EFFECTS OF MANIPULATING OPTIC FLOW GAIN ON DYNAMIC POSTURAL CONTROL DURING CONTINUOUS SUPPORT SURFACE TRANSLATIONS

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ABSTRACT:

Human postural control involves interactions between visual, vestibular, and somatosensory systems to maintain upright stance and prevent falls. Visual information can be amplified or reduced through changes in optic flow gain relative to sway. Previous work has shown increased postural sway when visual feedback is reduced during quiet stance. Dynamic postural control has also been assessed through use of moveable support surfaces (SS). There has been limited work utilizing VR-induced optic flow gain in combination with continuous support surface translations to examine visual contributions to balance in dynamic stance situations. In this study, optic flow gain factors were used to alter the visual information associated with participants' movement. SS oscillations were used to perturb the participant in the A-P direction. Kinetic and kinematic information was collected and thus balance behaviour was quantified through center of mass (COM) & center of pressure (COP) amplitude (root mean square, RMS), frequency (mean power frequency, MPF), COP velocity, as well as joint and segment angular changes.

It was hypothesized that exposure to increased optic flow gain would result in tighter regulation of postural control, reflected as a decrease in postural sway amplitude, and an increase in sway frequency and velocity. Conversely, as gain values decrease, the opposite effects were hypothesized. Analyses were split into the first and second minutes of each trial to assess the effects of perturbation duration within the experiment. It was found that as gain values increased, participants experienced a relative decrease in sway amplitude, and increases in both sway frequency and velocity, thus reflecting a tighter regulation of stance as greater visual information was provided. These changes were generally more evident in the second minute of trials, where participants were exposed to the perturbation for a greater duration. By further examining dynamic postural control and its relationship with optic flow through VR, this thesis demonstrated the effectiveness of utilizing visual information to impact postural behaviours in young, healthy adults.

DEDICATION

I’d like to dedicate the work within this thesis to those I’ve unfortunately lost throughout its completion.

To my grandfather, Avi Alon. The first man in his name to hold a graduate degree. One of the greatest minds and kindest hearts I’ll ever know, and I’m proud to carry his name through academia and onwards. Thank you for cheering me on, I know you’ve been watching. This one’s for you.

To my cousin Mitchell Laurenza. The most luminous, magical soul I’ve ever had the pleasure of meeting. The joy you brought to us all is forever missed, and your strength is something I carry with me every single day. I know you’d get a kick out of this. This one’s for you too.

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LIST OF ABBREVIATIONS

The following table highlights common abbreviations, acronyms, and short forms found throughout the thesis proposal., as well as their scientific significance. Item meanings marked with (b) indicate bony anatomical landmarks.

Abbreviation	Meaning
AC	acromion process of scapula (b)
ANOVA	analysis of variance
AP	anteroposterior
APA	anticipatory postural adjustment
BOS	base of support
CCI	co-contraction index
CLC	calcaneus (b)
CNS	central nervous system
COG	center of gravity
COM	center of mass
COP	center of pressure
FH	head of fibula (b)
FthMT	base of 5th metatarsal (b)
HMD	head-mounted display
GT	greater trochanter of femur (b)
GTO	Golgi tendon organ
HA	hip-ankle
HK	hip-knee
Hz	Hertz
KA	knee-ankle
LE	lateral epicondyle of humerus (b)
LFC	lateral condyle of femur (b)
LM	lateral malleolus of ankle (b)
MVel	mean velocity
ML	mediolateral
MP	mastoid process (b)
MPF	mean power frequency
PD	Parkinson’s disease
RMANOVA	repeated measures analysis of variance
RMS	root mean square
SD	standard deviation
SS	support surface
VCN	vestibulocochlear nerve
VR	virtual reality
ZY	zygomatic arch (b)

CHAPTER 1: INTRODUCTION

1.1 Overview of thesis

Maintaining control of upright stance is paramount for humans to avoid falls and go about activities of daily living. To maintain postural equilibrium, constant input from multiple relevant sensory systems is integrated together to cumulatively establish an accurate representation of body position and orientation relative to the external environment. The use of visual feedback via optic flow gain can provide additional sensory information for the visual system to analyze, and therefore integrate towards the maintenance of upright stance in both quiet and dynamic stance environments. Since the use of optic flow gain is a fairly novel experimental protocol, examining its utility in a laboratory setting can help educate and establish on the complex relationship that exists between sensory integration and control of dynamic balance while using external environmental stimuli. Chapter 1 of this thesis provides a literature review of previous work examining optic flow, vision, and standing balance control, as well as their applicability toward balance maintenance in dynamic environments. Chapter 2 of this thesis outlines the methodologies used to study the effects of optic flow gain on dynamic balance control during continuous support surface (SS) translations. Lastly, chapters 3 and 4 present the findings of this thesis study, as well as discuss their relevance and utility within the context of current literature surrounding posturography.

1.2 Control of upright posture

Upright bipedal stance, or posture, is an automatic and unconscious position that humans assume in everyday tasks (Carini et al, 2017). The position of the body in space during an upright posture is maintained for the purpose of upright balance while providing maximum stability and minimal energy expenditure (Pastorelli & Pasquetti, 2013), which is paramount to efficient biomechanical and physiological function. Postural control is often referred to as the capabilities of sensorimotor mechanisms to maintain the body's centre of gravity (COG) over the base of support (BOS), although the processes that perform this seemingly basic function, especially during balance perturbations, remain deeply complex (Carini et al, 2017; Olchowik et al, 2014; Winter et al, 1990). Upright posture, or balance, can be maintained due to the interactions of sensory systems, which allow an individual to remain upright and avoid a fall or loss of balance. The vestibular and somatosensory systems work very closely in unison with the visual system in order to maintain a given posture (Carini et al, 2017; Horak, 2006) which makes their functions essential to upright balance. The integration of these systems together can entail a constant re-weighting of their relative inputs throughout an everyday task for the goal of maintaining sensory equilibrium (Horak, 2006).

The vestibular system is responsible for interpreting information regarding head position in space, based on the position of the vestibular apparatus in the inner ear relative to the orientation of the head. Within an inner layer of the temporal bone, a net work of bony holes (labyrinth) surrounds a membranous labyrinth filled with endolymph, a potassium-rich lymph compound (Kandel et al, 2013). This fluid travels throughout anterior vertical, lateral, and posterior vertical semicircular canals within the vestibular apparatus, and these canals are able to sense the rotations of the head based on the locations of the fluid at any given point. Additionally, anterior to the semicircular canals, there exists the utricle and saccule (otherwise known as otoliths) which are able to sense all linear accelerations of the head, including gravity (Fernandez & Lindsay, 1962; Ramos de Miguel et al, 2020). The otoliths act as receptor organs as they contain specific hair cells that are responsible for sensing these fluid shifts. Any accelerations of the head cause shear force to be exerted between the layers of the otolith and thus the positions of the hair cells can change, causing changes in membrane potential. These excitation and inhibition signals can be converted into vestibular signals, which then travel to the brain stem through the vestibulocochlear nerve (VCN) (Highstein & Holstein, 2012; Ramos de Miguel et al, 2020). The vestibular portion of the VCN enters the brainstem at the junction of the pons and medulla, and projects signals into the vestibular nuclei located lateral to the sulcus limitans, and medial to the inferior cerebellar peduncle (Highstein & Holstein, 2012). From here, projections are sent into the thalamus and cerebellum, further down the spinal cord via the vestibulospinal tracts, and also to extraocular motor nuclei to activate the vestibulo-ocular reflex (Yoo & Mihaila, 2022).

The somatosensory system is made up of a network of neurons that process internal as well as external tactile stimuli to establish a sense of body positioning in space, otherwise known as proprioception (Kropf et al 2019). The two main functions of the somatosensory system include the transduction of tactile and proprioceptive signals into electrical stimuli, and the transmission of such signals towards the central nervous system (CNS) for processing. Dorsal root ganglia are clusters of bipolar cells that transmit signals from peripheral body structures (skin, organs, muscles, etc.) while they synapse with the CNS on the other end. These are also called the primary afferent nerve fibers as they carry information from the periphery to the CNS (Kandel et al, 2013). Sensory fibers within the somatosensory system with the highest conduction velocities are type Ia afferents from muscle spindles, and Ib afferents from Golgi tendon organs (GTOs) (Tsunozaki & Bautista, 2009). Muscle spindles, embedded within skeletal muscle bellies, act as mechanoreceptors which respond to stretch in muscle fibers and rate of change in muscle length. The CNS can efferently direct muscle spindles through use of gamma motor neurons, which subsequently control the quantity of afferent signal outputs towards each respective alpha motor neuron (Oliver et al, 2021). This control loop demonstrates how the CNS can modulate muscle spindle activity. GTOs are located within the myotendinous junction of skeletal muscles. Since they are innervated by type Ib afferents, they are phenotypically different mechanoreceptors, less sensitive towards multi-directional muscle stretch and often more selective towards motor units oriented in series with the respective tendon in which they innervate. GTOs can interpret changes in relative effort through muscle contractile force, and will often elicit inhibitory signals towards their innervated motor neurons to prevent excess loading of the tissue and subsequent injury (Oliver et al, 2021). Through these proprioceptive reflex circuits, the CNS can interpret afferent feedback generated by skeletal muscles at different joints, and produce a motor action that would be necessary for postural compensation in response to a stimulus (Hayes, 1982).

1.3 Neurophysiology of vision

The visual system works to provide essential sensory feedback to individuals and allow them to process visual information as they navigate everyday activities. The visual system is able to process different patterns of light as they enter the eye within various environments, and therefore allow an individual to interpret and appropriately react to any stimuli that they may face. As stimuli within a particular environment begin to change, the patterns of light that are reflected onto the eyes change over time as well, and the combinations of these changes in motion structures of light make up what can be recognized as flow of visual information, or optic flow (Ludwig et al, 2018). Optic flow can be generated by the relative movement of an individual within an environment, or simply just the environment itself, and these distinctions play an important role in how optic flow can be perceived by an observer (Royden et al, 1992). Therefore, optic flow is comprised of the changes in retinal image streams that come across the eyes as incoming information from the surrounding environment changes (Harris et al, 2000).

Once light from the environment reaches the eye, the pupil's structure permits the light to pass through the cornea and lens, and get refracted and focused onto the retina which lines the back of the eye. The photoreceptor cells within the retina absorb the light and convert the photons into an interpretable neural input through phototransduction. These signals travel into retinal ganglion cells at the back of the retina, whose axons converge to form the optic nerves (Kandel et al, 2013). Input from the optic nerves is directed to both the ipsilateral side of the visual field, and partially decussates contralaterally to carry information from respective sides of the visual field through both sides of the optic chiasm. Information is passed through the lateral geniculate nucleus in the thalamus to be interpreted, and terminates in the primary visual cortex within the occipital lobe (Rezaei & Saghazadeh, 2019).

There are many neural pathways that extend toward the primary visual cortex. The first is horizontally through the calcarine fissure of the occipital lobe. Another exists from the retina to the superior colliculus, into the pons and then directly into motor nuclei of the orbital muscles, which control movements of the eye (Yoo & Mihaila, 2022). The last pathway extends from the midbrain through neurons that control pupillary reflexes of the eyes (Rezaei & Saghazadeh, 2019). As for pathways originating from the primary visual cortex, certain pathways travel ventrally to the temporal lobe containing information on the nature of a stimulus, and others will travel dorsally to the parietal lobe containing spatial information about a stimulus (Kandel et al, 2013; Rezaei & Saghazadeh, 2019). The information that enters both hemispheres is integrated through the corpus callosum at the junction of both hemispheres (Rezaei & Saghazadeh, 2019).

1.4 Visual contributions to balance

When considering the relevant sensorimotor contributions to balance, the visual system plays an integral role in postural control. As displayed by the Romberg diagnostic test, when a subject is asked to remain standing still with eyes open and then closed, individuals will experience much greater postural sway with eyes closed, compared to eyes open. This is a simple test that highlights the dependence on the visual system for balance maintenance (Bronstein, 2019; Horak & Macpherson, 1996). More specifically, visual feedback paradigms can help further investigate the intricacies of the visual system. Visual information can be amplified or reduced through changes in optic flow magnification (or gain) relative to one's postural sway. Increasing visual biofeedback has been linked to greater control of posture in quiet stance, indicating a decrease in sway amplitude and COP variability (Cawsey et al, 2009; Jehu et al, 2016). This effect was supported in Assländer et al (2015), where stroboscopic illumination frequencies as low as 3 Hz suggest an attenuation of sway measures compared to eyes closed conditions. Conversely, decreased optic flow has been shown to negatively impact postural control and therefore increase sway (Bronstein, 2019; Jilk et al, 2014), as subjects have less visual information available to perform a necessary postural compensation (Horak, 1987).

Postural adjustment strategies may also be sensitive to visual positional and velocity cues. Inputs to the visual system provide an individual with orientation information relative to their baseline head position, as well as movement information about the environment (Kandel et al, 2013), or visual target they are focusing on. Stroboscopic illumination frequencies below 4 Hz can be processed by the visual system as positional cues in the form of displacement information (Assländer et al, 2015), whereas above

4 Hz, the system begins to place a greater dependence on visual velocity and orientational cues which tend to cause a reduction in sway characterized by reduced COM amplitude, thus eliciting tighter control of posture (Assländer et al, 2013; Paulus et al, 1984). This was supported by Jilk et al (2014) which observed greater COM variability and increased sway with decreased visual information present during a pseudorandom oscillation sequence. Additionally, if positional information about the stimulus is low, this suggests that the sway response will be even less significant (Assländer et al, 2015), though the visual contributions may be suppressed when the cue is in motion (Bronstein, 2019). Although these works display strong validity for the use of visual positional and velocity cues in balance control, there is no clear consensus on the exact cognitive mechanism that is able to interpret or integrate these mechanisms to an individual's benefit when performing a balance task.

1.5 Sensory re-weighting & integration during balance tasks

Multisensory integration is an essential, dynamic process that occurs in the processes of maintaining an upright posture. Certain studies have postulated that visual information is constantly being interpreted and integrated with inputs from the vestibular and somatosensory systems, to compensate for deviations from an upright posture (Bronstein, 2019; Horak, 2006; Peterka, 2018). As there are multiple afferent inputs being integrated throughout each task, the process of assessing the relative contributions of each sensory modality originally proved to be quite difficult and not universally reliable (Dietz et al, 1993). Earlier models of integration recognize the interactions between the three sensory systems, but generally tended to attribute relative re-weighting patterns toward inter-subject variability (Guerraz et al, 2001; Witkin 1959). In a healthy population, the weight that individuals placed on visual cues seemed to vary considerably between participants (Witkin, 1959). This idea seemed to hold more weight in certain experimental designs, where it seemed certain individuals could disregard visual motion or inertial cues and were said to be entirely visually independent in their control of posture, as could also be demonstrated by use of the Rod-and-Frame test (Morris & Shapiro 1973, Slaboda et al, 2011; Souder, 1972). It was then postulated that the amount of visual dependence can vary based on the reliability of a particular sensory input. For example, visual input would be prioritized within one's bandwidth of sensory integration if vestibulo-proprioceptive information was reduced, or not as readily available (Cheung et al, 1990).

More recent work has attributed the processes of sensorimotor integration towards a closed-loop feedback control system to stabilize a given body, integrating all of the system inputs at any given point (Peterka, 2018). Sensory systems detect their relevant stimuli, detect the direction of sway, and relay afferent information to the CNS (Peterka, 2018; Schut et al, 2017) to integrate within the basal ganglia and cerebellum (Edwards et al, 2019), to ultimately elicit motor action that will compensate for sway (Peterka, 2018). Motor actions that are generated from the primary motor cortex can also contribute towards developing an efference copy to be compared with real-time inputs within the cerebellum, which assesses the correctness and desired outcome of the movement, in relation to the expected sensory feedback that will come from it (Edwards et al, 2019; von Holst and Mittelstaedt, 1950). Compensations occur through selective muscle activation relayed from the motor cortex to the periphery, in the form of "corrective, stabilizing torques" which occur about certain joints proportional to the necessary response to remain upright (Schut et al, 2017).

1.6 Optic flow gain changes & VR effects on balance

Virtual reality (VR) technology is a relatively novel technology that utilizes computer-generated virtual environments to simulate real-world experiences under controlled conditions. VR simulations are often presented to an individual through a head-mounted display (HMD) that uses online updating to present three-dimensional interactive visual scene relative to their head position and orientation (Assländer & Streuber, 2020). When utilizing VR, the technical specifications of the HMD should emphasize a sense of immersion (Robert et al, 2016; Slater & Wilbur, 1997) as this helps promote behavioural realism within the virtual environment, which is important in ensuring the individual can respond similarly in the VR environment as they would in a real-world scenario (Slater & Wilbur, 1997). To preserve reliability of observable behaviours within VR, (Gumma et al, 2021), factors that can improve the immersive nature of an environment include a wide field-of-view, high pixel resolution, magnitude of multi-sensory stimulation, and vivid experiences (Assländer & Streuber, 2020).

Assländer & Streuber (2020) concluded that photo-realistic VR environments can be reliable in replicating a real-world setting, and this is further validated by Soltani & Andrade (2021) which highlights that HMD VR environments can be used to assess and train functional balance with a high degree of ecological validity. An advantage in utilizing VR to produce visual stimuli for assessing postural control lies in its ability to modify mechanisms of sensory feedback within a virtual setting, in order to replicate stimuli that an individual may experience in a real-world setting. Thus, manipulating optic flow gain within such environments can prove to be a useful tool in assessing the capabilities of the postural control system. Optic flow gain represents a ratio of the flow of visual information within a setting relative to head or body movement (Bronstein, 2019; Lee 1980), such that if gain is increased, the amplitude of movement experienced in space will be greater than normal, and conversely less if gain is decreased.

Postural responses can vary based on experimental design and optic flow patterns within VR (van Asten et al, 1988). Gumaa et al (2021) used an interactive visual stimulus to demonstrate that virtual visual stimuli can be useful in improving control of posture in healthy adults, although other work proposes that 3-dimensional transient oscillating stimuli through VR can negatively impact postural control (Morel et al, 2015). Further, the referencing of optic flow to one’s head position seems to increase postural sway compared to a normal visual input (Black et al, 1983). This can be seen as an example of sway-referencing during a stimulus, and it indicates that the magnitude and direction of optic flow can further modulate control of posture during quiet standing (Horlings et al, 2009; van Asten 1988). During dynamic stance, in support of Jilk et al (2014), reducing the magnitude of flow may negatively influence postural control (Bronstein, 2019), which can also be reflected as an increase in segmental velocity and displacement (Keshner et al, 2007). However, there is conflicting evidence of postural behaviours when both the subject and visual scene begin to move concurrently within one task. Keshner et al (2004) used anteroposterior (AP) translations in combination with AP visual scene motion. This group discovered that sway was not directly impacted when only the subject is translated forward relative to the scene, but sway significantly increased when visual flow was introduced via scene movement in the opposite direction. This shows that the response to optic flow was potentiated by the addition of physical movement, as likely due to the amplification of visual inputs with the physical translation characteristics of the moving platform (Keshner et al, 2004). On another hand, Akiduki et al (2003) demonstrated that an optic flow gain of 2x, where the environment moved double that of head movement in the opposite direction, can reduce sway and therefore instead have a positive impact on postural stability. These results display the differences between subject and scene motion behaviours within VR, as it highlights the importance of the magnitude and direction, and certainly the relative perspective of a stimulus, within a virtual scene.

