 values cannot be interpreted unequivocally as adaptation to either visual occlusion or the perturbations themselves. Likewise, the absence of a similar trend during optical flow modulation may reflect the perturbation itself, its position in the protocol, or both. The virtual environment may not have produced a perfect one-to-one correspondence between visual translation and physical motion, potentially introducing unintended discrepancies between perceived and actual movement. Finally, the sample consisted of eight healthy young adults, which limits the statistical findings and leaves generalizations to the clinical populations untested. The results reported here, therefore, should not be generalized to specific populations. Future work should consider reordering the trial types and visual occlusion levels to distinguish adaptation from order effects and include clinical populations.

Overall, these preliminary results suggest that gait variability and foot-placement control may be more sensitive than mean spatiotemporal measures to degraded visual information and perturbations during treadmill walking. Visual occlusion was most consistently associated with increased step width variability, whereas treadmill speed modulation produced larger changes in dynamic stability. The distinct pattern observed during optical flow modulation further suggests that visual motion cues may influence how participants coordinate CoM motion and subsequent foot placement, although a randomized experimental design will be necessary to isolate these effects.

CONCLUSION

Manipulation of visual conditions during VR walking provides a controlled framework for isolating how the CNS integrates sensory information during gait. These findings may help inform the design of future VR-based rehabilitation protocols and fall prevention strategies for populations with visual or sensory impairment. Optical flow and speed perturbations revealed differences in balance and gait behavior, reflected by higher step width and MoS variability. Foot placement predictability further showed that participants progressively strengthened their coupling between CoM state and foot placement across walking and speed modulation trials, but this adaptation was flat during the optical flow trials as vision became more occluded. One explanation for this difference is sensory conflict, in which optical flow perturbations provided the CNS with visual information that contradicted bodily motion and disrupted the reliability of visual feedback, although the present design cannot test this directly.

Targeting this reliance through visual perturbations in virtual reality is therefore a promising approach for assessing walking balance deficits in people with movement disorders compared to healthy subjects. These results indicate the need for larger studies with a wider range of ages, movement disorders, and balance impairments with additional opportunities for incorporating new perturbation training paradigms to improve walking balance and gait integrity.

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