CONTRIBUTION OF MODIFIED VISUAL GAIN TO HUMAN BALANCE CONTROL DURING QUIET, UPRIGHT STANCE

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Abstract

Visual feedback provides critical information to support postural stability. Previous work has shown that magnifying visual feedback indirectly can improve postural control, such as by providing individuals with biofeedback during balance tasks. When studies have manipulated vision directly, the conditions have been restricted to include an absence of visual feedback and sway referenced paradigms. Therefore, this thesis aimed to understand how the gain of optic flow contributes to balance control during quiet, upright stance among healthy adults. Optic flow was amplified or reduced relative to head motion while participants stood quietly on either a firm or foam surface. Overall, when there was an increased reliance placed on the visual system by standing on foam, a tighter regulation of upright stance was observed as optic flow gain increased. Further, this thesis provided evidence that visual contributions to balance control may extend to higher frequencies of postural sway than previously theorized.

Dedication

The work within this thesis is dedicated to my grandparents for igniting my curiosity and motivation to pursue balance research. Thank you for being my biggest cheerleaders, whether presently or in spirit.

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List of Abbreviations

AP: Anteroposterior

BOS: Base of support

COM: Center of mass

COP: Center of pressure

EMG: Electromyography

HeadPos: Head position

HMD: Head-mounted display

ML: Mediolateral

MPF: Mean power frequency

mVel: Mean velocity

RMS: Root mean square

VEPR: Visually evoked postural response

VR: Virtual reality

Chapter 1: Introduction

1.1 Overview of thesis

Maintaining balance control while standing is crucial for humans to remain upright and avoid a fall. To achieve postural stability, we require input from multiple sensory systems to gain an accurate representation of our position and orientation relative to the environment. Specifically, visual feedback offers a valuable source of sensory information to support balance control while standing; however, it currently remains unclear exactly how changes in optic flow contribute to quiet standing balance behaviour. Therefore, furthering our knowledge on the relationship between the gain of optic flow and human standing balance control is crucial. Chapter 1 of this thesis provides a literature review of optic flow, quiet standing balance control, and the relationship between vision and posture. Chapter 2 outlines the methodology used to study the effect of optic flow gain on quiet stance. Finally, Chapters 3 and 4 present the findings of this thesis and consolidate them in the context of the current literature.

1.2 Optic flow

The visual system provides critical sensory feedback for detecting objects, carrying out action, and navigating through an environment. Within an environment, varying patterns of light are constantly reflected by external surroundings and processed by the visual system; this creates an optic array which allows the observer to interpret information about the external world (Gibson, 1979). As the eye or environment is in a state of constant motion, the structure of the optic array reflected onto the retina changes over time and consequently generates what is known as optic flow (Gibson, 1979). Consequently, optic flow represents the retinal image shifts caused by relative motion between the environment and observer (Rogers, 2021). Optic flow provides crucial information used to perceive ego-motion, through supporting the observer in perceptually inferring self-motion relative to both stationary and moving objects in the environment (Luu, Zangerl, Kalloniatis, & Kim, 2021). Specifically, previous work has demonstrated that the strength of perceived self-motion is significantly influenced by changes in retinal motion patterns generated by optic flow (Seno et al., 2018).

1.3 Binocular visual processing and stereoscopic vision

Binocular visual processing involves interpreting a stimulus that is seen by both eyes simultaneously or in rapid succession (Howard & Rogers, 2008). Stereoscopic vision, a form of binocular visual cue, refers to the perception of the spatial structure of a three-dimensional environment (Howard & Rogers, 2008). The natural offset between the eyes in the head contributes to differences in image localization on the retina (Howard & Rogers, 2008). Specifically, when stimuli are projected onto the fovea of each eye, points of the stimuli that are closer or farther from the plane of fixation are projected onto different retinal locations (Howard & Rogers, 2008). The resulting disparities between these two monocular images provide information about the depth of a point, relative to the point of fixation (Howard & Rogers, 2008). Further, stereoscopic vision is valuable for spatial acuity, where acuity thresholds have been shown to be greater for patterns of images without disparity (Lappin & Craft, 2000).

1.4 Visual contributions to balance control

It is well known that the visual system provides necessary feedback to maintain postural stability during balance control tasks. By eliminating visual feedback through having individuals stand with their eyes closed, postural sway has been shown to increase by up to 40-50% compared to standing with eyes open (Diener, Dichgans, Bacher, & Gompf, 1984). Additionally, while exposing participants to intermittent visual feedback, such as through 4 Hz stroboscopic illumination, postural sway has been shown to increase relative to baseline (Assländer, Hettich, & Mergner, 2015).

Standard measures of vision are also predictive of an individual's level of postural stability and consequently, risk of suffering from falls and related injuries (Lord & Dayhew, 2001). Vision has been shown to be necessary for accurately assessing both distances and spatial relationships while adjusting postural control strategies. For example, poorer depth perception, as measured through the Howard-Dohlman depth perception test, can predict an individual's risk of suffering from multiple falls within a lifespan (Lord & Dayhew, 2001). Additionally, visual feedback provides necessary sensory information to detect and recognize objects during balance control tasks, also providing important information to coordinate an appropriate postural response. For example, one study has demonstrated that individuals with poor binocular low-contrast acuity have a greater risk of suffering from multiple falls when compared to those with normal visual acuity, while visual field loss in the lower periphery was also associated with fall risk (Lord & Dayhew, 2001).

Although it is well-known that vision is a key component to achieving postural stability, there is still no clear consensus on when vision is utilized during balance control tasks. Multiple studies have reported vision to be utilized during very slow frequencies of sway (less than 1 Hz) (Diener, Dichgans, Guschlbauer, & Bacher, 1986; Redfern, Yardley, & Bronstein, 2001). Further, work by Zettel et al. (2005) found no clear link between visual fixation and gaze shift with step placement while participants carried out rapidly generated stepping reactions; this suggests that vision may be too slow to influence such responses (Zettel, Holbeche, Mcilroy, & Maki, 2005).

1.5 Visual-based sensory integration and re-weighting

Visual information is constantly being integrated with feedback from the somatosensory and vestibular systems to detect slight deviations from upright stability and coordinate motor responses to correct for such deviations. Multi-sensory integration proves to be valuable in helping to increase the certainty of neural representations of body orientation and motion, while also providing an element of redundancy in case of sensory deficits (Peterka, 2018). Under a closed-loop feedback model, balance control is maintained by integrating feedback from these three sensory systems to generate an internal estimate of the body's orientation in space (Peterka, 2018). This internal estimate is then compared with the desired orientation outcome, which generates a sensory error (Peterka, 2018). From this, a neural controller is able to initiate and coordinate appropriate corrective torques proportional to the angular position and velocity experienced by the body (Peterka, 2018).

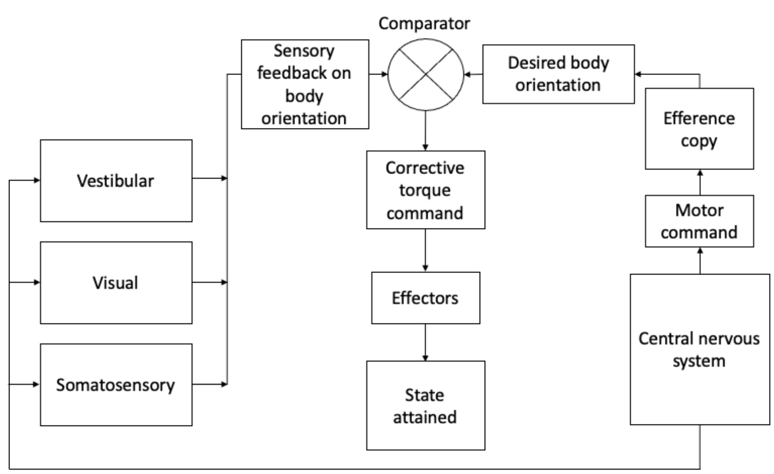

Figure. 1 Model of sensory integration to support upright stance (adapted from Peterka, 2018)

Corrective torque has previously been considered as both active and passive in nature; passive torque is generated by muscle tone without a time delay, such as that caused by muscle/tendon stretch during body sway, whereas active torque integrates negative feedback from sensory systems and involves sensory transduction with a time delay (Peterka, 2002). In addition to the negative feedback contributions from the three sensory systems, the sensory integration model proposed by Peterka (2018) illustrates the importance of positive feedback elicited through reactive torque, while Fitzpatrick et al. (1996) argue for the importance of feedforward predictive strategies in regulating a postural response (Fitzpatrick, Burke, & Gandevia, 1996). Further, information provided by torque, through contributors such as Golgi tendon organs, has been shown to relay lower frequency signals of sway (Peterka, 2018).

Different models have been proposed to estimate sensory re-weighting processes. In the Carver et al. (2006) model, a Kalman filter is used to generate an internal estimate of body position and velocity. This estimate is used to predict temporal asymmetry following changes to motion in the environment (Carver, Kiemel, & Jeka, 2006). As a result, re-weighting is probed to minimize the mean squared ankle torque defined by the neural controller (Carver et al., 2006). Previous work has supported this sensory re-weighting hypothesis through identifying a down-weighting of visual gain, as measured by a decrease in the amplitude of compensatory sway, in response to an increase in environmental motion (Oie, Kiemel, & Jeka, 2001, 2002). It was suggested that this response occurs in an effort to reduce the effect of sensory noise on sway and consequently, reduce the need for the neural controller to initiate corrective ankle torques (Carver et al., 2006; Jeka, Oie, & Kiemel, 2008).

Changes to sensory weighting is both variable and complex, with there being evidence that such re-weighting can be triggered through changes to external environmental conditions.

To measure instances of sensory re-weighting, it is common for studies to either withdraw or reduce one or more sensory inputs, or to induce a perturbation targeted to one system (Schut, Engelhart, Pasma, Aarts, & Schouten, 2017). For example, by having participants complete a quiet standing task on a compliant surface, such as foam, proprioception of the ankles becomes disturbed leading to a down-weighting of proprioceptive information; subsequent sensory re-weighting then occurs, placing greater reliance on the visual and vestibular systems (Schut et al., 2017).

It is also crucial to consider both the magnitude of gain changes and time scale over which changes to sensory weighting occur. For example, among the older adult population, there is evidence that time delays among the re-weighting process may drive sensory gain changes and, thus, changes to postural control strategies (Allison, Kiemel, & Jeka, 2006). To study gain changes with respect to time, Jeka et al. (2008) manipulated visual feedback through changing the amplitude of motion of a surrounding visual scene. As the amplitude of scene motion increased, the frequency of reactive sway consequently increased, accompanying a down-weighting of visual gain (Jeka et al., 2008). Researchers proposed that this was due to larger corrective ankle torques being highly sensitive to changes in the sensory re-weighting model (Jeka et al., 2008). Following this increase in scene motion, researchers then decreased the amplitude of motion of the visual scene and observed a slower up-weighting of visual gain (Jeka et al., 2008). It was proposed that because visual gain was already low, there was a smaller decrease in the frequency of sway and corrective ankle torques; the smaller ankle torques make the system less sensitive to changes in re-weighting causing a slower re-weighting response (Jeka et al., 2008). Therefore, the time constant involved in this re-weighting model depends on the previous starting level of gain, rather than the direction that gain changes. Further, work by

Peterka (2002) has suggested that there a saturation effect in response to increasing the amplitude of visual stimuli. Therefore sensory re-weighting can be considered to change as a function of stimulus amplitude, giving a quasi-linear relationship at each stimulus amplitude (Peterka, 2002).

Lastly, afferents from each sensory system have been shown to have different levels of sensitivity to higher level processing. For example, visually evoked postural responses appear to be cognitively modulated by expectation, such as when a participant is aware that their sense of vection represents external agency (Guerraz & Day, 2005). In contrast, the vestibular system has been shown to be less influenced by expectation (Guerraz & Day, 2005).

1.6 Vection and visually evoked postural responses

Linear vection, or illusory self-motion, occurs when an observer attributes movement from a scene as being self-generated, rather than externally generated (Wright, 2009). By experimentally inducing a perceived translation within an individual's surrounding environment, an internal conflict can be generated between sensory feedback provided by the vestibular and proprioceptive systems, which detect no change to one's position relative to the environment, and the visual system which detects a change in motion (Wright, 2009). When these systems are in conflict, it has been proposed that the central nervous system assumes that a component of gravity was exerted upon the utriculus to cancel the effects of linear acceleration, so the body feels like it is being inclined relative to its subjective vertical (Lestienne, Soechting, & Berthoz, 1977). As a result, a postural readjustment is elicited to realign the body with the perceived direction of gravity (Lestienne et al., 1977). When such a postural response is observed following the perception of self-motion, it is said that vection is elicited.

To observe the effects of vection on postural readjustments, Lestienne et al. (1977) projected an image moving linearly in the saggital plane onto the peripheral visual field of participants. It was observed that body inclination in the antero-posterior (AP) direction moved in the same direction as the motion of the visual scene (Lestienne et al., 1977). For example, if the visual stimulus moved in the backward direction, participants would perceive themselves to be moving forwards and consequently generate a postural readjustment in the backward direction (Lestienne et al., 1977). It was also shown that visually induced postural responses became in phase with the movement of the visual scene when the scene was moving a frequency between 0.2 - 0.25 Hz (Lestienne et al., 1977).

Further, temporal and spatial characteristics of the visual scene were examined for their contribution to vection-induced postural sway. The amplitude of participants' postural response was shown to be proportional to and dependent on the total area of the moving visual scene (Lestienne et al., 1977). Specifically, the relationship between the velocity of the visual stimulus and postural response was discovered to be non-linear and logarithmic in nature, where the amplitude of the postural readjustment depends logarithmically on the velocity of image motion (Lestienne et al., 1977). Through extrapolating this relationship, the minimum stimulus velocity needed to induce a sense of vection was discovered to be 0.02 m/s (Lestienne et al., 1977). Further, there was an observed plateau in the amplitude of the postural response when participants were exposed to visual stimuli moving at high velocities (greater than 1 m/s) (Lestienne et al., 1977). It was concluded that this saturation effect was a consequence of a limit in image motion perception, rather than biomechanical limitations about the ankle joint (Lestienne et al., 1977).

Visually evoked postural responses (VEPRs) counteract postural adjustments generated in response to vection in order to stabilize the body (Bronstein & Buckwell, 1997). Bronstein and Buckwell (1997) have provided evidence for environment-dependent VEPRs through having participants face a laterally moving visual scene while fixating either directly on the background, or on the background through a stationary object (window). While fixating directly on the background when there was no object present in the foreground, participants' VEPRs followed the direction of the visual motion; this allowed participants to stabilize themselves relative to the external reference (Bronstein & Buckwell, 1997). When fixating on the background through the window, the direction of the VEPRs was reversed, such that the direction of sway opposed the direction of background motion (Bronstein & Buckwell, 1997). From these results, researchers concluded that postural control is modulated by one's interpretation of the surrounding environment, rather than strictly being regulated by optokinetic reflexes, and that the direction of VEPRs is sensitive to motion parallax cues (Bronstein & Buckwell, 1997). Lastly, there is variation in the literature surrounding the latencies of visual evoked postural responses, with evidence of latencies ranging from less than 200ms up until 1s (Bronstein & Buckwell, 1997; Day, Muller, Offord, & Di Giulio, 2016).

