Comparing Biomechanical Variables in Older Adult Fallers and Non-Fallers

by

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ABSTRACT

Falling in the older adult population is a major concern among health care professionals. Several studies have compared the biomechanical differences between older adults with and without a history of fall. However, these studies have mainly focused on the participants' Center of Pressure (COP) and have neglected other biomechanical variables, like Ground Reaction Force (GRF) and Plantar Pressure measures. The purpose of this study was to compare different biomechanical variables (COP, GRF, and Plantar Pressure) between older adult fallers and non-fallers during walking and standing to gain further insight regarding falling in the aging population. 102 older adults (41 fallers, 61 non-fallers) were asked to walk on a walkway with embedded pressure mats, mounted on top of force plates. Additionally, the participants were asked to stand in two different conditions: Eyes Closed Narrow Stance (ECNS) and Eyes Open semi-Tandem Stance (EOTS), for 30 seconds. The results of the study indicated that the GRF and COP measures for the walking tests, and the COP and Plantar Pressure measures for the standing tests were significantly different between the two groups. Moreover, among the two standing test conditions the ECNS was more appropriate for measuring biomechanical variables. This study provides useful data with respect to falling in the older adult population and researchers can use this information to identify older adult who are at risk of falling. Consequently, through proper intervention the rate of falls can be reduced in the aging population.

DEDICATION

To Gene Roddenberry, may your vision of the future one day come true!

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List of Symbols, Nomenclature or Abbreviations

COP: Center of Pressure

COG: Center of Gravity

GRF: Ground Reaction Force

COM: Center of Mass

COF: Center of Force

BOS: Base of Support

ΔΦ: Rate of change in the direction of COP displacement

ECNS: Eyes Closed Narrow Stance

EOTS: Eyes Open semi-Tandem Stance

PkCP: Peak Contact Pressure

FTI: Force-Time Integral

PTI: Pressure-Time Integral

MP: Mean Pressure

PP: Peak Pressure

AP: Anterior-Posterior direction

ML: Medial-Lateral direction

AP GRF: Anterior-Posterior component of the GRF

V GRF: Vertical component of the GRF

Fz: The force in the vertical direction

Fx: The force in the AP direction

Fy: The force in the ML direction

Fz1: First peak of the vertical component of the GRF which occurs after heel strike in gait

Tz1: Time it takes for the first VGRF peak to occur

Fz2: Minimum GRF's vertical component which occurs during mid stance in gait

Tz2: Time it takes foe Fz2 to occur

Fz3: Second peak of the GRF's vertical component which occurs after heel strike in gait

Tz3: Time it takes for the second VGRF peak to occur

Fx1: First peak of the AP GRF, which represents the maximum breaking force

Tx1: Time it takes for Fx1 to occur

Fx2: AP GRF crossing zero

Tx2: Time it takes for Fx2 to occur

Fx3: Second peak of the AP GRF, which represents maximum propulsion force

LR: The Loading Response sub-phase of a walking trial

MSt: The Mid-Stance sub-phase of a walking trial

TSt: The Terminal-Stance sub-phase of a walking trial

PS: The Pre-Swing sub-phase of a walking trial

COP_T_Disp: The COP total displacement

COP_T_Disp_SD: The variability in the COP total displacements

COP_{ML}_Disp: The COP displacement in the ML direction

COP_{ML}_Disp_SD: The variability in the COP ML displacements

COP_{AP}_Disp: The COP displacement in the AP direction

COP_{AP}_Disp_SD: The variability in the COP AP displacements

Fz_{Stand}_SD: Standard Deviation of the average vertical force

Fy_{Stand}_SD: Standard Deviation of the average ML force

Fx_{Stand}_SD: Standard Deviation of the average AP force

COP_{T}-Vel_{Stand}: Average velocity of total COP displacement

COP_{AP}-Vel_{Stand}: Average velocity of the COP displacement in AP

COP_{ML}-Vel_{Stand}: Average velocity of the COP displacement in ML

$COP_{EllArea}$: 95% confidence ellipse area around COP coordinates during standing

SampEnt-$COP_{\Delta\Phi}$: The sample entropy of $\Delta\Phi$

BW: Body Weight

1. Introduction

Aging is a complex process that is associated with a progressive loss of muscle mass, strength, power, endurance, bone density, and overall functional capacity (Gregory & Travis, 2015). The loss of strength and general neuromuscular deconditioning that stems from aging, reduces an individual's ability to perform daily living tasks and increases the risk of injury. Falls pose considerable health problems among the older adult population including frailty, immobility, and decline in functional ability (Berg & Cassells, 1992). In 2005, Canada's Public Health Agency released a report indicating that the number of fatalities resulting from falls for people aged 65 and older is 9.4 per ten thousand population (Government of New Brunswick, 2013).

In order to properly identify people who are at risk of falling we must gain more insight regarding biomechanical differences between older adults who have experienced a fall and those who have not. The most commonly used laboratory equipment for this purpose are force plates. Force plates are the gold standard for analysing a person's Center of Pressure (COP), moments, and forces, through dynamic and static trials. Force plates, however, cannot provide information on an individual's plantar pressure distribution (Cavanagh & Rodgers, 1985).

Pressure sensitive devices, such as pressure mats and insoles, are designed to measure the pressure created under an individual's foot; they can provide researchers with useful information regarding gait disorder and falls

(Mickle, Munro, Lord, Menz, & Steele, 2010). These pressure sensitive devices can measure one's COP as well as plantar pressure distribution during different activities. Devices able to measure the plantar pressure distribution allow researchers to assess the interaction of the foot and the supporting surface and provide a quantifiable record of a person's pressure patterns, which can in turn be used to monitor changes following prevention programs (Orlin & McPoil, 2000). Finally, the presence of foot pain can cause gait and balance disturbance, and is linked with high plantar pressure generated during gait (Burns, Crosbie, Hunt, & Ouvrier, 2005; Hodge, Bach, & Carter, 1999). Increased plantar pressure during gait can predispose older individuals to falling (Mickle et al., 2010).

Mickle et al. (2010) investigated the pressure generated underneath older adults' feet during walking using a pressure platform. Their findings revealed that older individual fallers generated a significantly higher peak pressure and pressure-time integral under their foot in comparison to older adult non-fallers. This study claimed to be the first to assess the differences in plantar pressure between older adult fallers and non-fallers; however, it did not consider non-pressure variables, which can be obtained through force plates.

Tracking older adults' COP has been explored thoroughly by researchers and shown to be significantly different between fallers and non-fallers. Conceivably, the medial-lateral and anterior-posterior shear force that can be obtained from GRF, which has been found to be equal to the accelerations of an individual's Center of Mass (COM) when divided by the

body mass (Winter, 1995), can provide information regarding a participant's stability during walking. Additionally, the increased plantar pressure observed in older adult fallers during gait, when compared to non-fallers, can provide important information regarding falls in this population. Hence, incorporating additional quantitative laboratory measurements (GRF, peak pressure, and pressure integral) can lead to an improved understanding of biomechanical differences between older adult fallers and non-fallers.

Finally, force plates and pressure mats can each provide unique data that cannot be obtained by using one type of device alone. Although previous research has explored them individually, to the best of the author's knowledge, no other research has aimed to compare the data obtained from both pressure mats and force plates between older adult fallers and non-fallers. Investigating different biomechanical variables in this population can lead to future research that can use these variables to better identify fallers.

1.1 Purpose

The purpose of this study is to compare plantar pressure and ground reaction force data during gait and standing in older adult fallers and non-fallers. To this end, individuals over the age of 65 were divided into two groups of fallers and non-fallers based on their history of fall. Prior to data collection, the participants were assessed using the Berg Balance Scale (BBS), and Brief-Balance Evaluation Systems Test (Brief-BESTest). The biomechanical data were measured during standing and walking trials and the obtained data were

divided into three sets: COP measures, GRF measures, and plantar pressure measures. Two Tekscan HR mats (Tekscan Inc., Boston, MA, USA) and two Kistler piezoelectric platforms (Kistler Instrument Corp., Switzerland) were used for data collection. The information obtained in this study will be useful for:

- Examining differences in plantar pressure measures between older adult fallers and non-fallers.
- Examining differences in GRF measures between older adult fallers and non-fallers.
- Examining differences in COP-measures between older adult fallers and non-fallers.
- Gaining a better insight into differences between adult fallers and non-fallers during gait and standing.
- Providing information for future research to build predictive models for identifying older adults who are at risk of falling.

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2. Literature Review

One of the major concerns in countries with an aging population is the increasing rate of falling in their older adult population (World Health Organization, Noncommunicable Diseases and Mental Health Cluster, Noncommunicable Disease Prevention and Health Promotion Department, & Ageing and Life Course, 2002). As an individual ages, some physiological changes may occur, such as sight and hearing difficulties, changes in the musculoskeletal system, a decrease in the antigravity muscle strength, and changes in the central nervous system. These changes can lead to experiences such as an increase in postural sway with instability (Nochajski, 2000). Delbaere et al. (2006) assert, "The etiology of falls is multifactorial, since medical, sensory, psychological, postural control as well as physical variables provide a predictive value to identify frequent fallers" (p. 87).

Consistent with this position, other researchers also urge healthcare professionals to challenge the view of a single measurement being a surrogate for fall risk in older adult-dwelling communities, such as nursing homes. Instead, they encourage practitioners and researchers to embrace multiple measurements in order to gain the most accurate data that could both predict falls and ultimately prevent them (Thapa, Gideon, Brockman, Fought, & Ray, 1996a).

2.1 New Brunswick's Aging Population and Falling

The demographic shift in New Brunswick has resulted in the Maritime's province having the second oldest population among any region in North America. Considering that in New Brunswick, people over the age of 65 represent more than 18% of its residents, the province's aging population is affecting the healthcare system and impacting older adult care. By 2030, residents of New Brunswick who are 65 years and older and young people under the age of 14 will outnumber working-age adults (Government of Canada, 2014). Reflecting on the increasingly older population in New Brunswick, it is essential to consider some of the issues impacting the province's older adults, including preventative measures regarding falling.

According to Hausdorff, Edelberg, Mitchell, Goldberger, & We (1997), one in every three individuals over the age of 65 will suffer one or more falls in a given year. In a province with a growing aging population, like New Brunswick, this is problematic (Government of New Brunswick, 2013). The increase in the age of New Brunswick's population has led to a greater risk of falling and a more pronounced need for efficient ways of identifying those who are at high risk of falling. Fall among individuals 65 years and older also results in debilitative injuries in some instances, which in turn can lead to an anticipation of a greater number of injuries from falls every year. Therefore, it is crucial for healthcare professionals to detect subtle changes in postural stability in an effort to prevent falls. In this context, laboratory-based assessments offer a unique perspective regarding balance that could enable

researchers and healthcare professionals to gain insight in biomechanical differences between older adult fallers and non-fallers during gait and standing.

2.2 Falling and its Impact on the Healthcare Sector

The Center for Disease Control and Prevention (CDC) and the National Center for Injury Prevention and Control (NCIPC) categorize falls as intentional or unintentional events. Both agencies affirm that older adults are more likely to experience unintentional falls (Greenberg, 2014). In the United Kingdom, up to thirty-percent of mild to severe injuries result from falls among older adults; approximately 15% of all emergency visits are attributed to falls (Scuffham, Chaplin, & Legood, 2003). In Canada, over 50% of injury-related hospitalizations among individuals who are 65 years and older have been accredited to falls (Scott, Pearce, & Pengelly, 2005).

In the Canadian province of British Columbia, the duration of hospital stays following a fall injury is considerably longer than other injuries and could last up to 15 days (Herman, Gallagher, & Scott, 2006). Consequently, the prolonged hospitalization experience can impose enforced immobilization, accelerated bone loss, and sensory deficiencies, which can lead to irreversible functional decline (Creditor, 1993). However, the extent of stay in the hospital, does depend on the kind of injury incurred. In this respect, individuals suffering from hip fractures tend to stay the longest in the hospital, and 20% are expected to die within a year of sustaining their injuries (Zuckerman, 1996). In view of the relatively high mortality rate associated with hip fracture

(Roche, Wenn, Sahota, & Moran, 2005), a comprehensive review of studies over a period of ten years found that in comparison "to the expected mortality among people of similar age and sex … mortality one year after hip fracture was three to four times higher than expected" (Harris, Cumming, Watts, Ebeling, & Crowley, 1998).

Despite efforts on the part of healthcare professionals to minimize injuries from falls among the older adults, there is a paucity of data regarding the systematic assessment of the principle causes of falls and effective preventative measures (Close et al., 1999). Greater accuracy in identifying older adults who are at risk of falling can also curb the costs imposed on the healthcare sector, as well as decrease the physiological and psychological effects associated with these incidents among the older adults.

Some of the more prominent complications linked to falling include: intracranial hematomas; injury of internal organs; and fractures of the humerus, wrist, pelvis, and femur (Pasquetti, Apicella, & Mangone, 2014). In addition to these physical injuries, falls in the older adult community can lead to a gradual loss of confidence and independence, as well as intensify morbidity (Ilha, Gautério, & Backes, 2014). This can lead to self-imposed restrictions on the daily physical activities of the older adults, which can in turn increase the risk of falling (Tinetti, Leon, Mendes, Doucette, & Baker, 1994). Individuals who experience falls potentially end up living in a constant state of fear that they may fall again, which negatively impacts their quality of life. Fear of falling among the aging population can also become the root of more serious

problems by causing inactivity and deconditioning, subsequently increasing the risk of falls among this age group (Hindmarsh & Estes, 1989). The psychological ramifications of such fears have been associated with the "post-fall-syndrome", which usually leads to restrictions in activities and gradual loss of independence. The effect of these restrictions can directly lead to different forms of balance deterioration as well (Myers et al., 1996). Falls have also been shown to be a driving force behind social withdrawal, depression, and confusion among the older adults (Nevitt, Cummings, & Hudes, 1991). Also, falling leads to premature transferring of older adults to senior centers and institutions, which imposes extra costs on the healthcare sector. Another consequence of falling is the increased risk of hospitalization, which also results in an escalation of healthcare costs.

Therefore, it should be no surprise that medical health professionals have become increasingly concerned about the hazards of falling among older adults. This includes required post-discharge assistance from hospitals and rehabilitation costs for individuals who have experienced falls. According to the Public Health Agency of Canada, the direct costs associated with falling among older adults in 2004 has been estimated to be over $2 billion annually. Hence, falling not only burdens the healthcare system, but also inadvertently sends many older adults to nursing homes when they could still enjoy an independent life. This raises another problem: the potential lack of proper facilities to deal with the growing older adult population and the inevitable higher number of fall cases. In this context, higher healthcare demands and

costs on the New Brunswick Medicare system are among some of the issues that need to be addressed. Lack of professionals to properly deal with the needs of the older adult population who have suffered from falls, the negative effect of falling on the quality of life in a vulnerable population, as well as the issue of limited hospital space and the need for long-term hospice care are other concerns that must be effectively tackled.

2.3 Falling from a Biomechanical Perspective

Considering some of the many problems healthcare professionals and regulatory agencies have to deal with, it is imperative to understand the phenomenon of falling from a biomechanical perspective. Falls are some of the more common external causes of unintentional injury. In an extensive literature review, thirty definitions of falls were identified with the one offered by the Kellogg International Working Group on Fall Prevention in older adults being among the more prominent descriptions of falling. The Kellogg Group defines falling as “an event which results in a person inadvertently coming to rest on the ground or other lower level and other than as a consequence of the following: sustaining a violent blow, loss of consciousness, sudden onset of paralysis as in a stroke, epileptic seizure” (Zecevic, Salmoni, Speechley, & Vandervoort, 2006, p.368). Additionally, a compelling and illustrative three-phase description of falls has been presented in the literature:

The first phase is an initiating event that displaces the body's COM beyond its base of support. Initiating events involve extrinsic factors such as

environmental hazards; intrinsic factors such as unstable joints, muscle weakness, and unreliable postural reflexes; and physical activities in progress at the time of the fall (Berg & Cassells, 1992).

This is followed by the second phase where individuals are unable to prevent falls due to "a failure of the systems for maintaining upright posture to detect and correct this displacement in time to avoid a fall" (Berg & Cassells, 1992, p.264). Neuromuscular and sensory impairment including the loss of sensory function, muscle weakness, and impaired central processing are among several factors attributed to the failure in this phase. The next phase involves the impact the body has on "environmental surfaces, usually the floor or ground, which results in the transmission of forces to body tissue and organs. The potential for injury is a function of the magnitude and direction of the forces and the susceptibility of tissues and organs to damage" (Berg & Cassells, 1992, p.264)

While adopting an operational definition of falls has been problematic for researchers, the World Health Organization (WHO) has described falls as "inadvertently coming to rest on the ground, floor or other lower level, excluding intentional change in position to rest in furniture, wall or other objects" (World Health Organization, 2008, p.1). This definition of fall is adopted for the present study. Additionally, accidental events, such as furniture collapse, are excluded from the definition of a fall (Shumway-Cook, Brauer, & Woollacott, 2000).

2.4 The Fall Conundrum

Through an in-depth exploration of falls, researchers and health care professionals can gain deeper insight into the most effective intervention and preventative strategies in managing this problem. This is particularly valuable in dealing with intervention tactics that can help familiarize older individuals with the risk factors involved in falling, particularly factors that can be modified through the adoption of the appropriate personal or environmental strategies (Zecevic et al., 2006). Accordingly, aging can be regarded as the consequence of numerous functional, neural, muscular, and bone-related problems (Granacher, Gollhofer, Hortobágyi, Kressig, & Muehlbauer, 2013). For instance, the knowledge that lower extremity muscle strength can be reduced by up to 40% over the years (Arnold, 2008), can be useful in understanding the root causes of falls. Studies show that a number of risk factors including unsteady gait, muscle weakness, problems with balance, postural instability, and the use of different medications as well as environmental hazards can attribute to falling among older adults (Rubenstein, 2006). Additionally, increased risk of falling in older adults has also been associated with increased body sway and reduced postural stability (Carter, Kannus, & Khan, 2001; Lord, Ward, Williams, & Strudwick, 1995). As stated by Pasquetti, Apicella, & Mangone (2014), multidisciplinary and multidimensional approaches need to be adopted in evaluating the senior population at risk of falls.

2.4.1 Falls and Prevention Measures

According to (Rubenstein, 2006), systematic fall risk assessment and targeted interventions are among several programs that have proven to be both effective and cost-efficient. Different and even contradictory results have been reported on the effects of different types of fall prevention programs on older adults. While some studies claim that different types of trainings have shown to improve balance in older adults (Baker, Atlantis, & Singh, 2007; Häkkinen et al., 1998), others seem to disagree (Granacher, Gollhofer, Hortobágyi, Kressig, & Muehlbauer, 2013; Wolfson et al., 1996).

The inconsistencies in the research could be due to different prescribed intensity and frequency of the training programs (Chandler, Duncan, Kochersberger, & Studenski, 1998; Wolfson et al., 1996). In addition, these varied results may be attributable to the range of balance and postural assessment techniques used to investigate the balance improvements of their participants. Most of these training studies have used simple measures of balance, such as the length of time a person can stand on one leg (Schlicht, Camaione, & Owen, 2001), the ability to stand on a tilt board (Buchner et al., 1997), and the time required to perform a sit-to-stand motion (Schlicht et al., 2001; Wolfson et al., 1996). Clinical tests have certain limitations when it comes to identifying subtle changes in postural stability.

Consequently, some concerns regarding the validity of functional tests and their ability to identify potential fallers. One of the issues that has raised red flags is the subjective nature of these tests. Recent technological advances

have steered the direction of balance assessments to more objective approaches. Functional tests are subjective in nature, prone to show ceiling effect, and lack the sensitivity to detect if individuals have experienced progress or deterioration in their balance measures in small degrees. The high sensitivity of techniques that enable a more objective assessment of balance can positively impact the identification of those who are at risk of falling (Mancini & Horak, 2010).

According to Brauer, Burns, & Galley (2000), laboratory-based assessments can offer us insight into control processes and biomechanical changes related to balance. Individuals with a history of fall tend to show more postural sway than those who have not reported a fall (Fernie, Gryfe, Holliday, & Llewellyn, 1982). A study conducted by Maki, Holliday, & Fernie (1987), confirmed this statement by revealing that the Center of Pressure (COP) motion in the medial-lateral direction is among the most effectual predictors of fall within the elder community. It is in this context that laboratory measures of balance and gait can allow for a more personalized approach that can be used to measure and ultimately treat a person's gait or posture disturbances (Giladi, Horak, & Hausdorff, 2013).

2.4.2 Fall and Balance Assessments

In the field of mechanics, the term balance is described in accordance to Newton's First Law as the state of an object when the resultant load actions acting upon the said object reaches zero (Özkaya, Leger, Goldsheyder, &

Nordin, 2016). An object's ability to reach equilibrium and remain in a static state is directly linked to the position of the Centre of Gravity (COG) and the area of the Base of Support (BOS) of the object in question. An object can be considered balanced when its line of gravity falls within its BOS. An object's balance is disturbed, and consequently falls will occur, if the line of gravity no longer remains within the BOS of the object (Pollock, Durward, Rowe, & Paul, 2000). According to Mancini & Horak, (2010), sustaining balance requires restoring the body's Centre of Mass (COM) relative to the BOS. Therefore, to maintain balance, the body's COG, which is the vertical projection of the COM, must stay within the BOS. In addition, a specific posture needs to be maintained that enables voluntary movement and facilitates reaction to external disturbances. In this respect, it is paramount to embrace a more comprehensive and accurate assessment of balance to offset the physical (injuries due to falls) and psychological (fear of falling) impacts of balance disorders. Postural stability is attained when an individual is able to maintain, achieve, or restore a particular balance state (Geldhof et al., 2006). Postural stability is an essential component of physical function, particularly, the ability to stand and walk without relying on assisting devices. Postural stability can be compromised due to a number of factors, including aging and neurodegenerative diseases, which can potentially increase the risk of falls in everyday activities (Mancini & Horak, 2010).

One of the most common ways of quantifying postural stability is through the movements of the subject's Center of Pressure (COP) throughout

time (Granat, Barnett, Kirkwood, & Andrews, 1991). Due to the attraction of objects and bodies to the center of the earth, a force of attraction proportional to the body's mass is applied to the body. The resultant of all these forces, which are equal to the weight of the body, passes through a single point on the body, which is referred to as the COG. When the body's COG, which is the vertical projection of the COM, stays within the BOS during stance, balance is maintained. The BOS contains the plantar parts of the feet and the in-between space. Humans can manipulate the magnitude and location of the Ground Reaction Force (GRF) between the plantar part of the foot and the ground, allowing the COM to stay in the BOS (Winter, 1995). Small COM corrections that occur during the time an individual maintains their balance can be recorded and analyzed. However, estimating the location of the COM requires complex and expensive instrumentation; therefore, researchers use the simpler COP to indirectly measure postural sway. The COP is the point of application of the GRF vector and is defined as the projection of the Center of Gravity (COG) into the ground (Van Deursen, and Everett, 2010).

