INTERSENSORY VESTIBULAR CONTROL OF STANDING BALANCE

by

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Abstract

During standing balance, error signals delivered to the vestibular system through an electrical vestibular stimulus elicit compensatory muscle responses in the appendicular muscles involved in the control of standing. This response is only present in muscles that are active in balancing the whole-body but not present when muscle activity is unrelated to balancing the body. Previous work has shown that visual, vestibular, and proprioceptive cues that are congruent with efferent muscle signals through a robotic simulation of standing balance elicit this vestibular reflex response. The physiological connections between the visual, vestibular and neck somatosensory systems imply that congruency between any of the three sources of information and the efferent muscle signals could elicit a vestibular reflex response, but this has not been tested due to difficulties in isolating sensory feedback during standing balance. A newly constructed robotic balance simulator enabled an experimenter to control the congruency between afferent feedback and motor actions associated with standing balance. Here, we tested whether the vestibular reflexes rely on vestibular cues of self-motion being relevant to the control of standing balance. Eight healthy subjects maintained balance on the robotic balance simulator while vestibular cues of balance were minimized. To achieve this, the robotic balance simulator maintained the whole-body stable in space while providing visual and/or lower-limb somatosensory information that was congruent with the motor control of standing. It was shown that even without vestibular cues of self-motion, a vestibular reflex response can be elicited. These results will lead to a better understanding of the vestibular control of standing balance, and may be applicable to populations with balance deficits.
Preface

This dissertation is original, unpublished, independent work by the author, M. Shepherd.

The work proposed here has been approved by the Clinical Research Ethics Board of the University of British Columbia (H13-01951).
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¹ See APPENDIX B: Robotic Balance Simulator, Visual Delay
Introduction

Balance

Standing balance is a human ability that is often taken for granted. It is the ability to remain upright, while being supported only by the feet. To remain upright, the centre of gravity of the upright body must remain within the area between the feet known as the base of support. This is not a static process. Breathing, changes in body posture, gravity and external perturbations can cause the human body’s centre of mass to move, which can cause the centre of gravity to move. Passive structures like tendon stiffness resist these perturbations, but are not enough to maintain the centre of gravity within the base of support. Ankle stiffness can account for up to 90% of static toppling torque (Loram & Lakie, 2002). Muscle activity compensates for the remaining torque required to maintain the centre of gravity in the stable region. Postural sway describes the tendency of the body’s centre of mass to oscillate during balance. The body has many segments that move during postural sway; however, the physics of standing balance can be modelled as an inverted pendulum (Gage, Winter, Frank, & Adkin, 2004).

Loss of balance increases with age and can lead to injury, hospitalization, decreased independence and decreased quality of life (World Health Organization, 2007). To find a means of preventing a loss of balance, an understanding of the control of unperturbed balance is required. To understand unperturbed balance, the sensory components of balance, and their role in the balancing process must be understood.

Human standing balance relies on the vestibular system, vision, and somatosensory cues. The vestibular, visual, and somatosensory systems, and their role in human standing balance, will now be discussed.
The Vestibular System

The vestibular system can be sub-divided into central and peripheral components. The central vestibular system consists of the parts of the nervous system that relate to balance, gaze control, and location of self in space. It receives information from the peripheral vestibular system. The peripheral vestibular system contains two vestibular apparatuses, which are located in the inner ear on either side of the head (see figure 1). Each apparatus contains two otoliths, which encode linear acceleration of the head, and three semicircular canals which encode angular acceleration of the head. The otoliths consist of the utricle and the saccule. Both of these structures are small sacs of a fluid called endolymph. Along one side of each otolith is a macula, which is a jelly like substance containing hair cells. Each hair cell contains multiple stereocilia, and one kinocilium. The kinocilium is located at one edge of the hair cell. The orientation of the hair cell is defined by the location of the kinocilium relative to the stereocilia. When the stereocilia bend towards the kinocilium, the hair cell becomes depolarized, leading to an increase in firing rate of the associated vestibular afferent. When the stereocilia bend away from the kinocilium, the hair cell becomes hyperpolarized and the associated vestibular afferent fires less. It is hypothesized that this is accomplished by mechanical ion gating (Goldberg, et al., 2012). A stretch towards the

Figure 1: The inner ear
An illustration of the inner ear from the public domain (Wikipedia, 2014)
kinocilium would likely open the ion gates, causing a change in the potential of the hair cell. Inversely, a stretch away from the kinocilium would make the gates less likely to open. In both otolith organs, there is a curving axis called the striola that runs along the macula. This axis is defined by the orientation of the hair cells. In the utricle, the hair cells are oriented with the kinocilium closest to the striola. In the saccule, the kinocilium are located furthest away from the striola. Between the macula and the endolymph is a crystal-like layer. When the otoliths experience linear acceleration, the momentum of the crystals causes them to pull on the endolymph, which bends the stereocilia. The bending of the hair changes the firing rate of the afferents associated with the hair cells and is interpreted by the vestibular system as an acceleration of the otoliths. In each vestibular apparatus, the utricle lies horizontal, with the macula on the bottom. The utricle detects accelerations in the horizontal plane, as well as head tilts. The saccule, on the other hand lies perpendicular to the utricle and detects accelerations in the vertical plane. Due to the orientation of the hair cells around the striola, any motion will cause some hair cells to polarize, while others depolarize. It is thought that this could actually be beneficial in accurately determining the acceleration experienced by the sensory organs (Kandel & Schwartz, 1985).

The semicircular canals are three tori which are oriented at ~90 degrees to one another, and are known as the horizontal, the anterior (or superior), and the posterior semicircular canals. Each torus detects angular acceleration in one plane. This allows the vectors of angular acceleration detection to span a 3-dimensional space. Each canal has an ampulla, which is a bulging area at the base that contains the cupula. The cupula is a jelly-like structure that contains hair cells. The canals are filled with endolymph. The hair cells in the ampulla are aligned with each other to detect rotations in the canals. As the canal is rotated the endolymph pushes on the cupula, which causes the stereocilia to bend, and changes the likelihood that the vestibular afferent associated to the hair cell will fire. Firing of the afferents is interpreted by the vestibular system as an angular
acceleration. The otoliths and semicircular canals form two vestibular apparatuses, one on either side of the head, which are mirror images of each other. This system provides the brain with information on the location and movement of the head in space, and can act as an indicator of the position of the whole body in space.

In 1943, Adrian studied the vestibular system of cats (Adrian, 1943). Anaesthetized or decerebrate cats were placed on a hanging platform, which was then rotated horizontally or tilted laterally. The platform allowed the head to remain fixed relative to the body, and the brainstem was probed with an electrode to locate the vestibular nucleus. The firing of the vestibular nucleus was observed during the two mentioned motions. To investigate the behaviour of the horizontal canals in more detail, cats were also placed on a turntable. In cats, firing rate of the vestibular nuclei is influenced by the angular and linear acceleration of the vestibular apparatus. This is also the case in other mammals such as rats (Bush, Perachio, & Angelaki, 1993), and monkeys (Fernandez & Goldberg, 1971). Decerebrate rats were fixed in a device that allowed for rotation along a horizontal axis, translation along a perpendicular horizontal axis, and rotation around a vertical axis. The rats were initially placed so that the horizontal and vertical axis of the device aligned with their lateral, and vertical axes (respectively), and so that the axis of translation aligned with their longitudinal axis. In subsequent trials, the rats were furthermore affixed in the apparatus in different horizontal rotations that caused them to be translated along an axis that was horizontally offset from their initial longitudinal axis by -90, -45, 0, 45, or 90 degrees in the horizontal plane, and rotated around an axis that was horizontally offset from their initial lateral axis by -90, -45, 0, 45 or 90 degrees in the horizontal plane, respectively. Electrodes were stereotaxically placed in the vestibular nucleus. The rats were rotated, and neurons in the vestibular nuclei encoded both linear and angular acceleration. Fernandez and Goldberg (1971) similarly looked at vestibular nuclei in spider monkeys, and then modelled the relationship between the angular acceleration experienced by the vestibular labyrinth and the firing rate of the
primary vestibular afferents. Male squirrel monkeys had electrodes inserted into their first order vestibular afferents, and were attached to an apparatus that rotated their heads in space (Goldberg & Fernandez, 1971). The monkeys were rotated through a series of sine waves, ranging from 0.006 to 8.0Hz. The input accelerations and recorded firing rates were then used to fit coefficients of a transfer function that maps head acceleration to the firing rates of the vestibular neurons. A linear relationship was found between accelerations and firing rate. It was also shown that between ~0.1 to ~1Hz, the firing rate of the afferents is in phase with angular velocity (Fernandez & Goldberg, 1971)(see figure 2). These studies show that the firing of the vestibular afferents is indicative of head motion.

Figure 2: Bode plot for firing of semicircular canals relative to head velocity for squirrel monkeys

A bode plot is presented here for the transfer function for firing of the semicircular canal afferents in squirrel monkeys relative to head velocity. From the Fernandez and Goldberg paper (Fernandez & Goldberg, 1971): 

\[
\frac{32.4s^2(1+0.063s)}{(1+32.4s)(1+5.7s)(1+0.003s)}
\]
The Visual System

The visual system is also an important component of standing balance. The threshold for detecting a stimulus indicative of sway is an order of magnitude smaller with visual cues, than with vestibular cues (Fitzpatrick & McCloskey, 1994). To stand, infants rely more on visual information than on any other sense (Lee & Aronson, 1974). An adult can stand without vision, but the lack of vision does increase sway (Edward, 1946). The body sway of an adult can also be controlled by modifying their visual surroundings (Lee & Lishman, 1975). The anatomy of the eye allows it to explicitly detect motion. Whereas the parvocellular cells of the eye detect colour and fine detail, the magnocellular cells detect outlines and motion (Livingstone & Hubel, 1988). The firing rate of magnocellular cells is affected by both the velocity of the perceived movement and its magnitude (Xu, Ichida, Boyd, B, & Casagrade, 2001). It is further hypothesized that there are two separate visual pathways in the brain, one for object recognition, and one for spatial relation (Goodale & Milner, 1992). Spatial relation is indicative of body position, and is very relevant to balance. Both the parvocellular and magnocellular cells provide information to the lateral geniculate nucleus (Xu, Ichida, Boyd, B, & Casagrade, 2001), which is the nucleus of the thalamus that relates to vision. The thalamus and the cerebellum are likely sites of visual, vestibular, and proprioceptive cue integration (Cohen, 2013). This information is interpreted to provide information about the position of the head and body in space. Spatial relation of the head to objects in space is useful in inferring where the body is in space, and how the body is balancing.

