THE INFLUENCE OF AGE AND PHYSICAL ACTIVITY ON THE CONTROL OF MEDIOLATERAL DYNAMIC STABILITY DURING WALKING

by

Yash Ramesh Rawal

A Thesis submitted to the Faculty of Graduate Studies of

The University of Manitoba

in partial fulfilment of the requirements for the degree of

MASTER OF SCIENCE

Faculty of Kinesiology and Recreation Management

University of Manitoba

Winnipeg

ABSTRACT

The overall aim of this thesis was to understand how ageing and physical activity influences mediolateral stability during gait. The study recruited 14 inactive community-dwelling older adults, 14 community-dwelling highly active older adults, 14 inactive young adults, and 14 healthy highly active young adults. The outcome variables included the angle of divergence (θ_d) of the net and individual-limb ground reaction force along with minimum lateral distance between the whole-body centre of mass and the lateral aspect of the base of support (d_{min}). The θ_d was calculated as the difference between the inclination angles of 1) the ground reaction force and 2) a line joining the COP and COM, which has been purported to be an important variable regulating dynamic stability. The d_{min} was calculated as the difference between the whole-body COM and the most lateral aspect of the BOS during gait; d_{min} is believed to represent a kinematic measure of instability. The participants performed three walking conditions (normal, fast and modified tandem walking) for a total of 30 walking trials on a 10-meter pathway.

Older adults (OA) exhibited greater kinematic indices of stability across all gait conditions compared to young adults (YA). OA exhibited a smaller θ_d but only for the right limb and net GRF analyses across all gait conditions. There were no differences in kinematic indices of stability or the kinetic measures across all gait conditions between physical activity groups. The greater stability, as evidenced by a greater d_{min}, maybe a strategy used by OA to maintain stability; however, by reducing the GRF orientation, OA may not be able to recover after a perturbation as they may have challenges in regulating their angular accelerations. Stability was greater on the dominant limb compared to the non-dominant limb (left limb) during gait. There could be an increased number of falls on the non-dominant side because of the poorer kinematic stability on that side. It also suggests that running may not play a role in the control of mediolateral normal gait stability, however; runners may be more resistant to

gait perturbations. More specific balance training physical activity may be needed to be influence mediolateral stability during gait.

Keywords: gait, older adults, physical activity, instability, mediolateral stability.

ACKNOWLEDGEMENTS

My time spent as an undergraduate and graduate student at the University of Manitoba has made me realize that there are too many people who I would like to thank for their support. I was very fortunate to be surrounded by such amazing people.

To my advisor, Dr. Jonathan Singer: I would like to thank you, for allowing me to constantly bombard you with questions about my research and writing as well as constantly making me think outside the box. I have truly learned more in the past two and half years than I could ever think of. I would also like to thank the members of my committee, Dr. Cheryl Glazebrook and Dr. Kathryn Sibley, for all the support and guidance with this thesis.

To the entire biomechanics and HuMAnlab group: I would like to thank you all for all the encouragement, support, insights, and just for being amazing friends. You guys made my experience as a grad student that more incredible.

To my friends: I would like to thank you for putting up with me, allowing me vent, and telling me to put in more work when I had to. I will always be grateful for all the support you have provided, and just always being there for me.

Finally, to my parents: Words cannot express the appreciation for all that you have done for me. I would like to thank you for always believing in me, encouraging me and providing me with continuous support. I do not think I would have been able to reach this far in life and my career without it. I will always appreciate it.

DEDICATION

To my parents Paurvi Rawal and Ramesh Rawal.

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TERMS AND DEFINITIONS

1. *Fall:* “A sudden and unintentional change in position resulting in an individual landing at a lower level such as on an object, the floor, or the ground, with or without injury” (Public Health Agency of Canada, 2014, p. 3).
2. *Centre of mass (COM):* A point that represents the weighted average of COM of all the body segments (Winter, 1995).
3. *Centre of gravity (COG):* The vertical projection of COM onto the ground (Winter, 1995).
4. *Centre of pressure (COP):* The vector force representing the weight average of all the pressure over the surface of area that is in contact with the ground (Winter, 1995); the point of application of the ground reaction force
5. *Gait cycle:* A cyclical movement pattern that is used for bipedal locomotion in humans (Murray, 1967).
6. *Stride/step width (walking base):* The side to side distance between the feet measured between the right and left ankle joint centres (Perry, 1992).
7. *Balance*: Is the ability to maintain the body’s centre of gravity within its base of support (Pollock, Durward, Rowe, & Paul, 2000). In research, balance has been interchangeably used with postural stability/control.
8. *Stability*: “Refers to the inherent ability’ of an object to remain in or return to a state of balance” (Pollock et al., 2000, p. 404).
9. *Postural stability:* The act of maintaining, achieving or restoring a state of balance of the body during static or dynamic conditions (Pollock et al., 2000).
10. *Dynamic stability:* Is the regulation of the relationship between the centre of mass and base of support (Lugade, Lin, & Chou, 2011)**.**
11. *Angle of divergence (θ_d):* Is the difference between the inclination angles of 1) the net ground reaction force and 2) a line joining the net centre of pressure and centre of mass

calculated with respect to the mediolateral axis of the global coordinate system (Singer, Mcilroy, & Prentice, 2014).

12. *Lateral instability*: A kinematic measure quantified by the smallest lateral distance between the whole-body centre of mass (COM) and the side of the support envelope (area bound by the feet) (Singer et al., 2014; Singer, Prentice, & McIlroy, 2013).
13. *Linear momentum*: defined as the "product of an object's mass and its velocity" (Hall, 2011, p. 394).
14. *Angular Momentum*: defined as the product of the moment of inertia and the angular velocity (Hall, 2011, p. 463).

CHAPTER 1 – SCIENTIFIC FRAMEWORK

1.1 General Overview

Falls and fall-related injuries are very serious issues for adults and are a key public health problem in Canada especially for adults over the age of 65 (Canadian Institute for Health Information, 2010). The Canadian Community Health Survey has suggested that 35.1% and 62.9% of injuries sustained by individuals aged 20-64 and over the aged 65, respectively, were a direct consequence of a fall (Billette & Janz, 2011). With the percentage of the total population aged over 65 being predicted to increase from approximately 16.8% (5,935,635) in 2016 to approximately 27.8% by 2063 (Bohnert, Chagnon, & Dion, 2014; Statistics Canada, 2016), Canada will see a dramatic increase in the absolute number of individuals experiencing falls, fall-related injuries and subsequent hospitalization.

Many studies of fall rates among older adults have relied on self-reporting of falls. This is an issue because self-reporting is not the most reliable tool in assessing fall risk even though there is a clear definition of what constitutes a fall (Public Health Agency of Canada, 2014). Under-reporting of falls could be due to a variety of reasons: people may have forgotten about the fall, resulting injuries were too minor to require medical attention, or the individuals did not consider their fall incident to be a fall. Research has also shown a sex-based disparity in self-reporting of falls, with men being more prone to under-report falls or deny having fallen when compared to women (Campbell et al., 1990; Furuya et al., 2009). In Canada, the incidence of self-reported falls have increased from 179,000 in 2003 to 256,000 in 2010 (a 43% increase from 2003 to 2010) and women had a significantly higher rate of fall-related injuries compared to men across all survey years (Public Health Agency of Canada, 2014). With this information, the number of actual falls and fall-related injuries while walking could be even higher than what has been suggested from the self-reported falls data.

Walking is a fundamental element of everyday life and is a very complex task for humans compared to other animals. Since humans have only two limbs, the centre of mass (COM) falls outside the base of support for much of the gait cycle. Moreover, two-thirds of the body mass is usually located two-thirds of the body height above the ground. (Winter, Patla, Frank, & Walt, 1990; Winter, 1995). Coupled together, these factors make human gait inherently unstable (Maki & McIlroy, 1997). Poor control of stability is a major factor underlying the increased risk for falls in adults, particularly during gait (Guralnik et al., 1993). Indeed, 11% and 28.2% of all serious injuries among Canadians aged between 20 to 64 and over 65 years were sustained while walking (Billette & Janz, 2011). Further, mediolateral stability may be a good predictor of falls (Brauer, Burns, & Galley, 2000; Lizama et al., 2014; Lizama et al., 2015) and is considered an important component of balance recovery and falls prevention among community-dwelling older adults (Hilliard et al., 2008). Falls in any direction can lead to an injury, increase in fear of falling, and reduction of physical activity levels and quality of life (Maki, Holliday, & Topper, 1994). However, falls in the mediolateral (sideway) direction have been accompanied with an increased probability of injuries such as hip fractures which could lead to hospitalization and subsequent complications (Kannus, Leiponen, Parkkari, Palvanen, & Jarvinen, 2006).

According to the Public Health Agency of Canada (2014), 45% of older adults (65+) who fell and reported an injury were walking on a surface other than ice. This suggests that falls do not just occur from slips or perturbations but can also result from poor proactive control. This suggests that a better understanding of why older adults fall while walking could lead to new strategies to reduce the number of injuries due to falls**.** Fall-related injuries sustained while walking represent the largest proportion of injuries compared to other activities (sport/physical activity, rising from furniture, health problems, climbing up/down stairs, walking on ice or snow, skating, skiing, and snowboarding) that were associated with fall-related injuries (Public Health Agency of Canada, 2014).

The components of gait and balance are fundamental for physical function and mobility. Research has shown that physical activity and exercise programs have benefits for reducing fall risk and incidence especially among older adults (Barnett, Smith, Lord, Williams, & Baumand, 2003; Binder et al., 2002). As it stands, many types of physical and exercise programs are ‘prescribed’ for fall interventions. There are not enough guidelines which leads to ambiguity on which exercises to do and the guidelines on how much to prescribe are varied as well (Haas et al., 2012; Sibley et al., 2015). Despite this, physical activity and exercise have shown to be effective in reducing falls (Cadore, Rodríguez-Mañas, Sinclair, & Izquierdo, 2013; Gill et al., 2016). However, there remains considerable uncertainty regarding how exercise influences the underlying biomechanical mechanisms of balance during walking.

The present research used a biomechanical approach to understand how ageing influences mediolateral stability via the control of applied forces during walking. Further, this work aimed to quantify the influence of physical activity on mediolateral stability during walking. The above work laid the foundation for future work, which can examine specific forms of exercise to understand if there are differential effects of exercise on balance.

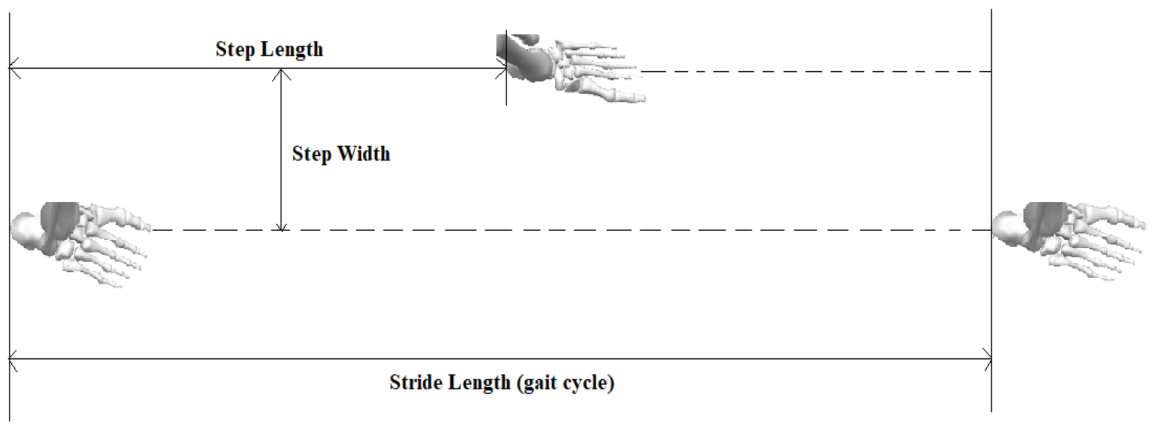
1.2 Review of Literature

1.2.1 Understanding the Gait Cycle and Gait Parameters

Gait parameters can be divided into temporal, spatial, kinetic and kinematic parameters. Temporal parameters consist, for example, of cadence, step time, stride time, stance time, swing time, single support time, and double support time while spatial parameters consist of step length, stride length, base of support area, and step width (Hollman & McDade, 2011). Kinetics is the study of movement and the forces involved in producing movement. Kinetic parameters that can be measured include ground reaction forces, joint reaction forces, centre of pressure, along with intersegment moments and moment power (Visser, Carpenter, van der Kooij, & Bloem, 2008). The study of kinetics parameters of gait helps explain how forces or moments of force are used to generate the observed movement of the body. Lastly, kinematics is the science that describes the motion of a body without consideration given to its mass and forces acting on it. Kinematic parameters that can be studied include the 3D positions of the segmental centres of mass, positions of joint centres, joint angles, angular velocities, and accelerations (Visser et al., 2008). These various parameters can be studied using various inertial sensors, and body/optical motion analysis system and force platforms.

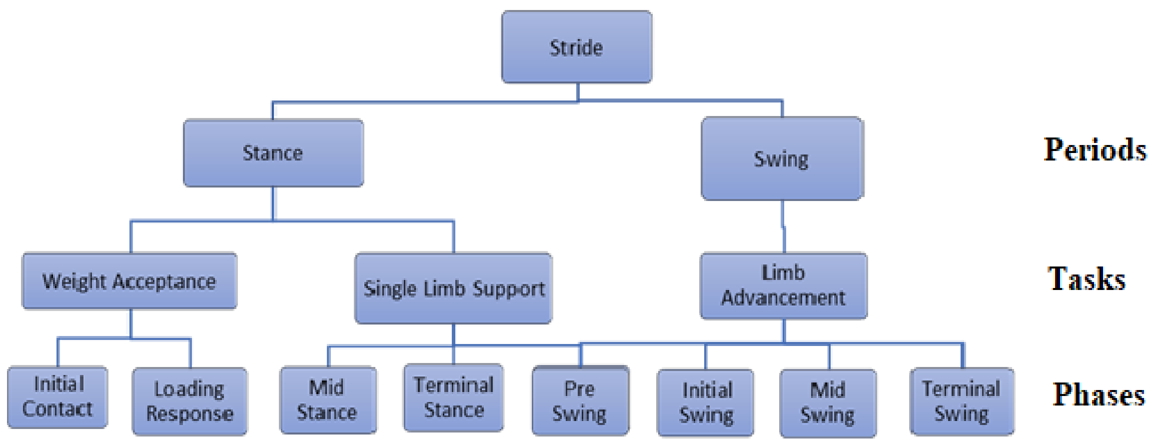
Normal gait consists of a series of consecutive cycles. One gait cycle (also referred to as a stride) consists of two consecutive heel strikes of the ipsilateral foot and is comprised of a stance period (foot is in contact with the floor) and swing period (the foot is moving to a new contact location) for each limb (Murray, 1967). Step length is measured as the anteroposterior distance between the points of contact of the heel with the ground of one foot followed by the point of contact of the contralateral foot (Murray, 1967). Stride length consists of the sum of two subsequent step lengths. Step/stride width is a measure of the medial-lateral distance between feet at heel strikes (Murray, 1967) (Figure 1.).

Figure 1 - Spatial gait parameters. Adapted from Bridenbaugh and Kressig (2011)



Further analysis of the stance and swing phases reveals a number of sub-phases (Figure 2.) (Whittle, 2002). During the stance period, two tasks occur: weight acceptance and single limb support while during the swing period, the task of limb advancement occurs in four phases: pre-swing, initial swing, mid-swing and terminal swing (Whittle, 2002). Given the fact that the body weight is transferred between the stance and swing limbs during every step, an inherent instability is introduced, particularly during the single limb support phase (Winter et al., 1990).

Figure 2 - Division of the gait cycle. Adapted from Burnfield (2010)

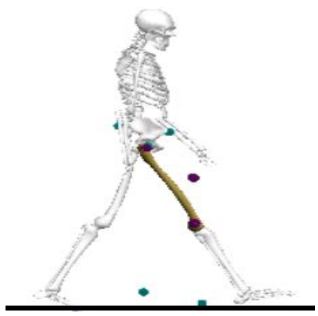


The stance period begins the instant the heel contacts the ground (heel strike) and ends the instant after the toe of the ipsilateral limb is off the ground (toe-off) (Kharb, Saini, Jain, & Dhiman, 2011). The foot is in contact with the supporting surface at all times during stance. It makes up 60% to 62% of the gait cycle (Novacheck, 1998). This period consists of 1) initial contact, 2) load response, 3) mid stance, and 4) terminal stance. Initial contact and load response occur during the task of weight acceptance.

Weight acceptance involves the transfer of the body weight onto a limb that has just finished the swing phase – the limb has an unstable alignment, with some degree of knee flexion (Kharb et al., 2011; Murray, 1967). The kinetic mechanisms that occur during weight acceptance aid in shock absorption and maintain the forward progression of the body (Kharb et al., 2011). During single limb support, one limb must support the entire body weight while also providing trunk stability during forward body progression. During a single stride, there are two periods of double limb support (both feet on the ground), which corresponds to the period between initial contact (stance limb) and toe-off (forthcoming swing limb) for each limb (Kharb et al., 2011; Murray, 1967).

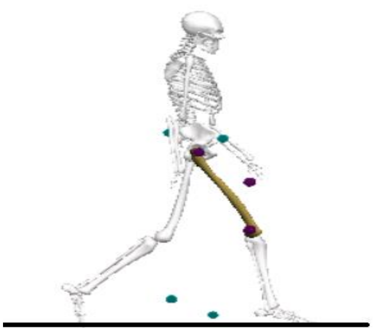
Initial contact of the foot represents the first sub-phase of weight acceptance and signifies the beginning of the stance phase (Whittle, 2002). This phase is frequently called ‘heel strike’, ‘heel contact’, ‘foot strike’ or ‘foot contact’ (Whittle, 2002). Initial contact is the moment when the ipsilateral heel of the foot touches the floor (Kharb et al., 2011) (Figure 3.) while the contralateral limb is in terminal stance (Whittle, 2002). Initial contact occupies about 2% percent of the gait cycle (occurs from 0-2% of the gait cycle) (Perry, 1992) . At heel strike, the hip is in a flexed position, the ankle is close to a neutral position and the knee is fully extended (Kadaba, Ramakrishnan, & Wootten, 1990). At the initial contact phase, the hip joint generates an eccentric extensor moment that limits forward trunk rotation and hip flexion under the superincumbent body weight. The knee joint has a concentric flexor moment to aid the hip joint in controlling the forward body movement and control hyper-extension. The ankle joint generates an eccentric dorsiflexor moment which controls the lowering of the foot (plantar flexion) (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995)**.**

Figure 3 - Initial contact phase



The loading response phase occupies about 10% percent of the gait cycle (occurs from 0-10% of the gait cycle) (Perry, 1992; Whittle, 2002) and constitutes the first period of double-limb support (Kharb et al., 2011). The loading response begins with the initial contact and ends when the contralateral limb is lifted (Perry, 1992). During this phase, the stance foot comes in full contact with the floor (Figure 4.), and the body weight is nearly fully transferred onto the stance limb (Kharb et al., 2011). The double-limb support period is important for shock absorption, control of the vertical COM position, weight-bearing, and forward progression (Kharb et al., 2011; Murray, 1967). The contralateral limb is in the pre-swing phase (Kharb et al., 2011). From initial contact to loading response, the hip flexion is slightly reduced, while the knee moves from being fully extended to slightly flexed, and the ankle plantarflexes which lowers the forefoot to the ground (Kadaba et al., 1990). During the loading response phase, there is a concentric extensor moment at the hip joint and a quick eccentric extensor moment of the quadriceps at the knee joint to ensure stability of knee, absorb the incoming impact force and control the rate of knee flexion. There is also an eccentric plantar-flexor moment to control the rate of dorsiflexion as the COM moves over the stance limb (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995).

Figure 4 - Loading response phase



The mid-stance phase begins when the contralateral limb leaves the ground and continues as the body weight travels parallel with the length of the foot until it is aligned over the forefoot of the stance limb (Kharb et al., 2011) (Figure 5.). It occupies about 20% of the gait cycle (occurs from the 10% to 30% of the gait cycle (Perry, 1992). This is a time of single limb support, during which, the contralateral limb is in mid-swing phase (Kharb et al., 2011). During this phase, the stance limb hip moves from a flexed position to 0 degrees (full extension - anatomical position). The stance knee moves from being in a flexion position to being fully extended, and stance ankle moves from a plantar-flexion to dorsiflexion position (Kadaba et al., 1990). From the loading response phase to mid-stance phase, there is an eccentric hip flexor moment to reduce/resist the increasing hip extension. While at the knee joint, the concentric knee extensor moment helps control knee flexion so as to control balance during the single limb support period. While at the ankle joint, there is an eccentric plantar-flexor moment which controls the dorsiflexion (Kirkwood, Gomes, Sampaio, Culham, & Costigan, 2007; Sadeghi, Prince, Zabjek, Sadeghi, & Labelle, 2002)**.**

Figure 5 - Mid-stance phase

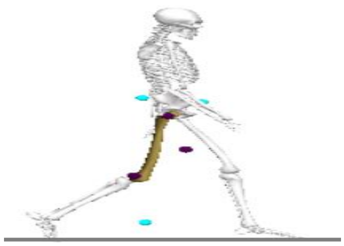
The terminal stance phase begins with heel rise of the stance limb and ends when the contralateral limb makes contact with the ground (Kharb et al., 2011) (Figure 6.). It is also called ‘heel rise’ or ‘heel off’ (Whittle, 2002). This phase occupies about 20% (occurs from the 30% to 50%) of the gait cycle (Perry, 1992; Whittle, 2002). During this phase, body weight moves ahead of the forefoot (right limb) (Kharb et al., 2011). It constitutes the second half of the double-limb support. During the phase, the stance hip is extended, while the ipsilateral knee is fully extended, and the ankle moves back into a plantarflexed position (Kadaba et al., 1990). From the mid-stance to terminal stance, the hip joint generates an eccentric flexor moment that continues to resist the increasing hip extension thus keeping the hip stable. The knee joint exhibits an eccentric extensor moment that controls the knee flexion. At the ankle joint, there is a concentric plantar-flexor moment that will forcefully push-off the limb (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995).

Figure 6 - Terminal stance phase

The swing period can be described as the interval in which the foot is not in contact with the ground (Kharb et al., 2011). It constitutes the remaining 38% to 40% of the gait cycle (Novacheck, 1998). The swing phase helps in the advancement of the limb which requires foot clearance from the floor (Kharb et al., 2011). The swing phase consists four sub-phases of 1) pre-swing, 2) initial swing, 3) mid-swing and 4) terminal swing (Kharb et al., 2011).

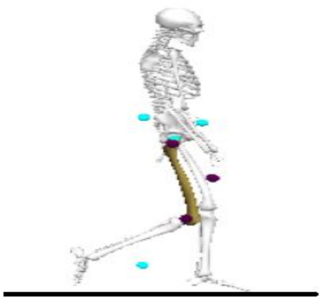
The pre-swing phase marks the end of the double-limb support period (Kharb et al., 2011) and occupies the last 10 percent of the contralateral stance phase (occurs from 50% to 60% of the gait cycle) (Baker, n.d.). It prepares the limb for swing advancement (Kharb et al., 2011) and in that sense, it could be considered a component of swing phase. This phase begins when the prior swing limb contacts the ground and ends with stance limb toe off (Kharb et al., 2011) (Figure 7.). During this period, the body weight on the stance limb is unloaded and it is transferred onto the contra-lateral limb. As a result, the pre-swing phase is also known as weight release or weight transfer (Kharb et al., 2011). The stance hip starts to return to 0 degrees from an extended position, while the knee starts to flex, and the ankle moves from dorsiflexion to plantar flexion (Kadaba et al., 1990). During the pre-swing phase, the hip joint generates a concentric flexor moment that provides a propulsive force to aid in the initial swing during the swing phase. The knee joint continues to exhibit an eccentric extensor moment that controls the knee flexion. At the ankle joint, there is a concentric plantar-flexor moment that aids in the forceful push-off the limb (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995).

Figure 7 - Pre-swing phase



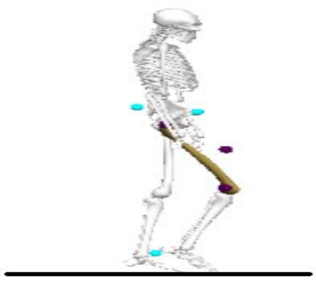
The initial swing begins once the toe leaves the ground and continues until mid-swing or the point at which the swinging limb is directly under the body (Kharb et al., 2011) (Figure 8.). This phase is also known as 'lift off' (Whittle, 2002). Approximately one-third of the swing period is spent in this phase (occurs from the 60% to 73% of the gait cycle) (Kharb et al., 2011; Perry, 1992). The stance hip is still in the process of returning to an anatomical position, while the knee is in a flexion position, and the ankle is in a dorsiflexion position. At the hip joint, there is a concentric hip flexor moment that results in the raising of the limb. At the knee joint, there is an eccentric extensor moment that will help control the knee flexion during the swing phase. At the ankle joint, there is a concentric dorsiflexor moment that helps in the toe clearance during the swing phase (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995).

