MODULATION OF STOCHASTIC VESTIBULAR STIMULATION-INDUCED REFLEXES WITHIN A DYNAMIC BALANCE PARADIGM: THE EFFECT OF RESPONSE PHASE AND EMOTIONAL STATE

by

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Abstract

The vestibular system is a complex network that plays an important role in balance control. When postural perturbations are exerted on the individual, it appears that the vestibular system plays a role in modulating the amplitude of the responses. The vestibular system is also susceptible to changes in psychosocial and autonomic states. Despite these findings, the inability to precisely record from and directly manipulate the system has hindered the field in completely understanding how the vestibular system is involved in balance. Therefore, the purposes of this thesis were 1) to investigate if there was phase-dependent modulation of the vestibular reflex during the postural responses and 2) to determine if the vestibular reflex was altered with postural threat.

Stochastic vestibular stimulation (SVS) was used to electrically probe the vestibular system while participants stood on a rotating platform. The vestibular reflex was analyzed by estimating the vestibulo-muscular (SVS-EMG) relationship using time-dependent SVS-EMG coherence throughout the postural response for the first purpose while, for the second purpose, SVS-EMG coherence, cumulant density, and gain were calculated between non-threatening and threatening conditions. Results from this thesis were unable to determine if there were phase-dependent modulations of the SVS-induced vestibular reflex. However, further testing and pilot data provides a promising method for further investigation. Furthermore, an increase gain in and coupling of the vestibular reflex was observed in the most muscles while a decrease in coupling was observed for the paraspinal muscles in the threatening situation. These results suggest that the central nervous system has the ability to prepare the body for responding to an upcoming postural perturbation by optimizing the vestibular output to the muscles.

Preface

The protocols used in these studies were reviewed by The University of British Columbia Clinical Research Ethics Board (UBC CREB# H06-04047; see Appendix A). All subjects provided written informed consent prior to participation in these studies and every effort has been made to ensure that the subjects are not identified in this thesis.

I was the lead investigator on the project, responsible for concept development, data collection and analysis, and document composition. Dr. Jean-Sébastien Blouin conducted the time-dependent coherence analyses and power spectral density estimates and was involved in concept development. Dr. Mark G. Carpenter and Dr. J. Timothy Inglis were the supervisory authors on the project and were involved in the concept formation and thesis revisions. The experiments contained in this thesis have not been submitted for publication at the time of thesis submission.

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List of Abbreviations

Cl:	Confidence limit
COM:	Centre of Mass
EDA:	Electrodermal Activity
EMG:	Electromyography
EO:	External Oblique muscle
FFT:	Fast-Fourier Transform
GM:	Gluteus Medius muscle
GVS:	Galvanic Vestibular Stimulation
mGAS:	Medial Gastrocnemius muscle
ML:	Medium Latency
PARA:	Paraspinal muscles
RMS:	Room Mean Square
SL:	Short Latency
SOL:	Soleus muscle
SVS:	Stochastic Vestibular Stimulation
VL:	Vastus Lateralis muscle

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For science

Chapter 1: Introduction

Sensory systems such as the visual, somatosensory, and vestibular systems are important in the maintenance of upright stance. These systems play a critical role in detecting a balance disturbance and in determining an appropriate response. Deficits or modifications to balance control are often seen when one of these systems is impaired or altered from their normal function. There is a vast amount of literature on each of these systems and their role in balance control. Despite the abundant work, the specific role of the vestibular system, in particular, is still debated. Evidence does suggest that this system's role differs across various phases of balance control and may change as a result of different arousing and psychosocial situations. However, methodological limitations have prevented these theoretical changes from being clearly determined. For this reason, the overall goal of this thesis is to implement a novel approach to investigate the vestibular system and its potential modulations.

1.1 Vestibular Anatomy

The vestibular system can be classified into its peripheral, central, and efferent networks. The peripheral vestibular organs within this system are located in the inner ear and are mirrored between the left and right sides of the head. As a result, they act as a paired system where, during a movement, activation of receptors on one side results in a complimentary response on the other side. These receptors are clusters of hair cells strategically situated within the organs and are affected by different types and directions of accelerations. The cells consist of 40-70 projections called stereocilia, aligned according to height with the tallest end adjacent to the largest and thickest hair, the kinocilium. When the stereocillia are displaced either towards or away from the kinocilium, the mechanical change causes an opening or a closing of

the cation channels, respectively. In a rest state, a small number of these ion channels remain open, resulting in a slightly depolarized state. A mechanical change in the cation channel allows the cellular membrane to further depolarize when opened or become hyperpolarized when closed (Goldberg, Walker, & Hudspeth, 2013).

The peripheral network consists of five organs (three semicircular ducts and two otolith organs), each of which utilizes the previously mentioned mechanoreceptor. The three semicircular ducts (anterior, horizontal, and posterior) are oriented orthogonally to each other, with the anterior portion of the horizontal canal tilted approximately 15-30 degrees above Reid's plane (auriculo-orbital plane) (Fitzpatrick & Day, 2004), and the anterior and posterior canals aligned 45 degrees from the sagittal plane (Krebs, Weinberg, & Akesson, 2012) (Figure 1A). These ducts are filled with endolymph, a fluid with a high potassium and low sodium composition. Due to the circular shape of the ducts and the inertial lag of fluid, the ducts best respond to angular accelerations of the head (Angelaki, Shaikh, Green, & Dickman, 2004). For example, when the head is quickly rotated clockwise (i.e. to the right), the endolymph in the horizontal duct lags and is displaced in a counterclockwise direction relative to the ducts. At the end of each duct the endolymph then flows into an enlarged region called the ampulla, which is lined with the hair bundles. Here, the hair cells are attached at the base to the crista ampullaris and are encased in a membrane called the cupula (Figure 1B). Movement of the endolymph displaces the cupula and, in turn, causes a mechanical change of the hair cells.

In addition to the semicircular ducts, the two otolith organs (the saccule and utricle) reside within vestibule of the bony labyrinth. The utricle is located directly adjacent to the ampulla of each canal whereas the saccule is connected to the cochlear duct (Figure 1A). The hair cells within these organs are encased in a gelatinous macula that is located at the bottom

and on the medial surfaces of the utricles and saccules, respectively. In addition, crystalized calcium carbonate fragments (otoconia) are embedded to the macular surface (Figure 1C). Movement of these dense otoconia through gravitational and linear accelerations creates shear forces on the membrane and mechanically activates specific hair bundles (Angelaki et al., 2004). These organs code for linear accelerations as opposed to the angular accelerations detected by the ducts.

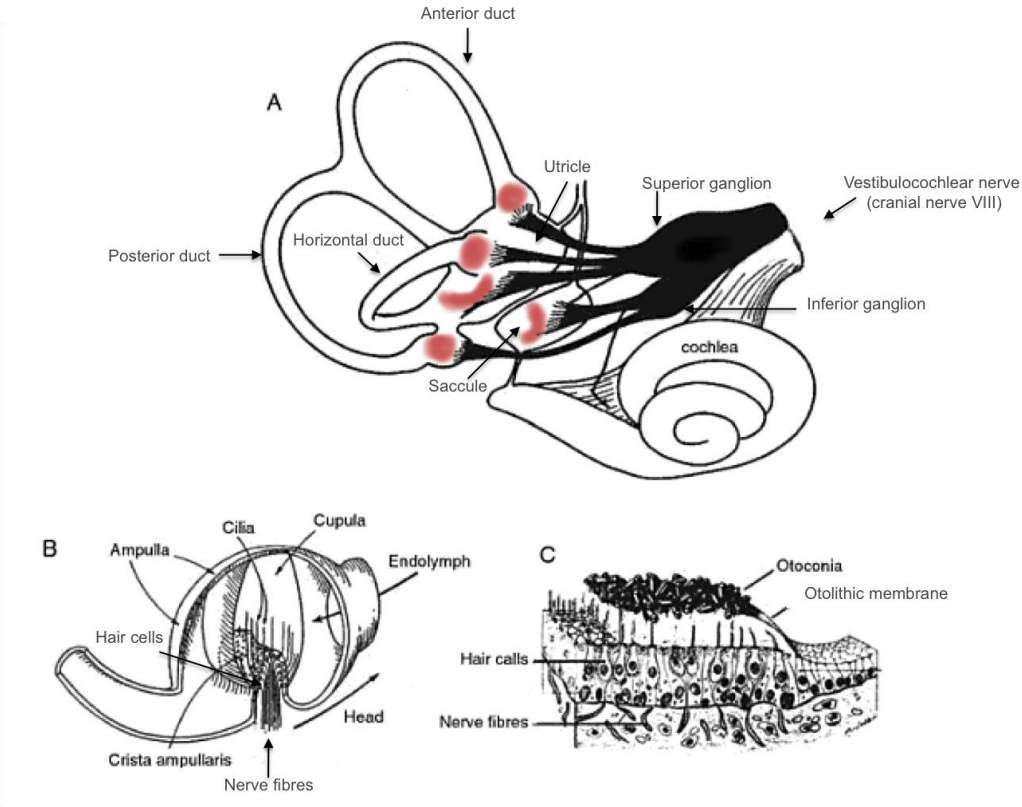

Figure 1.1 A The peripheral vestibular apparatus. The three semicircular ducts (anterior, posterior, and horizontal) are shown with the enlarged region (ampulla) adjacent to the utricle. The saccule, connected to the cochlea, lays adjacent to the utricle. The location of the hair bundles is indicated by the coloured regions and is innervated by the superior and inferior branches of the 8th cranial nerve.

B The hair bundles within an ampulla. C. The hair cells in the macular region of the otolith organs. *Figure adapted from Goldberg et al., (2012*).

The mechanical changes to the hair cells and their ion channels then increase action potential firing rates of the vestibular nerve. There are two general categories of vestibular afferent nerves: irregular and regular firing afferents. The irregular afferents typically respond to changes in accelerations, have larger axons, and are more sensitive. The regular afferents have opposite characteristics and generally respond to more steady-state information (Goldberg, Smith, & Fernandez, 1984). The vestibular division of the afferent vestibulocochlear, or 8th cranial nerve, innervates each hair cell. The vestibular cell bodies of these nerves are clustered in the superior and inferior vestibular ganglions, which lie within the internal auditory meatus (Goldberg et al., 2013). The superior vestibular ganglion receives information from the utricle, the anterior part of the saccule and the anterior and horizontal semicircular canals. The remaining input from the posterior section of the saccule and the posterior semicircular canals are sent to the inferior vestibular ganglion (Blumenfeld, 2002). These nerves then travel through the auditory canal and cranial cavity and connect to a number of different nuclei.

The majority of peripheral vestibular information is received by the ipsilateral vestibular nuclear complex (the central vestibular system) within the cerebellopontine angle of the brainstem (Martin, 2003) while a small number project directly to the cerebellum. The central vestibular system includes a pair of lateral, medial, inferior, and superior vestibular nuclei. The lateral vestibular nucleus primarily receives linear information from the otolithic organs whereas the medial vestibular nucleus processes angular information from the semicircular canals. Processed visual and proprioceptive information are also sent to the vestibular nuclei (Krebs et al., 2012). From here, all the information is then relayed to 1) the cerebellum through

the vestibulocerebellar tract, 2) the lower motor neurons through the descending medial and lateral vestibulospinal tracts and the reticulospinal tracts, 3) the eyes through the ascending medial longitudinal fasciculus for the vestibulo-ocular reflex or 4) the primary and secondary vestibular cortex via the ventral posterior thalamus through the ascending medial longitudinal fasciculus (Martin, 2003).

Most important to this thesis are the descending tracts. The medial and lateral vestibulospinal tracts are involved in maintenance of balance, orientation, and muscle tone (Carpenter, 1988). The majority of the information running through the medial vestibulospinal tract is relayed bilaterally from the medial and inferior vestibular nuclei. These bilateral projections carrying mainly angular information have both excitatory and inhibitory connections to alpha motor neurons in the cervical and thoracic regions (Carpenter, 1988). The majority of information traveling through the lateral vestibulospinal tract, on the other hand, comes from the ipsilateral medial and lateral vestibular nuclei. The lateral vestibulospinal tract preferentially sends a tonic facilitatory input onto the motor neurons of extensor muscles around the ankle, knee, back, and upper limb (Lund & Pompeiano, 1968; Uchino & Kushiro, 2011). Both angular and linear acceleration information is relayed through these tracts (Krebs et al., 2012). In addition, the reticulospinal tracts also relay vestibular information (as well as information from a number of other systems) to the spinal cord and influence muscle tone (Deliagina, Zelenin, & Orlovsky, 2012).

Through understanding the vestibular system's anatomy, it is clear that its ability to detect movement and transfer information to lower motor neurons make this system potentially critical for balance control. It is, however, still unclear how and when this system directly influences balance control during dynamic paradigms. Before detailing the possible roles of the

vestibular system, it is important to first understand dynamic balance control and, specifically, postural responses.

1.2 Feet-in-Place Postural Responses

Biomechanically, in order to maintain an upright stance position, the vertical projection of the body's centre of mass (COM) must remain within the individual's base of support. The COM is defined as a point that represents the body's total mass (Hamill & Knutzen, 2003; Macpherson & Horak, 2013). When the COM's vertical projection suddenly deviates towards an edge of the base of support, such as in a push, a necessary adjustment is needed to remain upright. This adjustment has been termed a postural response and could either involve stepping to change the base of support or a feet-in-place muscle response. For the purpose of this thesis, a postural response in this document will refer to the feet-in-place response.

The human postural response has been classified as a very distinct pattern of muscle activation (Carpenter, Allum, & Honegger, 1999a; Nashner, 1976, 1977). The first response to a support surface perturbation is a quick monosynaptic stretch reflex at approximately 40 ms after perturbation onset. This quick response is later followed by a balance-correcting response, approximately 120 ms after the perturbation, which is presumably too quick to be entirely voluntary (Nashner, 1976). These automatic responses are then followed by a later (240 ms), and arguably more voluntary, stabilizing response that brings the body to its new equilibrium point (Carpenter et al., 1999a).

This postural response is highly specific to the direction and amplitude of the perturbation as well as the initial biomechanical state of the individual. If the support surface is rotated in a toes-down direction, a bilateral stretch reflex is first seen in the tibialis anterior and quadriceps muscles while an unloading of the paraspinal muscles are observed. Following this,

a balance-correcting response is observed in the soleus and quadriceps (Carpenter et al., 1999a). On the contrary the same study showed that when the perturbation is in the roll plane with the right side down, an asymmetrical response is observed. The left soleus and right paraspinal muscles are stretched and the left paraspinals are unloaded. The balance-correcting response is then seen in the right and left tibialis anterior, right quadriceps, right soleus, and left paraspinal muscles. As the perturbation becomes faster and larger, the postural response amplitudes also scale accordingly in order to appropriately maintain the upright stance (Allum, Honegger, & Schicks, 1994; Inglis, Horak, Shupert, & Jones-Rycewicz, 1994; Park, Horak, & Kuo, 2004). Additionally, the biomechanical state of the individual can highly influence the postural response. Diener, Bootz, Dichgans, and Bruzek (1983) showed that leaning backwards prior to a toes-up rotation perturbation elicited greater and faster tibialis anterior activity in response to the greater postural disturbance compared to when standing normally without a backwards lean. Horak and Moore (1993) also showed similar changes in muscle activity when leaning prior a support surface translation (e.g. greater and faster tibialis anterior activity when standing normally compared to leaning forward prior to a forward translation). Although highly influenced by the type of the perturbation and state of the individual, under consistent conditions, the distinct phases and muscle activation patterns are quite robust and have clearly been shown in numerous dynamic balance situations (Allum, Honegger, & Schicks, 1993; Carpenter et al., 1999a; Diener, Dichgans, Guschlbauer, & Mau, 1984; Keshner, Allum, & Pfaltz, 1987; Keshner, Woollacott, & Debu, 1988).

1.3 The Vestibular System's Influence on Postural Responses

The potential role of the vestibular system in these postural responses has been widely debated. Previous work has suggested that the vestibular system could be involved in

triggering postural responses. This has been evidenced by numerous studies targeting the vestibular system through head perturbation with forehead taps (Bötzel, Feise, Kolev, Krafczyk, & Brandt, 2001; Bötzel, Kolev, & Brandt, 2006) and head translations (Horak, Earhart, & Dietz, 2001; Horak, Shupert, Dietz, & Horstmann, 1994; Horstmann & Dietz, 1988). With both of these perturbations, lower leg responses were triggered despite the absence of large ankle angle changes. When head taps were experienced by those with vestibular loss, the muscle activities were highly attenuated or absent in individuals with adult onset vestibular loss (Bötzel et al., 2001). Similarly, full body responses due to head translations were absent in individuals with adult onset vestibular loss (Horak et al., 1994). Another method used to independently stimulate the vestibular system was through drops or free falls (Greenwood & Hopkins, 1976a, 1976b). In this paradigm, independent of the height of the drop, responses in both the neck and lower leg muscles in vestibular intact individuals were consistently seen around 50 and 70 ms, respectively, providing evidence for an automatic vestibular reflex. When adult onset vestibular loss participates experienced this drop, these muscular responses disappeared (Greenwood & Hopkins, 1976b). In addition, this early response was still seen in canal-plugged felines but not when they were completely labrinthectomized, suggesting the response is of otolithic origin (Watt, 1976). The studies mentioned above all indicate the vestibular system's ability to trigger postural responses. However, these induced responses are frequently small compared to those resulting from full body perturbations.

When a full body perturbation is experienced, commonly through support surface platform perturbations, the vestibular system appears to play a greater role in modulating the amplitude of the response than in triggering the response. This modulation also appears to be dependent on the specific phase of the postural response. Although not directly recorded,

vestibular modulations during different phases of a postural response can be inferred through numerous experimental results. In these experiments, individuals with vestibular loss were perturbed during upright stance while their postural responses were recorded. Similar to healthy individuals, in most muscles, those with vestibular loss were still able to elicit postural responses at approximately the same latency after the onset of a platform rotation, which supports the idea of a non-vestibular trigger (Allum, Keshner, Honegger, & Pfaltz, 1988; Allum, Oude Nijhuis, & Carpenter, 2008; Allum & Pfaltz, 1985; Allum & Shepard, 1999; Carpenter, Allum, & Honegger, 2001; Keshner et al., 1987). Despite the regular onset times, consistent irregularities have been observed in the amplitude of muscle activation. Specifically, while the stretch reflex amplitudes remain the same, these individuals show a robust change in activation amplitude beginning in the balance-correcting window (120-220 ms). The responses of the leg (Carpenter et al., 2001a), the arm (Allum et al., 2008), and the neck (Allum & Honegger, 1998) muscles during this phase are significantly smaller while responses of the back muscles are larger (Allum, Honegger, & Acuña, 1995; Allum et al., 1994) in those with vestibular loss compared to normal controls. Following this, the secondary balancing correcting (240-340 ms) and stabilizing (350-700 ms) responses are often larger compared to normal controls. Similar changes in response amplitudes have been shown in individuals with vestibular deficit when they experience a platform translation (Horak, Nashner, & Diener, 1990; Runge, Shupert, Horak, & Zajac, 1998) or a combination of both rotation and translation (Allum et al., 1994). Although vestibular loss patients show impairments within numerous perturbation paradigms, these individuals have the greatest difficulty maintaining balance after a platform rotation. Compared to other perturbations, platform rotations result in the largest difference in postural response amplitude between vestibular loss patients and controls (Allum

et al., 1994). These impairments are also significantly different between the direction of perturbation with the greatest difference measured for lateral perturbations (Carpenter et al., 2001a). In addition, these individuals also have a high tendency to fall in the lateral direction (Martin, 2003). Overall this indicates that the vestibular system plays an important role in modulating the amplitude of a postural response following lateral platform rotations.

Changes in response amplitude have also been shown in feline bilateral labyrinthectomy studies (Inglis & Macpherson, 1995; Macpherson, Everaert, Stapley, & Ting, 2007). With the removal of the bilateral peripheral vestibular apparatus, these felines were still able to elicit postural responses to surface translations (Inglis & Macpherson, 1995) and surface rotations (Macpherson et al., 2007). These responses, however, were slightly different to those seen in humans with vestibular loss. In the response to a support-surface rotation, the amplitude change during the balance-correcting phase seemed to be of opposite polarity compared to the response in normal cats, where the activated muscles further perturbed the feline off balance. As noted by Allum et al. (2008), however, these discrepancies could be explained by the biomechanical differences between bipedal and quadrupedal mammals. These perturbations induced passive movements of the head in opposite directions that may have induced different neck reflexes, (e.g. vestibulo-colic and cervico-colic reflexes) and thus may relate to a different role of the vestibular system in quadrupeds compared to bipeds. The combined human and feline evidence indicates that a loss of peripheral vestibular function greatly impacts the amplitude of the platform-induced postural response at the balance-correcting phase and is not involved in triggering the response.

A major limitation to the work involving vestibular loss participants is the amount of compensation experienced by the recovering individual. As the central nervous system has

numerous sources of sensory information for body orientation and balance (e.g. visual, somatosensory, and vestibular), the loss of information from one sensory system could lead to a reweighting of information from other sources. This reweighting and adjusting for the loss of information is a function of compensation that occurs during recovery (Yardley & Redfern, 2001). Sensory compensation with vestibular loss individuals occurs quite rapidly (Igarashi, 1983). In fact, within several months of the injury, these individuals are able to stand upright and perform most tasks with the use of their visual and somatosensory systems. This quick compensation, therefore, makes it difficult to clearly determine what the vestibular system's role might be in controlling normal standing balance.

In order to determine the vestibular system's role in normal balance control, further investigation of the system in healthy individuals is needed. Some groups have attempted to temporarily distort the vestibular system in healthy individuals. One of the earlier studies of this involves the investigation of postural responses while in microgravity (Clement, Gurfinkel, Lestienne, Lipshits, & Popov, 1985). In this study, individuals were tested before space departure, while in space, and upon return to earth. Similar to that seen in individuals with vestibular loss, without the effect of normal gravitational forces, and thus normally functioning otolith organs, the balance-correcting response to a sudden forward translation was present at the same latency as that seen on earth but showed a significant decrease in amplitude of the lower leg muscles. Watt and colleagues (1986) used a similar paradigm where, instead of surface perturbations, they elicited drops at different levels of weightlessness. They also found a decrease in lower leg activity in microgravity. All the previously mentioned studies indicate that the vestibular system is involved in postural reactions, and more specifically, it may also be more involved during the balance-correcting phases of the response after perturbation onset.