1.7 Use of virtual reality in balance research

1.7.1 Virtual environmental design

Virtual reality (VR) technology can use a head mounted display (HMD) to immerse the user in a computer-generated stereoscopic virtual environment (Assländer & Streuber, 2020a). HMDs rely on online updating of the perceived scene using information about the motion of the user's head orientation and position in space (Assländer & Streuber, 2020a). One benefit of using

VR in behavioural experiments is being able to alter sensory feedback in a controlled environment that replicates a real-world setting; thus, VR has been praised for having a high level of ecological validity (Assländer & Streuber, 2020a). Different types of visual scenes have been used in VR experiments, such as photo-realistic scenes and abstract environments (Assländer & Streuber, 2020a). When designing a VR experimental paradigm that can be generalized to a real-world environment, it is important to optimize a sense of immersion, presence, and ecological relevance among the user to generate a high level of behavioural realism (Riecke, Schulte-Pelkum, Bülthoff, & der Heyde, 2006; Slater & Wilbur, 1997). To achieve this, factors such as technical specifications of the VR technology and visual properties of the developed virtual scene must be taken into account (Slater & Wilbur, 1997). For example, previous work by Riecke et al. (2006) has shown that VR-induced postural sway and the perception of vection is dependent upon how believable the visual stimulus is. Specifically, a direct correlation between spatial presence and vection was identified (Riecke et al., 2006).

Further, Assländer et al. (2020) found no significant differences between body sway induced through exposure to a VR photo-realistic environment compared to a real-world environment (Assländer & Streuber, 2020a). Thus, using virtual environments that replicate a real-world setting can induce adequate behavioural realism and create a greater sense of presence in comparison to abstract visual scenes. Similar conclusions were drawn by Riecke et al. (2006), where it was proposed that natural VR scenes provide the user with coherent pictorial depth cues and a spatial reference frame that increases the likelihood of perceiving visual motion as self-motion, rather than object motion (Riecke et al., 2006).

1.7.2 Use of virtual reality to induce VEPRs

Previous research has used VR technology to project a virtual scene around a participant and observe balance behaviour in response to manipulated visual motion (Greffou, Bertone, Hanssens, & Faubert, 2008). Greffou et al. (2008) created a virtual tunnel paradigm, where participants were exposed to a checkerboard pattern that oscillated with sinusoidal translating motion. The visual surround oscillated at frequencies ranging from 0.125 Hz to 0.5 Hz; however, it was found that the greatest amount of compensatory sway was elicited at a stimulus frequency of 0.25 Hz (Greffou et al., 2008).

Recent work has expanded upon the use of VR in balance research through providing evidence for the efficacy of HMDs in eliciting VEPRs. Nielsen et. al (2022) used a HMD to expose participants to a pseudorandom virtual stimulus in an environment replicating a real-world setting (Nielsen, Cleworth, & Carpenter, 2022). Continuous AP translations of the VR scene were induced between a frequency range of 0.006 – 1 Hz (Nielsen et al., 2022). It was discovered that the average frequency of sway increased with the frequency of visual stimulus between 0.049–0.977 Hz (Nielsen et al., 2022). Additionally, this work provided support for the overall clinical utility of VR HMDs by showing that participants perceived themselves to be equally stable with and without the visual stimulus, when not exposed to a simulated postural threat (Nielsen et al., 2022).

1.7.3 Potential limitations of virtual reality

One potential limitation identified in VR experimental paradigms is accommodation-vergence conflicts. Specifically, if the virtual scene is not arranged at the focal distance of the HMD, this can result in depth perception differing in a virtual compared to a real-world environment (Cooper, Cant, White, & Meyer, 2018). In a natural environment, the control signals for accommodation and vergence of the eyes onto the target are closely tied, whereas in VR, stereoscopic images are displayed on a single plane through the HMD (Cooper et al., 2018). One of the most common consequences of this accommodation-vergence conflict can be the feeling of discomfort and fatigue in VR (Hoffman, Girshick, Akeley, & Banks, 2008). So as not to compromise performance during VR experimental paradigms, appropriate rest periods should be introduced to minimize this discomfort. Additionally, VR can accompany the risk of cyber sickness, which encompasses a range of symptoms of discomfort and disorientation, such as nausea (Weech, Kenny, & Barnett-Cowan, 2019). Recent work has demonstrated that there may be an inverse relationship between individuals' sense of presence and cyber sickness in VR (Weech et al., 2019); therefore, photo-realistic VR environments may also be beneficial to reduce the risk of cyber sickness among the user.

Beyond discomfort and cyber sickness, an additional limitation with VR technology is the structural constraints surrounding the HMD field of view. VR HMDs often provide the user with a field of view ranging from 90-110º, which is less than what would naturally be observed by the human eye in a real-world environment (Nagata, 1996). A potential consequence for this reduced field of view in a HMD could be the reduction of peripheral visual cues, which have previously been shown to be vital for balance control during upright stance (O'Connell et al., 2017).

1.8 Methodological considerations

1.8.1 Manipulating the gain of visual feedback

Previous work has manipulated the gain of visual feedback by presenting participants with real-time biofeedback of their center of foot pressure (COP) trace (Cawsey et al., 2009; Jehu et al., 2015). Here, it was observed that a tighter regulation of postural control could be achieved with an increase in the magnification of biofeedback (Cawsey et al., 2009; Jehu et al., 2015). It was suggested that real-time biofeedback could be valuable as it offers additional artificial visual information to support balance control, supplementing the visual, vestibular, and somatosensory feedback gathered from the body (Jehu et al., 2015). Specifically, Cawsey et al. (2009) exposed participants to a trace of their COP position while completing a quiet standing task, and magnified the level of biofeedback received at 7 levels (1x, 4x, 8x, 16x, 32x, 48x, 64x). A relationship was discovered, where an increase in biofeedback magnification contributed to tighter regulation of COP (Cawsey et al., 2009). Specifically, a decrease in amplitude and an increase in frequency of COP was observed with an increase in biofeedback magnification (Cawsey et al., 2009). One possible explanation was that this amplification of visual feedback could have provided participants with increased sensory information necessary to elevate the control of corrective postural responses during quiet stance; when the result of an individual's postural adjustments were more easily detected, there may have been an increased ability to make smaller corrective adjustments and avoid larger deviations in COP trajectory (Cawsey et al., 2009). Additionally, this work showed a plateau in changes to COP amplitude between 4-8x magnification (Cawsey et al., 2009). It was suggested that even at high levels of visual feedback, there is a limit to the small deviations of COP that the body is able to detect and correct for (Cawsey et al., 2009). Jehu et al. (2015) extended this work by comparing COP behaviour at biofeedback magnifications of 5x and 10x and observed no difference between COP measures among the two conditions; thus, it was concluded that magnifying visual biofeedback to a degree of 5x can be sufficient in supporting a tighter regulation of upright stance (Jehu et al., 2015).

Within this work, it was also suggested that while standing on a non-compliant surface, vision may not be relied upon heavily enough to see significant changes in postural sway while manipulating visual feedback (Cawsey et al., 2009). Therefore, Cawsey et al. (2009) also aimed to assess COP behaviour while standing on a foam surface (Cawsey et al., 2009). Previous studies have shown that standing on a compliant surface can contribute to a greater sway response while probing sensory re-weighting (Schut et al., 2017). Specifically, standing on a foam surface has been shown to reduce the reliability of balance-related somatosensory feedback and increase the reliance on vision (Cawsey et al., 2009; Schut et al., 2017). In contrast to the non-compliant surface condition, a greater reduction in COP amplitude with an increase in the magnification of visual biofeedback was observed while standing on foam (Cawsey et al., 2009). Additionally, a plateau in COP amplitude was not observed until visual biofeedback was magnified to 16x, in comparison to 4-8x while standing on a non-compliant surface (Cawsey et al., 2009). Therefore, it was concluded that an increased reliance on vision to maintain balance control was experimentally induced by standing on foam in this study.

1.8.2 Quantifying postural stability during upright stance

Posturography can be used to quantify the postural response of an individual, often in response to the manipulation of various balance control tasks (Visser, Carpenter, van der Kooij, & Bloem, 2008). Specifically, static posturography analyzes an individual’s postural response while attempting to maintain quiet stance during an unperturbed condition (Visser et al., 2008).

The primary advantage of using posturography to quantify standing balance is that it is an objective measure in comparison to many widely used clinical analyses of balance, which are often more subjective forms of assessment (Visser et al., 2008). Kinetic-based posturography refers to the collection of reactive torques and shear/vertical ground reaction forces to calculate COP position, whereas kinematic analyses make use of motion sensors or optical motion analysis software to estimate center of mass (COM) position (Carpenter, Murnaghan, & Inglis, 2010). From kinetic and kinematic data, outcome measures can be derived from both time and frequency series COP recordings to identify velocity, frequency, and amplitude data; these outcome measures can then be used to characterize an individual's postural response during a balance control task (Winter, 1995).

A widely used measure to quantify the amplitude of postural sway across the time domain is root mean square (RMS), which is a form of standard deviation that reflects the average distance travelled by the signal from its mean location (Bendo, Kuvarati, Skënderi, & Veveçka, 2013). RMS has been shown to have strong intrasubject reliability across balance control studies, is highly sensitive to changes in the availability of visual information, and is not significantly influenced by slight deviations in the base of support (BOS) (Jancová, 2008; Kapteyn et al., 2008); all of which are valuable while assessing balance behaviour related to changes in visual information. An increase in RMS has been linked to greater neuromotor noise, poorer overall postural control, and increased fall risk, as seen in patients with Parkinson's disease (Kapteyn et al., 2008; Schmit et al., 2006). Consequently, it reflects an increase in sway amplitude and poorer regulation of the COM within the BOS (Kapteyn et al., 2008; Schmit et al., 2006).

Additionally, within the time domain mean velocity (mVel) has widely been used to quantify postural sway and reflects the average velocity moved in the AP and ML directions (Bendo et al., 2013). mVel has also been shown to have a high intrasubject reliability across balance control studies, specifically when compared against other measures such as total sway area (Ruhe, Fejer, & Walker, 2010). Although there is not a clear consensus on the functional significance of mVel when analyzed in isolation, in healthy adults who are not experiencing neurological or musculoskeletal deficits, an increase in mVel has generally been shown to reflect poorer overall postural control and an increase in fall risk (Palmieri, Ingersoll, Stone, & Krause, 2002). This has partially been attributed to the energetic inefficiency that accompanies the muscle force required to produce higher levels of sway velocity (Palmieri et al., 2002). Additionally, when mVel becomes too high, the ability to gather proprioceptive sensory feedback from the surrounding environment diminishes; without adequate sensory information about one's relationship to the surroundings, it becomes more difficult to produce a postural response complementary to the postural threat, leading to an increased risk of suffering from a fall (Carpenter et al., 2010).

Within the frequency domain, mean power frequency (MPF) is often used to quantify a postural response and represents the average of the power of the harmonic components within the signal's power spectrum (Phinyomark et al., 2012). A higher shift in MPF is generally indicative of poorer postural control and increased fall risk, as slower oscillations of sway better support exploratory inflow of sensory information from the surrounding environment (Visser et al., 2008; Zaback, Adkin, & Carpenter, 2019). Additionally, information can be gathered from the frequency domain by conducting separate mean power calculations within segmented frequency bins. This is because specific frequency ranges have been shown to be associated with different afferent sensory systems used to control quiet standing (Lin et al., 2019; Salsabili et al., 2013). Specifically, the visual system has been correlated with slower frequencies, often below 0.1 Hz, the vestibular system under 0.5 Hz, and the somatosensory system below 1.0 Hz (Salsabili et al., 2013; Taguchi, 1978). Additionally, changes to postural sway between 1.0 - 3.0 Hz are linked to disorders within the central nervous system (Salsabili et al., 2013). Therefore, by conducting frequency bin analyses, further insight into sensory re-weighting processes that occur during balance control tasks can be uncovered.

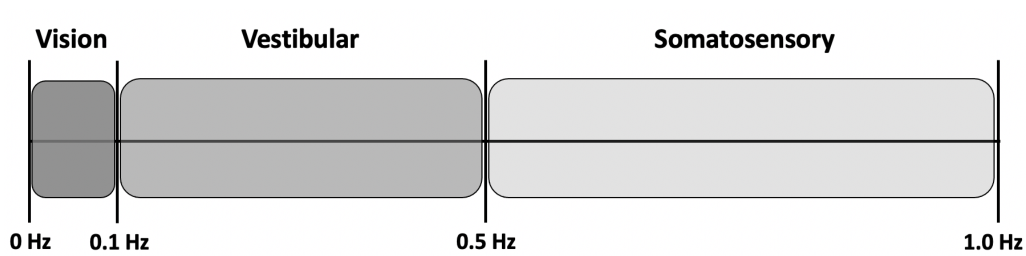

Figure 2. Proposed frequency-specific sensory contributions to balance control (adapted from Lin et al., 2019 and Salsabili et al., 2013)

Lastly, additional information about postural responses can be discovered when interpreting descriptive variables in combination, as opposed to in isolation. For example, an increase in RMS amplitude combined with a decrease in MPF has been shown to reflect a loss of sensory information contributing to the balance control strategy; this is seen through larger and slower oscillations of sway, deviating further towards the outskirts of the BOS (Carpenter, Frank, Silcher, & Peysar, 2001).

1.9 Current knowledge gap

Although previous work has manipulated the gain of visual feedback indirectly using a biofeedback approach, there is limited work that has manipulated optic flow gain directly to observe resulting balance behaviours. Additionally, the visual system has often been oversimplified in balance control studies by manipulating vision through strictly eyes open or eyes closed conditions, or by observing postural responses at optic flow gain values of 1 or 0 (sway referencing). This indirect manipulation of visual feedback combined with a traditionally coarse sampling of optic flow limits our understanding of the visual contributions towards balance control during quiet, upright stance.

