ASSESSING THE ATTENTIONAL DEMANDS OF ADDING HAPTIC INPUT DURING OVERGROUND WALKING

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ABSTRACT

Poor performance of two or more tasks have been linked to recurrent falls, lower attentional capacity and inability to allocate attention appropriately in older adults (Beauchet et al. 2008). Increasing attentional demands during walking through the addition of other tasks (i.e., modality use, cognitive tasks) can increase fall-risk in older adults, as the ability to achieve successful performance of two or more tasks is affected (Woollacott & Shumway-Cook, 2002).

The addition of sensory input in the form of haptic modalities, such as light touch (LT) of a rigid railing with less than 1 newton of force (Holden, Ventura, & Lackner, 1994), or haptic anchors (Mauerberg-deCastro et al., 2014), which involves pulling a light weight (~ 125 grams) attached to a string in each hand have been observed to improve dynamic stability, while not providing mechanical support. Determining the attentional demands of haptic modalities and the effect on dynamic stability will assist in better understanding their impact on fall-risk.

The primary objective of this thesis was to assess the attentional demands of haptic modalities during walking using a verbal reaction time (VRT) task in healthy, young adults. The secondary objective of this thesis was to assess the effect of haptic modalities during walking with an added VRT task on dynamic stability.

Twenty-two (12 male) healthy, young adults completed the testing protocol. Participants performed walking without haptic modalities (baseline), with LT of a rigid railing, and use of haptic anchors, with and without a VRT task that involved responding to a low or high frequency tone with the word "low" or "high", respectively. A one-way RM ANOVA [condition (Baseline/LT/Anchors)] was performed on VRTs to assess attentional demands. A 2 × 2 RM ANOVA [condition (baseline walking/haptic modality) × presence of VRT task (no VRT task/VRT task)] on all calculated kinematic variables for each haptic modality separately to measure dynamic stability and walking performance with the addition of the haptic modality and the VRT task.

No significant differences were observed (p = 0.506) between VRTs during walking conditions suggesting haptic modalities require similar attentional demands compared to baseline walking. It was observed that ML MOS was significantly decreased with LT (p < 0.001) and anchors (p = 0.010) suggesting using haptic modalities affects dynamic stability. There was little effect on dynamic stability measures with the added presence of a VRT task. The effect on dynamic stability observed when using haptic modalities may be associated with the arm position and the

lack of arm swing. Overall, these findings suggest haptic modalities may require similar attentional demands to baseline walking and that adding a VRT when using a haptic modality does not affect walking behaviour. Dynamic stability might be affected with modality use as indicated by changes in outcome measures related to stability and walking when the haptic modalities were used during walking.

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DEDICATION

I would like to dedicate this thesis to my parents for all their unconditional love and support of me to pursue my dreams.

TABLE OF CONTENTS

PERMISSION TO USE i

ABSTRACT ii

ACKNOWLEDGEMENTS iv

DEDICATION v

LIST OF TABLES viii

LIST OF FIGURES ix

LIST OF EQUATIONS xi

GLOSSARY xii

ABBREVIATIONS xiii

1. Introduction: 1

1.1. Walking Stability 1
1.1.1. Importance for Fall Prevention 1
1.1.2. Mechanical Requirements of Dynamic Stability 1
1.1.3. Sensory Contributions 2
1.1.4. Assessment of Dynamic Stability 5
1.2. Haptic Information and Effects on Static and Dynamic Stability 6
1.2.1. Addition of Haptic Input 6
1.2.2. Effects of Haptic Input on Standing Balance 7
1.2.3. Effects of Haptic Input on Dynamic Stability 9
1.3. Attentional Demands of Walking 10
1.3.1. Importance for Assessment and Awareness 10
1.3.2. Models of Attention 11
1.3.3 Assessment of Attentional Demands 12
1.4. Attentional Demands of Modality Use during Standing and Walking 16
1.5. Objectives and Hypotheses of Thesis 17

2. Methods 19

2.1. Participant Eligibility and Recruitment 19
2.2. Data Collection 19
2.2.1. Anthropometric Measures and Limb Dominance 19
2.2.2. Kinematics Data 19
2.2.3. Haptic Railing and Anchors 21
2.2.4. Kinetics Data and Verbal Reaction Times 21

2.3. Standing Calibration **22**
2.4. Walking Protocol **23**
2.5. Data Processing and Analysis **24**
2.5.1. Verbal Reaction Times 24
2.5.2. Kinematic Data – Standing Calibration 25
2.5.3. Kinematics Data – Walking Trials 26
2.6. Statistical Analysis **28**

3. Results **30**

3.1 – Attentional Demands of Haptic Modalities **30**
3.2 – Measures of Dynamic Stability with Use of Haptic Modalities and VRT **32**
3.2.1. Light touch on a railing 32
3.2.2. Anchors 33

4. Discussion **36**

4.1. Haptic Modalities and Effects on Attentional Demands **36**
4.2. Haptic Modalities and Effects on Dynamic Stability **39**
4.3. Limitations **45**
4.4. Future Directions **46**
4.5. Conclusion and Clinical Implications **47**

5. References **49**

Appendix A **60**

Appendix B **62**

Appendix C **64**

Appendix D **67**

Appendix E **77**

Appendix F **81**

LIST OF TABLES

Table 1.1. – Summary of Attention Models (Kahneman, 1973)

Table 2.1. – Marker Placement for Kinematic Data

Table 2.2. – Summary of Conditions

Table 2.3. – Summary of Dynamic Stability Measures

Table 3.1. – Summary of Findings for Dynamic Stability Measures

LIST OF FIGURES

Figure 1.1. – “Postural actions emerge from an interaction of the individual, the task with its inherent postural demands, and the environmental constraints on postural actions” *Figure from Motor Control: Translating Research Into Clinical Practice, 4th Edition by Shumway-Cook and Woollacott reproduced with the permission from Wolters Kluwer Health (Appendix A)

Figure 1.2. – Examples of an individual using haptic modalities: 1) Light touch (LT) on a rigid railing (Left) 2) Haptic anchors with square boxes around the anchors on the floor (Right)

Figure 2.1. – Configuration of force plates in lab, where grey ovals represent footfalls and ‘X’ indicates where VRT task is administered during walking trial with respect to participant’s starting position

Figure 2.2. – Representative Kinetic Data to Display Where VRT Task Was Administered during Walking Trial (Shown where black line meets graph and X)

Figure 2.3. – Representative Trial of Participant Performing Verbal Reaction Time Task and Depicting How Verbal Reaction Time is Determined

Figure 3.1. – Comparison of mean (± SE) verbal reaction times (s) between seated and baseline walking condition (Significant differences between ends of each respective line)

Figure 3.2. – Comparison of mean (± SE) verbal reaction times (s) between walking conditions

Figure Appendix E.1. – Posterior view of kinematic marker set up
* Indicates calibration marker

Figure Appendix E.2. – Anterior view of kinematic marker set up
* Indicates calibration marker

Figure Appendix E.3. – Lateral view of kinematic marker set up displaying foot cluster and markers
* Indicates calibration marker

Figure Appendix F.1. – View of proposed location for trunk cluster (grey square) (Adapted from Hedayat et al., 2017)

LIST OF EQUATIONS

Equation 3.1 – Extrapolated Centre of Mass (xCOM) position

Equation 3.2 – Medio-lateral Margin of Stability

Equation 3.3 – Normalization of Stride Velocity

GLOSSARY

Terms	Definitions
Anchors	- Two light weights of 125 grams each attached to strings that are pulled by an individual
Attention	- The allocation of cortical processes to successfully perform task relevant to the individual (Buschman & Kastner, 2015)
Base of support	- Area under a person and/or object from points of contact (e.g. feet) made with a supporting surface (e.g., ground)
Centre of mass	- A point in space where mass is distributed
Dual-task	- The act of performing two tasks at the same time
Dynamic stability	- Ability to control one's extrapolated centre of mass within a moving base of support
Extrapolated centre of mass	- A measure of centre of mass that takes into account the height and instantaneous velocity of centre of mass
Haptic input	- Tactile sensation received by cutaneous receptors in the skin due to changes in force
Light touch	- One newton or less of force applied by the index finger of the dominant hand
Margin of stability	- The distance from the extrapolated centre of mass position to a base of support boundary
Multi-task	- The act of performing three or more tasks at the same time
Verbal reaction time	- The duration from the onset of a cue (i.e. tone) to the onset of a response

ABBREVIATIONS

ANOVA: analysis of variance

AP: anterior-posterior

BOS: base of support

CNS: central nervous system

COM: centre of mass

COP: centre of pressure

DT: dual-task

FA: fast-adapting

LT: light touch

ML: medio-lateral

MOS: margin of stability

N: newton

RM: repeated measures

SA: slow-adapting

SD: standard deviation

SE: standard error

VRT: verbal reaction time

xCOM: extrapolated centre of mass

1. Introduction:

1.1. Walking Stability

1.1.1. Importance for Fall Prevention

Falls are defined as a "sudden and unintentional change in position resulting in an individual landing at a lower level such as on an object, the floor, or the ground, with or without injury" (Government of Canada, 2014). One in three older adults (65 years and older) will experience a fall at least once a year (Do, Chang, Kuran, & Thompson, 2015). Among older adults, fall-related injuries are the leading cause of hospitalization in Canada (Chang & Do, 2015; Government of Canada, 2014). In Canada, it is expected that by the year 2036, the older adult population will double. This increase may result in an added economic burden due to the costs of hospitalization and care needed from a significant increase in fall-related injuries (Chang & Do, 2015; Government of Canada, 2014). In older adults, a primary risk factor for increased falls is decreased stability during walking (Horak, Shupert, & Mirka, 1989), which can be improved through various fall prevention programming (American Geriatrics Society, 2010)

1.1.2. Mechanical Requirements of Dynamic Stability

Successful locomotion has three requirements: 1) the ability to generate a rhythmic stepping pattern through reciprocal flexor and extensor muscle activity; 2) the ability to maintain equilibrium by keeping the individual's centre of mass (COM) within the base of support (BOS), while resisting the force of gravity and other expected and unexpected forces to maintain the body upright in space; and 3) the ability to adapt locomotion according to the goals of the individual and/or surrounding environment (Grillner & Wallen, 1985; Morton & Bastian, 2004). The ability to maintain stability during walking has been identified as an important factor in fall-risk (Horak et al., 1989). A decline in walking function is related to an increase in fall-risk (Cesari *et al*., 2009). Walking plays a crucial role in daily life; therefore, factors that affect dynamic stability during walking require further investigation (Woollacott & Tang, 1997).

Maintaining stability during walking is complex. Standing balance is maintained by keeping the distribution of mass in space, known as COM, within the BOS, defined by the points of contact with a surface (i.e., left and right foot outline during standing

(Horak, Nashner, & Diener, 1990; Winter, 1995)). Walking is initiated by changing the pressure distribution under the feet, known as the centre of pressure (COP), to move the COM outside of the BOS limits (i.e., going in front of the feet). Stepping forward during walking allows the individual to 'catch' their COM by extending their BOS.

As outlined, walking requires a dynamic BOS to control the COM or the individual will fall (Winter, 1995). For progression of locomotion in the forward direction, COM may exceed the anterior and lateral limits of the BOS, requiring a step to change the BOS; (Winter, 1995) or a fall can occur (MacLellan & Patla, 2006; Young, Wilken, & Dingwell, 2012). Steady-state walking, therefore, requires a constantly changing BOS with subsequent steps to maintain forward progression and stability (Winter, 1995).

1.1.3. Sensory Contributions

Motor (musculo-skeletal components and neuro-muscular synergies), sensory (individual sensory systems, sensory strategies, and internal representations) and higher-level (adaptive mechanisms and anticipatory mechanisms) processes are integrated by the central nervous system (CNS) to maintain postural control (Shumway-Cook & Woollacott, 2007) (Figure 1.1). Sensory contributions from vestibular, visual, and somatosensory systems are used to aid mechanical requirements needed for successful locomotion (Horak *et al*., 1990). The integration of sensory input by the CNS assists in locomotion by maintaining equilibrium of an individual's COM within their BOS and adapting to the goals of the individual and/or surrounding environment.

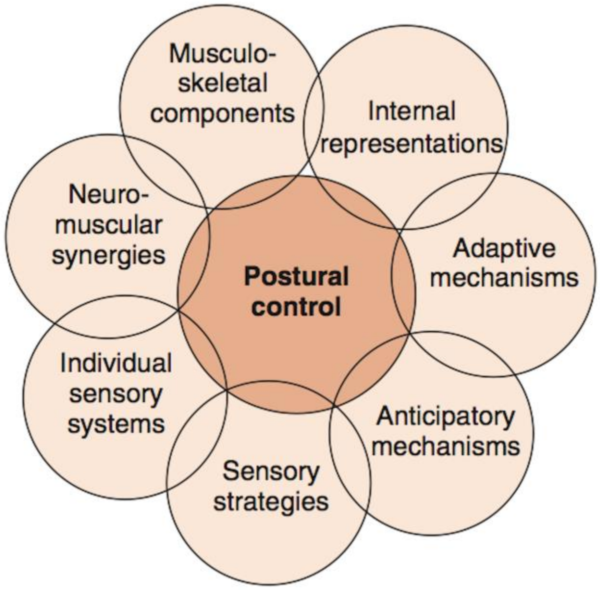

Figure 1.1. – "Postural actions emerge from an interaction of the individual, the task with its inherent postural demands, and the environmental constraints on postural actions" *Figure from Motor Control: Translating Research Into Clinical Practice, 4th Edition by Shumway-Cook and Woollacott reproduced with the permission from Wolters Kluwer Health (Appendix A)

Sensory input provided by vision contributes egocentric and allocentric information to the individual. Egocentric information provides the sense of where the body is relative to the external environment; allocentric information provides the sense of where objects within the environment are in relation to each other (Massion, 1994). Overall, egocentric and allocentric information provided from the visual system allows individuals to adapt to their environment in a proactive manner (Patla, 1997). Proactive mechanisms of the visual system dictate how individuals interact with their environment through avoidance and accommodation strategies. These strategies associated with the visual system include circumventing an obstacle within the path of travel or reducing gait velocity when encountering a slippery surface (Marigold & Patla, 2002; Patla, Prentice, Robinson, & Neufeld, 1991).

Sensory input provided by the vestibular system gives the individual egocentric information during locomotion by detecting linear and angular accelerations of the head (Massion, 1994). Sensing linear and angular head accelerations aids in specific reflexes, such as the vestibulo-occular reflex. This reflex integrates visual and vestibular information, allowing individuals to visually fixate on a point in space, while maintaining the ability to rotate their head (Laurutis & Robinson, 1986). In addition to specific reflexes, the vestibular system establishes a reference for the body's vertical orientation, with respect to where the head is oriented in space

(Bent, McFadyen, Merkley, Kennedy, & Inglis, 2000). In locomotion, vestibular input has been observed to be least used at mid-swing, while it is most used at initial heel contact (Bent, Inglis, & McFadyen, 2004). Vestibular input is most used at initial heel contact because a double support period is established, where the body is able to integrate vestibular and somatosensory information to create an accurate internal representation of the body in space (Bent *et al*., 2004).

The somatosensory system provides egocentric information, while also providing proprioception through cutaneous receptors (i.e., mechanoreceptors in the skin) (Jeka, Ribeiro, Oie, & Lackner, 1998). Within the skin of the foot and hand are dense populations of four different mechanoreceptors providing cutaneous sensation for touch (Vallbo & Johansson, 1984). These mechanoreceptors are grouped into two types: 1) fast-adapting (FA) and 2) slow-adapting (SA) determined by whether a discharge pattern is present when a force applied is constant (i.e., SA) or not (i.e., FA) (Vallbo & Johansson, 1984). Within these types, the mechanoreceptor is divided into types I and II, corresponding to the size of the receptive field, an area that can feel the sensation provided by the mechanoreceptor; therefore, an FAII receptor would be a fast-adapting mechanoreceptor, in which no discharge pattern is observed when the force applied remains constant with a large receptive field (Vallbo & Johansson, 1984).

Reduced input from the somatosensory system can result from diminished or loss of sensation (Van Deursen & Simoneau, 1999). Experimentally reducing cutaneous sensation through hypothermic anesthesia (i.e., cooling) of the feet shows the role of the somatosensory system during gait termination (Perry, Santos, & Patla, 2001). During gait termination, reduced cutaneous sensation affects feedforward information (e.g., increased foot placement variability) and feedback information (e.g., initiation of braking forces) (Perry *et al*., 2001). These forms of information are affected because the lack of cutaneous sensation impairs awareness about the foot's location in space (Perry *et al*., 2001). In contrast to gait termination, during steady state gait, reduced cutaneous sensation results in a more cautious behaviour characterized by decreased loading on the foot during stance and increased loading preceding toe off (Eils *et al*., 2004). Overall, the role of cutaneous sensation provided by the somatosensory system is important to locomotion because it provides the individual a perception of their BOS and plays a role in proprioception.

Proprioception involves perceiving the position and movement of one part of the body with respect to another, based on sensory input from muscles, joints, and skin (Goble, 2010;

Massion, 1994). Reductions in cutaneous sensation, experimentally induced via topical anesthesia, have been shown to affect proprioception (Mildren, Hare, & Bent, 2017). The effect is observed as an increased distance between limbs in a passive joint matching task (Mildren et al., 2017). The passive matching task requires an individual, with eyes closed to move one limb (i.e., left foot) to match the position of the other limb (i.e., right foot) moved by the researcher (Mildren et al., 2017). In older adults, loss of cutaneous sensation has been found to result in poor joint matching (i.e., greater measured distance between limbs) of both lower and upper limbs (Adamo, Martin, & Brown, 2007) These findings suggest the importance of cutaneous sensation to proprioception and its role in functional tasks (e.g., walking) (Adamo et al., 2007). The relationship between cutaneous sensation and proprioception suggests added sensory input via cutaneous sensation may provide an external frame of reference to aid an individual's standing and dynamic stability (further discussed in Section 1.3 and onwards).

1.1.4. Assessment of Dynamic Stability

Dynamic stability can be assessed with various parameters. Margin of stability (MOS) examines the distance of the COM from the boundaries of the BOS, typically during a static task, such as standing (Hof, Gazendam, & Sinke, 2005). To account for the velocity of COM during a motion task, such as walking, an extrapolated COM (xCOM) position is calculated. Changes in COM velocity influence the xCOM position, providing a better representation of balance control that may not be observed with COM position (Bruijn, Meijer, Beek, & van Dieen, 2013; Hof *et al*., 2005; Hof, 2008). MOS provides insight into how close the xCOM position comes to the boundaries of the BOS and; therefore, the dynamic stability of that individual. MOS values can indicate if dynamic stability is negatively affected (i.e., decrease in MOS), if there is a change in strategy to maintain stability from variables associated with MOS (i.e., decreasing walking velocity, reducing movement of the xCOM position and/or increasing the size of the BOS) or if a strategy/modality designed to improve stability is effective (i.e., increase in the MOS). The MOS is unique from other measures of dynamic stability, as it quantifies the distance between xCOM position and BOS that predicts instability of an individual at a specific instance in time or event.

As locomotion progresses, it is important to maintain stability by keeping the ML COM position within the lateral boundaries of the BOS (Young *et al*., 2012). Findings indicate aging results in increased ML COM displacement, which suggests decreased stability (Schrager, Kelly, Price, Ferrucci, & Shumway-Cook, 2008). Trunk mass represents a significant proportion (~ 40

%) of an individual's COM (Leva, 1996). Measuring trunk movement may represent how an individual's COM position moves to assess dynamic stability by providing insight regarding balance control, specifically in the ML direction (Gill et al., 2001). The inference regarding trunk and COM movement indicates that if increased ML COM displacement suggests decreased dynamic stability (Schrager et al., 2008), decreased ML trunk movement suggests improved dynamic stability (Mauerberg-deCastro et al., 2014).

Measures of ML COM variability have also been proposed to measure dynamic stability, as increased variability suggests decreased balance control (Maki, 1997). Furthermore, variability in gait parameters, such as both decreased and increased step width variability suggest decreases in dynamic stability (Brach, Berlin, VanSwearingen, Newman, & Studenski, 2005). Significant increases in step width variability is associated with decreased balance control, as the individual might lack the ability to compensate for instability, while a significant decrease in step width variability suggests an individual may not have the balance control to adapt their step width to maintain stability (Brach et al., 2005; Maki, 1997).

1.2. Haptic Information and Effects on Static and Dynamic Stability

1.2.1. Addition of Haptic Input

Somatosensory input as outlined above involves various proprioceptive and cutaneous receptors (Massion, 1994). Cutaneous receptors provide information via mechanoreceptors within the skin that sense changes in force (Turvey & Fonseca, 2014). Previous examples outlining somatosensory input examined the role of skin in the plantar surface of the foot; however, mechanoreceptors are included within the hand, which aids in tactile sensation (i.e., touch) in the form of haptic information (Johansson, Landstro, & Lundstro, 1982; Kennedy & Inglis, 2002). Skin on the hand utilizes haptic input for the purpose of proprioceptive and egocentric information (Turvey & Fonseca, 2014). Haptic information from hand contact with an external surface provides a frame of reference regarding where the individual's body segments are in space relative to the external surface (Lackner & DiZio, 2005).