Figure 8 - Initial swing phase



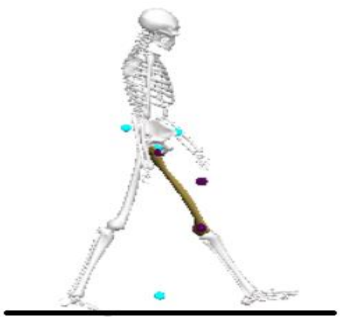
The mid-swing phase begins following maximum knee flexion and ends when the tibia is in a vertical position (Figure 9.). This phase is also known as ‘foot adjacent’ and ‘foot clearance’ (Whittle, 2002). The leg passes directly beneath the body. It occurs from the 73% to 87% of the gait cycle (Perry, 1992). The critical events that occur in this phase include continued limb advancement and foot clearance (Whittle, 2002). The contralateral limb is in mid-stance phase (Kharb et al., 2011). During mid-swing, the hip and knee are in flexion, and the ankle is in a neutral position (Kadaba et al., 1990). During mid-swing phase, there is an eccentric hip extensor moment to control passive flexion at the hip joint. There is an eccentric knee flexor moment to passively control the extension of the knee, while there is still a concentric dorsi-flexor moment at the ankle joint (Eng & Winter, 1995).

Figure 9 - Mid-swing phase



The terminal swing phase occurs when the tibia passes beyond perpendicular, and the knee fully extends in preparation for heel contact and ends when the right limb contacts the floor (Kharb et al., 2011) (Figure 10.). It occupies about 13% (from 87% to 100%) of the gait cycle (Perry, 1992). The contralateral limb is in the terminal stance phase (Kharb et al., 2011). By the end of the swing phase, the swing hip is still in a flexion position, while the knee becomes fully extended, and the ankle moves to neutral (Kadaba et al., 1990). From the mid-swing phase to the end of the swing phase, at the knee joint, the eccentric flexor moments move to become a concentric flexor moment that accelerates the thigh backwards and controls the knee. There is a concentric hip extensor moment just before heel contact at the hip joint so as to control the movement and prepare for body support. There is an eccentric dorsiflexor moment in preparation for initial contact (Bonnefoy-Mazure & Armand, 2015; Eng & Winter, 1995).

Figure 10 - Terminal swing phase



This section provides a general description for the gait cycle for healthy individuals. The gait cycle and gait parameters (kinetic, kinematic, temporal and spatial) usually tend to vary by age, and the presence of pathological conditions (Whittle, 2002).

1.2.1.1. Age-related challenges in Kinematic and Kinetic Parameters

There are differences in kinematic and kinetic parameters during gait between young and older adults (Anderson & Madigan, 2014; Bok, Lee, & Lee, 2013; Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998). Kinematic parameters such as joint angles during gait between young and older adults have been shown to be varied. Research has shown that older adults have slightly reduced range of motion (ROM) at both the knees, and ankles (Bok et al., 2013; Judge, Davis, & Ounpuu, 1996). The research has also suggested that older adults exhibit increased hip flexion, and a reduced peak hip extension ROM (Anderson & Madigan, 2014; Kerrigan et al., 1998). While kinetic parameters such as power have shown to be different between younger and older adults. For example, older adults exhibited a greater hip flexion power, reduced ankle plantarflexion power during the late stance phase, and reduced knee power during the mid-stance phase (Judge et al., 1996; Kerrigan et al., 1998). The reduction in ankle plantarflexion power, that can be attributed to ankle weakness, has been associated with a reduced step length in older adults (Judge et al., 1996; Winter et al., 1990).

1.2.1.2. Age-related Challenges in Spatial-Temporal Parameters

Spatial-temporal gait parameters have been used in many studies to identify gait stability/instability (Bisi, Riva, & Stagni, 2014; Debbi, Wolf, & Haim, 2012; Hollman, Kovash, Kubik, & Linbo, 2007; Hollman & McDade, 2011; Maki, 1997; Winter, 1990). Research has shown that the stance period proportion increases in men from 59% in 20-year-olds to 63% in 70-year-olds (Judge, Davis, & Ounpuu, 1996) and the double limb support phase has been seen to also increase with age from 18% in young adults to approximately 26% in healthy elderly people of a total gait cycle (Judge et al., 1996). Spatial-temporal parameters like gait speed, double-support time and stride length have been seen to be associated with fear of falling, especially in older adults (Maki, 1997). Step width, and step time are also good indicators for the prediction of risk of fall (Gabell & Nayak, 1984; Hausdorff, Rios, & Edelberg, 2001).

In healthy individuals, variability of spatial-temporal gait parameters remains somewhat stable throughout the lifespan (Hausdorff et al., 2001). An increased step-to-step variability during gait could indicate an inability to compensate for instability and increase the tendency to fall (Gabell & Nayak, 1984), which has been associated with aging and a reduced postural stability (Tucker, Kavanagh, Morrison, & Barrett, 2009). Several studies have investigated age-related differences in gait parameters and most of these studies have shown that there is a difference in walking speed between the age groups (Gabell & Nayak, 1984; Kang & Dingwell, 2008; Meyer & Ayalon, 2006; Owings & Grabiner, 2004; Sutherland, 1997). The studies have shown that older adults tend to walk at a consistent speed during steady state walking (Sutherland, 1997). They also tend to walk more slowly at both self-selected and maximal speed, and have a shorter step length, and broader walking base. This results in a gait cycle with a longer stance and/or longer double support time (Meyer & Ayalon, 2006). These increases have been suggested to be a shift towards a more stable gait pattern. An increase in double support time would reduce the amount of time where the COM is outside the base of support (BOS) (Winter et al., 1990). Older adults also exhibited greater stride-to-stride variability in step width when compared to young adults (Owings & Grabiner, 2004) which has been associated with an increased risk of falls (Hausdorff et al., 2001).

A one-year prospective cohort study was done by Maki (1997) on older adults to find out if spatial-temporal measures of foot placement during gait can predict the likelihood of future falls. The author reported that older adults who had a fear of falling also had reduced stride length, reduced speed, increased double-support time, and poorer clinical gait scores but these factors had no association with risk of falling. While the older adults who had an increased risk of falling showed increased stride-to-stride variability in stride length, speed, and double support time. However, the latter factors had no association with fear of falling. Stride-to-stride variability can influence a person's gait, for example, an increase in step/stance time during the stance phase increases the amount of time the COM is outside the BOS. This would increase the COM acceleration at the subsequent heel strike. This could lead to an increased necessity to control the COM by regulating the applied forces generated during the subsequent steps, which could lead to instability during gait. The author reached this conclusion by having participants complete a self-report questionnaire about their fear of falling. Baseline gait measurements using a series of balance tests were also measured. The baseline gait measurements included the one leg stance test for the static posturography test, spontaneous-sway and induced-sway using a movable perturbation platform for the dynamic posturography test, and the Tinetti balance and gait test was used for the clinical balance assessment. The static and dynamic posturography test comprised of eyes-open and eyes-closed trials.

In another one-year prospective study with older adults, Hausdorff et al. (2001) looked at gait variability and the association with falls. The authors reported that 40% of the participants fell during the 1-year study, the stride time variability was significantly greater in subjects who subsequently fell (106 ± 30 ms) when compared to those who did not fall (49 ± 4ms). The increase in stride time could be due to visual impairments, muscle strength, and the presence of gait pathological conditions. The results of Hausdorff et al. (2001) are similar to Maki (1997) as they both showed that stride-to stride variability was associated with risk of falling.

Thus far, the above section has focused on linear analysis to measure gait variability. Linear measures focus on the magnitude of variability that shows either an increase or decrease in the amount of variability and are reported using range, standard deviation and coefficient of variation (Kaipust, Huisinga, Filipi, & Stergiou, 2012). Another way of measuring gait variability is using non-linear tools that focus on understanding how gait variability occurs by finding the exact and hidden changes of the variability that cannot be detected by linear measures (Goshvarpour & Goshvarpour, 2012). Nonlinear measures of variability include Poincare Plot, wavelets, and detrended fluctuation analysis (DFA), and dynamic stability include Lyapunov exponents (LEs), the Hurst exponent, and Floquet multipliers (FMs) (Goshvarpour & Goshvarpour, 2012; Hamacher, Singh, Van Dieën, Heller, & Taylor, 2011). For example, Buzzi, Stergiou, Kurz, Hageman, and Heidel (2003) used the Lyapunov exponent along with linear measures (standard deviation and coefficient of variation) to look at how aging affects certain kinematic parameter variability during gait among young and older females. The results showed that older adults had a larger Lyapunov exponent for all parameters evaluated indicating local instability while the linear measures showed significantly higher stride-to-stride variability in the elderly population. The authors also indicated that the fluctuations in the time series for the nonlinear measures of certain kinematic parameters (hip, knee, and ankle joint angles) and temporal parameter (walking speed) are not random but display a deterministic behaviour (Buzzi et al., 2003). This means that older adults have a reduced ability to control joint motion indicating that there was local instability at the joints, which may cause older adults to fall. In another study, Kurz and Stergiou (2007) looked at the forward motion of COM using nonlinear analysis of gait dynamics. Participants walked at a self-selected speed for two minutes on a treadmill. A horizontal actuator assisted in the forward translation of the COM during the stance phase at various forces equal to 0, 3, 6, 9 percent of the participant's body. The authors used a Lyapunov exponent for the hip, knee, and ankle joint time series. The authors reported that as the percentage of assistance provided by the actuator increased, so did the magnitude of Lyapunov exponent. The authors suggested that this was the case because the horizontal propulsive forces influence nonlinear dynamics through the control of the forward progression of the COM. This shows that another advantage of nonlinear measures is that it measures the change in COM over time, which is not the case for linear measures. This is an advantage since the likelihood of detecting instability from a single step or a few steps is low, nonlinear measures allow for COM measures over longer durations, which may allow for easier analysis for gait dynamics. While useful in detecting instability, a main disadvantage of nonlinear measures is that they do not measure the underlying cause of instability, that is, how force regulates dynamic movements.

1.2.2 Current Balance Measure Tests

Several balance tests have been/are used to examine whether an individual has poor balance and potential fall risk. The various balance tests have been classified into clinical and laboratory measures. Examples of clinical tests include the Activities of Balance Confidence (ABC) (Powell & Myers, 1995), the Tinetti Balance and Gait Test (Tinetti, 1986), the Berg Balance Scale (Berg, Wood-Dauphinee, Williams, & Maki, 1992), Timed Up and Go (Podsiadlo & Richardson, 1991) and the Balance Evaluation Systems Test (BESTest) (Horak, Wrisley, & Frank, 2009). Each of the clinical tests try to mimic activities daily living. They challenge the stability of an individual by requiring them to shift their COM in various ways.

They are also easy to administer and score. Some of these tests put demand on the balance control by measuring dynamic balance and functional mobility of older adults (Berg, Wood-Dauphinee, Willams, & Maki, 1992; Podsiadlo & Richardson, 1991; Tinetti, 1986). However, clinical tests are based on scales, the balance scores provided by the tester are subjective thus could be exposed to tester bias (Mancini & Horak, 2010). In addition, clinical tests cannot detect minute changes in balance performance.

Laboratory tests are typically composed of force platform posturography (Duarte & Freitas, 2010). This type of balance test assesses the postural sway (centre of pressure displacements) of an individual (Visser et al., 2008). Posturography is further divided into two categories: static and dynamic posturography (Duarte & Freitas, 2010). Static posturography quantifies the centre of pressure of an individual during a quiet erect stance while dynamic posturography quantifies the centre of pressure of an individual by the use of balance perturbations (Duarte & Freitas, 2010). The use of balance perturbations or change in surface or visual conditions is used to study compensatory responses. Sensory perturbations can be done to the visual, vestibular, and somatosensory system (Patla, 2003)**.** The sensory perturbations require the central nervous system to adapt ongoing postural strategies to maintain stability (Patla, 2003). There have been various methods researchers have used to manipulate the environment, for example, dropping of obstacles, waist tugs and lean-and-release perturbations (King, Akula, & Luchies, 2012; Pai, Rogers, Patton, Cain, & Hanke, 1998; Schillings, van Wezel, Mulder, & Duysens, 2000). The problem with using these kinds of tests is that the types of equipment are costly and require technical knowledge of these systems. Although laboratory test of compensatory balance control in response to perturbations can help identify the elements of the control system responsible for instability (King et al., 2012; Pai et al., 1998; Schulz, Ashton-Miller, & Alexander, 2005), these techniques identify balance recovery using a single step after a single perturbation, rather than ongoing restabilization that occurs during gait.

Overall, both clinical and the laboratory test have had varying success to predict individual fall risk in adults, especially older adults. Reliability and validity studies have been performed on most of the balance tests. A test is considered reliable if the results are consistent and repeatable while a valid test measures what it is intended to measure (Rikli & Jones, 2013). There are many studies that have performed reliability and validity testing of the clinical balance tests and have been found to have a generally good reliability and validity ratings (Mancini & Horak, 2010). While for laboratory balance tests, there needs to be further research done assessing the reliability and validity of this type of balance test (Harstall, 1998).

While the clinical balance tests/assessments can provide an indication of balance problems that exist and the risk of falls in individuals, the ability of these tests to detect small changes in balance performance is unknown. Clinical tests are used to evaluate balance during functional tasks, thus reducing the need for equipment resources, time and complicated analysis. They, however, do not provide a clear indication of which elements of the balance control during gait are being measured (Mancini & Horak, 2010) and do not provide detailed information regarding the underlying mechanisms governing balance control. Since most falls occur during dynamic movement (Berg, Alessio, Mills, & Tong, 1997), high resolution laboratory-based measures are necessary to identify and understand the underlying the biomechanical mechanisms that may lead to falls. As most falls and fall-related injuries that are sustained by older adults occur while walking, the current study focused specifically on dynamic balance during gait using high-resolution laboratory-based measures to understand how the underlying biomechanical mechanisms could be responsible for on dynamic stability during gait.

1.2.3 Directional Stability

Gait instability has been seen in both the mediolateral and anterior-posterior directions (Winter, 1995). Mediolateral instability, however, may provide significant insight into gait instability as it requires the use of active control strategies rather than the passive control strategies that are predominant in anterior-posterior stability (O'Connor & Kuo, 2009). In a study done by Oates, Frank, Patla, and Greig (2005), changes in centre of pressure and centre of mass in the anterior-posterior and mediolateral directions were examined in younger adults while walking on irregular and regular surfaces. They found that while walking on slippery surfaces, there was an increase in the COM momentum in the mediolateral direction possibly due to a transfer of momentum from the anterior-posterior to mediolateral direction needed to help disperse the forward movement. In another study by O'Connor and Kuo (2009), they looked at the control of balance during walking and standing trials by applying low-frequency perturbations while measuring foot placement. They found that during walking, step variability was more sensitive to the mediolateral perturbations than the anterior-posterior perturbations. This was also seen in the tandem (heel-to-toe) stance. While in the stance trials, there was more sensitivity to the anterior-posterior perturbations than the mediolateral perturbations.

There is also evidence that older adults exhibit greater mediolateral instability during walking compared to younger adults (Maki, Edmonstone, & McIlroy, 2000; Schrager, Kelly, Price, Ferrucci, & Shumway-Cook, 2008). Schrager et al. (2008) tested the age-related differences in mediolateral stability during normal and narrow base walking. The authors looked at step error rates and spatial-temporal parameters, and gait variability during usual and narrow base walking. They found out that with increasing age, there was a greater variability in stride velocity and step length under both conditions. Together, such variation in stride velocity and step length could suggest that participants had varied single- and double-limb support times. This could lead the COM to being outside the base of support (BOS) for a greater duration of the single leg support phase or that COM could be closer to exceeding the lateral stability limits which could lead to mediolateral instability. Their data indicated that an age-related decline in the ability to successfully perform a narrow base walking task was associated with spatial-temporal parameters and mediolateral instability. The results of Schrager et al. (2008) are similar to the results of the study done by O'Connor and Kuo (2009) suggesting that the step variability was more prominent in the mediolateral direction and step variability also increases with age.

As stated earlier, mediolateral stability may be a good predictor for falls (Brauer et al., 2000). Brauer et al., (2000) recorded the centre of pressure (COP) in quiet stance, at the limits of stability (reaction-time step task) for 100 elderly women (65-86 years). The authors showed that mediolateral stability during a combination of quiet stance, rapid step task, and the movements to the limits of stability were best able to predict fall status. Falls that happen in the mediolateral direction have shown to be more prevalent and more devastating because it has been associated with the increased probability of an injury such as hip fractures (Kannus et al., 2006).

1.2.4 Centre of Mass - Centre of Pressure Relationship

Centre of mass (COM) can be defined as 'a point total body mass in the global reference system (GRS)' while centre of pressure (COP) is defined as 'the point location of the vertical ground reaction force vector' (Winter, 1995, p. 194). Displacements of the COP are believed to be a neuromuscular response to instability (Winter, 1995). The downward vertical projection of the COM has also been referred to as centre of gravity (Winter, 1995). When an individual is standing (static stance), both the vertical projection of the COM and COP are localized between the feet, however, there is still some movement of the COM and COP (Winter, 1995).

The relationship between the COM and COP is considered to be an important one. Winter (1995) described this relationship, which states that the COM projects a downward force onto the ground called the centre of gravity. The centre of gravity results in a ground reaction of equal magnitude, but opposite direction. When the COM undergoes small displacements as a result of internal perturbations (blood flow, breathing, muscle activity etc.), a COP displacement is required to maintain the COM within the BOS to prevent loss of balance. This is done by the generation of moments of forces (an upward directed ground reaction force) displaced in front of or behind the COM that will lead to a restabilizing moment. If the COM exceeds or comes close to exceeding the BOS, displacements of the COP may not be able to contain the COM and thus a change in support (e.g. a step) is required to generate a moment of force large enough to facilitate restabilization (Hasson, Van Emmerik, & Caldwell, 2008; Winter, 1995). The step taken redefines the support envelope and the area in which the COP is free to move.

There have been numerous studies examining ‘postural sway’, which is described as the sway of the COM in a standing position and is measured by the motion of the COP (Yamamoto et al., 2015). However, there are disagreements on the role of ‘postural sway’ in stability control. Kiemel, Oie, and Jeka (2002) suggested that the ‘postural sway’ has no practical role and just represents noise that is associated with the execution of movement and the neural control system while Mancini and Horak (2010) considered ‘postural sway’ as an excellent measure of the balance system because of its sensitivity. Work done by Riley, Wong, Mitra, and Turvey (1997) suggested that the ‘postural sway’ serves as an exploratory role by providing an individual’s perceptual system with a constant source of sensory information. Building on the above premises, Carpenter, Murnaghan, and Inglis (2010) suggested that there are two roles for ‘postural sway’: firstly, COP displacements allow for the regulation of stability, by controlling COM kinematics; secondly, COP displacements may allow for essential information to be gained from the environment for balance control, by generating a certain quality or volume of sensory information. This suggests that larger amounts of 'postural sway' (i.e. COP displacements) may be seen as a natural adaptation method by older adults to deal with age-related decline in sensory thresholds and integration capacity, that is, a larger COP displacements may provide greater amounts of sensory (plantar cutaneous) information for use in stability control to counteract the challenges from other sources of sensory information.

Studies showed that young adults have less 'postural sway' than older adults (Muir, Kiel, Hannan, Magaziner, & Rubin, 2013; Weirich, Bemben, & Bemben, 2010) and an increased COP displacement has been associated with increased risk of falling (Lafond, Corriveau, Hébert, & Prince, 2004). 'Postural sway' also varies under different conditions (e.g., eyes open or closed, or standing on a different support), it was reported that there was an increased COP displacement when subjects' sensory information was reduced i.e. subjects' eyes were closed (Woollacott, Shumway-Cook, & Nashner, 1986). A more recent study by Muir et al. (2013) looked at 'postural sway' of young and older adults during quiet stance. They reported that young adults had less COP displacement compared to older adults. This was noticeable in the anterior-posterior and mediolateral directions. In the young adults, the COP remained closer to the mean position while the COP displacement had increased variability for older adults.

In static stance, the criteria for balance are relatively small when compared to those of dynamic stability (Winter, 1995). When an individual is in static stance, the ability to maintain stability is relatively simple, as one only needs to maintain the COM within the BOS (Gatev, Thomas, Kepple, & Hallett, 1999; Winter, 1995) - this represents the condition for static postural stability (Taweetanalarp, Prasertsukdee, Vachalathiti, & Kaewkungwal, 2011). When comparing static stability and dynamic stability, Razavi (2017) looked at the comparison between static and dynamic stability in ‘postural sway’ in the mediolateral and anterior-posterior directions. For their study, the percent COG was used as an index of static stability. Static stability was measured in relation to the anthropometric range for centre of gravity, and base of support at various induced postural sway angles. While for the dynamic stability test, the participants stood on a gait analyser treadmill that measures the dynamic pressure and forces. The angle of the treadmill was varied to assess the postural sway at the various angles. The author calculated the Lyapunov exponent to assess stability at the various ‘postural sway’ positions. The results of the static stability tests and dynamic stability tests were then compared. The results showed that during the static stability test, when the sway angle was increased, the centre of gravity percentage increased as well. The results also showed that during the dynamic stability test, the variation in the centre of pressure path increased when the sway angle was also increased in every direction. This suggests that ‘postural sway’ can be seen in both static and dynamic conditions. There could be a relationship between static and dynamic control of stability.

1.2.4.1 COM-COP Relationship during Gait

Gait is characterized by a series of movements in which the body is propelled forward and maintains stability (Perry, 1992). As a person walks, the swing limb moves through space while the stance limb provides a support base to facilitate body-weight support and forward propulsion (DeLisa, 1998; Murray, 1967). The COM for humans is generally located anterior to the second sacral vertebra, midway between hip joint (DeLisa, 1998). During gait, the theoretically most mechanically efficient COM displacement is in a straight line without deviation up/down nor sideways (Kuo & Donelan, 2010; Saunders, Inman, & Eberhart, 1953). This is not possible in normal human gait because the motion of the stance and swing legs are like pendulums: heel strike redirects the COM, and the knee fully extends during the mid-stance phase**,** and thus the COM deviates both laterally and vertically from the straight line

(DeLisa, 1998; Kuo & Donelan, 2010). The automaticity of gait and minimization of energy during locomotion on level ground has been demonstrated by robots that are able to passively stabilize themselves and walk down a slope (in the anterior-posterior direction) with its only source of power is gravity and inertia (gravitational potential energy) without any control of joint angles (McGeer, 1990). Subsequent research on robotic gait has been able to demonstrate dynamic walking on level ground which requires a push off at the ankle or powering of the hip (Collins, Ruina, Tedrake, & Wisse, 2005). This process of walking is modelled as an inverted pendulum system in which the COM vaults over the rigid stance limb (Alexander, 1995). The average COM vertical displacement is approximately 5cm while the lateral displacement is also approximately 5 cm during gait in the adult male (DeLisa, 1998). The highest and lateral limits are reached during the mid-stance phase of gait (Cavagna, Heglund, & Taylor, 1977; Cavagna, Thys, & Zamboni, 1976; DeLisa, 1998). This suggests that the COM approaches the lateral limits of stability during the mid-stance phase of the gait cycle, which could increase the chance of lateral instability. As suggested earlier, mediolateral instability maybe a good predictor of falls (Brauer, Burns, & Galley, 2000; Lizama et al., 2015; Tirosh & Sparrow, 2005).

During normal steady-state gait, the criterion for the level of balance changes constantly. This is because balance during gait is a product of the ever-changing BOS and COM relationship (Maki & McIlroy, 1997; Winter, 1995). There are periods during the gait cycle (i.e. single limb support) where the COM moves outside the base of support. The BOS encloses the COM to prevent the person from falling during the subsequent double-limb support period. For example, a study done by Lugade, Lin, and Chou (2011) showed that during double-limb support, COM and COP remained inside the BOS. During single limb support, however, the COP remains within the BOS while the COM travels outside of the base of support. During gait, the body is considered to be dynamically stable in the anterior-posterior direction and unstable in the lateral direction (Kuo, 1999). In fact, walking has been described as a series of controlled falls because the whole-body COM regularly enters and exits an ever-changing base of support during each double-limb and single-limb support phase, respectively (Winter et al., 1990). Each successive fall is prevented with correct placement of the stepping limb along with the correct magnitude and orientation of the net ground reaction force to stabilize the whole-body COM during the double support stance phase (Winter et al., 1990). Therefore, an important determinant of balance could be the characteristics of force generation during successive heel contacts. When a force is applied at the COM of an object, the force will cause a linear motion of the object. However, if the force is not applied at the COM of the object (eccentric force), the force will cause a linear and angular motion of the object.

The current study examined whole-body instability in the mediolateral direction during walking by identifying the minimum distance between the whole-body COM and the lateral aspect of the BOS (d_{min}) and the orientation of the ground reaction force relative to the whole-body COM (angle of divergence - θ_d) during three walking conditions. These variables have been used in previous studies done by Singer et al., (2013, 2014, 2016) to examine mediolateral stability control during voluntary and perturbed stepping. Singer et al., (2016) suggested that older adults would require more time to regulate the net θ_d relative to the COM during the restabilization phase following foot-contact. The authors reported that there was a significant positive correlation between the timing of the second peak θ_d and mediolateral COM kinematic stability. Older adults who exhibited a poor timing of the second peak θ_d also exhibited poor COM kinematic stability during restabilization. The second peak θ_d has been associated with reactive control of stability (Singer et al., 2016).

