The Influence of Dual-tasking on Cortical Responses Associated with Instability

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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Abstract

Recent literature has revealed evidence of cortical involvement in reactive balance control. Using electroencephalography, event-related potentials (ERP) have been shown to be consistently evoked in response to a perturbation to balance. The most widely studied ERP, the N1, is speculated to be associated with resource reallocation in response to instability. A link between perturbation amplitude and N1 amplitude exists, suggesting exogenous attention can influence the N1 response. Thus, it may be expected that manipulating the focus of attention prior to the onset of a perturbation will change the amplitude of evoked cortical responses. In order to explore the link between attention and the N1 perturbation evoked cortical potential the current thesis explored the impact of dual-task performance on characteristics of the N1 and subsequent balance reactions. It was hypothesized attending to a cognitive task would lead to an N1 attenuation only in the small amplitude perturbation trials, not the large amplitude trials. Twelve, healthy adults participated in a study in which perturbations were applied using a lean-and-release protocol. A combination of two main tasks were assessed: perturbation amplitude and dual-task challenge. Changes in perturbation amplitudes were mediated by a hydraulic device slowing down the participant's acceleration, while dual-tasking was accomplished using a visuomotor tracking task. Results show dual-tasking led to the attenuation of the N1 peak amplitude regardless of the amplitude of the perturbation applied. However, the study did not reveal a scaling of the N1 to the different amplitudes of perturbation. The N1 appears to be associated with attention reallocation with little evidence towards the idea the N1 is related to the motor control in later phases of a balance reaction. Future studies should continue to draw parallels between attentional and reactive balance networks by exploring perturbation amplitudes and dual-task difficulty influences on the N1 peak amplitude.

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List of Abbreviations

A-P	Anterior posterior plane
ACC	Anterior cingulate cortex
APA	Anticipatory postural adjustment
APR	Automatic postural response
ANOVA	Analysis of variance
BoS	Base of support
CNS	Central nervous system
CoM	Center of mass
CoP	Center of pressure
EEG	Electroencephalography
EMG	Electromyography
EOG	Electrooculography
ERN	Error-related negativity
ERP	Event related potential
fMRI	Functional magnetic resonance imaging
LaR	Lean-and-release
MG	Medial gastrocnemius

SMA	Supplementary motor area
PET	Positron emission tomography
PEP	Perturbation evoked potential
RMS	Root-means-squared
TA	Tibialis anterior

Chapter 1: Literature Review

1.1 Introduction to Balance Control

1.1.1 The Problem of Falling

Maintaining upright balance is a pre-requisite for the successful completion of many activities of daily life. The process of aging in later life is known to lead to the deterioration of the physiological systems critical for the control of balance and may explain the observed association between aging and falls (Horak et al., 1989). The understanding of falls is relevant now more than ever as 15% of the Canadian population is composed of those who are 65 years or older (Statistics Canada, 2011). In parallel with the increase in the number of older adults in Canada, there has been an observed 43% increase in the number of self-reported falls in a span of 7 years, from 178,755 in 2003 to 256,011 cases in 2010 (Stinchcombe et al., 2014). Falls now account for 50% of accidental injuries that lead to hospitalizations, costing an estimated $4.5 billion to the Canadian health care system in 2004 (Smartrisk, 2009). About 20% of seniors living in the community fall each year; 37% of those report falling in more than one instance (Sibley et al., 2014). This prevalence of falls within older adults is known to increase with age and comorbidity (Sibley et al., 2014). The outcome of a fall can be devastating; common consequences include: significant losses in mobility, independence, and quality of life (Vellas et al., 1987; Stinchcombe et al., 2014); some even leading to death (Stevens and Dellinger, 2002). Falls can have life changing effects with 33% of those hospitalized quickly transitioning towards long-term care facilities (Scott et al., 2011). Fundamental to understanding the causes of falls is the need to better understand the control of balance, specifically in the reactions used to correct for instability. It remains important to advance this fundamental understanding as a way to

eventually enhance future treatments and diagnostic tools that could better inform clinical decision making.

1.1.2 Reactive vs. Predictive Balance Control

The concept of balance has a number of definitions (Pollock et al., 2000). Biomechanically, balance can be explained as the ability of the central nervous system (CNS) to keep the center of mass (CoM) within the base-of-support (BoS) (Winter, 1995). There are two distinct mechanisms through which the CNS controls balance: predictive and reactive control (Maki and McIlroy, 1997). Predictive control features muscular adjustments in anticipation of a postural disturbance to minimize instability, and reactive control, involves quick muscular activations to regain lost equilibrium in response to a moment of instability (Maki et al., 1994; Maki and McIlroy, 1997). In daily life, while predictive control is important, reactive balance control is the last available option to prevent a fall. Studying reactive balance control could further our understanding of falls by providing more functionally relevant information, as similarities exist between the stepping seen in reactive balance studies and gait dynamics; specifically, regarding step adjustment and gait initiation (Maki and McIlroy, 1997). In light of the importance and prevalence of falls in the Canadian aging population, the focus of the current thesis is directed to the control of reactive balance.

1.1.3 Characteristics of Reactive Balance Control

Within the scope of reactive balance, two classifications exist: 1) “fixed-support”, and 2) “change-in-support” reactions (Maki and McIlroy, 1997). “Fixed-support” strategies maintain the BoS for the duration of a reaction (Maki and McIlroy, 1997). Recovering balance then occurs through the deceleration of a moving CoM that follows the activation of muscles around the

ankle and hip joints; individually these are regarded as ankle and hip strategies (Horak and Nashner, 1986; Maki and McIlroy, 1997). In contrast, "change-in-support" strategies lead to the elongation of the BoS as individuals instead take a step or start using their upper limbs to affix to supporting structures (e.g. a rail, wall, etc.), therefore encompassing the CoM within a larger BoS (Maki and McIlroy, 1997). Both "fixed-support" and "change-in-support" strategies occur in parallel to one another, with each aiding to regain the control of balance through different mechanisms (Maki and McIlroy, 1997).

Typically, balance reactions are observed to have faster onset latencies than voluntary movements (Chan et al., 1979; Matthews, 1991). Platform motion cued stepping is, for example, observed to have significantly reduced latency, and step duration values, than visually cued, volitional stepping reactions (Burleigh et al., 1994; McIlroy and Maki, 1996). Balance reactions can in fact exhibit latencies, measured from onset of muscle activity, as early as 70 - 100 ms after a moment of instability (Jacobs and Horak, 2007; Maki and McIlroy, 2007). A consistent finding of lower-limb perturbations is that larger perturbations elicit significantly earlier, larger post-perturbation muscular amplitudes (Diener et al., 1988; Mochizuki et al., 2010). To be more specific, upright "fixed-support" or "change-in-support" balance reactions initially elicit the recruitment of muscles surrounding the ankle due to sensory information related to the applied perturbation; this first phase is believed to be automatic in nature and is known as the "automatic postural response (APR)" (Maki and McIlroy, 1997; McIlroy et al., 1999; Norrie et al., 2002). However, as an instability can be applied numerous ways, balance reactions are specific to the direction of the perturbation (Nashner, 1977; Moore et al., 1988; Misiaszek and Krauss, 2005; Safavynia and Ting, 2012). For a given direction (e.g. lateral), a number of different strategies may be used (i.e. cross-over stepping, side-stepping, or multiple steps) to recover balance (Maki

and McIlroy, 1997). Despite the fact that lateral stabilization is a critical component to the maintenance of balance, the complexity of electrophysiological testing, and lack of literature towards lateralized balance recovery makes it difficult to be further used for the purpose of this thesis. The current thesis instead focuses on reactive balance control within the anterior-posterior (A-P) plane.

Rapid postural reactions seem to be modifiable depending on the environment, instructions, previous exposure to stimuli, and other stimulus specific aspects (e.g. direction, predictability of onset, etc) (McIlroy and Maki, 1995). In addition to differences in the speed of a response when compared to voluntary stepping, reactive stepping often lacks anticipatory postural adjustments (APA) (Maki and McIlroy, 1997). APAs are lateral plane adjustments that occur before taking a step and lead to an increased loading in the stance leg as to ensure minimal falling towards the unloaded leg during swing phase (Maki and McIlroy, 1997). The absence of APAs in reactive movements does not ensure a faster balance recovery time following heel-strike (Maki and McIlroy, 1997). Instead, the APA is significantly reduced in time and amplitude, leading to more rapid foot-off time and the degree of modulation displayed is dependent on the amplitude of the perturbation and the stepping response (Maki and McIlroy, 1997). In trials where a perturbation is applied without any further instructions, APA is present in 61% of trials, though they are commonly absent in initial compensatory trials (McIlroy and Maki, 1995; Maki et al., 1996; Maki and McIlroy, 1997). Other adaptations that occur over repeated trials include decreases in the frequency of steps, and the length of steps taken; even when the direction is not known to the participant (McIlroy and Maki, 1995).

1.2 The Central Nervous System Control of Reactive Balance

1.2.1 Animal Models and the CNS Control of Balance Reactions

Falls are time sensitive and therefore require the rapid recruitment of reactive muscles for equilibrium to be regained. It has been previously depicted in literature that due to the speed of the compensatory movements that follow perturbations, balance reactions would have to originate from subcortical areas of the CNS (Diener, Ackermann, Dichgans, and Guschlbauer, 1985; Dietz, 1992; Keck et al., 1998; Mori, 1987). Decerebrate cats and rabbits have been observed to retain equilibrium, and even regain some postural features within days of a surgical lesion (Wood, 1964; Mori, 1987; Musienko et al., 2008; Honeycutt and Nichols, 2010). Early animal model studies emphasized that basic components of balance control are housed mainly in the brainstem, cerebellum and the spinal cord (Horak and Macpherson 1995). Other early animal model studies include inquiries in which the forebrain and motor cortex have been disconnected from other cortical sites (decorticates); as well as studies in which a lesion or an electrical stimulation is targeted to subcortical and spinal sites, showing how muscles essential for postural control can be affected and triggered without the need for the cerebral cortex (Chambers and Liu, 1957; Adkins et al., 1971; Dubrovsky et al., 1974; Mori, 1987; Asanome et al., 1998; Macpherson and Fung, 1999; Lyalka, 2005). Both lesion and stimulation approaches in animals have provided evidence of a reflex-driven theory of motor control that first sprouted from the works of Sherrington and Magnus in the early 1900s.

While balance control mechanisms are noted to derive from subcortical areas, the postural mechanisms associated with the aforementioned sites are often described as primitive, and do not encompass all aspects of balance control seen in intact animals (Bard, 1933; Fung and Macpherson, 1999; Horak and Macpherson 1995; Pratt, Fung, and Macpherson, 1994). Recent

evidence, using compact cortical recording technologies, has presented the possibility of higher-order contributions of the CNS to rapid balance reactions (Brooks, 1988; Beloozerova, 2003; Beloozerova et al., 2003; Deliagina et al., 2007; Karayannidou et al., 2009). New theories of balance control aim to provide an alternative to reflex-like models of motor control by incorporating cortical sites, and further conceptualizing balance as a distributed network in which a complex interaction exists between subcortical and cortical sites (Deliagina et al., 2007; Horak and Macpherson 1995).

1.2.2 Human Studies and the CNS Control of Balance Reactions

Neuroimaging techniques such as positron emission tomography (PET), functional magnetic resonance imaging (fMRI), and electroencephalography (EEG) have been used to reveal evidence of the involvement of the cerebral cortex during balance tasks (Dietz et al., 1984; Dimitrov et al., 1996; Fukuyama et al., 1997; Duckrow et al., 1999; Ouchi et al., 1999; Staines et al., 2001; Jahn et al., 2004; Quant et al., 2004a; Adkin et al., 2006; Slobounov et al., 2006). As explained by Maki and McIlroy (2007), the idea that balance is governed by automatic reflexes is not able to account for "online" stepping modifiability, or the distinctive speeds that exist between voluntary and reactive movements. Literature now acknowledges the cerebral cortex as an important contributor to the control of balance in humans (Jacobs and Horak, 2007; Maki and McIlroy, 2007). An emerging theory of balance control states that longer balance reaction latencies may be related to the involvement of a transcortical loop mediating rapid movements (Nashner, 1976; Chan et al., 1979; Horak and Nashner, 1986; Burleigh et al., 1994; Norrie et al., 2002; Beloozerova et al., 2003; Taube et al., 2006). An important aspect of understanding the cortical control of balance reactions derives from the measurement of the very rapid events that

underpin such reactions. The current study is focused on the use of EEG (details in ***Cortical Potentials***) largely due to the excellent temporal resolution it provides, and the ability to use this technique during rapid balance reactions with few constraints.

The importance of the cerebral cortex to balance control is additionally supported by the examination of those with specific cortical lesions, conditions and/or disease states (Maki and McIlroy, 1999). These individuals have difficulty with postural stability, even when critical sites (i.e. brainstem, cerebellum and the spinal cord) are spared (Chan et al., 1979; Diener et al., 1985; Rapport, 1993; Alexander et al., 1995; Chong et al., 1999; Pérennou et al., 2000; Garland et al., 2003, 2007; Corriveau et al., 2004; Geurts et al., 2005). Having suffered a stroke, for example, patients have been noted to have problems maintaining an upright stance (Ashburn, 1997; Geurts et al., 2005). Even when basic upright balance is regained, stroke patients find themselves heavily favoring their non-paretic leg, and therefore asymmetrically bearing body-weight (Geurts et al., 2005). Weakness, sensory perceptual changes, and behavioral strategy (e.g. pre-perturbation weight bearing asymmetry) are all related factors that hinder how efficiently stroke patients are able to respond to a perturbation (Maki and McIlroy, 1999; Lakhani et al., 2011a; Mansfield et al., 2013, 2015). The profound changes in reactive balance control after stroke reinforces the crucial role played by the different regions that make up the cerebral cortex (Lakhani et al., 2011a; Mansfield et al., 2011a, 2012).

Dual-task, or the concurrent performance of two tasks, is an approach that has been used to explore cognitive (and by extension cortical) contributions to balance control (Brown et al., 1999; McIlroy et al., 1999; Rankin et al., 2000; Brauer et al., 2002; Norrie et al., 2002; Redfern et al., 2002; Woollacott and Shumway-Cook, 2002; Quant et al., 2004b; Zettel et al., 2008; Little and Woollacott, 2014). Dual-task paradigms, and the task interference that emerges, can help

examine the cognitive substrate for the control of balance reactions (Brown et al., 1999; Maki and McIlroy, 2007). For example, when asked to complete a visuomotor task, arithmetic math problem, digit recall, auditory, or spatial memory tests, young healthy adults show signs of poor task performance and balance dysregulation if the two tasks are completed simultaneously (Kerr et al., 1985; Geurts et al., 1991; Lajoie et al., 1993, 1995; Maylor and Wing, 1996; Shumway-Cook et al., 1997; Schlesinger et al., 1998; Brown et al., 1999; Yardley et al., 1999; McIlroy et al., 1999; Rankin et al., 2000; Norrie et al., 2002; Quant et al., 2004b; Jehu et al., 2014; Little and Woollacott, 2014). Performance during dual-task can be examined through the exploration of attention shifting to show how effectively the CNS can alter the prioritization of ongoing tasks (Woollacott and Shumway-Cook, 2002; Maki et al., 2003). Some studies have utilized a continuous secondary, visuomotor tracking task in order to understand the timing of attention shifts during dual-task (McIlroy et al., 1999; Norrie et al., 2002; Quant et al., 2004b; Zettel et al., 2005, 2008). A decline in the performance of the visuomotor task, evident by a pause or absence of tracking, is hypothesized to indicate an attentional shift towards the balance reaction (McIlroy et al., 1999; Norrie et al., 2002; Quant et al., 2004b; Zettel et al., 2005, 2008). When perturbed while completing this continuous task in either a standing or seated position, a decrease (i.e. a pause) in secondary task performance is observed after the initial ankle electromyography (EMG) burst related to the earliest strategies to regain balance (APR) (McIlroy et al., 1999; Norrie et al., 2002). A delay in the secondary task interference indicates that attention may be critical for balance recovery following step initiation, meaning electrophysiological events prior to a measured task interference may be "automatic" in nature.

The study of attention shifting in a dual-task paradigm may provide a way to highlight how participants prioritize one task over another (McIlroy et al., 1999; Maki et al., 2001; Norrie

et al., 2002; Redfern et al., 2002; Müller et al., 2004; Quant et al., 2004b; Zettel et al., 2005, 2008; Maki and McIlroy, 2007). Attention reallocation could help characterize events that occur at higher-order cortical sites before and immediately following a perturbation. As previous literature shows that attention deteriorates with age, the study of attentional shifts could potentially translate to new approaches to help older adults retain and/or regain their ability to control balance (Shumway-Cook et al., 1997; Maki et al., 2001; Norrie et al., 2002; Zettel et al., 2005, 2008; Ruffieux et al., 2015). It could also be further used to examine those at higher risk of falling, such as stroke patients, whom are already known to have issues with attention (Hyndman and Ashburn, 2003).

1.3 Cortical Potentials

1.3.1 Perturbation Evoked Potentials

Due to the high temporal resolution requirements involved when exploring balance reactions, cortical responses to a perturbation have been mainly assessed using electroencephalography (Dimitrov et al., 1996; Duckrow et al., 1999; Staines et al., 2001; Quant et al., 2004a; Adkin et al., 2006; Mochizuki et al., 2008, 2010; Marlin et al., 2014). Electroencephalography is a neuroimaging technique that enables scientists to examine electrophysiological signals at the level of the outer layers of the cerebral cortex (Olejniczak, 2006). Summations of post-synaptic action potentials are recorded by electrodes placed at the scalp and pertain electrical activity deriving from the synapses of pyramidal neurons of layers III, IV and V of the cerebral cortex (Olejniczak, 2006). This technique is able to quantify externally-triggered cortical responses that are time-locked to a moment of instability (Kaiser, 2005). Time-locked fluctuations of cortical, electrical activity are known as event related potentials (ERPs); when derived from perturbations

these ERPs are described as perturbation evoked potentials (PEPs) (Xu and Ma, 2009). While the temporal resolution is very good, the spatial resolution of the EEG technique is limited due: 1) to the distance that exists between the neural event generator(s) and the scalp electrodes used to record and 2) due to volume conduction within the brain and how the structure of the brain affects the dispersion of cortical signals (Nunez, 1988; Luck et al., 2000). The continued development of this technique has led to new analysis that enables for the better localization of ERPs/PEPs and the isolation of relevant waveforms from raw EEG signals, further complementing cortical analysis and making EEG a suitable candidate for the study of reactive balance control (Dehaene et al., 1994; Miltner et al., 1994; Jung et al., 2000; Marlin et al., 2014).

In reactive balance control, both early and late cortical features have been previously characterized (Jacobs and Horak, 2007; Maki and McIlroy, 2007). Two separate potentials compose the earliest cortical responses to a perturbation: the P1 and N1 (Jacobs and Horak, 2007; Maki and McIlroy, 2007). The initial P1 potential is observed approximately 40 – 50 ms after an external perturbation, and is believed to be primarily a sensory-evoked cortical response (Quant et al., 2004a). Much like later potentials (i.e. the P2), the P1, is thought to be highly variable (Quant et al., 2004a, 2005). A more consistent, and well-studied potential, the N1, peaks ~100 – 200 ms after the onset of a perturbation (Quant et al., 2004a, 2005). The N1 is observed under a variety of externally-triggered protocols using: sitting whole-body perturbations (Mochizuki et al., 2009a), sitting translations (Staines et al., 2001), walking (Dietz et al., 1984, 1985a; Quintern et al., 1985), standing on a translating platform (Dietz et al., 1984, 1985a, 1985b; Ackermann et al., 1986; Dimitrov et al., 1996; Duckrow et al., 1999; Quant et al., 2004b, 2005; Little and Woollacott, 2014), sternal standing perturbations (Adkin et al., 2006), lean-and-release perturbations (Mochizuki et al., 2008, 2009b, 2010; Marlin et al., 2014), an ankle

pendulum protocol (Quant et al., 2004a), and even through the sway fluctuations that make up quiet standing (Varghese et al., 2015). The highest reported amplitudes for the N1 are recorded at frontocentral electrodes (i.e. Cz, FCz) (Dimitrov et al., 1996; Duckrow et al., 1999; Quant et al., 2004a; Adkin et al., 2006). Peak dipole source localization has been used to depict the supplementary motor area (SMA) as a main contributor for what composes the N1 (Marlin et al., 2014). Although well studied, the role(s) and generator(s) behind this potential are yet to be fully understood. Future research should expand upon the current knowledge of cortical contributions to balance reactions by establishing and examining the factors that may contribute to the N1 cortical response.

1.3.2 The Role of the N1

The early P1 potential is not always present in response to applied perturbations, while the late P2 cortical response tends to be more complex and variable in nature when compared to the N1 (Quant et al., 2004a, 2005; Jacobs and Horak, 2007). In contrast, the N1 is a very large amplitude potential (~ 30 μV) with a rapid onset and time to peak that is consistently evoked in response to imposed instability (Quant et al., 2004a, 2005; Mochizuki et al., 2008; Marlin et al., 2014). With respect to the potential role of the N1, it was initially thought the N1 was related to the processing of perturbation related sensory information (Dietz et al., 1984, 1985b). Using an ankle pendulum protocol, Quant et al. (2006) showed that even in the absence of muscular activity in passive trials, similar N1 PEPs were observed between those instructed to react and those told to let go of balancing the pendulum after being perturbed. As opposed to a motor act, the authors interpreted that the N1 may be a reflection of afferent information related to the applied perturbation. Indeed, research has shown increases in N1 amplitudes are observed when

preceded by larger perturbations (Staines et al., 2001). Alternatively, it has also been proposed that the N1 may be associated to changes in postural set, or CNS states between the predicted and the actual consequences of a perturbation (Dimitrov et al., 1996; Adkin et al., 2006; Mochizuki et al., 2008). Mochizuki et al. (2008) discuss that the temporal predictability of a perturbation led to an observed pre-stimulus negative drift in EEG activity and a largely reduced N1 potential, even though the amplitude of the perturbation remained the same throughout all trials (i.e. the changes recorded were independent of sensory information). In an attempt to evaluate this hypothesis, Marlin et al. (2014) examined if the N1 shared the same neural generator, the anterior cingulate cortex (ACC), as the well-documented error-related negativity (ERN) that occurs following errors in cognitive motor performance tasks (Scheffers and Coles, 2000). Source localization results of the dipole, peak amplitude of the ERN and N1 differed. The peak of the N1 was localized to the supplementary motor area (SMA), which led to the speculation that the N1 may be related to the execution/planning of motor responses at late/long-phases of a balance reaction (Marlin et al., 2014). This most recent hypothesis emerges from dual-tasking studies indicating that the earliest balance responses to a perturbation are not believed to require attention (McIlroy et al., 1999; Norrie et al., 2002). Instead, attention is deemed critical for the events that follow step initiation; a hypothesis in-line with the neural pathway model for cortical involvement proposed by Jacobs et al. (2007). Future research needs to disentangle contradictory literature in order to attain a more clear perspective of the contributions of the cerebral cortex to rapid, balance reactions.