1.10 Thesis purpose and hypotheses

The primary objective of this thesis is to develop our understanding of how optic flow gain contributes to postural stability during quiet, upright stance among healthy adults. Specifically, this thesis aims to address the following research question: How is balance behaviour, captured through COP and head kinematics, affected by manipulating the gain of optic flow in VR while standing on both a foam and non-compliant (firm) surface? This thesis will also examine if normal quiet standing balance behaviour is affected after undergoing a VR experimental paradigm. It was hypothesized that:

- as optic flow gain increases, participants will experience tighter control of upright stance, as measured by decreases in COP and head position (HeadPos) RMS and increases in mVel and MPF (Cawsey et al., 2009; Jehu et al., 2015)

- as optic flow gain decreases, postural sway will increase, as measured by increases in COP and HeadPos RMS and decreases in mVel and MPF (Cawsey et al., 2009; Jehu et al., 2015)
- an increase in mean power within the LOW (0-0.1Hz) frequency band will accompany an increase in optic flow gain, with no significant changes in mean power across the MED, MED-HIGH, or HIGH bands (Lin et al., 2019; Salsabili et al., 2013)
- the effect of the above changes to COP and HeadPos will amplified while standing on foam, compared to a firm surface (Cawsey et al., 2009)
- there will be no significant differences in balance behaviour before and after experiencing VR (Assländer & Streuber, 2020b; Riecke et al., 2006)

Chapter 2: Methods

2.1 Participants

33 healthy adults volunteered to participate in this study. Participants were recruited from York University and the surrounding community. Inclusion criteria required participants to be between the ages of 18-40 years. Participants were excluded if they self-reported any current neurological or orthopedic impairments that could influence balance or were currently taking medications that are known to influence balance. Additionally, participants were excluded from the study if they had a body mass index (BMI) of 30.0 or greater; previous work has shown that individuals within this BMI category may sway at a greater velocity while standing quietly on both firm and foam surfaces (Dutil et al., 2013; Son, 2016). Three participants were excluded from data analysis due to technical difficulties during data collection (optic flow gain did not change across conditions). Therefore, the sample size for data analysis consisted of 30 healthy adults (mean age (± SD): 22.9 (± 4.17) years, 21 female). The Human Participants Review Sub-Committee of York University's Ethics Review Board approved this study (#e2019-367), and all participants provided written, informed consent prior to participation.

2.2 Experimental procedure

For each trial, participants were asked to stand quietly on a force plate capable of recording ground reaction forces and moments for 60 seconds, with arms rested naturally at their sides (Figure 2). Stance width was standardized to the foot length of each participant and was marked on the force plate to ensure consistent foot placement across trials. There were two sets of surface conditions, each with four trials, and the order of the conditions was randomized across participants. One set of conditions required participants to stand on a piece of foam placed on top of the force plate, while the other had participants stand directly on the force plate. Before and after completing a set of trials, baseline quiet standing without VR was collected for 60 seconds.

Participants were asked to face forwards, stand quietly, and focus on a red target within the virtual environment for the duration of each of the eight experimental trials. For each VR trial, the gain values of 0.25, 2, 4, or 16 were presented in a counterbalanced order to amplify or reduce participants' optic flow, relative to head motion. These gain values were selected since previous studies that have examined balance behaviour in response to visual biofeedback have identified the optimal degree of visual magnification to be 2-4x to achieve an attenuated sway response (Litvinenoka & Hlavacka, 1973). Between each trial across both baseline and experimental conditions, participants were given a 60 second resting period where they were able stand freely, prior to commencing the subsequent trials.

2.3 Virtual reality setup

During experimental trials, participants wore a VR HMD (Oculus Rift, Oculus) with an approximate 110° field of view. The virtual environment, "gallery.osgb", was selected from a pre-designed, publicly available Vizard database (Vizard, WorldViz, USA). This scene simulated an art gallery environment (Figure 2) and was selected as it closely replicated an environment that could been observed in a real-world setting.

2.4 Manipulating optic flow gain

For each experimental trial, optic flow gain was manipulated by applying a gain factor of either 0.25, 2, 4, or 16 to the scene using Vizard python programming (Vizard, WorldViz, USA). Each gain factor changed the view position of the virtual scene in the AP and ML directions relative to participants' head position, as captured by the HMD. For example, a gain factor of 2 would cause twice the amount of scene motion compared to what would be experienced under normal conditions (Appendix A).

2.5 Data collection and signal processing

2.5.1 Kinetics

Ground reaction forces and moments were sampled at 100 Hz from a single force plate (AMTI, USA). COP displacements were then calculated in MATLAB in the AP and ML directions. The COP data were low pass filtered with a 5 Hz dual pass Butterworth filter and underwent bias removal by subtracting the mean COP position.

2.5.2 Head kinematics

Head kinematics (HeadPos) was captured in the AP direction using the HMD (Oculus, Oculus Rift). Data were outputted through Vizard (WorldViz, USA) using a digital lab interface (LabJack, Colorado, USA), and recorded at 100 Hz (Spike2, Cambridge Electronic Design, UK). HeadPos data were then low pass filtered with a 5 Hz dual pass Butterworth filter and underwent bias removal through subtracting the mean HeadPos position from the signal.

2.6 Quantifying balance behaviour

To quantify balance behaviour, MPF, RMS, mVel, and mean power across four different frequency bands (LOW: 0-0.1Hz, MED: 0.1-0.5Hz, MED-HIGH: 0.5-1.0Hz, HIGH: 1.0-5.0 Hz) were calculated for both AP and ML COP, and AP HeadPos.

Within the frequency domain, MPF was calculated using the following formula:

$$MPF = \frac{\sum f \ x \ P(f)}{\sum P(f)}$$

where f was the frequencies of the signal and P was the power amplitude at each respective frequency. Mean power was calculated using the *bandpower* function (MATLAB, MathWorks), which computes the average power within a specified frequency range.

For time series analysis, RMS was calculated using the following formula:

$$RMS = \sqrt{\frac{1}{n}} \sum_{i=1}^{n} x_i^2$$

where x was the data sample and n was the number of data points. mVel was calculated using:

$$\bar{v} = \frac{\Delta d}{\Delta t}$$

where d and t represented displacement and time data points, respectively.

2.7 Data analysis

Following data collection, 2 (surface) x 4 (gain value) and 2 (surface) x 2 (condition: quiet standing before or after VR) repeated measures analysis of variance (ANOVA) tests were conducted for each outcome measure (RMS, mVel, MPF, and mean power: LOW, MED, MED-HIGH, HIGH) in SPSS (IBM Corp., N.Y., USA). Shapiro-Wilks tests and histograms were used to assess normality across all variables and log transformations were applied where assumptions of normality were violated. Z-scores were used to detect outliers, where a z-score of greater or less than 3 indicated the presence of an outlier. To confirm whether these values were true outliers, time series data was plotted in MATLAB (Mathworks, USA) to identify if the data was representative of normal standing behaviour. All outlying values were then replaced to 3 standard deviations from the mean (Field, 2009). Transforming the data and replacing outliers where necessary did not correct for normality; therefore, due to the robust nature of repeated measures ANOVA and the central limit theorem approximating normality, no corrections for normality were used for analysis. Sphericity was evaluated using Mauchley's test of sphericity and Greenhouse-Geisser corrections were applied when the assumption of sphericity was violated. Sidak corrections were applied to correct for multiple comparisons and statistical significance was set at an α-level of 0.05. Significant interaction effects were explored using independent surface condition one-way repeated measures ANOVA tests, while significant main effects were explored post-hoc using non-parametric t-tests.

(A)

(B)

Figure 3. Experimental setup

(B) Participants’ approximate view inside the virtual environment; **(B)** Participant standing on the force plate wearing the VR HMD

Chapter 3: Results

3.1 General results

3.1.1 Amplitude of displacement

AP COP showed significant main effects of both gain and surface on RMS, with no significant interaction effect observed (Table 1). Post-hoc results showed RMS to be greatest at a gain of 0.25 and on average decreased up until a gain of 4 (Figure 3). There were significant main effects of both gain and surface observed on ML COP RMS, with no significant interaction effect (Table 1). Post-hoc results showed RMS to be greatest at a gain of 0.25 and lowest at a gain of 16 (Figure 3). Lastly, AP HeadPos showed significant main effects of both surface and gain on RMS, with no significant interaction effect observed (Table 1). Post-hoc results showed AP HeadPos RMS to be greatest at a gain of 0.25 and decreased to be lowest at a gain of 16 (Figure 3).

3.1.2 Amplitude of velocity

There were significant main effects of both surface and gain on AP COP mVel, as well as a significant interaction effect observed (Table 1). There was a significant main effect of gain on AP COP mVel among both the firm ($F_{2.377, 68.934} = 4.542$, $p = .005$) and foam ($F_{1.479, 42.899} = 18.195$, $p < .001$) surface conditions. Post-hoc results showed no significant pairwise comparisons among the firm surface condition; however, the foam surface condition showed AP COP mVel to be greatest at a gain of 16 (Figure 4). ML COP showed significant main effects of both surface and gain on mVel, with a significant interaction effect observed (Table 1). There was a significant main effect of gain on ML COP mVel among only the foam surface condition

($F_{1.835, 53.226} = 5.638$, $p = .007$). Post-hoc results showed no significant differences in ML COP mVel between any gain value (Figure 4). Lastly, significant main effects of both surface and gain on AP HeadPos mVel, as well as a significant interaction effect were observed (Table 1). There were significant main effects of gain on AP HeadPos mVel among both firm ($F_{3, 57} = 19.212$, $p < .001$) and foam surface conditions ($F_{3, 57} = 15.353$, $p < .001$). Post-hoc results showed AP HeadPos mVel to be greatest at a gain of 0.25, while on average decreasing with an increase in gain (Figure 4).

3.1.3 Frequency

There was a significant main effect of both surface and gain on AP COP MPF, with a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on AP COP MPF ($F_{1.967, 57.031} = 25.466$, $p < .001$). Post-hoc results showed AP COP MPF to be lowest at gain values of 0.25 and 2, while increasing to a gain of 16 while standing on foam (Figure 5). There was also a significant main effect of gain, but not surface, on ML COP MPF, with a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on ML COP MPF ($F_{2.112, 61.238} = 9.776$, $p < .001$). Post-hoc results revealed that ML COP MPF was lowest at a gain of 0.25 and 2 and on average, increased to be highest at a gain of 16 while on foam (Figure 5). Lastly, there was a significant main effect of surface, but not gain, on AP HeadPos MPF, with no significant interaction effect observed (Table 1).

3.1.4 Power spectrum

Within the LOW frequency band, a significant main effect of surface on AP COP mean power was observed (Table 1). There was no significant main effect of gain on AP COP mean power, or a significant interaction effect observed within the LOW frequency band (Table 1). Additionally, there was a significant main effect surface, but not gain, on ML COP mean power observed, with no significant interaction effect (Table 1). There were significant main effects of surface and gain on AP HeadPos mean power, with no significant interaction effect observed (Table 1). Post-hoc results showed AP HeadPos mean power to be greatest at a gain of 0.25 and on average, decrease to be lowest at a gain of 16 (Figure 6).

Within the MED frequency band, AP COP showed significant main effects of both surface and gain on mean power, with a significant interaction effect observed (Table 1). There were significant main effects of gain on AP COP mean power among both the firm ($F_{2.104, 61.008} = 10.717$, $p < .001$) and foam surfaces ($F_{2.283, 66.214} = 32.233$, $p < .001$). Post-hoc results showed AP COP mean power to be greatest at a gain of 0.25 and on average, decrease to a gain of 16 (Figure 7). There were also significant main effects of surface and gain on ML COP mean power within the MED frequency band, with a significant interaction effect observed (Table 1). There were significant main effects of gain on ML COP mean power among both the firm ($F_{2.348, 68.092} = 4.902$, $p = .007$) and foam surface conditions ($F_{2.146, 62.242} = 11.155$, $p < .001$). Post-hoc results revealed ML COP mean power to be greatest a gain of 0.25 and lowest at a gain of 16 while standing on foam (Figure 7). AP HeadPos showed significant main effects of both surface and gain on mean power, with a significant interaction effect observed (Table 1). There were significant main effects of gain on AP HeadPos mean power among both the firm ($F_{3, 57} = 14.831$, $p < .001$) and foam surface conditions ($F_{3, 57} = 19.614$, $p < .001$). Similar to COP measures, post-hoc results showed AP HeadPos mean power to be greatest at a gain of 0.25 and lowest at a gain of 16 (Figure 7).

Within the MED-HIGH frequency band, there were significant main effects of both surface and gain on AP COP mean power, with a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on AP COP mean power ($F_{1.517, 43.990} = 11.888$, $p < .001$), and post-hoc results showed AP COP mean power to be greatest at a gain of 16 (Figure 8). ML COP showed a significant main effect of surface on mean power (Table 1). Additionally, there was no significant main effect of gain on ML COP mean power observed and there was not a significant interaction effect (Table 1). There were significant main effects of both surface and gain on AP HeadPos mean power, with a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on AP HeadPos mean power ($F_{1.583, 30.077} = 6.450$, $p = .008$). Post-hoc results showed AP HeadPos mean power to be lowest at a gain of 0.25 and greatest at a gain of 16 while standing on foam (Figure 8).

Lastly, within the HIGH frequency band, there were significant main effects of both surface and gain on AP COP mean power, with a significant interaction effect observed (Table 1). There were significant main effects of gain on AP COP mean power among both firm ($F_{1.811, 52.516} = 4.845$, $p = .014$) and foam surface conditions ($F_{1.340, 38.864} = 16.500$, $p < .001$). Post-hoc results showed AP COP mean power to be greatest at a gain of 16 while standing on foam (Figure 9). There were also significant main effects of both surface and gain on ML COP mean power, with a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on ML COP mean power ($F_{1.122, 32.552} = 5.910$, $p = .018$). Post-hoc results showed ML COP mean power to be greatest at a gain of 16 within the foam surface condition (Figure 9). AP HeadPos showed significant main effects of surface and gain on mean power, in addition to there being a significant interaction effect observed (Table 1). Within the foam surface condition, there was a significant main effect of gain on AP HeadPos ($F_{1.652, 31.386} = 11.898$, $p < .001$). Similar to COP variables, post-hoc results showed AP HeadPos mean power to be greatest at a gain of 16 while standing on foam (Figure 9).

3.2 Comparing center of pressure pre and post VR

3.2.1 Center of pressure amplitude of displacement

There was a significant main effect of condition (pre and post VR) on RMS in the AP direction (Table 2). Additionally, in the AP direction there was a significant main effect of surface on RMS, with no significant interaction effect observed (Table 2). Specifically, AP COP RMS was greater post VR exposure (Figure 10). In the ML direction, there was no significant main effect of condition on RMS observed (Table 2). There was a significant main effect of surface on RMS, with no significant interaction effect observed (Table 2).

3.2.2 Center of pressure amplitude of velocity

There was no significant main effect of condition (pre and post VR) on mVel in the AP direction; however, there was a significant main effect of surface on mVel observed, with no significant interaction effect (Table 2). In the ML direction, there was a significant main effect of surface and condition on mVel observed, in addition to a significant interaction effect (Table 2). Within the foam surface condition, there was a significant main effect of condition on ML mVel

($F_{1, 29} = 8.554$, $p = .007$). Post-hoc results showed ML mVel to be greater pre VR while standing on foam.