The Somatosensory System

A third system useful for balance is the somatosensory system. Cues from lower-limb afferents alone has been shown to be sufficient to balance an inverted pendulum of similar dynamics to a
human body (Fitzpatrick, Rogers, & McCloskey, 1994). Somatosensory information originates from cutaneous and joint receptors, muscle spindles as well as Golgi Tendon Organs (GTOs). Cutaneous receptors, specifically those that deal with touch and pressure under the feet can provide information about centre of pressure, which relates to centre of mass (Morasso, Spada, & Capra, 1999), and therefore is very relevant to body position and standing balance.

There are four types of cutaneous receptors. For small precise detection of objects, there are Fast Adapting type 1 (FA1) receptors, which detect rate of change of pressure with respect to time, and Slowly Adapting type 1 (SA1) receptors, which detect both pressure and the change in pressure with respect to time. For large detection of pressure without precise borders, there are Fast Adapting type 2 (FA2) receptors, which detect acceleration of skin indentation, and Slowly Adapting type 2 receptors (SA2), which indicate skin stretch. Cutaneous receptors under the foot can give information about the pressure changes that result from body sway, and from plantarflexion and dorsiflexion (Kennedy & Inglis, 2002). Cutaneous receptors located in the skin around the ankle can also code for skin stretch, which may reflect ankle movement (Edin & Johansson, 1995). Stretching of the skin on the back of the ankle, coupled with a slackening of the skin on the front of the ankle likely indicates a sway in the forward direction, while slackening of the skin on the back of the ankle, coupled with a stretching of the skin on the front of the ankle joint can indicate a posterior sway.

Joint capsules contain 4 different types of joint receptors: low threshold/slowly adapting, low threshold/rapidly adapting, high threshold/slowly adapting, and nociceptors (labelled type I-IV, respectively). Joint capsules are mainly indicative of motion when the joint is at an extreme angle for its range of motion (Burgess & Clark, 1969). The joint capsule at the ankle detects deformations in the ankle joint, as well as motion of the joint. The motion detected is ambiguous, because it does not indicate in which direction the joint has moved. During human standing balance, joint receptors at the ankle may indicate ankle motion during both plantar flexion and
dorsiflexion. Since joint capsules are ambiguous in regards to direction of movement, and since the motion of the ankle joint during standing is not extreme it is unlikely that joint capsules play a large role in standing balance.

Muscle spindles typically give an indication of muscle length or rate of length change (Matthews, 1963). They are located within the whole muscle, parallel to muscle fibres, and are composed of multiple intrafusal muscle fibers. The intrafusal fibers of the muscle spindle are divided into three types: nuclear bag₁ fibers, nuclear bag₂ fibers, and chain fibers. Type II afferents are connected to nuclear bag₁ fibers and chain fibers. They provide information on the length of the muscle (Boyd, 1980). Ia afferents, which connect to all three of the intrafusal fibers, give information regarding the rate of change of the muscle, as well as the length (Boyd, 1980). During standing balance, the plantar flexor muscle-tendon unit (gastrocnemius and soleus) is lengthening while the muscle is contracting, and shortening when the muscle is relaxing (Loram, Maganaris, & Lakie, 2004). This is different from a muscle-tendon unit during a typical contraction which shortens when the muscle contracts, and lengthens when the muscle relaxes, and is due to the elasticity of the tendon. It has been proposed that the muscles activated during balance tighten the muscle-tendon unit, and the tension is used to compensate for the gravitational toppling torque (Loram, Maganaris, & Lakie, 2004). This behaviour implies that the muscle spindles in the gastrocnemius and soleus may indicate muscle activity, and not body position (Loram, Maganaris, & Lakie, 2008). The muscle spindles in the antagonistic muscle, the tibialis anterior, are likely correlated better with center of gravity of a balancing human, and might instead be used as an indication of body position in space (Di Giulio, Maganaris, Baltzopoulou, & Loram, 2009; Day, Lichtwark, & Cresswell, 2013).

GTOs are capsules that exist between muscle fibers and the tendon. Inside of the capsule, collagen fibers are interwoven with axons from a sensory neuron. When the muscle is contracted, the pressure on the sensory axon from the collagen fibers is modified. The pressure on the
sensory axon is an afferent signal indicative of the active force produced by a muscle (Stephens, Reinking, & Stuart, 1975). Since active force is an indication of muscle activity, GTOs likely provide indirect information about body position. They could be useful when confirming that the efferent muscle signal during balance is producing the appropriate amount of force, but will not indicate body position.

**Studying Balance**

To study how information from the visual, vestibular and somatosensory systems combine to influence balance, experiments are often performed while information from one or more of these systems is modified, or removed (Britton, Day, Rothwell, Thompson, & Marsden, 1993; Fitzpatrick, Rogers, & McCloskey, 1994; Day & Gueraz, 2007; Law, 2011; Lee & Lishman, 1975; Lee & Aronson, 1974; Luu, et al., 2012). To activate the peripheral vestibular apparatuses, the vestibular apparatus would need to be moved through space. This would require pushing or pulling on the head of a subject, and would also stimulate the somatosensory system. Because of this, it is difficult to independently activate the peripheral vestibular apparatus. One way around this problem is Galvanic Vestibular Stimulation (GVS).

**GVS**

GVS is often used to probe the balance system in humans (Fitzpatrick & Day, 2004). It is a method of artificially stimulating the vestibular system using a current that modifies the firing rate of the nerves connecting the vestibular apparatuses to the vestibular nuclei (Goldberg, Smith, & Fernandez, 1984). GVS is typically applied with electrodes positioned on the mastoid processes. The current applied through the electrodes increases or decreases the resting potential of the underlying vestibular nerves (Goldberg, Fernandez, & Smith, 1982), and respectively increases or decreases the firing rate of the regular and irregular primary afferents (Fitzpatrick & Day, 2004). Both regular and irregular afferents connect to vestibulospinal and vestibuloocular
centres; however the irregular contribute more to the vestibulospinal centres, while the regular contribute more to the vestibuloocular centres (Goldberg, Smith, & Fernandez, 1984). The primary afferents come from the otoliths and semicircular canals, and project to the vestibular nucleus. In binaural bipolar stimulation, the anode is associated with a decrease in firing rate of the vestibular afferents, and the cathode with an increase. The regular afferents are activated, but do not modulate as much as the irregular afferents. The simultaneous increase in firing rate of the vestibular afferents on the cathodal side, and decrease in firing rate of the afferents on the anodal side is erroneously interpreted by the vestibular system as a signal from the vestibular apparatuses that the semicircular canals and otoliths on one side of the head are more active than normal, while those on the other side are less active than normal. This non-physiological signal is interpreted by the head as an angular movement towards the cathode (Fitzpatrick & Day, 2004). More specifically, the rotation occurs about an axis that is in the sagittal plane (see figure 3), and is approximately 18 degrees above Reid’s plane (Day & Fitzpatrick, 2005). A slight variation on GVS is Stochastic Vestibular Stimulation (SVS), which consists of applying a stochastic vestibular stimulation signal composed of many randomly changing frequencies and amplitudes. When looking for signal responses to GVS in the body, cross-correlation of a SVS signal to the EMG signal in appendicular balancing muscles can be used to approximate the average GVS response of multiple square waves, but is reported as less irritating to subjects, and takes less time (Dakin, Lee Son, Inglis, & Blouin, 2007).
The Vestibular Reflex Response

GVS applied to volunteers maintaining standing balance elicits a response in leg muscles (Fitzpatrick, Burke, & Gandevia, 1994; Nashner, 1973; Britton, Day, Rothwell, Thompson, & Marsden, 1993). This response can also be observed using time and frequency correlations of surface EMG recordings of the muscle and the GVS signal (Dakin, Lee Son, Inglis, & Blouin, 2007). The EMG response to GVS is present during balancing tasks, such as standing unbraced, and absent during non-balancing tasks (Britton, Day, Rothwell, Thompson, & Marsden, 1993), such as being braced upright, and performing an isometric contraction with the legs. The response is present when a subject is balancing their own body, absent when they are braced in a rigid position, and is independent of being conscious of balancing (Luu, et al., 2012). Tasks simulating the control of standing, such as being braced upright and controlling an inverted pendulum with ankle torque also fail to show a response (Fitzpatrick, Burke, & Gandevia, 1994). The unconscious presence of this response exclusively during balance makes it an ideal indication

Figure 3: GVS vector of rotation

The vector of rotation is ~18° above Reid’s plane, and causes the subject to feel as if they are rotating towards the cathode – in this case, the right. Image courtesy of Dr. Ryan Peters.
of a vestibular control of standing balance. Given its dependency on a postural task, this reflex response will be used throughout the rest of this proposal as an indication of a vestibular control of balance, as it gives an indication of whether an induced vestibular error is corrected by the balancing muscles.

The vestibular reflex requires sensory and motor signals associated with standing balance to be congruent in order to observe it in human lower leg muscles (Luu, et al., 2012). There are three sources of afferent information that were congruent during the trials by Lou et al (2012) that evoked the vestibular response: visual, vestibular, and somatosensory. It is not clear which combination(s) of these sensory information need(s) to be congruent with the motor signals to engage the vestibular control of balance. Sensory information from all of these sources project to the vestibular nucleus (Barmack, 2003) and could therefore be key in eliciting the response.

