PREDICTIVE BALANCE CONTROL DURING BACKWARD WALKING AND EFFECTS OF A HAPTIC INPUT BASED INTERVENTION ON PREDICTIVE BALANCE CONTROL DURING WALKING

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By

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ABSTRACT

Background
Falls are a leading cause of injuries and hospitalizations in individuals globally and in Canada. Even though falls can occur during any activity, a majority of falls occur during walking. Understanding and improving balance control during walking can help reduce falls. One way of improving balance control may be to add haptic input during walking and including backward and tandem walking in gait training programs.
Purpose
The overall purpose of this study was to examine the balance control and sensorimotor integration during backward walking as well as study the effects of an intervention consisting of backward and tandem walking on balance control in healthy adults.
Methods and results
Study one: Test-retest reliability, standard error of measurement, and minimal detectable change were computed for spatiotemporal and balance control measures for forward, backward, and tandem walking for fifteen healthy adults. The results demonstrated moderate to excellent reliability for all spatiotemporal and balance measures but low to poor reliability for variability measures for forward, backward, and tandem walking.
Study two: Differences in spatiotemporal and balance control measures between forward and backward walking and the correlation of backward walking velocity with biomechanical balance control measures during forward and tandem walking were examined in fifty-five healthy adults. Backward walking was significantly different in terms of spatiotemporal and balance control measures compared to forward walking. Participants walked significantly slower and with a significant reduction in relative double support time during backward walking compared to forward walking. Step length and anteroposterior margin of stability were significantly reduced, and step width and mediolateral margin of stability were significantly increased during backward walking compared to forward walking. Backward walking was also significantly more variable compared to forward walking. Step length, step width, and anteroposterior and mediolateral margins of stability were significantly more variable during backward walking compared to forward walking. Velocity during backward walking showed a significant positive correlation with anteroposterior margin of stability and velocity during forward walking and a significant negative correlation with step length variability during forward walking.
Study three: The effects of vision and haptic input added with haptic anchors during backward walking was examined in 55 healthy adults. It was observed that walking backward with eyes closed significantly changed spatiotemporal and balance control measures compared to walking with eyes open. Participants walked slower, with an increased amount of double support time, reduced step length, and increased step width when walking backward with eyes closed compared to walking with eyes open. Variability of step width and margin of stability in the anteroposterior and mediolateral directions were also significantly higher when

walking backward with eyes closed. Margin of stability in the mediolateral direction was significantly lower when walking backward with the haptic anchors compared to walking without haptic anchors. An interaction between vision and haptic input revealed that step length was significantly lower when walking backward using the haptic anchors compared to walking without haptic anchors in the eyes open condition.

Study four: This study examined the effects of a six-week (three days/week) intervention on balance control during forward, backward, and tandem walking in a total of forty-five healthy adults. Fifteen participants completed the intervention using haptic anchors, another fifteen completed the same intervention without the haptic anchors, and a control group of fifteen participants did not complete the intervention. The intervention consisted of performing ten trials each of backward and tandem walking with eyes closed over a distance of ten meters in random order at the participants' preferred speed. During forward walking, change in step length variability was significantly higher in the eyes closed condition compared to the eyes open condition. During backward walking, velocity, %DS, and step length change scores were significantly higher in the eyes closed condition compared to eyes open and the change score for AP_MOS was significantly higher in the eyes closed condition compared to the eyes open condition only for the group that trained without the haptic anchors. During tandem walking, change score for ML_MOS was significantly lower in the eyes closed condition compared to the eyes open condition. No significant effects of the intervention were observed on any measures for forward, backward, and tandem walking except the AP_MOS change scores in the group that performed the intervention without using the haptic anchors.

Conclusion

This thesis provided novel evidence on the reliability of spatiotemporal and balance control measures across three different walking styles. The findings provide support in favour of using MOS measures as well as backward walking to assess mobility and integrity of the balance control system. The insignificant effects of the haptic input based intervention warrants further research on the long-term use of haptic anchors to improve balance control.

PREFACE AND AUTHOR CONTRIBUTION

I, Kirat Shukla, was the primary author of all chapters within this thesis. Chapters three to six represent manuscripts that will be submitted for publication in peer- reviewed journals. Author contributions have been discussed and approved by the research advisory committee. I designed the research questions, collected and analyzed data, and prepared the manuscript for each study under the supervision of Drs. Alison Oates, Joel Lanovaz, Jon Farthing, and Gary Linassi. Results from study 2 (chapter four) were presented at the International Society of Biomechanics conference (ISB) that was held virtually in July 2021. Results from study 4 (chapter six) were presented at the International Society of Posture and Gait Research (ISPGR) conference held in Edinburgh, Scotland in July 2019.

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DEDICATION

To mom (Nisha Shukla), dad (Hariom Shukla) and my brother (Bihag Shukla). For all your support and sacrifices.

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Conceptual definitions

Fall: An event that results in a person coming to rest inadvertently on the ground or floor or other lower level (World Health Organization) [1].

Centre of mass: The weighted average of the centre of mass of all body segments [2].

Base of support: The area beneath the feet that includes all the points of contact between the foot/feet and the surface on which an individual is standing or walking [3].

Balance: An instantaneous state where the position of the centre of mass is within the base of support [3].

Balance or postural control: The ability of the human system to achieve a state of balance when performing any activity [3].

Stability: The intrinsic ability of a body to resist becoming unbalanced from a state of balance or moving from an unbalanced to a balanced state [4].

List of common abbreviations used

%DS: Relative amount of time in the double support phase during one full gait cycle
AP_MOS: Anteroposterior margin of stability
AP_MOS_SD: Variability of anteroposterior margin of stability
BOS: Base of support
COM: Centre of mass
HA: Haptic anchors
ML_MOS: Mediolateral margin of stability
ML_MOS_SD: Variability of mediolateral margin of stability
MOS: Margin of stability
nSL: Normalized step length
nSV: Normalized stride velocity
QOL: Quality of life
SD: Standard deviation
SL_SD: Step length variability
SW: Step width
SW_SD: Step width variability
xCOM: Extrapolated centre of mass

List of appendices

Formulas and calculations used in this thesis

Normalised stride velocity (nSV) = stride velocity/$\sqrt{H*g}$)
where H = participant's leg length, and g = 9.81 m/s^2 [5]

Normalised step length (nSL) = step length/participants' leg length [5]

Margin of stability (MOS) = Distance between edge of BOS and xCOM where BOS = base of support, and xCOM = extrapolated centre of mass [6]

ICC (Intraclass correlation coefficient) = Two-way mixed analysis of variance of absolute agreement with average of the measures (3,k) [7]

Standard error of measurement (SEM) = SD * $\sqrt{1-ICC}$
where SD = pooled standard deviation, and ICC = intraclass correlation coefficient [8, 9]

Minimal detectable change (MDC95) = SEM * $\sqrt{2}$ * 1.96)
where SEM = standard error of measurement [8, 9]

Effects sizes
Cohen's d (for paired samples t-tests) = mean difference/standard deviation of mean difference) [10]
r (for Wilcoxon signed rank test) = Z score/$\sqrt{total\ number\ of\ observations}$ [11]

1 Introduction

This chapter introduces the concepts involved in balance control during walking, the functions of the balance control system, and the overall purpose of this thesis. Each of the concepts discussed in this section are discussed in further detail in the literature review section (Chapter 2) of this thesis.

Walking is an essential requirement for completing activities of daily living and having an increased quality of life [12]. Humans can adapt to walk in different environments such as their homes, parks, malls, and across hard, soft, slippery wet, and uneven surfaces such as hardwood, concrete, ice, snow, grass, and sand. Walking without falling appears exceptionally coordinated and automatic, something that individuals accomplish without much attention. At a neural level, an overly complex set of processes and interactions materialize to generate synergistic muscle activity and adapt to imposed demands by perturbations.

Human walking is bipedal, i.e., walking on two lower limbs. Bipedal walking has advantages over quadrupedal walking (four limbs), including non-reliance on the upper extremities to move from one place to another. The upper extremities are free to concurrently accomplish tasks such as talking on a phone, holding a cane, pushing a stroller while walking, or carrying grocery bags. Bipedalism is advantageous but unstable [13]. Continuous control needs to be exercised to support a large head, arms, and trunk mass on a small base of support formed by the feet [2, 14]. Due to the structure of the human body, human walking has also been described as ‘controlled falling’ in which individuals avoid a fall by the execution of each step. During walking, the position of the centre of mass (COM) in the anteroposterior and mediolateral direction has to be kept within the base of support (BOS) formed by the feet to remain stable and avoid a fall. The size of the BOS is an important factor for stability. Mechanically, a larger BOS leads to greater stability and a reduction in the size of the BOS reduces stability requiring greater control of the COM by the central nervous system (CNS). During walking, the size of the BOS is reduced to the area under one foot during the single support phase, and the vertical projection of the COM passes through the medial border of the stance foot. This reduction in the size of the BOS induces instability during the single support phase. This instability is corrected by the placement of the swing limb during the subsequent step that brings the COM back within the BOS [2, 15]. The subsequent step after the single support phase increases the size of the BOS since both the feet are in contact with the surface (double support phase). Failure to execute a step successfully can lead to a fall that can have disastrous and sometimes fatal consequences [16, 17].

Balance control is a complex motor skill that is necessary to execute tasks of daily living safely. The functions of the balance control system during walking as described by Patla et al. are [18]:

a) maintain the posture and orientation of the body

b) initiate beginning and complete ending the process of walking when required

c) activation and generation of appropriate muscle synergies depending on the direction of walking

d) counter the forces of gravity and environmental forces on the body during walking

e) modulate walking speed and foot placement in response to different terrains and obstacles

f) walk successfully towards endpoints that are not visible

g) optimize energy consumption during walking

g) safeguard the integrity of the bodily systems involved in walking.

Balance control can be affected by normal ageing and neurological disorders such as stroke, Parkinson's disease, multiple sclerosis, and traumatic events such as concussions and spinal cord injuries. This impairment in balance control makes individuals vulnerable to falls and an increased risk of frequent falls [19]. In addition to fall-related injuries, poor balance control can lead to fear of falling that further leads to a reduction in physical activity and self-efficacy in performing physical tasks such as stair negotiation and other nonhazardous activities of daily living [20]. To address the difficulties mentioned above relating to balance, understanding, and finding novel ways to assess and improve balance control is vital in improving quality of life (QOL) in individuals at risk of falling and preventing subsequent future falls. The purpose of this thesis was to: i) identify balance control strategies employed by healthy adults during backward walking, to examine whether backward walking velocity is associated with balance control measures; ii) assess the effects of vision and added haptic input during backward walking and, iii) to investigate the effects of an intervention with added haptic input on balance control during forward, backward, and tandem walking.

1.1 Gaps in literature

The current thesis aimed to address the following gaps that exist in literature:

1. The test retest reliability of the commonly used biomechanical measures of balance control during three different walking styles (forward, backward, and tandem (heel-toe) walking).

2. The differences in balance control strategies adopted by healthy adults between forward and backward walking and correlation between backward walking velocity and biomechanical measures of balance control during forward and tandem walking.

3. The role of visual input and the effects of added haptic input on spatiotemporal and balance control parameters during backward walking.

4. The effects of an intervention using haptic input on balance control during forward, backward, and tandem walking.

   The aims, purpose, and the hypotheses for each study aimed at addressing the above mentioned gaps are discussed in section 2.14.

2 Literature review

2.1 Epidemiology of falls

Globally, falls are the second leading cause of accidental death in individuals after road traffic injuries [1]. The World Health Organization defines a fall as ”an event which results in a person coming to rest inadvertently on the ground or floor or other lower-level” [1]. Although most fall-related injuries are non-fatal, they can lead to fractures, head trauma, prolonged hospitalizations, and increased permanent disabilities [1, 21, 22]. Older individuals and individuals living with neurological disorders are at an increased risk of experiencing a fall, but fall-related injuries can occur irrespective of age and health status [1, 21, 22]. A single fall incident can have long-lasting consequences of social isolation and a reduced QOL [23].

Within Canada, falls in adults over 65 years increased significantly from 2005-13, with women experiencing significantly higher falls than men [24]. The highest number of falls occurred when walking on a non-icy surface (45.2% of falls), followed by walking on snow or ice (15.5% of falls) [24]. The population of seniors in Canada is expected to reach more than ten million by 2036 [25]. Seeing that most falls occur during walking and with a rising population of seniors both globally and in Canada, it is imperative to devise strategies and plans that improve balance during walking and potentially mitigate falls.

Falls are multifactorial, with various intrinsic and extrinsic risk factors that lead to a fall [26]. Intrinsic risk factors are specific to individuals, including age, sex, gender, race, balance control, muscle strength, sensory input, cognitive abilities, and diseases [26]. Extrinsic risk factors include medications, footwear, and other environmental factors such as stair design, slipping and tripping hazards, presence or absence of handrails, the amount of lighting, and walking surfaces [26]. Whereas factors such as age and sex are non-modifiable, factors such as muscle strength, cognition, sensory integration, and balance control can be modified with some form of intervention. The following section in chapter two will focus on literature examining balance control and interventions to improve balance control.

2.2 Analysis of balance control during walking

2.2.1 Gait variability

Gait variability is defined as the step-to-step fluctuations of performance measures during walking [27]. Variability of parameters provides valuable information about the integrity of the motor control system and the sensorimotor integration during walking. A small amount of variability evident during walking in healthy adults indicates the motor system’s ability to produce a highly consistent walking pattern. Step variability has often been examined in literature to identify retrospective and prospective fallers. In a one-year prospective study, Hausdorff et al. [28] found that individuals who experienced falls had a higher stride time variability. Maki et al. [29] found that an increase in variability of stride length, walking speed, and double support were predictors of future falls. Along with an increased likelihood of falling, increased gait variability is also associated with a reduction in cortical gray matter volume in healthy older adults [30]. Whereas too much variability indicates an irregular walking pattern and the inability of the motor system to control COM movement, too little variability reflects a deficient motor control system that cannot respond appropriately to perturbations [31]. Collectively, it can be suggested that gait variability demonstrates a U-shaped graph where too little or too much variability during walking can be disadvantageous and lead to falls [31]. The amount of variability must be interpreted in the context of the task and the environment in which the task is performed. An increase in variability to recover from perturbations during walking or when walking in an unfamiliar or novel environment is necessary to maintain stability, but high variability when walking on a firm surface in a well-lit environment might indicate a pathology.

2.2.2 Margin of stability

The concept of margin of stability (MOS) is based on the principle that controlling the COM alone within the BOS is not sufficient to maintain balance during dynamic tasks such as walking. The projected COM position taking into account the COM velocity has to be within the boundaries of the BOS during walking to avoid a fall [6]. During standing, a person is considered stable if the vertical projection of the velocity accounted COM is within the boundaries of the BOS [2]. During walking the COM velocity is directed forward and outward causing the COM to fall outside the BOS. This construct of the position and velocity of the COM is termed as the extrapolated centre of mass (xCOM), and the spatial and temporal association between the xCOM and the BOS is defined as MOS [6]. The spatial component is the instantaneous distance between the xCOM and the edge of the BOS [6]. Based on the equation used to calculate the MOS, a positive value indicates a greater distance between the xCOM and the BOS and that the xCOM is within the BOS, and a negative value indicates that the xCOM is beyond the boundaries of the BOS. An individual is deemed more stable and less likely to fall if they have a positive MOS and a negative MOS implies an increased likelihood of a fall [32]. Margins of stability in the sagittal (anteroposterior direction) and frontal (mediolateral direction) planes are primarily controlled via foot placement during walking [33]. Changing the step length and step width are common strategies used to overcome challenges to balance control during walking [16]. The margin of stability can be used as an indicator of balance control strategies adopted by individuals in response to small internal perturbations during steady-state walking and more significant external perturbations such as slips and trips [16, 34]. Internal perturbations are caused when the COM-BOS relationship is challenged during voluntary movement [35].

MOS has been widely used in the literature to identify balance control differences between individuals with balance impairments and healthy controls [36, 37] and examine balance control strategies in response to perturbations [16]. The use of MOS to assess balance control in neurologically impaired and healthy individuals has shown mixed results. Whereas most neurologically impaired individuals walk with a smaller MOS than healthy controls, certain studies have shown that neurologically impaired individuals have a larger MOS than healthy controls [36, 38]. A higher MOS in individuals with a neurological condition is attributed to a compensatory strategy to increase stability during walking. The compensatory strategy to increase the MOS is usually accomplished by changing the step length and/or step width.

The stability margin is regulated at each step during walking to avoid falling and keep progressing in the preferred direction. If a moment of instability is encountered during the stance phase, the following swing limb is appropriately placed to regain stability during the next step [16, 34, 37]. Adjusting foot placement (changing step length or step width) in response to perturbations increases the instantaneous MOS by bringing the xCOM within the BOS, after which the individual can continue walking with their preferred MOS. Even though the mean MOS during walking may be positive, instantaneous values greater or smaller than the mean or sometimes negative MOS values are prevalent during instances of instability [16, 34]. This suggests that, in addition to mean MOS, examining MOS variability is also necessary to identify the consistency with which individuals can maintain a particular MOS during walking.

2.3 Forward walking

The definition of balance control states that walking without falling requires the centre of mass to be within the base of support [2]. Any changes in the BOS impacts balance control [39–41]. A frequently used strategy to avoid falls is changing the size of the BOS. Individuals change their step length and width in response to external and internal perturbations encountered during walking [16, 37]. The changing of step length and step width is used to control the size of the BOS in the anteroposterior (sagittal plane) and mediolateral (frontal plane) directions, respectively [16, 37]. The changing of step size in response to perturbations is a strategy utilized to control the COM within this new BOS to continue walking without a fall after encountering the perturbation. Dynamic control can be examined by inducing biomechanical constraints during walking tasks. One task that can be used to restrain the size of the BOS is tandem (heel- to-toe) walking. The ability to control the movement of the COM within a restricted BOS while walking in tandem can shed light on balance control capabilities [42–44]. Tandem walking is explained further in section 2.5.

During walking, information from the visual, somatosensory, and vestibular systems is utilized by the central nervous system to generate appropriate motor outputs and walk without falling in different environments [18]. The process that allows individuals to perform specific motor tasks based on combining sensory input from multiple sources is known as sensorimotor integration [45]. Even though each sensory system has an individual role in maintaining balance during walking, information from all systems is integrated into the central nervous system to generate anticipatory, reactive, and predictive balance strategies [46]. Information from each sensory system is reweighted, (i.e., the reliance on each sensory system is altered) based on the demands imposed by the task at hand and the external environment [44]. For example, when walking on a surface that is non-rigid where the somatosensory input is challenged, the reliance on visual systems is increased to maintain balance [47].

Similarly, when walking in a low-lit environment or without vision, the reliance on somatosensory systems increases to maintain a stable walking pattern [48]. Sensory reweighting is also seen in older adults and individuals with neurological populations where reliance on one sensory system is increased to compensate for the lack of or impairments in another sensory system(s) [49]. Posturography studies have shown that individuals with stroke and incomplete spinal cord injury rely more on vision to maintain balance during standing [50, 51]. During walking, the addition of haptic input improves balance control when visual input is restricted and in neurological conditions such as stroke and incomplete spinal cord injury where lower limb somatosensation is reduced or impaired [43, 48, 52].

2.4 Backward walking

Backward walking is not undertaken frequently in daily life but backward stepping and control of COM movement during backward stepping is an important component when performing ADLs. Examples include stepping backward when opening a house or car door, stepping backward after picking an object placed in front or above, taking backward steps to sit on a chair or a couch and stepping backward as a protective mechanism to avoid a fall.

The assessment of walking described in the previous sections are in context of forward walking. Walking in the forward direction remains the primary choice of mobility assessment by clinicians and researchers. Walking forward normally and in tandem (heel to toe) are frequently utilized tasks in clinical tests and research studies to assess balance control [42–44, 53]. While certain clinical tests and studies do include backward walking, it is not as frequently utilized as forward and tandem walking [54–57]. Results from previous literature present a compelling argument in favour of utilizing backward walking as both a task to assess balance control and as an exercise in balance training programs [54, 55, 58–63].

Hackney et al. [59] compared spatiotemporal and gait variability parameters between individuals with Parkinson's disease (PD) and controls. The authors found that individuals with PD and controls had similar velocity when walking forward but individuals with PD walked slower than control when walking backwards. Variability of the amount of time in the stance phase was also higher during backward walking in individuals with PD compared to controls. Backward walking velocity was negatively correlated with the Unified Parkinson's Disease Rating Scale (UPDRS) – a clinical scale used to rate the severity of symptoms in PD. Individuals that scored higher on UPDRS had a slower backward walking velocity. In addition to UPDRS, backward walking velocity was also positively correlated with the Berg Balance Scale – a clinical test used to evaluate balance control. Individuals with PD that obtained a higher score on the BBS (better balance control) also had a higher backward walking velocity. In another study involving older adults, backward walking velocity was significantly correlated with fall predictor clinical measures including the Timed Up and Go (TUG) test, the Activities-specific Balance Confidence (ABC) scale, and the Four Square Step Test (FSST) [58]. Identifying individuals that are likely to fall can help tailor rehabilitation programs to avoid future falls. Findings from a study by Edwards et al. [55] showed that backward walking velocity significantly distinguished between retrospective and prospective fallers and non-fallers in a group of people with Multiple Sclerosis (MS). Carter et al. [64] have also proposed a novel 3-metre backward walking test to identify retrospective fallers. The authors found that individuals who took longer than 4.5 seconds were more likely to have reported falls and individuals that took 3 seconds or less to complete the test were unlikely to have reported falls. The 3-metre backwards walk test has also demonstrated excellent reliability on older adults [65] and individuals with stroke [66]. The 3-m backward walk test can also be used to identify backward walking velocity and determine optimal ranges and cut-off scores for individuals that are at a risk of falls.

Along with identifying individuals with balance deficits, the use of backward walking as an exercise has also shown benefits on balance control and gait measures. Children with cerebral palsy that received backward walking training along with a conventional physical therapy program showed a greater improvement in stability compared to a group that received only the conventional physical therapy program [62]. Individuals with acute stroke that performed body weight supported backward walking training along with conventional therapy showed significantly greater improvements in walking velocity as well as performance on the Rivermead Mobility Index score compared to a group that received body weight supported forward walking training in addition to conventional therapy [61]. In another study involving people with acute stroke, the group of participants that underwent an overground backward walking training showed higher improvements in forward and backward walking velocity and the ABC scale compared to a group that received a standing balance training program [60]. Maritz et al. [63] examined the effects of a backward walking training program on older adults and found that a twice/week backward walking program for five weeks significantly improved performance on the TUG test, the dynamic gait subsection of the Mini-BESTest, and heel rise test compared to a control group that did not undergo the intervention. Other than older adults and neurological populations mentioned above, backward walking training also improved balance control during standing in young healthy boys after twelve weeks of backward walking training and the improvements in balance control were retained twelve weeks after the end of the training program [67]. Taken together, the results of previous research indicate the usefulness of using backward walking as a measure of mobility as well as an exercise to improve balance control.

A study on joint kinematics by Winter et al. [68] identified that knee and hip angle patterns were similar and ankle angle patterns were different in backward walking compared to forward walking. Another study on joint kinematics in children indicated that upper extremity joint movements during forward walking were similar to a time reversed backward walking [69].

In terms of muscle activation, Thorstensson [70] noted significant differences in activity of the tibialis anterior (TA), rectus femoris (RF), hamstrings, lateral gastrocnemius (LG), vastus lateralis (VL), and gluteus maximus (GM) during backward walking compared to forward walking. Analysis of medial gastrocnemius (MG), vastus medialis (VM), soleus, tibialis anterior (TA), rectus femoris (RF), hamstrings by Winter et al. [68] also found differences in the amplitude of muscle activation between forward and backward walking. The changes in amplitude were attributed to the change in muscle contraction from concentric to eccentric when walking backward. Similar results were obtained by Grasso et al. [71] when comparing muscle activity between forward and backward walking at different speeds. While the mean EMG activity increased with walking speed for both forward and backward walking, the mean EMG at a given speed was higher for backward walking compared to forward walking. The EMG results suggest that backward walking is generated by the same muscles involved in forward walking but the timing, the type of contraction, and muscle activation amplitude are different when walking backward compared to forward walking.

Supraspinal control of backward walking has been examined using functional near-infrared spectroscopy (fNIRS) and for imagined backward walking using functional magnetic resonance imaging (fMRI). During imagined backward walking, higher activations in the primary motor cortex, supplementary motor area, parietal cortex, thalamus, putamen and lower activations in cerebellum and brainstem were observed compared to forward walking [72]. An fNIRS study by Kurz et al. [73] showed that oxygen uptake during backward walking was greater in the supplementary motor area, precentral gyrus, and superior parietal lobule in relation to forward walking. The authors hypothesized that the increase in cortical activity was due to an increased challenge to the balance control system during backward walking. Other reasons proposed for an increase in cortical activity are increased demands for generating motor synergies for backward stepping, absence of visual information about oncoming obstacles that is otherwise available during forward walking, and the novelty of backward walking, a task that is not as well learnt as forward walking. In terms of movement control, forward walking is controlled via a feedforward mechanism where information about the external environment and potential obstacles is obtained by the visual system and walking is adapted to navigate or avoid the obstacle altogether [46, 74, 75]. In comparison, backward walking is controlled by a feedback mechanism where information from the vestibular and somatosensory systems during each step is used to plan and execute the subsequent steps.

Taken together, the results from biomechanical measures, and measures of muscle activity, and brain activation show that the alternate stepping movement during backward walking is achieved by the same structures as forward walking, but the balance control aspect of backward walking requires a separate or a more intense neural control.

Certain studies have also used backward walking to identify differences in mobility between clinical populations and their controls as well as to identify balance impairments between fallers and non-fallers. Backward walking has been shown to be better than forward walking to distinguish mobility between young and older adults, individuals living with Parkinson's disease, stroke, multiple sclerosis, and cerebral palsy [55–57, 59, 76]. As previously mentioned, balance is controlled through foot placement and specific information about balance control strategies adopted by individuals during backward walking is lacking. Identifying strategies used by healthy individuals to maintain balance through step placement can provide information on how backward walking is regulated and whether older adults and individuals with neurological injuries use similar strategies. Furthermore, balance control strategies employed during backward walking can be used to identify the impairments in balance control that are probably not identified using forward walking alone. For example, the ability to execute a backward step in response to a backward loss of balance or the ability to take multiple backward steps to accomplish a task are not examined during forward walking. In addition to balance control measures during backward walking, information on the test-retest reliability of spatiotemporal and balance control measures during backward walking has not been examined.

The reliability of performance during backward walking is necessary if backward walking is to be recommended as a clinical test to identify balance impairments. The postural challenge created by backward walking can be exploited to identify the integrity of the balance control system. Backward walking is a task that is easy to administer, can be completed within a limited area, and provides vital information about dynamic balance control. Identifying the strategies used to maintain balance during backward walking and the effects of practicing backward walking can provide further results in support of using backward walking as a measure of balance control as well as an exercise in balance training programs.

2.5 Tandem walking (forward direction)

Tandem (heel-to-toe) walking is used as a screening tool to assess the integrity of balance control as well as an exercise in balance training programs [77, 78]. In the context of this thesis, tandem walking refers to heel-to-toe walking performed in the forward direction.