Haptic input can be added in a variety of different forms during walking; however, the focus of this thesis will be placed on light touch (LT) and haptic anchors. LT is operationally defined as 1 Newton (N) or less of force applied by the index finger of the dominant hand to a rigid railing. This level of force provides the individual with haptic input without providing mechanical support (Holden, Ventura, & Lackner, 1994). Haptic anchors consist of two light weights, approximately 125 grams each, attached to a string (Mauerberg-Decastro, 2004). The individual holds one string in each hand and drags the light weights while they are on the ground (Figure 1.2)

Figure 1.2. – Examples of an individual using haptic modalities: 1) Light touch (LT) on a rigid railing (Left) 2) Haptic anchors with square boxes around the anchors on the floor (Right)

A pulling force of approximately 1.2 N is transduced through the string to each hand during standing (Mauerberg-Decastro, 2004) and approximately 0.6 N during walking (Unpublished Results). Differences between these two haptic modalities are apparent and might play important roles in how they improve stability: Anchors require a pulling force (Mauerberg-deCastro *et al*., 2014), while LT requires the use of a downward force . The effects LT and haptic anchors have during standing balance and dynamic stability will be examined in the following sections.

1.2.2. Effects of Haptic Input on Standing Balance

The effect of haptic input on standing balance has been well documented for both LT and anchors. One of the earlier studies investigating LT in young, healthy adults showed LT with eyes closed had similar postural sway, as measured by mean sway amplitude and path length compared to no touch with eyes open (Holden *et al*., 1994). Findings suggest sensory input from

LT provided a representation of where the individual was in space, similar to vision present (Holden *et al*., 1994). Follow-up studies in healthy, young adults displayed similar findings and provided insight into the underlying mechanisms of how LT reduced postural sway compared to no touch. Results showed increased muscle activity amplitude and a temporal relationship between the change in muscle activity and fingertip contact force suggesting a supra-spinal long-loop pathway (Jeka & Lackner, 1995). A subsequent study found another temporal relationship between changes in postural sway and force levels at the fingertip: Changes of force at the fingertip preceded changes in postural sway, suggesting a feed-forward mechanism (Rabin, Bortolami, DiZio, & Lackner, 1999). In older adult and clinical populations (i.e., individuals with bilateral vestibular loss, diabetic neuropathy and incomplete spinal cord injury) where sensory systems are compromised, postural sway is significantly reduced with LT (Arora, Musselman, Lanovaz, & Oates, 2017; Baccini et al., 2007; Dickstein, Peterka, & Horak, 2003; Lackner et al., 1999). These findings (Arora et al., 2017; Baccini et al., 2007; R Dickstein et al., 2003; Lackner et al., 1999) in clinical populations indicate sensory input provided by LT improves standing balance and can be used as an aid to compensate for loss of sensory input. Findings from Arora *et al*. further elucidated results related to LT use in an incomplete spinal cord injury population. Individuals with incomplete spinal cord injury that had significant impairments to lower extremity proprioception had significant improvements to standing balance (i.e., reduced COP variability) with eyes closed using LT. These findings suggest input from mechanoreceptors in the fingertip were sufficient to improve awareness in space, despite impairments in lower limb proprioception (Arora et al., 2017). Furthermore, it was observed that reduced upper limb cutaneous sensation diminished any improvements with LT, suggesting cutaneous sensation from the finger was necessary to improve standing balance (Arora *et al*., 2017).

In addition to LT on a rigid railing as a haptic modality, haptic anchors represent a non-rigid haptic tool that transmits haptic input through the pulling force created from light weights dragging on the ground. The role of haptic anchors during standing has not been as extensively researched when compared to LT. The first study utilizing haptic anchors found 125 grams per anchor were the optimal weight to improve standing balance in young, healthy adults (Mauerberg-Decastro, 2004). Follow-up studies indicated anchor use improved standing balance in older adults and individuals with cognitive impairments by reducing postural sway in challenging postural conditions (e.g., low/high balance beam, semi-tandem stance) (Mauerberg-deCastro *et al*., 2014).

1.2.3. Effects of Haptic Input on Dynamic Stability

Unlike haptic input during standing balance, the effects of haptic input on dynamic stability have not been extensively investigated. In an early study on walking with LT, participants walked on a treadmill at a pre-determined speed (0.83 m/s) with and without eyes open, while not touching, lightly touching, and heavily touching the rail of the treadmill (Dickstein & Laufer, 2004). Findings indicated LT was similar to vision in providing spatial orientation during treadmill walking and decreased ML COM variance, suggesting improved dynamic stability (Dickstein & Laufer, 2004). A more recent study had similar results, indicating that ML COM displacements decreased when walking with LT on a soft railing (Bingenheimer *et al*., 2015). Another study examining LT on a static and dynamic external surface found that LT on both surfaces reduced step width variability, suggesting improved dynamic stability (Kodesh, Falash, Sprecher, & Dickstein, 2015). In related research, ML trunk sway in young, healthy adults was much less in individuals using anchors than those using LT, indicating that haptic modalities, particularly anchors, improve dynamic stability (Hedayat, Moraes, Lanovaz, & Oates, 2017). Although the research has been sparse, the studies that have been done suggest that haptic modalities improve dynamic stability in healthy, young adults.

Recent studies have investigated haptic input on dynamic stability in populations with greater instability during walking, such as older adults (da Silva Costa, Manciopi, Mauerberg-deCastro, & Moraes, 2015) and individuals with Parkinson's disease (Rabin *et al*., 2015). Reduced ML body sway was found in individuals with Parkinson's disease during overground walking when examining the effects of a non-moving compared to a moving railing (Rabin *et al*., 2015). These findings suggest LT improved dynamic stability in a population with Parkinson’s disease (Rabin *et al*., 2015). Similarly, studies of anchor use during tandem walking in older adults observed reduced ML trunk acceleration compared to conditions without anchors, suggesting that anchors improve dynamic stability during a challenging walking scenario in an older adult population (da Silva Costa et al., 2015). Overall, then, similar findings among these studies (Bingenheimer et al., 2015; da Silva Costa et al., 2015; Dickstein & Laufer, 2004; Hedayat et al., 2017; Kodesh et al., 2015; Mauerberg-deCastro et al., 2014) suggest that haptic input improves dynamic stability in young adult, older adults, and selected clinical populations.

1.3. Attentional Demands of Walking

1.3.1. Importance for Assessment and Awareness

Distractors (e.g., technology) individuals are exposed to and use during walking have increased in recent years (Neale, Dingus, Klauer, Sudweeks, & Goodman, 2005). Researchers have observed that standing and walking require cortical input (Little & Woollacott, 2015; Woollacott & Shumway-Cook, 2002). Input required from cortical structures during standing and walking indicates the role of attention in these processes (Woollacott & Shumway-Cook, 2002). Attention in the context of this thesis will be defined as the allocation of cortical processes to successfully perform tasks relevant to the individual (Buschman & Kastner, 2015). How attention is allocated to one or more tasks is complex and warrants further investigation. Understanding these processes may provide insight into fall prevention by addressing how individuals attend to other tasks while maintaining balance (Horak, 2006).

Walking requires attention (Lajoie, Teasdale, Bard, & Fleury, 1996). Attention placed on another task during walking can reduce performance in one or both tasks. In young adults, the addition of another task (e.g., texting while walking) may affect performance in the primary task (e.g., walking) (Marone, Patel, Hurt, & Grabiner, 2014). Overall, the impact on the primary task may result in increased rates of accidents, commonly resulting in injury (Caird, Johnston, Willness, Asbridge, & Steel, 2014; Nasar & Troyer, 2013; Saltos, Smith, Schreiber, Lichenstein, & Lichenstein, 2015). The impact of the additional task(s) on the primary task relates to the complexity of both; however, whether attention allocated to another task is affecting the performance of the primary task and/or destabilizing the individual is not understood (Dingwell, Robb, Troy, & Grabiner, 2008).

The importance of understanding how attention is allocated is illustrated in older adult and clinical populations (e.g., Multiple Sclerosis, Parkinson’s Disease) (Vance, Healy, Galvin, & French, 2015; Wajda, Motl, & Sosnoff, 2013). Poor performance of two or more tasks have been linked to recurrent falls, lower attentional capacity, and an inability to allocate attention appropriately (Beauchet *et al*., 2008). As identified earlier, the older adult population is expected to increase by 2036 (Section 1.1.1.), and activities of daily living while walking involve dual tasking, such as talking or carrying objects. Understanding how attention is allocated and the potential impacts on dynamic stability is critical. Insights may provide an understanding to the increase in fall-risk observed in aging and/or pathology and the development of fall-prevention initiatives.

1.3.2. Models of Attention

Performance of two tasks simultaneously, referred to as a dual-task (DT), or performance of three or more tasks simultaneously, referred to as a multi-task, may result in interference between those tasks due the attention required to execute each task. To execute a single task, three components must be achieved: 1) Perception (obtain information appropriate to task from environment); 2) Response selection (selection of a suitable response to the information perceived regarding the task); and 3) Motor response (execution of response following appropriate selection) (Strobach, Liepelt, Pashler, Frensch, & Schubert, 2013; Worden, Mendes, Singh, & Vallis, 2016). During a DT and multi-task scenario, researchers have observed that interference arising between tasks occurs in the response selection component of executing a task (Strobach *et al*., 2013).

The capacity interference model (attention models summarized in Table 1.1) views attention as finite and examines attentional demands associated with non-specific stimuli (ex. non-similar tasks) (Abernethy, 1988). In the capacity interference model, decreases in performance occur when capacity is reached/exceeded: Both tasks become mutually interfering resulting in decreased performance in one or both tasks (Kahneman, 1973). In this model, interference between tasks occurs at the response selection stage because the participant has to determine the sequence to perform the tasks, which can occur in concurrent tasks (i.e., counting backwards while walking) and discrete tasks (i.e., reaction time task when walking) (Worden *et al*., 2016).

Structural interference is an attention model that examines interference of tasks performed at the same time that require the same cortical structure (Abernethy, 1988; Wright & Kemp, 1992). Decreased performance in one or more of the tasks is expected, if a common cortical structure is required in the execution of the tasks (ex. naming pictures on a screen while having to step on a new tile that lights up compared to naming pictures on a screen alone – there will be decreased performance in one or both tasks during the DT scenario, as both tasks require visual stimuli that need to be perceived and responded to) (Kahneman, 1973). Structural interference results in greater decrement to performance in one or more tasks compared to capacity interference. The greater decrease in task performance is associated with both perception and response selection components affected (Marteniuk, 1986; Temprado, Zanone, Monno, & Laurent, 2001).

Table 1.1. – Summary of Attention Models (Kahneman, 1973)

Model:	**Capacity Interference**	**Structural Interference**
Form of Stimuli (Cortical Structure Used):	Different	Same
Phase of Task Execution Where Interference Occurs:	1) Response Selection	1) Perception 2) Response Selection

1.3.3 Assessment of Attentional Demands

The framework for attention developed within cognitive psychology allows research in biomechanics to understand how the allocation of attention may affect motor tasks, such as standing and walking (Lajoie *et al*., 1996). Understanding attention in static and dynamic postural tasks is accomplished by DT and multi-task paradigms. In DT and multi-task paradigms, tasks are added in sequence to observe how they interfere with each other. The amount of interference can be assessed by adding another task, either concurrent or discrete, and measuring performance in one or all of the tasks (Al-Yahya *et al*., 2011).

Concurrent tasks range in complexity, such as counting backwards by 1s to an n-back task, where an individual is required to remember a series of numbers and stop when a number is repeated (ex. for a 2-back task in the number sequence **3**, 5, 2, **3**, the individual would say stop when they hear 3, as it was said 2 positions before) (Schaefer, Schellenbach, Lindenberger, & Woollacott, 2015). A discrete task is administered at a specific point in time and commonly referred to as a probe reaction time task (Lajoie *et al*., 1996). The probe reaction time is measured from the onset of the cue to the onset of an individual’s response (Abernethy, 1988; Kahneman, 1973). The role of the additional task is based on the capacity interference model; therefore, each subsequent task added results in affected performance of the primary task and/or the additional task(s) (Woollacott & Shumway-Cook, 2002; Yogev‐Seligmann, Hausdorff, & Giladi, 2012).

Effects of concurrent and discrete DTs in biomechanics were first examined during quiet standing of individuals. One of the earliest studies (Kerr, Condon, & McDonald, 1985) examining different secondary tasks while maintaining standing balance in a tandem stance (primary task) had young adults perform conditions involving a Brooks' spatial or non-spatial memory task (secondary task). These memory tasks involved verbally organizing numbers (i.e., spatial) or pairs of words and numbers (i.e., non-spatial) in a 4 x 4 matrix (Brooks, 1967; Kerr et al., 1985). Results showed increased difficulty in the secondary task had greater interference with the primary task (i.e., standing) because error rates in the memory task increased while balance was maintained (Kerr et al., 1985). These findings suggest posture and spatial memory share neural mechanisms (Kerr et al., 1985).

A follow-up to Kerr et al. (1985) examined differences between young and older adults with different standing tasks (primary task) but the same secondary task (i.e., pushing a button) (Teasdale, Bard, LaRue, & Fleury, 1993). When the postural task became more challenging, reaction time to push a button increased, with the greatest changes (i.e., increases) in the older adult group (Teasdale et al., 1993). These findings indicate attention was allocated to the postural task to maintain balance, affecting the ability to perform the secondary task (i.e., pushing button) (Teasdale et al., 1993). Furthermore, findings suggest older adults have a lowered attention capacity compared to young adults because reaction times were longer (Teasdale et al., 1993). A subsequent study expanded on the previous findings by examining attentional demands, measured via verbal reaction time (VRT), in young and older adults during different standing postures and walking (Lajoie et al., 1996). Lajoie et al. (1996) found increasing the complexity of the postural task resulted in increased VRT. In addition, walking had the longest VRT in both groups, with the largest effect in older adults (Lajoie et al., 1996). Overall, findings were similar to Teasdale et al. (1993) and suggests walking requires greater attentional demands compared to standing (Lajoie et al., 1996).

1.3.3.1. Concurrent Secondary Tasks

Concurrent secondary tasks can be performed on a treadmill (Lövdén, Schellenbach, Grossman-Hutter, Krüger, & Lindenberger, 2005) or overground (Harley, Wilkie, & Wann, 2009). Performance of the primary task (i.e., walking) is observed with and without the secondary task (Harley et al., 2009; Lövdén et al., 2005). In treadmill walking studies, speed of the treadmill might be fixed to observe changes in attentional demands with different secondary tasks and/or assess changes between walking with and without a secondary task (Grabiner & Troy, 2005). Maintaining the same treadmill speed with the concurrent task provides researchers the ability to assess how the individual compensates for the addition of the secondary task (Dingwell *et al*., 2008; Grabiner & Troy, 2005; Maki, 1997).

Concurrent secondary tasks are easily administered during overground walking within clinical settings to assess how an individual integrates another task into their walking (Campbell, Rowse, Ciol, & Shumway-Cook, 2003). Concurrent secondary tasks are commonly used in fall-risk assessment tests (e.g., Timed Up and Go (TUG) Test) in older adult and clinical populations (Campbell *et al*., 2003; Pettersson, Olsson, & Wahlund, 2005). A concurrent task can be administered in a variety of ways, such as the use of a motor or verbal task. A motor task has an individual balancing (i.e., a tray) or manipulating (i.e., doing up buttons on a shirt) an object, while a verbal task involves a person generating answers (i.e., counting backward, naming objects, etc.) (Al-Yahya *et al*., 2011). The addition of a secondary task affects the individual's walking performance and may reduce gait velocity (Al-Yahya et al., 2011). Significant increases in time to complete the TUG test with a DT (cognitive or motor) suggests a lower attentional capacity (Pettersson et al., 2005; Shumway-Cook, Brauer, & Woollacott, 2000). Increased time to complete the TUG test is linked to increased fall-risk in older adults due to impairments in functional mobility during everyday encounters with cognitive tasks (Beauchet *et al*., 2009).

1.3.3.2. Discrete Secondary Tasks

Discrete secondary tasks are used to assess attentional demands of one or more tasks (Abernethy, 1988; Kahneman, 1973). The task can be administered through use of a button or a verbal response to a tone, word, or event (Lajoie et al., 1996; Vuillerme, Isableu, & Nougier, 2006). Difficulty of tasks can vary from simply saying a word when a tone is heard to distinguishing a high or low-pitched voice, as the voice says either "high" or "low". Usually, the VRT task is administered as a single-task to determine baseline VRT, then with the addition of other tasks (Siu, Catena, Chou, van Donkelaar, & Woollacott, 2008). VRT increases proportionally to the amount of attentional demand required by each subsequent task added (Abernethy, 1988). This trend suggests, as the number of tasks (i.e., stimuli) increase (increased attentional demands), VRT will increase because less attention will be available to allocate to the VRT task due to greater interference among tasks.

Walking is observed to have increased attentional demands compared to sitting (Lajoie *et* *al*., 1996; Siu *et al*., 2008a). As outlined by the capacity interference model, delay of response selection and corresponding output results when attentional capacity is reached, commonly referred to as cognitive-motor interference (Al-Yahya *et al*., 2011; Springer *et al*., 2006; Woollacott & Shumway-Cook, 2002). A VRT task provides insight into how attention is allocated when an individual is presented with one or more task(s) during walking (Abernethy, 1988). Brown, McKenzie, & Doan (2005) examined attentional demands using a VRT task (saying "top" when hearing a buzzer) during walking and obstacle crossing in young and older adults. Results indicate older adults had longer VRTs during walking compared to young adults, suggesting increased attentional demands in the older group. Furthermore, prior to obstacle crossing both age groups had significantly increased VRTs compared to no obstacle present (Brown, McKenzie, & Doan, 2005). These results suggest both age groups placed greater attention on the postural threat (i.e., obstacle) than the auditory task (Brown et al., 2005). Overall, these findings (Brown et al., 2005) were consistent with previous research (Chen *et al*., 1996; Lajoie *et al*., 1996; Siu, Chou, Mayr, Donkelaar, & Woollacott, 2008).

1.3.3.3 Posture-First Response

An emerging theme in attention and walking literature is as postural threat increases, attentional demands required increase (M. Woollacott & Shumway-Cook, 2002). The increase in attentional demands is observed by increases in reaction times (i.e., discrete cognitive task) or decreases in walking velocity (i.e., concurrent cognitive task) to allocate attention to the postural threat and maintain dynamic stability (Harley *et al*., 2009; Lajoie *et al*., 1996; Marone *et al*., 2014; Schaefer *et al*., 2015; Siu *et al*., 2008a; Springer *et al*., 2006; Woollacott & Shumway-Cook, 2002; Yogev‐Seligmann *et al*., 2012). The posture-first response in DT and multi-task research within biomechanics suggests an individual presented with other tasks while walking, such as motor and/or cognitive will focus on maintaining their balance rather than the other tasks (Woollacott & Shumway-Cook, 2002; Yogev‐Seligmann *et al*., 2012).

The posture-first response is observed in both concurrent and discrete secondary task studies. Harley et al (2009) investigated young and older adults walking a track with and without an obstacle present while saying words related to a category (e.g. concurrent secondary task). Results showed young adults decreased the number of words and increased foot clearance of the obstacle to avoid contact (Harley *et al*., 2009). These findings indicate a posture-first response because attention was reallocated from the word task to focus on the obstacle by increasing toe clearance to avoid contacting the obstacle (Harley *et al*., 2009). Results with older adults showed increased number of words generated and decreased toe obstacle clearance (Harley *et al*., 2009). These findings suggest older adults may not implement a posture-first response because attention is shifted towards the word task resulting in decreased obstacle clearance (Harley *et al*., 2009). Other studies using concurrent secondary tasks found similar results in young adults suggesting a posture-first response is maintained during walking with additional tasks (Marone *et al*., 2014; Schaefer *et al*., 2015). Interestingly, Marone *et al.* (2014) found participants texting and walking had a significant reduction in gait velocity with a significantly increased MOS. Decreased gait velocity may occur due to structural interference because vision is utilized to guide the participant walking and texting (Marone et al., 2014; Woollacott & Shumway-Cook, 2002). Alternately, significant decreases in gait velocity may be a proactive strategy to improve dynamic stability (Marone *et al*., 2014).