While lower-limb afferents are sufficient to control an inverted pendulum with similar dynamics to a human (Fitzpatrick, Rogers, & McCloskey, 1994), they are unable to independently engage the vestibular control of balance, but do engage the control when combined with vestibular cues of self-motion (Fitzpatrick, Burke, & Gandevia, 1994). This could imply that vestibular cues of self-motion are required to elicit a vestibular reflex. I will now present evidence that vision and neck somatosensory information might also be able to engage the vestibular control of standing.

**The Visual System and the Vestibular Control of Standing**

The visual system can cause circular and linear vection from optokinetic signals (Goldberg, et al., 2012). This implies a possible link between the visual system and vestibular nuclei. In both goldfish and rhesus monkeys, vestibular nuclei can respond to optokinetic input. In the case of the monkeys, this is true even without the vestibular apparatuses experiencing a motion stimulus. In one study, goldfish were attached to a rotating surface, surrounded by a rotating background, with recording electrodes inserted into their vestibular nuclei (Dishgans, Schmidth, & Graf, 1973). Rotating the fish in the dark caused an increase in the firing rate of the vestibular nuclei.
during acceleration, which then decreased back to the baseline during constant velocity.

Performing the same rotation with light and a stationary background caused the firing rate of the vestibular nuclei to increase during acceleration, then to stay constant during constant velocity. Keeping the fish stationary, and rotating the visual background in the opposite direction (so the vision cues are the same as if the background were stationary, and the fish were spinning) indicated that visual information without the presence of vestibular information was not informative, as the firing rates of the vestibular nuclei did not reflect the motion of the visual field. A similar experiment was done on non-human primates (Waespe & Henn, 1977). Six macaca mulatta had electrodes inserted to record their vestibular nuclei, and were placed on a rotating platform, with a visual backdrop that could be rotated around the non-human primate, and the three conditions from the goldfish experiment were performed. Throughout all trials, the non-human primates were alert and able to move their eyes. The first two conditions were very similar to those reported in goldfish. Rotating in the dark yielded an increase in firing rate in the vestibular nuclei during acceleration, and a return to base level during constant velocity. Rotating in the light with a stationary background yielded an increase in firing rate of the vestibular nuclei during acceleration, which maintained during constant velocity. The third condition, rotating the visual scene while keeping the non-human primate stationary, showed the firing rate of the vestibular nuclei increased during acceleration of the visual scene and remained high during a constant velocity rotation of the scene, which contrasts with the goldfish experiment from above.

It seems the firing rates of the vestibular nuclei in non-human primates can be influenced by visual information of self-motion, even in the absence of vestibular information. A change in the levels of visual cues has also been shown to modify the responses of the human body to GVS during balance – removing visual cues increases the magnitude of the vestibular response to GVS (Day & Guerraz, 2007). This may result from the influence of a higher level brain function on the vestibular nucleus, as visual and vestibular cortices have an inhibitory relationship with each other; an increase of activity in one corresponds with a decrease of activity of the other (Brandt,
et al., 2002). Other studies have shown similarly that a vestibular reflex occurs even when the eyes are closed, but that the response is larger (Fitzpatrick, Burke, & Gandevia, 1994; Britton, et al., 1993). This implies that vision is not required to elicit the response, but it does not clarify if it is possible for vision, in absence of vestibular cues of self-motion, to engage the vestibular control of standing balance.

**The Somatosensory System and the Vestibular Control of Standing**

The somatosensory system provides information about body position as well, and may also be sufficient to engage the vestibular control of standing. Neck somatosensory signals are also a possible contributor to a sense of head motion, and therefore may be able to elicit the vestibular response. Vestibular neurons in precollicular decerebrate cats have been shown to indicate motion of the head relative to the body (Boyle & Pompeiano, 1980). A similar experiment recorded vestibular neurons in squirrel monkeys during whole body rotation, neck rotation, and head-on-trunk rotation (Gdowski & McCrea, 2000). They found that neurons in the vestibular nuclei exhibited responses to head rotations as well as neck rotation. Roy and Cullen (2004) performed an experiment where they independently rotated the neck and body of monkeys, and showed that the vestibular nuclei could distinguish between voluntary and involuntary rotations of the head (Roy & Cullen, 2004). This demonstrated that neck somatosensory signals can activate the vestibular nuclei, and also shows that the vestibular system has a way of determining congruency between anticipated feedback for an intended action, and the actual feedback during that intended action. This ability to establish afferent vs. efferent signals could be crucial in detecting a congruency between the intended and perceived sway during balance.

As mentioned earlier, vestibular nuclei receive information from the vestibular apparatus, optokinetic cues, and neck somatosensory afferents, and cannot distinguish where the information came from (Barmack, 2003). We hypothesize that if correct feedback from vision, neck somatosensory, or the vestibular systems during a standing balance task is enough to elicit the
vestibular reflex, then correct feedback from any of those three senses during a standing balance task should be enough to engage the vestibular control of standing balance.

**The Next Step**

To determine if vision alone is enough to elicit a vestibular reflex response, the contribution of the peripheral vestibular apparatuses and somatosensory system to the vestibular system should be minimized. We have constructed an apparatus to help isolate the contribution of sensory information from the visual, vestibular and somatosensory systems in a balance-like setting. With this apparatus, we are able to independently simulate balance-related sensory feedback from the visual, vestibular, and lower-limb proprioceptive systems as if they were involved in maintaining standing balance. The device measures ankle torque to determine what angle the subject would be at if they were standing freely, then indicates that angle to them through one (or combination) of the aforementioned sensory cues.

We intend to determine whether a balance task providing visual feedback, but minimal feedback from the vestibular apparatus and somatosensory systems can engage the vestibular control of standing balance.

The first experimental condition will consist of balancing an inverted pendulum while experiencing visual cues of self-motion during a simulated standing balance task. The vestibular cues of self-motion will be kept to a minimum by keeping the head and body stationary in space. The somatosensory cues from the ankle motion will also be kept to a minimum by keeping the footplate and backboard fixed in space. It is expected that isolated visual cues of self-motion can elicit the vestibular response.

Initial pilot tests showed that balancing only visual cues of self-motion to be difficult, and to require conscious effort. We have added a second experimental condition -- with vision and ankle proprioception -- that is more intuitive, since the response that we are looking for occurs
subconsciously (Luu, et al., 2012), and a conscious effort during a difficult task may override any reflex response that is present.

The second experimental condition will consist of balancing an inverted pendulum while experiencing visual and lower-limb somatosensory cues of self-motion during a simulated standing balance task. The vestibular cues of self-motion will be kept to a minimum by keeping the head and body stationary in space. The ankle angle will change throughout this standing balance condition, providing the subject somatosensory cues from the lower limbs. Since we expect that vision alone can elicit the vestibular response, we expect vision with additional congruent somatosensory information to also elicit the response.

We hypothesize that if vision alone, or vision and ankle proprioception, is enough to engage the vestibular control of standing, then we will see a vestibular-evoked muscle response during simulated balance control while all other senses are minimized. We expect this response to be present in the condition with vision and in the condition with vision and ankle proprioception.
Statement of Question

Can visual cues of self-motion congruent with the efferent motor signals related to standing balance engage the vestibular control of standing if all other senses are kept to a minimum?

Can visual and lower-limb somatosensory cues of self-motion congruent with the efferent motor signals related to standing balance engage the vestibular control of standing if all other senses are kept to a minimum?
Methodology

Figure 4: Apparatus

Experimental setup. The subject is strapped to the backboard with two straps – one across the waist, and one across the chest. They are standing on the force plate, and facing the screen. Not visible are the electrodes on the back of the right calf. GVS electrodes are placed on the mastoid processes. 3D goggles are worn, with fabric to block out peripheral vision. A headband is worn, with a laser pointer attached. This allows the experimenter to monitor the pitch angle of the head throughout the experiment. Not shown are the earplugs and noise cancelling headphones that were also worn.

A- The force plate is fixed to the footplate, which is rotated around the subject’s ankles by a motor.

B- The backboard is independently rotated around the subject’s ankles by a second rotary motor.

C- Unprocessed SVS and medial gastrocnemius surface EMG signals
Subjects

8 healthy male subjects aged 26.1 ± 3.8 (µ ± SD) years, with mean weight 81.2 kg (S.D. 14.1 kg) participated in the present experiments. Subjects were screened to ensure they had no history of musculoskeletal or neurological disorders, balance disorders, or hearing deficits. Exclusion criteria also included frequent or severe headaches, pregnancy, respiratory diseases, cardiovascular diseases, or hearing deficits. Subjects were also screened based on the inability to give informed consent, inability to communicate in English, and for the taking of any medications that affect any of the above or whose side-effects include dizziness, lack of motor control, or slowed reaction time. All subjects gave written informed consent prior to participation. The experiments were approved by the Clinical Research Ethics Board of the University of British Columbia (H13-01951).

Experimental Set-up

A robotic balancing apparatus was designed and built for this experiment (see figure 4: Apparatus). The device was programmed with a model of quiet standing balance based on the work of Luu et al. (Luu, Huryn, Van der Loos, Croft, & Blouin, 2011). The model simulated an inverted pendulum with a distributed mass, as opposed to a point mass, since this has been found to better approximate the load of a standing human (Luu, Huryn, Van der Loos, Croft, & Blouin, 2011). Each simulation used the subject’s mass and height of the centre of mass to simulate the load they would experience during standing balance. The mass of the subject was measured with the force plate, and was determined from the force in the z direction (see figure 4, A). Subject 4 was erroneously given a simulation with a mass of 70 kg, instead of their true mass of 90 kg. The centre of mass of the subjects was approximated with the aid of a board and piece of dowelling (28.5 millimetres in diameter). The subject lay supine on the board, which was resting on the dowelling. The subject and board were then rolled back and forth on top of the dowelling until the

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2 For more information on the model, see APPENDIX B: Robotic Balance Simulator, Transfer Function
system balanced. The horizontal distance from the dowelling to the ankles was then used as the CoM. Subjects applied positive moments to the force plate through plantar flexion and negative moments through dorsiflexion\(^3\) (see figure 4, A). These moments were then used to update the model of the inverted pendulum through a transfer function described by Luu et al. (Luu, Huryn, Van der Loos, Croft, & Blouin, 2011).