Unlike backward walking, tandem walking is not directionally different from forward walking. The challenge associated with tandem walking is controlling the COM within a small base of support. The size of the BOS is a vital factor in determining stability during walking [39, 79]. Controlling the COM within a limited area during tandem walking requires generating an appropriate motor output to keep progressing in the desired direction without falling. Lark and Pasupuleti [79] demonstrated that the amount of narrowing of the BOS was related to balance control performance during walking. Decreasing step width led to an increase in time required to perform the parallel walk test (walking between parallel lines of different widths) over a distance of six meters. Tandem walking performance is analyzed by counting the number of missteps, the amount of time required to traverse a certain distance, the amount of trunk movement, and margin of stability [42, 43, 53, 80–83]. Moderate to excellent test-retest reliability has been observed for time to complete tandem walking [81, 82] but test-rest reliability of MOS during tandem walking has not been examined. Due to the common use of tandem walking to identify and improve balance control capacity in individuals, the examination of balance control strategies when walking in tandem are warranted. Studies that have examined the effects of practice on tandem walking have shown mixed results. Examining the effects of repeated practice is necessary to obtain information on the feasibility of the task being practiced and to know whether repeated practice of the task can induce adaptations in walking.

One study by Dozza et al. [84] showed that repetition of tandem walking in participants with unilateral vestibular loss improved performance as evidenced by a reduced variability in COM movement, trunk tilt variability, and a reduction in step width. In another study by Costa et al. [83] participants reduced step speed during tandem walking with repeated practice. A commonality in the studies by Dozza et al. [84] and Costa et al. [83] is that participants practiced tandem walking with somatosensory feedback. Walking in tandem with somatosensory feedback improved balance control but the short-term effects of the somatosensory feedback during practice sessions were not retained. Further investigation is required to assess whether increasing the amount of practice trials, increasing task difficulty, and providing somatosensory feedback in the form of added haptic input improves balance control during tandem walking.

2.6 Balance control during walking

Balance can be used as an umbrella term to describe the ability of an individual to resist perturbations and avoid a fall [3]. In mechanical terms, a body is considered in a state of balance or equilibrium if the body's vertical COM projection lies within the BOS [2, 3]. The body becomes unbalanced or loses equilibrium if the COM moves outside the BOS [3].

Postural control can be defined as a combination of postural equilibrium and postural orientation [39]. Postural equilibrium is the ability to control the COM movement during a task. Postural orientation is the ability to control and align body segments with respect to one another, gravity, and the external environment [2, 39]. An intact balance control system allows individuals to maintain task-appropriate postural orientation and counter perturbations associated with the task.

Stability is defined as the intrinsic property of a body to resist moving from a balanced to an unbalanced state or move from an unbalanced to a balanced state [4]. Stability can also be defined as the ability to resist perturbations [4]. An individual is considered stable if they are able to resist perturbations of varying magnitude and avoid a fall or recover their state of balance when they are unbalanced through sensory and motor systems [3]. Individuals with an intact balance control system allow them to modify their posture to increase or decrease their resistance to perturbations [4].

Balance control during walking is achieved by employing a mix of anticipatory, reactive, and predictive strategies [46]. Each of the strategies is used in a mixed manner when undertaking locomotion. To better understand the neuromechanics of these strategies, they are often examined individually by movement scientists. An anticipatory balance control strategy is primarily based on sensory information obtained from visual input [46]. The exteroceptive information from the environment is sampled in a feedforward manner by the visual system to guide walking. For example, information about an obstacle obtained through vision leads to a change in stepping where an individual either steps over or around the obstacle [75]. Anticipatory strategies are therefore employed *in advance or before* a perturbation is encountered.

A reactive balance control strategy is primarily based on sensory feedback received in response to a perturbation. Reactive balance strategies involve changes in step size, swinging of the arm/s, and grabbing an object in the external environment to ensure stability during walking [16, 85, 86]. Reactive strategies consider the input obtained from the visual, vestibular, and somatosensory systems when the balance is challenged due to an unexpected perturbation. An unexpected trip over an obstacle provides input from the somatosensory system of the foot, which leads to an elevation strategy (hyperflexion of the hip, knee, and ankle) of the perturbed limb before the subsequent step [87]. In contrast to anticipatory strategies, reactive strategies are employed *after* experiencing a perturbation.

Predictive balance control is based on integrating previous experiences and sensory information obtained during ongoing movement [46]. In terms of walking, the internal perturbation generated by the movement of the limbs during various phases of the gait cycle and the sensory input obtained during each stance and swing phase dictates the size and timing of subsequent steps. Similarly, when walking on a slippery sidewalk during winter, individuals change their stepping strategy based on the previous knowledge or previous experience of having suffered a slip on an icy surface [88]. A predictive strategy is applied *in advance and during* each step of unperturbed walking.

2.7 Neural structures involved in walking and balance control

Information regarding the role of neural structures involved in balance control has primarily been gained from animal studies. In humans, the neurophysiology of walking is obtained indirectly by comparing individuals with neurological injuries to their healthy counterparts and by analyzing brain activation during imagined walking [89–91].

Walking is controlled through an intricate network of neurons and neural structures within the spinal cord and cerebral cortices. Neuroimaging and behavioural studies in individuals with neurological injuries have highlighted the specific role of distinct neural structures involved in walking and balance control [89, 91]. The structural connections between (and within) the spinal cord, cerebellum, brainstem, thalamus, basal ganglia, and cerebral cortices interact in series and parallel fashion to control various aspects of walking such as initiation, steady-state walking, balance control, and termination [89, 90]. The highest level in the hierarchy is the cerebral cortex. The motor area of the cortex is divided into the premotor cortex, supplementary motor area, and primary motor cortex. Based on the integrated sensory information obtained from the somatosensory cortex, the premotor and supplementary motor cortices plan and coordinate movements, and the execution is accomplished by the primary motor cortex via the pyramidal tracts [92, 93].

Brainstem structures regulate the initiation of walking and balance control via the extrapyramidal system. Extrapyramidal tracts originate from various structures of the brainstem, including vestibular nuclei (vestibulospinal tract), red nucleus (rubrospinal tract), reticular system (reticulospinal tract) and synapse directly on the spinal cord [92]. The mesencephalic locomotor region within the brainstem engages in the initiation of walking and control of muscle tone. The pontomedullary reticular formation participates in predictive and reactive balance reactions, controlling walking speed and muscle tone regulation [89, 90, 92, 93].

The basal ganglia are a specialized group of subcortical nuclei located deep within the cerebral hemispheres that control walking via direct and indirect pathways [94]. The role of the basal ganglia is examined by analyzing behavioural changes in animal models and individuals with basal ganglia dysfunction. Pathology of the basal ganglia leads to Parkinson's disease (PD), a neurological movement disorder [95]. Studies with individuals who have PD have revealed balance impairments during standing and walking [96–98]. Parkinsonian gait is characterized by a reduced BOS, festination, and freezing [95]. The basal ganglia control walking and balance through intrinsic connections within the basal ganglia nuclei via the direct and indirect pathways [92,93]. Based on impairments observed in people with Parkinson's Disease, basal ganglia are thought to contribute to balance control by executing motor actions, integrating somatosensory information, regulating muscle tone, and scaling automatic postural reactions [99].

The cerebellum does not directly send projections to the spinal cord, but it receives input from sensory tracts via the spinal cord and sends out efferent connections to the thalamus, brainstem, basal ganglia, and cortex [94]. Balance control is regulated using a feedforward mechanism via the cerebellum, where sensory information obtained during each step is used to control the subsequent foot placement [100, 101]. The cerebellum is also responsible for coordination during walking. Lesions to the cerebellum affect foot placement during walking when navigating obstacles and lead to a highly uncoordinated gait (ataxic gait), with a widened BOS, and with an increased step variability [100, 101].

The spinal cord can generate rhythmic activation of flexor and extensor muscles in the lower limb, even in the absence of supraspinal input [93]. This rhythmic activation of flexor and extensor musculature is conducted by interneurons in the spinal cord consisting of mutually inhibiting interneurons or half-centres [93, 102]. This network of interneurons that generate alternate movement of the limbs is known as central pattern generators (CPGs) [103]. During walking, CPG activity is modulated from proprioceptive and cutaneous input from the lower limbs [87]. Sensory inputs from the proprioceptive and cutaneous systems are integrated and utilized at the spinal level in a phase-dependent manner during walking [87]. Stimulation of cutaneous afferents of the foot dorsum during the early swing phase leads to an elevation strategy achieved by flexion at the knee and plantar flexion at the ankle. In contrast, stimulation of the foot during late swing causes a lowering strategy accomplished by landing with a reduced step length, flatter foot, and flexed knee at heel strike [87]. Proprioceptive information about the lower limb joint position controls the transition from the stance to swing phase during walking [102, 104]. The amount of hip extension in the stance leg regulates the initiation of the swing phase during a gait cycle [102, 104]. The hip joint of the stance leg must be extended and unloaded to initiate the swing phase [102, 104]. Though alternate activation of flexors and extensors is achieved by the CPGs, balance control during walking is complex where CPG activity alone is not sufficient to maintain stability. Supraspinal structures also modulate the activity of CPGs during walking via the pyramidal tracts from the motor cortex and extrapyramidal tracts from the brainstem structures [92]. The tracts originating from the cortical and subcortical structures project on the CPGs and are responsible for regulating muscle tone, balance control, and in navigating complex environments [94]. For example, cats that are transected at the level of the cortex regain their ability to walk on a flat surface but are unable to step over obstacles or walk over a ladder [105].

2.8 Changes in visual control during backward walking

Vision is one of the most important sensory inputs to control balance during walking [106]. During forward walking, sensory input from vision is used in an anticipatory manner to traverse over different terrains, to use avoidance strategies when approaching a potential obstacle to maintain balance, and to plan the walking route or path [106]. When visual input during walking is manipulated or removed, it significantly affects balance control as seen by an increase in variability of step measures [107, 108] and increase in COM movement and COM movement variability [48]. Spatiotemporal measures are also altered to maintain dynamic stability when walking with eyes closed [109, 110]. There is an increase in step width and relative amount of time in the double support phase and a decrease in step length, walking speed, and cadence when walking with eyes closed compared to walking with eyes open [109, 110]. When walking over short distances where the end goal or end point is visible, visual input can singularly guide walking [106]. When the end goal is not visible from the beginning, visual input uses information about landmarks and spatial maps stored in memory to guide walking behaviour [106]. When walking within a lab environment, where the end goal is visible before walking begins, visual information can be used to walk from one end of the lab to another and to stop walking at the end of the walkway. When walking over short distances within a lab environment, removal of vision leads to underestimation of the walked distance [111]. The likely reason for the underestimation of the distance when walking within a lab could be the fear of colliding into the lab walls. Collision into a static or oncoming object is controlled by estimating time to contact (TTC) [112]. TTC is an aspect used to control walking behaviour through visual input [113]. TTC is the amount of time available or remaining before an object collides with an individual [112]. The TTC is determined through the visual system by the changing size of the colliding object on the retina. As an individual approaches an object or vice-versa, the size of the object on the retina increases which allows an individual to take action to avoid collision [112]. Different behaviours such as hitting an incoming ball in tennis or cricket, braking a car at a stop sign, decelerating when approaching a door are all performed using TTC information [112]. When walking in the forward direction, TTC information can be used by individuals to control the position and timing of foot placement to regulate stability [113]. During backward walking, where the endpoint is not visible from the beginning of the movement, individuals have to rely more on the vestibular and somatosensory system to regulate foot placement, thereby making backward walking more challenging in terms of balance control compared to forward walking.

2.9 Haptic input and balance control

Haptic input is the sensory input obtained through the cutaneous and proprioceptive systems when lightly touching an object (applying less than 1 Newton force) placed in the environment [114]. It is theorized that haptic input improves balance control by providing additional information about body position in relation to the support surface and the source of haptic input [115]. When touching an object in the environment, additional cutaneous and proprioceptive information is integrated into the central nervous system, and balance control is improved by either increasing or decreasing muscle activity to better control the COM movement [115].

Haptic input can be provided via various modalities such as railings, canes, walkers, and haptic anchors [43, 52, 91, 98, 116]. Haptic anchors are light weights (125 grams) attached to strings and dragged by individuals during walking [117]. Haptic input is provided via skin mechanoreceptors in the hand when the weights produce tension in the string, which provides sensory input about where one is in space [117]. Whereas modalities such as railings and canes have been used often to examine the effects of haptic input on balance control, haptic anchors are a relatively new modality that need further examination.

2.9.1 Effects of haptic input during standing

The effects of haptic input on balance control during standing have been well established in the literature [40, 51, 114, 118, 119]. Lightly touching an object such as a railing during regular and tandem standing improves balance control as seen by reductions in trunk and COP sway parameters [51, 115, 119]. Integration of haptic input also improves balance control when sensory input from the visual and vestibular systems and the somatosensory system from the lower limb is reduced or impaired [49, 50]. Additionally, the dependence on haptic input for balance control is proportional to the difficulty of the task. A study by Magre et al. [119] on healthy adults found that as the level of task difficulty increased, so did the reliance on haptic input to control postural sway. Increasing the postural challenge associated with a task leads the CNS to place greater requirements on haptic input as a source of sensory information to improve balance control during the task [119].

2.9.2 Arm orientation and attention associated with added haptic input

Similar to standing, lightly touching a railing or touching a stable surface when walking on a treadmill, and dragging haptic anchors improves balance control during walking [42–44, 53, 83, 120–122]. Even though most studies provide evidence favouring haptic input in improving balance control during walking, the effects of haptic input on spatiotemporal measures of walking have shown mixed results. Gait velocity, the amount of time in the double support phase, step length, and step width can increase, decrease, or show no significant change when walking with haptic input [123].

One reason for the mixed effects of haptic input on spatiotemporal measures is the type of haptic modality. Different modalities require different arm configurations that can have varying effects on spatiotemporal measures [122]. The varied results of haptic input on spatiotemporal measures may also be attributed to attentional demands when using haptic modalities [120, 123]. Sustaining a force below a certain threshold on a railing during walking may require more attentional demands [122]. Also, walking with a fixed arm configuration at a certain distance from the source of haptic input (railing) could further exacerbate attentional demands and affect the spatiotemporal measures [120]. Canes and anchors are actively moved along by individuals during walking, whereas railings and plates are stationary where individuals place their fingers during walking, which can change walking behaviour [120]. Moving the haptic modalities assumes a different arm movement and arm configuration compared to lightly touching a stationary object. Also, the arm configuration by itself can have stabilizing effects by increasing the moment of inertia, leading to an improvement in balance control [121, 122].

Haptic input has improved balance control measures examined using trunk movement, COM movement, and variability measures. In particular, variability measures during walking showed a decrease indicating that movement consistency improves with haptic input [42–44, 48, 83, 121]. As previously mentioned, the reliance on haptic input increases with increasing task difficulty. Tandem walking is a particularly challenging task where step width is constrained, and the COM has to be kept and controlled within a narrow BOS. Haptic input improves balance control during tandem walking, demonstrated by a reduction in trunk acceleration, trunk acceleration variability, and MOS variability [42, 43, 83]. Backward walking is another challenging task that requires walking without visual information of an end goal. The effects of haptic input during backward walking are not known. Examining the effects of added haptic input during backward walking will add to the literature on the positive effects of added haptic input when performing tasks that challenge the balance control system.

2.10 Balance assessment

2.10.1 Assessment of Balance control

The efficiency and integrity of the balance control system are examined by asking individuals to perform challenging tasks, modifying the external environment where the task is being performed, or modifying input/s from one or more sensory systems. The performance on the task is measured qualitatively and quantitatively to obtain an estimate of the balance control abilities of individuals. Commonly used measures to evaluate balance are clinical tests and biomechanical outcomes obtained from technological systems such as 3-D motion capture systems, inertial measurement units (IMUs), and pressure-sensitive mats. Clinical and biomechanical techniques of balance evaluation are discussed further in the following sections.

2.10.2 Balance assessment using biomechanical measures

The advancement of technology has made it possible to assess human behaviour during standing and walking with high accuracy and minimal error. Biomechanical analyses of walking provide information on spatiotemporal measures and more complex concepts such as COM movement that are not captured by clinical tests of balance. A variety of biomechanical measures are used to identify balance control strategies during walking [124]. The changes in biomechanical measures in response to physical or sensory perturbation shed light on the sensorimotor integration of balance control during walking. Challenges are provided by changing sensory input or the delivery of external perturbations. Sensory challenges include walking with reduced or absent vision [48, 121], walking on a compliant surface [74], walking with anesthetized feet [125, 126], walking after muscle or tendon vibration [127], and walking with external stimulation of the vestibular system [128]. Perturbation studies are typically undertaken to assess reactive balance control by delivering unexpected perturbations through slip devices [129], surface translations on a treadmill [16], or by pushing or pulling through a mechanical device [130]. Anticipatory and predictive balance control are typically examined by asking participants to walk over known perturbations such as an expected slippery surface [88].

Identifying the changes or lack thereof in spatiotemporal measures, COM movement, spatiotemporal variability, COM-BOS interaction, and trunk movements between different sensory conditions provide insight into the contributions of different sensory systems used by individuals to maintain stability during walking. Comparison of biomechanical balance measures between neurologically impaired and healthy individuals provides indirect evidence on the neural structures involved in balance control (e.g., comparing people with Parkinson's Disease, stroke, SCI to age and sex matched controls) [37, 43, 131]. In this thesis, the biomechanical measures used to assess predictive balance control include margins of stability and gait variability, discussed in the previous sections.

2.10.3 Clinical assessment

Various clinical tests have been developed and are currently used to assess balance control by having individuals perform challenging tasks similar to those performed in real life [132]. The tasks are designed to evaluate the anticipatory, reactive, and predictive aspects of balance control. In addition, clinical tests can evaluate the integrity of sensorimotor integration by performing tasks in altered sensory conditions (e.g., standing with eyes closed on a foam surface). Commonly used tests of balance control during standing and walking are Berg Balance Scale (BBS) [4, 133] mini-Balance Evaluation Systems Test (Mini-BESTest) [134], Dynamic Gait Index (DGI) [133], the Timed Up and Go (TUG) [132, 133], and the 10-metre walk test [132]. The advantages of using clinical tests are that they are easy to learn and administer, require relatively minimal time and equipment, and can be performed in various clinical settings. Scores are assigned for each task performance, and the final score is used to estimate an individual's balance control capabilities. One concern using clinical tests to identify the integrity of balance control is that these tests tend to have a ceiling effect making them challenging to use in individuals with balance impairments that are not severe and in identifying minor changes in balance control in response to therapy [135]. The clinical tests preferred by interprofessionals to assess balance control vary, and the test(s) used may not comprehensively evaluate all aspects of balance [132, 136]. The choice of the test may depend on the experience of the interprofessional as well as the quality of information gained from the test [136]. Other factors that influence interprofessionals' use or non-use of a test are insufficient knowledge of the tests and a shortage of time and equipment to conduct these tests [137]. The concerns pertaining to clinical tests require the addition of novel tasks to assess balance control. Tasks that require minimal time, space, and training and could provide valuable information to therapists about the balance control system would be a valuable addition to the existing measures.

2.11 Interventions to improve balance during walking

Strategies for improving balance control to reduce falls and fall-related injuries is essential for individuals' QOL and reducing the healthcare system's economic burden [138]. To date, there is no consensus on a specific training program that is effective in improving balance control and reducing falls [139]. A variety of interventions have been implemented to address balance deficits and reduce falls. Other than balance training, programs such as dance and Tai- chi have also been used to rehabilitate balance deficits [140–142]. Based on a review by Sherrington et al. [143], the most effective type of training that reduces falls includes a component of balance and functional training and resistance training. In addition to the type of balance training, the dose of the training is an equally crucial factor when designing and implementing a training program. Similar to the type of balance training, there is no consensus on the length and duration of a balance training program. A review by Howe et al. [139] found that balance training was effective in three-month-long programs with three sessions per week. The review did not disclose whether the time duration for the balance training program was for group or individual training. Though ample research has been conducted on developing strategies to improve balance, more work is required to identify a specific type and duration of training programs that can improve balance control and reduce fall risks [139, 143].

2.11.1 Balance training

Balance training programs often involve individuals performing challenging tasks over a period of time, either under the supervision of a therapist or home-based exercises [144]. Performing challenging tasks forces individuals to activate postural muscles and balance control strategies to avoid falling. Researchers have hypothesized that the motor system is altered to meet the demands of the new challenging task or the challenging environment with repeated practice [145]. Repetition of the tasks also compels the CNS to generate a motor output that minimizes or negates the postural disturbances caused by the challenging tasks, leading to improved balance control [146, 147]. At a cellular level, the improvement in balance control is attributed to neural plasticity that occurs in response to training [148, 149].

A variety of interventions have been used in an attempt to improve performance in balance compromised populations. Though most interventions have shown improvements in balance control, the frequency and length of the intervention vary. A review by Howe et al. [139] found that interventions ranging from four weeks to twelve months were sufficient in improving balance control in older adults. Another review by DiStefano et al. [150] found that training for four weeks was sufficient to induce changes in balance control in healthy adults. Though recommendations are made based on findings from previous literature, no specific recommendations exist for the duration and frequency of balance training programs for different populations.

2.11.2 Adaptation and post adaptation

An approach to improve balance control during walking is to cause adaptations in walking behaviour by introducing constraints or changes in the external environment. The purpose of causing adaptations is to improve balance control and sustain those improvements after the adaptation inducing stimuli are removed.

The 'broken escalator effect', described by Reynolds and Bronstein [151] provides an example of adaptation and post-adaptation effect by changing the sensory input obtained from the somatosensory system. In the study, participants were asked to step on a sled from a stationary platform. During the first ten trials, participants stepped on the sled as it remained stationary. For the following twenty trials, participants stepped on the sled while it was moving. Then the researchers stopped the sled, informed participants that the sled would remain stationary, and asked the participants to step again on the stationary sled. The researchers found that participants adapted their walking patterns to walk over a sliding surface. The CNS changed the motor output to account for the postural instability caused due to the sliding surface such that participants stepped on a moving surface without falling. Following the trials of stepping on the moving sled, the participants continued walking in an adapted manner when the surface was stationary and prior knowledge that the surface would remain stationary. The participants did not revert back to the original walking pattern when the surface stopped sliding, indicating a post adaptation effect. This principle of walking adaptation has an important role in chapter six (study four) of this thesis. In chapter six, adaptation and post-adaptation effects on walking have been investigated when haptic input is provided for six weeks using haptic anchors to induce a change in walking behaviour.

2.11.3 Adaptation-based training

Balance during walking can be improved by causing adaptations in walking patterns through task and environmental constraints. Adaptability can be exploited by therapists and researchers where a certain balance training paradigm provided within a lab or clinic can change predictive and proactive balance control that can be used in a real world setting to walk successfully without falling. Balance performance during walking can be changed by modifying the environment, by inducing constraints, and augmenting sensory input during task performance. As previously explained by the 'Broken escalator effect', individuals do not immediately revert back to the pre-adaptation walking pattern after the adaptation effects are removed [152]. Walking can be adapted to a new pattern by imposing task demands that challenge balance control. Training in an unfamiliar environment or with constraints such as forcing participants to walk at a certain walking speed causes the CNS to generate a motor program to create a walking pattern under the imposed constraints, thereby leading to increased balance control. Researchers have used multiple strategies to induce adaptations in walking behaviour in healthy adults, older adults, and clinical populations. Overground walking, virtual reality, treadmill, and split-belt treadmill are frequently used for gait adaptation [96, 153, 154]. A split-belt treadmill consists of two belts that can be programmed to move at different speeds. Gait adaptations can be achieved by asking individuals to walk on the treadmill at different speeds for each leg over the course of five to fifteen minutes. Adaptations are preserved as evidenced by participants walking with a similar pattern when the belt speeds are equal or during overground walking [155]. This principle of gait adaptability on a split-belt treadmill has been exploited in individuals with pathological gait such as people with stroke [154].

2.11.4 Perturbation-based training

Balance control is task-specific, and therefore, one aspect of balance training is targeted towards the specificity of balance control [156]. Repetition of a particular task improves performance of that task. This principle of specificity is one of the factors included in rehabilitation programs for individuals with a neurological injury [157]. The goal of this type of training is to rehabilitate balance capacity during a specified task. One such example of task- specific training frequently reported in the literature is perturbation training [158]. Perturbation training is primarily targeted towards improving reactive balance control.

During walking, internal perturbations caused by the movement of the body and external perturbations from the environment have to be overcome to walk without falling. External perturbations include slipping on an icy sidewalk or tripping over a step. Recovering from an unexpected external perturbation requires the activation of the reactive balance control mechanism. Training individuals by providing unexpected perturbations can be useful for improving reactive balance mechanisms, whereas expected perturbations can improve proactive balance control strategies [88, 158]. The improvement in reactive balance control can then be transferred to a real-world situation where an individual can overcome an unexpected perturbation during walking.

2.11.5 Feedback-based training

Providing sensory feedback during training can also lead to changes in walking and improvements in balance control. Biofeedback is a technique of providing sensory information to individuals in addition to what is already available to them [159]. The supplementary sensory information provides feedback about any ongoing movement and can be used by individuals to improve their performance by modifying movement patterns [84]. Biofeedback can be provided using visual, auditory, and vibrotactile systems during walking [159]. Virtual reality is a form of immersive biofeedback that is used to improve balance control during walking [160]. Biofeedback has been used previously to improve dynamic stability during walking in young and older adults [161], people with Parkinson's Disease [162], and stroke [163]. Though the use of biofeedback in gait training has demonstrated immediate beneficial effects, the long-term effects and retention effects after biofeedback training are still relatively unknown [164]. One way of providing biofeedback is through haptic input. Haptic input includes sensory information via skin mechanoreceptors and joint proprioception when touching an object in the external environment [115]. Commonly used tools to provide haptic input are railings, canes, walkers, rollators, and haptic anchors. Similar to other forms of feedback, the use of haptic input via the haptic anchors has demonstrated beneficial short-term effects on balance control during walking [42, 120–122], but long-term effects of haptic input are yet to be determined. One study that used haptic input as biofeedback in individuals with vestibular pathology demonstrated improvements in walking and balance control in the group that trained using additional haptic input [165]. Additionally, only the group that trained using haptic input retained the effects after six weeks [165]. Study 4 in this thesis examines the role of training using added haptic input via haptic anchors on walking and balance control in healthy adults.

The purpose of using and examining the effects of haptic input using haptic anchors was primarily based on the ease of using haptic anchors and haptic anchors possibly being a more equitable option in the context of rehabilitation. Even though haptic input can be provided using canes and walkers, individuals need to be trained by specialists on the proper technique of using these modalities [166]. Improper and chronic use of the commonly prescribed walking aids can lead to discomfort, pain, injuries, and tendon and joint inflammation which can then lead to individuals completely abandoning the use of the prescribed aid(s) [166]. Compared to canes and walkers, haptic anchors are lighter in weight, easy to use, and require minimal training by specialists on the technique of using them. Even though the recommended use of haptic anchors requires that individuals actively grip the strings and drag the weights during walking, similar improvements in balance control have been observed when the anchor strings are tied at different locations of the upper extremity and the individuals do not actively hold or grip the anchors [167]. This beneficial aspect of haptic anchors can be exploited to improve balance control in individuals that have impaired hand function, low grip strength, or are unable to actively grip objects such as people with stroke. Furthermore, haptic anchors do not require a sophisticated construction and design. Haptic anchors can be easily made using strings and weights at a significantly lower cost compared to canes and walkers. Additional rehabilitation services require out of pocket expenses or additional insurance coverage which might not be an affordable option for certain individuals. In such cases, haptic anchors can be a viable alternative for improving balance control.