3.2.3 Center of pressure frequency

In the AP direction, there was a significant main effect of condition (pre and post VR) on MPF observed (Table 2). There was no significant main effect of surface observed on MPF, nor was there a significant interaction effect (Table 2). Specifically, AP COP MPF was greater pre VR exposure (Figure 11). In the ML direction, there was no significant main effect of surface on MPF observed (Table 2). There was a significant main effect of condition on ML MPF observed, as well as a significant interaction effect (Table 2). Within the foam surface condition, there was a significant main effect of condition on ML MPF ($F_{1, 29} = 22.443$, $p < .001$). While standing on foam, post-hoc results showed ML MPF to be greater pre VR exposure (Figure 11).

3.2.4 Center of pressure power spectrum

Within the LOW frequency band, there was a significant main effect of condition (pre and post VR) on mean power observed in the AP direction (Table 2). Additionally, there was a significant main effect of surface on mean power observed in the AP direction, as well as a significant interaction effect (Table 2). Within the foam surface condition, there was a significant main effect of condition on AP mean power ($F_{1, 29} = 11.843$, $p = .002$). On foam, AP mean power in the LOW band was greater post VR exposure. In the ML direction, there were significant main effects of both surface and condition on mean power, with a significant interaction effect observed (Table 2). Within the foam surface condition, there was a significant main effect of condition on ML mean power ($F_{1, 29} = 7.641$, $p = .010$). ML mean power in the LOW band was greater post VR exposure while standing on foam.

In the MED frequency band, there was a significant main effect of surface observed on mean power in the AP direction; however, there was no significant main effect of condition (pre and post VR) observed, nor was there a significant interaction effect (Table 2). In the ML direction, there was a significant main effect of surface on mean power observed (Table 2). Further, there was no significant main effect of condition on mean power observed, and there also was no significant interaction effect in the ML direction (Table 2).

In the MED-HIGH frequency band, there was a significant main effect of both surface and condition (pre and post VR) on mean power observed in the AP direction (Table 2). Additionally, there was a significant interaction effect observed in the AP direction (Table 2). Within the foam surface condition, there was a significant main effect of condition on AP mean power ($F_{1, 29} = 14.790$, $p < .001$). Within the MED-HIGH frequency band, while standing on foam AP mean power was greater pre VR exposure. In the ML direction, there was a significant main effect of surface on mean power observed. There was additionally a significant main effect of condition on mean power, with a significant interaction effect observed in the ML direction (Table 2). Within the foam surface condition, there was a significant main effect of condition on ML mean power ($F_{1, 29} = 9.707$, $p = .004$). In the MED-HIGH band, ML mean power was greater pre VR exposure while standing on foam.

Within the HIGH frequency band, there was a significant main effect of surface on mean power observed in the AP direction (Table 2). In the AP direction, there was no significant main effect of condition (pre and post VR) on mean power observed, and there also was no significant interaction effect (Table 2). In the ML direction, there was a significant main effect of surface on mean power observed (Table 2). There was also a significant main effect of condition on mean power observed, with no significant interaction effect in the ML direction (Table 2). Specifically, within the HIGH frequency band ML mean power was greater pre VR exposure.

Table 1. Summary of repeated measures ANOVA results for experimental trials

Results shown for RMS (mm), MPF (Hz), mVel (mm/s), and mean power (mm^2/Hz) across each frequency band (LOW, MED, MED-HIGH, HIGH). Bolded p-values denote statistical significance.

		Surface			Gain			Surface*Gain		
		F	df	p-value	F	df	p-value	F	df	p-value
AP COP	RMS	44.418	1, 29	< **.001**	5.635	2.266, 65.717	**.004**	.315	3, 87	.814
	MPF	22.238	1, 29	< **.001**	16.272	3, 87	< **.001**	8.298	3, 87	< **.001**
	mVel	93.195	1, 29	**<.001**	20.443	1.746, 50.634	**<.001**	11.727	1.419, 41.154	**<.001**
	LOW	13.141	1, 29	**.001**	2.558	2.418, 70.108	.074	1.621	3, 87	.190
	MED	121.444	1, 29	< **.001**	39.259	2.364, 68.568	< **.001**	21.425	2.397, 69.504	< **.001**
	MED-HIGH	56.985	1, 29	**<.001**	12.316	1.512, 43.837	**<.001**	11.332	1.540, 44.654	**<.001**
	HIGH	58.979	1, 29	**<.001**	18.577	1.311, 38.030	**<.001**	13.012	1.395, 40.444	**<.001**
ML COP	RMS	130.913	1, 29	< **.001**	4.915	3, 87	**.003**	1.000	3, 87	.987
	MPF	0.220	1, 29	.643	6.195	3, 87	**<.001**	2.803	3, 87	**.045**
	mVel	90.010	1, 29	**<.001**	5.373	1.932, 56.042	**.008**	4.948	1.931, 56.007	**.011**
	LOW	70.533	1, 29	**<.001**	4.915	2.152, 62.404	.067	2.045	2.016, 58.467	.138
	MED	58.368	1, 29	**<.001**	12.237	2.162, 62.685	**<.001**	9.509	2.155, 62.489	**<.001**
	MED-HIGH	41.659	1, 29	**<.001**	.995	2.422, 70.235	.387	1.049	2.429, 70.451	.366
	HIGH	32.676	1, 29	**<.001**	6.408	1.137, 32.984	**.013**	5.348	1.122, 32.549	**.024**
HeadPos	RMS	23.147	1, 19	**<.001**	6.543	3, 57	**<.001**	.237	2.193, 41.666	.810
	MPF	7.965	1, 19	**.011**	.566	3, 57	.639	1.886	3, 57	.142
	mVel	245.036	1, 19	**<.001**	22.963	2.344, 44.531	**<.001**	5.334	3, 57	**.005**
	LOW	9.146	1, 19	**.007**	3.646	3, 57	**.018**	1.119	2.088, 39.669	.339
	MED	64.022	1, 19	**<.001**	28.575	3, 57	**<.001**	9.435	2.408, 45.751	**<.001**
	MED-HIGH	40.432	1, 19	**<.001**	5.534	1.612, 30.626	**.013**	6.960	1.648, 31.317	**.005**
	HIGH	43.582	1, 19	**<.001**	12.965	1.887, 35.852	**<.001**	8.415	1.591, 30.226	**.002**

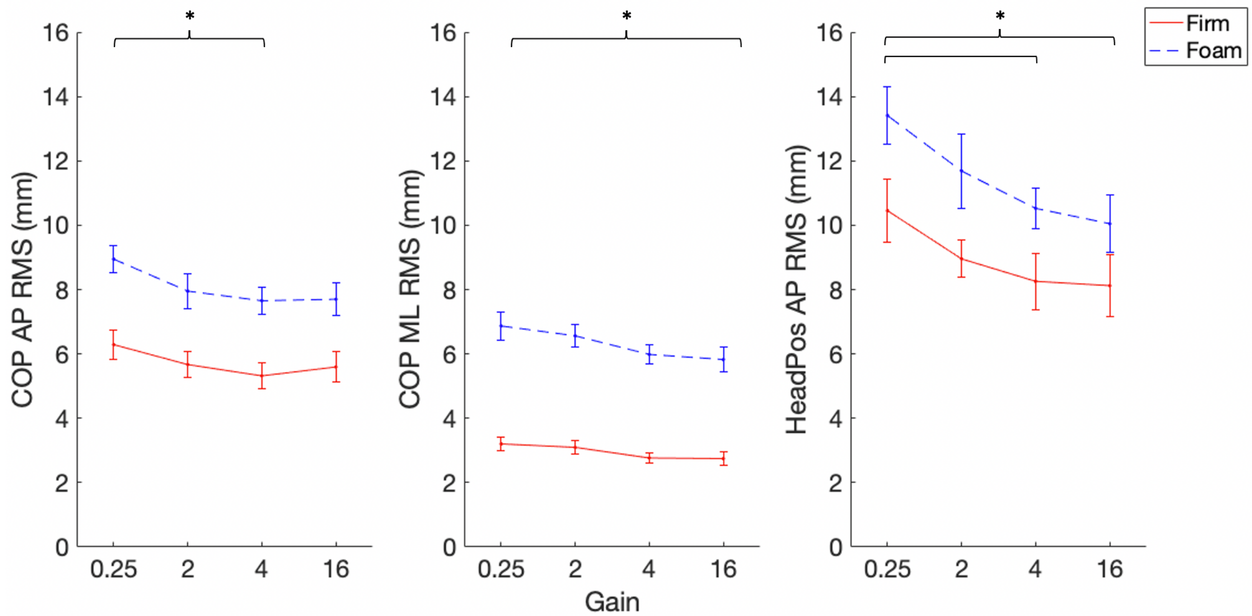

Figure 4. RMS across experimental trials

Mean (SEM) RMS plotted for COP AP, COP ML, and HeadPos AP. Significant pairwise comparisons displayed between optic flow gain values, with black lines of significance used for collapsed surface conditions.

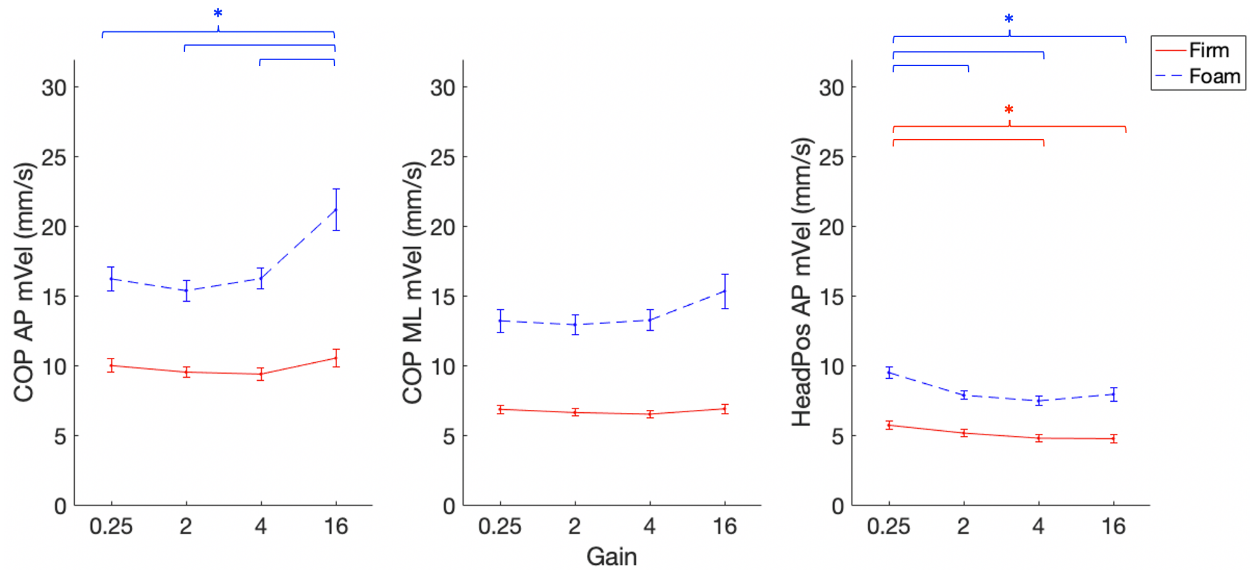

Figure 5. mVel across experimental trials

Mean (SEM) mVel plotted for COP AP, COP ML, and HeadPos AP. Significant pairwise comparisons displayed between optic flow gain values.

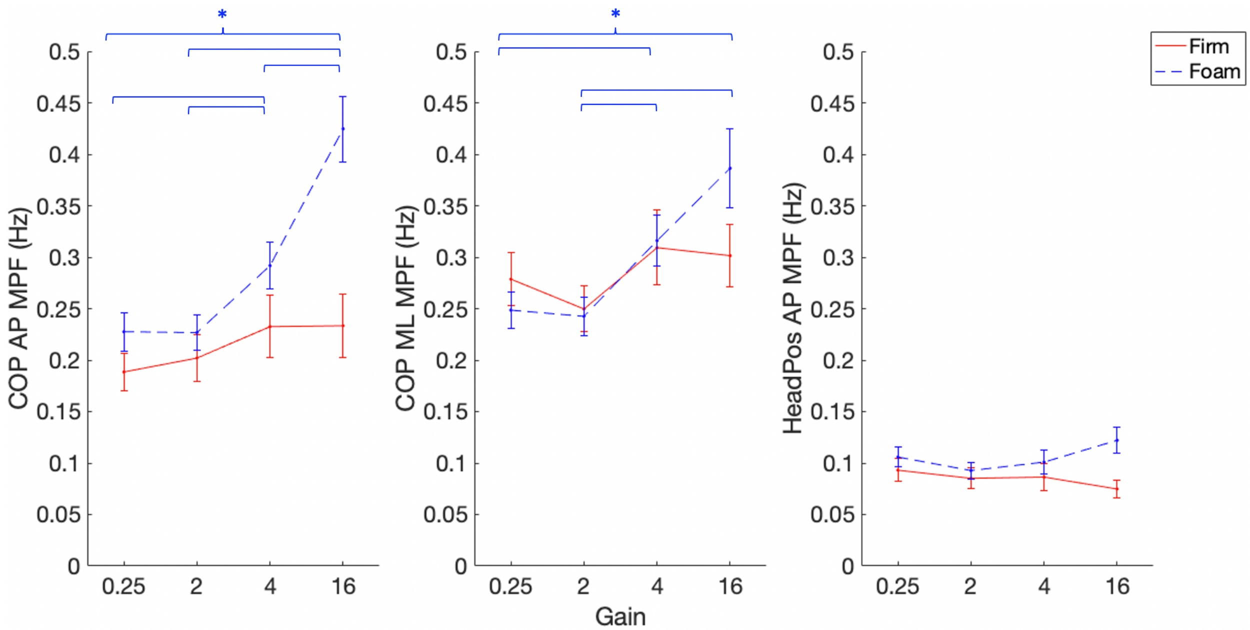

Figure 6. MPF across experimental trials

Mean (SEM) MPF plotted for COP AP, COP ML, and HeadPos AP. Significant pairwise comparisons displayed between optic flow gain values.