1.3.3 Factors that influence the timing and amplitude of the N1

While the timing of the N1 response appears immutable to variations in task conditions, it should be noted that aging appears to result in a temporal delay of the N1 response (Duckrow et al., 1999). In contrast, the amplitude of the N1 is highly modifiable and has been linked to several factors: 1) perturbation amplitude or more challenging postural tasks in standing (Staines et al., 2001; Mochizuki et al., 2010), 2) perturbation predictability (Dietz et al., 1985b; Adkin et al., 2006; Mochizuki et al., 2008, 2010), 3) aging (Duckrow et al., 1999), 4) stance width (Dimitrov et al., 1996), and 5) concurrent sensory information (Staines et al., 2001). More recently there has been speculation that the N1 response may be linked to frontocentral activation related to resource reallocation in response to a perturbation (Quant et al., 2004b; Little and Woollacott, 2014). Specifically, the N1 response seems to be influenced by the dynamics of attention allocation before and immediately following a perturbation using a dual-task protocol (Quant et al., 2004b; Little and Woollacott, 2014). A change in perturbation amplitude, a more significant afferent volley, leads to a believed increase in the recruitment and temporal synchrony within neural pools that reflect onto the cortical processes related to reactive balance control. Examining how factors influence the N1 amplitude may prove beneficial for the purpose of understanding its role(s) in the processes associated with balance reactions.

1.4 Attention and the N1 Perturbation Evoked Potential

1.4.1 Attentional Networks and Attention Allocation

Attention can be defined as the information processing for the purpose of assimilating relevant incoming afferent information (Herrmann and Knight, 2001; Woollacott and Shumway-Cook, 2002; Petersen and Posner, 2012). However, such a definition dismisses the so-called

"signal domains" described by Posner (2011) to which attention is allocated to; or the idea that attention is thought to be a distributed network with sites that carry out attentional processes believed to be distinct but associated with the data processing involved in part by the motor and sensory systems (Posner, 1990; Petersen and Posner, 2012). In this more complex case, attention can be divided into different subsystems with separate attentional functions, such as: orienting, alerting, target detection, and maintenance of attention (Posner, 1990; Petersen and Posner, 2012). Pertaining to reactive balance, two main factors are relevant and thought to control attention. Firstly, attention can be modified or "biased" by previous experience, current desires and goals (Corbetta and Shulman, 2002). This cognitive factor is endogenous in nature, and reflects top-down processing from higher-order cortical areas (Corbetta and Shulman, 2002). In contrast, a sensory stimulus can also have a large effect on attention. Bottom-up processing pertains to the alerting and automatic orientation of attention to a "salient" sensory stimulus (Corbetta and Shulman, 2002). It is important to note that attention and its reallocation are critical for the successful completion of reactive movements. In a dual-task protocol, attention can be thought to be allocated through these two different mechanisms before and immediately following a perturbation.

Attention in reactive balance may be considered important for the processing of sensory information and/or in the involvement of planning and execution of motor responses. Dual-task paradigms are used to highlight the interaction between cognitive and postural tasks. Attention is known to be necessary for the successful completion of a postural task in young, healthy adults (Kerr et al., 1985). In a perturbation-based dual-task study, endogenous mechanisms are believed to drive attention towards the completion of the secondary task before a perturbation; while exogenous mechanisms related to the applied perturbation are believed to lead to the reallocation

of attention towards the successful completion of balance reactions. Note that attentional demands increase with the difficulty of the primary postural task (Lajoie et al., 1993). However, the study of attention and postural control remains underdeveloped. Confounding factors, such as arousal, the secondary task used, and the instructions used can affect the observed values of studies inquiring about the association between attention, through a dual-task paradigm, and balance control (Maki and McIlroy, 1996; Yardley et al., 1999; Fraizer and Mitra, 2008). It is then important to advance our understanding of the role of the cerebral cortex and its association with the attentional components associated with reactive balance reactions.

1.4.2 The Influence of Dual-Task on PEPs and Balance Reactions

The study of attentional dynamics is an inquiry into the temporal features of processing related to an externally-triggered perturbation (Maki and McIlroy, 2007). In reactive balance control, attention is believed to be critical, especially in the late phases of a balance reaction (McIlroy et al., 1999; Norrie et al., 2002). When a perturbation is applied while dual-tasking, a pause in the completion of the secondary task is observed about 300 ms after the onset of a perturbation (Maki and McIlroy, 2007). The length of the pause in secondary task performance is believed to be an associated requirement for reallocation of attention possibly linked to the control of the balance task (Norrie et al., 2002). Some dual-task studies have also stated that balance responses following a perturbation may be affected significantly by the attentional demands of the secondary task (McIlroy et al., 1999; Rankin et al., 2000; Norrie et al., 2002; Quant et al., 2004b). The task related change in balance responses can be seen both through the observations of EMG muscles around the ankle, tibialis anterior (TA) and the medial gastrocnemius (MG), and CoP parameters (Peak displacement). An increase in EMG and CoP

amplitude is seen after the initial APR phase (~ 80 -150 ms post onset) of balance recovery under a range of dual-task, perturbation-based paradigms (McIlroy and Maki, 1993; McIlroy et al., 1999; Rankin et al., 2000; Norrie et al., 2002; Quant et al., 2004b). Cognitive involvement is believed to occur after the APR phase and be present until balance is fully recovered; within a window termed the late/long-phased balance reactions (Jacobs and Horak, 2007; Marlin et al., 2014). Authors Rankin et al. (2000) and Quant et al. (2004b) analyzed EMG activity around the ankle, under dual-tasking paradigms, in small epochs in order to better depict changes in balance reactions associated with divided attention. These small intervals, 0 - 50, 50 - 100, and 100 - 150 ms post-muscular activation for Quant et al. (2004b) and 36 – 70, 70 – 150, 150 – 350, and 350 – 500 ms post-muscular activation for Rankin et al. (2000) enable a more specific look into continuous EMG data. Only the events that follow the APR (+350 ms) are commonly termed late-phased balance reactions, although no definite boundaries have been previously described (McIlroy and Maki, 1993; Jacobs and Horak, 2007; Maki and McIlroy, 2007; Bolton, 2015b). In dual-task paradigms these late-phased balance reactions occur between the time attention shifts away from the secondary task (i.e. pause onset), up until the point the individual regains balance and returns attention to the secondary task.

While the motor responses vary, there is more predictable changes in the N1 response during dual-task studies. Typically the N1 is attenuated compared to non-dual-task trials (Quant et al., 2004b; Little and Woollacott, 2014). These recent studies highlight the possibility that the N1 may be importantly associated with the allocation of attentional resources. As the role(s) and generator(s) behind the N1 potential are not fully understood, dual-tasking paradigms, alongside EEG, may help to characterize the factors that influence the early N1 cortical response; thus, expanding our current knowledge of cortical contributions to reactive balance control.

It has been previously proposed that attention is a limited resource (Egeth and Kahneman, 1975). Arguably the attenuation of the N1 response may arise due to the lack of available resources under dual-task conditions. For example, as proposed by Quant et al (2006) when completing a challenging cognitive task prior to a perturbation, in this case a visuomotor tracking task, attention is allocated to the performance of the secondary task. The diminished N1 post-perturbation potential may highlight that, due to the difficulty of the secondary task, decreased neural activity linked to the allocation of attentional resources when balance and mental task share some of the same cognitive resources. The authors further note that the N1 attenuation was observed alongside a change in later phases of balance recovery reactions. This indicates that the N1 amplitude reduction related to an attentional shift may be associated with the altered motor responses that follows a perturbation. However, this explanation of the observed N1 attenuation does not take into account the interaction that exists between endogenous and exogenous attentional subsystems under a dual-tasking, perturbation-based paradigm. While completing the secondary task, attention is allocated endogenously and maintained through the same mechanisms until the point at which a perturbation is presented. Immediately following a perturbation, attention is automatically allocated to the applied perturbation, through exogenous mechanisms. Reactive balance related cortical processes, reflected by the N1, may be modulated by the interaction existing between endogenous and exogenous processes before and immediately after a perturbation.

Understanding of the dual-task influences on perturbation cortical potentials and balance recovery has the potential to provide insight into the role of attention and the specific role of the evoked N1 response. Of very specific interest is the role that attention/resource reallocation may play in the generation of the cortically evoked responses to balance perturbations. The work

conducted within this thesis is intended to provide a fundamental understanding of the role of the cortex in reactive balance control.

1.5 Study Rationale

Dual-task studies provide indirect evidence that cortical, higher-order sites of the CNS are involved in the process of balance control. When dual-tasking, task interference effects are thought to reflect that both the cognitive and balance task share some of the same cognitive resources. In a reactive strategy protocol, challenges in dual-tasking are associated with transient periods of instability, specifically the moments associated with generating a balance reaction (McIlroy et al., 1999; Norrie et al., 2002). While the behavioural consequences of dual-tasking are well described, the underlying mechanisms or processes are less understood; specifically those related to the temporal dynamics of resource allocation believed to be important to the recovery of balance after a perturbation is applied (Maki and McIlroy, 2007). The current study combines an attention demanding cognitive task during reactive balance control along with EEG, to examine attention allocation and its relationship to the N1 response thought to be influenced by attention.

It is likely attentional processing underlying reactive balance control is complicated by a mix of endogenous and exogenous processes. Attentional resources can be allocated endogenously, by an individual's goals and intentions, or exogenously, through an automatic process evoked by an external stimulus (Corbetta and Shulman, 2002). Endogenous and exogenous attention reflect distinct aspects of CNS processing and are believed to each encompass dominance over the allocation of attention at different times during balance recovery evoked by a perturbation. When dual-tasking, attentional resources are endogenously allocated to

the secondary task. Upon a perturbation, an automatic shift in attentional resources is exogenously evoked to the external "salient" stimulus threatening equilibrium. In the case of reactive control, the perturbation (stimulus), evokes exogenous attention reallocation. Larger amplitudes of a perturbation presumably require a greater redirect of resources (neural activity). This is proposed to account for the larger amplitude N1 response associated with larger amplitude perturbations (Staines et al., 2001), but also the smaller N1 seen under predictable stimulus conditions when attention is already directed to the stimulus prior to perturbation (Mochizuki et al., 2008; Jacobs et al., 2009). However, the N1 amplitude is also attenuated when the subject is dual-tasking (Quant et al. 2004b, Little and Woollacott (2014). This appears to conflict with the idea of smaller N1 when stimulus is temporally predictable since one might imagine that a larger N1 would occur in conditions where attention is directed to a secondary task. To explain this, it may be that endogenous orienting associated with the secondary task suppresses or masks the electrophysiological expression of exogenous resource allocation associated with the perturbation. However, it is also possible that the amplitude of the N1 response in the single task condition may have been an artifact of the task condition. Specially, the perturbation amplitude was small in both the Quant et al. (2004b) and the Little and Woollacott (2014) studies. The smaller N1 potential during dual-task trials may not impact balance control, since the stimulus and challenge of control was modest. A larger perturbation may demand greater resource allocation impacting the balance between exogenous and endogenous attention. The current study focused on whether under higher amplitude perturbations the N1 response would be augmented in spite of a concurrent dual task. This would reinforce the idea that the N1 is potentially associated with exogenous attention shifts.

1.6 Objectives and Hypotheses

The main objective of this thesis was to characterize the relationship between attention, perturbation evoked cortical activity, and the associated balance reaction. Specifically, it was proposed that the amplitude of the N1 response reflects the stimulus driven exogenous reallocation of attention. This was explored by varying both the background state by manipulating endogenous attention via dual-tasking, and the amplitude of perturbation, the stimulus that evoked exogenous responses.

1.6.1 Primary Objective and Hypothesis

The primary objective of this thesis was to investigate how dual-task performance influences cortical responses associated with instability evoked by different perturbation amplitudes (***Figure 1***). As previously mentioned, it was hypothesized that the amplitude of the N1 cortical response is related to the allocation of exogenous attention in response to the applied perturbation. Higher amplitude perturbations are known to lead to larger N1 amplitudes as illustrated by Mochizuki et al. (2010), and Staines et al. (2001). Therefore, it was proposed that greater exogenous reallocation, driven by larger perturbation amplitudes, would reflect the attentional demands necessary for the successful completion of balance reactions (McIlroy et al., 1999; Norrie et al., 2002). The N1 response is reduced under conditions when the stimulus is temporally predictable, independent of motor or sensory responses (Dietz et al., 1985b; Adkin et al., 2006; Mochizuki et al., 2008, 2010). This supports, indirectly, the potential link to exogenous attention since under predictable conditions there would be a reduction in exogenous reallocation as the participant is cued and preparatory processes related to the perturbation could be taken possibly explained by what is already observed to occur cortically before a predictable

perturbation (Mochizuki et al., 2008; Jacobs et al., 2009). However, as noted, the amplitude of the N1 response appears to decrease when individuals perform a concurrent secondary task when the participants experience an external perturbation (Quant et al., 2004b, Little and Woollacott, 2014). This would appear, on the surface, to conflict with the idea that the N1 is linked to exogenous attention since a greater sense of temporal unpredictability should be experienced by the participant when presented with an unpredictable perturbation *while* dual-tasking. An alternate explanation for this observation is that N1 amplitude in single-task trials exhibit an added attentional precaution or "safety net" by the CNS in light of a small perturbation. The mechanisms that drive attention following a perturbation are stimulus driven, and of greater priority if larger perturbations were to be presented.

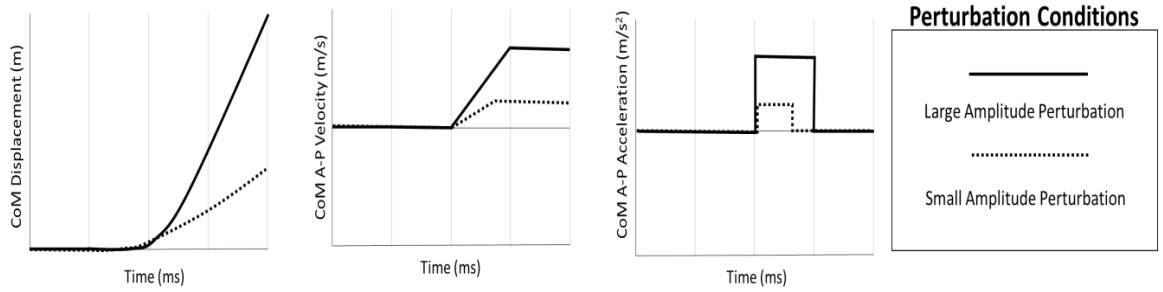

Figure 1- Expected Displacement, Velocity, and Acceleration Profiles for Experimental Conditions. Experimental conditions in which the initial acceleration was modulated: small perturbation amplitude (S) and large perturbation amplitude (L). All measures (displacement, velocity, and acceleration) are in relation to the center of mass (CoM) in the anterior-posterior (A-P) plane.

It was proposed that the N1 response amplitude would be affected by perturbation amplitude, inferred to be associated with a demand for exogenous attention. The effects of dual-task on secondary task performance were largely expected to be dependent on the characteristics of the perturbation linked to changes in attention. During large perturbation, high acceleration

trials leading to a step response, it was hypothesized that there would be no dual-task influence on the N1 response. This is due to the greater balance challenge, and greater reallocation of attention through exogenous processes. However, in small perturbation, low acceleration trials eliciting a feet-in-place response, an attenuated N1 potential would be observed for dual-tasking trials, much like those recorded by Quant et al. (2004b), and Little and Woollacott (2014). Large perturbations demand greater exogenous attention reallocation due to exogenous processes and its relationship with greater balance challenge. This will translate into large perturbation dual-tasking trials exhibiting earlier and longer task interference (i.e. pauses in secondary task performance), much like those previously observed by McIlroy et al. (1999), and Norrie et al. (2002), when compared to small perturbation dual-tasking trials.

1.6.2 Secondary Objective and Hypothesis

While dual-tasking, only minimal changes in the quality of late-phase reactions were seen by Little and Woollacott (2014), and Quant et al. (2004b); even though, moderate N1 attenuations were observed immediately after a perturbation. The suggestion is that the attenuated N1 response had no influence over the control of the evoked balance reaction leading to the view that such shifts in attention were unrelated to the reactions evoked by instability. The present thesis sought to explore if changes to the N1 cortical response due to dual-tasking were associated with behavioral consequences that may be more evident under greater challenges to stability higher amplitude perturbations. With the presentation of large perturbations while dual-tasking, this would entail a quicker step onset and foot-off when compared to the small perturbation trials, a less prioritized event. This would also translate into large perturbation dual-tasking trials exhibiting earlier and longer task interference (i.e. pauses in secondary task

performance), much like those previously observed by McIlroy et al. (1999), and Norrie et al. (2002).

Chapter 2: Methods

2.1 Pilot Studies

Overall, six young, healthy adults volunteered to be collected. During the study, the first pilot consisted of two participants tested to explore how suitable two different methodologies could be at answering questions regarding dual-tasking under a LaR protocol. The first method tested was the methodology of Mochizuki et al. (2010), who utilized a secondary cable that takes the body weight load and is immediately taut after the release of the main supporting cable leading to a very small amplitude perturbation. This method proved to be difficult to modify as the amount of acceleration post-perturbation was dependent on the length of the secondary cable, leading to small precision. A greater amount of precision was pursued during the second method in which a bungee like, elastic cable was used, resulting in a slower initial and peak acceleration. Much like the rigid secondary cable by Mochizuki et al. (2010), the bungee cord led to a feet-in-place reaction (no stepping response required) where the release without secondary cable always resulted in stepping reactions generally confirming the difference between perturbation amplitudes. However, this method additionally meant that the participant could decipher which condition they were in since the elastic cable would have to be pre-stretched prior to perturbation onset. Both cords made it difficult to attribute any changes that were seen to be related to the slowing down of the participant or due to knowing which trial condition was being completed. While it was possible to evoke N1 responses there was considerable challenge in controlling the amplitude of perturbation between subjects without the subject being able to anticipate the trial condition.

The results from the first pilot study and the proposal feedback led to the development of a third system using a hydraulics that was tested in four additional pilot subjects. The hydraulic

pipe enabled the participant to be decelerated post-perturbation, without cueing the participant of the condition that was to be presented. There appeared, based on A-P acceleration profiles, to be improved control over the amplitude of perturbation. During this phase of testing the specifics regarding the dual-tasking paradigm were also evaluated (**see *Dual-Task Protocol***). It is important to also note that the introduction of a dual-tasking paradigm to a LaR protocol has not been previously published; nor has the introduction of different perturbation amplitudes while recording electrical cortical activity.

2.2 Main Study

2.2.1 Participants

Twelve young, healthy adults (aged 18 – 35 years, 22.5 SD 2.43 years) were recruited for the study. A sample size calculation (Eng, 2003; Vanvoorhis and Morgan, 2007) was completed using the size effect average from Quant et al. (2004b), Little and Woollacott (2014), and pilot data. Twelve volunteers places the present study within ranges of previous lean-and-release (LaR) and/or dual-tasking literature in which between 7 to 14 participants have been tested (Quant et al., 2004b; Mochizuki et al., 2010; Little and Woollacott, 2014; Marlin et al., 2014). This study received ethics clearance through a University of Waterloo Research Ethics Committee and all participants completed a consent form prior to testing.

Participants were excluded if they had a self-reported history of lower limb injuries and/or neurological impairments according to a health status form. Demographic data collected included: 1) age, 2) biological sex, 3) height, 4) weight, 5) past health problems, 6) present health problems, 7) medications within the last 2 weeks, 8) symptoms within the last 2 weeks, 9)

current physical activity status, and 10) limb dominance. Limb dominance was assessed using a foot preference survey (Chapman et al., 1987).

2.2.2 Experimental Procedure

All experimental procedures were set prior to testing and were completed in one session. The experiment consisted of two blocks, 60 trials each, in which dual-task challenge (single-task (ST) vs. dual-task (DT)) and perturbation amplitude (small (S) vs. large (L)) were randomly presented. Dual-tasking trials involved a visuomotor tracking task before and after a perturbation (McIlroy et al., 1999; Norrie et al., 2002; Quant et al., 2004b).

Prior to the experimental blocks, participants were introduced to the protocol through the completion of 5 randomized standing, practice trials per condition (20 trials total). To confirm participants were fully attending during trials, an additional two seated blocks of the secondary task were completed before standing practice trials (5 – 10 trials) and after the experimental blocks (5 trials). Seated block trials were completed until a plateau in secondary task performance was reached. A plateau entailed a low variability score, better performance (***see Dual-Task Protocol***). Seated trials allowed the visual tracking task to be collected in while the participant was not being perturbed or maintaining balance. Seated rest took place after every experimental block or whenever a participant expressed the need for a break. In total, 30 experimental trials per condition (e.g. STS, STL, DTS, DTL) were recorded: 60 large perturbations leading to a step (L), 60 small perturbations leading to a feet-in-place reaction (S), and 150 – 155 trials grand total.

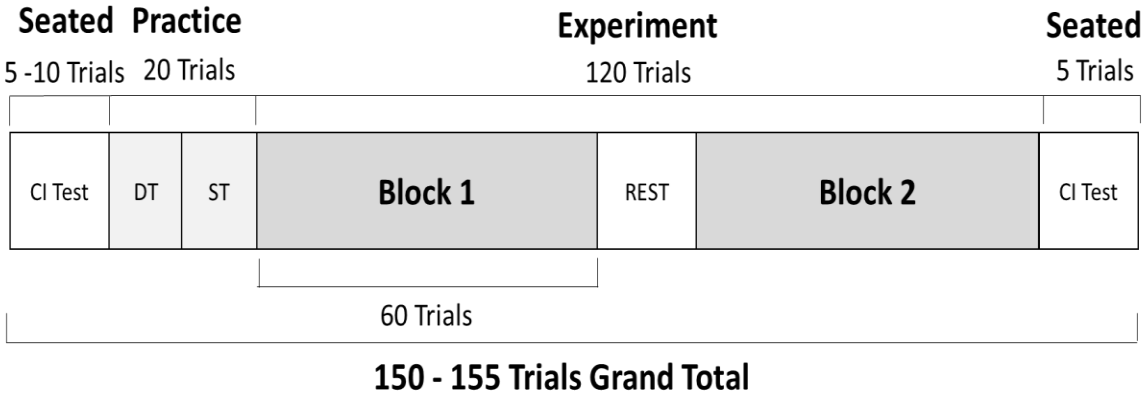

Figure 2- Experimental Procedure. Twenty practice trials took place before experimental recordings as well as seated secondary task trials before the practice and after the second experimental block. Each experimental block consisted of 60 randomized single-task (ST) or dual-task (DT) trials in which perturbation amplitude was variant (small vs. large).