1.2.5 Effect of Age on Factors Underlying Stability

Even though young children and young adults have higher probability of sustaining an injury from falls from various activities, older adults had an increased risk of sustaining an injury from a fall while walking (Billette & Janz, 2011). Age-related changes in the ability to control balance during walking have been associated with falls (Schultz, Ashton-Miller, & Alexander, 1997). Impaired balance during gait could be due to loss of muscle mass, muscle force output and power, or visual, vestibular and somatosensory decline or a combination of these factors (Janssen, Samson, & Verhaar, 2002; Macaluso & De Vito, 2004; Pijnappels, van der Burg, Reeves, & van Dieën, 2008; Qiu et al., 2012). Moreover, these factors could be modulated by maintaining high levels of physical activity throughout the lifespan (Granacher, Zahner, & Gollhofer, 2008; Henwood & Taaffe, 2005; Rocha, Santos, Vasconcelos, & Santos, 2016).

1.2.5.1 Muscle mass, muscle strength, and power

There are three types of muscle tissues: cardiac, smooth and skeletal muscles (Whittle, 2002). The skeletal muscles are responsible for the movement of the limbs (Whittle, 2002). For an individual to move or perform a task, there is a minimum amount of strength or power required (Marsh et al., 2006) and as a person ages, muscle strength and power decreases (Keller & Engelhardt, 2013). Muscle strength is defined as "the maximal force or tension level that can be produced by a muscle group" (Heyward & Gibson, 2014, p. 48) while muscle power is defined as 'the product of muscular force and velocity of muscle shortening' (Hall, 2011, p. 170)**.**

Humans generally tend to lose muscle mass as they age, from the age of 25, muscle strength and power tend to decrease with the loss occurring more rapidly from the age of 65 (Metter, Conwit, Tobin, & Fozard, 1997). The loss of muscle mass, strength and power could be attributed by a reduced participation in physical activity and exercise. The decrease in muscle strength and power can be reduced by taking part in physical activity and/or exercise (Law, Clark, & Clark, 2016). The decrease in muscle mass can affect individuals performing activities of daily living such as walking (Hamill & Knutzen, 2008). Reduction in the size and type of fast twitch muscle fibres are the primary reasons for the reduction in muscle mass (Clarkson, Kroll, & Melchionda, 1981). The decrease in muscle mass is not uniform across the muscles in the body and is more evident in the lower extremities where the rate of loss was twice that that of the upper extremities (Janssen et al., 2000; Jennekens, Tomlinson, & Walton, 1971).

As individuals age, the loss of the muscle mass is directly related to the loss of muscle strength and power. Studies have shown that a decrease in muscle power can result in delayed postural reactions (e.g. correction step) to external perturbations probably leading to loss of balance and falls (Rubenstein, 2006; Woollacott & Shumway-Cook, 1990). A study by Lee and Chou (2007) examined the association between strength reduction in leg muscles due to aging and balance control during gait in younger and older adults. They hypothesized that a reduction in muscle strength in the leg muscles is associated with changes in the COM-COP inclination angles. Participants (38 older adults, 19 younger adults) were asked to walk at a self-selected pace across a walkway for two conditions (obstructed and unobstructed). Isometric muscle strength of the hip abductors, knee extensors, and ankle plantar flexors were also performed on the participants. The result showed that older adults had significantly greater and smaller inclination angles in the mediolateral and anterior-posterior directions, respectively, when compared to younger adults (Lee & Chou, 2007). The inclination angles in the mediolateral direction are a direct result of the increased step width. The greater inclination angles in mediolateral direction in older adults could lead one to infer that there is also greater mediolateral instability during walking when compared to younger adults. The results also showed that older adults had a 34% to 50% reduction in isometric joint moments in the hip abductors, knee extensors and ankle plantar flexors when compared to younger adults. The authors also performed a regression analysis to see if the isometric joint moments could predict the gait measures for both task conditions and for all participants (Lee & Chou, 2007). The results showed that there is a relationship between the reduction in lower extremity muscle strength and COM-COP inclination angles in the mediolateral direction and anterior-posterior direction. Overall, the above studies suggest that physical activity and exercises reduces the rate loss of muscle mass, strength and power, which is required to perform activities of daily living.

1.2.5.2 Visual, Vestibular and Somatosensory decline

The ability to maintain posture and balance during both standing and walking requires the integration of the visual, vestibular, and somatosensory systems (Poole, 1992). For example, vision plays a role in the control of dynamic stability during the safe navigation through the environment and identifying obstacles from sensory receptors (rods and cones) in the retina (Lin, Tsubota, & Apte, 2016; Patla, 1997). Research has shown that the visual system is the primary sensory system used to maintain postural balance during quiet stance (Liaw, Chen, Pei, Leong, & Lau, 2009). While the vestibular system gets information about motion, equilibrium and spatial orientation from the vestibular apparatus in each ear (Guerraz & Day, 2005). Lastly, information from proprioceptive and cutaneous inputs is needed to maintain a normal stance and to safely perform activities of daily living such as walking, reaching and grasping, and maintenance of posture (Horak, Nashner, & Diener, 1990).

The different sensory systems begin to develop and refine starting from a young age and begin to decline as people age, especially for older adults (Era et al., 2006; Poole, 1992). For example, results from a study done by Era et al. (2006) showed that there was difference in postural control amongst young and middle-aged adults in more demanding test conditions (standing with eyes closed) than during normal standing with the eyes open. In another study, Liaw et al., (2009) compared the balance characteristics among different age groups using computerized dynamic posturography. The participants stood on a fixed platform and performed six conditions: 1) eyes open on the fixed platform, 2) eyes closed on the fixed platform, 3) swaying visual surround on the fixed platform, 4) eyes open but the platform swayed, 5) eyes closed and the platform swayed, and 6), swaying visual surround on a swaying platform. The purpose of the six conditions was to isolate the different sensory systems (i.e., visual, vestibular, and somatosensory) used for balance (Nashner, 1982). The authors showed that the elderly group had the lowest average scores for maximal and average stability in all sensory organization test (SOT) subtests (Nashner, 1982)**.** Liaw et al.'s research also showed that most risk of postural imbalance occurred when the visual and somatosensory inputs were isolated. In another study, Hallemans et al. (2009) looked at postural stability during eyes open and eyes closed conditions while walking and compared the results with eyes open and eyes closed during a quiet stance. Participants (20 adults, and 40 children) performed eyes open and closed balance test during a quiet stance on a force plate, as well as walked at a self-selected speed on a walkway with their eyes open and closed. The authors used Spearman correlation to test whether the visual deprivation in gait was related to the role of vison in postural control. Their results showed that during the quiet stance, 'postural sway' was significantly larger in the eyes closed, condition and more so in the younger participants. The authors also showed that all participants, when walking with eyes closed, had shorter stride length, increased step frequency, slower walking speeds and increased duration in the double support phase. There was also a correlation between 'postural sway' and walking speeds, step frequency, and stride length. The latter results could suggest that gait is affected by visual deprivation and more so for children. In summary, the above studies suggest that the integration of the visual, vestibular and somatosensory systems are important for postural control during normal standing and gait and that the decline of these systems can lead to instability and thus contribute to an increased possibility of falls.

1.2.6 Effects of Physical Activity and Exercise on Stability

Physical activity can be described as any body movement produced by the use of skeletal muscles that substantially increases energy expenditure and can be divided into occupational leisure activity, and activities of daily living (Caspersen, Powell, & Christenson, 1985). Physical activity can range from everyday actions such as walking, housework or gardening to leisure activities such as swimming, cycling or gym activities (Caspersen et al., 1985). While exercise can be described as planned/structured activities where repetitive body movements produced by skeletal muscles are made to improve/maintain components of fitness such as strength training or fitness programmes (Caspersen et al., 1985). Questionnaires, heart rate monitors, and diaries have been used to measure physical activity levels (Skelton, 2001).

In general, performing physical activity and exercise has shown to have many health benefits that include postural balance and proprioception improvements, reduction in fall risk and fall incidence, and reduction and prevention of injuries (Gillespie et al., 2003; Gioftsidou et al., 2012; Perrin, Gauchard, Perrot, & Jeandel, 1999). But what we do not know is which exercises to do and how much to prescribe (Haas et al., 2012; Sibley et al., 2015). A significant amount of research investigating the benefits of exercise for reducing fall risk and fall incidence rates in the older adult population, either as a single exercise programs (e.g., resistance exercise, walking, tai chi) or as a multicomponent exercise programs (e.g., aerobic endurance, flexibility, strength, and balance training) has been published over the past two decades (Barnett et al., 2003; Binder et al., 2002; Chang et al., 2004; Gillespie et al., 2003; Haines, Bennell, Osborne, & Hill, 2004). These studies had similar findings showing that both single exercise programs or as a multicomponent exercise programs have benefits for reducing fall risk and incidence among older adults which again suggests that there is an ambiguity on which exercises to do and how much to prescribe.

Studies have provided evidence that physical activity had an impact on postural stability (Brooke-Wavell, Athersmith, Jones, & Masud, 1998; Perrin et al., 1999). The participation of regular practice of physical activities such as walking, swimming, cycling, and yoga, had a positive impact on balance control during a static balance test (Perrin et al., 1999) and is also considered an important measure against falls (O'Loughlin, Robitaille, Boivin, & Suissa, 1993). Those individuals who had regularly taken part in physical activity throughout their life performed best in measures of postural stability while those individuals who had only recently started to incorporate physical activity into their lifestyle had scores that approached those of the lifelong group (Perrin et al., 1999). At the other end of the spectrum, those individuals who were active earlier in life, but stopped their participation showed less postural stability than either of the currently active groups while those who have led a sedentary lifestyle showed the poorest performance on measures of balance (Perrin et al., 1999) and had an increased risk for falls (Campbell, Borrie, & Spears, 1989). This suggests that physical activity improves postural stability.

Even in young adults, balance is an important factor in high-performance sports because many sports are performed standing up and balance is required to be able to perform the sport (Ibuki et al., 2017; Sugiura et al., 2014). The relationship between the COM-COP and 'postural sway' is important for balance as discussed earlier (in section 1.2.4). In addition, it is well documented that physical activity and exercise help improve balance. It can be suggested that people who participate in high levels of physical activity and exercise will have better balance. For example, Ibuki et al., (2017) examined the characteristics of COM and COP fluctuations in female ballet dancers and non-ballet dancers standing quietly. Participants underwent 3 standing conditions (2-legged stance, 1 legged stance, and tiptoe stance) with their eyes open. The COM-COP displacements were calculated in the anterior-posterior and mediolateral directions. The results showed that during the one leg stance, the COP fluctuated more closely and evenly in both directions in the ballet dancers. Sugiura et al. (2014) examined static and dynamic balance in competitive swimmers and general university students by examining centre sway of foot pressure (COP displacements) for static balance and stability on an unstable stool for dynamic balance for both the legs. The authors reported that there was no difference in static balance between the two groups, but the swimmers had a better dynamic balance in both the legs when compared to the general university students. This suggests that there is a difference in dynamic balance between highly active younger adults and general younger adults. It also suggests that physical activity improves balance control.

Balance has also been seen as an indicator for injury risk amongst athletes (Mcleod, Armstrong, Miller, & Sauers, 2009). It is therefore important to identify various mechanisms that prevent injuries and improve balance. Mcleod et al. (2009) performed a study that looked to see if balance can be improved in a neuromuscular-training program in high school female basketball players. The participants participated in a 6-week program with various components such as plyometrics, strength training, balance, and stability-ball exercises. The participants completed the Balance Error Scoring System (BESS) and Star Excursion Balance Test (SEBT) before and after the 6-week exercise intervention. The results showed a significant decrease in total BESS errors at the post-test than in the pre-test. The improvements were seen in the SEBT in all four directions (lateral, anteromedial, medial, and posterior). In another study that involved soccer players, Gioftsidou et al. (2012) compared two different frequency-based balance training programs with the aim of improving proprioceptive ability. The results were similar for participants in both of the frequency-based balance groups. The participants improved their balance ability for both lower limbs similarly despite having different programs. This shows that the different frequency-based balance programs had the same outcome i.e. improving balance. Overall, these studies suggest that physical activity and exercise programs improve balance. The studies also reiterate the fact that we do not know what exercises to prescribe, how much exercise to prescribe and if exercise prescription will ever be the same between people.

1.3 Rationale

Gait and balance are important to be able to perform activities of daily living. While stability control is influenced by both age and amount of physical activity, falls remain one of the major problems associated with aging (Perrin et al., 1999; Schultz, Ashton-Miller, & Alexander, 1997). Falls in the mediolateral direction appear to be particularly devastating because of their association to hip fracture. Therefore, examining stability control in the mediolateral direction may provide considerable insight into falls. Physical activity and exercise have been seen to reduce the risk of falls in older adults but there are limited guidelines regarding the type or amount to prescribe. The proposed study will include highly active younger and older adults to better understand the influence of physical activity on the biomechanical mechanisms underlying mediolateral stability. Since the proposed research will be using new force-derived biomechanical measures of mediolateral stability, it will provide a biomechanical understanding of how physical activity and exercise influences balance control during walking. This is important because these biomechanical measures could give insight into the underlying stability control challenges that are faced by older adults. It is also important because it could lead to the development of specific physical activity interventions for older adults.

Despite our knowledge of the link between exercise and falls, we still lack the understanding of the specific mechanisms responsible for lateral balance dyscontrol among adults during walking and how such mechanisms may be modulated by physical activity. By examining the whole-body COM kinematics and the associated kinetic parameters influencing the kinematics, the proposed work could provide insight into how force-related mechanisms are modified by both age and physical activity levels. That is, how do these factors specifically alter the way in which individuals apply force to their environment to regulate mediolateral stability during walking? Ultimately, if we can determine the mechanisms by which age and physical activity and exercise influences mediolateral stability control, we can perhaps begin to design interventions that specifically augment those factors.

1.4 Aims and Objectives of the Study

The main aim of the study was to understand how ageing and physical activity influence mediolateral stability during walking. The objectives of the study were to (a) quantify mediolateral kinematic instability among younger and older adults; (b) determine the underlying force-related mechanisms related to such instability during walking; (c) quantify the influence of physical activity on mediolateral stability during walking among younger and older adults.

1.5 Hypotheses

It is hypothesized that:

(a) Across all levels of physical activity, older adults will exhibit greater instability relative to young adults. Across all age groups, individuals who are inactive will exhibit greater instability relative to individuals who are highly active. It is expected that physical activity will have a greater effect on stability in older adults compared to young adults. Instability will be evidenced by a smaller minimum distance between the whole-body centre of mass and the lateral aspect of the base of support during walking (quantified by d_{min}). Individual limbs will be analysed.

(b) Across all levels of physical activity, older adults will exhibit greater challenges in regulating the orientation of the ground reaction force during walking. Across all age groups, individuals who are inactive will exhibit greater challenges in regulating the orientation of the ground reaction force relative to individuals who are highly active. It is expected that physical activity will have a greater effect in regulating the orientation of the ground reaction force in older adults compared to young adults. Challenges in the regulating the orientation of the ground reaction force will be quantified by a smaller angle of divergence (θ_d). Individual limbs will be analysed.

(c) Overall, I expect that differences in stability measures (d_{min} and θ_d) as a function of age as well as physical activity and exercise will be accentuated when participants perform two additional walking conditions: 'as fast as possible' and modified tandem walking. Across all levels of physical activity, older adults will exhibit a smaller d_{min} and smaller θ_d in the fast and modified tandem walking conditions relative to the normal walking condition compared to young adults. Across all age groups, individuals who are inactive will exhibit a smaller d_{min} and smaller θ_d in the fast and modified tandem walking conditions relative to the normal walking condition compared to individuals who are highly active.

CHAPTER 2 - METHODS

2.1 Participants

This study recruited 14 inactive community-dwelling healthy older adults and 14 community-dwelling highly active older adults (≥ 65 years of age; 50% female for the inactive group and 57% female for the highly active group). The study also recruited 14 healthy inactive young adults and 14 healthy highly active young adults (18 to 31 years of age**;** 50% female per group). Average age, height, and weight are listed in Table 1 in the results section for all four groups.

2.1.1 Sample Size and Selection of Sample.

A sample size of 56 participants was obtained using $\alpha = 0.05$, $\beta = 0.2$, $f^2 = 0.52$ (estimated using η_p^2 from previous work (Singer et al., 2014, 2013). The alpha value is the probability of a Type I error occurring in any hypothesis test (incorrectly claiming that there is a statistical significance) while the beta value is the probability of a Type II error occurring in any hypothesis test (incorrectly concluding that there is no statistical significance) (Portney & Watkins, 2009). Cohen's f^2 measures the effect size when using methods like ANOVA (Portney & Watkins, 2009).

2.1.2 Consent

The participants provided informed consent to be able to participate in the study in accordance with ethical approval sought from the University of Manitoba Research Ethics Board and was performed per the Declaration of Helsinki (World Medical Association, 2013).

2.1.3 Inclusion and Exclusion Criteria

Participants had no diagnosed lower body anatomical, neurological impairment, or injury affecting their gait during testing within the last 6 months. Participants did not have any lower body surgery within one year. The participants were able to stand and walk without aid.

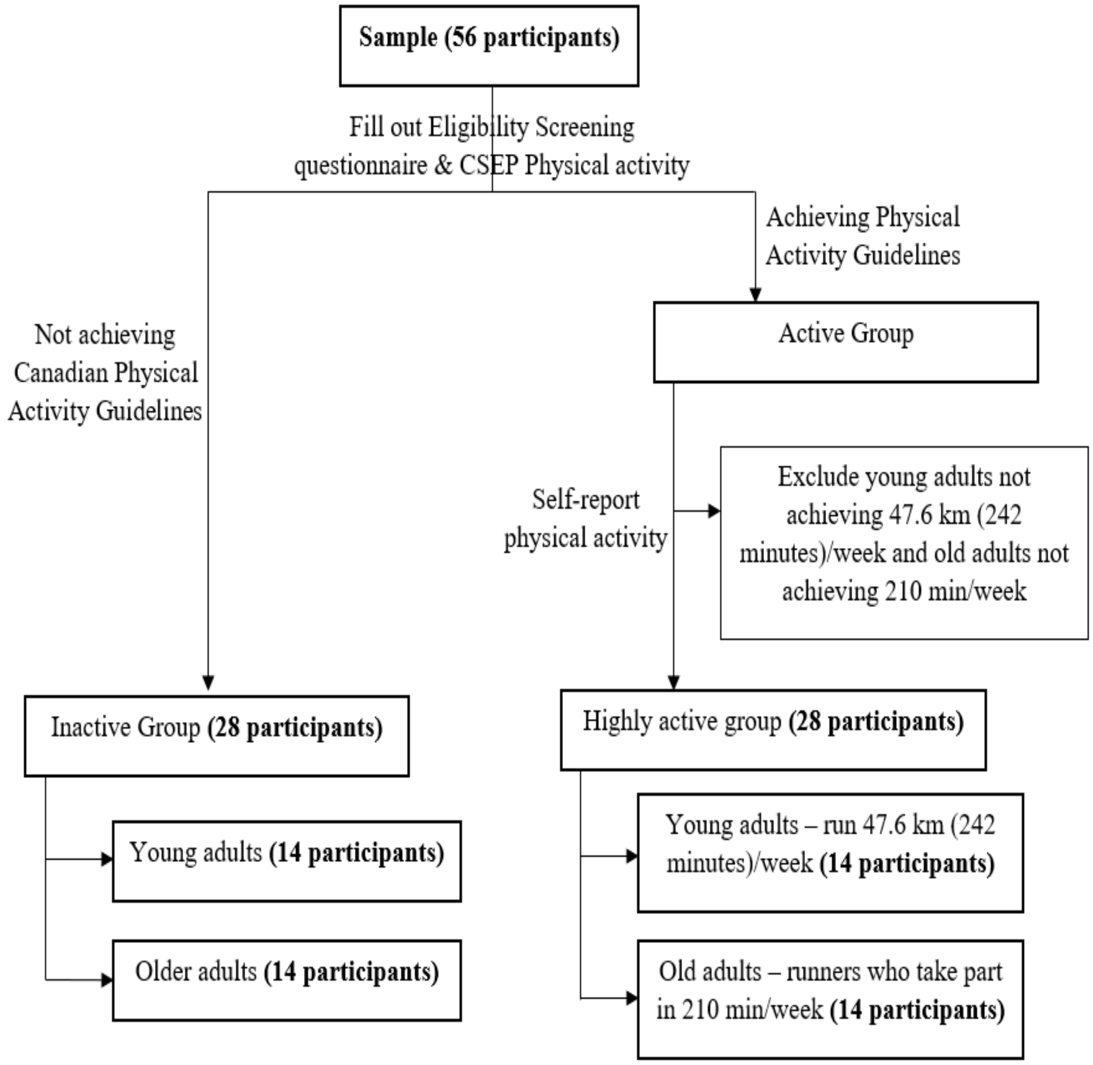
The participants had no previous history of falls during normal activities of daily living within the last six months. Participants were not taking any medication for at least a week before recruitment. Several medications have been associated with increased fall risk (de Jong, Van der Elst, & Hartholt, 2013) (see Appendix A). Participants taking the medications specified in appendix B were excluded. Participants did not have any head injuries. If they did, they had to have been completely asymptomatic for at least two weeks and be cleared by sport medicine physician or a health care practitioner that is familiar with concussion management (Echemendia et al., 2017; McCrory et al., 2017). Participants were able to understand and follow instructions to rule out cognitive impairment.

Participants were screened for falls during normal activities of daily living, age, head injuries, and medication use during initial telephone contact/meeting/email (see Appendix B). Participants were asked to fill out the physical activity and sedentary behaviour questionnaire (PASB-Q) set out by the Canadian Society for Exercise Physiology (CSEP, 2013) (see Appendix C). The PASB-Q is a valid and reliable measure for physical activity (Fowles, O'Brien, Wojcik, d'Entremont, & Shields, 2017). The eligibility screening questions were asked to see if the participants are healthy and meet the inclusion requirements. While the PASB-Q was filled out to find out the physical activity level (PAL) of the participants and differentiate between those achieving and not achieving the Canadian physical activity guidelines.

The inactive healthy younger adults' group and the inactive 'community-dwelling' healthy older adults group consisted of participants not achieving the aerobic portion of the Canadian physical activity guidelines for their respective groups. The Canadian physical activity guidelines state that adults (age 18 to 64) need to engage in at least 150 minutes of moderate to vigorous aerobic physical activity/week with bouts of 10 minutes or more, at least 2 days/week for strengthening activities, and engaging in more physical activity and strength training is better (CSEP, 2013). While for older adults (age 65+) should engage in 150 minutes of moderate to vigorous aerobic physical activity/week with bouts of 10 minutes or more, at least 2 days/week for strengthening activities, and engaging in more physical activity and strength training is better. Older adults with poor mobility should perform physical activity to enhance balance and prevent falls (CSEP, 2013)**.**

The young active participants were then asked to self-report on average how much they ran during the previous week and month. The young adult participants were required to be middle to long distance runners. Participants were considered as middle to long distance runners if they take part in 800 meters distance running or more (Martin & Coe, 1997). The younger adults were considered as 'highly active' if they are achieving at least an average of 47.6 kilometres (km) of running per week during the preparation phase (Martin & Coe, 1997). The qualifying time for the age group 18 to 34 years for the Boston marathon (42.2 km) is 3 hours 35 minutes (215 minutes). This makes the average pace for the marathon 11.78 km/h. This allows for the conversion of 47.6 km into minutes by dividing the total distance over speed. This equates to 4 hours 2 minutes (242 minutes) of running every week. The older adults were considered as 'highly active' if they took part in at least 210 minutes of moderate to vigorous physical activity per week (Peterson et al., 2009). The active participants that did not achieve the criteria for their respective groups i.e. younger and older adults were excluded (Figure 11.).