1.7 Dynamic Posturography

Posturography is the process of quantitatively analyzing the behaviours of the postural control system through controlled tasks utilizing various postural assessment equipment. Static posturography analyzes postural responses when an individual is maintaining a quiet stance, essentially placing them in a relatively unperturbed state upon some sort of fixed surface (Visser et al, 2008). While assessments of quiet stance can objectively inform researchers about postural compensations and certain sway behaviours, it has appeared to be more ecologically valid and reliable to assess postural control in response to some form of external perturbation, or what has come to be known as dynamic posturography (Allum et al, 2002; Furman 1994; Winter et al, 1990). Kinetic-based information can be collected during postural assessment through use of mountable force plates capable of measuring forces, moments, and reactive torques being applied to the body upon the SS (Carpenter et al, 2010; Visser et al, 2008). Kinematic-based information can also be collected through use of motion sensors and motion capture systems to estimate relative displacements and positioning of COM, as well as joint angular motion (Bloem et al, 2003; Carpenter et al, 2010). The combination of kinetic and kinematic information collected through such modalities can be compiled into analysis software in order to identify different movement parameters, including but not limited to frequency and amplitude of acceleration, velocity, or displacement data which can help characterize postural strategies and sway behaviours during a balance task (Visser et al 2008; Terry et al, 2011; Winter, 2009). Changes in postural responses can also be quantified by alternative response amplitudes, and other compensatory strategies such as agonist/antagonist muscle activation and bodily angular changes, that are necessary to maintaining postural control throughout a certain task (Bloem et al, 2003; Jilk et al, 2014; Visser et al, 2008).

Dynamic posturography entails the use of mechanical stimuli, such as platforms or SS capable of multi-planar movement, to elicit balance perturbations for which compensations will be needed to maintain stance (Allum et al, 2002; Furman, 1995; Visser et al, 2008). The direction in which the SS will move to perturb posture is variant among the literature. SS rotations (tilting) will involve movement in the pitch (AP) and/or roll (mediolateral; ML) planes (Akram et al, 2008; Beylergil et al, 2019; Carpenter et al, 2004; Slaboda et al, 2011), whereas SS translations can occur in the sagittal (AP) or transverse (ML), as well as vertical planes (Bax et al, 2020; Phanthanourak et al, 2016, Van Ooteghem et al, 2008). Certain models or testing paradigms may implement a combination of movements to assess postural control capabilities during multi-directional translations or rotations (Carpenter et al, 2004; Kaminishi et al, 2019).

While many investigators have elected to use rapid, transient perturbations to assess reactive postural control, slower and longer oscillatory SS translations may be more effective when assessing feed-forward postural control (Diener et al 1982, Dietz et al, 1993), which can hold further utility in assessing participants’ postural control in a dynamic stance situation (Schmid et al, 2011, Visser et al, 2008). Continuous sinusoidal perturbations can be characterized as successive transient perturbations that are linked through time (Kennedy et al, 2013), and permit investigators to perturb a SS at different frequencies and amplitudes throughout a single trial (Mills & Sveistrup, 2018). Further, sinusoidal perturbations have greater capability to induce sensorimotor adaptation when done repeatedly (Kennedy et al, 2013) over transient ones, and can more closely simulate typical perturbation situations that individuals may encounter throughout everyday life (Ghulyan et al, 2005). Longer sampling durations should be used during sinusoidal oscillations, at least 60 seconds long, as longer oscillation periods preserve the reliability of sway measures including the amplitude and frequency of COP, and larger amplitude changes in COM sway (Carpenter et al, 2001). Additionally, Schmid et al (2011) highlights that COM oscillations will become higher at lower oscillation frequencies, though COP variations seem to become more robust at higher frequencies. Thus, there is contesting debate upon which oscillation frequencies will help in eliciting the most informative sway response, and how many are productive to assess within one paradigm. In accordance with Diener et al (1982) and Dietz et al (1993), the shift towards anticipatory feedforward control is even more evident in longer sampling durations, where the shift in postural control strategy becomes advantageous as the stimulus becomes predictable over time (Buchanan & Horak 1999, Van Ooteghem et al, 2008). This aspect may further encourage the use of a pseudorandom frequency cycle instead, featuring a range of frequencies in one cycle. Within a range of 0.25-0.5 Hz, this design can be informative when assessing postural control with AP translations in combination with optic flow environments (Ghulyan et al, 2005; Kennedy et al, 2013; Keshner et al, 2004; Peterson et al, 2001) as potential to predict the stimulus can be minimized, and thus reliance on compensatory strategies will be greater due to the unexpected change in frequency at a given time (Jilk et al, 2014). Further, Mills & Sveistrup (2018) displayed that pseudorandom perturbation frequencies can have further utility when assessing postural control of a young, healthy population, as these individuals can quickly learn to predict necessary postural adjustments during fixed-frequency sinusoidal translations fairly easily. Lastly, more recent work by Sozzi et al (2020) supports that postural control may be further modulated by performing visual tasks concurrently with continuous platform translations within a moderate frequency range.

1.8 COP & COM behaviour within postural control

The interplay between COP and COM has been long debated, especially so within dynamic stance. The behaviours of each measure can be largely variable based on experimental design and population characteristics. The addition of visual gain within a dynamic perturbation provides a comparison between behaviours that are foundational in postural control as opposed to those that are displayed simply due to an outright dominance of vision as a sensory modality. It is important to recognize that certain postural control strategies will be constant and arise due to the nature of sensorimotor control, and rather other behaviours only seem evident when one piece of the puzzle is assessed (i.e. only COP) instead of all components.

Based on the inverted pendulum model, Carpenter (1999) hypothesized that increased MPF and decreased SD of COP signified an ankle stiffening strategy in response to postural threats done at height. Since this study purely relied on kinetic measures without accompanying COM measures to validate, this was widely accepted as a valid model of postural control during quiet stance. Despite this, there was earlier work done by Winter et al (1997) that highlighted differences between behaviours of COP and COM, and that these can be highly correlated with horizontal accelerations of the COM, indicating that the behaviour of the COM may be the primary control variable. Later, Winter et al (1998) postulated that the COP lags behind the COM in the same direction, supporting that the lag between COP and COM was too short to indicate a monosynaptic reaction between the movement and the perceived reaction. This suggested a passive control mechanism, such that the COP passively tracked the COM as there was not enough time for negative feedback to occur and have a meaningful adjustment. From these conclusions, the stiffness model was created by Winter, and newer studies that utilize electromyography to assess muscle activation in conjunction with kinetics and kinematics suggest there may be a feedforward approach to show anticipation for the COM and COP displacement through increased co-contraction at the ankle joint (Adkin et al, 2002; Kennedy et al 2013; Schmid et al 2011).

1.9 Reactive vs. anticipatory postural control

Coordinated muscle actions are an essential component to maintaining a given stance within a postural task. Reactive postural control involves the use of compensatory strategies in response to an unexpected postural perturbation to maintain a given stance (Maki & McIlroy, 1997). These strategies are integrated with visual inputs to coordinate an appropriate response to the perturbation (Dietz et al, 1993; Jilk et al, 2014; Zettel et al, 2005). Compensatory action often consists of a fixed response, where balance is maintained with feet in place, or a change-in-support strategy, either of which involve immediate muscle contraction to facilitate rapid limb movement, usually in the form of reactive stepping to maintain balance (Maki & McIlroy, 1997; Ting et al, 2009; Zettel et al, 2008). Anticipatory postural control entails the use of predictive strategies that occur in anticipation of a perturbation, in attempt to minimize the postural disturbance and avoid stepping or falling (Adkin et al, 2002; Bax et al, 2020; Maki & McIlroy, 1997). Anticipatory postural adjustments (APAs) are predictive adjusostments that act as a mechanism of feedforward balance control, as individuals are able to pre-emptively initiate muscle actions that minimize body movement in anticipation of an upcoming perturbation (Adkin et al, 2002; Zettel et al, 2002). Co-contraction of lower leg musculature to facilitate an APA is a common and effective strategy to maintain a fixed-position stance, by initiating body sway in an AP or ML direction, depending on the nature of the perturbation (Allum & Pfaltz, 1985; Buchanan & Horak 1999; Keshner et al 1987). While an APA may not always be followed by co-contraction in the lower limb, adjustments can be made in individual muscles as well to facilitate the necessary muscle action required for compensation.

Within a dynamic perturbation, Adkin et al (2002) postulates that sway be initiated in the same direction to assist the initiation of movement, or conversely a stiffening strategy may be used in attempt to counteract potential destabilizing forces and stabilize the body to maintain the position of the COM. During a continuous perturbation, individuals will attempt to best modulate postural control through reactive adjustments initially, though it seems to be more advantageous to shift over to anticipatory strategies throughout the perturbation. APAs have greater potential to attenuate changes in whole-body COM and postural sway compared to reactive postural adjustments (Adkin et al, 2002; Dietz et al, 1993; Kennedy et al 2013; Schmid et al 2011), thus indicating the effectiveness of APAs as a strategy for postural control within a dynamic stance paradigm.

1.10 Current knowledge gap

The visual system's role in balance control is integral to the maintenance of balance in dynamic environments. The role of the visual system in postural control during quiet stance has been previously explored, but its roles within postural control in dynamic environments warrants further investigation. In reality, our activities of daily living often require stimulus inputs and feedback travelling through multiple systems, and the integration of said sensorimotor modalities is the basis for postural maintenance. There is an evident lack of clear consensus on how novel methods such as VR-based optic flow gain changes, in combination with continuous mechanical stimuli, can impact balance in dynamic stance environments. Further, there is limited work examining joint and segment angular changes that are present throughout these paradigms, and how these may interact with changes in center of mass & center of pressure changes to maintain an upright stance during an oscillating perturbation routine.

1.11 Purpose and hypothesis

The purpose of this thesis is to aid in developing our understanding of how different optic flow gain factors and SS translations contribute to postural control within a dynamic stance environment amongst young, healthy adults. This thesis aims to address the following research questions: 1) How may balance behaviour have become impacted by applying various optic flow gain factors in VR, simultaneously with a continuous, pseudorandom platform perturbation sequence?, and 2) How might an individual’s postural responses to mechanical perturbations be influenced by visual information produced through a VR environment versus a real-view environment (no VR)?

It was hypothesized that the amplitude of optic flow gain exposure would be inversely related to postural sway. Exposure to increased visual gain throughout the experiment would result in a tighter regulation of postural control, which would be reflected as decreased sway. This simply meant that participants’ postural control would be influenced as the magnitude of optic flow gain changed through VR. More specifically, as gain values increased above 1, it was expected that measures of postural sway would likely decrease to compensate for the increased flow of visual information experienced compared to the baseline trial. This was to be characterized by a decrease in COM and COP amplitude, and an increase in COP frequency and MVel. Conversely, as gain values decreased below 1, participants would experience increased sway, where COM and COP amplitude would increase, and COM frequency and MVel would decrease.

CHAPTER 2: METHODOLOGY

2.1 Participant characteristics

Twenty-eight healthy, young adults (14 male and 14 female) between the ages of 18-40 (mean age (+/-SD): 25.04 (+/- 4.66) years) were recruited as participants for this study from York University and the surrounding area. Participants were required to have normal or corrected-to-normal vision, be free of any musculoskeletal or neurological deficits, and be free of taking any prescription medications that may have influenced their ability to maintain their balance. Participants were excluded from this study if they failed to meet one or more of these criteria. In addition, this study did not consider participants' biological or socio-cultural identifications of gender.

2.2 Experimental procedure

Participants were instructed to stand barefoot upon a force plate (AMTI, Advanced Mechanical Technology Inc., USA), capable of recording kinetic information in the form of ground reaction forces and moments, with their arms resting naturally by their side, head facing forward. While standing upon the force plate, participants were equipped with a safety harness, secured through the upper body and pelvis, and attached to the ceiling above which ensured their safety in the case of a fall. A designated spotter stood by the participant during the experiment. The force plate was mounted securely upon a translating SS (Single Rail Stage, H2W Technologies Inc, USA) (Figure 1), so that appropriate measures could be collected while the participant experienced the dynamic balance tasks. The SS utilized was capable of mechanical translations in a single plane of movement, through the sagittal plane, or the anterior-posterior direction relative to the participant. Stance width was determined through measurement and standardization of the foot length of each participant, and this was marked on the force plate to ensure consistent stance width and foot orientation across all trials. Participants would subsequently undergo a series of 6 experimental trials with varying optic flow gain values in a randomized sequence, each of 2 minutes in duration. Both before and after performing their respective set of trials, each participant would also complete an additional trial without VR.

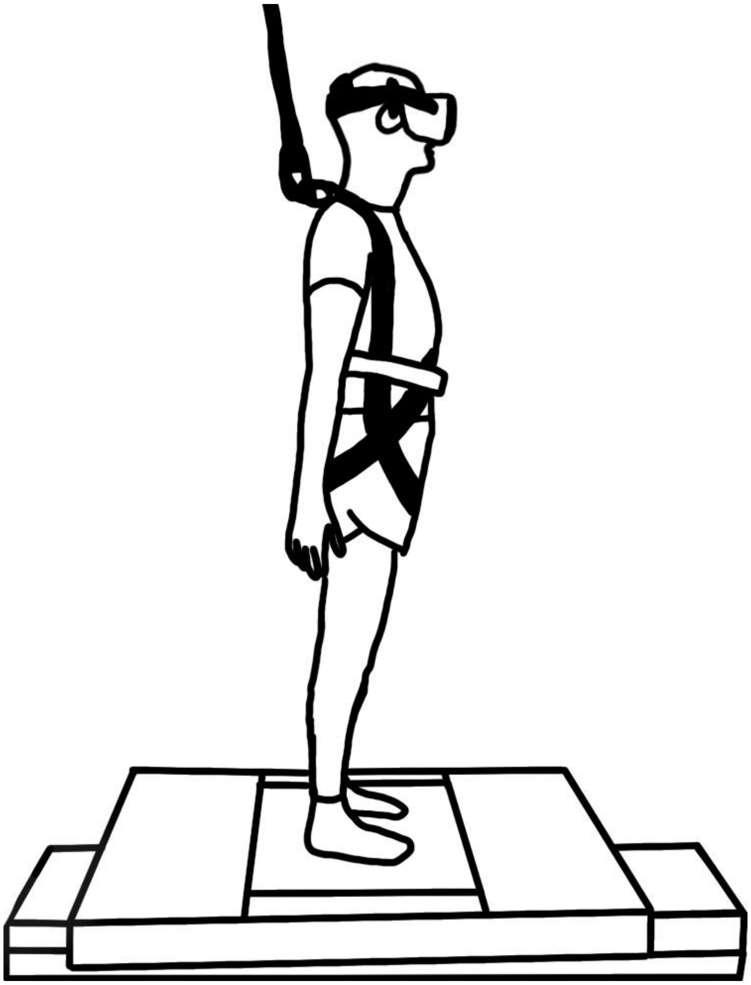


Figure 1

Experimental equipment setup

Equipment setup diagram with the head-mounted display worn by the participant. Force plate was mounted upon the translating platform, with a harness secured above that the participant could be secured into throughout the duration of testing.

During each of the 6 experimental trials, participants were instructed to wear a VR HMD (VIVE Pro 2, HTC Corporation). This headset exposed participants to a full field of view of around 110° horizontally in a virtual setting, which was intended to replicate a real-world environment (Figure 3).

Participants were instructed to focus on a red target ahead of them within the virtual environment (Figure 3), and asked to best maintain their current stance throughout the duration of each trial. To begin, a VR baseline condition was completed where participants experienced the dynamic perturbation cycle with normal optic flow through the VR headset. Next, Vizard Python programming (WorldViz) was used in conjunction with the VR headset to apply the optic flow gain of different values to the participant. Trials with gain values of 0.25, 1, 2, 4, or 16 counterbalanced were completed which would either amplify or reduce participants' optic flow, relative to their head motion, within VR. Participants also had the opportunity to take a 60-second rest between any of the trials, where they were able to sit or stand freely, prior to continuing with each subsequent trial.

The pseudorandom perturbation cycle consisted of a series of anterior-posterior translations of various amplitudes within an oscillation frequency range of 0-0.5 Hz. Peak acceleration and deceleration were programmed to 3 m/s^2, and translation velocity was programmed not to exceed 3 m/s during the cycle, and displacement range was set at +/- 10 cm from the starting position (Appendix A). Participants were also equipped with infrared emitting markers which were used to collect 3D motion capture data for kinematic analysis (Optotrak, NDI). There were 2 markers attached to the SS to monitor its movement throughout trials, and the participant was equipped with a total of 10 markers on their right side at different bodily landmarks to permit for 2-dimensional tracking of segment endpoints. The locations of the landmarks included the right mastoid process (MP), zygomatic arch (ZY), acromion (AC), lateral epicondyle of the humerus (LE), greater trochanter of the femur (GT), lateral condyle of the femur (LFC), fibular head (FH), lateral malleolus of the ankle (LM), base of the 5th metatarsal (FthMT), and calcaneus (CLC) (Figure 2).

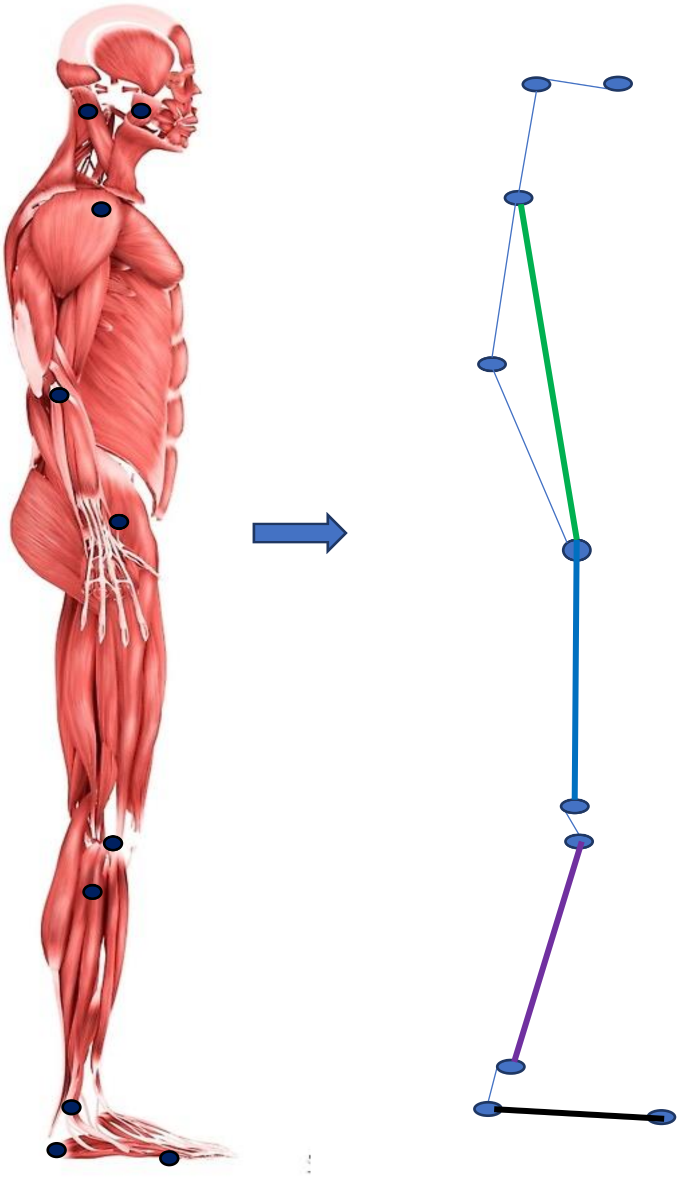


Figure 2

Kinematic model & link segment model

Left: Kinematic model diagram displaying how the model was developed using 10 kinematic markers (black dots) placed on bony landmarks (MP, ZY, AC, LE, GT, LFC, FH, LM, FthMT, CLC). Right: Link segment model derived from kinematic model. Each of the 10 bony landmarks (blue dots) helped derive the four segments being used in the model. Segments are outlined using coloured, bolded lines: Trunk (green), Thigh (blue), Shank (purple), Foot (black).