2.2.3 Perturbation Amplitude

In order to elicit cortical responses, researchers have used a wide range of different methods, most of which are used to modulate the amplitude of the applied perturbation (Dietz et al., 1984, 1985a, 1985b; Quintern et al., 1985; Ackermann et al., 1986; Dimitrov et al., 1996; Duckrow et al., 1999; Staines et al., 2001; Quant et al., 2004a, 2004b, 2005; Adkin et al., 2006; Mochizuki et al., 2010, 2008, 2009a, 2009b; Varghese et al., 2015). Lean and release (LaR) protocols are currently the method to evoke rapid onset perturbations as an alternative to platform translations (Mochizuki et al., 2008, 2009b, 2010; Lakhani et al., 2011b; Marlin et al., 2014; Varghese et al., 2014). LaR protocols enable the application of perturbations that are low cost, yet lead to similar behavioral outcomes when compared to platform translations. The release of the supporting cable was automatically triggered by a solenoid device. To restrict temporal predictability of the upcoming perturbation, the timing of the cable release was randomized by computer to release the solenoid over a window of 20-seconds. Two perturbation conditions were tested: 1) small perturbation (S) condition leading to a feet-in-place reaction, and

2) large perturbation (L) condition leading to a stepping reaction (***Figure 3***). Note, all perturbations were presented at the same leaning angle, and therefore body weight percentage (~ 5% BW recorded by a force transducer, ***see Figure 12***), so that participants would not be able to be cued, predicting the amplitude of the perturbation. To elicit different perturbation amplitudes (small vs. large), while maintaining the same body weight percentage on the supporting cable, a hydraulic pipe was used to slow the initial post-release acceleration. Although a secondary "catch" cable for a lean-and-release protocol has been previously described by Mochizuki et al. (2010), it was depicted as rigid in nature and described as being taut after a small window of time post-perturbation. A hydraulic pipe was used in this study in order to decelerate the participant immediately, without a delay, due to the possibility that the N1 is a representation of post-perturbation afferent information (Adkin et al., 2006). Furthermore, before the initiation of any trial, a minimal amount in muscular activity (EMG bursts) surrounding the dominant ankle (tibialis anterior and medial gastrocnemius) was confirmed by the experimenter via display in order to minimize any anticipatory postural adjustments (Maki and McIlroy, 1999). Participants were instructed to keep body segments aligned to maintain consistency throughout every trial and diminish the possibility of any upper body movements. Extraneous upper body movements were limited by an upper limb sling, firmly securing the dominant (DT trials, tracking) hand to the body, even when no secondary task was completed (ST trials). The use of a sling diminished muscular activity and any movement related disruption of tracking performance; both of which increase noise in the bioelectrical signals of interest and are unrelated to the goals of this thesis (Maki and McIlroy, 2007).

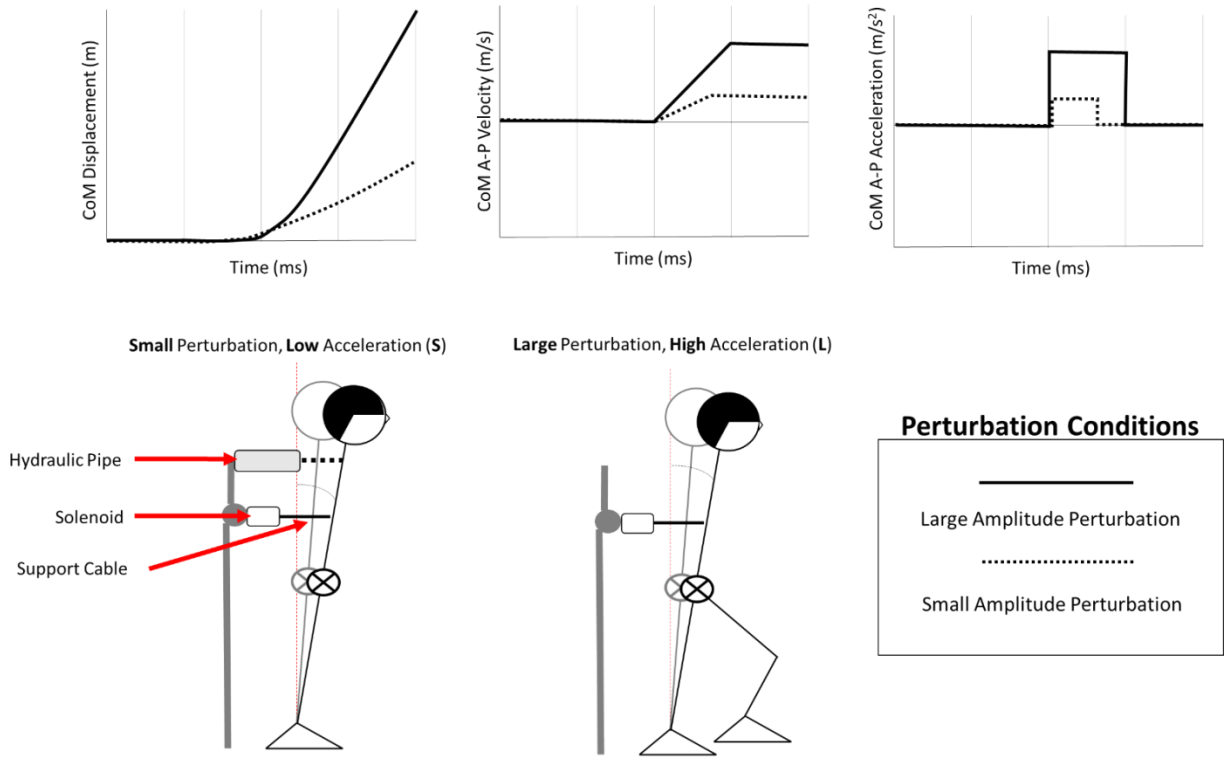

Figure 3- Perturbation Conditions. Perturbation conditions and their expected CoM displacement (m), velocity (m/s) and acceleration (m/s^2) profiles in the anterior posterior (A-P) plane. Conditions include a small perturbation, low acceleration condition (left) and a large amplitude, high acceleration condition (right). No stepping reactions were elicited by small amplitude perturbations due to the utilization of a secondary hydraulic pipe.

2.2.4 Single-Task Protocol (posture only)

The single-task protocol consisted of a temporally unpredictable perturbation leading to either a stepping or feet-in-place reaction. A standardized foot position was set before each trial to allow for between subject comparisons (McIlroy and Maki, 1997). At the beginning of an experimental block, participants were instructed to react to an upcoming perturbation as naturally as possible. At the beginning of every trial, participants were instructed to fixate on a small target centered on a computer screen 2 meters away. During the first 10-seconds of a trial, no perturbation was applied; however, following this time, a perturbation was applied within a

random 20-second time window. The length of this window was set to minimize temporal predictability as it is known to influence N1 amplitudes (Dietz et al., 1985b; Adkin et al., 2006; Mochizuki et al., 2008, 2010). Participants were asked to react naturally post-perturbation and when presented with a large perturbation, participants were asked to hold their step position until the end of the trial (30-seconds post-perturbation onset) while fixating on the screen. During small perturbations, participants regained balance and returned to their original position after being released.

2.2.5 Dual-Task Protocol

At the beginning of a dual-task trial, participants were instructed that they would need to respond as naturally as possible to an upcoming perturbation, but that if the screen showed the dual-tasking protocol, they would need to focus on the accurate completion of said cognitive task. At the beginning of each trial, participants were first instructed to fixate on a small target centered on a computer screen 2 meters away. This same screen was used for the tracking task and displayed all relevant information for the subject (***see Figure 4***). The visuomotor task took 30 seconds and consisted of tracking a random waveform using a cursor guided by a trackball wireless mouse held by the participant's dominant hand which was supported by a sling (McIlroy et al., 1999; Norrie et al., 2002; Quant et al., 2004b). Three sinusoids at random phases (degrees) were summed to create the presented waveform (Quant et al., 2004b). The first 10 seconds of visuomotor task performance, pre-perturbation, were used to evaluate task performance by comparing values to the post-plateau of the seated block. A randomly timed solenoid release occurred within a 20 second window following the initial 10 seconds. Performance after each trial was expressed on the same display as the visuomotor task, and was represented as the error

rate (root-mean-squared, RMS) measured from the start of the secondary task up until the onset of the perturbation. After each trial, the participants were instructed to try and improve performance if the trial error rate was larger than the average error rate from the final trials (post-plateau) of the seated practice block (***Figure 2***). The seated practice (post-plateau) error rate was known to the experimenter and was used to encourage feedback for subjects to attend to the secondary task throughout the length of the study. Lastly, participants were asked to maintain their post-perturbation stepping position (when presented with a large perturbation) until the end of the 30 second secondary task. This was done to limit attention shifts in the 30-seconds data collection window that may have been associated with foot placement back to the starting location.

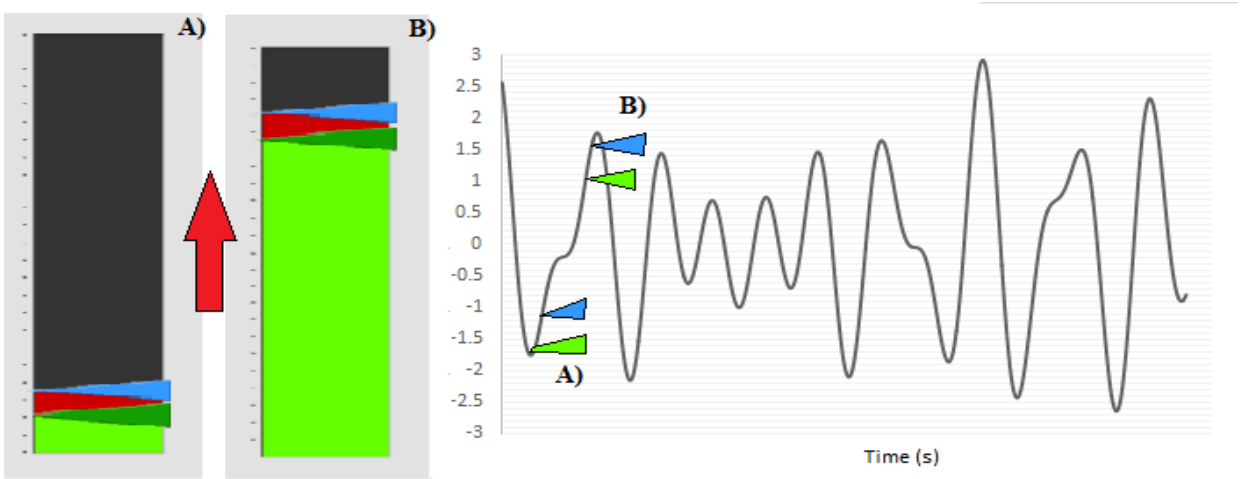

Figure 4- Visuomotor Tracking Task. Visuomotor tracking task as seen by participant (left and middle). Participant instructed to track waveform (blue cursor) using trackball wireless mouse (green cursor). A) (before) and B) (after) refer to different timepoints during a trial (right).

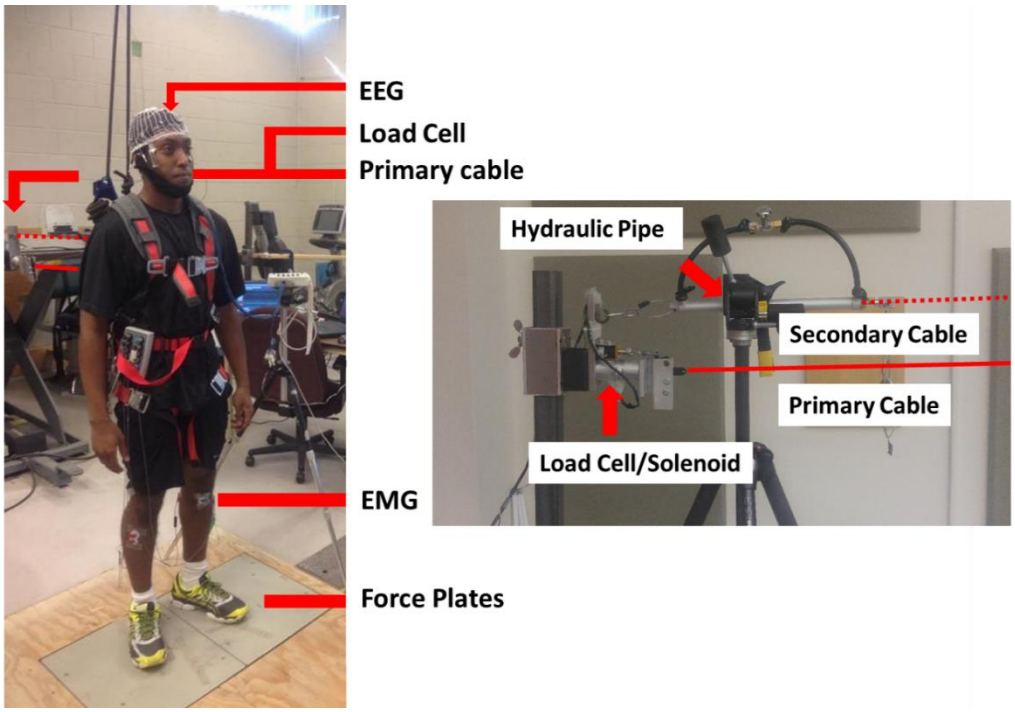

Figure 5- Experimental Set-up. The primary cable connected to the solenoid and load cell used to collect BW%. The secondary cable connected to the hydraulic pipe used to elicit a feet-in-place reaction.

2.3 Measures and Data Acquisition

2.3.1 Postural Perturbation

Force supported by the main cable holding the leaning position was recorded by an in-line force transducer secured to a platform (Transducer Techniques, Temecula, CA, USA). Perturbation onset was determined as the moment the amplitude of the recorded values of the in-line force transducer dropped below the pre-perturbation 99% confidence band (***see Figure 6***) (Marlin et al., 2014).

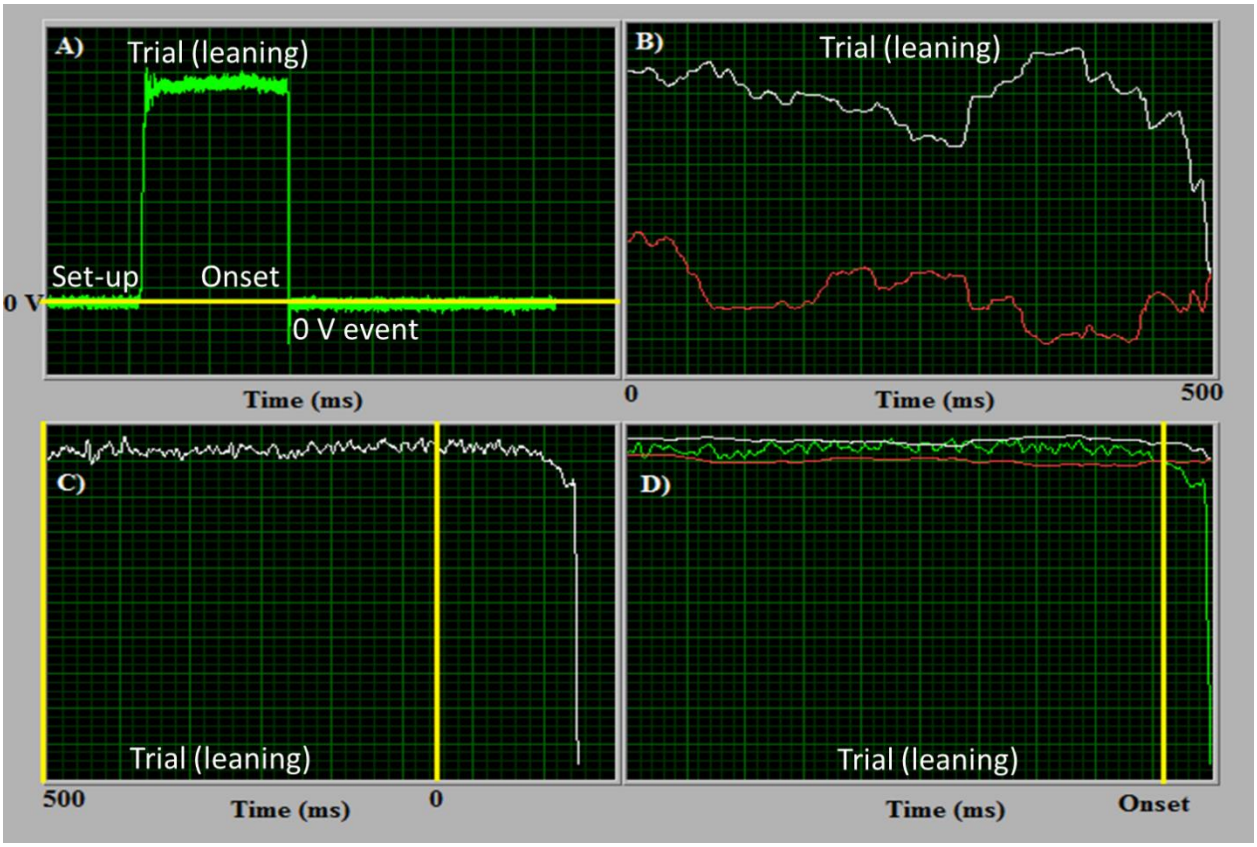

Figure 6- Load Cell Processing and Perturbation Onset Calculation. Filtered load cell data (A). Epoched 500 ms window before load cell reaches 0 V (B). Load cell 750 ms before 0 V event; highlighted 500 ms epoch used to calculate 99% confidence interval (CI) (C). Load cell 750 ms window (D) including upper CI (white) and lower CI (red) values used to calculate perturbation onset (yellow line).

2.3.2 Electroencephalography (EEG)

A 32 channel EEG cap (Synamps Quikcap, Neuroscan, Charlotte, NC, USA) was used alongside a Synamps^2 amplifier (Neuroscan, Charlotte, NC, USA) to record raw cortical brain activity at 1000 Hz (NeuroScan v4.3, El Paso, TX, USA). Channel layouts for this study followed the international 10-20 system, with each channel measuring an impedance level below 5kΩ prior to recordings. Impedance levels were checked before any collection and after each experimental block. Linked mastoids served as the reference system to all other recording

electrodes on the EEG cap. Eye movements were recorded using four surface electrodes and stored offline for the purposes of noise reduction during EEG analysis.

2.3.3 Head and Trunk Accelerometry

Head and trunk (sternum) acceleration was captured by two tri-axial accelerometers (Shimmer3, Shimmer Sensing, Dublin, IRE). Accelerometers were attached to the participant using Velcro straps. The head accelerometer was placed over the EEG cap, above the ear, in-line with midline electrodes, and parallel to the ground. The sternum accelerometer was placed over the sternum, 7.5 cms inferior to the jugular notch and running perpendicular to the ground. Each accelerometer sampled at a rate of 102.4 Hz and experimental trials were recorded using a customized LabVIEW program (LabVIEW v14.0f1, National Instruments, TX, USA). The accelerometers were temporally synchronized to the onset of perturbations.

2.3.4 Electromyography (EMG) and Ground Reaction Forces (GRFs)

Balance reactions were assessed using ground reaction forces through the use of two force plates (Advanced Medical Technology Inc., Watertown, MA); while muscular activity surrounding the ankle joint was gathered using EMG. Participants commenced each trial standing on the force plates. Muscular activity was collected from the tibialis anterior (TA) and the medial gastrocnemius (MG) muscles of the dominant/stepping foot (x1,000 amplification; AMT-8, Bortec Biomedical, Calgary, AB). Two self-adhesive Ag/AgCl electrodes were placed longitudinally on each muscle belly with the ground being placed at the proximal tibia. All EMG and force plate data was sampled at a rate of 1000 Hz with the use of a digitizer (BNC-2111, National Instruments Corporation, Austin, TX, USA).

2.4 Data Analysis

2.4.1 Electroencephalography (EEG)

Event-related potentials were analyzed at the FCz and Cz electrodes as the largest N1 amplitudes over various conditions and protocols have been localized to this frontocentral area (Adkin et al., 2006; Dimitrov et al., 1996; Duckrow et al., 1999; Marlin et al., 2014; Mochizuki et al., 2009; Quant et al., 2004b). EEG data was processed using EEGLAB (Delorme and Makeig, 2004). Raw EEG signals were band-pass filtered (high-pass: 1 Hz; low-pass: 30 Hz), and down sampled to 250 Hz, before being epoched around the perturbation onset (before: -500 ms; after: 500 ms). Independent component analysis (ICA) was used to remove noise components of the raw EEG related to eye and muscular activity (Jung et al., 2000). Four electrooculography (EOG) electrodes enabled for the elimination of ICA components which had scalp topography typical of eye blinks and eye movements. Similarly, ICA components related to muscular activity were removed based on both topography and frequency composition. Once the dataset was filtered and artifacts removed, epochs were subjected to a baseline removal (– 250 ms to 0 ms). If noise was unable to be removed, corrupted epochs were discarded manually from future analysis. N1 amplitude was calculated by comparing the pre-perturbation baseline measures to the peak of the first negative post-perturbation potential that occurred before the first 300 ms post onset. The N1 latency was calculated by measuring the time between perturbation onset and the peak of the first electrical negativity peak.

2.4.2 Accelerometry

The gravitational component was eliminated through the subtraction of the mean of the signal and then zero-phase high-pass filtered (0.3 Hz; Martinez-Mendez et al., 2011, 2012). Data

was then aligned and epochs were created around the perturbation onset (before: -500 ms; after: 500 ms). The acceleration parameters of interest (average 100ms acceleration, peak acceleration and peak acceleration time) were isolated for the A-P plane due to the nature of the LaR protocol used.

2.4.3 Secondary Task in Dual-task Conditions

To determine if participants were engaged in the secondary task, tracking variability (root-mean-squared, RMS) values were calculated. The RMS is a measure of how much a participant's waveform deviated from the waveform generated/presented. An enhancement was expected during practice trials indicating participants actively sought to increase their performance on the secondary task. Allocation of attention towards the secondary task prior to perturbations was expected to result in no difference in the RMS between seated conditions, after a plateau is reached, and dual-task trials. A record was made of relative pre-perturbation performance (RMS) for secondary analysis. Previous literature has shown that when reacting to a perturbation while attending to a secondary task, a pause in tracking performance is observed 325 ms after the onset of a perturbation under a seated protocol, and 480 ms under a standing protocol (McIlroy et al., 1999; Norrie et al., 2002). Pauses are considered the moment in which a participant discontinues waveform tracking, up until they return to the task at hand (***see Figure 7***). This was measured by averaging all seated trials (pre and post experimental blocks) and then calculating a 95% confidence interval (CI) band. All experimental trials were then compared to the calculated CI band to highlight lapses in performance. The 'Start' of the cognitive task interference pause occurred once a deviation from the CI band was detected, meaning they were believed to be attending to the balance task. The 'End' of task interference was defined when the

tracking signal (mouse tracing) re-entered the CI band due to tracking initiation. The difference between the 'Start' and 'End' of task interference was then calculated as the task interference pause length.