1.4 Galvanic Vestibular Stimulation

Another way to investigate the vestibular system's role in balance control is through a technique called galvanic vestibular stimulation (GVS). GVS is delivered by passing current through electrodes on the mastoid processes behind each ear. This current conducts through the skin and into the peripheral vestibular nerves. Although the exact point of stimulation is not fully understood, it is generally believed that GVS modulates the vestibular afferents by changing the firing rates of the vestibular nerve, and in particular, the highly sensitive irregularly firing afferents (Goldberg et al., 1984). Placement of the positive electrode (the anode) on the mastoid is thought to decrease neural firing whereas the nerves under the negative electrode (the cathode) are thought to increase their firing rate. The electrical stimulation affects the population of nerves below the electrode and, therefore, afferents originating from all peripheral vestibular organs are affected. As reviewed by Fitzpatrick and Day (2004), the stimulation of otolith afferents results in an acceleration along the line between the mastoids, the inter-aural line, towards the anode, while stimulation of semicircular canal afferents exhibits a rotation around Reid's plane towards the anode. This model is consistent with numerous behavioural effects of GVS such as those changes observed in perception (Zink, Bucher, Weiss, Brandt, & Dieterich, 1998), posture (Ali, Rowen, & Iles, 2003) or gait trajectory (Bent, McFadyen, French Merkley, Kennedy, & Inglis, 2000; Iles, Baderin, Tanner, & Simon, 2007).

As the orientation of the vestibular organs is locked within the head, the effect of GVS is directly affected by position of the head. With the head forward, the effect will be a lateral sway, whereas turning the head 90 degrees to the left or right results in an anterior or posterior sway (Britton et al., 1993). To a certain point, similar to that seen after a postural platform

perturbation (Inglis et al., 1994), an increase in stimulus intensity or amplitude results in an increased postural response (see review by Fitzpatrick & Day, 2004).

There are various patterns of electrical stimulations that can be applied through the electrodes. The square pulse is a discreet stimulus that elicits a directional reflexive contraction mainly in muscles involved in postural tasks (Britton et al., 1993). Within this reflex are two peaks of opposite polarity. In EMG, the initial short-latency peak occurs approximately 50-70 ms after the stimulus, while the medium-latency peak is seen 100-120 ms after the stimulus (Britton et al., 1993; Dakin, Lee Son, Inglis, & Blouin, 2007; Fitzpatrick & Day, 2004; Lee Son, Blouin, & Inglis, 2008). For example, with the head turned to the side and the anode electrode now on the posterior mastoid a square wave pulse would evoke a short-latency inhibitory response followed by a medium-latency excitatory response in the soleus muscles and a sway backwards (Ali et al., 2003). The opposite response amplitudes would be seen in the antagonist muscles (Lee Son et al., 2008). Currently, the nature of these peaks have not been clearly defined. It has been postulated that the peaks may represent different origins (e.g. canals or otoliths (Cathers, Day, & Fitzpatrick, 2005; Mian, Dakin, Blouin, Fitzpatrick, & Day, 2010)) or different pathways (e.g. vestibulospinal or reticulospinal (Britton et al., 1993)). In fact, although the two peaks cannot be independently controlled, each peak is more sensitive to certain frequency ranges compared to the other (Dakin, Inglis, & Blouin, 2011). On the contrary, both peaks seem to respond similarly to changes in head orientation and stimulation orientation (Mian et al., 2010).

Despite the effectiveness of using a square wave, this technique becomes less powerful when investigating dynamic balance paradigms. Previous work has used long-duration square wave pulses to look at the effects of additional vestibular stimulus on the postural response due

to platform translations (Hlavacka, Shupert, & Horak, 1999; Inglis, Shupert, Hlavacka, & Horak, 1995). Even with implementation of GVS at different intervals before or at the perturbation onset, no changes were noted in the automatic postural response phase. Instead, this 8 second GVS stimulus affected the later components (>1.5 s) of the response, which resulted in a change of the subject's final position. This equilibrium change from long GVS pulses, however, is not surprising as GVS has previously been shown to induce leaning (Fitzpatrick & Day, 2004). Short latency muscle responses were also not recorded in these experiments.

Perhaps implementation of short duration pulses, as commonly seen in previous GVS studies (Britton et al., 1993; Lee Son et al., 2008; Reynolds, 2011), appropriately timed to different phases of the postural response would provide a better indication of the transient effects of the vestibular system. However, this may present a new confound, as the shutting off of GVS also evokes a postural response. Additionally, for the effect of GVS to be reliable and visible, numerous pulses are commonly presented (e.g. 1024 pulses in Lee Son et al. 2008; 760 in Dakin et al. 2007). With the large number of trials needed and the numerous windows of stimulation, using this square wave pulse to investigate the vestibular system's role across the whole postural response would be inefficient and fatiguing for the subject.

A more effective method to use is stochastic vestibular stimulation (SVS). SVS is similar to the GVS technique in that paired electrodes are placed on the mastoid processes. However, instead of a single pulse, the stochastic stimulus is a continuous stimulus of varying amplitudes, polarity, and frequencies. This type of stimulation is also advantageous because the constantly varying signal does not result in noticeable vestibular responses in any specific direction and is less predictable. Although a directionally specific response is not observable,

short and medium latency components, similar to those seen with GVS, can be calculated through correlation estimates in the time domain (i.e. cumulant density analysis). In addition to this, the relationship between the inputted SVS signal and the outcome measures (e.g. COP or EMG) can be estimated through coherence and gain calculations. In fact, SVS has been used in numerous studies to investigate the vestibulo-muscular coupling in postural control (Dakin et al., 2007; Pavlik, Inglis, Lauk, Oddsson, & Collins, 1999; Reynolds, 2010) and has been shown to evoke a response only in the muscles actively involved in balance control (Luu et al., 2012). This technique has also been used to assess the vestibulo-muscular relationship during different phases of gait (Blouin et al., 2011; Dakin, Inglis, Chua, & Blouin, 2013). Results from these gait studies indicate that the strength of vestibulo-muscular coupling is dependent on the phase of gait. Based on these results, SVS can be an efficient technique to investigate dynamic balance paradigms and, for the purpose of this thesis, it can be used to investigate the vestibulo-muscular relationship in postural responses.

1.5 Modulation of the Vestibular Reflex with a Threat of Perturbation

The previous sections highlighted the possible changes of the vestibular system's contributions to different phases of the postural response. In addition to these modulations, the vestibular system's overall influence also has the potential to be modulated before a perturbation occurs. When standing in an environment with increased postural threat, such as standing at height (Carpenter, Frank, & Silcher, 1999b; Cleworth, Horslen, & Carpenter, 2012) or in anticipation of a perturbation (Horslen, Murnaghan, Inglis, Chua, & Carpenter, 2013; Shaw, Stefanyk, Frank, Jog, & Adkin, 2012), increases in psychosocial and autonomic measures can be induced. Psychosocial measures, such as fear and anxiety, and autonomic arousal levels have been neuroanatomically linked to the vestibular system through reciprocal

connections via the parabrachial nucleus, the locus coeruleus, and the raphe nuclei (Balaban, 2002; Balaban & Thayer, 2001).

Through these connections, there are two possible ways for threat-induced changes to affect the vestibular system's influence on postural responses. The first is an increase in the gain of the vestibular reflex. Recent work in humans has provided evidence for this proposed influence through a height induced postural threat paradigm. When the vestibular system is stimulated while the individual is standing at height vestibular reflexes are significantly larger than the reflexes observed at ground level (Horslen, Dakin, Inglis, Blouin, & Carpenter, 2014; Naranjo et al. in progress; yet see Osler, Tersteeg, Reynolds, & Loram 2013 for contrast). Secondly, it is possible that these induced changes to the vestibular system under an increased threat increases the tonic input from the vestibular system and directly affects the muscular and biomechanical state of the individual prior to the perturbation. In fact, in a decerebrated individual, the absence of a rubrospinal tract leads to a tonic increase in extensor muscles tone, and thus stiffness, through increased descending input via the vestibulo- and reticulospinal tracts (Krebs et al., 2012; Richerson, Aston-Jones, & Saper, 2013). On the other hand, inhibition of lateral vestibular nucleus, and thus the vestibulospinal tract, through stimulating the inhibitory connection from the cerebellum results in decreased extensor rigidity (Lund & Pompeiano, 1968).

Based on these changes, any modulation of the vestibular system prior to the perturbation can then influence the postural response. Tonic input that affects the muscular and biomechanical state of the individual prior to the perturbation, as previously mentioned, can directly influence the postural response. Behavioural evidence observed an increase in stiffness when standing at height (Carpenter, Frank, et al., 1999; Carpenter, Frank, Silcher, & Peysar,

2001b) and increased postural response amplitudes to perturbations when standing at height (Carpenter, Frank, Adkin, Paton, & Allum, 2004). In order to determine how these postural changes occur, further investigation of the vestibular system under different postural threats need to be examined.

1.6 Summary and Hypothesis

The previous sections indicated a clear involvement of the vestibular system in postural responses. Although there are many theories on how this system may influence these responses (e.g. triggering, amplitude modulation, tonic input, etc.), there is a need to further classify when the vestibular system is coupled to muscles within a dynamic balance control paradigm. For this reason, the first purpose of this thesis was to investigate the phase-dependent relationship between the vestibular system and the postural muscles involved in support surface rotations. In order to gain a clearer insight into possible changes in this relationship, I measured the input-output relationship between stochastic vestibular stimulation and the muscles involved in the postural response. Based on the characteristic changes seen in both individuals with vestibular loss and those in microgravity, I hypothesized that after the platform perturbation onset, the vestibulo-muscular relationship (coherence) would strengthen beginning at the balance-correcting phase of the postural response. The vestibular system can also be influenced through changes in central input. As these changes have the potential to modulate the vestibular system's influence on postural responses, the second purpose of this thesis was to determine if an increased postural threat led to changes in the SVS-induced vestibular reflex. The same technique used to address the first purpose (i.e. SVS-EMG coupling) was used to address this secondary purpose. With this paradigm, I hypothesized that significant vestibulo-musclar coupling (coherence and cumulant density peaks) would first be

observed in muscles involved in standing balance and I also hypothesized that this relationship as well as the gain between the two signals would become stronger and larger with increases in postural threat.

Chapter 2: Methods

EXPERIMENT 1

2.1 Participants

Thirteen healthy individuals between the ages of 19 and 35 (4 females, 9 males; age: 23 ± 3 yrs; height: 171.94 ± 13.75 cm; mean ± SD) volunteered for this study. Each individual had no known neurological, vestibular, postural, skeletal, or orthopaedic deficits based on self-report. Due to the nature of the stimulation and task, individuals who were prone to motion sickness or had ankle weakness were excluded from this study. All individuals were informed of the study prior to participation and provided signed consent in accordance to the ethics board of the University of British Columbia (certificate number: H06-04047). Participants were completely naïve to the perturbation parameters (i.e. velocity, amplitude, direction) and had no prior experience on a rotating platform. Participants received a $30 honorarium for completing the study.

2.2 Data Collection Parameters

2.2.1 Platform Perturbations

Participants stood barefoot on a dual-axis rotating platform. Feet were lightly strapped in position with the heels of each foot placed 15 cm apart. The heel brackets were also adjusted so each lateral malleoli was aligned with the pitch rotation axes of the platform (Figure 2.1A). For safety purposes, handrails were placed on both sides of the platform and a spotter was present at all times to ensure a fall did not occur.

Platform rotations were used based on previous comparisons of the different types of perturbations elicited in vestibular loss participants (Allum et al., 2008)(see section 1.3 for

details). These rotations had a constant maximum displacement amplitude of 7.7° and an average angular velocity of 55°/s (Allum et al., 2008; Carpenter et al., 2001a; Carpenter et al., 2004). Average angular velocity was calculated as the maximum displacement divided by the time it took to reach that displacement. Platform rotations occurred in a pure rightwards roll direction (Figure 2.1B). This direction was chosen because the largest postural response differences seen between individuals with vestibular deficit and healthy controls were observed in the lateral directions (Carpenter et al., 1999a). With the head facing forward, this direction was also optimal as the postural response was in line with the largest vestibular-induced reflex (Britton et al., 1993; Day & Fitzpatrick, 2005). The majority of perturbations were automatically triggered once participants stood in their initial stance for a minimum of 5 seconds[1]. For some trials, participants were not able to maintain initial stance for 5 seconds and would often quickly move out of their threshold values and immediately return to their initial stance. Under these circumstances, platform perturbations were manually triggered to avoid any fatigue from standing for long periods. 10% of the perturbations occurred in the opposite, roll left direction, and acted as catch trials to avoid any anticipation of perturbation direction. Catch trials were not analysed. After the 7.7° displacement was reached, the platform was held at that amplitude for approximately 5 seconds before slowly (1.1°/s) returning back to the original, 0° position. The total number of perturbations presented was dependent on the participants' ability to regain balance without assistance (i.e. without grabbing the safety rail or without help from the spotter). These ranged from 127 to 150 perturbations. The first perturbation of each 10 minute block occurred 15 seconds after SVS onset.

[1] Maintenance of initial stance was determined through two force transducers built into the rotating platform. The combined anterior-posterior (AP) torques and medial-lateral (ML) torques were displayed through an oscilloscope placed in front of the participant and was used by the spotter to guide the participant to their initial stance.

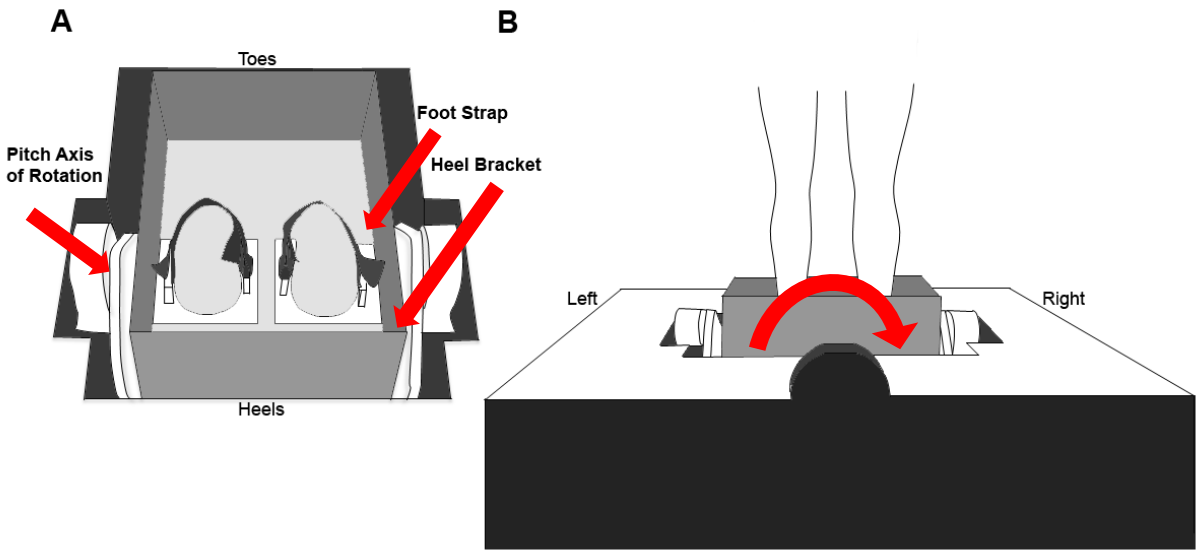

Figure 2.1 **A** Schematic of the foot placement on the platform. Brackets are positioned behind the heels of each foot and are adjusted so each participant's foot is aligned to the axis of rotation. Velcro straps are lightly placed across the top of each foot to maintain the foot position.

B Schematic of a subject standing on the rotating platform. The majority of perturbations were elicited in the rightwards roll direction, as indicated by the arrow, while 10% of the trials were in the opposite, roll left direction.

2.2.2 Stochastic Vestibular Stimulation

Two 9 cm^2, carbon rubber electrodes, one behind each ear, were placed on the participant. A conductive gel (Spectra 360 electrode gel, Parker Laboratories, USA) was used to ensure proper electrical conduction between the electrodes and the vestibular system. These electrodes were secured and held in place by a tensor band during the whole experiment. Continuous SVS was sent binaurally to each electrode. This stimulus contained white noise filtered to frequencies between 0 and 25 Hz, had peak amplitudes of ± 4.5 mA, average root mean square (RMS) of approximately 1.05 mA (Blouin et al., 2011; Dakin et al., 2013), was created using the Labview software (National Instruments, USA), and was delivered as an analog signal (NI PCI-MIO-16E-1 with NI BNC-2110, National Instruments, USA) to an isolated constant-current unit (Model 2200 Analog Stimulus Isolator, AM Systems, USA).

In order to obtain the maximum SVS-induced vestibular reflex during perturbations, the participant's head was positioned facing forwards with an upward tilt of 18° from Reid's plane (Fitzpatrick & Day, 2004). This head position (i.e. pitch rotation) was maintained and monitored online through motion capture markers on the head (Northern Digital Inc., Canada). A laser pointer was also used to as a second reference for the head position target.

The SVS signal was sampled at 2000 Hz (Spike2, Cambridge Electronic Design, UK) then offline low-pass filtered at 100 Hz using a 4^{th} order dual-pass butterworth filter. These data were then resampled to 500 Hz for data reduction (Matlab R2012b, MathWorks Inc, USA).

2.2.3 Electromyography

Surface electromyographic (EMG) data were collected from bilateral soleus (SOL), medial gastrocnemius (mGAS), vastus lateralis (VL), external oblique (EO), and paraspinals at

the L1-L2 level (PARA) muscles. As reflexes induced by galvanic stimulation have only been observed in muscles that are involved in balance control (Britton et al., 1993; Luu et al., 2012), these specific muscles were chosen due to their role in regaining balance, and in particular, regaining balance from lateral perturbations. Pairs of surface electrodes were placed on the muscle bellies with a 2 cm separation. Data were collected at 3000 Hz, amplified 500×, and bandpass filtered between 10 and 500 Hz (Telemyo 2400R, Noraxon, USA). The data were then A/D converted (Power 1401, Cambridge Electronic Design, UK) and sampled at 2000 Hz (Spike2, Cambridge Electronic Design, UK). Offline, the bias was removed from the entire signal by subtracting the mean 500 ms prior to the first perturbation onset (Matlab R2012b, MathWorks Inc, USA). To remove any heart rate and movement artifacts, all EMG data were high-pass filtered at 50 Hz using a 4^{th} order dual-pass butterworth filter. The EMG data were then full-wave rectified, low-pass filtered (100 Hz, 4^{th} order dual-pass butterworth), and resampled to 500 Hz for data reduction prior to analysis.

2.2.4 Angular Head Movements

Perturbation-induced angular head movements were recorded using an active motion capture system (Optotrak Certus, Northern Digital Inc., Canada). Three light emitting diodes were fixed on a rigid body that was placed on the right side of the head approximately above the coronal suture of the skull. Optotrak data were sampled at 250 Hz (NDI First Principles, Canada).

2.2.5 Autonomic arousal measure

Electrodermal activity (EDA) was collected throughout the entire experiment to measure the physiological arousal changes (Venables, 1991; Venables & Mitchell, 1996).

Electrodes were placed on the thenar and hypothenar eminences of the non-dominant hand and were sampled at 2000 Hz (model 2502, CED, UK).

2.2.6 Psychosocial Questionnaires

Psychosocial questionnaires were used to assess the state of the individual. These questionnaires were presented before and after the No Threat conditions and the first Threat condition. Prior to each condition, on a scale of 0% - 100%, participants reported how confident they were in their ability to avoid falling and maintain balance (0%: not confident, 100%: completely confident). After each block, participants filled out a series of questionnaires on fear and anxiety. The fear of falling questionnaire was answered on a scale of 0% - 100% where 0% referred to no fear and 100% referred to feeling completely fearful. The 15-item anxiety questionnaire had a scale ranging through (1) "I don't feel this at all", (5) "I feel this moderately", and (9) "I feel this extremely". Each item on the anxiety questionnaire was then totaled for each subject leaving a grand score out of 135 for each condition where a larger score represented higher anxiety. This questionnaire was modified from previous experiments (Adkin, Frank, Carpenter, & Peysar, 2002) and has since been used in several experiments (Cleworth et al., 2012; Horslen et al., 2013). One item was removed from the previously used, 16-item questionnaire as it was not applicable for eyes closed conditions. See Appendix A for questionnaires.

2.3 Procedure and Conditions

At the start of the experiment, initial stance was calibrated in order to obtain a triggering window for platform perturbations. Participants were blindfolded and asked to stand relaxed on the rotating platform with a slight lean in the forward direction, their arms by their sides, and

their head tilted 18° upwards from Reid's plane. No vestibular stimulation was presented at this time.

After 1 minute of standing for the calibration, a seat was placed behind the participants and they were asked to sit down while keeping their feet in place on the platform. Participants were then informed about the details of the first condition, the No Threat condition, and were asked to fill out the balance confidence questionnaire before standing up. For this condition, participants stood in their initial stance for 3 minutes while receiving the SVS. No perturbations were expected or triggered during this time. Participants were then seated and asked to complete the remaining fear and anxiety questionnaires (Figure 2.2). After a minimum 5-minute seated rest period, participants were informed that throughout the upcoming condition (Threat condition) rapid platform perturbations would be presented at different times and in any direction. Their task was to quickly respond to the perturbation in order to maintain upright stance. During this condition, several blocks of SVS and platform perturbations were simultaneously presented to the participant. For each block, SVS was presented in two 5 minute bouts (10 minutes total). In order to remove the confounding transient effects from the onset of the SVS stimulation on EDA, SVS always started 10 seconds prior to platform perturbations. Participants were asked to complete the psychosocial questionnaires prior to and following the first and second blocks (Figure 2.3). In order to avoid fatigue, a mandatory seated-rest was provided in between each block. After experiencing approximately 150 perturbations, participants completed a final No Threat condition.