2.3 Data collection and processing

Following data collection, the data was processed through MATLAB software (MATLAB R2022b, MathWorks Inc.). In this study, outcome measures were analyzed in the AP direction only. For each trial, primary outcome measures to assess postural control included COM and COP. COM was calculated using the subsets of kinematic marker data of the modelled body segments in combination with anthropometry. Marker data was sampled at 100 Hz, and low pass filtered at 5 Hz with a dual pass Butterworth filter. Bias removal was completed by subtracting the position of the SS from the position of each marker at each respective frame through each trial. Anthropometric measures including the relevant segment COM proportions (Winter, 2009) were combined with the filtered marker data to calculate both segment and whole-body COM. If any markers went out of view of the motion capture camera for durations of <0.5 seconds (<50 frames of data), cubic spline interpolation was used.

AP COP was calculated using ground reactions forces and moments sampled from the force plate. Sampling rate was set at 100 Hz, and low pass filtered with a dual pass Butterworth filter set at 5 Hz. Bias removal for COP data was be done through subtraction of mean position from the signal for each trial.

MPF, RMS, and MVel were calculated from COP and COM data to help quantify postural behaviour throughout each trial. MPF was calculated using the following formula:

$$MPF = \frac{\sum f \times P(f)}{\sum P(f)} \tag{1}$$

where f represented signal frequency, and P represented power amplitude at each respective frequency, and RMS was calculated using:

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

where x represented each data sample, and n represented the number of data points from the respective sample. Lastly, MVel was calculated using:

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

(3)

where d represented displacement data points for COP and COM in each signal, and t represented time reflected as frame rate during the trial.

Relative and absolute joint angles were calculated from kinematic data based on joint dot products and vectors of the segments both proximal and distal to the joint throughout each trial. The following formula was utilized:

$$\theta = \frac{Jd}{Lp * Ld} \tag{4}$$

where Jd represented the respective joint dot product, Lp represented the length of the segment proximal to the joint, and Ld represented the length of the segment distal to the joint, and absolute segment angles were calculated using the inverse tan of the “y” and “z” components of each segment vector. The following formula was utilized:

$$\theta = tan^{-1}\,(\frac{Sy}{Sz}) \tag{5}$$

where Sy represented the “y” vector component of the relevant segment, and Sz represented the “z” vector component of the relevant segment.

Mean angles were calculated from joint and segment angular data, where the following formula was used:`

$$Mean = \frac{\sum x}{n} \tag{6}$$

where x represented each data sample, and n represented the number of data points in each respective sample, and standard deviation was calculated using:

$$SD = \sqrt{\frac{\sum(x - \bar{x})^2}{n - 1}} \tag{7}$$

Where x represented each data point, $\bar{x}$ represented the sum of data points, and n represented the number of data points from the respective sample.

2.4 Data analysis

Statistical analysis for the processed data was completed using SPSS software (IBM Corp.). Significance was set at $\alpha = 0.05$ for all analyses. Analysis began with 1 x 5 repeated measures analysis of variance (RMANOVA) to assess the main effects of various optic flow gain factors (0.25, 1, 2, 4, 16) on postural outcome measures such COP and COM, reflected quantitatively through RMS, MPF, and MVel. To assess further interactions of gain alongside the impact of time on posture, 2 (minute: M1 vs. M2) x 5 (gain: 0.25, 1, 2, 4, 16) RMANOVAs were completed to assess the effects of gain in the first (M1) and second minute (M2) of each trial. The same process was performed for all joint (hip, knee, ankle) and segment (trunk, thigh, shank, foot) body angles, as well as examining joint angle coherence between gain values. Shapiro-Wilk tests of normality and histograms were used to assess normality across all variables, as well as the presence of outliers. Outliers were identified as any values that were 3 or more standard deviations (SD) from the mean. Transforming the data and correcting for outliers where necessary did not correct for normality; therefore, due to the robust nature of RMANOVAs, no corrections for normality were used for analysis. Assumptions of sphericity were evaluated using Mauchly's test of sphericity, and in the case of violations, Greenhouse-Geisser corrections were applied to correct the variance. If $p < 0.05$, Bonferroni corrections were performed to assess multiple comparisons. Significant interaction effects of gain and minute were explored using independent minute condition 1 x 5 RMANOVAs, while significant effects between minutes were explored post-hoc using non-parametric t-tests. A summary of statistical significance for each outcome measure is reported below (Appendix B).

CHAPTER 3: RESULTS

3.1 Center of pressure

3.1.1 Amplitude of COP displacement

There was a significant effect of gain on COP RMS, as well as an interaction effect present. There was a significant interaction effect present between minute and gain ($F_{2.959, 79.888} = 3.129$, $p = 0.031$), and no main effect of minute was found ($F_{1,27} = 0.908$, $p = 0.349$). Subsequent analysis of each minute revealed a significant effect of gain in both M1 ($F_{2.661, 71.853} = 3.145$, $p = 0.036$) and M2 time frames ($F_{3.196, 86.298} = 5.957$, $p = < 0.001$). Post-hoc results for pairwise comparisons showed that in M2, COP RMS at a gain of 0.25 was significantly greater than all other gains (Figure 3). In M1, COP RMS at a gain of 0.25 was significantly greater than gains of 1 and 2 (Figure 3). Additionally, there was a statistically significant difference within a gain of 16 between M1 and M2 ($p = 0.023$), where COP RMS was significantly greater in M1 (Figure 3). No other significant differences of gain between minutes were found.

3.1.2 Amplitude of COP velocity

For COP MVel, there was a significant interaction effect observed between minute and gain ($F_{2.277, 61.476} = 15.686$, $p = < 0.001$), as well as a significant main effect of minute ($F_{1, 27} = 78.079$, $p = < 0.001$). Subsequent analysis of gain revealed that there was a significant effect of gain in both M1 ($F_{2.272, 61.353} = 36.085$, $p = < 0.001$) and M2 time frames ($F_{2.129, 57.488} = 23.653$, $p = < 0.001$). Post-hoc results showed that on average, MVel was lowest at a gain of 1 and greatest at a gain of 16. In M1, gains of 4 and 16 were both significantly greater than 0.25, 1, and 2 (Figure 4). In M2, similar trends were revealed and to add, 0.25 was significantly greater than 1 and 2 (Figure 4). Further, there was a statistically significant difference across all gain values between M1 and M2 ($p = < 0.001$), with MVel being significantly greater in M1 (Figure 4).

3.1.3 Frequency of COP displacement

For COP MPF, there was a significant interaction effect observed between minute and gain ($F_{2.987, 80.650}$ = 5.650, p = < 0.001), as well as a significant main effect of minute ($F_{1, 27}$ = 12.364, p = < 0.001). Subsequent analysis of each minute revealed a significant main effect of gain in both M1 ($F_{2.857, 77.131}$ = 40.162, p = < 0.001) and M2 ($F_{2.641, 71.299}$ = 25.833, p = < 0.001) time frames. Post-hoc results showed that on average, COP MPF was lowest at a gain of 1 and highest at a gain of 16. In M1, gains of 4 and 16 were significantly greater than 0.25, 1, and 2 (Figure 5). In M2, similar trends were displayed, where 4 was significantly greater than 1 and 2. In both minutes, 16 was significantly greater than all other gains (Figure 5). Additionally, there was a statistically significant difference across gains of 2 (p = 0.003), 4 (p = 0.000), and 16 (p = 0.011) between the time frames, with COP MPF being significantly greater in M1 (Figure 5). There were no differences in gains of 0.25 (p = 0.172) or 1 (p = 0.227) between minutes.

3.2 Center of mass

3.2.1 Amplitude of COM displacement

For COM RMS, there was a significant interaction effect observed between minute and gain ($F_{4,108}$ = 6.162, p = < 0.001), as well as a significant main effect of minute ($F_{1,27}$ = 4.612, p = 0.041). Subsequent analysis revealed a significant main effect of gain for both M1 ($F_{2.586, 69.820}$ = 5.114, p = 0.005) and M2 ($F_{2.770, 74.782}$ = 4.105, p = 0.011). Post-hoc results for pairwise comparisons revealed that in M1, COM RMS was highest at a gain of 0.25 while lowest at a gain of 4, and gains of 0.25 and 1 were both significantly greater than 4 (Figure 6). Gain of 0.25 was also significantly greater than 16 in M1. In M2, COM RMS was highest at 1 while lowest at 16, and there was a significant difference between gains of 1 and 4, as well as 1 and 16 (Figure 6). No other differences of gain within M2 were found. Additionally, there was a significant difference for gains of both 0.25 (p = 0.005) and 16 (p = 0.003) between the two time frames, where RMS was significantly greater in M1 for both gain values (Figure 6). There were no differences in gains of 1 (p = 0.750), 2 (p = 0.227) or 4 (p = 0.495) between minutes.

3.2.2 Amplitude of COM velocity

For COM MVel, there was a significant interaction effect observed between minute and gain ($F_{2.868, 77.432} = 2.921$, p = 0.041), and no main effect of minute was found ($F_{1,27} = 1.424$, p = 0.243). Subsequent analysis of each minute revealed a significant main effect of gain for both M1 ($F_{2.476, 66.843} = 8.268$, p = < 0.001) and M2 ($F_{2.585, 72.391} = 7.662$, p = < 0.001). Post-hoc results revealed that in M1, MVel was highest at 1 and lowest at 4, and gains 0.25 and 1 were both significantly different than 2, 4, and 16 (Figure 7). In M2, similar results were found, where MVel was also greatest at 1 and lowest at 4, and 0.25 and 1 were both significantly greater than 2, 4, and 16 (Figure 7). Additionally, there was a significant difference for a gain of 16 (p = 0.048) between the two time frames, where MVel was significantly greater in M1 than M2 (Figure 7). There were no other differences in gains of 0.25 (p = 0.466), 1 (p = 0.810), 2 (p = 0.631), or 4 (0.296) between minutes.

3.2.3 Frequency of COM displacement

For COM MPF, there was no interaction effect present between minute and gain ($F_{4,108} = 1.145$, p = 0.340), and there was no significant main effect of minute present ($F_{1,27} = 3.437$, p = 0.075). However, there was a significant main effect of gain present ($F_{2.920,78.844} = 3.224$, p = 0.028). Post-hoc testing revealed that COM MPF for a gain of 1 was significantly greater than 4 (p = 0.008), with no other significant pairwise comparisons found (Figure 8).

3.3 Joint and segment angular changes

3.3.1 Joint angle mean and standard deviation

For both mean and SD of hip angle, there was no significant effect of minute ($F_{1,27}$ = 0.91, p = 0.766; $F_{1,27}$ = 0.976, p = 0.332) or gain ($F_{4,108}$ = 0.899, p = 0.437; $F_{4,108}$ = 2.502, p = 0.072) observed, nor was there an interaction effect present ($F_{4,108}$ = 0.569, p = 0.686; $F_{4,108}$ = 1.252, p = 0.290) (Figures 9a & 9b). For mean knee angle, there was a significant main effect of minute ($F_{1,27}$ = 13.686, p = < 0.001) and gain ($F_{3.014,81.368}$ = 2.858, p = 0.042), but no interaction effect was present ($F_{4,108}$ = 0.827, p = 0.511). Post-hoc results displayed that M2 was greater than M1 across gains on average, and there was a significant difference between 4 and 16 where 4 was greater than 16 (p = 0.039) (Figure 10a). Additionally, there was a significant main effect of minute on knee angle SD ($F_{1,27}$ = 15.982, p = < 0.001), where SD was greater on average in M1 than M2 (Figure 10b). There was no main effect of gain ($F_{4,108}$ = 1.499, p = 0.207) or interaction effect ($F_{4,108}$ = 0.805, p = 0.525) present for knee angle SD. For mean ankle angle, there was a significant main effect of minute observed ($F_{1,27}$ = 4.796, p = 0.037), where the angle in M2 was greater than M1 on average (Figure 11a). There was no main effect of gain ($F_{4,108}$ = 2.150, p = 0.129) or interaction effect ($F_{4,108}$ = 1.018, p = 0.401) present for mean ankle angle. Lastly, for ankle angle SD, there was a significant main effect of minute ($F_{1,27}$ = 16.243, p = 0.000), where SD was greater on average in M1 than M2 (Figure 11b). There was no main effect of gain ($F_{4,108}$ = 1.165, p = 0.330) or interaction effect ($F_{4,108}$ = 0.219, p = 0.928) present for ankle angle SD.

3.3.2 Joint angle coherence

There was a significant main effect of gain observed on hip-knee (HK) coherence ($F_{1.952,52.704}$ = 3.080, p = 0.038). Post-hoc testing revealed a significant difference between gains of 4 and 16 (p = 0.011), where 4 was greater than 16 (Figure 12). There was also a significant main effect of gain observed on hip-ankle (HA) coherence ($F_{1.771,47.824}$ = 2.541, p = 0.045). Post-hoc results displayed a significant difference between gains of 4 and 16 (p = 0.049), where 4 was greater than 16 (Figure 12). Additionally, there was a significant main effect of gain observed for knee-ankle (KA) coherence ($F_{2.813,75.948}$ = 2.845,

$p = 0.041$). Similar to HK and HA, post-hoc testing displayed a significant difference between gains of 4 and 16 ($p = 0.026$), where 4 was greater than 16 (Figure 12). No other significant differences between gains were found within each variable.

There was a significant difference between HK and HA coherence ($F_{1,28} = 28.738$, $p = < 0.001$), where HK coherence was significantly greater than HA across all gain values. Post-hoc results revealed that there was a difference across each of the gain values, which all reached statistical significance ($p < 0.001$) (Figure 12). There was also a significant difference between HA and KA coherence ($F_{1,28} = 34.251$, $p = < 0.001$), where KA coherence was significantly greater than HA. Post-hoc results revealed that there was a significant difference ($p < 0.001$) across all gain values between the two variables as well (Figure 12). No significant differences were observed between HK and KA coherence ($F_{1,28} = 0.552$, $p = 0.464$) (Figure 12).

3.3.3 Segment angle mean and standard deviation

For mean trunk angle, there was a significant main effect of minute ($F_{1,27} = 7.036$, $p = 0.013$) where M1 was greater on average than M2 (Figure 13a), although no main effect of gain ($F_{4,108} = 0.477$, $p = 0.753$) or interaction effect was present ($F_{4,108} = 2.091$, $p = 0.087$). For trunk angle SD, there was a significant main effect of minute ($F_{1,27} = 4.284$, $p = 0.048$), a significant main effect of gain in both M1 ($F_{2.996,80898} = 4.442$, $p = 0.045$) and M2 ($F_{3.061,82.650} = 6.888$, $p = < 0.001$), as well as a significant interaction effect present ($F_{3.130,84.500} = 5.074$, $p = < 0.001$). Post-hoc results displayed that SD was significantly highest at a gain of 0.25 and lowest at a gain of 4 in M1. In M2, a gain of 1 was significantly greater than 2, 4, and 16, but not 0.25 (Figure 13b). Additionally, there was a significant difference with a gain of 16 between M1 and M2, where SD was significantly greater in M1 ($p < 0.05$) (Figure 12b). For mean thigh angle, there was a significant main effect of minute ($F_{1,27} = 13.349$, $p = < 0.001$) where M1 was greater on average than M2 (Figure 14a), but no main effect of gain ($F_{4,108} = 1.455$, $p = 0.221$) or interaction effect ($F_{4,108} = 0.280$, $p = 0.890$). For thigh angle SD, a similar trend was found where there was a main effect of minute ($F_{1,27} = 5.626$, $p = 0.025$) with M1 being greater on average than M2 (Figure

14b), with no effect of gain ($F_{4,108}$ = 1.722, p = 0.150) or interaction effect ($F_{4,108}$ = 1.363, p = 0.252). For mean shank angle, there was a significant main effect of minute ($F_{1,27}$ = 15.694, p = < 0.001) as well as gain ($F_{3.015,81.402}$ = 6.736, p = < 0.001), but no interaction effect was present ($F_{4,108}$ = 1.677, p = 0.161). Post-hoc results displayed that were significant differences between gains of 2 and 16, as well as 4 and 16 (Figure 15a). Additionally, there was a significant main effect of minute on shank angle SD ($F_{1,27}$ = 10.137, p = 0.004), where SD was greater on average in M1 than M2 (Figure 15b). No significant effect of gain ($F_{4,108}$ = 0.254, p = 0.329) or interaction effect ($F_{4,108}$ = 0.392, p = 0.814) was present for shank angle SD. Lastly, for both mean and SD of foot angle, there was no significant main effect of minute ($F_{1,27}$ = 1.025, p = 0.320; $F_{1,27}$ = 0.025, p = 0.876) or gain ($F_{4,108}$ = 1.607, p = 0.178; $F_{4,108}$ = 0.613, p = 0.654), and no interaction effects ($F_{4,108}$ = 0.664, p = 0.618; $F_{4,108}$ = 0.312, p = 0.869) (Figure 16a and 16b).