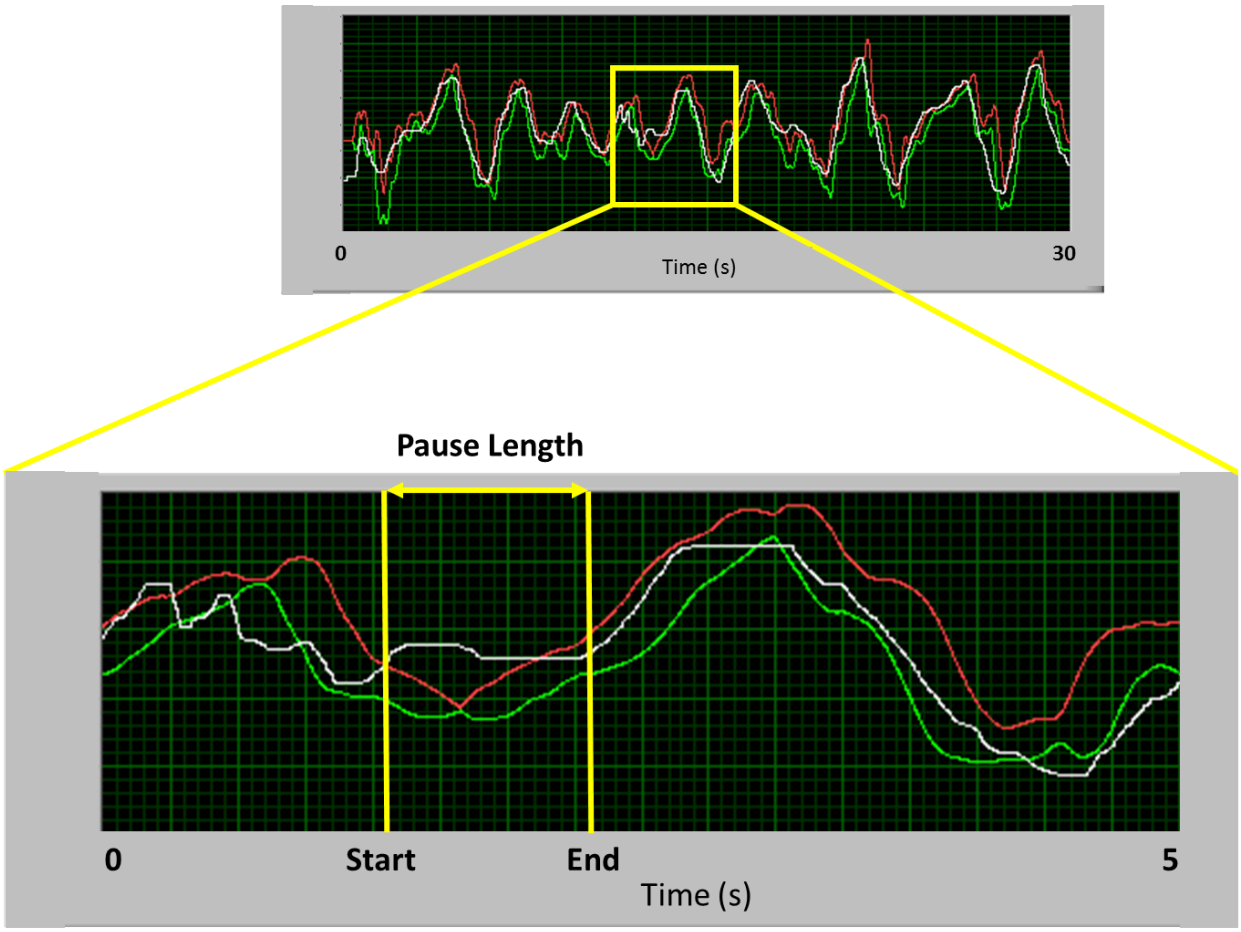

Figure 7- Dual-Tasking Interference. A participant's visuomotor tracking performance during a large perturbation dual-task trial (white line). Calculated upper (red) and lower (green) CI bands appear in both the entirety of the trial (above) and an epoched window (below). The deviation from tracking is highlighted by the 'Start' vertical cursor and the reinstatement of attention towards the secondary task is highlighted by the 'End' vertical cursor. The length of the task interference pause is denoted by the difference in time between the 'Start' and 'End' events.

2.4.4 Electromyography (EMG)

Electromyographic activity of TA and MG muscles was first hard-ware filtered (band-pass 10-10,000 Hz). Digitized data was then band-pass filtered (second order Butterworth, 20-450 Hz), baseline corrected, full-wave rectified, and then low-pass filtered (2^{nd} order, 5 Hz Butterworth). The threshold detector used to detect EMG onset consisted of the measurement of a 99% CI using baseline EMG activity prior to the perturbation onset (750 to 250 ms before onset). Full-wave rectified EMG data was further analyzed to determine the magnitude of EMG signals. This included EMG peak amplitude, EMG peak amplitude time, and the area under the curve for three time intervals: EMG onset - 50ms, EMG onset - 100 ms, and EMG onset - 200 ms.

2.4.5 Center of Pressure (CoP)

All force plate data was filtered using a low-pass filter (4^{th} order, 10 Hz Butterworth). Center of pressure parameters (step onset and foot-off time) were calculated using a customized LabVIEW program (LabVIEW v14.0f1, National Instruments, TX, USA). This customized program was used to analyze the effects of dual-tasking challenge on force plate parameters for each perturbation condition separately. The anterior-posterior (A-P) center of pressure baseline and onset was calculated based on previous work by Norrie et al. (2002), Marlin et al. (2014), and Mochizuki et al. (2008). Baseline, mean (cm) and standard deviation values were determined for each trial using a pre-perturbation window of 1000 ms. The step onset was determined as the time (ms) after a perturbation in which the AP-CoP amplitude exceeded 5 standard deviations from the baseline. An additional inclusion of timing for foot-off was determined for the conditions in which a stepping-reaction was evoked (L). Foot-off was considered the time point

at which the vertical load recorded by the force plate on the stepping leg was below 1% body weight (Lakhani et al., 2011). All timing values for force plate reactions are displayed relative to the onset of the applied perturbation.

2.5 Statistical Analysis

2.5.1 Task Effects

A two-way repeated measures analysis of variance (ANOVA) was used to determine the effect of factors: dual-task challenge (ST vs. DT) and perturbation amplitude (Small vs. Large) on the following dependent measures: cortical activity (N1 amplitude and latency), perturbation amplitude (pre-perturbation load cell body weight, averaged 100 ms A-P acceleration, peak acceleration, and peak time acceleration for accelerometer data), behavior responses (EMG onset, EMG amplitude, CoP step onset, and CoP foot off). All statistical analysis included the determination of possible interaction effects among factors of interest. An alpha value of 0.05 was used to denote statistical significance. All statistical calculations were completed using the Statistical Analysis System software (SAS v9.4, Cary, NC, USA).

2.5.2 Secondary Task Performance

Secondary task performance, average participant error rates (RMSs), associations to the number of trials were analyzed using a Pearson product moment correlation coefficient. Error rates were derived from the first 10-seconds of tracking in which no perturbations were presented and these were compared using a t-test against the averaged error rates for each of the blocks that compose the experiment. The t-test (Dunnet) assigned as control the initial seated Block1 (CI block, ***see Figure 2***) used to attain the error rate (RMS) that the experimenter relied on to

encourage dual-tasking participation throughout the experiment. Lastly, two separate t-test comparisons (DTS vs. DTL) were used to determine if different perturbation amplitudes led to differences in dual-task interference (pause onset and pause length).

Chapter 3: Results

A total of 12 young healthy adults (aged 18 – 35 years, 22.5 SD 2.43 years) consented to participate. A summary of the participant demographics is presented in ***Table 1***. The majority of participants (8 in total) were male (4 female) with an average self-reported physical training status, and right dominant stepping leg. Due to equipment issues, one participant's data was unable to be used for this thesis. No participants reported having a history of lower limb injuries and/or neurological impairments as per the health status form.

Table 1- Subject demographics, anthropometrics, lower limb dominance, and self-reported physical activity level. Individual and averaged data presented along with standard deviations.

ID#	Age (years)	Sex (M or F)	Height (cm)	Weight (Kg)	Stepping Leg (L or R)	Physical Training Status
01	22	M	179	80.54	R	Average
02	25	M	183	87.33	R	Average
03	19	M	180	79.18	R	High
04	25	M	170	74.66	R	Average
05	23	M	185	90.95	R	Average
06	26	M	168	75.11	R	Average
07	20	M	183	76.92	R	Average
08	21	F	168	67.87	R	Average
09	19	F	165	52.03	R	Average
10	23	M	172	62.44	R	Average
11	22	F	172	61.99	R	Average
12	25	F	167	58.82	R	Average
Mean (n = 12)	**22.5 ± 2.43**	**4F/8M**	**174.33 ± 7.19**	**72.32 ± 11.83**		

3.1 The N1 PEP

N1 responses, evoked by perturbation, were identified from all subjects and task conditions. The grand average topographic representation from all subjects and tasks is presented (***see Figure 8, Bottom)***. The spatiotemporal distribution confirms previous studies revealing the

peak negativity occurring in frontocentral electrode locations at approximately 150 ms after onset of perturbation. As a result, the analysis regarding N1 amplitude was restricted to FCz and Cz electrode locations.

The grand averaged N1 event-related potentials across all subjects and presented for each task condition are presented in ***Figure 8 (Top)***. The data reveals the large amplitude N1 responses, little to no preceding P1 and a more variable P2 after 300 ms. Overall, the amplitude of the N1 responses across task was significantly smaller for the dual-task versus the single-task condition (FCz: $F(1,10) = 31.12$, $p = 0.0002$; Cz: $F(1,10) = 19.16$, $p = 0.0014$). The mean N1 amplitude was 21.4% lower in the dual-task versus the single-task conditions (***Table 2***). There was however no statistically significant difference in the N1 response when comparing between small amplitude and large amplitude perturbations (FCz: $F(1,10) = 0.42$, $p = 0.5327$; Cz: $F(1,10) = 1.52$, $p = 0.2460$). Note, that there was no significant interaction between dual-task condition and perturbation amplitude (FCz: $F(1,10) = 1.53$, $p = 0.2449$; Cz: $F(1,10) = 0.09$, $p = 0.7686$). ***Figure 9*** provides the average N1 amplitude for each task condition and perturbation amplitude on both electrode sites.

Table 2- Individual N1 peak amplitude (μV) and latency (ms) across task conditions at both FCz and Cz electrodes. Individual means with grand averages provided at the bottom along with standard deviations.

	Single-Task (FCz)				Dual-Task (FCz)			
	Small Amplitude		Large Amplitude		Small Amplitude		Large Amplitude	
ID#	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)
01	-31.48	140	-37.40	156	-28.88	144	-28.49	132
02	-27.41	132	-26.70	168	-16.87	132	-18.24	168
03	-33.25	144	-25.00	160	-31.92	132	-22.53	156
04	-30.65	156	-31.12	160	-18.39	156	-18.40	176
05	-9.10	128	-13.12	116	-10.96	128	-6.210	156
06	-54.58	176	-64.69	188	-44.82	200	-42.88	168
07	-19.23	132	-17.87	140	-14.00	144	-10.56	148
08	-36.68	140	-33.96	136	-30.57	136	-24.90	108
09	-48.47	148	-56.62	144	-33.12	148	-57.86	132
10	-18.33	144	-36.42	144	-17.77	120	-21.69	140
11	-32.59	164	-28.02	156	-30.54	132	-23.85	156
Mean (n = 11)	**-31.07 ± 13.01**	**145.82 ± 14.57**	**-33.72 ± 15.31**	**151.64 ± 18.80**	**-25.26 ± 10.31**	**142.91 ± 21.47**	**-25.06 ± 14.43**	**149.09 ± 19.85**

	Single-Task (Cz)				Dual-Task (Cz)			
	Small Amplitude		Large Amplitude		Small Amplitude		Large Amplitude	
ID#	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)	Amplitude (μV)	Latency (ms)
01	-21.87	156	-28.60	160	-19.08	144	-19.51	132
02	-25.16	132	-26.18	148	-12.52	132	-20.07	168
03	-27.74	144	-25.22	156	-30.28	132	-22.46	156
04	-27.79	156	-28.82	160	-15.81	156	-15.55	176
05	-9.93	128	-14.28	132	-13.87	132	-9.68	132
06	-55.32	172	-60.16	188	-46.30	172	-45.78	168
07	-17.54	132	-14.17	140	-11.42	144	-11.37	168
08	-36.09	140	-32.28	136	-30.77	136	-26.92	112
09	-44.98	148	-51.67	148	-31.14	148	-58.00	156
10	-16.74	144	-37.09	140	-15.04	140	-22.62	144
11	-18.51	164	-16.17	148	-16.89	148	-15.23	148
Mean (n = 11)	**-27.42 ± 13.40**	**146.91 ± 13.98**	**-30.42 ± 14.71**	**150.55 ± 15.52**	**-22.10 ± 10.99**	**144 ± 12.13**	**-24.29 ± 14.80**	**150.91 ± 19.60**

With respect to timing (N1 latency) there were no main effects found related to dual-tasking (FCz: $F(1,10) = 0.59$, $p = 0.4613$; Cz: $F(1,10) = 0.13$, $p = 0.7251$) or perturbation amplitude (FCz: $F(1,10) = 1.83$, $p = 0.2063$; Cz: $F(1,10) = 2.14$, $p = 0.1738$). Moreover, no interaction effects were found between conditions (FCz: $F(1,10) = 0.00$, $p = 0.9640$; Cz: $F(1,10) = 0.49$, $p = 0.4990$). N1 peak amplitude latency ranged between 142.9 – 151.6 ms across tasks (***Figure 10***) with comparable within subject task related differences.

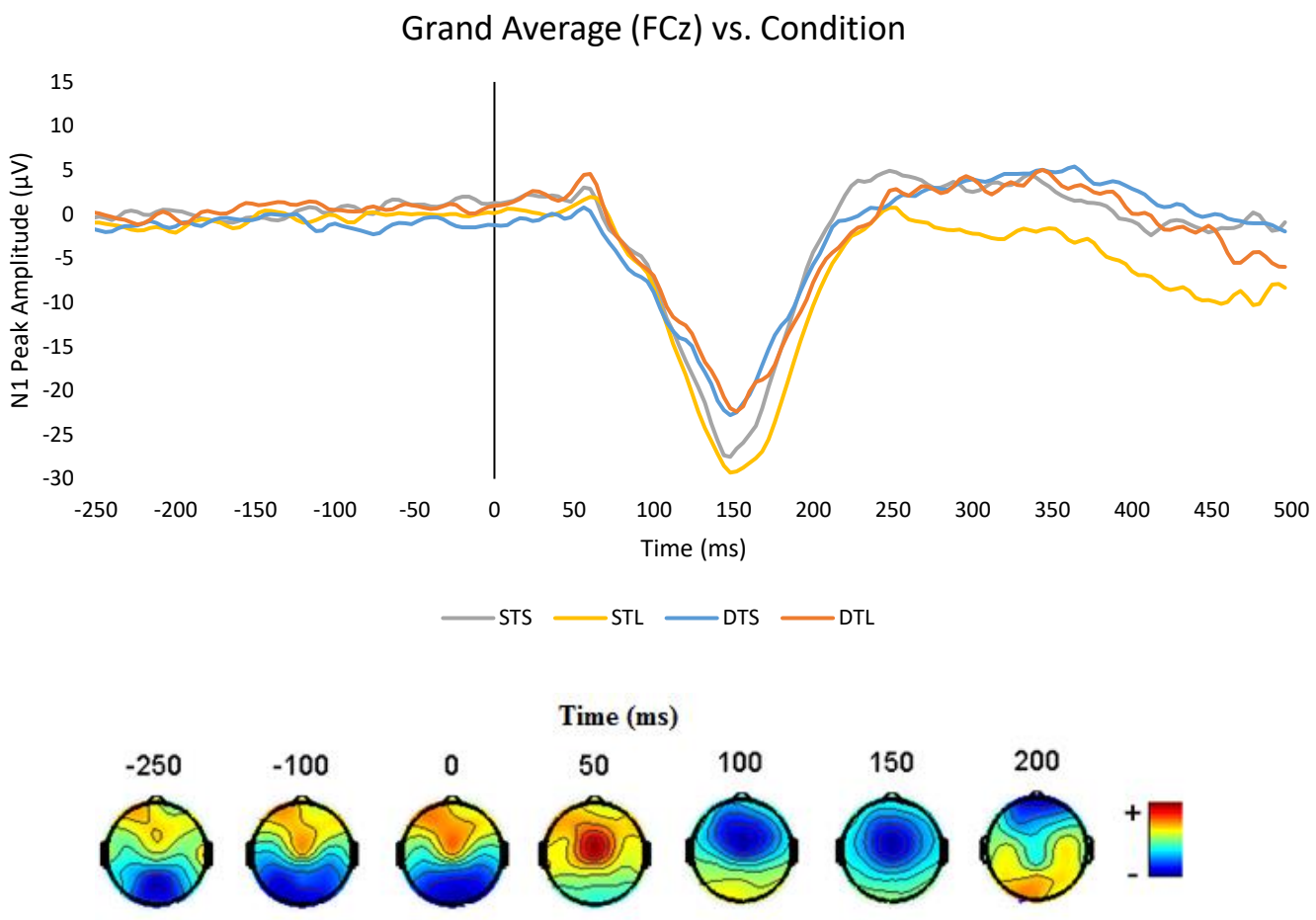

Figure 8- Top: Grand averaged N1 Epochs at the FCz site. Data are average across all subjects for each task condition. ST: Single-task, DT: Dual-task; S: Small (No step) perturbation trial, L: Large (Step) perturbation trial. Bottom: Compiled N1 Peak Amplitude Topography Across Conditions. Blue represents a negative amplitude. A frontocentral topography of the N1 PEP is first observed at 100 ms post-perturbation.

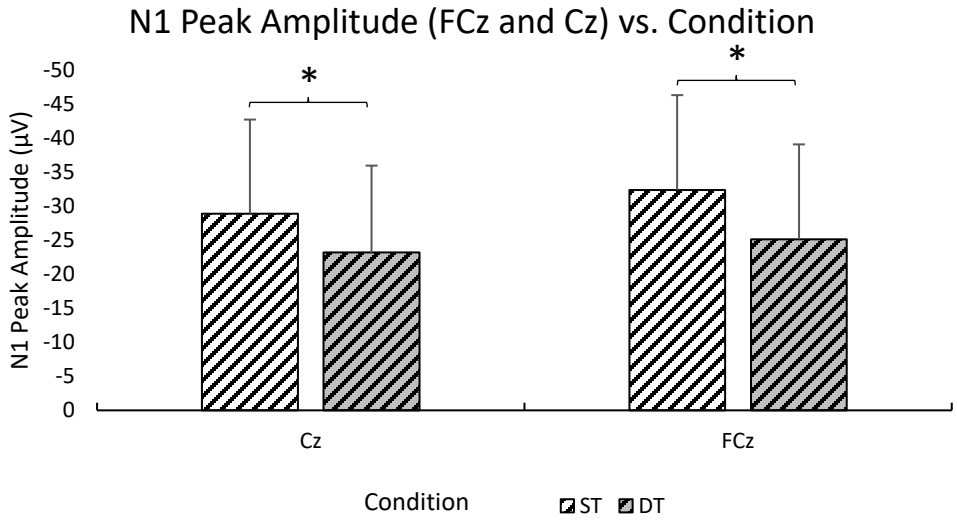

Figure 9- N1 Peak Amplitude at FCz and Cz vs. Condition. Data collapsed across dual-tasking challenge. ST: Single-task, DT: Dual-task. The (*) denotes a statistical significance. A main effect of dual-tasking on the N1 amplitude is observed for the FCz (p = 0.0002) and Cz (p = 0.0014) electrodes.

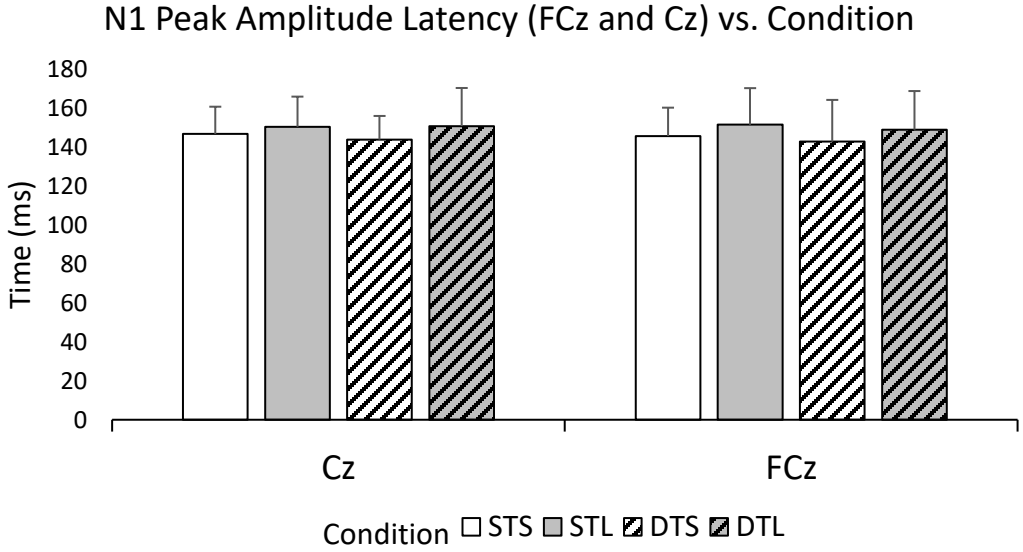

Figure 10- N1 Grand Average Latency at FCz and Cz for each task condition (average across subjects with standard deviation). ST: Single-task, DT: Dual-task; S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial.

Across subjects, the task effects on N1 peak amplitude were comparable; while the absolute amplitude of the N1 in the single-task condition ranged considerably across subjects (***Figure 11A and B***). The main effect related to dual-tasking on N1 amplitude was consistent across subjects. Overall, all the 11 subjects were characterized by lower N1 amplitude in the dual-task versus single-task condition (FCz: 23.17% attenuation; Cz: 19.72% attenuation) when collapsing across perturbation amplitudes. However, only a small number of participants (FCz: 2, Cz: 4) were observed to have a greater attenuation in the small perturbation condition comparisons (STS vs. DTS. FCz: 15.72%; Cz: 15.71%) than in the large perturbation condition comparisons (STL vs. DTL. FCz: 28.48%; Cz: 21.56%). Meaning a larger attenuation seemed to derive from large perturbation conditions.

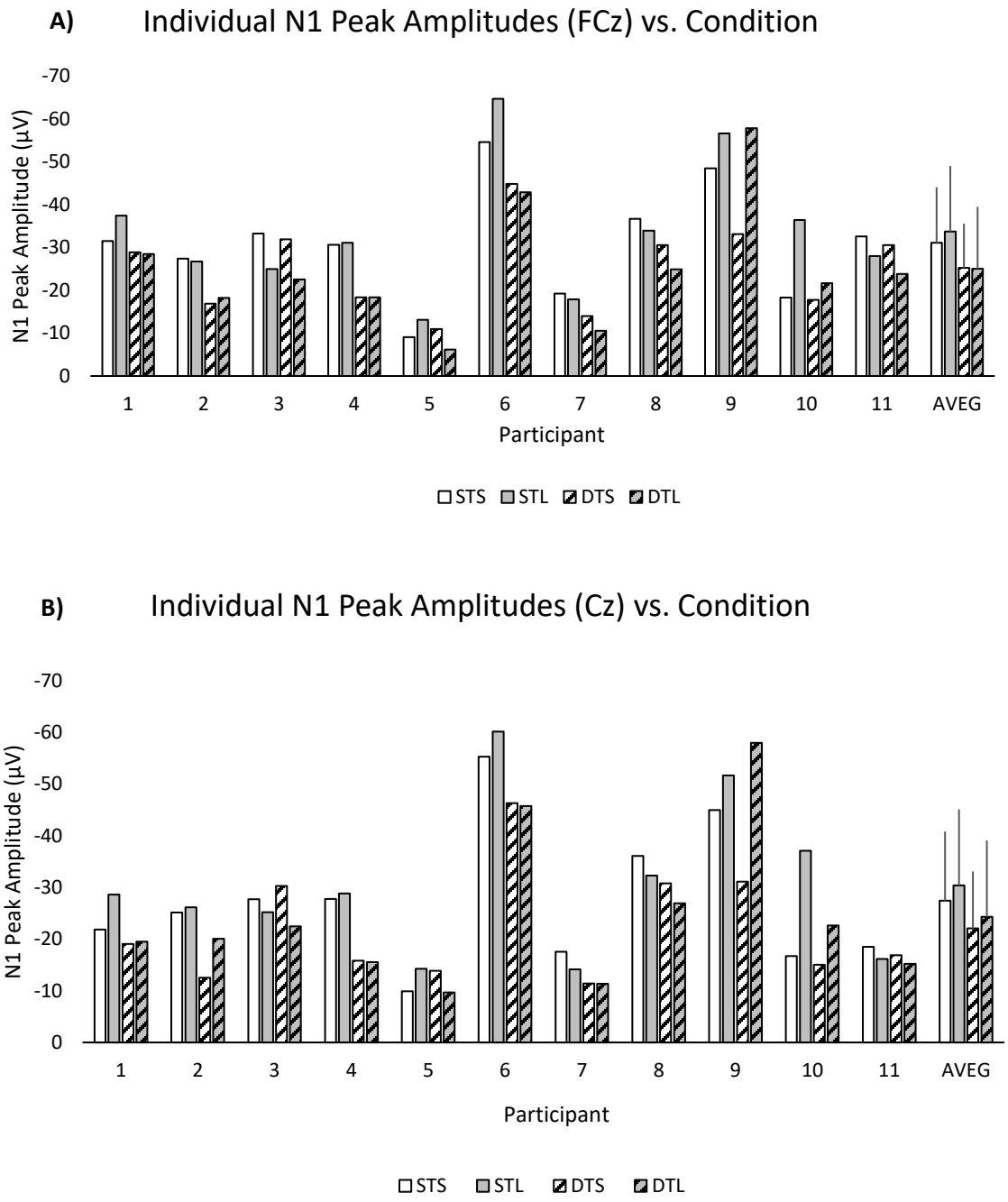

Figure 11- N1 Peak Amplitude Variability (average for subjects with standard deviations) at the A) FCz and B) Cz vs. Condition. ST: Single-task, DT: Dual-task; S: Small (No step) perturbation trial, L: Large (Step) perturbation trial. Individual dots represent individual averaged N1 peak amplitudes for said conditions, while grey bars represent the average.