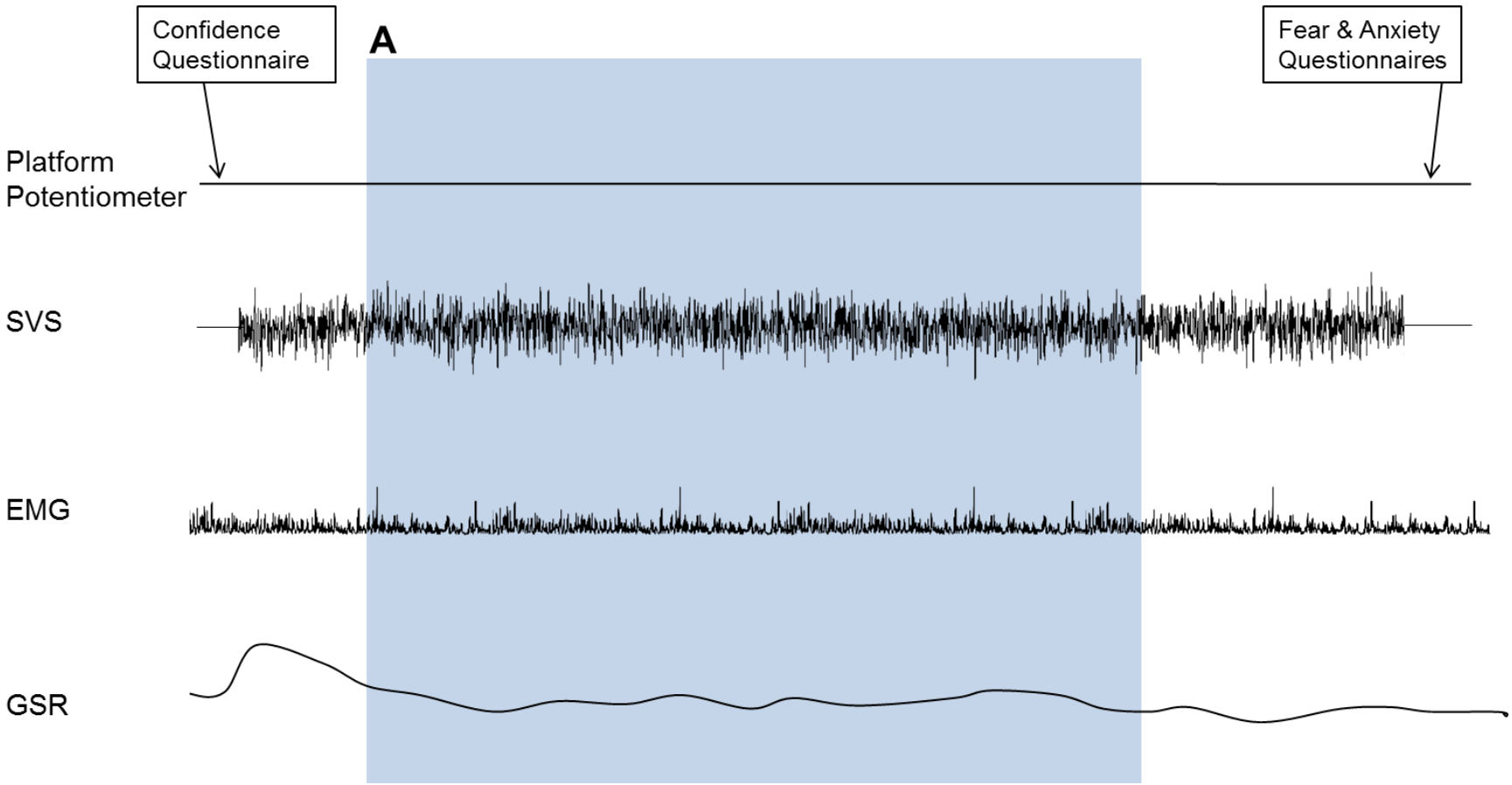

Figure 2.2 Schematic of the procedure for the No Threat condition. Window A indicates the clipped data throughout the quiet stance period. Questionnaires were answered before and after the trial.

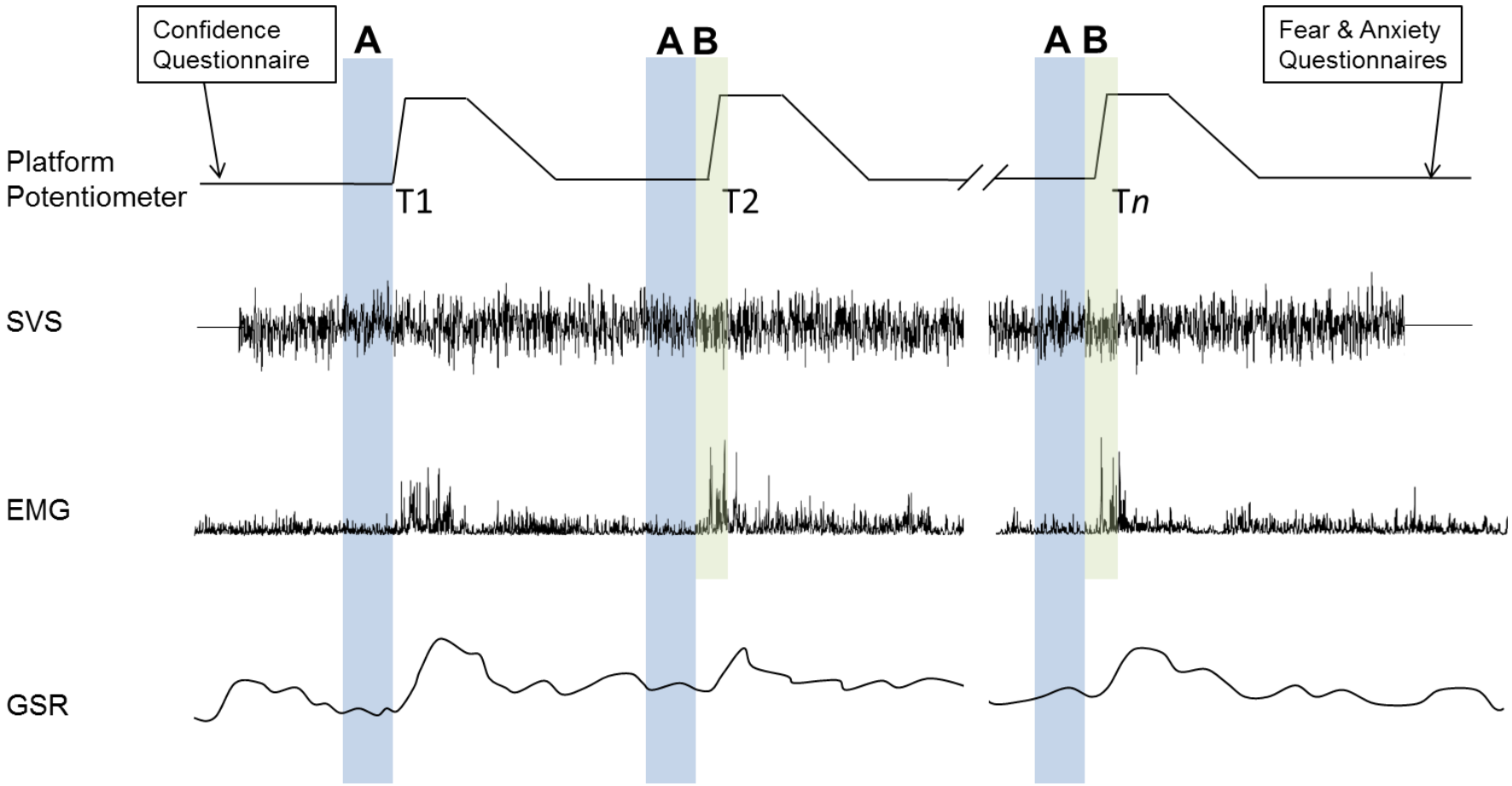

Figure 2.3 Schematic of the procedure for the first block in the Threat condition. Window A indicates the 5 seconds of data clipped prior to perturbation onset. Window B indications the 2 seconds of data analyzed around the perturbation onset. Questionnaires were answered before and after the first two blocks.

2.4 Calculations and Statistical Analyses

2.4.1 Phase-Dependent Modulation

Platform perturbation onsets were determined by the time the platform potentiometer recording exceeded the mean background recording plus 2 SD. For one subject, potentiometer recordings were not correctly read in for the first 5 perturbations and the platform trigger pulse was used to determine platform onset. This pulse occurred 111 ms prior to the perturbation onset and this timing was confirmed by calculating the difference between the trigger and the potentiometer recording onset for the remaining perturbations in the first block (111 ± 1 ms). The first trial was excluded from all analyses in order to ensure that a large first-trial effect did not skew the data (Tang, Honegger, & Allum, 2012). The postural response in each muscle was visually inspected to ensure that the patterns of activation were similar to previous work. A significantly larger number of platform perturbations were presented in this study compared to previous work. In order to compare these results other studies, only data from trials 2 to 11 were averaged and inspected.

Displacement, velocity, and acceleration onsets and magnitudes of angular head movement in the pitch and roll planes were also calculated. Head displacement was calculated from the rotation vector created by the rigid body placed on the participant's head. Head velocity was determined through calculating the derivative of the rigid body's rotation vector while head accelerations were calculated through taking the derivative of the calculated velocities. Onsets were defined as the point at which the trace exceeded 2 SD above the mean background value calculated 100 ms prior to the perturbation onset.

In order to investigate the evoked vestibular response in the muscles during different phases of the postural response, coherence between the SVS input and the EMG output was

calculated as a function of time (Blouin et al., 2011). A Morlet wavelet transform was created by performing continuous wavelet decompositions of the non-stationary EMG signal (Zhan, Halliday, Jiang, Liu, & Feng, 2006). The total EMG and SVS traces were split into disjoint segments containing the recordings starting at 5000 ms prior to and ending at 8000 ms after perturbation onset. To avoid improper interpretation of the data due to the cone of influence effect (Torrence & Compo, 1998) at the beginning and end of each segment, a significant amount of data were added to each end of segment. This additional data were then clipped out before interpretation and data 1 second prior to and after the perturbation was analyzed (Figure 2.3). Significant coherence throughout the postural response was determined when values exceed a 99% confidence limit (*Cl*) (Halliday et al., 1995). For a 99% confidence interval, the limit was determined as $Cl = 1 - 0.01^{1/(t-1)}$ where t is the minimum number of useable perturbations experienced by all participants.

As significant SVS-EMG coherence is typically seen throughout the 0 to 20 Hz bandwidth (Dakin et al. 2007; Luu et al. 2012; and also confirmed in this experiment with the pre-perturbation section of this thesis), total coherence was calculated as the average coherence values between 0 to 20 Hz for all muscles with an approximate resolution of 0.8 Hz. A change in coherence was determined by the time at which the average coherence values were outside of the mean ± 3 SD coherence values calculated 1000 – 500 ms prior to perturbation onset.

2.4.2 Threat of Perturbation Effects

All variables were compared between the No Threat and Threat conditions. To control for any averaging biases, data were appropriately clipped to the same time frame between conditions and participants. As the number of perturbations experienced by each participant was dependent on individual performance, the amount of data attained was variable. For this

reason, the participant who experienced the lowest number of perturbations dictated the amount of data clipped for each subject. For the No Threat condition, the clipped data began 10 seconds after SVS onset. In the Threat condition, the data consisted of concatenated 5 second bins from the quiet stance portion prior to perturbation onset. During this time, guiding instructions (i.e. leaning and head tilt adjustments) were minimal and were therefore most comparable to the No Threat conditions. For those who experienced more than the minimum number, data were concatenated from the start of block 1 until the minimum number of trials was reached. In order to maximize the effect of threat, responses were only taken from the first two blocks of the Threat condition. A minimum of 120 seconds of data obtained from each subject.

Autonomic arousal and psychosocial questionnaires responses for the Threat condition were averaged between the first and second block and then compared to the responses from the No Threat condition. The results from each questionnaire (Confidence, Fear, and Anxiety) and the EDA were then separately compared between the conditions using paired t-tests. A Kolmogorov-Smirnov test of normality was conducted on each variable and their between-conditions difference. If assumptions of normality were not met, the data were compared using a non-parametric Wilcoxon Signed Rank Test. Significance levels were set to $\alpha=0.05$.

To compare the vestibulo-muscular relationship, individual differences between the two conditions were compared. Similar to the clipping that was done for the EDA response, the total amount of data used by each participant was dictated by the participant that experienced the lowest number of perturbations. A 5 second segment was then taken prior to each perturbation. During these 5 seconds, all participants were standing quietly within their original stance boundaries.

SVS-EMG comparisons were done using correlations, in both the frequency (coherence) and time (cumulant density) domains, and gain of the signals. The total concatenated subject data were used for the coherence and gain estimates in order to obtain a general range of significance for each muscle. Within a condition, significant coherence was classified as the values that exceeded the 95% confidence level (Halliday et al., 1995). The pooled data resulted in correlation analyses using 1521 segments and a frequency resolution of 0.977 Hz (1.024 s/seg). With 1521 segments, the 95 % confidence limit was 0.0020. As significant SVS-EMG coherence is typically seen across bands of multiple frequencies and not only at a single frequency, coherence needed to remain above the 95% confidence level for more than 2 frequency points for it to be included. For each muscle, significant coherence differences between the two conditions (No Threat vs. Threat) were calculated using a Difference of Coherence test (Halliday et al., 1995). To investigate significant differences in gain between the conditions, a pointwise 95% confidence interval was calculated across each frequency and the points at which the confidence intervals did not overlap indicated a significant difference in gain. Differences in gain were only calculated for the frequencies that showed significant coherence in at least one of the two conditions. The percent difference between the Threat compared to No Threat condition were also calculated.

Cumulant density responses were analyzed on a subject-by-subject basis to determine the presence of significant responses for each muscle. Significant responses were determined when the values exceeded the 95% confidence interval. Peak and troughs were calculated for the short and medium latency (SL and ML, respectively) responses. In order to reduce the chances of Type I statistical error that can occur when conducting multiple statistical tests and because the two peaks are not completely independent, only the ML response was compared

between the two conditions. The peak ML response amplitude was calculated within 90 and 200 ms lag. If an individual's muscle showed no significant peaks during this time window, a non-significant peak was still calculated in order to run statistical tests. In a normal quiet standing position there is currently no evidence suggesting vestibular reflex differences between muscles on the right and left side of the body. However, due to the nature of the perturbation condition (90% of perturbations are in the rightwards direction), an asymmetrical response may occur. For this reason, a 2 (No Threat, Threat) x 2 (Left side, Right side) repeated measures ANOVA was conducted to compare the peak magnitude differences. Any significant interaction effects were then compared with a Tukey's HSD post hoc test. Alpha was set to 0.05.

In order to assess the relationship between the vestibular reflex and the emotional state, for the muscles that exhibited a significant difference in peak ML response amplitude between conditions, correlations between the change in peak ML magnitude and the changes in GSR were calculated. Of the different emotional factors, responses were only compared to GSR as this measure is most representative of the individual's state specifically during the quiet stance portion prior to the perturbations in the Threat condition.

EXPERIMENT 2: Simulating Slower Perturbation Velocities

There are several possible limitations that could arise from the wavelet transform of the data in experiment 1 (see results and discussion for more detail). Decomposition of the EMG response using the Fast-Fourier Transform (FFT) introduces power at low frequencies due to the sharp rise and gradual fall-off of the EMG burst evoked by platform perturbations. These low frequency components would result in a decrease in temporal resolution of the wavelet

transform. To address this limitation, time-dependent power spectral density estimates of the EMG recordings were conducted on all muscles in experiment 1.

Additionally, due to the large EMG bursts, vestibular reflexes may have become masked. In order to further understand and overcome this analytical limitation, data were created to simulate EMG responses to different platform perturbation velocities. Theoretically, a decrease in platform perturbation velocity would result in decreased muscle response activation. At a certain point, the EMG burst would no longer overpower the vestibular reflexes and any changes in coherence could then be observed.

The number of trials for each velocity was matched to the total number of trials analyzed in experiment 1 (105 trials, see results). Artificial input and output signals were created using MATLAB (R2012b, MathWorks Inc). Input signals were copied from the SVS signal generated in experiment 1. The output signal included a filtered copy of each input signal, randomly generated noise, and a single copy of an EMG trace taken from one muscle in a participant from experiment 1. See Figure 2.4 for a visual representation of the output signal components and the following section for a detailed description of each output component.

To simulate the actual EMG response in experiment 1, certain transformations of the data were conducted to match the theoretical relative contribution of each output component (SVS, noise, and EMG) to the overall power of the signal. Largest SVS-EMG coherence is often observed around 5 Hz with slowly decreasing coherence and power at smaller frequencies and rapidly decreasing coherence and power larger frequencies (Blouin et al., 2011; Carriot, Jamali, Chacron, & Cullen, 2014; Dakin et al., 2007). In order to capture the peak coherence around 5 Hz, a 1st order bandpass butterworth filter between 0.5-5 Hz was applied to the SVS output signal (Figure 2.2 B). This resulted in large power between 2-10 Hz.

The randomly generated noise had a trend of increasing power towards the higher frequencies. Thus, to ensure that the additional noise was evenly distributed throughout all frequencies, a 15 Hz 1st order low pass butterworth filter was applied. This noise was then amplified by 10, resulting in peak noise power 3 times larger than SVS peak power and a decrease in the input-output coherence (see Figure 2.5). Finally, a 950 ms EMG trace was extracted from a muscle response in experiment 1, amplified by 1000 to result in a peak power 3 times greater than the combined SVS and noise signal. This EMG trace was added to the combined SVS + noise trace 50 ms after the hypothetical perturbation onset (timed according to the data from experiment 1). Instead of including a full 2 seconds of EMG data, only a segment of the EMG response was added to the output signal to ensure that the following signal manipulations for each condition were directly related to the change in EMG burst amplitude and not background activity.

Ten different conditions were created to simulate different sizes of muscle responses. The first condition consisted of all the components mentioned above. In following 9 conditions, the EMG trace was reduced by 90, 80, 70, 60, 50, 40, 30, 20, and 10% of the original EMG trace. All other signals remained the same. Time-dependent coherence analysis using continuous wavelet transforms were conducted between the original SVS input signal and the 10 output signals. Time-dependent power analyses were also conducted on the data.

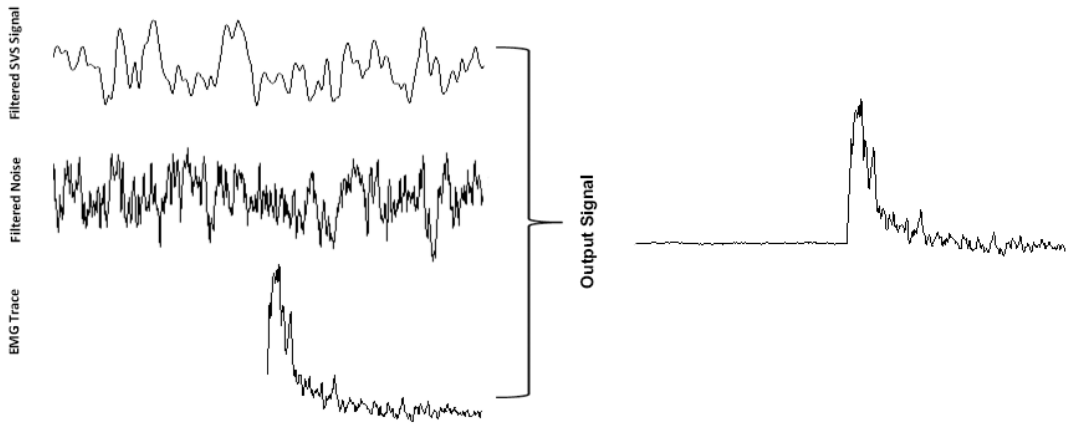

Figure 2.4 Simulated output signal and its components: SVS, noise, EMG. Each trace is optimized to each signal.

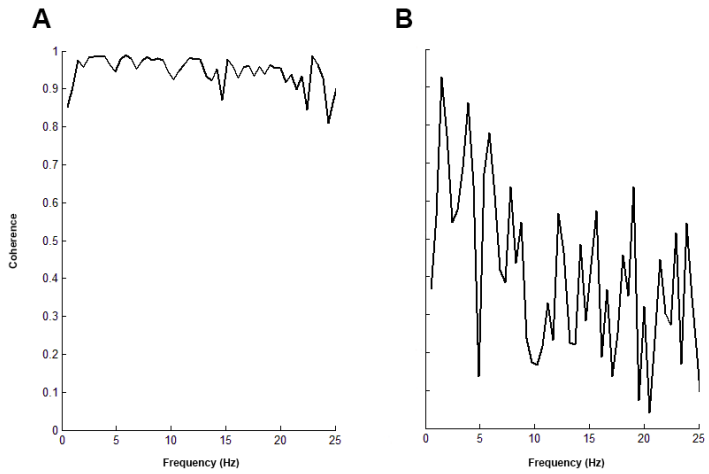

Figure 2.5 Change in coherence when relative contribution of each signal is manipulated. The first trace (A) is the coherence estimate between the input SVS and the output SVS + unamplified noise. The second trace (B) is the coherence estimate between the input SVS and the output SVS + amplified noise.

EXPERIMENT 3: Slower Velocity Perturbations

The simulated data revealed continuous coherence throughout the segment when the EMG trace amplitude was 10% of the original trace (see results for more detail). Based on these results, a single subject (Female, Age: 22, Height: 157.48 cm) was tested to determine if changes in platform perturbation velocities elicited changes in muscle response amplitudes comparable to the simulated data. 10 trials of 5 different perturbation velocities (55, 28, 18, 13, 11.5°/s) were presented. The first trial was excluded from the data set and the average response for each muscle at each perturbation velocity was then measured. Results from this subject indicated that the size of the response decreased with decreased perturbation velocities (Figure 2.6) while the onset latency and shape of the response remained relatively similar. This was consistent with the simulated data. With the slowest perturbation velocity (11.5°/sec), numerous muscles showed an approximate 90% decrease in peak EMG activity. In some muscles, such as the mGAS, PARA, left GM and right SOL, EMG activity did not decrease by the same amount. These muscles (except mGAS), however, still showed a similar scaling effect with the perturbation velocities.

Based on these findings and the results from the simulated data, a subgroup of participants was tested at the slowest velocity. Three participants (Female: 1; Age: 24 ± 3 Height: 167.55 ± 7.49 cm) experienced a protocol similar to experiment 1. Participants first experienced a 3-minute No Threat condition followed by the Threat conditions with approximately 90 perturbations (10% of these were catch trials). A smaller number of perturbations at the same velocity as in experiment 1 were presented to avoid fatigue during the following perturbations. Correlation analysis of the data from experiment 1 also revealed similar results when 75 compared to 105 perturbations were included (see Appendix C for

these results). Participants then experienced the second No Threat condition for another 3 minutes. Finally, participants experienced an additional 90 perturbations at a slow velocity (11.5°/sec). All aspects of the collection, calculation, and analysis were the same as experiment 1, however, due to technological failure, SVS stimulation was provided through a new stimulator (Stimsola, BioPac Systems Inc., CA, USA) that produced the same output. As the protocol for the threat of perturbation questions was similar to experiment 1, these three participants were included in the overall data analysis of that portion in experiment 1.