Discrete reaction time tasks have similar effects to concurrent tasks, indicating that these too have a posture-first response (Woollacott & Shumway-Cook, 2002). Also, discrete reaction time tasks allow researchers to investigate how individuals reallocate their attention based on different instructions. In one study, young adults performed obstacle crossing with a VRT task. Participants were instructed, according to three different conditions: first, to focus equally on both tasks; second, to focus on the obstacle; and third, to focus on the VRT task (Siu et al., 2008b). Results showed that participants had similar VRTs when focusing equally on both tasks and when prioritizing the obstacle. Interestingly, when explicitly told to focus on the VRT task, young adults can flexibly reallocate attention to the VRT task and significantly decrease reaction time (Siu et al., 2008b). Overall, these findings indicate that young adults prioritize allocating attention to a postural threat and support the posture-first response (Siu et al., 2008a; Siu et al., 2008b; Woollacott & Shumway-Cook, 2002).

1.4. Attentional Demands of Modality Use during Standing and Walking

The use of modalities, such as walkers and canes is common in older adult and clinical populations to assist with stability for fall-prevention purposes (Edelstein, 2013). These modalities can vary in design. Various designs require different attentional demands (Wright & Kemp, 1992). Standard walkers require more attentional demand because VRT increased compared to rolling walkers in healthy, young (Wright & Kemp, 1992) and older adults (Wellmon, Pezzillo, Eichhorn, Lockhart, & Morris, 2006). These results can be attributed to the accuracy demand differences between modalities because more attention is required to lift, move forward, and accurately place a standard walker compared to a rolling walker (Wright & Kemp, 1992). Attention required by a walking aid may pose a fall-risk to individuals with reduced attentional capacity (Bateni & Maki, 2005).

LT has been shown to require attention and act as a DT during quiet standing (Vuillerme *et al*., 2006). The attentional demand associated with LT could be a result of the accuracy needed, as light contact must be maintained on a surface is that is often very small. It has been suggested that improved standing balance (i.e., decreased postural sway) with LT results from attention being placed on the task (Riley, Stoffregen, Grocki, & Turvey, 1999). When individuals used LT without directing attention towards touching the external surface (i.e., being told to stand in a different position where their fingers coincidentally made contact with the external surface) during quiet standing, balance (i.e., postural sway) was unchanged (Riley *et al*., 1999).

DT scenarios can be classified as competitive or non-competitive based on the attentional demand required and interference created with the primary task (i.e., gait). A competitive DT results in interference with the primary task (i.e., gait), as observed in young adults walking with LT or anchors because stride velocity significantly decreases (Hedayat et al., 2017). Alternatively, reductions in stride velocity in Hedayat *et al*. (2017) might be a result of the novelty using LT and anchors with minimal familiarization by the participants, which can affect walking behaviour. It has been suggested that anchors are non-competitive, indicating minimal to no attentional demand requirements (Mauerberg-deCastro *et al*., 2014). This suggests attentional demands of the anchors will not reach attentional capacity that results in interference affecting gait (Mauerberg-deCastro *et al*., 2014).

1.5. Objectives and Hypotheses of Thesis

The **primary objective** of this thesis was to investigate the attentional demands with a VRT task when using either LT on a railing or haptic anchors during walking compared to baseline walking (i.e., walking with no modality).

For the primary objective, it is hypothesized LT will require more attention (i.e., significantly longer VRT) compared to anchor use and baseline walking. Furthermore, it is hypothesized anchors will have similar (i.e., not significantly different) VRTs when compared to baseline walking.

The **secondary objective** was to investigate the effect on dynamic stability of an added VRT task in the multi-task scenario created using either LT on a railing or the haptic anchors compared to baseline walking.

For the secondary objective, it is hypothesized dynamic stability will be increased when using haptic modalities compared to baseline walking, with the greatest increase observed in haptic anchors.

2. Methods

2.1. Participant Eligibility and Recruitment

For participant eligibility, the inclusion criteria were individuals who were 18 – 30 years old and able to walk independently for at least 10 metres. Exclusion criteria included: Individuals with visual and sensory impairments (i.e., neuropathies, hearing impairments, etc.), musculoskeletal (i.e., current/past broken bones, torn ligaments, etc.) and neurological injuries (i.e. concussion, stroke, etc.). Recruitment occurred at the University of Saskatchewan through word of mouth, classroom announcements, and postings on the campus-wide posting board. The experimental procedures and recruitment were approved by the University of Saskatchewan Research Ethics Board (Appendix B).

2.2. Data Collection

Interested participants completed a screening questionnaire administered over the phone or email to ensure eligibility (Appendix C). If eligible, participants were scheduled for data collection on a date and time of their choosing. Upon entering the lab, participants reviewed the consent form (Appendix D) and were allowed to ask any questions/concerns they may have had prior to or after giving informed consent. After informed consent, participants changed into shorts, t-shirt, and their own comfortable pair of walking shoes (i.e., running shoes).

2.2.1. Anthropometric Measures and Limb Dominance

Participant height was measured using a stadiometer to the nearest 0.1 cm. Participants had mass measured using a digital scale to the nearest 0.1 kg. For both height and mass measurements, participants left the shoes they were wearing for the study on. Hand dominance was self-reported by the participant. The dominant leg of the participant was determined by asking which leg they preferred when stepping up onto a platform (Coren & Porac, 1978). The dominant leg was measured from the greater trochanter to the floor to determine leg length (Hof, 1996).

2.2.2. Kinematics Data

Kinematics data were captured using an eight-camera 3D motion capture system (VICON

Nexus, Centennial, CO) sampled at 100 Hz. The motion capture system was calibrated to an error of 0.20 mm or less, the day of data collection prior to the participant's arrival. The marker set used was modified from a previous COM model (Tisserand, Robert, Dumas, & Cheze, 2016) to create a 12 segment COM model. The segments for the COM model included: 1) Head, 2) Trunk, 3) and 4) Upper Arm (Left and Right), 5) and 6) Lower Arm (Left and Right), 7) and 8) Thigh (Left and Right), 9) and 10) Shank (Left and Right), 11) and 12) Foot (Left and Right). The marker set consisted of 34 passive reflective markers attached to specific anatomical landmarks on the body, eight of which were used as virtual markers. Locations of these markers are summarized in Table 2.1 and figures in Appendix E. Infrared cameras tracked the position of these markers in space.

Table 2.1. – Marker Placement for Kinematic Data

Body Part/Region	**Landmark**
Head	Centre of Forehead
	Left Part of Head above Ear (level of forehead marker)
	Right Part of Head Above Ear (level of forehead marker)
Shoulders	Left and Right Acromioclavicular Joint
Elbows	Left and Right Lateral Epicondyle of the Humerus
Wrists	Left and Right Styloid Process of Ulna
Finger	Dominant Index Finger
Pelvis Cluster (used to retain positions of calibration (virtual) markers necessary to create hip joint centres)	4 markers on a rigid rectangular cluster held around the participant's pelvis using a belt
Hip Markers (calibration only)	Left and Right Anterior Superior Iliac Spine
	Left and Right Posterior Superior Iliac Spine
Knee	Left and Right Lateral Femoral Epicondyles
Ankle	Left and Right Lateral Malleoli
Heel	Left and Right Calcaneus (Heel) of foot

Foot Clusters (used to retain position of calibration markers on foot to create base of support and longitudinal axis of foot)	3 Markers Placed on Lateral Side of Left and Right Foot non-collinearly
Anterior Boundary of Base of Support (calibration markers)	Most anterior point on left and right shoe (tip of first distal phalanx)
Lateral Boundary of Base of Support (calibration markers)	Most lateral point on left and right shoe (lateral point on the fifth metatarsal)

2.2.3. Haptic Railing and Anchors

A railing instrumented with load sensors (Futek Advance Sensory Technology, Inc., CA, USA; Range: 0 – 5 N) provided participants an external rigid surface to lightly touch with the index finger of their dominant hand during walking. The railing was 85 cm in height and has been used in previous studies (Arora et al., 2017; Hedayat et al., 2017). Load sensors allowed researchers to monitor the force applied and ensure it was less than 1 N (Dickstein *et al*., 2003; Hedayat *et al*., 2017). For haptic anchors, two light weights (each weighing 125 g) were attached to a string (Hedayat et al., 2017; Mauerberg-deCastro et al., 2014), which created a pulling force of approximately 0.6 N (Unpublished Results).

2.2.4. Kinetics Data and Verbal Reaction Times

Kinetics data were collected by two force plates (AMTI, Watertown, MA) embedded within ground that were sampled at 2000 Hz. These force plates were located at the midway point of the walkway to record ground reaction forces during walking. The force plate closest to the participant when starting a trial triggered a VRT task (described in section 2.4). The VRT task was administered by a speaker with a standardized volume (i.e. maximum volume settings) located at the side of the walkway when 10% of the participant's body weight was reached (Figure 2.1 and 2.2 provide layout of force plates in ground and force graph of where VRT was initiated based on starting position, respectively). Tones and responses to the VRT tasks were captured using a wireless voice recorder (Philips Voice Tracer 2000) sampled at 44,100 Hz.

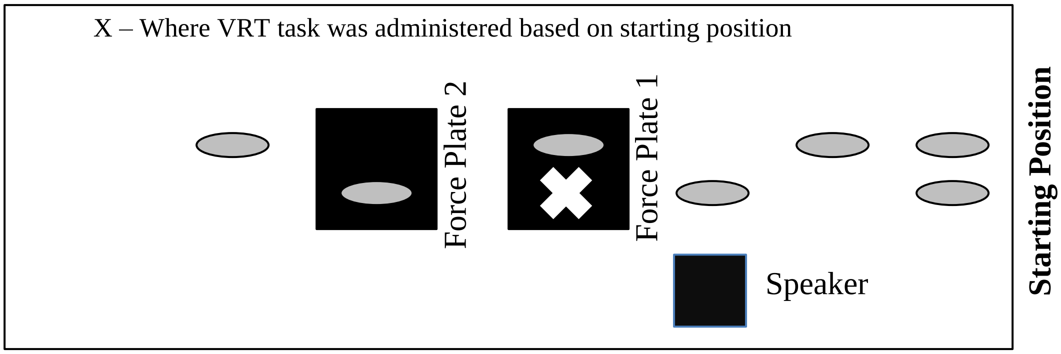

Figure 2.1. – Configuration of force plates in lab, where grey ovals represent footfalls and ‘X’ indicates where VRT task is administered during walking trial with respect to participant’s starting position

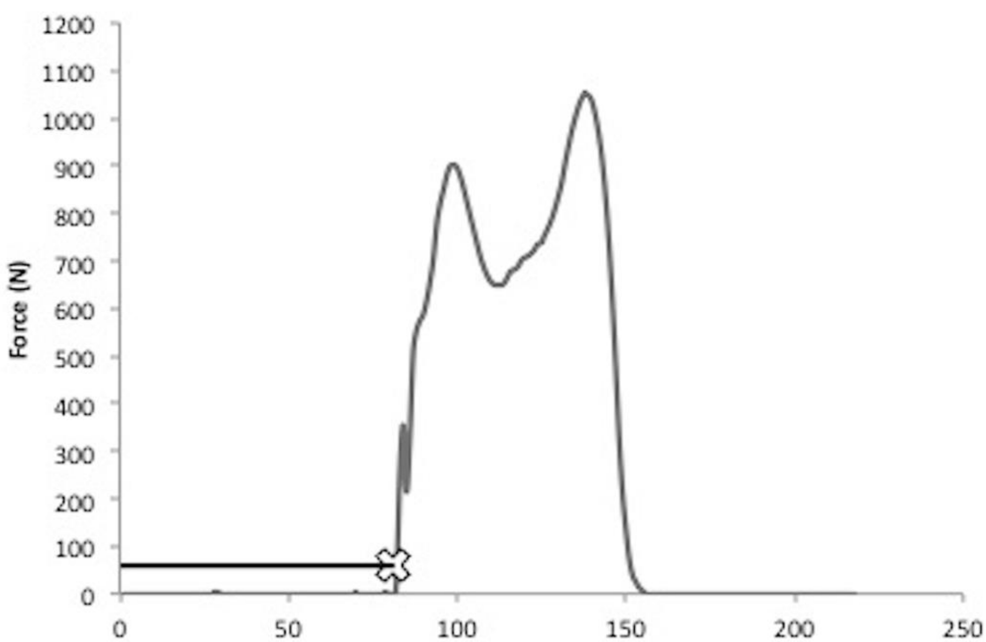

Figure 2.2. – Representative kinetic data to display where VRT task was administered during walking trial (Shown where black line meets graph and X).

2.3. Standing Calibration

Before seated VRT tasks and walking protocol could be conducted, the participant had to undergo a standing calibration. Standing calibration involved having all passive markers (including calibration) on the body of the participant and in the correct locations. When all markers were secured and in the right position, the participant was instructed to step onto a wooden platform located in the view of all cameras. Heels of the participant were in contact with the back of the platform, while medial sides of the feet were in contact with a ridge located in the middle of the platform that separated the feet by 20 cm. When positioned on the wooden platform, participants were instructed to raise their arms by their sides (i.e., abduction), keep their elbows slightly flexed and hands parallel to the ground. Calibration markers were checked to ensure no movement from original locations and participants stood still (i.e., static) for 2 – 3 seconds, while a standing trial was recorded. Following a successful trial, participant's data were calibrated and calibration markers were removed. A subsequent standing trial for 60 seconds was recorded, as the participant stood as still as possible with their arms by their sides to obtain a COM position for the MOS calculations detailed in section 2.6.

2.4. Walking Protocol

Prior to the start of collection for each block, participants practiced using the haptic modality of that block, with and without VRT task (~ 10 – 20 trials total per block). This familiarization period ensured participants felt comfortable with both modalities and the VRT task. Furthermore, the familiarization provided the opportunity to fine tune the starting position for each participant to ensure the VRT task was triggered approximately at heel strike on a force plate.

For the VRT task used to assess attentional demands, participants distinguished if a tone (~ 200 milliseconds in duration) was high (1000 Hz) or low (200 Hz) frequency and responded aloud with the answer of either "high" or "low". Participants' seated baseline VRTs were determined prior to walking conditions to determine their non-walking VRTs. Participants were seated, had their arms placed by their sides, not touching the chair and had their feet in contact with the floor. Participants performed eight seated randomized VRT task trials that were administered by the researcher at randomized intervals at least three seconds apart. Seated VRTs were compared to their walking with no modality VRTs to determine if participants were prioritizing the VRT task when walking.

Following seated baseline, to investigate the attentional demands required by LT and anchors, participants walked under three conditions (16 trials per walking condition), with or without a VRT task: 1) baseline walking (no haptic input); 2) lightly touching the railing (LT); and 3) dragging the anchors. The VRT task occurred during 50% of walking trials to ensure the cognitive task was unexpected (Wright & Kemp, 1992). The VRT task occurred at approximately the midway point of the walkway, upon foot contact with a force plate, to ensure the participant was in steady-state gait. Participants were instructed to "walk at a comfortable pace and try to keep equal focus on all tasks." The order of trials was blocked by modality (2 blocks) with the starting block (i.e., LT or anchors) counterbalanced. Trial order was randomized within each block. Half of the total number of baseline walking trials (n = 8), with and without VRT tasks, were in each block. (Table 2.2 provides a summary of conditions and number of trials).

Table 2.2. – Summary of Conditions

Seated (Secondary Task only)	**Anchors Block**	**Light Touch Block**
VRT Only (8 trials)	Baseline (4 trials)	Baseline (4 trials)
	Baseline – VRT (4 trials)	Baseline – VRT (4 trials)
	Anchors (8 trials)	LT (8 trials)
	Anchors – VRT (8 trials)	LT – VRT (8 trials)

2.5. Data Processing and Analysis

2.5.1. Verbal Reaction Times

Audio data was cropped into individual VRT trials using Audacity 2.1.2 (Audacity Team, Pittsburg, Pennsylvania) and processed using Praat v6.0.19 (Boersma and Weenink, Amsterdam, Netherlands). VRTs were defined from the onset of the tone to the onset of the participant's response (Figure 2.3 provides an example of how VRT was determined) and trials were listened to with the aid of spectrograms and formants to ensure onsets were correctly identified (Gould, Cummine, & Borowsky, 2012).

Intra-rater reliability and percent agreement were assessed using reanalyzed VRTs from five randomly selected participants compared to their original VRT values. These analyses were performed to determine the accuracy and reliability of the method used by the researcher. A two-way mixed effects intraclass correlation coefficient analysis revealed the intraclass correlation was 0.996 (F (1,123) = 490.822, $p < 0.001$). The percent agreement calculated was approximately 99 %. These values are in line with the reliability of visually inspecting VRTs, which had a percent agreement of 96 % between two researchers in a previous study (Worden & Vallis, 2014).

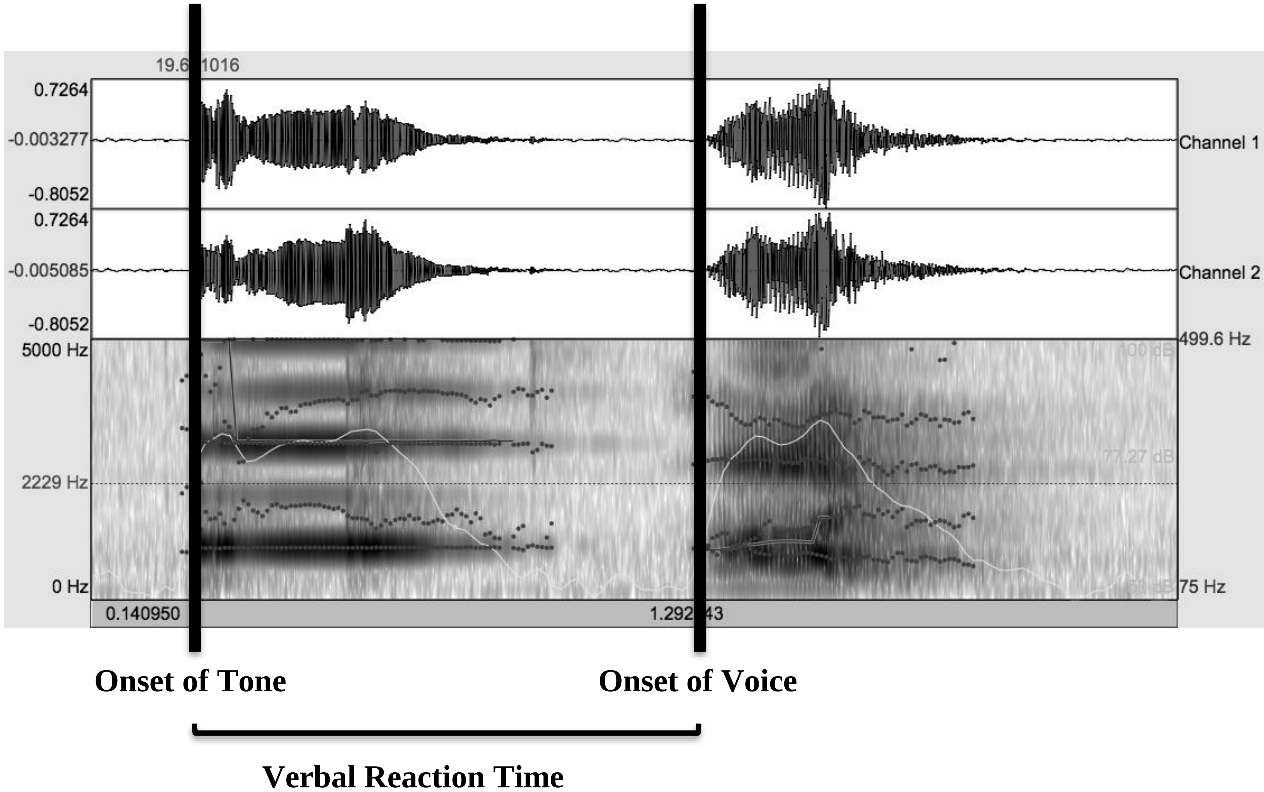

Figure 2.3. – Representative Trial of Participant Performing Verbal Reaction Time Task and Depicting How Verbal Reaction Time is Determined

2.5.2. Kinematic Data – Standing Calibration

The standing calibration trial was processed in VICON Nexus 2.3 (Vicon Motion Systems, Centennial, CO), which was used to save the positions of the calibration markers. Markers on the participant's left and right anterior superior iliac spine and posterior superior iliac spine were represented in the coordinate system of the pelvis cluster that remained on the participant for the remainder of data collection. Similarly, calibration markers placed on the foot were represented in the coordinate system of the respective foot cluster that remained on the participant. The positions of these markers were expressed in the global coordinate system from their respective coordinate systems. This allowed for calculation of hip joint centres using the positions of where the calibration markers would be on the pelvis (Harrington, Zavatsky, Lawson, Yuan, & Theologis, 2007) and determining BOS using the position of where the calibration markers would be on the foot. Hip joint centres allowed for a better estimation of the upper leg segment COM location and length (Leva, 1996) during standing and walking. The BOS was calculated during walking to be used for the MOS calculation.