3.2 Pre-perturbation state

The average pre-release body weight percentage (BW%, ***Figure 12***) significantly differed due to perturbation amplitudes ($F(1,10) = 12.53$, $p = 0.0054$), but not dual-task challenge ($F(1,10) = 4.27$, $p = 0.0657$). No interaction effects was observed ($F(1,10) = 0.01$, $p = 0.9164$). Although BW% was controlled before a release, it appears that across tasks during small perturbations, participants placed 4.52% of their BW onto the support cable, while during large perturbation trials, 5.60% of BW was placed onto the same support cable. Highest average BW% derived from STL trials (5.72%). Lowest BW% derived from DTS trials (4.41%).

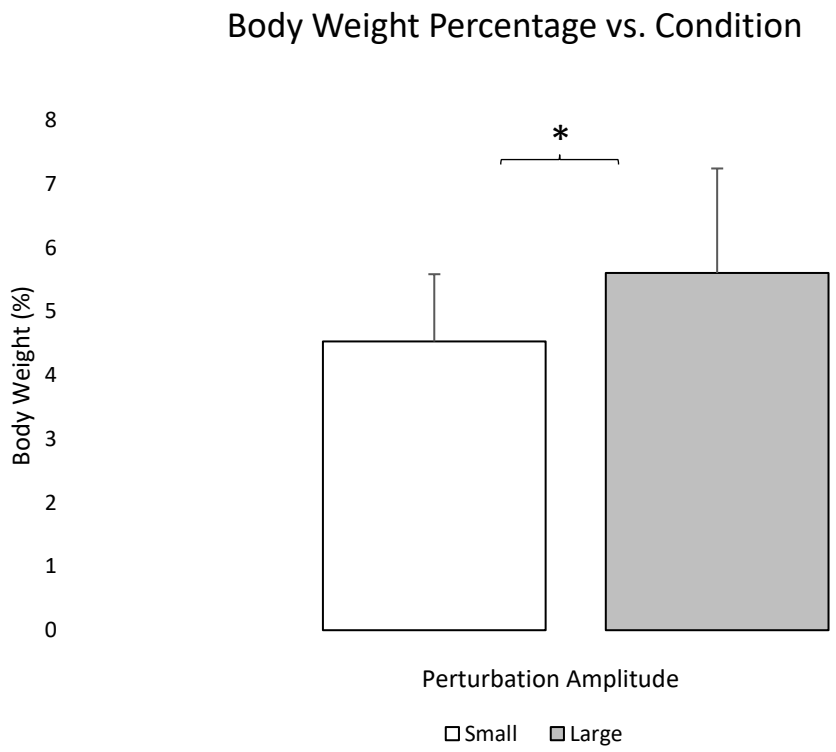

Figure 12- Body Weight Percentage vs. Conditions. Data collapsed across perturbation amplitude. S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. A statistically significant (*) main effect of perturbation amplitude is observed (p = 0. 0054).

3.3 EMG Onset and Amplitude

As expected, EMG onsets were found to be different between recorded sites due to the leaning position in which the participant was asked to maintain prior to the supporting cable release (***Figure 13***). Meaning EMG onsets at the MG (Range: 76.1 – 87.7 ms) occurred earlier than in the TA muscle (Range: 107.5 – 118.0 ms). EMG onset significantly differed due to the site, tibialis anterior vs. medial gastrocnemius muscle ($F(1,10) = 20.12$, $p = 0.0012$), but not between dual-task challenge (MG: $F(1,10) = 2.54$, $p = 0.1423$; TA: $F(1,10) = 0.11$, $p = 0.7425$) or perturbation amplitudes (MG: $F(1,10) = 4.05$, $p = 0.0718$; TA: $F(1,10) = 0.31$, $p = 0.5873$). No interaction effects were observed (MG: $F(1,10) = 0.57$, $p = 0.4690$; TA: $F(1,10) = 3.50$, $p = 0.0910$).

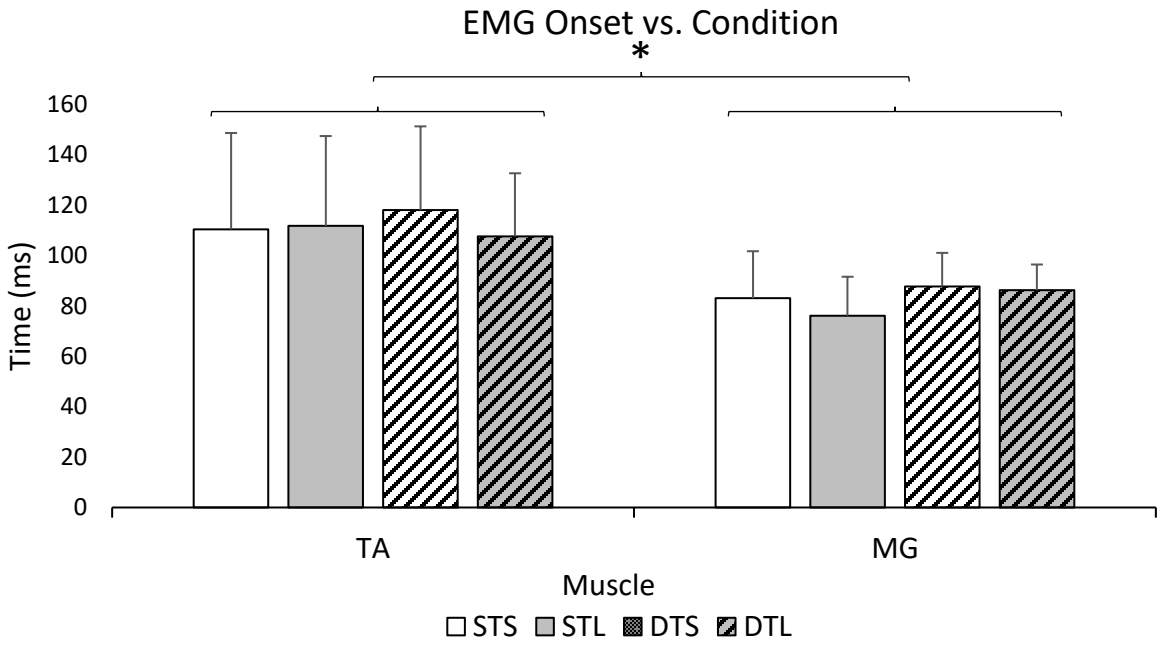

Figure 13- EMG Onset for TA and MG Muscles vs. Conditions. ST: Single-task, DT: Dual-task; S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. A statistically significant (*) main effect of muscle site is observed (p = 0.0012).

At the TA muscle the two perturbation amplitudes, small (***Figure 14A***; Onset – 50 ms: 0.53 mVs; Onset – 100 ms: 1.19 mVms; Onset – 200 ms: 2.15 mVs) and large (Onset – 50 ms: 0.75 mVs; Onset – 100 ms: 1.82 mVs; Onset – 200 ms: 3.90 mVs), appear to show distinct EMG amplitude values. TA EMG amplitude significant differences were found related to perturbation amplitude ($F(1,10) = 9.86$, $p = 0.0105$) and time frame ($F(2,10) = 28.60$, $p = X < 0.001$), but not due to dual-task challenge ($F(1,10) = 0.77$, $p = 0.4011$). An interactions effect exists between perturbation amplitude and time frame ($F(2,10) = 16.69$, $p = X < 0.001$). A post-hoc analysis determined that a statistical significance exists between Onset – 50 ms and Onset – 100 ms ($F(1,10) = 5.37$, $p = 0.0222$), and between Onset – 50 ms and Onset – 200 ms ($F(1,10) = 32.56$, $p = X < 0.0001$), but not between Onset – 100 ms and Onset – 200 ms ($F(1,10) = 0.61$, $p = 0.4351$). Regarding perturbation amplitudes, as expected in the MG site, large perturbation amplitudes trials (Onset – 50 ms: 4.30 mVs; Onset – 100 ms: 10.82 mVs; Onset – 200 ms: 21.41 mVs) are observed to be associated with higher EMG amplitude values when compared to small perturbation amplitudes (***Figure 14B***; Onset – 50 ms: 3.47 mVs; Onset – 100 ms: 7.92 mVs; Onset – 200 ms: 12.69 mVs). MG EMG amplitude significant differences were found related to perturbation amplitude ($F(1,10) = 9.15$, $p = 0.0128$) and time frame ($F(2,10) = 18.42$, $p = X < 0.001$), but not due to dual-task challenge ($F(1,10) = 2.57$, $p = 0.1402$). An interactions effect exists between perturbation amplitude and time frame ($F(2,10) = 9.78$, $p = 0.0011$). A post-hoc analysis determined that a statistical significance exists between Onset – 50 ms and Onset – 100 ms ($F(1,10) = 7.73$, $p = 0.0063$), and between Onset – 50 ms and Onset – 200 ms ($F(1,10) = 33.26$, $p = X < 0.0001$), but not between Onset – 100 ms and Onset – 200 ms ($F(1,10) = 0.27$, $p = 0.6034$).

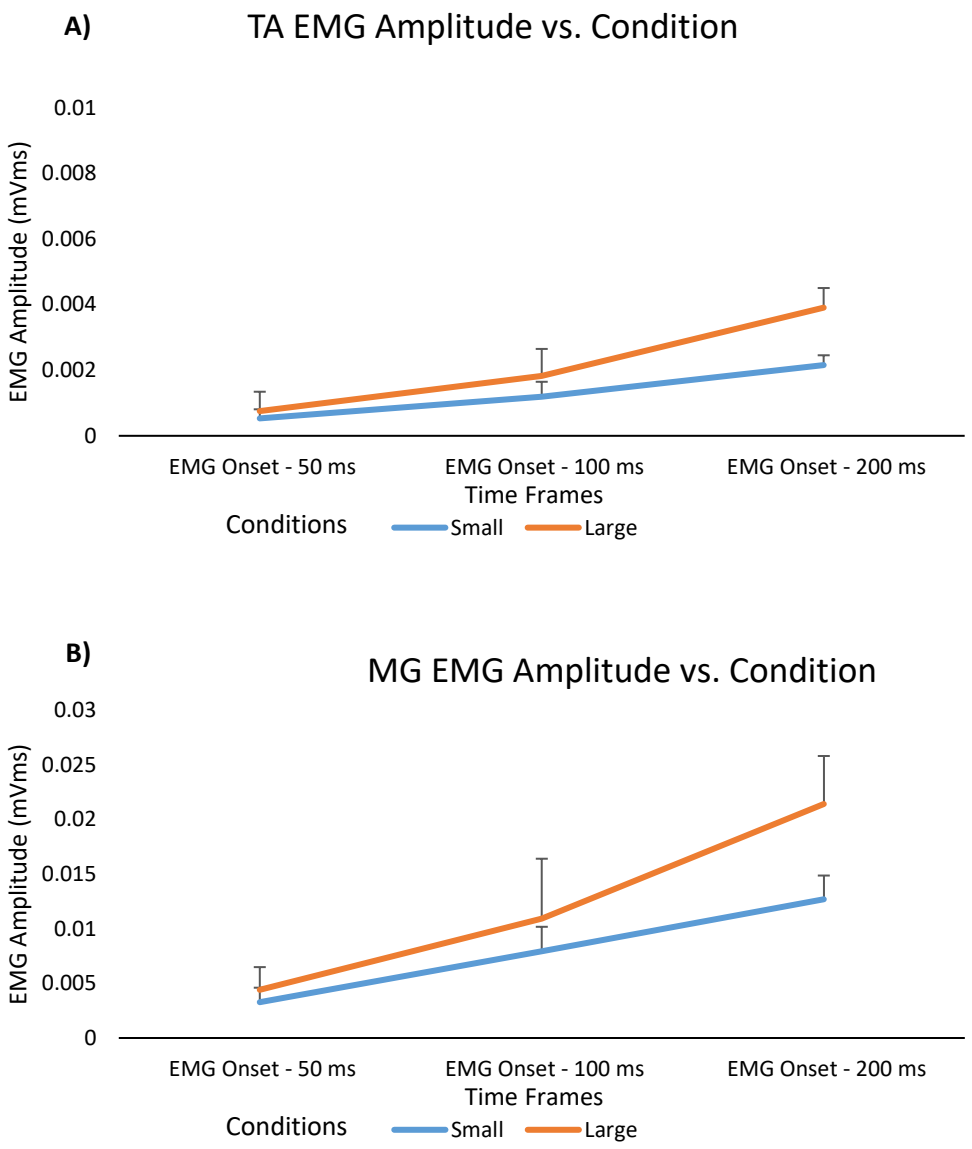

Figure 14- A) TA EMG B) MG EMG Amplitude at Different Time Frames vs. Condition. Data collapsed across perturbation amplitude. S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial.

3.4 Head and Sternum Accelerometry

To elicit different amplitude perturbations, a hydraulic device was used to slow down the participant's anterior acceleration (AP plane) after the release of the main cable holding them. The accelerometry analysis indicated a mixed success for said device (***Figure 15 -Figure* 17**). No changes related to dual-task challenge ($F(1,10) = 0.03$, $p = 0.8591$) or perturbation amplitude ($F(1,10) = 3.87$, $p = 0.0776$) were observed in the first 100ms of the averaged AP acceleration at the sternum (***Figure 15A***). However, head accelerometry (***Figure 16A***) shows a significant difference due to perturbation amplitude in the same early time frame ($F(1,10) = 9.38$, $p = 0.0120$), but not dual-task challenge ($F(1,10) = 0.07$, $p = 0.7977$). Significant differences are further observed beyond the first 100 ms post-perturbation. At the sternum, a significant difference was found in the peak AP acceleration (***Figure 15B***; $F(1,10) = 34.56$, $p = 0.0002$) and peak latency (***Figure 17A***; $F(1,10) = 21.74$, $p = 0.0009$) due to perturbation amplitude but not due to dual-tasking (peak acceleration: $F(1,10) = 0.92$, $p = 0.3599$; peak latency: $F(1,10) = 3.14$, $p = 0.1068$). In contrast at the head, a significant difference (***Figure 16B***) in peak AP acceleration due to perturbation amplitude ($F(1,10) = 35.93$, $p = 0.0001$) and dual-task challenge ($F(1,10) = 7.79$, $p = 0.0191$). Finally, significant differences in peak AP acceleration latency resulted due to perturbation amplitude ($F(1,10) = 7.65$, $p = 0.0199$) but not dual-tasking challenge ($F(1,10) = 0.60$, $p = 0.4563$) was found at the head (***Figure 17A***). No interaction effects were found in any of the acceleration statistical comparisons (Head|100 ms acceleration: $F(1,10) = 2.08$, $p = 0.1799$; peak acceleration: $F(1,10) = 0.01$, $p = 0.9396$; peak acceleration time: $F(1,10) = 0.21$, $p = 0.6544$. Sternum|100 ms acceleration: $F(1,10) = 0.01$, $p = 0.9359$; peak acceleration: $F(1,10) = 1.52$, $p = 0.2459$; peak acceleration time: $F(1,10) = 0.02$, $p = 0.8852$).

On average the acceleration recorded within the first 100 ms ranged from 0.06 – 0.09 m/s^2 for the sternum and 0.10 – 0.21 m/s^2 for the head (***Figure 15AFigure 16A***). The highest accelerations were observed from the head accelerometers during the presentation of large perturbation trials. The peak A-P acceleration was found to take place between 225.5 – 238.0 ms after the cable release for the sternum and after 234.5 – 243.8 ms for the head accelerometers (***Figure 17A and B***). Peak acceleration values were always found to be higher than those seen in the averaged 100 ms subset (Sternum: 0.39 – 0.81 m/s^2; Head: 0.64 – 1.18 m/s^2). It appears that different perturbation amplitude trials show very little difference at the 100 ms mark at both sites when averaged across dual-task challenge (Head| Small: 0.11 m/s^2, Large: 0.20 m/s^2; Sternum| Small: 0.07 m/s^2, Large: 0.09 m/s^2). Differences in acceleration between perturbation amplitude trials appear to become more apparent at peak, where head accelerometry shows a 0.47 m/s^2 difference (Small: 0.67 m/s^2, Large: 1.14 m/s^2); sternum accelerometry shows a similar 0.39 m/s^2 difference (Small: 0.41 m/s^2, Large: 0.81 m/s^2). Large perturbation trials tend to display a longer time to peak acceleration (Head| Small: 236.1 ms, Large: 243.5 ms; Sternum| Small: 227.5 ms, Large: 239.6 ms). When comparing between dual-task conditions, regardless of perturbation amplitude, accelerometry parameters look to be much more similar: averaged 100 ms acceleration (Head| ST: 0.15 m/s^2, DT: 0.15 m/s^2; Sternum| ST: 0.085 m/s^2, DT: 0.08 m/s^2), peak acceleration (Head| ST: 0.94 m/s^2, DT: 0.87 m/s^2; Sternum| ST: 0.62 m/s^2, DT: 0.60 m/s^2), peak acceleration time (Head| ST: 240.4 ms, DT: 239.2 ms; Sternum| ST: 235.3 ms, DT: 231.8 ms).

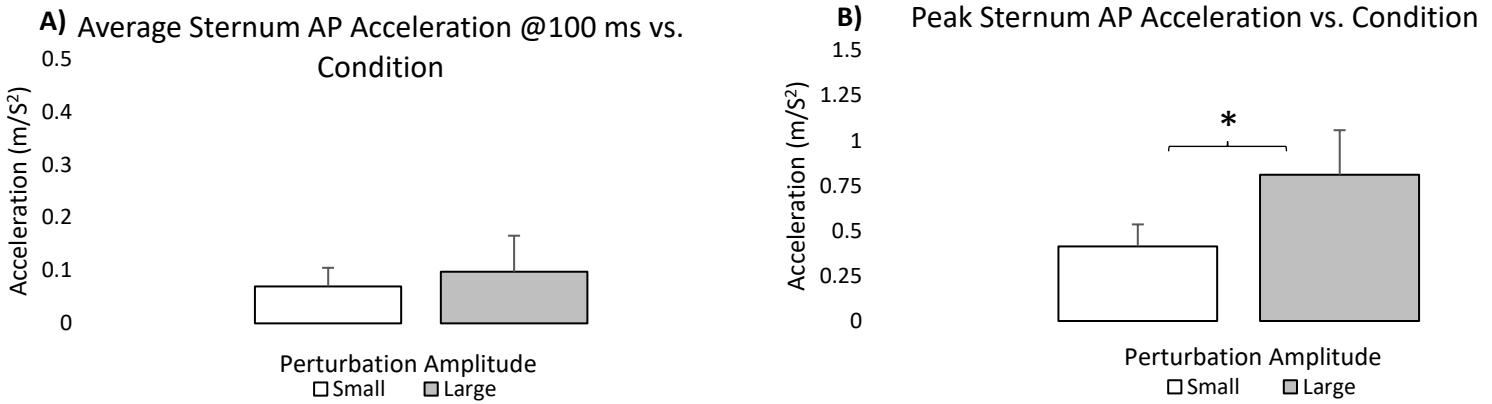

Figure 15- A) Average Sternum AP Acceleration @100 ms vs. Condition. B) Peak Sternum AP Acceleration vs. Condition. C) Peak Sternum AP Acceleration Time vs. Condition. Data collapsed across perturbation amplitude. S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. The (*) denotes a statistical significance. B) A main effect of perturbation amplitude was found in the peak sternum AP acceleration (p = 0.0002).

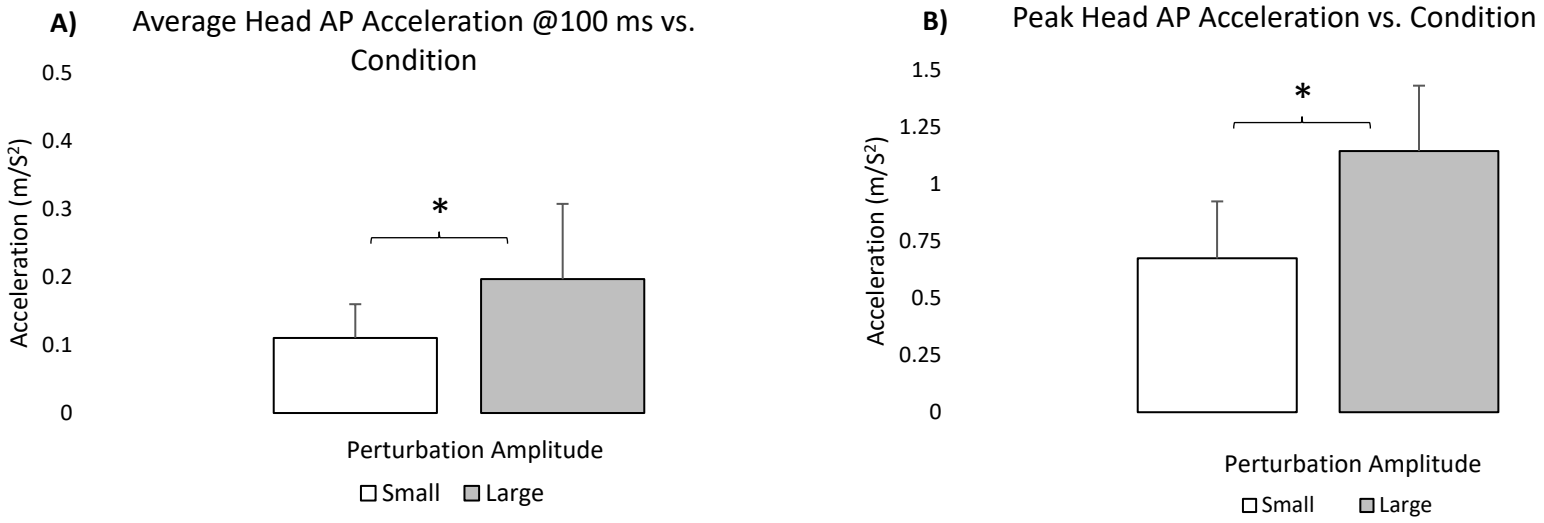

Figure 16- A) Average Head AP Acceleration @100 ms vs. Condition. B) Peak Head AP Acceleration vs. Condition. C) Peak Head AP Acceleration Time vs. Condition. Data collapsed across perturbation amplitude. S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. The (*) denotes a statistical significance. A) A main effect of perturbation amplitude for the average head AP acceleration @100 ms was found (p = 0.0120). B) Furthermore, a main effect of both perturbation amplitude (p = 0.0001) and dual-task challenge (p = 0.0191) were found for the head peak AP acceleration.