2.12 Reliability

Using reliable measures in research and clinical practice is vital to obtain accurate assessment of performance parameters. In terms of walking and balance control, walking performance is measured using spatiotemporal parameters of walking velocity, amount of time in the double support phase, step length and step width. As mentioned in the previous sections, balance control during walking can be evaluated using variability of step and MOS measures as well as the magnitude of the MOS. Evidence exists on the test-retest reliability for walking measures and step variability measures for forward walking [168–170] but similar information for backward and tandem walking is lacking. Furthermore, test-rest reliability of MOS and its variability is also not examined. It is therefore necessary to examine the test-retest reliability of walking performance and balance control measures across walking styles that are used by researchers to assess balance control.

Reliability can be described as the consistency or the repeatability of a measure [7, 9]. Reliability is defined in terms of relative or absolute reliability [9]. Relative reliability refers to the ability of a measure to differentiate performance between individuals and absolute reliability is the consistency in scores for each individual [9]. Various methods have been used to estimate relative and absolute reliability but often used measures to estimate relative and absolute reliability are intraclass correlation coefficient (ICC) using an ANOVA and standard error of measurement respectively (SEM) respectively [17, 65, 81, 168, 171, 172].

Relative reliability can be measured in terms of interrater, intrarater, and test-retest reliability [173]. Interrater reliability measures the consistency of a measure when the same measure is administered by two or more raters, intrarater reliability is the consistency of scores when the same rater scores the same sample multiple times, and test-retest reliability identifies the consistency of a test or measure in a sample when the same test or measure is administered to the same sample multiple times [173]. Test-retest reliability is similar to intrarater reliability where the same rater administers the measure, but the term test-retest is used when the raters are not involved in scoring the performance [7, 173], e.g.,spatiotemporal gait measures obtained from motion capture systems, IMUs, or pressure sensitive mats. For identifying the reliability of continuous variables such as spatiotemporal and balance control measures during walking obtained from a 3-D motion capture system, the most appropriate measure would be test-retest reliability [173]. ICC can be calculated using a variety of formulas depending on the type of reliability analysis and the research design [7]. For test-retest reliability, it is recommended that researchers use the two-way mixed effect ANOVA model with absolute agreement. In this approach, only the participants in the sample are of interest and the reliability values do not need to be generalized to other participants [7].

Absolute reliability is the consistency when a measure or a test is administered to the same person multiple times [9]. Measured using the SEM, absolute reliability provides an estimate of an individual's true score or the precision of the test or measure [9]. A smaller SEM would indicate a greater absolute reliability of a measure or instrument. One method of calculating the SEM is using the standard deviation of the sample and the test-retest reliability SEM = SD * $\sqrt{1-ICC}$ [9]. The SEM is also used to calculate the minimal detectable change $(MDC_{95} = SEM * \sqrt{2} * 1.96)$ which is a value that indicates whether a change in scores of a measurement is a true change or change attributed to error [9].

2.13 Summary of literature and gaps in literature

Falls are a cause of concern in the Canadian and the global population [1, 25]. A vital and modifiable risk factor for falls is balance control [26]. One method to improve balance control is through provision of added haptic input [123]. Investigating strategies used by individuals to maintain stability across different walking styles and effects of added haptic input to improve balance during walking can reduce future falls and fall related injuries. Currently, gaps exist in literature regarding the reliability of balance control measures employed during forward, backward, and tandem walking. Kinematic and kinetic differences between forward and backward walking have been examined in the past but the information on the balance control strategies utilized during backward walking is lacking. The ability of backward walking to identify deficits in mobility and the positive effects on balance control by practicing backward walking warrants investigation into how individuals maintain balance when walking backward. Identifying balance control strategies during backward walking is also required to examine whether sensory input from the visual, vestibular, and somatosensory systems is utilized in a comparable manner as forward walking. Knowledge about balance control strategies can also provide further support in using backward walking as a measure of mobility and balance control along with forward walking.

Balance control during walking is maintained by generating motor output based on sensory information from the visual, somatosensory, and vestibular systems [39, 45]. In addition to the balance control strategies employed during backward walking, the process of sensorimotor integration during backward walking is also not known [174]. Finally, use of haptic anchors has shown beneficial but short-term effects on balance control during walking [42, 120–122]. The beneficial effects disappear when walking without haptic anchors [83]. Data are needed to identify the effects of using haptic anchors over a period of time and whether these effects persist when walking without the haptic anchors. These gaps in the literature will be addressed in the following sections through chapters 3 to 6. This thesis focuses on examining the balance control strategies and sensorimotor integration during backward walking. This thesis also examines the effects of a haptic input-based intervention on balance control during forward, backward, and tandem walking.

2.14 Aims, hypotheses, and anticipated outcomes

Study one (Chapter 3)

Examining the test-retest reliability and measurement error for spatiotemporal and balance control measures during forward, backward, and tandem walking in healthy adults.

Aims

The aim of this study was to examine the test-retest reliability, measurement error, and minimal detectable change for spatiotemporal, MOS, and variability measures before and after six weeks for forward, backward, and tandem walking.

Hypothesis

All spatiotemporal and balance control measures will demonstrate moderate to excellent reliability as measured by intraclass correlation coefficient (ICC) for forward, backward, and tandem walking similar to previous ICC values for spatiotemporal measures for forward walking [170, 175].

Anticipated outcomes for study one

The outcomes from this study will provide measurement error and minimal detectable change values for the spatiotemporal, margin of stability, and variability measures to quantify clinically meaningful changes in balance control during different walking styles. These analyses are important first steps towards validating the use of MOS and step parameters during walking (forward, backward, and tandem) in conjunction with clinical tests of balance as a potential measure in identifying individuals that are at a higher risk of sustaining a fall.

Study two (Chapter 4)

Differences in balance control strategies between forward and backward walking and correlation of backward walking velocity with biomechanical balance measures.

Aims

The first aim of this study is to examine differences in strategies to maintain balance control between forward and backward walking. The second aim is to examine the association of backward walking velocity with biomechanical measures of balance control during forward and tandem walking.

Hypothesis one

Step length will be lower, step width will be higher and balance control will be lower during backward walking compared to forward walking.

Hypothesis two

A significant positive correlation will be present between velocity during backward walking and the magnitude of balance control parameters during forward and tandem walking. A higher velocity during backward walking will be associated with increased balance control (i.e., higher MOS) values and lower variability values (of step and MOS measures) during forward and tandem walking.

Anticipated outcomes for study two

The results of this study will help identify how balance is controlled by healthy adults via foot placement when walking backward. Examining step size (step length and step width) and its effects on MOS will highlight differences in balance control strategies between backward and forward walking. Correlation results will highlight associations between backward walking velocity and biomechanical balance control parameters during forward and tandem walking. Any association between backward walking velocity and biomechanical balance measures will provide further evidence and support for using backward walking as a measure of balance control by therapists and researchers.

Study three (Chapter 5)

The effects of vision and added haptic input on spatiotemporal and balance control measures during backward walking.

Aims

The aims of this study are to examine whether the availability of vision and added haptic input change step and balance control parameters during backward walking and whether utilization of added haptic input changes with the availability of vision.

Hypothesis one

Velocity and step length will reduce, the relative amount of time in the double support phase and step width will increase, and balance control will be lower when visual input is absent during backward walking. Velocity, the relative amount of time in the double support phase, step length and step width will be unchanged whereas step and MOS variability will be lowered when walking with added haptic input.

Hypothesis two

The effects of added haptic input on step and balance control parameters will be significantly greater during backward walking with eyes closed.

Anticipated outcomes

Vision is vital for step placement and balance control during forward walking [75, 106]. The outcomes from this study will highlight the role of vision during backward walking and whether the provision of haptic input can alter backward walking behaviour.

Study four (Chapter 6)

The effects of an intervention using haptic anchors on spatiotemporal and balance control parameters during forward, backward, and tandem walking.

Aims

Previous studies have found that somatosensory feedback during walking immediately improves walking performance, but the improvements are not retained when the somatosensory feedback is removed [83, 84]. The lack of retention effects of the added somatosensory feedback could be due to an insufficient amount of practice trials [83, 84]. This study aims to examine if practicing walking tasks using haptic anchors over six weeks improves balance control when added haptic input is removed i.e., walking without the haptic anchors. Using a pretest posttest study design, walking performance and balance control measures were compared across three groups before and after six weeks during forward, backward, and tandem walking. One group underwent a thrice/week, six-week intervention using haptic anchors (wHA), another group performed the same intervention without the haptic anchors (nHA), and the third group did not undergo the intervention (CTL). Participants walked with and without haptic anchors and with eyes open and eyes closed for each walking style during the pre- and posttest sessions. Each walking performance and balance control outcome variable for forward, backward, and tandem walking was converted to a change score using the formula: Change score = (Posttest score – pretest score/ pretest score).

Hypothesis one

Changes scores for MOS measures will increase and change scores for variability of MOS and step parameters will decrease in the group that underwent the intervention using haptic anchors compared to the group that underwent the intervention without the haptic anchors and the group that did not complete the intervention for forward, tandem, and backward walking.

Hypothesis two

a). Change scores for gait velocity, the relative amount of time spent in the double support phase, step length, and step width will show no significant differences during forward and backward walking between the three intervention groups because participants were asked to walk at their preferred speed.

b). Change scores for gait velocity will increase, the relative amount of time spent in the double support phase will decrease, and step length and step width will show no significant differences for tandem walking in the group that underwent the intervention using haptic anchors compared to the group that underwent the intervention without the haptic anchors and the group that did not complete the intervention.

Anticipated outcomes

Previous research has demonstrated that the beneficial effects of using haptic anchors are acute, meaning that the effects cease to exist when haptic anchors are removed [83, 120–122]. The outcomes from this study will demonstrate whether acute adaptation effects observed when walking with haptic anchors persist when walking without haptic anchors after practicing walking with the haptic anchors for six weeks. The results will also highlight whether an intervention with added haptic input has the capacity to influence the specificity of balance control as examined by changes in outcome measures during backward and tandem walking and whether the effects transfer to forward walking.

Chapter three

Study 1: Examining the test-retest reliability of spatiotemporal and balance control measures for forward, backward, and tandem walking.

Abstract

Margin of stability (MOS) and variability of step measures are used to examine balance control capabilities during walking. MOS has been validated against clinical tests of balance, but the test-retest reliability has not been examined. Test-retest reliability for MOS and variability measures during backward and tandem walking have not been examined. The purpose of this study was to examine the test-retest reliability of MOS and step measures and the variability of those same measures during forward, backward, and tandem walking. MOS and step data were obtained from fifteen participants (11 females, age: M = 27.2, SD = 10.8 years), (mass: M = 69.6, SD = 13.1 kg), (height: M = 1.7, SD = 0.1 m) who completed five trials each of forward, backward, and tandem walking twice, six weeks apart. The intraclass correlation coefficient (ICC) was used to assess the test-retest reliability for the outcome measures. Standard error of measurement (SEM) and minimal detectable at 95% confidence (MDC_{95}) were also calculated for each outcome measure. ICC values for MOS and step measures during forward (ICC range: 0.83- 0.95), backward (ICC range: 0.76-0.93), and tandem walking (ICC range: 0.66-0.93) showed moderate to excellent reliability except for step width during tandem walking that demonstrated a poor reliability (ICC=0.26). Variability of MOS and step measures showed poor reliability for forward (ICC range: -0.29-0.12), backward (ICC range: -0.07-0.69), and tandem walking (ICC range: 0.07-0.39). Results from the current study show that MOS is a reliable outcome measure across different walking styles whereas variability measures should be assessed with caution when trying to distinguish balance control between individuals. Across all measures, SEM values ranged from 0.01-43.91 for forward walking, 0.02-23.82 for backward walking, and 0.02-19.08 for tandem walking. Similarly, MDC_{95} values ranged from 0.02-63.12 for forward walking, and 0.04- 66.02 for backward walking and 0.04-52.90 for tandem walking.

3.1:Introduction

Spatiotemporal measures, margins of stability (MOS), and their variability are commonly used to assess walking performance and balance control during walking [124]. The MOS is a frequently utilized lab-based biomechanical measure that has spatial and temporal components [6]. A dynamic MOS is based on the control of the extrapolated centre of mass (xCOM) with respect to the base of support (BOS) [6]. The xCOM includes the position and velocity of the centre of mass (COM), providing a better estimation of balance control ability in dynamic tasks such as walking where the COM is not always within the BOS. The spatial component of the MOS is the instantaneous distance between the xCOM and the BOS [6]. A higher MOS value means that the xCOM has a greater distance to travel before crossing the edge of the BOS, after which, corrective action is required to avoid falling. Based on the equation used to calculate the MOS, a greater MOS value during walking indicates that an individual is less likely to incur a fall.

Step variability can be analyzed using standard deviation or coefficient of variation of step parameters over multiple steps or strides. Step variability is a measure that indicates the flexibility as well as the robustness of the central nervous system to achieve a consistent and predictable walking pattern with appropriate modifications in response to perturbations [27]. Some variability during walking is necessary in order to move successfully over different surfaces, under varying light conditions, and to avoid obstacles [27]. Excess step variability is detrimental and indicates the inability of the motor system to achieve a consistent gait pattern leading to an increased risk of falls [27, 29, 176]. Brach et al. [31] found that individuals with step width variability greater than a coefficient of variation of 30% and less than 7% were more likely to report a fall compared to individuals with step width variability values between 7 and 30%. Step variability should be interpreted as beneficial or unfavourable depending on the task and the environment in which the task is performed. For example, when responding to an external perturbation such as slip or a trip, an increase in step variability is necessary to regain stability and continue walking but an increase in step variability when walking on a firm surface in a well-lit obstacle free environment is an indicator of a reduced balance control [27].

A test or measure used to analyze performance has to be consistent over time with minimal error when administered repeatedly. A weak consistency and high amount of error make interpretation and generalizability difficult. MOS and step variability measures during forward walking have demonstrated validity [124], but information on the test-retest reliability and error of MOS during forward normal walking is sparse. A recent study by de Jong et al. [171] has shown good to excellent test-retest reliability for MOS in the mediolateral direction in individuals with spinal cord injury, stroke, and other neurological conditions; however, participants in de Jong et al's study walked on a self paced treadmill and not overground. Additionally, the test-retest reliability and error of MOS and stepping parameters during backward walking and tandem walking have not been examined. ICCs measure the relative or between-participant reliability, which is the ability of a measure to distinguish differences between individuals [7, 9]. The absolute or within-participant reliability is measured using SEM that identifies the amount of variability between trials and provides an estimate about an individual's true score on a measure [9]. The SEM is also used in the calculation of minimal detectable change (MDC) that determines whether a change in the score of a measure is true change or attributed to error [9]. Since the measures of MOS and step variability are widely used in studies of balance and gait, it is necessary to examine the between and within-participant reliability of these measures.

MOS and step variability measures have mainly been used to examine forward normal walking, with some studies also examining forward tandem walking [44, 53, 177]. Tandem walking challenges the balance control system by narrowing the BOS and is used as a test for dynamic balance and as an exercise in gait rehabilitation [77, 78]. In addition to forward and tandem walking, backward walking has demonstrated clinical utility with its ability to distinguish individuals with balance deficits [55, 56, 178]. Backward walking has also gained popularity as a potential exercise in balance training programs [174]. Reliability studies that have examined backward and tandem walking have focussed on assessing the reliability of the time required to complete backward and tandem walking over a fixed distance. The test-retest reliability of stepping and balance control measures during backward and tandem walking have not been examined. It is essential to examine the reliability and error of commonly used balance control measures when performing various walking tasks since variations in walking styles are used in the research environment to evaluate balance control.

The purpose of this study was to analyze the test-retest reliability, SEM, and minimal detectable change at 95% confidence interval (MDC_{95}) for spatiotemporal and balance control parameters during forward normal, backward, and forward tandem walking. It was hypothesized that all spatiotemporal and balance control parameters will demonstrate good to excellent test- retest reliability for forward, backward, and tandem walking. The SEM and MDC_{95} values obtained from this study can be used as reference for future studies examining similar outcome variables in different environments and walking protocols.

3.2: Methods

3.2.1: Participants

Participants were recruited through word of mouth and announcements on university web portals. Participants were recruited after screening for inclusion criteria using a questionnaire. The inclusion criteria were a) age between 18 to 55 years; b) not currently involved in a balance training program or activities; c) not living with any conditions that affect balance; d) not having any visual impairment that could not be corrected with eyewear;

e) and not having reduced or lost sensation in their upper and lower extremities. The study was approved by the university research ethics board (BIO 17-157). Written informed consent was obtained from all participants before data collection.

3.2.2: Protocol

Demographic data were collected during both data collection sessions. Age was calculated from participants' self-reported birth year and month. Mass was measured using a weigh scale and height was measured using a stadiometer. Data presented in this study are from participants who were part of a six-week intervention study and who were pseudo-randomly assigned to the control group who received no intervention during the six-week period between pre- and post-testing. As such, data here are from participants who were tested twice – before and after six weeks. Participants performed five trials each of forward, backward, and tandem walking with eyes open at their preferred speed in random order on a rigid 10-metre long walkway in a lab. During backward walking, participants were asked to stop walking by the researcher two steps before the end of the platform to avoid stepping off the platform or reaching the walls of the lab.

3.2.3: Instrumentation

Kinematic data were obtained using a 3-D motion capture system (Vicon Nexus, Vicon Motion Systems, Centennial, CO) with a sampling frequency of 100 Hz. Sixty-three reflective markers were placed on anatomical landmarks of the body to generate a 12-segment full-body model (Figure 3.1). The full-body model was used to calculate the total body centre of mass (COM) based on anthropometric tables [179].

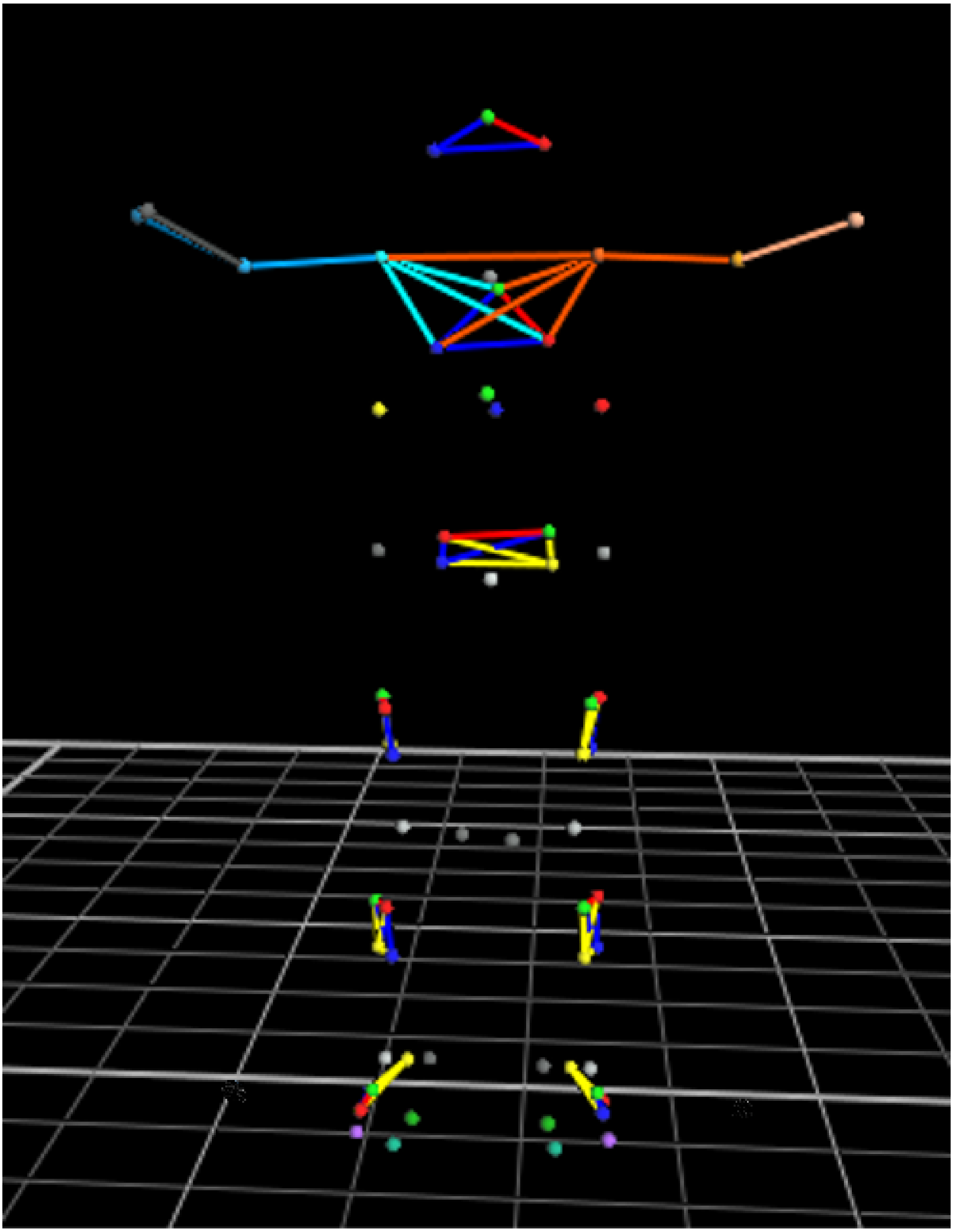

Figure 3.1: Marker set and full body model for 3-D data capture.

3.2.4: Data analysis

Raw marker data were filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 8Hz. Filtering of data and calculation of outcome variables were performed using customized scripts (MATLAB R2019b for PC, MathWorks, Natick, MA). All outcome variables were calculated post the data filtering process. Spatiotemporal parameters included stride velocity, the relative amount of time in double support during a gait cycle (%DS), step length (SL), and step width (SW). Stride velocity was calculated by dividing stride length by stride time and was normalized using the formula: (normalized stride velocity (nSV) = stride velocity/$\sqrt{H * g}$) where H is the participant's leg length, and g is the acceleration due to gravity (9.81 m/s^2) [5]. The relative amount of time in double support was calculated as the time duration in percentage when both the feet were on the ground during one (100%) gait cycle. Step length was calculated as the anteroposterior distance between the right and left heel markers at each foot strike and normalized to participants' leg length (nSL) [5]. Leg length was calculated as the distance from the heel to the hip joint centre. Step width was calculated as the mediolateral distance between the right and left heel markers at each foot strike. Balance control was examined using the MOS and variability of nSL, SW, and MOS. MOS was calculated as suggested by Hof et al. [6] as the distance between the extrapolated centre of mass (xCOM) and the base of support (BOS). Instantaneous MOS values over each stride were averaged for each trial. The mean values across the trials were used for analysis. The MOS in the anteroposterior direction (AP_MOS) for forward and tandem walking was calculated over the duration of each stride as the distance between the xCOM and the heel of the trailing limb; the AP_MOS for backward walking was calculated over each stride as the distance between the xCOM and the toe of the trailing limb. The MOS in the mediolateral direction (ML_MOS) was calculated over the duration of each stride as distance between the xCOM and the closest lateral edge of the BOS. The standard deviation (SD) value was used to calculate the variability of the MOS (AP_MOS_SD) (ML_MOS_SD) and step (nSL_SD) (SW_SD) parameters. The number of strides/trial for each participant in this study ranged from 5-10 strides for forward walking, 5-20 strides for backward walking, and 10-20 strides for tandem walking obtained by adding the number strides across five trials.

3.3: Statistical analysis

Test-retest reliability was assessed using ICC (3, k) which was calculated using the two-way mixed model with absolute agreement [7]. ICCs were calculated using the analysis of variance models for each outcome variable for each walking style.

ICC values were interpreted as poor (<0.5), moderate (0.5-0.75), good (0.75-0.90), and excellent (ICC>0.9) [7]. Absolute reliability was calculated using the standard error of measurement (SEM). The SEM indicates the distribution of values around a participant's true score [8, 9, 180]. SEM was calculated using the formula SEM = SD * $\sqrt{1-ICC}$ where SD difference is the standard deviation of the difference in scores from session one and session two and ICC is the square root of the test-retest reliability coefficient. Minimal detectable change at 95% confidence interval (MDC_{95}) was calculated using the formula MDC_{95} = SEM x $\sqrt{2}$ x 1.96. The SEM and MDC_{95} were also expressed as a percentage of the difference in mean score between the two test sessions [9].

3.4: Results

A total of fifteen participants completed two data collection sessions six weeks apart. The participants had a mean age of 27.1±13.1 years, mean mass of 69.5±7.9 kg, and a mean height of 1.7±0.08 metres.

3.4.1: Forward walking

The ICC, SEM, and MDC values for outcome measures during forward normal walking are provided in Table 3.1. All outcome measures demonstrated good to excellent reliability with ICC values ranging from 0.83 to 0.95. nSV demonstrated the highest test-retest reliability (ICC = 0.95) followed by nSL (ICC = 0.93), AP_MOS (ICC = 0.91), SW (ICC = 0.84), %DS (ICC = 0.83), and ML_MOS (ICC = 0.83). Variability measures demonstrated poor reliability with ICC values ranging from -0.29 to 0.12. SW_SD (ICC = -0.29) and nSL_SD (-0.26) had the lowest test-retest reliability followed by AP_MO_SD (ICC = 0.05), and ML_MOS_SD (ICC = 0.13). SEM values ranged from 0.01-43.91 and $MDC_9$5 values ranged from 0.02-63.12 across all measures.

Table 3.1: ICC, SEM, and MDC values for outcomes variables during forward walking.

Variable	Pretest Mean (SD)	Posttest Mean (SD)	Difference between test sessions (Posttest-pretest)	ICC (95% CI)	SEM (SEM %)	MDC_{95} (MDC_{95}%)
nSV (a.u.)	0.41 (0.07)	0.42 (0.06)	0.01	0.95 (0.86-98)	0.02 (5.16)	0.06 (14.30)
%DS	29.34 (3.11)	29.14 (2.99)	-0.20	0.83 (0.49-0.94)	1.68 (5.73)	4.65 (15.89)
ML_MOS (mm)	98.75 (14.73)	99.31 (10.97)	0.56	0.83 (0.48-0.94)	7.20 (7.27)	19.94 (20.14)
ML_MOS_SD (mm)	7.81 (1.97)	7.95 (3.89)	0.14	0.13 (-1.96-0.72)	2.99 (37.92)	8.28 (105.10)
AP_MOS (mm)	633.2 (79.05)	643.54 (73.60)	10.34	0.91 (0.74-0.97)	31.05 (4.86)	86.06 (13.48)
AP_MOS_SD (mm)	23.64 (10.30)	27.64 (10.46)	4.00	0.05 (-1.77-0.68)	10.24 (39.95)	28.39 (110.74)
SW (mm)	91.51 (31.23)	87.8 (27.71)	-3.71	0.84 (0.54-0.95)	15.57 (17.37)	43.16 (48.14)
SW_SD (mm)	20.52 (5.68)	23.31 (4.53)	2.79	-0.29 (-2.46-0.55)	5.80 (26.46)	16.07 (73.34)
nSL (a.u.)	0.74 (0.07)	0.76 (0.07)	0.02	0.93 (0.78-0.98)	0.03 (3.51)	0.07 (9.73)
nSL_SD (a.u.)	0.03 (0.01)	0.02 (0.007)	-0.01	-0.26 (-2.85-0.58)	0.01 (59.03)	0.04 (163.62)

nSV = normalized stride velocity, %DS = relative amount of time in the double support phase, ML_MOS = mediolateral margin of stability, ML_MOS_SD = mediolateral margin of stability variability, AP_MOS = anteroposterior margin of stability, AP_MOS_SD = anteroposterior margin of stability variability, nSL = normalized step length, SW = step width, nSL_SD = step length variability, SW_SD = step width variability.