The biomechanical importance of the COP has been obvious to researchers even before it was quantified. Utilizing the COP provides researchers with constant values, regardless of the operator or measuring device (Han, Paik, & Im, 1999). The COP has been used as a measurement of balance and foot function during standing and walking. The COP can be described as the origin of the ground reaction force vector, or the centroid of external forces acting on the plantar surface of the foot and the shoe. Therefore,

there are several devices that can be used to capture the COP, including: force plates, pressure plates, and in-shoe pressure measurement systems. A number of studies have illustrated that the measurement of the position and displacement of COP with the use of a force plate is among the most common techniques employed by practitioners and researchers (Geldhof et al., 2006). It is noteworthy to mention that only a few studies have used pressure measuring devices instead of force plates to measure the postural sway of older adults (McGrath et al., 2012).

2.4.3 Common methods and metrics in obtaining COP

As stated by Piirtola & Era (2006), the measurements extracted from a participant's COP can help identify future falls and recurrent fallers. However, the difference in the methods used in measuring the participants' postural sway and the various ways of analysis, calls for more studies in this area. Various methods have been proposed for analyzing the COP of individuals to estimate their postural stability. According to Norris, Marsh, Smith, Kohut, & Miller, (2005), the "traditional" COP measures **(**COP range, sway path length, sway area, and average sway velocity**)** are significantly different between individuals who are at high risk or low risk of falling. In addition, the average velocity of the COP displacement in both anterior-posterior and medial-lateral directions are among the most frequently used methods (Piirtola & Era, 2006). The sway area has also been the subject of interest for many researchers for measuring postural steadiness while standing. (Hasselkus & Shambes, 1975;

Hughes, Duncan, Rose, Chandler, & Studenski, 1996; Lichtenstein, Burger, Shields, & Shiavi, 1990; Vieira, Oliveira, & Nadal, 2009; Wollseifen, 2011).

Rhea, Kiefer, Haran, Glass, & Warren (2014), recently introduced a new metric ($\Delta\Phi$) for analyzing postural sway. $\Delta\Phi$ describes the rate of change in the direction of COP displacement and provides more insight into postural steadiness. Rhea and his colleagues claimed that by only using the magnitude of the COP resultant vector and neglecting the direction of these vectors, we might overlook abrupt changes in the direction of the vector that could provide valuable information regarding postural stability. The authors compared this measure between a clinical population with an anterior cruciate ligament (ACL) rupture and healthy subjects. The healthy subjects demonstrated more irregular behavior in their $\Delta\Phi$ measures compared to the clinical population. The authors attributed this finding to the healthy subjects' ability to produce adaptive behavior, exploring their boundaries of postural control. On the other hand, the clinical population demonstrated a lower sample entropy value for the $\Delta\Phi$ measure, which they interrupted as a maladaptive behavior, demonstrating difficulty in adjusting their behavior to respond to perturbations. The sample entropy of $\Delta\Phi$, however has yet to be used for the evaluation of stability in the older adult population with respect to falling.

The most common methods to track the COP of an individual during different activities are through force plates and pressure measuring devices, such as pressure mats, pressure insoles, and electronic walkways. Force platforms use the COG and track the vertical component of the GRF, while

pressure measuring devices have many sensory elements and use the distribution of the force across the sensor grid (Gerbino, Griffin, & Zurakowski, 2007). Previous research has shown, however, that the COP obtained from pressure measuring devices and force plates are nearly identical and can be used interchangeably (Barnett, Cunningham, & West, 2000; Chesnin, Selby-Silverstein, & Besser, 2000).

2.4.4 Force Plates

A force plate is a rigid platform that measures the reaction of the foot on its surface using strain gauges. These gauges are arranged in a manner to produce outputs dependent on the forces and moments in three mutually perpendicular axes on the surface of the plate (Grundy, Tosh, McLeish, & Smidt, 1975). A force plate is an equipment specialized in measuring the GRF of any object standing or moving across its surface. Advanced models of force plates are capable of measuring the three-dimensional components of GRF (Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2013). These components include, the vertical force, the medial-lateral and anterior posterior shear force, and the corresponding moments about each axis during a gait trial. The piezoelectric measurement technology utilized in certain types of force plates, makes these instruments capable of detecting the slightest changes in COP and gait patterns. Force platforms use the COG and track the vertical component of the GRF for measuring the COP (Gerbino et al., 2007). Variables that could be assessed using the force curve produced by force plates during

gait include loading rates, COF patterns, and moments applied by the foot (Chester, Biden, & Tingley, 2005).

Walking is one of the most frequent dynamic activities in a person's daily life—which is why it is important for an individual's stability to be assessed during walking. In clinical settings, the force plate is deemed as a widespread tool used in the study of locomotion and it is used to measure different components of the ground reaction force (Chester, Biden, & Tingley, 2005).

The majority of the studies that have focused on older adult fallers using laboratory measurements have mainly focused on the oscillation of COP during standing or while performing balance tests (Boulgarides, McGinty, Willett, & Barnes, 2003; Brauer et al., 2000; Stel, Smit, Pluijm, & Lips, 2003; Thapa, Gideon, Brockman, Fought, & Ray, 1996).

In addition to many studies that have assessed older adults' postural sway using force plates while standing, the gait of this population has also been investigated. Studies have examined age related differences in the variability of COP movement during normal walking (Bizovska et al., 2014; Stergiou, Giakas, Byrne, & Pomeroy, 2002), and focused on recognizing specific velocity-independent, characteristics of gait that change with age (Kerrigan et al., 1998). The differences between the gait of older individuals who are at risk of falling and those who are not at risk has been also assessed by researchers: for instance, Uemera et al. (2012), evaluated the difference in gait initiation between older adults with fear of falling and those with no fear of falling. The

researchers assessed the COP data of 57 participants (age: 79.2 ± 6.8) under single and dual task conditions using a force platform, and showed that subjects with fear of falling show specific deficiencies in balance control compared to subjects with no fear of falling. Additionally, Bizovska et al. (2014), suggest that differences in COP during gait between fallers and non-fallers, could be associated with fall risk and calls for more research in this area. According to de-Souza, Svoboda, Bizovská, & Lehnert (2017), the potential of variability of COP displacement during walking for fall risk prediction has not yet been sufficiently observed.

Notwithstanding the ability of force plates to measure the distribution of vertical load under the foot, there is still a need for greater accuracy regarding how the pressure on the plantar surface of the foot is distributed with respect to the supporting surface. The precision displayed by force plates fails to offer the necessary insight into the distribution of the loads over the planar surface of the foot (Cavanagh & Rodgers, 1985). This is while a plantar pressure measurement device is capable of offering detailed information on plantar pressure as the foot makes contact with the floor. Force plates offer significantly faster sampling rates and greater accuracy, but they do not provide insight into the distribution of the load over the plantar surface of the foot (Rosenbaum & Becker, 2003).

The utilization of pressure measuring devices can also give researchers the opportunity to gain a better understanding of the pressure generated under an individual's feet during various activates. The data derived through the

distribution of pressure between the sole of the foot and the ground can help researchers gain further insight regarding the function of the foot during walking, standing, and many more activates.

2.4.5. Pressure measuring devices

Before the usage of floor-mounted transducer matrix platforms, the force plates were the chosen primary device for measuring COP (Orlin & McPoil, 2000). Force platforms use the COG and track the vertical component of the GRF, while other systems, such as pressure measuring devices, have many sensory elements and use the distribution of the force across the sensor grid (Gerbino et al., 2007). Pressure sensitive platform systems contain several sensory elements and use the distribution of the force across the sensor grid, which enables them to quantify the COP and the pressure patterns under the feet over time. However, these devices do not directly measure the applied force vector and are unable to quantify horizontal components of the applied force (Robertson et al., 2013). The COP recorded from a force plate is the "instantaneous point of application of the resultant foot-floor reaction vector"; thus, the COP derived from a force plate represents the vertical section of the GRF and not the pressure acting on the plantar surface of the foot (Orlin & McPoil, 2000, p.334). The COP derived from pressure sensitive platforms is determined by calculating the centroid of the total number of active transducers for each frame.

The data obtained from these devices has been used for the assessment and evaluation of lower limb functions, such as assisting patients with diabetic peripheral neuropathy, and providing them with footwear modifications and redistributive insoles aimed at offloading areas of high pressure. The information gained in this manner has been leveraged in different areas, ranging from predicting the occurrence of ulceration to designing orthopedic insoles that can provide relief for the maximum pressure experienced by an individual while walking. Most of the research using platform pressure measurements has focused on factors, such as type of insole material (McPoil & Cornwall, 1992), external and internal shoe modification (Childs, Olson, McPoil, & Cornwall, 1996), effect of therapeutic footwear (Mueller, Strube, & Allen, 1997), insoles that decrease high pressure areas on the foot (Mueller, et al., 1997), and shoe modifications (Nawoczenski, Birke, & Coleman, 1988; Orlin & McPoil, 2000).

Research regarding plantar pressure measures and age are controversial: Deepashini et al. (2014), believes that age has a significant influence on the plantar pressure of the foot and Bosch, Nagel, Weigend, & Rosenbaum, (2009) agree with this statement. They investigated plantar pressure patterns in different age-related stages. Their findings indicated an increase in plantar pressure measures with age, which the authors attributed to stiffness in the elastic tissue underneath the older adults' foot. Conversely, Pisciotta, (2015) reported no differences in the magnitude of PP and PTI between older and younger adults; while, Hessert et al., (2005) reported lower

plantar pressure measures for older adults in comparison to young adults. Thus, more research is needed to understand the changes in Plantar Pressure measures in older adults.

Nevertheless, the relationship between plantar pressure and foot pain in older adult fallers and non-fallers was investigated in a research using a pressure platform. The study recruited 312 participants aged 60 and older. Participants were asked to look straight ahead (to avoid targeting the pressure platform) and to walk at a self-selected comfortable speed. The plantar pressure generated underneath each participant's feet was assessed while they walked over the pressure platform. The contact time, total peak pressure and total pressure-time integral values extracted from the pressure platforms were used for comparison. Their results indicated that older adult fallers showed higher values for both variables in comparison to older adult non-fallers. The researchers also found that fallers had a significantly higher prevalence in foot pain compared to non-fallers. These findings led the researchers to believe that the high plantar pressure generated by older adult fallers can increase the pressure load under anatomical structures of the foot, which can consequently result into higher prevalence of foot pain among this population and act as a contributing factor to risk of fall (Mickle, Munro, Lord, Menz, & Steele, 2010). This study claimed to be the first to investigate the peak pressure and pressure time integral created under older adult faller and non-faller feet. To shed more light on plantar pressure distribution and its impact on identifying older adults who are at risk of falling, there is a need for more research in this area.

Finally, it is important to acknowledge that pressure mats provide limited information regarding vertical and shear components of the ground reaction force for each individual during foot contact. In order to gain a more comprehensive understanding of the differences between older adult fallers and non-fallers, it is important to investigate other biomechanical variables during gait and standing balance tests as well. Tracking an individual's COP, along with variables such as peak pressure, pressure time integral, loading rate and shear force can provide valuable information with regards to the balance of older adult fallers and non-fallers. Therefore, this research investigates COP, plantar pressure and GRF variables of its participants. Essentially, the sets of data obtained from the two types of devices can complement each other.

2.4.6 Effect of Walking Speed on Pressure data

Among several methods that researchers have used to show the sensitivity of COP measures, quiet standing seems to be the most popular approach (Mancini et al., 2012). However, it is not sufficient to assess one's postural control during static trials. Individuals often fall during walking as opposed to standing; O'Connor & Kuo, (2009), argue that improving the qualitative assessment of postural sway during static trials entails a number of changes. These changes include reduction in the size of the base support, reducing visual and proprioceptive feedback, as well as the application of a secondary task during the time the participant is trying to keep their balance. Although, the introduction of these tasks can work to make the trials more

challenging, they fail to provide adequate information regarding dynamic balance during gait. The addition of a dynamic balance test can help offer a more accurate description of potential fallers considering that the majority of falls happen when a person is either walking, turning, or sitting. In order to gain a more effective understanding of older adults' balance with respect to falling, this research project acquired measurements during both standing and walking trials.

When evaluating an individual during walking it is common to ask them to walk in a comfortable speed. This is because forcing a participant to walk outside their natural walking speed can alter their gait characteristics. According to Kerrigan et al., (1998) an alteration in gait characteristics was observed when older adult fallers were asked to walk at a faster speed than their comfortable gait velocity. In order to avoid participants changing the way they normally walk, most studies ask their participants to walk in their comfortable speed (Bizovska et al., 2014; Giakas & Baltzopoulos, 1997; Mickle et al., 2010; Wu, Wang, & Liu, 2007).

Plantar pressure distribution has also been shown to change with change in walking speed. Studies that have compared slow, fast, and normal velocity of gait, have shown that walking speed can significantly affect the peak plantar pressure (Burnfield, Few, Mohamed, & Perry, 2004; Segal et al., 2004). According to Segal et al., (2004), higher gait velocity results in higher peak pressure values in most regions of the foot—especially in the heel area. Additionally, Burnfield et al., (2004) concluded that walking with faster

speeds, results in higher mean and peak plantar pressures in the older adult population: One needs to produce more power to walk faster, which results into more force and higher plantar pressure being produced.

Nevertheless one of the most constant findings in research regarding older adults' gait is that as an individual ages, their gait velocity decreases (Burnfield et al., 2004; Knoblauch, Pietrucha, & Nitzburg, 1996; Oberg, Karsznia, & Oberg, 1993; Peel, Kuys, & Klein, 2013; Prince, Corriveau, Hébert, & Winter, 1997). For instance, a study conducted by Kerrigan et al., (1998) indicated that in addition to showing reduced stride length compared to young individuals, older adults exhibited a significantly lower walking speed when asked to walk in their comfortable speed.

Research also suggests that, due to the low walking speed adopted by older adults, the findings regarding gait velocity and pressure data should not be extrapolated to this group. For instance, Hausdorff, (2007,) found that the walking speed of older adult fallers and non-fallers is similar and Taylor, Menz, & Keenan, (2004), found no significant differences between self-selected slow and normal speeds for maximum force and peak pressure values. The authors suggested asking participants to walk at a self-selected speed, to prevent gait alteration.

Moreover, when Mickle et al., (2010) observed a significant statistical difference in the pressure data between the investigated elderly fallers and non-fallers, they ruled out the possibility of their findings being attributed to difference in walking speed. The researchers stated that fallers have been

shown to walk more slowly than non-fallers, and in that study, fallers recorded longer contact times than the non-fallers, which is characteristic of a slower average walking speed. In order to capture the participants normal foot to ground interaction and avoid changes in their gait characteristics (Bosch, Nagel, Weigend, & Rosenbaum, 2009), the participants of the current study were asked to walk with a self-selected speed. The gait velocity of the participants was calculated by dividing the distance traveled by the time it took the participants to travel the distance.

2.5 Hypotheses

- COP measures during walking (COP displacements and their variability) - and during standing (COP velocity, COP ellipse area) - will be significantly higher in fallers compared to non-fallers; the $\Delta\Phi$ of the COP during standing will be significantly different between fallers and non-fallers[1].
- The GRF measures during walking (first vertical peak force, minimum vertical force, second vertical peak force, maximum breaking force, mid stance, maximum propulsion force, and the relative time to reach each of these forces) and standing (standard deviation of the average vertical force, ML force, and AP force)-will be significantly different between fallers and non-fallers[1].
- Plantar pressure measures (peak contact pressure, force time integral pressure time integral, mean pressure, and peak pressure) will be

significantly different between fallers and non-fallers during walking and standing[1].

[1] Since very little is known about the differences in the sample entropy of the ΔΦ of the COP, GRF data and plantar pressure data between fallers and non-fallers, the hypothesis for these variables was kept broad.

2.6 Summary

The literature that investigates differences between older adult fallers and non-fallers, has mainly focused on tracking the COP of participants while standing with varying rates of success. To better understand the biomechanical differences in older adult fallers and non-fallers, researchers should explore different variables that can be obtained during an older individual's gait, such as: COP measures, GRF measures and plantar pressure measures. There has only been one previous study that compared plantar pressure measures between these two groups, and no other study has investigated these measures during standing. To better understand biomechanical differences between older adult fallers and non-fallers during walking and standing, there is a need for more research in this area. Assessing the differences in various biomechanical variables between these groups can provide valuable information for researchers.

3. Method

The following section provides information regarding the research project's participants, procedure, instrumentation, and data analysis.

3.1 Participants

All data was collected in the Andrew and Marjorie McCain Human Performance Laboratory located in the Richard J. CURRIE CENTER at the University of New Brunswick. 102 participants over the age of 65 were recruited from older-adult community centers located in Fredericton, New Brunswick.

The participants were informed via word of mouth and posters that were posted in nursing homes, rehabilitation centers, physiotherapy clinics, orthopedic clinics, the YMCA Fredericton, and the campus of the University of New Brunswick.

The participants were categorized into two groups based on their history of fall (Dibble & Lange, 2006; Lajoie & Gallagher, 2004; Lajoie, Girard, & Guay, 2002; MacRae, Lacourse, & Moldavon, 1992; Thomas & Lane, 2005). Forty one of the participants who had experienced one or more falls in the previous year were placed into the "Faller" (F) group; those who had not fallen in the previous year were placed in the "Non-Faller" (NF) group. A fall was defined as "inadvertently coming to rest on the ground, floor or other lower level, excluding intentional change in position to rest in furniture, wall or other objects" (World Health Organization. Ageing & Life Course

Unit, 2008). Additionally, inevitable events like furniture collapse were excluded from the definition of a fall (Shumway-Cook et al., 2000).

The participants were asked to participate in the study once for a duration of one hour. There were two chairs at both ends of the walkway for the participants to sit on, in case they felt the need to rest. Additionally, there was a Kinesiology student present during the data collection who walked and stood alongside the participants to ensure no falls occurred during data collection.

Participants were briefed orally on the nature, procedures, purpose, and potential risks of the experiment. An information sheet detailing the potential risks associated with the experiment was given to each participant. Also, participants were asked to read and sign an informed consent form, a PAR-Q+ questionnaire, and answer questions pertaining to their medical history and fall history before the data collection. Additionally, a Mini-Cog test was administrated as a measure of participant intellectual capacity to consent without third-party assent. The inclusion and exclusion criteria for this research are indicated in Table 1.

Table 1: Inclusion and exclusion criteria for the participation in the study.

Inclusion Criteria	Exclusion Criteria
Individuals over the age of 65.	Leg and/or foot amputation.
Able to stand and ambulate with or without walking aids.	Need external support other than walking aids to stand.
Able to walk without walking aid for 5 meters.	Cannot walk without walking aid for 10 minutes.
	Any physical or neurological condition that results in abnormal gaits such as: Hemiplegic gait, Diplegic gait, steppage gait, Hyperkinetic gait, Ataxic gait and Parkinsonian gait prohibiting participation.
	Any mental or psychiatric illness prohibiting participation.

3.2 Measurements and Instrumentation

The following section provides information regarding the instrumentation used in this research.

3.2.1 Force plates

The Andrew and Marjorie McCain Human Performance Laboratory is equipped with six piezoelectric force plates [Kistler Instruments Ltd, Switzerland]. The force plates can measure three force vectors of F_x, (the force in the anterior-posterior direction), F_y (the force in the medial-lateral direction) and F_z (the force in the vertical direction) and the moment components of the three forces. Additionally, the plates record a three-dimensional projection of

the forces and the COP coordinates of a subject's foot during the time they are in contact with them. The force plate data were sampled at a frequency of 1600 Hz. Since 2 pressure mats were utilized in this study, only 2 of the 6 available force plates were used for data collection.



Figure 1: Six piezoelectric force plates embedded in the lab floor.

3.2.2 High Resolution pressure mat

Two High Resolution (HR) pressure sensitive floor mats [Tekscan Inc., Boston, MA, USA] were used for this research. The Tekscan HR foot Mat software collects plantar pressure data from the sensor and makes that information available to the software. The sensor is sampled as the subject stands, walks, runs or jumps. The software allows the user to view the collected sensor pressure data in real time, record this information as a "movie", and

review and analyze it at a later time. The data was recorded with a sample frequency of 160 Hz and synchronized with the force plates using an A/D board. The Tekscan HR Mat (Tekscan Inc., Boston, MA, USA) has a sensel area of 44.35 × 48.55 cm^2 with 8448 sensels. Each sensel has an area of 0.25 cm^2, and these sensels are connected to two main sensors. The sensors transfer the information to the analysis software through a 2-port hub.



Figure 2: Tekscan HR mat (Tekscan Inc., Boston, MA, USA). *Two High Resolution (HR) pressure sensitive floor mats were used for this research.*



Figure 3: Components of Tekscan HR mat. *From left to right, VersaTek, 2-Port Hub, Power Supply Cord, and USB Mini Cable.*

3.3.3 Walkway

A 1.2 × 5.8 meter walkway covered the force plates (except for the two plates that contained the pressure mats) and the lab floor to disguise the pressure mats and account for their height. The walkway consisted of Mannington platinum flooring. Figure 4 illustrates the walkway mounted on top of the lab floor and the plates.

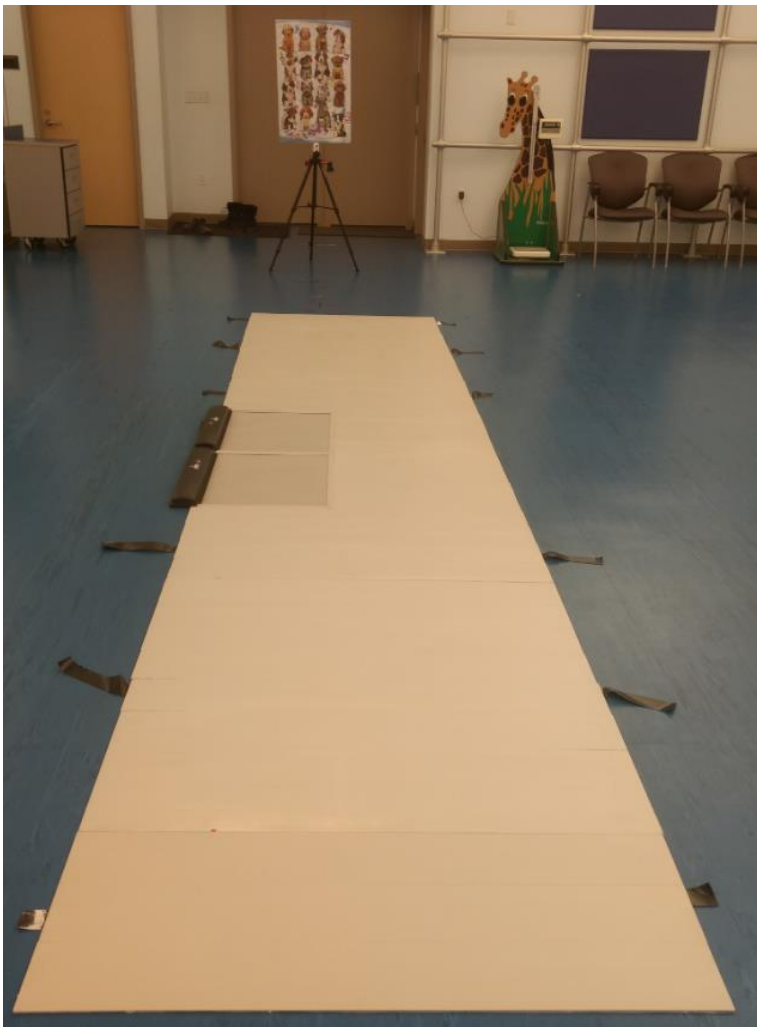

Figure 4: The Walkway with two embedded pressure mats. *The width and length of the walkway are 1.2 × 5.8 meters; it is mounted on top of the six piezoelectric force plates.*

To assess the effect of the walkway on the forces recorded from the force plates, an experiment with 10 stationary objects with different weights was conducted. These objects were placed on the force plate and the same objects were placed on the walkway mounted on top of the force plate. The differences between the two conditions of "force plate-alone" and "force plate-walkway" were calculated and the results are presented in Appendix I.