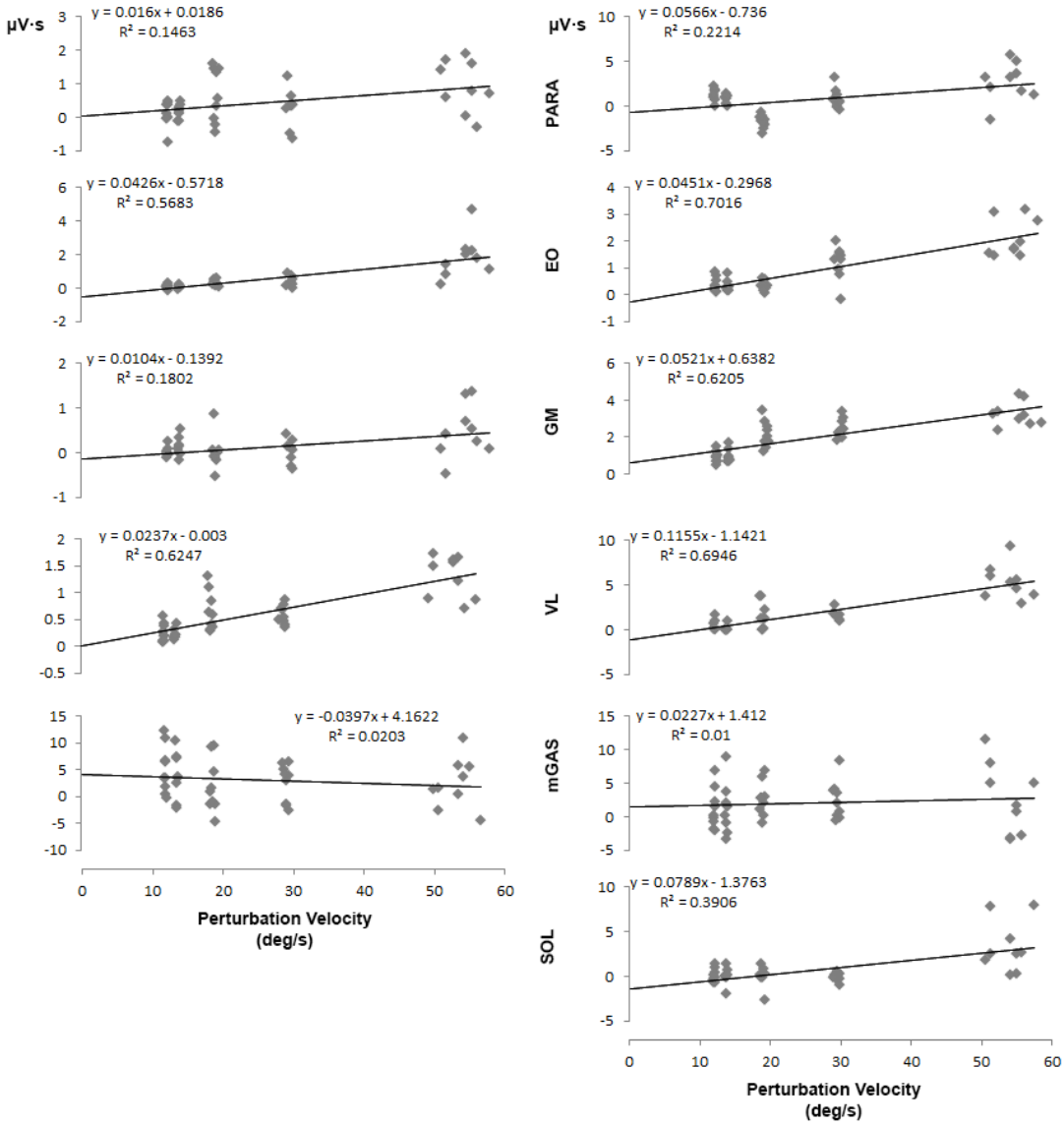

Figure 2.6 Muscle activation during the balance-correcting window (120-220 ms) for each muscle during different perturbation velocities. The left column displays muscles of the left side of the body while the right column displays muscles on the right side of the body.

Chapter 3: Results

3.1 Participants and Perturbations

All 16 participants (13 from experiment 1 + 3 from experiment 3) did not express any discomfort with the platform perturbations and, after the first block, all participants were able to recover and maintain upright stance without assistance. In general, participants were able to tolerate the vestibular stimulation and many participants did not perceive the stimulation after experiencing several blocks. One person withdrew from experiment 1 because they felt uncomfortable tenderness on their mastoid processes after 20 minutes of stimulation. For 2 other participants, there were technical difficulties with the vestibular stimulation system resulting in an incomplete data set. Therefore, data from 13 participants were included in this study.

3.2 Phase-Dependent Modulation

EXPERIMENT 1

3.2.1 Angular Head Movement

The platform perturbation induced head movements in both the pitch and roll planes with larger movements observed in the roll direction. The perturbation resulted in a pitch angular displacement in the chin down direction with average onset latency of 84 ms and maximum displacement of 2° at 596 ms. Velocity onset occurred 60 ms after perturbation onset and reached a maximum of 13°/s at 132 ms while acceleration onset occurred at 36 ms with a maximum of 473°/s^2 at 280 ms. Head roll angular displacement towards the left occurred 64 ms after perturbation onset with a maximum displacement of 2° and 188 ms. Velocity onset occurred at 48 ms with a maximum of 23°/s at 112 ms and acceleration onset was 40 ms with a maximum of 762°/s^2 and 80 ms (Figure 3.1).

3.2.2 Postural Response

Distinct patterns of muscle activation were observed for all recorded muscles. For some muscles such as the left SOL and right PARA, a distinct stretch reflex and balance-correcting response onset was observed while for other muscle, such as the right mGAS and right SOL for example, separate onsets could not be determined. An unloading response was also observed in the left PARA, right GM, and right VL shortly after the onset of the perturbation. Onsets of muscle activation are displayed by an arrow in Figure 3.2. If two distinct onsets were observed, then the onset of only the second burst (the balance-correcting response) was reported.

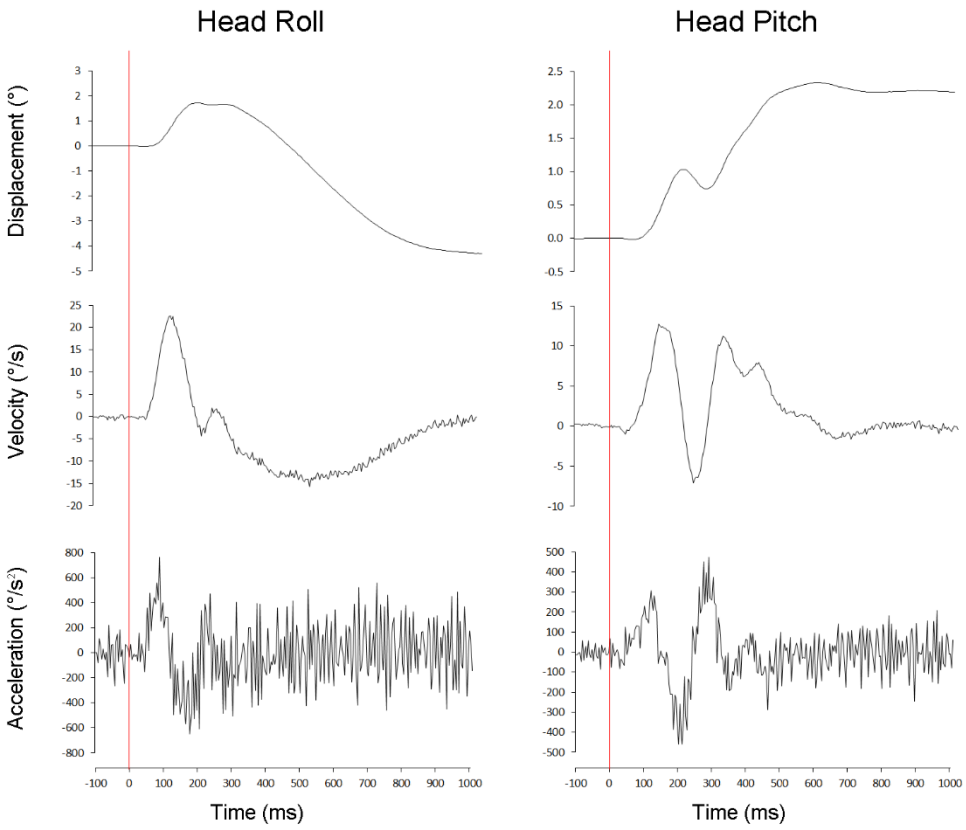

Figure 3.1 Average angular head movement and velocity traces from the first ten perturbations in each subject. Positive values for the displacement traces indicate head down for the pitch plane and head left for the roll plane. The vertical line indicates the onset of platform perturbation.

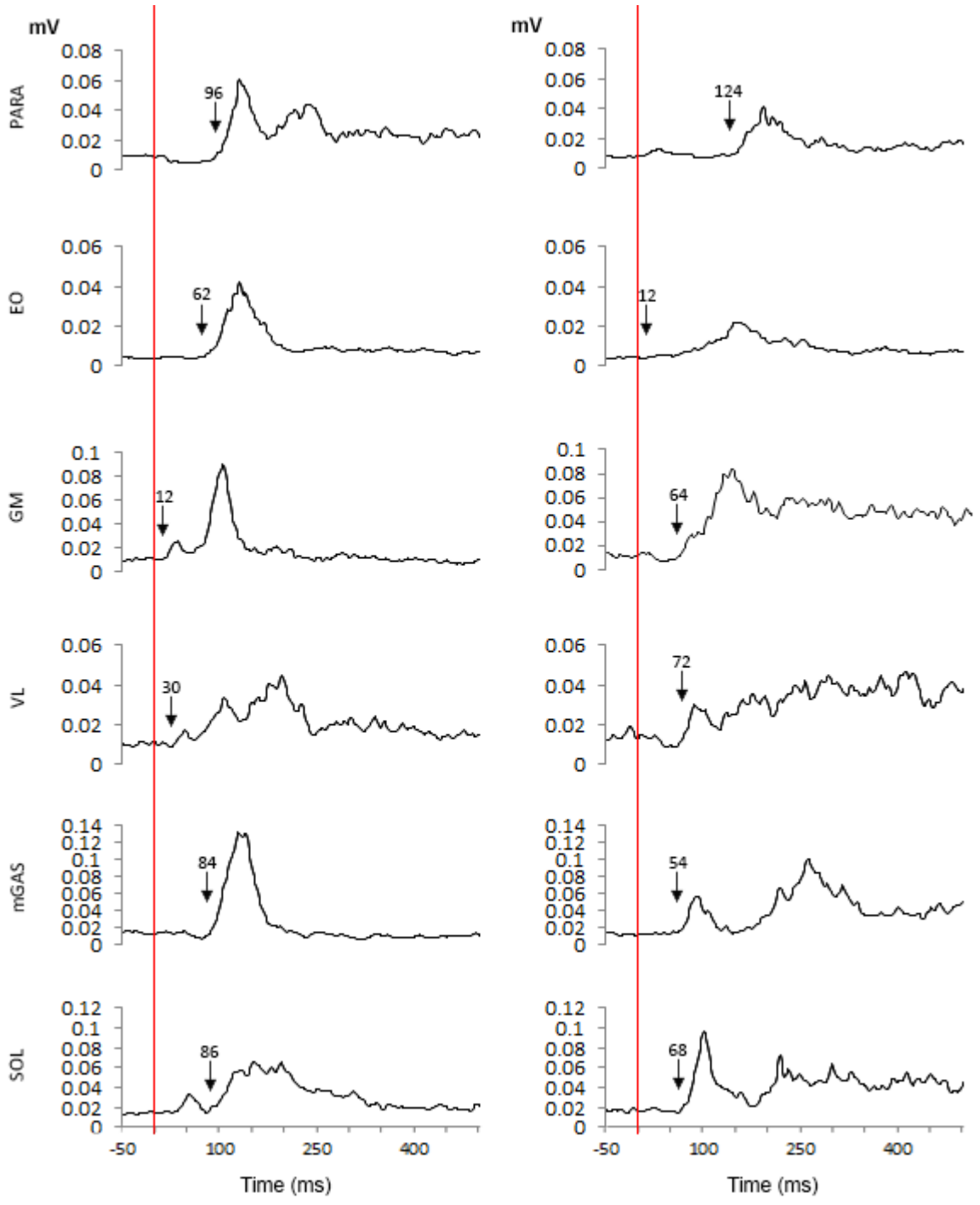

Figure 3.2 Average EMG traces (mV) of 10 trials from each participant. The left column displays responses from muscles on the left side while the right column displays muscles on the right side of the body. Muscle onsets are noted in ms by the arrows on the trace. The red line indicates the onset of the platform perturbation at time 0 ms.

3.2.3 Vestibular Reflex

Time-dependent coherence analysis revealed changes in the vestibular reflex during postural responses. In all muscles, peak coherence values were observed in frequencies between 0.5-4.4 Hz. Left GM showed the highest coupling with a peak coherence value of 0.12 while left EO showed the lowest coherence peaking at 0.01 (Figure 3.3 and Figure 3.4). Postural perturbations induced significant decreases in coherence in the majority of muscles compared to their baseline coherence values prior to the perturbation. When the coherence values were collapsed across frequencies 0-20 Hz, this drop in coherence occurred as early as 412 ms prior to the perturbation in the left GM muscle and as late as 24 ms after the perturbation in right VL. It is important to note that the time resolution at the lower frequencies may result in a shifting of the actual timing in coherence change. For this reason, the frequency that exhibited the highest baseline coherence was determined and used to calculate time-dependent modulations of the vestibular reflex. Drops in coherence occurred at different times for each muscle relative to the perturbation: -36 to +642 ms for the left PARA, -88 to +180 ms for the right PARA, -144 to +668 ms for right EO, +32 to +272 ms for the left GM, +24 to +286 ms for the right GM, +194 to +578 ms for the right VL, +114 to +294 ms for the left mGAS, +294 to +312 ms for the right mGAS, -102 ms onwards for the left SOL and +82 to +530 ms for the right SOL. No significant coherence values were observed in the left EO and therefore, changes in coherence were not calculated for this muscle. For the left VL, background coherence was not significant before the perturbation but increased above the 95% confidence level at 296 ms after the perturbation onset.

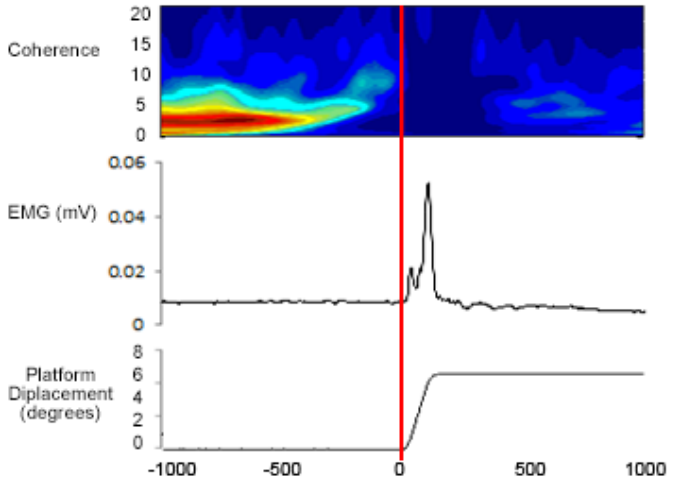

Figure 3.3 Relative timing of coherence offset compared to perturbation and EMG onset. Coherence, muscle activity, and platform displacement plots for the left GM muscle. The colour scale for the coherence plot indicates the strength of the coherence with red being the highest and dark blue being the lowest coherence. The vertical line indicates the onset of the platform perturbation.

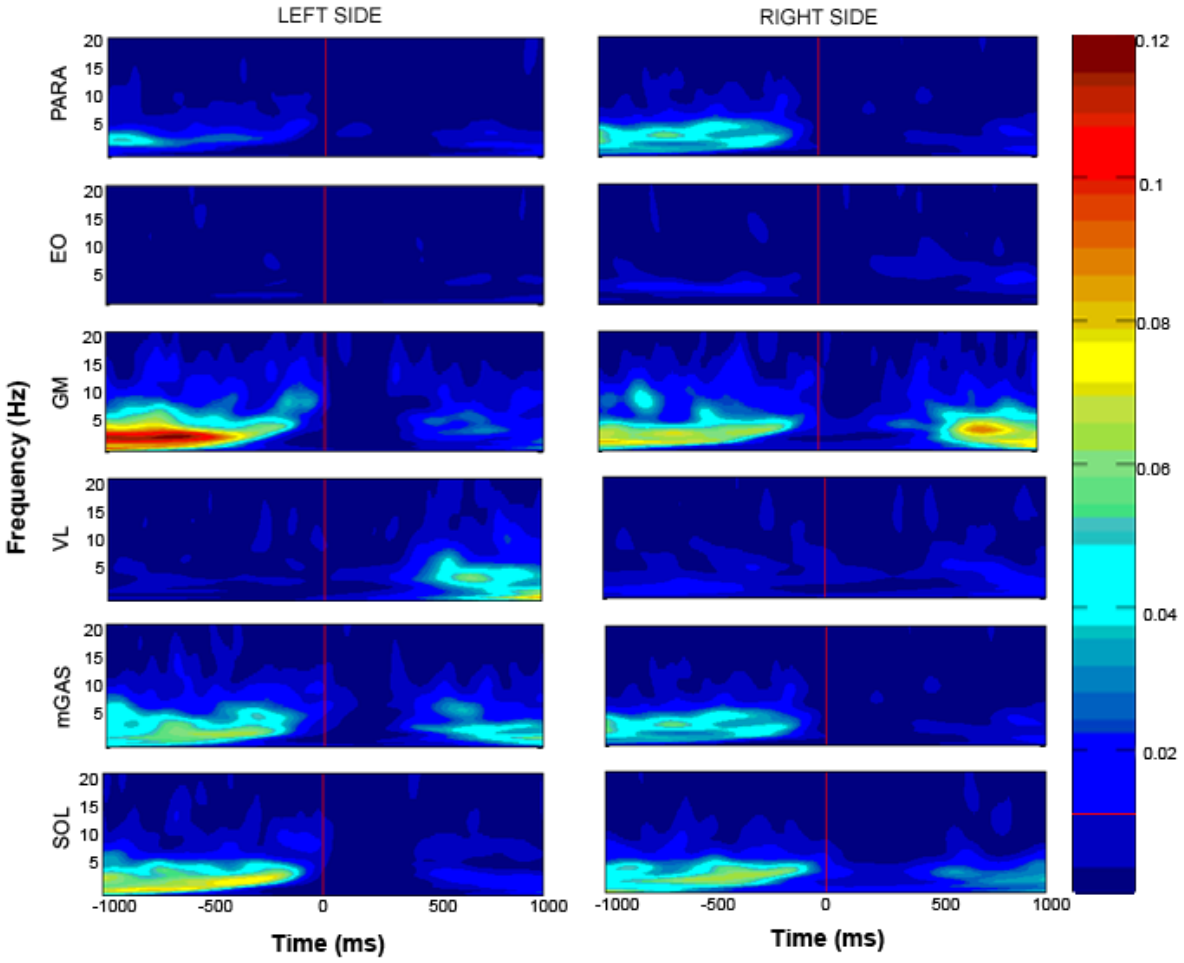

Figure 3.4 Time-dependent coherence plots from wavelet transform results. Each box represents the response from a single muscle across time for frequencies 0-20 Hz. The left column displays muscles on the left side of the body while the columns on the right display muscles on the right side of the body. The red vertical line crossing the middle of each box indicates the onset of the platform perturbation. The color scale indicates the strength of the coherence estimate with red being the highest coherence observed and dark blue indicating no coherence. The colour indicating the 99% confidence limit (0.00442) is marked by a red line on the colour scale.

EXPERIMENT 2

Time-dependent power spectral density estimates of the EMG responses in experiment 1 (Figure 3.5) reveal an increase in power at a time that corresponded to the decrease in coherence observed in Figure 3.4.

Simulated data revealed time-dependent coherence estimates in all output signals. Largest coherence occurred below 5 Hz and little coherence occurred above 20 Hz. Simulated data revealed an offset of coherence that coincided with the increase in power for each frequency (Figure 3.6). As the size of the EMG trace decreased from 100 % to 5 %, the power at each frequency also decreased while the timing of coherence offset matched the timing of power increase. The first level at which the output signal remained coherent with the input signal throughout the whole response window occurred at 10 % of the maximum EMG trace.

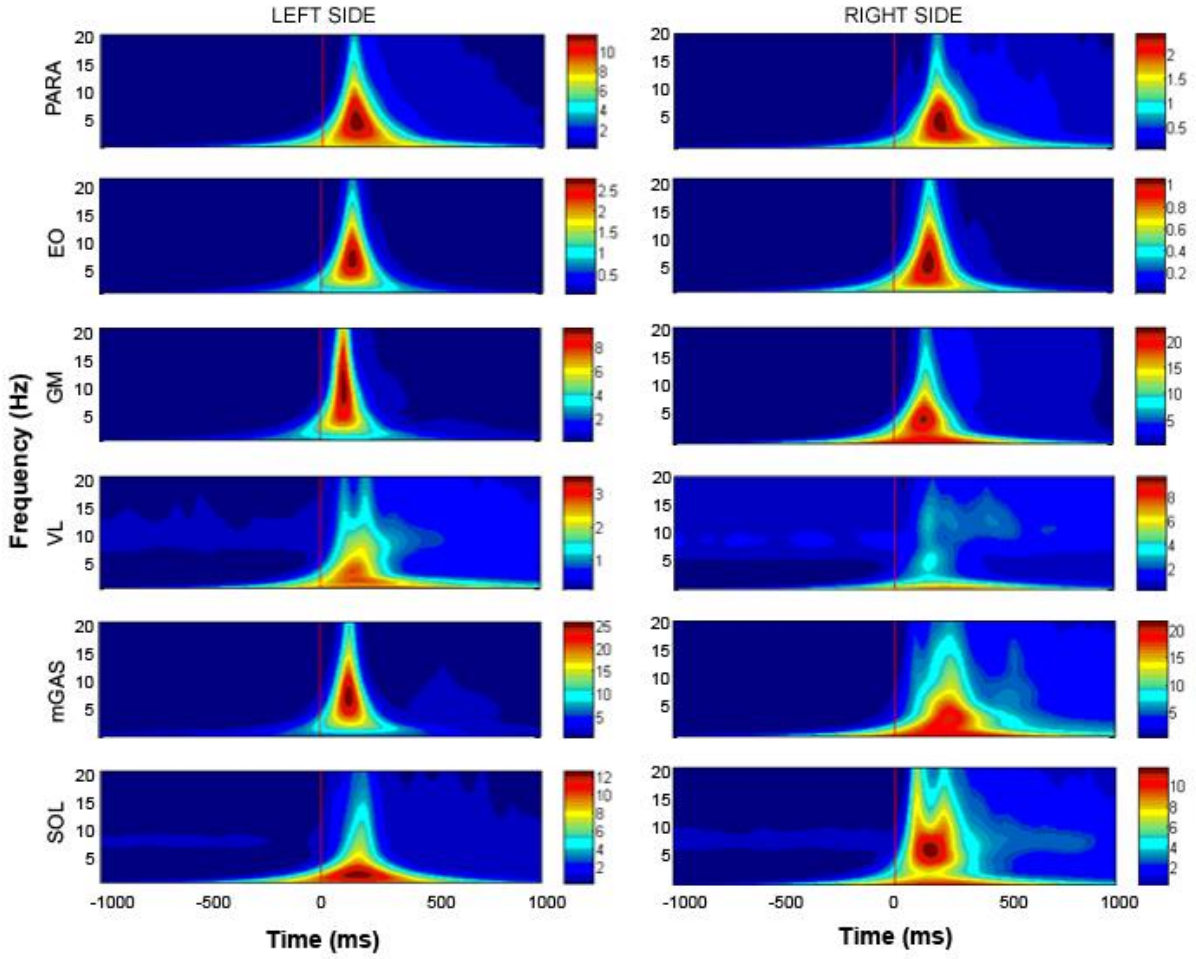

Figure 3.5 Time-dependent power spectral density estimates for each muscle. The column on the left represents the muscles on the left side of the body while the column on the right represents the muscles recorded from the right side of the body. Responses are optimized for each muscle response.