3.4.2: Backward walking

The ICC, SEM, and MDC values for outcome measures during backward walking are provided in Table 3.2 with ICC values ranging from 0.76 to 0.93. Similar to forward walking, the highest test-retest reliability was observed for nSV (ICC = 0.93), followed by nSL (ICC= 0.92), AP_MOS (ICC = 0.90), %DS (ICC = 0.88) ML_MOS (ICC = 0.82), and SW (ICC = 0.76). Variability measures demonstrated poor to moderate reliability ranging from -0.07 to 0.69. AP_MOS_SD had the lowest reliability (ICC = -0.07) followed by nSL_SD (ICC = 0.25), ML_MOS_SD (ICC = 0.55), and SW_SD (ICC = 0.69). SEM values ranged from 0.02-23.82 and MDC_{95} values ranged from 0.04-66.02 across all measures.

Table 3.2: ICC, SEM, and MDC values for outcomes variables during backward walking.

Variable	Pretest Mean (SD)	Posttest Mean (SD)	Difference between test sessions (Posttest-pretest)	ICC (95% CI)	SEM (SEM %)	MDC_{95} (MDC_{95}%)
nSV (a.u.)	0.29 (0.06)	0.32 (0.06)	0.03	0.93 (0.40-0.98)	0.02 (5.61)	0.05 (15.54)
%DS	22.04 (4.58)	22.05 (4.53)	0.01	0.88 (0.63-0.96)	2.20 (9.97)	6.09 (27.63)
ML_MOS (mm)	119.89 (18.22)	126 (16.63)	6.11	0.82 (0.47-0.94)	9.21 (7.49)	25.52 (20.75)
ML_MOS_SD (mm)	13.15 (6.03)	11.55 (4.25)	-1.6	0.56 (-0.28-0.85)	4.09 (33.08)	11.33 (91.71)
AP_MOS (mm)	503.67 (79.97)	534.68 (70.67)	31.01	0.90 (0.50-0.97)	26.02 (5.01)	72.13 (13.89)
AP_MOS_SD (mm)	36.71 (15.85)	30.7 (12.54)	-6.01	-0.07 (-2.09-0.64)	14.68 (43.56)	40.70 (120.75)
SW (mm)	144.04 (35.98)	147.49 (39.98)	3.45	0.76 (0.27-0.92)	24.06 (16.51)	66.69 (45.75)
SW_SD (mm)	30.03 (11.50)	28.73 (7.09)	-1.3	0.70 (0.09-0.90)	6.59 (22.43)	18.26 (62.17)
nSL (a.u.)	0.56 (0.07)	0.59 (0.08)	0.03	0.92 (0.63-0.98)	0.03 (4.80)	0.08 (13.31)
nSL_SD (a.u.)	0.05 (0.02)	0.06 (0.03)	0.01	0.26 (-1.43-0.76)	0.02 (45.00)	0.07 (124.73)

nSV = normalized stride velocity, %DS = relative amount of time in the double support phase, ML_MOS = mediolateral margin of stability, ML_MOS_SD = mediolateral margin of stability variability, AP_MOS = anteroposterior margin of stability, AP_MOS_SD = anteroposterior margin of stability variability, nSL = normalized step length, SW = step width, nSL_SD = step length variability, SW_SD = step width variability.

3.4.3: Tandem walking

The ICC, SEM, and MDC values for outcome measures during forward walking are provided in Table 3.3. All outcome measures except step width demonstrated moderate to excellent reliability with ICC values ranging from 0.66 to 0.93. ML_MOS (ICC = 0.93) had the highest test-retest reliability followed by SL (ICC = 0.91), AP_MOS (ICC = 0.88), stride velocity (ICC = 0.87), and %DS (ICC = 0.67). Step width had a poor ICC value of 0.26. Variability showed poor reliability with measures ranging from 0.07 to 0.39. ML_MOS_SD (ICC = 0.07) had the lowest test-retest reliability followed by SL_SD (ICC = 0.28), AP_MOS_SD (ICC = 0.39), and SW_SD (ICC = 0.40). Across all measures, SEM values ranged from 0.02-19.08 and MDC_{95} values ranged from 0.04-52.90 across all measures.

Table 3.3: ICC, SEM, and MDC values for outcomes variables during tandem walking.

Variable	Pretest Mean (SD)	Posttest Mean (SD)	Difference between test sessions (Posttest-pretest)	ICC (95% CI)	SEM (SEM %)	MDC_{95} (MDC_{95}%)
nSV (a.u.)	0.16 (0.04)	0.15 (0.04)	-0.01	0.87 (0.62-0.96)	0.02 (14.37)	0.06 (39.82)
%DS	39.75 (3.96)	39.79 (5.04)	0.04	0.67 (-0.03-0.89)	3.26 (8.19)	9.02 (22.69)
ML_MOS (mm)	56.37 (9.82)	55.6 (11.29)	-0.77	0.93 (0.80-0.98)	3.89 (6.95)	10.79 (19.27)
ML_MOS_SD (mm)	6.99 (2.73)	5.79 (1.93)	-1.2	0.07 (-1.45-0.68)	2.32 (36.28)	6.43 (100.55)
AP_MOS (mm)	347.44 (53.53)	343.94 (56.65)	-3.5	0.88 (0.64-0.96)	26.08 (7.54)	72.28 (20.91)
AP_MOS_SD (mm)	37.46 (19.80)	21.24 (8.58)	-16.22	0.39 (-0.30-0.76)	12.19 (41.52)	33.78 (115.09)
SW (mm)	22.91 (13.06)	21.81 (9.32)	-1.1	0.26 (-1.44-0.76)	10.52 (47.03)	29.15 (130.37)
SW_SD (mm)	14.28 (4.40)	15.53 (5.92)	1.25	0.40 (-0.85-0.80)	4.55 (30.52)	12.61 (84.60)
nSL (a.u.)	0.38 (0.08)	0.36 (0.07)	-0.02	0.91 (0.74-0.97)	0.03 (8.71)	0.09 (24.15)
nSL_SD (a.u.)	0.04 (0.03)	0.02 (0.01)	-0.02	0.28 (-0.52-0.72)	0.02 (71.98)	0.06 (199.53)

nSV = normalized stride velocity, %DS = relative amount of time in the double support phase, ML_MOS = mediolateral margin of stability, ML_MOS_SD = mediolateral margin of stability variability, AP_MOS = anteroposterior margin of stability, AP_MOS_SD = anteroposterior margin of stability variability, nSL = normalized step length, SW = step width, nSL_SD = step length variability, SW_SD = step width variability.

3.5: Discussion

The purpose of this study was to assess the test-retest reliability of spatiotemporal and balance control parameters across different walking styles. To the best of our knowledge, this is the first study to examine the test-retest reliability of spatiotemporal parameters and MOS for forward, backward, and tandem walking. The reliability of spatiotemporal variables was similar to previous research showing good to excellent reliability during forward normal walking [169, 170, 175]. The examination of test-retest reliability of the mean of spatiotemporal parameters during backward and tandem walking is novel in that it has not been previously examined and showed good to excellent reliability (except step width during tandem walking).

MOS measures also showed good to excellent test-retest reliability across all walking styles. The study by de Jong et al. [171] found partly comparable results to our study where ML_MOS had good to excellent reliability in individuals with neurological conditions but a moderate reliability in healthy adults. The combined results from the current study and de Jong et al. [171] together provide support in favour of ML_MOS being a reliable measure in healthy and clinical populations as well as during overground and treadmill walking. The current results showing good to excellent test-retest reliability further support the use of spatiotemporal variables and MOS to examine balance control over different testing sessions and across different walking styles.

Variability outcomes demonstrated poor to fair reliability for all measures and across all walking styles, similar to results in previous literature [168, 169, 172, 181]. The poor ICC scores for variability can be attributed to multiple factors. First is the homogeneity among the participants. The calculation of ICC is based on the ratio of between-participant variance to the total amount of variance [7, 9]. Therefore, if the between-participant variance is low, it leads to an ICC value closer to zero. The cohort of participants in the current study were healthy adults (primarily university students) without any conditions affecting their walking and balance (a more homogenous sample), and hence, might have had a similar amount of variability during walking.

Another reason for poor reliability in the variability measures could be the testing protocol. A higher number of strides and continuous walking over a longer distance improves the consistency of gait measures [170, 182]. Literature suggests using 15-50 strides of continuous walking to obtain a reliable measure of gait outcomes [183, 184]. The amount of strides/trial for each participant in this study ranged from 5-10 strides for forward walking, 5-20 strides for backward walking, and 10-20 strides for tandem walking obtained by adding the number strides across five trials. Data obtained by averaging values over multiple trials instead of continuous walking could have negatively affected the reliability values. Since values of standard deviation are influenced by the number of samples and the mean [185], the low reliability of variability measures could be explained due to a low number of strides for certain participants as well as standard deviation calculated by averaging values from five short passes over a walkway instead of continuous walking over a longer distance.

Future studies need to compare variability outcomes obtained by walking multiple times over a short distance to walking continuously over a longer distance such as in a hallway or walking outdoors. However, caution should be exercised since increasing the amount of walking trials or walking over longer distances can cause fatigue which can impact the variability of outcome measures [186, 187] and might not be ethically feasible for certain populations.

Absolute reliability was calculated using the SEM. The SEM values for forward normal walking obtained in this study were similar to previous studies [168]. The SEM provides an estimate of how far a score is spread around a person's true score [9]. Therefore, SEM provides within-subject reliability values and is a better measure of reliability for gait studies to assess changes in individual performance at multiple time points or if a test or measure is administered multiple times to the same individual. The SEM values obtained in this study provide a reference for future studies that examine changes in MOS and variability across multiple walking tasks and in response to interventions.

The MDC_{95} is the value that indicates the minimum amount of change required in a variable for it to be considered real change and not a change due to error. The MDC_{95} values obtained in this study provide a novel insight into what is considered true change in terms of spatiotemporal and balance control measures in healthy adults. The values obtained from this study can be used as a guide to examine change in similar outcomes measures in clinical populations. The MDC_{95} values can also be used to examine the efficacy of rehabilitation programs to assess whether changes in behaviour are true changes or due to error.

There were certain limitations to this study. Participants were healthy adults (most of whom were university students) who are less likely to change their walking over a period of six weeks due to the absence of injury and being free of any health complications. Future investigations need to include older adults and individuals living with neurological conditions such as stroke, spinal cord injury, Parkinson's disease, and multiple sclerosis to establish reliability since these populations might show a lower consistency compared to healthy adults in their walking behaviour across time points [168]. Secondly, a larger sample size is required to adequately establish test-retest reliability for use in clinical research [188]. A post hoc power analysis revealed that a sample size of fifteen, an alpha value of .05 and 80% power was sufficient to obtain an ICC value of 0.6. The current study had a sample size of fifteen participants that completed two test sessions before and after six weeks. To obtain a similar ICC of 0.6 at 90% power requires a sample size of twenty participants [188]. Future studies should include an a-priori power analysis to achieve an adequate sample size if the study includes clinical populations. A third limitation was reporting of the sex of the participants. The sex was assumed by the researcher based on expression of gender and the information about sex and gender was not obtained from the participants directly. A fourth limitation was that sex-based analyses for reliability was not performed for any of the three walking styles. Future work is needed to identify differences in walking and balance control based on sex and gender by obtaining information about sex and gender directly from the participants instead of assuming sex based on gender expression.

Conclusion

In summary, spatiotemporal and MOS parameters demonstrated good to excellent test-retest reliability for forward, backward, and tandem walking in an overground lab setting in healthy adults that can be used to assess change in walking performance and balance control in response to gait and balance training programs. Researchers and clinicians can utilize changes in MOS to examine fluctuations in balance control capacity across repeated test sessions in longitudinal studies or during follow up sessions after completing rehabilitation programs. The assessment and interpretation of variability should be undertaken with caution as variability is error-prone due to factors such as homogeneity of participants, data obtained from continuous walking, and the number of steps used in the calculation of variability outcomes.

Relevance of Study 1 to the thesis

Relevance of Study 1 to the thesis The purpose of study one was to evaluate the test-retest reliability and error of spatiotemporal and balance control measures across three walking styles. For a measure to be generalized to different populations and across studies, it must assess a construct consistently over time with minimal error [180]. MOS is used in literature to assess differences in balance control between healthy and clinical populations and to examine strategies to regain balance in response to perturbation [16, 88]. Gait variability predicts falls and an increase in variability is indicative of an impairment in balance control [28]. MOS and variability measures have the ability to distinguish between balance impaired and healthy populations, but the reliability of these measures is not known [16, 43, 169]. Furthermore, time required to walk over a fixed distance is often used to evaluate performance for tandem and backward walking, but spatiotemporal MOS and variability are not frequently examined [65,66,81]. The findings from this study provide novel evidence on the reliability of MOS and variability measures across three distinct walking styles. This study demonstrated that, in a spatially confined setting such as a lab, the MOS is a reliable measure of balance control. Spatial variability measures assessed over multiple time points should be interpreted with caution or variability data should be obtained from a minimum of fifteen consecutive steps by walking on a treadmill, across a long hallway, or walking outdoors.

Chapter four

Study 2: Balance control strategies between forward and backward walking and correlation of backward walking velocity with biomechanical balance measures.

Abstract

Balance control is primarily examined using forward walking by researchers and clinicians. Recent evidence has suggested that backward walking distinguishes fallers from non fallers and backward walking velocity is correlated to performance on clinical tests of balance [49], [50]. The purpose of this study was to examine differences in balance control between forward normal and backward walking and to examine the correlation of backward walking velocity with biomechanical measures of balance control during forward and tandem walking. Fifty-five adults (37 females, age: M = 28.1, SD = 9.9 years), (mass: M = 71.4, SD = 14.8 kg), (height: M = 1.7, SD = 0.09 m) completed forward, backward, and forward tandem walking trials. Walking performance was evaluated using spatiotemporal parameters including stride velocity, relative time in double support, step length, and step width. Balance control was evaluated using margin of stability (MOS) and variability of MOS, step length, and step width. Outcome variables were compared between forward normal and backward walking. The correlation between backward walking velocity with balance control measures during forward normal and forward tandem walking was also examined.

Participants walked slower, and with shorter and wider steps during backward walking compared to forward normal walking. Backward walking was more variable compared to forward walking characterized by an increased variability of MOS, step length, and step width. Backward walking velocity was positively correlated with MOS in the anteroposterior direction and negatively correlated with step length variability during forward normal walking. Backward walking presents a challenge to the balance control system and backward walking velocity is associated with anteroposterior stability during walking. Future work could potentially use backward walking performance as a measure to assess balance control capabilities after establishing validity of backward walking in clinical populations.

4.1: Introduction

Dynamic balance control during walking has most often been examined during forward walking or performing tasks that involve forward movement, such as turning during walking [189]. Forward walking velocity is used to assess gait performance, identify individuals that are at risk of falls, monitor the progress of individuals undergoing rehabilitation and is deemed the 'sixth vital sign' of human health [190]. Although forward walking is the primary choice of assessment, some studies do use tandem walking to measure balance control [42, 53]. Tandem walking is used as a task to evaluate balance control as well as an exercise in gait rehabilitation [77, 78]. Tandem walking challenges balance control by reducing the size of the base of support (BOS) during a gait cycle. The size of the BOS is a crucial factor in balance control and any change or alteration in the BOS can affect balance [39]. Individuals need to constantly regulate the movement of COM within a small BOS that requires continuous activation of balance control strategies. In addition to forward walking, tandem walking is another task used to highlight the balance control capabilities of an individual.

Recent literature that has incorporated backward walking in addition to forward walking has shown the spatiotemporal parameters of backward walking such as gait velocity, stride length, percentage of double support, and size of BOS are more sensitive in detecting fallers from non-fallers than forward walking and can distinguish mobility impairments between healthy and clinical populations [55–59, 76, 178]. Furthermore, backward walking velocity is also highly correlated with clinical tests of balance, is better at identifying individuals with diminished balance control, and revealing walking impairments compared to forward walking [58, 64].

The study of joint movements between forward and backward walking has revealed that backward walking is just a reversal of forward walking [68]. Hip and knee movements during backward walking follow a reversed pattern of forward walking [68]. Along with lower limbs, movement of upper extremities also follow a time reversed pattern during backward walking [69]. These findings support the hypothesis that backward walking is simply a directionally reversed forward walking; however, analyses of neural activity using electroencephalography (EEG) have demonstrated that backward walking requires greater cognitive control and with significant differences in activation of cortical areas compared to forward walking [72, 73, 191]. An adaptation study by Choi et al. [192] further suggests that independent neural structures control backward walking. Using a split belt paradigm for backward walking with different belt speeds, post adaption effects persisted in backward walking, but no effects were observed during forward walking. Since altering backward walking behaviour had no impact on forward walking behaviour, it can be postulated that backward and forward walking are regulated individually through different neural structures.

Taken together, the outcomes of previous research show that even though kinematic joint patterns during backward walking might be similar to time-reversed forward walking, the control of stepping accomplished by muscle activation during backward walking may differ from forward walking. Step size (step length and width) and step consistency during walking are vital factors for balance control. Changes in step size ensures that the body's COM is contained within the BOS during walking [16]. An increase in step variability is related to an increase in risk of falls [28].

Backward walking has been analyzed in terms of kinematics, muscle activation, and brain activity, but balance control strategies utilized to keep walking backward without falling have not been examined. Significant associations are observed between backward walking velocity and clinical tests of balance [66] but information about correlations of backward walking velocity with biomechanical measures of balance during forward and tandem walking is lacking. The rationale for examining the association of backward walking velocity and biomechanical measures of balance control was based on findings from previous literature on forward walking where velocity has a high correlation with measures of mobility and various other aspects of health such as future morbidity and even mortality [190, 193]. Similar associations between backward walking velocity and measures of balance control can lend support in favour of using backward walking velocity as an overall measure of physical function, mobility, and balance control. The current study aims to address these gaps in literature. The choice of forward and tandem walking biomechanical measures is because of their frequent use in literature to identify mobility and balance control problems.

The first purpose of this study was to compare spatiotemporal and balance control parameters between forward and backward walking. It was hypothesized that spatiotemporal and balance control parameters during backward walking will be significantly different from forward walking. The second purpose was to examine the correlation between backward walking velocity and lab-based balance control measures during forward and tandem walking. Tandem walking was included as a challenging task to assess how individuals maintain stability when walking with a reduced BOS. The association between backward walking velocity and tandem walking outcomes could provide an insight into whether individuals with a higher backward walking velocity possess a higher functioning balance control system as evidenced by larger MOS and lower variability values during tandem walking. It was hypothesized that backward walking velocity will be positively correlated with MOS measures and negatively correlated with variability measures of forward and tandem walking.

4.2: Methods

4.2.1: Participants

Participants were recruited through word of mouth, by placing posters across the university campus, and through announcements on university web portals. Participants were screened to determine eligibility with an inclusion/exclusion questionnaire. Participants that were between the ages of 18 to 55 years; not currently participating in a balance training program or activities; not living with any conditions that affect balance; not having any visual impairment that could not be corrected with eyewear, and not having reduced or lost sensation in their upper and lower extremities were recruited for the study. The study was approved by the university research ethics board (BIO 17-157). Informed consent was obtained from each participant before commencing the data collection session.

4.2.2: Protocol

Participants performed five trials each of forward, backward, and tandem (heel-toe) walking, with their eyes open for a total of fifteen trials. The number of strides ranged from 1-3 strides, 1-4 strides, and 1-7 strides/trial/participant for forward, backward, and tandem walking, respectively. The walking trials were performed at the participants' preferred speed in a random order over 10-metres in a lab environment. Data obtained from the capture volume within the middle of the walkway were used for analyses. Individuals with a longer stride length resulted in fewer strides being recorded within the capture volume. During backward walking, participants were asked to stop walking by the researcher two steps before the end of the walkway or if the participants started to veer toward the walls of the lab. In the case where participants had to be stopped from veering towards the wall, the trial was repeated.

The sections below in italics are the same as mentioned in study 1 (Chapter three)

4.2.3: Instrumentation

Kinematic data were obtained using a 3-D motion capture system (Vicon Nexus, Vicon Motion Systems, Centennial, CO) with a sampling frequency of 100 Hz. Sixty-three reflective markers were placed on anatomical landmarks of the body to generate a 12-segment full-body model (Figure 3.1). The full-body model was used to calculate the total body centre of mass (COM) based on anthropometric tables [179].

4.2.4: Data analysis

Raw marker data were filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 8Hz. Filtering of data and calculation of outcome variables were performed *using customized scripts (MATLAB R2019b for PC, MathWorks, Natick, MA). All outcome variables were calculated post the data filtering process. Spatiotemporal parameters included stride velocity, the relative amount of time in double support during a gait cycle (%DS), step length (SL), and step width (SW). Stride velocity was calculated by dividing stride length by stride time and was normalized using the formula: (normalized stride velocity (nSV) = stride velocity/$\sqrt{H * g}$) where H is the participant's leg length, and g is the acceleration due to gravity (9.81 m/s^2) [5]. The relative amount of time in double support was calculated as the time duration in percentage when both the feet were on the ground during one (100%) gait cycle. Step length was calculated as the anteroposterior distance between the right and left heel markers at each foot strike and normalized to participants' leg length (nSL) [5]. Leg length was calculated as the distance from the heel to the hip joint centre. Step width was calculated as the mediolateral distance between the right and left heel markers at each foot strike. Balance control was examined using the MOS and variability of nSL, SW, and MOS. MOS was calculated as suggested by Hof et al. [6] as the distance between the extrapolated centre of mass (xCOM) and the base of support (BOS). Instantaneous MOS values over each stride were averaged for each trial. The mean values across the trials were used for analysis. The MOS in the anteroposterior direction (AP_MOS) for forward and tandem walking was calculated over the duration of each stride as the distance between the xCOM and the heel of the trailing limb; the AP_MOS for backward walking was calculated over each stride as the distance between the xCOM and the toe of the trailing limb. The MOS in the mediolateral direction (ML_MOS) was calculated over the duration of each stride as distance between the xCOM and the closest lateral edge of the BOS. The standard deviation (SD) value was used to calculate the variability of the MOS (AP_MOS_SD) (ML_MOS_SD) and step (nSL_SD) (SW_SD) parameters. Values for all spatiotemporal and balance control measures were averaged across the five trials for each walking style.*

4.3: Statistical analysis

The purpose and focus of the current study did not involve analysing sex-based differences in walking and balance control and therefore, no analyses were conducted to investigate differences between males and females for age, height, and mass. Furthermore, because sex of the participants was assumed, a sex-based analysis is not warranted. Information about gender was also not obtained from the participants directly and hence, a gender-based analysis was not performed.

All data were examined for normality using the Shapiro-Wilk test. Analyses for nSV, AP_MOS, AP_MO_SD, ML_MOS, and ML_MOS were performed for fifty-three out of fifty-five participants since values for two participants were lost due to data collection error.

Differences between forward and backward walking were examined using paired samples t-tests for normally distributed data (results shown with *p*) and the Wilcoxon signed-rank tests for non-normal data (results shown with p_w). Effects sizes were calculated using Cohen's *d* (*d* = mean difference/standard deviation of mean difference) and [10] r = Z score/$\sqrt{total\ number\ of\ observations}$ [11] for the paired samples t-test and Wilcoxon signed-rank test, respectively. Effect sizes were interpreted as small (0.2-0.5), medium (0.51-0.8), and large (>0.8) [10]. A conservative alpha value of .01 was used to account for multiple comparisons. The association between balance control variables during forward and tandem walking and backward walking were examined using Pearson's r for normally distributed data and Spearman's ρ for non-normal data.

4.4: Results

A total of 55 healthy adults (37 females) with a mean age of 28.1 $\pm$ 9.8 years, mean height of 1.71 $\pm$ 0.09 m, and mean mass of 71.3 $\pm$ 14.8 kg participated in the study. The sex of the participants was assumed by the researcher based on gender expression. Significant differences were observed in all spatiotemporal and balance control variables between forward and backward walking (Table 4.1).

Compared to forward walking, participants walked slower (t (52) = 18.4, p <.001, d = 2.52), spent less time in the double support phase (t (52) = 11.8, p <.001, d = 1.62), reduced their step length (Z = 6.4, p <.001, r = 0.87), and increased their step width (t (54) = 15.6, p <.001, d = 2.10) during backward walking. The differences in stepping behaviour were reflected in the values obtained for the MOS measures where participants had a significantly lower MOS in the anteroposterior direction (t (52) = 17.8, p <.001, d = 2.45) and a significantly higher MOS in mediolateral direction (t (52) = 14.2, p <.001, d = 1.96) during backward walking.

Table 4.1: Mean (standard deviation) values for outcome variables between forward and backward walking examined using the paired samples t-test and Wilcoxon signed rank test.

	Forward Walking	Backward Walking	*p*-value	Effect size
nSV (a.u.)	0.41 (0.06)	0.30 (0.06)	.001	2.52
%DS	29.61 (2.80)	22.21 (4.41)	.001	1.62
SW (mm)	85.38 (27.17)	140.76 (35.34)	.001	2.10
#nSL (a.u.)	0.76 (0.07)	0.60 (0.08)	.001	0.87
AP_MOS (mm)	635.37 (69.72)	524.11 (70.92)	.001	2.45
ML_MOS (mm)	101.01 (14.13)	126.58 (18.06)	.001	1.96

***Note: Significance was set at p <.01.** (nSV = normalized stride velocity, %DS = relative amount of time in the double support phase, nSL = normalized step length, SW = step width, AP_MOS = anteroposterior margin of stability, ML_MOS = mediolateral margin of stability).
# indicates variable(s) analyzed using the Wilcoxon signed rank test.

Backward walking was also significantly more variable compared to forward walking (Table 4.2). nSL_SD (Z = 5.6, p_w <.001, r = .76), SW_SD (Z = 5.3, p <.001, r = .71), AP_MOS_SD (Z = 3.1, p_w <.001, r = .43), and ML_MOS_SD (Z = 5.1, p <.001, r = .70), were significantly higher during backward walking compared to forward walking.

Table 4.2: Mean values for variability between forward and backward walking.

	Forward walking	Backward walking	p-value	Effect size
nSL_SD	9.49	17.26	.001	.77
SW_SD	21.34	29.82	.001	.71
AP_MOS_SD	23.61	33.54	.002	.43
ML_MOS_SD	7.94	13.46	.001	.70

***Note: Significance was set at p <.01.** (nSL_SD = step length variability, SW_SD = step width variability, AP_MOS_SD = anteroposterior margin of stability variability, ML_MOS_SD = mediolateral margin of stability variability)

Significant correlations between backward walking velocity and balance control parameters were observed for forward walking but not for tandem walking (Table 4.3). The AP_MOS showed a significant positive correlation (r = .751, p <.001) (Figure 4.1) and nSL_SD showed a weak negative correlation (ρ = -.292, p <.05) (Figure 4.2) with backward walking velocity.

Table 4.3: Correlation between backward walking velocity and balance control measures for forward and tandem walking.

	Forward walking		Tandem walking	
		Correlation coefficient		Correlation coefficient
nSV during backward walking	ML_MOS	0.204	ML_MOS	0.225
	ML_MOS_SD	-0.018	ML_MOS_SD	-0.091
	AP_MOS	.751*	AP_MOS	0.042
	AP_MOS_SD	0.051	AP_MOS_SD	-0.203
	SW_SD	0.11	SW_SD	0.04
	SL_SD	-.293**	SL_SD	-0.242

***Note: Significance was set at $p < .05$.** (ML_MOS = mediolateral margin of stability, ML_MOS_SD = mediolateral margin of stability variability, AP_MOS= anteroposterior margin of stability, AP_MOS_SD = anteroposterior margin of stability variability, SW_SD = step width variability, nSL_SD = step length variability). * indicates Pearson's correlation coefficient and ** indicates Spearman's correlation coefficient.