2.5.3. Kinematics Data – Walking Trials

Kinematics data for walking trials were processed in VICON Nexus 2.3 (Vicon Motion Systems, Centennial, CO). Kinematics data were used to calculate variables related to dynamic stability and walking (Summarized in Table 2.3) using custom Matlab (R2006b for PC, MathWorks, Natick, MA) routines. Variables were calculated during the stride where the VRT task was triggered to ensure steady state gait and determine changes that could occur from administration of the VRT task. Heel contact and toe-off events were determined using velocity-based thresholds from a percentage of walking speed using the velocity of the toe marker in the sagittal plane (Bruening & Ridge, 2014).

Whole-body COM was calculated from kinematics data using a 12-segment COM model (described in section 2.2.3.). Body segments were calculated using anthropometric tables (Leva, 1996). COM data were filtered with a 4^{th} order low pass Butterworth digital filter at 8 Hz before calculating the derivative of the COM position data to obtain COM velocities. For the calculation of xCOM position, the height of the participant's COM was calculated during the standing trial and used in the following equation (3.1) (Young et al., 2012):

$$xCOM\ Position = COM\ Position + \frac{Instantaneous\ COM\ Velocity}{\omega_0} \quad \text{.........(3.1)}$$

where, $\omega_0 = \sqrt{\frac{g}{\ell}}$

Gravity (g) = 9.81 m/s^2, ℓ = distance from COM to floor

One of the variables calculated associated with dynamic stability was the ML MOS. The lateral boundary of the individual's BOS was determined by obtaining the relative position of the calibration marker placed on the lateral side of the fifth metatarsal from the standing calibration trial. The ML xCOM position was subtracted from the right and left lateral boundaries of the BOS using the following equation (3.2) (Young et al., 2012):

$$ML\ MOS = Lateral\ Boundary\ of\ BOS - ML\ xCOM\ position \quad \text{.........(3.2)}$$

The minimum distances between the ML xCOM position and either left or right lateral BOS boundary were obtained across the stride and all minimum distances obtained were averaged.

Changes in dynamic stability when walking can predict the likelihood of instability for an individual (Bruijn et al., 2013; Worden & Vallis, 2015; Young et al., 2012). If dynamic stability is negatively affected, a decrease in ML MOS would be expected, while if dynamic stability improved, an increase in ML MOS is expected (Worden & Vallis, 2015; Young et al., 2012).

To understand changes in and/or how ML MOS is maintained in different conditions, the following variables were assessed: the ML xCOM range, step width, and stride velocity. The ML xCOM range provided an understanding of the range xCOM moved in the frontal plane during the stride. The ML xCOM range was determined by calculating the absolute difference between the maximum and minimum value during the stride. If the lateral boundaries of the BOS are similar between conditions, an increase in ML MOS is expected to have a decrease in the ML xCOM range, while a decrease in ML MOS is expected to have an increase in the ML xCOM range.

Step width provided a representation of the width of an individual's BOS. Step width was calculated by determining the absolute difference between the ML components of heel contact to the subsequent heel contact (Kodesh *et al*., 2015); therefore, two step widths were calculated per stride. If the ML xCOM range was the same, with an increase in ML MOS, it is expected step width would increase, while with a decrease in ML MOS, it is expected that step width would decrease.

Stride velocity was calculated by dividing the individual's stride length over the time it took to complete that stride. Stride velocity was normalized to account for the influence different individual leg lengths can have on the measure by using the following equation 3.3 (Hof, 1996):

$$Normalized\ Stride\ Velocity = \frac{Stride\ Velocity\ (m/s)}{\sqrt{Leg\ Length\ (m) \times g}} \quad \text{(3.3)}$$

Gravity (g) = 9.81 m/s^2, Dominant leg was measured for leg length.

Overall, ML MOS was used to provide an understanding of an individual's dynamic stability. It was hypothesized dynamic stability will be increased with LT and anchors compared to baseline walking with an added VRT task; therefore, a significant increase in ML MOS is expected. As ML MOS is expected to increase, it is expected for variables related to ML MOS: A significant decrease in the ML xCOM range for both LT and anchors compared to baseline walking, no significant differences between walking conditions for step width.

Variability, such as step width variability and ML COM SD, can provide insight into balance control to understand dynamic stability (Maki, 1997). Step width variability was determined by calculating the step width SD within trial during the stride where the VRT task occurred. Furthermore, ML COM SD was calculated and normalized by the participant's ML COM range during the stride within that trial. ML COM SD was normalized to the ML COM range because an individual with a reduced range may inherently have a decreased ML COM SD observed. If dynamic stability was increased, it is expected step width SD and ML COM SD would be decreased, while for decreased dynamic stability, it is expected step width SD and ML COM SD would be increased. Increases in step width SD and ML COM SD are associated with decreased dynamic stability because increased variability in these measures are seen to have no fundamental stabilizing influence (Maki, 1997). It is hypothesized step width SD and ML COM SD would be significantly decreased during walking with an added VRT task with LT or anchor use compared to baseline walking.

Table 2.3. – Summary of Dynamic Stability Measures

Measure:	**Relationship to Dynamic Stability:**	**Expectation with Increase in Dynamic Stability**
ML MOS	Dynamic stability	Increase
ML xCOM Range	Measures associated with ML MOS	Decrease
Step Width		Increase
Step Width SD	Balance control	Decrease
Normalized ML COM SD		Decrease

2.6. Statistical Analysis

Statistical Package for Social Science (SPSS) version 24.0 (SPSS Inc., Chicago, IL) was used with significance was set at $\alpha = 0.05$ for all statistical analyses.
Normality tests were not conducted, as the repeated measures (RM) ANOVA is robust to not meeting the assumption of normality (Lix, Keselman, & Keselman, 1996).

The order of haptic modality use was blocked and counterbalanced, resulting in half of participants starting data collection using LT on the railing first and the other half starting with anchors. To assess if there was a difference in VRTs between participants that started with anchors first compared to LT, a two-way RM ANOVA was conducted [block (anchors first/LT first) × modality (anchors/LT)]. Independent t-tests were used to analyze interactions. If post hoc comparisons were not significant, the baseline walking VRTs for the anchors and LT blocks were collapsed together.

The VRT tests involved both a “low” and “high” frequency cue. To assess if there were differences between low and high VRTs, paired samples t-tests were conducted. If all paired t-tests were not significant, low and high VRTs were collapsed together. Outliers, identified as any value within participant and condition that were ± 2 SDs away from the calculated mean, were removed and means were recalculated without outliers for the remaining statistical analysis.

The primary objective was to determine the attentional demands of adding haptic input in the forms of LT and anchors during overground walking. To assess the change in attentional demands of the VRT between sitting and walking, a paired-samples t-test was conducted. To examine the attentional demands of different haptic modalities during walking (Primary Objective), a one-way RM ANOVA was conducted [condition (baseline walking /LT/anchors)] with VRT values.

The secondary objective was to investigate the effect of an added VRT on dynamic stability during walking while using either LT on a railing or the haptic anchors compared to baseline walking. To determine the difference between the use of each haptic tool to no tool and the difference between the absence and presence of the VRT task, a 2 × 2 RM ANOVA [condition (baseline walking/LT) x presence of VRT task (no task/with task)] and a 2 × 2 RM ANOVA [condition (baseline walking/anchors) × VRT task (no task/with task)] were conducted for the ML MOS, ML xCOM range, step width, normalized stride velocity, step width SD and normalized ML COM SD.

RM ANOVA results were checked for sphericity using Mauchly’s test. If Mauchly’s test was significant, a Greenhouse-Geisser correction was used. Post hoc analyses included pairwise comparisons with Bonferroni corrections.

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3. Results

Twenty-two young, healthy adults (12 males) aged (24.5 +/- 2.9 years), with mass (75.6 +/- 17.5 kg) and height (1.71 +/- 0.18 m) participated. Twenty-one participants were right hand dominant and 14 were right leg dominant.

3.1 – Attentional Demands of Haptic Modalities

Paired *t*-tests revealed that there were no significant differences between the VRTs of low and high tones for all conditions in which the VRT task was given seated ($t(21) = 1.885$, $p = 0.073$), baseline walking (no modality use) within anchor block ($t(21) = -0.090$, $p = 0.929$), baseline walking within the LT block ($t(21) = -0.629$, $p = 0.536$), LT use ($t(21) = 0.586$, $p = 0.564$), and anchor use ($t(21) = -1.323$, $p = 0.200$). As no significance was observed, the low and high VRT data were collapsed.

A 2 × 2 [Start (anchors first/LT first) × modality (LT/ anchors)] RM ANOVA to assess the difference in VRTs between starting with LT or the anchors revealed a significant interaction ($F(1,20) = 9.334$, $p = 0.006$). Independent samples *t*-tests revealed no significant differences between blocks (i.e., starting first or second with respective modality) for VRTs while walking with the anchors ($t(20) = -0.484$, $p = 0.634$) or LT ($t(20) = 1.928$, $p = 0.068$). No significant differences were observed in the post hoc analysis of the interaction, which allowed for baseline walking conditions from both the anchors and LT blocks to be collapsed. The false positive for the interaction was a result of the 2 × 2 RM ANOVA design. The 2 × 2 RM ANOVA design can detect a false positive for interaction based on any intersection between the two groups analyzed.

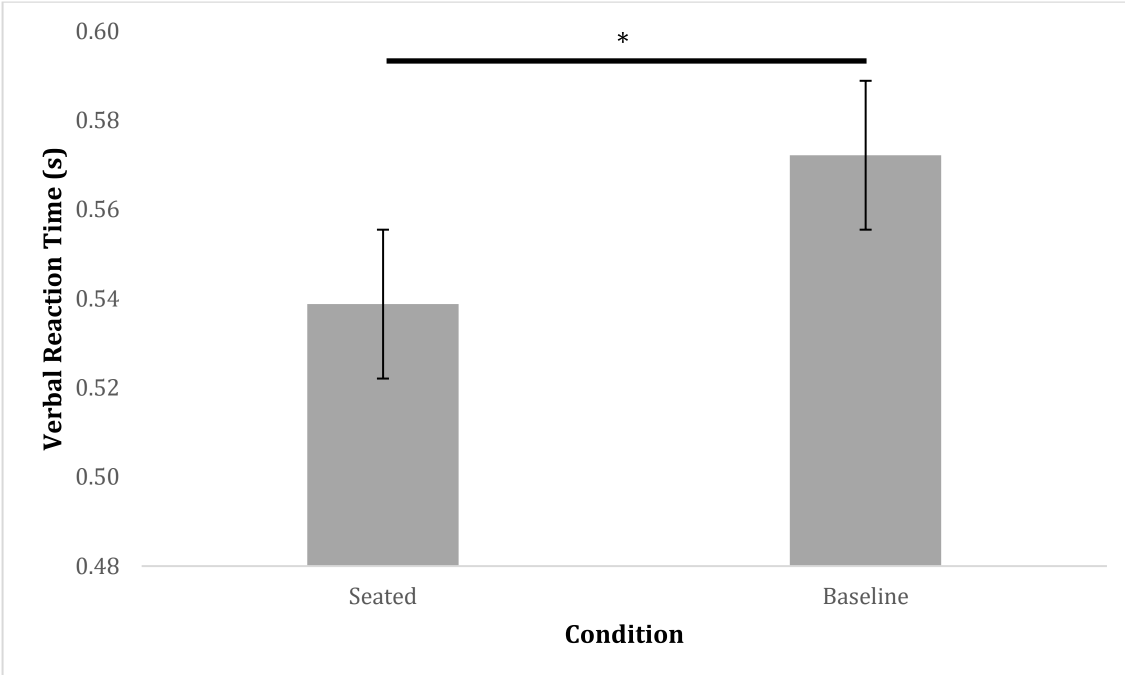

Figure 3.1. – Comparison of mean (± SE) verbal reaction times (s) between seated and baseline walking condition (* indicates a significant difference between means).

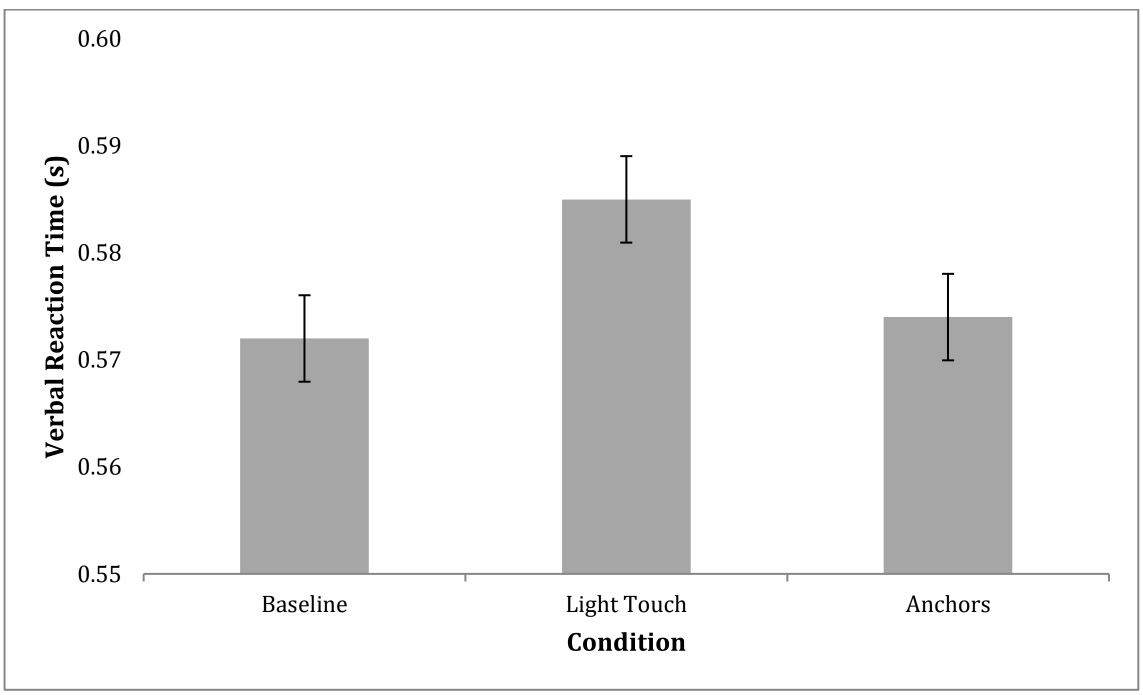

Figure 3.2. – Comparison of mean (± SE) verbal reaction times (s) between walking conditions

A paired samples *t*-test comparing VRTs between sitting and walking revealed a significant increase in VRT when walking ($t(21) = -2.081$, $p = 0.050$) (Figure 3.1). One-way RM ANOVA [condition (baseline walking/LT/anchors)] of VRTs revealed no significant changes between walking conditions (Greenhouse-Geisser corrected $p = 0.506$) (Figure 3.2).

3.2 – Measures of Dynamic Stability with Use of Haptic Modalities and VRT

3.2.1. Light touch on a railing

There was a significant main effect of condition for average ML MOS for LT ($F(1,21) = 21.904$, $p < 0.001$) when compared to baseline walking. Post hoc analysis revealed that baseline walking had significantly greater average ML MOS compared to LT ($p < 0.001$) (Table 3.1). There was no significant main effect for presence of VRT task for the ML MOS in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking ($F(1,21) = 4.078$, $p = 0.056$). For the 2 × 2 RM ANOVA analysis for ML MOS, there was no significant interaction between condition and presence of VRT task ($F(1,21) = 0.051$, $p = 0.824$)

There was a significant main effect of condition for ML xCOM range for LT use compared to baseline walking ($F(1,21) = 14.278$, $p = 0.001$). Post hoc analysis revealed that LT use had a significantly decreased ML xCOM range compared to baseline walking ($p = 0.001$) (Table 3.1). There was a significant main effect for presence of VRT task for the ML xCOM range in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking ($F(1,21) = 6.191$, $p = 0.021$). Post hoc analysis revealed that presence of a VRT task had a significant increase in the ML xCOM range compared to no VRT task ($p = 0.021$) (Table 3.1). For the 2 × 2 RM ANOVA analysis for the ML xCOM range, there was no significant interaction between condition and presence of VRT task ($F(1,21) = 0.339$, $p = 0.567$)

There was a significant main effect of condition for normalized stride velocity for LT use compared to baseline walking ($F(1,21) = 20.891$, $p < 0.001$). Post hoc analysis revealed that LT had significantly decreased normalized stride velocity compared to baseline walking ($p < 0.001$) (Table 3.1). There was no significant main effect for presence of VRT task for normalized stride velocity in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking ($F(1,21) = 0.000$, $p = 0.983$). For the 2 × 2 RM ANOVA analysis for normalized stride velocity, there was no significant interaction between condition and presence of VRT task ($F(1,21) = 0.051$, $p = 0.824$)

There was a significant main effect of condition for step width for LT use (*F* (1,21) = 32.318, $p < 0.001$) compared to baseline walking. Post hoc analysis revealed that LT use ($p < 0.001$) had a significantly decreased step width when compared to baseline walking (Table 3.1). There was no significant main effect for presence of VRT task for step width in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking (*F* (1,21) = 2.175, $p = 0.155$). For the 2 × 2 RM ANOVA analysis for step width, there was no significant interaction between condition and presence of VRT task (*F* (1,21) = 0.015, $p = 0.905$)

There was a significant main effect of condition for step width SD for LT (*F* (1,21) = 5.411, $p = 0.030$) compared to baseline walking. Post hoc analysis revealed that LT ($p = 0.030$) had significantly increased step width SD when compared to baseline walking (Table 3.1). There was no significant main effect for presence of VRT task for step width SD in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking (*F* (1,21) = 0.526, $p = 0.894$). For the 2 × 2 RM ANOVA analysis for step width SD, there was no significant interaction between condition and presence of VRT task (*F* (1,21) = 0.723, $p = 0.405$)

There was a significant main effect of condition for normalized ML COM SD for LT use compared to baseline walking (*F* (1,21) = 17.080, $p < 0.001$). Post hoc analysis revealed that LT had significantly increased normalized ML COM SD compared to baseline walking ($p < 0.001$) (Table 3.1). There was no significant main effect for presence of VRT task for normalized ML COM SD in the 2 × 2 RM ANOVA analysis that examines LT use to baseline walking (*F* (1,21) = 0.018, $p = 0.894$). For the 2 × 2 RM ANOVA analysis for normalized ML COM SD, there was no significant interaction between condition and presence of VRT task (*F* (1,21) = 0.291, $p = 0.595$)

3.2.2. Anchors

There was a significant main effect of condition for average ML MOS for anchors (*F* (1,21) = 8.002, $p = 0.010$) when compared to baseline walking. Post hoc analysis revealed that baseline walking had significantly greater average ML MOS compared to anchors ($p = 0.010$) (Table 3.1). There was no significant main effect for presence of VRT task for the ML MOS in the 2 × 2 RM ANOVA analysis that examines anchors use to baseline walking (*F* (1,21) = 1.478, $p = 0.238$). For the 2 × 2 RM ANOVA analysis for ML MOS, there was no significant interaction between condition and presence of VRT task (*F* (1,21) = 0.202, $p = 0.658$)

There was no significant main effect of condition for ML xCOM range for anchors compared to baseline walking (F (1,21) = 1.927, p = 0.180). There was no significant main effect for presence of VRT task for the ML xCOM range in the 2 × 2 RM ANOVA analysis that examines anchor use to baseline walking (F (1,21) = 1.734, p = 0.202). For the 2 × 2 RM ANOVA analysis for the ML xCOM range, there was no significant interaction between condition and presence of VRT task (F (1,21) = 0.050, p = 0.826)

There was no significant main effect of condition for normalized stride velocity for anchors compared to baseline walking (F (1,21) = 0.067, p = 0.798). There was no significant main effect for presence of VRT task for normalized stride velocity in the 2 × 2 RM ANOVA analysis that examines anchors use to baseline walking (F (1,21) = 0.148, p = 0.704). For the 2 × 2 RM ANOVA analysis for normalized stride velocity, there was no significant interaction between condition and presence of VRT task (F (1,21) = 0.072, p = 0.791)

There was a significant main effect of condition for step width with anchors (F (1,21) = 4.590, p = 0.044) compared to baseline walking. Post hoc analysis revealed that anchors (p = 0.044) had resulted in significantly decreased step widths when compared to baseline walking (Table 3.1). There was no significant main effect for presence of VRT task for step width in the 2 × 2 RM ANOVA analysis that examines anchors use to baseline walking (F (1,21) = 4.027, p = 0.058). For the 2 × 2 RM ANOVA analysis for step width, there was no significant interaction between condition and presence of VRT task (F (1,21) = 1.262, p = 0.274)

There was a significant main effect of condition for step width SD for anchor use (F (1,21) = 5.153, p = 0.034) compared to baseline walking. Post hoc analysis revealed that anchors (p = 0.034) had significantly increased step width SD when compared to baseline walking (Table 3.1). There was no significant main effect for presence of VRT task for step width SD in the 2 × 2 RM ANOVA analysis that examines anchors use to baseline walking (F (1,21) = 1.028, p = 0.322). For the 2 × 2 RM ANOVA analysis for step width SD, there was no significant interaction between condition and presence of VRT task (F (1,21) = 4.003, p = 0.059)

There was no significant main effect of condition for normalized ML COM SD for anchors compared to baseline walking (F (1,21) = 0.137, p = 0.715). There was no significant main effect for presence of VRT task for normalized ML COM SD in the 2 × 2 RM ANOVA analysis that examines anchors use to baseline walking (F (1,21) = 0.423, p = 0.522). For the 2 ×

2 RM ANOVA analysis for normalized ML COM SD, there was no significant interaction between condition and presence of VRT task (F (1,21) = 0.001, p = 0.981)

Table 3.1. – Summary of dynamic stability and walking measures

Measure	Baseline	LT	Anchors
Average ML MOS (mm)	111 (2.5)	106 (2.4)*	109 (2.5) *
Normalized stride velocity	0.42 (0.03)	0.40 (0.03)*	0.42 (0.03)
ML xCOM range (mm)	36.5 (2.0)	33.6 (2.0)*	35.0 (1.9)
Step width (mm)	85.9 (6.6)	75.3 (5.9)*	82.4 (6.2)*
Step width SD	10.7 (1.2)	14.1 (1.3)*	13.5 (1.5)*
Normalized ML COM SD	31.9 (0.5)	32.9 (0.5)*	31.8 (0.4)

* indicates a significant difference from baseline walking

For additional analyses regarding dynamic stability (i.e. ML trunk range of motion and peak ML trunk velocity), refer to Appendix F.