Figure 11 - Flow chart for participant recruitment



2.1.4 Recruitment

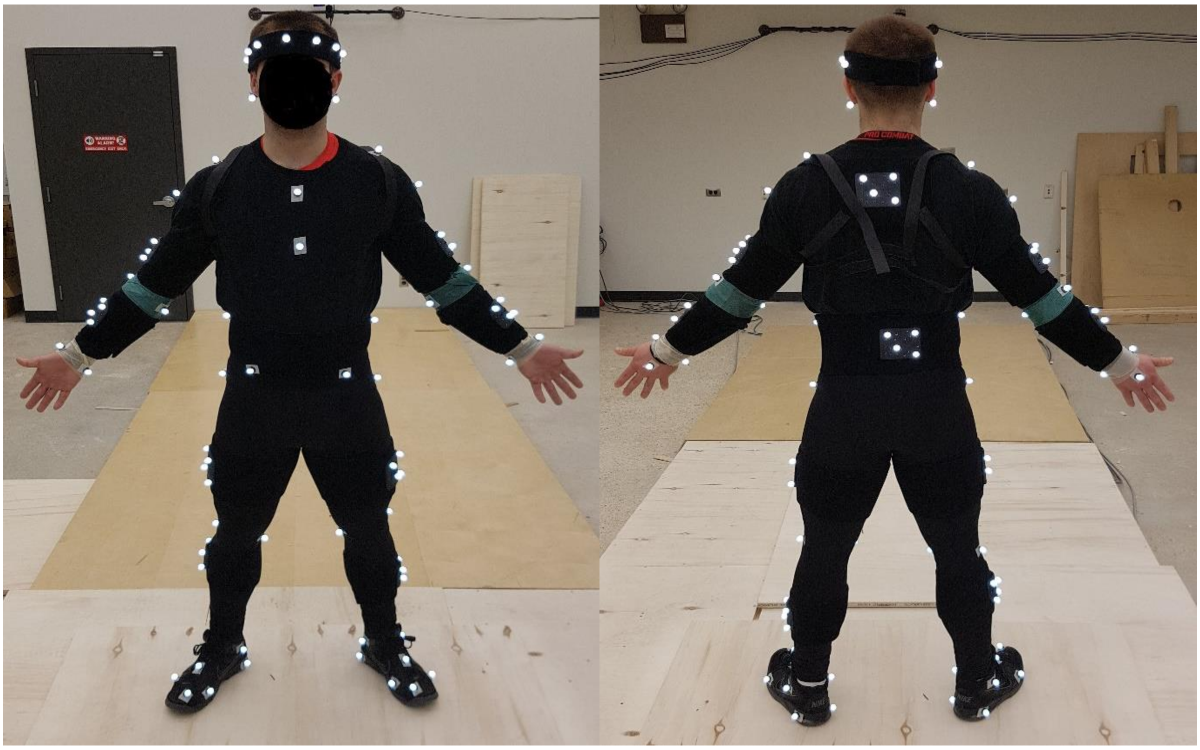
Posters were placed around campus to recruit younger adults and older adults. Emails were also sent to the various running organizations (e.g. Manitoba runners association, Winnipeg run club, running room, etc.) and sporting organizations (e.g. Sport Manitoba, Canadian Sport Centre Manitoba, Bison strength training room, Active living Centre, etc.) about the study. Older adult participants were also recruited via information obtained from the electronic database maintained by the Centre on Aging at the University of Manitoba. Participants were also recruited by word of mouth.

2.2 Instrumentation

Whole-body kinematic data were obtained using eight Vicon Vero cameras (sampling frequency = 100 Hz) synchronously obtained via a motion analysis system (Vicon Motion Systems, Los Angeles, CA). Kinematic data was used to model the position of the whole-body COM. Ground reaction force and moment components were obtained from four force platforms (Kistler, sampling frequency = 1000 Hz) embedded in a 10-meter walkway**.** Net ground reaction force (GRF) data, from the force platforms, and from each individuual limb were analysed.

One centimetre retroreflective calibration markers were placed over anatomical landmarks on the participants' upper limbs, lower limbs and pelvis (Singer et al., 2014). To define segment endpoints for the pelvis, trunk, head and upper limbs, additional calibration markers were placed bilaterally on the iliac crests, anterior superior iliac spines, acromioclavicular joints, anterior to the external auditory meatus, greater tubercles of the humerus, medial and lateral epicondyles of the humerus, radial and ulnar styloid processes and the head of the 3rd metacarpal. Rigid clusters containing four markers were placed on the sacrum and trunk, and bilaterally on the upper arms, lower arms, lower legs, and thighs (Figure 12.). The purpose of the calibration markers was to define the local coordinate system for each segment– these were removed following subject calibration. Tracking markers have a fixed position within the local coordinate system and remain on the participant during experimental trials – these track the exact position and orientation of the segmental local coordinate system.

Figure 12 - Complete marker set, consisting of both calibration and tracking markers



2.3 Procedures

Participants were asked about which leg they considered as their dominant leg (determined by which leg the participant would use to kick a ball (King et al., 2012)), aerobic activities and miles/kilometres (minutes) ran in a typical week for the highly active participants prior to the start of the study. The participants took part in three walking conditions. They walked (a) at their typical speed (normal walking (NW)), (b) 'as fast as possible' without running (fast walking (FW)), and (c) modified tandem walking (TW) across a 10-meter platform. Performing the FW and TW conditions was done to present a challenge to the participants’ gait stability (Ko, Hausdorff, & Ferrucci, 2010; Matthew A. Schrager, Kelly, Price, Ferrucci, & Shumway-Cook, 2008).

Each condition consisted of ten trials. Any mistrials that occurred in each condition were removed and an additional trial was performed so as to have consistency amongst the three conditions for all the participants. A trial was considered as a mistrial if: a) the participant partially stepped on the force plate. b) If they did not perform the condition correctly, for example, a participant did not take a narrow step during the TW condition or not walking as fast as they can during the FW condition. c) If a marker fell off or changed position during the walking trial. The walking conditions were counter-balanced between the participants. For example, for one participant, the NW condition was performed first followed by the FW and TW conditions. For another participant, the FW condition was performed first followed by the NW and TW conditions. The participants were allowed to rest if needed for as long as they wanted to between trials or conditions. Participants were allowed to practice each walking condition so that they felt comfortable when performing the conditions and to ensure that they were able to make full contact on the force plates.

A kinematic measure of instability was quantified by the smallest lateral distance between the whole-body COM and the side of the support envelope (area bound by the feet) (Singer et al., 2014, 2013) during gait (d_{min}). The biomechanical, force-related, mechanisms underlying such instability was quantified by examining how the GRF was directed, relative to the whole-body COM, within the frontal plane. The relative orientation of the net GRF has been shown to be directly related to the movement of the centre of mass and subsequent instability during stepping (θ_d) (Singer et al., 2014, 2013).

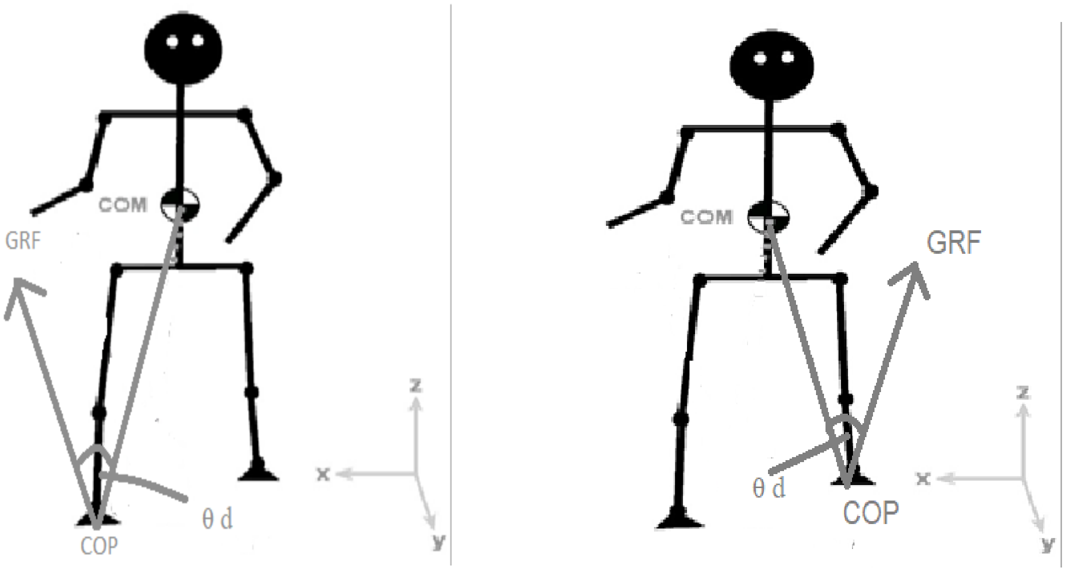
2.4 Data Analysis

The kinematic and kinetic data were processed and analysed using Visual 3D (C-Motion, MD, USA Version 6) and Matlab (Mathworks, MA, USA Version 9.1 - R2016b). The kinematic and kinetic data were captured using a motion capture camera and force plates, respectively. The data were then exported to Visual 3D. In visual 3D, a series of scripts were implemented: first, the creation of a model and the application of the model to all of the movement trials of the participants was run. The entire body was modelled as a rigid system of individual segments. The segment masses for older adults were estimated using parameters and segment COM positions using the geometrical model proposed by Dempster (1955) and Hanavan (1964) (cited in Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2013). While the segment masses for younger adults were estimated using Zatsiorsky, Seluyanov, and Chugunova (1990) with modifications by (de Leva, 2016). This was done because (Zatsiorsky et al.'s (1990) segment parameters were obtained using regression equations that are not defined relative to segment endpoints and thus (de Leva, 2016) modified Zatsiorsky et al.'s (1990) equations so as to make the model congruent with typical biomechanical models consisting of segments modelled by their endpoints.

Force platform data were then low pass filtered using a zero-lag, fourth-order, Butterworth filter with a cut-off frequency of 15 Hz. That specific cut-off frequency was used because it has been used in a previous study by Singer et al. (2014). Kinematic data was low pass filtered using a zero-lag, twentieth-order critically damped filter, with a cut-off frequency of 6Hz (used in a previous study by Singer et al. (2014)).

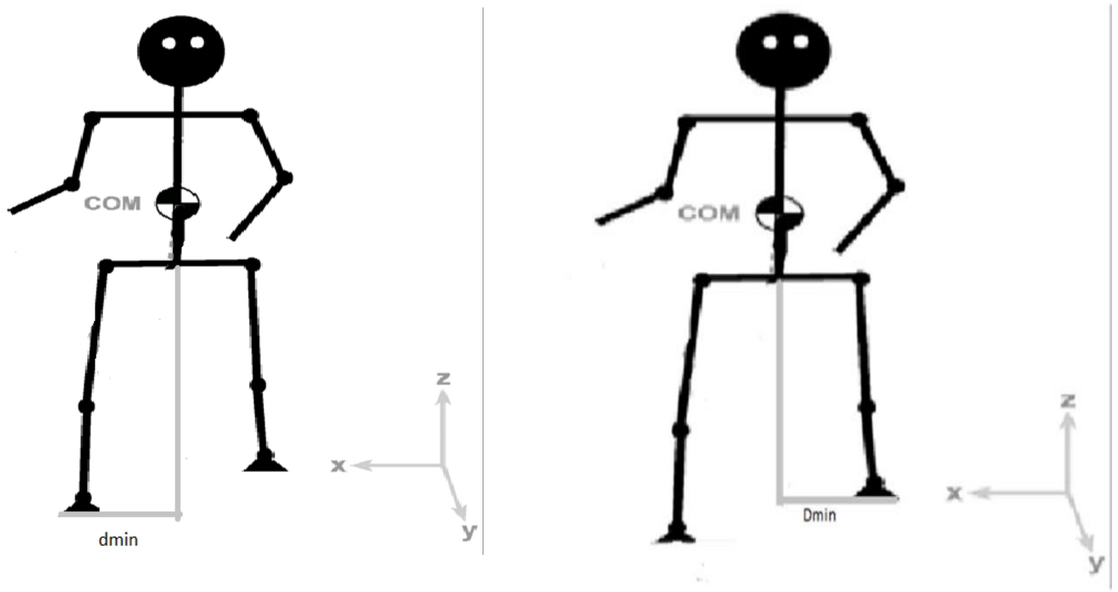
The creation of the inclination angle of the GRF and a line joining the COP and COM in the frontal plane was ran. The filtered ground reaction force data from all force platforms was combined to yield the ground reaction force and centre of pressure (i.e. the point of application of the ground reaction force). The inclination angle of 1) the ground reaction force and 2) a line joining the centre of pressure and centre of mass was calculated with respect to the mediolateral axis of the global coordinate system (Figure 13.). The angle of divergence (θ_d) was calculated as the difference between these two waveforms.

Figure 13 - Angle of divergence (θ_d) between the ground reaction force vector (GRF) and of a line joining the centre of pressure (COP) and whole-body COM for the right and left limb. **(J. Singer, personal communication, December 6, 2016)**



A script, in Visual 3D, for the identification of the heel strike and toe off for the individual limbs was run. This allowed for identification of the minimum lateral distance (d_{min}) because it allowed for the calculation of the difference between the mediolateral coordinates of the COM (at its peak lateral position) and the lateral border of the lead limb (head of the 5th metatarsal) during the stance phase (Figure 14.). The d_{min} for the left limb, right limb and the average of the two limbs (composite d_{min}) was calculated. The composite d_{min} is related the net θ_d since the net θ_d is a measure of θ_d over one stride (i.e. one right and one left step). The results from Visual 3D provided information on d_{min} for the left limb, right limbs for each trial for each walking condition. The results were exported to an excel file. The variables were averaged for each participant and walking condition and were entered into SPSS for analysis. The composite d_{min} was calculated in SPSS by averaging the d_{min} for the left and right limb for each participant and walking condition.

***Figure 14 - The Minimum Lateral Distance (d_{min}) between the whole-body centre of mass (COM) and lateral aspect of the BOS for the right and left limb* (J. Singer, personal communication, December 6, 2016)**



The absolute mean difference (AMD) of the θ_d represents the extent of eccentricity of the GRF relative to the COM. AMD is defined 'as the average of the absolute differences of all pairs of values in a population" (Glasser, 1962, p. 648). It is calculated as $\frac{1}{N-1}\sum_{j=1}^{N}|X_j - X_i|$ or $|X_i - X_j|$ (Glasser, 1962). For the current study, X_i is the perfectly centric GRF orientation and X_j is the actual eccentricity of the GRF vector. The AMD was calculated as the absolute difference from zero (a perfectly centric GRF orientation) and the actual eccentricity of the GRF vector (regardless of polarity) over the stance phase. The magnitude of the θ_d for the GRF_{net} and the GRF beneath the left and right limbs separately were extracted for analysis. The AMD of the θ_d of the GRF_{net} was extracted for one stride, whereas the AMD of the θ_d of the GRF of the left and right limbs were exported for one step.

The waveforms for the left, right and the net θ_d were exported to Matlab. Finally, the code in Matlab provided the AMD value of the waveform for the individual limbs and the net limb value. The AMD for the positive and negative phases for the left and right limb were extracted from the waveforms for every trial and from the three walking conditions. The positive and negative values were analysed separately. The average of the positive phase, negative phase and the AMD value of the θ_d for the left limb, right limb, and the net value of θ_d for the three walking conditions, for each participant, were entered into SPSS.

A secondary analysis of AMD of the θ_d was also calculated independently for the positive and negative regions of the θ_d waveform. When a force is applied at the COM of an object, the force will cause a linear motion of the object. However, if the force is not applied at the COM of the object (eccentric force), the force will cause a linear and angular motion of the object. Thus, in the current study, the positive phase AMD values are believed to signify a GRF orientation tending to cause restabilizing angular acceleration because it redirects the COM back within the BOS. While the negative phase AMD values are believed to signify a GRF orientation tending to cause destabilizing angular acceleration because it suggests that the COM is redirected away from the BOS. While the secondary analysis of the spatial-temporal variables was calculated based of the heel strikes of the individual limbs. Step length was measured as the distance between the feet at successive heel strikes. Stride length was measured as the sum of two subsequent step lengths. Step width was measured as the distance between the feet at heel strike. Composite step width was calculated by averaging the the step width for the left and right limb. Stance time was calculated as the duration the stance leg was on the ground (i.e. heel strike to toe off). Stride time was calculated as the time to complete the stride length.

Upon visually inspecting the data, the kinematic and kinetic values for the individual limb were different. The exploratory analysis was performed to see if the individual limb values were statistically different from each other. The exploratory analysis of the differences in the kinetic measures and kinematic measure between the left and right limbs between age groups were performed in SPSS.

Figure 15 - Representative ensemble average θ_d for all NW trials from a single participant. Right limb (Black) and left limb (Blue). Positive values signify a GRF orientation tending to cause restabilization. Negative values signify a GRF orientation tending to cause destabilization

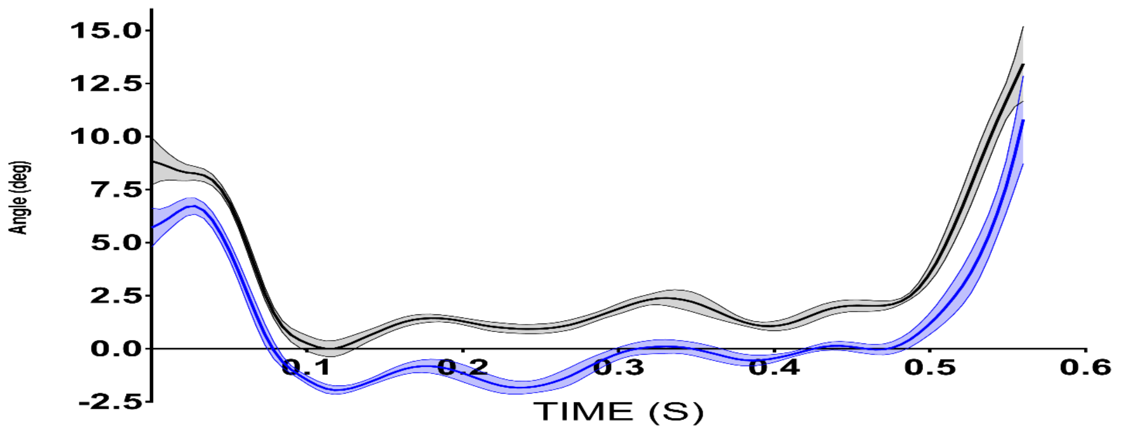
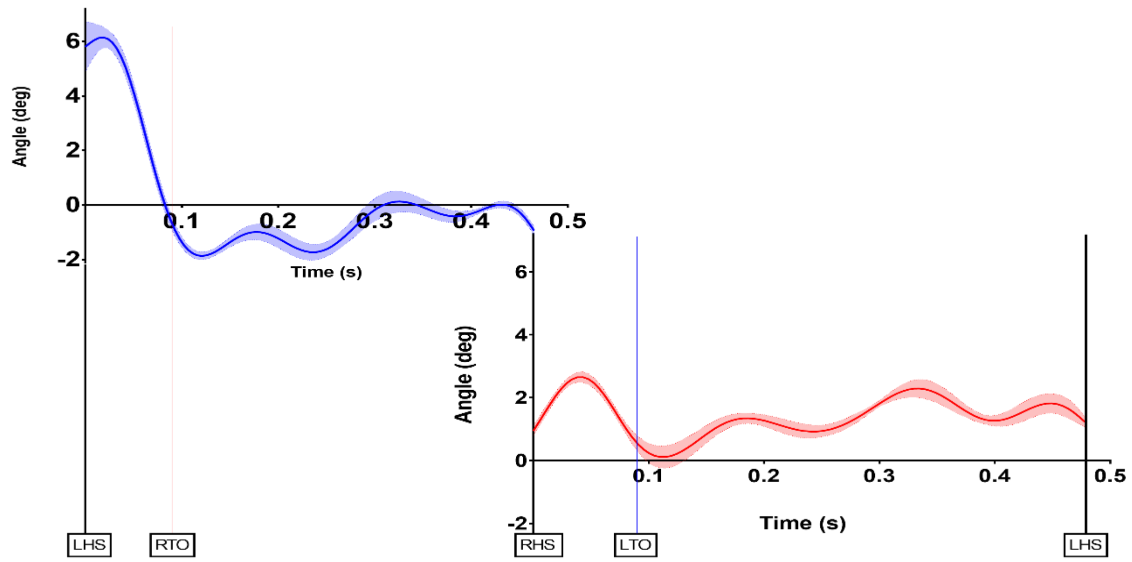


Figure 16 - Representative ensemble average net θ_d for all NW trials from a single participant. Top: θ_d from the left heel strike to right heel strike. Bottom: θ_d from the right heel strike to left heel strike. Positive values signify a GRF_{net} orientation tending to cause restabilization. Negative values signify a GRF_{net} orientation tending to cause destabilization



***Polarity of top (blue) time series has been reversed to maintain these conventions.*

2.5 Statistical Analysis

2.5.1 Primary Analysis

This study looked at the effect of amount of physical activity in young and older adults. There are three independent variables: the experimental condition, age, and amount of physical activity. The experimental condition has three categories: normal walking, fast walking and modified tandem walking; while age has two categories: young adults and older adults and level of physical activity has two levels: inactive and highly active. The within subject factor is the walking conditions while the between subject factors are age and physical activity. The dependent variables include the AMD of the θ_d along with d_{min}. The composite d_{min} and the d_{min} for the individual limbs as well as the AMD of the net θ_d along with the AMD of the θ_d values for the individual limbs were averaged within subject for each experimental condition.

A three-way ANOVA with repeated measures was used, initially, to examine the main effects and interactions for each variable (composite d_{min}, net θ_d, along with the d_{min} and θ_d for each limb). More specifically, an age by physical activity with repeated measures on the gait condition was used to examine the main effects and interactions for each variable**.** Significant interactions were analysed with additional ANOVA's, or independent and paired t-tests, as required. Significant main effects of gait condition were analysed with paired sample t-test. A Bonferroni correction (an adjustment made to the p-value for the number of contrasts being performed within a data set) was used to protect against a type 1 error (Portney & Watkins, 2009). To perform this correction, the critical P value (alpha) was divided by the number of comparisons being made within that family (Portney & Watkins, 2009). All statistical analyses were run in SPSS. Mauchly's test of sphericity was used. If the assumption of Mauchly's test was violated, the Greenhouse-Geisser test was used. The initial p-value for the ANOVA model was set at $p < 0.05$. Post hoc alpha level for most follow up comparisons was set as $p < 0.017$ unless otherwise specified. Cohen's d was used to measure the effect size in the follow-up t-tests. Only the main effects and interactions that were significant were reported unless otherwise specified.

2.5.2 Secondary Analysis

The positive and negative AMD values for the individual limbs were averaged within subject for each experimental condition. Since the initial analysis were composite measures, including both positive (believed to be restabilizing) and negative (believed to be destabilizing) effects of the GRF, the secondary analysis was carried out to gain further insight on the specific phases of the kinetic measure (θ_d) and how it could affect kinematic stability for the individual limbs (Table 7 and 8.).

There could have been age-related and walking condition-related differences in the spatial-temporal parameters, which may have affected the measures of kinematic stability. The spatial-temporal parameters were subjected to secondary analysis (Table 9, 10 and 11.). Secondary statistical analyses were performed using the ANOVA model and follow-up analyses detailed above. Only the main effects and interactions that were significant were reported unless otherwise specified. All statistical analyses were performed using the follow-up analyses detailed above.

2.5.3 Exploratory Analysis

A statistical comparison was also performed to see if there were any differences between the left and right limbs in the kinetic measure (including the positive and negative phases of the θ_d) and the measure of kinematic stability (d_{min}). Exploratory analyses were performed to assess age- and physical activity- related differences in the variables during the gait cycle for each limb. A three-way ANOVA with repeated measures was used, initially, to examine the main effects and interactions for each variable (d_{min} and θ_d for each limb). More specifically, an age by physical activity with repeated measures on the limb was used to examine the main effects and interactions for each variable during the normal walking condition only. Only the main effects and interactions that were significant were reported unless otherwise specified.

CHAPTER 3 – RESULTS

3.1 Participant Characteristics

There were 56 participants in the study. All the participants managed to complete the walking conditions. No participants fell during any trials of the three walking conditions. All participants indicated that they were right leg dominant.

Table 1 - Participant characteristics for Inactive Young Adults and Older Adults

	Inactive Young Adults		Inactive Older Adults	
Mean (SD)	**Males**	**Females**	**Males**	**Females**
N	7	7	6	8
Age (years)	24.00 (2.89)	22.70 (1.70)	73.83 (6.88)	73.88 (7.36)
Height (m)	1.84 (0.08)	1.65 (0.07)	1.77 (0.07)	1.61 (0.08)
Weight (kg)	79.20 (11.5)	61.8 (8.87)	81.31 (3.73)	63.28 (12.30)

Table 2 - Participant characteristics for Highly Active Young Adults and Older Adults

	Highly Active Young Adults		Highly Active Older Adults	
Mean (SD)	**Males**	**Females**	**Males**	**Females**
N	7	7	7	7
Age (years)	21.60 (3.31)	22.70 (4.54)	69.90 (2.34)	67.4 (3.55)
Height (m)	1.78 (0.07)	1.66 (0.05)	1.77 (0.09)	1.65 (0.08)
Weight (kg)	65.6 (7.29)	55.5 (6.79)	78.7 (7.91)	62.9 (9.38)

Table 3 - Self-reported average number of minutes of aerobic activities at a moderate to vigorous intensity per week for each group.

Group	Time (Mean(SD)) (minutes)
Highly active young male	504 (168)
Inactive young male	64 (39)
Highly active young female	301 (44)
Inactive young female	92 (51)
Highly active older male	373 (131)
Inactive older male	117 (18)
Highly active older female	276 (58)
Inactive older female	75 (54)

3.2 Primary Analysis

3.2.1 Minimum Lateral Distance (D_{min})

3.2.1.1 Composite d_{min}

The results showed that there was significant main effect of gait condition (F (1.73, 90.03) = 433.96, $p < 0.001$, $\eta_p{}^2 = .893$), sphericity violated (Table 4.). Results of the follow-up paired t-tests revealed that there was a significantly greater composite d_{min} for the NW than the TW condition; t (55) = 22.88, p < 0.001, d =3.40. There was also a significantly greater composite d_{min} for the FW than the TW condition; t (55) = 21.81, p < 0.001, d = 3.98. There was no significant difference between the composite d_{min} for NW and the FW condition.