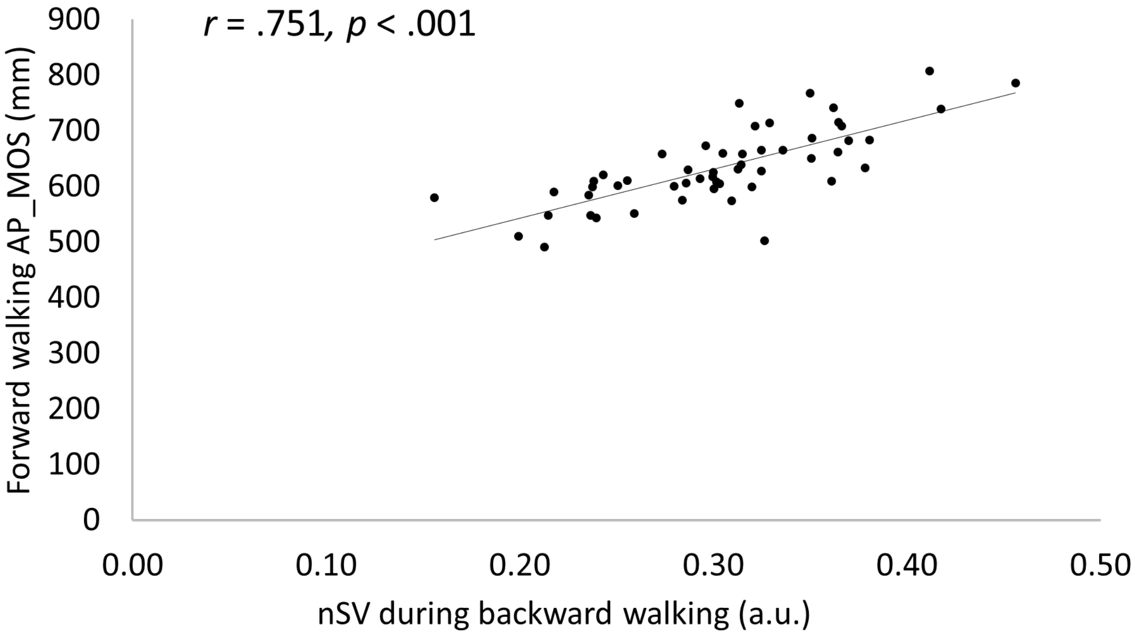

Figure 4.1: Scatterplot showing correlation data for AP_MOS during forward walking with normalized backward walking velocity.

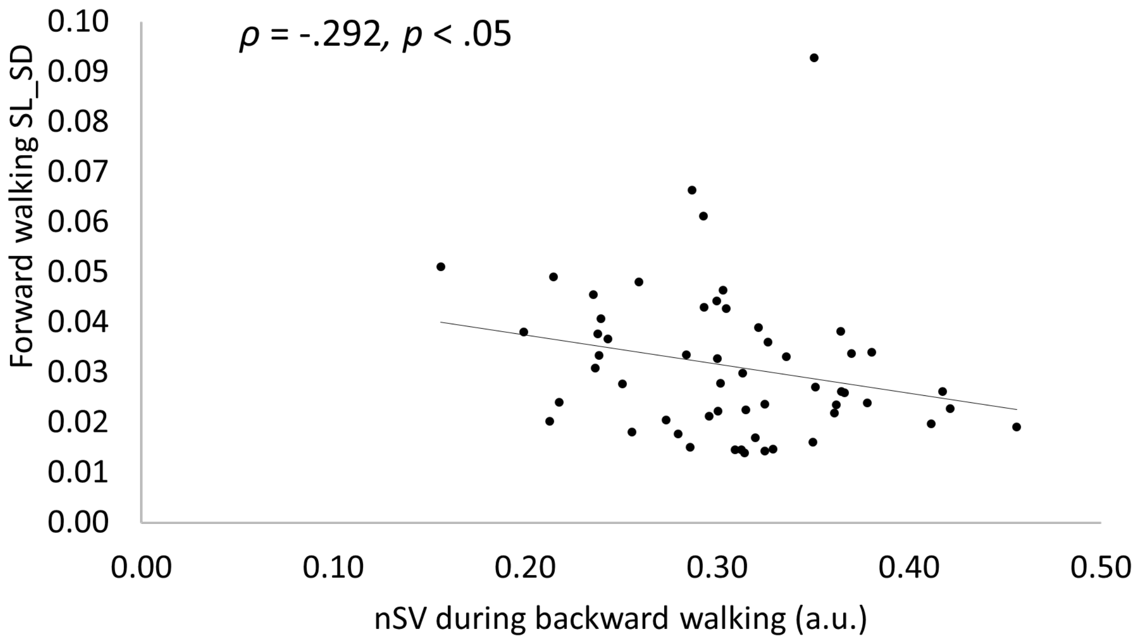

Figure 4.2: Scatterplot showing correlation data for step length variability during forward walking with normalized backward walking velocity.

4.5: Discussion

The purpose of this study was two-fold: The first purpose was to investigate differences in spatiotemporal and balance control parameters between forward and backward walking. The second purpose was to identify correlations between backward walking velocity and balance control measures not related to backward walking i.e. during forward and tandem walking. The overall results of this study demonstrate that backward walking is different from forward walking in terms of stepping and balance control, backward walking provides an increased challenge to the balance control system, and backward walking velocity is associated with balance control in the anteroposterior direction during forward walking.

Spatiotemporal measures of backward walking were similar to the stereotypical 'cautious gait' adopted by older adults and individuals with neurological deficits in response to a challenge to their balance control system [194]. Backward walking was characterized by a reduced velocity and step length, and an increase in step width compared to forward walking. The results in this study were similar to results from previous studies where individuals with neurological conditions and healthy adults reduced their walking velocity and stride length during backward walking [55, 140]. A reduction in step length and the consequent reduction in AP_MOS during backward walking might be attributed to fear of falling that may have arisen due to not knowing the end of the walkway, fear of running into obstacles, or stepping off the walkway. This fear during backward walking might have led participants to proactively walk with a shorter step length and subsequently a reduced AP_MOS compared to forward walking. A similar compensatory strategy was reported by Arora et al. [195] where people with iSCI walked forward with a shorter step and reduced velocity compared to healthy controls. In another study, Yang et al. [196] hypothesized that healthy adults reduced their step length and adopted a cautious gait strategy when walking forward on a treadmill compared to walking forward overground due to the stability challenges associated with treadmill walking. In the present study, participants adopted a similar strategy of reducing step length to keep the COM position closer to the posterior edge of BOS (as seen by a reduction in AP_MOS) which could be due to instability experienced during backward walking.

Based on the equation for MOS, an increase in step width leads to an increase in the ML_MOS and could therefore imply improved balance control. However, walking with wider steps is also a feature of 'cautious gait' where individuals increase their step width to widen the BOS to increase stability in the mediolateral direction. Similar to step length, fear of falling and perceived instability might have contributed to an increase in step width and ML_MOS during backward walking [197]. Another reason for increasing step width could be in response to an increase in xCOM movement. A study on stroke survivors found that people with stroke walked with an increased step width and a high xCOM movement leading to a similar ML_MOS compared to controls [37]. The xCOM excursion in people with stroke was significantly higher than controls [37]. These results imply that people with stroke increase their step width in response to an increase in xCOM movement to maintain a certain MOS value [37]. In this study, a similar strategy of increasing step width might have been in response to maintaining the preferred MOS value during backward walking. Even though xCOM movement was not examined in this study, future work should look at the motion and trajectory of the COM in the sagittal and frontal planes during backward walking and whether the COM movement follows the same pattern as forward walking.

Walking at a higher velocity is usually associated with a decrease in %DS [198]. Therefore, a decrease in the %DS along with a reduction in velocity during backward walking was an interesting finding. A likely reason for this finding is that participants executed the foot-off phase rapidly after foot strike. Another reason could be that participants spent more time in the single support phase in the process of exploring the area of the subsequent foot placement before executing each backward step.

Step and MOS parameters during backward walking were also significantly more variable compared to forward walking which could be due to the visual control and/or novelty of the task. Backward walking involves performing a task in the absence of a visual target, which can lead to an increase in step variability [107]. The increase in stepping variability was reflected in a significant increase in both anteroposterior and mediolateral MOS variability during backward walking. An increase in gait variability is suggestive of an increased risk of falls and can be indicative of impairment in balance control during walking [28]. Since the participants in the current study were healthy and free of any pathology, the increase in variability suggests that backward walking was a significantly more challenging task compared to forward walking. Even though backward walking is readily performed by individuals, it is a task that is not frequently undertaken in daily life. Because backward walking is not frequently performed, it is not as well-practiced and learnt. Due to the relative novelty of backward walking, the CNS could also have increased variability to better explore the region of stability. i.e. the balance control system is constantly gathering information about its current position and movement due to lack of experience in backward walking. The effect size observed for variability measures was small to moderate, suggesting that a learning effect could have occurred for backward walking over five trials, reducing the mean variability.

Significant correlations with backward walking velocity were observed only with balance control variables during forward walking. The absence of significant correlations with balance measures during tandem walking might be attributed to multiple reasons. First, the health status of the participants and the tandem walking protocol. Since all the participants were healthy adults, tandem walking did not significantly challenge the balance control system and therefore, the balance measures during tandem walking did not co-vary with backward walking velocity, leading to insignificant correlations. Second, differences in strategies during backward and tandem walking could also have led to insignificant correlations. Backward walking consists of stepping in a direction without anticipatory visual feedback with a change in direction whereas tandem walking consists of generating an accurate heel-to-toe foot placement and controlling the COM movement within a restricted BOS. The differences between backward and tandem walking may require separate motor control strategies - meaning that backward and tandem outcomes are unrelated to one another. Third, Pearson's and Spearman's correlation examines whether a linear relationship exists between two or more variables [199]. A lack of correlation between backward walking velocity and balance control measures during tandem walking also suggests that a non-linear relationship might exist between those outcome variable.

A negative correlation with nSL_SD suggests that participants with a higher backward walking velocity had more consistent step length values during forward walking. The correlation results are partially in line with previous literature where backward walking velocity was significantly correlated with performance on the Timed Up and Go (TUG) test, a clinical measure that is also related to fall risk as variability [58]. Together, these results provide further support in favour of exploring backward walking velocity to obtain an estimate of dynamic balance control. Further work can be directed towards using backward walking velocity to predict falls and identify prospective fallers.

A significant positive correlation of backward walking velocity with AP_MOS during forward walking indicates that participants who walked faster backwards also walked with a higher AP_MOS in the forward direction. These findings were in contrast to Hackney et al. [59] who found that people with PD walked at a similar speed during forward walking but at a slower speed during backward walking compared to controls. Additional work is needed to establish whether a similar correlation between backward and forward walking velocity exists for older adults and other balance compromised populations. The absence of significant correlations with balance control in the mediolateral directions should be investigated further since balance control in the mediolateral direction has shown to be a predictor of falls [200]. Apart from velocity, stride length, size of the BOS, total stance time, %DS, and step time variability during backward walking also successfully distinguish fallers from non-fallers [55]. In addition to velocity, correlation between balance measures and spatiotemporal parameters of backward walking also need further examination.

One limitation of this study was the health status of the participants. All were healthy (self-reported) with intact sensorimotor functioning. The nonsignificant correlations between backward walking velocity and with all balance control parameters during tandem walking as well as balance control in the mediolateral direction during forward walking indicate that the forward and tandem walking might not have imposed significant challenges to the balance control system. Another factor could also be the tandem walking protocol. Participants were asked to perform the tandem walking trials using their preferred arm orientation which could have affected the balance control measures [122]. Walking in tandem with arms across the chest (thereby reducing the moment of inertia of the body in the frontal plane and subsequently reducing the resistance for trunk sway) would have been a more challenging method along with a consistent level of difficulty for all participants. Another limitation of the study was the reporting on the sex of the participants. The reported sex of the participants was assumed by the researcher, and information about the sex and gender was not explicitly obtained from the participants. Future studies that report data about sex and gender should do so after obtaining information from the participants and describing how those data were collected.

Conclusion

This study compared spatiotemporal and balance control parameters between forward and backward walking and investigated the association between backward walking velocity and balance control variables during forward and tandem walking. Balance control during backward walking is maintained by adopting a cautious gait strategy characterized by short and wide steps. The results also address the gap in the existing literature about balance control strategies used by healthy adults during backward walking. Backward walking challenges the balance control system in healthy adults and backward walking velocity is correlated with stability during forward walking in the anteroposterior direction. The instability created during backward walking could be used to challenge and assess balance control in older adults and individuals with compromised balance by clinicians and researchers.

Relevance of Study 2 to the thesis

Study 2 aimed to identify differences in spatiotemporal and balance control measures between forward and backward walking. A second purpose of the study was to examine the correlation between backward walking velocity and balance control measures during forward and tandem walking. Backward walking was significantly different in terms of spatiotemporal parameters and balance control strategies compared to forward walking. Backward walking velocity was also indicative of balance control in the anteroposterior direction during forward walking. Study 2 showed that healthy adults are unstable and step more variably when walking backwards. Backward walking was challenging to the balance control system as evidenced by an increased variability compared to forward walking. The results provide further evidence to support the use of backward walking as a measure of walking performance and balance control. In terms of clinical significance, backward walking may unmask balance impairments that are not otherwise revealed during forward walking.

Chapter five

Study 3: The effects of vision and added haptic input on spatiotemporal and balance control measures during backward walking.

Abstract

Vision plays a significant role in maintaining stability and modulating step placement during forward walking. Adding haptic input during forward walking has shown to improve stability as measured by trunk movement, margin of stability, and step variability. The purpose of this study was to examine the effects of vision and added haptic input on spatiotemporal and balance control measures during backward walking. Fifty-five healthy adults (37 females, age: M = 28.1, SD = 9.9 years), (mass: M = 71.4, SD = 14.8 kg), (height: M = 1.7, SD = 0.09 m) completed five backward walking trials each with eyes open and closed and with and without using the haptic anchors. The results demonstrated that participants walked slower, with a greater amount of time in the double support phase, shorter step length, and increased step width when walking backward with eyes closed compared to backward walking with eyes open. Backward walking was also more variable in terms of MOS and step measures when walking with eyes closed compared to backward walking with eyes open. MOS in the mediolateral direction was reduced when walking backwards with haptic anchors. A significant interaction was observed between haptic anchor use and vision for step length. Step length was shorter when walking with eyes closed irrespective of haptic anchor use. Step length was shorter when walking with haptic anchors compared to walking without haptic anchors in the eyes open condition. Whereas the effects of added haptic input during backward walking need further examination by providing a further challenge during backward walking, the instability and challenge caused by walking backward with eyes closed can be used to assess dynamic balance control in older and balance compromised populations as well as an exercise in gait training interventions.

5.1: Introduction

The sensorimotor control of walking is achieved through motor output based on sensory inputs from the visual, somatosensory, and vestibular systems [39]. One method to improve balance control is adding haptic input during walking. Haptic input is the sensory information obtained by cutaneous receptors of the hand and the proprioceptors of the upper extremity when touching an external object in the environment [115]. One such method of providing haptic input is via haptic anchors. Haptic anchors consist of weights ($\approx$125 grams) that are attached to a string and dragged by individuals during walking [117]. Haptic input via anchors is provided by a combination of skin mechanoreceptors that sense tension produced in the strings by the weights and arm proprioceptors that sense the position of the arms while dragging the anchors [117]. The information gathered from the skin mechanoreceptors and arm proprioceptors is then integrated into the central nervous system to improve balance control. Haptic input orients an individual with respect to the surface and with respect to the source of the haptic input thus improving balance control [114]. Haptic anchors have demonstrated a positive impact on balance control by reducing trunk velocity and trunk acceleration during walking [83, 121, 122].

Balance during walking is also controlled by sensory input from the visual system. Vision provides an egocentric and allocentric frame of reference with respect to the individual and their environment [106]. The effects of reduced or absent vision on walking is evidenced by individuals decreasing their walking velocity, shortening their step length, and increasing their step width [201]. Balance control is also negatively affected when walking with reduced or absent vision, seen as an increase in variability of spatiotemporal parameters and trunk movement [108, 201]. Information about the external environment provided through visual input helps maintain balance control during walking in a feedforward manner by modulating step placement, changing the travel path, and changing the toe clearance height when navigating obstacles [106].

Backward walking has recently gained popularity in gait and balance research [202]. Backward walking has been used successfully in gait training programs to improve dynamic balance control [60–63], and recent findings have demonstrated that backward walking is a better discriminator of balance performance compared to forward walking in healthy as well as in clinical populations [55–57, 59, 76]. Backward walking is more challenging since visual information of the environment in the direction of progression is limited. For example, when walking forward within a lab environment, vision provides information about the beginning and end of the walkway, allowing an individual to decelerate and eventually coming to a stop at the end of the walkway. Unlike forward walking, information about approaching obstacles is unavailable in an anticipatory manner during backward walking even when walking with eyes open. As mentioned above, the use of added haptic input has shown beneficial effects during forward walking by reducing gait variability and MOS [42, 53, 107, 121]. Backward walking was performed with haptic anchors to examine whether added haptic input could provide similar stabilizing effects as forward and tandem walking during backward walking.

The purpose of this study was to observe the effects of added haptic input and availability of vision on spatiotemporal and balance control parameters during backward walking and whether the effects of added haptic input vary with the availability of vision. Based on the findings from previous literature [48, 203], it was hypothesized that backward walking will be significantly more challenging when vision is absent, balance control parameters during backward walking will improve when using haptic anchors, and balance control parameters will significantly improve when using haptic anchors in the eyes closed condition [83, 121, 122].

5.2: Methods

5.2.1: Participants

Participants were recruited through word of mouth, placing posters across the university campus, and announcements on university portals. Participants were recruited after completing a questionnaire for inclusion and exclusion criteria. Participants that were between the ages of 18 to 55 years; not currently participating in a balance training program or activities; not living with any conditions that affect balance; not having any visual impairment that could not be corrected with eyewear; and not having reduced or lost sensation in their upper and lower extremities were recruited for the study. The study was approved by the university research ethics board (Bio 17-157). All participants provided informed consent before commencing the data collection session.

5.2.2: Protocol

Participants performed five trials each of backward walking with and without the haptic anchors, and with their eyes open and eyes closed for a total of twenty trials. The trials were performed at the participants' preferred speed on a rigid 10-metre walkway in random order in a lab environment. Participants were asked to stop walking by the researcher two steps before the end of the walkway or if the participants started to veer toward the walls of the lab. If the participants veered towards the walls midway through the trial, data were recollected by asking participants to complete another trial of the same condition.

The sections below in italics are the same as mentioned in study 1 (Chapter three)

5.2.3: Instrumentation

Kinematic data were obtained using a 3-D motion capture system (Vicon Nexus, Vicon Motion Systems, Centennial, CO) with a sampling frequency of 100 Hz. Sixty-three reflective markers were placed on anatomical landmarks of the body to generate a 12-segment full-body model (Figure 3.1). The full-body model was used to calculate the total body centre of mass (COM) based on anthropometric tables [179].

5.2.4: Data analysis

Raw marker data were filtered using a 4^{th} *order Butterworth filter with a cutoff frequency of 8Hz. Filtering of data and calculation of outcome variables were performed using customized scripts (MATLAB R2019b for PC, MathWorks, Natick, MA). All outcome variables were calculated post the data filtering process. Spatiotemporal parameters included stride velocity, the relative amount of time in double support during a gait cycle (%DS), step length (SL), and step width (SW). Stride velocity was calculated by dividing stride length by stride time and was normalized using the formula: (normalized stride velocity (nSV) = stride velocity/*$\sqrt{H * g}$*) where H is the participant's leg length, and g is the acceleration due to gravity (9.81 m/s*^{2}*) [5]. The relative amount of time in double support was calculated as the time duration in percentage when both the feet were on the ground during one (100%) gait cycle. Step length was calculated as the anteroposterior distance between the right and left heel markers at each foot strike and normalized to participants' leg length (nSL) [5]. Leg length was calculated as the distance from the heel to the hip joint centre. Step width was calculated as the mediolateral distance between the right and left heel markers at each foot strike. Balance control was examined using the MOS and variability of nSL, SW, and MOS. MOS was calculated as suggested by Hof et al. [6] as the distance between the extrapolated centre of mass (xCOM) and the base of support (BOS). Instantaneous MOS values over each stride were averaged for each trial. The mean values across the trials were used for analysis. The MOS in the anteroposterior direction (AP_MOS) for forward and tandem walking was calculated over the duration of each stride as the distance between the xCOM and the heel of the trailing limb; the AP_MOS for backward walking was calculated over each stride as the distance between the xCOM and the toe of the trailing limb. The MOS in the mediolateral direction (ML_MOS) was calculated over the duration of each stride as distance between the xCOM and the closest lateral edge of the BOS. The standard deviation (SD) value was used to calculate the variability of the MOS (AP_MOS_SD) (ML_MOS_SD) and step (nSL_SD) (SW_SD) parameters. Values for all spatiotemporal and balance control measures were averaged across the five trials for each walking style.*

5.3: Statistical analysis

Data were examined for normality using the Shapiro Wilk test. Non-normal data were transformed using a two-step process [204]. Each outcome variable was examined using a within-subject 2 (no anchors/anchors) by 2 (eyes open/eyes closed) analysis of variance (ANOVA). A conservative alpha value of .01 was used to account for multiple comparisons. Significant interaction effects were analysed further using paired samples t-test with a Bonferroni correction. Effect sizes for RM-ANOVA and follow up t-tests are reported as partial eta squared (η_p^2) and Cohen's *d* values, respectively. Cohen's *d* was calculated using the formula (*d* =mean difference/standard deviation of mean difference). The effect sizes were interpreted as small ($\leq$0.2), medium (0.21-0.8), and large ($\geq$0.8) for Cohen's *d* and as small (0.01), medium (0.06), and large ($\geq$ 0.14) for partial eta squared [10, 199].

5.4: Results

A total of 55 healthy adults (37 females) with a mean age of 28.1 $\pm$ 9.8 years, a mean height of 1.71 $\pm$ 0.09 m, and a mean mass of 71.3 $\pm$ 14.8 kg participated in the study. Differences between males and females for age, height, and mass were not examined since this study did not aim to examine sex-based differences in walking behaviour. The ANOVA showed a significant main effect of vision for normalized stride velocity (F(1,54) = 168.121, p<.001, η_p^2 = .757), %DS (F(1,54) = 47.065, p<.001, η_p^2 = .466), ML_MOS_SD (F(1,54) = 16.072, p<.001, η_p^2 = .229), AP_MOS (F(1,54) = 165.393, p<.001, η_p^2 = .754), AP_MOS_SD (F(1,54) = 8.797, $p = .004$, η_p^2 = .140), SW (F(1,54) = 13.699, p< .001, η_p^2 = .202), and SW_SD (F(1,54) = 50.356, p< .001, η_p^2 = .483) (Table 5.1). Participants walked significantly slower, with more time in %DS, an increased SW, higher ML_MOS and lower AP_MOS while eyes closed compared to walking backward with eyes open. Variability of SW, AP_MOS, and ML_MOS was also higher when walking backward with eyes closed compared to walking with eyes open. A significant main effect of anchors was present for ML_MOS (F(1,54) = 22.632, p< .001, η_p^2 = .295). ML_MOS was significantly lower when walking backward with anchors compared to walking backward without anchors. A significant interaction effect between anchor use and vision was present for nSL (F(1,54) = 10.757, $p = .002$, η_p^2 = .166) (Figure 5.1). (Table 5.2: Marginal means (standard error) for outcome variables for each anchor use and vision conditions.). Four separate follow-up t-tests were performed for nSL for each anchor use and vision conditions with significance set at α = .002 (.01/4) (Table 5.3). nSL was significantly shorter when walking with eyes closed compared to eyes open for both the no anchor (p< .001, d = 1.623) and anchor (p< .001, d = 1.349) conditions. nSL was significantly shorter when using haptic anchors compared to walking without anchors for the eyes open condition (p< .001, d = .584). No significant differences were observed for anchor use for the eyes closed condition (p< .073, d = .247)

(Table 5.4). No significant effects were observed for nSL_SD.

Table 5.1: Main effects and interaction for haptic anchors and vision for outcome variables during backward walking.

	Main effect of anchors p value (effect size (η_p^2))	Main effect of vision p value (effect size (η_p^2))	Interaction effect between anchor use and vision p value (effect size (η_p^2))
nSV	.069 (.061)	**<.001 (.757)**	.088 (.053)
%DS	.090 (.052)	**<.001 (.466)**	.345 (.017)
ML_MOS	**<.001 (.295)**	.024 (.091)	.142 (.040)
ML_MOS_SD	.115 (.045)	**<.001 (.229)**	.526 (.007)
AP_MOS	.205 (.030)	**<.001 (.754)**	.378 (.014)
AP_MOS_SD	.068 (.060)	**.004 (.140)**	.025 (.090)
SW	.014 (.106)	**<.001 (.202)**	.073 (.058)
SW_SD	.224 (.027)	**<.001 (.483)**	.091 (.000)
nSL	.002 (.160)	**<.001 (.718)**	**.002 (.166)**
nSL_SD	.844 (.000)	.017 (.101)	.051 (.069)

***Note: Significance was set to *p*p < .01.** Bolded values indicate a significant result. Bolded values indicate significant difference between walking with and without anchors, significant difference between walking with eyes open and eyes closed, and a significant interaction between the haptic and vision conditions.

Table 5.2: Marginal means (standard error) for outcome variables for each anchor use and vision conditions.

	No anchors	Anchors	Eyes open	Eyes closed
nSV (a.u.)	0.28 (.008)	0.27 (.008)	0.30 (.008)	0.25[b] (.008)
%DS	24.07 (.584)	24.49 (.584)	22.40 (.633)	26.16[b] (.633)
ML_MOS (mm)	128.88 (2.445)	123.86[a] (2.445)	124.69 (2.495)	128.06 (2.495)
ML_MOS_SD (mm)	15.40 (.641)	14.26 (.641)	13.11 (.686)	16.55[b] (.686)
AP_MOS (mm)	500.38 (9.789)	495.09 (9.789)	520.58 (9.824)	474.89[b] (9.824)
AP_MOS_SD (mm)	39.91 (1.765)	35.77 (1.765)	34.26 (1.827)	41.42[b] (1.827)
SW (mm)	146.89 (5.014)	141.26 (5.014)	139.23 (5.062)	148.93[b] (5.062)
SW_SD (mm)	34.53 (1.075)	33.23 (1.075)	29.78 (1.101)	37.98[b] (1.101)
nSL_SD (a.u.)	0.07 (0.005)	0.07 (0.005)	0.06 (0.005)	0.07 (0.005)

***Note: Significance was set to $p < .01$.** a = significant difference between no anchors and anchor use; b = significant difference between eyes open and eyes closed.

Table 5.3: Follow up t-test values for each condition of anchor use and vision for step length.

Step length conditions	*t* statistic	*p* value	Effect size (Cohen's *d*)
No anchor-eyes open – No anchor-eyes closed	12.03	**<.001**	1.62
Anchor-eyes open – Anchor-eyes closed	10.00	**<.001**	1.35
No anchor-eyes open – Anchor-eyes open	4.33	**<.001**	0.58
No anchor-eyes closed – Anchor-eyes closed	1.83	0.073	0.25

***Note: Significance was set to $p < .002$.** Bolded values indicate statistically significant difference between conditions.