4. Discussion

The primary objective of this thesis was to use a VRT task to investigate the attentional demands of using either LT on a railing or haptic anchors during walking. The secondary objective was to investigate the effect of an added VRT task on dynamic stability during walking while using either LT on a railing or the haptic anchors. To my knowledge, this is the first study to examine these objectives in young adults. For the primary objective, it was hypothesized that LT would require a significant increase in attentional demands (i.e., a significantly longer VRT) compared to baseline walking and anchor use, while anchors would require similar attentional demands (similar VRTs) compared to baseline walking. The hypothesis was partially supported, as anchors required similar attentional demands to those of baseline walking; however, LT was also found to have similar attentional demands to those of baseline walking.

4.1. Haptic Modalities and Effects on Attentional Demands

It was observed that VRT during baseline walking increased more than VRT in the seated condition (Figure 3.1). The increase in VRT suggests that walking requires greater attentional demands than sitting, which has been observed in previous studies (Lajoie *et al*., 1996; Siu *et al*., 2008a; Siu *et al*., 2008b). Increased attentional demands during walking corresponds to increased cortical input compared to that required for sitting (Gwin, Gramann, Makeig, & Ferris, 2011; Jain, Gourab, Schindler-Ivens, & Schmit, 2013).

I found VRTs during walking were similar between LT, anchors and baseline walking (Figure 3.2). These findings suggest that walking with haptic modalities requires similar attentional demands to baseline walking. In comparison, a study by Vuillerme *et al*. (2006) examined attentional demands of LT during standing. In the study, participants pressed a button when a tone was heard, while performing LT with the dominant hand or not performing LT (Vuillerme et al., 2006). Results showed that the use of LT significantly increased reaction time compared to that with no LT during standing (Vuillerme et al., 2006). Differences between my study and Vuillerme *et al*. (2006) may be due to the type of attentional resources required and the ability to practice the tasks in advance of testing. Participants in Vuillerme *et al*. (2006) had to push a button with one hand and perform LT with the other, which required cutaneous input to maintain contact with both. The overlap of sensory information between LT and holding a button, while each hand performs a different task may result in greater structural interference between tasks (Kahneman, 1973). In my study, the VRT task required different attentional resources than walking and/or using the haptic modalities, thereby minimizing structural interference (Abernethy, 1988). Structural interference results in longer reaction times, as similar attentional resources are used (Kahneman, 1973). Similar attentional resources affect both perception and response selection phases of task execution, overall, affecting output (Abernethy, 1988; Kahneman, 1973; Worden et al., 2016). Alternatively, capacity interference requires different attentional resources because different cortical structures are used (Abernethy, 1988; Kahneman, 1973). Interference between tasks occurs when attentional capacity is reached because the response-selection phase of each task creates a bottleneck that delays output (i.e., increased reaction time) (Kahneman, 1973).

In addition to the capacity and structural interference models of attention, familiarization with tasks may account for different findings between studies because it can improve action-selection. Researchers have observed that DT training prior to data collection results in reduced VRTs from improved action-selection (Worden & Vallis, 2014). The action-selection model of attention suggests tasks are executed as one and practicing tasks together reduces interference during the response selection phase; therefore, reducing attentional demands (Neumann, Heuer, & Sanders, 1987; Strobach et al., 2013). In my study, participants familiarized with the haptic modalities and VRT task, while participants in Vuillerme *et al*. (2006) did not. I provided familiarization to ensure participants were comfortable and consistent with how they used the haptic modalities throughout data collection. Similar VRTs between haptic anchors and LT with a short familiarization period (~ 10 – 20 trials per block) prior to collection with each modality is interesting because it contrasts other studies that examined walking aids. In studies that have examined DT with the use of walking aids (i.e., canes, walkers), these findings have not been observed (Wellmon *et al*., 2006; Wright & Kemp, 1992).

In the first study to assess attentional demands of rolling and standard walkers, participants were provided the opportunity to practice (Wright & Kemp, 1992). Furthermore, the first half of VRT task trials out of 10 trials collected were considered practice (Wright & Kemp, 1992). With familiarization and practice trials, the use of walkers still resulted in significantly increased VRTs, suggesting increased attentional demands compared to baseline walking.

Subsequent research compared older adult groups that use different walking aids (i.e., standard canes, rolling walkers) to those that used no modality during standing and walking (Wellmon *et al*., 2006). Standing revealed no significant differences in VRT between groups; however, differences were observed during walking (Wellmon et al., 2006). The group that used rolling walkers had significantly increased VRTs compared to groups using no modality or a standard cane. These findings indicate that even with experience and familiarization, walking aids requiring increased complexity to maneuver require greater attentional demands during walking (Wright & Kemp, 1992).

Increased attentional demands placed on walking aids may be explained by the phases of task execution (Strobach *et al*., 2013). Standard walker use involves lifting, moving, placing and walking towards the walker, consistently engaging the individual in the response selection stage to perform the next phase of maneuvering the walker (Strobach *et al*., 2013; Wright & Kemp, 1992). Familiarization to improve the action-selection process (i.e., decrease attentional demands) becomes challenging, as each phase maneuvering the modality would require practice with the additional task (Neumann et al., 1987); however, it has also been observed with concurrent motor tasks that stride velocity decreases (Al-Yahya et al., 2011). The decrease in stride velocity has also been used to suggest attentional demands are required with a task, in addition to the increases in VRTs (Al-Yahya et al., 2011; Woollacott & Shumway-Cook, 2002). These findings suggest attentional demands of a task may be observed differently, depending on the requirements of the task.

The decrease observed in normalized stride velocity with LT suggests attentional demands might still be required. The decrease in stride velocity can be due to the concurrent nature of the motor task created with LT because the individual is required to maintain connection with a small contact surface (Al-Yahya et al., 2011). Anchors appear to not exhibit a decline in stride velocity compared to LT because anchors are passively dragged alongside the individual. The similar VRTs with LT and anchors compared to baseline walking observed in my study may also be a result of task execution not varying during walking. With haptic modalities, task execution does not vary during walking because prior to the start of each trial the participant places their finger on the railing and does not need to make adjustments to ensure contact is made, and with the anchors the person ensures they are positioned comfortably for walking.

4.2. Haptic Modalities and Effects on Dynamic Stability

The secondary objective of this thesis was to investigate the effect of an added VRT task on dynamic stability during walking while using either LT on a railing or the haptic anchors compared to baseline walking. The hypothesis was not supported because measures related to dynamic stability for walking were similar between the presence of a VRT task and no task, while no significant interactions were observed. Interestingly, when comparing use of haptic modalities to baseline walking, it was observed LT and anchors use affected dynamic stability compared to baseline walking (Table 3.1). For ML MOS, which assesses dynamic stability of the individual, there was a significant decrease when using LT and anchors compared to baseline walking. Furthermore, measures related to ML MOS, such as the ML xCOM range and step width were significantly decreased with LT compared to baseline walking, while anchors had a significant decrease in step width compared to baseline walking. Overall, decreases in step width would result in the participant's ML xCOM position being closer to the lateral boundaries of the BOS. The decrease in distance between the ML xCOM position and the lateral BOS boundary would result in a reduction in ML MOS. With LT, where a larger reduction in ML MOS is seen, the decrease in the ML xCOM range was observed to be significantly decreased. The decrease in the ML xCOM range would result in less movement of the ML xCOM position across the stride. A decrease in movement of the ML xCOM position across the stride may occur to minimize the decrease in ML MOS because the ML xCOM position would be moving closer to the lateral BOS boundaries associated with the decreased step width observed.

Both LT and anchors had significant increases for step width SD when compared to baseline walking, while LT also had a significant decrease in normalized ML COM SD when compared to baseline walking, suggesting balance control was most affected with LT. It has been suggested an increase in step width SD is associated with a decrease in dynamic stability because the individual is unable to control their COM (Brach et al., 2005). With LT use, it was observed that normalized ML COM SD was significantly increased, as well, which may suggest dynamic stability might be affected. The increase in step width SD may indicate an effect on dynamic stability with anchors; however, similar normalized ML COM SD to baseline walking may suggest the effect is less profound than those observed with LT. Overall, these findings suggest LT and anchors had decreased dynamic stability compared to baseline walking.

My findings suggest the addition of a VRT task when using haptic modalities did not affect dynamic stability when compared to no VRT task present for ML MOS, step width SD and normalized ML COM SD. Comparisons among measures of dynamic stability (e.g., ML MOS, step width SD and normalized ML COM SD) assessed suggest the addition of the VRT task did not affect the individual's dynamic stability and may have not resulted in the changes observed in dynamic stability between haptic modalities and baseline walking. The lack of change in dynamic stability measures may be due to the simplicity of the VRT task present, as has been observed in previous literature (Siu et al., 2008a; Woollacott & Shumway-Cook, 2002; Worden & Vallis, 2015). Furthermore, similarities in measures related to dynamic stability may indicate individuals were not prioritizing the VRT task, which has been observed in the DT literature (Harley et al., 2009) ; however, the decreases observed in dynamic stability measures (e.g., ML MOS, step width SD and normalized ML COM SD) may have been associated with the use of haptic modalities.

When comparing the use of haptic modalities to baseline walking, my findings suggest LT affected the participants' dynamic stability because ML MOS significantly decreased compared to baseline walking (Table 3.1). These findings suggest LT use may predict greater instability of an individual during a perturbation, as indicated by previous studies that assessed ML MOS during overground walking in conditions that challenged dynamic stability (e.g., obstacle crossing) (Worden & Vallis, 2015; Young et al., 2012). As ML MOS considers an individual's ML xCOM position in relation to their BOS, the ML xCOM range can provide insight regarding changes in ML MOS. I found the ML xCOM range significantly reduced with LT compared to baseline walking (Table 3.1). These results suggest individuals are moving their ML xCOM position less across the stride. It was hypothesized that this strategy would increase ML MOS. This expectation was hypothesized because the reduced ML xCOM range suggests the ML xCOM position distance from the lateral boundary of the BOS would increase, if step width was similar between conditions, thereby increasing ML MOS. Contrary to expectations, the size of the individual's BOS decreased (i.e., smaller step width) for both LT and anchors use. It was observed that no changes in step width occurred with the presence of the VRT task compared to no task. Overall, the reduced step width observed with haptic modalities may explain the observed decrease in ML MOS with LT and anchors, as the ML xCOM position has a narrower BOS in which to move, possibly moving closer to the lateral boundary of the BOS (Schrager *et al*., 2008).

These findings suggest that a relationship may exist between the ML xCOM range and step width; however, the underlying mechanism that would explain such a relationship is unknown. Reduced step width may occur because individuals using haptic modalities perceive their stability has improved (Ortega, Fehlman, & Farley, 2008). It has been shown when external mechanical support in the frontal plane is provided, step width decreases (Ortega *et al*., 2008). In a study that compared walking aids (i.e., crutches, cane, rolling walker) in healthy, young adults, crutches and rolling walkers significantly decreased step width compared to using no aid (Youdas, Kotajarvi, Padgett, & Kaufman, 2005). Reduced step width with rolling walkers and crutches may have occurred because individuals significantly increased their BOS from the mechanical support provided by the walking aid, since approximately 50 % of their body weight was offloaded (Youdas et al., 2005). Cane use compared to no cane use during walking in healthy, young and older adults was observed to have similar step widths, instead of the significant decreases observed with the other walking aids (Boonsinsukh, Saengsirisuwan, Carlson-Kuhta, & Horak, 2012; Youdas et al., 2005). These findings were found in healthy populations that did not have any instabilities and/or asymmetries that are common in clinical populations, such as stroke, to have a benefit from a cane (Bateni & Maki, 2005). In a stroke population with hemiplegia that could benefit from unloading of their affected side to improve stability due to mechanical support, use of a cane had a significant decrease in step width (Kuan, Tsou, & Su, 1999). Overall, these studies (Boonsinsukh et al., 2012; Kuan et al., 1999; Youdas et al., 2005) suggest walking aids that provide mechanical support result in decreased step width. The decreased step width may be associated with increased mechanical support from increasing the individual's BOS (Bateni & Maki, 2005).

Compared to walking aids, haptic modalities do not provide mechanical support (Holden *et al*., 1994). Sensory input may have provided the perception of improved stability leading to the change in walking behaviour (i.e., decreased step width) with anchors and most noticeably, with LT (Ortega et al., 2008). It is possible that LT of a rigid railing may provide a context where the individual perceives if they lost their balance, the railing could be used to provide mechanical support, which has been observed with walking aids, such as canes (Bateni, Zecevic, McIlroy, & Maki, 2004). The context provided may have resulted in the larger decrease in step width using

LT compared to anchors. In addition to the context the railing may provide the individual, my results regarding step width may also be due to a relationship with the ML xCOM range. It appears changes in step width parallel changes in the ML xCOM range because when the ML xCOM range is reduced with LT (Table 3.1), step width decreased. Similar trends between step width and the ML xCOM range may be due to the individual restricting the movement of their ML xCOM position because it is closer to the lateral boundary of their BOS. Reducing the ML xCOM range may be an attempt to prevent further reduction in ML MOS.

Normalized stride velocity significantly decreased with LT compared to baseline walking. (Table 3.1). An explanation for LT decreasing normalized stride velocity has been associated with the capacity interference model of attention, as outlined in section 4.1. (Yogev-Seligmann et al., 2010). These attentional demands associated with LT may result in walking behaviour (i.e. stride velocity) changes that affected measures of dynamic stability (i.e., ML MOS, step width SD and normalized ML COM SD), as has been observed in the literature with other dynamic stability measures (Dingwell et al., 2008; Hak et al., 2012).

Overall, changes in ML MOS suggests LT and anchors may have affected dynamic stability compared to baseline walking. These findings were unexpected because previous literature examining measures of balance control (i.e., variability of COM and step width, ML trunk ROM and peak ML trunk velocity) have shown LT and anchors significantly improved dynamic stability (Bingenheimer et al., 2015; da Silva Costa et al., 2015; Dickstein & Laufer, 2004; Hedayat et al., 2017; Kodesh et al., 2015; Mauerberg-deCastro et al., 2014). My findings are unlikely to be a result of differences in measures because I calculated balance control measures used in previous studies to provide a comprehensive understanding of the changes in dynamic stability. Differences observed in previous studies might be due to the 12 segment COM model used in my study compared to previous studies that used single marker COM models (discussed later in this section).

Variability measures were calculated to assess balance control, such as step width SD and a normalized ML COM SD because significant decreases suggest improvements in dynamic stability (Dickstein & Laufer, 2004; Kodesh *et al*., 2015). Our findings showed step width SD with LT and anchors were significantly increased compared to baseline walking. In comparison to my findings, Kodesh *et al*. (2015) observed LT use reduced step width SD suggesting LT improved dynamic stability. These findings were observed when participants used LT with their eyes closed, where loss of visual input would affect foot placement during walking (Kodesh et al., 2015). The sensory input provided by LT when the individual has their eyes closed allows them to have an awareness of where they are in space, reducing the variability of their foot placement (Holden et al., 1994; Kodesh et al., 2015). The different findings in my study may have resulted because participants had their eyes open for all conditions, which did not provide an opportunity to compare use of modalities without vision compared to with vision present (i.e., eyes open). Without vision, it would be expected that haptic modalities would improve variability measures (i.e. decrease variability) compared to no modality because the individual would have better awareness of where they are in space from the proprioceptive information provided by the haptic modalities (Bingenheimer et al., 2015; Dickstein & Laufer, 2004; Kodesh et al., 2015). Overall, my study focused on how the sensory input could assist an individual in an environment, where vision is present.

Similar to step width SD, it was also found in my study that normalized ML COM SD was significantly increased with LT compared to baseline walking, indicating variability of COM movement was affected with LT, while anchors were similar to baseline walking. Different results to my study were found in a study where dynamic stability was assessed using COM displacement approximated from an accelerometer placed on the participant's lower back region (Bingenheimer et al., 2015). It was found that ML COM displacement was similar between different walking modality conditions compared to baseline walking with eyes opened (Bingenheimer et al., 2015). Furthermore, results from Bingenheimer *et al*. (2015) are comparable with those observed in another study that found ML COM variance was similar between LT use and baseline walking with eyes open (Dickstein & Laufer, 2004). The different results found here in normalized ML COM SD for LT use compared to baseline walking may be a result of my 12 segment COM model compared to the single point (i.e. reduced marker set) COM models used in other studies (Bingenheimer et al., 2015; Dickstein & Laufer, 2004). Differences in ML COM SD would be observed because single marker COM models used in other studies (Bingenheimer et al., 2015; Dickstein & Laufer, 2004) are predisposed to error, since these reduced COM models cannot account for changes in arm position (Tisserand et al., 2016; Yang & Pai, 2014). Use of haptic modalities involve changes in arm position and constraining arm swing, which can affect the COM variables calculated from single marker COM models that do not include arm segments; therefore, resulting in differences from my study.

The multi-segment COM model I used in my study considered various body segments, including the arms. When participants use LT, the dominant arm position is constrained, while the other arm can swing normally. The asymmetry created by constraining the arm position of the dominant arm, while allowing the non-dominant to move naturally may have created the increased normalized ML COM SD with LT compared to baseline walking because arm swing was affected (Meyns, Bruijn, & Duysens, 2013).

Lastly, the significant decreases observed in ML MOS observed with LT and anchors may have been attributed to arm position and decreased arm swing. With anchors, arm swing is constrained when holding onto the strings, which can result in a change in the individuals ML xCOM position that can bring the ML xCOM closer to the lateral boundary of BOS. In contrast, LT has only one arm held out, unlike anchor use, which may result in the ML xCOM position moving closer towards the dominant arm; therefore, moving closer to the lateral BOS boundary, decreasing ML MOS. When using haptic modalities, arm position is fixed, which as previously mentioned may result in the ML xCOM position moving closer to a lateral BOS boundary. In a study that examined cell phone text messaging while walking, where arm position was placed in the front of the participant, there was a significant increase in ML MOS (Marone et al., 2014). The significant increase in ML MOS observed in Marone et al. (2014) may have been associated with the arms being moved closer to the trunk; therefore, resulting in the ML xCOM position moving closer to the trunk. The ML xCOM position moving closer to the trunk would result in a greater distance to the lateral BOS boundary, resulting in an increased ML MOS, despite arm swing being constrained. Also, it has been shown that arm swing plays a role in dynamic stability. In a study that examined the role of arm swing related to dynamic stability during walking, it was observed that increased arm swing led to improved dynamic stability. An inference that can be made from this study is if arm swing is reduced or constrained, dynamic stability may be affected (Punt, Bruijn, Wittink, & van Dieën, 2015). Overall, the arm position and lack of arm swing associated with use of haptic modalities may have resulted in the significant decreases observed in ML MOS.

Overall, my findings suggest dynamic stability is affected with use of haptic modalities; however, the use of haptic modalities did not affect dynamic stability with the addition of a VRT task compared to no VRT task. The effect using haptic modalities may have on dynamic stability might be a result of arm swing being affected. In addition, the arm position may influence the xCOM position to shift closer to the lateral BOS boundaries, in comparison to baseline walking. Overall, constrained arm swing with the use of haptic modalities may have resulted in the decreases observed in dynamic stability. It should be noted that changes observed in dynamic stability measures were observed in young, healthy adults, which do not have impairments in dynamic stability. As a result of the young adult population, they may not be using the sensory input to improve their dynamic stability because the overground walking and VRT task would not provide challenges to their dynamic stability. Furthermore, differences between haptic modalities and baseline walking in dynamic stability, such as in ML MOS, step width SD and normalized ML COM SD may not be clinically significant. The magnitude of changes between haptic modalities and baseline walking may not represent clinical significance (e.g., ~ 5 mm difference between LT and baseline walking) because it is not known how these changes would affect a healthy population. In addition, the significant differences observed in dynamic stability measures between haptic modalities and baseline walking in my study, although suggesting decreased dynamic stability, may not be observed in populations with instability, who may be able to utilize the sensory input to improve their dynamic stability during walking.