**Footplate, Backboard, and Control**

The apparatus was controlled by a real-time system (PXI-8108; National Instruments, TX, USA) running at 2000Hz. The simulations were controlled by a subject’s ankle torque, as measured by a force plate (OR6-7-1000; AMTI, MA, USA) attached to the footplate (see figure 4, C). Force plate data were amplified by a factor of 4000 (MSA-6; AMTI, Watertown, MA, USA), before being acquired at 2000Hz (PXI-6289, National Instruments, TX, USA). Compensation for torques due to movements of the footplate, and not ankle torque, was performed as per Huryn et al. (Huryn, Luu, Van der Loos, & Blouin, 2010).\(^4\) As the simulation ran, the apparatus updated the angle of the inverted pendulum presented to the subject through motion of the footplate, motion of the backboard, visual cues or a combination of these inputs. The footplate was rotated about the subject’s ankles by a rotary motor (SCMCS-2ZN3A-YA21, Yaskawa, Japan). The range of motion of the footplate was limited to ±19.5 degrees from horizontal with hard limits. The backboard consists of two parts: the board, and the metal frame. The board was adjusted relative to the frame to account for the subject’s natural standing angle. The backboard was rotated about the subject’s ankles by a motor (SCMCS-2ZN3A-YA21, Yaskawa, Japan). The range of motion was limited to ±10.5 degrees from vertical with hard limits. Both motors had a resolution of 1048576 encoder readings per revolution (i.e. 0.00034 degree). Soft motion limits of 50 degrees/s and 1000 degrees/s\(^2\) were programmed to encompass the physical limits of sway during standing balance (Pospisil, Luu, Blouin, 2011).

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\(^3\) See APPENDIX B: Robotic Balance Simulator, Ankle Torque and Force Plate Compensation

\(^4\) See APPENDIX B: Robotic Balance Simulator, Ankle Torque and Force Plate Compensation
Van der Loos, & Croft, 2012). The delay between a motion command, and the motor responses was found to be approximately 20ms\(^5\).

**Visual Display and Control**

The projection screen was located 0.77 metres to the left of the subject. The screen rested 0.18 metres above the ground, and was 2.4 metres tall. It was a 55.13 degree arc of a 4.36 metre radius horizontal circle. A projector (W1080ST; BenQ, Taipei, Taiwan) was located 3m behind the screen at ground level(resolution of 1920x1080p). Subjects wore active 3D glasses (DLP Link 3D Glasses; BenQ, Taipei, Taiwan), modified to block out peripheral vision and limit the subject’s field of view to approximately ± 45 degrees horizontally and ± 30 degrees vertically. The visual scene was controlled by a desktop computer with a dedicated graphics card (Quadro K4000, Nvidia, CA, USA), which received commands and angle updates from the real time system over a network connection. The projection was created so that when viewed from the subject's height and eye location, the visual cues simulated those received when standing in a courtyard, next to a fountain\(^6\) (Vizard 2013\(^7\); WorldViz, CA, USA). Rendering and projection of the visual scene took approximately 70ms, which was longer than the response time of the motors. A linear least-squares predictor algorithm was used to predict the visual angle by 50ms, and synchronize the visual motion with the motors at a delay of 20ms.\(^8\)

**Electrical Vestibular Stimulation**

For electrical vestibular stimulation, two carbon rubber electrodes (9cm\(^2\)), coated in Spectra 360 electrode gel (Parker Laboratories, NJ, USA) were placed on the mastoid processes behind each ear of the subject, and secured with tape. Binaural bipolar stimulation was used. For all experiments, a stochastic vestibular signal with a bandwidth of 0-25Hz and amplitude of ±5mA(root mean square of 1.24mA) was used. The

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5 See APPENDIX B: Robotic Balance Simulator, Motor Delay
6 See APPENDIX B: Robotic Balance Simulator, Frustum and Notional Camera
7 Vizard is a python IDE and wrapper for openGL
8 See APPENDIX B: Robotic Balance Simulator, Visual Delay
electrical signal was generated offline and all subjects received the same signal. It had a temporal resolution of 2000Hz and was delivered to the subject through an isolated constant current stimulator (DS5, Digitimer, Hertfordshire, England). By convention, a positive value of electrical vestibular stimulus corresponds to the electrode behind the right ear being the anode, and the electrode behind the left ear being the cathode.

Data Acquisition

EMG was recorded differentially from the right medial gastrocnemius using Ag/AgCl surface electrodes (Blue Sensor M, M-00-S/50; Ambu A/S Ballerup, Denmark), amplified by 5000 and filtered between 30 and 1000Hz (NL 844 preAmp and NL 136 Low Pass Filter; Neurolog, Welwyn Garden City, England). All force plate, EMG, and electrical vestibular stimulus data were lowpass filtered at 40 kHz prior to being digitized at 2000Hz (PXI-6289; National Instruments, TX, USA).
Experimental Protocol

The control condition with vestibular, visual and ankle proprioception (A): the footplate is stationary, the backboard moves, and the head is moving relative to the visual scene. This results in a balancing task that provides visual, vestibular and somatosensory feedback. The control condition with only ankle proprioception (B): the backboard is stationary, the head is stationary relative to the visual scene, and the footplate moves. This results in a balancing task that only provides relevant ankle somatosensory cues. The experimental condition with only vision (C): the backboard and footplate remain stationary, and the visual scene is moving relative to the head. This results in a balancing task that only provides relevant visual cues. The experimental condition with vision, and ankle proprioception, but no vestibular cues (D): The backboard remains stationary, the footplate moves, and the visual scene moves relative to the head. This provides a balancing task that provides relevant visual and somatosensory cues.
All experimental conditions consisted of one 120 second trial. For each subject, all experimental conditions were performed on the same day and were randomized. One subject had difficulty performing the balance task during the majority of the conditions. He reported that the task did not initially feel intuitive and as a result they spent a large portion of the condition with visual and proprioceptive cues on the limits of the robotic device. The subject was brought back on a different day to balance the robot until it felt intuitive. This process took about 15 minutes. The subject was then retested on a subsequent day. After the second testing, the subject reported that the process felt easier, and he stayed within the balance limits for that trial. Only data from the second day was used in the analysis. Both sets of data from this subject are presented for comparison in Appendix A: Experimental Data, Vestibular Responses. Subjects were not informed of what order the conditions would be given and most were naïve as to which combination of vision, footplate and backboard would move for any condition. For all conditions, subjects stood naturally, without shoes, on the force plate. Subjects were then strapped to the backboard (see figure 4, B) using two seatbelts: one across the shoulders and one around the waist. While at this natural standing angle, the force plate was zeroed. Subjects wore active 3D glasses and viewed a 3D projection, 90 degrees over their left shoulder, with their head tilted 18 degrees up from horizontal (see figure 4). This head position was to maximize the response from electrical vestibular stimulation in ankle plantar flexor muscles during anterior-posterior sway (Day & Fitzpatrick, 2005; Cathers, Day, & Fitzpatrick, 2005). To ensure that their head stayed at this position (relative to their body), an active laser pointer was strapped to their head, and the resultant beam was observed by the experimenter. To minimise auditory cues that might give an indication to the subject of their position in space or movement, subjects wore earplugs (NRR32 - 32 decibels disposable earplugs, Stanley, CA, USA) and noise cancelling headphones (QC25 - QuietComfort 25 Acoustic Noise Cancelling Headphones, Bose, MA, USA) that delivered background noise of a gurgling water fountain.

Control Conditions

There were two controls conditions. Each consisted of a single 120 second trial.
The first control condition confirmed the presence of the vestibular reflex when quiet standing balance is simulated with realistic ankle somatosensory, vision, and vestibular cues, while still being on the robot. To provide a realistic simulation of standing balance, the ankle platform remained stationary while the angle of the backboard was changed to the angle specified by the model of standing balance. The angle of the subject’s body was then determined by the transfer function used in the model. Visual cues indicative of motion were provided to the subject. This setup was intended to show that it was possible to elicit a vestibular reflex on the device by giving appropriate visual, vestibular and somatosensory cues, as previously shown by Luu et al. (Luu, et al., 2012).

The second control checked the absence of the vestibular reflex during a balance-like task not involving the vestibular control of standing, similar to Fitzpatrick et al. (Fitzpatrick, Burke, & Gandevia, 1994). The backboard remained stationary to keep the vestibular cues constant and no visual cues of self-motion were provided to the subject. The ankle-tilt platform was rotated in the opposite direction of the predicted standing balance model’s angle. This simulated the ankle angle that would be experienced during quiet standing balance while keeping vision and vestibular cues constant. This control condition was intended to show that a vestibular reflex is not evoked on the device in a task that is not expected to elicit a vestibular reflex, as described by Fitzpatrick et al. (Fitzpatrick, Burke, & Gandevia, 1994).

**Experimental Conditions**

The first experimental condition, vision alone, examined the presence of a vestibular reflex when only visual cues of balancing were available to the subject. The backboard and ankle tilt platform remained stationary to keep the vestibular and ankle angle cues constant. Since both the backboard and footplate were kept fixed in space, the angle between the feet and the legs was also

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9 See APPENDIX B: Robotic Balance Simulator, Frustum and Notional Camera
10 See APPENDIX B: Robotic Balance Simulator, Frustum and Notional Camera
kept constant and the somatosensory cues from the lower limbs were kept to a minimum. The cues from pressure receptors in the foot and tendon/muscle stretching in the lower leg were assumed to be too small to trigger the response, since the second control showed that large coherent cues to those systems could not independently trigger the response. Visual cues indicative of self-motion were presented to the subject.\textsuperscript{11} Because of the strong link between vision and the vestibular system (Waespe & Henn, 1977), we expected subjects to show a vestibular reflex in this condition.