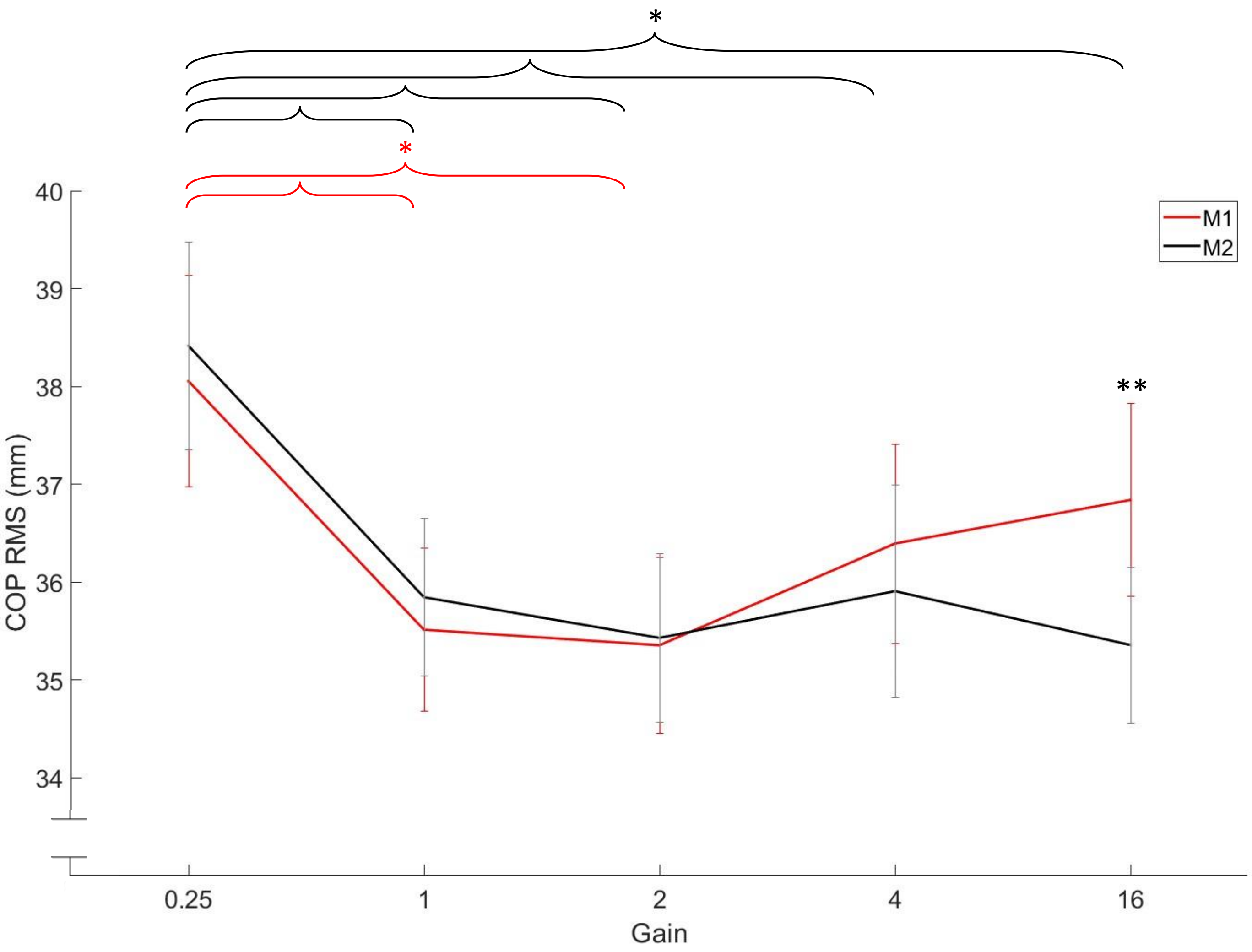


Figure 3. COP RMS across gain trials and between minutes.

Mean (SEM) RMS plotted for COP during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

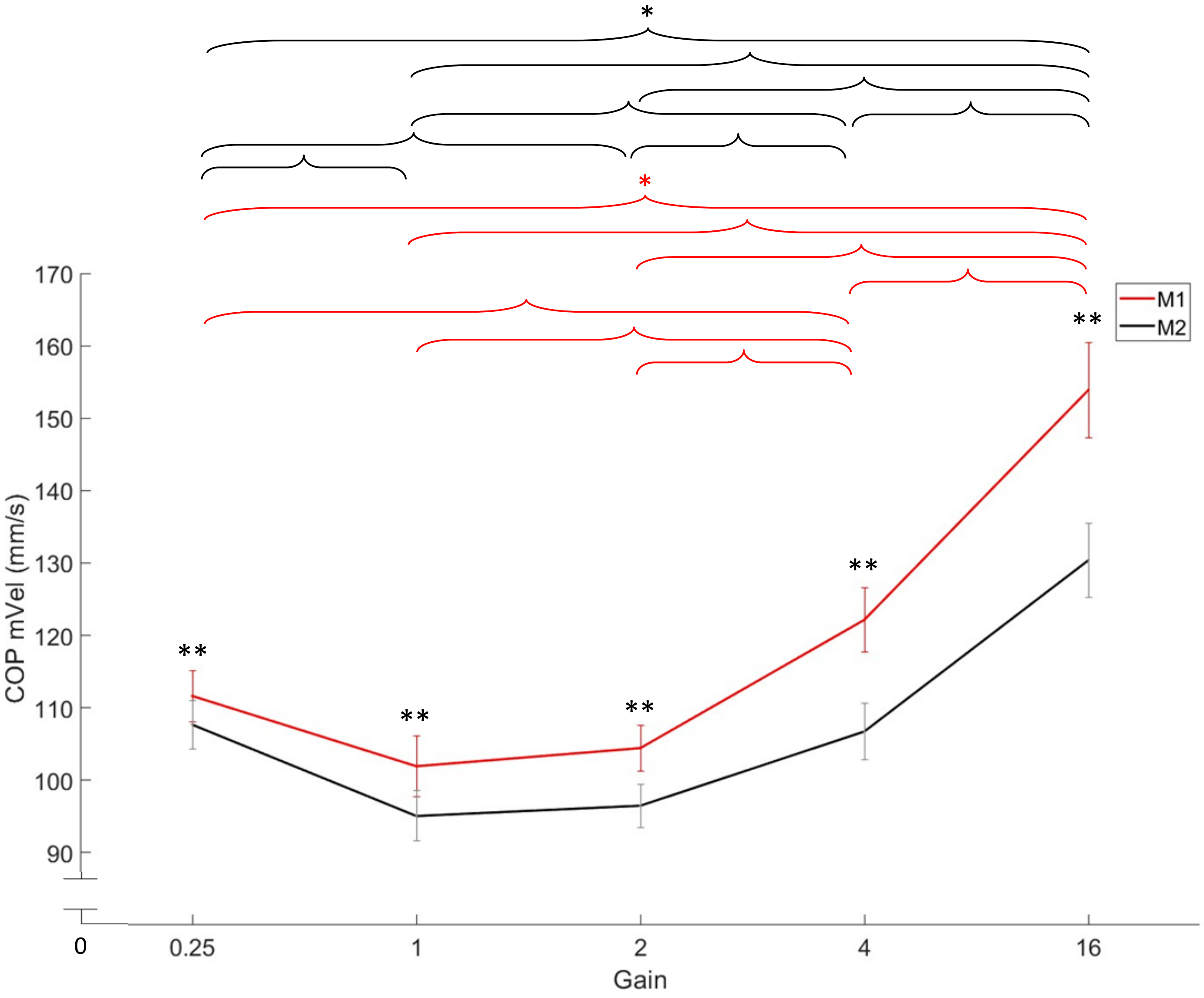


Figure 4. COP MVel across gain trials and between minutes.

Mean (SEM) MVel plotted for COP during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

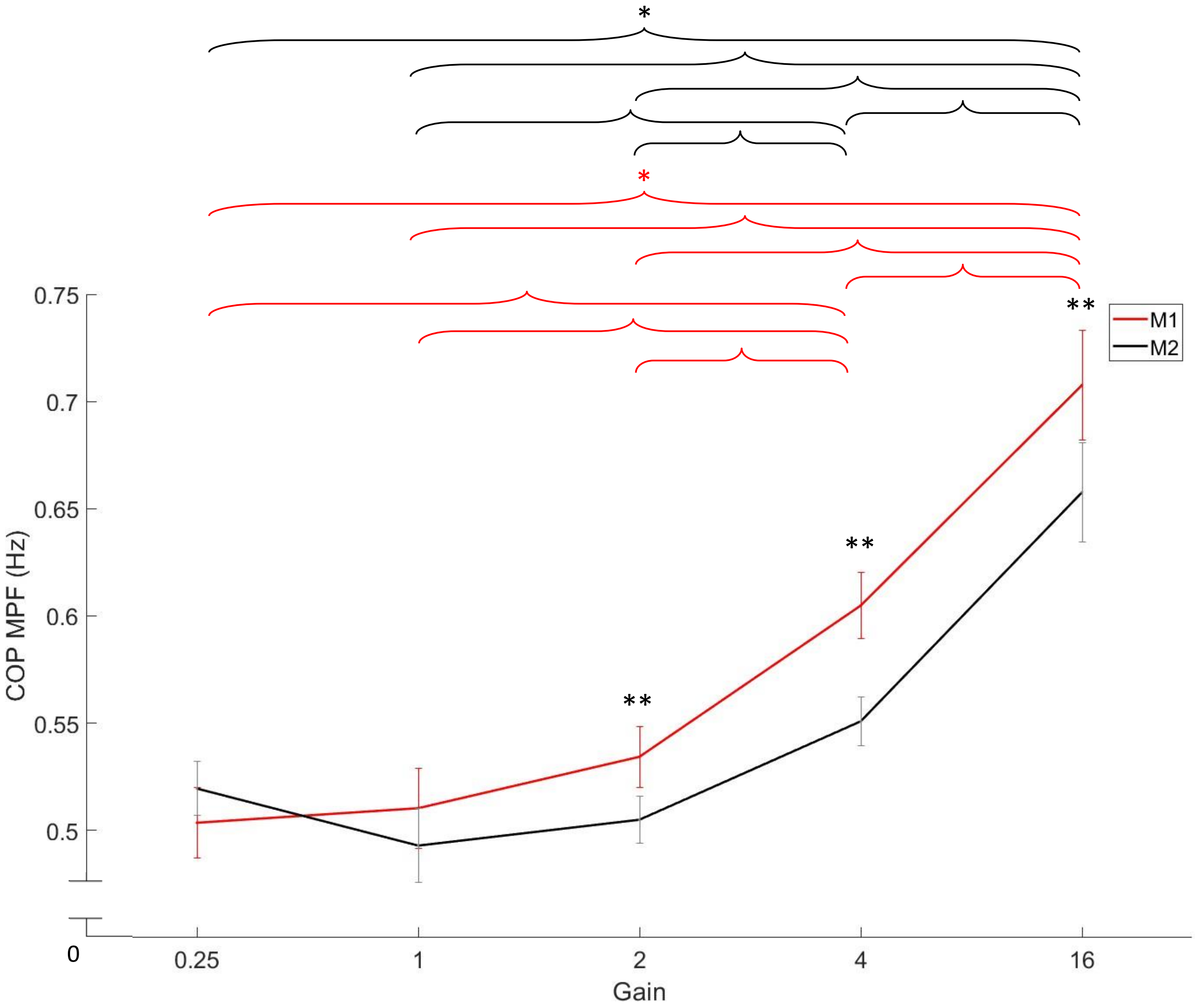


Figure 5. COP MPF across gain trials and between minutes.

Mean (SEM) MPF plotted for COP during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

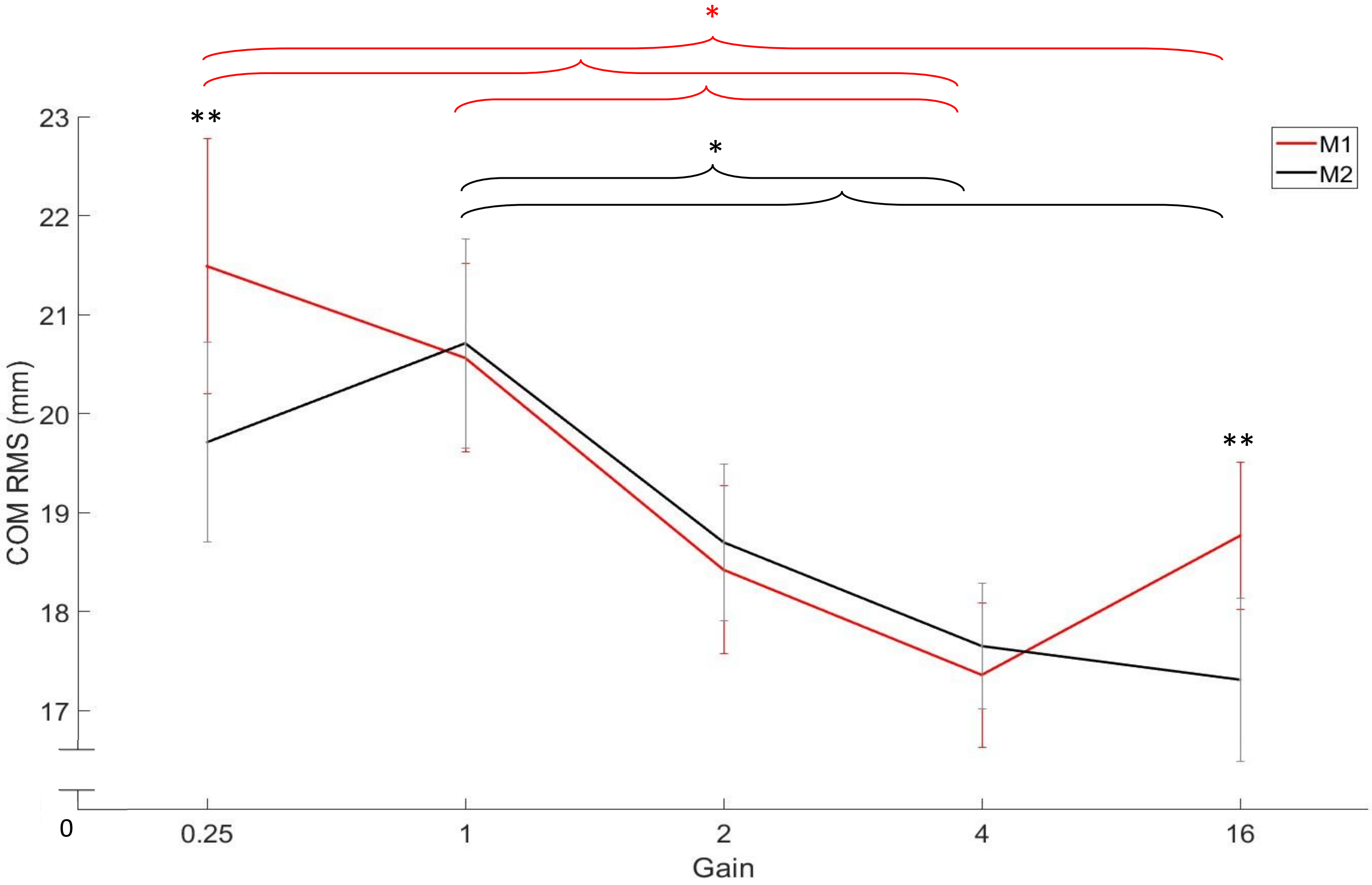


Figure 6. COM RMS across gain trials and between minutes.

Mean (SEM) RMS plotted for COM during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

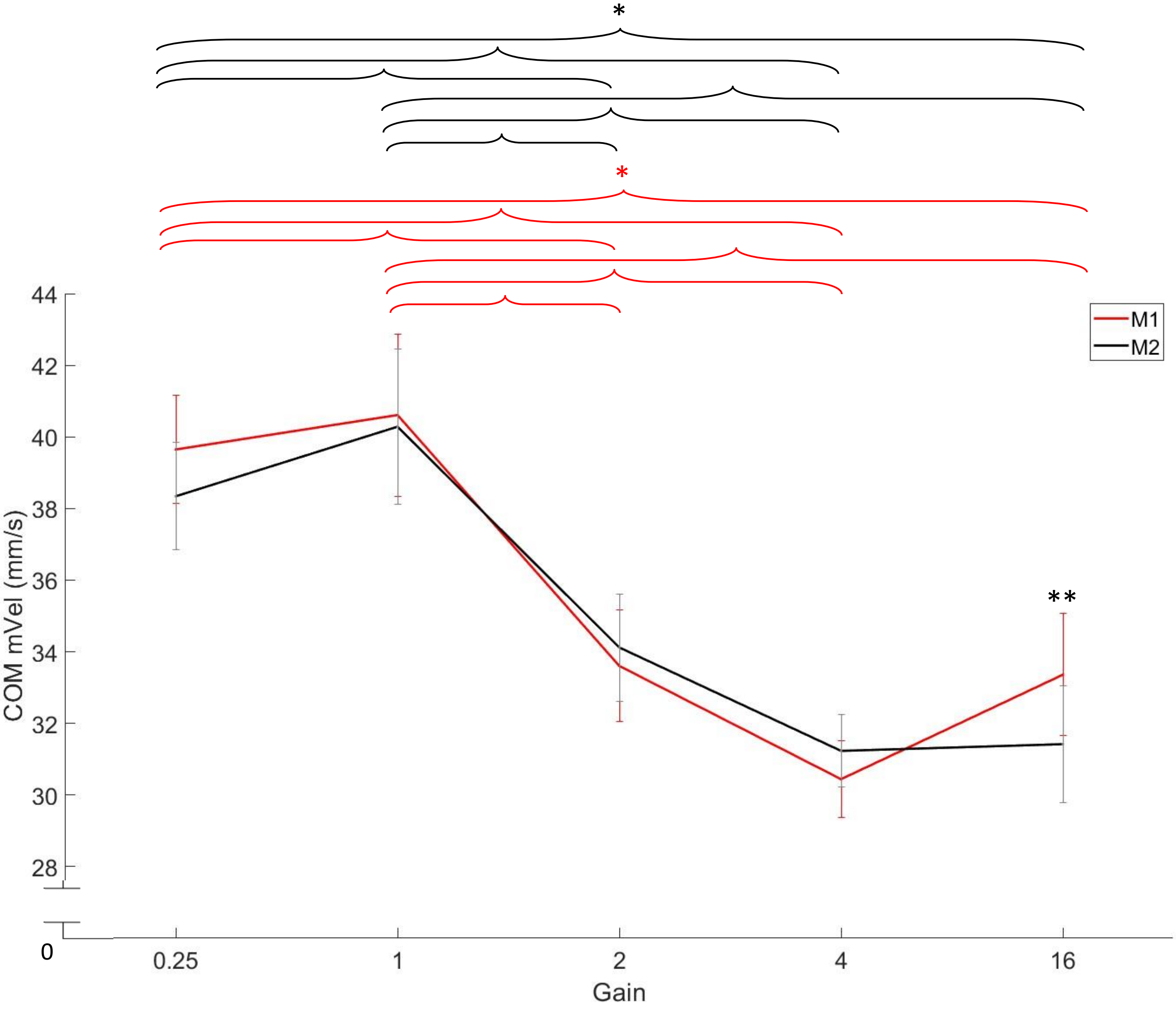


Figure 7. COM MVel across gain trials and between minutes.