4.3. Limitations

A limitation in this study is not collecting conditions where the participants pretended (i.e., maintained a similar arm position) to use the haptic modalities, as in Hedayat *et al* (2017). These conditions would provide insight into if the arm placements were resulting in the decrease in ML MOS observed with LT by altering the ML xCOM position. In addition, to these conditions that constrain arm position, a limitation in my study would include that use of haptic modalities affect natural arm swing, which may also play a role in dynamic stability and if arm swing is restricted, this could result in a decreased dynamic stability (Meyns et al., 2013).

Another limitation regarding the study is not collecting kinematics and VRT data during the familiarization period. Data collection during the familiarization period would have provided an understanding of if a training effect occurred during that period with the haptic modalities. An observed training effect would be able to provide further evidence that action-selection was improved, resulting in minimal attentional demand differences observed between walking conditions. If no difference was observed during the familiarization period, it would also suggest no capacity interference occurred and may suggest that the response selection phase is not affected with haptic modalities. The lack of interference may be a result of the participant not needing to maneuver/manipulate the modality while walking, as can be seen with walking aids (i.e., walkers). Furthermore, no conditions were performed where the participants were asked to allocate their attention to specific tasks, such as focusing on the haptic modality only, VRT task only or walking only. These conditions would verify, in addition to my instructions to maintain equal focus on all tasks, if a posture-first response was maintained, since similar VRTs would be expected between the equal focus and walking only conditions.

For motion capture, there are several limitations that should be noted. As passive markers could not always be placed on the skin, they were placed on pieces of clothing and taped down. The movement of skin and clothing can create movement artifacts that can affect marker position. To minimize movement artifact of skin and clothing, reflective markers were taped down.

As we did not use a cluster-based system to determine joint centres of the knees and ankles, this can influence the segment lengths calculated for the lower limbs. Overall, using single markers to approximate joint centres can affect the accuracy of the whole-body COM calculated because it may result in under- or overestimating segment lengths, depending on the placements of the reflective markers. Furthermore, the hand segments were approximated in the calculation of whole-body COM. By approximating the hand segments, the accuracy of the whole-body COM calculated can be affected. Also, the use of a trunk segment that was calculated using the midpoints of the shoulders and hip joint centres, instead of a multi-segment trunk, can affect the COM location of the trunk, which can result in a larger whole-body COM error because approximately 40 % of an individual's body mass is accounted for by the trunk (Leva, 1996).

Lastly, use of the force plates to trigger the VRT tasks can affect the reliability of where the VRT task was occurring. Although, the participant's starting positions were determined during familiarization periods to ensure similar foot placement on the force plates, it could not be guaranteed foot placement was consistent throughout the entirety of data collection. Furthermore, there could be a slight delay between the force plate detecting 10 % of the participant's mass to the computer triggering the tone for the VRT task. This delay could also affect the consistency of when the VRT task was administered.

4.4. Future Directions

The current study has numerous future directions that can further the understanding of haptic modalities and attention demands during walking. One of these possible directions would be to expand this study to examine the response to a postural threat to gait (i.e., obstacle crossing, slip perturbation). The challenges these tasks present during gait would provide insight into how the modalities affect attentional demands during a more challenging walking condition. Furthermore, it would provide an understanding if these modalities improve stability during a more challenging task, in comparison to overground walking.

Another direction would be consideration of the cognitive task. In this study, a simple VRT task was used; however, it would be interesting to see the impact a more challenging reaction time task, such as a Stroop task, could have with respect to attentional demands. Furthermore, examining different instructions on what task to focus on for the participant would provide insight to if the posture-first response is maintained when using haptic modalities, as the perception of improved stability could result in the allocation of attention towards the VRT task.

Lastly, using the established study design in older adult and clinical populations with balance impairments would be important to assess if the attentional demands of haptic modalities remain minimal, as observed in young adults. Furthermore, if haptic modalities require attentional demands in older adult and clinical populations, it would be essential to assess how the attentional demands compare to the use of walking aids, such as canes and walkers. In addition, if attentional demands are observed with haptic modalities in older adult and/or clinical populations, the basis for a training intervention would be important for two key reasons: 1) to assess if attentional demands of haptic modalities can be attenuated, and; 2) comparing results to other dual-task training intervention studies. If results are similar or better than previously performed dual-task training studies, the use of haptic modalities may extend beyond assisting during walking but rather be used to maintain the ability to dual- and multi-task in at-home training interventions.

4.5. Conclusion and Clinical Implications

In summary, it was observed that haptic modalities during overground walking did not have a significant effect on VRT, suggesting minimal attentional demands; however, the significant decrease in ML MOS and other measures may suggest decreased dynamic stability with LT and anchors with or without a VRT task. Overall, the similar VRTs associated with haptic modalities suggest that LT and the anchors may provide an individual the ability to successfully engage in other tasks; however, further investigation is necessary. For dynamic stability with an added VRT task, it was observed that measures of dynamic stability were unaffected when compared to conditions where no VRT task was administered. The decreases observed in dynamic stability associated with haptic modalities may be associated with a lack of arm swing that can occur when using these modalities.

Future studies should focus on the effects of increasing the difficulty of the VRT task, change to a concurrent cognitive task of varying difficulty, and/or a threat to gait, with respect to attentional demands and dynamic stability when using haptic modalities. Furthermore, investigation is required to assess if VRT and dynamic stability results observed in young adults are seen in older adults and clinical populations, where decreased attentional capacity may have a greater impact on stability and fall risk. If attentional demands are required and no impairments to dynamic stability are observed with the modalities in older adult and/or clinical populations, further investigation is needed. Further investigation should be performed to assess if training with the modalities can reduce the attentional demands, while maintaining or improving dynamic stability and walking performance, and how these results compare to other studies that examine dual- and multi-task training.

5. References

Abernethy, B. (1988). Dual-task methodology and motor skills research: some applications and methodological constraints. *Journal of Human Movement Studies*, *14*(3), 101–132.

Adamo, D. E., Martin, B. J., & Brown, S. H. (2007). Age-related differences in upper limb proprioceptive acuity. *Perceptual and Motor Skills*, *104*(3_suppl), 1297–1309.

Al-Yahya, E., Dawes, H., Smith, L., Dennis, A., Howells, K., & Cockburn, J. (2011). Cognitive motor interference while walking: a systematic review and meta-analysis. *Neuroscience & Biobehavioral Reviews*, *35*(3), 715–728.

American Geriatrics Society, B. G. S. (2010). AGS/BGS Clinical Practice Guideline: Prevention of Falls in Older Persons. American Geriatrics Society New York, NY.

Arora, T., Musselman, K. E., Lanovaz, J., & Oates, A. (2017). Effect of haptic input on standing balance among individuals with incomplete spinal cord injury. *Neuroscience Letters*, *642*, 91–96.

Baccini, M., Rinaldi, L. A., Federighi, G., Vannucchi, L., Paci, M., & Masotti, G. (2007). Effectiveness of fingertip light contact in reducing postural sway in older people. *Age and Ageing*, *36*(1), 30–35. https://doi.org/afl072 [pii]

Bateni, H., & Maki, B. E. (2005). Assistive devices for balance and mobility: benefits, demands, and adverse consequences. *Archives of Physical Medicine and Rehabilitation*, *86*(1), 134–145.

Bateni, H., Zecevic, A., McIlroy, W. E., & Maki, B. E. (2004). Resolving conflicts in task demands during balance recovery: does holding an object inhibit compensatory grasping? *Experimental Brain Research*, *157*(1), 49–58.

Beauchet, O., Annweiler, C., Allali, G., Berrut, G., Herrmann, F. R., & Dubost, V. (2008). Recurrent Falls and Dual Task–Related Decrease in Walking Speed: Is There a Relationship? *Journal of the American Geriatrics Society*, *56*(7), 1265–1269.

Beauchet, O., Annweiler, C., Dubost, V., Allali, G., Kressig, R. W., Bridenbaugh, S., … Herrmann, F. R. (2009). Stops walking when talking: a predictor of falls in older adults?

European Journal of Neurology, *16*(7), 786–795.

Bent, L. R., Inglis, J. T., & McFadyen, B. J. (2004). When is vestibular information important during walking? *Journal of Neurophysiology*, *92*(3), 1269–1275. https://doi.org/10.1152/jn.01260.2003 [doi]

Bent, L. R., McFadyen, B. J., Merkley, V. F., Kennedy, P. M., & Inglis, J. T. (2000). Magnitude effects of galvanic vestibular stimulation on the trajectory of human gait. *Neuroscience Letters*, *279*(3), 157–160.

Bingenheimer, K., Temprado, J. J., Harnagea, M., Bricot, N., Villani, P., & Berton, E. (2015). Effects of a light touch on fixed or mobile supports on gait parameters in visually restricted young adults. *Neuroscience Letters*, *589*, 176–180.

Boonsinsukh, R., Saengsirisuwan, V., Carlson-Kuhta, P., & Horak, F. B. (2012). A cane improves postural recovery from an unpracticed slip during walking in people with Parkinson disease. *Physical Therapy*, *92*(9), 1117.

Brach, J. S., Berlin, J. E., VanSwearingen, J. M., Newman, A. B., & Studenski, S. A. (2005). Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. *Journal of Neuroengineering and Rehabilitation*, *2*(1), 21.

Brooks, L. R. (1967). The suppression of visualization by reading. *The Quarterly Journal of Experimental Psychology*, *19*(4), 289–299.

Brown, L. A., McKenzie, N. C., & Doan, J. B. (2005). Age-dependent differences in the attentional demands of obstacle negotiation. *The Journals of gerontology.Series A, Biological Sciences and Medical Sciences*, *60*(7), 924–927. https://doi.org/60/7/924 [pii]

Bruening, D. A., & Ridge, S. T. (2014). Automated event detection algorithms in pathological gait. *Gait & Posture*, *39*(1), 472–477.

Bruijn, S. M., Meijer, O. G., Beek, P. J., & van Dieen, J. H. (2013). Assessing the stability of human locomotion: a review of current measures. *Journal of the Royal Society, Interface*, *10*(83), 20120999. https://doi.org/10.1098/rsif.2012.0999 [doi]

Buschman, T. J., & Kastner, S. (2015). From Behavior to Neural Dynamics: An Integrated Theory of Attention. *Neuron*, *88*(1), 127–144.

Caird, J. K., Johnston, K. A., Willness, C. R., Asbridge, M., & Steel, P. (2014). A meta-analysis of the effects of texting on driving. *Accident Analysis & Prevention*, *71*, 311–318.

Campbell, C. M., Rowse, J. L., Ciol, M. A., & Shumway-Cook, A. (2003). The Effect of

Cognitive Demand on Timed Up and Go Performance in Older Adults With and Without Parkinson Disease. *Journal of Neurologic Physical Therapy*, *27*(1), 2–7.

Cesari, M., Kritchevsky, S. B., Newman, A. B., Simonsick, E. M., Harris, T. B., Penninx, B. W., … Bauer, D. C. (2009). Added Value of Physical Performance Measures in Predicting Adverse Health-Related Events: Results from the Health, Aging and Body Composition Study. *Journal of the American Geriatrics Society*, *57*(2), 251–259.

Chang, V. C., & Do, M. T. (2015). Risk factors for falls among seniors: implications of gender. *American Journal of Epidemiology*, *181*(7), 521–531. https://doi.org/10.1093/aje/kwu268 [doi]

Chen, H.-C., Schultz, A. B., Ashton-Miller, J. A., Giordani, B., Alexander, N. B., & Guire, K. E. (1996). Stepping over obstacles: dividing attention impairs performance of old more than young adults. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *51*(3), M122.

Coren, S., & Porac, C. (1978). The validity and reliability of self-report items for the measurement of lateral preference. *British Journal of Psychology*, *69*(2), 207–211.

da Silva Costa, A. A., Manciopi, P. A. R., Mauerberg-deCastro, E., & Moraes, R. (2015). Haptic information provided by the “anchor system” reduces trunk sway acceleration in the frontal plane during tandem walking in older adults. *Neuroscience Letters*, *609*, 1–6.

Dickstein, R., & Laufer, Y. (2004). Light touch and center of mass stability during treadmill locomotion. *Gait & Posture*, *20*(1), 41–47.

Dickstein, R., Peterka, R. J., & Horak, F. B. (2003). Effects of light fingertip touch on postural responses in subjects with diabetic neuropathy. *Journal of Neurology, Neurosurgery, and Psychiatry*, *74*(5), 620–626.

Dingwell, J. B., Robb, R. T., Troy, K. L., & Grabiner, M. D. (2008). Effects of an attention demanding task on dynamic stability during treadmill walking. *Journal of Neuroengineering and Rehabilitation*, *5*(1), 12.

Do, M. T., Chang, V. C., Kuran, N., & Thompson, W. (2015). Fall-related injuries among Canadian seniors, 2005--2013: an analysis of the Canadian Community Health Survey. *Health Promotion and Chronic Disease Prevention in Canada: Research, Policy and Practice*, *35*(7), 99.

Edelstein, J. E. (2013). Assistive devices for ambulation. *Physical Medicine and Rehabilitation*

Clinics of North America, *24*(2), 291–303.

Gill, J., Allum, J. H. J., Carpenter, M. G., Held-Ziolkowska, M., Adkin, A. L., Honegger, F., & Pierchala, K. (2001). Trunk sway measures of postural stability during clinical balance tests effects of age. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *56*(7), M438--M447.

Goble, D. J. (2010). Proprioceptive acuity assessment via joint position matching: from basic science to general practice. *Physical Therapy*, *90*(8), 1176.

Gould, L., Cummine, J., & Borowsky, R. (2012). The cognitive chronometric architecture of reading aloud: semantic and lexical effects on naming onset and duration. *Frontiers in Human Neuroscience*, *6*, 287.

Government of Canada, P. H. A. of C. (2014). *Seniors' Falls in Canada: Second Report - Public Health Agency of Canada*. Retrieved from http://site.ebrary.com/lib/usask/docDetail.action?docID=10943172&ppg=1

Grabiner, M. D., & Troy, K. L. (2005). Attention demanding tasks during treadmill walking reduce step width variability in young adults. *Journal of Neuroengineering and Rehabilitation*, *2*(1), 25.

Grillner, S., & Wallen, P. (1985). Central pattern generators for locomotion, with special reference to vertebrates. *Annual Review of Neuroscience*, *8*(1), 233–261.

Gwin, J. T., Gramann, K., Makeig, S., & Ferris, D. P. (2011). Electrocortical activity is coupled to gait cycle phase during treadmill walking. *NeuroImage*, *54*(2), 1289–1296.

Hak, L., Houdijk, H., Steenbrink, F., Mert, A., van der Wurff, P., Beek, P. J., & van Dien, J. H. (2012). Speeding up or slowing down?: Gait adaptations to preserve gait stability in response to balance perturbations. *Gait & Posture*, *36*(2), 260–264.

Harley, C., Wilkie, R. M., & Wann, J. P. (2009). Stepping over obstacles: attention demands and aging. *Gait & Posture*, *29*(3), 428–432.

Harrington, M. E., Zavatsky, A. B., Lawson, S. E. M., Yuan, Z., & Theologis, T. N. (2007). Prediction of the hip joint centre in adults, children, and patients with cerebral palsy based on magnetic resonance imaging. *Journal of Biomechanics*, *40*(3), 595–602.

Hedayat, I., Moraes, R., Lanovaz, J. L., & Oates, A. R. (2017). Different haptic tools reduce trunk velocity in the frontal plane during walking, but haptic anchors have advantages over lightly touching a railing. *Experimental Brain Research*, 1–9.

Hof, A. L. (1996). Scaling gait data to body size. *Gait & Posture*, *3*(4), 222–223.

Hof, A. L. (2008). The "extrapolated center of mass" concept suggests a simple control of balance in walking. *Human Movement Science*, *27*(1), 112–125.

Hof, A. L., Gazendam, M. G. J., & Sinke, W. E. (2005). The condition for dynamic stability. *Journal of Biomechanics*, *38*(1), 1–8.

Holden, M., Ventura, J., & Lackner, J. R. (1994). Stabilization of posture by precision contact of the index finger. *Journal of Vestibular Research*, *4*(4), 283–285.

Horak, F. B. (2006). Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? *Age and Ageing*, *35 Suppl 2*, ii11. https://doi.org/35/suppl_2/ii7 [pii]

Horak, F. B., Nashner, L. M., & Diener, H. C. (1990). Postural strategies associated with somatosensory and vestibular loss. *Experimental Brain Research*, *82*(1), 167–177.

Horak, F. B., Shupert, C. L., & Mirka, A. (1989). Components of postural dyscontrol in the elderly: a review. *Neurobiology of Aging*, *10*(6), 727–738.

Jain, S., Gourab, K., Schindler-Ivens, S., & Schmit, B. D. (2013). EEG during pedaling: evidence for cortical control of locomotor tasks. *Clinical Neurophysiology*, *124*(2), 379–390.

Jeka, J. J., & Lackner, J. R. (1995). The role of haptic cues from rough and slippery surfaces in human postural control. *Experimental Brain Research*, *103*(2), 267–276.

Jeka, J. J., Ribeiro, P., Oie, K., & Lackner, J. R. (1998). The structure of somatosensory information for human postural control. *Motor Control*, *2*(1), 13–33.

Johansson, R. S., Landstro, U., & Lundstro, R. (1982). Responses of mechanoreceptive afferent units in the glabrous skin of the human hand to sinusoidal skin displacements. *Brain Research*, *244*(1), 17–25.

Kahneman, D. (1973). *Attention and effort*. Englewood Cliffs, N.J., Prentice-Hall: Englewood Cliffs, N.J., Prentice-Hall.

Kennedy, P. M., & Inglis, J. T. (2002). Distribution and behaviour of glabrous cutaneous receptors in the human foot sole. *The Journal of Physiology*, *538*(3), 995–1002.

Kerr, B., Condon, S. M., & McDonald, L. A. (1985). Cognitive spatial processing and the regulation of posture. *Journal of Experimental Psychology: Human Perception and Performance*, *11*(5), 617.

Kodesh, E., Falash, F., Sprecher, E., & Dickstein, R. (2015). Light touch and medio-lateral postural stability during short distance gait. *Neuroscience Letters*, *584*, 378–381.

Kuan, T.-S., Tsou, J.-Y., & Su, F.-C. (1999). Hemiplegic gait of stroke patients: the effect of using a cane. *Archives of Physical Medicine and Rehabilitation*, *80*(7), 777–784.

Lackner, J. R., & DiZio, P. (2005). Vestibular, proprioceptive, and haptic contributions to spatial orientation. *Annu.Rev.Psychol.*, *56*, 115–147.

Lackner, J. R., DiZio, P., Jeka, J., Horak, F., Krebs, D., & Rabin, E. (1999). Precision contact of the fingertip reduces postural sway of individuals with bilateral vestibular loss. *Experimental Brain Research*, *126*(4), 459–466.

Lajoie, Y., Teasdale, N., Bard, C., & Fleury, M. (1996). Upright standing and gait: are there changes in attentional requirements related to normal aging? *Experimental Aging Research*, *22*(2), 185–198.

Laurutis, V. P., & Robinson, D. A. (1986). The vestibulo-ocular reflex during human saccadic eye movements. *The Journal of Physiology*, *373*(1), 209–233.

Lee, S. W., Verghese, J., Holtzer, R., Mahoney, J. R., & Oh-Park, M. (2014). Trunk sway during walking among older adults: norms and correlation with gait velocity. *Gait & Posture*, *40*(4), 676–681.

Leva, P. De. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*, *29*(9), 1223–1230.

Little, C. E., & Woollacott, M. (2015). EEG measures reveal dual-task interference in postural performance in young adults. *Experimental Brain Research*, *233*(1), 27–37.

Lix, L. M., Keselman, J. C., & Keselman, H. J. (1996). Consequences of assumption violations revisited: A quantitative review of alternatives to the one-way analysis of variance F test. *Review of Educational Research*, *66*(4), 579–619.

Lövdén, M., Schellenbach, M., Grossman-Hutter, B., Krüger, A., & Lindenberger, U. (2005). Environmental topography and postural control demands shape aging-associated decrements in spatial navigation performance. *Psychology and Aging*, *20*(4), 683.

MacLellan, M. J., & Patla, A. E. (2006). Adaptations of walking pattern on a compliant surface to regulate dynamic stability. *Experimental Brain Research*, *173*(3), 521–530.