Table 5.4: Mean (standard error) values for follow-up t-tests.

Step length conditions	*t* statistic	*p* value	Effect size (Cohen's *d*)
No anchor-eyes open – No anchor-eyes closed	12.03	**<.001**	1.62
Anchor-eyes open – Anchor-eyes closed	10.00	**<.001**	1.35
No anchor-eyes open – Anchor-eyes open	4.33	**<.001**	0.58
No anchor-eyes closed – Anchor-eyes closed	1.83	0.073	0.25

***Note: Significance was set to $p < .01$.** a = significant difference between eyes open and eyes closed; b = significant difference between no anchors and anchor use.

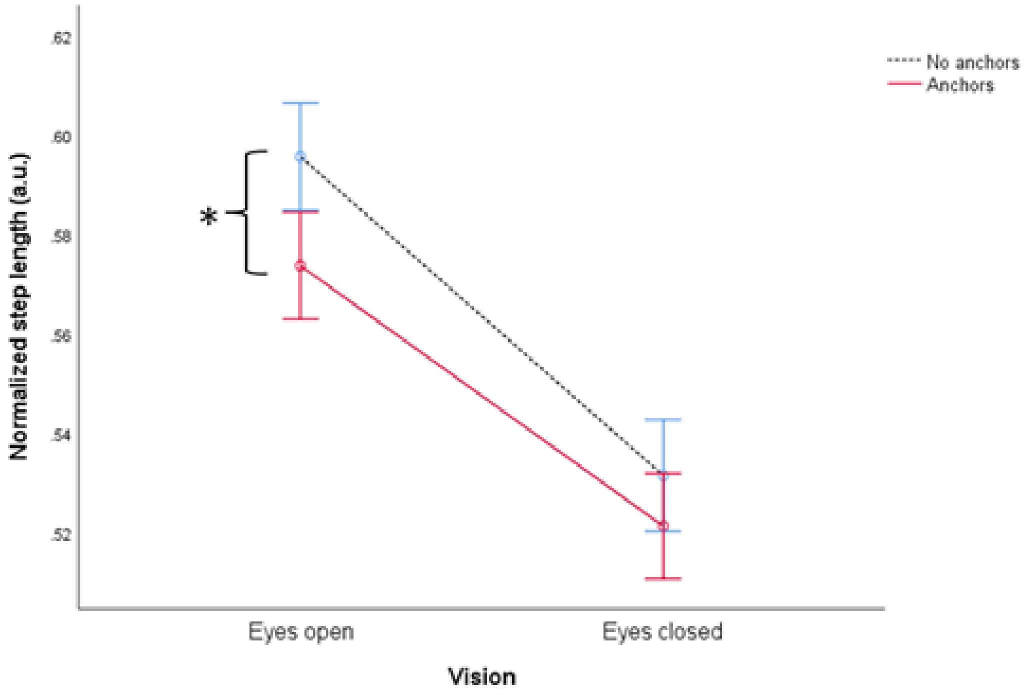

Figure 5.1: Interaction effects between haptic input and visual conditions for normalised step length during backward walking. Error bars indicate standard error. *Significant difference between anchor use and non-use in the eyes open condition (p <.002).

5.5: Discussion

The purpose of this study was to examine the effects of haptic anchors and vision on backward walking. The results suggest that both haptic anchors and vision significantly alter behaviour during backward walking similar to what has been reported for forward and tandem walking [83, 121, 122]. The results supported the proposed hypothesis of the effects of vision, but the hypothesis for the effects of using haptic anchors during backward walking was partially supported.

5.5.1: Significant main effects of vision

Compared to walking with eyes open, participants walked slower, with more time in the double support phase, increased step width, reduced step length, and a reduced AP_MOS when walking backward with eyes closed. The difference in parameters suggests that removal of vision challenged the balance control system during backward walking, and participants compensated by reducing their velocity, widening their base of support, and bringing their COM closer to their anterior edge (trailing limb) of the BOS. During forward walking, a similar strategy is used as a proactive measure to avoid using reactive strategies in case a perturbation is unexpectedly encountered [195] and to avoid slip-related loss of balance on a low-friction surface [205]. The shortened step length with a subsequent decrease in AP_MOS may have been achieved by a reduction in hip extension during backward walking [56].

Walking backward with eyes closed also caused a significant increase in variability during backward walking. The increase in variability was observed for step width, AP_MOS, and ML_MOS with small to medium effect sizes. An increase in step width variability along with an increase in ML_MOS variability suggests that the increase in step width variability led to an increase in MOS variability in the mediolateral direction. An increase in AP_MOS variability without a significant increase in SL variability suggests that an increase in COM movement variability led to the increase in MOS variability in the anteroposterior direction. Even though differences in variability between eyes open and eyes closed conditions were significant, the small to medium effect sizes suggest that factors other than vision such as fear of falling or colliding into the walls of the lab might have contributed to the increase in variability during backward walking [197].

5.5.2: Main effects of haptic anchors

A main effect of haptic anchors was observed only for the ML_MOS. The use of haptic anchors led to a significant reduction in ML_MOS during backward walking with a small effect size. Based on the theory of MOS as proposed by Hof et al. [6], a reduced MOS suggests an increased fall risk. As previously mentioned, an increase in step width resulting in an increase in ML_MOS is a compensatory strategy to account for instability caused due to high COM excursions [37]. A significant reduction in ML_MOS without a notable change in SW suggests that participants may have maintained their step width when walking with haptic anchors. Haptic anchors could have led to a change in postural strategy, causing participants to walk with an unchanged step width but greater movement of the trunk, leading to a COM position closer to the edge of the BOS and decreased ML_MOS. A similar result was observed in another study by Awdhan et al. [120] where participants reduced their ML_MOS when walking with haptic anchors compared to walking without the haptic anchors. Awdhan et al. [120] hypothesized that a reduction in ML_MOS without a change in step width when using haptic anchors could be due to a change in postural strategy when walking with the anchors or adopting a walking pattern that minimizes energy costs. Participants in the current study walked backwards in contrast to participants in Awdhan et al.'s [120] study where participants performed forward walking trials. The similarity in results; however, suggests that participants used similar strategies of walking with an unchanged step width that could have led to the COM position closer to the boundaries of the BOS when walking with haptic anchors for both backward and forward walking.

5.5.3: Interaction effects between anchor use and vision.

Participants reduced their step length when walking with their eyes closed compared to walking with eyes open for both the no anchor and anchor use conditions (Table 5.3). Vision modulates stepping during walking in a feedforward manner by utilizing information from the external environment to maintain balance [106]. The unavailability of visual input combined with a different walking style forced participants to modify their stepping behaviour in order to walk without falling. Participants reduced their step length in the eyes closed condition when walking with and without haptic anchors implying that augmentation of haptic input did not modulate step length during backward walking when visual input was absent. The presence of a large effect size for step length indicates that participants significantly relied on vision to control their step length during backward walking.

Follow-up t-tests comparing backward walking between haptic anchor use for each visual condition revealed that participants walked with a shorter step length when walking with the anchors compared to walking without anchors only during the eyes open condition with a medium effect size. Walking backward with and without the haptic anchors with eyes closed had no significant effect on step length. These results suggest that haptic anchors did not compensate for the absence of vision in modulating step length during backward walking and that participants relied more on vision than haptic input to control step length during backward walking.

These results suggest that when walking backwards using the haptic anchors during the eyes open condition, the postural strategy and postural orientation that was imposed upon the participants due to the task of dragging the haptic anchors could have led to participants walking with a reduced step length to keep the COM position closer to the anterior edge of the BOS to avoid and counter a fall in the backward direction.

Limitations

The study had certain limitations to consider. The participants were healthy individuals with intact sensorimotor functions. Therefore, the scope of generalizing the results to individuals with an impaired balance control system is limited. Even though instability was generated with the removal of vision, reliance on the added input from the haptic anchors was not required as the intact vestibular and somatosensory functions may have compensated for the lack of vision. Another limitation of the study was the reporting on the sex of the participants. The reported sex of the participants was assumed on the basis of gender expression, and information about the sex and gender was not obtained from the participants. Assumption of the sex of the participants does not provide a comprehensive picture about the differences in sex and gender-based analyses. Future studies that report data about sex and gender should do so after obtaining information from the participants.

Conclusion

In conclusion, the results of this study suggest that altering sensory input via removal of vision or augmentation of haptic input affects backward walking. Restriction of vision challenges balance control during backward walking, leading to increased variability and instability. The negative effects of haptic anchors on balance control in the mediolateral direction and step length were opposite to our expectations. Contrary to previous studies examining the effects of haptic input during forward walking, haptic anchors had no significant effect on the variability of margins of stability and stepping measures in backward walking [44, 53]. The role of added haptic input requires further examination in the context of backward walking since the hypotheses for the effects of haptic input were not supported.

Future work can examine the effects of haptic anchors by increasing the task difficulty by walking backwards on a compliant surface or by asking participants to walk backwards as fast as possible. The effects of haptic input during backward walking could be also examined in a population with impaired balance such as older adults and individuals living with neurological conditions that are more likely to utilize the additional sensory input provided by the haptic anchors to compensate for the decline or impairment in function of the visual, vestibular, and proprioceptive systems [43, 44].

Relevance of Study 3 to the thesis

Relevance of Study 3 to the thesis The purpose of study 3 was to investigate the role of vision and effects of added haptic input during backward walking. In study 3, we saw that backward walking was more variable with an increased variability of AP_MOS, ML_MOS, step length, and step width compared to forward walking. Based on results from the study in chapter four, it was hypothesized in the current study that removal of vision would further reduce balance control and provision of haptic input would improve balance control during backward walking. The hypotheses were partially supported whereby removal of vision during backward walking led to an increased variability of step and MOS parameters. Walking backward with added haptic input impacted balance control in the mediolateral direction as evidenced by a change in ML_MOS and a reduction in step length in the eyes open condition when walking with the haptic anchors.

Removal of vision during forward walking places a significant challenge on the balance control system that requires activation of appropriate postural strategies to walk without a fall [47, 201]. Performing tasks in the absence of vision is also a strategy used by therapists to improve balance control [78]. Combining results from studies 2 and 3 lends support to the idea that backward walking places a challenge to the balance control system and removal of vision further exacerbates that challenge. The instability encountered during backward walking and backward walking with eyes closed may be used to assess as well as improve dynamic balance control in neurological populations.

The purpose of providing additional haptic input is to supplement sensory information that can be used to improve dynamic balance control. Based on the theory of added haptic input and results from previous literature, balance control is improved, and variability of a given movement decreases when the task is performed with additional haptic input. Given the instability induced by backward walking itself and removal of vision during backward walking, haptic input should have improved balance control and reduced the variability of step and MOS measures during backward walking. In the current study, haptic input had an opposite impact on balance control in the mediolateral direction. An increase in step width during forward walking is a part of the 'cautious gait' strategy adopted to increase the size of the BOS. An increase in step width during backward walking was observed compared to forward walking (study 2) and in the current study when walking backward with eyes closed. When walking backwards with haptic anchors, participants did not increase their step width, suggesting that input from haptic anchors was not sufficient to induce a change in step width. Participants did not utilize haptic input and relied more on vision to regulate step width during backward walking. Margin of stability in the mediolateral direction was reduced when walking with the haptic anchors. Analysing of the movement of xCOM and step amplitude separately in future work could highlight the change in postural strategy adopted by individuals when walking backward using the haptic anchors.

Based on the results from studies 2 and 3, and results of backward walking training in previous literature [60, 61, 63], the intervention in study 4 consisted of performing backward walking trials with eyes closed. A second task in the same intervention consisted of walking in tandem with eyes closed. Tandem walking provides a mechanical challenge during walking by reducing the size of the BOS and compelling participants to walk with a narrow step width (heel to toe). It was hypothesized that practicing backward and tandem walking with eyes closed over six weeks would improve balance control measures during backward and tandem walking as well as forward walking. The hypotheses in study 4 were based on the concept that participants will adapt to these challenging walking styles through repeated practice and the adaptation effects will be transferred to forward walking.

Chapter six

Study 4: The effects of an intervention using haptic anchors on spatiotemporal and balance control measures in healthy adults.

Abstract

The purpose of this study was to examine the effects of an intervention using haptic anchors on balance control. Forty-five healthy adults were randomly allocated to one of three intervention groups such that each group consisted of fifteen participants. One group performed the intervention with the haptic anchors (wHA) (7 females, age: M = 28.1, SD = 9.9 years), (mass: M = 79.2, SD = 17.3 kg), (height: M = 1.7, SD = 0.1 m), another group performed the same intervention without the haptic anchors (nHA) (12 females, age: M = 29.5, SD = 11.3 years), (mass: M = 68.2, SD = 14.5 kg), (height: M = 1.7, SD = 0.1 m), and a control group (CTL) (11 females, age: M = 27.2, SD = 10.8 years), (mass: M = 69.6, SD = 13.1 kg), (height: M = 1.7, SD = 0.1 m) did not perform the intervention. The intervention tasks consisted of walking forward in tandem (heel to toe) and backwards (normally) with eyes closed for ten meters, three times/week, for six weeks. Spatiotemporal and balance control measures were calculated for forward, backward, and tandem walking with eyes open and eyes closed, and walking with and without the haptic anchors. Spatiotemporal measures included walking velocity, relative amount of time in the double support phase (%DS), step length, and step width. Balance control was measured using margin of stability (MOS) in the anteroposterior (AP_MOS) and mediolateral (ML_MOS) directions and variability of MOS and step measures. Variability was calculated as the standard deviation (SD) value. The assessment of all the measures was performed twice, before (pretest) and after (posttest) six weeks. Change scores for each outcome measure for each walking style were analyzed using a linear mixed effects model.

In addition, change scores were also compared to the MDC_{95} values from study 1. The results demonstrated that change scores for selected spatiotemporal measures were significantly higher in the eyes closed condition compared to the eyes open conditions for each walking style. AP_MOS showed a significant increase in the eyes closed condition compared to the eyes open condition in the nHA group. Comparison of change scores to the MDC_{95} values showed mixed results for each walking condition for forward, backward, and tandem walking. No significant effects on spatiotemporal or balance control measures were observed for the wHA group. A likely reason for the findings in this study is that haptic input

is utilized as natural sensory input to enhance balance control but is not sufficient to cause permanent changes in motor performance. The effects of long-term use of haptic anchors requires further examination before they can be recommended as a rehabilitation tool.

6.1: Introduction

Haptic anchors, developed by Mauerberg de Castro, are an accessible and inexpensive modality to improve balance control [117]. Haptic anchors consist of light weights (125 grams) attached to strings [117]. The use of haptic anchors requires individuals to hold the strings in each hand and drag the weights along the ground during walking [117]. Haptic input is provided by the cutaneous mechanoreceptors that sense the tension produced in the strings from the weights and the arm position in relation to the torso when holding the anchors [117]. The combined input of the arm configuration and cutaneous mechanoreceptors provide information on where an individual is in space with respect to the surface on which the individual is walking [117]. Studies that have examined the use of haptic anchors have demonstrated positive but acute effects on various balance control parameters during walking [42, 83, 120–122]. The beneficial effects of walking with the anchors on balance control during walking have been observed as a reduction of trunk movement velocity [121, 122] reduction of the variability of trunk acceleration [83], and as an increase in margins of stability [120]; however, these effects mentioned above recede with the removal of the anchors [83, 121, 122]. Understanding whether these beneficial effects can be retained with practice and whether these effects persist when walking without the haptic anchors is essential to determine whether haptic anchors can be recommended and used as a rehabilitation tool.

One study by Costa et al. [83] examined the effects of practicing tandem walking using the anchors and found that, after practicing tandem walking for nine trials, the effects of the anchors did not transfer when participants performed tandem walking trials without the anchors immediately after the practice trials. The authors hypothesized that the lack of transfer of effects of the anchors might be due to the practice sessions not being sufficiently challenging and an insufficient amount of practice trials.

Based on the known beneficial acute effects of haptic anchors and findings by Costa et al. [83], this study was designed to examine the effects of using haptic anchors over six weeks on predictive balance control during forward, backward, and tandem walking. Based on the suggestions proposed by Costa et al. [83], tandem walking was made more challenging by performing tandem walking with eyes closed. An additional task of backward walking with eyes closed was also included in the intervention to increase challenge. Balance training principles dictate that the balance control system should be sufficiently challenged to improve balance and to transfer the improved effects to other tasks [206]. It was hypothesized that participants who performed the intervention using the haptic anchors would demonstrate an improvement in balance control (as measured by MOS and variability measures) compared to the group that performed the intervention without using the haptic anchors and the control group that did not perform the intervention.

6.2: Methods

6.2.1: Participants

A total of forty-five healthy adults from the city of Saskatoon and the University of Saskatchewan participated in the study after providing informed consent. Participants were recruited after completing an inclusion/exclusion screening questionnaire where participants between the ages of 18 to 55 years, not currently involved in balance training programs or activities, not living with any conditions that affect balance, not having any visual impairment that could not be corrected with eyewear, and not having reduced or lost sensation in upper and lower extremities were recruited for the study. The study was approved by the university research ethics board (Bio 17-157).

6.2.2: Protocol

Participants completed two data collection sessions six weeks apart. During each data collection, participants performed five trials of forward, backward, and tandem (heel-toe) walking each with and without haptic anchors and with their eyes open and closed (60 trials total) at their preferred speed on a rigid 10-metre long platform in random order. Participants that were unable to walk heel to toe during tandem walking were asked to walk with an increased heel-toe distance (step length). During backward walking trials and trials performed with eyes closed, participants were asked to stop walking by the researcher as they approached the end of the walkway.

Following the first data collection session, participants were assigned to one of three intervention groups in a pseudorandom order. The allocation was based on a random list generated to allocate each participant to a specific intervention group. All participants were assigned randomly to the intervention groups except some participants that were interested in the study but were not able to commit to the intervention and so were placed in the control group. This pseudorandom allocation was done to avoid participants dropping out from the study.

Participants in one group performed the intervention with the haptic anchors (wHA), participants in the second group performed the same intervention without the haptic anchors (nHA), and the third group (CTL) did not perform the intervention.

6.2.3: Intervention

The intervention consisted of performing ten trials of tandem and backward walking each (20 trials total) with eyes closed over ten metres in a hallway, three times/week for six weeks. A researcher walked alongside the participant to provide support during instances of significant instability. Participants were asked to stop walking and open their eyes if they were unable to continue the trial. If the participants stopped or opened their eyes, they were asked to resume the trial from the same position where they stopped. Participants performed the trials in random order at their self-selected walking speed.

The intervention was designed based on a combination of balance training principles as stated above and results from a previous study by Costa et al. [83] who hypothesized that the beneficial effects of using the anchor system were not transferred partly because the number of practice trials were insufficient. The intervention was performed over the duration of six weeks with a total of 360 trials compared to nine trials in one session in Costa et al's study [83].

The duration of six weeks in this study was chosen based on the suggested value that was within the recommended range [139, 150]. The frequency (3 times/week) was chosen based on previous recommendations and since this frequency was most commonly reported in literature [139]. Also, the frequency of the training sessions (3 times/ week) was to ensure that the practice sessions were spaced out since divided practice leads to a better retention of a skill or a task compared to massed practice [207].

According to the theory of motor control, repetition of a task over time improves performance of the task as well as reduces variability when performing the task [207]. The intervention in this study was designed with the aim that practicing challenging walking tasks will improve performance on those tasks and reduce variability after six weeks. A second aspect of motor learning is transfer, i.e., ability to perform a learnt skill in a new task, context, or environment [207]. The design of the intervention was also based on the hypothesis that improvements in balance control gained by practicing tandem and backward walking with eyes closed with transfer to forward walking and walking with eyes open.

The sections below in italics are the same as mentioned in study 1 (Chapter three)

6.2.4: Instrumentation

Kinematic data were obtained using a 3-D motion capture system (Vicon Nexus, Vicon Motion Systems, Centennial, CO) with a sampling frequency of 100 Hz. Sixty-three reflective markers were placed on anatomical landmarks of the body to generate a 12-segment full-body model (Figure 3.1). The full-body model was used to calculate the total body centre of mass (COM) based on anthropometric tables [179].

6.2.5: Data analysis

Raw marker data were filtered using a 4^{th} order Butterworth filter with a cutoff frequency of 8Hz. Filtering of data and calculation of outcome variables were performed using customized scripts (MATLAB R2019b for PC, MathWorks, Natick, MA). All outcome *variables were calculated post the data filtering process. Spatiotemporal parameters included stride velocity, the relative amount of time in double support during a gait cycle (%DS), step length (SL), and step width (SW). Stride velocity was calculated by dividing stride length by stride time and was normalized using the formula: (normalized stride velocity (nSV) = stride velocity/*$\sqrt{H * g}$*) where H is the participant's leg length, and g is the acceleration due to gravity (9.81 m/s*^{2}*) [5]. The relative amount of time in double support was calculated as the time duration in percentage when both the feet were on the ground during one (100%) gait cycle. Step length was calculated as the anteroposterior distance between the right and left heel markers at each foot strike and normalized to participants' leg length (nSL) [5]. Leg length was calculated as the distance from the heel to the hip joint centre. Step width was calculated as the mediolateral distance between the right and left heel markers at each foot strike. Balance control was examined using the MOS and variability of nSL, SW, and MOS. MOS was calculated as suggested by Hof et al. [6] as the distance between the extrapolated centre of mass (xCOM) and the base of support (BOS). Instantaneous MOS values over each stride were averaged for each trial. The mean values across the trials were used for analysis. The MOS in the anteroposterior direction (AP_MOS) for forward and tandem walking was calculated over the duration of each stride as the distance between the xCOM and the heel of the trailing limb; the AP_MOS for backward walking was calculated over each stride as the distance between the xCOM and the toe of the trailing limb. The MOS in the mediolateral direction (ML_MOS) was calculated over the duration of each stride as distance between the xCOM and the closest lateral edge of the BOS. The standard deviation (SD) value was used to calculate the variability of the MOS (AP_MOS_SD) (ML_MOS_SD) and step (nSL_SD) (SW_SD) parameters. Values for all spatiotemporal and balance control measures were averaged across the five trials for each walking style.*

6.3: Statistical analysiss

Forward, backward, and tandem walking styles were analyzed individually. Change scores were obtained for each outcome variable for each walking style. Change scores were calculated using the formula: *Change score = (Posttest score - Pretest score)/Pretest score*. A positive change score indicates an increase in value during the posttest compared to pretest and a negative change score indicates a decrease in posttest value compared to pretest value. Change scores of the outcome variables were examined using linear mixed effects models with the intervention group, anchor use, and vision as fixed factors and participants as random factors. Significant main effects and interactions were further analyzed using simple effects and post-hoc tests.

A separate analysis was also conducted to identify whether changes in outcome variables for both the training groups were beyond the MDC_{95} values obtained in study 1. Change scores across each walking condition were compared against the MDC_{95} values from study 1. As mentioned in the methods section, walking conditions included walking without anchors with eyes open (NA-EO), without anchors with eyes closed (NA-EC), using the haptic anchors with eyes open (A-EO), and using the haptic anchors with eyes closed (A-EC).

6.4: Results

A total of 45 healthy adults (30 females) with a mean (standard deviation) age of 28.2 (10.5) years, mean mass of 72.3 (15.6) kg, and a mean height of 1.71 (9.4) metres participated in this study. Each intervention group consisted of fifteen participants. The mean values for age, mass, and height for each group are provided in Table 6.1. A one-way analysis of variance found no significant differences between the three groups for age (F (2,42) = 0.17, $p >. 05$), mass (F (2,42) = 1.3, $p > .05$), and height (F (2,42) = 2.37, $p > .05$).

Table 6.1: Means (SD) of demographic data for participants in each intervention group.

Group	Female:Male (number of participants)	Age (SD) (years)	Mass (SD) (kg)	Height (SD) (metres)
wHA	7:8	28.1 (9.9)	79.2 (17.3)	1.7 (0.1)
nHA	12:3	29.5 (11.2)	68.2 (14.5)	1.7 (0.08)
CTL	11:4	27.2 (10.8)	69.6 (13.1)	1.7 (0.07)

*Note: wHA: with haptic anchors, nHA: no haptic anchors, CTL: control, SD: standard deviation

Forward walking

Results from the linear mixed effects model

A significant main effect of vision was observed nSL_SD ((F (2, 42) = 1.76, $p < .01$) during forward walking. The change in nSL_SD was significantly higher in the eyes closed condition compared to the eyes open condition during forward walking (Figure 6.1.). No other significant effects or interactions were observed for any other outcome variables during forward walking.

Results comparing change scores to MDC_{95} values (Appendix C)

For the wHA group, step length variability showed a reduction beyond the MDC_{95} values in the NA-EO, NA-EC, and A-EO conditions and an increase beyond the MDC_{95} values for the A-EC condition. For the nHA group, the change in nSV showed an increase beyond the MDC_{95} value in the A-EO and A-EC conditions, and SL_SD showed an increase beyond the MDC_{95} values for all the walking conditions during forward walking.

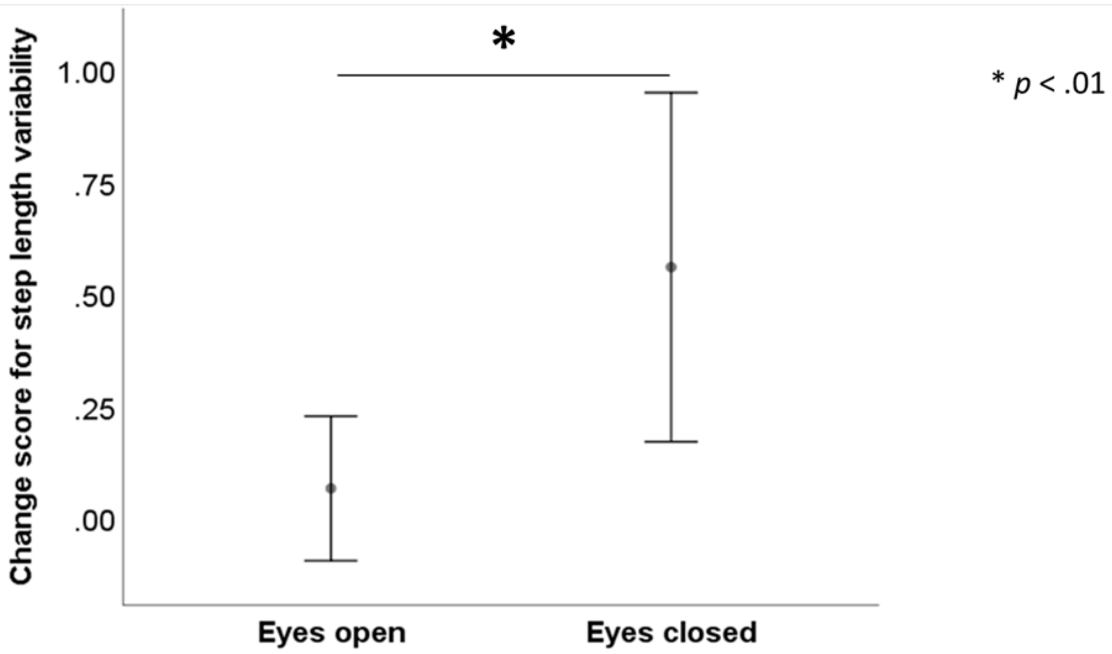

Figure 6.1: Mean of change scores for step length variability between eyes open and eyes closed during forward walking. Individual data points around the mean represent change score values for each participant. Error bars standard error.