In addition, there was a significant main effect of age (F (1, 52) = 7.03, $p = 0.011$, $\eta_p{}^2 = .119$) with OA having a greater composite d_{min} (M=0.099, SD=0.011) compared to YA (M=0.090, SD=0.013).

The main effects were qualified by an interaction of gait condition and the physical activity level (F (1.73, 90.03) = 3.94, $p = .028$, $\eta_p{}^2 = .070$), sphericity violated (Table 6.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the composite d_{min} between the highly active and inactive groups. The results of the paired sample t-test performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (HA; t (27) = 18.06, p < 0.001, d =4.65, IA; t (27) = 16.05, p < 0.001, d = 2.65). Specifically for FW and TW comparisons (HA; t (27) = 16.10, p < 0.001, IA; t (27) = 16.05, p < 0.001, d = 3.63).

3.2.1.2 d_{min} Left

The results showed that there was significant main effect of gait condition, (F (2,104) = 241.90, $p < 0.001$, $\eta_p{}^2 = .823$) (Table 4.). The results of the follow-up paired sample t-test revealed that there was a significantly greater d_{min} in the NW than the TW condition; t (55) = 19.39, p < 0.001, d =3.08. There was also a significantly greater d_{min} in the FW than the TW condition; t (55) = 17.16, p < 0.001, d =3.17. There was no significant difference in d_{min} between NW and FW condition.

In addition, there was a significant main effect of age (F (1, 52) = 6.14, p = 0.017, η_p^2 = .106) with OA having a greater d_{min} (M=0.090, SD=0.012) compared to YA (M=0.082, SD=0.012).

3.2.1.3 d_{min} Right

The results showed that there was significant main effect of gait condition (F (2, 104) = 321.39, p < 0.001, η_p^2 = .861) (Table 4.). The results of the follow-up paired sample t-test revealed that there was a significantly greater d_{min} in the NW than the TW condition; t (55) = 20.68, p < 0.001, d = 3.17. There was also a significantly greater d_{min} in the FW than the TW condition; t (55) = 19.48, p < 0.001, d = 3.59. There was no significant difference in d_{min} between NW and the FW condition.

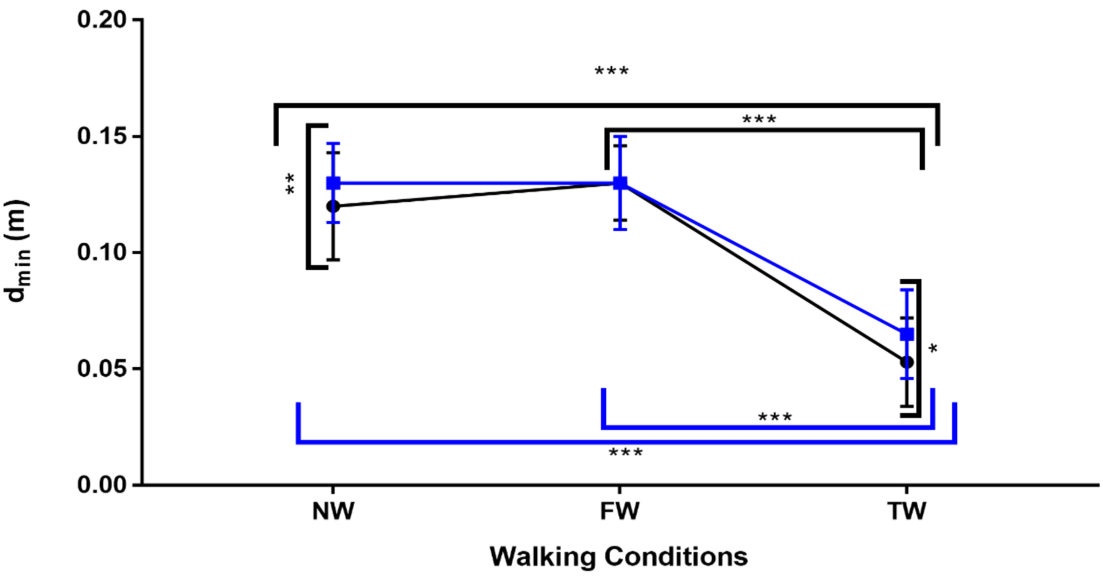
In addition, there was a significant main effect of age (F (1, 52) = 5.74, p = 0.020, η_p^2 = .099) with OA having a greater d_{min} (M=0.11, SD=0.014) compared to YA (M=0.099, SD=0.013).

The main effects were qualified by an interaction of gait condition with age and physical activity level (F (2, 104) = 4.01 p = 0.021, η_p^2 = .072). The interaction was further investigated with three two-way univariate ANOVA's, one for each gait condition. In the NW condition, there was a significant main effect of age (F (1, 52) = 7.30, p = 0.009, η_p^2 = .123) with OA having greater d_{min} (M=0.13, SD=0.017) compared to YA (M=0.12, SD=0.023). In the FW condition, there was no significant interaction of age by PAL interaction. In the TW condition, there was a trend towards a main effect of age (F (1, 52) = 5.82, p = 0.019, η_p^2 = .123) with OA having a greater d_{min} (M=0.065, SD=0.019) compared to YA (M=0.053, SD=0.019). Note: the p value approaches significance for follow-up comparison for the TW condition.

There was also a two-way interaction between gait condition and age ($F(2, 104) = 3.43$, $p = .036$, $\eta_p^2 = .062$) (Table 5.). The results of the follow-up independent sample t-test revealed that in the NW condition, there was a significantly greater d_{min} for OA than YA participants; $t(54) = -2.73$, $p = 0.009$, $d = 0.72$. There was no significant differences in d_{min} in the FW condition between the YA and OA participants. There was a trend to a greater d_{min} in the TW condition for OA than YA participants; $t(54) = -2.40$, $p = 0.020$, $d = 0.01$. Note: the p value approaches significance for follow-up comparison for the TW condition.

The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (OA; $t(27) = 15.93$, $p < 0.001$, $d = 3.69$, YA; $t(27) = 13.34$, $p < 0.001$, $d = 3.03$). Specifically for FW and TW comparisons (OA; $t(27) = 11.07$, $p < 0.001$, $d = 3.13$, YA; $t(27) = 18.74$, $p < 0.001$, $d = 4.25$).

Figure 17 – d_{min} for the Right Limb across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



***: significant difference at P < 0.05**: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

3.2.2 Angle of Divergence (θ_d)

3.2.2.1 Net θ_d

The results showed that there was significant main effect of gait condition (F (1.67, 86.99) = 45.57, $p < 0.001$, $\eta_p^2 = .47$), sphericity violated (Table 4.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the net θ_d for the NW and the TW condition. There was a significantly greater net θ_d for the FW than the net θ_d TW condition; t (55) = 7.37, $p < 0.001$, $d = 1.26$). There was also a significantly greater net θ_d for the FW than the NW condition; t (55) = -7.62**,** $p < 0.001$, $d = 1.23$.

In addition, there was a main effect of age (F (1, 52) = 4.02, $p = 0.050$, $\eta_p^2 = .44$) with OA having a smaller net θ_d (M=1.77, SD=0.16) compared to YA (M=1.87, SD=0.20). Note: the p value approaches significance.

3.2.2.2 Left Leg θ_d

The results showed that there was significant main effect of gait condition (F (1.67, 86.57) = 67.55, $p < .001$, $\eta_p^2 = .565$), sphericity violated (Table 4.). The results of the follow-up paired sample t-tests revealed that there was a significantly greater θ_d for the NW than the TW condition; t (55) = 4.79, $p < 0.001$, $d = 0.86$. There was also a significantly greater θ_d for the FW than the TW condition; t (55) = 11.27, $p < 0.001$, $d = 1.78$. There was also a significantly greater θ_d for the FW than the NW condition; t (55) = -6.38, $p < 0.001$, $d = 1.23$.

3.2.2.3 Right Leg θ_d

The results showed that there was significant main effect of gait condition (F (2, 104) = 4.73, $p = 0.011$, $\eta_p^2 = .083$) (Table 4.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the θ_d for the NW and the TW conditions. There was a significantly greater θ_d for the FW than the TW conditions; t (55) =

2.50, p = 0.016, d = 0.41. There was also a significantly greater θ_d for the FW than the NW conditions; t (55) = -2.74, p = 0.008, d = 0.42.

In addition, there was a significant main effect of age (F (1, 52) = 8.64, p = 0.005, $\eta_p{}^2$ = .44) with OA having a smaller θ_d of the right leg (M=3.00, SD=0.30) compared to YA (M=3.25, SD=0.34).

Table 4 – Comparison of the Primary Variables across the Gait Conditions

Variables (Mean (SD))	NW	FW	TW
Composite d_{min} ^{2,3}	0.11 (0.018)	0.12 (0.014)	0.053 (0.018)
d_{min} Left ^{2,3}	0.10 (0.019)	0.11 (0.019)	0.047 (0.019)
d_{min} Right ^{2,3}	0.12 (0.021)	0.13 (0.018)	0.059 (0.020)
Net θ_d ^{1,3}	1.71 (0.20)	2.05 (0.34)	1.68 (0.35)
Left Limb θ_d ^{1,2,3}	1.77 (0.25)	2.23 (0.47)	1.54 (0.28)
Right Limb θ_d ^{1,3}	3.06 (0.46)	3.25 (0.50)	3.06 (0.43)

1 - Denotes a significant difference between NW and FW conditions

2 - Denotes a significant difference between NW and TW conditions

3 - Denotes a significant difference between FW and TW conditions

Table 5 – Comparison across Age groups stratified by Gait Condition collapsed across Physical Activity Level

Variables Mean (SD)	Young Adults			Older Adults		
	NW	**FW**	**TW**	**NW**	**FW**	**TW**
Composite d_{min} ^{AA}	0.12 (0.012)	0.12 (0.014)	0.050 (0.017)	0.11 (0.022)	0.12 (0.014)	0.055 (0.018)
d_{min} left ^{AA}	0.10 (0.021)	0.10 (0.020)	0.039 (0.020)	0.11 (0.017)	0.11 (0.018)	0.055 (0.013)
d_{min} right ^{AA} ^{1,3}	0.12 (0.023)	0.13 (0.016)	0.053 (0.019)	0.13 (0.017)	0.13 (0.021)	0.065 (0.019)
Net θ_d ^{AA}	1.72 (0.21)	2.14 (0.36)	1.73 (0.25)	1.70 (0.19)	1.96 (0.28)	1.64 (0.23)
Left limb θ_d	1.75 (0.25)	2.36 (0.51)	1.56 (0.29)	1.78 (0.25)	2.10 (0.40)	1.53 (0.27)
Right limb θ_d ^{AA}	3.22 (0.40)	3.36 (0.55)	3.16 (0.47)	2.89 (0.47)	3.15 (0.41)	2.96 (0.38)

AA- Main effect of age

1 - Denotes a significant difference between YA and OA in the NW condition

2 - Denotes a significant difference between YA and OA in the FW condition

3 - Denotes a significant difference between YA and OA in the TW condition

Table 6 – Comparison of Composite d_{min} across the Gait Conditions between the Highly Active and Inactive Group collapsed across Age groups

Variables (Mean (SD))	NW	FW	TW
HA Composite d_{min} (m) ^{2,3}	0.12 (0.012)	0.12 (0.014)	0.059 (0.017)
IA Composite d_{min} (m) ^{2,3}	0.11 (0.022)	0.12 (0.014)	0.056 (0.018)

1 - Denotes a significant difference between NW and FW conditions

2 - Denotes a significant difference between NW and TW conditions

3 - Denotes a significant difference between FW and TW conditions

3.3 Secondary Analysis

3.3.1 Individual Limb Analysis

3.3.1.1 Angle of Divergence (θ_d) for the Left Leg

3.3.1.1.1 Positive Phase of the θ_d

The results showed that there was significant main effect of gait condition ($F(2, 104) = 40.08$, $p < 0.001$, $\eta_p^2 = .44$) (Table 7.). The results of the follow-up paired sample t-test revealed that there was a significantly more positive θ_d for the NW than the TW conditions; $t(55) = 8.52$, $p < 0.001$, $d = 1.40$. There was also a significantly more positive θ_d for the FW than the TW conditions; $t(55) = 7.92$, $p < 0.001$, $d = 1.18$. There was no significant difference between the positive θ_d for the NW and the FW conditions.

3.3.1.1.2 Negative Phase of the θ_d

The results showed that there was significant main effect of gait condition ($F(1.46, 75.70) = 37.58$, $p < 0.001$, $\eta_p^2 = .419$), sphericity violated (Table 7.). The results of the follow-up paired sample t-test revealed that there was a significantly more negative θ_d for the FW than the NW conditions; $t(55) = -6.81$, $p < 0.001$, $d = 1.23$. There was also a significantly more negative θ_d for the FW than the TW conditions; $t(55) = 6.24$, $p < 0.001$, $d = 1.11$. There was no significant difference between the negative θ_d for the NW and the TW conditions.

3.3.1.2 Angle of Divergence (θ_d) for the Right Leg

3.3.1.2.1 Positive Phase of the θ_d

The results showed that there was significant main effect of the gait condition (F (2, 104) = 15.65, p < 0.001, $\eta_p{}^2$ = .231) (Table 7.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the positive θ_d for the NW and the TW conditions. There was a significantly more positive θ_d for the FW than the TW conditions; t (55) = 4.78, p < 0.001, d = 0.78. There was also a significantly more positive θ_d for the FW than the NW conditions; t (55) = -4.66, p < 0.001, d = 0.72.

In addition, there was a significant main effect of age (F (1, 52) = 10.93, p = 0.002, $\eta_p{}^2$ = .17) with OA having a smaller (less positive) θ_d of the right leg (M=3.07, SD=0.30) compared to YA (M=3.35, SD=0.34).

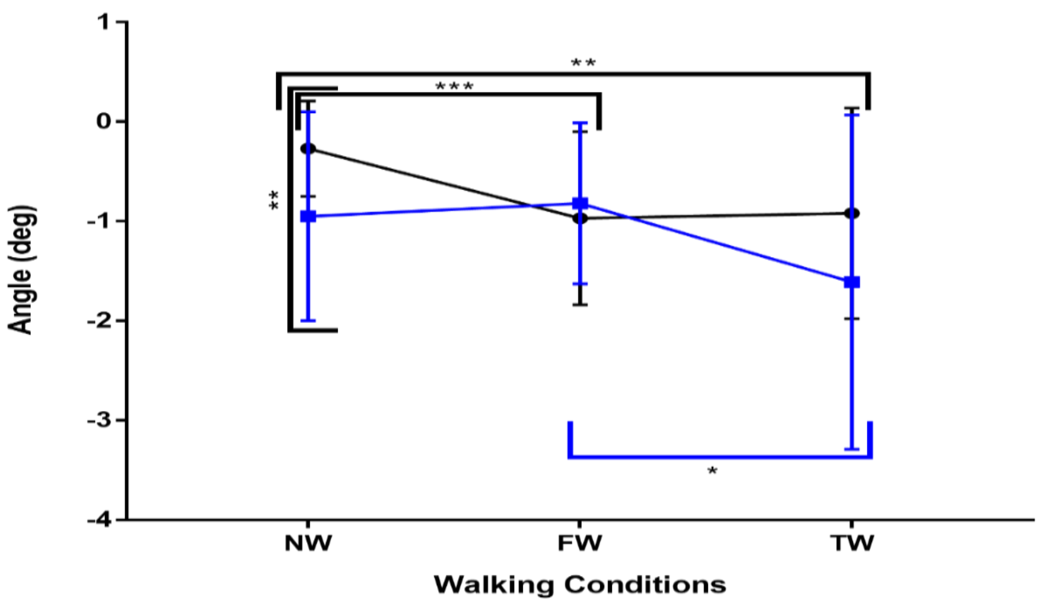
3.3.1.2.2 Negative Phase of the θ_d

The results showed that there was significant main effect of gait condition (F (1.65, 85.89) = 7.64, p = .002, $\eta_p{}^2$ = .13), sphericity violated (Table 10.). The results of the follow-up paired sample t-test revealed that there was a significantly more negative θ_d for the TW than the NW conditions; t (55) = -3.55, p = 0.001, d = 0.55. There was no significant difference between the negative θ_d for the FW and the TW conditions and between the negative θ_d for the NW and the FW conditions.

The main effects were qualified by an interaction of gait condition and age (F (1.65, 85.89) = 4.15, p = .025, $\eta_p{}^2$ = .0074), sphericity violated (Table 8.). The results of the follow-up independent sample t-test revealed that there was a significantly more negative θ_d in the NW for the OA than YA participants; t (37.83) = -3.138, p = 0.003, d = 0.84. There was no significant difference in the negative θ_d in the FW and TW between YA and OA participants.

The results of the paired sample t-test, in the young adult category, revealed that there was a significantly more negative θ_d in the TW than the NW conditions; t (27) = -3.22, p = 0.003, d = 0.79. There was no significant difference between the negative θ_d for the FW and the TW conditions. There was a significantly more negative θ_d in the FW than the NW conditions; t (27) = -4.12, p < 0.001, d = 1.00. The results of the paired sample t-test, in the old adult category, revealed that there was no significant difference between the negative θ_d for the NW and TW conditions. There was a significantly more negative θ_d in the TW than the FW conditions; t (27) = -2.72, p = 0.011, d = 0.60. There was also no significant difference between the negative θ_d for the NW and the FW conditions.

Figure 18 – Negative Phase of the θ_d for the Right Leg across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



***: significant difference at P < 0.05, **: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

Table 7 – Comparison of the Positive and Negative Phases of the θ_d across the Gait Conditions for the Individual limbs

Variables (Mean (SD))	NW	FW	TW
Positive phase Left Limb θ_d ^{2,3}	2.96(0.65)	2.86(0.68)	2.13(0.54)
Negative phase Left Limb θ_d ^{1,3}	0.93(0.31)	1.56(0.66)	0.98(0.35)
Positive phase Right Limb θ_d ^{1,3}	3.11(0.45)	3.46(0.52)	3.08(0.45)
Negative phase Right Limb θ_d ^{2}	0.61(0.88)	0.90(0.84)	1.27(1.44)

1 - Denotes a significant difference between NW and FW conditions

2 - Denotes a significant difference between NW and TW conditions

3 - Denotes a significant difference between FW and TW conditions

Table 8 – Comparison of the Positive and Negative Phases of the θ_d for the Individual limbs between Age Groups collapsed across Physical Activity Level

Variables Mean (SD)	Young Adults			Older Adults		
	NW	**FW**	**TW**	**NW**	**FW**	**TW**
Positive phase Left Limb θ_d	2.98 (0.64)	2.89 (0.73)	2.08 (0.46)	2.94 (0.66)	2.83 (0.64)	2.19 (0.61)
Negative phase Left Limb θ_d	0.87 (0.25)	1.67 (0.68)	0.93 (0.40)	0.99 (0.36)	1.44 (0.64)	1.02 (0.29)
Positive phase Right Limb θ_d $^{AA\ 1}$	3.26 (0.40)	3.61 (0.56)	3.19 (0.47)	2.95 (0.45)	3.31 (0.45)	2.97 (0.40)
Negative phase Right Limb θ_d $^{AA\ 1}$	0.27 (0.48)	0.97 (0.87)	0.92 (1.06)	0.95 (1.05)	0.82 (0.81)	1.61 (1.68)

AA- Main effect of age

1 - Denotes a significant difference between YA and OA in the NW condition

2 - Denotes a significant difference between YA and OA in the FW condition

3 - Denotes a significant difference between YA and OA in the TW condition

3.3.2 Spatial-temporal Parameter Characteristics

A three-way ANOVA with repeated measures (age by physical activity with repeated measures on the gait condition) was performed for the left step length, right step length, left step width, composite step width, left stance time, right stance time, stride length, stride time and stride velocity. This analysis was done to see if the differences seen in stability related outcome variables could be explained by the temporal-spatial modifications to gait. The only spatial-temporal parameter believed to directly influence the outcome variables (d_{min} and θ_d) were the step width variables. The results of all remaining spatial-temporal parameters appear in the appendix D.

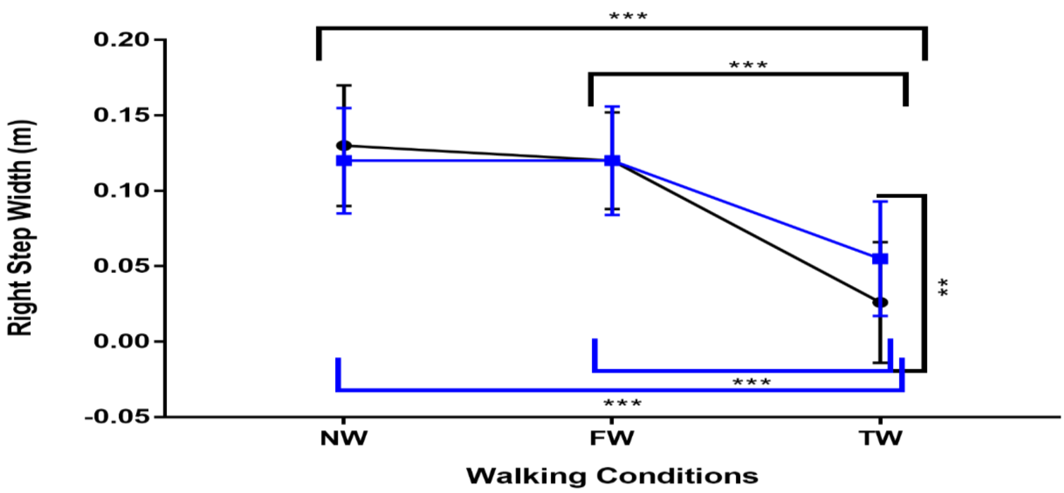
3.3.2.1 Right Step Width

The results showed that there was significant main effect of gait condition (F (1.68, 87.55) = 77.78, $p < .001$, $\eta_p{}^2 = .653$), sphericity violated (Table 9.). The results of the follow-up paired t-tests revealed that there was a significantly greater right step width for the NW than the TW conditions; t (55) = 10.46, p < 0.001, d = 2.17. There was also a significantly greater right step width for the FW than the TW conditions; t (55) = 10.46, p < 0.001, d = 2.14. There was no significant difference between the right step width for the NW and the FW.

The main effects were qualified by an interaction of gait condition and age (F (1.68, 87.55) = 4.82, $p = .015$, $\eta_p{}^2 = .085$), sphericity violated (Table 10.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the right step width in the NW and FW conditions between the OA and the YA participants. However, there was a significantly greater right step width in the TW condition for the OA than the YA participants; t (54) = -2.79, p = 0.007, d = 0.75.

The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (OA; t (27) = 6.37, p < 0.001, d = 1.79, YA; t (27) = 2.75, p < 0.001, d = 2.67). Specifically for FW and TW comparisons (OA; t (27) = 6.29, p < 0.001 d = 1.80, YA; t (27) = 8.81, p < 0.001, d = 2.61).

Figure 19 – Right Step Width across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

3.3.2.2 Left Step Width

The results showed that there was significant main effect of gait condition (F (1.75, 90.92) = 184.92, p < 0.001, $\eta_p{}^2$ = .781), sphericity violated (Table 9.). The results of the follow-up paired t-tests revealed that there was a significantly greater left step width for the NW than the TW conditions; t (55) = 13.82, p < 0.001, d =2.61. There was also a significantly greater left step width for the FW than the TW conditions; t (55) = 16.65, p < 0.001, d = 2.84. There was no significant difference between the left step width for NW and the left step width for FW conditions.

The main effects were qualified by an interaction of gait condition and age (F (1.75, 90.92) = 3.67, p = .035, $\eta_p{}^2$ = .066), sphericity violated (Table 10.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the left step width in the all three-gait condition for the OA and the YA participants. The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (OA; $t(27) = 14.86$, $p < 0.001$, $d = 3.46$, YA; $t(27) = 7.29$, $p < 0.001$, $d = 2.05$). Specifically for FW and TW comparisons (OA; $t(27) = 10.63$, $p < 0.001$, $d = 2.78$, YA; $t(27) = 12.97$, $p < 0.001$, $d = 2.89$).