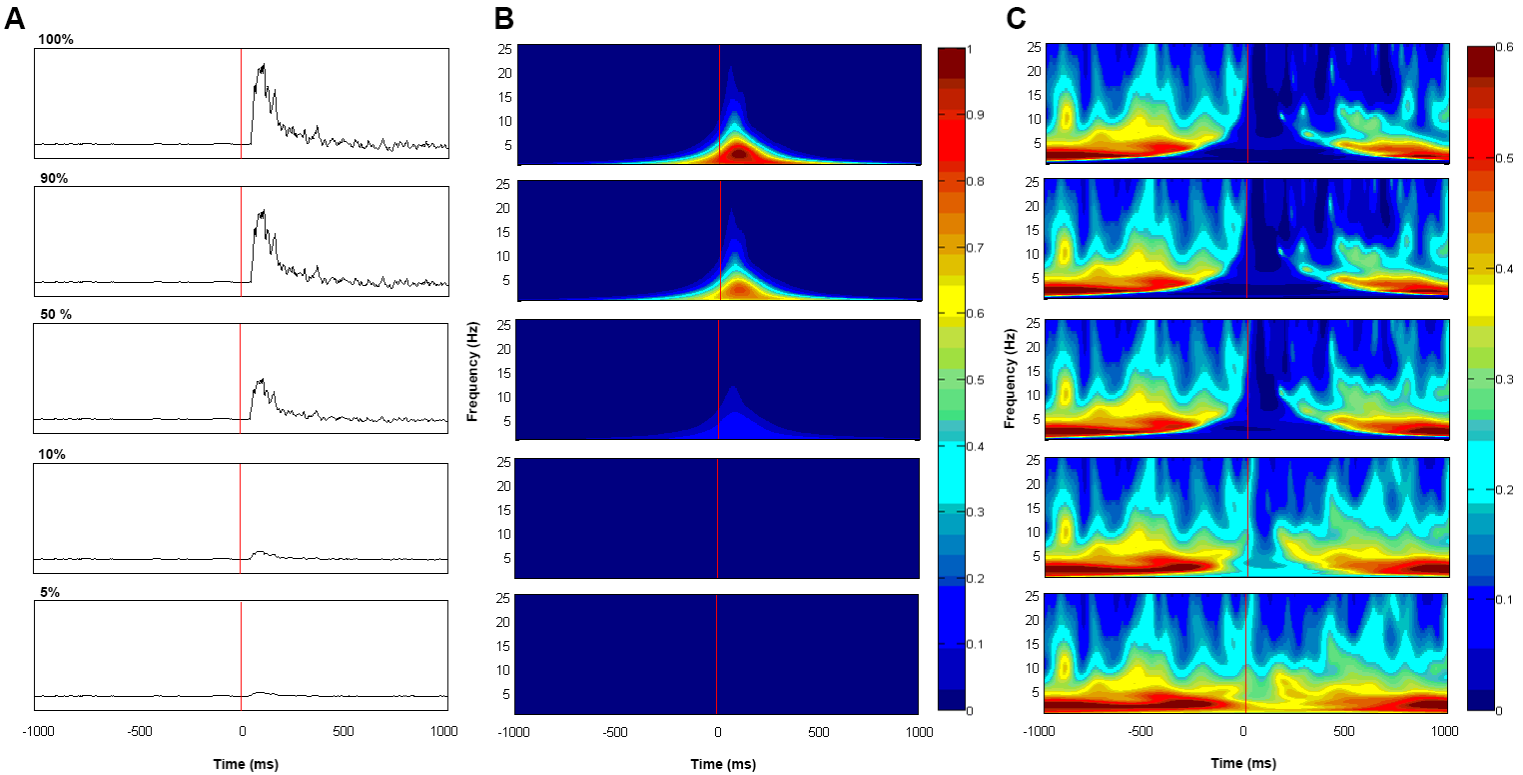

Figure 3.6 **A** Generated output signal. **B** Time-dependent power spectral density estimates of the output signal. **C** Time-dependent coherence analysis between the input and the output signals. Each row represents the results from a subset of simulated data. The first row contains a simulated muscle trace at 100 %, the trace in the second row is 90% of the original, third is 50 %, fourth is 10 %, and the fifth row represents a signal with 5 % of the original EMG trace. The vertical line at time zero of each plot represents the theoretical onset of platform perturbation.

EXPERIMENT 3

3.2.4 Angular Head Movement

In the pitch plane, the FAST platform perturbation induced a head pitch displacement in the chin down direction with an average onset latency of 88 ms and a maximum displacement of 3° at 568 ms. Velocity onset occurred 68 ms after perturbation onset and reached a maximum of 19°/s at 288 ms. Pitch acceleration occurred at 48 ms with a maximum of 729°/s^2 at 236 ms. Head roll displacement towards the left occurred 52 ms after perturbation onset with a maximum displacement of 3° at 200 ms, velocity onset occurred at 48 ms with maximum of velocity 35°/s at 116 ms, and acceleration onset was 44 ms with a maximum of 932°/s^2 at 76 ms (Figure 3.7).

Average angular head movements for the SLOW platform perturbations were smaller than the FAST perturbations. Pitch displacement had a maximum displacement of 2° at 980 ms. No specific onset could be detected. Roll displacement onset occurred at 64 ms with a maximum and 1° at 244 ms, velocity onset was 48 ms with a maximum of 8°/s at 128 ms. With the current methods, no clear onset of acceleration could be detected.

3.2.5 Postural Response

For the FAST perturbations, patterns of muscle activation were similar to those observed in experiment 1 (Figure 3.8 and Table 3.1). Postural responses for the SLOW perturbations were much smaller than the responses induced by the FAST perturbations (see Table C1 in Appendix C for exact values). In general, the EMG patterns were similar to those induced by the FAST perturbations, however, stretch reflexes were absent in right VL and right mGAS and very little activity was observed in the right PARA and bilateral EO muscles.

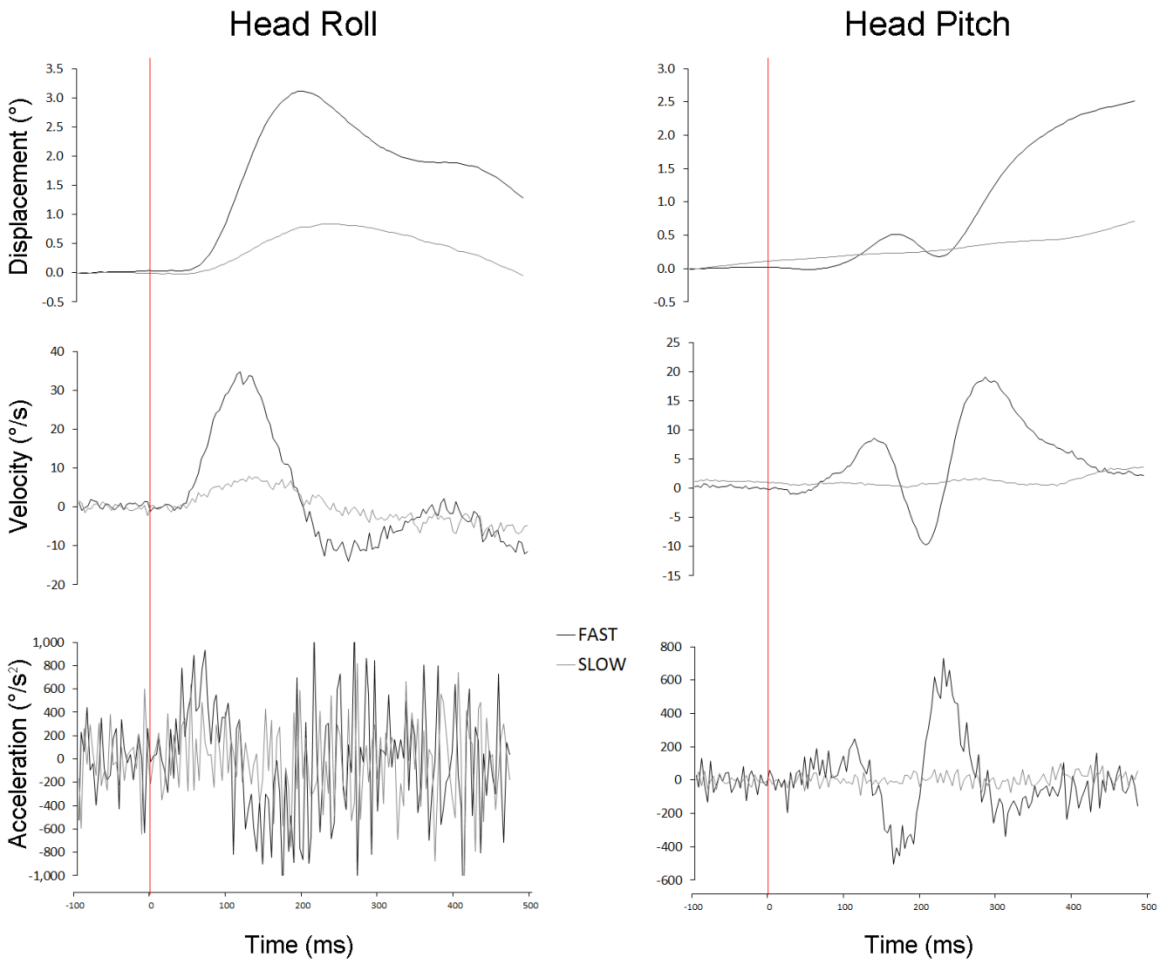

Figure 3.7 Average angular head movements induced by the first ten FAST and SLOW platform perturbations. The black lines are the movements induced by the FAST perturbations while the grey lines are the movements induced by the SLOW perturbations. The vertical line indicates the onset of the platform.

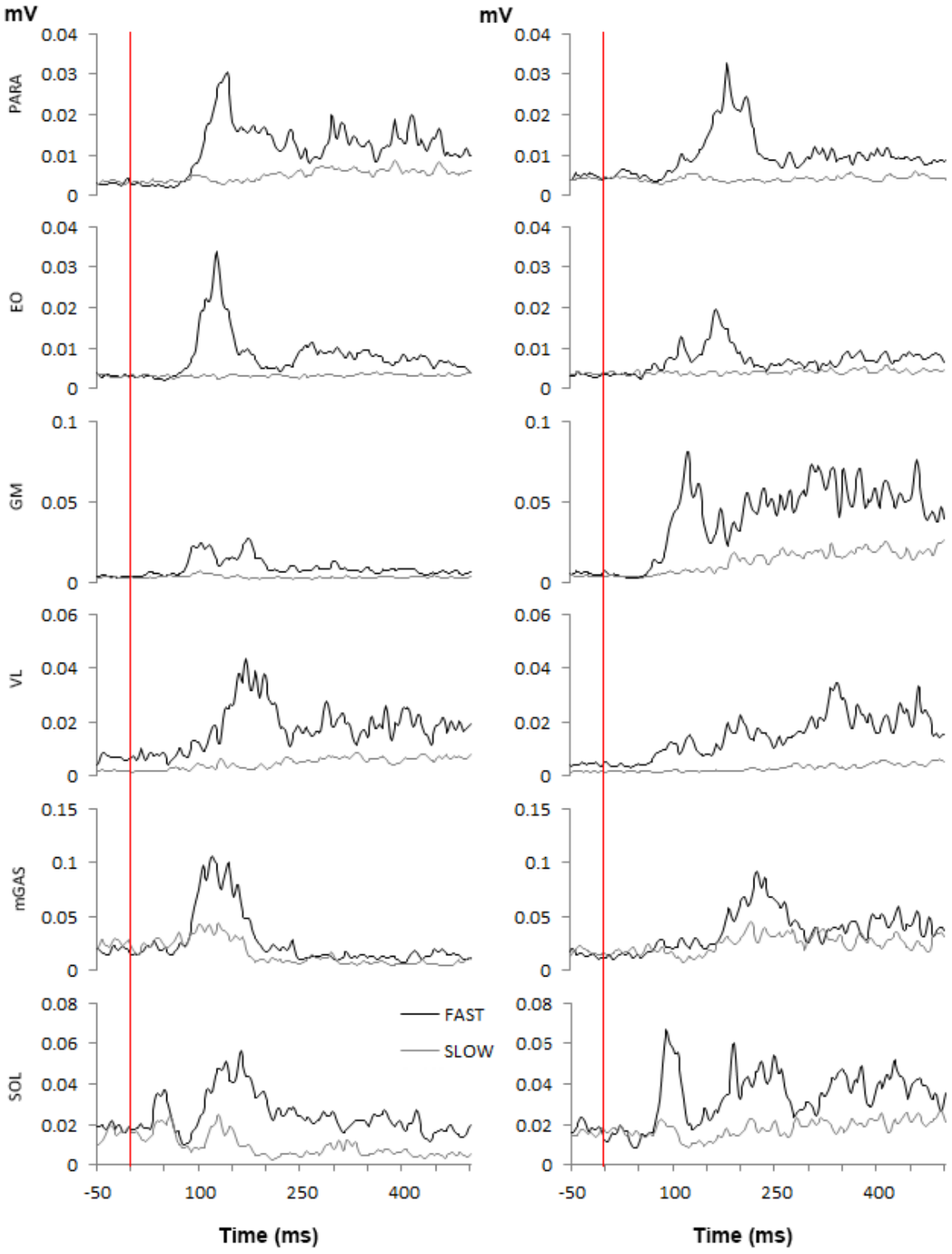

Figure 3.8 Average (N=3) postural responses resulting from the FAST (black) and SLOW (grey) perturbations.

Table 3.1 Onset latencies of muscle response after the perturbation onset. n.s. indicates no significant onset detected 2SD above the mean background value.

	Left Side		Right Side	
	Fast Perturbation	**Slow Perturbation**	**Fast Perturbation**	**Slow Perturbation**
PARA	81 ms	86 ms	90 ms	n.s.
EO	69 ms	n.s	59 ms	n.s.
GM	55 ms	51 ms	65 ms	84 ms
VL	87 ms	91 ms	69 ms	n.s.
mGAS	83 ms	85 ms	73 ms	n.s.
SOL	32 ms	34 ms	74 ms	77 ms

3.2.6 Vestibular Reflex

Time-dependent power-spectral density and coherence estimates for the FAST perturbation revealed similar results to those from the first experiment. A decrease in coherence was observed around the onset of platform perturbation where the timing of offset was related to the power of the EMG response (Figure 3.9 and 3.10). Note that the wavelet transform is not as clear as experiment 1 and is noisier due to the lower number of subjects included in this experiment (N=3 versus N=10).

When the participants experienced the SLOW perturbation, the EMG power spectral density profile was drastically smaller and no longer included high overall power around the onset of platform perturbation. In all muscles, except mGAS, the highest power shifted to a later time. The highest EMG power from the mGAS muscles were still observed around the onset of the platform, however they were smaller than the observed EMG power from the FAST perturbations (Figure 3.9).

The vestibular responses for these SLOW perturbations were also different than those calculated from the FAST perturbations. Due to the amount of noise observed with analysis of only three subjects, changes in coherence were only visually inspected and significant drops from the collapsed frequencies were not calculated. For several muscles, coherence did not drop until after the perturbation (PARAs), cohered around the onset of perturbation (GMs and mGASs), or maintained coherence throughout the whole window (SOLs) (Figure 3.10).

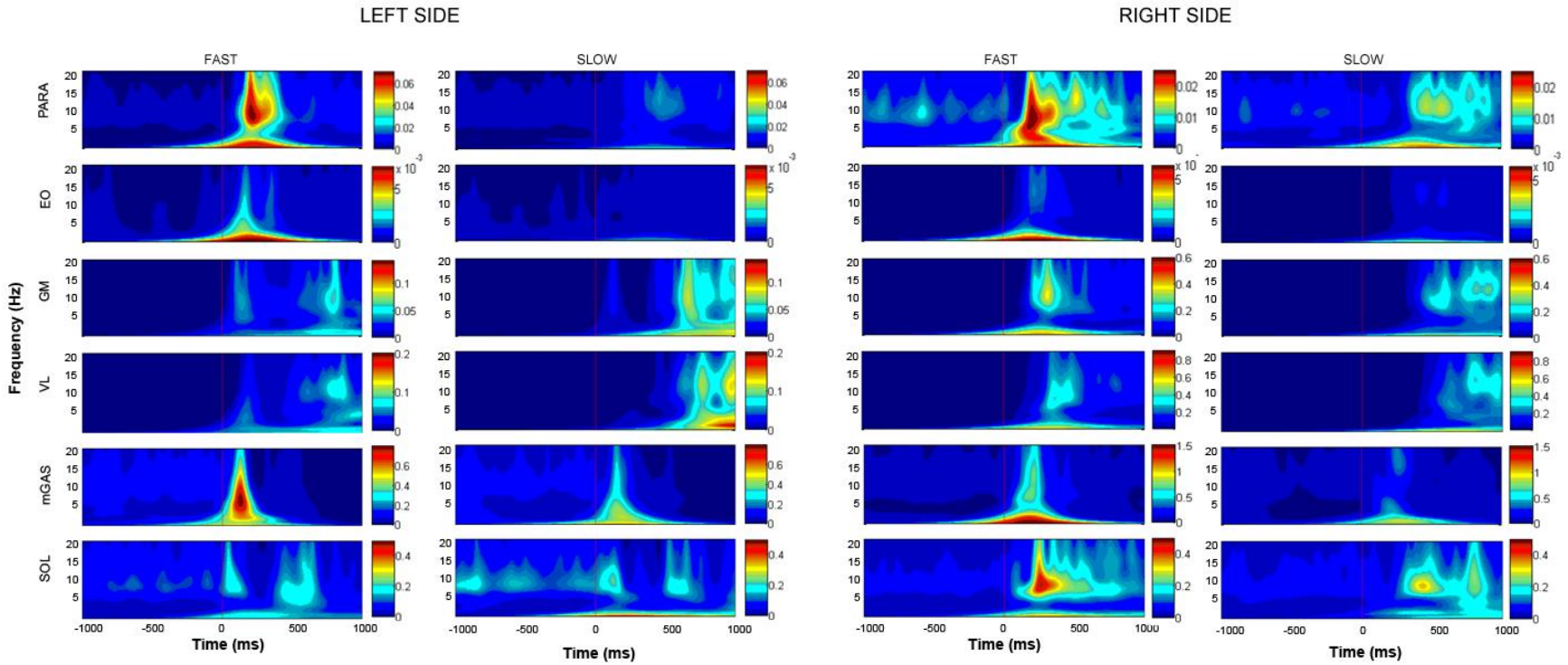

Figure 3.9 Time-dependent power spectral density estimates for FAST and SLOW perturbations. Plots represent the average response from three subjects. Colours indicate the strength of the power where red represents the highest value and blue represents the lowest

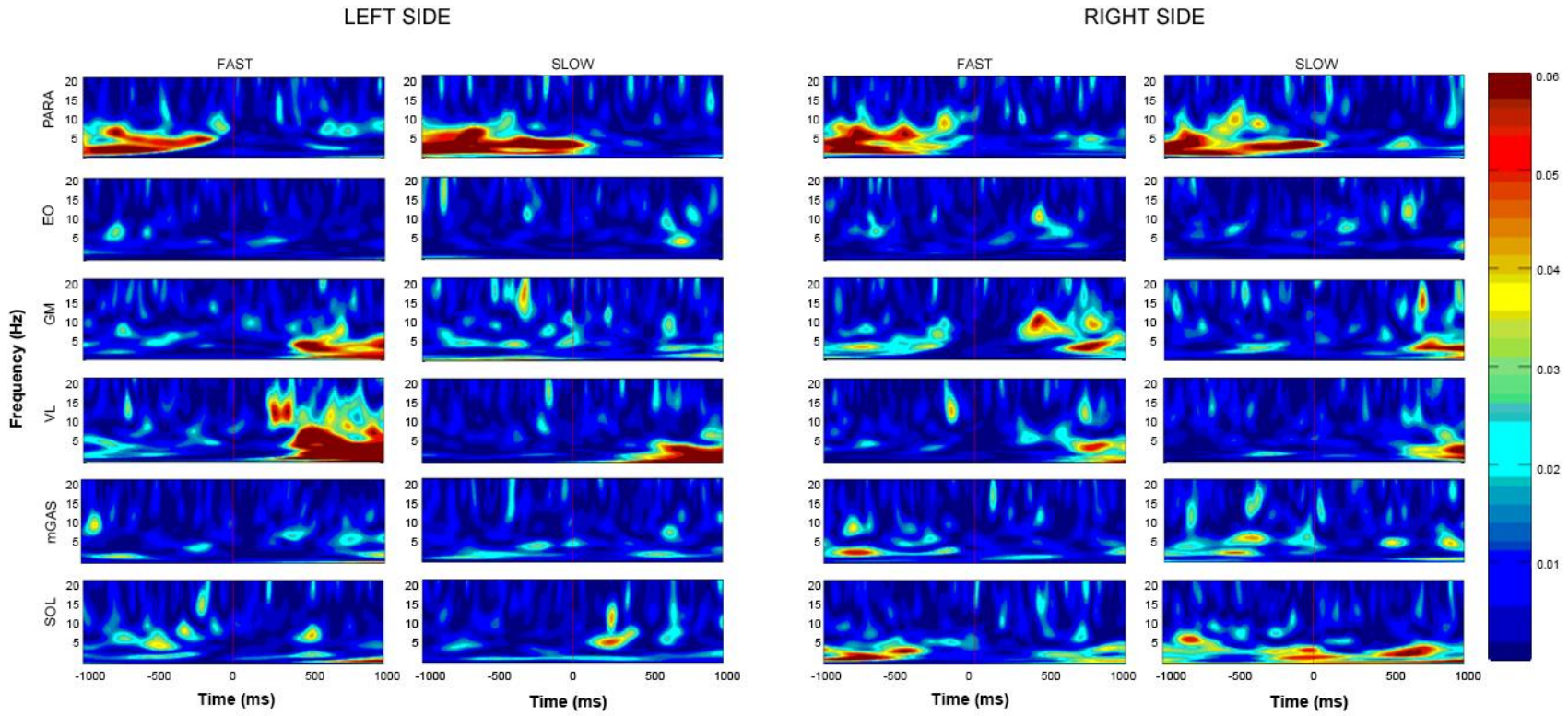

Figure 3.10 Time-dependent coherence estimates for FAST and SLOW perturbations. Plots represent the average response from three subjects. Colours indicate the strength of the coherence where red represents the highest value and blue represents the lowest.