The second experimental condition, vision and ankle proprioception, was performed to determine the presence of a vestibular reflex when visual and ankle cues of balancing were available to the subject. Pilot testing revealed the first experimental condition to be difficult and potentially relied on conscious control. This condition -- vision and lower limb somatosensory cues -- allowed for a more natural (automatic) balancing task. The backboard remained stationary to keep the vestibular cues constant but the ankle-tilt platform was rotated in the opposite direction of the predicted standing balance model’s angle. This mimicked the ankle angle and visual cues of self-motion\textsuperscript{12} that would be experienced during quiet standing balance. Because there is a strong link between vision and the vestibular system (Goldberg, et al., 2012), and because this condition appeared easier to control than the previous case, we expected a vestibular reflex for this condition.

Throughout all conditions, the backboard, even when ‘stationary’, were being held in space by motors. To ensure that the backboard was stationary, the encoder of the backboard motor was tracked, and the RMS of the backboard’s actual motion, as well as the RMS of the angle in the balance simulation was computed for all trials. To compare the two, one was divided by the other. To check if the RMS of the

\textsuperscript{11} See APPENDIX B: Robotic Balance Simulator, Frustum and Notional Camera
\textsuperscript{12} See APPENDIX B: Robotic Balance Simulator, Frustum and Notional Camera
backboard was providing vestibular or proprioceptive cues relative to the balancing task, a correlation was run between the angle of the simulation, and the actual position of the backboard.

**Data Analysis**

After acquisition, all EMG data were highpass filtered at 10Hz (zero-phase shift, second order Butterworth filter) and rectified (Dakin, Dalton, Luu, & Blouin, 2014).

To check for a vestibular reflex, the delivered electrical vestibular signal was correlated to the recorded EMG signal using two methods: coherence and scaled cross-correlation. Both of these measures were performed in Fourier space. These measures were computed individually for each condition, for each subject. All trials for all subjects were the same length, and were recorded at the same rate. They were each independently converted to Fourier space with a windows size of $2^{11} = 2048$ (~1 second), yielding a resolution of ~0.98hz across 117 windows.

Coherence (or magnitude squared coherence) gives a measure of a linear relationship (across frequencies) between two signals. Its values range from zero (indicating no correlation between the two signals) to one (indicating complete linear relationship between the two signals). A higher value of coherence corresponds to a stronger likelihood of a linear relationship between the two signals (Halliday, et al., 1995). Confidence limits were calculated from the number of windows as per Halliday et al (Halliday, et al., 1995), and the 95% value was 0.0255.

Scaled cross-correlation, sometimes called scaled cumulant density, gives the correlation coefficient of two signals as a function of potential time delays between the signals (Halliday, et al., 1995; Dakin, Luu, van den Doel, Inglis, & Blouin, 2010). In this case, we computed the scaled cross-correlation between the filtered, rectified EMG, and the SVS signal. Confidence limits were set at 95% and were calculated using the input data and the number of samples, and were scaled by the norms of the EMG and GVS signals.
To be considered a vestibular reflex, a relationship must have existed in both the coherence and the cross-correlation above the 95% confidence interval. In the cross-correlation, there needed to be a significant positive short and a negative medium latency peaks (around 60 and 100 milliseconds, respectively) (Dakin, Lee Son, Inglis, & Blouin, 2007; Luu, et al., 2012) that surpassed the 95% confidence intervals. This is similar to the behaviour of the cumulant density in previous studies that has been indicative of that the vestibular control of standing was engaged (Dakin, Lee Son, Inglis, & Blouin, 2007; Luu, et al., 2012). The presence of a vestibular reflex was determined using the latencies and peak-to-peak values of the short and medium latency peaks, and was computed on a subject-by-subject basis.

After the coherence and cumulant density for each subject were computed, a one-way repeated measures ANOVA was performed on the peak-to-peak amplitude of the cumulant densities. For any condition that yielded no statistically significant peaks at the expected times for the short and medium latency responses, a value of zero was used for the peak-to-peak amplitude while performing the ANOVA. If the repeated measures ANOVA revealed a main effect (p<0.05), its decomposition was performed using a Tukey HSD test to determine which conditions exhibited a significant difference from the control with only ankle somatosensory cues, which in previous work has been taken to show a lack of vestibular reflex (Fitzpatrick, Burke, & Gandevia, 1994).

All descriptive statistics in this work will be presented as mean plus/minus the standard deviation.
Results

Results shown for all conditions: visual, vestibular, proprioceptive cues (VVP); proprioceptive cues (P); visual cues (V); visual and proprioceptive cues (VP). Horizontal lines show 95% confidence intervals. Vertical lines show timing of short and medium latency peaks in the first control (visual, vestibular, and proprioceptive cues). Note the statistically significant peaks in the cumulant density for the first and last rows. The timings and size of the peaks indicate a vestibular reflex response.

Data for a typical subject are illustrated in figure 6. The peak-to-peak amplitude of the cumulant density’s short and medium latency peaks (58ms and 92.5ms) was 0.091 in the control condition (vision, vestibular and ankle proprioceptive) and was not different in magnitude to the 0.086 peak-to-peak amplitude of the short and medium latency peaks (54.5ms and 90ms) of the cumulant density in the
condition with only vision and ankle proprioception. The latencies of the peaks were indicative of a vestibular control of standing as described in previous studies (Dakin, Lee Son, Inglis, & Blouin, 2007; Luu, et al., 2012). For this subject, vision and proprioception condition elicited a vestibular reflex because the latency and amplitude of the vestibular-evoked response were similar to the control condition with all sensory feedback. No vestibular response was observed for the control condition with only ankle somatosensory feedback. The condition with vision feedback alone did exhibit peaks at the appropriate times (55ms and 94.5ms), but the magnitude of the peak-to-peak amplitude was less than a third of the peak-to-peak amplitude of the control with visual, vestibular and somatosensory feedback (0.028 vs. 0.091).
Figure 7: Average across subjects

Grey lines indicate an individual subject, dark black lines indicate an average of all subjects (n=8).

The dotted horizontal line in the coherence indicates the 95% confidence interval for any single subject. It does not apply to the average. The 95% confidence interval for the cumulant density is unique for each subject, and is therefore not presented.
Figure 8: Group means

Means and standard deviations of peak-to-peak values for all subjects, for the four conditions: proprioceptive (P); visual (V); visual and proprioceptive (VP); and vision, vestibular and proprioceptive cues (VVP). ‘*’ Represents a statistically significant p value (p<0.005) from the post-hoc Tukey HSD test.
Table 1: Peak-to-peak amplitudes of the four conditions

<table>
<thead>
<tr>
<th>Subject</th>
<th>VVP</th>
<th>P</th>
<th>V</th>
<th>VP</th>
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</thead>
<tbody>
<tr>
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<td>0.09</td>
<td>0.00</td>
<td>0.03</td>
<td>0.09</td>
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<tr>
<td>S02</td>
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Peak-to-peak amplitudes of the cumulant density for the short and medium latency peaks typical of a vestibular reflex. All conditions that did not have peaks crossing the 95% confidence interval at the appropriate time were given a value of 0. Values shown for: vision, vestibular and proprioceptive cues (VVP); proprioceptive cues (P); visual cues (V); visual, and proprioceptive cues (VP). A vestibular reflex was identified in the trials that are highlighted in grey.

These observations were typical for most subjects (see figure 7 and table 1). In the control condition with feedback from vision, vestibular & ankle proprioception, all subjects exhibited a positive short latency peak at 61.9 ± 4.4ms, and a negative medium latency peak at 99.2 ± 7.4ms, both of which crossed their respective 95% confidence intervals. The average peak-to-peak amplitudes are presented in figure 8 (0.13 ± 0.05,n=8). All but one subject exhibited statistically significant peak-to-peak amplitudes for the condition with vision and proprioception (0.13 ± 0.05,n=7), two of the subjects exhibited responses in the condition with proprioceptive cues alone (0.10 ± 0.002, n=2), and one of those subjects had a comparable peak-to-peak amplitude in the condition with only vision (0.084). The one-way repeated measures ANOVA revealed a main effect of Condition for the peak-to-peak amplitude of the vestibular-evoked responses (F_{3,21}=13.304; p=0.00004). Compared to proprioception alone, the condition with visual, proprioceptive, and vestibular cues, and the condition with vision, and proprioception showed statistically
significant values ($p = 0.0044$ and $0.0008$ respectively), while the condition with only vision did not ($p = 0.99$).

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<th>VP</th>
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Table 2: RMS of simulation throughout the four conditions

Vision, vestibular and proprioceptive cues (VVP); proprioceptive cues (P); visual cues* (V); visual, and proprioceptive cues (VP). * vision trials had subjects spending most of their time at the limits of the device, and very little time balancing in between limits. The value of the root mean square is presented in units of degrees. A vestibular reflex was found in the trials that are highlighted in grey.

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Table 3: RMS of backboard throughout the four conditions

Vision, vestibular and proprioceptive cues (VVP); proprioceptive cues (P); visual cues* (V); visual, and proprioceptive cues (VP). * vision trials had subjects spending most of their time at the limits of the device, and very little time balancing in between limits. The value of the root mean square is presented in units of degrees. A vestibular reflex was found in the trials that are highlighted in grey.
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Table 4: Correlation of backboard to virtual model throughout the four conditions

Vision, vestibular and proprioceptive cues (VVP); proprioceptive cues (P); visual cues (V); visual, and proprioceptive cues (VP). A vestibular reflex was found in the trials that are highlighted in grey.