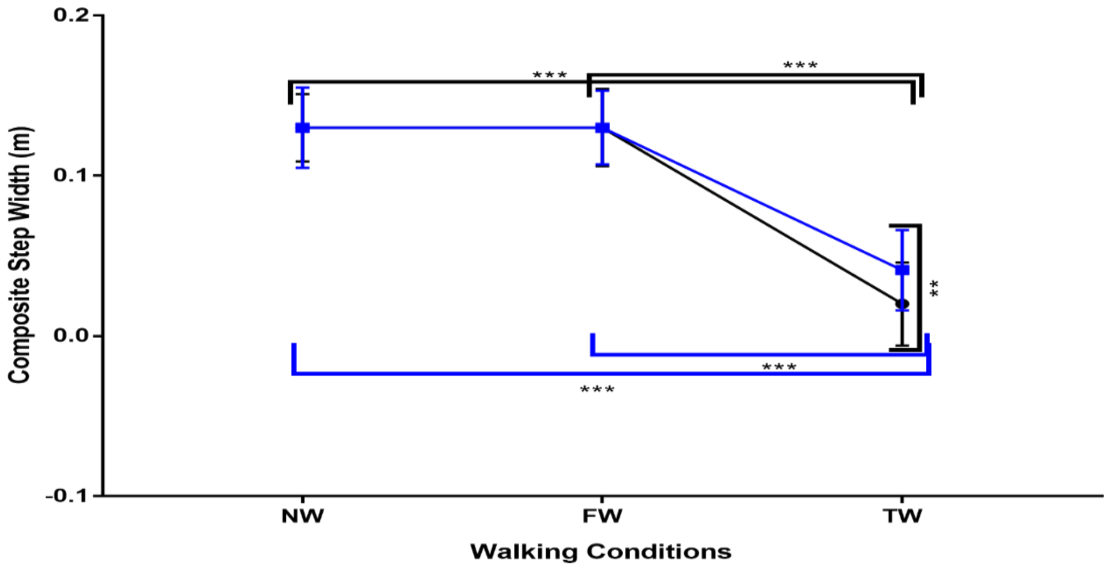
3.3.2.3 Composite Step Width

The results showed that there was significant main effect of gait condition ($F(1.38, 71.56) = 436.68$, $p < 0.001$, $\eta_p{}^2 = .894$), sphericity violated (Table 9.). The results of the follow-up paired sample t-test revealed that there was a significantly greater composite step width for the NW than the TW condition; $t(55) = 22.01$, $p < 0.001$, $d = 3.92$. There was also a significantly greater composite step width for the FW than the TW conditions; $t(55) = 20.28$, $p < 0.001$, $d = 3.78$. There was also a significantly greater composite step width for the NW than the FW conditions; $t(55) = 0.59$, $p = 0.56$, $d = 0.06$.

The main effects were qualified by an interaction of gait condition and physical activity ($F(1.38, 71.56) = 5.01$, $p = 0.018$, $\eta_p{}^2 = .088$), sphericity violated (Table 11.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the composite step width in the NW and FW conditions between the highly active and inactive groups. There was a significantly greater composite step width in the TW condition for the inactive participants than the highly active participants; $t(54) = -3.16$, $p = 0.004$, $d = 0.84$.

The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (HA; $t(27) = 16.80$, $p < 0.001$, $d = 4.57$, IA; $t(27) = 15.69$, $p < 0.001$, $d = 3.61$). Specifically for FW and TW comparisons (HA; $t(27) = 16.24$, $p < 0.001$, $d = 4.26$, IA; $t(27) = 13.48$, $p < 0.001$, $d = 3.59$).

Figure 20 - Composite Step Width across the Gait Conditions. Highly Active Adults (Black circles) and Inactive Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

Table 9 – Comparison of Spatial-Temporal Characteristics across the Gait Conditions

Variable (Mean (SD))	NW	FW	TW
Left step length (m) [1,3]	0.78 (0.08)	0.91 (0.12)	0.77 (0.08)
Right step length (m) [1,3]	0.78 (0.09)	0.92 (0.12)	0.77 (0.10)
Left step width (m) [2,3]	0.13 (0.04)	0.13 (0.04)	0.02 (0.04)
Right step width (m) [2,3]	0.13 (0.04)	0.12 (0.04)	0.04 (0.04)
Composite step width [2,3]	0.13 (0.022)	0.13 (0.023)	0.030 (0.027)
Left step time (s) [1,3]	0.64 (0.12)	0.51 (0.10)	0.67 (0.16)
Right step time (s) [1,2,3]	0.61 (0.10)	0.52 (0.09)	0.68 (0.14)
Stride length (m) [1,3]	1.56 (0.18)	1.83 (0.24)	1.54(0.18)
Stride time (s) [1,2,3]	1.25 (0.12)	1.03 (0.13)	1.36 (0.17)
Stride velocity (m/s) [1,2,3]	1.26 (0.18)	1.80 (0.30)	1.16 (0.21)

1 - Denotes a significant difference between NW and FW conditions

2 - Denotes a significant difference between NW and TW conditions

3 - Denotes a significant difference between FW and TW conditions

Table 10 – Comparison of the Spatial-Temporal Characteristics for Young and Older adults across the Gait Conditions collapsed across Physical Activity Level

Variables	**Young Adults**			**Older Adults**		
Mean (SD)	**NW**	**FW**	**TW**	**NW**	**FW**	**TW**
Left step length (m) ***AA 1,2,3***	0.811 (0.075)	0.973 (0.091)	*0.811 (0.68)*	0.751 (0.086)	0.852 (0.116)	0.723 (0.065)
Right step length (m) ***AA 2,3***	0.807 (0.082)	0.991 (0.086)	0.825 (0.086)	0.754 (0.089)	0.853 (0.118)	0.723 (0.096)
Left step width (m)	0.117 (0.046)	0.136 (0.032)	0.024 (0.045)	0.137 (0.034)	0.124 (0.041)	0.019 (0.034)
Right step width (m) ***3***	0.133 (0.040)	0.121 (0.032)	0.026 (0.040)	0.120 (0.035)	0.122 (0.036)	0.055 (0.038)
Left stance time (s)	0.648 (0.098)	0.489 (0.084)	0.622 (0.103)	0.623 (0.138)	0.534 (0.111)	0.711 (0.193)
Right stance time (s)	0.584 (0.082)	0.514 (0.073)	0.658 (0.110)	0.641 (0.115)	0.525 (0.113)	0.722 (0.158)
Stride length (m) ***AA 1,2,3***	1.618 (0.154)	1.964 (0.210)	1.636 (0.147)	1.494 (0.178)	1.705 (0.229)	1.446 (0.157)
Stride time (s) ***AA 3***	1.322 (0.088)	1.003 (0.076)	1.280 (0.092)	1.263 (0.147)	1.059 (0.164)	1.433 (0.197)
Stride velocity (m/s) ***AA 1,2,3***	1.321 (0.151)	1.969 (0.210)	1.287 (0.145)	1.201 (0.188)	1.642 (0.287)	1.036 (0.192)

AA - Main effect of age

1 - Denotes a significant difference between YA and OA in the NW condition

2 - Denotes a significant difference between YA and OA in the FW condition

3 - Denotes a significant difference between YA and OA in the TW condition

Table 11 - Comparison of the Composite Step Width across the Gait Conditions between the Highly Active and Inactive Group collapsed across Age groups

Variables Mean (SD)	**Highly active** [2,3]			**Inactive** [2,3]		
	NW	**FW**	**TW**	**NW**	**FW**	**TW**
Composite step width (m) [c]	0.13 (0.021)	0.13 (0.024)	0.020 (0.026)	0.13 (0.023)	0.13 (0.023)	0.041 (0.025)

1 - Denotes a significant difference between NW and FW conditions

2 - Denotes a significant difference between NW and TW conditions

3 - Denotes a significant difference between FW and TW conditions

AA - Main effect of age

a - Denotes a significant difference between HA and IA in the NW condition

b - Denotes a significant difference between HA and IA in the FW condition

c - Denotes a significant difference between HA and IA in the TW condition

3.4 Exploratory Analysis of the Between Limb Comparison for the NW condition

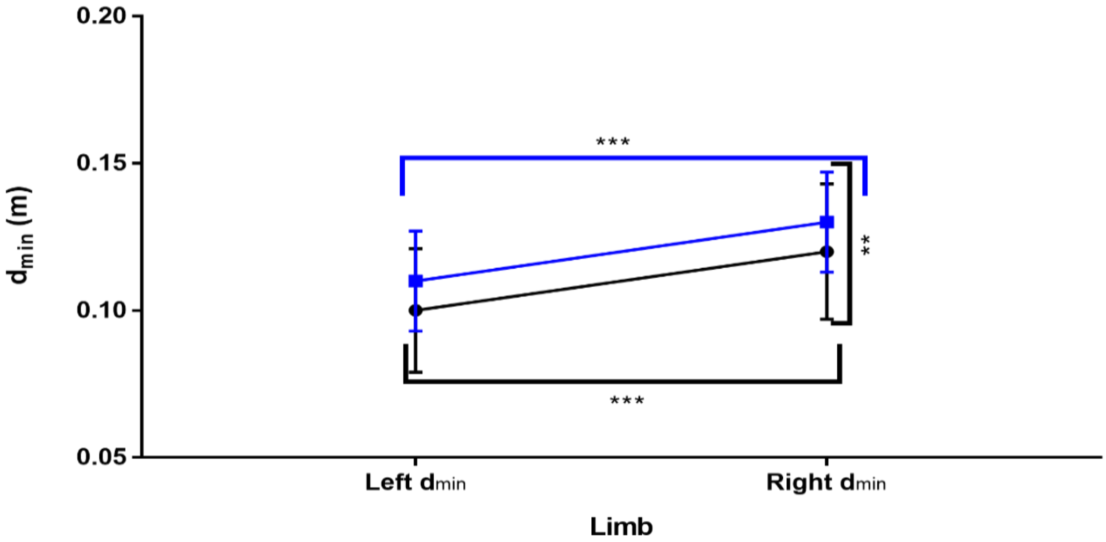
3.4.2.1 d_{min} Comparison for the Left and Right Limb

The results showed that there was significant main effect of limb, ($F(1, 52) = 71.13$, $p < 0.001$, $\eta_p^2 = .578$). The results revealed that there was a significantly greater right d_{min} than the left d_{min} (Table 4.). In addition, there was a significant main effect of age ($F(1, 52) = 4.35$, $p = 0.042$, $\eta_p^2 = .77$) with older adults having a greater d_{min} (M=0.12, SD=0.014) compared to younger adults (M=0.11, SD=0.021).

The main effects were qualified by an interaction of limb by age ($F(1, 52) = 4.47$, $p = .0.039$, $\eta_p^2 = .079$). The results of the follow-up independent sample t-test revealed that there was a significantly greater right limb d_{min} in the OA than the YA participants; $t(54) = -2.71$, $p = 0.009$, $d = 0.72$. There were no significant differences in the left limb d_{min} between YA and the OA (table 5.). The results of the paired sample t-test performed independently on both groups revealed differences consistent with the main effect of limb for the initial ANOVA.

Specifically for left and right limb comparisons (OA; t (27) = -6.70, p < 0.001, d = 1.40, YA; t (27) = -5.10, p < 0.001, d = 0.65).

Figure 21 - d_{min} Limb Comparison during NW Condition. Young Adults (Black circles) and Older Adults (Blue Squares)



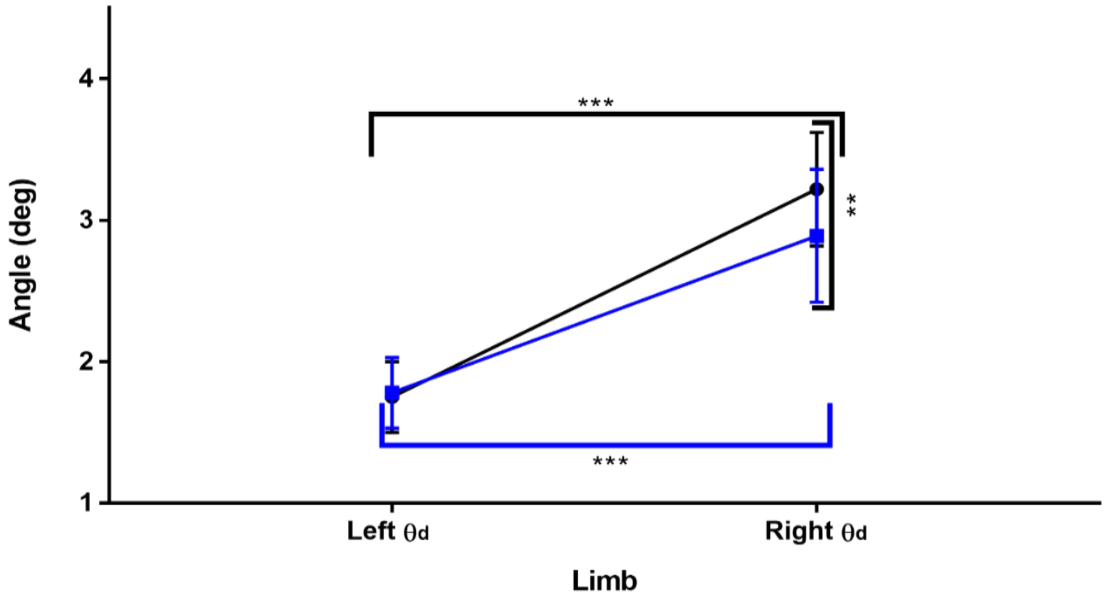
****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

3.4.2.2 Angle of Divergence (θ_d) Comparison of the Left and Right Limb

The results showed that there was significant main effect of limb, (F (1, 52) = 467.56, p < 0.001, η_p^2 = .900). The results revealed that there was a significantly greater θ_d for the right limb than the θ_d for the left limb (Table 4.). In addition, there was a significant main effect of age (F (1, 52) = 4.35, p = 0.042, η_p^2 = .77) with OA having a smaller θ_d (M=2.34, SD=0.28) compared to YA (M=2.49, SD=0.25).

The main effects were qualified by an interaction of limb by age (F (1, 52) = 9.24, p = .0.004, η_p^2 = .151). The results of the follow-up independent sample t-test revealed that there was a significantly smaller θ_d for the right limb in the OA than the YA participants; t (54) = 2.86, p = 0.006, d = 0.76. There were no significant differences in the θ_d for the left limb between YA and the OA (table 5.). The results of the paired sample t-test performed independently on both groups revealed differences consistent with the main effect of limb for the initial ANOVA. Specifically for left and right limb comparisons (OA; $t(27) = -12.08$, $p < 0.001$, $d = 2.96$, YA; $t(27) = -17.81$, $p < 0.001$, $d = 4.39$).

Figure 22 – θ_d Limb Comparison during NW Condition. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

3.4.2.2.1 Comparison of Left and Right Limb for the Positive phase of the θ_d

The results showed that there was no significant main effect of limb, age or limb by age interaction.

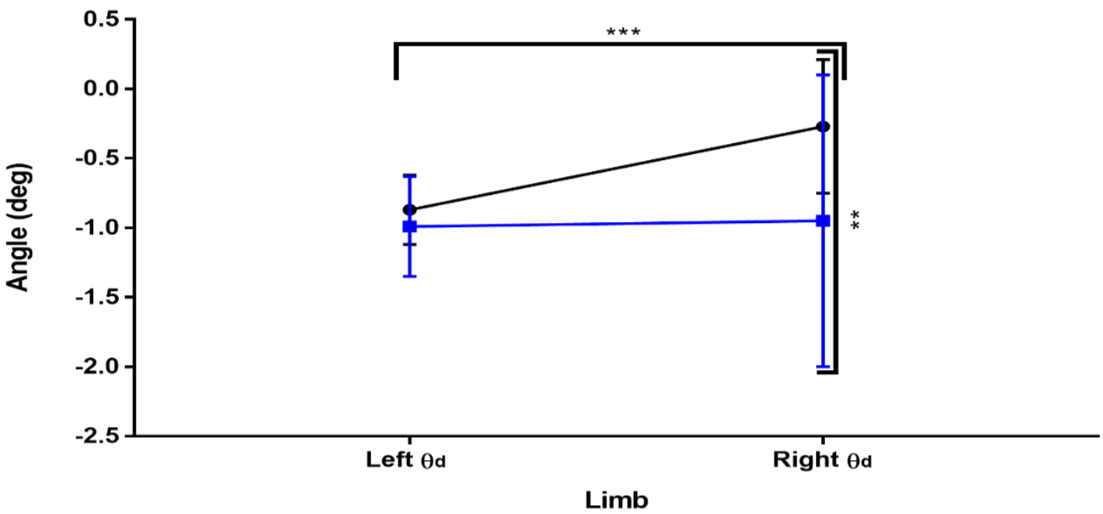
3.4.2.2.2 Comparison of Left and Right Limb for the Negative phase of the θ_d

The results showed that there was significant main effect of limb, ($F(1, 52) = 7.54$, $p = 0.008$, $\eta_p^2 = .127$). The results revealed that there was a significantly more negative θ_d for the left limb than the right limb (Table 7.). In addition, there was a significant main effect of age

(F (1, 52) = 11.71, p = 0.001, $\eta_p{}^2$ = .184) with OA having a more negative θ_d (M=0.97, SD=0.54) compared to YA (M=0.57, SD=0.31).

The main effects were qualified by an interaction of limb by age (F (1, 52) = 5.96, p = .0.0018, $\eta_p{}^2$ = .103). The results of the follow-up independent sample t-test revealed that there was a significantly more negative θ_d for the right limb in the OA than the YA; t (37.83) = -3.14, p = 0.003, d = 0.84. There were no significant differences in the θ_d for the left limb between YA and OA (Table 8.). The results of the paired sample t-test performed independently on both groups revealed that there were no limb differences for the OA. While for the YA, there was a significantly more negative θ_d for the left limb than the right limb (t (27) = 7.09, p < 0.001, d = 1.57).

Figure 23 - Limb Comparison during the NW Condition for the Negative phase of the θ_d. Young Adults (Black circles) and Older Adults (Blue Squares)



: significant difference at P < 0.01, *: significant difference at P < 0.001.**

CHAPTER 4 – DISCUSSION

Previous work examining the eccentricity of the net ground reaction force (GRF_{net}) has elucidated the importance of this mechanism underlying increased mediolateral instability among older adults (OA) during single step voluntary and compensatory stepping responses (Singer et al., 2014, 2016). While those studies looked at the restabilization phase of stepping after foot contact during a single step, the findings by Singer et al. (2014, 2016) suggest that a greater mediolateral centre of mass (COM) displacement would place the COM closer to the lateral stability limits placing an individual at an increased risk of instability. Such challenges in controlling the COM kinematics are believed to arise via challenges regulating the eccentricity of the GRF (Singer et al., 2013). The present study examined the control of COM kinematics via the eccentricity of the ground reaction force (GRF) to better understand mechanisms underlying mediolateral dynamic stability during gait. As physical activity is believed to improve stability, this thesis sought to understand how the effects of ageing on stability control during gait may be mitigated by physical activity. The effects of age and physical activity on mediolateral stability during gait were examined by: (a) quantifying mediolateral kinematic instability among younger and older adults; (b) determining the underlying force-related mechanisms related to such instability during walking; (c) Quantifying how high levels of physical activity may influence the control of mediolateral stability during walking among both younger and older adults.

I tested the following hypotheses: (1) across all levels of physical activity, older adults would exhibit greater kinematic instability relative to young adults. Across all age groups, individuals who are inactive would exhibit greater instability relative to individuals who are highly active. It was expected that physical activity will have a greater effect on stability in older adults compared to young adults. Greater instability would be evidenced by the COM travelling closer to the lateral limits of the base of support (BOS) (i.e. smaller d_{min}). (2) Across all levels of physical activity, older adults would exhibit greater challenges in regulating stability via the direction of applied forces relative to the COM. Across all age groups, individuals who are inactive would exhibit greater challenges in regulating stability via the direction of applied forces relative to the COM relative to individuals who are highly active. It was expected that physical activity would have a greater effect on the ability to regulate the orientation of the ground reaction force in older adults compared to young adults. Challenges in the regulating the orientation of the ground reaction force was quantified by a smaller angle of divergence (θ_d). (3) Overall, I expected that differences in stability as a function of age as well as physical activity and exercise would be accentuated in the dependent measures (d_{min} and θ_d) when participants perform two additional walking conditions: 'as fast as possible' (FW) and modified tandem walking (TW). Across all levels of physical activity, older adults would exhibit a smaller d_{min} and smaller θ_d in the FW and TW conditions relative to the normal walking (NW) condition compared to young adults. Across all age groups, individuals who are inactive would exhibit a smaller d_{min} and smaller θ_d in the FW and TW conditions relative to the normal walking (NW) condition compared to individuals who are highly active. FW and TW conditions challenge the individuals' stability and are considered to be more destabilizing in the mediolateral direction (Ko et al., 2010; Schrager et al., 2008). During unconstrained level-walking, healthy adults can maintain dynamic stability by regulating GRF orientation or by modifying foot placement (Arvin, van Dieën, & Bruijn, 2016; Hof, 2007; Redfern & Schumann, 1994). The FW condition is considered challenging, as increased mediolateral COM accelerations require participants to modify their foot placement or generate larger applied forces/eccentric forces to redirect the increased COM accelerations. The current study showed that, for the FW condition, participants generated larger eccentric forces to maintain dynamic stability for the net measure, left and right limb. TW condition is challenging because the step placement is constrained (reduced BOS) thus participants must regulate the direction of the applied forces precisely to maintain stability. The current study showed that, during the TW condition, participants may have been more unstable as a consequence of a reduced step width and the inability to generate the eccentric forces to maintain stability.

Contrary to the initial hypothesis, older adults (OA) did not exhibit markers of greater kinematic instability compared to younger adults (YA). There were not physical activity related differences in the kinematic measures of stability. Our results were partially consistent with the second hypothesis, in that OA did exhibit a smaller eccentricity of the GRF (θ_d), which has been suggested to indicate challenges regulating applied forces during each step and, consequently, instability. Interestingly, such kinetic outcomes are inconsistent with our kinematic findings of increased stability. There were not physical activity related differences in the regulation of applied forces.

Our results were partially consistent with the third hypothesis, in that on average, across all gait conditions, OA exhibited greater kinematic stability (i.e. larger d_{min}) for the left limb and composite measures. For the right limb, however, OA exhibited greater kinematic stability (i.e. larger d_{min}) during NW and TW conditions, but were not different from YA during the FW condition. When comparing strictly across gait conditions, all d_{min} values were greater in both the NW and FW conditions, relative to the TW condition, while there was no difference between the NW and FW conditions. Specifically, across all gait conditions, OA exhibited a smaller θ_d, but only for the right limb and net GRF analyses. For both age groups, θ_d was largest in the FW condition, followed by NW and TW conditions. On average, across all gait conditions, there was no evidence to suggest that the inactive groups exhibited more kinematic instability compared the highly active group. For both activity level groups, all d_{min} values were greater in both the NW and FW conditions, relative to the TW condition, while there was no difference between the NW and FW conditions. On average, across all gait conditions, there were no significant differences in the kinetic measures between the inactive and highly active groups.

To better understand the meaning of the age-related differences in kinetics, we further decomposed the angle of divergence waveform into positive (believed to be restabilizing) and negative (believed to be destabilizing) phases for each the right and left stance phases. For the left stance phase, there were no age-related differences in either the positive or the negative GRF orientation. Across gait conditions, the positive GRF orientation for the left side was largest in the NW condition, followed by FW condition and then TW condition. The negative GRF orientation was largest in the FW condition, followed by NW and TW conditions. For the right stance phase, there were age-related differences in the positive and negative phases. On average, across all gait conditions, YA had a greater positively oriented GRF while OA having a greater negatively oriented GRF. The positive GRF orientation was largest in the FW condition, followed by NW condition and then TW condition. The negative GRF orientation was largest in the TW condition, followed by FW condition and then NW condition. As the interpretation of positive and negative θ_d values is in contrast to indices of stability as indicated by the kinematic variable (d_{min}), we present a new interpretation of the influence of the GRF orientation on mediolateral dynamic stability during gait.

4.1 Age-related Differences during Walking

OA exhibited a significantly greater composite d_{min} and an overall reduced net θ_d compared to YA. OA having a greater composite d_{min} would suggest that they are kinematically stable compared to YA because the COM is further away from the lateral limits of the BOS. However, the smaller net θ_d exhibited by OA, given current interpretations (Singer et al., 2016) indicates challenges regulating stability, as the GRF_{net} is directed closer to the COM thereby limiting the ability to oppose destabilizing angular momentum. Contrary to compensatory stepping, which has been shown to require the regulation of angular momentum for balance recovery (Pijnappels, Bobbert, & Van Dieën, 2004), steady-state gait may require different control mechanisms such as the regulation of linear momentum. Previous studies have shown that OA tend to walk with a larger step width compared to YA (Grabiner, Biswas, & Grabiner, 2001), however in the current study, the data showed that there were no age-related differences in composite step width. As there were no age-related differences in composite step width across the gait conditions, the greater composite d_{min}, exhibited by OA, suggests a minimization of the mediolateral oscillations of the COM within BOS, as a consequence of kinetic mechanisms. The reduction in the mediolateral COM displacement is likely related to limited lateral accelerations of the COM (Hernández, Silder, Heiderscheit, & Thelen, 2009), via kinetic mechanisms, described below, which in turn may reduce the risk of lateral instability because the COM is further away from the limits of the BOS.