Mean (SEM) MVel plotted for COM during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

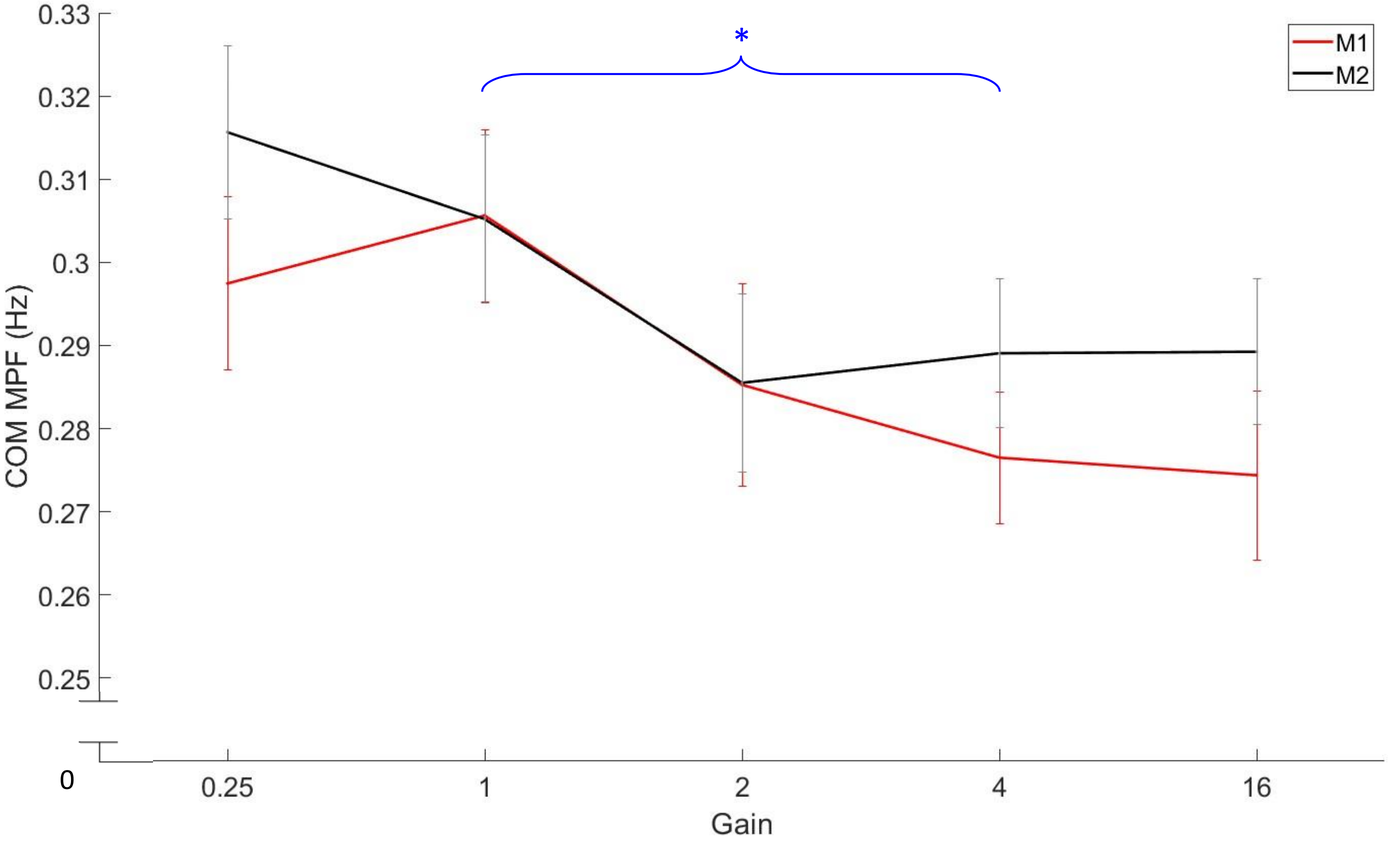


Figure 8. COM MPF across gain trials and between minutes.

Mean (SEM) RMS plotted for COP during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**). Blue lines of significance indicate significant pairwise comparisons for collapsed minute conditions.

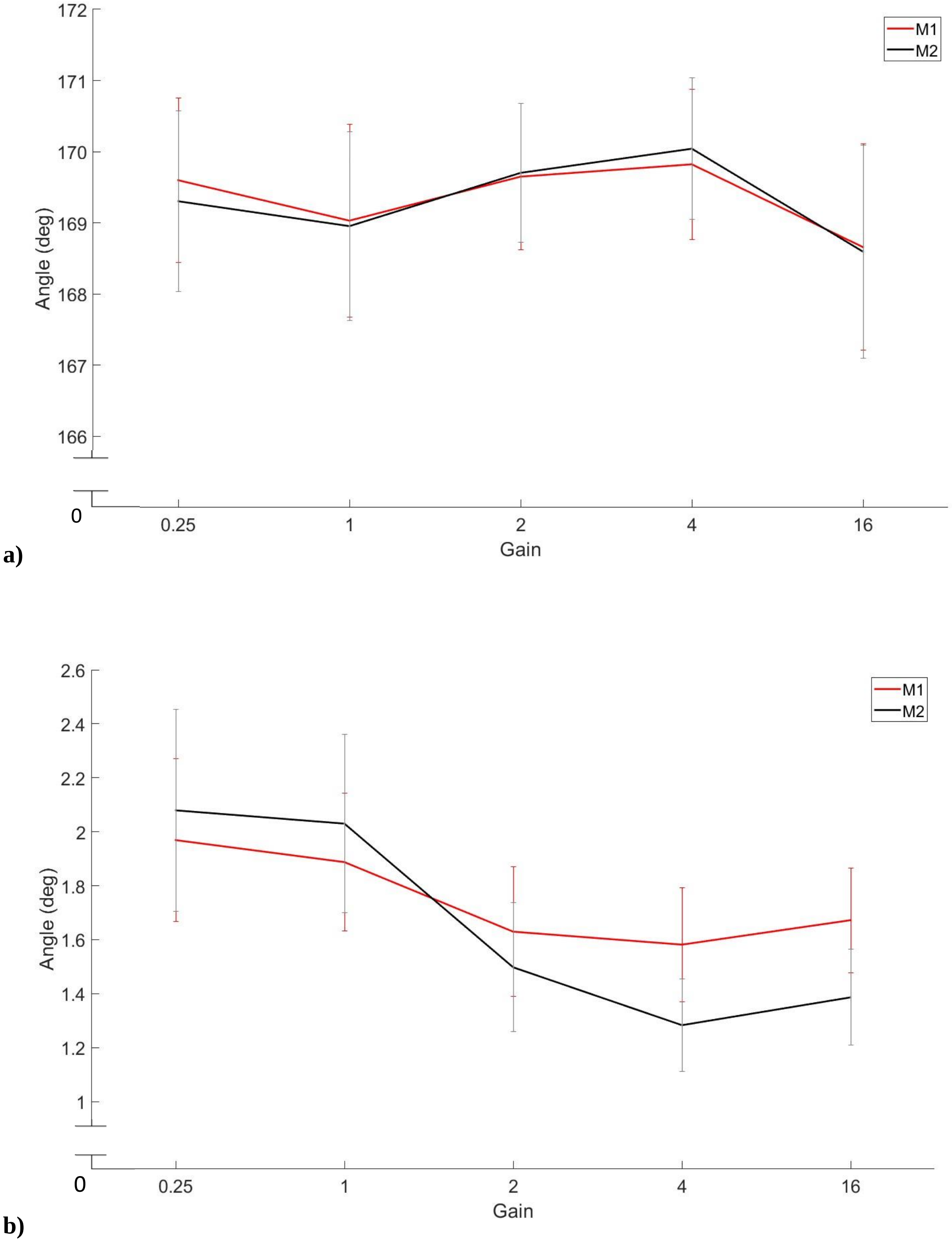


Figure 9. Hip joint across gain trials and between minutes.

a) Mean (SEM) hip joint angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*).

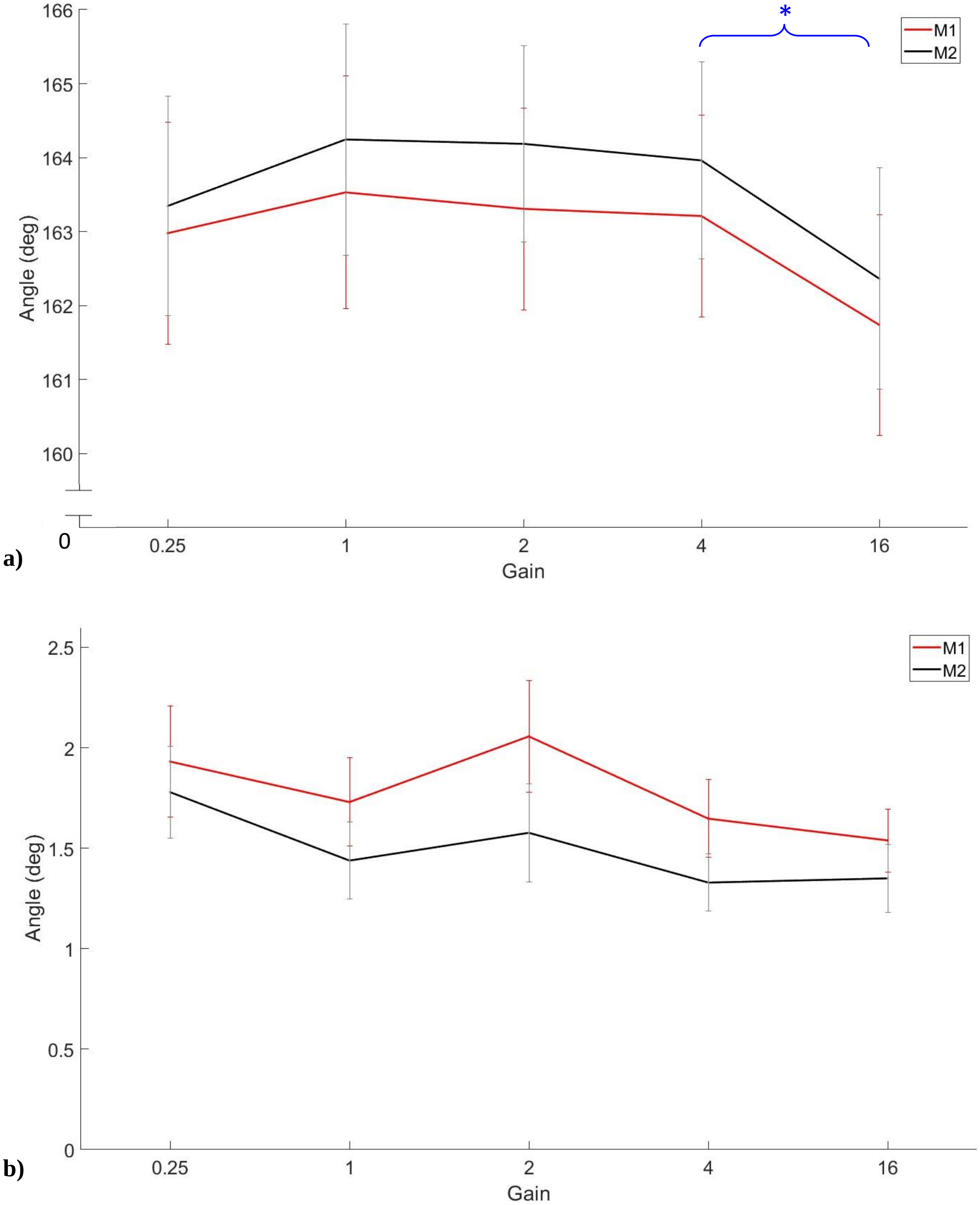


Figure 10. Knee joint across gain trials and between minutes.

a) Mean (SEM) knee joint angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Blue lines of significance indicate significant pairwise comparisons for collapsed minute conditions.

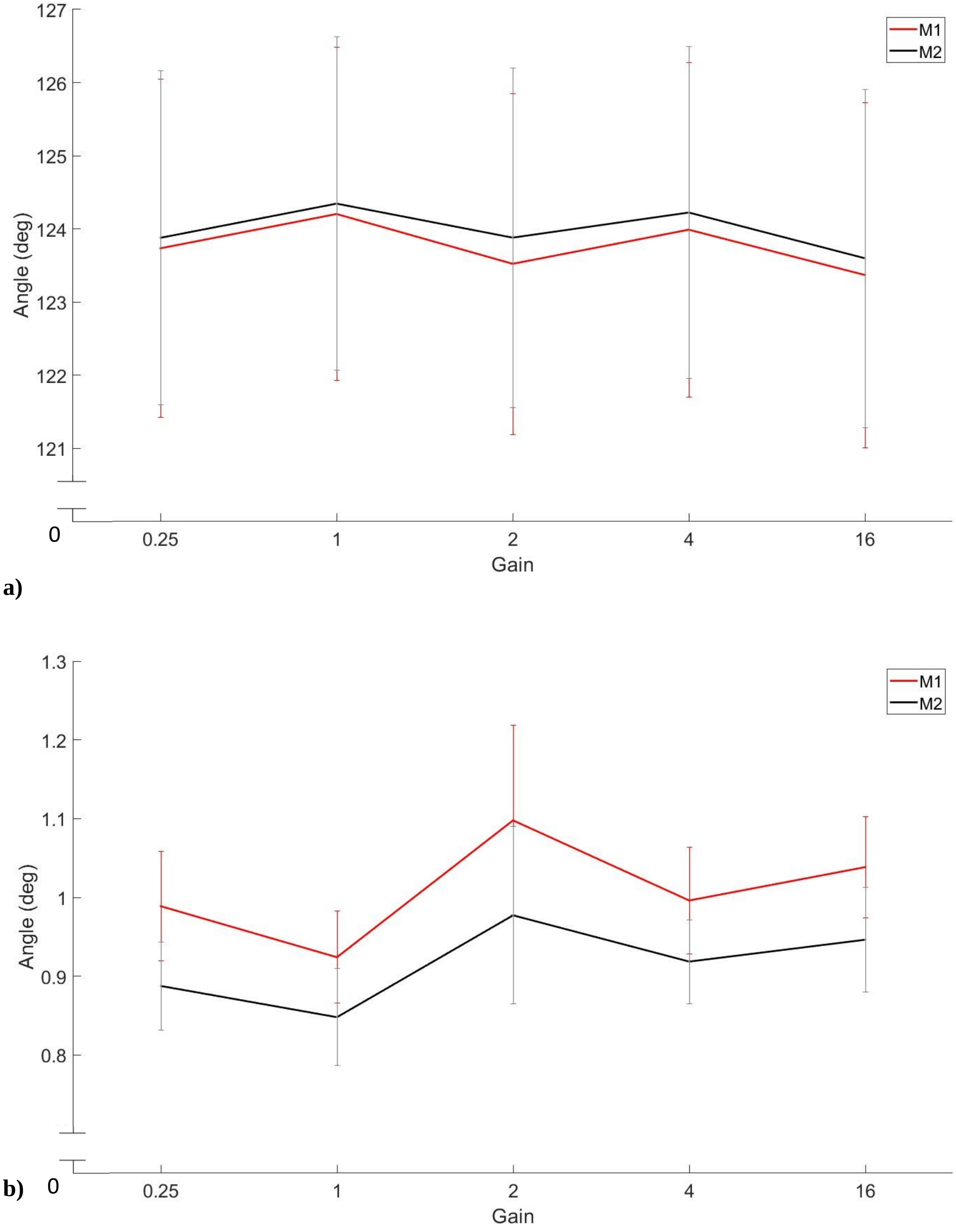


Figure 11. Ankle joint across gain trials and between minutes.

a) Mean (SEM) ankle joint angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*).

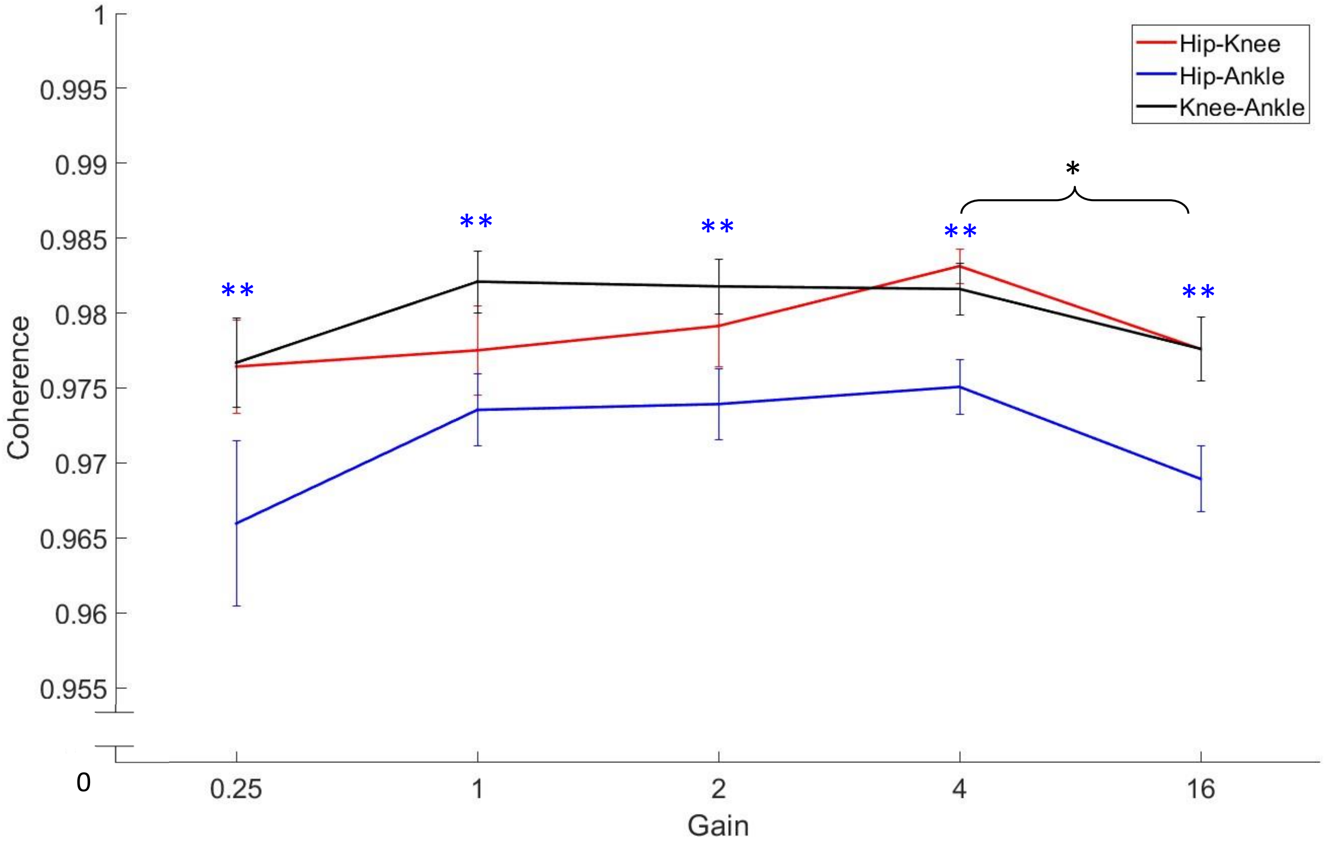


Figure 12. Hip-knee-ankle joint coherence across gain trials.

Mean (SEM) joint coherence plotted across full trial for each gain value. Significant pairwise comparisons are displayed between optic flow gain values (*). Significant pairwise comparisons are also displayed between coherence measures of each gain value (**).