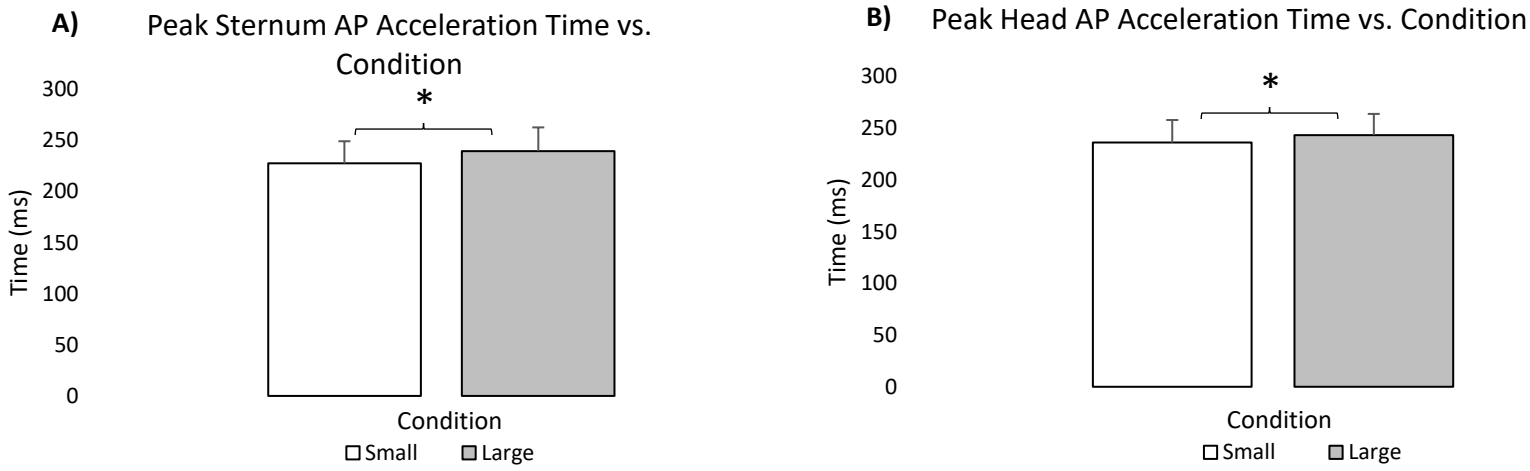

Figure 17- A) Peak Sternum B) Head AP Acceleration Time vs. Condition. Data collapsed across perturbation amplitude. S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. The (*) denotes a statistical significance. A main effect of perturbation amplitude was found in both the peak latency A) sternum (p = 0.0009) and B) head (p = 0.0199) sites.

3.5 CoP Step Onset and Foot-off

During trials in which a large perturbation amplitude was delivered, the average step onset after a perturbation took place at 195.9 ± 75.8 ms for the ST condition and 221.0 ± 58.1 ms for the DT condition (***Figure 18***). As expected, foot-off values took place around the peak acceleration time (STL: 407.9 ± 118.2 ms; DTL: 447.79 ± 85.7 ms). Statistically significant chances were recorded for step onset due to dual-task challenge ($t(10) = -2.48$, $p = 0.0327$). No changes in foot-off related to dual-task challenge ($t(10) = -1.55$, $p = 0.1511$).

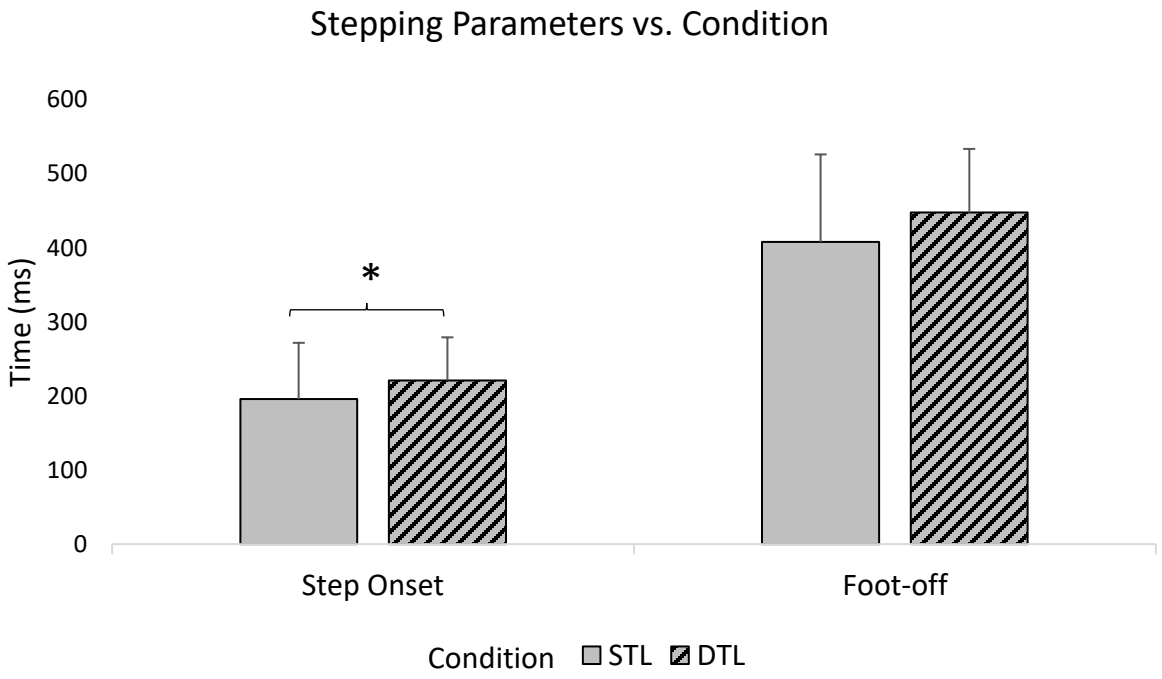

Figure 18- Stepping Onset and Foot-off vs. Dual-task Challenge. ST: Single-task, DT: Dual-task; L: Large amplitude (Step) perturbation trial. The (*) denotes a statistical significance where p = 0.0327.

3.6 Dual-tasking Pre-perturbation Performance

Overall there was an increase in pre-release dual-task performance as reflected by the significant correlation between pre-release dual-task performance and the trial number count ($r(10) = -0.06555$; $p = 0.0250$). Moreover, a significant difference in pre-release dual-task performance was recorded between the initial seated1, block and practice, block1, block2, but not the second seated2 block condition ($t(10) = 2.36$, $p < 0.05$, ***Figure 19***).

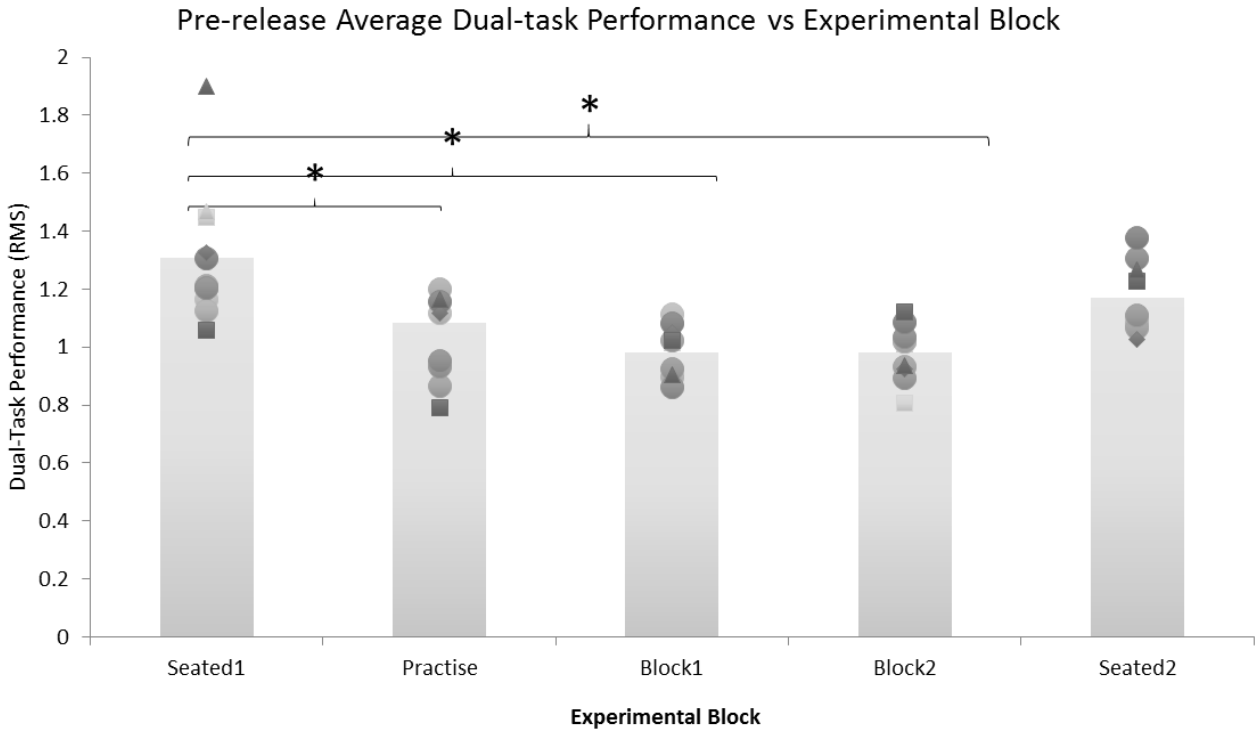

Figure 19- Pre-release Average Dual-task Performance vs. Experimental Block. Each dot represents averaged individual data. Bar graphs represent grand averaged data. The (*) denotes a statistical significance where p < 0.005. Pre-perturbation dual-task performance was not statistically significant between seated trials, suggesting standing may have led to increased performance.

3.7 Dual-tasking Interference

No statistically significant changes were observed in dual-task interference start ($t(10) = -0.92$, $p = 0.3796$), or length ($t(10) = -0.78$, $p = 0.4519$) related to perturbation amplitude (***Figure 20***). The average start of the dual-task interference took place 1225.2 ± 319.6 ms post-perturbation for the DTS trials and 1336.1 ± 306.7 ms for the DTL trials. The length of these periods of inattention was generally shorter for the DTS condition (447.3 ± 67.0 ms) than in the DTL condition (472.8 ± 88.7 ms).

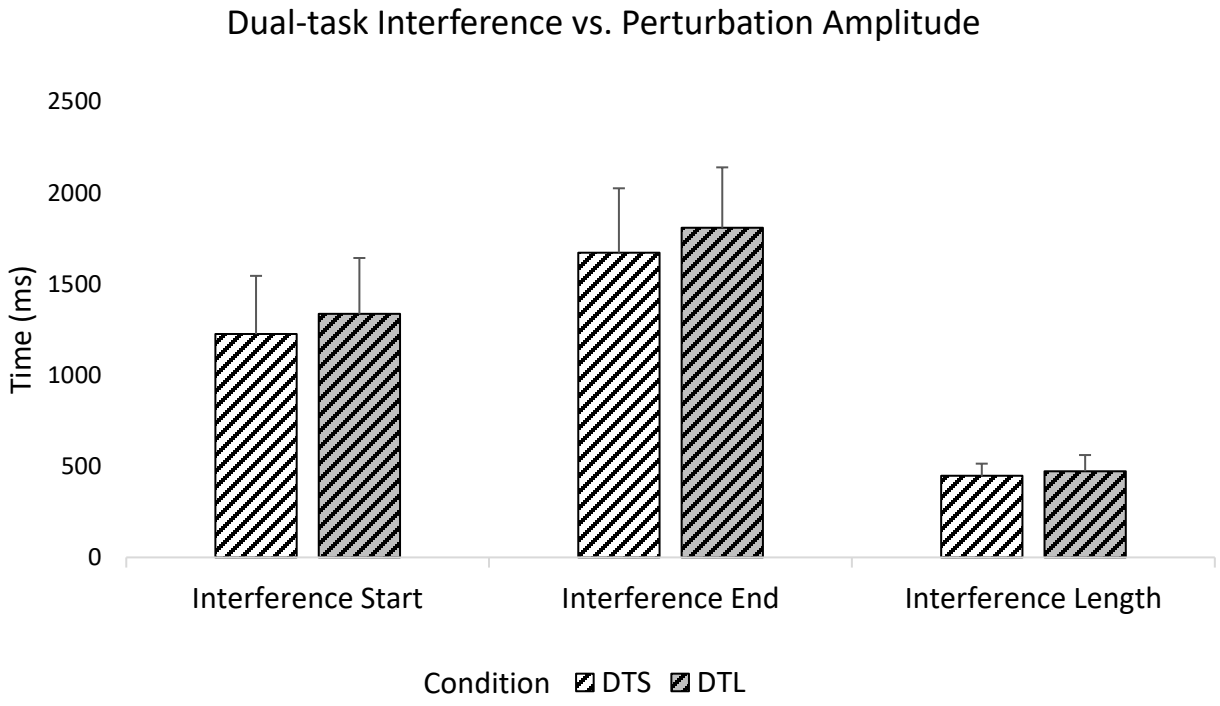

Figure 20- Dual-task Interference vs. Perturbation Amplitude. DT: Dual-task; S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial. No statistical changes observed in dual-task interference were observed when contrasting small vs. large perturbation amplitudes (DT start: p = 0.3796; DT length: p = 0.4519).

Chapter 4: Discussion

Cortical responses associated with instability are characterized by a complex waveform that is dominated by the N1 perturbation-evoked potential. The present thesis sought to expand our understanding regarding the relationship between attention, the perturbation-evoked N1 potential, and the associated balance reaction post-perturbation. This was conducted, in part, to provide a better understanding of the role of the cerebral cortex in reactive balance control. Three main hypotheses regarding the N1 and its association with dual-tasking were explored (***Figure 21***). Firstly, it was hypothesized that the introduction of a large perturbation amplitude would lead to an observed large N1 peak amplitude in contrast with cortical responses from small perturbation trials. As larger perturbations were expected to be of higher priority to the CNS, demanding greater attentional resources, it was secondly hypothesized that upon dual-tasking no attenuation of the N1 peak amplitude would be observed in large perturbation dual-tasking trials. Due to the high prioritization of large perturbation trials, posing a greater threat to balance control than small perturbation trials, it was thirdly hypothesized that behavioral data would depict earlier stepping. Whenever dual-tasking prior to a large perturbation, then dual-task tracking interference would occur earlier and last longer as a step is needed is regain balance.

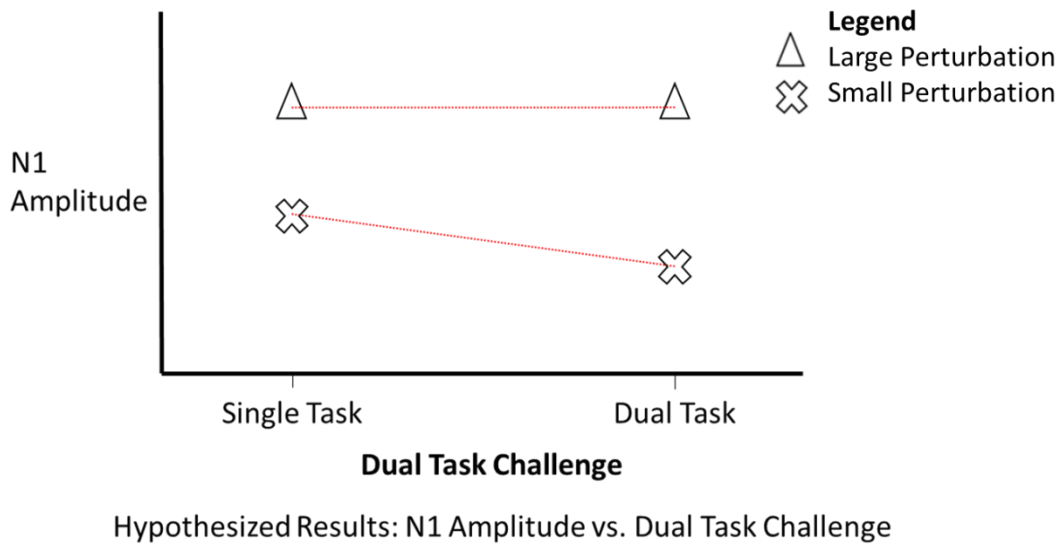

Figure 21- Hypothesized N1 Peak Amplitude at Cz versus Task Condition. S: Small amplitude (no step) perturbation trial, L: large amplitude (step) perturbation trial.

Figure 22 allows for the visualization of the main cortical findings in the present study. It is first important to state that different perturbation amplitudes elicited very different balance reactions. In one, large perturbation trials, a step was taken by participants while in small perturbation trials only a feet-in-place reaction was elicited. Although different behaviors were observed, no scaling effect is present between the different perturbation amplitudes and the N1 cortical potential. Depicted by the downward trend of points in ***Figure 22***, an attenuation of the N1 peak amplitude was observed whenever participants completed a cognitive task right before receiving a postural perturbation. Furthermore, and not in line with hypothesized results, the introduction of large perturbation amplitudes while dual-tasking led to attenuations of the N1 to the same degree as small perturbation trial attenuations. Lastly, some early and late evidence of behavioral changes related to perturbation amplitude, but not dual-tasking, exists pointing towards the idea that perturbation amplitudes were not different enough.

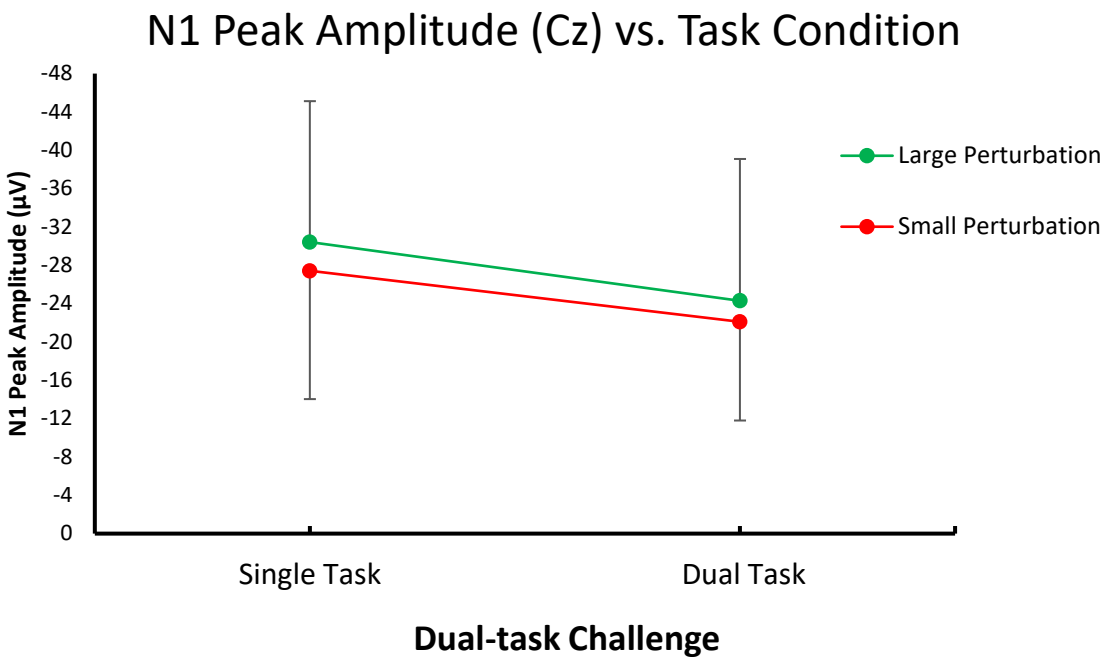

Figure 22- Actual N1 Peak Amplitude at Cz versus Task Condition. DT: Dual-task; S: Small amplitude (No step) perturbation trial, L: Large amplitude (Step) perturbation trial.

4.1 Scaling of N1 cortical response to perturbation amplitude

The current results did not support the hypothesis that increasing amplitude of perturbation would lead to an increased N1 peak amplitude (***Figure 22***). In literature, various factors are known to lead to changes in the peak N1 amplitude, such as: aging, stance width, perturbation amplitude, perturbation predictability, and concurrent sensory information (Dietz et al., 1985b; Dimitrov et al., 1996; Duckrow et al., 1999; Staines et al., 2001; Quant et al., 2004b; Adkin et al., 2006; Mochizuki et al., 2008, 2010; Little and Woollacott, 2014). There has only be one study that specifically explored the N1 amplitude for different perturbation amplitudes (Staines et al., 2001). This study relied on chair (seated) translations and observed cortical response changes while applying different perturbation amplitudes. Staines et al. 2001 describes the faster the backwards platform translations (0.5, 1.0, and 1.5 m/s^2), the larger the N1 peak

amplitudes. In more recent literature, it appears that the N1 is present in a significantly smaller amplitude even while participants are quietly standing (Varghese et al., 2015). Although it is difficult to compare across studies due to between-subject variability, experimental protocol, etc; both studies suggest that although the N1 is modulated by numerous factors, it appears that, when all other experimental aspects are controlled, the N1 peak is scaled to the amplitude of the perturbation applied.

The absence of scaling of N1 response amplitude was unexpected in the current study. The N1 amplitude observed in other studies that have used a lean-and-release protocol has been about 30 μV for control trials at the FCz and Cz electrodes (Mochizuki et al., 2008, 2009b, 2010; Marlin et al., 2014). The N1 amplitude was comparable in this study with average amplitudes between 31 and 34 μV for the small and large amplitude perturbations control conditions. While the responses were comparable to other lean and release studies the N1amplitude for the small perturbation was much larger than that recorded by Staines et al. (2001) where the small amplitude responses were approximately 10 μV. The interesting aspect remains that in current literature it is not the studies in which a step is elicited, due to larger perturbation being applied, were the highest N1 peak amplitudes are generally observed. Both LaR studies in which a feet-in-place reaction was attained show an average N1 peak amplitude of 31 μV (Mochizuki et al., 2008, 2009b; Marlin et al., 2014), while the present study and that by Mochizuki et al. (2010), in which a stepping reaction was elicited, have an average N1 peak amplitude of 31 μV. It is likely that the similarity between N1 amplitudes can be mainly explained by between-person variability; however, the same similarities also highlighting that environment and/or experimental factors such as temporal predictability, amplitude predictability, direction predictability, and experimental instructions that can have a large effect on the shaping of PEPs.