The average of the relative error between the two conditions was ± 0.76% for force plate 2 and ±0.61% for force plate 3. In other words, the walkway can cause an error up to 0.61% and 0.76% resulting the force plates to overestimate or underestimate the measured forces.

3.3.4 A/D board

A 128-channel analog to digital (A/D) board allowed the Vicon Nexus software to communicate with the Tekscan software. A programmable delay allowed timing adjustments on the trigger and synchronization signals before they were sent to buffers for output through the output connectors. This A/D board synchronized the recording between the pressure mats and the force plates.

3.3.5 Berg Balance Test

A Berg Balance Scale (BBS) test was completed for each participant. This scale entails 14 items that evaluate the ability to stand up and maintain the standing position during different tasks. For each item of the BBS, the

examiner can score the participant from 0-4 and the maximum score that can be obtained is 56. The 14 items of BBS are presented in Appendix II.

3.3.6 Balance Evaluation Scale Test

The Balance Evaluation Scale Test (BESTest) is a 36-item clinical balance tool that is targeted at identifying the components contributing to balance dysfunction in 6 subsystems of postural control (O'Hoski et al., 2014). This test has also been recognized as a standardized balance measure that assesses all components of balance (Sibley, Beauchamp, Van Ooteghem, Straus, & Jaglal, 2015). However, the long duration that is required to administer a BESTest (30-60 minutes) has made this test not feasible to use in many clinical settings. Therefore, two abbreviated versions of this test have been developed as alternatives to the original test (O'Hoski, Sibley, Brooks, & Beauchamp, 2015). The Mini Balance Evaluation Scale Test (Mini-BESTest) and the Brief Balance Evaluation Scale Test (Brief-BESTest); these tests can be administered in half the time required for a total BESTest (O'Hoski et al., 2014; Padgett, Jacobs, & Kasser, 2012).

The Mini-BESTest can only provide a total score for dynamic balance and does not have the ability to identify the underlying components contributing to balance impairments. On the other hand, the Brief-BESTest is more theoretically consistent with the total BESTest. The test consists of 8 items: one item from each subsection of the total BESTest, with two items [Single Leg Stance (SLS) and functional reach forward] scored bilaterally

(Padgett et al., 2012). Each item is scored from 0 to 3 and a maximum possible score of 24 points can be obtained (O'Hoski, Sibley, Brooks, & Beauchamp, 2015). The Brief-BESTest, which has been designed to preserve the construct validity of the total Best test, has been recommended by the Neurology Section of the American Physical Therapy Association (Padgett et al., 2012). A Brief-BESTest was therefore completed for each participant; the 8 items of BBS are presented in Appendix III.

3.4 Procedure

The participants were given an introduction of the lab and the experimental procedure. Following that, the participants were asked to read and sign an informed consent form. A PAR-Q+ questionnaire was administered to determine if the participants require doctor consultation prior to tests**.** Any questions and/or concerns from the participants were addressed before the data collection process began. Demographics and data regarding the volunteers' fall history was collected. Afterwards, the participants were assessed by the researcher while performing the 14 items of the Berg balance scale and the 8 items of the Brief-BESTest.

The data collection began after the participants signed the informed consent form, received a score for their BBS and Brief-BESTest performance, and provide information about their demographics and fall history. The data collection contained two evaluation procedures: walking and standing trials. After the participants were familiarized with the lab and the walkway, they

were asked to perform the walking trials. Following that, the participants were asked to partake in the standing trials. Unlike the walking trials, which had an automatic calibration, the standing trials required calibration of the pressure mats and standing on the mats in two different conditions. Therefore, the aforementioned sequence was chosen to avoid drawing attention to the pressure mats, and the participants targeting the mats during the walking trials.

3.4.1 Walking test

The participants were asked to walk at a self-selected speed on a 5.83-meter walkway that contained 2 embedded pressure mats and was mounted on top of force plates. There was a 2.29-meter distance on each side before the participants reached the pressure mats. This distance allowed more than three steps before reaching the pressure mats and force plates and was enough for the participants to reach a steady gait speed. In order to ensure the participants would strike the pressure mats and force plates, differently colored start lines were placed at the beginning of the walkway. If the participant failed to provide an acceptable foot contact, they were asked to start again on a new start line.

To avoid targeting by the participants, the walkway was painted in a color similar to the pressure mats. Additionally, the participants were asked to look at a visual target while walking on the walkway. There were two posters sized 87× 57 cm. Each poster was placed 184 cm away from each end of the walkway to prevent the participants from targeting the pressure mats during the dynamic trials.

Initially, the participants were asked to stand behind the start-line and after the count of three followed by the "go" command, they walked until they reached the end of the walkway with their self-selected speed. Prior to each trial, the force plates were zeroed and the actual capture began on the count of "one". Once the participant reached the end of the walkway they stepped off the walkway and the force plates were zeroed again. Afterwards, the participants turned around and stood behind a line that was symmetrical to the original starting line. Subsequently, they walked back towards their starting point. The recording ended once the subjects returned to the original starting point. This was to ensure that both left and right feet would come in contact with the plates and mats.

Many studies suggest that obtaining three successful trials can ensure acceptable reliability of pressure data (Bryant, Singer, & Tinley, 1999; T. G. McPoil, Cornwall, Dupuis, & Cornwell, 1999; Meyers-Rice, Sugars, McPoil, & Cornwall, 1994; van der Leeden, Dekker, Siemonsma, Lek-Westerhof, & Steultjens, 2004; White, Agouris, Selbie, & Kirkpatrick, 1999). Therefore, this procedure was repeated until 3 successful strikes from the left foot and right foot were attained. The participants were provided with a 2-minute rest period between each trial, during which they could rest on a comfortable chair. Figure 5 shows the walking test.

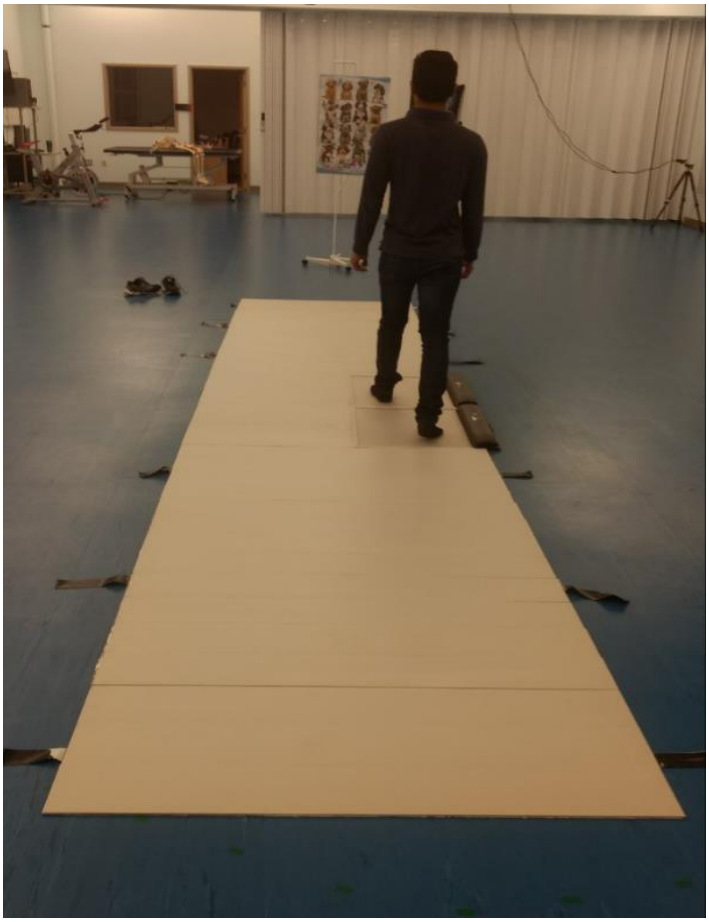

Figure 5: Demonstration of the walking test. *During the walking trials the participants walk back and forth on the walkway*

3.4.2 Standing test

For this part of the experiment, the participants were asked to step on one of the pressure mats mounted on top of a force plate. After calibrating the mat, they were asked to perform two standing balance tests for 30 seconds; semi tandem stance (place one foot directly in front of the other) with eyes open; narrow stance (BOS less than the width of the shoulder) with eyes closed. The tests were repeated three times with 1-minute rest period for the

eyes open condition and 3-minute rest period for the eyes closed condition; the participants were asked to sit on a chair during the rest periods.

These tests have been used for fall risk assessment in previous research (Ejupi, Gschwind, Valenzuela, Lord, & Delbaere, 2016). Additionally, these test conditions have been reported to be reliable balance tests for discriminating between older adults with and without balance impairments (Brotherton et al., 2005; Melzer, Kurz, & Oddsson, 2010a). Figures 6 and 7 illustrate the Eyes Closed Narrow Stance (ECNS) and the Eyes Open semi-Tandem Stance (EOTS) conditions of the Standing trials respectively.



Figure 6: Demonstration of the ECNS condition of the Standing test. *The ECNS standing test condition includes participants standing with their eyes closed, such that their BOS is smaller than their shoulder width*

.



Figure 7: Demonstration of the EOTS condition of the Standing test. *The EOTS standing test condition includes participants standing with their eyes open, such that the toes of their back-foot touches the heel of their front-foot.*

3.5 Data analysis

The data obtained through each trial was divided into three sections for analysis; plantar pressure data, which was acquired from the pressure mats; GRF data, which was obtained from the force plates; and COP data which was attained from both force plates and pressure mats. Despite the different methods used in obtaining the COP from force plates and pressure measuring devices; research in this area has confirmed that the COP obtained from force plates and pressure measuring devices are nearly identical (Brauer, Burns, & Galley, 2000b; Meyers-Rice et al., 1994). The COP acquired from force plates is more commonly used in the literature; therefore, for the purposes of this study, the force plates' COP was used for analysis. In order to compare the

similarity of the COP data that was obtained from both devices, an experiment was conducted (Appendix IV) and a transformation matrix was created to transform the coordinates of the pressure mat to the coordinates of the force plate (Appendix V).

Additionally, apart from the variables of the plantar pressure, which remained the same for the walking and standing trials, different COP and GRF variables were used for analyzing the walking and standing trials.

3.5.1 Plantar pressure variables

The plantar pressure data were processed using the Tekscan software (FootMat Research 7.0) for both walking and standing trials. The program calculates values for force, area and pressure. The following variables were used to compare the differences between older adult fallers and non-fallers: Peak Contact pressure (PkCP), Force Time Integral (FTI), Pressure Time Integral (PTI), Mean Pressure (MP) and Peak Pressure (PP). In order to normalize the pressure data, all the data obtained from the mats were divided by each participant's weight. Table 2 displays the plantar pressure dependent variables and their definitions. Additionally, to compare the differences between older adult fallers (F) and non-fallers (NF), independent sample t-tests were conducted between the two groups for each of the dependent variables presented in Table 2.

Table 2: The plantar pressure Dependent Variables (DVs) and their definitions.

Independent variable	Plantar pressure Dependent variables
Groups (F and NF)	**PkCP** *(N/mm^2)*: The small area (two-by-two sensel array) that contains the highest values for each frame. The value is then averaged throughout the stance.
	FTI *(N*Sec)*: The force in each cell while considering the amount of time the force was present in each cell.
	PTI *(N*Sec/mm^2):* The pressure in each cell while considering the amount of time the pressure was present in each cell.
	MP ***(****N/ mm^2)*: The average of the pressure throughout the stance.
	PP ***(****N/mm^2):* The Maximum force throughout the whole stance (e.g., at a certain frame force reached its maximum) divided by the contact area.

3.5.2 GRF & COP data

The Vicon software (Vicon Nexus 2.1.5) provides data regarding the three components of the GRF. From these data the coordinates of the COP can be obtained. The following sections discuss the variables that were extracted from the force plates for the walking and standing trials.

3.5.2.1 Walking trials-GRF variables

Assessment of GRF components can provide valuable information regarding their locomotion characteristics during the stance phase of walking. To assess the GRF components, all the GRF component graphs are presented

in % of stance duration (horizontal axis) and % of bodyweight (vertical axis) (Hasan, Jailani, Md Tahir, & Ilias, 2017). Additionally, the heel contact and toe-off is defined as the points in time when the vertical GRF exceeds 5% of body weight and goes below 5% of body weight respectively (Kim & Eng, 2003; McCrory, Chambers, Daftary, & Redfern, 2011). The Vertical, Anterior Posterior, and Medial Lateral components of the GRF and the variables that were chosen from each of these components are elaborated further.

i. Vertical component of GRF

The following vertical components of the GRF were used for further analysis based on previous research (Chen, Chen, Wong, Tang, & Chen, 2001; Chockalingam, Dangerfield, Rahmatalla, Ahmed, & Cochrane, 2004; Claeys, 1983; Giakas & Baltzopoulos, 1997b; Goh, Thambyah, & Bose, 1998; Hasan, Jailani, Md Tahir, & Ilias, 2017b; Jarrett, Moore, & Swanson, 1980; White et al., 1999): first peak force that occurs after heel strike (Fz1), minimum force that occurs during mid stance (Fz2), second peak force that occurs before toe-off (Fz3), and the relative time to reach each of these forces (Tz1, Tz2, Tz3). Figure 8, shows the Fz1, Fz2, Fz3, Tz1, Tz2, and Tz3 during one walking trial of a participant.

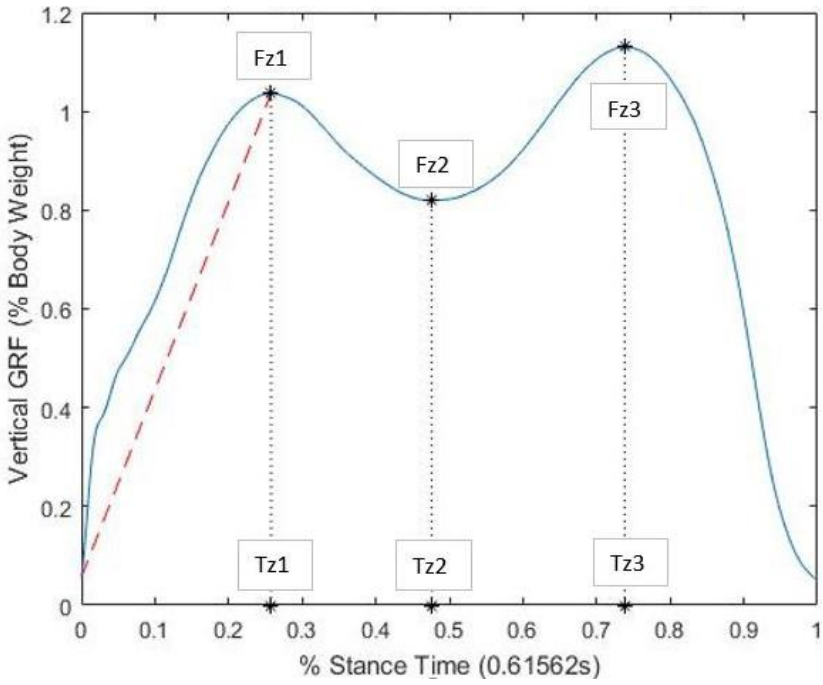

Figure 8: The Vertical component of the GRF's DVs for walking tests. *The figure displays each DV during a participant's walking trial: Fz1 (first peak force), Fz2 (minimum force), Fz3 (second peak force), and the relative time to reach each of the forces (Tz1, Tz2, Tz3).*

Loading rate and decay rate have also been used by researchers as part of the vertical component assessment of the GRF (Jarrett et al., 1980). However, according to (Diss, 2001a), decay rate shows low reliability and requires at least 7 trials to achieve an acceptable reliability. Since fewer than 7 trials were recorded for the participants of this study, the decay rate was excluded from our analysis. Loading rate was defined as the absolute value of the slope of the straight line connecting the zero intercept (at time T_0) of the force curve at heel contact with the first maximum peak (at time T_1) (Creaby, May, & Bennell, 2011; Sławomir Winiarski & Rutkowska-Kucharska, 2009).

Figure 2 and Equation 1 show the loading rate and how it was calculated respectively.

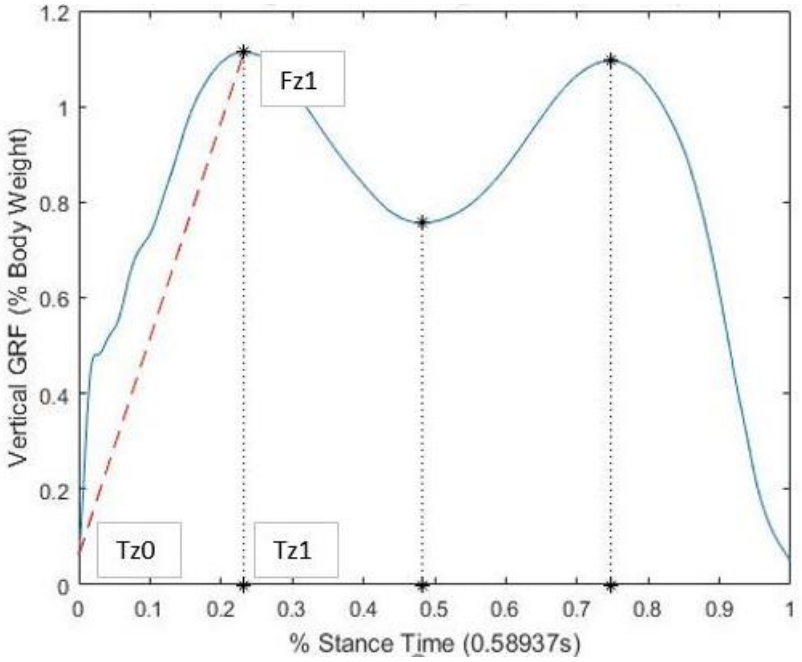

Figure 9: The loading rate during a participant's walking trial. *The red line displays the loading rate, which is defined as the absolute value of the slope of the straight line connecting Tz0 of the force curve with Fz1 at time Tz1.*

Equation 1:

$$Loading\ rate\ (\frac{\%BW}{\%Stance}) = \left|\frac{Fz_1}{T_{z1} - T_{z0}}\right|$$

ii. Anterior -Posterior component of GRF

The following Anterior –Posterior (AP) components of the GRF were used for further analysis based on the literature (Claeys, 1983; Diss, 2001b; Giakas & Baltzopoulos, 1997b; Goh et al., 1998; Hasan et al., 2017b; Jarrett et al., 1980; Kim & Eng, 2003; White et al., 1999): The first peak (Fx1), which represents the maximum breaking force; mid stance (Fx2); the second peak (Fx3), which represents the maximum propulsion force; and the relative time to reach each of the forces (Tx1, Tx2, Tx3). Figure 10 shows the Fx1, Fx2, Fx3, Tx1, Tx2, and Tx3 during a walking trial of a participant.

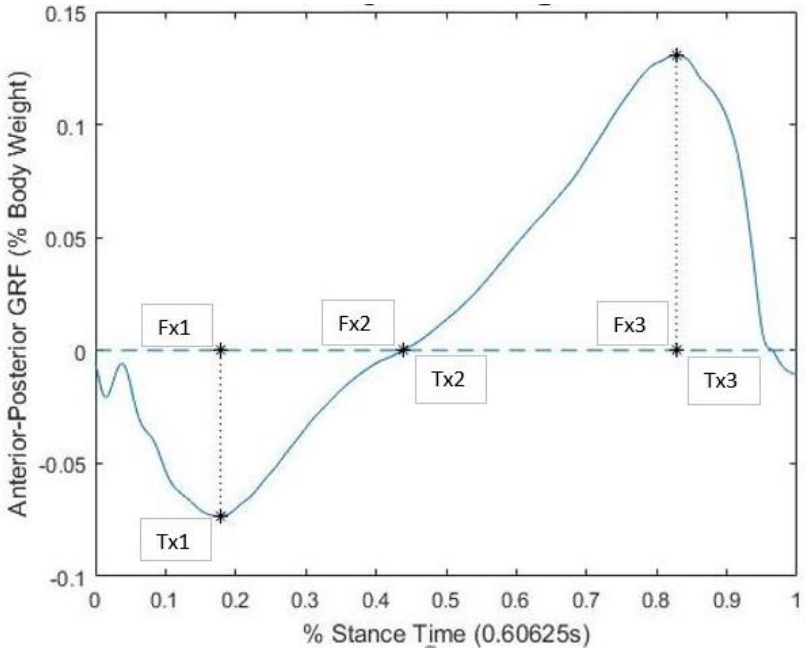

Figure 10: The AP component of the GRF's DVs for the walking tests. *The figure displays each DV during a participant's walking trial: Fx1 (maximum breaking force), Fx2 (mid stance), Fx3 (maximum propulsion force), and the relative time to reach each of the forces (Tx1, Tx2, Tx3).*

iii. Medial -Lateral component of GRF

With respect to the Medial-Lateral (ML) force, many researchers believe that this force component is small, variable and has no consistent pattern from individual to individual (Chockalingam et al., 2004; Claeys, 1983; Goh et al., 1998; Herzog, Nigg, Read, & Olsson, 1989).Therefore, the ML components of the GRF were excluded from the analysis.

Overall, two of the GRF components (Vertical and AP) were selected for further analysis; from each of these components different variables were chosen, which resulted in 13 GRF-dependent variables being compared between the two groups for the walking trials. The GRF variables that were extracted from each of the participant's footsteps during the walking trials are presented in Table 3. In order to investigate the differences between older adult fallers (F) and non-fallers (NF), independent sample t-tests were conducted for each of the GRF dependent variables presented in Table 3.

Table 3: The GRF Dependent Variables (DVs) for the walking tests.

Independent variable	GRF Dependent variable	
Groups (F and NF)	Loading rate (%BW/second)	
	Fz1 (%Bw)	Tz1 (%stance)
	Fz2 (%Bw)	Tz2 (%stance)
	Fz3 (%Bw)	Tz3 (%stance)
	Fx1 (%Bw)	Tx1 (%stance)
	Fx2 (%Bw)	Tx2 (%stance)
	Fx3 (%Bw)	Tx3 (%stance)

The walking-GRF DVs were extracted from the force plates for each walking trial.