The ratio of the RMS of the backboard’s movement in the trials where it was assumed to be stationary compared to the RMS of the simulation’s movement was 0.03±0.015 (n=24), compared with 1.02±0.015 (n=8) in the trials where the backboard was intended to move (see table 2 and table 3). The RMS of the backboard’s motion divided by the RMS of the motion of the simulation in the condition with only vision and lower-limb proprioception (which elicited a vestibular response) was 0.028±0.016. The movement of the backboard was highly correlated with the movement of the simulation during the condition with all sensory cues, where the backboard was supposed to move ($\bar{r}_{VVP}=1.00 \pm 0.0$, see table 4). The three remaining trials also showed good correlation ($\bar{r}_P=0.93\pm0.03, \bar{r}_V=0.55\pm0.19, \bar{r}_{VP}=0.90\pm0.08$).
Discussion

The present work supports the idea that the vestibular control of standing balance can be engaged with minimal vestibular cues of self-motion. This is exemplified by the presence of a vestibular reflex during the balancing trial with vision and ankle proprioception, but limited vestibular cues. Since the vestibular reflex has already been shown during balancing tasks that used vestibular cues of motion on similar robots (Luu, et al., 2012), and since the secondary vestibular neurons are unable to distinguish between optokinetic and vestibular cues of motion (Barmack, 2003), this was an expected result. The same reasoning would imply that the trial with vision alone should also elicit a vestibular reflex; however, that trial did not elicit a vestibular reflex. As a result, this work is unable to show support that visual cues of self-motion congruent with the efferent motor signals related to standing balance can engage the vestibular control of standing if all other sensory cues are kept to a minimum.

Subjects showed a vestibular reflex during the trial with vision and lower-limb proprioception. This goes against the claim by Fitzpatrick et al. (Fitzpatrick, Burke, & Gandevia, 1994), that the presence of the response is dependent on the task, i.e. and the need for vestibular information to be relevant to the control of standing. In this experiment minimal information was provided by the vestibular apparatuses, yet the gain of the response was not different in amplitude to the control condition (where the vestibular apparatus was providing information relevant to the control of standing). Based on this claim, we would expect a negligible response to a vestibular error signal, in a task where the vestibular apparatuses are not being used. Fitzpatrick et al. (Fitzpatrick, Burke, & Gandevia, 1994), however, did not test a balance condition with stationary vestibular cues and congruent visual cues of self-motion. Animal literature would suggest that since both vision and the vestibular afferents feed into the vestibular nuclei (Barmack, 2003), reliance on one could positively influence the dependence on the other, which would explain the results that were found in this work. Subjects had visual and lower-limb somatosensory cues of motion that were congruent with efferent motor signals during a balancing task. Since the visual cues of motion were relevant to the task, the visual signals projecting to the vestibular nuclei potentially contributed to
the muscle activity required to balance. Since the vestibular nuclei are unable to distinguish between visual and vestibular cues of motion, a vestibular error signal (induced by the SVS) was corrected by the balance centres as if the vestibular apparatus had been giving congruent information to the balancing task. Only one subject did not exhibit a vestibular-evoked response to this condition. The subject did not exhibit as much EMG for the maintenance of standing as the other subjects (see figure 9). It looks as if for large segments of time, the medial gastrocnemius was not active. This is likely because the subject was balancing with their dorsiflexors (only the medial gastrocnemius was recorded). Without enough EMG, the data analysis would be heavily biased towards a null result. It is also possible that without enough activation of the muscle, the reflex does not occur.
The lack of vestibular reflex during the condition with only vision is unexpected, but could be explained by the conscious effort required to balance the visual scene in that trial. All subjects commented that the condition was very difficult. All subjects spent the majority of their time at the physical limits of the device (88.35s / 120s ± 21.36s / 120s, n = 8\(^1\)), and little time maintaining a stable balance. The balancing that did occur during this trial required a conscious effort from the subject, and was not automatic. It is proposed that the conscious effort of balancing during this trial engaged higher level brain functions.

\(^1\) For a full list of the balancing times, see Appendix A: Experimental Data, Time in Limits.

Figure 9: Typical EMG and Anomalous EMG

Rectified and filtered EMG during the condition with Visual and Proprioceptive cues from a typical subject (above) and an atypical subject (below). Note that the atypical subject has large intervals with minimal EMG activity.
which overrode any potential control of standing that could occur in the brainstem. This explanation would also explain the results seen when retesting one of the subjects (Appendix A – S05). The subject initially exhibited no vestibular reflex for the condition that combined vision and ankle proprioception, and described the task as non-intuitive. The same subject, after familiarization of the trial over multiple days, described the same condition as much easier, and exhibited a vestibular reflex. It is possible that the vestibular reflex may only be obtainable during an intuitive balancing task that does not require conscious control of standing. This could explain why only one subject did show a response on the condition with only visual cues.

The explanation of why vision alone does not elicit a vestibular reflex is further reinforced by the majority of subjects (n=7) showing a vestibular reflex during the condition with vision and ankle proprioception. This trial was added as an alternative visual condition that was more intuitive. For most subjects, balancing with ankle proprioception alone was not enough to elicit response vestibular reflex, but all subjects were able to easily balance when given only ankle proprioceptive feedback. Combining ankle proprioception with visual cues allowed for an intuitive balancing task that provided visual feedback with no cues of balance from the peripheral vestibular apparatuses.

Two subjects showed a vestibular reflex for the condition with ankle proprioception alone. A vestibular reflex for a proprioception-only balancing task was not reported by Fitzpatrick et al. (Fitzpatrick, Burke, & Gandevia, 1994), but the data from the averaged EMG of their six subjects did show a very small indication of what might have been a small vestibular response in a few subjects. The new technique of SVS has a higher signal to noise ratio (Reynolds, 2011), and gives a confidence interval for each subject (Halliday, et al., 1995). It is possible that performing the same experiment, but using SVS instead of square wave GVS, might yield a measurable vestibular reflex in some of the subjects in the Fitzpatrick study.
One subject showed a response during all four of the trials. It is unlikely that this is due to any type of vestibular cues of motion. The subject was firmly strapped to the backboard, and for all subjects, the RMS of the backboard movement was the same order of magnitude.

**Limitations**

This experiment assumed the backboard was stationary, while it was being held by motors. The maximum angular velocity for the backboard when it was supposed to be stationary was 0.65 deg/s ± 0.45 deg/s (n = 24)\(^{14}\). This is below the angular velocity detection threshold of 2.07deg/s (Benson, Hutt, & Brown, 1989). The RMS value of the backboard’s motion when it was supposed to be stationary was 0.07 deg ± 0.03 deg (n=24). Throughout all trials, the backboard’s position had a positive correlation with the simulation; however, since the backboard’s movement was on average less than 3% of the motion experienced by the other sensory systems, and since the velocity of the movement was below previously reported detection thresholds, it is assumed that the motion-induced vestibular activity was negligible. In addition, the stationary trial that produced a vestibular reflex, had the smallest RMS of backboard motion. If the small amount of backboard motion the subject experienced was able to trigger a vestibular reflex, a reflex should also have been seen in the other two trials that had a stationary backboard (and larger RMS of movement), which is not the case.

The motors did have an audible noise that changed when they were moving, and might have given cues to subjects about whether the backboard was moving or not. Noise from the motors was minimised for the subjects by having them wear earplugs and noise cancelling headphones. No subjects commented on the noise of the motors. The presence or absence of a vestibular reflex are not directly indicative of the movement of the backboard, so it is assumed that the noise from the backboards did not affect the results.

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\(^{14}\) For a full list of the maximum velocities, see Appendix A: Experimental Data, Maximum and Average Velocities
The visual feedback from the subjects was projected onto a screen, and therefore had a somewhat pixelated resolution. It appeared to subjects as if they were looking through a screen door. Antialiasing was used to alleviate some of the pixelation. Since the magnocellular pathway is believed to be in charge of motion cues, and since the magnocellular cells are not sensitive to fine detail, the slight pixelation from a digital image were likely insignificant for this experiment.

Due to an error, subject 4 was balancing a mass of 70kg instead of their true mass of 90kg. This did not appear to affect their results. That subject followed the same behaviour as other subjects. This implies that the presence of a vestibular reflex may not need the feedback to follow the exact physics of the balanced body in the real world. It is possible, however, that this influenced the magnitude of the vestibular responses in all conditions.

**Future Directions**

To confirm if ease of balance is necessary for the vestibular reflex, the trial with vision alone could be performed again under easier conditions. The condition might be made easier by extensively training the subjects to balance the robot with only visual cues. Pilot attempts to train subjects to balance with vision alone showed this to be a difficult task. Even after multiple sessions practicing the trial, subjects were able to stay off the limits for at most one or two seconds, and found it difficult to do so. Another alternative to make this condition easier would be to change the dynamics of the model. Changing the coefficient of the damping term in the inverted pendulum transfer function equation could limit the angular velocity of the balancing model. This could prevent subjects from rapidly falling over, and would provide a more forgiving balancing task. Changing the dynamics of the model, or spending further time training may allow balancing with only visual cues to become more intuitive, and might show a response in future tests.

The current experiment investigated if vision was enough to engage a vestibular control of standing. An interesting complement would be to see if vision is enough to suppress a vestibular control of standing. A
proposed follow-up experiment would give vestibular and ankle somatosensory cues of motion while the visual cues of motion would be non-congruent with the motion. This experiment could be performed on the current device. The control for the experiment would include every sensory feedback giving congruent cues of motion; the backboard would move to give vestibular cues of motion, the footplate would stay stationary to give somatosensory cues of ankle motion, and vision would move relative to the head to give visual cues of motion. This is the same as one of the controls used for this experiment, and shows a vestibular reflex. The intervention would be very similar to the control, but would give non-congruent visual cues; the backboard would move (giving vestibular cues of motion), the footplate would be left stationary (giving somatosensory cues of ankle motion), and the visual scene would move in some non-congruent motion. In all cases, the focus of the visual scene would follow the subject’s head (so that the visual cues from the scene are immersive, and the subject is not reminded that they are looking at a projection on a screen), but the motion of the notional camera would not follow the movement of the subject. If the notional camera stays fixed, it would appear visually to the subject that they are not moving in space. Visually, this would be similar to the visual feedback from the stable vision condition of the Sensory Organization Test (Nashner, Black, & Wall III, 1982). This intervention would switch the role of the cues from the vestibular apparatus and those from vision in the current experiment. In the proposed intervention it would be the vestibular apparatus that is giving cues congruent with motion, while the visual inputs are giving cues of being stationary. Since the vestibular nuclei are unable to distinguish between these two inputs, the results from this proposed trial should be similar to the results from the condition in the current experiment with only vision and ankle proprioception, and a vestibular reflex would be expected. To shut off the reflex, it is possible that visual cues should not just indicate a lack of motion to the subject, but should indicate improper cues of self-motion. To accomplish this, a motion signal could be recorded from a balancing task, and in one intervention, this motion could be played to the subject while they are simultaneously balancing the backboard. Since the vestibular nuclei are unable to distinguish where the signal came from, it could cause the sum of the cues from vision and
the vestibular apparatuses to become incongruent from the efferent motor signal, and the vestibular reflex may be suppressed.