While the absolute amplitude of the evoked N1 response were comparable with other lean and release studies the N1 peak amplitude did not scale to the amplitude of the perturbation applied. The challenge in a lean and release paradigm is to achieve differences in perturbation amplitude without prior knowledge since prediction of stimulus amplitude can alter the N1 response (Mochizuki et al., 2010). As the result of pilot studies a new novel approach was adopted to achieve variations in the initial acceleration when subjects adopted the same starting lean angle. The initial lean angles were different between the task conditions, although all remained approximately 5% BW. Mochizuki et al. (2010) recorded around 10% BW to elicit stepping reactions, while Marlin et al. (2014) observed values around ~5% BW, eliciting feet-in-place reactions. The hydraulic system used in this study was intended to limit the initial acceleration and perturbation amplitude in the small amplitude condition. In the large amplitude condition (no hydraulic attached) the current lean angle results in stepping responses consistently. While there were clear differences in behavior, stepping in response to the large amplitude versus small amplitude perturbations, the important characteristic of the perturbation that drives the initial reactions and the N1 response is related to the initial rate of acceleration (Bolton, 2015b). Thus, a key measure of the amplitude of perturbation was considered to be the initial body acceleration. At the sternum, there were no statistically significant differences in the acceleration at 100 ms post-perturbation comparing between the small and large amplitude perturbations. However, this does not mean different perturbation amplitudes were not achieved as head accelerometry did show statistical difference related to perturbation amplitude at 100 ms post-perturbation, and sternum acceleration results were close to statistical significance (p = 0.0776). Behaviorally, EMG onset was found to be similar across tasks regardless of which muscle was analysed (TA vs. MG). The earliest reactions to a perturbation may not have been

affected by perturbation amplitude changes however throughout all time frames a statistical difference in EMG amplitude related to the perturbation amplitude used was found. With time, differences in EMG amplitude responses related to perturbation amplitude become more pronounced in early (50 ms), mid (100 ms), and late (200 ms) time frames. Peak accelerations for the head accelerometers (occurring around 230 ms after perturbation) were significantly greater for large compared to small perturbations. After this time it is difficult to contrast balance reactions between perturbation amplitude trials as step onsets and foot off were only seen in the large amplitude perturbation trials (single-task and dual-task). Overall, the use of a hydraulic pipe to limit the amount of forward movement after the release of the supporting cable appears to be effective in impacting the earliest characteristics of the induced instability.

The methodological limitations in the present study are important as they impact the potential utility of the lean and release approach to provide unpredictable stimulus to explore balance reactions. Although inexpensive compared to platform translation systems, lean and release protocols are limited by achieving perturbation amplitude largely based on initial lean angle; meaning, subjects would be typically aware of the perturbation amplitude based on the initial lean angle. Previous literature has demonstrated that perturbation predictability leads to a significant attenuation of the N1 peak amplitude, alongside the appearance of a pre-perturbation cortical response (Dietz et al., 1985b; Adkin et al., 2006; Mochizuki et al., 2008, 2010). In a study by Mochizuki et al. (2010), knowing the amplitude and direction of a perturbation did not lead to differences in the N1 peak amplitude, the only significant changes were observed in pre-perturbation cortical events. Pre and post cortical windows are thought to be independent from each other (Mochizuki et al., 2010). However, Mochizuki et al. (2010) introduced perturbations at a known time to the participant, making the results hard to interpret (Marlin et al., 2014). In

the case of the Mochizuki et al. (2010) study, it can be argued that as the lean angle was the same between conditions (small vs. large), the initial rate of acceleration may have also been the same. The methodology in this study sought to control for as many aspects related to the perturbation as possible with the intention that some of these (e.g. predictability, instructions, direction) may lead to changes in cortical responses that is outside of the question of interest. Perturbation amplitude predictability was controlled through the leaning angle by having all participants lean so that a feet-in-place reaction was elicited through the inclusion of a hydraulic pipe. Other studies that have probed the use of different perturbation amplitudes indicate that a 300 ms square-wave pulse acceleration was utilized to generate perturbations (Maki et al., 1996; Staines et al., 2001; Norrie et al., 2002; Quant et al., 2004b). In the platform translations by Staines et al. (2001) the acceleration was specifically controlled and the initial accelerations were significantly different early after the onset of perturbation (0.5 m/s^2 (small), 1.0 m/s^2 (medium), 1.5 m/s^2 (large)). Changes in perturbation amplitude, as depicted by the platform accelerations, resulted in scaled changes to the N1 peak amplitude. However, such cortical changes published by Staines et al. (2001) show that the perturbation amplitude to N1 amplitude scaling is not linear. To explain, it appears small perturbations (0.5 m/ s^2) resulted in smaller (10 µV) N1 amplitudes when compared to large perturbation trials (1.5 m/ s^2, 20 µV). Yet an obvious difference is difficult to be observed when comparing medium to large perturbation trials (1.0 m/ s^2 and 1.5 m/s^2, respectively). Meaning, that although a large difference in acceleration existed between the three perturbation amplitudes, the differences in the resultant N1 peak amplitudes are only seen when comparing extremes (small vs. large). Furthermore, the scaling was only observed in two of three participants tested with the third participant having "saturated" N1 peak amplitudes. It is therefore possible that the lack of perturbation amplitude to N1 scaling in the present study may

be have derived from initial accelerations that although different are not reflected by the N1 cortical response. It is likely that small perturbations in this study were not small enough, or large perturbations were not large enough, making the contrast between conditions seem non-differentiated. Due to the steps elicited in the large perturbation trials, it is likely that the small perturbation trials were indeed not small enough. The threat imposed by the stimulus delivered in the present study is also a factor which may need to be further studied. In the Staines et al. (2001) study, the stimulus can be argued to never have posed a postural threat. Participants were secured into a chair that did not tilt or become unstable upon translation. In contrast, a lack of a step (or a misstep) in the present study may have led to a near-fall experience, a more threatening event to the CNS. Previous studies on arousal show that increased postural threat is linked to an increased cortical response independent of the amplitude of the perturbation applied (Sibley et al., 2010). It is possible that the amplitude of the N1 was comparable for different amplitudes of perturbation as they may have been perceived as a similar threat. In this study, since participants started each trial in a similar lean angle, the resultant N1 amplitudes may reflect equal pre-perturbation perceived threat. A non-linear afferent scaling on part of the N1 may provide insight as to what the role of the N1 is to reactive balance control, while methodological limitations highlight the practical challenges for the use of a LaR system for the study of cortical involvement in reactive balance control.

The current study reinforced the importance of the techniques used to introduce postural instability. The initial responses, including the N1, are importantly determined by the initial acceleration imposed on the individual by the instability. Not all perturbation techniques are able to achieve a desired level of control. The advantages of precise platform accelerations are an accurate regulation of the imposed instability (Diener et al., 1984). However, it should be noted

that many studies using such devices do not concurrently collect the acceleration of the body to confirm the imposed instability. The benefit of ease of use and cost effectiveness has increased use of lean and releases including in clinical settings (Mansfield et al., 2011b, 2013, Inness et al., 2014, 2015). However, a lean and release protocol is potentially confounded by the predictability of perturbation amplitude and direction. Knowing the N1 is modulated by numerous environmental factor surrounding experimental protocols, means it may be likely that these two characteristics of an applied perturbation can affect the N1 independently. As shown by Varghese et al. (2015), there is a possibility researchers could investigate placing participants in challenging stances to see if being more challenged would lead to an N1 scaling similar to perturbation amplitude scaling. Future studies in this field should feel compelled to investigate the effect of perturbation amplitude and direction predictability as it may be another factor that needs to be more widely controlled in literature.

4.2 Attenuation of the N1 cortical response while dual-tasking

Of most relevance to this study was the influence of dual-tasking on the cortical responses associated with instability. The current results were comparable to those of Quant et al. (2004b) who reported dual-tasking resulted in a 32.5% reduction in the N1 peak amplitude at the Cz electrode. In the present study, the N1 peak amplitude attenuation ranged from 19.40 to 20.1% at the Cz electrode and from 18.7 to 25.6% at the FCz electrode. To elicit a perturbation, the study by Quant et al. (2004b) utilized a platform translation as opposed to a lean-and-release event. Little and Woolacott (2014) similarly published a dual-tasking platform translation study and results showed attenuations around 21% over multiple motor and sensory electrode sites. Little and colleagues (2014) utilized a visual memory secondary task instead of a visuomotor

tracking task. The variability of dual-task N1 peak amplitude attenuations between the studies may be related to both the characteristics of the perturbation used, the type of task/instructions given, and/or the type analysis used. However, in spite of the task related differences, the degree of attenuation was remarkably comparable.

Attending to a secondary task was hypothesized to elicit a difference in the characteristics of the N1 cortical response associated with a time-locked moment of instability (***Figure 21***). The use of a larger perturbation was set to examine if exogenous attention would lead to the same attenuation seen by Quant et al. (2004b) and Little and Woolacott (2014). As depicted by ***Figure 22***, a decrease in cortical activity is present even when perturbation amplitudes are clearly different. As previously interpreted by Quant et al. (2004b), the N1 may be bound by the allocation of attentional resources immediately after a perturbation. Attention in this case is the potential to assimilate sensory information (Herrmann and Knight, 2001; Woollacott and Shumway-Cook, 2002; Petersen and Posner, 2012). Attentional resources can be allocated as attention is believed to be a distributed network of subsystems, each overlapping and depicting different attentional processes (Posner, 1990; Petersen and Posner, 2012). The limits of said resource allocation, and therefore sensory processing, are then highlighted whenever a participant is asked to perform a secondary task prior to a perturbation. The voluntary act of endogenous orienting may have suppressed or masked the electrophysiological expression of the exogenous resource allocation associated with the perturbation. Electrophysiologically, the emergence of the N1 derives from the spatial and temporal summation of extracellular postsynaptic potentials that depolarize pyramidal neurons in layers III, IV and V of the cerebral cortex (Olejniczak, 2006). The attenuation of the N1 due to dual-tasking may then be a result of either an increased inhibition or a decreased excitation of neural pools (generators) of the N1. As

the N1 is thought to be a complex waveform with various generator sites it is difficult to isolate the N1 to just one site, although its peak was localized to the SMA, a premotor cortical region (Marlin et al., 2014; Varghese et al., 2014). In previous imaging literature, balance tasks have been associated with the activation of frontocentral areas of the cerebral cortex, such as the prefrontal cortex (Fukuyama et al., 1997; Duckrow et al., 1999; Ouchi et al., 1999; Jahn et al., 2004; Slobounov et al., 2006; Mihara et al., 2008). An overlap may exist between areas believed to be associated with attentional processes and areas related to balance control. It is possible that whenever completing the secondary task prior to the perturbation, attentional networks overlapping balance related processes are already in use. Once perturbation onset occurs, then it appears the N1 attenuates however other cortical analysis (e.g. phasic or connectivity tests) may be able to illustrate these changes in a way averaged data and single dipoles at one point in time cannot. The N1, to some extent, is believed to be a reflection of the allocation of attention, and the capability of the areas involved in the generation of the N1 may be altered, in terms of cortical excitability, due to endogenous processes and network overlaps when completing an attention demanding task just prior to a perturbation.

Whenever dual-tasking, particularly when the task is challenging, previous literature has shown that one of the two tasks being completed will show a decrease in performance (Kerr et al., 1985; Geurts et al., 1991; Lajoie et al., 1993, 1995; Maylor and Wing, 1996; Shumway-Cook et al., 1997; Schlesinger et al., 1998; Brown et al., 1999; Yardley et al., 1999; McIlroy et al., 1999; Rankin et al., 2000; Norrie et al., 2002; Quant et al., 2004b; Jehu et al., 2014; Little and Woollacott, 2014). Dual-task interference is a phenomenon attributed to the temporal changes in attention and the bottlenecking effect of resource allocation (Woollacott and Shumway-Cook, 2002; Maki et al., 2003). Quant et al. (2004b) and Little and Woolacott (2014) interpreted that

observed N1 peak amplitude attenuations seen with dual-task paradigms may in turn be related to the changes observed in the participant's postural control (post-APR). Quant et al. (2004b) saw both a change in AP CoP excursions (displacement and velocity) and an increase of TA EMG amplitude after the first 50 ms post-EMG onset when compared to single-task trials. Similarly, Little and Woolacott (2014) recorded changes in AP COP excursion, but saw no differences related to dual-tasking on either the EMG onset or magnitude. In the current study, behavioral changes due to dual-tasking were only found in step onset, but not foot-off, EMG latencies or EMG amplitudes. Both Quant et al. (2004b) and Little and Woolacott (2014) argue that postural CoP differences seen in dual-tasking trials are related to an inefficacy on part of the CNS whenever confronted with multiple attentional demanding tasks. This inefficacy, they believe, is highlighted by the attenuation of the N1. However, it is important to note that although a moderate attenuation of the N1 peak amplitude is observed whenever completing a task prior to an instability, the behavioral changes noted in reactive balance control within literature have all been minimal. If the N1 was to be critical to reactive balance control, it would be expected for a more pronounced deviation of both postural and behavioral parameters.

4.3 What is the role of the N1?

The N1 potential remains the most consistently evoked and studied perturbation-related cortical event. Although the role of the N1 is currently unknown, it is widely accepted that the N1 is more than a reflection of afferent information resultant from the application of an instability. Instead, the N1 is believed to be a more complex waveform shaped by a distributed network of neural pools and therefore depicting numerous, bound cortical processes related to a perturbation (Varghese et al., 2014). Two opposing models are presently used to describe what

the role of the N1 might be in the context of reactive balance control (Bolton, 2015b). Firstly, due to the influence of temporal predictability, perturbation amplitude, and the emergence of an N1 independent of motor control, the N1 has been hypothesized to function as an error detection signal between the predicted and the actual outcome of an instability, much like the error-related negativity (ERN) observed in behavioral neuroscience (Dimitrov et al., 1996; Scheffers and Coles, 2000; Adkin et al., 2006; Mochizuki et al., 2008). An N1 perspective that heavily relies on environmental contrasts is described to be an error model of the role of the N1. However, a discrepancy exists as the ERN peak appears to be localized to the anterior cingulate cortex (ACC), while the N1 peak appears to be localized to the supplementary motor area (SMA, Marlin et al., 2014). The SMA, part of Brodmann area 6, relays information gathered by the basal ganglia (through the anterior nucleus of the thalamus), the prefrontal cortex, the association cortex and the limbic cortex towards primary motor cortex (Kandel, R. Eric et al., 2012). The SMA is known to play a role in the preparation of voluntary movements with some evidence also pointing towards direct motor control as 30 - 40% of neurons that compromise the corticospinal tract are estimated to emerge from the SMA (Kandel, R. Eric et al., 2012). Wide connections to various association and motor cortices, alongside of evidence of dual-tasking studies indicating that the earliest balance responses to a perturbation does not require attention, led to the speculation that the N1 may be related to the execution/planning of motor responses at late/long-phases of a balance reaction (McIlroy et al., 1999; Norrie et al., 2002; Marlin et al., 2014). A second, motor model for the role of the N1 falls in-line with the neural pathway hypothesis for cortical involvement proposed by Jacobs et al. (2007). The neural pathway model for cortical involvement explains that the longer a latency is to a motor reaction, the more likely that reaction is to involve the cerebral cortex (Jacobs and Horak, 2007). Dual-task paradigms are often used as

evidence of cortical involvement due to the behavioral and performance changes that are observed whenever a balance task is paired with a cognitive task (Brown et al., 1999; McIlroy et al., 1999; Rankin et al., 2000; Brauer et al., 2002; Norrie et al., 2002; Redfern et al., 2002; Woollacott and Shumway-Cook, 2002; Quant et al., 2004b; Zettel et al., 2008; Little and Woollacott, 2014). Attention is known to be a factor that influences the N1 (Quant et al., 2004b; Little and Woollacott, 2014). The present thesis aimed at manipulating attention, by applying large perturbations, in order to better understand the interaction between endogenous/exogenous attentional processes and the N1 perturbation-evoked potential.

Marlin et al. (2014) has provided a different perspective as to what the role of the N1 is, hypothesizing motor preparation as the main role of the N1 due to its localization to the SMA, a cortical pre-motor cortex site; specifically, in the execution/planning of the motor responses. Knowing that attention is not believed to be a critical component of early balance reactions (i.e. APR), Marlin et al. (2014) suggested the N1 is involved in later phases of the balance reactions, though the specific definition of such responses is not well defined. Authors Jacobs and Horak (2007), explain that late-phased balance reactions are post-APR and likely emergent with reaching/stepping responses. However, in the large perturbation dual-tasking trials of the present study, no changes in foot-off values were observed even though the latency of such a response took place at about 450 ms post-perturbation. The only changes due to dual-tasking were observed during step onset at approximately 200 ms. Thus, the results of this study do not appear consistent with the idea the N1 is related to the motor planning/execution at the later phases of a balance reaction. Instead, the results of the present study points towards the idea the role of the N1 is a reflection of event detection (Maki and McIlroy, 2007; Mochizuki et al., 2009a). If the role of the N1 is to indeed plan for the motor execution of later phases of balance reactions post-

perturbation, then moderate changes should have been observed in parallel with moderate N1 peak attenuations during dual-tasking trials. Although previous literature has hypothesized a link between cortical and muscular activity post-perturbation, only small changes in EMG amplitude have been observed in trials in which the N1 is found to be significantly attenuated due to dual-tasking (Quant et al., 2004b; Little and Woollacott, 2014). It can be argued that EEG and EMG appear to be influenced by different factors related to the perturbation applied. The N1 specifically appears to be shaped by a mixture of endogenous and exogenous processes, with factors such as: postural threat, temporal predictability, and dual-tasking leading to significant changes to cortical responses (Adkin et al., 2006; Mochizuki et al., 2008; Sibley et al., 2010). In contrast, EMG measures are significantly shaped by the amplitude and direction of the perturbation to an extent the EEG measures do not appear to be influenced (Diener et al., 1984; Maki and McIlroy, 1997). Future research needs to be able to specify the characteristics governing EEG and EMG signals. Such information may be able to help guide researchers as to what the role of the cerebral cortex is to reactive balance control.

The N1 is observed over a wide variety of protocols, it can be argued that it is a generic event detection signal (Dietz et al., 1984, 1985a, 1985b; Quintern et al., 1985; Ackermann et al., 1986; Dimitrov et al., 1996; Duckrow et al., 1999; Staines et al., 2001; Quant et al., 2004b, 2005; Adkin et al., 2006; Mochizuki et al., 2009b, 2010, 2008, 2009a, Varghese et al., 2014, 2015; Little and Woollacott, 2014). In light of the need for reactive balance control to be accurate and time efficient, the N1 may be cortical processing depicting contrasts of predicted outcomes (Bolton, 2015a). As explained by Bolton et al. (2015a), a predictable "experienced" system would not need complete afferent information in order for appropriate responses to be made. This idea fits well with the non-linear scaling between perturbation amplitude and the N1 peak

amplitude observed in the present study. It may be that initial rates of acceleration, and their respective afferent volleys, have to be drastically different to be reflected by the N1; making the N1 cortical responses heavily reliant on predicted outcomes, much more than previously thought. A system that is always predicting outcomes would therefore be difficult to study from single snap shot perspective. The expand on this point, the peak of the N1 and the ERN do not appear to share the same generators (Scheffers and Coles, 2000; Marlin et al., 2014). In fact, the N1 does not appear to have just one neural generator, meaning a single dipole localization may not provide enough insight to direct further speculations about what the role of the N1 is to reactive balance control (Marlin et al., 2014; Varghese et al., 2014). This, coupled with evidence of overlapping between balance and attentional networks points towards the idea that the current analysis of cortical responses associated with instability, specifically the N1, are flawed and should be reevaluated. The N1 attenuations seen during dual-tasking in literature all show a 21 to 33% cortical decrease (Quant et al., 2004b; Little and Woollacott, 2014). It may be that the attenuations of the N1 are a result of the limitations of single dipole observations of a multisite cortical process. A greater understanding might derive from an analysis of factors known to affect the N1 that is based on connectivity and the phasic changes between different cortical areas. Future research should further aim at exploring parallels between attention related processes and the N1 PEP. The present research highlights the interaction between attentional and balance related networks that should be further pursued in order to specify how divided attention shapes cortical events related to reactive balance control.

EEG has been the most reliable neuroimaging technique for reactive balance control research for scientists due to a high amount of temporal precision. EEG enables the recording of electrophysiological signals at the outer layers of the cerebral cortex. However, EEG contains

minimal spatial resolution that is limited to the recordings of neural pools (dipoles) in layers III, IV and V of the cerebral cortex (Olejniczak, 2006). Future studies pairing EEG and a dual-tasking should inquire about various factors related to research methodology. Firstly, the level of secondary task challenge modulation relative to the level of perceived threat posed by the perturbation needs to be further explored. Moreover, more information needs to be gathered regarding how factors surrounding the perturbation (i.e. instructions, level of temporal and amplitude predictability, trial-to-trial analysis) independently alter the N1. Especially as these factors may not be easily manipulated within a LaR system; meaning results could eventually guide researchers as to which technique is most suitable to answer the question at hand (e.g. seated translations, standing translations, platform tilts, LaR, etc.). Lastly, due to the evidence gathered in this study, future research should aim at exploring reactive balance control through the use of network analysis, cortical areas connectivity and information flow.

4.4 Conclusions

Dual-tasking paradigms have served as a tool to examine a patient's ability to perform under divided attention, particularly in older adults (Woollacott and Shumway-Cook, 2002). The loss of performance that emerges with the concurrent execution of a mental task, alongside a balance task, infers about the contributions that exist from higher-order cortical areas to the success of balance control. The present study sought to use the EEG technique paired with a dual-task paradigm to directly characterize cortical involvement; specifically, following a moment of instability. However, prior to examining those of older age, a better understanding of factors influencing the widely studied N1 cortical response needs further development; specifically, in controlling for all the aspects that may shape cortical events. The goal of the

present study was to pair electroencephalography alongside a cognitive task while on a lean-and-release (LaR) protocol in the hopes to expand on the specifics of said cortical involvement in reactive balance control. It was hypothesized that due to the balance threat of large amplitude perturbation trials, a more attentional demanding exogenously driven event, the N1 PEP would not be attenuated as previously seen in literature (Quant et al., 2004b; Little and Woollacott, 2014). The hypothesis was based on the fact that independently, perturbation amplitude is known to be scaled to the amplitude of the peak N1 potential (Staines et al., 2001). Due to the speculated attentional demands of large perturbation trials, it was further hypothesized that behavioral parameters would be affected to reflect a faster response to instability. In summary, results from this study reinforce previous literature in which an attenuation of the N1 cortical response is observed whenever dual-tasking. The study was not able to confirm that this attenuation was associated a surplus of resources available in single task conditions since it was evident even under large amplitude perturbations. The mechanism and cause of this inhibition remains to be determined. Surprisingly, a scaling effect between perturbation amplitude and N1 peak amplitude was not observed although there were differences in evoked reactions comparing between perturbations of different amplitude. Evidence collected within the study points towards the possibility of a non-linear perturbation amplitude to N1 scaling. Overall, the results speak on the importance of attentional resource allocation to the N1 composition. The lack of significant behavioral changes related to dual-tasking in balance control, especially in the later phases, depicts the role of the N1 as a more general event detection process in-line with an environmental/error related model. Future research needs to further explore the parallels between attentional and balance related networks. The study of exogenous processes related to specific aspects of the applied perturbation may enable the possibility of understanding the attention

influences on cortical responses, and may even help disentangle what the role of the N1 is to rapid balance reactions; further enabling the possibility to research cortical markers associated with reactive balance in those with sensory and/or motor deficits as a result of a cortical accident or disease.

References

Ackermann H, Diener HC, Dichgans J (1986) Mechanically evoked cerebral potentials and long-latency muscle responses in the evaluation of afferent and efferent long-loop pathways in humans. Neurosci Lett 66:233–238.

Adkin AL, Quant S, Maki BE, McIlroy WE (2006) Cortical responses associated with predictable and unpredictable compensatory balance reactions. Exp brain Res 172:85–93.

Adkins RJ, Cegnar MR, Rafuse DD (1971) Differential effects of lesions of the anterior and posterior sigmoid gyri in cats. Brain Res 30:411–414.

Alexander NB, Mollo JM, Giordani B, Ashton-Miller J a, Schultz a B, Grunawalt J a, Foster NL (1995) Maintenance of balance, gait patterns, and obstacle clearance in Alzheimer's disease. Neurology 45:908–914.