3.5.2.2 Walking trials-COP variables

When considering the COP movement during gait, researchers divide the stance phase into 4 sub-phases based on the behavior of the vertical component of the GRF in each stance phase (Bizovska et al., 2014b; de Souza, Svoboda, Bizovská, & Lehnert, 2017b; J. Lee, Hong, & Shin, 2018; Zdenek Svoboda et al., 2017a; Zdeněk Svoboda et al., 2015; van Kooten, Hettinga, Duffy, Jackson, & Taylor, 2018).The sub-phases of stance are: loading response (LR), mid stance (MSt), terminal stance (TSt) and Pre-swing (PSw). LR is the time interval between initial contact (heel strike) and the first peak (Fz1); MSt is the time interval from Fz1 to the minimum of the vertical GRF (Fz2); TSt is the time interval from FZ2 to the second peak (FZ3); and PS is the time interval from FZ3 to toe-off. Figure 11 shows the sub-phases of stance.

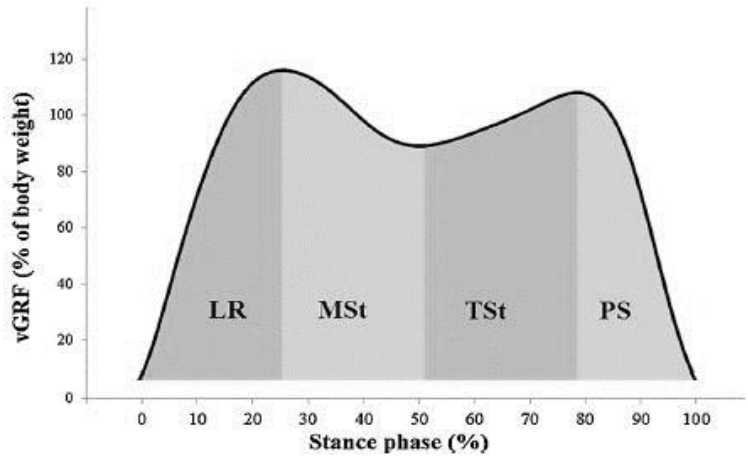

Figure 11: The sub-phases of a stance phase during gait. (Bizovska et al, 2014, p.400.) *Each sub-phase of the vertical ground reaction force is defined as: Loading Response (LR): the time between heel strike and first peak, Mid-Stance (MSt): the time*

between first peak and minimum force, Terminal Stance (TSt): the time between minimum force and second peak, Pre-Swing (PS): the time between second peak and toe-off.

For each sub-phase the instantaneous displacement in the ML direction, instantaneous displacement in the AP direction, the total displacement of COP movement (Bizovska et al., 2014b; J. Lee et al., 2018), and their variability (Standard Deviations) were calculated (Bizovska et al., 2014b; de Souza et al., 2017b; J. Lee et al., 2018; Zdenek Svoboda et al., 2017a; Zdeněk Svoboda et al., 2015; van Kooten et al., 2018).

The calculation for the COP's Total displacement (COP_T_Disp), ML displacement (COP_{ML}_Disp), and AP displacement (COP_{AP}_Disp) during the walking trials are shown in Equations 2, 3, and 4 respectively.

Equation 2:

$$COP_{AP}\text{_Disp_Walk} = \sqrt{(x_2 - x_1)^2}$$

Equation 3:

$$COP_{ML}\text{_Disp_Walk} = \sqrt{(y_2 - y_1)^2}$$

Equation 4:

$$COP_{T}\text{_Disp_Walk} = \sqrt{(x_2 - x_1)^2 + (y_2 - y_1)^2}$$

Afterwards, to account for the variability in the COP displacements the Standard Deviations (SD) of the COP_T, $COP_{AP,}$ and COP_{ML} displacements were calculated for each walking trial (COP_T_Disp_SD, COP_{AP}_Disp_SD, COP_{ML}_Disp_SD). Overall, for each sub-phase (LR, MSt, TSt, and PS) the COP displacement and the variability of the displacements were calculated, which resulted in 24 COP-dependent variables being compared between the two groups for the walking trials.

The COP variables that were extracted from each sub-phase of the stance during the walking trials are presented in Table 4. In order to investigate the differences between older adult fallers (F) and non-fallers (NF), independent sample t-tests were conducted for each of the COP dependent variables presented in Table 4.

Table 4: The COP Dependent Variables (DVs) for the walking tests.

Independent variable	COP Dependent variable (cm)	
Groups (F and NF)	LR_COP_T_Disp	LR_COP_T_Disp_SD

Independent variable	COP Dependent variable (cm)	
	$LR_COP_{ML}_Disp$	$LR_COP_{ML}_Disp_SD$
	$LR_COP_{AP}_Disp$	$LR_COP_{AP}_Disp_SD$
	$MSt_COP_{T}_Disp$	$MSt_COP_{T}_Disp_SD$
	$MSt_COP_{ML}_Disp$	$MSt_COP_{ML}_Disp_SD$
	$MSt_COP_{AP}_Disp$	$MSt_COP_{AP}_Disp_SD$
	$TSt_COP_{T}_Disp$	$TSt_COP_{T}_Disp_SD$
	$TSt_COP_{ML}_Disp$	$TSt_COP_{ML}_Disp_SD$
	$TSt_COP_{AP}_Disp$	$TSt_COP_{AP}_Disp_SD$
	$PS_COP_{T}_Disp$	$PS_COP_{T}_Disp_SD$
	$PS_COP_{ML}_Disp$	$PS_COP_{ML}_Disp_SD$
	$PS_COP_{AP}_Disp$	$PS_COP_{AP}_Disp_SD$

The walking-COP DVs extracted from the force plates for each walking trial.

3.5.2.3 Walking trials-Participants walking Speed

In order to avoid altering the participants' gait characteristics, they were asked to walk with a comfortable speed. However, when an individual walks faster higher force/pressure is produced during gait. Therefore, the participants' walking speed was calculated for further evaluation during the analysis. The participants' estimated walking speed was calculated by using the location of the GRF vector for two consecutive steps. The distance between the GRF locations was calculated and divided by the time interval. In order to do so, the average COP for the first step (x_1, y_1) and the second step (x_2, y_2)

were calculated (Equation 5). To calculate t_1 & t_2, the time that heel contact occurred was added to half of the stance duration.

Equation 5:

$$\boldsymbol{Speed}\left(\frac{\boldsymbol{m}}{\boldsymbol{s}}\right) = \frac{\sqrt{(x_2 - x_1)^2 + \left(y_2 - y_1\right)^2}}{t_2 - t_1}$$

3.5.2.4 Standing trials-GRF variables

Regarding the GRF during standing trials, researchers use the standard deviations about the mean force in the Vertical ML, and AP, and planes (Goldie, Bach, & Evans, 1989a; Önell, 2000). Therefore, for the GRF during standing trials, Standard Deviation (SD) of the average vertical force (Fz_{Stand}_SD), ML force (Fy_{Stand}_SD), and AP force (Fx_{Stand}_SD) were calculated.

The GRF variables that were extracted from each of the Standing trials are presented in Table 5. In order to investigate the differences between older adult fallers (F) and non-fallers (NF), independent sample t-tests were conducted for each of the GRF dependent variables presented in Table 3, for both ECNS and EOTS conditions.

Table 5: The GRF Dependent Variables (DVs) for the standing tests.

Independent variable	GRF Dependent variable
Groups (F and NF)	Fz_{Stand}_SD (%Bw)
	Fy_{Stand}_SD (%Bw)
	Fx_{Stand}_SD (%Bw)

The standing-GRF DVs were extracted from the force plates for both ECNS and EOTS conditions of the standing trials.

3.5.2.5 Standing trials-COP variables

For the COP calculations during the Standing trials, the following were calculated: the COP Total Velocity (COP_T-Vel_{Stand}), the average velocity of the COP displacement in both AP (COP_{AP}-Vel_{Stand}) and ML directions (COP_{ML}-Vel_{Stand}). Equations 6, 7, and 8 show how each of these variables were calculated.

Equation 6:

$$COP_{AP}_Vel_{Stand}(\frac{cm}{s}) = \frac{\sum_{i=t_1}^{t_2}|y_i - y_{i-1}|}{t_2 - t_1}$$

Equation 7:

$$COP_{ML}_Vel_{Stand}\ (\frac{cm}{s}) = \frac{\sum_{i=t_1}^{t_2}|x_i - x_{i-1}|}{t_2 - t_1}$$

Equation 8:

$$COP_{T_}Vel_{Stand}(\frac{cm}{s}) = \frac{\sum_{i=t_1}^{t_2}\sqrt{(x_i - x_{i-1})^2 + (y_i - y_{i-1})^2}}{t_2 - t_1}$$

Additionally, the following two variables were calculated using the COP coordinates during the Standing trials: 95% confidence ellipse area ($COP_{EllArea}$), and the sample entropy of ΔΦ (SampEnt-$COP_{\Delta\Phi}$). The calculation of 95% confidence ellipse around the COP coordinates was based on the work of (Hoover & Rockville; 1984), which is presented in Appendix VI. In order to obtain the $COP_{\Delta\Phi}$, first, a vector was calculated from two sequential points of the COP.

Equation 9:

$$\mathrm{V_i} = \mathrm{COP_i} - \mathrm{COP_{i-1}}$$

Afterwards, the angle between two successive vectors was calculated:

Equation 10:

$$\Delta\Phi_\mathrm{i} = \mathrm{atan2}(\mathrm{V_i}, \mathrm{V_{i-1}}) \qquad 1$$

The Sample Entropy (SampEnt) of the $\Delta\Phi$ was calculated using an online source[2], the validation of the SampEnt calculations is presented in Appendix VII. The COP variables that were extracted from each of the standing trials are presented in Table 6. In order to investigate the differences between older adult Fallers (F) and non-fallers (NF), independent sample t-tests were conducted for each of the COP dependent variables presented in Table 6, for both ECNS and EOTS conditions.

Table 6: The COP Dependent Variables (DVs) for the standing tests**.**

Independent variable	COP Dependent variable
Groups (F and NF)	COP_T-Vel_{Stand} (cm/sec)
	COP_{AP}-Vel_{Stand} (cm/sec)
	COP_{ML}-Vel_{Stand} (cm/sec)
	$COP_{EllArea}$(cm^2)
	SampEnt-$COP_{\Delta\Phi}$

The standing-COP DVs were extracted from the force plates for both ECNS and EOTS conditions of the standing trials.

3.5.3 Statistical analysis

All data were analyzed using the MATLAB R2017 software; the calculated variables, were compared between older adult fallers and non-fallers using IBM SPSS Statistics 25 (IBM® SPSS Statistics Software). Multiple independent-samples t-tests were conducted to investigate the differences in the means of three sets of variables (COP variables, GRF variables, and plantar

[2] https://physionet.org/physiotools/sampen/

pressure variables) between older adult fallers and non-fallers, during walking and standing tests. However, conducting multiple t-tests can increase the likelihood of making a Type-I error (Mittelhammer, Judge, & Miller, 2000). Bonferroni correction compensates for this increase by examining each individual hypothesis at a significance level of α/m, where alpha (α) is the desired overall alpha level and m is the number of dependent variables (Miller, 1981). Hence, a Bonferroni correction was applied to the alpha level of 0.05 for each set of measures by dividing 0.05 by the number of dependent variables present in each set of variables (COP variables, GRF variables, and plantar pressure variables). Additionally, prior to conducting the independent-samples t-tests, preliminary analyses were performed to ensure non-violation of the assumptions of normality and homogeneity of variance.

Furthermore, to investigate the effect of speed on the data, univariate analysis of variance was conducted using the participant's speed as a covariant (ANCOVA). Apart from normality and homogeneity of variance, the overall relationship between the dependent variables and the covariant were also checked. The preliminary analysis indicated that all of the required assumptions except for the homogeneity of regression slopes were satisfied. However, the violation of this assumption was only in some cases and with the exception of Fz2 none of these cases showed significant differences in the independent sample t tests.

3.6 Summary

Falling in older adults is a multi-dimensional and complex issue that needs to be explored from different aspects. Studies that have investigated biomechanical differences between older adult fallers and non-fallers, have mainly focused on tracking the COP of their participants while standing. To gain a better understanding of falls, researchers should also explore different variables that can be obtained during an older individual's gait. Incorporating additional variables such as plantar pressure and GRF data can potentially lead to enhancing the identification of older adult fallers. Additionally, there is scarcity of data in terms of the plantar pressure generated during gait and the risk of falling in this population. So far only, only one study has investigated the differences in plantar pressure between older adult fallers and non-fallers (Mickle, et al., 2010) In order to better understand the plantar pressure differences between older adult fallers and non-fallers, there is a need for more research in this area.

The goal of this research was to compare biomechanical data during gait and standing in older adult fallers and non-fallers. Participants were asked to partake in walking and standing trials. During the walking trials they walked on a walkway that incorporated two pressure mats and were mounted on top of two force plates. For the Standing trials the participants stood on top of one of the pressure mats, in two different conditions of ECNS and EOTS. This research provides information regarding variables that can help distinguish older adult fallers from older adult non-fallers.

4. Results

This section provides information regarding the participants' demographics and the results of the statistical analysis.

4.1 Participant Demographics

In order to compare biomechanical data between older adult fallers and non-fallers during walking and standing, 102 participants were recruited for this study. Among these participants, 41 had experienced a fall in the previous year (faller) and 61 have not experienced a fall (non-faller). However, due to some technical difficulties that occurred during data collection, the walking trials of 99 participants were used for analysis. Furthermore, one of the participants did not take part in the standing trials due to issues that occurred when calibrating the pressure mat, the Standing trial data of 4 participants were not used for analysis. Therefore, for the standing tests data of 97 participants were used for analysis. The demographics of the fallers and non-faller for the walking and standing trials are presented in Table 7.

Table 7: Participants' demographics.

	Faller (Mean± SD)		Non-Faller (Mean± SD)	
	Standing	Walking	Standing	Walking
Age (years)	73.9±12.5	74.0 ±12.3	72.0 ± 5.4	72.3 ± 5.7
Height (cm)	165.2 ± 8.9	164.8 ± 8.8	164.8 ± 10.1	164.5 ± 10.0
Weight (kg)	77.0 ± 13.4	77.2 ± 13.3	73.6 ± 13.0	73.1 ± 13.3
Number (Male; Female)	N=39 (12 M; 27Fe)	N=41 (13M; 28Fe)	N=58 (23M; 35Fe)	N=58 (23M; 35Fe)

Mean and Standard Deviation (SD) of the Age (years), Height (cm), Weight (Kg) for the Fallers and Non-Faller during the walking and Standing trials.

Additionally, the age, height and weight of the participants were compared between the two groups (F vs. NF) using independent sample t-tests. The results showed no significant differences between the Fallers and Non-Fallers regarding their age height and weight.

Furthermore, Data collection took place in Fredericton New Brunswick from September to January. Despite the winter weather, the. most frequent reason for falling reported by our participants was tripping on objects not slipping on ice. The older adults who participated in this study reported that the causes of their falls were:

1. Tripping on uneven surface or objects such as mats, tree roots, small steps, etc. (53.65%)
2. Slipping on ice or in the bathroom (19.51%)
3. Ascending or descending stairs or slopes (17.07%)

4. Other reasons such as, reaching forward, carrying heavy objects, getting out of bed, getting out of chair, getting down a latter, and sudden loss of balance while moving fast (9.75%)

4.2 Balance Scores

At the start of the data collection session, each participant took part in two balance tests (BBS and Brief-BESTest) and received a score for each test. The participants' balance scores were compared between males vs females and Faller vs Non-Faller, which are presented in the following sections

4.2.1 Balance scores of Males vs Females

To investigate the differences in the BBS and Brief-BESTest scores between male and female participants, independent-sample *t*-tests were conducted. Preliminary analysis were performed to ensure non violation of the assumptions of normality, linearity, and homoscedasticity. The results indicated that there were no significant difference in the BBS for males (M=51.19, SD=4.97) and females (M=51, SD=5.3); t (76.18) =-0.18, p=0.85 (two-tailed); no significant difference in the Brief-BESTest for males (M=15.61, SD=4.71) and females (M=15.69, SD=4.34); t (99) =-0.08, p=0.93 (two-tailed).

4.2.2 Balance scores of Faller vs Non Fallers

To compare the BBS and Brief-BESTest scores between the Fallers and Non-Fallers, independent-samples *t*-tests were conducted. Preliminary

analyses were performed to ensure non violation of the assumptions of normality, linearity, and homoscedasticity. The results indicated that Non-Fallers (M=54.15, SD=1.6) had a significantly higher BBS score than Fallers (M=46.49, SD=5.2); t (45.24) =9.06, p<0.001 (two-tailed); Non-Fallers (M=18.45, SD=2.9) also had a significantly higher Brief-BESTest score than Fallers (M=11.59, SD=2.8); t (99) =11.62, p<0.001 (two-tailed). Figure 12 and 13 present the difference in the average BBS and Brief-BESTest scores of Fallers and Non-fallers respectively.

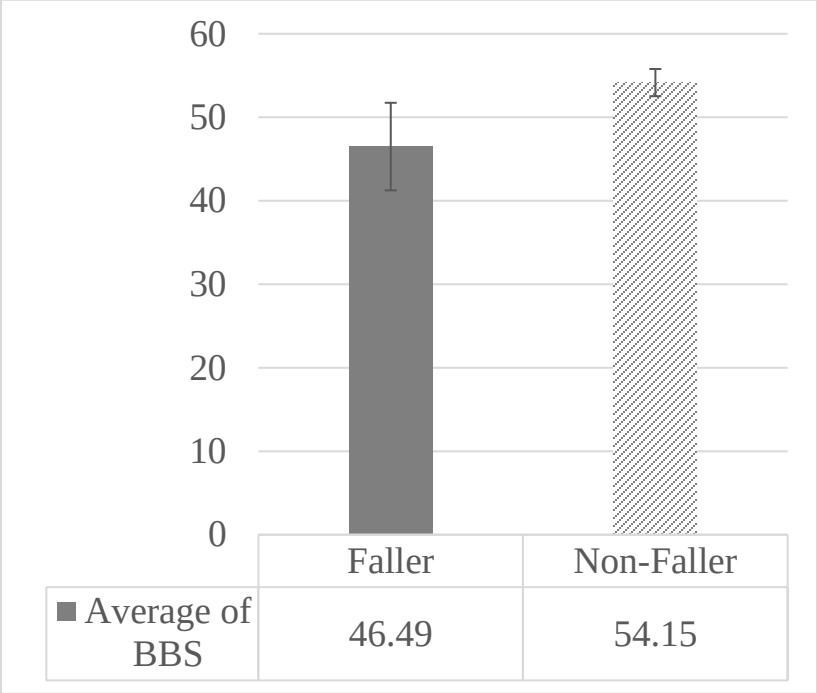

Figure 12: Average BBS Scores of Fallers and Non-Fallers.

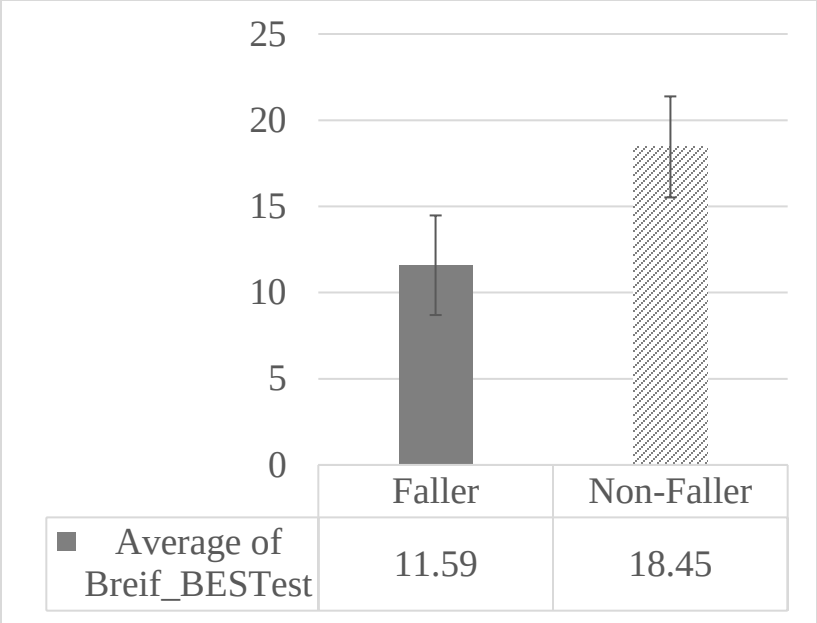

Figure 13: Average Brief-BESTest Scores of Fallers and Non-Fallers.

4.3 Walking tests

After the participants were familiarized with the lab and the walkway, they were asked to walk on the walkway until 3 successful strikes from the left foot and right foot were attained (6 total foot strikes). The data obtained through each trial were divided into three parts: Walking-COP variables, Walking-GRF variables, and Walking-plantar pressure variables and were compared between groups (F vs NF). The participants were asked to walk with a comfortable speed, however, higher speed can result in higher force/pressure being produced during gait. Therefore, the participants speed was also compared between Sex (Male vs Female) and groups (F vs NF) for the Walking tests.

4.3.1 Speed of Male vs Female

To investigate if there were differences in walking speed between male and female participants, an independent-samples *t*-test was conducted. Preliminary analyses were performed to ensure non violation of the assumptions of normality, and homogeneity of variance. There was no significant difference in the walking speed for males (M=1.03, SD=0.21) and females (M=1.05, SD=0.18); t (99) =-0.35, p=0.72 (two-tailed). Figure 14 presents the difference in the average speed of male and female participants.

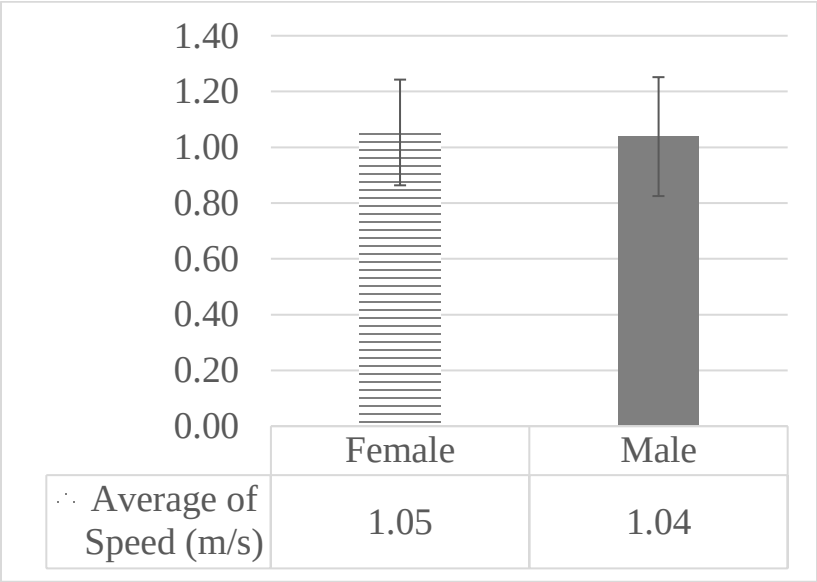

Figure 14: Average of the walking speed of males and females.