3.3 Threat of Perturbation

3.3.1 Autonomic Arousal and Psychosocial Measures

For several participants EDA activity recordings were inaccurate for the last No Threat condition as the electrodes either became fully saturated with sweat or physically shifted on the palm of the hand. Therefore, comparisons for all variables were only done between the first No Threat condition and the Threat condition.

Two participants were excluded from the EDA analysis because their EDA exceeded the maximum recordable value. The Kolmogorov-Smirnov test indicated a violation of normality for the EDA measure and fear of falling questionnaire results. Therefore, the non-parametric Wilcoxon Signed Rank test was performed and indicated a significant increase in both EDA ($z(11) = -2.312$, $p=0.021$, $r = 0.70$) and fear of falling ($z(13) = -2.596$, $p = 0.009$, $r = 0.72$) response for the Threat compared to No Threat conditions. No violation of normality was observed for the anxiety and balance confidence questionnaires and a paired t-test was performed. The Threat condition resulted in a significantly larger anxiety score for the Threat versus No Threat conditions ($t(12) = 4.91$, $p < 0.000$, $r = 0.82$) while a significant decrease in balance confidence ($t(12) = -3.759$, $p=0.003$, $r = 0.74$) was observed in the Threat condition.

3.3.2 Vestibular Reflex

For the No Threat condition, significant coherence was observed in all recorded muscles except bilateral EOs. Highest coherence was calculated in the right PARA with a peak of 0.08 at 2.9 Hz with a significant coherence range between 0-10 Hz (Figure 3.11). The peaks and ranges of significant coherence varied between muscles (Table 3.2).

There was a significant increase in coherence in the Threat condition for all muscles, except PARAs, that had SVS-EMG coherence in the No Threat condition (Table 3.2).

Likewise, although no significant coherence was present in the right EO muscles during the No Threat condition, SVS-EMG coherence became significant for a small frequency bandwidth (1-5 Hz) under the Threat condition. For several muscles, the frequency bandwidth of significant coherence also increased in the Threat condition (see Table 3.2). The Difference of Coherence tests indicated significant differences at several frequencies within each muscle's range of significant coherence. Unlike previous work, these significant differences were not grouped around specific frequency ranges. For this reason, an alternative method was used to classify the differences between the two conditions. A percentage value was calculated for each muscle to represent how much of the coherent spectrum was statistically different across the threat conditions. A significant increase in coherence in the Threat condition compared to the No Threat condition was seen in 25 % for right EO, 12 % for left GM, 13 % for right GM, 31 % for left VL, 24 % for right VL, 11 % for left mGAS, 5 % for right mGAS, 38 % for left SOL, and 29 % of the significant frequencies for right SOL. Significant decreases in coherence were seen in 7 % for left PARA, 12 % for right PARA, 8 % for left GM, and 11 % for left mGAS.

In the majority of muscles, cumulant density estimates revealed a biphasic response with a SL and ML peak (Figure 3.12). The lag of the responses ranged from 51-84 ms and 93-163 ms for SL and ML responses, respectively for each muscle. For bilateral GMs, VLs, mGASs, and SOLs, the magnitude of the ML peak showed a significant main effect of condition where the Threat condition was larger than the No Threat condition (GM: $F(1,12) = 11.57$, $p = 0.005$, $r = 0.70$; VL: $F(1,12) = 7.19$, $p = 0.02$, $r = 0.61$; mGAS: $F(1,12) = 7.13$, $p = 0.020$, $r = 0.61$; SOL: $F(1,12) = 13.13$, $p = 0.003$, $r = 0.72$). Bilateral PARAs also showed a significant main effect of condition with the Threat condition showing a decrease in ML peak

magnitude compared to the No Threat condition, $F(1,12) = 5.19$, $p = 0.042$, $r = 0.55$. Of these muscles, there were no significant main effects of muscle side or any interaction effects. There was a significant interaction effect for the EOs ($F(1,12) = 6.685$, $p = 0.024$, $r = 0.60$). Tukey's HSD revealed a significantly larger response for the Threat compared to No Threat condition for the right EO. This is significant difference between the two sides is also evident through the lack of ML peaks outside of the 95 % confidence interval for the left EO (Figure 3.13).

Table 3.2 Peak coherence value taken from data concatenated from all subjects (N=10) within the ranges of significant coherence for each muscle in the No Threat and Threat condition. "n.s." indicates no significant values above the 95% confidence limit

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	Left Side				Right Side			
	Peak		Range		Peak		Range	
	No Threat	Threat	No Threat	Threat	No Threat	Threat	No Threat	Threat
PARA	0.06 at 3.9 Hz	0.04 at 3.9 Hz	0-14 Hz	0-14 Hz	0.08 at 2.9 Hz	0.06 at 3.9 Hz	0-10 Hz	0-17 Hz
EO	n.s.	n.s.	n.s.	n.s.	n.s.	0.01 at 3.9 Hz	n.s.	1-5 Hz
GM	0.07 at 7.8 Hz	0.07 at 5.9 Hz	0-24 Hz	0-25 Hz	0.05 at 5.9 Hz	0.07 at 6.8 Hz	0-24 Hz	0-24 Hz
VL	0.01 at 5.9 Hz	0.01 at 2.9 Hz	4-12 Hz	0-16 Hz	0.01 at 5.9 Hz	0.02 at 2.9 Hz	4-8 and 10-13 Hz	0-21 Hz
mGAS	0.05 at 6.8 Hz	0.05 at 3.9 Hz	0-19 Hz	0-19 Hz	0.01 at 5.9 Hz	0.05 at 3.9 Hz	0-22 Hz	0-20 Hz
SOL	0.04 at 7.8 Hz	0.06 at 6.9 Hz	0-21 Hz	0-21 Hz	0.03 at 5.9 Hz	0.05 at 4.9 Hz	0-14 Hz	0-14 Hz

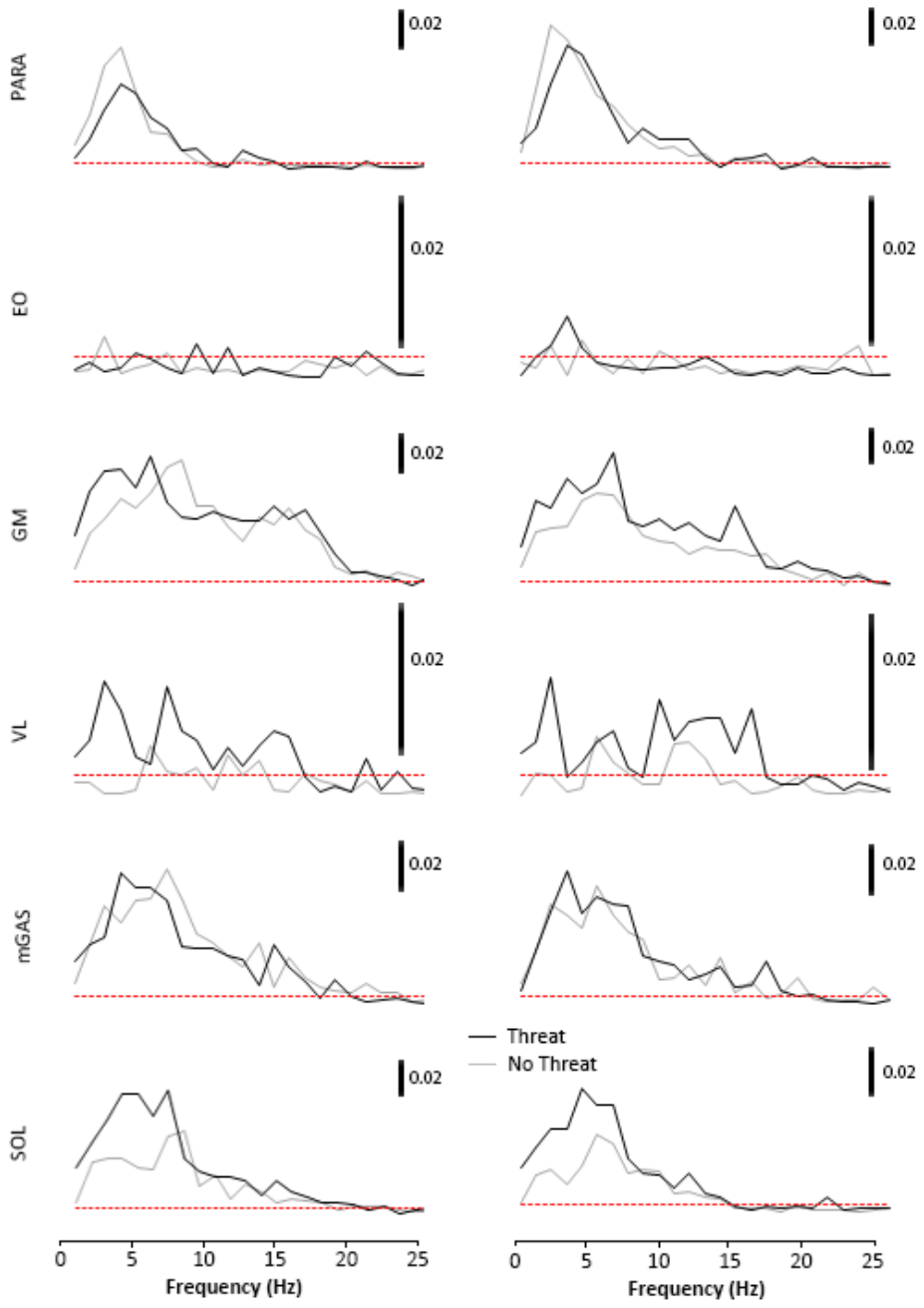

Figure 3.11 Coherence estimates for the measured muscles. The left column displays the coherence for the muscles on the left side and the right column displays the coherence for the right side. Gray lines indicate the No Threat condition while the black lines indicate the Threat condition. The red horizontal line indicates the 95% confidence level calculated for 1521 segments (0.0026).

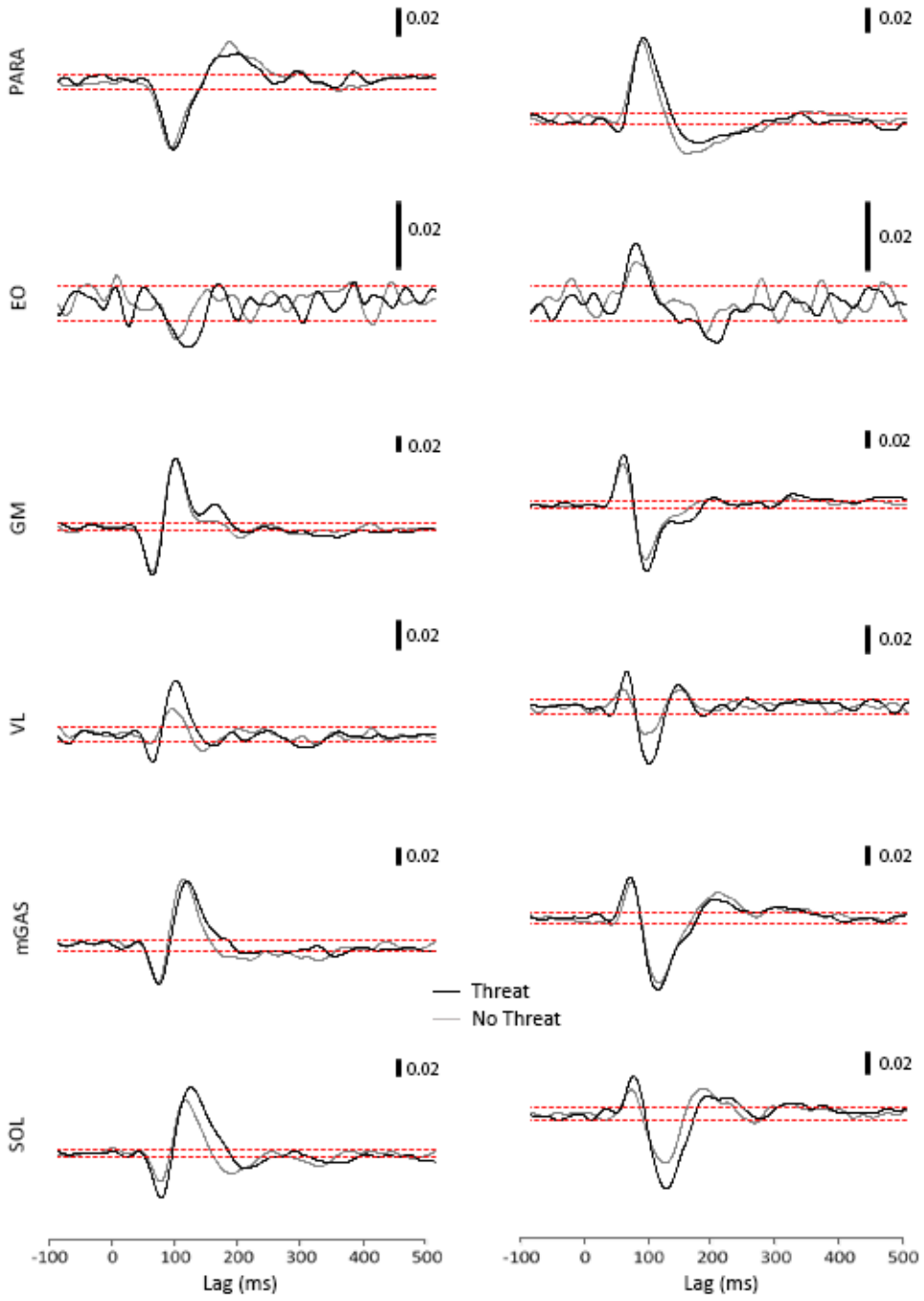

Figure 3.12 Cumulant density estimates for the measured muscles. The left column displays the cumulant for the muscles on the left side and the right column displays the cumulant for the right side. Gray lines indicate the No Threat condition while the black lines indicate the Threat condition. The red horizontal lines indicate the 95% confidence interval for each muscle.

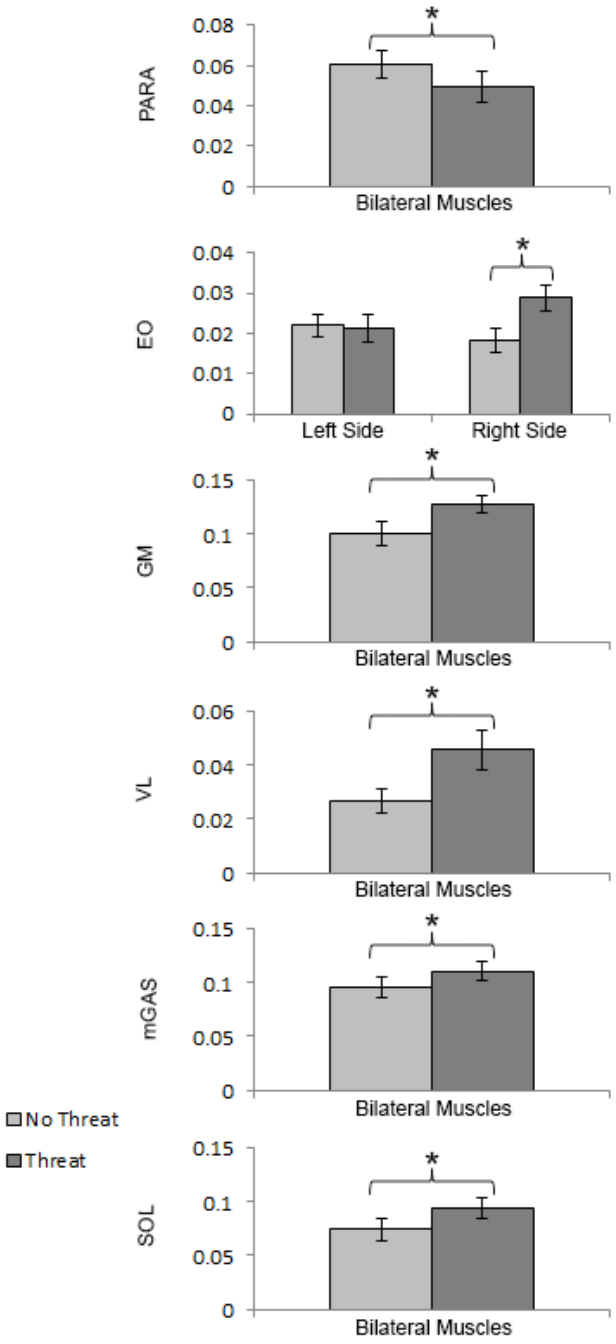

Figure 3.13 Magnitude (mean ± SE) of the medium latency (ML) cumulant density peak in each muscle for the No Threat and Threat conditions. * indicates a significant difference ($p < 0.05$). All muscles but the EO showed a significant main effect of condition and no effect of the muscle side or any interaction effect. A significant interaction effect was observed in the EO muscles and post-hoc analysis revealed an effect of condition for the right side.

Significant correlations between the change in peak ML response amplitude and the change in GSR were only observed for the GM ($r=0.672$, $p=0.0167$) and SOL ($r=0.742$, $p=0.0058$) muscles (Figure 3.14).

Significant differences in gain between the two conditions were observed in bilateral GMs, mGASs, and SOLs and right VL. Of these muscles, the frequencies at which the gains were significantly different were only consistent between the bilateral EO and SOL muscles. These significant differences are displayed in Figure 3.15. Despite an absence of a distinct and consistent difference for some muscles, the gain calculations showed a trend for larger gains in the Threat condition compared to No Threat condition (Table 3.3).

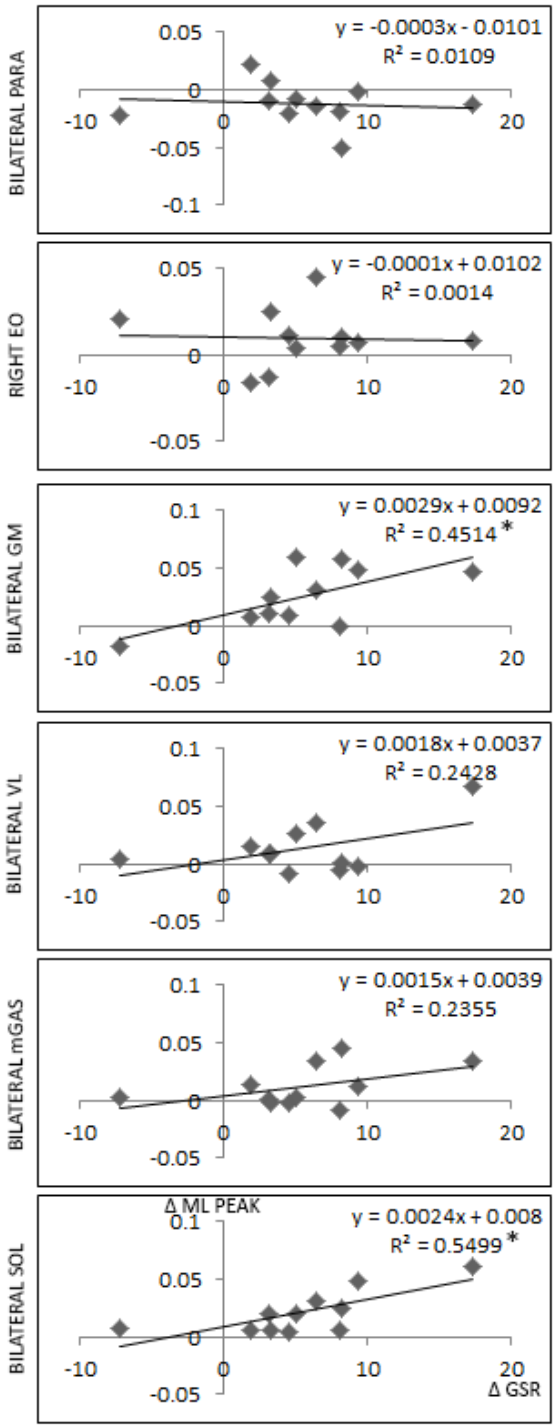

Figure 3.14 Change (Threat-No Threat condition) in GSR amplitude in relation to the change in peak ML response amplitudes for muscles that showed a significant difference in peak ML response amplitude across conditions. * indicates a significant correlation.