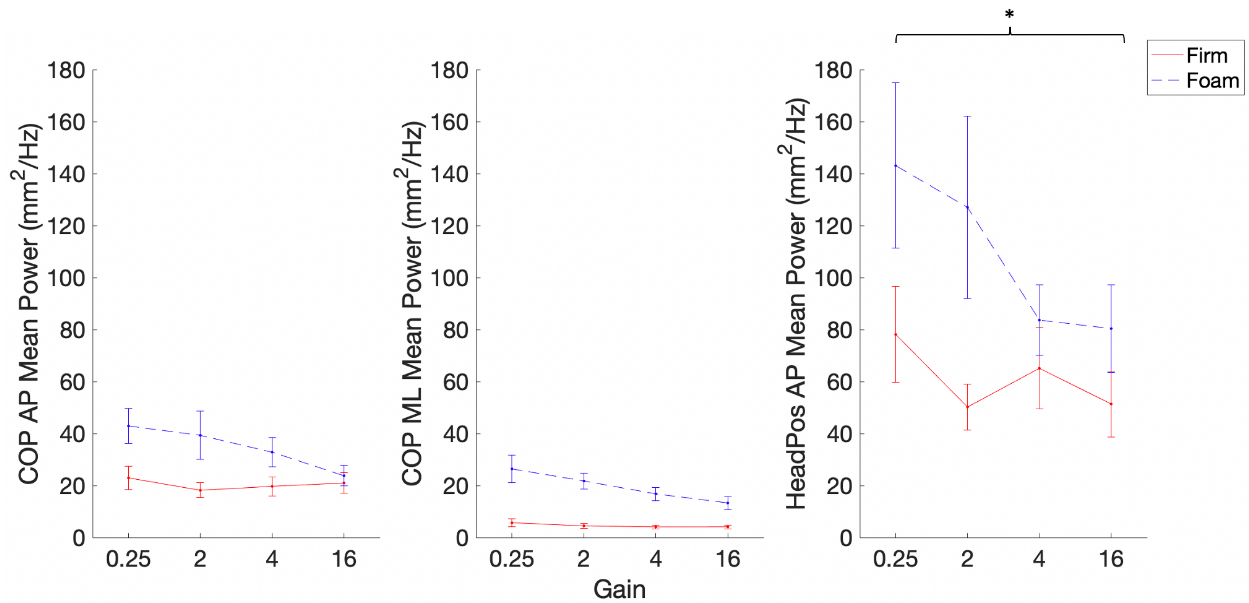

Figure 7. LOW (0-0.1 Hz) mean power across experimental trials

Mean (SEM) mean power plotted within the LOW frequency band. Significant pairwise comparisons displayed between optic flow gain values for COP AP, COP ML, and HeadPos AP. Black lines of significance indicate significant pairwise comparison for collapsed surfaces.

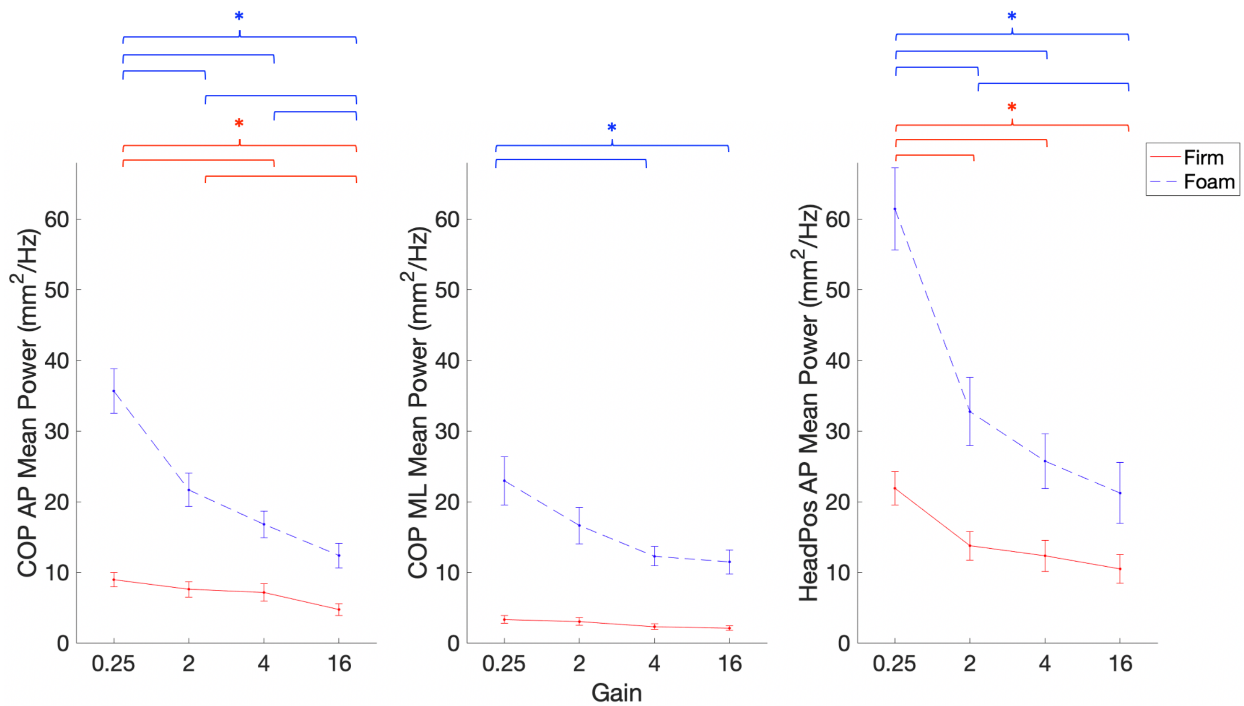

Figure 8. MED (0.1-0.5 Hz) mean power across experimental trials

Mean (SEM) mean power plotted within the MED frequency band. Significant pairwise comparisons displayed between optic flow gain values for COP AP, COP ML, and HeadPos AP.

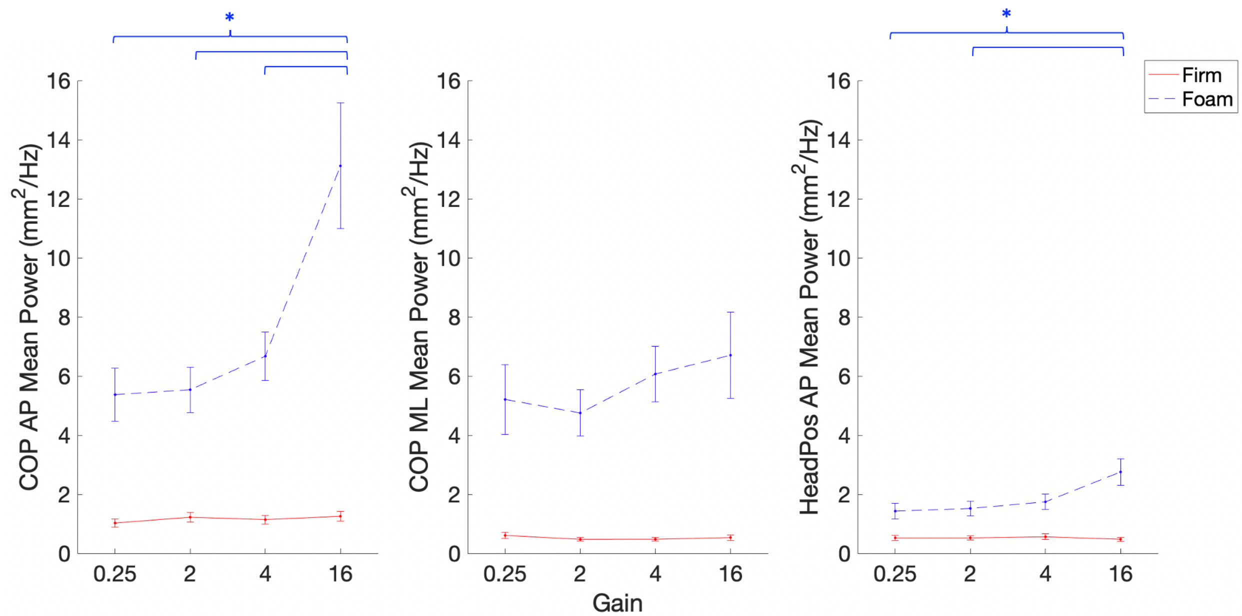

Figure 9. MED-HIGH (0.5-1.0 Hz) mean power across experimental trials

Mean (SEM) mean power plotted within the MED-HIGH frequency band. Significant pairwise comparisons displayed between optic flow gain values for COP AP, COP ML, and HeadPos AP.

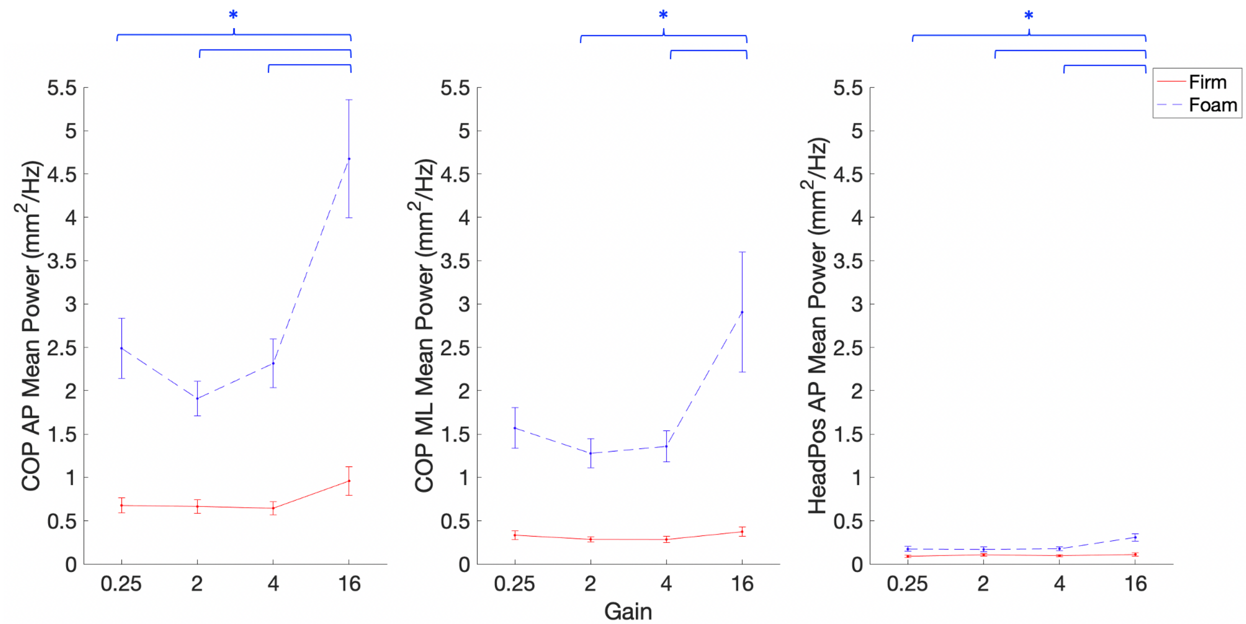

Figure 10. HIGH (1.0-5.0 Hz) mean power across experimental trials

Mean (SEM) mean power plotted within the HIGH frequency band. Significant pairwise comparisons displayed between optic flow gain values for COP AP, COP ML, and HeadPos AP.

Table 2. Summary of repeated measures ANOVA results for baseline quiet standing

Results shown for RMS (mm), MPF (Hz), mVel (mm/s), and mean power (mm^2/Hz) across each frequency band (LOW, MED, MED-HIGH, HIGH). Bolded p-values denote statistical significance.

		Surface			Gain			Surface*Gain		
		F	df	p-value	F	df	p-value	F	df	p-value
AP COP	RMS	9.604	1, 29	**.004**	5.597	1, 29	**.025**	.669	1, 29	.420
	MPF	3.408	1, 29	.075	9.981	1, 29	**.004**	.435	1, 29	.515
	mVel	36.828	1, 29	**<.001**	3.160	1, 29	.086	2.329	1, 29	.074
	LOW	4.440	1, 29	**.044**	9.216	1, 29	**.005**	6.204	1, 29	**.019**
	MED	50.283	1, 29	**< .001**	1.819	1, 29	.188	.492	1, 29	.489
	MED-HIGH	28.535	1, 29	**<.001**	13.782	1, 29	**<.001**	5.150	1, 29	**.031**
	HIGH	20.378	1, 29	**<.001**	.761	1, 29	.390	.297	1, 29	.590
ML COP	RMS	132.484	1, 29	**< .001**	2.551	1, 29	.121	.724	1, 29	.402
	MPF	.332	1, 29	.569	5.169	1, 29	**.031**	10.225	1, 29	**.003**
	mVel	62.499	1, 29	**<.001**	6.588	1, 29	**.016**	4.761	1, 29	**.037**
	LOW	28.698	1, 29	**<.001**	4.785	1, 29	**.037**	10.606	1, 29	**.003**
	MED	75.096	1, 29	**<.001**	3.045	1, 29	.092	.623	1, 29	.436
	MED-HIGH	34.596	1, 29	**<.001**	11.646	1, 29	**.002**	7.520	1, 29	**.010**
	HIGH	41.016	1, 29	**<.001**	4.751	1, 29	**.038**	2.847	1, 29	.102

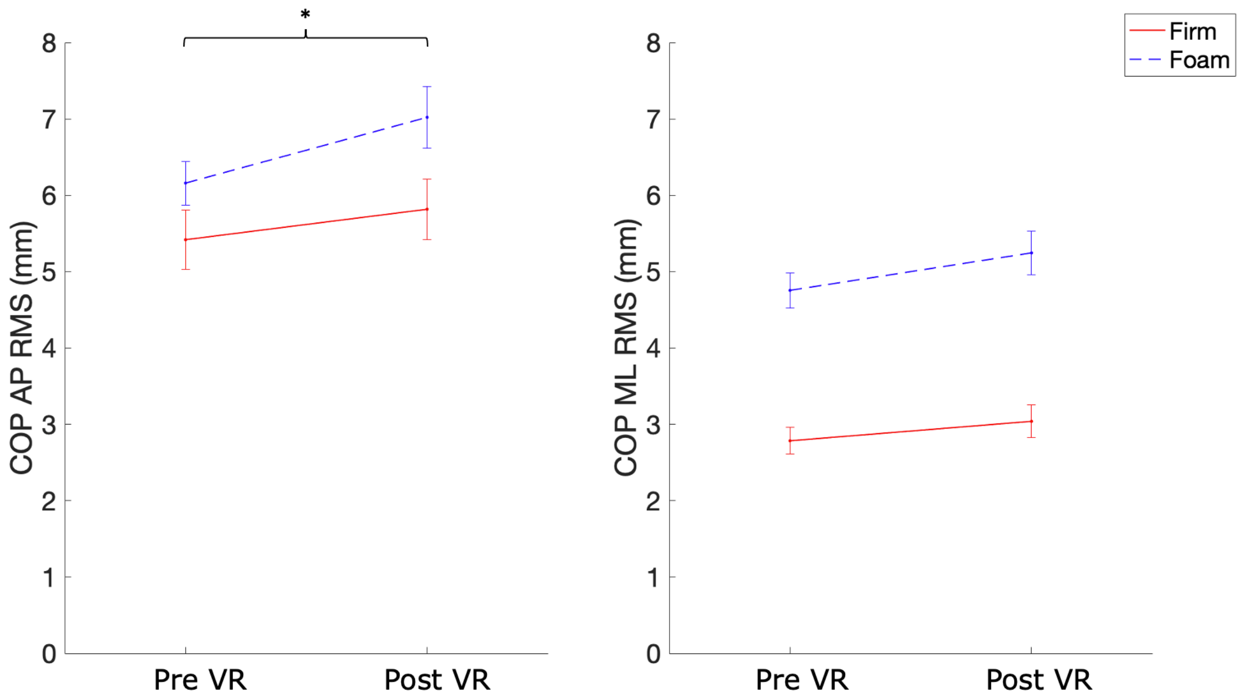

Figure 11. RMS pre and post VR

Mean (SEM) COP RMS plotted for baseline quiet standing pre and post VR. Black line of significance illustrating significant main effect of optic flow gain for collapsed surface conditions in the AP direction.