4.3.2 Speed of Faller vs Non-Faller

Additionally, to further investigate the effect of speed on the differences observed between the F and NF an independent-samples *t*-test was conducted to compare the walking speed of the faller and non-faller participants. Preliminary analyses were performed to ensure non violation of

the assumptions of normality and homogeneity of variance. The results indicated that non-fallers (M=1.13, SD=0.15) had a significantly higher walking speed than fallers (M=0.92, SD=0.18); t (97) =6.07, p<0.001 (two-tailed). Figure 15 presents the difference in the average speed of F and NF.

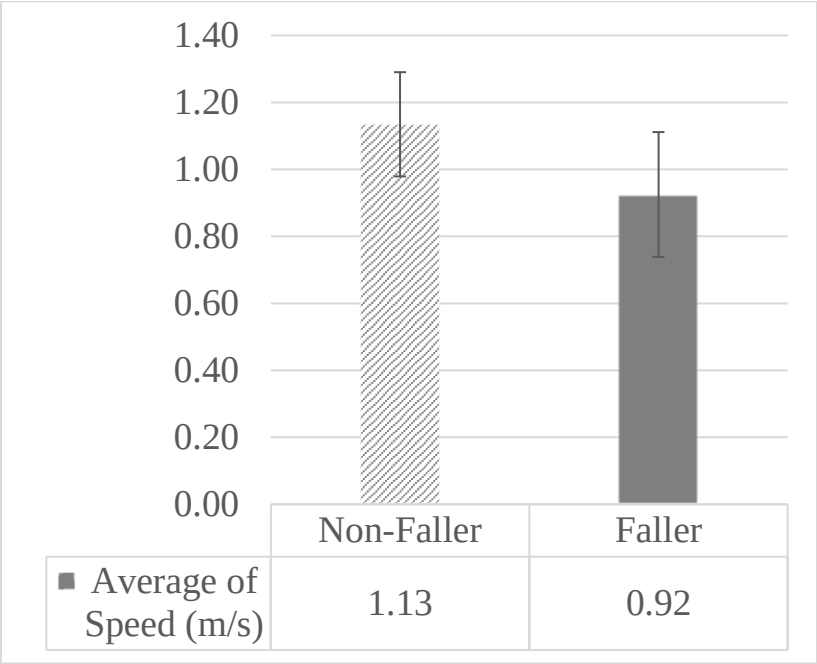

Figure 15: Average walking speed of Fallers and Non-fallers.

4.3.3 Stance time of Fallers vs Non-Fallers

The stance time- the time (sec) one foot is in contact with the floor- was also compared between the two groups; an independent-samples *t*-test was conducted to compare the stance time of the F and NF participants. Preliminary analyses were performed to ensure non violation of the assumptions of normality and homogeneity of variance. The results indicated that fallers (M=0.51, SD=0.09) had a significantly higher stance time than non-fallers

(M=0.46, SD=0.11); *t* (97) = -2.15, *p*=0.03 (two-tailed). Figure 16 presents the difference in the average speed of F and NF.

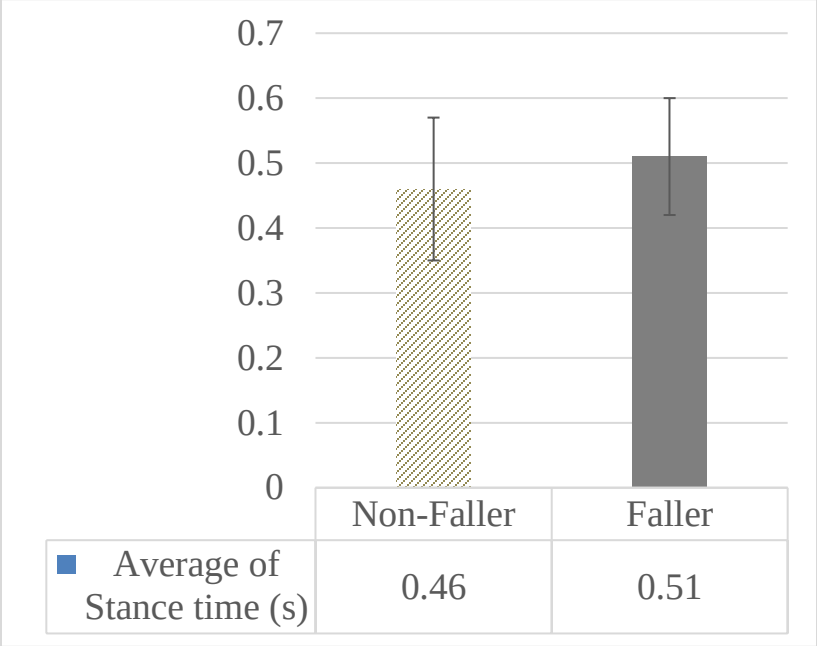

Figure 16: Average stance time (s) of Fallers and Non-fallers.

4.3.4 Walking-COP variables

The Walking-COP variables that were extracted from the walking trials were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the Walking-COP variables for the two groups and the results of the independent sample t-tests are presented in Table 8.

Table 8 : The mean ± SD of each of the Walking-COP variables.

COP Variable	Fallers	Non Fallers	P-Value
LR_COP_T_Disp (cm)	0.03 ± 0.01	0.05 ± 0.01	<0.001*
LR_COP_T_Disp_SD	0.92 ± 0.31	1.24 ± 0.40	<0.001*
LR_COP_{ML}_Disp (cm)	0.01 ± 0.00	0.02 ± 0.01	0.001*
LR_COP_{ML}_Disp_SD	0.47 ± 0.21	0.63 ± 0.30	0.003
LR_COP_{AP}_Disp (cm)	0.03± 0.01	0.04 ± 0.01	<0.001*
LR_COP_{AP}_Disp_SD	0.72 ± 0.27	0.96 ± 0.38	0.001*
MSt_COP_T_Disp (cm)	0.02 ± 0.01	0.02 ± 0.00	0.883
MSt_COP_T_Disp_SD (cm)	0.00 ± 0.01	0.00 ± 0.00	0.095
MSt_COP_{ML}_Disp (cm)	0.00 ± 0.00	0.00 ± 0.00	0.337
MSt_COP_{ML}_Disp_SD	0.00 ± 0.00	0.00 ± 0.00	0.099
MSt_COP_{AP}_Disp (cm)	0.02 ± 0.01	0.02 ± 0.00	0.946
MSt_COP_{AP}_Disp_SD	0.00 ± 0.00	0.00 ± 0.00	0.106
TSt_COP_{ML}_Disp	0.02 ± 0.00	0.01 ± 0.00	0.080
TSt_COP_T_Disp_SD	0.01 ± 0.00	0.00± 0.00	0.024
TSt_COP_{ML}_Disp_SD	0.00 ± 0.00	0.004 ± 0.00	0.189
TSt_COP_{ML}_Disp_SD	0.00 ± 0.00	0.00 ± 0.00	0.009
TSt_COP_{AP}_Disp (cm)	0.02 ± 0.00	0.01 ± 0.00	0.099
TSt_COP_{AP}_Disp_SD	0.01 ± 0.00	0.00 ± 0.00	0.039
PS_COP_T_Disp (cm)	0.02 ± 0.00	0.02 ± 0.00	0.255
PS_COP_T_Disp_SD	0.02 ± 0.01	0.02 ± 0.00	0.782
PS_COP_{ML}_Disp (cm)	0.00 ± 0.00	0.00 ± 0.00	0.366
PS_COP_{ML}_Disp_SD	0.01 ± 0.00	0.01 ± 0.00	0.354
PS_COP_{AP}_Disp (cm)	0.02 ± 0.00	0.02 ± 0.00	0.536
PS_COP_{AP}_Disp_SD	0.02 ± 0.00	0.02 ± 0.00	0.607

The results of the independent t tests, comparing the Walking-COP variables between fallers and non-fallers.

The result of the independent t-tests with an adjusted alpha level of 0.002 (Bonferroni correction: $^{0.05}/_{24} = 0.002$) indicated that from the 24 Walking-COP variables the following 5 variables showed significant differences between older adult fallers and non-fallers:

1. The total displacement of the COP movement in the LR sub-phase of gait ($LR_COP_T_Disp$) was significantly *higher* in the NF group (0.053 ± 0.017) compared to the F group (0.039 ± 0.013); $p<0.001$, Cohen's d=0.94.
2. The variability (SD) in the total displacement of the COP movement in the LR sub-phase of gait (LR_COPT_Disp_SD) was significantly *higher* in the NF group (1.244 ± 0.40) compared to the F group (0.921 ± 0.317); $p<0.001$, Cohen's d=0.87.
3. The instantaneous displacement in the ML direction of COP movement in the LR sub-phase of gait ($LR_COP_{ML}_Disp$) was significantly *higher* in the NF group (0.021 ± 0.01) compared to the F group (0.015 ± 0.006); $p=0.001$, Cohen's d=0.66.
4. The instantaneous displacement in the AP direction of COP movement in the LR sub-phase of gait (LR_COPAP_Disp) was significantly *higher* in the NF group (0.044 ± 0.014) compared to the F group (0.033 ± 0.011); $p<0.001$, Cohen's d=0.87.
5. The variability (SD) in instantaneous displacement in the AP direction of COP movement in the LR sub-phase of gait (LR_COPAP_Disp_SD) was significantly *higher* in the NF group (0.960 ± 0.382) compared to the F group (0.725 ± 0.277); $p=0.001$, Cohen's d=0.68.

Overall, significant differences for Walking-COP variables between the two groups (F vs. NF) were observed during the Loading Response (LR) sub-phase of gait. Non-fallers demonstrated higher COP displacement and COP displacement variability across most Walking-COP measures during the LR sub-phase of gait. The Cohen d value for all the significant COP measures were larger than 0.8, which indicates a large effect size. The only exceptions were LR_COPAP_Disp_SD and LR_COP_{ML}_Disp, which had a medium effect size.

4.3.5 Walking-GRF variables

The Walking-GRF variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the Walking-GRF variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 9.

Table 9: The mean ± SD of each of the Walking-GRF variables.

GRF Variable	Fallers	Non Fallers	P-Value
Loading Rate (BW/Stance)	3.78 ± 0.60	4.59 ± 0.85	<0.001*
Fz1 (BW)	1.02 ± 0.04	1.056± 0.06	0.004
Tz1 (Stance)	0.27 ± 0.03	0.24 ± 0.03	<0.001*
Fz2 (BW)	0.86 ± 0.04	0.80 ± 0.06	<0.001*
Tz2 (Stance)	0.48 ± 0.04	0.46 ± 0.04	0.080
Fz3 (BW)	1.05 ± 0.04	1.08 ± 0.04	0.001*
Tz3 (Stance)	0.75 ± 0.02	0.76 ± 0.01	0.005
Fx1 (BW)	-0.06 ±0.01	-0.07 ± 0.01	<0.001*
Tx1 (Stance)	0.16± 0.047	0.14 ± 0.03	0.016
Fx2 (BW)	0.00 ± 0.00	0.00± 0.00	0.059
Tx2 (Stance)	0.41 ± 0.04	0.40 ± 0.03	0.275
Fx3 (BW)	0.11 ± 0.01	0.13 ± 0.01	<0.001*
Tx3 (Stance)	0.82 ± 0.04	0.84 ± 0.01	0.013

The results of the independent t tests, comparing the Walking- GRF variables between groups (F vs NF)

The result of the independent t tests with an adjusted alpha level of 0.003 (Bonferroni correction: $^{0.05}/_{13} = 0.003$) indicated that from the 13 Walking-GRF variables, the following 6 variables showed significant differences between older adult fallers and non-fallers:

1. The loading rate was significantly *higher* in the NF group (4.592 ± 0.858) compared to the F group (3.787 ± 0.606); $p<0.001$, Cohen's d=0.84.
2. The relative time to reach the vertical component of the GRF's first peak (Tz1) was significantly *lower* in the NF group (0.240 ± 0.033) compared to the F group (0.278 ± 0.039); $p<0.001$, Cohen's d=-1.05.

3. The minimum vertical component of the GRF, which occurs during mid stance (Fz2) was significantly *lower* in the NF group (0.803 ± 0.061) compared to the F group (0.862 ± 0.040); $p<0.001$, Cohen's d=-1.11.

4. The second peak of the vertical component of the GRF, which occurs before toe off (Fz3) was significantly *higher* in the NF group (1.083 ± 0.046) compared to the F group (1.051 ± 0.045); $p=0.001$, Cohen's d=0.69.

5. First peak of the AP component of the GRF, which represents the maximum breaking force (Fx1) was significantly *higher* in the NF group (-0.077 ± 0.016) compared to the F group (-0.061 ± 0.012); $p<0.001$, Cohen's d=-1.15.

6. Second peak of the AP component which represents the maximum propulsion force (Fx3) was significantly *higher* in the NF group (0.133 ± 0.018) compared to the F group (0.110 ± 0.018); $p<0.001$, Cohen's d=1.24.

Overall, significant differences between fallers and non-fallers were observed among the Walking-GRF variables in the loading rate, Tz1, Fz2, Fz3, Fx1, and Fx3; with the exception of Tz1 and Fz2, Non-fallers demonstrated significantly higher measures for the mentioned variables. Additionally, the Cohen d value for all the significant GRF measures were larger than one with the exception of Fz3 (medium effect size). This shows that the difference between the two means in all these measures were larger than on standard deviation.

4.3.6 Walking-plantar pressure-variables

The walking- plantar pressure variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the walking- plantar pressure variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 10.

Table 10: The mean ± SD of each of the Walking- plantar pressure variables.

Plantar pressure Variable	Fallers	Non Fallers	P-Value
PkCP(N/mm^2)	105.89 ± 24.28	110.50 ± 23.7	0.33
FTI (N*Sec)	5.93 ± 0.64	5.71 ± 0.60	0.05
PTI (N*Sec)	0.07 ± 0.01	0.06 ± 0.01	0.51
MP(N/ mm^2)	0.09 ± 0.01	0.10 ± 0.01	0.08
PP(N/mm^2)	14.26 ± 2.79	15.69 ± 3.26	0.02

The results of the independent t-tests, for the Walking-plantar pressure variables

The result of the independent t-tests with an adjusted alpha level of 0.01 (Bonferroni correction: $0.05/5 = 0.01$) indicated there were no significant differences in the five Walking-plantar pressure variables between the two groups (F vs NF).

4.3.7 Walking speed as a covariate

To investigate the effect of participant's speed on the COP, GRF, and plantar pressure measures during gait univariate analysis of variance with a covariate (ANCOVA) was conducted. The mean ± SD for each of the

walking variables for the two groups, and results of the ANCOVA are presented in Table 11.

Table 11: The ANCOVA results and mean ± SD of all the walking variables.

Variable	Fallers	Non Fallers	ANCOVA P-Value
COPLRDisp (cm)	0.03 ± 0.013	0.05± 0.017	0.14
COPLRDispStd	0.921 ± 0.317	1.244 ± 0.400	0.52
COPLRyDisp (cm)	0.015 ± 0.006	0.021 ± 0.010	0.27
COPLRYDispStd	0.475 ± 0.212	0.631 ± 0.300	0.26
COPLRxDisp (cm)	0.033 ± 0.011	0.044 ± 0.014	0.23
COPLRXDispStd	0.725 ± 0.277	0.960 ± 0.382	0.13
COPMStDisp (cm)	0.023 ± 0.012	0.023 ± 0.006	0.68
COPMStDispStd (cm)	0.009 ± 0.011	0.006 ± 0.003	0.47
COPMStyDisp (cm)	0.006 ± 0.006	0.005 ± 0.002	0.53
COPMStyDispStd (cm)	0.004 ± 0.006	0.002 ± 0.001	0.9
COPMStxDisp (cm)	0.022 ± 0.011	0.022 ± 0.006	0.57
COPMStxDispStd (cm)	0.009 ± 0.009	0.006 ± 0.003	0.38
COPTStDisp (cm)	0.021 ± 0.006	0.019 ± 0.005	0.38
COPTStDispStd (cm)	0.011 ± 0.006	0.009 ± 0.003	0.06
COPTStyDisp (cm)	0.004 ± 0.002	0.004 ± 0.001	0.87
COPTStyDispStd (cm)	0.004 ± 0.003	0.002 ± 0.001	0.59
COPTStxDisp (cm)	0.020 ± 0.006	0.018 ± 0.005	0.058
COPTStxDispStd (cm)	0.011 ± 0.006	0.008 ± 0.003	0.41
COPPSDisp (cm)	0.023 ± 0.006	0.025 ± 0.006	0.85
COPPSDispStd (cm)	0.024 ± 0.011	0.024 ± 0.009	0.82
COPPSyDisp (cm)	0.008 ± 0.003	0.009 ± 0.004	0.91
COPPSyDispStd (cm)	0.011 ± 0.008	0.012 ± 0.008	0.7
COPPSxDisp (cm)	0.021 ± 0.005	0.021 ± 0.005	0.65
COPPSxDispStd (cm)	0.021 ± 0.009	0.020 ± 0.007	0.46
LoadingRate (BW/Frame)	3.787 ± 0.606	4.592 ± 0.858	0.04*
Fz1 (BW)	1.021 ± 0.048	1.056 ± 0.066	0.1
Tz1 (Stance)	0.278 ± 0.039	0.240 ± 0.033	0.008*
Fz2 (BW)	0.862 ± 0.040	0.803 ± 0.061	0.018*
Tz2 (Stance)	0.484 ± 0.043	0.468 ± 0.044	0.08
Fz3 (BW)	1.051 ± 0.045	1.083 ± 0.046	0.03*
Tz3 (Stance)	0.755 ± 0.026	0.768 ± 0.016	0.14
Fx1 (BW)	-0.06 ± 0.01	-0.077 ± 0.016	0.005*
Tx1 (Stance)	0.166 ± 0.047	0.145 ± 0.032	0.36

Variable	Fallers	Non Fallers	ANCOVA P-Value
Fx2 (BW)	0.000 ± 0.000	0.000 ± 0.000	0.9
Tx2 (Stance)	0.414 ± 0.043	0.405 ± 0.036	0.37
Fx3 (BW)	0.110 ± 0.018	0.133 ± 0.018	0.002*
Tx3 (Stance)	0.829 ± 0.044	0.848 ± 0.016	0.89
PkCP(N/cm^2)	105.89± 24.28	110.50 ± 23.73	0.34
FTI (N*Sec)	5.936 ± 0.641	5.713 ± 0.608	0.16
PTI (N*Sec)	0.071 ± 0.013	0.069 ± 0.013	0.07
MP(N/mm2)	0.09 ± 0.01	0.10 ± 0.01	0.2
PP(N/cm^2)	14.26 ± 2.79	15.69 ± 3.26	0.07

The mean ± SD and the results of the ANCOVA test, for all the Walking variables (COP, GRF, and Plantar Pressure).

The result of the test indicated that only the GRF measures remained significantly different between the two groups after the participant's walking speed was added as a covariate. Similar to the results of the independent t tests the Loading rate: F (1, 99) = 4.12, p=0.04; Tz1: F (1, 99) = 7.40, *p*=0.008; Fz2: (1, 99) = 5.82, *p*=0.01; Fz3: F (1, 99) = 4.46; *p*=0.03; Fx1: F (1, 99) = 8.28, , *p*=0.005; Fx2: F (1, 99) = 9.96, *p*=0.002 were significantly different between the fallers and non-fallers.

4.4 Standing tests

The participants were asked to perform two standing tests for 30 seconds: Eyes Closed Narrow Stance (ECNS) and Eyes Open semi-Tandem Stance (EOTS). Each test condition was repeated three times and the data was divided into 3 parts (COP variables, GRF variables, and plantar pressure variables). The standing tests' variables are compared between groups (F vs NF) for each test condition (ECNS and EOTS) in the following sections.

4.4.1 ECNS-COP Variable

The ECNS-COP variables that were extracted from the standing trials were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the ECNS-COP variables for the two group and the results of the independent sample t tests (two-tailed) are presented in Table 12.

Table 12: The mean ± SD of each of the ECNS-COP variables.

COP Variable	**Fallers**	**Non-Fallers**	**P-Value**
SampEnt-$COP_{\Delta\Phi}$	1.13 ± 0.14	1.18 ± 0.14	0.07
$COP_{EllArea}$ (cm^2)	13.79 ± 6.98	10.46 ± 5.8	0.01*
COP_T-Vel_{Stand} (cm/sec)	6.05 ± 1.93	5.40 ± 1.80	0.12
COP_{AP}-Vel_{Stand} (cm/sec)	5.11 ± 1.81	4.44 ± 1.71	0.10
COP_{ML}-Vel_{Stand} (cm/sec)	2.34 ± 0.75	2.32 ± 0.79	0.84

The results of the independent t tests, comparing the ECNS-COP variables.

The result of the independent t tests with an adjusted alpha level of 0.01 (Bonferroni correction: $^{0.05}/_{5} = 0.01$) indicated that from the 5 ECNS-COP variables the COP 95% confidence ellipse around the COP coordinates ($COP_{EllArea}$) showed significant differences between older adult fallers and non-fallers. The $COP_{EllArea}$ was significantly smaller for the NF group (13.790 ± 6.981) compared to the F group (10.466 ± 5.822); p=0.01 Cohen's d=-0.50 (medium effect size).

4.4.2 ECNS-GRF Variable

The ECNS-GRF variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the ECNS -GRF variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 13.

Table 13: The mean ± SD of each of the ECNS-GRF variables.

GRF Variable	Fallers	Non Fallers	P-Value
Fz_{Stand}_SD (BW)	0.013 ± 0.001	0.013 ± 0.002	0.36
Fy_{Stand}_SD (BW)	0.003 ± 0.003	0.004 ± 0.004	0.47
Fx_{Stand}_SD (BW)	0.002 ± 0.001	0.002 ± 0.001	0.10

The results of the independent t tests, comparing the ECNS-GRF variables.

The result of the independent t-tests with an adjusted alpha level of 0.016 (Bonferroni correction: $^{0.05}/_{3} = 0.016$) indicated there were no significant differences in the three ECNS-GRF variables between the two groups (F vs NF).

4.4.3 ECNS-plantar pressure Variable

The ECNS - plantar pressure variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the ECNS-plantar pressure variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 14.

Table 14: The mean ± SD of each of the ECNS- plantar pressure variables.

plantar pressure	**Fallers**	**Non Fallers**	**P-Value**
PkCP (N/mm^2)	0.49 ± 0.18	0.40 ± 0.13	0.009*
FTI (N*Sec)	141.00 ± 37.90	150.39 ± 34.45	0.20
PTI (N*Sec)	0.95 ± 0.24	0.99 ± 0.24	0.37
MP (N)	0.07± 0.01	0.06± 0.01	0.28
PP (N/mm^2)	0.09 ± 0.02	0.09 ± 0.01	0.19

The results of the independent t tests, comparing the ECNS- plantar pressure variables between groups (F vs NF).