Backward walking

Results from the linear mixed effects model.

The change in nSV showed a significant main effect of vision ($F(1, 124.50) = 31.53$, $p < 0.01$). significantly increased in the eyes closed conditions compared to the eyes open conditions (Figure 6.2). The change in %DS showed a significant main effect of vision ($F(1, 124.76) = 14.17$, $p < 0.01$) and was significantly reduced in the eyes closed conditions compared to the eyes open conditions (Figure 6.3). AP_MOS change showed a significant interaction between group and vision ($F(2, 124.31) = 8.10$, $p < 0.01$). Follow up analysis showed that the change in AP_MOS was significantly higher in the eyes closed condition compared to the eyes closed condition for the nHA group (Figure 6.4). The change in nSL showed a significant main effect of vision ($F(1, 126) = 26.28$, $p < 0.01$) and significantly increased in the eyes closed condition compared to the eyes open condition (Figure 6.5). No significant main effects or interactions were observed for ML_MOS, ML_MOS_SD, AP_MOS_SD, SW, SW_SD, and nSL_SD.

Results comparing change scores to MDC_{95} values (Appendix D)

For the wHA group, nSV showed an increase greater than the MDC_{95} values for the NA-EC and A-EC conditions, nSL showed an increase greater than the MDC_{95} values for the NA-EC and A-EC conditions, and nSL_SD showed an increase greater than the MDC_{95} values for the NA-EO, NA-EC and the A-EO conditions. nSL_SD showed a decrease beyond the MDC_{95} values for the A-EC condition.

For the nHA group, nSV showed an increase greater than the MDC_{95} values for all the walking conditions, nSL showed an increase greater than the MDC_{95} values for the NA-EC and A-EC conditions, and nSL_SD showed an increase greater than the MDC_{95} values for all walking conditions.

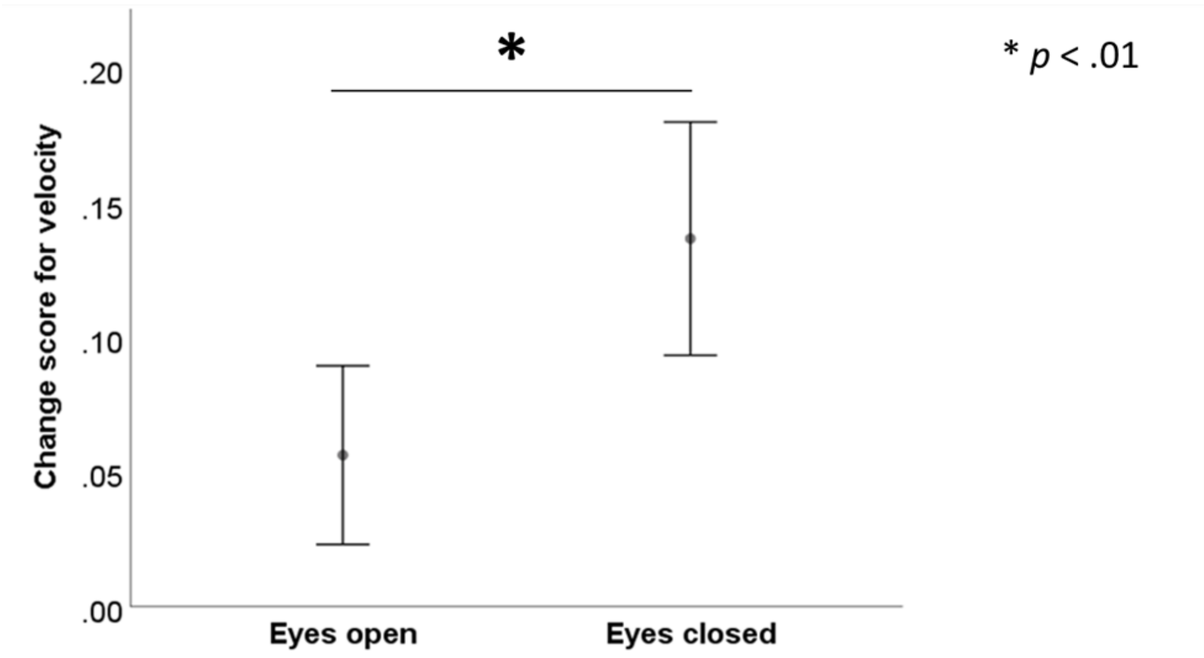

Figure 6.2: Mean of change scores for backward walking velocity between eyes open and eyes closed. Individual data points around the mean represent change score values for each participant. Error bars represent standard error.

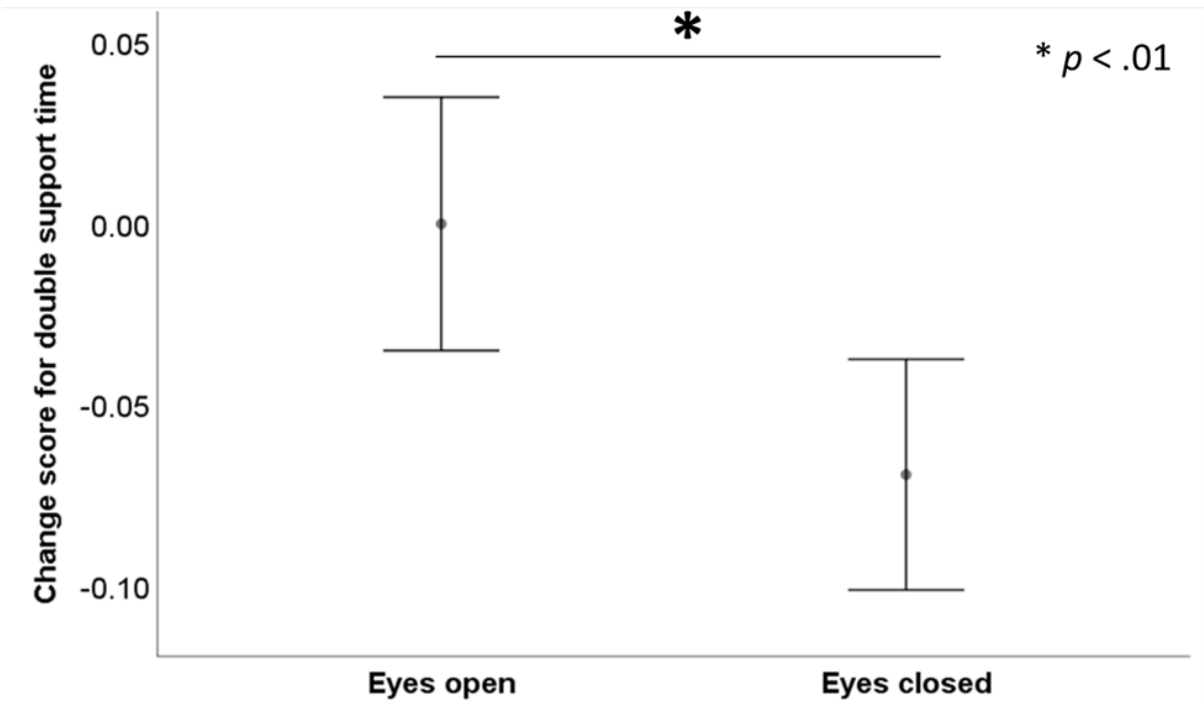

Figure 6.3: Mean of change scores for relative amount of time in the double support phase between eyes open and eyes closed during backward walking. Individual data points around the mean represent change score values for each participant. Error bars represent standard error.

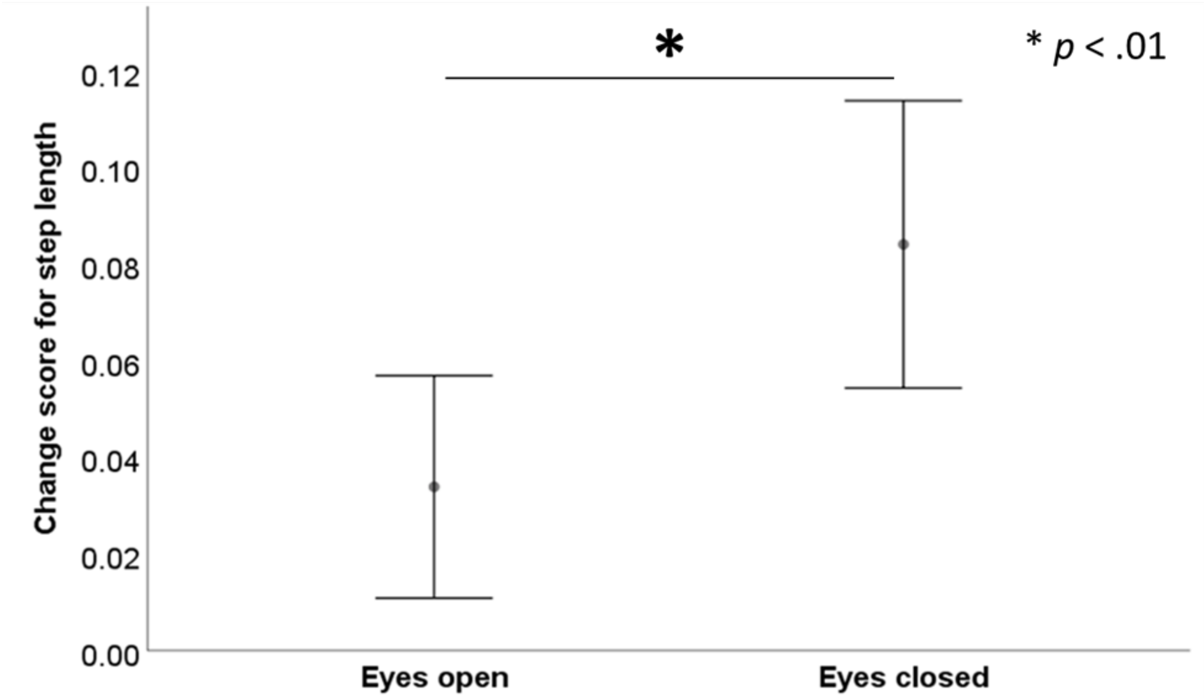

Figure 6.4: Mean of change scores for step length between eyes open and eyes closed during backward walking. Individual data points around the mean represent change score values for each participant. Error bars represent standard error.

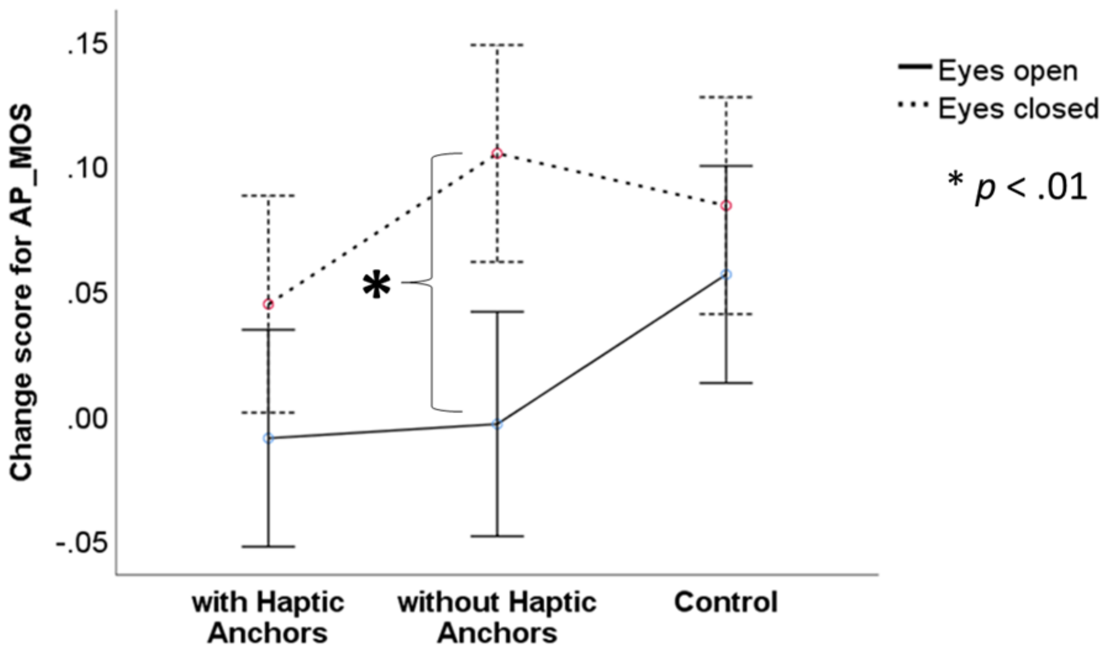

Figure 6.5: Mean of change scores for AP_MOS between groups and visual conditions during backward walking. Individual data points around the mean represent change score values for each participant. Error bars represent standard error.

Tandem walking

Results from the linear mixed effects model. The change in ML_MOS showed a significant main effect of vision ($F(1, 126) = 15.30$, $p < 0.01$) during tandem walking and was significantly reduced in the eyes closed condition compared to the eyes open condition. No significant main effects or interactions were observed for any other outcome variables during tandem walking.

Results comparing change scores to MDC_{95} values (Appendix E)

For the wHA group, (nSV) showed an increase greater than the MDC_{95} values for the NA-EC, A-EO and A-EC conditions. nSL_SD showed an increase greater than the MDC_{95} values for the NA-EC condition and a decrease greater than the MDC_{95} values for the NA-EO, A-EO, and A-EC conditions.

For the nHA group, nSV showed an increase greater than the MDC_{95} values for all the walking conditions. nSL_SD showed an increase greater than the MDC_{95} values for the NA-EO, NA-EC, and A-EC conditions.

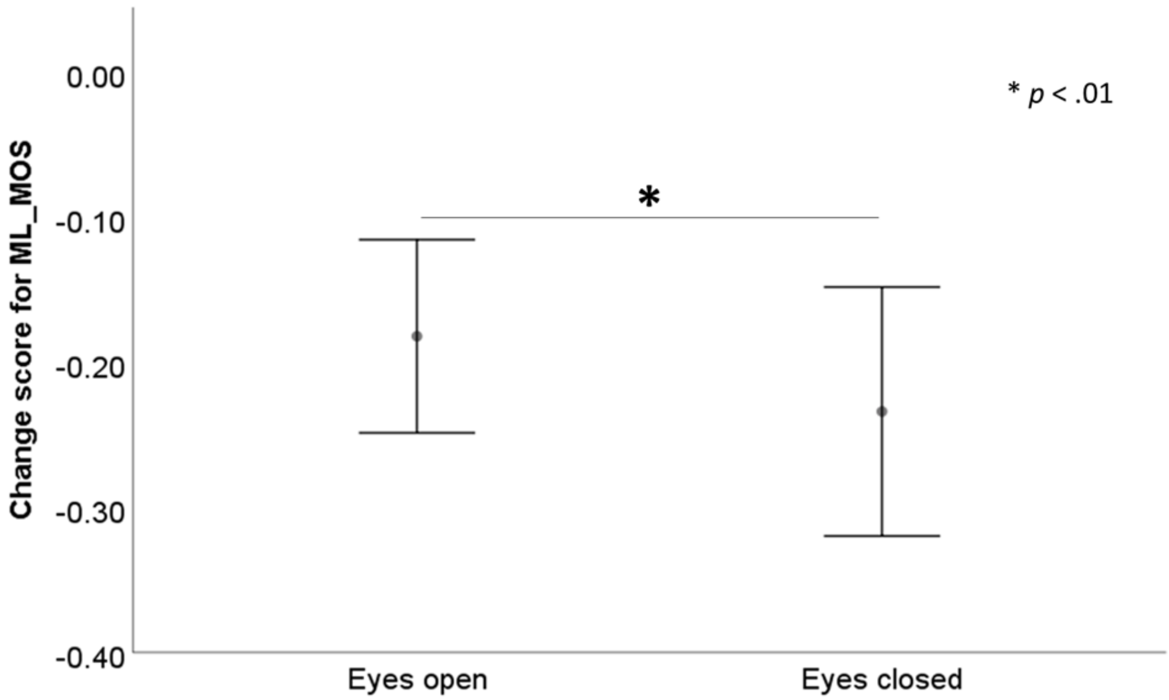

Figure 6.6: Mean of change scores for ML_MOS between groups and visual conditions during backward walking. Individual data points around the mean represent change score values for each participant. Error bars represent standard error.

6.5: Discussion

The purpose of this study was to examine the effects of a haptic intervention on spatiotemporal and balance control parameters in healthy adults. The results of the current study were contrary to our expectations where significant changes in spatiotemporal and balance control measures were not observed for the group that performed the intervention using the haptic anchors. A significant main effect of the wHA group or interaction effects between the wHA group and vision or wHA group and haptic anchor use were not observed. Change scores that were beyond the MDC_{95} values were observed for both the training groups (wHA and nHA). The overall results from the linear mixed effects analysis and comparison between change scores and MDC_{95} values indicate that the intervention was successful in inducing changes in walking behaviour but training with added haptic input had no additional effects on the outcome variables.

The participant dropout rate for the intervention study was low (ten participants dropped out of the fifty-five that performed the pretest session) and adherence to the intervention was high with participants completing an average of 17.13/18 (95%) training sessions in the wHA group and 16.26/18 (90%) training sessions in the nHA group. A probable reason for the low dropout and high adherence rate was the duration and adjustability of the intervention. The intervention sessions were conducted at a time and place convenient to the participants. Whereas the majority of the intervention sessions were conducted in the same hallway outside the test lab, intervention sessions for certain participants were conducted at a place of the participants' choosing including hallways outside their office, and open areas across different buildings on the university campus that allowed walking for a minimum distance of ten metres.

Forward walking

The change score for step length variability was significantly higher when walking with eyes closed compared to walking with eyes open. Comparing change scores for step length variability to MDC_{95} values showed mixed results with both an increase and decrease in step length variability in both training groups. An important factor that affects step length variability is walking speed. Step length variability is minimal at an individual's preferred walking speed and increases when walking speed is higher or lower than the preferred walking speed [208] [196]. Since change scores for velocity during forward walking did not reach significance, the change in step length variability was not likely due to a change in velocity.

Backward walking

Backward walking showed the most change in terms of the number of outcome variables with significant change values. There was a greater increase in velocity during the eyes closed condition compared to the eyes open condition. The amount of double support time usually decreases with an increase in velocity [198] as seen here. Change scores for

%DS paralleled the changes in velocity as they were significantly higher during the eyes closed condition compared to the eyes open condition. One way to increase or decrease walking velocity is modulating the size of step length. The change score for step length significantly increased in the eyes closed condition compared to eyes open condition during backward walking. The significant change in velocity during backward walking could have been due to an increase in step length when walking backward with eyes closed.

An increase in the AP_MOS change score was considered an improvement in performance in this study. The change in AP_MOS was significantly higher in the eyes closed condition compared to the eyes open condition in the nHA group. The nHA group performed backward walking trials with eyes closed during the intervention. A significant increase in AP_MOS when walking with eyes closed suggests a learning effect due to the intervention. The fact that individuals in the wHA group did not show further change in AP_MOS compared to nHA and CTL groups after performing the same intervention suggests that practicing with haptic anchors did not offer any additional advantage compared to training without the anchors. Results from study 3 also demonstrated that except nSL and ML_MOS, no other spatiotemporal or balance control measures were affected when walking with haptic anchors during backward walking.

Changes scores were beyond the MDC_{95} values for velocity, step length, and step length variability for both training groups. These results suggest that the individuals who performed the intervention irrespective of using the haptic anchors during the intervention showed a change in outcome variables.

Tandem walking

A significant reduction in the change score for ML_MOS during the eyes closed condition compared to eyes open condition without a significant reduction in change score for SW suggests that participants walked with a COM position closer to the edge of the BOS when walking with eyes closed.

Comparing the change scores to the MDC_{95} values showed that velocity increased post-intervention during tandem walking for both training groups except for the NA-EO condition in the wHA group. A possible reason for this exception is that participants in the wHA group walked at their peak velocity during both the pretest and posttest sessions leading to a minimal change between the two test sessions. Similar to backward walking, an increase in velocity during tandem walking for both training groups indicates a learning effect due to the intervention.

Haptic input obtained from the anchors is a form of somatosensory feedback where the combined information from the cutaneous and proprioceptive systems of the upper limbs is used as feedback to guide movement during walking [117]. Somatosensory feedback can be used as a substitute for deterioration in sensory systems or as an augmentation of sensory input(s). Even though provision of somatosensory feedback has shown immediate and long-term improvements in balance control [84, 160, 209], results from this study and another study by Lim et al. [161] are in contrast to previous literature. Lim et al. [161] found that a group who completed training with a combined audio, visual, and vibrotactile feedback for nine sessions did not show additional beneficial effects on balance control measures during standing and walking compared to a control group that completed the training without somatosensory feedback. In study 3, the effects of haptic input during backward walking were only evident on ML_MOS and nSL in the eyes open condition, both of which showed a reduction, contrary to the proposed hypotheses. Haptic anchors did not induce acute beneficial effects during backward walking and therefore, it seems that participants in the wHA groups did not use haptic input during backward walking throughout the intervention leading to insignificant changes in spatiotemporal and balance control measures after the intervention. The results from the current study can be attributed to several reasons:

Age and health status of the participants: The participants were healthy adults within the age range of 18- 55 years with more than 50% of participants below 25 years of age. Owing to the age group and health status of the participants, the spatiotemporal, balance control, and variability measures likely experienced ceiling effects. Due to (assumed) intact sensorimotor function, it is possible that participants did not utilize the added haptic input during the intervention. A similar intervention using haptic anchors improved balance control after training in individuals with vestibular pathology and the effects of the haptic intervention were retained after three months [165]. Unlike healthy individuals, participants with vestibular pathology utilized the added haptic input from the haptic anchors to compensate for the loss of input from the vestibular system, leading to an improvement and retention in balance control after the intervention [165].

The intervention: Balance training principles dictate that the difficulty of tasks within an intervention should progressively increase to observe improvements in balance control [78, 206]. The intervention in the current study consisted of performing the same tasks over a period of six weeks that could have led to early adaptations during the initial period of the intervention without further changes in walking behavior after six weeks. Analysis of spatiotemporal and balance control measures every two weeks might provide insights into whether early adaptations occurred with the intervention, whether these adaptations were beyond the measurement error, and whether the training needs to be progressively challenging over the duration of the intervention [210].

Provision of haptic input: Motor learning literature shows that though provision of feedback during practice is beneficial in learning a motor task, the frequency of feedback is vital in learning and retention of the task [207]. Continuous feedback during each trial or repetition of a motor task leads to an increased reliance on that feedback which hinders learning and long-term retention of the skill [207]. The increased reliance prohibits individuals from utilizing the strategies required to accomplish a task without the feedback. This ultimately leads to a reduced performance when performing the task without receiving feedback [207]. A study by Freitas et al. [211] on balance control during standing showed that older adults that used haptic anchors for 50% of the time during practice trials retained the benefits of haptic input at follow-up test sessions compared to individuals that did not use the anchors and individuals that used the anchors for all (100%) the practice trials. In the current study, participants in the wHA group were provided haptic input throughout the duration of the intervention for all the trials. A similar study of using haptic anchors with varying frequency during an intervention during walking is needed to further corroborate the findings of the current and Freitas et al's [211] studies. Insignificant findings can also be attributed to the statistical approach in analysing the data. The significance value was set at α ¡ 0.01 as opposed to the usual practice of α ¡ 0.05. The stringent alpha value was necessary due to the number of outcome variables and the amount of comparisons that were performed in the study and to avoid a Type 1 error.

Limitations

The age to participate in the study was set at 18-55 years to obtain a comprehensive idea of balance control. The age of more than fifty percent of the individuals that participated in the study was less than 25 years and hence, the results are not representative of healthy adults across different age groups. Sex of the participants was assumed by the researcher based on physical gender expression. It is important to obtain sex and gender information directly from the participants to identify sex and gender-based differences in balance control. A recent study by Coelho et al. [165] has shown that individuals with chronic dizziness who underwent balance training with and without haptic anchors both showed improvements in balance control but only the group who trained with the haptic anchors retained those effects three months after the intervention. Based on comparisons to the MDC_{95} values, both training groups showed improvements in certain spatiotemporal measures during forward, backward, and tandem walking. A similar follow-up test session as Coelho et al. [165] would demonstrate whether the changes in spatiotemporal measures are maintained for the group that completed the intervention using haptic anchors. Identifying the amount of practice sessions and the duration of each practice session it takes to observe a change in walking behavior can further assist in defining the time period between test sessions as well as establishing the duration of balance training programs.

Conclusion

A six-week balance control intervention using haptic anchors did not alter spatiotemporal and balance control parameters in healthy adults except AP_MOS in the group that practiced without the haptic anchors. Based on evidence from previous literature [83,165] and the current study on the effects of haptic anchors, further investigation into the long-term effects of haptic anchors are needed before they can be included in rehabilitation programs.

Relevance of study 4 to the thesis

The primary purpose of study four was to examine whether an intervention using haptic anchors has an effect on spatiotemporal and balance control measures during forward, backward, and tandem walking. Interventions that improve balance control during walking are necessary to prevent falls and morbidity associated with falls. Providing additional haptic input during walking is one way of targeting and improving dynamic balance control. Given the short-term improvements in balance control induced by haptic anchors during walking [42, 121, 122], investigations into the effects of long-term use of haptic anchors are warranted. Haptic anchors are an inexpensive and easy to use modality that can be used to provide somatosensory feedback to cause changes in walking behaviour and improve balance control [117].

Practice with haptic anchors did not cause any long-term effects during forward, backward, and tandem walking in this study. Continuous use of haptic anchors can lead to an increased reliance on haptic input [211] when corrective actions such as changing step length or step width are required during instances of instability during walking [15, 16]. Concerning the use of haptic anchors in clinical practice, further research is needed to establish whether long-term use of haptic anchors improve balance control and whether the improvement is retained when walking without the haptic anchors before recommendations can be made for use as a rehabilitation tool.

7:General discussion

Balance control is one of the modifiable factors that can be targeted during rehabilitation to avoid falls and promote independent ambulation [143]. Recognizing how individuals maintain their balance during walking and understanding the sensory contributions to balance control are necessary to inform balance training interventions. Balance control is a complex skill requiring activity from multiple bodily systems. This thesis focussed on aspects of biomechanics, sensory integration, and dynamic control of balance. Examining the combined roles of vision and haptic input provided through a novel modality is a small step in understanding the complex multifactorial aspect of balance during walking. Furthermore, performing exploratory studies with healthy participants lays the groundwork to perform similar studies in older adults and balance compromised individuals.

The integrity of balance control during walking is examined using a variety of measures [124]. Measures used to examine a construct need to be reliable and have minimal error to help identify differences between and within individuals [9, 180]. In study 1 (chapter 3), this thesis examined the reliability and measurement error for the spatiotemporal and balance measures used in the subsequent studies for forward, backward, and tandem walking. Whereas the spatiotemporal and MOS measures demonstrated moderate to excellent reliability for forward, backward, and tandem walking, variability measures demonstrated a poor to good reliability for each walking style. The results from study 1 provide evidence in favour of using MOS as a measure of balance control during different walking styles whereas obtaining and assessing variability measures require further attention and examination.