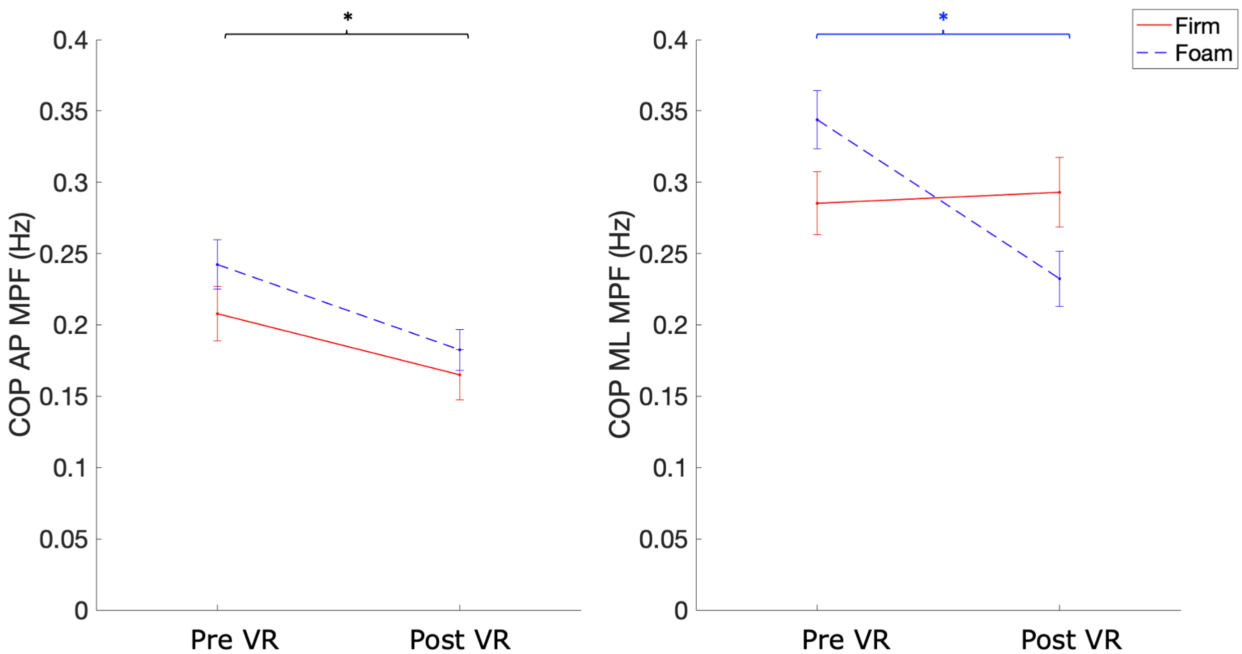

Figure 12. MPF pre and post VR

Mean (SEM) COP MPF plotted for baseline quiet standing pre and post VR. Black line of significance illustrating significant main effect of optic flow gain for collapsed surface conditions.

Chapter 4: Discussion

4.1 Overview

The overarching aim of this thesis was to further our understanding of the visual contributions to balance control during quiet standing. Specifically, this thesis aimed to examine the influence of modified optic flow gain on quiet, upright stance among healthy adults, while also comparing balance behaviour on both firm and foam surfaces. This was addressed through exposing participants to a VR environment and applying gain factors to manipulate optic flow relative to head motion captured through a HMD. The results of this thesis demonstrate that an increase in VR-generated optic flow gain can contribute to a tighter regulation of postural control, specifically while standing on a foam surface.

4.2 General findings

On average, as optic flow gain was increased in VR, COP amplitude decreased while COP velocity and frequency increased. Specifically, while standing on foam COP RMS decreased between optic flow gain values of 0.25 and 4. Additionally, while on foam COP mVel, and MPF were greatest at an optic flow gain value of 16 in comparison to all other gain values.

4.2.1 Changes to center of pressure

COP amplitude in the AP direction decreased with an increase in optic flow gain, as hypothesized; however, significant changes in amplitude were only observed between optic flow gain values of 0.25 and 4, meaning that no significant changes in RMS were observed between gain values of 4 and 16. These results suggest that a gain value of 4 may be sufficient in reducing the amplitude of postural sway, since providing participants with additional visual feedback at a gain value of 16 did not influence COP amplitude. This contradicts results observed by Cawsey et al. (2009), where a plateau in COP AP amplitude occurred at a biofeedback magnification of 16x. However, the results of this study were irrespective of surface condition, whereas the plateau effect identified by Cawsey et al. (2009) was specific to standing on foam. Collectively, these results suggest that standing on foam increases the degree of visual feedback needed to observe a plateau in the reduction of COP amplitude.

Additionally, as hypothesized, COP frequency increased with an increase in optic flow gain in both directions and was greatest at a gain value of 16. Specifically, in the AP direction significant differences in MPF were observed while standing on foam between each gain value, except between 0.25 and 2. Comparing these results to amplitude measures, COP frequency appeared to be more sensitive to changes in optic flow gain. This is supported by previous work which showed that a plateau in MPF under conditions of amplified visual biofeedback only occurred at 48x magnification, rather than 16x for COP RMS (Cawsey et al., 2009). The observation that there were no significant differences to COP frequency between the two smallest gain values also suggests that smaller levels of visual feedback may not be sufficient to induce changes in the postural response. Previous work has shown that an increase in frequency of COP oscillations is a consequence of exerting more stabilizing moments to maintain the COM within the BOS (Sweeny et al., 2021). Further, providing individuals with increased levels of visual feedback may allow for smaller deviations in sway to be detected and corrected for through these stabilizing moments. Considering the results of this study, smaller levels of visual feedback may not provide enough supplementary information about an individual's position to be able to make these corrections.

Lastly, COP velocity was also greatest at an optic flow gain value of 16 in the AP direction. These results were partially supported by the hypothesis which predicted COP velocity to increase with an increase in optic flow gain; however, significant differences in COP mVel were only observed between optic flow gain values of 16 with each other gain value (0.25, 2, and 4). In comparison to COP frequency and amplitude, this further suggests that velocity may be less sensitive to changes in optic flow, as greater visual feedback needed to be provided to observe any changes among this outcome variable. Previous work has shown COP mVel and the standard deviation of COP velocity to increase while standing on foam compared to a non-compliant surface (Jeka, Kiemel, Creath, Horak, & Peterka, 2004). Additionally, there is evidence to suggest that proprioception is especially sensitive to controlling COM velocity, as opposed to position or acceleration (Masani, Popovic, Nakazawa, Kouzaki, & Nozaki, 2003). In this study, proprioceptive reliability was experimentally reduced while participants stood on foam; therefore, when proprioceptive feedback is reliable, such as when standing on a non-compliant surface, COP velocity may not be sensitive to changes in optic flow gain. An increase in velocity of postural sway has been explained by previous work that compared balance behaviour between healthy individuals and patients with Parkinson's disease (PD) (Carpenter, Allum, Honegger, & Adkin, 2004). Specifically, patients with PD have demonstrated a decrease in angular trunk velocity in response to perturbation during quiet stance; this has in part been explained by increased latencies of balance correction accompanying changes to ankle torque with a stiffening strategy (Carpenter, Allum, Honegger, & Adkin, 2004). Therefore, an increase in COP velocity with elevated visual feedback could be indicative of a tighter regulation of postural control due to reduced balance correction latencies.

4.2.2 Sensory contributions to balance control

There was no main effect of optic flow gain on mean power observed within the LOW frequency band. This contradicted the hypothesis which stated that there would be significant changes in mean power below frequencies of 0.1 Hz when optic flow gain was manipulated. Previous literature examining the contributions of each sensory system to balance control have suggested that vision may help to stabilize postural sway below frequencies of 0.1 Hz (Salsabili et al., 2013; Taguchi, 1978); however, results of this study demonstrate that visual contributions to balance control may exist above 0.1 Hz. Further, mean power was observed to significantly decrease with an increase in optic flow gain within the MED frequency band, on both firm and foam surfaces. This was the only frequency band where significant pairwise comparisons were observed while standing on a firm surface, which was a condition where the reliance on vision may have been reduced compared to while standing on foam. Therefore, there is evidence that visual contributions to balance control were strongest in this study between frequencies of sway of 0.1-0.5 Hz. Although these results conflict with earlier work suggesting that vision stabilizes postural sway at frequencies below 0.1 Hz, recent work has demonstrated that VEPRs can be elicited by delivering pseudorandom stimuli through a VR HMD at frequencies between 0.195 - 0.977 Hz (Nielsen et al., 2022). Additionally, phase-coupling to visual perturbations in VR has been demonstrated at a frequency of 1.5 Hz (Engel et al., 2020). Therefore, the discrepancy between these results could be explained by a difference in visual contributions to balance control when exposed to VR-generated, compared to naturally occurring, visual stimuli. Specifically, visual contributions to balance control under the constraints of a VR HMD may exist beyond frequencies of 0.1 Hz.

Additionally, while standing on the foam surface, significant main effects of optic flow gain on mean power were observed within the MED-HIGH and HIGH frequency bands, contrary to the hypothesis. Specifically, mean power was greatest at an optic flow gain value of 16 compared to gain values of 0.25, 2, and 4 within each of these bands, corresponding to frequencies between 0.5-5 Hz. Previous work has suggested that higher frequencies of postural sway are modulated by the vestibular and somatosensory systems (Salsabili et al., 2013; Taguchi, 1978); however, the discrepancy in the results of our study with this previous work could be due to our study manipulating proprioceptive feedback by standing on foam. Having healthy individuals stand on foam to reduce the reliability of proprioceptive feedback could have induced similar sensory re-weighting to clinical populations with sensory and balance deficits. Previous work has demonstrated that both stroke and concussion patients show elevated higher frequency power of postural sway during quiet standing (Schinkel-Ivy, Singer, Inness, & Mansfield, 2016; Sweeny et al., 2021). Stoke patients with high frequency RMS amplitude also had a reduced ability to recover from external perturbations and control their COM within the BOS (Schinkel-Ivy et al., 2016). Further, previous work has shown mean power to increase among higher frequency bands (> 1.8 Hz) when individuals are exposed to a postural threat, such as by standing at an elevated surface height (Zaback et al., 2019). It was therefore concluded that higher frequency bands may be linked to emotional regulation of postural sway (Zaback et al., 2019).

Therefore, although amplified optic flow gain may support a tighter regulation of upright stance, the results of this power spectrum analysis suggest that too much optic flow may be maladaptive to improving postural stability. This is because when optic flow was amplified to a gain value of 16, the increase in mean power within the higher frequency bands could be indicative of poorer reactive balance control by having to exert greater stabilizing moments to control the COM. Additionally, similar results have been seen when individuals are experiencing anxiety at elevated surface heights, therefore a gain of 16 may have reduced balance confidence among participants. This theory that too much visual feedback may be maladaptive is supported by previous work conducted by Jehu et al. (2015), who identified that an optimal magnification of visual biofeedback to support postural stability occurs between 5-10x.

4.3 Functional significance of changes to balance behaviour

Collectively, these results provide evidence that elevated visual feedback supports a tighter regulation of upright stance. As previously mentioned, the results of this study showed that an increase in optic flow gain increases the frequency and decreases the amplitude of postural sway. This suggests that the increased stabilizing moments contributing to an increase in MPF were successful in minimizing the trajectory of sway. Similar changes to frequency and amplitude variables of postural sway have been observed when individuals were elevated to increased surface heights (Adkin, Frank, Carpenter, & Peysar, 2000; Carpenter, Frank, & Silcher, 1999). Specifically, Carpenter et al. (1999) suggested that this decrease in amplitude and increase in frequency is indicative of a stiffening strategy about the ankle joint, where the individual passively controls the COM within a smaller area through activating appropriate muscle tone of ankle plantarflexors. Additionally, experimentally amplifying visual feedback could provide the central nervous system with greater opportunity to generate appropriate postural adjustments. Specifically, this elevated sensory feedback may allow for smaller oscillations of sway to be detected and corrected for before the COM is able to deviate outside of the BOS.

Considering the results of amplitude and frequency changes in combination can also reveal information about sensory feedback loss. Previous work has consistently demonstrated that a loss of visual information, as modelled in this study by a decrease in optic flow gain, contributes to an overall increase in the amplitude and decrease in the frequency of postural sway (Black, Wall, Rockette, & Kitch, 1982; Carpenter et al., 2001). Similarly, the variability of COP and COM trajectory has also been shown to increase with the removal of vestibular feedback (Simoneau, Leibowitz, Ulbrecht, Tyrrell, & Cavanagh, 1992). Therefore, the findings of this study further support the literature suggesting that when feedback from one sensory system used for balance control is reduced, individuals experience larger oscillations of sway at lower frequencies.

Lastly, previous work conducted on patients with multiple sclerosis (MS) has demonstrated a similar decrease in power within lower frequencies of sway (< 0.3 Hz) and an increase in power within higher frequencies of sway (> 1.0 Hz). Specifically, this postural control strategy was uncovered when the reliance of one sensory system was experimentally reduced (Kanekar, Lee, & Aruin, 2014). MS patients shifted reliance away from vision when standing with eyes closed; this was similar to the methodology used in this study, where one sensory system was interfered with by having participants stand on a compliant surface. Interestingly, changes to power within these frequencies were additionally coupled with an increase in sway velocity, which led researchers to conclude that this adapted postural control strategy was not efficient in maintaining balance homeostasis (Kanekar et al., 2014). Similarly, in this present study at an optic flow gain value of 16 mean power among higher frequency bands also increased along with velocity. Therefore, although a tighter regulation of upright stance was achieved at a gain of 16, as seen through a decrease in the amplitude of COP, there is evidence that individuals did not experience improvements to postural stability at such high levels of optic flow.