Maki, B. E. (1997). Gait changes in older adults: predictors of falls or indicators of fear? *Journal of the American Geriatrics Society*, *45*(3), 313–320.

Marigold, D. S., & Patla, A. E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge. *Journal of Neurophysiology*, *88*(1), 339–353. https://doi.org/10.1152/jn.00691.2001 [doi]

Marone, J. R., Patel, P. B., Hurt, C. P., & Grabiner, M. D. (2014). Frontal plane margin of stability is increased during texting while walking. *Gait & Posture*, *40*(1), 243–246.

Marteniuk, R. G. (1986). Information processes in movement learning: Capacity and structural interference effects. *Journal of Motor Behavior*, *18*(1), 55–75.

Massion, J. (1994). Postural control system. *Current Opinion in Neurobiology*, *4*(6), 877–887.

Mauerberg-Decastro, E. (2004). Developing an “ anchor” system to enhance postural control. *Motor Control*, *8*(3), 339.

Mauerberg-deCastro, E., Moraes, R., Tavares, C. P., Figueiredo, G. A., Pacheco, S., & Costa, T. D. A. (2014). Haptic anchoring and human postural control. *Psychology & Neuroscience*, *7*(3), 301.

Meyns, P., Bruijn, S. M., & Duysens, J. (2013). The how and why of arm swing during human walking. *Gait & Posture*, *38*(4), 555–562.

Mildren, R. L., Hare, C. M., & Bent, L. R. (2017). Cutaneous afferent feedback from the posterior ankle contributes to proprioception. *Neuroscience Letters*, *636*, 145–150.

Morton, S. M., & Bastian, A. J. (2004). Cerebellar control of balance and locomotion. *The Neuroscientist*, *10*(3), 247–259.

Nasar, J. L., & Troyer, D. (2013). Pedestrian injuries due to mobile phone use in public places. *Accident Analysis & Prevention*, *57*, 91–95.

Neale, V. L., Dingus, T. A., Klauer, S. G., Sudweeks, J., & Goodman, M. (2005). An overview of the 100-car naturalistic study and findings. *National Highway Traffic Safety Administration, Paper*, (05–0400).

Neumann, O., Heuer, H., & Sanders, A. F. (1987). Beyond capacity: A functional view of attention. *Perspectives on Perception and Action*, *14*, 361–394.

Ortega, J. D., Fehlman, L. A., & Farley, C. T. (2008). Effects of aging and arm swing on the metabolic cost of stability in human walking. *Journal of Biomechanics*, *41*(16), 3303–3308.

Patla, A. E. (1997). Understanding the roles of vision in the control of human locomotion. *Gait & Posture*, *5*(1), 54–69.

Patla, A. E., Prentice, S. D., Robinson, C., & Neufeld, J. (1991). Visual control of locomotion: strategies for changing direction and for going over obstacles. *Journal of Experimental*

Psychology: Human Perception and Performance, *17*(3), 603.

Perry, S. D., Santos, L. C., & Patla, A. E. (2001). Contribution of vision and cutaneous sensation to the control of centre of mass (COM) during gait termination. *Brain Research*, *913*(1), 27–34.

Pettersson, A. F., Olsson, E., & Wahlund, L. O. (2005). Motor function in subjects with mild cognitive impairment and early Alzheimer's disease. *Dementia and Geriatric Cognitive Disorders*, *19*(5–6), 299–304. https://doi.org/84555 [pii]

Punt, M., Bruijn, S. M., Wittink, H., & van Dieën, J. H. (2015). Effect of arm swing strategy on local dynamic stability of human gait. *Gait & Posture*, *41*(2), 504–509.

Rabin, E., Bortolami, S. B., DiZio, P., & Lackner, J. R. (1999). Haptic stabilization of posture: changes in arm proprioception and cutaneous feedback for different arm orientations. *Journal of Neurophysiology*, *82*(6), 3541–3549.

Rabin, E., Demin, A., Pirrotta, S., Chen, J., Patel, H., Bhambri, A., … DiFrancisco-Donoghue, J. (2015). Parkinsonian gait ameliorated with a moving handrail, not with a banister. *Archives of Physical Medicine and Rehabilitation*, *96*(4), 735–741.

Riley, M. A., Stoffregen, T. A., Grocki, M. J., & Turvey, M. T. (1999). Postural stabilization for the control of touching. *Human Movement Science*, *18*(6), 795–817.

Saltos, A., Smith, D., Schreiber, K., Lichenstein, S., & Lichenstein, R. (2015). Cell-phone related injuries in the united states from 2000–2012. *Journal of Safety Studies*, *1*(1), 1–14.

Schaefer, S., Schellenbach, M., Lindenberger, U., & Woollacott, M. (2015). Walking in high-risk settings: Do older adults still prioritize gait when distracted by a cognitive task? *Experimental Brain Research*, *233*(1), 79–88.

Schrager, M. A., Kelly, V. E., Price, R., Ferrucci, L., & Shumway-Cook, A. (2008). The effects of age on medio-lateral stability during normal and narrow base walking. *Gait & Posture*, *28*(3), 466–471.

Shumway-Cook, A., Brauer, S., & Woollacott, M. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. *Physical Therapy*, *80*(9), 896.

Shumway-Cook, A., & Woollacott, M. H. (2007). *Motor control: translating research into clinical practice*. Lippincott Williams & Wilkins.

Siu, K.-C., Catena, R. D., Chou, L.-S., van Donkelaar, P., & Woollacott, M. H. (2008). Effects of a secondary task on obstacle avoidance in healthy young adults. *Experimental Brain Research*, *184*(1), 115–120.

Siu, K. C., Chou, L. S., Mayr, U., Donkelaar, P., & Woollacott, M. H. (2008). Does inability to allocate attention contribute to balance constraints during gait in older adults? *The Journals of gerontology.Series A, Biological Sciences and Medical Sciences*, *63*(12), 1364–1369. https://doi.org/63/12/1364 [pii]

Springer, S., Giladi, N., Peretz, C., Yogev, G., Simon, E. S., & Hausdorff, J. M. (2006). Dual-tasking effects on gait variability: The role of aging, falls, and executive function. *Movement Disorders*, *21*(7), 950–957.

Strobach, T., Liepelt, R., Pashler, H., Frensch, P. A., & Schubert, T. (2013). Effects of extensive dual-task practice on processing stages in simultaneous choice tasks. *Attention, Perception, & Psychophysics*, *75*(5), 900–920.

Teasdale, N., Bard, C., LaRue, J., & Fleury, M. (1993). On the cognitive penetrability of posture control. *Experimental Aging Research*, *19*(1), 1–13.

Temprado, J. J., Zanone, P. G., Monno, A., & Laurent, M. (2001). A dynamical framework to understand performance trade-offs and interference in dual tasks. *Journal of Experimental Psychology: Human Perception and Performance*, *27*(6), 1303.

Tisserand, R., Robert, T., Dumas, R., & Cheze, L. (2016). A simplified marker set to define the center of mass for stability analysis in dynamic situations. *Gait & Posture*, *48*, 64–67.

Turvey, M. T., & Fonseca, S. T. (2014). The medium of haptic perception: A tensegrity hypothesis. *Journal of Motor Behavior*, *46*(3), 143–187.

Vallbo, a B., & Johansson, R. S. (1984). Properties of cutaneous mechanoreceptors in the human hand related to touch sensation. *Human Neurobiology*, *3*(1), 3–14.

Van Deursen, R. W. M., & Simoneau, G. G. (1999). Foot and ankle sensory neuropathy, proprioception, and postural stability. *Journal of Orthopaedic & Sports Physical Therapy*, *29*(12), 718–726.

Vance, R. C., Healy, D. G., Galvin, R., & French, H. P. (2015). Dual tasking with the timed “up & go” test improves detection of risk of falls in people with Parkinson disease. *Physical Therapy*, *95*(1), 95–102. https://doi.org/10.2522/ptj.20130386 [doi]

Vuillerme, N., Isableu, B., & Nougier, V. (2006). Attentional demands associated with the use of a light fingertip touch for postural control during quiet standing. *Experimental Brain*

Research, *169*(2), 232–236.

Wajda, D. A., Motl, R. W., & Sosnoff, J. J. (2013). Dual task cost of walking is related to fall risk in persons with multiple sclerosis. *Journal of the Neurological Sciences*, *335*(1), 160–163.

Wellmon, R., Pezzillo, K., Eichhorn, G., Lockhart, W., & Morris, J. (2006). Changes in dual-task voice reaction time among elders who use assistive devices. *Journal of Geriatric Physical Therapy (2001)*, *29*(2), 74–80.

Winter, D. A. (1995). Human balance and posture control during standing and walking. *Gait & Posture*, *3*(4), 193–214.

Woollacott, M. H., & Tang, P. F. (1997). Balance control during walking in the older adult: research and its implications. *Physical Therapy*, *77*(6), 646–660.

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), 1–14.

Worden, T. A., Mendes, M., Singh, P., & Vallis, L. A. (2016). Measuring the effects of a visual or auditory Stroop task on dual-task costs during obstacle crossing. *Gait & Posture*, *50*, 159–163.

Worden, T. A., & Vallis, L. A. (2014). Concurrent Performance of a Cognitive and Dynamic Obstacle Avoidance Task: Influence of Dual-Task Training. *Journal of Motor Behavior*, *46*(5), 357–368.

Worden, T. A., & Vallis, L. A. (2015). Stability control during the performance of a simultaneous obstacle avoidance and auditory Stroop task. *Experimental Brain Research*, 1–10.

Wright, D. L., & Kemp, T. L. (1992). The dual-task methodology and assessing the attentional demands of ambulation with walking devices. *Physical Therapy*, *72*(4), 5.

Yang, F., & Pai, Y.-C. (2014). Can sacral marker approximate center of mass during gait and slip-fall recovery among community-dwelling older adults? *Journal of Biomechanics*, *47*(16), 3807–3812.

Yogev-Seligmann, G., Rotem-Galili, Y., Mirelman, A., Dickstein, R., Giladi, N., & Hausdorff, J. M. (2010). How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. *Physical Therapy*, *90*(2), 177.

Yogev-Seligmann, G., Hausdorff, J. M., & Giladi, N. (2012). Do we always prioritize balance when walking? Towards an integrated model of task prioritization. *Movement Disorders*, *27*(6), 765–770.

Youdas, J. W., Kotajarvi, B. J., Padgett, D. J., & Kaufman, K. R. (2005). Partial weight-bearing gait using conventional assistive devices. *Archives of Physical Medicine and Rehabilitation*, *86*(3), 394–398.

Young, P. M. M., Wilken, J. M., & Dingwell, J. B. (2012). Dynamic margins of stability during human walking in destabilizing environments. *Journal of Biomechanics, 45*(6), 1053–1059.

Appendix A

WOLTERS KLUWER HEALTH LICENSE
TERMS AND CONDITIONS

Jun 01, 2017

This Agreement between University of Saskatchewan -- Aaron Awdhan ("You") and Wolters Kluwer Health ("Wolters Kluwer Health") consists of your license details and the terms and conditions provided by Wolters Kluwer Health and Copyright Clearance Center.

License Number	4120540060178
License date	Jun 01, 2017
Licensed Content Publisher	Wolters Kluwer Health
Licensed Content Publication	WK Health Book
Licensed Content Title	Motor Control
Licensed Content Author	
Licensed Content Date	Jan 1, 2016
Type of Use	Dissertation/Thesis
Requestor type	academic/educational
Format	print and electronic
Portion	figures/tables/illustrations
Number of figures/tables/illustrations	2
The ID numbers of the figures/tables/illustrations are...	FIGURE 1.2 Factors within the individual, the task, and the environment affect the organiza- tion of movement. Factors within the individual include the interaction of perception, cognition, and action (motor) systems. Environmental con- straints on movement are divided into regulatory and nonregulatory factors. Finally, attributes of the task contribute to the organization of func- tional movement. FIGURE 7.4 Conceptual model representing the many components of postural control that have been studied by researchers. Postural control is not regulated by a single sys- tem, but emerges from the interaction of many systems.
Will you be translating?	no
Reusing current or a previous edition	current edition
Circulation/distribution	5
Order reference number	
Title of your thesis / dissertation	ASSESSING THE ATTENTIONAL DEMANDS OF ADDING HAPTIC INPUT DURING OVERGROUND WALKING
Expected completion date	Jul 2017
Estimated size (number of pages)	110
Requestor Location	University of Saskatchewan

Appendix B

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB)

Certificate of Approval

PRINCIPAL INVESTIGATOR	DEPARTMENT	Bio #
Alison Oates	Kinesiology	16-139

INSTITUTION(S) WHERE RESEARCH WILL BE CARRIED OUT
College of Kinesiology
87 Campus Drive
Saskatoon SK S7N 5B2

STUDENT RESEARCHER(S)
Aaron Awdhan

FUNDER(S)
GOVERNMENT OF SASKATCHEWAN
UNIVERSITY OF SASKATCHEWAN

TITLE
Protocol : Assessing the Attentional Demands of Adding Haptic Input during Walking

ORIGINAL REVIEW DATE	APPROVED ON	APPROVAL OF	EXPIRY DATE
02-Jun-2016	13-Jun-2016	revised Application for Human Research Biomedical Research Review (rec'd June 13, 2016) response to Notice of Ethical Review (rec'd June 13, 2016) revised Participant Information and Consent Form (rec'd June 13, 2016) revised PAWS Announcement (rec'd June 13, 2016) revised Recruitment Flyer (rec'd June 13, 2016) Attention Questionnaire (rec'd May 11, 2016) Screening Questionnaire (rec'd May 11, 2016)	12-Jun-2017

Delegated Review ☒ Full Board Meeting ☐

CERTIFICATION

The study is acceptable on scientific and ethical grounds. The principal investigator has the responsibility for any other administrative or regulatory approvals that may pertain to this research study, and for ensuring that the authorized research is carried out according to governing law. This approval is valid for the specified period provided there is no change to the approved protocol or consent process.

FIRST TIME REVIEW AND CONTINUING APPROVAL

The University of Saskatchewan Biomedical Research Ethics Board reviews above minimal studies at a full-board (face-to-face) meeting. If a protocol has been reviewed at a full board meeting, a subsequent study of the same protocol may be reviewed through the delegated review process. Any research classified as minimal risk is reviewed through the delegated (subcommittee) review process. The initial Certificate of Approval includes the approval period the REB has assigned to a study. The Status Report form must be submitted within one month prior to the assigned expiry date. The researcher shall indicate to the REB any specific requirements of the sponsoring organizations (e.g. requirement for full-board review and approval) for the continuing review process deemed necessary for that project. For more information visit http://research.usask.ca/for-researchers/ethics/index.php.

REB ATTESTATION

In respect to clinical trials, the University of Saskatchewan Research Ethics Board complies with the membership requirements for Research Ethics Boards defined in Part 4 of the Natural Health Products Regulations and Part C Division 5 of the Food and Drug Regulations and carries out its functions in a manner consistent with Good Clinical Practices. Members of the Bio-REB who are named as investigators, do not participate in the discussion related to, nor vote on such studies when presented to the Bio-REB. This approval and the views of this REB have been documented in writing. The University of Saskatchewan Biomedical Research Ethics Board has been approved by the Minister of Health, Province of Saskatchewan, to serve as a Research Ethics Board (REB) for research studies involving human participants under section 29 of The Health Information Protection Act (HIPA).

Ildiko Badea, PhD., Vice-Chair
University of Saskatchewan
Biomedical Research Ethics Board

Please send all correspondence to:

Research Services and Ethics Office
Room 223 Thorvaldson Building
110 Science Place
Saskatoon, SK Canada S7N 5C9

Appendix C

Screening Questionnaire (to be answered via email/over the phone/in person)

Date of screening: _______________

1. Are you between the ages of 18 – 30 years old?

Yes / No

2. Do you have any current lower/upper limb injuries (i.e., broken bones, ligament tears/repairs, muscle tears, etc)?

Yes / No

If yes, please describe:

____

____

3. Do you have any neurological and/or musculoskeletal impairments that may affect your standing and walking balance?

Yes / No

4. Do you have any visual impairments that cannot be corrected with corrective eyewear?

Yes / No

5. Do you have any reduced or lost sensation (feeling) in your lower/upper extremities?

Yes / No

6. Do you have any hearing impairments that may affect your ability to distinguish a high pitch tone from a low pitch tone?

Yes / No

If participant is eligible (Yes to Q1, NO to Q2-6), schedule testing and assign participant ID:

____

If participant is not eligible, inform individual of ineligibility delete/shred this form.

Appendix D

Research participant information and consent form

Assessing the attentional demands of adding haptic input during walking

Principal Investigator:	Alison Oates, PhD College of Kinesiology University of Saskatchewan Ph: 306-966-1080 Email: alison.oates@usask.ca
Co-investigator:	Aaron Awdhan, MSc Candidate College of Kinesiology University of Saskatchewan Email: aaron.awdhan@usask.ca

Introduction

You are invited to take part in this research study because you are a healthy young adult between the ages of 18 – 30 years old. Your participation is voluntary and it is your decision to take part or not take part in this study. If you choose to participate, at any time you are able to withdraw without any reason(s) for your decision. If you do choose to not participate, your academic standing, if enrolled at the University of Saskatchewan will not be affected.

Please take time to read the following information carefully. You can ask the study investigators to explain any information that you do not clearly understand and may ask as many questions as you need. In addition, feel free to discuss this with your family, friends, or family physician before you decide to proceed.

Who is conducting the study?

This study is being conducted by a Masters student, Aaron Awdhan. Neither the institution nor any of the investigators, staff or students will receive and direct financial benefit from conducting this study.

Why is this study being done?

In the older adult population (65 years +) falls are a serious concern to both the person and health care system. Good balance control may help prevent falls. Balance control relies on the sensory information from our body and motor signals to our muscles to maintain stability and prevent a fall. Walking requires some of our attention. Adding other tasks during walking (i.e., stepping up a curb and/or talking to someone else) can negatively affect walking performance, which may increase fall-risk in older adults. Adding haptic information, in the forms of anchoring (dragging light weights of ~ 125 g) and light touch on a railing provides sensory information about where your body is in space may improve walking balance. The purpose of this research project is to study the attention needed to use haptic modalities during overground walking and while stepping over an obstacle and/or responding to a cognitive task. The effect of haptic input on balance control during walking will be studied in young, healthy adults (~ 30 participants).

Who can participate in this study?

You are eligible to participate in this study if you are between the ages of 18 – 30 years with no existing medical conditions that may affect your ability to walk safely over ground (i.e., current musculoskeletal injuries, neurological impairments) and no visual, hearing, or sensory (loss of feeling in limbs) impairments. You must be able to walk for at least 10 meters without the use of any aid, such as a walker.

What does this study involve?

This study involves performing a series of walking trials during one data collection session, scheduled at your convenience in the Biomechanics of Balance and Movement (BBAM) laboratory in the College of Kinesiology. You will be asked to walk normally on a flat surface, with or without an obstacle (approximately curb height) and/or a cognitive task (responding to the pitch of a tone you hear over a speaker).

With your permission, we may take photos and/or video recordings to be used for research and/or teaching purposes. To be sure your identity remains confidential, we will cover or remove any identifying parts of the image, such as your. You will be asked if photographs or video recordings can be made of you before doing so. You can request to not be photographed or videotaped at any time in the session. Even if you originally agreed to be video-recorded or photographed, you can request visual recordings to be stopped.

Main data collection

At the main data collection visit, we will measure your body dimensions, such as height and weight. We will ask for your birth month and year to record your age. You will then be asked to step up onto a platform with the foot that you would use to step up onto a small stool. Next, you will be asked to fill out a questionnaire related to factors that may influence your ability to divide your attention between tasks. Afterwards, you will be provided with an opportunity to become comfortable lightly touching a railing and using the haptic while walking.

After you feel comfortable walking and either touching the railing or using the haptic anchors, you will be outfitted with reflective spheres (markers) over various anatomical landmarks and a wireless microphone to record your response for trials where a verbal task is present. These reflective markers will allow us to measure body movement using the VICON 3D motion captures system. The markers will be placed on your shoes, on medical grade plastic forms that will be secured to your lower legs, thighs, and pelvis, and to their shoulders, elbows, wrists, dominant fingertip, and on a headband worn around the head. The markers will be secured in place using hypo-allergenic, double-sided wig tape and/or fabric wrap and strips. It may take up to 30 minutes to secure all of the kinematic markers. To ensure the markers can be placed on specific areas of your body, you will be asked to wear shorts, a form-fitting t-shirt that does not have sleeves, and comfortable walking shoes. Ideally this clothing is non-reflective. Changing rooms are available outside of the laboratory and data collection will take place in a closed laboratory with only yourself and the researcher(s) present. If you do not have shorts and/or a form fitting t-shirt without sleeves, either/both will be provided for your use during the research study.