More follow up experiments could look at the role of neck somatosensory cues in engaging a vestibular control of standing. Neck somatosensory afferents also feed into the vestibular nuclei where it is integrated with visual cues of motion, and signals from the vestibular apparatuses (Boyle & Pompeiano, 1980; Gdowski & McCrea, 2000). These experiments could investigate the role of neck somatosensory cues in engaging the vestibular control of standing. Ideally, a new device would be constructed for this. The device would have the ability to keep the head stationary, and to rotate the body underneath the head. The rotation of the body would appear as a normal pendulum with a rotation point in the head. To provide accurate somatosensory cues of ankle motion, a second motor would rotate a footplate (and force plate) about the ankle. Because of this, the distance between the two rotation points would be different for any two subjects that have different heights. The device would use torque to move the body and feet in space. Since the head is stationary, there would be no vestibular cues of motion. A visual screen could provide stationary cues of motion to the visual system. This device would then use the subject’s plantar flexor torque and the same transfer functions presented here to move a subject’s body in space (providing congruent cues of motion to neck somatosensory cues), and also to move a subject’s feet relative to their body (providing congruent somatosensory cues of ankle motion). The presence or absence of a vestibular reflex would then be checked. Due to the integration of neck somatosensory cues shown in the vestibular nuclei, it is expected that a vestibular reflex would be observed. For a control, the same device would have to fix the body underneath the head, then use torque to move only the footplate. This would cause no vestibular cues of motion (since the head is fixed), no congruent neck somatosensory cues (since both the head and body and fixed), no visual cues of motion (since the visual system would stay stationary) but would give ankle somatosensory cues (since the feet are rotating around the ankle). This would be similar to the work of Fitzpatrick et al (Fitzpatrick, Burke, & Gandevia, 1994), as well as the ankle proprioception only trial here, and should yield no vestibular reflex.
Conclusion

Since there was a vestibular reflex present in the majority of the subjects during a trial with minimal vestibular cues, this study suggests congruent visual cues, lower-limb somatosensory cues, and motor signals engage the vestibular control of standing.

The lack of a vestibular reflex in the majority of the subjects during a trial with visual cues, but minimal vestibular and ankle somatosensory cues was inconclusive. It is possible that with proper training the balancing during this trial could be more intuitive, which may allow the vestibular reflex to occur.

These results imply that the vestibular control of standing balance can be engaged with minimal vestibular cues of self-motion, and could be beneficial to balance disorder research, and rehabilitation for clinical populations with balance deficits.
Bibliography


APPENDICES

Appendix A: Experimental Data

Vestibular Responses

Figure 10: Vestibular responses for subject 01
Figure 11: Vestibular responses for subject 02

Figure 12: Vestibular responses for subject 03
Figure 13: Vestibular responses for subject 04
Figure 14: Vestibular responses for subject 05

Data from first and second testing of Subject 5. The first set of data (grey lines) was discarded because the subject commented that the process was unintuitive, and was having difficulty balancing.
Figure 15: Vestibular responses for subject 06

Figure 16: Vestibular responses for subject 07
Figure 17: Vestibular responses for subject 08
### Time in Limits

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Table 5: Time in limits for condition with vestibular, visual, and proprioceptive cues

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Table 6: Time in limits for condition with proprioceptive cues

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<td>120.00</td>
<td>1.00</td>
</tr>
<tr>
<td>3</td>
<td>120.00</td>
<td>1.00</td>
</tr>
<tr>
<td>4</td>
<td>120.00</td>
<td>1.00</td>
</tr>
<tr>
<td>5</td>
<td>118.98</td>
<td>0.99</td>
</tr>
<tr>
<td>6</td>
<td>102.31</td>
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<td>0.85</td>
</tr>
<tr>
<td>8</td>
<td>120.00</td>
<td>1.00</td>
</tr>
</tbody>
</table>

Table 8: Time in limits for condition with visual, and proprioceptive cues

**Maximum and Average Velocities**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Max Velocity (deg/s)</th>
<th>Average Velocity (deg/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2.1E+00</td>
<td>-1.7E-02</td>
</tr>
<tr>
<td>2</td>
<td>1.3E+00</td>
<td>3.5E-03</td>
</tr>
<tr>
<td>3</td>
<td>7.6E-01</td>
<td>7.8E-04</td>
</tr>
<tr>
<td>4</td>
<td>8.9E-01</td>
<td>2.1E-03</td>
</tr>
<tr>
<td>5</td>
<td>1.6E+00</td>
<td>5.0E-03</td>
</tr>
<tr>
<td>6</td>
<td>2.0E+00</td>
<td>2.2E-02</td>
</tr>
<tr>
<td>7</td>
<td>2.1E+00</td>
<td>1.1E-02</td>
</tr>
<tr>
<td>8</td>
<td>9.7E-01</td>
<td>3.9E-03</td>
</tr>
</tbody>
</table>

Table 9: Velocities for condition with visual, vestibular, and proprioceptive cues

<table>
<thead>
<tr>
<th>Subject</th>
<th>Max Velocity (deg/s)</th>
<th>Average Velocity (deg/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>3.3E-01</td>
<td>4.0E-04</td>
</tr>
<tr>
<td>2</td>
<td>6.8E-01</td>
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<tr>
<td>3</td>
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<td>2.7E-04</td>
</tr>
<tr>
<td>4</td>
<td>1.2E-01</td>
<td>-6.1E-04</td>
</tr>
<tr>
<td>5</td>
<td>5.5E-01</td>
<td>-1.4E-03</td>
</tr>
<tr>
<td>6</td>
<td>7.3E-01</td>
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<td>7</td>
<td>5.5E-01</td>
<td>1.1E-03</td>
</tr>
<tr>
<td>8</td>
<td>3.0E-01</td>
<td>5.1E-04</td>
</tr>
</tbody>
</table>

Table 10: Velocities for condition with proprioceptive cues
<table>
<thead>
<tr>
<th>Subject</th>
<th>Max Velocity (deg/s)</th>
<th>Average Velocity (deg/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>9.7E-01</td>
<td>-2.7E-07</td>
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<tr>
<td>2</td>
<td>1.3E+00</td>
<td>-6.4E-04</td>
</tr>
<tr>
<td>3</td>
<td>4.8E-01</td>
<td>6.1E-04</td>
</tr>
<tr>
<td>4</td>
<td>2.9E-01</td>
<td>-8.6E-05</td>
</tr>
<tr>
<td>5</td>
<td>1.4E+00</td>
<td>-1.9E-04</td>
</tr>
<tr>
<td>6</td>
<td>5.0E-01</td>
<td>-8.0E-05</td>
</tr>
<tr>
<td>7</td>
<td>1.4E+00</td>
<td>5.1E-04</td>
</tr>
<tr>
<td>8</td>
<td>1.5E+00</td>
<td>2.0E-04</td>
</tr>
</tbody>
</table>

Table 11: Velocities for condition with visual cues

<table>
<thead>
<tr>
<th>Subject</th>
<th>Max Velocity (deg/s)</th>
<th>Average Velocity (deg/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>4.0E-01</td>
<td>-7.3E-05</td>
</tr>
<tr>
<td>2</td>
<td>9.2E-01</td>
<td>1.8E-04</td>
</tr>
<tr>
<td>3</td>
<td>9.7E-02</td>
<td>1.3E-04</td>
</tr>
<tr>
<td>4</td>
<td>9.4E-02</td>
<td>3.0E-04</td>
</tr>
<tr>
<td>5</td>
<td>4.4E-01</td>
<td>-3.8E-04</td>
</tr>
<tr>
<td>6</td>
<td>1.3E+00</td>
<td>-3.1E-04</td>
</tr>
<tr>
<td>7</td>
<td>8.6E-01</td>
<td>-1.6E-04</td>
</tr>
<tr>
<td>8</td>
<td>2.4E-01</td>
<td>2.7E-04</td>
</tr>
</tbody>
</table>

Table 12: Velocities for condition with visual, and proprioceptive cues
APPENDIX B: Robotic Balance Simulator

Ankle Torque and Force Plate Compensation

The force plate consists of two sections. The top, which moves around slightly when forces are applied, and the base, which houses the top, and is stationary relative to whatever it is fixed to. When the force plate changes angular acceleration, as is caused consistently by the motors in this experiment, the top lags slightly behind the base, and causes the force plate to read a moment that is due solely to the acceleration of the motors, and is completely separate from the moment applied by the subject. This lag is also increased by the weight of the feet on the force plate. To account for this, and extrapolate the moment due solely to ankle torque (and not rotation), we use equations 1-6 from Huryn et al. (Huryn, Luu, Van der Loos, & Blouin, 2010):

\[ \ddot{\theta} = \frac{(M_{\text{grav}} + M_{\text{ankle}})}{I_m} \]

\[ M_p^{\text{measured}} = -M_p^{\text{feet}} \cdot (d_{ZCM} + d_{RCM}) + (m_{fp} a + m_{fp} g \sin \theta) \cdot d_{ZCM} + M_p^{\text{feet}} - I_{fp} \ddot{\theta} M_p^{\text{feet}} \]

\[ M_p^{\text{feet}} = -M_{\text{ankle}} + 0.02 m_m g L_m \sin \theta \]

\[ a = \theta \cdot \ddot{d}_{RCM} \]

\[ M_p^{\text{measured}} = -F_s^{\text{measured}} \cdot Z_0 - M_{\text{ankle}} \]

Once this moment is measured, it is used as the input torque for the transfer function.