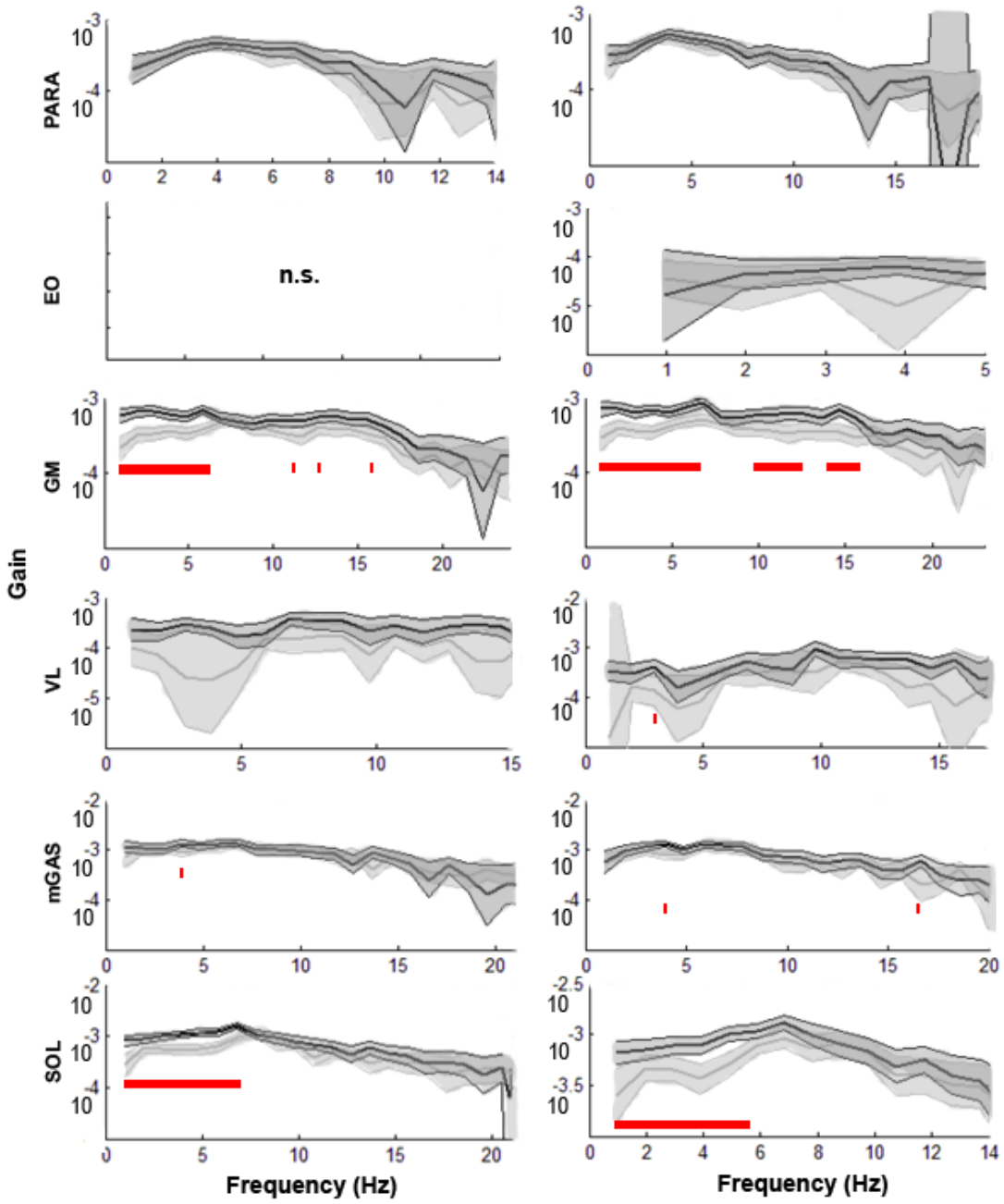

Figure 3.15 Vestibulo-muscular gain for the No Threat (light grey) and Threat (darker grey) conditions. Point-wise gain is plotted with 95% confidence intervals shaded in light grey (No Threat) and a darker grey (Threat). Areas where the confidence intervals do not overlap, as marked with a solid red line, indicate a significant difference in gain between the two conditions. Comparisons were only made for frequencies that revealed significant coherence in at least one of the conditions (plotted).

Table 3.3 Gain increases calculated as a percentage of the average SVS-EMG gain across all significant frequencies between the two conditions (Threat/No Threat). Positive percentages indicate a larger gain for the Threat compared to No Threat condition.

	Left Side	**Right Side**
PARA	1.18%	1.11%
EO	n.s.	1.49%
GM	1.51%	1.65%
VL	2.60%	1.95%
mGAS	1.05%	1.22%
SOL	1.46%	1.44%

Chapter 4: Discussion

In order to further investigate the nature of the SVS-induced vestibular reflex, this thesis was separated into two major purposes: to determine 1) if there is phase-dependent modulation of the vestibular reflex during the platform-induced postural response and 2) if a threat of perturbation induces changes of the vestibular reflex. This thesis provided inconclusive results for the presence of a phase-dependent modulation during the expected phase. However, modeling data and pilot work provide promising results for further investigation. On the contrary, results from the second purpose of this thesis were partially in line with the hypothesis. Changes in psychosocial and autonomic states resulted in significant changes to the vestibular reflex where an increase in threat was generally followed by an increase in the SVS-EMG relationship and gain in most muscles.

4.1 Phase-Dependent Modulation

4.1.1 Angular Head Movement

A 55°/s perturbation in the pure rightward roll direction elicited responses similar to those seen in previous work. The onset of head accelerations in the roll plane occurred at 40 ms with an average maximum acceleration of 762°/s^2 at 70 ms. Although these onsets were similar to those recorded in previous work (Carpenter et al., 1999a), the maximum head acceleration recorded in this experiment was almost 4 times larger than previously reported. One important difference between the experiments is the methods of recording angular head movements. Carpenter et al. (1999a) used an array of linear accelerometers to record head accelerations while the current experiment used differentiated data from a set of position markers. The accuracy of measuring accelerations using position markers is not analogous to measurements obtained through accelerometers and may account for some of the discrepancies between the

studies. Nevertheless, the onset of head acceleration recorded from both studies occurred before the balance-correcting response and therefore, when perturbed, the natural vestibular information received through the head movements still had the potential to influence the response.

4.1.2 Postural Response

These results also revealed postural responses that were similar to those analyzed in previous studies (Carpenter et al., 1999a). The perturbation in a pure rightward roll direction evoked asymmetrical responses with distinct stretch reflexes and unloading responses followed by balance-correcting responses.

4.1.3 Vestibular Reflex

When collapsed across frequencies between 0-20 Hz, the time-dependent coherence results from the first experiment showed a significant decrease in coherence starting as early as 412 ms before the perturbation and as late as 24 ms after the perturbation. Within our paradigm subjects were not able to definitively anticipate the perturbation as there were no vibrotactile or auditory cues preceding the platform perturbation. Catch trials in the opposite direction were also presented to decrease the preparation of a direction specific response. Therefore, it was surprising that a decrease in coherence occurred before the platform perturbation onset.

Similar decreases in coherence around the stimulus onset have previously been observed when investigating corticomuscular coherence (McClelland, Cvetkovic, & Mills, 2012). McClelland et al. (2012) investigated the influence of afferent feedback on corticomuscular synchrony by implementing a transient external electrical or mechanical stimulation to the hand while gripping an object. Although the authors did not focus on this aspect in their paper, a similar decrease in coherence was observed around the onset of the stimulus (i.e.

perturbation). During this period, a reflexive response to the stimulus was observed, which interfered with the corticomuscular coherence until approximately 400 ms after the stimulus. Since the platform perturbation in our experiment also induced a response, it is possible that the response in this experiment also interfered with our calculated coherence.

Within the current paradigm, the perturbation itself would have resulted in a large influx of sensory information from multiple receptors. Perception of ankle movement, and thus activation of somatosensory receptors in the ankles, would have been able to detect the 7.7° movement of the platform (Fitzpatrick & McCloskey, 1994). Additionally, although kinematic data was not recorded in this experiment, previous work using similar perturbations indicated that trunk movements occurred around 20 ms after the perturbation and moved at approximately 12°/s (Carpenter et al., 1999a). This movement was enough to elicit stretch reflexes in the PARA 40 ms later. In addition, the head was not displaced to an extent that would significantly affect the SVS-induced vestibular reflex (<3°). Based on a vestibular reflex transformation study by Luu et al. (in progress), the amplitude of the vestibular reflex degrades by the cosine of head angle in the yaw rotation from the optimal head position. From these results, at 3° the reflex would still be 99% the amplitude of the optimal size. A similar cosine relationship has also been calculated for perceptual GVS-induced rotations at different head pitch angles (Day & Fitzpatrick, 2005). Despite the small head displacements, it is still possible that the natural movement of the head interfered with the SVS-induced response. However, although not tested in the same planes and at the same velocities calculated in this thesis, previous work suggests that these platform-induced head movements should not affect the SVS-induced vestibular reflex as they were not affected by movements of the head at peak velocities of 57 ± 2.5°/s in the yaw plane (Dakin, 2012). Since the vestibular receptors are

known to detect linear and angular accelerations of the head, these large platform-induced accelerations would have been detected and encoded by the vestibular receptors. As a whole, the information gathered from all of these receptors had the potential to be involved in the platform-induced response. This large motor output generated from the natural sensory information then theoretically competed with and overwhelmed the relatively small SVS signal and its corresponding reflex.

Further analysis and testing in experiment 2 and 3 confirmed that this decrease in coherence was confounded by the large EMG burst induced by the platform perturbation. The large, natural platform-induced input, whether it was somatosensory from stretched muscles or vestibular from passive head movements, resulted in a response that was too large for the SVS-induced response to be observed. Simulated data were created in experiment 2 to replicate the response observed in the first experiment. Although a related signal was present in the simulated input and output signals, the presence of an additional large burst of activity in the output signal resulted in a significant decrease in coherence similar to that seen in experiment 1. This indicated a large limitation in investigating the vestibular reflex with SVS during the different phases of the postural response since any connection or phasic change of the SVS-induced vestibular reflex would have been washed out by the large platform-induced response.

One possible way to investigate the phase-dependent modulations was to attenuate the platform-induced response. Again, I tested this theory using simulated data with gradually smaller bursts within the related signals and observed increasing persistence of the input-output relationship with smaller bursts. At 10% of the original burst size, the input-output relationship remained coherent throughout the time window. This revealed that a smaller transient burst would be more ideal for investigating phase-dependent modulations of the vestibular reflex.

In experiment 3, I tested the results of the simulated data on 3 human participants. Previous work suggested that, while the pattern and onset of response remain the same, the size of the postural responses scale with the size of the perturbation velocity (Allum et al., 1994). With the results from our simulated data we tested five different platform perturbation velocities to determine the point at which the response sizes were at least 10% of the response observed with the fast perturbation from experiment 1. For the majority of muscles, an 11.5°/s perturbation resulted in a balance-correcting response that was less than 10% the size observed with the 55°/s perturbation. With this velocity, some muscles no longer showed a postural response and I suspected that a further decrease would result in very little or no activity in more muscles. Therefore, this velocity was chosen for experiment 3.

Compared to the fast, 55°/s perturbation, the slower 11.5°/s perturbation resulted in smaller responses. Since it was possible that natural vestibular input could contribute to the postural response and affect the transfer of the SVS-induced vestibular reflex, it was very important to reduce head movement and reduce the amount of natural vestibular input passing through the vestibular system. Smaller head movements were observed and, as expected, similar but smaller postural responses between the two perturbation velocities were seen in most muscles. This postural response difference was also evident through a transient power shift away from the perturbation onset and through a clear decrease in transient power. Additionally, vestibulomuscular coherence no longer disappeared before the start of the perturbation or followed the timing of the transient power onset. Moreover, in some cases such as the SOL muscles, a decrease in coherence was not apparent.

Although the parameters used in the first experiment cannot clearly determine phase-dependent modulations of the reflex, the novel techniques supported by the pilot results from

experiment 2 and 3 provide a promising method for further investigation. Further work should also aim to determine if other perturbation parameters, such as platform rotations with slower accelerations, elicit a response that is broader (i.e. does not have a sharp rise in EMG) and is not limited by the analysis. Furthermore, to optimize the ability to detect the SVS-induced vestibular response, more work needs to be done to investigate the ideal vestibular stimulation parameters. As the amplitude of the vestibular reflex is related to the amplitude of the stimulation intensity, optimizing the SVS characteristics to a higher overall intensity (i.e. higher RMS) may also increase the ability to identify any phase-dependent modulations of the SVS-induced vestibular reflex.

It is also possible that the phase-dependent differences previously observed in vestibular loss patients reacting to a postural perturbation is a result of an absent tonic vestibular drive to the muscles. Based on the results from the threat of perturbation portion of this thesis (see next section for a discussion of these results), prior to the perturbation, the vestibulo-muscular relationship was optimized and upregulated in all muscles but the PARAs. If the vestibular system were involved with setting the tone or excitability of the muscles prior to the perturbation, then the absence of a vestibular input would result in a decreased muscular response and an increased PARA response. This is indeed observed when vestibular loss patients respond to postural perturbations (Carpenter et al., 2001; Allum et al., 1995; Allum et al., 1988). To further test this theory, other muscles that show increased response amplitude with vestibular loss such as the TA in toes-down perturbations could be tested (Carpenter et al., 2001a).

4.2 Threat of Perturbation

4.2.1 Autonomic Arousal and Psychosocial Measures

The Threat of perturbation condition successfully induced psychosocial and autonomic changes that were similar to those induced by a height paradigm. Electrodermal recordings measures sweat gland secretions and are thought to reflect sympathetic discharge that is in part influenced by supraspinal centres (Venables 1991 for review). One of these supraspinal mediators is the amygdala, which has been linked to arousing (McGaugh, 2004) and fearful (Öhman, 2005) situations. Activation of the amygdala is highly correlated to the adrenal stress hormones released under arousing situations and its facilitating effects on long-term and enhanced memory consolidation and retrieval (McGaugh, 2004). Changes in anxiety, on the other hand, have been associated with activation of areas such as the locus coeruleus (Gorman, Liebowitz, Fyer, & Stein, 1989). An increase in EDA, fear, and anxiety with increased postural threat has consistently been shown in both studies involving a perturbation-induced threat (Horslen et al., 2013) as well as numerous studies using a height-induced threat (Cleworth et al., 2012; Davis, Campbell, Adkin, & Carpenter, 2009). In this thesis, similar changes were observed for all autonomic and psychoscial measures during the Threat compared to No Threat condition.

4.2.2 Vestibular Response

Two characteristics of the vestibular response were measured between the No Threat and Threat conditions: correlation, in the frequency (coherence) and time (cumulant) domains, and gain. As hypothesized, significant correlations in both domains were observed in all muscles except the EOs during quiet stance. Animal work has found strongest vestibular connections to extensor muscles throughout the body with minimal excitatory connections to flexor muscles

(Grillner, Hongo, & Lund, 1970). As the EOs are the only pure flexor muscles recorded in this experiment, it is possible that the lack of correlation was due to a weak vestibular connection with these muscles. GVS work also suggests that the induced vestibular reflex is only observed in muscles when they are actively engaged in balance (Britton et al., 1993; Luu et al., 2012). A lack of vestibular reflex could also suggest that the EOs are not strongly engaged in balance during quiet stance.

As previously discussed, changes in fear, anxiety, and arousal are mediated and related to activation of various supraspinal areas such as the amygdala and locus coeruleus. Other structures involved in these emotional states include the infralimbic cortex, bed nucleus of stria terminalis, and hypothalamus. Animal work shows anatomical connections between these areas and the vestibular nuclei through multiple direct and indirect connections from the parabrachial nucleus, locus coreuleus, and dorsal raphe nuclei (Balaban 2002; Balaban & Thayer 2001). Additionally, neural imaging of human cortex also suggests that areas of the vestibular cortex, such as the parieto-insular vestibular cortex, could mediate the vestibular output (Dieterich & Brandt, 2008). These anatomical connections, as proposed in previous work (Carpenter et al., 1999; Carpenter et al., 2004; Horslen et al., 2014; Yardley & Redfern, 2001), provide a way for psychosocial and autonomic state changes to modulate the vestibular system's output.

The numerous anatomical links and theories are supported by various behavioural studies. In humans, several vestibulo-occular deficits have been recorded in individuals with panic disorders with or without agoraphobia (Jacob, Furman, Durrant, & Turner, 1996; Jacob, Redfern, & Furman, 1996). Additionally, drowsy, or non-aroused, subjects show decreased or absent nystagmus when placed in a rotational chair (Kasper, Diefenhardt, Mackert, & Thoden, 1992). While there has been a vast number of evidence for vestibulo-occular changes related

pychosocial and autonomic factors, there are only a limited number of studies that have observed increases in the vestibular response with increased psychosocial and autonomic states (Horslen et al. 2014; Naranjo et al., in progress; yet see Osler et al., 2013). Horslen et al. (2014) calculated significantly larger coupling and gain between SVS and ground reaction force while Naranjo et al. observed increases in vestibular evoked myogenic potential amplitudes when using loud acoustic bursts at height compared to ground level.

Results from this thesis are in line with the previous findings. When the participants were placed in a condition with an increased postural threat, the SVS-EMG correlations increased in most muscles and, interestingly, appeared in several muscles that previously lacked correlation. These changes are presumably influenced by the supraspinal emotional centres as observed by significant correlations between the vestibular reflex and arousal for GM and SOL. The absence of a significant correlation for the other muscles was likely due to the low number of participants in this study and thus a lack of power with the calculation. Although vestibular information predominately propagates to extensor muscles, the appearance of significant correlations for the right EO indicates that the vestibular system is connected to and has the ability to transfer vestibular information to this muscle. The increased frequency bandwidths observed in most muscles, and most prominently observed in the VL muscles, also indicates a change in the type of vestibular information that is being propagated. Current results suggest that threatening situations may change the vestibulomusclar connection in preparation of a stabilizing postural perturbation. Thus, under threatening situations, the information transferred to muscles involved in the upcoming postural response may be optimized or upregulated in anticipation of the perturbation. This upregulation is also evident through the increases in gain observed in all muscles. These changes would then arguably result in a larger than normal

postural response, as shown in previous work (Carpenter et al., 2001b; Brown & Frank, 1997; Cleworth et al., in progress).

Although most muscles showed an increase in correlation with threat, the PARAs showed a significant decrease in correlation. At first, this seemed contrary to the theories and previous works on the effects of arousal, fear, and anxiety; however, this is in line with previous work on vestibular loss patients. When an individual with vestibular deficit experiences a rapid perturbation, balance-correcting responses are attenuated in all muscles but the PARAs (Carpenter et al., 2001a; Allum & Honegger, 1998). The muscular differences observed and the proposed explanations in the previous experiments are supported by the results of this current experiment. The decrease in vestibular reflex with threat of perturbation in this thesis provides further evidence for a different vestibular spinal connection to the trunk muscles. Indeed, Carpenter et al. (2001a) proposed that these trunk muscles might receive strong inhibitory inputs from the vestibular system. Animal work shows that the most common monosynaptic connections from the vestibular system to extensor muscles are excitatory while connections to the flexor muscles are typically inhibitory (Grillner et al., 1970). Animal work also describes strong monosynaptic and polysynaptic inhibitory connections within the medial vestibulospinal tract (Carpenter, 1988). As the PARAs encompass numerous muscles, some of which span the back and may receive input from cervical or upper thoracic collaterals of the medial tract, it is possible that a large number of these muscles receive inhibitory input from the vestibular system (Wilson, Yoshida & Schor, 1970). When standing with anticipation of a perturbation, results from this thesis show that the fearful, anxious, and arousing effects of the upcoming perturbation prepare the muscles for the postural response by increasing the vestibular gain and coupling. For most muscles, this increase comes as an excitatory input

through the vestibular output; while for the PARA muscles, the inhibitory input is increased. Under normal conditions, the PARA response to a postural perturbation would be attenuated; in bilateral vestibular loss patients, the absence of the vestibular input would arguably remove the inhibition to the PARAs and result in a larger response to the perturbation.

Functionally, because a human's center of mass is located approximately around the midsection, any sudden displacement to the body's trunk segments would lead to large changes in the center of mass. Under threatening situations, an increase in center of mass deviations would not be beneficial to the individual, as it would result in a greater destabilization. This decrease in center of mass deviation with a perturbation at height compared to ground level has been documented in previous studies (Carpenter et al., 2004). Despite this center of mass decrease, Carpenter et al. (2004) observed an increase in PARA activity. This difference could be explained by the location of increased threat and the direction of perturbation. With the height paradigm, the threat of falling is only increased anterior to the participant. Therefore, irrespective of the perturbation direction, an increase in PARA activity would bring the center of mass away from the threat. On the other hand, in the threat of perturbation paradigm, the threat of falling could be prominent in all directions.

One limitation of the current experiment is the method of inducing threat. Although I was successful in inducing significant changes in fear of falling, anxiety, balance confidence, and arousal the threat-induced changes in the vestibular reflex were not as large as those previously seen (Horslen et al., 2014). This smaller difference is evident in the inability to distinguish a broad frequency bandwidth in the difference of coherence test and in the difference of gain calculations. Although there are clear trends in the expected directions, the strength of the reflex differences under the perturbation-induced threat were not large enough to be

statistically different. In fact, although not directly compared, previous work utilizing the two induced threat paradigms have also shown larger response differences with a height-induced threat compared to a perturbation-induced threat (Horslen et al., 2013).

Despite these differences, I believe it is important to conduct further studies using the perturbation-induced threat paradigm. External perturbations often occur in everyday situations and a person's ability to react to these perturbations is critical for injury prevention. If an individual is fearful, anxious, or aroused by a looming perturbation, it is important to determine how these psychological and autonomic changes can affect the sensory systems involved in balance control.

Further work should also look at the effects of different severities of threats. The SLOW perturbations used in experiment 3 may have been perceived as a less threatening than the FAST ones. Previous work has shown a scaling effect with the psychosocial and autonomic measures at different levels of threat (Davis et al., 2009; Sibley, Lakhani, Mochizuki, & McIlroy, 2010). Preliminary data from this thesis suggest that a scaling effect may also be observed at different perturbation velocities. The three participants who experienced both No Threat, threat of slow perturbations (LowThreat), and threat of fast perturbations (HighThreat) showed a scaling effect for the fear of falling, balance confidence, and arousal measures (Figure 4.1). No consistent difference in the SVS-induced vestibular reflex was observed between the threat conditions. These results, however, are very preliminary as the LowThreat condition occurred last for all participants and the results may be confounded by fatigue or familiarity of the perturbation. Therefore, future work needs to be done in order to determine if there is a scaling effect of the vestibular reflex.