4.4 Changes to head position

Both HeadPos RMS and mVel decreased with an increase in optic flow gain. Specifically, in comparison to AP COP, HeadPos showed an additional significant decrease in RMS between optic flow gain values of 0.25 and 16. These results suggest that amplitude with respect to head kinematics may be more sensitive to changes in optic flow gain compared to COP. Additionally, there is evidence that the plateau effect of decreasing COP amplitude with optic flow gain, as observed within this study and previous work (Cawsey et al., 2009), may not apply to head kinematics; alternatively, such a plateau may exist beyond a gain value of 16, which was not captured within this study. HeadPos mVel also followed an opposite trend compared to COP mVel, where it was observed to decrease with an increase in optic flow gain, specifically when comparing gain values of 2, 4, and 16 with 0.25. Collectively, these decreases in amplitude and velocity measures could suggest that a head stabilization strategy accompanied an increase in optic flow. These results could be considered under the inverted pendulum model of postural control, which suggests that the body above the ankle acts as a rigid segment and rotates about this joint (Winter et al., 1998). The displacement of each COM segment increases linearly with height above the ankle joint (Gage, Winter, Frank, & Adkin, 2004); therefore, under the assumption of this model, there is greater freedom of head sway which could contribute to the lack of plateau in HeadPos RMS observed in this study.

Further, it is possible that the body attempted to stabilize retinal image motion when optic flow gain was experimentally elevated. To achieve this, head motion could have been attenuated to reduce optic flow to what would be experienced in a natural setting. Previous work has suggested that retinal slip velocity must remain below 2-4º/s to maintain adequate visual acuity (Demer, Honrubia, & Baloh, 1994). Therefore, image stabilization could have been achieved through reducing head translations and rotations to minimize eye movement (Crane & Demer, 1998).

Lastly, there was no main effect of optic flow gain on HeadPos MPF, which was contradictory to the hypothesis which stated that all MPF variables would increase with an increase in optic flow. Extending the above explanation of head stabilization, previous work has suggested that the vestibulo-ocular reflex (VOR) may not be well-calibrated to support the lower frequencies of postural sway that accompany dynamic standing tasks, as opposed to during gait (Crane & Demer, 1998). Specifically, visual pursuit and adequate head stability among healthy populations permit the VOR to function at a lower degree of precision under these balance conditions to keep retinal slip velocity below 2-4º/s (Crane & Demer, 1998). This study focused on quiet standing, which naturally contributes to even lower frequencies of sway than dynamic standing; therefore, it is possible that sway frequency about the ankle joints was not great enough to initiate high frequency, corrective head stabilization strategies through the VOR.

4.5 Comparing AP and ML sway

Across the ML direction for COP motion, there was no main effect of optic flow gain on velocity observed. Although there was a significant main effect of optic flow gain on frequency, less significant pairwise comparisons were observed than in the AP direction. Similarly, there was a significant main effect of optic flow gain on amplitude; however, COP ML RMS was only statistically different between the largest and smallest gain values: 0.25 and 16. Cumulatively, there is evidence that ML sway was less sensitive to changes in VR HMD-manipulated optic flow than AP sway. VR HMDs expose participants to a reduced field of view compared to what would be observed naturally in a real-world environment (Nagata, 1996). As a result, this experimental paradigm could have reduced the availability of peripheral visual information that participants use to stabilize balance. Previous work has suggested that the peripheral, as opposed to central, visual field is especially sensitive to changes in optic flow (Horiuchi, Ishihara, & Imanaka, 2017). For example, among patients with diabetic retinopathy, an increase in peripheral retina damage is related to poorer overall postural stability (Piras, Perazzolo, Scalinci, & Raffi, 2021). Additionally, peripheral vision is often characterized by high temporal frequency, whereas central vision is often characterized by high spatial frequency (Kelly, 1984). This suggests that peripheral vision is sensitive to detecting changes in movement (Horiuchi et al., 2017; O'Connell et al., 2017). Therefore, it is possible that the use of a HMD could have reduced participants' sensitivity to changes in visual feedback.

This study sample also consisted of young, healthy adults. Aging and specific diseases, such as Parkinson's disease, are closely linked to deterioration of postural stability, specifically in the ML direction; reduced trunk flexibility and a deficient in hip abductor–adductor muscle torque-time capacity have been cited as possible factors for this decline (Horak, Dimitrova, & Nutt, 2005; Mille, Johnson, Martinez, & Rogers, 2005). Additionally, due to biomechanical constraints, it is well-known that displacements of postural sway occur primarily within the AP compared to ML plane (Fearing, 1924). Therefore, having a young, healthy sample combined with using a VR HMD could suggest that more extreme changes in optic flow gain needed to be induced to see changes to balance behaviour in this plane of motion. This explanation is supported by Cawsey et al. (2009) who showed changes to COP ML amplitude only between visual biofeedback magnifications of 0 and 1, with 0 representing an absence of biofeedback.

Lastly, previous work examining individuals' postural response at elevated surface heights has also identified less significant changes to ML, compared to AP sway (Carpenter et al., 1999). Carpenter et al. (1999) suggested that emotional regulation may contribute to this discrepancy, where the CNS perceives a greater risk of falling in the forwards direction and thus, attempts to stabilize COM within this plane to protect itself from falling in this direction (Carpenter et al., 1999). Therefore, participants in this study may have been more concerned about falling in the forwards direction, so greater adaptations to AP sway were applied as the gain of optic flow changed across trials.

4.6 Influence of standing on a firm and foam surface

While standing on a firm surface, no significant main effect of optic flow gain on COP measures was observed across either mVel or MPF, while RMS showed a significant main effect of gain irrespective of surface condition. Collectively, these results provide evidence that standing on a non-compliant surface did not allow for vision to be relied upon heavily enough for balance behaviour to be influenced by optic flow gain. These findings contradict the hypothesis which stated that balance behaviour would be influenced by optic flow gain on both surface conditions, but that these effects would be greater while standing on foam. While exposing participants to manipulated visual biofeedback of their COP trace, previous work did identify significant changes in COP outcome measures while standing on a firm surface (Cawsey et al., 2009; Jehu et al., 2015); however, this method of changing visual feedback did not manipulate participants' optic flow directly, as visual feedback was delivered through having participants view their COP trace on a screen. In contrast, previous work that directly reduced the overall availability of optic flow through visual occlusion techniques found that balance behaviour was not affected unless proprioceptive information was concurrently manipulated (O'Connell et al., 2017). In this study, optic flow was manipulated directly through a VR HMD; therefore, while proprioceptive information is reliable, such as by standing on a non-compliant surface, it is possible that healthy, young adults do not place a great enough reliance on vision for balance behaviour to be affected under conditions of manipulated optic flow. Lastly, while standing on a firm surface, previous work showed that significant changes in COP amplitude were observed between biofeedback magnifications of 0 and 1 (Cawsey et al., 2009), whereas the lowest optic flow gain value used in this study was 0.25. Thus, while standing on firm surfaces, where there may be less reliance on vision, postural stability may only be compromised (as demonstrated by an increase in sway amplitude) when visual feedback is severely reduced.

Lastly, the results of this study further support previous work which has suggested that standing on a compliant surface may increase visual contributions to balance control (Kars, Hijmans, Geertzen, & Zijlstra, 2009; Schut et al., 2017). Main effects of optic flow gain on mean power were observed across each frequency band, excluding the LOW band, while standing on foam. In contrast, while standing on a firm surface, a main effect of optic flow gain on mean power was only observed within the MED frequency band. This supports the widely held theory that compliant surfaces, such as foam, reduce the gain of proprioceptive feedback through disrupting feedback surrounding ankle rotation, and that this upregulates the gain of intact sensory systems, such as the visual system (Kars et al., 2009; Schut et al., 2017).

4.7 Comparing the postural response before and after VR

Before VR exposure, baseline quiet standing results showed that participants experienced a decrease in AP COP amplitude and increase in frequency compared to quiet standing after VR. Further, this change in COP frequency was explained by an increase in mean power within the LOW frequency band and decrease in mean power within the MED-HIGH band, as well as HIGH band within the ML direction. Collectively, these results suggest that when first exposed to new balance task, participants maintained tighter control of COP within the BOS. It is possible that this postural control strategy was due to heightened arousal or anxiety during the first trial of this experimental paradigm. For example, when individuals are first exposed to a postural threat, such as by standing at an elevated surface height, a decrease in amplitude and increase in frequency of postural sway was observed (Zaback, Reiter, Adkin, & Carpenter, 2021; Zaback et al., 2019). Further, after repeated exposure to this threat, individuals' emotional response was attenuated, in addition to mean power among sway frequencies above 1.8 Hz (Zaback et al., 2019). Therefore, participants in this study could have experienced a first trial effect, where exposure to an unfamiliar balance task contributed to a tighter regulation of upright stance.

Additionally, previous work has cited the influence of the "white coat effect" during clinical assessments of balance control (Geh, Beauchamp, Crocker, & Carpenter, 2011). Specifically, when participants feel as though they are being evaluated or closely observed, social anxiety could be elevated from the desire to perform well on the task at hand. In balance control tasks, when under assessment by a clinical evaluator, older adults showed higher frequencies of COP oscillations (Geh et al., 2011). Thus, it is also possible that participants in this study were influenced by social anxiety and feeling the need to perform well in front of the researchers. As a result, higher frequencies of COP oscillations could have been generated to maintain a tighter control of COM within the BOS. After completing the VR trials, participants may have then felt more comfortable under observation; this would explain why COP frequency decreased and amplitude increased during the second baseline quiet standing condition.

Lastly, this postural control strategy has been cited as being beneficial to balance control, rather than compromising overall stability. Patients with phobic postural vertigo, a diagnosis combining dizziness with psychological factors such as anxiety and obsessions, demonstrate an increase in sway at 3.5–8 Hz while standing on foam; however, patients' amplitude of sway is often equivalent to healthy controls (Krafczyk, Schlamp, Dieterich, Haberhauer, & Brandt, 1999). This suggests that both these patients and participants from this present study adapted their postural strategy when faced with increased anxiety, and that this strategy successfully kept their COP kept well within the BOS.

4.8 Limitations

4.8.1 Technological constraints

The first limitation of this thesis encompasses technological constraints. First, head kinematics could only be captured in the AP direction. As a result, the postural response in the ML direction could only be interpreted from COP measures. This is a limitation as it was not possible to understand whole body mechanics as it relates to ML sway in response to manipulated optic flow gain. For example, in the AP direction, COP combined with HeadPos data provided additional information as to a potential head stabilization strategy that could have influenced the postural response. Further, there is evidence to suggest that central and peripheral visual cues may play distinct roles in stabilizing both AP and ML sway (O'Connell et al., 2017; Piras et al., 2021; Radvay et al., 2007). If ML head kinematics were captured, additional information could have been uncovered as to if a VR HMD limits peripherally controlled ML sway.

A second technological limitation within this thesis was not being able include an optic flow gain value of 1 within analyses. Although initially included within the Vizard python script, a technical error during data collection led this gain value to be excluded from the experimental protocol. An optic flow gain value of 1 would have provided a baseline comparison in VR, through providing participants with equivalent visual feedback through scene motion as was generated by head motion captured through the HMD. This trial would have allowed the remaining gain values (0.25, 2, 4, and 16) to be compared directly to “normal” optic flow generated in VR. Additionally, comparing balance behaviour at a gain value of 1 to both before and after VR baseline trials would have provided additional information as to if this VR HMD was able to produce seemingly equivalent optic flow as seen in a real-world setting, especially while standing on foam. This would help to extend previous work conducted by Assländer & Streuber (2020), which showed that photo-realistic VR environments can produce similar postural responses compared to real-world settings, while standing on a non-compliant surface. Although a gain value of 1 was not included within the analyses of this thesis, pilot data collection was conducted on seven participants (mean age (±SD): 22.0 (±2.94) years, five female), including the following gain values delivered in a counterbalanced order: 0.25, 1, 2, 4, 16. On average, in continuation with the trends observed in this thesis, COP and HeadPos RMS decreased with an increase in optic flow gain (Appendix C).

4.8.2 Lack of psychological assessment

This thesis was also limited by not containing a psychological assessment to understand participants' balance confidence throughout the study. Specifically, this study demonstrated that an increase in VR-generated optic flow can create a tighter regulation of postural control; however, combining a qualitative assessment of balance confidence would provide further information as to whether this postural control strategy equated to overall improved postural stability, or if it was a compensatory strategy coupled by fear of falling. This thesis provided evidence that at very high levels of optic flow (gain value of 16), postural stability may not have been improved, even though a tighter control of upright stance was achieved. Specifically collecting information regarding participants' task-specific balance efficacy (Hauck, Carpenter, & Frank, 2008) would have helped to support this theory. Lastly, an assessment of balance confidence would have helped to support the overall clinical utility of VR HMDs in improving postural control among patients with balance deficits. Specifically, understanding individuals' perception of stability throughout this experimental paradigm would further validate the use of amplified optic flow in rehabilitation settings.

4.9 Recommendations for future research

This thesis specifically evaluated quiet stance while being exposed to VR-manipulated optic flow gain. To build upon this work, future research could be directed towards dynamic standing conditions, such as by exposing participants to surface support translations or visual perturbations. VR HMDs have been used in the past to generate visual perturbations with the goal of assessing and training dynamic balance (Heidner et al., 2020; Peterson, Rios, & Ferris, 2018). Therefore, by using VR HMDs to amplify optic flow gain in dynamic standing conditions, valuable insight could be gained as to how we use visual feedback to maintain postural stability during recovery from external perturbations.

Further, this work focused on examining balance behaviour among healthy young adults; therefore, future work could be directed to specific populations of interest. For example, the foam surface condition used within this thesis may have probed similar sensory reweighting as seen among clinical populations, such as older adults with proprioceptive deficits (Sundermier et al., 1996). However, rather than generalizing the results of this foam condition to such clinical populations, it is important to specifically assess the relationship between balance behaviour and VR-manipulated optic flow among these groups. This is especially important if this work is to be used in a rehabilitative context in the future to improve postural stability among patients with balance deficits.

Lastly, the final recommendation for future research is to incorporate EMG and eye-tracking technology within similar VR and balance control experimental paradigms. Specifically, EMG around ankle stabilizing muscles and eye-tracking software built into a HMD could be added. Previous work has used EMG activity to provide evidence for improved corrective postural adjustments while standing with eyes open compared to eyes closed on foam (Fransson, et al., 2007). Being able to assess EMG activity in response to manipulated optic flow of varying degrees would provide useful knowledge as to whether elevated feedback is truly beneficial in supporting postural stability. Specifically, adding EMG recordings about ankle stabilizing muscles could provide additional information as to whether a compensatory stiffening strategy is adopted at high levels of optic flow gain. Additionally, recent work has demonstrated the efficacy of using eye-tracking software built into HMDs as a way to assess visual fatigue (Souchet, Philippe, Lourdeaux, & Leroy, 2021). For example, characteristic blinking and eye movement features have been discovered to accompany VR HMD use, leading to visual fatigue and eventually visually induced motion sickness (Wang et al., 2019). Thus, eye-tracking data could provide an objective measure to assess the overall clinical utility of using VR HMDs to modify optic flow gain for rehabilitative purposes.