The result of the independent t tests with an adjusted alpha level of 0.01 (Bonferroni correction: $^{0.05}/_{5} = 0.01$) indicated that from the 5 ECNS-plantar pressure variables, the Peak Contact Pressure (PkCP) showed significant differences between older adult fallers and non-fallers. The F group demonstrated significantly higher PkCP (0.491 ± 0.188) compared to the NF group (0.404 ± 0.132); p=0.009, Cohen's d=-0.55 (medium effect size).

4.4.4 EOTS-COP Variable

The EOTS-COP variables that were extracted from the walking trials were averaged for each participant and compared between the Faller (F) and the Non-Faller (NF) groups. The mean ± SD of each of the EOTS-COP variables for the two group and the results of the independent sample t-tests (two-tailed) are presented in Table 15.

Table 15: The mean ± SD of each of the EOTS-COP variables.

COP Variable	Fallers	Non-Fallers	P-Value
SampEnt-$COP_{\Delta\Phi}$	1.23 ± 0.08	1.24 ± 0.09	0.84
$COP_{EllArea}$ (cm^2)	15.73 ± 5.55	13.23 ± 5.00	0.02
COP_T-Vel_{Stand} (cm/sec)	6.34 ± 1.36	6.05 ± 1.60	0.81
COP_{AP}-Vel_{Stand} (cm/sec)	4.90 ± 1.39	4.57 ± 1.58	0.48
COP_{ML}-Vel_{Stand} (cm/sec)	3.01 ± 0.81	3.09 ± 0.75	0.35

The results of the independent t tests, comparing the EOTS-COP variables.

The result of the independent t-tests with an adjusted alpha level of 0.01 (Bonferroni correction: $^{0.05}/_{5} = 0.01$) indicated that there were no significant differences in the five EOTS-COP variables between the two groups (F vs NF).

4.4.5 EOTS-GRF Variable

The EOTS-GRF variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the EOTS -GRF variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 16.

Table 16: The mean ± SD of each of the EOTS-GRF variables.

GRF Variable	Fallers	Non Fallers	P-Value
Fz_{Stand}_SD (BW)	0.013 ± 0.001	0.013 ± 0.001	0.41
Fy_{Stand}_SD (BW)	0.005 ± 0.005	0.006 ± 0.005	0.23
Fx_{Stand}_SD (BW)	0.002 ± 0.000	0.002 ± 0.001	0.95

The results of the independent t tests, comparing the EOTS-GRF variables.

The result of the independent t-tests with an adjusted alpha level of 0.016 (Bonferroni correction: $^{0.05}/_{3} = 0.016$) indicated there were no significant differences in the three EOTS-GRF variables between the two groups (F vs NF).

4.4.6 EOTS-plantar pressure Variable

The EOTS - plantar pressure variables that were extracted from each walking trial were averaged for each participant and compared between the F and the NF groups. The mean ± SD of each of the EOTS-plantar pressure variables for the two groups, and results of the independent t-tests (two-tailed) are presented in Table 17.

Table 17: The mean ± SD of each of the EOTS-plantar pressure variables.

plantar pressure	**Fallers**	**Non-Fallers**	**P-Value**
PkCP (N/mm^2)	0.60 ± 0.19	0.54 ± 0.17	0.09
FTI (N*Sec)	143.74 ± 34.81	163.05± 115.30	0.42
PTI (N*Sec)	0.93 ± 0.23	0.97 ± 0.23	0.39
MP ((N/mm^2)	0.07± 0.01	0.06± 0.01	0.57
PP (N/mm^2)	0.09 ± 0.02	0.10 ± 0.09	0.64

The results of the independent t tests, for the EOTS-plantar pressure variables.

The result of the independent t tests with an adjusted alpha level of 0.01 (Bonferroni correction: $^{0.05}/_{5} = 0.01$) indicated that there were no significant differences in the 5 EOTS-plantar pressure variables between the two groups (F vs NF).

4.5 Summary

To investigate the differences between F and NF in the means of three sets of biomechanical variables (COP variables, GRF variables, and plantar pressure variables) independent-samples t-tests with Bonferroni corrections were conducted. The two groups (F and NF) participated in walking trials and standing trials; the latter had two conditions of Eyes Closed Narrow Stance (ECNS) and Eyes Open semi-Tandem Stance (EOTS). The results of the comparisons of the three sets of biomechanical variables between the two groups, for each test are as following:

1. Walking

- Fallers demonstrated significantly lower COP displacement and COP displacement variability across most Walking-COP measures during the LR sub-phase of gait
- Fallers demonstrated significantly lower loading rate, Fz3, Fx1, and Fx3; significantly higher Tz1 and Fz2 across the Walking-GRF measures.
- Fallers had a significantly lower walking speed and higher stance time, and there was a significantly medium negative correlation between participant's walking speed and their stance time ($r=-0.45$, $n=99$, $p<0.001$).

2. Standing-ECNS

- Fallers demonstrated a significantly larger $COP_{EllArea}$ across the ECNS-COP measures.
- There were no significant differences between fallers and non-Fallers across the ECNS-GRF measures.

- Fallers demonstrated a significantly higher PkCP across the ECNS-plantar pressure measure.

3. Standing-EOTS

- There were no significant differences between F and NF across the EOTS-COP measures.
- There were no significant differences between F and NF across the EOTS-GRF measures.
- There were no significant differences between F and NF across the EOTS-plantar pressure measures.

The summary of the significant findings for the walking and standing tests are shown in Table 18. The table also displays whether the fallers had a higher (↑) or lower (↓) average compared to the non-faller for each of the significant variables.

Table 18: Dependent variables with significant differences between Faller and Non-fallers.

Measures		**Walking Test**	**Standing-ECNS**
COP	LR_COP_T_Disp (cm)	Faller ↓	
	LR_COP_T_Disp_SD	Faller ↓	
	LR_COP_{ML}_Disp (cm)	Faller ↓	
	LR_COP_{AP}_Disp (cm)	Faller ↓	
	LR_COP_{AP}_Disp_SD	Faller ↓	
	$COP_{EllArea}$		Faller ↑
GRF	Loading Rate (Bw/Stance)	Faller ↓	
	Tz1 (Stance)	Faller ↑	
	Fz2 (BW)	Faller ↑	
	Fz3 (BW)	Faller ↓	
	Fx1 (BW)	Faller ↓	
	Fx3 (BW)	Faller ↓	
Plantar pressure	PkCP		Faller ↑
Other	Walking speed (m/s)	Faller ↓	
	Stance time (sec)	Faller ↑	

The Faller's measures were significantly higher (↑) or lower (↓) than Non-Faller.

5. Discussion

The purpose of this study was to compare biomechanical data (COP, GRF and plantar pressure) during gait and standing in older adult fallers and non-fallers. To accomplish this objective, data from 102 participants over the age of 65 were analyzed. The outcome measures of the standing and walking tests were compared between older adult fallers (N=41) and non-fallers (N=61), using independent sample t-tests and Bonferroni corrections.

5.1 Balance Scores

At the beginning of each data collection session, the participants were examined by the researchers using two balance tests. Each participant received two balance scores, one for the Berg Balance Scale (BBS) and one for the Brief -BESTest. The scores were then used to compare between F and NF.

5.1.1 Berg Balance Score

This balance scale evaluates the ability to stand up and maintain the standing position during different tasks. The non-fallers (M=54.1, SD=1.6) had a significantly higher BBS score than fallers (M=46.4, SD=5.2). Higher BBS scores indicate better balance, while, lower scores are associated with increased risk of injurious falls (Downs, Marquez, & Chiarelli, 2014). Additionally, Franchignoni & Velozo (2005), suggest that individuals who scores higher than 45 on this test are more likely to pass challenging balance tasks such as stand on one leg and tandem stance.

Moreover, clinicians consider a score below 45 in the BBS an indication of the need for an aid or supervision while walking (Stevenson, Connelly, Murray, Huggett, & Overend, 2010). The average score of fallers was 46.5; this score is close to 45 , which indicates poor balance; non-fallers average score was 54.1; this score is close to 56, which indicates functional balance (Berg, Wood-Dauphinee, Williams, & Maki, 1992). Therefore, fallers showed a significantly poorer balance compared to non-fallers. This finding is consistent with other researchers who have attributed poor balance to be one of the main risk factors for falling in the older adult population (Arfken, Lach, Birge, & Miller, 1994; Austin, Devine, Dick, Prince, & Bruce, n.d.; Joshi & Sadhale, 2014; Stel, Smit, Pluijm, & Lips, 2003; Tinetti, Speechley, & Ginter, 1988).

Furthermore, according to some researchers a BBS score of 45 and less has been associated with high risk of falling (Mbourou, Lajoie, & Teasdale, 2003; Shumway-Cook, Brauer, & Woollacott, 2000b; Thorbahn, D, & Newton, 1996).Nevertheless, only 37% of older adult fallers who participated in this study scored less than 45 in the BBS. Figure 17 displays the percent_cumulative-distribution of F and NF with respect to their BBS score. The mass distribution of fallers and non-fallers was calculated for each BBS score. Afterwards, the cumulative distribution was calculated by dividing cumulative mass by the number of participants in each group (The values are presented in percent for simplicity).

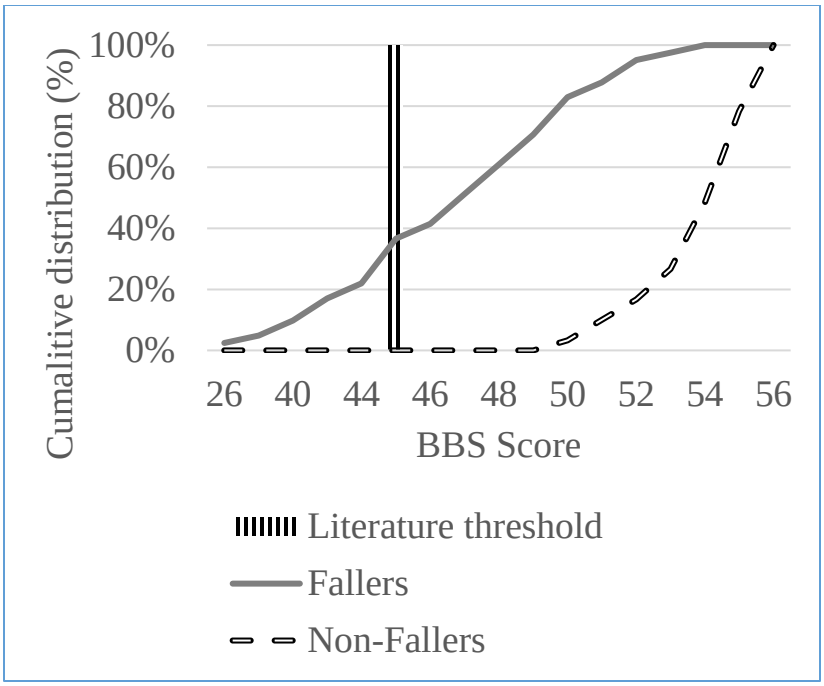

Figure 17: The %cumulative-distribution of F and NF with respect to their BBS. *The filled line represents the faller group the dotted line represents the non-faller group, the vertical line represents the cutoff point recommend by the literature.*

As shown in figure 17, 63% of the fallers who participated in this study had scored higher than 45 in their BBS, which shows that the recommended threshold of 45 was not able to properly discriminate between older adult fallers and non-fallers. This may be explained by the ceiling effect in the BBS that has been reported by other researchers (Blum & Korner-Bitensky, 2008; Lemay & Nadeau, 2010; Oliveira et al., 2011). According to Lemay & Nadeau (2010), the presence of ceiling effects limits a test's practicality, as comparison among better functioning subjects may not be possible. Although, scoring below 45 in the BBS is an indication of poor balance and a sign for potential clinical intervention, it does not perform well in

discriminating between older adult fallers and non-fallers. Therefore, some researchers suggest using the BBS in conjunction with other balance measures (Blum & Korner-Bitensky, 2008; Oliveira et al., 2011). Our results indicate that the BBS is not challenging enough to discriminate between older adult fallers and non-fallers. Perhaps a higher threshold for this test could perform better regarding this matter. According to our data, 83% of fallers who participated in this study scored below 50 in the BBS. More research among community dwelling older adults is required to find an adequate BBS threshold that can discriminate between older adult fallers and non-fallers.

5.1.2 Brief-BESTest

The Brief-BESTest, which is an abbreviated version of the Balance Evaluation Scale Test (BESTest), examines the components contributing to balance dysfunction in 6 subsystems of postural control. The non-fallers (M=18.45, SD=2.9) had a significantly higher Brief-BESTest score than fallers (M=11.59, SD=2.8). Gaining low scores in the Brief-BESTest is an indication of poor balance (Fay B. Horak, Wrisley, & Frank, 2009). However, few studies have assessed the balance of older adult fallers and non-fallers using the Brief-BESTest. Duncan et al. (2013), conducted a study to discriminate between older adult fallers and non-fallers with Parkinson's disease (PD). The authors stated that gaining a score of less than 11 in the Brief-BESTest was able to adequately detect retrospective fallers in the PD population. In the

present study, the fallers average score in the Brief-BESTest (11.6±2.8) was closer to the suggested threshold compared to non-fallers average score (18.4±2.9) in this test. Therefore, fallers showed a significantly poorer balance compared to non-fallers which is consistent with the literature regarding perceiving poor balance as one of the main risk factors for falling in the older adult population. (Arfken et al., 1994; Austin et al., n.d.; Joshi & Sadhale, 2014; Stel et al., 2003b; Tinetti et al., 1988).

According to Duncan et al. (2013) a threshold of 11 and lower was able to correctly identify 45.8% of older adult fallers with PD. Consistent with their findings our results showed that approximately half (51%) of the older adult fallers who participated in this study had a Brief-BESTest score of 11 and lower. Figure 18 displays the percent cumulative-distribution of F and NF with respect to their Brief-BESTest score. The mass distribution of fallers and non-fallers was calculated for each Brief-BESTest. Afterwards, the cumulative distribution was calculated by dividing cumulative mass by the number of participants in each group.The values are presented in percent for simplicity.

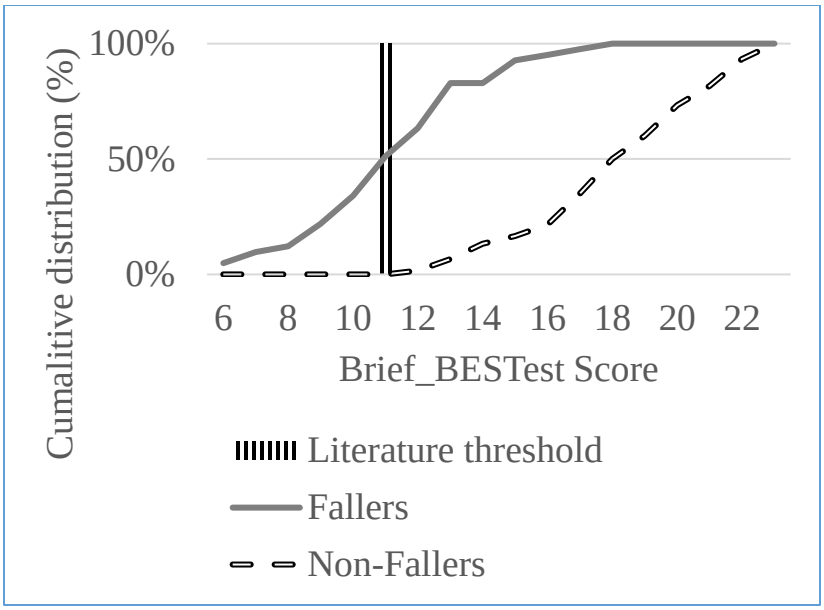

Figure 18: The %cumulative-distribution of F and NF with respect to their Brief_BESTest scores. *The filled line represents the faller group the dotted line represents the non-faller group, the vertical line represents the cutoff point recommend by the literature.*

As shown in Figure 18, 93% older adult fallers who participated in this study scored 15 or lower on the Brief-BESTest, which suggests that a higher threshold could perform better in discriminating between older adult fallers and non-fallers. A few researchers have shown that the Brief-BESTest is a valid tool for discriminating between those at low and high risk of falls (Duncan et al., 2013; O'Hoski et al., 2015); our findings also indicate that the Brief-BESTest is a useful tool for discriminating between community dwelling older adult fallers and non-fallers. Future research can shed light on a Brief-BESTest threshold that can effectively discriminate and identify older adult fallers.

5.2 Walking test

The data obtained through each walking trial was divided into three measures: Walking-COP variables, Walking-GRF variables, Walking-plantar pressure variables, and were compared between the two groups (F vs NF). The following sections will discuss the findings for each measure.

5.2.1 Walking-COP variables

For the Walking-COP measures, each walking trial was divided into 4 sub-phases (LR, MSt, TSt, and PS). The results of the independent t-tests with an α level of 0.002 (Bonferroni correction) indicated that the only sub-phase that showed significant differences, with a medium to large effect size between the two groups, for the Walking-COP measures was the LR sub-phase. The fallers showed significantly lower COP measures in the LR sub-phase compared to the non-fallers. Studies that have investigated Walking-COP measures between healthy older adults and young individuals have reported greater COP variability during the LR (Bizovska et al., 2014; Kwon, Kwon, Park, & Kim, 2018) and PS (Bizovska et al., 2014) sub-phase of gait, which may be attributed to the weight transfer from one limb to another that occurs during these sub-phases. The results of our study revealed significant differences only in the LR sub-phase of gait between older adult fallers and non-fallers.

Additionally, Svoboda et al. (2017), investigated variability of COP displacements between older adult fallers and non-fallers during gait at self-

selected, defined, and fast speeds. After applying the Bonferroni correction for multiple comparisons, the researchers found no significant differences in COP movement variability between the two groups. Conversely, in our study, fallers demonstrated significantly lower COP displacement and COP displacement variability across the Walking-COP measures during the LR sub-phase of gait. One of the reasons for this discrepancy may be because the current study did not control for speed. To avoid gait alteration, the participants were asked to walk with their comfortable speed. However, walking speed affects a number of biomechanical gait parameters (Riley, Della Croce, & Casey Kerrigan, 2001). Furthermore, the comparison of the walking speed between the two groups indicated that the NF walking speed (1.13 ± 0.15) was significantly higher than F walking speed (0.92 ± 0.18), $p<0.001$); further analysis indicated significant correlation between the participants speed and their LR walking-COP measures. The Pearson product-moment correlation coefficient between these variables and the participants speed are shown in Table 19 for fallers and non-fallers and all the participants (Total).

Table 19: The correlation between speed and Walking-COP variables.

	COP_T_ Disp	**COP_T_Disp_ SD**	**COP_{ML}_Di sp**	**COP_{AP}_Di sp**	**COP_{AP}_Disp_ SD**
NF	.555**	.360**	.317*	.571**	.296*
F	.670**	.491**	.547**	.585**	.359*
Total	.674**	.518**	.388**	.650**	.414**

The table displays the Pearson product-moment correlation coefficient between the participants speed and their walking-COP measures during the LR sub-phase.
***. Correlation is significant at the 0.01 level (2-tailed).*
**. Correlation is significant at the 0.05 level (2-tailed).*

To visually demonstrate these correlations, Figure 19 displays one of the F and NF Walking COP measures (LR-COPT-Disp) plotted against speed.

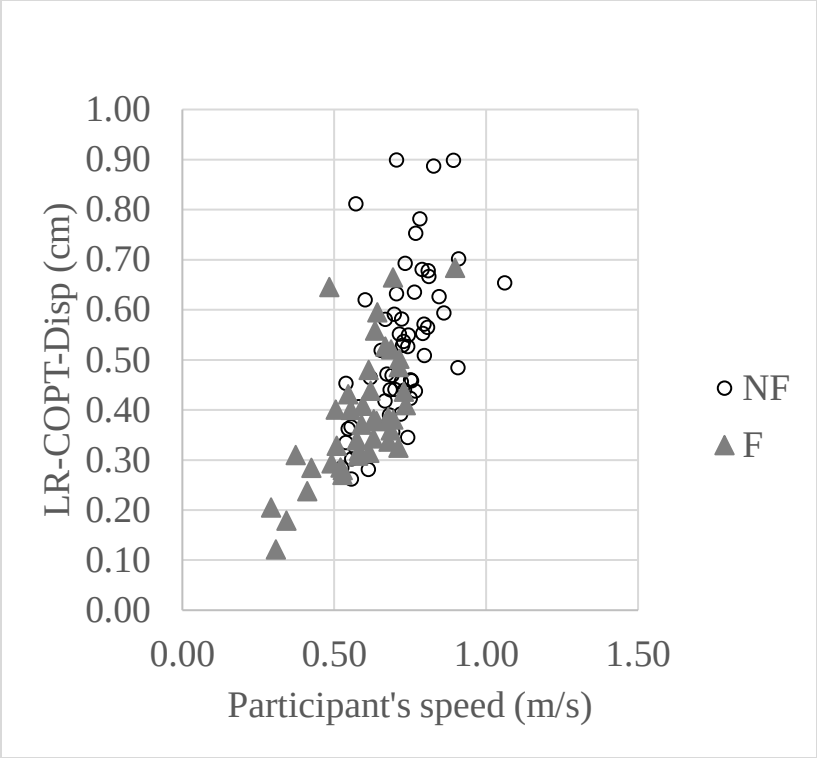

Figure 19: LR-COPT-Disp plotted against participants' speed.
The figure shows Faller's and Non-Faller's COP total displacement during the LR phase of gait plotted against their speed

The plot shows that with an increase in the participants speed, the COP total displacement in the LR sub-phase also increased for both fallers and non-fallers. Additionally, most of the fallers are in the area less than 0.70 m/s, which is an indication of their slower speed. The rest of the LR Walking-COP measures showed a similar effect when plotted against speed. Contrary to our results some studies have found significant differences in walking speed between younger and older adults, but no significant differences between fallers and non-fallers (Hausdorff et al. 1977; Pijnappels, Bobbert, van Dieën, 2005). This inconstancy may be due to the small sample size that were used in

those studies. Additionally, other clinical studies of falls risk factors confirm our finding of older adult fallers walking slower than non-fallers (Nevitt, Cummings, Kidd, & Black, 1989; Thapa, Gideon, Brockman, Fought, & Ray, 1996b; Tinetti et al., 1988).

Maki (2015), states that reduced walking speed in fallers may be a stabilizing adaptation related to fear of falling. Moreover, our results indicated that stance time (the time one foot is in contact with the floor) was significantly higher in fallers (0.51 sec ±0.09) than non-fallers (0.46 sec ±0.11), p=0.03. Similarly, a cross-sectional study investigating the gait of older adult fallers, also found that the stance time of fallers was higher compared to non-fallers (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997b). Further analysis of our data revealed that there was a significantly medium negative correlation between participant's walking speed and their stance time (r=-0.45, n=99, p<0.001). Stance time has been related to the automatic stepping mechanism; increased stance time could possibly indicate that the person is "thinking about taking each step" (Brach, Studenski, Perera, VanSwearingen, & Newman, 2007). Therefore, it is possible that the fallers higher stance time could be an indication of impairment in their automatic stepping mechanism.