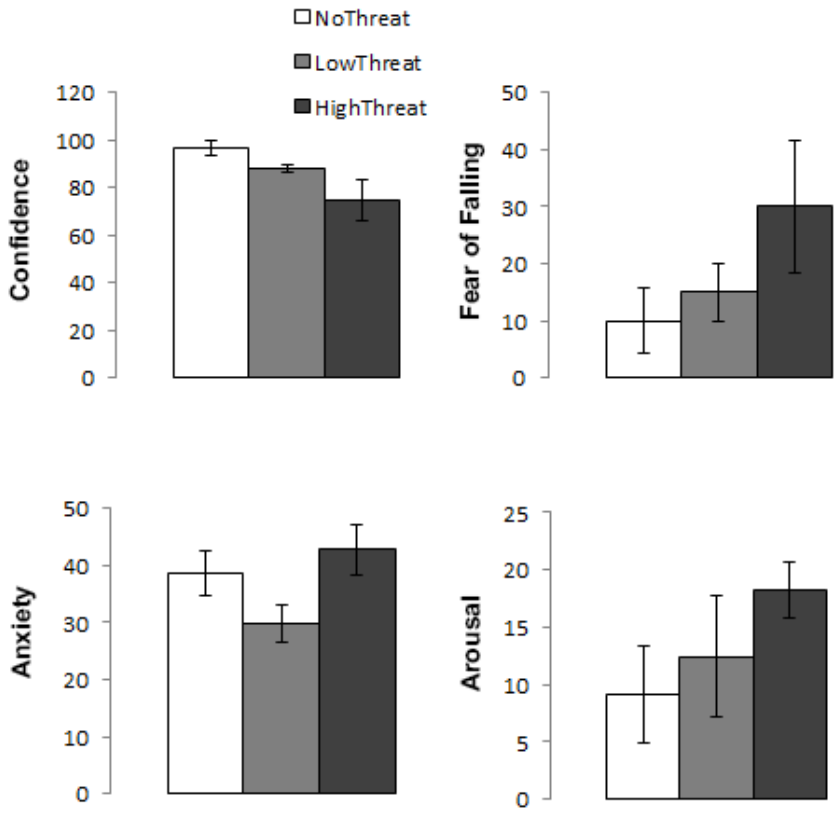

Figure 4.1 Preliminary (N=3) psychosocial and autonomic arousal results (mean ± SE) from the NoThreat, LowThreat, and HighThreat conditions.

Chapter 5: Conclusion

The first experiment in this thesis indicated that the methods used were not reliable enough to determine vestibular changes during fast perturbations. The platform perturbations used in this current and several previous studies are too fast and induce postural responses that are too large for an SVS-induced vestibular reflex to be observed under the current parameters. The simulated data and pilot results from three participants provided a promising method for further investigation. Results indicated decreased postural response amplitudes and a clear change or shift in transient power changes away from the onset of a slow platform perturbation. With slower perturbations, the vestibulomuscular coherence also no longer completely washed out before the start of the perturbation and, for some muscles, the coherence did not shut off during the postural response window. Although I was unable to confidently determine if there were any phase-dependent modulations of the vestibular reflex during the platform-induced postural response, further analyses and testing did provide a possible way to investigate these changes in the future.

Results from the second portion of this thesis partially confirmed my hypotheses and provided further support for psychosocial and autonomic modulations of the SVS-induced vestibular reflex. Placing participants in an environment where a postural perturbation could occur, probed the vestibular system by optimizing and upregulating the information that is transferred to the postural muscles. The modulation of the vestibular response could have been mediated by various brain stem centres, such as the parabrachial nucleus, locus coeruleus, or raphe nucleus, or through subcortical (amygdala) or cortical (vestibular cortex) areas. Interestingly, the PARAs exhibited an opposite response showing decreased coupling with

threat. This reverse response could suggest a different tonic vestibulo-muscular connection that is more inhibitory in nature compared to the excitatory connections typically observed in other postural muscles. Results from this thesis provide important evidence for how posture is affected by psychosocial and autonomic factors and provide further insights on possible mechanisms of balance impairments in various populations.

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Appendices

Appendix A Questionnaires

A.1 Balance Confidence Scale

Please use the following scale to rate how confident you are that you can maintain your balance and avoid a fall during the balance task:

0……10……20……30……40……50……60……70……80…..90……100

I do not feel confident at all — I feel moderately confident — I feel completely confident

A.2 Fear Questionnaire

Using the following scale, please rate how fearful of falling you felt when performing the balance task:

0……10……20……30……40……50……60……70……80…..90……100

I did not feel fearful at all — I felt moderately fearful — I felt completely fearful

A.3 Perceived Anxiety Subscale

Please answer the following questions about how you honestly feel just after standing under this condition using the following scale:

1	2	3	4	5	6	7	8	9
I don't feel at all				I feel this moderately				I feel this extremely

1. I felt nervous when standing at this height

* 2. I had lapses of concentration when standing at this height

3. I had self doubts when standing at this height

4. I felt myself tense and shaking when standing at this height

* 5. I was concerned about being unable to concentrate when standing at this height

6. I was concerned about doing the balance task correctly when standing at this height

7. My body was tense when standing at this height

* 8. I had difficulty focusing on what I had to do when standing at this height

9. I was worried about my personal safety when standing at this height

10. I felt my stomach sinking when standing at this height

* 11. My heart was racing when standing at this height

12. Thoughts of falling interfered with my concentration when standing at this height

* 13. I was concerned that others would be disappointed with my balance performance at this height

14. I found myself hyperventilating when standing at this height

15. I found myself thinking about things not related to doing the balance task when standing at this height

Appendix B Consent Form

SUBJECT INFORMATION AND CONSENT FORM

The role of supraspinal structures in triggering automatic postural responses (Experiment 7)

Principal Investigator: Mark G. Carpenter, Ph.D
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1. INTRODUCTION
You are being invited to take part in this research study because you are a healthy person between 19-35 years of age.

2. YOUR PARTICIPATION IS VOLUNTARY
Your participation is entirely voluntary, so it is up to you to decide whether or not to take part in this study. Before you decide, it is important for you to understand what the research involves. This consent form will tell you about the study, why the research is being done, what will happen to you during the study, and the possible benefits, risks and discomforts.

If you wish to participate, you will be asked to sign this form. If you decide to take part in this study, you are still free to withdraw at any time and without giving any reasons for your decision.

If you do not wish to participate, you do not have to provide any reason for your decision not to participate, nor will you lose the benefit of any medical care to which you are entitled or are presently receiving.

Please take time to read the following information carefully and to discuss it with your family, friends, and doctor before you decide.

3. WHO IS CONDUCTING THE STUDY?
The study is being conducted by the Neural Control of Posture and Movement Laboratory at the UBC School of Kinesiology. This study is funded by a Research Discovery Grant from the Natural Sciences and Engineering Research Council of Canada (NSERC).

4. BACKGROUND
Falls are the leading cause of accidental injury and death in older adults. Efforts to design effective training and rehab programs to reduce falls are currently hampered by a lack of understanding of how balance responses (i.e. the patterns of muscle activity and body movemnts made to prevent a fall) to unexpected balance disturbances (i.e. events causing a loss of balance) are normally triggered. It was initially thought that balance reflexes begin at the level of the lower leg and are controlled by the spinal cord. More recent research, however, has suggested that the brain also plays an important role in producing a balance response. Such knowledge should allow new and specifically tailored treatments to be developed to reduce the incidence and impact of falls in individuals with balance disorders.

5. WHAT IS THE PURPOSE OF THE STUDY?
The purpose of these studies are to determine how balance responses are triggered and modulated in healthy young adults following a sudden movement of the surface on which you are standing.

6. WHO CAN PARTICIPATE IN THE STUDY?
Healthy subjects between 19-35 years of age are being invited to participate in this project.

7. WHO SHOULD NOT PARTICIPATE IN THE STUDY?
If you meet any of the following criteria, you should not participate in this study:

Neurological diseases such as:
Parkinson's disease, cerebellar ataxia, Huntington's disease, or inner-ear disorders (deafness)
Orthopaedic issues such as:
Chronic foot, leg, hip, back, or neck pain
Joint pain/arthritis
Surgical treatment of the foot, ankle, leg, hip, back, or neck

8. WHAT DOES THE STUDY INVOLVE?
This study is taking place at the Neural Control of Posture and Movement Laboratory at UBC.

If you agree to take part in this study, the procedures you can expect will include the following:

Setup: You will have your height and weight measured. In order to record the movements of your head, a rigid set of infra-red markers will be placed by your temple. The cameras used to indicate the location of the infra-red lights are not video cameras. Rather, they are specialized devices only capable of detecting the location of the infra-red light markers. Therefore, in no way will you be made identifiable through their use. You will also be fitted with twenty surface electrodes to measure muscle activity. These electrodes consist of small adhesive foam pads containing a conducting gel. These electrodes will be stuck to the skin of your stomach, back, and hips on both sides of your body and you lower and upper leg on the right side of your body. The electrodes will then be attached to a small pack worn on a belt placed around your waist. Prior to applying the electrodes, the skin under the electrodes will be shaved and then lightly sanded (abraded) to improve contact. A female experimenter will be available at all times during the experiment should you feel more comfortable having her place the various markers onto your body. A twenty-first electrode will serve as a reference to measure the muscle activity and will be placed on your ankle. Two electrodes will also be placed on the palm of your non-dominant hand to measure autonomic arousal. To aid in processes of locating body landmarks and applying the markers to the necessary muscle groups, you will be asked to wear a pair of shorts and a short-sleeve t-shirt to the experimental session. Finally, electrodes with hypo-allergenic gel will also be attached to the bony surface directly each ear; these electrodes will be taped in place and will also be further secured with a headband. The electrodes behind the ears will be attached to wires and will be used to electrically stimulate the vestibular system.

Procedures:
After the electrodes have been checked to ensure they are functioning properly, you will be asked to stand on a forceplate that has been mounted onto the platform with your arms hanging naturally at your sides. The forceplate is a device that can determine whether you are leaning forwards onto your toes, leaning backwards onto your heels, or to your left or right side. You will be asked to refrain from making any unnecessary arm or body motions prior to, and following the end of each experimental trial.

While on the forceplate, you will be asked to stand quietly with your eyes closed for 3 minutes. During this time, the platform will be locked in place and you will receive a continuous, low electrical current behind your ears. After the three minutes, you will be given a 5 minutes rest period before the next condition. For the second condition, you will be asked to stand quietly on the forceplate with your eyes closed. At periodic times, the platform will sudden rotate in a random direction and there will be at least 10 seconds between each platform perturbation. During this time, you will be exposed to the electrical current for blocks of 5 minutes. At all times, a spotter will be present to assist you should you begin to lose your balance. There will be a 5 minute seated rest period every two blocks. Your goal throughout the experiment will be to stay standing after each platform movement. You will experience a total of 150 perturbations (approximately 75 minutes of total electrical stimulation). Finally, the last condition will be another 3 minute quiet standing trial, similar to the first condition.

The experiment will take approximately 3.5 hours to complete.

9. WHAT ARE MY RESPONSIBILITIES?

If you decide to take part in this study, you will be required to stand quietly on the forceplate and to react to unexpected rotations of the surface on which you are standing. Before each standing trial you will be required to answer a questionnaire about your confidence to maintain your balance for that particular condition. After each standing trial you will be required to answer a questionnaire about your perceived anxiety during the previous trial. Please note that you do not have to answer any questions in the questionnaires that you do not feel comfortable with.

10. WHAT ARE THE POSSIBLE HARMS AND SIDE EFFECTS OF PARTICIPATING?

The imposed balance disturbance is a quick rotation of your support surface that has been designed specifically to allow you develop a balance response without causing you to fall over. There is minimal risk of falling during the experiment. A spotter will be present at all times to stand behind you during the experiment to assist you in the event of a loss of balance.

Small infra-red lights (the size of a watch battery) will be attached to your skin or clothing using non-allergenic tape. Numerous surface electrodes to record muscle activity will be placed on your trunk, hip, and leg muscles using common electrode adhesive and a hypo-allergenic conducting gel. You will have the area immediately under the electrodes shaved and cleaned with rubbing alcohol to improve conductivity. Allergic responses, such as redness and itching of the skin, may occur in response to the electrode adhesive and gel. You may also experience mild redness and itching of the skin from the shaving and rubbing of your skin.

The electrical stimulation applied behind your ears will produce a skin sensation of very mild tingling at the site of the stimulating electrodes and may generate dizziness in rare occasions. This technique has been extensively used in our laboratory and most subjects only report skin sensation. If you feel any discomfort during the stimulation, you should inform the experimenter and the experiment will be stopped. There are no known physical or psychological risks associated with this type of non-invasive vestibular stimulation technique. Some subjects who are highly susceptible to car motion sickness may possibly experience mild

nausea or light-headedness for a brief period (up to 1 hour) following the experiment (in about 20% of subjects we have tested in the past). If you feel sick or have mild nausea, you should not drive until the side effects have worn off. Therefore, you may rest in the lab following the experiment until the symptoms pass.

In the unlikely event that you experience any muscle soreness or dizziness following an imposed disturbance, the experiment will be stopped. Also, if you do not comply with the experiment's instructions, the investigator can stop the experimental session.

If you experience side effects other than minor muscle soreness lasting less than one day, please notify Dr. Mark Carpenter

11. WHAT ARE THE BENEFITS OF PARTICIPATING IN THIS STUDY?
You will not receive any direct benefit from participating in this study.

12. WHAT IF NEW INFORMATION BECOMES AVAILABLE THAT MAY AFFECT MY DECISION TO PARTICIPATE?
If new information regarding the procedures or risks of this study becomes available, you will be advised of this information.

13. WHAT HAPPENS IF I DECIDE TO WITHDRAW MY CONSENT TO PARTICIPATE?
Your participation in this research is entirely voluntary. You may withdraw from this study at any time. If you decide to enter the study and to withdraw at any time in the future, there will be no penalty or loss of benefits to which you are otherwise entitled.

The study investigators may decide to discontinue the study at any time, or withdraw you from the study at any time, if they feel that it is in your best interests. If you choose to enter the study and then decide to withdraw at a later time, all data collected about you during your enrolment in the study will be retained for analysis. By law, this data cannot be destroyed.

14. WHAT HAPPENS IF SOMETHING GOES WRONG?
By signing this form, you do not give up any of your legal rights and you do not release the study doctor, participating institutions, or anyone else from their legal and professional duties. If you become ill or physically injured as a result of participation in this study, medical treatment will be provided at no additional cost to you. The costs of your medical treatment will be paid by your provincial medical plan.

15. CAN I BE ASKED TO LEAVE THE STUDY?
If you are not complying with the requirements of the study or for any other reason, the study investigator may withdraw you from the study.

16. AFTER THE STUDY IS FINISHED
You will not be directly informed of the results of this study. The results will be analyzed and published in a scientific journal.

17. WHAT WILL THE STUDY COST ME?
You will be provided a $30 honorarium for your participation. Payment will be dispensed at completion of the study. If you do not complete this study, the honorarium will be prorated according to the amount of the completed research.

18. WILL MY TAKING PART IN THIS STUDY BE KEPT CONFIDENTIAL?
Your confidentiality will be respected. However, research records and health or other source records identifying you may be inspected in the presence of the Investigator or his or her designate by representatives of the UBC Clinical Research Ethics Board for the purpose of monitoring the research. No information or records that disclose your identity will be published without your consent, nor will any information or records that disclose your identity be removed or released without your consent unless required by law.

You will be assigned a unique study number as a subject in this study. This number will not include any personal information that could identify you (e.g., it will not include your Personal Health Number, SIN, or your initials, etc.). Only this number will be used on any research-related information collected about you during the course of this study, so that your identity [i.e. your name or any other information that could identify you] as a subject in this study will be kept confidential. Information that contains your identity will remain only with the Principal Investigator and/or designate. The list that matches your name to the unique study number that is used on your research-related information will not be removed or released without your consent unless required by law.

Your rights to privacy are legally protected by federal and provincial laws that require safeguards to insure that your privacy is respected and also give you the right of access to the information about you that has been provided to the sponsor and, if need be, an opportunity to correct any errors in this information. Further details about these laws are available on request to your study doctor.

19. WHO DO I CONTACT IF I HAVE QUESTIONS ABOUT THE STUDY DURING MY PARTICIPATION?
If you have any questions or desire further information about this study before or during participation, you can contact Shannon Lim or Dr Mark G. Carpenter

20. WHO DO I CONTACT IF I HAVE ANY QUESTIONS OR CONCERNS ABOUT MY RIGHTS AS A SUBJECT DURING THE STUDY?
If you have any concerns about your rights as a research subject and/or your experiences while participating in this study, contact the Research Participant Complaint Line in the University of British Columbia's Office of Research Services

21. SUBJECT CONSENT TO PARTICIPATE
This section of the consent form is not a contract and as such you do not give up any legal rights by signing it.

I have read and understood the subject information and consent form.
I am aware what experiment I am volunteering to participate in.
I have had sufficient time to consider the information provided and to ask for advice if necessary.
I have had the opportunity to ask questions and have had satisfactory responses to my questions.
I understand that all of the information collected will be kept confidential and that the result will only be used for scientific objectives.
I understand that my participation in this study is voluntary and that I am completely free to refuse to participate or to withdraw from this study at any time without changing in any way the quality of care that I receive.
I understand that I am not waiving any of my legal rights as a result of signing this consent form.
I have read this form and I freely consent to participate in this study.
I have been told that I will receive a dated and signed copy of this form.

I have been told I will receive a copy of this consent form for my own records.
I consent to participate in this study.

_____	_____	_____
Subject Signature	Print Name	Date

_____	_____	_____
Principal Investigator Signature	Print Name	Date

Appendix C Additional Results

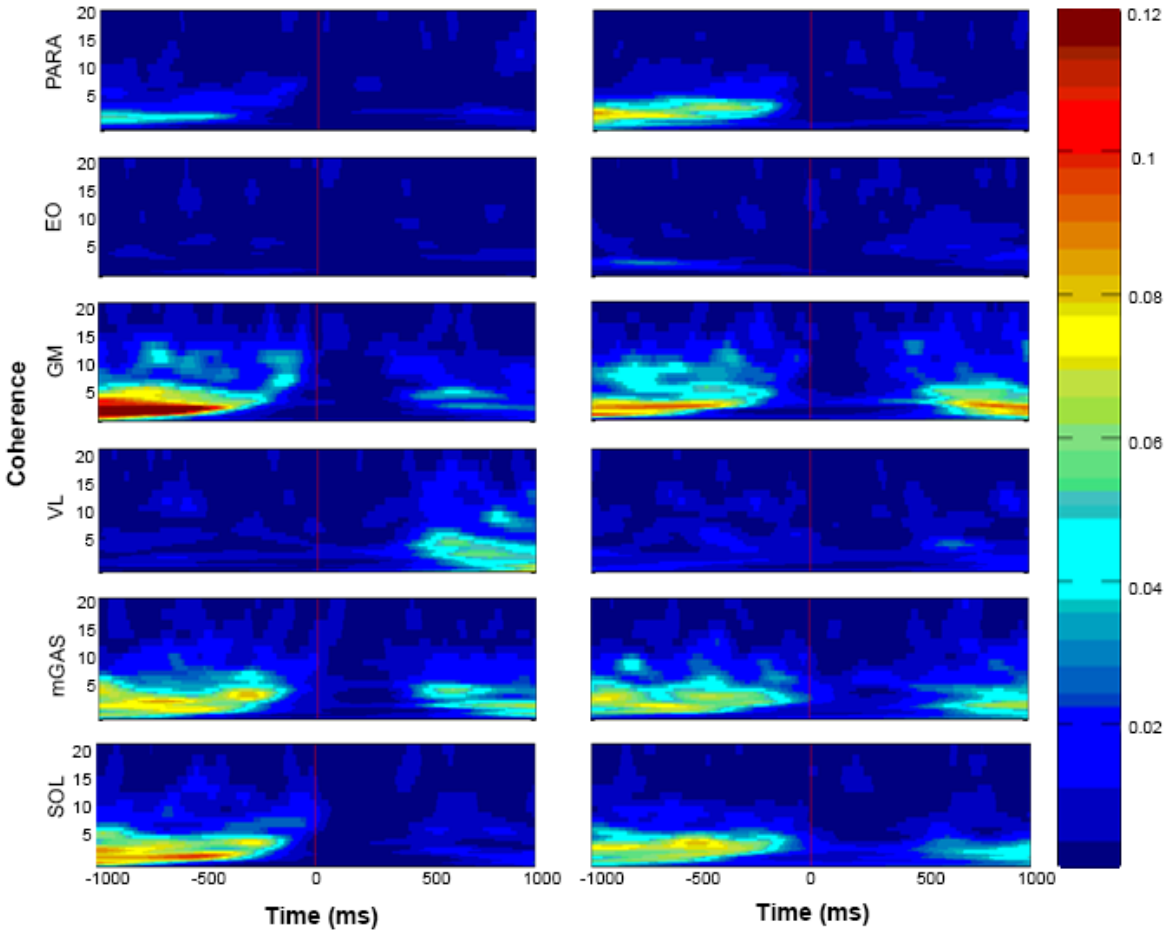

Figure C1 Time-dependent coherence estimates using 75 trials per participant. Similar trends are observed when 75 compared to 105 trials are compared. The left column indicates the estimates for muscles on the left side of the body while the right column indicates the coherence for the right muscles.

A	Left		% SLOW / FAST	Right		% SLOW / FAST
	FAST	SLOW		FAST	SLOW	
PARA	2.30	0.32	13%	0.53	-0.17	0%
EO	3.48	0.06	2%	1.44	0.05	3%
GM	5.32	0.70	13%	9.48	0.69	7%
VL	1.37	0.40	29%	1.93	0.05	3%
mGAS	4.31	1.07	25%	2.24	0.67	30%
SOL	1.30	-0.46	0%	5.28	0.43	8%

B	Left		% SLOW / FAST	Right		% SLOW / FAST
	FAST	SLOW		FAST	SLOW	
PARA	12.57	0.51	4%	10.93	0.00	0%
EO	7.47	0.00	0%	6.20	-0.09	0%
GM	8.90	0.14	2%	29.37	4.97	17%
VL	17.91	1.18	7%	7.48	0.18	2%
mGAS	39.74	2.20	6%	14.72	-0.27	0%
SOL	17.45	-2.42	0%	11.94	-3.16	0%

C	Left		% SLOW / FAST	Right		% SLOW / FAST
	FAST	SLOW		FAST	SLOW	
PARA	7.82	2.34	30%	3.59	-0.06	0%
EO	4.70	0.06	1%	2.29	0.30	13%
GM	3.95	-0.07	0%	40.70	11.01	27%
VL	9.46	3.50	37%	10.91	1.43	13%
mGAS	-4.36	-12.85	0%	25.96	10.86	42%
SOL	2.81	-6.11	0%	14.102	1.95	14%

D	Left		% SLOW / FAST	Right		% SLOW / FAST
	FAST	SLOW		FAST	SLOW	
PARA	20.78	7.71	37%	10.65	1.36	13%
EO	7.72	0.78	10%	10.24	1.84	18%
GM	6.77	1.13	17%	111.31	45.55	41%
VL	32.65	13.28	41%	43.54	10.68	25%
mGAS	-17.43	-48.32	0%	65.75	18.05	27%
SOL	-3.39	-24.93	0%	57.53	15.93	28%

Table C1 Muscle activation (μV·S) for different windows within the postural response after a FAST (55 °/s) and SLOW (12 °/s) platform perturbation.

A Stretch Reflex **B** Balance-Correcting Response **C** Secondary Balance-Correcting Response **D** Stabilizing Response