Given the kinetic results (i.e. reduced net θ_d among OA), the only plausible explanation for an increased composite d_{min} is that OA are producing a greater mediolateral GRF component, which could decrease mediolateral COM displacements and yield the greater kinematic stability among OA. The size of the GRF orientation generated could vary depending on the relationship between the horizontal and vertical GRF forces produced. An increased vertical GRF component and/or reduced horizontal GRF component would maximize the eccentricity of the GRF relative to the COM (θ_d), while an increased horizontal GRF component and/or reduced vertical GRF component would minimize the eccentricity of the GRF relative to the COM (θ_d). A greater medially (more horizontal) directed GRF component (consistent with a reduced θ_d) - could be causing the larger composite d_{min} among OA, since the more horizontal, medially directed, GRF component would limit the lateral oscillations and accelerations of the COM. On the other hand, the reduced θ_d (and consequent increased composite d_{min}) exhibited by OA, could also be caused by a reduction in the vertical GRF component, as a consequence of an effort to reduce vertical loading on foot contact (independent of a strategy to maintain mediolateral stability) (Brooshak, Asadi, & Hosseini, 2017; Chung & Wang, 2010). As compressive forces act three times the body weight on the knee joint (Kettlekamp & Jacobs, 1972), for example, it could be suggested that a reduced vertical GRF component could be a strategy to reduce the amount of compressive forces acting on the knee joint. The reduced vertical GRF component could similarly reduce the influence of compressive forces on the other joints of the lower limbs.

During walking, mediolateral dynamic stability has been proposed to be maintained by the control of linear momentum (Simoneau & Krebs, 2000) and minimizing the amount of angular momentum generated (Herr & Popovic, 2008; Neptune & McGowan, 2016). The linear momentum is controlled via foot placement while angular momentum is controlled by directing the GRF_{net} around the COM via muscle force generation (Neptune & McGowan, 2016; Simoneau & Krebs, 2000). The current results suggest that linear momentum can also be controlled by directing the GRF_{net} orientation given that there were no age-related differences in foot placement. In addition, the current results suggest OA are tightly regulating/minimizing their whole body angular momentum by directing the GRF_{net} at or nearer the COM and also regulating/minimizing whole body linear momentum by potentially generating a larger medially directed GRF component. A strict regulation of angular momentum may be a strategy used by OA to maintain steady state gait stability; however, it may come at the expense of an ability to regain stability if they experience a perturbation. Specifically, a reduced eccentricity of the GRF_{net} may limit the ability to generate enough restabalizing angular momentum to oppose the destabalizing angular momentum induced by perturbation itself (Pijnappels et al., 2004).

4.1.1 Individual Limb Analysis

The initial analysis of the net kinetic (net θ_d) and composite kinematic (composite d_{min}) variables provided information on how OA regulate applied forces as a strategy to maintain stability. Breaking down the net kinetic and the composite kinematic variables to the individual limbs could potentially provide information on the presence of a differential contribution from each limb to the regulation of dynamic stability. Upon investigating the individual limbs, OA had an increased kinematic stability (d_{min}) on both the left and right limbs (Figure 17.). OA exhibited a significantly smaller θ_d compared to YA for the right limb. There were no age-related differences in the θ_d for the left limb. The secondary analysis of the positive and negative θ_d showed that for the right limb, OA had a significantly less positive θ_d and significantly more negative θ_d in the NW condition (Figure 18.). While for the left limb, there were no significant differences in the positive and negative θ_d between the age groups. Even though there were no significant differences in the positive or negative θ_d between the age groups, the left limb showed a similar trend to the right limb where OA had a less positive θ_d and a more negative θ_d.

As mentioned earlier in the discussion section, healthy adults maintain dynamic stability by either regulating GRF orientation or by modifying foot placement during walking. The individual limb differences in d_{min} between the age groups could be explained by the regulation of applied forces. OA exhibited a significantly more negative θ_d on the right limb. The more horizontal, medially directed, GRF component limits the lateral acceleration of the COM thus causing the increased kinematic stability (increased d_{min}) on the right limb. For the left limb, even though there were no significant age-related differences in both the positive and negative θ_d, OA had a less positive θ_d and a less negative θ_d.

The differences that arise from the individual limbs could be also explained spatial-temporally (step width). Increasing the step width from one side could increase the kinematic stability (d_{min}) for the ipsilateral limb. Alternatively, increasing the step width from one side could increase the kinematic stability (d_{min}) for the contralateral limb by limiting the lateral COM displacement. Studies have shown that OA tend to have a wider step width compared to

YA(Grabiner et al., 2001; Owings & Grabiner, 2004). The present results show that, despite the lack of age-related differences in composite step width measures described above, there were age-related differences on the right step width during the TW condition, with OA exhibiting a greater right step width. This could suggest that for the TW condition, OA had an increased kinematic stability (d_{min}) for the right limb by increasing the step width for the right limb. Alternatively, the increased right step width seen in OA, compared to YA, could have reduced the lateral COM displacement towards the contralateral side. The reduced lateral COM displacement increases the kinematic stability (d_{min}) for the left limb for the OA.

Given the results (OA exhibiting a greater step width for only the TW condition, coupled with significant age-related differences in the kinetic measures for only the right limb), the question arises, how OA exhibit greater kinematic stability (d_{min}) in the left limb during the NW and FW conditions. Even though the differences in the kinetic and spatial-temporal parameters were non-significant, a plausible explanation is that OA may use the various mechanisms (kinetic and spatial-temporal), together, to maintain kinematic stability for the left limb. OA exhibited a trend towards a less positive θ_d and a less negative θ_d. In addition, there were no significant age-related differences in the left stance time, the results showed a trend towards an increased left stance time for OA which increases the duration in which the forces (i.e. impulse) are applied. All the small differences between the YA and OA, even though non-significant, could have added up and resulted into a slightly greater kinematic stability (increased d_{min}) for the left limb in the OA compared to YA, which is consistent with the present results. Taken together, these results suggest that OA may be required to employ multiple mechanisms (kinetic and temporospatial) to regulate dynamic stability, and these strategies may be particularly relevant for the management of dynamic stability on the non-dominant side.

4.1.2 Between Limb Analysis during Normal Walking

Further analysis of the differences between the right and left limbs could potentially provide information on whether OA are more susceptible to instability on the dominant or non-dominant side. All the participants reported that they were right leg dominant. Upon investigating the between limb differences, all subjects exhibited greater kinematic stability (d_{min}) on the right side. OA specifically had a significantly greater d_{min} for the right limb but also had a trend towards an increased d_{min} for the left limb (Figure 21.), relative to YA. OA exhibited a significantly smaller θ_d compared to YA for the right limb while there were no age-related differences in the θ_d for the left limb (Figure 22.). The exploratory analysis also showed that there were no differences of the positive θ_d between limbs and between age groups. However, OA exhibited a significantly more negative θ_d for the right limb while there were no age-related differences in the negative θ_d for the left limb (Figure 22.). However, there was a trend showing that OA had a more negative θ_d compared to YA for the left limb.

The differences between the left and right limbs and between the age groups could be explained by the regulation of applied forces. OA had a significantly greater kinematic stability for the right limb and had a non-significant trend towards an increased kinematic stability for the left limb. This is consistent with the kinetic findings (significantly greater negative θ_d on the right side among OA; non-significantly greater negative θ_d on the left side among OA). The interpretation being a greater negative θ_d and associated larger medially directed GRF component is responsible linear momentum regulation and, hence, the greater kinematic stability.

All the participants exhibited a greater kinematic stability on the right side relative to the left side. Only OA exhibited a consistently negative θ_d between limbs; YA exhibited a greater negative θ_d on the left side relative to the right side. This suggests that OA seem to be attempting to tightly regulate stability, equally on both limbs, by directing the θ_d more negatively (i.e. greater medially directed GRF). The more horizontal, medially directed, GRF component would limit the lateral acceleration of the COM and increased kinematic stability (increased d_{min}). However, given the smaller d_{min} on the left side, relative to the right side among OA, this group may be producing insufficient horizontally directed forces to regulate the COM kinematics on their non-dominant side, despite the between-limb similarities in the negative θ_d. This could occur if there was a minimization of both the vertical and horizontal GRF components on the left side, and this minimization of both components occurred in proportion. In this case, even though the negative θ_d is similar between the left and ride sides, the horizontal GRF component it may still be insufficient to constrain the mediolateral COM kinematics and lead to the smaller kinematic stability on the left side.

For YA, the θ_d was more negative for the left limb compared to right limb, despite fact that YA had markers of greater kinematic stability on the right limb. This may be a case where the θ_d fails to capture the magnitudes of the individual horizontal and vertical GRF components. In the present situation, YA may be producing a large medially directed GRF component on the right side (despite a smaller negative θ_d, relative to the left side) and a small medially directed GRF component on the left side (despite a larger negative θ_d, relative to the right side). As the θ_d captures the orientation of the GRF relative to the COM, independent of the resultant force magnitude, the right-left difference in the θ_d could manifest if there were correspondingly large and small vertical force components from right and left limbs, respectively. The between-limb differences underscore the necessity to analyse individual force components in addition to θ_d, to understand both the control of applied forces and the generation of applied force, via their individual magnitudes.

The present results suggest there may be natural dominant/non-dominant side asymmetries in both force control and force generation, given that YA and OA have a smaller d_{min} on the left side. Coupled with identical right and left limb step placement for both age groups, it may be suggested that the stability margin is reduced for the left limb compared to the right limb during the NW, potentially increasing the susceptibility to lateral instability toward the left (non-dominant) side. As lateral instability may be associated with increased lateral fall risk (Maki et al., 1994; Schulz et al., 2005; Tirosh & Sparrow, 2005), understanding the role that leg dominance may play in control the mediolateral stability during gait may be an important future direction to reduce falls risk among OA.

4.2 Physical Activity-related Differences during Walking

There could be several possibilities why there were no differences in the kinematic and kinetic indices of stability for the inactive participants (IYA and IOA) when compared to highly active participants (HAYA and HAOA). Firstly, running is a cardiovascular sport (Cantwell, 1985); the results of the current study could suggest that more balance specific physical activity is required to see a difference in mediolateral stability when comparing active and inactive individuals. Ringhof and Stein (2018) examined dynamic stability in young females engaging in regular training in either gymnastics or swimming. The participants took part in a single leg jump landing, perturbations on an unstable platform, simulated forward falls and single leg stance. The participants were instructed to recover balance as quickly as possible. The results showed that out of the four tests, gymnasts regained stability significantly faster in the jump-landing task compared to the swimmers. Gymnasts regularly include landing exercises in their routine thus could perform better in the jump landing task. In another study, Shimada, Uchiyama, and Kakurai, (2003) compared the effect of balance exercises, gait exercises on two groups and a control group on frail, elderly population. Both training groups had significantly improved physical function, however, the balance group showed more improvements in static balance measures and the gait group showed more improvement in the dynamic balance and mobility measures. The results from these studies suggest that appropriate balance training is required for specific tasks. Performing a more specific balance training physical activity may provide a greater benefit to improving stability. These studies also suggest that incorporating balance-specific components may be necessary to improve gait stability. The results from the studies follow the principle of specificity of learning (Henry, 1968) which suggests that learning is the most effective when practice sessions resembles the skills and the environment conditions that is similar to the task.

Secondly, based on the present results, it may appear that PAL does not affect gait stability among healthy participants. The present study did not test perturbation responses. Perturbations challenge the reactive control system. There is no evidence in the data to suggest that running and other forms of physical activity and exercise programs could not help individuals recover from a perturbation. The data only suggest that PAL has no or minimal impact on dealing with self-imposed perturbations experienced during gait.

Thirdly, the aerobic physical activity values in the present study were self-reported. In a systematic review, Prince et al., (2008) assessed the current and changing physical activity levels and to evaluate the effectiveness of the interventions by reviewing self-reported data and directly measured data of adults engaging in physical activity. The authors concluded that self-reporting measures of physical activity varied and thus could affect the reliability and accuracy of the data. In the current study, the participants could have either over-reported or under-reported their aerobic physical activity values. This would have made the participants, especially the OA, fit the criteria of either as being highly active or inactive when probably they should not have been categorized into those groups.

Lastly, some of the HAOA consisted of individuals who did not achieve the strength-training portion of the CSEP guidelines. The CSEP guidelines considered brisk walking as a moderate to vigorous intensity exercise (CSEP, 2013). The HAOA also consisted of individuals who included walking as part of their aerobic activities. Tully, Cupples, Chan, McGlade, and

Young (2005) described brisk walking as a physical activity that is moderately intensive that increases the heart rate. According to the Ainsworth et al. (2000), depending on the speed and whether an individual is carrying a load, walking can be considered as light, moderate or vigorous. Thus the interpretation of brisk walking may have been misinterpreted by the OA and thus they could have over-reported the number of minutes of moderate-to-vigorous physical activity they performed. In a systematic review, Sherrington, Tiedemann, Fairhall, Close, and Lord (2011) reported that a higher exercise dose that includes balance training that included walking as part of their training had less reduction in falls compared to higher exercise dose that includes balance training that did not include walking as part of their training. The authors also recommended that the exercises should provide a moderate-to-vigorous challenge to balance. This could suggest that some HAOA may have not performed exercises that challenged their balance or those HAOA who included walking as part of their aerobic physical activity and the could have had a slightly increased susceptibility to lateral instability compared to those HAOA that did not include walking as part of their aerobic physical activity and performed exercises that challenged their balance. The data from HAOA who included walking as part of their aerobic physical activity or performed exercises that did not challenge their balance stability could have affected the overall values of the HAOA group.

4.3 Study Limitations

A number of factors limited this study. Firstly, due to the location of the force plates in the walkway, data from only two steps (1 left footstep and 1 right footstep) was captured and analyzed. The likelihood of detecting instability from a single step or a few steps is low, more steps increases the chances of detecting instability. Secondly, the size of the testing laboratory may not have been long enough to allow for steady-state gait (gait maintained at a steady speed before the first contact on the force plate). It takes around three steps to achieve steady state gait (Mann, Hagy, White, & Liddell, 1979). Achieving steady-state gait may not have been possible for the taller participants, especially during the FW condition. Muir, Rietdyk, and Haddad (2014) suggested that there could be a higher step length and step width variability during the initial steps to reach steady-state gait.

Thirdly, it may have been difficult to get an accurate COM measure from the marker placements for each segment; however accurately the markers were placed to define the segment endpoints, the soft tissue may cause a slight location error in the segment COM. When net location of the COM was calculated by getting the weighted average of each segment, it may have led to a slightly inaccurate location of the whole-body COM. This could affect the kinematic measure value, as the values could be made smaller or larger depending on the location of the whole-body COM. Fourthly, the speed for the walking conditions were self-selected. Therefore, the indifferences in the kinematic variable between the FW and NW conditions could be attributed to the possibility that participants performing the FW condition were not performed at their fastest walking speed. Lastly, I did not directly analyse vertical and horizontal components of the GRF because the individual force magnitudes could be smaller or larger but if the horizontal and vertical components scale in proportion, the θ_d could be identical. I did not specifically analyse the vertical and horizontal forces because the present work sought to study the control of applied forces, rather than the ability to generate forces with sufficient magnitude.

CHAPTER 5 – CONCLUSION/FUTURE DIRECTIONS

Walking is a fundamental element of everyday life, which poses a complex stability control challenge. The components of gait and balance are fundamental to physical function and mobility. The maintenance of mediolateral stability is challenging for older adults (OA) and is critically important since lateral falls carry an increased probability of hip fracture. Using new biomechanical measures of mediolateral stability allowed for a better understanding of how age-related challenges and physical activity contributes to stability during the different types of walking.

5.1 Future Direction

The results of this thesis lay the foundation for future work that could provide a better understanding of the age- and physical activity-related differences in mediolateral dynamic stability control during walking. Future studies could include examining other forms of physical activity and exercise such ballet dancing or soccer to understand their effects of on balance. These activities require balance while performing dynamic movements. In addition, using electromyography to see the association between muscle activation and the applied forces regulating stability will be an important step toward developing targeted interventions. Examining the d_{min} and the θ_d of the ground reaction force for multiple strides or for specific instances in time may provide a more accurate analysis of the kinetic and kinematic variables in addition to the provision of variability-based measures. The inclusion of mediolateral perturbations while performing a similar study may allow for the determination of how physical activity and exercise may aid individuals in balance recovery after a perturbation. Lastly, even though the study looked at individual limb contribution in the control of mediolateral stability, a future study could look at having a better measure for leg dominance and performing a similar analysis.

5.2 Conclusion

The age-related differences suggest that greater kinematic stability is a strategy used by older adults to maintain stability by directing the GRF orientation more towards the COM during walking. However, by reducing the GRF orientation, older adults may not be able to recover after a perturbation as they may have challenges to regulate their stability. In addition, all the participants were right leg dominant; the dominant limb corresponded better with kinematic stability compared to the non-dominant leg (left limb) during gait. There could be an increased number of falls on the non-dominant side because of the poorer kinematic stability on that side. This may suggest that there may be a need for more balance and strength training exercises on the non-dominant limb than the dominant limb. It also suggests that running may not play role in the control of mediolateral normal gait stability. More specific balance physical activity and/or exercise training may be needed to be performed to influence mediolateral stability.

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APPENDICES

Appendix A - List of Medication

A partial list of drugs that has been associated with gait disturbances (Barbara Cadario and BC Falls and Injury Prevention, 2011):

- ACE inhibitors,
- Alpha receptor blockers,
- Anticoagulants,
- Antipsychotic,
- Eye drops,
- Muscle relaxants,
- NSAIDs,
- Analgesics,
- Antiarrhythmics,
- Anticonvulsants,
- Antidepressants, and
- Antihypertensives.

Appendix B - Eligibility Screaning Questions

1. Sex – Male Female

2. Age –

3. Height – Weight –

4. Have you had any surgeries on your lower body (waist and below) within the last one year? YES NO

If yes, specify:

____

____

5. Have you been diagnosed with any anatomical, neurological impairements within the last 6 months? YES NO

6. Do currently have an injury or soreness on your lower body? YES NO

If yes, specify:

____

____

7. In the last 6 months, have you had a fall* while performing activities of daily living (walking, household chores, etc.)? YES NO

If yes, how many? __________

***A Fall is defined as a sudden and unintentional change in position resulting in an individual landing at a lower level such as on an object, the floor, or the ground, with or without injury.**

8. Do you walk with any sort of aide (i.e. walker, cane)? YES NO

9. Are you currently taking any substances or medications or have you taken any substances or medications during the previous week that could affect your walking? (i.e., medication for depression, anxiety, pain killers, etc.) YES NO

If yes, please specify:

____

____

10. Have you experienced any head injuries within the last 2 weeks? YES NO

If yes, please specify (severity of injury and date of occurrence):

____

____

Appendix C – Physical Activity Questionnaire

CSEP-PATH: PHYSICAL ACTIVITY AND SEDENTARY
BEHAVIOUR QUESTIONNAIRE (PASB-Q)
Adult (18 and over)

Please answer the following questions based on what you do in a typical week. To increase accuracy, you may wish to log your physical activity and sedentary behavior for one week prior to answering the questions.

Aerobic Physical Activity

1. Frequency: In a typical week, how many days do you do moderate-intensity (like brisk walking) to vigorous intensity (like running) aerobic physical activity?

___ days/week

2. Time or Duration: On average for days that you do at least moderate-intensity aerobic physical activity (as specified above), how many minutes do you do?

___ minutes/day

Total: Multiply your average number of days per week by the average number of minutes per day.

___ minutes/week

Muscle Strengthening Physical Activity

3. In a typical week, how many times do you do muscle strengthening activities (such as resistance training or very heavy gardening)?

__ times/week

Perceived Aerobic Fitness

4. In general, would you say that your aerobic fitness (ability to walk/run distances) is:

__ Excellent, __ Very Good, __ Good, __Fair, __ Poor

Sedentary Behaviour

5. On a typical day, how many hours do you spend in continuous sitting: at work, in meetings, volunteer commitments and commuting (i.e., by motorized transport)?

… None	… < 1 hour	… 1 to < 2	… 2 to < 3
… 3 to < 4	… 4 to < 5	… 5 to < 6	… > 6

6. On a typical day, how many hours do you watch television, use a computer, read, and spend sitting quietly during your leisure time?

… None	… < 1 hour	… 1 to < 2	… 2 to < 3
… 3 to < 4	… 4 to < 5	… 5 to < 6	… > 6

Total Sedentary Behaviour (add responses to questions 5 and 6) __ hours/day

7. When sitting for prolonged periods (one hour or more), at what interval would you typically take a break to stand and move around for two minutes?

… < 10 minutes

… 10 to < 20 minutes

… 20 to < 30 minutes

… 30 to < 45 minutes

… 45 to < 1 hour

… 1 to < 1.5 hours

… 1.5 to < 2 hours

… > 2 hours

Appendix D – Spatial-Temporal Parameters

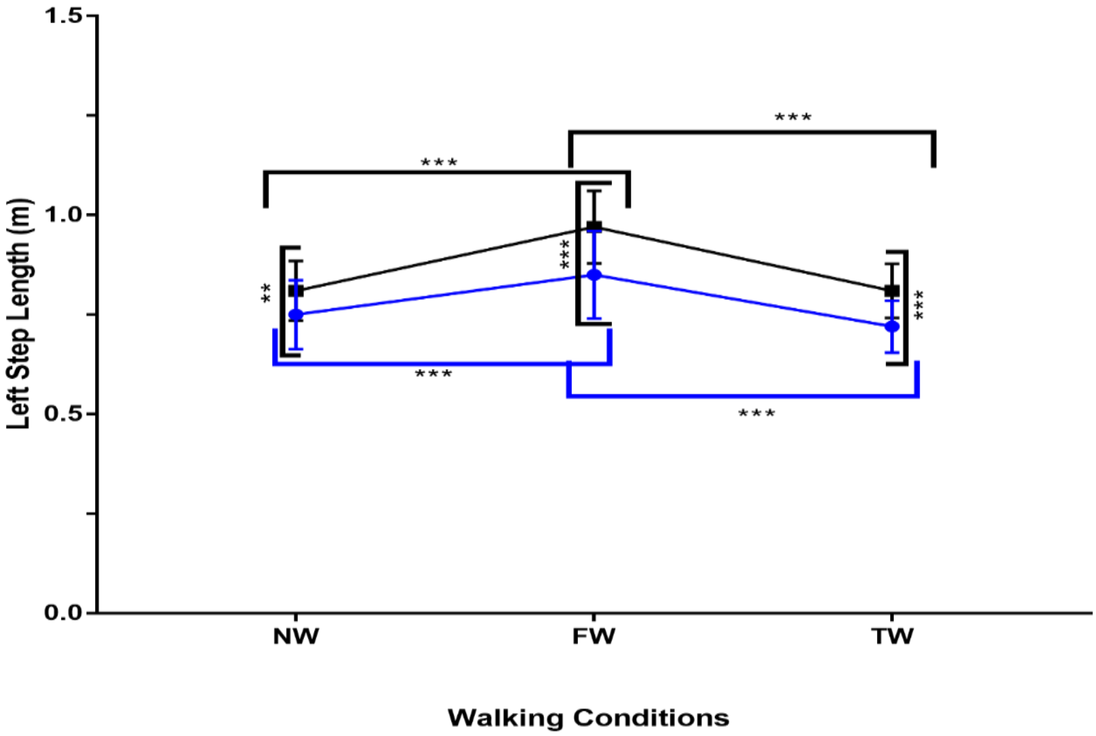
Left Step Length

The results showed that there was significant main effect of gait condition ($F_{(2,104)}$ = 122.45, $p < .001$, $\eta_p{}^2 = .702$) (Table 9.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the left step length for NW and the TW condition. There was a significantly greater left step length for the FW than the TW condition; $t(55) = 12.61$, $p < 0.001$, $d = 0.17$. There was also a significantly greater left step length for the FW than the NW condition; $t(55) = -11.080$, $p < 0.001$, $d = 1.26$.

In addition, there was a significant main effect of age ($F_{(1, 52)} = 20.49$, $p < 0.001$, $\eta_p{}^2 = .283$) with OA having a smaller left step length (M=0.76, SD=0.077) compared to YA (M=0.87, SD=0.068).

The main effects were qualified by an interaction of gait condition and age ($F_{(2, 104)} = 4.52$, $p = .013$, $\eta p\ 2 = .080$) (Table 10.). However, as with the main effect of age, OA exhibited a consistently smaller step length in NW ($t(54) = 2.79$, $p = 0.007$, $d = 0.74$), FW ($t(54) = 4.37$, $p < 0.001$, $d = 1.17$) and TW conditions ($t(54) = 4.95$, $p < 0.001$, $d = 1.32$). The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for FW and TW comparisons (OA; $t(27) = 7.03$, $p < 0.001$, $d = 1.37$, YA; $t(27) = 11.89$, $p < 0.001$, $d = 2.02$). Specifically for NW and FW comparisons (OA; $t(27) = -5.88$, $p < 0.001$, $d = 0.99$, YA; $t(27) = -13.62$, $p < 0.001$, $d = 1.94$).