Study 2 (chapter 4) investigated the differences in spatiotemporal and balance measures between forward and backward walking as well as the association of backward walking velocity to biomechanical measures of balance during forward and tandem walking. Backward walking was significantly different compared to forward walking, both in terms of spatiotemporal and balance control parameters. Changes in step behaviour were reflected in MOS values where a 'cautious gait' strategy of walking with short and wide steps [125, 196, 203] was used by participants during backward walking. Backward walking was also significantly more variable in terms of step and MOS measures suggesting that it challenged the balance control of the participants. Backward walking velocity was positively correlated with AP_MOS and negatively correlated with nSL_SD during forward walking. The results from study 2 provide further support for using backward walking in addition to performance on forward and tandem walking as a measure of dynamic balance control. Study 3 (chapter 5) provided further evidence on the importance of visual input on balance control when performing any walking tasks. The findings from this study revealed that exteroceptive information from the visual system affects step placement during backward walking. Backward walking was significantly more variable, and participants walked significantly slower, and with shorter and wider steps when walking backward with eyes closed compared to walking with eyes open. The effects of added haptic input on backward walking were evident only for ML_MOS and nSL in the eyes open condition, both of which showed a reduction when walking with haptic anchors. A probable reason could be a change in postural strategy of walking with greater trunk movement when walking backward with the haptic anchors. The reduction in nSL might be due to a reduced arm swing when walking with the haptic anchors [212, 213]. A lack of interaction between vision and added haptic input for most measures suggests that input from haptic anchors was not sufficient to correct the instability caused by lack of vision during backward walking.

Study 4 (chapter 6) examined the effects of an intervention using haptic anchors for six weeks on balance control during forward, backward, and tandem walking. The results from the study showed that performing the intervention with added haptic input had no effect on any measures during forward, backward, and tandem walking except the AP_MOS during backward walking in the nHA group when data were analyzed using a linear mixed effects model. The changes in outcome variables between eyes open and eyes closed during forward, backward, and tandem walking could be due to a learning effect from the pretest to the posttest session. Results were different when outcomes were compared to the $(MDC_{95}$ values from study 1 where velocity, step length, as well as step length variability showed changes beyond the $(MDC\_95$ for forward, backward, and tandem walking for both the training groups.

While haptic anchors have shown to improve balance control measures in the past, those effects have been short-term [42, 83, 121, 122]. The findings from the previous studies on the short-term effects of haptic anchors and from this study hint at the possibility that input from haptic anchors is integrated by the CNS to improve balance control when an individual is using it, but practicing with added haptic input is not sufficient to cause a permanent change in walking behaviour.

7.1: Strengths

The combined results of all the four studies provide evidence on balance control strategies used by healthy adults during backward walking and the effects of an intervention with added haptic input on balance control during forward, backward, and tandem walking. One major strength of this thesis is the sample size in each study. Though a larger sample size is always beneficial to increase the external validity of the results, the current thesis had a sufficient sample size to infer conclusions at 80% power and a significance value of $\alpha <$ .05. To ensure a robust difference between groups and conditions for each walking style, statistical analyses for all the studies were performed using a conservative approach ($\alpha <$ .01) to accommodate multiple comparisons and avoid a Type 1 error in significance testing. This thesis examined reliability and error for balance measures across three different walking styles. Even though the reliability and error for spatiotemporal measures during forward walking has been studied in the past [168–171, 175], this thesis added information to the literature by examining the reliability of MOS measures during forward walking. In addition to forward walking, the reliability of spatiotemporal and balance during backward and tandem walking also lends support to use these measures to distinguish dynamic balance capacity between participants and to identify changes in balance control within participants.

Backward walking has been examined in the past in terms of joint kinematics, muscle activation, and differences in spatiotemporal measures between healthy and balance impaired populations [55–57, 59, 68, 70]. This thesis provided novel information on differences in balance control strategies between forward and backward walking as well as the sensorimotor control of backward walking in the absence of visual input and addition of haptic input. The results from studies 2 and 3 provide insight into how the MOS is controlled by changing step length and step width during backward walking. A change or reduction of visual input increases variability of movement and affects balance control during forward walking [207, 211]. Results from chapter five demonstrated that vision supported walking stability in backward walking similar to forward walking. Walking backward with eyes closed leads to a significant increase in movement variability and changes in step size to control the COM movement.

The findings in study 4 were contrary to the hypotheses but the results do provide certain aspects to consider when examining the long-term effects of added haptic input. Based on the theory of motor learning, intermittent provision of added haptic input could be more beneficial in improving balance control compared to continuous haptic feedback [207, 211]. An intervention that is progressively challenging may cause a significant change in balance control in healthy adults, compared to an intervention consisting of the same tasks performed for a certain time (six weeks for this thesis). Another approach in analyzing change in spatiotemporal and balance measures is by examining whether the change in values is beyond the minimal detectable change ((MDC_{95}) [210]. The (MDC_{95} values obtained from the results in study 1 can be used to assess whether changes in spatiotemporal and balance measures are due to the intervention or due to measurement error. Whereas change scores were compared to the (MDC_{95} for outcome variables in study 4, a similar comparison was not undertaken for results in studies 2 and 3. The SEM values obtained in studies 2 and 3 are from trials completed on the same day (within trial differences) for each walking condition whereas SEM and MDC_{95} values in study 1 were obtained from trials between two test sessions for each walking style.

Balance control was comprehensively evaluated in the anteroposterior and mediolateral directions using MOS and variability measures. MOS provides information on how the COM is controlled by step placement [214]. Variability of MOS and step measures demonstrate the ability of the motor system to consistently control the movement of COM and step size [27, 169]. However, based on results from study 1, variability measures should be interpreted with caution, especially when data are obtained from less than 50 strides collected by multiple passes over a short distance instead of continuous walking. The use of MOS for three different walking styles provides insight into how healthy adults maintain their balance control when walking with a restricted BOS in the forward direction (tandem walking) and when the walking direction is reversed (backward walking). The significant difference in MOS values between forward and backward walking demonstrates how healthy adults change their stepping strategy to maintain balance during backward walking.

7.2: Limitations

The study aimed to recruit healthy individuals between 18-55 years to obtain a representative sample across different age groups. A majority of the participants were below the age of 25 years and the sample was therefore not representative of the entire 18-55 years age range. To obtain normative values for reliability and balance constructs in a healthy population, it is necessary to include participants from across various age groups. This limitation in the current study could be addressed by targeted recruitment of participants that encompass the complete range of 18-55 years. A majority of the participants in the studies were students from the university. This could be because the information about the studies were posted on university portals and advertisement boards, and in-person announcements during course lectures. A sample more diverse in age could have been obtained if recruitment strategies included announcements and postings outside the university campus within the community.

An adequate amount of muscle activity and neuromuscular function is necessary for balance control [2]. Analysis of muscle activity during walking provides information regarding the timing and intensity of specific muscles involved during the stance and swing phases of the gait cycle. Information about muscle activity was not obtained in the studies that make up this thesis. Inclusion of electromyographic measures in addition to biomechanical measures provides a more comprehensive analysis of the efficiency of the balance control system. The increase or decrease in lower limb muscle activity in response to removal of vision, addition of haptic input, and across different walking styles could provide information about how healthy adults modulate step placement to maintain balance control during walking [44, 53].

Recent research has demonstrated that position of the arms can affect balance control measures during walking [121, 122]. The positions of the upper limbs when walking without the anchors during the test sessions and intervention was not normalized across participants. Participants were asked to perform each walking style with their preferred arm position and a fixed arm orientation was not imposed across all participants. Removing the potential stabilizing effects of the arms and to make the task equally challenging for all participants, walking trials should be performed with arms positioned across the chest. Analyzing the position and movement of the arms during each walking style could also provide a picture on the strategies used by participants to maintain balance during walking with eyes closed. Tandem walking was performed on a flat rigid surface where participants might have increased their step width to compensate for the instability to continue the tandem walking trial. Walking on a narrow beam where step width is restricted would be a more challenging task to accomplish compared to walking on a walkway where participants can change their step width to compensate for instability. Walking on a beam can also be more challenging since walking on a surface which is elevated from the floor can induce a fear of falling [41]. Also, failure to control the COM movement within the position of the feet and inability to execute steps within the constraints of the beam can cause a "fall" off the beam [215]. In this study, beam walking would have been a better task at challenging the balance control in healthy adults compared to tandem walking.

Retention tests are administered 24 hours or later after the practice sessions have ended to identify whether performance on a task or a skill persists after not practicing the task or skill [207]. Retention is said to have occurred if the performance on the task does not deteriorate or reduce compared to the immediate post practice performance [207]. The possible reasons for improved retention include improvement in neural connectivity, and consolidation of new neural connections during the interval between the post practice and retention tests [207]. Retention tests in study 4 were not planned a-priori. A follow up test session allowing time for neural consolidation could have indicated whether individuals that practiced with the haptic anchors retained the effects of training more than the group that practiced without the anchors and the control group.

7.3: Future work

The current thesis included exploratory studies aimed at identifying the reliability of spatiotemporal and balance control measures during walking, identifying balance control strategies between forward and backward walking, association of backward walking velocity with biomechanical measures of balance during tasks other than backward walking, combined effects of vision and haptic input using haptic anchors on balance control during walking, and the effects of an intervention using haptic anchors on balance control during walking. Future work should focus on practical applications from the findings of these studies. Results from study one can be used to identify changes in balance control over repeated test sessions and in response to therapy using MOS measures during forward, backward, and tandem walking. Studies two and three provided information on the predictive ability of backward walking velocity and sensorimotor control of balance during backward walking respectively. The results from studies two and three provide further support for using backward walking as a task to assess dynamic balance control and as an exercise during gait rehabilitation. Results from study four did not show beneficial effects of an intervention with added haptic input on balance control. Group rehabilitation programs and individual client-based rehabilitation consist of a combination of exercises, treatment of vision, and modifications in the external environment to prevent falls [216]. Novel methods to make rehabilitation more accessible and equitable are required to include a majority of the population that does not have access to hospital or clinic-based rehabilitation. The addition of haptic input via haptic anchors provides one such novel and cost-effective option of rehabilitating individuals with balance deficits. The effects of an intervention using haptic input also needs investigation in older age groups and individuals with reduced balance and increased fall risk.

7.4:Conclusion

In summary, this thesis adds knowledge about the reliability of balance control measures and balance control during walking with a focus on backward walking. Spatiotemporal and MOS measures show a high degree of reliability over time for forward, backward, and tandem walking. Backward walking has the potential to identify underlying balance deficits that are otherwise not highlighted with forward walking alone. The immediate effects of haptic anchors were observed for mediolateral balance control and step length during backward walking but the effects of long-term use of haptic anchors are yet to be ascertained.

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Appendix A: Anatomical sites for marker placement to generate a 3-D full body model.

Anatomical site	**Marker placement**
Head (3 markers)	Front of the forehead, temple above the left ear, temple above the right ear
Shoulder (2 markers)	Right and left acromioclavicular joint
Sternum (1 marker)	Sternal notch
Back (3 markers)	3 markers placed non-collinearly on the upper back
Thorax (4 markers form a circle from front to back)	Front – just below the xiphoid process of the sternum Right – in line with the front marker on the right side of the thoracic cage Back – in line with the right marker on the back Left – in line with the front marker on the left side of the thoracic cage
Pelvis (7 markers)	Pelvis cluster consisting of 4 markers placed non-collinearly on a rectangular plate which is secured using a belt such that the plate rests on the posterior part of the pelvis Front – just below the navel Right – in line with the front marker in the center of the right iliac crest Left - in line with the front marker in the center of the left iliac crest
Thigh (8 markers)	Marker cluster consisting of 4 non-collinearly placed markers on the lateral side on each of the right and left thigh
Knee (4 markers)	Medial and lateral epicondyles of right and left knee joint
Leg (8 markers)	Marker cluster consisting of 4 non-collinearly placed markers on the lateral side of the right and left leg each
Ankle (4 markers)	Medial and lateral malleolus of the right and left ankle joint
Foot (12 markers)	On the most anterior point of the right and left foot On the head of the first metatarsal joint of the right and left foot On the most lateral aspect of the right and left foot On the calcaneus of the right and left foot 3 markers placed non-collinearly on the lateral surface of the right and left foot
Finger (1 marker)	On the dorsal surface of the distal phalanx of the index finger of the dominant hand

Appendix B: Comparison of change scores for each outcome variable to the MDC_{95} values obtained from study 1 for forward walking.

			Group		Change	
	Condition	MDC_{95}	wHA (a.u.)	nHA (a.u.)	wHA (a.u.)	nHA (a.u.)
nSV		0.04				
	NA-EO		0.00	0.02	NS	NS
	NA-EC		-0.02	0.02	NS	NS
	A-EO		-0.01	0.05	NS	Sig
	A-EC		-0.02	0.04	NS	Sig
%DS		3.47				
	NA-EO		-0.03	-0.01	NS	NS
	NA-EC		0.07	-0.01	NS	NS
	A-EO		0.03	-0.02	NS	NS
	A-EC		0.10	0.00	NS	NS
ML_MOS		14.77				
	NA-EO		-0.02	0.02	NS	NS
	NA-EC		-0.04	0.05	NS	NS
	A-EO		-0.02	0.04	NS	NS
	A-EC		-0.02	-0.02	NS	NS
ML_MOS_SD		7.59				
	NA-EO		0.29	0.20	NS	NS
	NA-EC		0.13	0.58	NS	NS
	A-EO		0.33	-0.09	NS	NS
	A-EC		0.28	0.29	NS	NS
AP_MOS		63.12				
	NA-EO		-0.01	0.00	NS	NS
	NA-EC		0.01	0.01	NS	NS
	A-EO		0.00	0.02	NS	NS
	A-EC		0.00	0.02	NS	NS
AP_MOS_SD		28.01				
	NA-EO		0.22	0.14	NS	NS
	NA-EC		-0.10	0.29	NS	NS

	A-EO		0.36	-0.07	NS	NS
	A-EC		0.46	0.09	NS	NS
SW		32.26				
	NA-EO		-0.08	0.03	NS	NS
	NA-EC		-0.07	0.10	NS	NS
	A-EO		-0.04	0.04	NS	NS
	A-EC		-0.04	0.05	NS	NS
SW_SD		16.07				
	NA-EO		0.02	0.04	NS	NS
	NA-EC		0.15	0.26	NS	NS
	A-EO		0.26	-0.01	NS	NS
	A-EC		0.07	0.20	NS	NS
nSL		0.05				
	NA-EO		-0.01	0.01	NS	NS
	NA-EC		-0.01	0.00	NS	NS
	A-EO		-0.01	0.02	NS	NS
	A-EC		-0.01	0.00	NS	NS
nSL_SD		0.03				
	NA-EO		-0.08	0.52	Sig	Sig
	NA-EC		-0.05	1.12	Sig	Sig
	A-EO		-0.08	-0.15	Sig	Sig
	A-EC		0.23	0.97	Sig	Sig

*Note: NS (Non significant) - Change value within the range of MDC95 values, Sig (Significant) - Change beyond the range of MDC_{95} values. a.u.: arbitrary units

Appendix C: Comparison of change scores for each outcome variable to the MDC_{95} values obtained from study 1 for backward walking.

			Group		Change	
	Condition	MDC_{95}	wHA (a.u.)	nHA (a.u.)	wHA (a.u.)	nHA (a.u.)
Stride velocity		0.04				
	NA-EO		0.03	0.06	NS	Sig
	NA-EC		0.11	0.16	Sig	Sig
	A-EO		0.01	0.07	NS	Sig
	A-EC		0.11	0.19	Sig	Sig
%DS		4.45				
	NA-EO		-0.02	-0.01	NS	NS
	NA-EC		-0.05	-0.08	NS	NS
	A-EO		0.05	-0.03	NS	NS
	A-EC		-0.09	-0.07	NS	NS
ML_MOS		20.61				
	NA-EO		0.01	0.09	NS	NS
	NA-EC		-0.01	0.13	NS	NS
	A-EO		0.06	0.08	NS	NS
	A-EC		0.05	0.11	NS	NS
ML_MOS_SD		9.49				
	NA-EO		0.05	0.41	NS	NS
	NA-EC		0.40	-0.11	NS	NS
	A-EO		0.16	0.35	NS	NS
	A-EC		0.13	0.05	NS	NS
AP_MOS		66.02				
	NA-EO		0.00	0.00	NS	NS
	NA-EC		0.05	0.10	NS	NS
	A-EO		-0.02	-0.01	NS	NS
	A-EC		0.04	0.11	NS	NS
AP_MOS_SD		40.70				
	NA-EO		-0.14	0.22	NS	NS
	NA-EC		0.03	-0.39	NS	NS

	A-EO		-0.06	0.19	NS	NS
	A-EC		-0.03	-0.25	NS	NS
SW		51.47				
	NA-EO		0.07	0.10	NS	NS
	NA-EC		0.06	0.19	NS	NS
	A-EO		0.13	0.19	NS	NS
	A-EC		0.15	0.27	NS	NS
SW_SD		14.16				
	NA-EO		0.06	0.02	NS	NS
	NA-EC		0.14	-0.05	NS	NS
	A-EO		0.04	0.20	NS	NS
	A-EC		0.12	0.03	NS	NS
nSL		0.06				
	NA-EO		0.03	0.01	NS	NS
	NA-EC		0.08	0.08	Sig	Sig
	A-EO		0.03	0.03	NS	NS
	A-EC		0.07	0.09	Sig	Sig
nSL_SD		0.06				
	NA-EO		0.17	0.58	Sig	Sig
	NA-EC		0.21	0.06	Sig	Sig
	A-EO		0.32	0.58	Sig	Sig
	A-EC		-0.08	0.13	Sig	Sig

*Note: NS (Non significant) - Change value within the range of MDC95 values, Sig (Significant) - Change beyond the range of MDC_{95} values. a.u.: arbitrary units

Appendix D: Comparison of change scores for each outcome variable to the MDC_{95} values obtained from study 1 for tandem walking.

			Group		Change	
	Condition	MDC_{95}	wHA (a.u.)	nHA (a.u.)	wHA (a.u.)	nHA (a.u.)
Stride velocity		0.04				
	NA-EO		0.02	0.07	NS	Sig
	NA-EC		0.07	0.13	Sig	Sig
	A-EO		0.12	0.07	Sig	Sig
	A-EC		0.07	0.15	Sig	Sig
%DS		7.18				
	NA-EO		-0.02	-0.04	NS	NS
	NA-EC		-0.03	-0.07	NS	NS
	A-EO		-0.03	-0.02	NS	NS
	A-EC		-0.04	-0.08	NS	NS
ML_MOS		7.69				
	NA-EO		-0.20	-0.31	NS	NS
	NA-EC		-0.33	-0.34	NS	NS
	A-EO		-0.25	-0.29	NS	NS
	A-EC		-0.32	-0.37	NS	NS
ML_MOS_SD		6.23				
	NA-EO		0.14	0.24	NS	NS
	NA-EC		0.22	0.13	NS	NS
	A-EO		0.27	0.33	NS	NS
	A-EC		0.04	0.16	NS	NS
AP_MOS		52.90				
	NA-EO		0.01	0.01	NS	NS
	NA-EC		0.05	0.05	NS	NS
	A-EO		0.07	0.03	NS	NS
	A-EC		0.05	0.06	NS	NS
AP_MOS_SD		30.85				
	NA-EO		-0.13	-0.03	NS	NS
	NA-EC		-0.09	-0.23	NS	NS

	A-EO		0.01	0.27	NS	NS
	A-EC		0.03	-0.02	NS	NS
SW		26.70				
	NA-EO		0.13	0.63	NS	NS
	NA-EC		0.20	0.34	NS	NS
	A-EO		0.47	0.91	NS	NS
	A-EC		0.41	0.58	NS	NS
SW_SD		11.12				
	NA-EO		0.19	0.07	NS	NS
	NA-EC		0.09	-0.07	NS	NS
	A-EO		0.26	0.17	NS	NS
	A-EC		0.30	-0.02	NS	NS
nSL		0.06				
	NA-EO		-0.03	-0.01	NS	NS
	NA-EC		-6.804e -4	0.01	NS	NS
	A-EO		-0.01	-0.01	NS	NS
	A-EC		-0.04	0.02	NS	NS
nSL_SD		0.05				
	NA-EO		-0.07	0.36	Sig	Sig
	NA-EC		0.85	1.21	Sig	Sig
	A-EO		-0.10	0.02	Sig	NS
	A-EC		-0.08	1.38	Sig	Sig

*Note: NS (Non significant) - Change value within the range of MDC95 values, Sig (Significant) - Change beyond the range of MDC_{95} values. a.u.: arbitrary units

Appendix E: Ethics approval certifications

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB)

Certificate of Approval

PRINCIPAL INVESTIGATOR	DEPARTMENT	Bio #
Alison Oates	Kinesiology	17-157

INSTITUTION(S) WHERE RESEARCH WILL BE CARRIED OUT

Biomechanics of Balance and Movement Laboratory
University of Saskatchewan

College of Kinesiology
University of Saskatchewan
Saskatoon SK S7N 5B2

FUNDER(S)

UNIVERSITY OF SASKATCHEWAN

TITLE

Protocol: Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Difficult Walking Tasks in Young Healthy Adults? A Pilot Study

ORIGINAL REVIEW DATE	APPROVED ON	APPROVAL OF	EXPIRY DATE
16-Jun-2017	12-Jul-2017	Notice of Ethical Review Responses, rec'd 07-July-2017 Revised Application for Biomedical Research Ethics Review, rec'd 07-July-2017 Recruitment Flyer v.2, rec'd 07-July-2017 Master List, rec'd 07-July-2017 PAWS Announcement v.2, rec'd 07-July-2017 Research Participant Information and Consent Form v.2, rec'd 07-July-2017	11-Jul-2018

Delegated Review ☒ Full Board Meeting ☐

IRB 1 Registration #00001471 ☐ IRB 2 Registration #00008358 ☐ Not Applicable ☒

CERTIFICATION

The University of Saskatchewan Biomedical Research Ethics Board (Bio-REB) has reviewed the above-named research study. The study was found to be 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

PRINCIPAL INVESTIGATOR	DEPARTMENT	Bio #
Alison Oates	Kinesiology	17-157

approval and the views of this REB have been documented in writing. The University of Saskatchewan Biomedical Research Ethics Board is constituted and operates in accordance with the current version of the *Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans* (TCPS 2 2014).

Please send all correspondence to:

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

Biomedical Research Ethics Board (Bio-REB)

Certificate of Approval
Study Amendment

PRINCIPAL INVESTIGATOR	DEPARTMENT	Bio #
Alison Oates	Kinesiology	17-157

INSTITUTION(S) WHERE RESEARCH WILL BE CARRIED OUT

Biomechanics of Balance and Movement Laboratory University of Saskatchewan	College of Kinesiology University of Saskatchewan Saskatoon SK S7N 5B2

FUNDER(S)
UNIVERSITY OF SASKATCHEWAN

TITLE
Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Difficult Walking Tasks in Young Healthy Adults? A Pilot Study

APPROVAL OF	APPROVED ON	CURRENT EXPIRY DATE
Biomedical Amendment Form, rec'd 30-May-2018 Removal of Student Researcher, Abhishek Kumar Revised Application for Biomedical Research Ethics Review, rec'd 30-May-2018 Revised Recruitment Flyer, rec'd 30-May-2018 Revised PAWS Announcement, rec'd 30-May-2018 Research Participant Information and Consent Form, Version: 2	18-Jun-2018	11-Jul-2018

Delegated Review ☒ Full Board Meeting ☐

IRB 1 Registration #00001471 ☐ IRB 2 Registration #00008358 ☐ Not Applicable ☒

CERTIFICATION
The University of Saskatchewan Biomedical Research Ethics Board (Bio-REB) has reviewed the above-named research study. The study was found to be 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. 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
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Digitally Approved by Ildiko Badea
Vice-Chair, Biomedical Research Ethics Board
University of Saskatchewan

Please send all correspondence to:

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University of Saskatchewan
Room 223 Thorvaldson Building
110 Science Place
Saskatoon SK Canada S7N 4J8

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB)

Certificate of Re-Approval

Internal ID:	
Ethics Number:	17-157
Principal Investigator:	Alison Oates
Department:	College of Kinesiology
Student(s):	
Funder(s):	University of Saskatchewan
Sponsor(s):	
Title:	Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Difficult Walking Tasks in Young Healthy Adults? A Pilot Study
Protocol Number:	
Re-Approval Date:	31/07/2018
Expiry Date:	30/07/2019
Approval Of:	
Review Type:	Delegated Review
IRB Registration Number:	Not Applicable

* This study, inclusive of all previously approved documents, has been re-approved until the expiry date noted above

CERTIFICATION

The University of Saskatchewan Biomedical Research Ethics Board (Bio-REB) has reviewed the above-named project.The project 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 project, and for ensuring that the authorized project is carried out according to governing law. This approval is valid for the specified period provided there is no change to the approved project.

FIRST TIME REVIEW AND CONTINUING APPROVAL

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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 is constituted and operates in accordance with the current version of the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2 2014).

Digitally Approved by Gordon McKay, Ph.D.
Chair, Biomedical Research Ethics Board
University of Saskatchewan

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB) 17-Jul-2019

Certificate of Re-Approval

Ethics Number:	17-157
Principal Investigator:	Alison Oates
Department:	College of Kinesiology
Locations Where Research Activities are Conducted:	College of Kinesiology, Canada Biomechanics of Balance and Movement Laboratory, Canada
Student(s):	
Funder(s):	University of Saskatchewan
Sponsor:	
Title:	Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Walking Tasks in Young Healthy Adults? A Pilot Study
Protocol Number:	
Approved On:	11/07/2019
Expiry Date:	10/07/2020
Acknowledgment Of:	
Review Type:	Delegated Review
IRB Registration Number:	Not Applicable

* This study, inclusive of all previously approved documents, has been re-approved until the expiry date noted above

CERTIFICATION

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Digitally Approved by Gordon McKay, Ph.D.
Chair, Biomedical Research Ethics Board
University of Saskatchewan

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB) 23-Jun-2020

Certificate of Re-Approval

Ethics Number:	17-157
Principal Investigator:	Alison Oates
Department:	College of Kinesiology
Locations Where Research Activities are Conducted:	College of Kinesiology, Canada Biomechanics of Balance and Movement Laboratory, Canada
Student(s):	
Funder(s):	University of Saskatchewan
Protocol Number:	
Sponsor:	
Title:	Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Difficult Walking Tasks in Young Healthy Adults? A Pilot Study
Approved On:	10/07/2020
Expiry Date:	09/07/2021
Acknowledgment Of:	
IRB Registration Number:	Not Applicable
Meeting Date:	17/06/2020

* This study, inclusive of all previously approved documents, has been re-approved until the expiry date noted above

CERTIFICATION

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Digitally Approved by Dr. Gordon McKay, Ph.D.
Chair, Biomedical Research Ethics Board
University of Saskatchewan

UNIVERSITY OF SASKATCHEWAN

Biomedical Research Ethics Board (Bio-REB) 08-Jul-2021

Certificate of Re-Approval

Ethics Number: 17-157

Principal Investigator: Alison Oates

Department: College of Kinesiology

Locations Where Research Activities are Conducted: College of Kinesiology, Canada
Biomechanics of Balance and Movement Laboratory, Canada

Student(s):

Funder(s): University of Saskatchewan

Sponsor:

Title: Does Practice with the Haptic Anchors Have Any Effect On Dynamic Stability During Difficult Walking Tasks in Young Healthy Adults? A Pilot Study

Approval Effective Date: 09-Jul-2021

Expiry Date: 09-Jul-2022

Acknowledgment Of:

Review Type: Delegated Review

IRB Registration Number: Not Applicable

* This study, inclusive of all previously approved documents, has been re-approved until the expiry date noted above

CERTIFICATION

The University of Saskatchewan Biomedical Research Ethics Board (Bio-REB) has reviewed the above-named project.The project 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 project, and for ensuring that the authorized project is carried out according to governing law. This approval is valid for the specified period provided there is no change to the approved project.

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