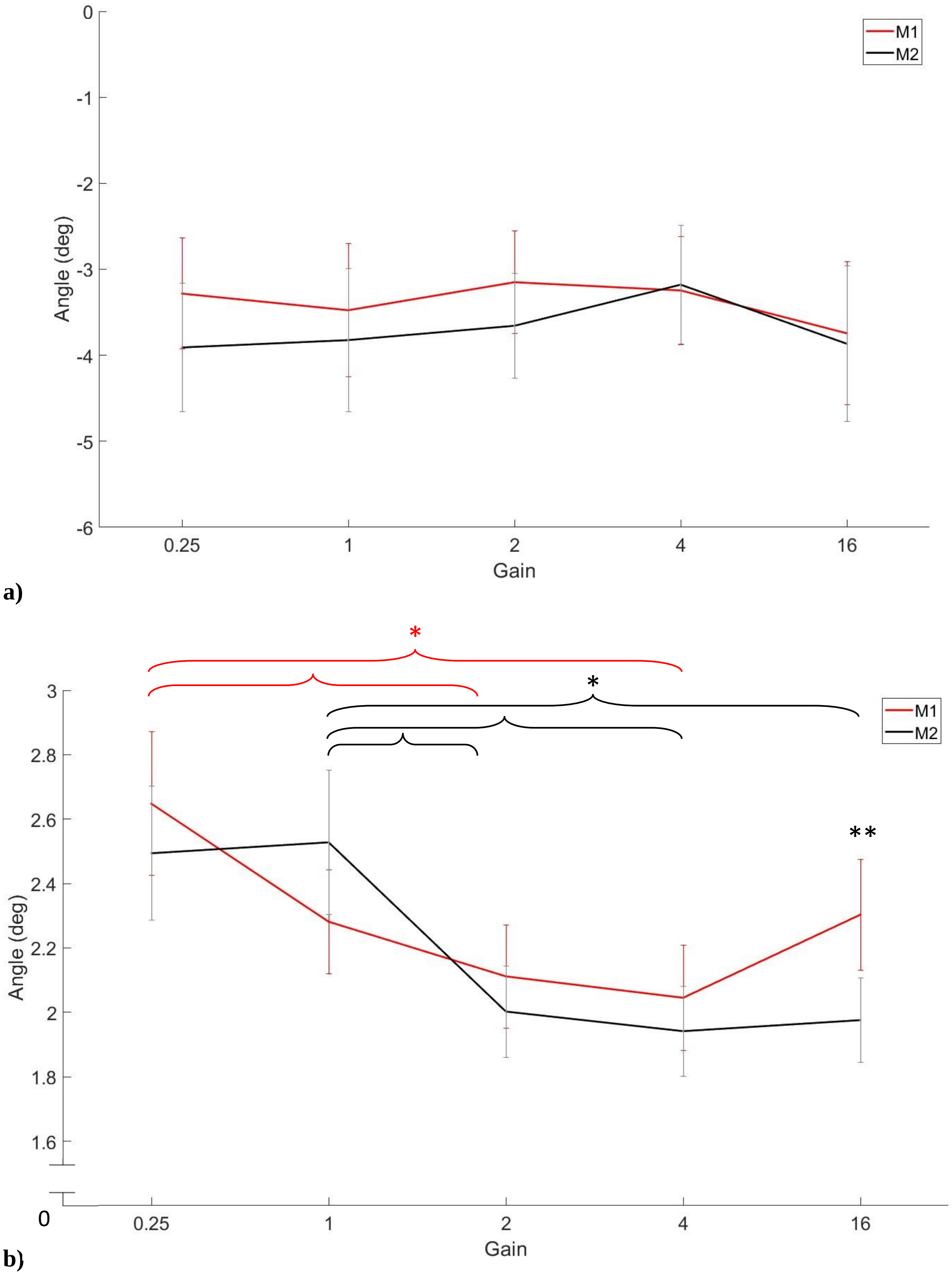


Figure 13. Trunk segment across gain trials and between minutes.

a) Mean (SEM) trunk segment angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Significant interaction effects of each gain between minutes are displayed (**).

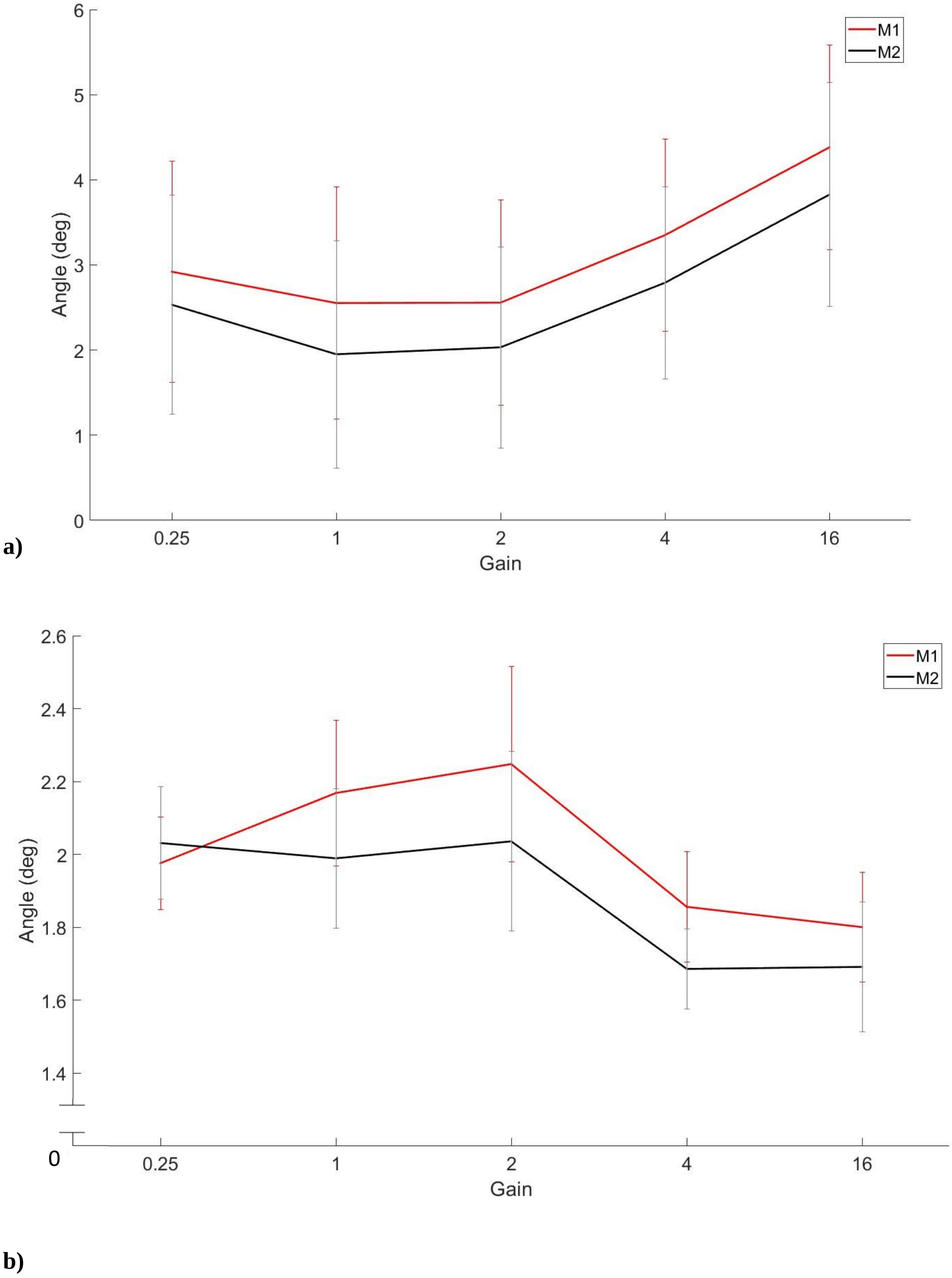


Figure 14. Thigh segment across gain trials and between minutes.

a) Mean (SEM) thigh segment angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*).

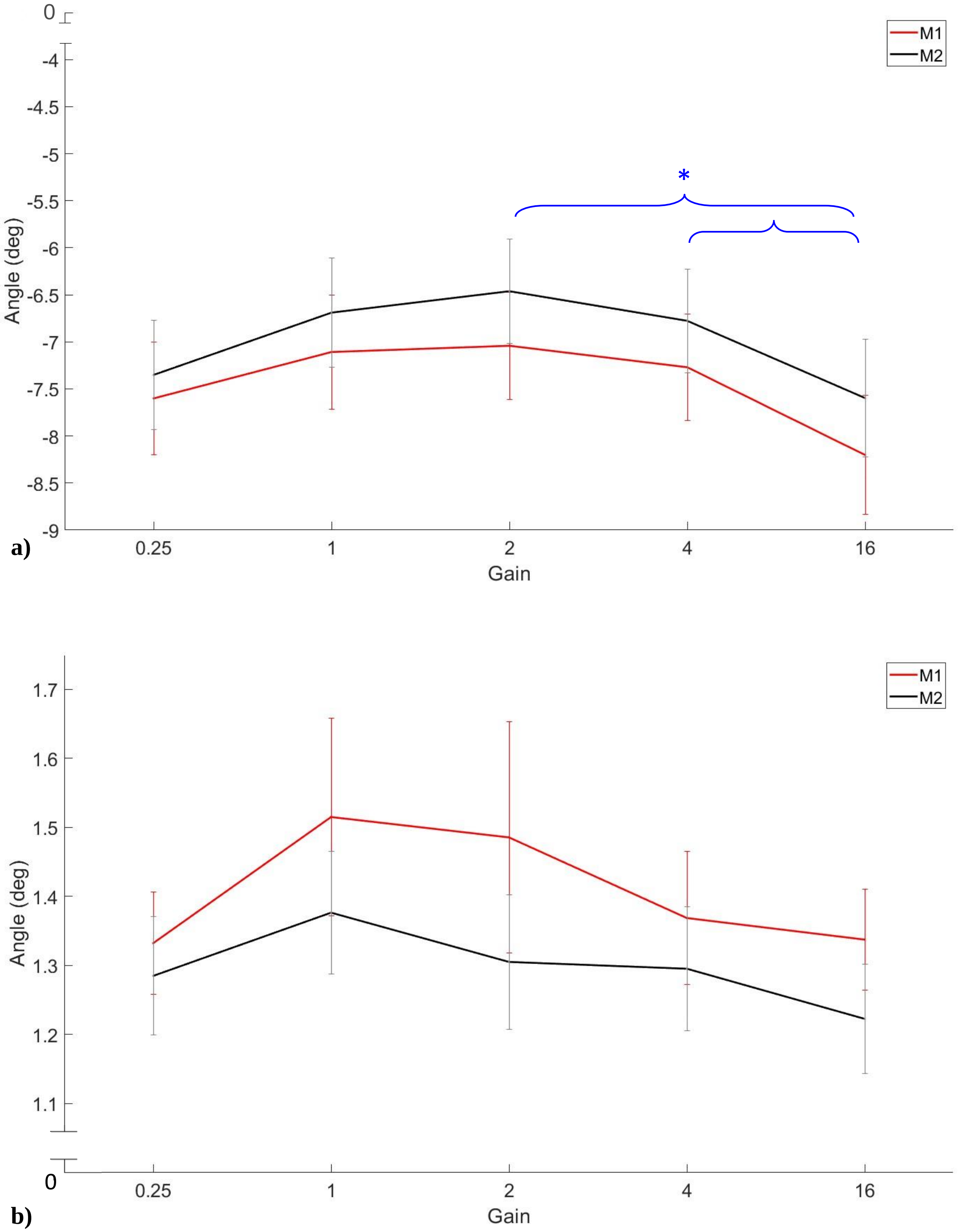
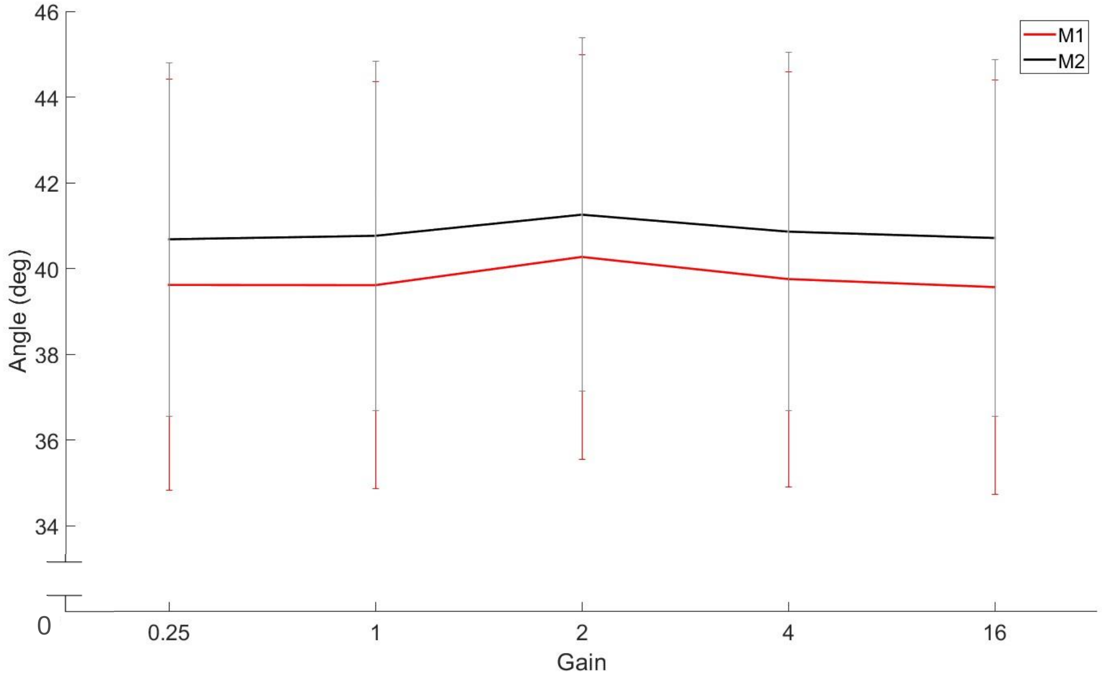
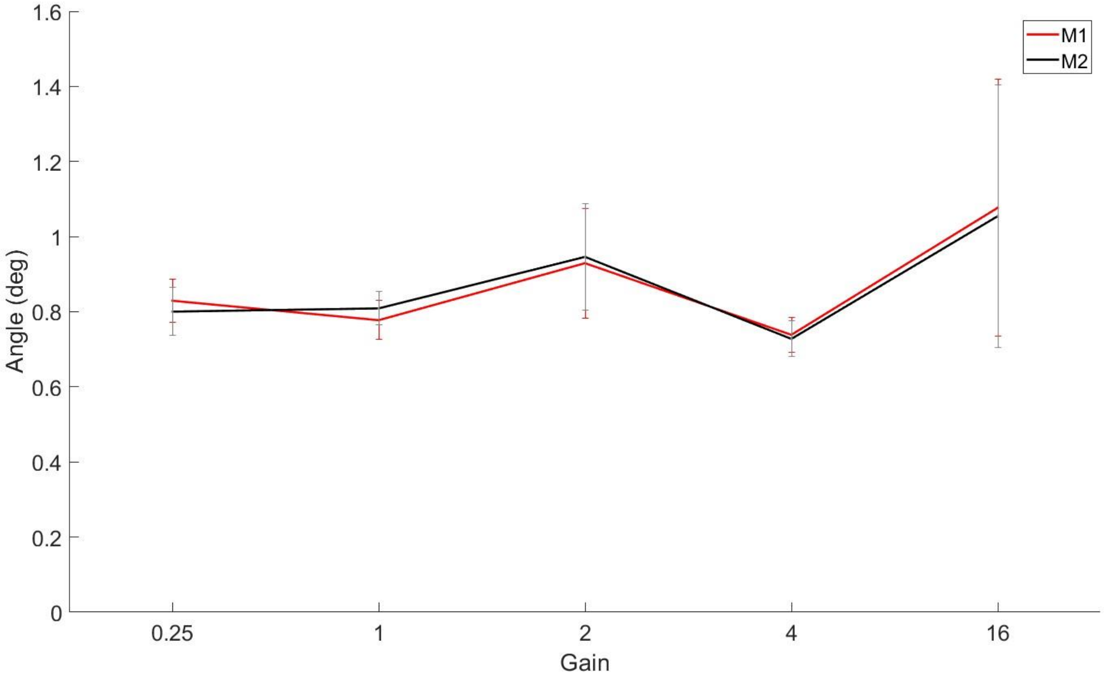


Figure 15. Shank segment across gain trials and between minutes.

a) Mean (SEM) shank segment angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*). Blue lines of significance indicate significant pairwise comparisons for collapsed minute conditions.



a)



b)

Figure 16. Foot segment across gain trials and between minutes.

a) Mean (SEM) foot segment angle and b) standard deviation plotted during both M1 and M2. Significant pairwise comparisons are displayed between optic flow gain values during each respective minute (*).

CHAPTER 4: DISCUSSION

4.1 Overview

The overarching aim of this thesis was to further investigate the relationship between visual feedback and balance control throughout a continuous, dynamic stance perturbation routine. This thesis utilized the manipulations of optic flow gain through VR to specifically increase or decrease the amount of visual feedback received by the participant. Concurrently, as visual feedback was provided, subjects were instructed to maintain an upright stance throughout a novel pseudorandom, continuous SS perturbation routine. A comparison between the first and second minute of each trial was performed to assess the capacities of participants to maintain an upright stance position throughout the duration of each trial. Overall, the results of this thesis demonstrate that an increase in VR-generated optic flow gain during continuous SS translations can generally elicit tighter regulation of postural control where participants perform sets of smaller-amplitude adjustments during trials. This was done by adjusting the COP and COM at a higher and lower frequency, respectively. Within a two-minute perturbation, these changes appear to be more evident during the first minute of trials as opposed to the second, displaying that the time to which one is exposed to the perturbation can have an impact on postural control strategies and sway patterns as well. Additionally, results in the second minute of the perturbation generate an adaptive response to the perturbation, in which participants were able to maintain postural control more effectively in the form of decreased postural sway. These findings are explored further within this section.

4.2 Changes to center of pressure

4.2.1 Amplitude of center of pressure

In M1, COP RMS decreased with an increase in optic flow until a gain of 2, and then increased back upward in higher gain values of 4 and 16. In M2, COP RMS significantly decreased with an increase in optic flow across all gains as hypothesized. This difference emphasizes the role that temporal dynamics have on postural control, as the influence of gain was not consistent across both minutes. Carpenter et al (2001) suggests that sampling durations must be at least 60 seconds long to have increased reliability of COP RMS measures within a signal. Although this study analyzed sampling durations during a quiet stance task, this study supports the use of a 120-second sampling duration in our current study to preserve the reliability of postural outcome measures. This suggests that there was likely not enough reliable visual feedback during the perturbation to elicit accurate postural adjustments within M1 during gains of 4 and 16. Additionally, Ko & Newell (2016) suggests that as acute dynamic task complexity increases, COP RMS can become more variable in a young, healthy adult sample. They demonstrated that younger adults could experience a slightly increased COP variability compared to a static balance task. This helps demonstrate the difference between COP RMS at a gain of 16 between M1 and M2. As hypothesized, the results in M2 suggest that the amplification of visual feedback toward a gain of 16 permitted subjects to have consistently greater control of their posture and minimize the movement of the COP, which supports previous work (Cawsey et al, 2009; Lavalle & Cleworth, 2023). These studies illustrate the utility of increased visual feedback in a quiet stance environment, although it is important to highlight that this is not the only situation where this increased feedback can be effectively utilized. The results from M2 can be further supported by other work using dynamic stance in combination with optic flow gain as well. Cheung et al (1990) suggested that if a particular sensory input is more readily available, then this information can be used more reliably. Similarly, Oie et al (2001) suggested that the gain toward a particular sensory input would be amplified if the particular input becomes more reliable over time. The prolonged presence of the increased visual information during the sinusoidal SS motion can permit a decrease in sway amplitude when the tasks are done concurrently, as the combination of inputs can help reinforce one another (Keshner et al, 2004). The presence of the physical motion of the SS further modulates the postural response and allows for increased dependency on the visual information, even as optic flow gain went above baseline and provided disproportionate feedback relative to participant motion (Keshner et al 2004; Sozzi et al 2020). Collectively, these results suggest that the increase in visual information during a dynamic balance task plays a key role in reducing COP amplitude.