Figure 24 – Left Step Length across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

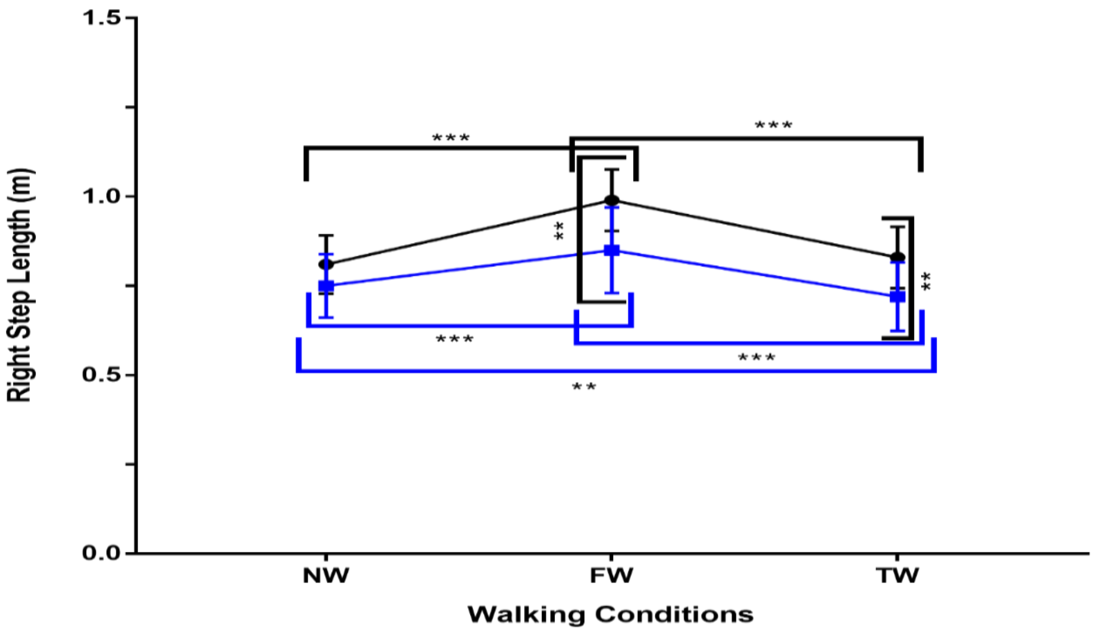
Right Step Length

The results showed that there was significant main effect of gait condition (F (1.69, 88.11) = 136.34, p < .001, $\eta_p{}^2$ = .724), sphericity violated (Table 9.). The results of the follow-up paired sample t-tests revealed that there was no significant difference between right step length for the NW the TW conditions. There was a significantly greater right step length for the FW than the TW conditions; t (55) = 12.92, p < 0.001, d = 1.29. There was also a significantly greater right step length for the FW than the NW conditions; t (55) = -11.29, p < 0.001, d = 1.32.

In addition, there was a significant main effect of age ($F(1, 52) = 18.90$, $p < 0.001$, $\eta_p^2 = .267$) with OA having a smaller right step length (M=0.78, SD=0.090) compared to YA (M=0.87, SD=0.075).

The main effects were qualified by an interaction of gait condition and age ($F(1.69, 88.11) = 8.93$, $p = .001$, $\eta_p^2 = .147$), sphericity violated (Table 10.). There was no significant difference in the step length in the NW between OA and YA participants. However, OA exhibited smaller step length in FW ($t(54) = 5.00$, $p < 0.001$, $d = 1.34$) and TW ($t(54) = 4.19$, $p < 0.001$, $d = 1.12$). The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (OA; $t(27) = 7.27$, $p < 0.001$, $d = 1.21$, YA; $t(27) = 11.99$, $p < 0.001$, $d = 1.93$). Specifically for NW and FW comparisons (OA; $t(27) = -5.50$, $p < 0.001$, $d = 0.95$, YA; $t(27) = -13.64$, $p < 0.001$, $d = 2.20$).

Figure 25 – Right Step Length across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

Left Stance Time

The results showed that there was significant main effect of gait condition ($F (1.66, 86.22) = 28.90$, $p < 0.001$, $\eta_p^2 = .357$), sphericity violated (Table 9.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the left stance time for the NW and the left stance time for the TW conditions. There was a significantly greater left stance time for the TW than the FW conditions; $t (55) = -6.26$, $p < 0.001$, $d = 1.16$. There was also a significantly greater left stance time for the NW than the FW conditions; $t (55) = 7.28$, $p < 0.001$, $d = 1.12$.

The main effects were qualified by an interaction of gait condition and age ($F (1.66, 86.22) = 3.54$, $p = 0.042$, $\eta_p^2 = .064$), sphericity violated (Table 10.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the left stance time for all three-gait condition for the OA and the YA participants. The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for FW and TW comparisons (OA; $t (27) = -3.93$, $p < 0.001$, $d = 1.12$, YA; $t (27) = -6.42$, $p < 0.001$, $d = 1.42$). Specifically for NW and FW comparisons (OA; $t (27) = 3.56$, $p < 0.001$, $d = 0.71$, YA; $t (27) = 7.37$, $p < 0.001$, $d = 1.73$).

Right Stance Time

The results showed that there was significant main effect of gait condition ($F (2, 104) = 43.70$, $p < 0.001$, $\eta_p^2 = .457$) (Table 9.). The results of the follow-up paired sample t-test revealed that there was a significantly greater right stance time for the NW than the FW condition; $t (55) = 5.76$, $p < 0.001$, $d =0.94$. There was also a significantly greater right stance time for the TW than the NW condition; $t (55) = -3.88$, $p < 0.001$, $d =0.63$. There was also a significantly greater right stance for the TW than the FW condition; $t(55) = -9.40$, $p < 0.001$, $d = 1.33$.

In addition, there was a significant main effect of age ($F(1, 52) = 4.43$, $p = 0.040$, $\eta_p^2 = .078$) with OA having a greater right stance time (M=0.63, SD=0.098) compared to YA (M=0.59, SD=0.057). There was also a significant main effect of physical activity levels ($F(1, 52) = 4.31$, $p = 0.043$, $\eta_p^2 = .076$) with inactive participants having a greater right stance time (M=0.63, SD=0.083) compared to highly active participants (M=0.59, SD=0.077).

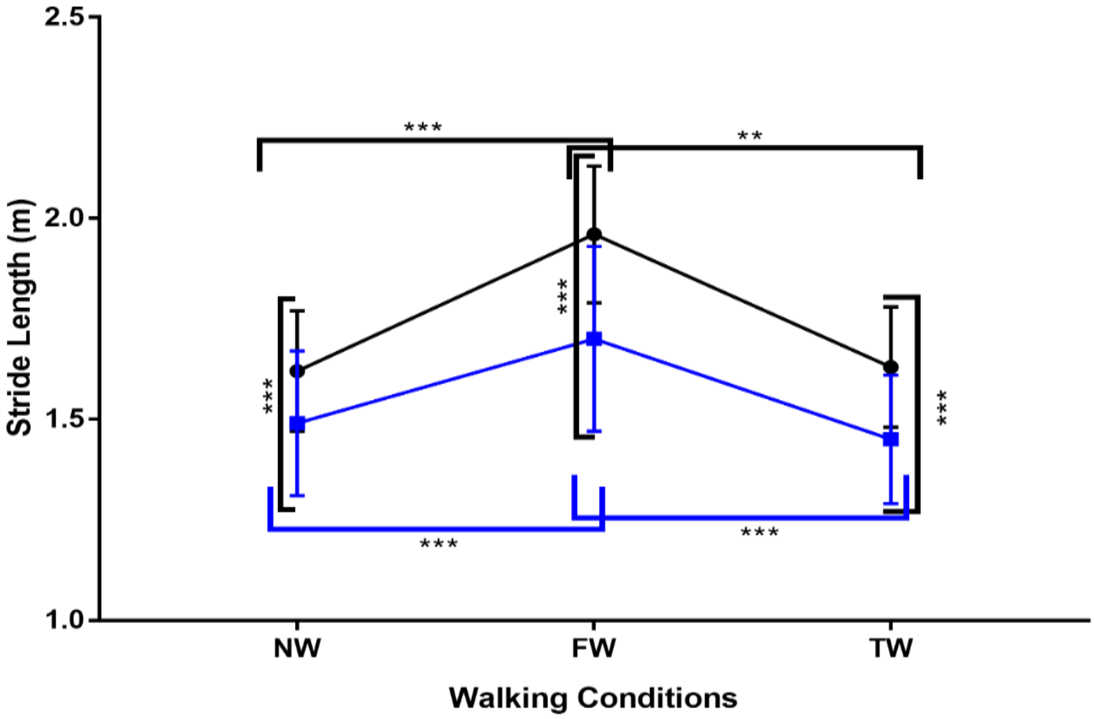
Stride Length

The results showed that there was significant main effect of the gait condition ($F(1.80, 93.54) = 144.39$, $p < 0.001$, $\eta_p^2 = .083$), sphericity violated (Table 9.). The results of the follow-up paired sample t-test revealed that there was no significant difference between the stride length for NW and the TW conditions. There was a significantly greater stride length for the FW than the TW condition; $t(55) = 13.87$, $p < 0.001$, $d = 1.39$. There was also a significantly greater stride length for the FW than the NW condition; $t(55) = -11.97$, $p < 0.001$, $d = 1.33$.

In addition, there was a significant main effect of age ($F(1, 52) = 8.64$, $p < 0.001$, $\eta_p^2 = .290$) with OA having a smaller stride length (M=1.54, SD=0.16) compared to YA (M=1.74, SD=0.14).

The main effects were qualified by an interaction of gait condition and age ($F(1.80, 93.54) = 6.17$, $p = .004$, $\eta_p^2 = .106$), sphericity violated (Table 10.). However, as with the main effect of age, OA exhibited a consistently smaller stride length in the NW ($t(54) = 2.77$, $p = 0.008$, $d = 0.74$), FW ($t(49.27) = 4.84$, $p < 0.001$, $d = 1.29$) and TW conditions ($t(54) = -1.85$, $p < 0.001$, $d = 1.39$). The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for FW and TW comparisons (OA; $t(27) = 7.40$, $p < 0.001$, $d = 1.32$, YA; t (27) = 14.50, p = 0.003, d = 2.09). Specifically for NW and FW comparisons (OA; t (27) = -5.70, < 0.001, d = 1.02, YA; t (27) = -15.50, p < 0.001, d = 2.16).

Figure 26 – Stride Length across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

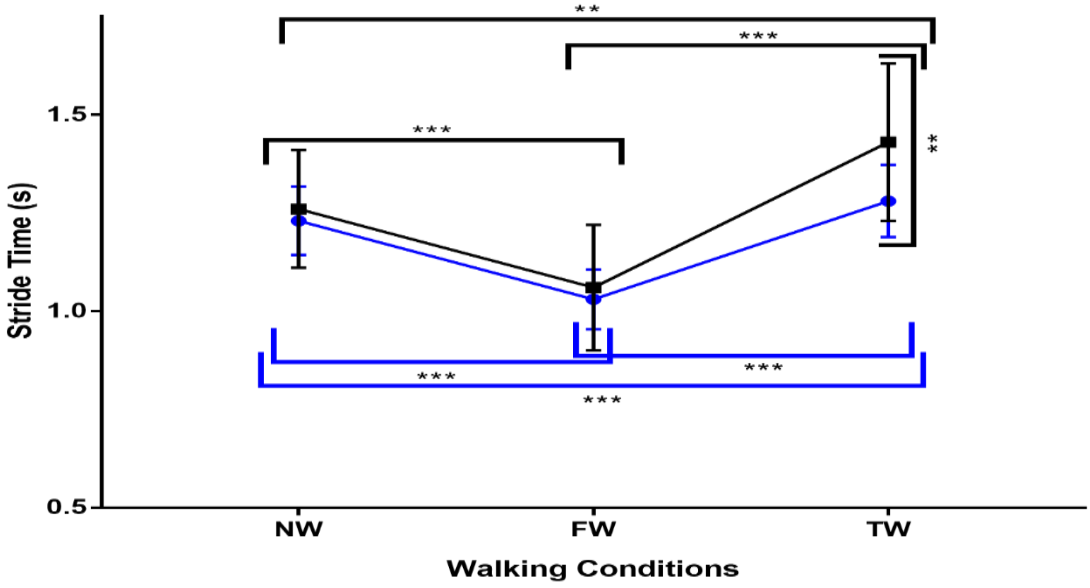
Stride Time

The results showed that there was significant main effect of gait condition (F (1.60, 83.08) = 116.74, $p < 0.001$, $\eta_p^2 = .692$), sphericity violated (Table 9.). The results of the follow-up paired sample t-test revealed that there was a significantly greater stride time for the TW than the NW condition; t (55) = -5.00, p < 0.001, d = 0.74. There was also a significantly greater stride time for the TW than the FW condition; t (55) = -12.17, p < 0.001, d = 2.60. There was also a significantly greater stride time for the NW than the FW condition; t (55) = 12.91, p < 0.001, d = 1.72.

In addition, there was a significant main effect of age ($F (1, 52) = 9.55$, $p = 0.003$, $\eta_p{}^2 = .155$) with OA having a greater stride time (M=1.25, SD=0.12) compared to YA (M=1.17, SD=0.068).

The main effects were qualified by an interaction of gait condition and age ($F (1.60, 83.08) = 4.15$, $p = .023$, $\eta_p{}^2 = .077$), sphericity violated (Table 10.). The results of the follow-up independent sample t-test revealed that there were no significant differences in the stride time in the NW and FW conditions for the OA and the YA participants. However, there was a significantly greater stride time in the TW condition for the OA than the YA participants; $t (38.30) = -3.71$, $p = 0.001$, $d = 0.99$. The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA. Specifically for NW and TW comparisons (OA; $t (27) = -4.54$, $p < 0.001$, $d = 0.97$, YA; $t (27) = -2.93$, $p = 0.007$, $d = 0.55$). Specifically for FW and TW comparisons (OA; $t (27) = -7.76$, $p < 0.001$, $d = 2.06$, YA; $t (27) = -13.54$, $p < 0.001$, $d = 3.28$). Specifically for NW and FW comparisons (OA; $t (27) = 6.56$, $p < 0.001$, $d = 1.31$, YA; $t (27) = 17.93$, $p < 0.001$, $d =2.80$).

Figure 27 – Stride Time across the Gait Conditions. Young Adults (Black circles) and Old Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

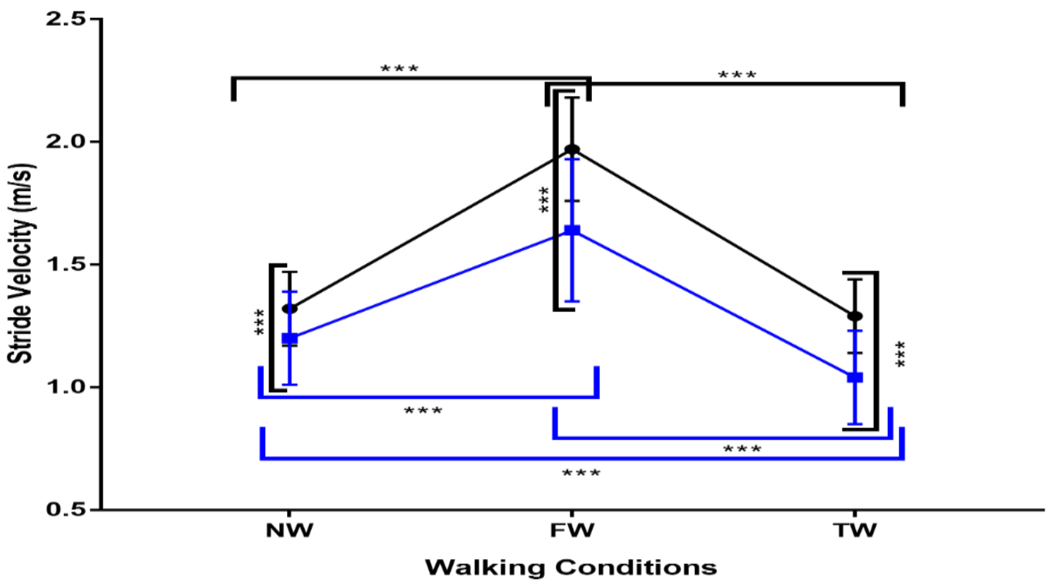
Stride Velocity

The results showed that there was significant main effect of gait condition (F (1.70, 88.56) = 279.91, p < 0.001, $\eta_p{}^2$ = .843), sphericity violated (Table 9.). The results of the follow-up paired sample t-test revealed that there was a significantly greater stride velocity for the NW than the TW conditions; t (55) = 4.08, p < 0.001, d = 0.51. There was a significantly greater stride velocity for the FW than the TW conditions; t (55) = 18.43, p < 0.001, d = 2.49. There was also a significantly greater stride velocity for the FW than the NW condition; t (55) = -16.71, p < 0.001, d = 2.21.

In addition, there was a significant main effect of age (F (1, 52) = 30.95, p < 0.001, $\eta_p{}^2$ = .373) with OA having a smaller stride velocity (M=1.29, SD=0.17) compared to YA (M=1.53, SD=0.14).

The main effects were qualified by an interaction of gait condition and age (F (1.70, 88.56) = 6.35, $p = .004$, $\eta_p{}^2 = .109$), sphericity violated (Table 10.). However, as with the main effect of age, OA exhibited a consistently smaller stride length in the NW (t (54) = 2.65, p = 0.010, d = 0.71), FW (t (54) = 4.88, $p < 0.001$, d = 1.30) and TW conditions (t (54) = 5.54, $p <$ 0.001, d = 1.48). The results of the paired samples t-tests performed independently on both groups revealed differences consistent with the main effect of gait condition for the initial ANOVA for the OA. Specifically for NW and TW comparisons; t (27) = 4.42, $p < 0.001$, d = 0.87. Specifically for FW and TW comparisons; t (27) = 10.44, $p < 0.001$, d = 2.49. Specifically for NW and FW comparisons; t (27) = -8.53, $p < 0.001$, d = 1.82. However, for the YA, there was no significant difference between the stride velocity for the NW and the TW conditions. There was a significantly greater stride velocity for the FW than the NW conditions; t (27) = -22.09, $p < 0.001$, d = 3.54 and for the FW than the TW conditions; t (27) = 17.61, $p < 0.001$, d = 3.78.

Figure 28 – Stride Velocity across the Gait Conditions. Young Adults (Black circles) and Older Adults (Blue Squares)



****: significant difference at P < 0.01, ***: significant difference at P < 0.001.**

Appendix E – Participant Informed Consent Form

Faculty of kinesiology and recreation management

UNIVERSITY OF MANITOBA

Research Project Title: The influence of age and physical activity on the control of mediolateral dynamic stability during walking

Student Investigator: Yash Rawal

University of Manitoba, Faculty of Kinesiology and Recreation Management

Lab: 179B Frank Kennedy Centre

(204) 887-2955

Sponsor: Manitoba Medical Service Foundation (MMSF)

This consent form, a copy of which will be left with you for your records and reference, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more detail about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

Purpose of this Study:

Sideways falls during walking are a common and significant health concern for many older adults. These falls often lead to hip fracture, which reduces independent mobility and can increase the risk of mortality. Such reduction of individual independence can place a large burden on family members, caregivers and the Canadian healthcare system.

Research has shown that physical activity and exercise programs have benefits for reducing fall risk and incidence especially among older adults. Despite considerable research, we lack specific understanding of why older adults fall during walking. From a biomechanical perspective, the control of muscle force output from the lower limbs not only generates forward movement during walking, but also controls the side-to-side movement of the whole-body centre of mass. This side-to-side control is challenging because the combined force output from the lower limbs must first stop and then reverse the direction of centre of mass during each step. Age- and physical activity related changes in lower limb force output may change centre of mass movement, which can lead to instability during walking.

The proposed work uses newly developed measures to better understand the biomechanical mechanisms underlying age- and physical activity related challenges in sideways balance control during walking. Better understanding of the specific mechanisms that contribute to sideways instability among older adults is important because it can improve the specificity of interventions, to target specific balance control challenges and reduce the risk of falls, injury and subsequent mortality.

Procedures Involved in this Study:

Being screened for falls during normal activities, age, head injuries, medication use, and filling out a PASB-Q to see whether an individual is achieving the Canadian Physical Activity Guidelines or not. This study consists an experimental set-up, followed by two walking conditions: (1) normal paced walking, (2) fast paced walking and (3) modified tandem walking. Total time including experimental set-up and the two walking conditions will take no more than three hours of your time.

Experimental Set-up:

- Anthropometric measurements (height, body weight and segment lengths) will be recorded at this time for use in the biomechanical model (the biomechanical model is a mathematical model of your skeleton that we use to determine the position of your whole-body centre of mass and the force transmitted across your joints).
- Reflective surface markers will be placed over various locations on your feet, legs, thighs, pelvis, trunk, arms and head, using double-sided tape placed on top of your clothing.
- You will be asked to stand upright and as motionless as possible, with your arms at your sides, for approximately 60 seconds. We will record the position of all the reflective markers on your body. This information will be used to build the biomechanical model (a mathematical representation) of all your body segments (e.g. arms, legs, trunk, and head) in space.

Normal paced walking:

In this condition, you will start with feet side-by-side and will be asked to begin walking at your normal pace, along a 10-metre walkway. Once you reach the end of the walkway, you can turn around and return to the start. Five trials of normal paced walking will be collected.

Fast paced walking:

In this condition, you will also start with feet side-by-side and will be asked to begin walking as fast as you can (without running), along a 10-metre walkway. Once you reach the end of the walkway, you can turn around and return to the start. Five trials of fast paced walking will be collected.

Modified Tandem Walking:

In this condition, you will also start with feet side-by-side and will be asked to begin walking at a normal speed but you will be walking with narrow step width as you possibly can, along a

10-metre walkway. Once you reach the end of the walkway, you can turn around and return to the start. Five trials of the modified tandem walking will be collected.

Recording Devices:

During all trials, a motion analysis system will record the position of each reflective spherical marker you have placed on your body. The cameras that record the position of these reflective markers only respond to infrared light and are not capable of recording images of anything other than the reflective markers (i.e. it is not possible to see images of your person, as you would see with a typical video camera). The information we obtain from the position of these reflective markers is fed into the biomechanical model and used to compute the position of your whole-body centre of mass and the forces that are transmitted across your joints.

We will also record the forces that you exert on the ground, using a force platform. A force platform is similar to a typical bathroom scale, except that a force platform also responds to forces applied in the front-to-back and side-to-side directions, in addition to forces in a downward direction.

Benefits of Participation:

Apart from the opportunity to learn about how humans control their balance during walking, there are no direct benefits to you from participating.

Risks to Participation and Associated Safeguards

As you are walking along a level walkway with no tripping hazards, the risk to you is no different than if you were walking outside on a level surface. Nevertheless, the risk of falling does always exist when you are moving. A research assistant or I will be constantly observing your walking and will be within an arm's reach of you, should you experience any instability.

It should be noted that the principal investigator has observed over 500 walking trials (of similar characteristics to those being administered here), without an incidence of falling.

In some individuals, the adhesive tape used to affix the reflective markers to the skin has caused some redness and discomfort. If you have an allergy to adhesive tape or bandages, please make one of the investigators in the room aware of this and testing will not proceed. If, during testing, you notice that you are experiencing redness or discomfort at the sites of the reflective markers, please make of the investigators in the room aware of this and testing will stop immediately.

Anonymity and Confidentiality of Data:

As the motion analysis cameras only record the position of the reflective markers located on your body, this data is completely anonymous. There is also no way to identify you from the forces you apply to the force platforms. You will be identified only by a participant identification code, which contains no personally identifiable information. These codes contain only a number and cannot be linked back to any specific person.

This consent form, the physical activity questionnaire and the bottom portion of the participant feedback form, which will contain your name and signature, will be kept in a locked filing cabinet in the principal investigator's office for three years after the completion of the study (the principal investigator's office is located behind two locked doors). Only the principal investigator will have access to the consent and feedback forms. After this time, the consent, the physical activity questionnaire and feedback forms will be destroyed via a file shredder.

The principal investigator and research assistant (graduate student) will have access to the biomechanical data (motion analysis and force platform data) collected during this study. This information, which is anonymous, will be retained indefinitely on a password-protected computer in the principal investigator's office.

Remuneration:

Upon arriving to the university, we will have provided a parking pass to cover the cost of your parking (Lot X). If you took public transportation to the university, we will have provided bus tickets to reimburse the cost of travelling to the university and to cover the cost of your return trip home.

Changing Your Mind about Participation

You may withdraw from this study at any time without any negative consequences. To do so, indicate this to the researcher or one of the research assistants by saying, "I no longer wish to participate in this study", or similar statement.

Participant Feedback

After your participation in the study, you will have the opportunity to discuss the research with the principal investigator, should you have any questions. As the raw data collected take some time to process in order to obtain any interpretable results, you will have the option of indicating that you would like a summary of the research results following the completion of the study. Results should be available during the summer of 2018. On the participant feedback form, you will be able to either provide a mailing address or email address to which we will send the results.

Dissemination of Results:

Results of this study will be presented at academic conferences (such as the Congress of the Canadian Society for Biomechanics or the International Society for Posture and Gait Research). Data will also be published in manuscript format (such as in the Journal of Biomechanics or the journal Gait and Posture). Data will be presented as group average values – there will be no information presented that could identify you as a participant in this study.

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal rights nor release the researchers, sponsors, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time, and /or refrain from answering any questions you prefer to omit, without prejudice or consequence. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation.

The University of Manitoba may look at your research records to see that the research is being done in a safe and proper way.

This research has been approved by the Education/Nursing Research Ethics Board. If you have any concerns or complaints about this project, you may contact any of the above-named persons or the Human Ethics Coordinator at 204-474-7122. A copy of this consent form has been given to you to keep for your records and reference.

Participant’s Signature ________________________ Date ____________

Researcher and/or Delegate’s Signature ___________________ Date ___________