4.10 Clinical applications

Falls among older adults are a leading contributor of injury, death, and overall poorer quality of life (Hindmarsh & Estes, 1989); thus, designing fall prevention and rehabilitation programs that support patients with balance deficits, such as older adults, is invaluable. There is evidence that older adults may place an increased reliance on vision during balance control tasks (Peterka & Loughlin, 2004; Sundermier et al., 1996). Further, this increase in visual dependency has been linked to a reduced ability to adapt to changes in visual conditions, which can increase the risk of falling (Teasdale et al., 1991). Aging has been linked to a decline in vestibular function and somatosensory impairment at the lower limb which can probe sensory reweighting (Henry & Baudry, 2019; Rosenhall & Rubin, 1975), similar to what was observed in this study when healthy adults stood on foam. Therefore, manipulating visual feedback through VR technology may be a promising intervention to support older adults who have an increased dependency on vision during balance tasks. Specifically, future work with VR could be used to develop fall prevention and rehabilitation programs that expose older adult patients to elevated optic flow gain with the overall goal of developing tighter regulation of upright stance.

4.11 Conclusions

In conclusion, the overall findings of this thesis demonstrate that manipulating optic flow gain in VR influences balance control during quiet stance among healthy adults. First, when there is an increased reliance placed on vision, such as by standing on a compliant surface, an increase in optic flow gain can support tighter regulation of upright stance; however, when optic flow is amplified to very high degrees (gain of 16), too much visual feedback may no longer improve postural stability. Second, it was discovered that visual contributions to balance control may extend to higher frequencies of postural sway than previously theorized, specifically above frequencies of 0.1 Hz. This present study showed how a first trial effect may also lead to a tighter regulation of COP within the BOS. Overall, future research can continue to use VR to manipulate optic flow among balance deficit patients, such as older adults, to assess the utility incorporating such VR interventions within fall prevention programs.

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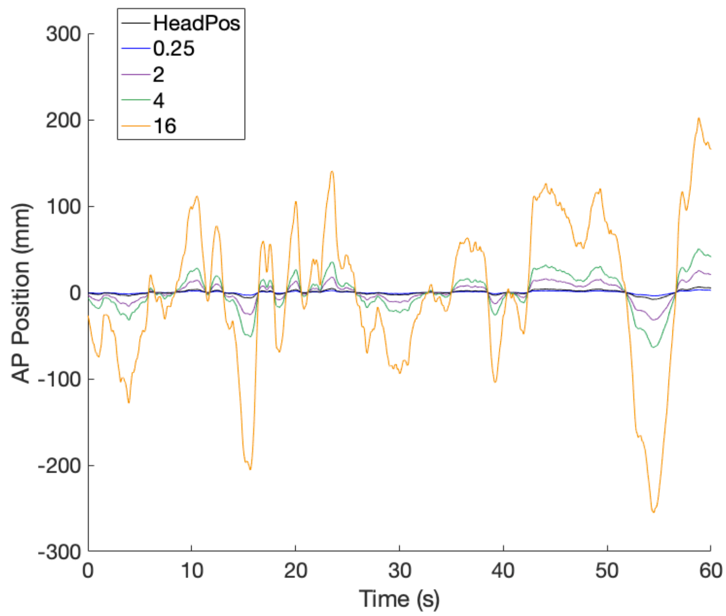
Appendix A – Manipulating optic flow gain

(A)



(B)

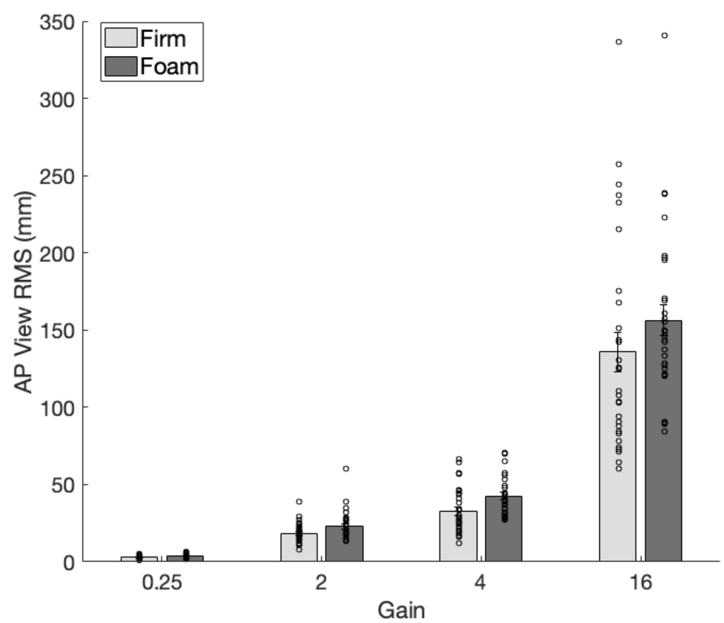


Figure 13. Head and view position

(A) Head position of a representative participant across a 60-second trial, with view position amplified and reduced to demonstrate scene motion at each gain value. **(B)** View position amplitude across each gain value for all participants.

Appendix B – Mean (SEM)

Table 3. Mean (SEM) summary for baseline quiet standing

Results shown for RMS (mm), MPF (Hz), mVel (mm/s), and mean power (mm^2/Hz) across each frequency band (LOW, MED, MED-HIGH, HIGH).

		AP COP		ML COP	
		Before	**After**	**Before**	**After**
RMS	Firm	5.34 (0.40)	5.72 (0.41)	2.79 (0.18)	2.99 (0.22)
	Foam	6.10 (0.30)	6.93 (0.42)	4.74 (0.23)	5.27 (0.33)
MPF	Firm	0.21 (0.02)	0.17 (0.02)	0.29 (0.02)	0.30 (0.03)
	Foam	0.25 (0.02)	0.19 (0.02)	0.34 (0.02)	0.23 (0.02)
mVel	Firm	8.81 (0.38)	8.72 (0.37)	6.94 (0.30)	6.90 (0.29)
	Foam	11.89 (0.57)	11.14 (0.56)	11.33 (0.69)	10.21 (0.52)
LOW	Firm	19.27 (4.25)	23.86 (4.20)	4.54 (0.90)	5.54 (1.66)
	Foam	17.22 (2.77)	42.91 (6.70)	8.82 (1.20)	19.44 (3.78)
MED	Firm	6.38 (1.24)	5.62 (0.65)	2.67 (0.54)	2.82 (0.53)
	Foam	15.83 (1.74)	13.89 (1.66)	8.35 (0.96)	9.63 (1.08)
MED-HIGH	Firm	1.07 (0.29)	0.81 (0.12)	0.52 (0.07)	0.51 (0.13)
	Foam	3.13 (0.39)	1.91 (0.25)	3.66 (0.70)	1.83 (0.25)
HIGH	Firm	0.57 (0.09)	0.54 (0.07)	0.31 (0.04)	0.30 (0.04)
	Foam	1.04 (0.12)	0.96 (0.11)	1.02 (0.14)	0.88 (0.11)

Table 4. Mean (SEM) summary for experimental trials

Results shown for RMS (mm), MPF (Hz), mVel (mm/s), and mean power (mm^2/Hz) across each frequency band (LOW, MED, MED-HIGH, HIGH).

		AP COP				ML COP				AP HeadPos			
		0.25	**2**	**4**	**16**	**0.25**	**2**	**4**	**16**	**0.25**	**2**	**4**	**16**
RMS	Firm	6.29 (0.44)	5.67 (0.41)	5.32 (0.40)	5.59 (0.47)	3.20 (0.22)	3.10 (0.20)	2.77 (0.16)	2.75 (0.20)	10.45 (0.97)	8.95 (0.57)	8.26 (0.87)	8.12 (0.96)
	Foam	8.94 (0.44)	7.95 (0.54)	7.65 (0.42)	7.70 (0.50)	6.87 (0.44)	6.56 (0.36)	5.98 (0.31)	5.83 (0.39)	13.41 (0.89)	11.68 (1.15)	10.52 (0.63)	10.05 (0.88)
MPF	Firm	0.19 (0.02)	0.20 (0.02)	0.23 (0.03)	0.23 (0.03)	0.28 (0.03)	0.25 (0.02)	0.31 (0.04)	0.30 (0.03)	0.09 (0.01)	0.09 (0.01)	0.09 (0.01)	0.07 (0.01)
	Foam	0.23 (0.02)	0.23 (0.02)	0.29 (0.02)	0.42 (0.03)	0.25 (0.02)	0.24 (0.02)	0.32 (0.02)	0.39 (0.04)	0.11 (0.01)	0.09 (0.01)	0.10 (0.01)	0.12 (0.01)
mVel	Firm	20.82 (1.29)	20.00 (1.04)	19.95 (1.24)	21.04 (1.32)	14.49 (0.70)	14.12 (0.61)	14.08 (0.73)	14.05 (0.69)	5.78 (0.26)	5.22 (0.26)	4.85 (0.23)	4.82 (0.29)
	Foam	24.77 (1.18)	23.60 (1.17)	24.65 (1.15)	29.21 (1.54)	19.56 (0.93)	19.20 (0.77)	19.47 (0.89)	21.14 (1.21)	9.55 (0.38)	7.93 (0.31)	7.52 (0.33)	8.00 (0.50)
LOW	Firm	22.97 (4.40)	18.22 (2.87)	19.72 (3.69)	21.01 (3.98)	5.70 (1.51)	4.52 (0.90)	4.11 (0.74)	4.12 (0.73)	78.15 (18.55)	50.23 (8.93)	65.13 (15.69)	51.41 (12.66)
	Foam	42.91 (6.70)	39.34 (9.28)	32.85 (5.64)	23.80 (3.99)	26.43 (5.30)	21.79 (3.08)	16.82 (2.54)	13.34 (2.57)	143.08 (31.85)	127.02 (35.09)	83.68 (13.64)	80.40 (16.90)
MED	Firm	8.95 (0.99)	7.60 (1.09)	7.14 (1.20)	4.73 (0.83)	3.31 (0.55)	3.02 (0.54)	2.29 (0.40)	2.08 (0.31)	21.89 (2.35)	13.76 (2.02)	12.34 (2.23)	10.48 (2.03)
	Foam	35.66 (3.18)	21.65 (2.34)	16.79 (1.89)	12.37 (1.74)	22.96 (3.39)	16.62 (2.56)	12.26 (1.38)	11.47 (1.72)	61.44 (5.80)	32.75 (4.81)	25.73 (3.88)	21.21 (4.31)
MED-HIGH	Firm	1.03 (0.13)	1.23 (0.16)	1.15 (0.15)	1.26 (0.17)	0.61 (0.10)	0.48 (0.06)	0.48 (0.06)	0.54 (0.09)	0.53 (0.08)	0.53 (0.06)	0.56 (0.10)	0.48 (0.07)
	Foam	5.37 (0.90)	5.54 (0.77)	6.67 (0.81)	13.11 (2.13)	5.21 (1.18)	4.75 (0.79)	6.07 (0.94)	6.71 (1.45)	1.43 (0.26)	1.52 (0.25)	1.75 (0.26)	2.76 (0.45)
HIGH	Firm	0.68 (0.09)	0.67 (0.08)	0.64 (0.08)	0.96 (0.16)	0.33 (0.05)	0.29 (0.03)	0.28 (0.04)	0.37 (0.05)	0.09 (0.01)	0.11 (0.02)	0.10 (0.01)	0.11 (0.02)
	Foam	2.49 (0.34)	1.91 (0.20)	2.32 (0.28)	4.67 (0.68)	1.57 (0.24)	1.28 (0.17)	1.36 (0.18)	2.91 (0.69)	0.17 (0.03)	0.17 (0.03)	0.18 (0.02)	0.31 (0.04)

Appendix C – Pilot work including optic flow gain of 1

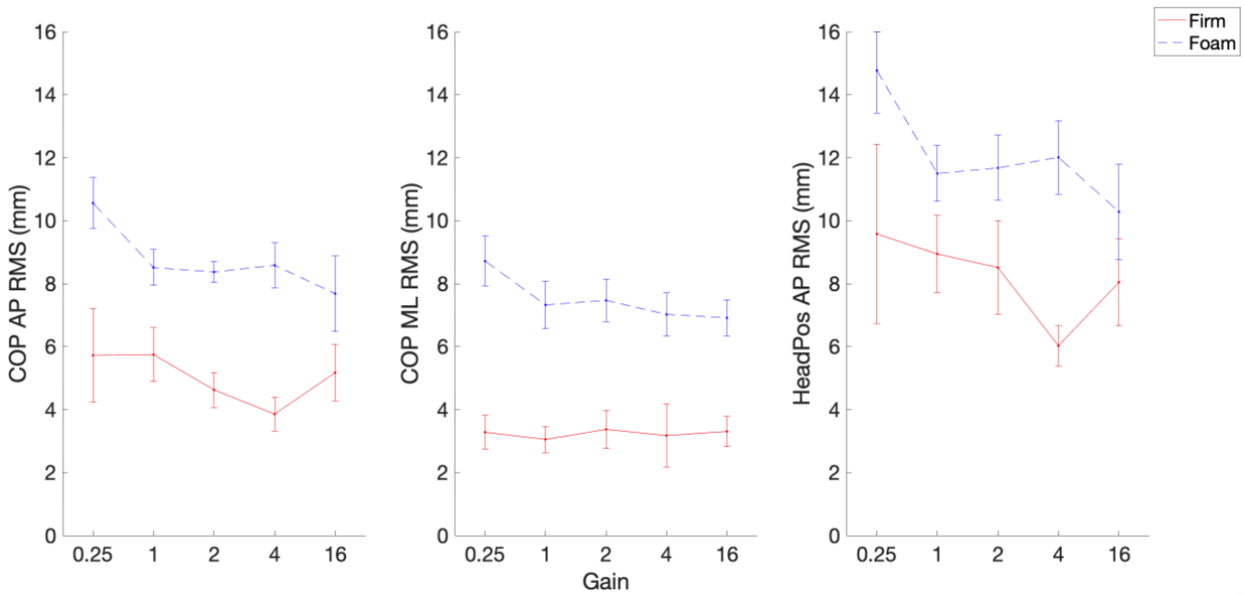


Figure 14. RMS across experimental trials for pilot work

Changes to AP and ML COP and AP HeadPos RMS amplitude across optic flow gain values of 0.25, 1, 2, 4, and 16.