4.2.2 Frequency of center of pressure

Additionally, COP frequency increased directly alongside increases in optic flow gain and was greatest at a gain of 16 in both M1 and M2. In M1, 4 and 16 were significantly different from all other gain values, and the same was present in M2, except for 0.25 and 4. COP frequency was significantly greater in M1 compared to M2 at gains of 2, 4, and 16, again illustrating the importance of temporal considerations when assessing the impacts of gain. Comparing these results to COP amplitude suggests a greater robustness and sensitivity of frequency toward changes in optic flow gain (Schiepatti et al, 2002; Sozzi et al, 2019). This is supported by Carpenter et al (2001), where it is postulated that increased COP frequency and decreased COP amplitude were a consequence of feeling unstable, thus creating increased stiffness through increased co-contraction of lower limb muscles. In this current study, subjects had to maintain their stance and avoid stepping. If they felt that they were experiencing instability at higher gain values, they may have utilized lower limb muscle co-contraction as a primary mechanism to initiate a postural adjustment such that they can maintain their current stance position. Further, Schmid et al (2011) displayed that oscillations in COP during a dynamic perturbation are more sensitive towards changes in visual information, and these can be used to dictate compensatory adjustments of the COM through muscle co-contraction as well. This demonstrates that the increased visual information provided at higher gain values has potential to directly impact the observed COP response.

4.2.3 Velocity of center of pressure

Lastly, as hypothesized, COP MVel was also significantly greatest at a gain of 16 in both M1 and M2. In M1, only gains of 4 and 16 were significantly different from all other gains, whereas in M2, 0.25 was also greater than 1 and 2. Across all gain values, MVel is significantly greater in M1 compared to M2. This highlights the differences that amplification or reduction of gain can have compared to baseline, rather than a single-direction effect based purely on optic flow magnitude. These results contradict some earlier work suggesting that reducing visual information during a continuous support-surface translation did not have any significant impact on COP velocity (Maki & Ostrovski, 1993). Similarly, Day et al

(1993) assessed changes in COP position and velocity at various stance widths while removing visual input and found that velocity would only decrease as a function of increasing stance width, indicating COP was decreasing due to a wider and thus intuitively more stable BOS. However, more recent work by Lavalle & Cleworth (2023) highlighted that in quiet stance, COP MVel as well as SD of COP velocity would increase as more visual information became available during quiet stance as a result of decreased response latency and therefore a faster adjustment, done at a greater velocity. Their study saw similar results to this one, as a gain of 16 had a significantly greater COP MVel compared to lower gain values. During dynamic stance, Ghulyan et al (2005) highlighted that amplitude of COP velocity can be further dependent upon translational velocity and oscillation frequency. Lower frequencies (0.1 - 0.25 Hz) can be easily adaptable for young, healthy adults as they are able to anticipate the corrective postural adjustment more accurately, which would prevent any large increases in COP velocity (Baloh et al, 1994; Ghulyan et al, 2005). Collectively, within a moderate oscillation frequency range as used in this study, there is a stronger perturbation input and therefore a greater challenge for the participant to remain upright and modulate velocity of sway (Mills & Sveistrup, 2018). The significant decrease from M1 to M2 across gains may suggest that participants were able to control and be more deliberate with their postural adjustments, which is even more evident at higher gains where a greater difference exists between minutes. As visual information increases, a decrease in COP amplitude and an increase in COP frequency could reinforce that participants can create smaller postural adjustments on a more frequent basis throughout the perturbation. This would be done at a higher velocity, providing a rationale for the upward trend in COP velocity and thus a tighter regulation of posture.

4.3 Changes to center of mass

4.3.1 Amplitude of center of mass

In M1, COM RMS was greatest at gain of 0.25 and lowest at a gain of 4, indicating a decreasing trend in RMS as gain approached 4. In M2, COM RMS was greatest a gain of 1 and lowest at 16, maintaining a decrease in COM RMS as gains went above baseline as hypothesized. In both M1 and M2, there were significant difference between minutes, where COM RMS was significantly greater in M1 for both 0.25 and 16. This result is interesting as it displays not only a time-dependent behaviour of posture, but that these differences are amplified at the ends of the gain spectrum that we assessed. These results are supported by previous work which suggested that changes in COM would generally decrease when more visual information was available. Foundationally, Buchanan & Horak (1999) demonstrated that increased visual information permitted a reduction in head, trunk, and whole-body COM position as translation frequency progressively increased from low (0.1 Hz) to moderate (0.1 – 0.5 Hz). The effects of vision at this frequency range also seemed to be most optimal in Dietz et al (1993). This was primarily presumed to be due to the position of the head in response to the moving platform. When the head was more in phase with platform translations as frequency increased, head pitch angle decreased displaying decreased oscillation of the segment (Buchanan & Horak, 1999), which would negate the need for adjustment of its position and the trunk would presumably follow. To add, this behaviour was also observed in quiet stance environments in response to increased visual gain through VR, where RMS of head position in the AP direction decreased as visual gain increased (Lavalle & Cleworth, 2023). In support of these observations, other work by Keshner et al (2004) observed that during continuous translations in addition to visual motion cues, vision became the primary sensory modality to rely on for maintaining stance, and that subjects were more responsive to the visual information when combined with physical scene motion. Similar to the observed decrease in COP amplitude with increasing gain in our study, the tendency to favour a particular sensory input will be amplified if the particular input becomes more reliable over time (Oie et al, 2001). In the current study, when higher gain values provided more reliable visual information to respond to, subjects demonstrated a potentiation of the COM response and thus a greater regulation of stance when combining the visual feedback with the other sensory inputs they were obtaining during the perturbation cycle. This helps demonstrate why a decrease in COM is observed as gain increases across trials. The fact that the trials were pseudorandomized only further reinforces that the visual information is objectively beneficial in maintaining the COM within the BOS, which therefore provides the basis for a reduction in change in COP. Additionally, Schmid et al (2011) suggested that the

COM oscillations were not only a product of the visual condition, but of perturbation frequency as well. During an unfamiliar perturbation, the main objective of the CNS is to maintain the COM within the BOS, even at the cost of additionally expended energy. Schmid et al (2011) observed this through increased tibialis anterior and soleus muscle activation as frequencies changed from low to high. Once the CNS was able to interpret the perturbation and the required response, the muscle activity quickly diminished thus demonstrated an effectiveness of the response and a downregulation of the behaviour as time goes on, showing evidence for adaptation to the perturbation (Schmid et al, 2011). Even when vision is not perturbed, increases in COP frequency, increased muscle activation amplitude, and decreased muscle onset latencies are progressively modified throughout the perturbation to prioritize control of the COM more efficiently (Kennedy et al, 2013). Although muscular response amplitudes and latencies were not evaluated in this study, this behaviour is very likely responsible for the reduction in COM amplitudes at higher gains. This is likely also the very mechanism responsible for the reductions in relative COM amplitude between M1 and M2. If increased ankle muscle stiffness was present to counteract a change in COM, we would expect to see a decreased range of COM displacement (Carpenter et al, 2001) initially, followed by a greater difference as the perturbation progresses due to the adaptive nature of the CNS during a dynamic perturbation, and resultantly more consistently accurate and reliable APAs. It is important to note that the decrease in COM amplitude and increase in stabilizing moments produced by increased ankle stiffness are accompanied by the increase in COP frequency discussed earlier. This was demonstrated in both quiet stance (Sweeny et al, 2021) and dynamic stance environments (Kennedy et al, 2013; Schmid et al, 2011).

Conversely, Jilk et al (2014) displayed that within a perturbation frequency range of 0.08 - 0.5 Hz, COM sway relative to the perturbation decreased in eyes closed conditions compared to eyes open, displaying variability in COM during a continuous pseudorandom perturbation. These results postulate that when vision was completely removed, that the changes in COM were almost or entirely dependent on vestibular and proprioceptive feedback to make corrective postural adjustments and modulate sway. The differences in reliability of visual information when it is not as readily available versus completely unavailable can dictate the mode of motor control that the system can proceed with. If vision is still available but reduced, individuals can maintain a feedforward approach to their postural adjustments, and predominantly rely on proprioceptive cues in conjunction with the remaining visual information available (Vitorio et al, 2014). In the current study, subjects were able to regulate their stance more effectively and reduce amplitude of COM sway as more visual information was provided, within the given perturbation frequency range.

However, the importance of the reliability of vestibular and proprioceptive cues in the observed population is not to be downplayed. A young, healthy adult population, such as in this study, without any visual or musculoskeletal impairments can reliably integrate and generate feedback from each system without concern. In contrast, individuals with vestibular deficits may have to rely further on alternative sensory inputs to achieve any modulation of sway. Nardone et al (2010) observed increased baseline sway in patients with unilateral vestibular loss compared to healthy subjects, as well as increased head displacement. Similarly, patients with vestibular neuritis would experience increases in AP sway due to a deficit in vestibulo-ocular reflex responses via diminished lateral semicircular canal feedback, and this would consequently place a greater reliance on vision and may impact postural performance if the mode of visual input is variable or not reliable (Esteban-Sanchez & Martin-Sanz, 2022; Cousins et al, 2014). It was outlined in previous work that adaptations to sinusoidal translations of a SS will be more evident in healthy subjects, and sway would generally decrease as a function of successive oscillation cycles, but individuals with a vestibular sensory deficit may require some sort of further training or exposure to the dynamic perturbations to obtain this same effect (Nardone et al, 2010). Alternatively, individuals with Parkinson's disease (PD) could still have relied further on visual cues, however the proprioceptive feedback and elicited postural adjustments could be less reliable due to the decreased accuracy of motor actions (Vitorio et al, 2014). Even if PD patients were to attempt to utilize APAs as a strategy in response to the perturbation, and even if the vision was present to potentiate the maintenance of the COM within the BOS, feedforward control of visual information and processing in the primary somatosensory cortex is known to be abnormal in PD patients (Nelson et al, 2018). Similarly, individuals with peripheral neuropathies will rely more heavily on APAs to make postural adjustments, and shift their dependence toward visual and vestibular systems for balance maintenance (Horak & Hlavacka, 2001; Nardone et al, 2010). In any case, this would lead to a less accurate and less reliable motor response, even if vision attempts to overcompensate for proprioception (Konczak et al, 2009). Therefore, although this study does not provide direct evidence of specific increase in neuromechanical efficiency in young and healthy adults, it represents a potential complementary mechanism in which subjects were able to further regulate their COM sway when more visual information became available.

4.3.2 Velocity of center of mass

In both M1 and M2, COM MVel was greatest at a gain of 1 and lowest at a gain of 4. We observed a significantly greater difference in MVel in gain values above a gain of 1 rather than below, where no significant change was seen from 0.25 to 1 in M1 or M2. However, we observed significant decreases in COM velocity as gain increased above baseline. Jeka et al (2004) highlighted that the velocity of the COM during quiet stance provides more reliable feedback to the CNS as opposed to position or acceleration and can therefore be prioritized during perturbations to help further aid in maintenance of the COM. Masani et al (2003) agrees with this finding, as they observed a modulation of COM displacement as a function of velocity, through the use of anticipatory modulation of ankle extensor muscle activity. They proposed that a sufficient amount of initial velocity information indicates the direction and amplitude with which the current COM position will be changed into the next perturbation, thus making COM velocity a key anticipatory measure for minimizing COM displacement after perturbation onset. This can also help rationalize the very close relationship in trend observed in COM RMS and MVel in our study. Schmid & Sozzi (2016) observed that muscular compensations to changes in COM position and velocity via increased tibialis anterior activation were much more apparent earlier on in the trial (within the first 60 seconds), and after this the response would be diminished, which can rationalize the significant increase in COM MVel in M1 compared to M2. Additionally, during dynamic perturbations, the compliance of the surface may not necessarily impact the amount of sway, but rather has the potential to increase the complexity of the sway response due to the changes in proprioceptive feedback, which can often be reflected as a decrease in COM sway velocity (Jeka et al, 2004). Additionally, Cleworth et al (2016) suggested that there would be no need for significant changes in COM velocity, and that segmental kinematic changes such as arm flexion and abduction can also be used to compensate for the perturbation without experiencing any major change in COM position. This suggests that when the subject is moving, additional muscle activation may be required from the upper body as well to counteract larger changes in AP COM. Schmid & Sozzi (2016) agrees with this, and further suggests that smaller changes in COM velocity and therefore slower, more controlled compensatory adjustments may be more effective in maintaining stance position during repetitive perturbations. The modulation of muscular responses and COM displacement mitigates the risk of movement inefficiencies, and permits for accurate displacement and velocity of COP changes to follow as well, which would verify the behaviours observed in the COP response in our study.

4.3.3 Frequency of center of mass

Interestingly the trends for COM MPF also seem to follow the trends of COM RMS. There is no significant difference in respective gains between each minute, although a significant difference was present between gains 1 and 4 when minute conditions were collapsed due to a lack of significant interaction, where MPF was significantly greater at a gain of 1. These results suggest that as more visual information was provided, subjects experienced some small decreases in frequency of sway. Alongside the changes that were seen for COM RMS, this could suggest that subjects were able to maintain stance at higher gains with lower-amplitude adjustments on a less frequent basis. The decreases in COM MPF were also accompanied by increases in COP MPF as gain increased, indicating that participants could maintain COM position while increasing stiffness at the ankle joint through plantarflexion. If the reductions in COM MPF were causing increased sway, there'd likely be an associated increase in COM RMS as the adjustments that were made had to be larger, although this was not necessary as the COP could be adjusted more frequently to prevent the need for a larger-amplitude change in COM. There are implications for COM MVel as well (Figure 7), which may also play a role in the lack of changes seen in COM MPF. Jeka et al (2004) observed that subjects experienced minor increases in COM velocity and SD of COM on a stable, non-compliant surface. This suggested that compared to a compliant surface such as foam, subjects could better maintain the positions of the COM with smaller adjustments, and this was also demonstrated by the least changes in mean path length (Jeka et al, 2004). Additionally, Brown & Frank (1997) suggested that the compensations for COM can be mitigated if participants begin in a neutral, upright stance, which they were instructed to do each time in this study. They also outline that there only may be an initial need for adjustment initially, and once the perturbation begins, the CNS initiates a control strategy for the COM. The constant relay of feedback and slow progression toward a more stable COM negates the need for additional adjustments later on (Brown & Frank, 1997), which would be reflected as a decrease in frequency over time. The difference from baseline gain of 1 to 4 highlight a clear, significant reduction in MPF, while a change in gain of 1 to 16 was almost the same, indicating that the additional visual information may not prove to be any more productive in postural control after a gain of 4 on average. Since gains above 16 were not assessed, this is an interesting component to assess in future work, as any possible differences or effects in additional visual information could not be seen here.

4.4 Functional significance of COP-COM behaviour

Within the young healthy adult population in this study, the addition of kinematic information in the form of COM and body angles helps to contextualize behaviours that are seen in previous work that exclusively use COP to assess posture. Carpenter et al (2001) observed a decrease in gastrocnemius and soleus muscle activation when the COP amplitude increased, displaying an inverse relationship between shank muscle activation and COP position. This therefore indicated that there was increased stiffness at the ankle joint when COP amplitude decreased, which allowed for tighter regulation of posture and therefore a more stable COP position throughout. This supports the stiffness model by Winter, wherein the body would become stiffer due to an expected perturbation as the perturbation was anticipated (Winter et al, 1998).

In a dynamic perturbation with continuous SS translations, there are likely increases in stiffness at the ankle muscles to counteract the changes in COM. This would illustrate an anticipatory mechanism as participants could pre-emptively induce ankle stiffening to reduce large movements in the COM once the perturbation begins, effectively reducing reactive sway. Similar behaviour was also observed by Keshner et al (2007) and Schmid et al (2011). The muscle stiffening strategy allowed them to not only mitigate COM movement once the perturbation began, but continue to modulate sway through this mechanism once they realized that it produced a desired result (i.e. maintaining position or remaining upright). This adaptation was especially prevalent in our study with young, healthy adults, as they could effectively utilize additional visual information to reduce motion of the head and trunk and translate the COM at a similar velocity as the perturbation progressed. This would negate the need for more frequent adjustments of the COM, and therefore be reflected as a reduction in amplitude for both COM and COP. The COP would not have to herd the COM as far to maintain the same stance position. As the larger amplitude and frequency adjustments in COP were greatest at higher gain values, this may suggest that the muscular response to increased visual information is present in the form of compensation occurring below the COM in lower limb musculature. Although the oscillatory signal in the experimental design was composed of several frequencies, it seems that the population was able to habituate fairly well, as predicted by Dietz et al (1993) and Mills & Sveistrup (2018), especially entering into M2 where significant decreases were observed in several postural measures that we analyzed compared to M1. Additionally, the increase in COM RMS at lower gain values is likely present due to the lack of visual information initially. Once the individual can integrate enough information from the perturbation, regardless of condition, the decrease in visual information encourages the system to minimize COM movement, and this causes a subsequent decrease in frequency of COP as a change is not needed yet when the COM has not moved significantly (Carpenter et al, 2001). Since the participants were instructed to complete a VR baseline trial, they were exposed to the mechanics of the visual perturbation in VR prior to experiencing the gain values, and were therefore able to anticipate a functional response pattern and modulate their COM more effectively as gains increased above baseline.

As hypothesized, this behaviour suggests the presence of a feedforward control mechanism of the COM which was performed through greater use of APAs, due to both increased ankle stiffness and increased presence of reliable visual information. Allum et al (2002) postulated that this stiffening could be an active protective strategy in response to the anticipation of a dynamic perturbation, and that this could be achieved through pre-perturbation muscle activation in the trunk and shank muscles. If the trunk is innately stiffer due to the increased stiffness, then it is initially assumed that this would result in greater instability as the trunk would now follow the direction of sway. However, Allum et al (2002) highlighted that the compensations to an impending perturbation can be better regulated in young adults as they are able to down-regulate the muscle synergies in the lower limbs to compensate for any changes in the COM if the trunk is stiffer and thus less variable due to an APA. This could provide insight as to why we observed significant decreases in shank angle SD in higher gain trials, indicating increased plantarflexion in the ankle muscles as the variation in trunk angle decreased. Further, Grüneberg et al (2004) observed early changes in trunk pitch angle prior to forward and backward perturbations, as well as decreased trunk angle after perturbation onset. This type of behaviour would not be attributed to reactive control as this would imply the presence of a negative feedback loop (Maki & McIlroy, 1997) during the perturbation which would reflect real-time, intentional reactions and compensations to the COP in response to each movement of the COM after the perturbation occurred, which were not observed nor reflected in previous work. Reinforced by previous work in quiet stance, Cawsey et al (2009) demonstrated that when more visual information became available, smaller movements of the COP, reflected through decreases in COP RMS, would be preferred over larger amplitude changes in the COM in an attempt to preserve the current stance and avoid an unnecessary adjustment. The constant comparative nature of the COP-COM provides subjects with the ability to pre-emptively minimize the movement of the COP in response to the COM, as they can extract and compare the relevant information they receive at higher gain values to avoid unnecessarily large and destabilizing postural movements at lower gains. Lastly, the potentiation of the

COM response through additional visual feedback outlined in this study, and reinforced by Keshner et al (2004), is notably dependent on the measures that were collected. Keshner et al (2004) outlined that the same behaviour of the COM would not have been observed if they solely assessed COP alone, indicating that there are further components to the postural response in a dynamic perturbation alongside visual cues. It was crucial that they were able to capture the enhanced muscular and proprioceptive responses that were observed in the head and trunk that subsequently aided in maintaining their stance. Having this basis for our current study, especially in dynamic stance, this provided additional contextualization for the results that were observed, and help rationalize the presumed "low amplitude-high frequency" response patterns that were initially anticipated.