Asanome M, Matsuyama K, Mori S (1998) Augmentation of postural muscle tone induced by the stimulation of the descending fibers in the midline area of the cerebellar white matter in the acute decerebrate cat. Neurosci Res 30:257–269.

Ashburn A (1997) Physical Recovery Following Stroke. Physiotherapy 83:480–490.

Beloozerova IN (2003) Postural Control in the Rabbit Maintaining Balance on the Tilting Platform. J Neurophysiol 90:3783–3793.

Beloozerova IN, Sirota MG, Swadlow H a, Orlovsky GN, Popova LB, Deliagina TG (2003) Activity of different classes of neurons of the motor cortex during postural corrections. J Neurosci 23:7844–7853.

Bolton D a. E (2015a) The role of the cerebral cortex in postural responses to externally induced perturbations. Neurosci Biobehav Rev 57:142–155.

Bolton DAE (2015b) The role of the cerebral cortex in postural responses to externally induced perturbations. Neurosci Biobehav Rev 57:142–155.

Brauer SG, Woollacott M, Shumway-Cook A (2002) The influence of a concurrent cognitive task on the compensatory stepping response to a perturbation in balance-impaired and healthy elders. Gait Posture 15:83–93.

Brooks CM (1988) Localized representation of hopping and placing reactions in the rat. Arch Neurol Psychiatry 2:1–24.

Brown L a, Shumway-Cook A, Woollacott MH (1999) Attentional demands and postural recovery: the effects of aging. J Gerontol A Biol Sci Med Sci 54:M165–M171.

Burleigh a L, Horak FB, Malouin F (1994) Modification of postural responses and step initiation: evidence for goal-directed postural interactions. J Neurophysiol 72:2892–2902.

Chambers WW, Liu C-N (1957) Cortico-spinal tract of the cat. An attempt to correlate the pattern of degeneration with deficits in reflex activity following neocortical lesions. J Comp Neurol 108:23–55.

Chan CW, Jones GM, Catchlove RF (1979) The "late" electromyographic response to limb displacement in man. II. Sensory origin. Electroencephalogr Clin Neurophysiol 46:182–188.

Chapman JP, Chapman LJ, Allen JJ (1987) The measurement of foot preference. Neuropsychologia 25:579–584.

Chong RK, Jones CL, Horak FB (1999) Postural set for balance control is normal in Alzheimer's but not in Parkinson's disease. J Gerontol A Biol Sci Med Sci 54:M129–M135.

Corbetta M, Shulman GL (2002) Control of goal-directed and stimulus-driven attention in the

brain. Nat Rev Neurosci 3:201–215.

Corriveau H, Hébert R, Raîche M, Prince F (2004) Evaluation of postural stability in the elderly with stroke. Arch Phys Med Rehabil 85:1095–1101.

Dehaene S, Posner MI, Tucker DM (1994) Localization of a Neural System for Error Detection and Compensation. Psychol Sci 5:303–305.

Deliagina TG, Zelenin P V, Beloozerova IN, Orlovsky GN (2007) Nervous mechanisms controlling body posture. Physiol Behav 92:148–154.

Delorme A, Makeig S (2004) EEGLAB: an open source toolbox for analysis of single-trial EEG dynamics including independent component analysis. J Neurosci Methods 134:9–21.

Diener HC, Ackermann H, Dichgans J, Guschlbauer B (1985) Medium- and long-latency responses to displacements of the ankle joint in patients with spinal and central lesions. Electroencephalogr Clin Neurophysiol 60:407–416.

Diener HC, Dichgans J, Bootz F, Bacher M (1984) Early stabilization of human posture after a sudden disturbance: influence of rate and amplitude of displacement. Exp Brain Res 56:126–134.

Diener HC, Horak FB, Nashner LM (1988) Influence of stimulus parameters on human postural responses. J Neurophysiol 59:1888–1905.

Dietz V (1992) Human neuronal control of automatic functional movements: interaction between central programs and afferent input. Physiol Rev 72:33–69.

Dietz V, Quintern J, Berger W (1984) Cerebral evoked potentials associated with the compensatory reactions following stance and gait perturbation. Neurosci Lett 50:181–186.

Dietz V, Quintern J, Berger W (1985a) Afferent control of human stance and gait: evidence for

blocking of group I afferents during gait. Exp Brain Res 61:153–163.

Dietz V, Quintern J, Berger W, Schenck E (1985b) Cerebral potentials and leg muscle e.m.g. responses associated with stance perturbation. Exp Brain Res 57:348–354.

Dimitrov B, Gavrilenko T, Gatev P (1996) Mechanically evoked cerebral potentials to sudden ankle dorsiflexion in human subjects during standing. Neurosci Lett 208:199–202.

Dubrovsky B, Garcia Rill E, Surkes MA (1974) Effects of discrete precruciate cortex lesions on motor behavior. Brain Res 82:328–333.

Duckrow RB, Abu-Hasaballah K, Whipple R, Wolfson L (1999) Stance perturbation-evoked potentials in old people with poor gait and balance. Clin Neurophysiol 110:2026–2032.

Egeth H, Kahneman D (1975) Attention and Effort. Am J Psychol 88:339.

Eng J (2003) Sample Size Estimation: How Many Individuals Should Be Studied? Radiology 227:309–313.

Fraizer E V., Mitra S (2008) Methodological and interpretive issues in posture-cognition dual-tasking in upright stance. Gait Posture 27:271–279.

Fukuyama H, Ouchi Y, Matsuzaki S, Nagahama Y, Yamauchi H, Ogawa M, Kimura J, Shibasaki H (1997) Brain functional activity during gait in normal subjects: A SPECT study. Neurosci Lett 228:183–186.

Garland SJ, Ivanova TD, Mochizuki G (2007) Recovery of standing balance and health-related quality of life after mild or moderately severe stroke. Arch Phys Med Rehabil 88:218–227.

Garland SJ, Willems D a, Ivanova TD, Miller KJ (2003) Recovery of standing balance and functional mobility after stroke. Arch Phys Med Rehabil 84:1753–1759.

Geurts a C, Mulder TW, Nienhuis B, Rijken R a (1991) Dual-task assessment of reorganization of postural control in persons with lower limb amputation. Arch Phys Med Rehabil 72:1059–1064.

Geurts ACH, De Haart M, Van Nes IJW, Duysens J (2005) A review of standing balance recovery from stroke. Gait Posture 22:267–281.

Herrmann CS, Knight RT (2001) Mechanisms of human attention: event-relates potentials and oscillations. Neurosci Biobehav Rev 25:465–476.

Honeycutt CF, Nichols TR (2010) The decerebrate cat generates the essential features of the force constraint strategy. J Neurophysiol 103:3266–3273.

Horak F, Shupert C, Mirka A (1989) Components of postural dyscontrol in the elderly: A review. Neurobiol Aging 10:727–738.

Horak FB, Nashner LM (1986) Central programming of postural movements: adaptation to altered support-surface configurations. J Neurophysiol 55:1369–1381.

Hyndman D, Ashburn A (2003) People with stroke living in the community: Attention deficits, balance, ADL ability and falls. Disabil Rehabil 25:817–822.

Inness EL, Mansfield A, Biasin L, Brunton K, Bayley M, McIlroy WE (2015) Clinical implementation of a reactive balance control assessment in a sub-acute stroke patient population using a “lean-and-release” methodology. Gait Posture 41:529–534.

Inness EL, Mansfield A, Lakhani B, Bayley M, McIlroy WE (2014) Impaired Reactive Stepping Among Patients Ready for Discharge From Inpatient Stroke Rehabilitation. Phys Ther 94:1755–1764.

Jacobs J V, Fujiwara K, Tomita H, Furune N, Kunita K, Horak FB (2009) Response

Modification When Warned of a Perturbation. 119:1431–1442.

Jacobs J V, Horak FB (2007) Cortical control of postural responses. J Neural Transm 114:1339–1348.

Jahn K, Deutschländer A, Stephan T, Strupp M, Wiesmann M, Brandt T (2004) Brain activation patterns during imagined stance and locomotion in functional magnetic resonance imaging. Neuroimage 22:1722–1731.

Jehu D a, Desponts A, Paquet N, Lajoie Y (2014) Prioritizing attention on a reaction time task improves postural control and reaction time. Int J Neurosci 125:1–7.

Jung TP, Makeig S, Humphries C, Lee TW, McKeown MJ, Iragui V, Sejnowski TJ (2000) Removing electroencephalographic artifacts by blind source separation. Psychophysiology 37:163–178.

Kaiser D a. (2005) Basic Principles of Quantitative EEG. J Adult Dev 12:99–104.

Kandel, R. Eric; Schwartz, H. James; Jessell, M. Thomas; Siegelbaum, A. Steven; Hudspeth AJ. (2012) The Principles of Neural Science. McGraw-Hill Education.

Karayannidou a, Beloozerova IN, Zelenin P V, Stout EE, Sirota MG, Orlovsky GN, Deliagina TG (2009) Activity of pyramidal tract neurons in the cat during standing and walking on an inclined plane. J Physiol 587:3795–3811.

Keck ME, Pijnappels M, Schubert M, Colombo G, Curt A, Dietz V (1998) Stumbling reactions in man: Influence of corticospinal input. Electroencephalogr Clin Neurophysiol - Electromyogr Mot Control 109:215–223.

Kerr B, Condon SM, McDonald L a (1985) Cognitive spatial processing and the regulation of posture. J Exp Psychol Hum Percept Perform 11:617–622.

Lajoie Y, Teasdale N, Bard C, Fleury M (1993) Attentional demands for static and dynamic equilibrium. Exp Brain Res 97:139–144.

Lajoie Y, Teasdale N, Bard C, Fleury M (1995) Upright standing and gait: are there changes in attentional requirements related to normal aging? Exp Aging Res 22:185–198.

Lakhani B, Mansfield A, Inness EL, McIlroy WE (2011a) Compensatory stepping responses in individuals with stroke: A pilot study. Physiother Theory Pract 27:299–309.

Lakhani B, Mansfield A, Inness EL, McIlroy WE (2011b) Characterizing the determinants of limb preference for compensatory stepping in healthy young adults. Gait Posture 33:200–204.

Little CE, Woollacott M (2014) EEG measures reveal dual-task interference in postural performance in young adults. Exp Brain Res 233:27–37.

Luck S, Woodman GF, Vogel EK (2000) Event-related potential studies of attention. Trends Cogn Sci 4:432–440.

Lyalka VF (2005) Impairment and Recovery of Postural Control in Rabbits With Spinal Cord Lesions. J Neurophysiol 94:3677–3690.

Macpherson JM, Fung J (1999) Weight support and balance during perturbed stance in the chronic spinal cat. J Neurophysiol 82:3066–3081.

Maki BE, Holliday PJ, Topper a K (1994) A Prospective Study of Postural Balance and Risk of Falling in An Ambulatory and Independent Elderly Population. J Gerontol 49:M72–M84.

Maki BE, McIlroy WE (1996) Influence of arousal and attention on the control of postural sway. J Vestib Res 6:53–59.

Maki BE, McIlroy WE (1997) The role of limb movements in maintaining upright stance: the

"change-in-support" strategy. Phys Ther 77:488–507.

Maki BE, McIlroy WE (1999) Control of compensatory stepping reactions: Age-related impairment and the potential for remedial intervention. Physiother Theory Pr 15:69–90.

Maki BE, McIlroy WE (2007) Cognitive demands and cortical control of human balance-recovery reactions. J Neural Transm 114:1279–1296.

Maki BE, McIlroy WE, Fernie GR (2003) Change-in-Support Reactions for Balance Recovery. IEEE Eng Med Biol Mag 22:20–26.

Maki BE, McIlroy WE, Perry SD (1996) Influence of lateral destabilization on compensatory stepping responses. J Biomech 29:343–353.

Maki BE, Zecevic A, Bateni H, Kirshenbaum N, McIlroy WE (2001) Cognitive demands of executing postural reactions: does aging impede attention switching? Neuroreport 12:3583–3587.

Mansfield a., Wong JS, McIlroy WE, Biasin L, Brunton K, Bayley M, Inness EL (2015) Do measures of reactive balance control predict falls in people with stroke returning to the community? Physiotherapy:1–8.

Mansfield A, Inness EL, Komar J, Biasin L, Brunton K, Lakhani B, McIlroy WE (2011a) Training rapid stepping responses in an individual with stroke. Phys Ther 91:958–969.

Mansfield A, Inness EL, Komar J, Biasin L, Brunton K, Lakhani B, McIlroy WE (2011b) Training Rapid Stepping Responses in an Individual With Stroke. Phys Ther 91:958–969.

Mansfield A, Inness EL, Lakhani B, McIlroy WE (2012) Determinants of limb preference for initiating compensatory stepping poststroke. Arch Phys Med Rehabil 93:1179–1184.

Mansfield A, Inness EL, Wong JS, Fraser JE, McIlroy WE (2013) Is impaired control of reactive

stepping related to falls during inpatient stroke rehabilitation? Neurorehabil Neural Repair 27:526–533.

Marlin A, Mochizuki G, Staines WR, McIlroy WE (2014) Localizing evoked cortical activity associated with balance reactions: does the anterior cingulate play a role? J Neurophysiol 111:2634–2643.

Martinez-Mendez R, Sekine M, Tamura T (2011) Detection of anticipatory postural adjustments prior to gait initiation using inertial wearable sensors. J Neuroeng Rehabil 8:17.

Martinez-Mendez R, Sekine M, Tamura T (2012) Postural sway parameters using a triaxial accelerometer: comparing elderly and young healthy adults. Comput Methods Biomech Biomed Engin 15:899–910.

Matthews PBC (1991) The human stretch reflex and the motor cortex. Trends Neurosci 14:87–91.

Maylor E a, Wing a M (1996) Age differences in postural stability are increased by additional cognitive demands. J Gerontol B Psychol Sci Soc Sci 51:P143–P154.

McIlroy W, Maki B (1995) Adaptive changes to compensatory stepping responses. Gait Posture 3:43–50.

McIlroy WE, Maki BE (1993) Changes in early "automatic" postural responses associated with the prior-planning and execution of a compensatory step. Brain Res 631:203–211.

McIlroy WE, Maki BE (1996) Influence of Destabilization on the Temporal Characteristics of "Volitional" Stepping. J Mot Behav 28:28–34.

McIlroy WE, Maki BE (1997) Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. Clin Biomech (Bristol, Avon) 12:66–

70.

McIlroy WE, Norrie RG, Brooke JD, Bishop DC, Nelson a J, Maki BE (1999) Temporal properties of attention sharing consequent to disturbed balance. Neuroreport 10:2895–2899.

Mihara M, Miyai I, Hatakenaka M, Kubota K, Sakoda S (2008) Role of the prefrontal cortex in human balance control. Neuroimage 43:329–336.

Miltner W, Braun C, Johnson R, Simpson G V., Ruchkin DS (1994) A test of brain electrical source analysis (BESA): A simulation study. Electroencephalogr Clin Neurophysiol 91:295–310.

Misiaszek JE, Krauss EM (2005) Restricting arm use enhances compensatory reactions of leg muscles during walking. Exp Brain Res 161:474–485.

Mochizuki G, Boe S, Marlin a, McIlRoy WE (2010) Perturbation-evoked cortical activity reflects both the context and consequence of postural instability. Neuroscience 170:599–609.

Mochizuki G, Sibley KM, Cheung HJ, Camilleri JM, McIlroy WE (2009a) Generalizability of perturbation-evoked cortical potentials: Independence from sensory, motor and overall postural state. Neurosci Lett 451:40–44.

Mochizuki G, Sibley KM, Cheung HJ, McIlroy WE (2009b) Cortical activity prior to predictable postural instability: is there a difference between self-initiated and externally-initiated perturbations? Brain Res 1279:29–36.

Mochizuki G, Sibley KM, Esposito JG, Camilleri JM, McIlroy WE (2008) Cortical responses associated with the preparation and reaction to full-body perturbations to upright stability. Clin Neurophysiol 119:1626–1637.

Moore SP, Rushmer DS, Windus SL, Nashner LM (1988) Human automatic postural responses: responses to horizontal perturbations of stance in multiple directions. Exp Brain Res 73:648–658.

Mori S (1987) Integration of posture and locomotion in acute decerebrate cats and in awake, freely moving cats. Prog Neurobiol 28:161–195..

Müller MLTM, Jennings JR, Redfern MS, Furman JM (2004) Effect of preparation on dual-task performance in postural control. J Mot Behav 36:137–146.

Musienko PE, Zelenin PV, Lyalka VF, Orlovsky GN, Deliagina TG (2008) Postural performance in decerebrated rabbit. Behav Brain Res 190:124–134.

Nashner LM (1976) Adapting reflexes controlling the human posture. Exp Brain Res 26:59–72.

Nashner LM (1977) Fixed patterns of rapid postural responses among leg muscles during stance. Exp Brain Res 30:13–24.

Norrie RG, Maki BE, Staines RW, McIlroy WE (2002) The time course of attention shifts following perturbation of upright stance. Exp Brain Res 146:315–321.

Nunez PL (1988) Methods to improve spatial resolution of EEG. In: Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, pp 972–973 vol.2. IEEE.

Olejniczak P (2006) Neurophysiologic Basis of EEG. J Clin Neurophysiol 23:186–189.

Ouchi Y, Okada H, Yoshikawa E, Nobezawa S, Futatsubashi M (1999) Brain activation during maintenance of standing postures in humans. Brain 122:329–338.

Pérennou DA, Leblond C, Amblard B, Micallef JP, Rouget E, Pélissier J (2000) The polymodal sensory cortex is crucial for controlling lateral postural stability: Evidence from stroke

patients. Brain Res Bull 53:359–365.

Petersen S., Posner M (2012) The Attention System of the Human Brain: 20 Years After. Annu Rev Neurosci 21:73–89.

Pollock AS, Durward BR, Rowe PJ, Paul JP (2000) What is balance? Clin Rehabil 14:402–406.

Posner P (1990) The attention system of the human brain. Annu Rev Neurosci 13:24–42.

Quant S, Adkin a L, Staines WR, McIlroy WE (2004a) Cortical activation following a balance disturbance. Exp brain Res 155:393–400.

Quant S, Adkin AL, Staines WR, Maki BE, McIlroy WE (2004b) The effect of a concurrent cognitive task on cortical potentials evoked by unpredictable balance perturbations. BMC Neurosci 5:18.

Quant S, Maki BE, McIlroy WE (2005) The association between later cortical potentials and later phases of postural reactions evoked by perturbations to upright stance. Neurosci Lett 381:269–274.

Quintern J, Berger W, Dietz V (1985) Compensatory reactions to gait perturbations in man: short- and long-term effects of neuronal adaptation. Neurosci Lett 62:371–376.

Rankin JK, Woollacott MH, Shumway-Cook A, Brown L a (2000) Cognitive influence on postural stability: a neuromuscular analysis in young and older adults. J Gerontol A Biol Sci Med Sci 55:M112–M119.

Rapport L (1993) Predictors of falls among right-hemisphere stroke patients in the rehabilitation setting,. Arch Phys Med Rehabil 74:621–626.

Redfern MS, Müller MLTM, Jennings JR, Furman JM (2002) Attentional dynamics in postural control during perturbations in young and older adults. J Gerontol A Biol Sci Med Sci

57:B298–B303.

Ruffieux J, Keller M, Lauber B, Taube W (2015) Changes in Standing and Walking Performance Under Dual-Task Conditions Across the Lifespan. Sport Med 45:1739–1758.

Safavynia S a., Ting LH (2012) Task-level feedback can explain temporal recruitment of spatially fixed muscle synergies throughout postural perturbations. J Neurophysiol 107:159–177.

Scheffers MK, Coles MG (2000) Performance monitoring in a confusing world: error-related brain activity, judgments of response accuracy, and types of errors. J Exp Psychol Hum Percept Perform 26:141–151.

Schlesinger A, Redfern MS, Dahl RE, Jennings JR (1998) Postural control, attention and sleep deprivation.

Scott V, Wagar L, Elliott S (2011) Falls & Related Injuries among Older Canadians: Fall-related Hospitalizations & Prevention Initiatives. Div Aging Seniors, Public Heal Agency Canada:1–43.

Shumway-Cook a, Woollacott M, Kerns K a, Baldwin M (1997) The effects of two types of cognitive tasks on postural stability in older adults with and without a history of falls. J Gerontol A Biol Sci Med Sci 52:M232–M240.

Sibley KM, Mochizuki G, Frank JS, McIlroy WE (2010) The relationship between physiological arousal and cortical and autonomic responses to postural instability. Exp Brain Res 203:533–540.

Sibley KM, Voth J, Munce SE, Straus SE, Jaglal SB (2014) Chronic disease and falls in community-dwelling Canadians over 65 years old: a population-based study exploring

associations with number and pattern of chronic conditions. BMC Geriatr 14:22.

Slobounov S, Wu T, Hallett M (2006) Neural basis subserving the detection of postural instability: an fMRI study. Motor Control 10:69–89.

Smartrisk (2009) The Economic Burden of Injury in Canada.

Staines WR, McIlroy WE, Brooke JD (2001) Cortical representation of whole-body movement is modulated by proprioceptive discharge in humans. Exp Brain Res 138:235–242.

Statistics Canada (2011) Analytical document The Canadian Population in 2011 : Age and Sex. :27.

Stevens J a, Dellinger a M (2002) Motor vehicle and fall related deaths among older Americans 1990-98: sex, race, and ethnic disparities. Inj Prev 8:272–275.

Stinchcombe A, Kuran N, Powell S (2014) Seniors' falls in Canada: Second report: Key highlights. Chronic Dis Inj Can.

Taube W, Schubert M, Gruber M, Beck S, Faist M, Gollhofer A (2006) Direct corticospinal pathways contribute to neuromuscular control of perturbed stance. J Appl Physiol 101:420–429.

Vanvoorhis CRW, Morgan BL (2007) Understanding Power and Rules of Thumb for Determining Sample Sizes. Tutorials Quant Methods 3:43–50.

Varghese JP, Beyer KB, Williams L, Miyasike-daSilva V, McIlroy WE (2015) Standing still: Is there a role for the cortex?.

Varghese JP, Marlin A, B Beyer K, Staines WR, Mochizuki G, McIlroy WE (2014) Frequency characteristics of cortical activity associated with perturbations to upright stability. Neurosci Lett 578C:33–38.

Vellas B, Cayla F, Bocquet H, de Pemille F, Albarede JL (1987) Prospective study of restriction of activity in old people after falls. Age Ageing 16:189–193.

Winter D (1995) Human balance and posture control during standing and walking. Gait Posture 3:193–214.

Wood JW (1964) Behaviour of chronic decerebrate rats. J Neurophysiol 27:635–644.

Woollacott M, Shumway-Cook A (2002) Attention and the control of posture and gait: A review of an emerging area of research. Gait Posture 16:1–14.

Xu J, Ma H (2009) An Introduction to Event-Related Potentials and Their Neural Origins. Acta Mech Sin 25:319–324.

Yardley L, Gardner M, Leadbetter A, Lavie N (1999) Effect of articulatory and mental tasks on postural control. Neuroreport 10:215–219.

Zettel JL, Holbeche A, McIlroy WE, Maki BE (2005) Redirection of gaze and switching of attention during rapid stepping reactions evoked by unpredictable postural perturbation. Exp Brain Res 165:392–401.

Zettel JL, McIlroy WE, Maki BE (2008) Effect of competing attentional demands on perturbation-evoked stepping reactions and associated gaze behavior in young and older adults. J Gerontol A Biol Sci Med Sci 63:1370–1379.