Furthermore, aging is associated with an increased risk of cognitive impairments and decreased performance of cognitive tasks (Bherer, Erickson, & Liu-Ambrose, 2013); this may be due to decline in brain volume, regional atrophy and loss of white matter integrity (Soumaré et al.; 2009). The cognitive decline that is associated with aging can manifest itself in reduction

in walking speed (Watson et al., 2010). For instance, Holtzer, Verghese, Xue, & Lipton (2006), revealed that verbal IQ, speed/executive, attention, and memory were significant predictors of variance in gait velocity. In the current study, despite a Min-Cog test, which was administrated for consent purposes, no other data regarding the participants cognitive status were collected. The Mini-Cog test has a maximum score of 5 and a scoring below 3 in this test is considered "possibly impaired" (Borson, Scanlan, Chen, & Ganguli, 2003). All the participants of this study scored above 3 and there was no need for a third-party consent. Therefore, this may rule out the likelihood of dementia and cognitive impairment in the participants of our study. It is important to mention, that in addition to walking speed (Kooten, Hettinga, Duffy, Jackson, & Taylor, 2018; Zdenek Svoboda et al., 2017), COP measures can also be influenced by foot morphology (Lugade & Kaufman, 2014); this study did not control for either, thus, it is possible that the observed variations were related to the different walking speed adopted by the two groups.

5.2.2 Walking-GRF-measures

The following measures of the Vertical (V) and Anterior-Posterior (AP) components of the GRF were compared between older adult fallers and non-fallers: loading rate; first V peak force, which occurs after heel strike (Fz1); minimum V force, which occurs during mid stance (Fz2); second V peak force, which occurs before toe-off (Fz3); first AP peak (Fx1), which represents the maximum breaking force; mid stance (Fx2); second AP peak (Fx3), which

represents the maximum propulsion force; and the relative time to reach each of these forces (Tz1, Tz2, Tz3,Tx1, Tx2, Tx3). The results of the independent t-tests with an α level of 0.003 (Bonferroni correction) indicated that fallers demonstrated significantly lower loading rates, Fz3, Fx1, and Fx3; significantly higher Tz1 and Fz2 across the Walking-GRF measures with a medium to large effect size.

Loading rate, which is the manifestation of the time it takes for the V GRF to rise by one's BW (Diss, 2001), was significantly lower in fallers compared to non-fallers. This may be a result of the fallers producing smaller force and/or requiring more time to produce the force. This explanation is supported by our results, which indicate that it took the fallers significantly more time to reach the first V GRF peak (Tz1), and although not significant, the average of the Fz1 (first V GRF peak) was also lower for the fallers (1.02 BW ±0.04) compared to the non-fallers (1.056 BW ± 0.06), *p=0.004*.

During gait, the V GRF first peak (Fz1), exceeds the participants' Body Weight (BW) as full weight-bearing takes place and represents upward acceleration of the COM; then, as the participant enters mid-stance and flexes their knee, the force plates are partially unloaded and the V GRF drops below the BW and the minimum force (Fz2) happens and represents downward acceleration of the COM. The V GRF second peak (Fz2) occurs during push-off which also exceeds the body weight and represents upward acceleration of the COM; finally, the weight drops to zero as the other leg takes up the BW (Marasovic, Cecic, & Zanchi, 2009). It is noteworthy to mention that GRF

measures are speed-mediated variables. Therefore, walking speed may influence the V GRF measures, for instance walking with a higher speed may produce larger V GRF peaks (Slawomir Winiarski & Rutkowska-Kucharska, 2009). Figure 20 displays the mean V GRF of F vs NF for the significant measures.

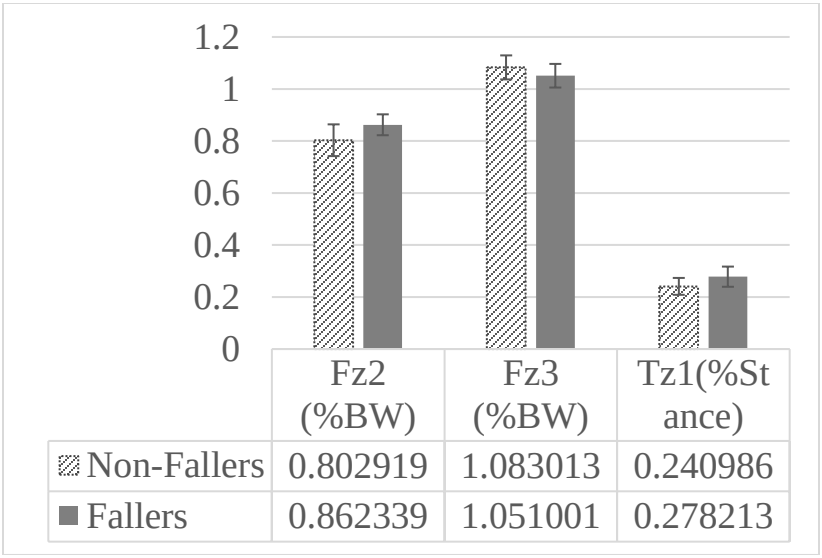

Figure 20: Average of significant V GRF measures for fallers vs non-fallers. *The graph shows that fallers exhibited a higher minimum value at mid-stance, lower Fz2 (second V GRF peak, and took longer to reach the first V GRF (Tz1)*

To date, only one recent study conducted by Kwon et al., (2018), has investigated the variation of GRF gait patterns of older adult fallers and non-fallers.The reserarchers measured the gait of 78 older adults (38 fallers and 38 non-fallers), using a portable walkway. They found that fallers showed a significantly higher Tz1 and Tz2 compared to non-fallers, however our results only showed significant difference between the two groups in Tz1; although

fallers demonstrated lower Tz2 values compared to non-fallers, the difference was not significant p=0.080. Additionally, Kwon et al., (2018), found no significant difference in Fz1 and Fz2 between F and NF. Conversely, our results indicated significantly higher Fz2, and lower Fz3 for the fallers. The variation between our results and the Kwon study could be due to the different equipment being used for the measurement of the GRF. The authors used a portable walkway (GaitRite CIR Systems Inc.Clifton, NJ, USA); our study utilized two piezoelectric force plates (Kistler Instruments Ltd, Switzerland). Since, pressure sensitive platform systems do not directly measure the applied force vector (Robertson et al., 2013), the researchers calculated the V GRF peaks using a self-developed analysis; in comparison to force plates, which have the ability of direct measurement of applied force, their device lacks the accuracy of force plates. Additionally, this disadvantage also prevented the researchers from measuring other GRF components such as the AP GRF, which was investigated in our study.

Other researchers, who assessed GRF gait patterns in different populations, have attributed the reduction in Fz3 to the participant's difficulty in supporting their own body weight during terminal stance (Ambrosia, Courchesne, & Kaufman, 1998; Hasan, Jailani, Md Tahir, & Ilias, 2017; Williams, Gibbs, Meadows, & Abboud, 2011). Perhaps, older adult fallers struggle in supporting their BW during terminal stance, hence show significantly lower Fz3 compared to non-fallers. The V GRF patterns during gait may be affected by aging; in particular, older adults show lower Fz1, Fz3,

and higher Fz2 across V GRF measures when compared to younger adults (Kwon et al., 2018). As suggested by Hessert et al., (2005), the forces needed to stabilize the ankle during heel contact are reduced in older people, and they have lesser ability to push-off in anticipation of the swing phase. Our results confirmed the same patterns with respect to older adult fallers and non-fallers. This may be interpreted as a sign of instability during gait, which increases as an individual ages and becomes more predominant when they experience a fall. Further research is needed to confirm this idea by exploring GRF gait pattern between older adult fallers and non-fallers; younger adults should also be included as a baseline.

Regarding the AP GRF peaks, which represent the maximum breaking force and maximum propulsion force respectively, fallers demonstrated significantly lower Fx1 and Fx3 than non-fallers. To the best of the author's knowledge, no other study has compared the AP GRF force between older adult fallers and non-fallers. Diss, (2001), states that an understanding of body's COM relative to the point of force application may be obtained through the maximum breaking and propulsion forces; on impact with the ground, the COM is behind the point of force application. As the COM moves forward, the breaking force prevents the foot from sliding forward as it happens when one is walking on icy or slippery surfaces (Marasovic et al., 2009). During this period, the body's COM decelerates from initial foot contact to the mid stance (Hasan et al., 2017). The propulsion is caused by the foot pushing back against the force plate (Marasovic et al., 2009), during which the COM moves forward

of the point of force application (Diss, 2001). Similar to the V GRF the AP GRF measures are also speed-dependent and may be influenced by walking slow or fast (Marasovic et al., 2009). Figure 21, displays the mean AP GRF of F vs NF for the significant measures.

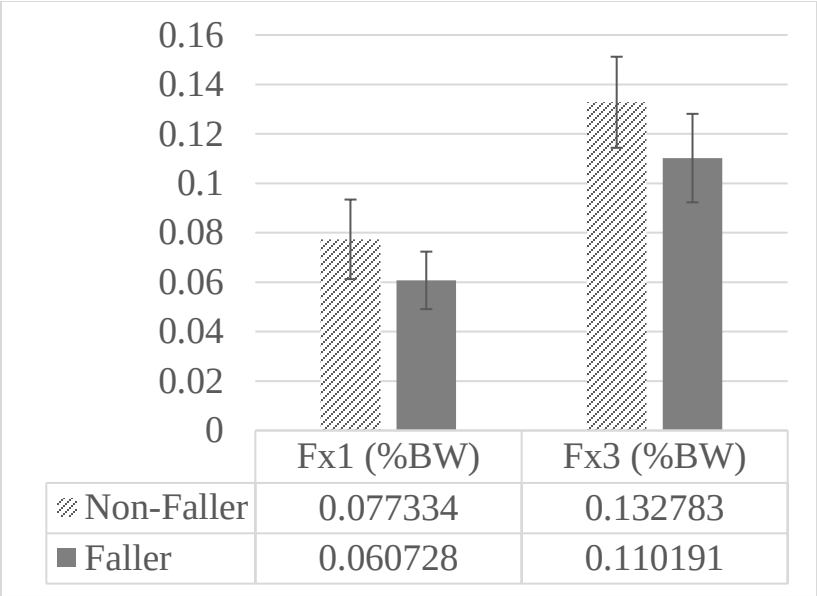

Figure 21. Average of significant AP GRF measures for fallers and non-fallers. *The graph shows that fallers exhibited lower breaking force (Fx1) and propulsion force (Fx2) than non-fallers.*

As mentioned previously, the observed AP and V GRF differences between the two groups may also be attributed to the difference in the participant's speed. NF showed a significantly higher walking speed than F; which may cause higher GRF peak values. Further analysis was conducted to investigate the relationship between gait speed and GRF measures.The Pearson product-moment correlation coefficient between the significant GRF measures

and the participants speed are shown in Table 20 for fallers, non-fallers and all the participants (Total).

Table 20: Correlation between participants speed and Walking-GRF variables.

	Tz1	Fz2	Fz3	Fx1	Fx3	Fz1
NF	-.416**	-.736**	.284*	-.545**	.566**	.472**
F	-.440**	-.358*	.074	-.485**	.557**	-.230
Total	-.558**	-.679**	.315**	-.627**	.671**	.310**

The table displays the Pearson product-moment correlation coefficient between participants speed and their significant walking-GRF measures for the LR phase

***. Correlation is significant at the 0.01 level (2-tailed).*

**. Correlation is significant at the 0.05 level (2-tailed).*

To visually demonstrate these correlations, Figure 22, displays one of the fallers and non-fallers Walking GRF measures (Fz2) plotted against speed.

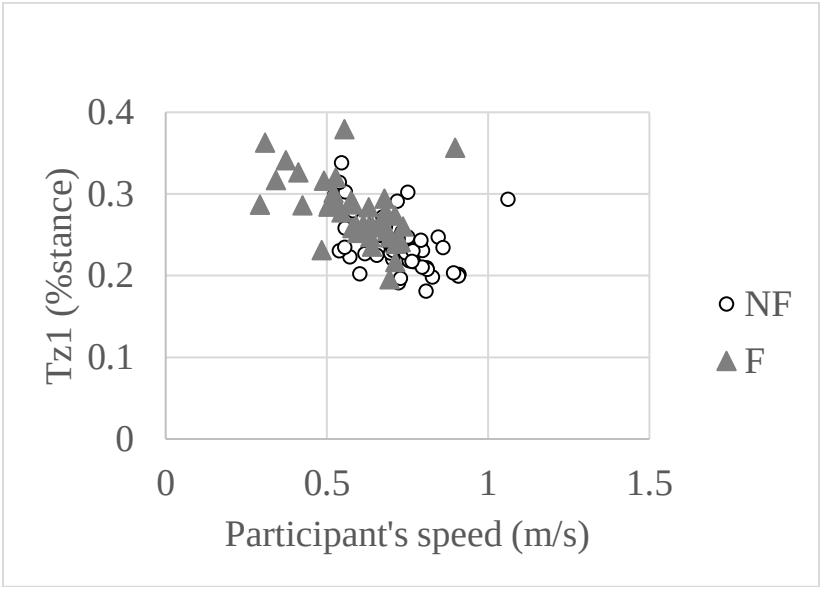

Figure 22: Time to reach the first V GRF peak (Tz1) against participants' speed. *The figure shows the time it takes Faller's and Non-Faller's to reach their first V GRF peak (Tz1) plotted against their speed.*

The plot shows that with an increase in the participants' speed, they reach the first V GRF peak faster (Tz1 decreases). Additionally, most of the fallers are in the area less than 0.70 m/s, which is an indication of their slower speed. The rest of the GRF peaks produced similar graphs when plotted against speed. Nevertheless, when the participant's speed was added as a covariate to the statistical analysis the GRF measures still remained significant. This suggests that the participant's walking speed may not have affected the GRF measures.

The lower speed adopted by fallers can be an indication of a more conservative and cautious gait strategy to maintain dynamic balance by reducing gait velocity (Kwon et al., 2018). Additionally, gait characteristics are associated with distinct brain networks in older adults (Rosano et al., 2008); these extensive neural networks are vulnerable to ischemic changes within the deep white matter, which may cause declines in both executive and motor functions (Watson et al., 2010). This can be manifested in change of gait characteristics, for instance, slower speed and longer stance time in older adults is significantly associated with reduced functioning of the visual, vestibular and sensorimotor systems (Lord, Lloyd, & Li, 1996). Furthermore, reduced gait speed can be a consequence of different muscular and/or neurological

factors such as decrease in motor units, impaired muscular activation, substitution of type II by type I fibers, diminished cutaneous sensitivity, decreased nerve conduction, decreased grey matter volume with functional brain impairment, and the presence of white matter lesions (Abellan Van Kan et al., 2009). The results of our study indicated that fallers had a significantly slower gait speed and higher stance time compared to non-fallers, which can be an indication of deficiencies in the cognitive, muscular, neurological, visual, vestibular and sensorimotor systems.

5.2.3 Walking-plantar pressure variables

The results of this study, with an adjusted alpha level of 0.01 (Bonferroni correction), indicated no significant differences in the plantar pressure measurements between F and NF during walking; therefore, our study's original hypothesis was rejected. Conversely, the only other study that compared plantar pressure measures between older adult F and NF during gait, found a significant difference in the PTI and PkCP between the two groups. Mickle et al. (2010), found that fallers had a significantly higher prevalence in foot pain compared to non-fallers. These findings led the researchers to believe that the higher prevalence of foot pain among this population is caused by the increase in pressure load under their foot. However, our results do not confirm their claim; this may be due to the different frequencies used by the studies. Mickle et al (2015),collected the plantar pressure data with a pressure platform sampling at a frequency of 25 Hz, whereas, the pressure mat used for our data

collection, it was operating with a much higher frequency of 160 Hz. According to Sullivan, Burns, Adams, Pappas, & Crosbie (2015), it is possible for platforms with lower frequencies to make errors regarding peaks in pressure or force. Therefore, the results of their study might have been affected by their platform limitations. Additionally, our study did not collect data regarding the presence of pain in the participants' feet. Foot pain may alter plantar pressure measures (Sullivan, Burns, Adams, Pappas, & Crosbie, 2015); therefore, the differences between the findings could be due to the presence of foot pain.

Though not significant, fallers showed slightly lower Peak contact Pressure (Pkcp), Mean Pressure (MP), Peak Pressure (PP), and slightly higher PTI and FTI values compared to non-fallers. Additionally, fallers exhibited a significantly higher stance time compared to non-fallers (p=0.001). Since, both the PTI and FTI measures are time-dependent variables, the fallers' slightly higher values for these measures may be attributed to their stance time. Further analysis revealed that FTI indeed had a strong positive correlation with the participant's stance time (r=0.61, n=99, p<0.001); PTI had a medium positive correlation with the participant's stance time (r=0.39, n=99, p<0.001).

A recent study investigated the effects of an exercise intervention on the improvement of plantar pressure gait parameters in an older adult group. The authors credited the decline in plantar pressure values, which was observed after the 6 month intervention, to their participants' balance improvement (Ramalho et al., 2018). According to our results, fallers obtained

significantly lower balance scores than non-fallers, which is an indication of poor balance. Therefore, Ramalho et al's., (2018) interpretation of plantar pressure gait parameters does not align with our findings. Several factors, such as anatomical differences (Cavanagh et al., 1997) and Body Mass Index (BMI), may influence plantar pressure measures. For instance, low arched foot and high BMI may increase plantar pressure variables (Obrien & Tyndyk, 2014). An independent sample t-test revealed no significant difference in the F and NF BMI (p=0.068); however, data about the participant's foot arch type was not collected.

Higher speed may also increase plantar pressure's measure (Rosenbaum et al., 1994). Some researchers, therefore believe that it is necessary to control the speed through external methods, such as treadmill walking (Cavanagh et al., 1997; Segal et al., 2004). Further analysis of the current results revealed that across the measured plantar pressure variables, only the time-dependent variables (PTI and FTI) had a significant correlation with speed. The FTI had a strong negative correlation with the participant's speed (r=0.57, n=99, p<0.001); the PTI had a medium negative correlation with the participant's speed (r=0.42, n=99, p<0.001). Yet, despite the difference in speed, there were no significant differences in the plantar pressure measures between the two groups. This could be an indication of speed being less influential on plantar pressure measures than what some research has claimed.

5.2.4 Participant's speed as a covariate

When the participant's speed was added as a covariate to the statistical analysis the COP measures were no longer significantly different, while the GRF measures remained significantly different between the two groups. Overall, the differences between ANCOVA and the independent sample t tests were:

- There were no significant differences between the COP measures for the ANCOVA, while significant differences were observed in the LR sub-phase of gait for the independent sample t tests.
- There were significant differences in the GRF measures (Loading rate, Tz1, Fz2, Fz3, Fx1, and Fx3) for both the ANCOVA and the independent sample t tests.
- There were no significant differences observed in the plantar pressure measures for both ANCOVA and t tests.

This can be interpreted as the participant's walking speed having more effect on the COP measures than the GRF measures.

5.3 Standing test

The standing tests consisted of two conditions: Eyes Closed Narrow Stance (ECNS) and Eyes Open semi-Tandem Stance (EOTS). The data obtained through each test condition were divided into three measures per condition: Standing-COP, Standing-GRF, and Standing-plantar pressure. The

measures were compared between the two groups (F vs. NF) and are discussed in the following sections for both conditions (ECNS and EOTS).

5.3.1 Standing-COP variables

The results of the independent t-tests with an alpha level of 0.01 (Bonferroni correction) indicated that across the COP measures, only the COP 95% confidence ellipse area ($COP_{EllArea}$) in the ECNS condition was significantly different between the two groups. Fallers showed a significantly larger $COP_{EllArea}$ in the ECNS condition; they also exhibited a larger $COP_{EllArea}$ for the EOTS condition as well, but the difference was not significant (p=0.02). This is congruent with previous research that have found significance differences in $COP_{EllArea}$ between F and NF (Melzer, Kurz, & Oddsson, 2010b; Norris et al., 2005; Thapa et al., 1996). The increase in elliptical area sway has been associated with weakness in the upper limbs and need for assistance with activity (Lichtenstein, Shields, Shiavi, & Burger, 1988; Thapa, Gideon, Fought, Kormicki, & Ray, 1994). Additionally, Chaudhry, Bukiet, Ji, & Findley (2011), found that when comparing between two groups those who have higher magnitude of COP_{EllAre} are considered to be less stable. According to Horak (2006), maintaining balance while standing relies on the capability of an individual's postural control system and may be affected by deficiencies in underlying physiological systems. For instance, cutaneous and joint somatosensory information from the feet and ankles (Horak, Nashner, & Diener, 1990), ankle proprioception (Bisson, McEwen, Lajoie, & Bilodeau,

2011), and hip and ankle strategies (Chaudhry et al., 2011; Horak & Nashner, 1986) have been identified as regulatory mechanisms for stabilization of the body; therefore, damage in any of these mechanisms may result in reduced stability. The increased COP_{EllAre} observed in older adult fallers indicates less stability in this group, which may be the consequence of deficiencies in the stabilization mechanisms.

Regarding SampEnt-$COP_{\Delta\Phi,}$ Rhea et al. (2014), observed a decrease in this measure in patients with an ACL rupture and attributed it to their deviation away from the healthy complexity (i.e., adaptive behavior) exhibited by healthy adults. Although Fallers demonstrated a lower SampEnt-$COP_{\Delta\Phi}$ for the ECNS condition, the difference was not significant between the two groups (p=0.07). However, the slight decrease in entropy is an indication of fallers exhibiting more regularity in their $COP_{\Delta\Phi}$ behavior, which may be interpreted as a minor decrease in flexibility regarding the adjustment of their COP to respond to perturbations than non-fallers.

Contrary to some studies, we did not find significant differences in the COP velocity measures such as: the COP Total velocity (Boulgarides, McGinty, Willett, & Barnes, 2003b) or the COP velocity in the AP direction (Brian E. Maki, Holliday, & Topper, 1994). However, as Piirtola & Era (2006), suggests, these differences may be explained by the participants' having different levels of functional ability. For instance, Maki et al. (1994), recruited their participants from frail older adults living in nursing homes, as for the current study, the participants were community-dwelling older adults.

It is possible that for frailer fallers, who have substantial balance issues, the COP measures during a simple task like standing is more sensitive than fallers who are active members of their community.

These differences could be accounted for by the differences in the test durations and participants' BOS while standing (Vieira, Oliveira, & Nadal, 2009). Furthermore, Thapa et al. (1996), found that the $COP_{EllArea}$ has a more pronounced estimated linear trend in distinguishing F and NF than the COP mean velocity. Moreover, Melzer et al. (2010), found a significant difference in $COP_{EllArea}$ between fallers and non-fallers, but no significant difference in other traditional CoP measures such as COP mean velocity.