4.5 Examining joint and segment angular changes

Changes in body angles over the course of the perturbation was an important strategy in regulating posture throughout the dynamic perturbation. Initially, the consistency in mean ankle angle or standard deviation suggests rigid control being present at the ankle joint throughout the perturbation, further enforcing the idea of an inverted pendulum-based control model. We observed a noticeable, but not significant change in mean hip angle as well, which supports the idea that the ankle and hip joints would more likely move in-phase at lower movement frequencies (below 1 Hz) (Creath et al, 2005). The rather fixed ankle position across trials could also permit for increases in lower body joint moments and increased dynamic loading at the ankle joint, which could therefore allow for increased support during the perturbation (Tajima et al, 2018). Importantly to note, Creath et al (2005) highlights that there can still be "simultaneously co-existing modes" of control strategies, such that the rigidity of the ankle joint does not mean that the joints above it must be innately stiffer as a result. Gage et al (2004) suggested that in such a model during quiet stance, horizontal movement of the COM would therefore mostly be attributed to changes in knee angle indicating strategies of contraction in the lower limbs. This strategy appears to be evident in our study as reflected by the relative reduction in knee angle across gains, and the increased variation in the knee compared to the ankle below. However, it is possible that this response may have been amplified due to the nature of the perturbation, especially at higher gains. We observed a significant decrease in mean knee angle from gains 4 to 16, which most likely indicated the need for compensation to modulate sway in the form of increased knee flexion indicating increased concentric hamstring and eccentric quadriceps activation. Interestingly, the increase in knee flexion seems to not have been complemented by significant hip extension as the hip angle remained relatively constant throughout. This would be an expected biomechanical response to maintain a stable COM position and prevent antero-lateral displacement of the COM (Gage et al, 2004; Horak & Nashner, 1986), although it seems that subjects were able to maintain an appropriate COM position until they reached higher gain values where greater adjustment of both COM and COP was needed to maintain their support. This was also reflected by the significant change in hip-knee coherence between gains of 4 and 16. Gage et al (2004) suggested that when more knee joint movement is present, this would permit for a higher correlation between lower-body joint angular displacement and AP COM compared to when there was limited to no change in lower limb angle. Postural responses to forward sway would also be primarily corrected through knee flexion and stiffening at the ankles (Allum et al, 2003; Grünebeg et al, 2004) These findings indicate that the significant changes in knee angle are a regulatory strategy that occur simultaneously with the changes in COM throughout the trial. Additionally, these would be likely be more evident in the earlier components of the trial where COM RMS was higher, suggesting moments of greater sway and therefore greater need for up-regulation of control in the lower body. Maki et al (2003) also determined that this type of strategy would be more prevalent in a young, healthy population, whereas an older adult population would likely require some sort of change-in-support to replicate a similar effect.

When examining segmental angular changes, we observed no significant changes in mean trunk angle. However, the significant decreases in trunk SD were notable, which highlights the adjustment of trunk position as more visual information was provided. During sinusoidal translations, Buchanan & Horak (1999) demonstrated that visual information was able to assist in stabilizing posture by helping control the head and trunk in space. As translation frequencies rise above 0.1 Hz, increased inertial forces begin to act on the head alongside an increase in translational motion. This instinctively provides subjects with an inertial cue to decrease AP head and trunk motion to remain upright and in place, and to shift towards a primary utilization of visual information to make postural adjustments. Across all frequencies, they demonstrated that fixing the head and trunk in space was functionally associated with decreased sway and increased control of posture. In our study, this same effect is demonstrated by the increase in visual information available, displaying that subjects were able to effectively utilize the increased information provided above a gain of 1 to increase postural stability. This also may rationalize the significant decreased in trunk standard deviation as gain increased. As more visual information was provided, the subjects could effectively utilize this information to decrease segmental response amplitudes and didn't require as much variation in trunk angle as the perturbation progressed. Further, the strategy of perturbation prediction and "platform riding" suggested by Mills & Sveistrup (2018) supports these observation, and reinforces that there is increased neuromechanical efficiency demonstrated by young, healthy adults during dynamic perturbations, and especially during sinusoidal translations. Keshner et al (2007) also compared head and trunk kinematics and found similar results, although instead they suggested that changes in visual scene velocity in the pitch plane would have a greater impact on head and trunk angle and angular velocity, despite any anticipations in perturbation. This implies that the differences may have been more evident as a product of different visual scene characteristics, although we did not assess behaviours with pitch plane scene velocity in our study. Lastly, Cleworth et al (2016) showed that compensations to SS translations were mostly attributed to changes in arm angular displacement, and shoulder flexion and abduction. Again, muscular response amplitudes were not evaluated in this study, although evaluation of deltoid activation and response pattern may have helped confirm this similar behaviour during our dynamic perturbation paradigm.

The lack of change in thigh angle or SD is consistent with the results observed in hip angle. Additionally, the ankle strategy that was presumed would negate the need for much change in foot segment angle, especially due to this study consisting of fixed-support perturbations. We did however observe some significant changes in shank angle. The production of increased knee flexion and anticipatory ankle stiffening permit for this observation, but we also must consider the impact of the SS.

Keshner et al (2004) highlight increased shank power and AP RMS when translating SS motion was present during a perturbation, and a lack of much change with purely visual scene motion. By analyzing the results when stimuli were not combined, they were able to display the isolation of the shank angular contributions and that it could be controlled by individual inputs. Since our study utilized both SS and relative scene motion, it initially demonstrates a combined effect. The Keshner work does support the assumption that the shank changes are controlled by the SS and the BOS characteristics in general, although its important to consider that the motion of the scene was isolated compared to our study where motion of the visual scene was relative to actual motion. In contrast, the previous results discussed regarding the head and trunk can be more attributed to the visual characteristics as these are closer in proximity and more responsive to changes in the visual system. Once again, the regulation of shank position seemed to be more evident later in trials during M2, suggesting that the anticipatory adjustment of the segment became more effective as the perturbation progressed.

4.6 Limitations

4.6.1 Fixed-support vs. change-in-support strategies

The first limitation of this thesis pertains to the study design. We opted to utilize a fixed-support strategy for the dynamic perturbation, wherein subjects were specifically instructed to maintain the foot positioning atop the SS and were not allowed to step or move their feet during any of the trials. Certain trials were re-attempted or omitted due to the participant stepping mid-trial, or utilizing a change in support strategy in some way. Firstly, while the use of a fixed-support strategy is fairly common in research, it does not encompass the entire basis of dynamic postural control. These types of perturbations are intended to elicit certain responses that are observed regarding postural control strategies for remaining upright, and how the behaviours of subjects' bodily kinetics and kinematics will change over the course of a perturbation in order to best maintain posture and attenuate sway. The use of change-in-support strategies such as reactive stepping or reach-to-grasp movements require and subsequently elicit different sensory and motor control mechanisms (Maki et al, 2003). Secondly, these types of perturbations will often assess a greater component of reactive postural control of an individual in response to the step or grasp to return to a baseline position. Since this study did not assess all types of paradigms, we are not able to make conclusions towards all types of dynamic perturbations, but rather towards a more specific type of fixed-support perturbations and balance behaviours.

4.6.2. Lack of psychological assessment

Another way that this study was limited was in that did not contain any sort of psychological assessment to understand subjects' feelings or confidence towards their own balance capabilities. Subjects did fill out a form prior to data collection where they were to inform of any neurological, musculoskeletal, or visual impairments that would impede their participation in the trial, although psychological measures were not considered nor assessed. The reason this is relevant is because there exists work that illustrates differences in postural control that are attributable towards the fear of imbalance or falling rather than purely towards bodily compensations. For example, work done by Zaback et al (2015) provided subjects with questionnaires to assess their anxiety, self-consciousness, and physical risk-taking prior to participating in the study. Other work by Cleworth et al (2016) utilized questionnaires projected to subjects through VR to specifically assess balance confidence towards each task they were about to perform. This provided researchers with the ability to assess participant readiness to perform balance tasks and would also help to further contextualize their results and establish tangible relationships between fear or anxiety of falling and balance performance. Lastly, the assessment of psychological factors and balance confidence would allow for the further validation of such work in a rehabilitative or long-term care setting. Individuals with impairments in posture or balance may be able to relate more closely to the experimental design, and this may help to prove even further utility in a clinical setting.

4.7 Recommendations toward future research

This thesis evaluated dynamic stance behaviours while exposed to a novel dynamic perturbation paradigm upon an AP translating SS. To build upon this work, future research could first be directed towards the alteration of various platform characteristics. The velocity and acceleration of the SS was constant across all trials and all subjects, and the displacement was within a fixed range. The translation frequency was indeed pseudorandom in nature, but the frequency range did not exceed 0.5 Hz. The use of a pseudorandom frequency range still holds plenty of clinical utility and has been used in the past, although a next step would be to assess higher frequency ranges (0.5-1 Hz, > 1 Hz, etc.). This work would help uncover postural control strategies that are different, or that may change from the ones that were observed in this study or in previous studies of similar nature.

Another recommendation for future work would be to verify the muscular components of this study that would be reflected through EMG analysis. Since the kinematic contributions were such a key component of this study, incorporating the use of EMG recordings and analysis would permit not only for further investigation of the postural behaviours in a new domain, but to help verify the conclusions put forth by this study that were drawn from kinematic measures such as changes in COM as well as joint and segment angular changes. In this study, the changes in joint and segment angles helped demonstrate postural control strategies and from these we were able to infer muscular response behaviours. Although, measuring muscular response amplitudes, latencies, and co-contraction of upper limb musculature in the arms and trunk, and particularly lower limb musculature surrounding the hips, knees, and ankles could help validate these observations and carry them further in the literature.

Lastly, this work examining the postural behaviours of young, healthy adults proved to be very informative about postural control in a dynamic stance environment with the addition of visual cues. It would be helpful to take this further and explore populations of interest such as populations with neurological conditions where the relevant sensory systems for balance are affected, or older adults to explore the impacts of aging on dynamic postural control. Alongside these changes would of course come some methodological considerations for safety purposes, although this work is especially important to continue to establish training programs and/or rehabilitative protocols for these groups. This would present a new and enticing way of identifying issues with, and improving postural stability among those that are at-risk of balance deficit due to physiological condition(s).

4.8 Real-world and clinical applications

The goal of this research was to continue to explore the impact that visual information can have on postural control in dynamic stance environments, as the use of visual information is paramount to the regulation of postural control. When the capabilities of the visual system are assessed within this type of environment while manipulating optic flow gain, this can help further explore the regulation that the visual system has in conjunction with the postural control system in a dynamic setting, which is a critical relationship for balance maintenance within many activities of daily living. In addition, this work helps in providing the utility of dynamic balance training in preventing falls in at-risk populations such as older adults or neurological populations. This type of knowledge or training can help eliminate some potential risk for such populations and helps establish how a visual gain routine during a dynamic perturbation, such as one similar to this study, can be useful in preventing falls and establishing physical competency and safety. The efficacy of such interventions is evident in Nardone et al (2010), where decreases in postural sway were seen in both quiet and dynamic balance tasks after performing different balance rehabilitation exercise protocols. Additionally, the ability to manipulate visual feedback through VR provides a new basis for assessing visual contributions to balance. VR paradigms are customizable and portable modalities that can have real impact as an intervention for populations at risk of falling due to inadequate postural control. Exposure to visual conditions and training of the sensory systems relevant to balance can help to mitigate the prevalence of such falls. Overarchingly, through further examining the flow of visual information and its relationship with other balance mechanisms in dynamic stance environments, we can further understand the contributions towards sensorimotor integration that help regulate balance in everyday life.

4.9 Conclusions

In conclusion, the overall findings of this thesis demonstrate the utility of optic flow gain through VR, and how the use of such paradigms can be used in combination with balance platforms in order to assess the influences of visual feedback on dynamic postural control and stance behaviour. This study found that an increase in visual feedback through VR permitted individuals to have an attenuation of sway reflected by an overall decrease in COM and COP amplitude, and an increase in COP frequency and velocity as more visual information was provided across gains. A decrease in COM frequency was also observed. This behaviour was able to reflect that subjects generally experienced a tighter control of posture and therefore less sway. We also determined that the contributions of different body segments and kinematic behaviour was directly related to the behaviours of the COM during a dynamic perturbation, and that there exists an anticipatory nature towards dynamic postural control wherein individuals will make pre-emptive, anticipatory postural adjustments to compensate for perturbations or interruptions to their stance that they may experience. This study provided insight into the postural behaviours of a young, healthy adult population, and future research can very much benefit from the use of VR paradigms in combination with dynamic stance environments to assess the balance of older adults and neurological populations.

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Appendix A – Dynamic perturbation cycle amplitude vs. frames

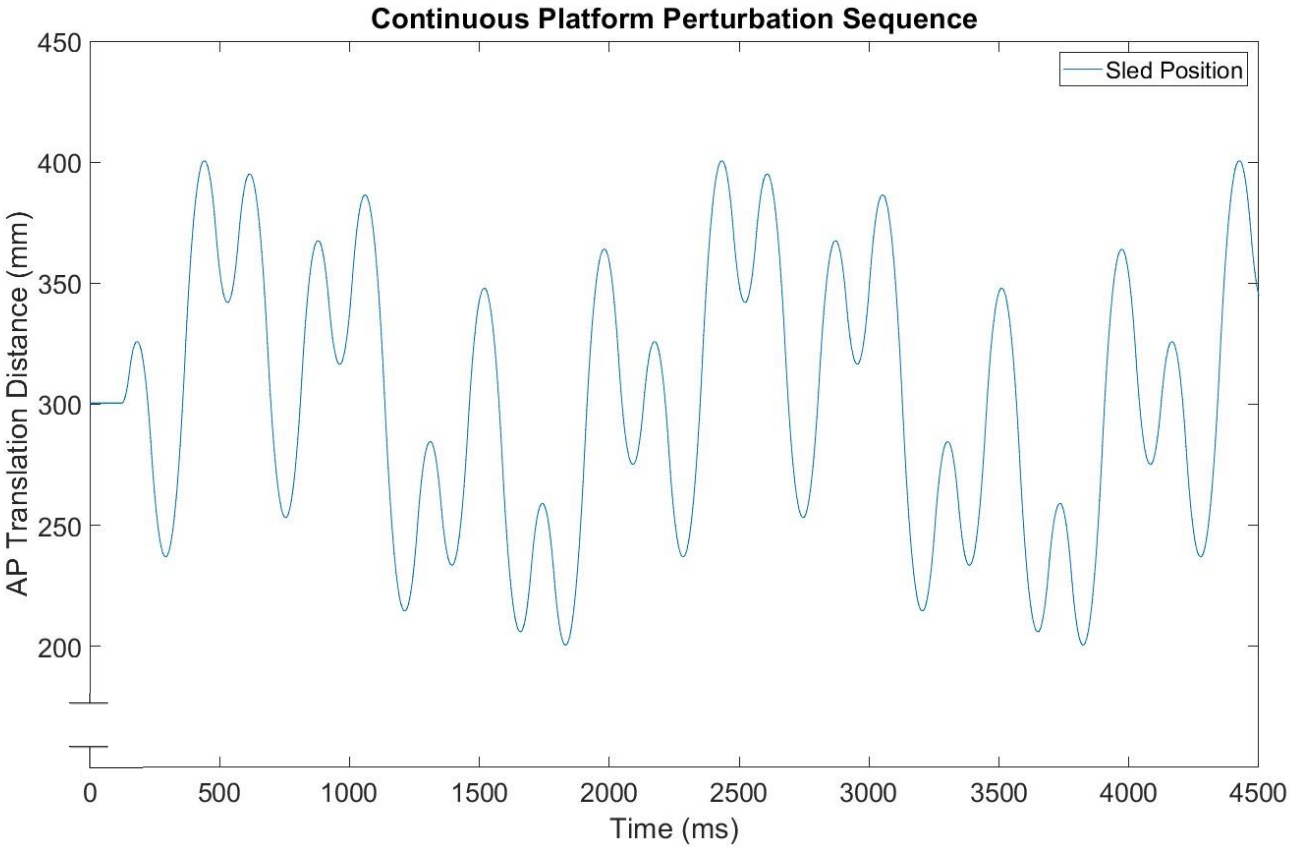

Figure 17. Pseudorandom platform perturbation cycle

One full cycle of the pseudorandom perturbation cycle utilized during experimental trials, displaying perturbation amplitude (mm) over time (frames) throughout the perturbation.

Appendix B – Summary of 2x5 repeated measures ANOVA results for gain trials

Results summary displayed for RMS (mm), MPF (Hz), MVel (mm/s), joint angles (deg), and segment angles (deg). Bolded p-values denote statistical significance.

	Minute			**Gain**			**Minute*Gain**		
	F	**df**	**p-value**	**F**	**df**	**p-value**	**F**	**df**	**p-value**
COP RMS	0.908	1,27	0.349	4.511	2.940,79.368	**0.006**	3.120	2.959,79.888	**0.031**
COP MVel	78.079	1,27	**<0.001**	32.661	2.192,59.172	**<0.001**	15.686	2.277,61.476	**<0.001**
COP MPF	12.364	1,27	**<0.001**	40.127	2.674,72.189	**<0.001**	5.650	2.987,80.650	**<0.001**
COM RMS	4.612	1,27	**0.041**	4.502	2.595,70.066	**0.009**	6.162	4,108	**<0.001**
COM MVel	1.424	1,27	0.243	7.949	2.484,67.057	**<0.001**	2.921	2.868,77.432	**0.041**
COM MPF	3.437	1,27	0.075	3.224	2.920,78.844	**0.028**	1.145	4,108	0.340
Hip Mean	0.91	1,27	0.766	0.899	4,108	0.437	0.569	4,108	0.686
Hip SD	0.976	1,27	0.332	2.502	4,108	0.072	1.252	4,108	0.290
Knee Mean	13.686	1,27	**<0.001**	2.858	3.014,81.368	**0.042**	0.827	4,108	0.511
Knee SD	15.982	1,27	**<0.001**	1.499	4,108	0.207	0.805	4,108	0.525
Ankle Mean	4.796	1,27	**0.037**	2.150	4,108	0.129	1.018	4,108	0.401
Ankle SD	16.243	1,27	**0.000**	1.165	4,108	0.330	0.219	4,108	0.928
Trunk Mean	7.036	1,27	**0.013**	0.477	4,108	0.753	2.091	4,108	0.087
Trunk SD	4.284	1,27	**0.048**	5.807	3.059,82.604	**0.001**	5.074	3.130,84.500	**<0.001**
Thigh Mean	13.349	1,27	**0.001**	1.455	4,108	0.221	0.280	4,108	0.890
Thigh SD	5.626	1,27	**0.025**	1.722	4,108	0.150	1.363	4,108	0.252
Shank Mean	15.694	1,27	**<0.001**	6.736	3.015,81.402	**<0.001**	1.677	4,108	0.161
Shank SD	10.137	1,27	**0.004**	0.254	4,108	0.329	0.392	4,108	0.814
Foot Mean	1.025	1,27	0.320	1.607	4,108	0.178	0.664	4,108	0.618
Foot SD	0.025	1,27	0.876	0.613	4,108	0.654	0.312	4,108	0.869