Transfer Function

\[ \dot{x} = Ax + Bu \rightarrow [\dot{\theta}] = \begin{bmatrix} 0 & 1 \end{bmatrix} \begin{bmatrix} \theta \\ \dot{\theta} \end{bmatrix} + \begin{bmatrix} 0 \\ 1 \end{bmatrix} [\tau] \]

\[ y = Cx + Du \rightarrow [\theta] = \begin{bmatrix} 1 & 0 \end{bmatrix} [\theta] + [0][\tau] \]
Where $\theta$ is the angle of the virtual inverted pendulum, $\tau$ is the plantarflexor torque from the subject, $l$ is the length from the subject’s ankles to their hips, $g$ is the gravitational constant (9.8N/kg), $m$ is the mass of the subject (0.971 $m$ is used for the virtual pendulum, because the weight of the feet has been removed), $I$ is the inertial mass of the virtual inverted pendulum ($I = 1.119 m^2 l^2$), and $b$ is the viscous damping ($b = 0.1066 Nm/s/deg$).

**Power Circuit for Motors**

![Diagram for Power Cabinet](image)

**Figure 18: Diagram for Power Cabinet**

For the second motor, almost everything must be doubled. Exceptions include the wall plug, on/off switch, and tdk-lambda, none of which need to be doubled. The output from this part of the circuit can be branched. A second Servo pack is required, but only a single PXI and FPGA is required (all non-clock
Motor Delay

The Yaskawa SCMCS-2ZN3A-YA21 motors have a delay of ~20ms. A stochastic signal (0-1Hz) was sent to the motors, then compared to the resulting encoder position. The delay was estimated through visual inspection of the lag between the input signal and the encoder position.

Motor Noise

The Yaskawa SCMCS-2ZN3A-YA21 motors were selected for their torque ratings (200Nm Stall Torque, 600Nm Instantaneous Peak Torque). These motors have a switching frequency of 16.667 kHz, which gets propagated as noise through the room, and all equipment. To counteract this noise, the following precautions have been included:

- Debounce filters on all digital inputs (includes encoders and limit switches)
- Twisted pair wires with shielding for all digital signals
- Two-way lowpass filtering on all recorded data

Visual Delay

Rendering of the visual scene takes a non-trivial amount of time, and required a predictor algorithm for synchronization. The time between a new angle being sent to the rendering computer, and the moment the new angle is displayed to the user depends on the scene. To determine this time, a basic program was created that simultaneously illuminates a led, and sends a new angle to the display. A high speed camera recorded both the led and the visual display. The frame rate, and number of frames between the led illuminating, and the change in angle of the scene was used to determine the
lag. A linear predictor has been used to synchronize the visual response with the motors at a delay of 20ms. The prediction method used was designed and implemented by Dr. Kees van den Doel. It is a linear prediction model based on 6 data points. The coefficients are selected by fitting data of human sway to an offset of the same data. The fitting is done using a linear least squares method.

**Frustum and Notional Camera**

The frustum is calculated for a flat screen that lies on the screen and exists between the subject and the projector. The front plane of the vision frustum is set to 4m x 5m, the near clipping plane is set at 0.77m, and the far clipping plane is set at a depth of 300m. The apex of the frustum is determined by the distance between a subject’s ankles and their eyes, and their standing angle. The apex of the frustum is continuously updated to reflect the current location of the subject’s head and eyes. To give a sense of movement, the standing angle is used to translate the notional camera as if it were on top of an inverted pendulum with the same ground-to-eye height as the subject. The horizontal location of the notional camera will be set to the height of the subject’s eyes scaled by the negative cosine of the angle. The vertical location of the notional camera will be set to the height of the subject’s eyes scaled by the sine of the angle. In rotations of the human head, there is counter-roll in the eyes (Collewijn, Van der Steen, Ferman, & Jansen, 1985). Pilot testing showed 30% to feel like a believable amount of roll to display to the eye. The notional camera will therefore be rotated by 30% of the standing angle. If the subject’s head is not actually moving, but motion is being simulated, the visual ocular reflex is simulated by rotating the camera by 30% of the actual rotation angle.

**APENDIX C: Code**

There are four devices that run code

- Host computer
This computer provides the only interface between the user and the PXI

- **Real-time Computer**

  Runs all transfer functions and interacts with all other systems. This system never interacts directly with the user. This ensures that all priority is given to maintaining the transfer functions and data acquisition with minimal delay.

- **FPGA**

  Controls the input and output of the motors, as well as the timing of the real-time computer. Controlled by the real time computer

- **Rendering Computer**

  Renders all images for the projection screen (for active 3D projection), and sends rendered images to the projector. Receives commands from the PXI via UDP packets.

Since some of these machines use object oriented programming (OOP), a concept which some of the readers may not be familiar with, a brief introduction to OOP will be given before the code is discussed.

**Object Oriented Programming**

OOP is built on the concept of ‘classes.’ A class is a collection of variables and methods. A class is often modelled after real-world items, like a force-plate or a motor. A specific instance of a class is referred to as an ‘object.’ This is similar to how a specific implementation of a model is referred to as a simulation. OOP (Object Oriented Programming) revolves around the idea of objects: keeping variables and methods neatly bundled in their appropriate classes. This contrasts with functional programming, where the programmer keeps track of all variables.
Note that while ‘methods’ and ‘functions’ are similar, a ‘method’ is a snippet of code that is implicitly called on an object, while a ‘function’ is a snippet of code that could exist without objects or classes. The two are not mutually exclusive, and can be used together.

After an object has been initialized, updating its values is accomplished by calling a method on that object.

Eg

Suppose there is a class, ‘forceplate.’ It contains two variables: ‘calibratrionMatrix,’ and ‘offset.’ It has a method, ‘getMomentX’ that takes an array, ‘analogSig,’ and returns a value based on ‘analogSig, calibrationMatrix, and offset.’ Suppose there is a second method called ‘zero’ which takes an input, ‘analogSig,’ and updates ‘offset.’

Here is pseudocode that demonstrates the use of this class in OOP:

```java
fpObj = new forceplate(); //initialize the forceplate object
analogSig = … //get analogSig
fpResult = fpObj.getMomentX(analogSig); //obtain result
fpObj.zero(analogSig); //update 'offset' value
fpResult = fpObj.getMomentX(analogSig); //obtain result with new offset value
```

Contrast this with code that does the same task with functional programming

```java
calibrationMatrix = ... //initialize the calibration matrix
offset = ... //initialize the offset
analogSig = … //get analogSig

//compute 'result'
result = computeMomentX(analogSig, calibrationMatrix, offset);
//update offset
offset = updateOffset(analogSig, calibrationMatrix, offset);
//compute 'result' with updated offset
result = computeMomentX(analogSig, calibrationMatrix, offset);
```
Note that in the second case, there are two variables, ‘calibrationMatrix,’ and ‘offset,’ that are explicitly passed to all the functions. In the first example these values are managed by the forceplate class, and are passed implicitly whenever a method is called on a forceplate object. If more variables were required for each of those computations, the OOP example would look the same; however, the functional programming example would have more variables that it would need to initialize (and possibly update), as well as more variables that it would have to explicitly pass to functions. Complex tasks will also have a larger difference in number of variables that must be tracked at the high levels.

OOP excels at conceptual groupings. All of the variables that relate specifically to the forceplate class are stored in the forceplate class. In the functional case, a similar connection between ‘calibrationMatrix,’ and ‘offset’ is not immediately apparent. OOP can simplify designing a program from a written description of the programming task. Nouns either become variables, or objects which group the variables (and sometimes objects) together. Verbs become methods that act on the objects. In functional programming, verbs become functions, and only some nouns become variables. OOP relates more directly with spoken language, and is often easier to describe and conceptualize. The conceptual structure of OOP classes is often displayed in structures called ‘UML Diagrams,’ where UML stands for ‘Unified Modeling Language.

OOP has three main concepts, often referred to as the ‘Three Pillars of Object Oriented Programming:’ encapsulation, inheritance, and polymorphism. The above only touches on encapsulation. Inheritance and Polymorphism are also very strong and compelling reasons to use OOP, but did not play a large role in this thesis, and will not be discussed here. quod erat demonstrandum.
FPGA

The FPGA runs at 40MHz. It outputs a 2 kHz square wave (which is used to trigger the execution of the real-time controller), observes the encoder of the motors, receives position commands from the PXI (at an expected rate of 2 kHz), and through the use of a cubic spline and a PID controller, computes the setpoint for each motor at 20 kHz.

Host Computer

The host computer does not have a fixed execution frequency. Commands are sent from the host computer to the real-time computer by updating variables over the network. Buffered data from the real-time computer is read by the host computer, then displayed to the user.
Real-time Computer

The real-time computer has multiple classes that are used. For ease of reference, the UML class diagrams are presented here.

Figure 19: UML Class Diagram 1/2
The real-time computer uses the moment from the force plate to update the simulation of an inverted pendulum. The angle of the inverted pendulum is used to update the motor position, to compute the level of GVS that is sent to the subject, and is also sent to the rendering computer.
Rendering Computer

The rendering computer is running a script which listens for UDP packets containing commands. When instructed, this script will open a 3D scene.

The 3D scene is rendered in openGL for active3D. Different environments can be selected (or constructed). Each scene is an implementation of the ‘myWorld’ abstract class:

<table>
<thead>
<tr>
<th>myWorld</th>
</tr>
</thead>
<tbody>
<tr>
<td>+init():void</td>
</tr>
<tr>
<td>+updateAngle(angle:double,coherence:double):void</td>
</tr>
<tr>
<td>+updateFrustum(angle:double):void</td>
</tr>
</tbody>
</table>

Figure 21: UML diagram for rendering computer