For this study you will be asked to walk along a level walkway normally at your own comfortable pace. The haptic information conditions will involve either lightly touching a railing (~ 1 N of force) that measures the force under the index finger of your dominant hand or holding onto strings attached to small bean bags (~ 125 g each) that will rest on the floor as you drag them behind you during walking. In addition to these conditions, there will be trials when you will step over an obstacle that is approximately curb height and/or respond to a verbal reaction time task. For the, verbal reaction time task, you will be asked to say the word ‘high’ when you hear a high pitched tone or ‘low’ when you hear a low pitched tone as soon as possible. Each walking condition will be performed a minimum of eight times in a random order for a total approximately 96 trials. You may rest in between walking trials as needed. The study should take 2 – 3 hours to complete.

After you have finished all of the walking trials, all of the reflective markers and the wireless microphone will be removed from your body.

What are the benefits of participating in this study?

If you choose to participate in this study, there will be no direct benefit to you. It is hoped that the information gained from this study can be used in the future to improve balance control during walking. At the conclusion of the study, if you agree to provide contact information, you will receive feedback on your performance in relation to the other participants in the research study.

What are the possible risks and discomforts?

There are some risks and discomforts related to the study procedures. There is a chance that you could contact the obstacle; however, the obstacle will not be secured to the ground and will fall over if you contact it with your foot. You may experience some fatigue and soreness as a result of the walking, which is temporary and should disappear in a few days. Additionally, the adhesive tape and fabric wrap and strips used to secure the VICON reflective markers are hypoallergenic but may cause a mild, temporary skin irritation similar to a band-aid that should disappear within a few days.

What happens if I decide to withdraw?

Your participation in this research is voluntary. You may withdraw from this study at any time. You do not have to provide a reason. Your future academic status and/or relationships with the University of Saskatchewan will not be affected. If you choose to enter the study and then decide to withdraw at a later time, all data collected about you during your enrollment will be retained for analysis.

What happens if something goes wrong?

In the unlikely event of an adverse effect arising related to the study procedures, trained staff will be available throughout the conduct of the study that can respond immediately. Necessary medical treatment will be made available at no additional cost to you. By signing this document, you do not waive any of your legal rights against the sponsor, investigators or anyone else.

What will the study cost me?

You will not be charged for any research-related testing. You will not be paid for participating in this study or reimbursed for any expenses related to participating in this study (e.g., parking).

Will my participation be kept confidential?

In Saskatchewan, the Health Information Protection Act (HIPA) protects the privacy of your personal health information. Your privacy will be respected as your name will not be attached to any information nor mentioned in any study report, nor be made available to anyone except the research team. It is the intention of the research team to publish results of this research in scientific journals and to present the findings at related conferences, workshops or teaching opportunities, but your identity will not be revealed. Only the researchers are involved in recruitment, and names and contact information of prospective participants are kept confidential.

Your contact information (email or mailing address) will be requested to be able to contact you regarding future studies and/or to send you the results of the study once it has finished. Your contact information will be kept in a secure location separate from your data. Group emails will not be sent, nor mass mail outs, to protect your identity. Note that you are not required to provide this information should you not want to be contacted following completion of the study.

Many times, video or photographic data is used to explain the research protocol and results. If you agree to have your video and/or photograph taken during data collection, any identifying aspects (i.e., your face) will be concealed to maintain your anonymity if used for teaching and/or research purposes. If you do not wish to be photographed or videotaped during the research study you are still able to participate without penalty.

All digital data will be recorded on password-protected digital media and all other data will be stored in locked rooms (PAC 333 & 355) at the University of Saskatchewan for a minimum of five years after the study is completed. Long-term data retention is under responsibility of the Primary Investigator. This data will only be accessible to the researchers. All participant data will be coded with a participant number at the time of collection and this code will be used during all subsequent analysis. Once all data has been processed and analyzed, all data will be destroyed appropriately when it is no longer needed. Electronic data will be deleted permanently, and paper documents will be destroyed through confidential shredding.

Who do I contact if I have questions about this study?

If you have any questions or desire further information about this study before, during or after participation, you can contact Alison Oates, PhD at 306-966-1080.

If you have concerns about your rights as a research participant and/or your experiences while participating in this study, contact the Chair of the University of Saskatchewan Research Ethics

Board at 306-966-6538. The Research Ethics Board is a group of individuals (scientists, physicians, ethicists, lawyers and members of the community) that provide an independent review of human research studies. This study has been reviewed and approved on ethical grounds by the University of Saskatchewan Research Ethics Board.

TITLE: Assessing the attentional demands of adding haptic input during walking

Consent to participate

- I have read (or someone has read to me) the information in this consent form
- I understand the purpose and procedures and the possible risks and benefits of the study
- I was given sufficient time to think about it
- I had the opportunity to ask questions and have received satisfactory answers
- I am free to withdraw from this study at any time for any reason and the decision to stop taking part will not affect my academic standing
- I agree to follow the investigator's instructions and will tell the investigators immediately if I feel I have any unexpected or unusual side effects
- I have been informed that this study will provide no benefits to me
- I give permission for the use and disclosure of my de-identified personal health information collected for research purposes described in this form
- I understand that by signing this document I do not waive any of my legal rights
- I will be given a signed and dated copy of this consent form
- I give permission to be contacted, using the contact information provided below, for future studies conducted by this research group if they suspect I would be eligible

  Yes ☐ No ☐

- I give permission to be contacted, using the contact information provided below, so that the investigators can provide results of the study

  Yes ☐ No ☐

- I agree to be photographed and/or have video taken of me during this study

  Yes ☐ No ☐

Contact information (to be used to send feedback based on your participation in the study and/or contact you about future studies according to your responses above):

Email address: ____

Mailing address:

____

I agree to participate in this study:

Printed name of participant: ___________________________ Date: ________________

Signature of participant: ____

Appendix E

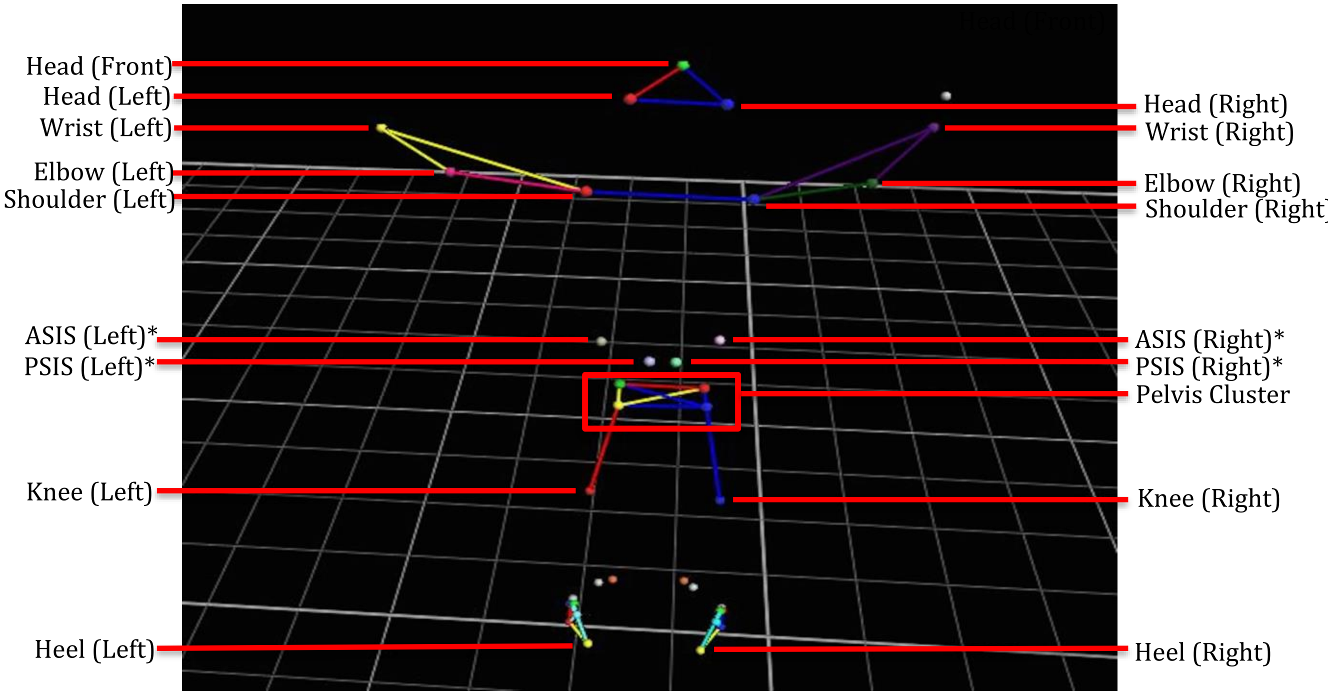

Figure Appendix E.1. – Posterior view of kinematic marker set up
* Indicates calibration marker

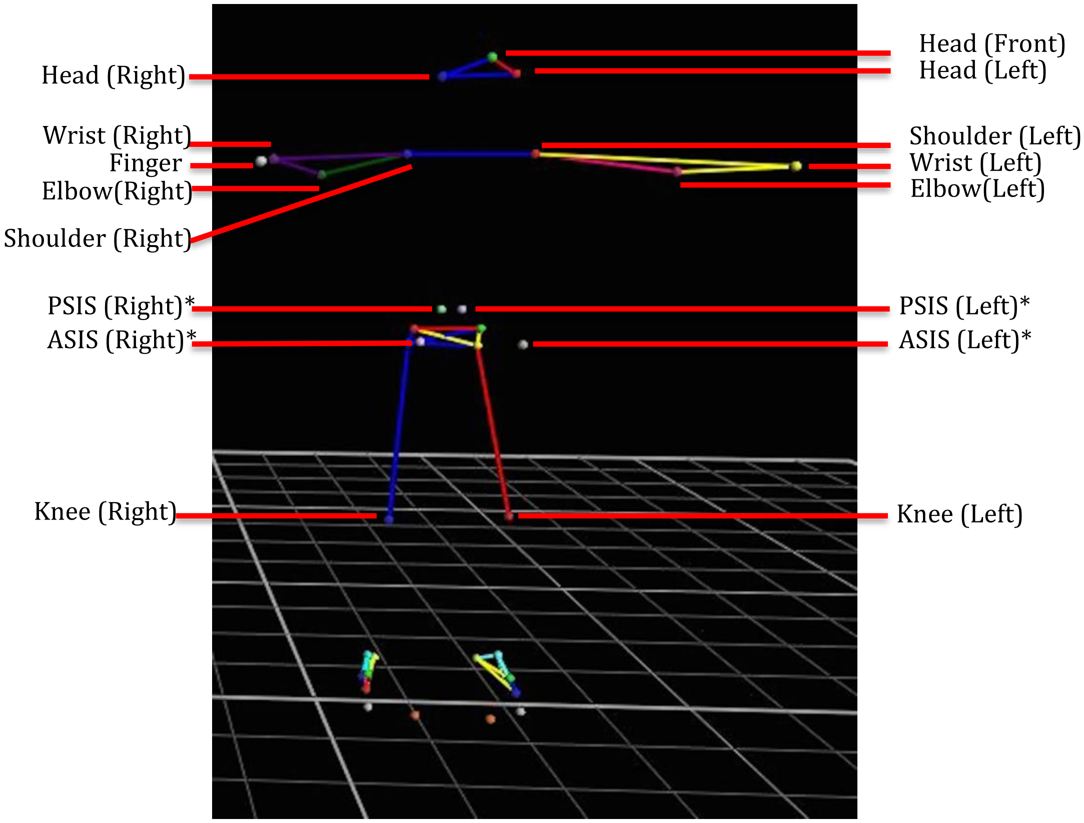

Figure Appendix E.2. – Anterior view of kinematic marker set up
* Indicates calibration marker

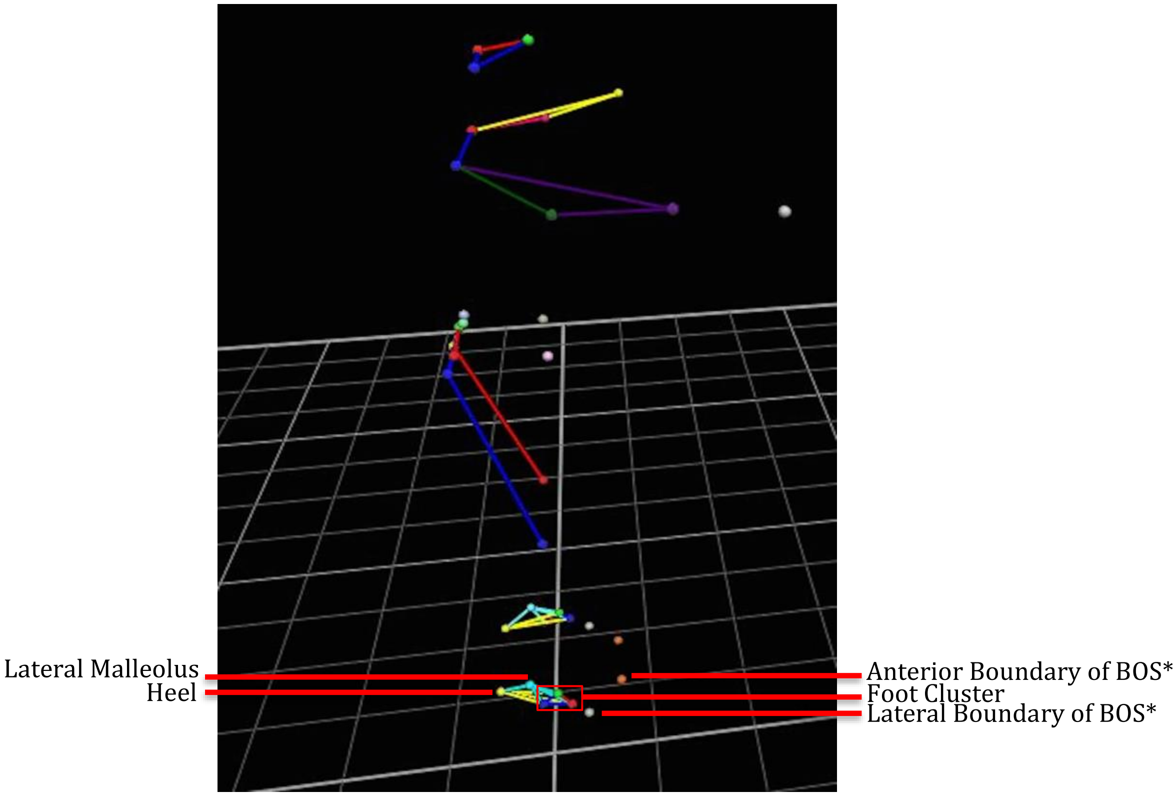

Figure Appendix E.3. – Lateral view of kinematic marker set up displaying foot cluster and markers
* Indicates calibration marker

Appendix F

This section contains additional analyses examining trunk movement variables to assess balance control associated with dynamic stability.

Methods:

Changes in dynamic stability may also be examined using trunk movement because it might illustrate movement of COM position. As outlined in section 1.1.4. trunk mass represents a significant proportion of whole-body COM that might provide an understanding of balance control. Trunk angle was calculated in the frontal plane because ML movement of the trunk has been shown to provide an important role in balance control related to dynamic stability (Gill et al., 2001). Calculation of trunk angle was determined by assessing the movement of the trunk COM to the midpoint of the hip joint centres, both expressed in the global coordinate systems, relative to the global vertical across the stride. ML trunk range of motion (ROM) was determined by obtaining the absolute difference between the maximum and minimum trunk angle values within the stride. The derivative of the trunk angle was obtained to calculate trunk velocity in the frontal plane. To determine the peak ML trunk velocity, the maximum value was determined within the stride. Normalization of the ML trunk ROM and peak ML trunk velocity was performed by dividing the values with the normalized stride velocity within trial. Normalization was performed to account for the velocity of the individual because differences in stride velocity can inherently affect movement of the trunk (i.e., increased stride velocity would result in increased trunk movement) (Hedayat et al., 2017; Lee, Verghese, Holtzer, Mahoney, & Oh-Park, 2014). With increased dynamic stability, it is expected normalized ML trunk ROM and peak ML trunk velocity would decrease, while decreased dynamic stability would be associated with increased normalized ML trunk ROM and peak ML trunk velocity. Dynamic stability during walking with an added VRT task was hypothesized to be increased with haptic modality use, so it is expected normalized ML trunk ROM and peak ML trunk velocity will be significantly decreased when using LT and haptic anchors compared to baseline walking.

Results:

There was no significant main effect of condition for normalized ML trunk ROM for LT (F (1,21) = 0.732, p = 0.402) and anchor use (F (1,21) = 3.526, p = 0.074) compared to baseline walking. There was no significant main effect of condition for normalized peak ML trunk velocity for LT (F (1,21) = 1.882, p = 0.185) and anchor use (F (1,21) = 3.145, p = 0.091) compared to baseline walking.

There was no significant main effect for presence of VRT task for normalized ML trunk ROM in the 2 × 2 RM ANOVA analysis that examines LT use (F (1,21) = 1.955, p = 0.177). There was no significant main effect for presence of VRT task for normalized ML trunk ROM in the 2 × 2 RM ANOVA analysis that examines anchors use (F (1,21) = 0.050, p = 0.825).

There was no significant main effect for presence of VRT task for normalized peak ML trunk velocity in the 2 × 2 RM ANOVA analysis that examines LT use (F (1,21) = 0.650, p = 0.429). There was no significant main effect for presence of VRT task for normalized peak ML trunk velocity in the 2 × 2 RM ANOVA analysis that examines anchors use (F (1,21) = 0.995, p = 0.330).

Discussion:

When assessing balance control with normalized ML trunk ROM and peak ML trunk velocity measures, my findings indicated no significant differences between haptic modalities to baseline walking and no difference between presence of a VRT task to no VRT task. These findings suggest that dynamic stability was similar with and without a VRT task and similar between haptic modalities and baseline walking. These findings related to haptic modalities compared to baseline walking were different from a similar study in our lab (Hedayat et al., 2017). Hedayat *et al*. examined anchors, LT, pretending (i.e., mimicking arm position) to use anchors, pretending to use LT and baseline walking. Hedayat *et al*. found haptic modalities improved dynamic stability by decreasing peak ML trunk velocity compared to baseline walking, with a significantly greater decrease seen with anchors compared to LT. Furthermore, pretending to use the railing had a similar decrease in peak ML trunk velocity compared to LT, while pretending to use anchors compared to anchor use did not (Hedayat et al., 2017). These findings suggest anchors improved dynamic stability due to sensory input and improvements with LT were possibly due to mechanical changes from arm placement (Hedayat et al., 2017).

My findings may differ from Hedayat et al. (2017) and other studies that calculated trunk-based measures (i.e., ML trunk SD, ML trunk ROM, peak ML trunk velocity) (da Silva Costa et al., 2015; Hedayat et al., 2017; Mauerberg-deCastro et al., 2014) likely due to differences in equipment. In previous studies (da Silva Costa et al., 2015; Hedayat et al., 2017; Mauerberg-deCastro et al., 2014), an inertial sensor was used to calculate trunk measures, which were positioned on the participants' sternum (Hedayat et al., 2017) or 7th cervical vertebrae (da Silva Costa et al., 2015; Mauerberg-deCastro et al., 2014). Use of an inertial sensor can affect values calculated, if the orientation is not aligned properly on the body (i.e., turned or twisted). My calculation of trunk measures utilized the trunk COM position to the midpoint of the hip joint centres with respect to the global vertical. The trunk COM position was chosen instead of the midpoint between the shoulders to approximate the position where the trunk cluster would be. The differences in equipment may explain why trunk measures in my findings were lower than those reported in studies that used an inertial sensor, as our trunk measures examined movement relative to another point on the body (i.e., midpoint of the hips).

One limitation of this study is calculating trunk-based measures without using a marker cluster on the sternum to approximate where the inertial sensor would be, as outlined in previous studies (Hedayat et al., 2017). An alternative would have been to use an inertial sensor on the participant, in addition to the passive markers to calculate trunk-based measures in a similar manner to previous studies (da Silva Costa et al., 2015; Hedayat et al., 2017; Mauerberg-deCastro et al., 2014) to see if similar results would be obtained (Figure F.1. for outline of where a trunk cluster could be placed on the participant).

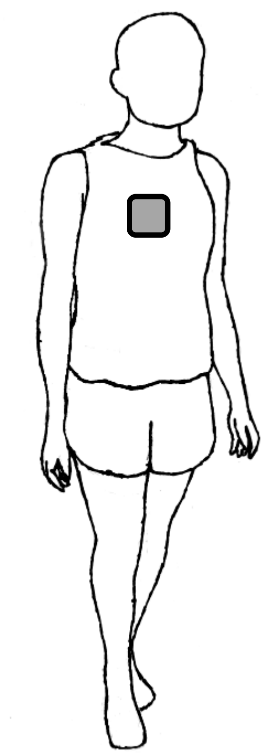

Figure Appendix F.1. – View of proposed location for trunk cluster (grey square) (Adapted from Hedayat et al., 2017)