Finally, our results indicated no significance difference in the COP measures between the two groups for the EOTS condition and our original hypothesis was rejected. The EOTS condition did not appear to be adequate in noticing differences between F and NF. This agrees with McGrath et al. (2012), who assert that despite previous research, they did not find the EOTS condition to be very discriminative between F and NF. Era et al. (2006), suggested that results of the EOTS condition may suffer from floor effect in the older adult population, which could be the reason for observing no statistical differences between the two groups for this condition.

5.3.2 Standing-GRF variables

The results of the independent t-tests with an alpha level of 0.01 (Bonferroni correction) indicated that there were no significant differences in

the GRF measures between the two groups for any of the standing test conditions; therefore, we reject our original hypothesis. Contrary to (Karlsson & Frykberg, 2000; Önell, 2000b) claims, the magnitude of GRF does not seem to be a good measure for assessing balance during standing. This is also supported by (Goldie, Bach, & Evans, 1989b), who recommend more challenging positions like standing on one-leg in order to observe GRF variations. However, standing on one leg may be a difficult task for older adult fallers to complete, while standing on both legs is not challenging enough to observe changes in the GRF measures. Therefore, the magnitude of GRF are not suitable measures for detecting differences between older adult fallers and non-fallers.

5.3.3 Standing-plantar pressure variables

The results of the independent t-tests with an alpha level of 0.01 (Bonfrroni correction) indicated that across the plantar pressure variables, only the Peak Contact Pressure (PkCP) in the ECNS condition was significantly different between the two groups. Fallers showed a significantly higher PkCP (peak plantar pressure averaged for the entire standing trial) in the ECNS condition, while the MP (average pressure) and PTI (average pressure considering the amount of time the pressure was present) were not significantly different between the two groups. This may be interpreted as the fallers applying more pressure underneath certain areas of their feet while trying to maintain their balance, which results into an increase in the peak plantar

pressure. Previous research has stated that decline in lower limb proprioception (Martínez-Amat et al., 2013), and sensorimotor deficiencies increases the rate of falls in the older adult population (Lord et al., 1996; Sorock & Labiner, 1992). Therefore, it is possible that the fallers apply more pressure underneath certain areas of their feet to compensate for lower limb proprioception, and sensory-motor impairments.

The lack of significant differences in the plantar pressure measures during the EOTS condition between the groups, further confirms this notion. When Meyer, Oddsson, & Luca, (2004), investigated the role of plantar cutaneous sensation in balance control during standing, they found that reduced plantar sensation had no effect on standing when vision was available. Therefore, it is possible that during the EOTS condition, the fallers did not need to increase their PkCP due to the presence of their vision.

Finally, the lack of significance could also be accounted for by the EOTS condition suffering from floor effect (Era. et al., 2006). Further analysis revealed that the participants had significantly higher PkCP for the EOTS condition (0.56 ± 0.18) compared to the ECNS condition (0.43 ± 0.16), $p<0.001$. This may be a testimony to the EOTS' challenging nature, which is manifested in the increase of PkCP by the older adults applying more pressure underneath their feet to maintain their balance. To the best of the author's knowledge, no other study has investigated such plantar pressure variables between older adult fallers and non-fallers while standing; hence, more research is required to confirm these notions.

5.5 Summary

This research compared several biomechanical measures between F and NF during walking and standing trials. The major points of the discussion section are summarized for the balance, walking, and standing tests.

i. Balance

The fallers scored significantly lower in the BBS and Brief-BESTest, which indicates diminished balance compared to the non-fallers. This finding is consistent with previous research investigating the balance of older adult fallers (Arfken et al., 1994; Austin et al., n.d.; Joshi & Sadhale, 2014; Stel et al., 2003; Tinetti et al., 1988).

The BBS did not perform well in categorizing the participants into fallers and non-fallers with respect to their history of fall. More than 60% of the fallers who participated in this study scored above the recommended cut-off score of 45. This is probably due to the ceiling effect that is associated with this test (Lemay & Nadeau, 2010). The items of this test were not challenging enough for the participants, therefore, most participants scored higher than the suggested 45. The presence of ceiling effect in the BBS limits the tests ability to compare among better functioning older adult with respect to their history of fall. Therefore, we do not recommend this test for discriminating fallers among community dwelling older adults.

The Brief-BESTest performed well in categorizing F and NF with respect to their history of fall. The recommended cut-off point of 11 (Duncan et al., 2013) was only able to correctly categorize approximately half of the older adult fallers who participated in this study. However, according to our results, 92.68% of the older adult fallers scored 15 or less on the Brief-BESTest. Perhaps a higher cut-off score for the Brief-BESTest can be more sensitive in discriminating between older adult fallers and non-fallers. Future research can investigate the discriminatory power of this test with respect to falling and provide a more sensitive cut off point. The short time it takes to administer the test and its sensitivity to categorizing fallers and non-fallers makes it an appealing test for discriminating fallers among community dwelling older adults.

ii. Walking

The fallers walked at a significantly lower speed than non-fallers and had a significantly higher stance time, which may be an indication of impairment in their automatic stepping mechanism (Brach et al., 2007); reduced functioning of the visual, vestibular and sensorimotor systems (Lord et al., 1996); decline in attentional resources (Holtzer et al., 2006); deficiencies in the cognitive, muscular, and neurological systems (Van Kan et al., 2009). Additionally, the reduction of speed and increased stance time has also been attributed to fear of falling (Maki, 2015). Perhaps the fallers walk in a more conservative manner and adopt a more cautious gait strategy to avoid falling (Kwon et al., 2018).

The COP measures were significantly different between F and NF in the loading response (LR) sub-phase of gait. During this period, weight is transferred from one leg to another, which makes it a suitable stage for investigating biomechanical differences.

The two groups showed the most significant differences in their GRF measures. For instance, it took fallers a longer time to reach their first V GRF peak and they exhibited lower loading rates. Additionally, the observed reduction in Fz3 and increase in Fz2 in the F group may be interpreted as their reduced ability for push-off; since propulsion force (Fx3) is caused by the foot pushing back against the force plate, the decrease in Fx3, also shows that in comparison to non-fallers, they produce smaller force during push-off. Likewise, the breaking force (Fx1), which prevents the foot from sliding forward, was reduced in fallers compared to non-fallers, which may be interpreted as the fallers, exhibiting less prevention force with respect to sliding forward.

It is important to mention that both COP and GRF measures are speed dependent. Therefore, the observed differences in these measures between the two groups may be attributed to the different walking speed adopted by the fallers and non-fallers. However, according to the results of the ANCOVA, COP measures seemed to be more affected by the participant's walking speed than the GRF measures. This suggests that the partcipant's walking speed may not influence the GRF measures.

Finally, there were no significant differences in the plantar pressure measures between the two groups during gait; this could be due to the presence of foot pain (Mickle et al., 2010), difference in speed (Burnfield et al., 2004) or foot structural differences (Scott, Menz, & Newcombe, 2007).

Lastly, a reduction in V GRF peaks has also been attributed to compensatory responses for managing pain (Katoh, Chao, Morrey, & Laughman, 1983; Lee, Simmonds, Etnyre, & Morris, 2007; Sullivan et al., 2015). Increases in GRF time variables such as Tz1 (the time it takes to reach the first V GRF peak) are thought to be a protective mechanism against overloading the contact area of the foot (Gimunova et al., 2015), which may be a response to avoid pain. Also, Bosch, et al., (2009) suggest that speed reduction is being utilized as a global strategy to reduce pain; thus, it is possible that fallers applied less pressure on their foot to avoid pain.

iii. Standing

The only COP measure that was significantly different between fallers and non-fallers during the standing tests was the $COP_{EllArea}$ in ECNS condition. The observed increase in this measure in F shows that, compared to NF, they had more difficulty in maintaining their balance during the ECNS condition.

There were no significant differences in the magnitude of GRF between the two groups for any of the standing test conditions. In order to observe GRF variations while standing, more challenging positions—such as standing on one leg—are recommended (Goldie et al., 1989), which, is not practical for the older adult population. Therefore, we do not recommend the magnitude of

GRF as suitable measures for finding biomechanical differences between F and NF.

The only plantar pressure measure that was significantly different between the two groups during the standing tests was the Peak Contact Pressure (PkCP) in the ECNS condition. The increase in PkCP in fallers, while all the other plantar pressure variables remained similar to non-fallers, may be an indication that older adult fallers apply more pressure underneath certain areas of their feet to compensate for lower limb proprioception, and sensory-motor impairments. To date, no other study has investigated these plantar pressure variables between F and NF during standing; thus, further research is needed to confirm this notion.

The lack of significance between the two groups for any measure in the EOTS condition was somewhat surprising. This may be due to the floor effect that is associated with the EOTS condition when utilized in the older adult population (Era et al., 2006). The challenging nature of the EOTS test condition makes it difficult to complete by older adults and therefore prevents a proper investigation. Similar to McGrath et al. (2012), we do not recommend the EOTS to be an adequate test for observing biomechanical differences between F and NF.

Finally, we propose that pressure mats are suitable tools for assessing standing balance of older adults in clinical settings. They showed to be useful for investigating postural sway and plantar pressure data during standing tests.

The PkCP and $COP_{EllArea}$[3] were significantly different between fallers and non-fallers. The plantar pressure data provided by this portable equipment can be used alongside other information such as foot pain, lower limb proprioception, and plantar cutaneous sensation to identify those who are at risk of falling in the aging population.

5.6 Limitations

The current study was not without limitations. For instance, we did not collect any data regarding the presence of foot pain, the participants' foot arch and/or their foot morphology, which could have influenced our planar pressure measures. This information could have been useful in shedding more light on the observed differences between the two groups and a more comprehensive interpretation of the pressure data.

Additionally, the pressure data for the sub-regions of the foot were not included in our analysis due to the template provided by the software not being aligned with the foot steps.

Another limitation of this study was the analysis of gait while the participants walked at their own self-selected speed; most researchers adopt this strategy mainly to avoid alerting gait characteristics (Kerrigan et al., 1998), and partially due to the difficulty of imposing a designated speed while

[3] The $COP_{EllArea}$ can be extracted from both force plates and pressure mats. -

walking on the ground (Stoquart, Detrembleur, & Lejeune, 2008). However, walking speed influences several biomechanical gait parameters, such as increasing COP measures (Zdenek Svoboda et al., 2017), GRF measures (Winiarski & Rutkowska-Kucharska, 2009), and plantar pressure measures (Rosenbaum, Hautmann, Gold, & Claes, 1994). Since, the walking speeds adopted by fallers and non-fallers in this study were different, it is possible that their gait analysis showed variations related only to these dissimilar speeds.

Older adults cognitive status has a significant impact on their walking speed; however, in the current study the only data collected regarding the participant's cognitive status was a Mini-Cog test, which provides little information regarding their, verbal IQ, attention, memory, which have shown to be significant predictors of variance in gait velocity (Holtzer,et al., 2006).

When several dependent variables are being assessed a Multi-Level Analysis of Variance (MANOVA) can be used for statistical analysis to avoid type 1 error. However, our data had violated assumptions required for conducting a MANOVA. For instance, the assumptions of multivariate normality and homogeneity of variance-covariance matrices were violated. Additionally, MANOVA works best when dependent variables are only moderately correlated (Pallant, 2007, P.282), which was not the case for several depended variables that were being assessed in this study. Therefore, independent sample t tests with Bonferroni corrections were used for comparing differences between the two groups.

Other limitations of this study include: the potential targeting of the mats, the walkway causing error in the force plate measurements, and not standardizing the foot placements for the standing tests. Also, this study only investigated COP measures, GRF measures, and plantar pressure measures, and stance time -the only temporal-spatial gait parameter- between F and NF. Further investigation of these variables and other gait parameters, such as: joint kinematics, joint kinetic, electromyography, and other temporal-spatial gait parameter could have provided more information regarding differences in gait characteristics of fallers and non-fallers.

Despite these limitations, this study provides useful information regarding the biomechanical differences between F and NF during walking and standing. The findings of our study can be used in future research for discriminating between older adult fallers and non-fallers. Additionally, the findings of our study can be incorporated in predictive models.to identify older adults who are at risk of falling. Consequently, through proper intervention the rate of falls can be reduced in the aging population.

6. Conclusion

Falling in the older adult population is a major concern, which can lead to the loss of independence, depression, and mortality. The goal of this study was to compare different biomechanical measures between F and NF. In order to achieve this goal, data from 102 participants with 65 years of age and older (41 Fallers; 61 Non-Fallers) were collected, using force plates and pressure mats during walking and standing trials. Prior to data collection, the participants' balance was assessed using the BBS and Brief-BESTest; results showed that the Brief-BESTest is more capable in categorizing participants as fallers and non-fallers with respect to their history of fall.

For the walking trials, our findings indicated significant differences between the two groups in the COP measures during the loading response phase of gait, significant differences in the GRF measures, and no significant differences in the plantar pressure measures. The comparison of GRF measures between F and NF, which has been neglected by researchers, provided insightful information regarding their gait patterns. For instance, the reduction in fallers VGRF peaks, the longer time it took them to reach the peaks, and the smaller force produced by them during push-off, seems to suggest the presence of foot pain among fallers. Additionally, the reduction of breaking and propulsion force in the F group seems to suggest that compared to older adult non-fallers, the fallers produce smaller magnitude of the force that prevents them from sliding forward. Not many researchers have investigated the

magnitude of GRF between older adult fallers and non-fallers; therefore, there is a need for more research to confirm these speculations.

Older adult fallers had a significantly lower walking speed and higher stance time compared to non-fallers, which could be an indication of impairments in their muscular and/or neurological systems. Future research can consider assessing the relationship between gait characteristics and different sensorimotor systems between these two groups. Additionally, when the participant's walking speed was added as a covariant, the GRF measures remained significant between the two groups, while the COP measures lost their significance.

For the standing trials, the only significant differences between the two groups were observed in the ECNS condition, for COP measures and plantar pressure measures. Our results supported Melzer et al. (2010) and Thapa et al, (1996), findings regarding the $COP_{EllArea}$ being a more sensitive measure than COP velocity in detecting significant differences between fallers and non-fallers during standing trials. Additionally, our results suggest that the GRF measures during standing and the EOTS condition are not suitable for detecting differences between older adult fallers and non-fallers. Furthermore, the significantly higher PkCP observed in the plantar pressure measures of fallers, seem to suggest lower limb proprioception, and/or sensory-motor deficiencies. To the best of the author's knowledge, no other research has investigated these plantar pressure measures in F and NF during standing trials;

therefore, more research is needed to confirm our finding with respect to plantar pressure measures.

The portable nature of the pressure mats makes them an ideal candidate for assessing standing balance of older adults in clinical settings. They can provide useful information regarding postural sway and plantar pressure data, which can be used for discriminating between older adult fallers and non-fallers. Additionally, our results suggest that standing on pressure mats in the ECNS condition for 30 seconds may be used for comparing balance in older adults. This method can assist in detecting if a specific fall prevention program or intervention has an effect on an individual's balance. The biomechanical variables that were found to be significantly different between older adult fallers and non-fallers during walking and standing can be used by researchers to build predictive models with respect to falling. For instance, during standing trials the COP ellipse area and the peak contact pressure can potentially be a useful variable for improving a predictive models' outcome. Additionally, GRF measures during walking and COP measures during the LR sub-phase of gait can be useful variables for building models that can identify older adult fallers with respect to their history of fall.

Finally, this research sought to provide insight on some well-researched and some neglected biomechanical measures that can be used for discriminating between older adult fallers and non-fallers. The information provided by this research can be used in future studies for building effective predictive models to identify older adults who are at risk of falling;

subsequently, through proper, adequate, and targeted interventions the number of falls in older adults can be reduced.

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Appendix I

Berg Balance Scale

Name: _____ Date: _____

Location: _____ Rater: _____

ITEM DESCRIPTION	SCORE (0-4)
Sitting to standing	_____
Standing unsupported	_____
Sitting unsupported	_____
Standing to sitting	_____
Transfers	_____
Standing with eyes closed	_____
Standing with feet together	_____
Reaching forward with outstretched arm	_____
Retrieving object from floor	_____
Turning to look behind	_____
Turning 360 degrees	_____
Placing alternate foot on stool	_____
Standing with one foot in front	_____
Standing on one foot	_____

Total _____

GENERAL INSTRUCTIONS

Please document each task and/or give instructions as written. When scoring, please record the lowest response category that applies for each item.

In most items, the subject is asked to maintain a given position for a specific time. Progressively more points are deducted if:

- the time or distance requirements are not met
- the subject's performance warrants supervision
- the subject touches an external support or receives assistance from the examiner

Subject should understand that they must maintain their balance while attempting the tasks. The choices of which leg to stand on or how far to reach are left to the subject. Poor judgment will adversely influence the performance and the scoring.

Equipment required for testing is a stopwatch or watch with a second hand, and a ruler or other indicator of 2, 5, and 10 inches. Chairs used during testing should be a reasonable height. Either a step or a stool of average step height may be used for item # 12.

Berg Balance Scale continued…

REACHING FORWARD WITH OUTSTRETCHED ARM WHILE STANDING
INSTRUCTIONS: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. (Examiner places a ruler at the end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the fingers reach while the subject is in the most forward lean position. When possible, ask subject to use both arms when reaching to avoid rotation of the trunk.)
( ) 4 can reach forward confidently 25 cm (10 inches)
( ) 3 can reach forward 12 cm (5 inches)
( ) 2 can reach forward 5 cm (2 inches)
( ) 1 reaches forward but needs supervision
( ) 0 loses balance while trying/requires external support

PICK UP OBJECT FROM THE FLOOR FROM A STANDING POSITION
INSTRUCTIONS: Pick up the shoe/slipper, which is in front of your feet.
( ) 4 able to pick up slipper safely and easily
( ) 3 able to pick up slipper but needs supervision
( ) 2 unable to pick up but reaches 2-5 cm(1-2 inches) from slipper and keeps balance independently
( ) 1 unable to pick up and needs supervision while trying
( ) 0 unable to try/needs assist to keep from losing balance or falling

TURNING TO LOOK BEHIND OVER LEFT AND RIGHT SHOULDERS WHILE STANDING
INSTRUCTIONS: Turn to look directly behind you over toward the left shoulder. Repeat to the right. (Examiner may pick an object to look at directly behind the subject to encourage a better twist turn.)
( ) 4 looks behind from both sides and weight shifts well
( ) 3 looks behind one side only other side shows less weight shift
( ) 2 turns sideways only but maintains balance
( ) 1 needs supervision when turning
( ) 0 needs assist to keep from losing balance or falling

TURN 360 DEGREES
INSTRUCTIONS: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.
( ) 4 able to turn 360 degrees safely in 4 seconds or less
( ) 3 able to turn 360 degrees safely one side only 4 seconds or less
( ) 2 able to turn 360 degrees safely but slowly
( ) 1 needs close supervision or verbal cuing
( ) 0 needs assistance while turning

PLACE ALTERNATE FOOT ON STEP OR STOOL WHILE STANDING UNSUPPORTED
INSTRUCTIONS: Place each foot alternately on the step/stool. Continue until each foot has touched the step/stool four times.
( ) 4 able to stand independently and safely and complete 8 steps in 20 seconds
( ) 3 able to stand independently and complete 8 steps in > 20 seconds
( ) 2 able to complete 4 steps without aid with supervision
( ) 1 able to complete > 2 steps needs minimal assist
( ) 0 needs assistance to keep from falling/unable to try

STANDING UNSUPPORTED ONE FOOT IN FRONT
INSTRUCTIONS: (DEMONSTRATE TO SUBJECT) Place one foot directly in front of the other. If you feel that you cannot place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the subject's normal stride width.)
( ) 4 able to place foot tandem independently and hold 30 seconds
( ) 3 able to place foot ahead independently and hold 30 seconds
( ) 2 able to take small step independently and hold 30 seconds
( ) 1 needs help to step but can hold 15 seconds
( ) 0 loses balance while stepping or standing

STANDING ON ONE LEG
INSTRUCTIONS: Stand on one leg as long as you can without holding on.
( ) 4 able to lift leg independently and hold > 10 seconds
( ) 3 able to lift leg independently and hold 5-10 seconds
( ) 2 able to lift leg independently and hold ≥ 3 seconds
( ) 1 tries to lift leg unable to hold 3 seconds but remains standing independently.
( ) 0 unable to try of needs assist to prevent fall

( ) TOTAL SCORE (Maximum = 56)

Berg Balance Scale continued…

REACHING FORWARD WITH OUTSTRETCHED ARM WHILE STANDING
INSTRUCTIONS: Lift arm to 90 degrees. Stretch out your fingers and reach forward as far as you can. (Examiner places a ruler at the end of fingertips when arm is at 90 degrees. Fingers should not touch the ruler while reaching forward. The recorded measure is the distance forward that the fingers reach while the subject is in the most forward lean position. When possible, ask subject to use both arms when reaching to avoid rotation of the trunk.)
( ) 4 can reach forward confidently 25 cm (10 inches)
( ) 3 can reach forward 12 cm (5 inches)
( ) 2 can reach forward 5 cm (2 inches)
( ) 1 reaches forward but needs supervision
( ) 0 loses balance while trying/requires external support

PICK UP OBJECT FROM THE FLOOR FROM A STANDING POSITION
INSTRUCTIONS: Pick up the shoe/slipper, which is in front of your feet.
( ) 4 able to pick up slipper safely and easily
( ) 3 able to pick up slipper but needs supervision
( ) 2 unable to pick up but reaches 2-5 cm(1-2 inches) from slipper and keeps balance independently
( ) 1 unable to pick up and needs supervision while trying
( ) 0 unable to try/needs assist to keep from losing balance or falling

TURNING TO LOOK BEHIND OVER LEFT AND RIGHT SHOULDERS WHILE STANDING
INSTRUCTIONS: Turn to look directly behind you over toward the left shoulder. Repeat to the right. (Examiner may pick an object to look at directly behind the subject to encourage a better twist turn.)
( ) 4 looks behind from both sides and weight shifts well
( ) 3 looks behind one side only other side shows less weight shift
( ) 2 turns sideways only but maintains balance
( ) 1 needs supervision when turning
( ) 0 needs assist to keep from losing balance or falling

TURN 360 DEGREES
INSTRUCTIONS: Turn completely around in a full circle. Pause. Then turn a full circle in the other direction.
( ) 4 able to turn 360 degrees safely in 4 seconds or less
( ) 3 able to turn 360 degrees safely one side only 4 seconds or less
( ) 2 able to turn 360 degrees safely but slowly
( ) 1 needs close supervision or verbal cuing
( ) 0 needs assistance while turning

PLACE ALTERNATE FOOT ON STEP OR STOOL WHILE STANDING UNSUPPORTED
INSTRUCTIONS: Place each foot alternately on the step/stool. Continue until each foot has touched the step/stool four times.
( ) 4 able to stand independently and safely and complete 8 steps in 20 seconds
( ) 3 able to stand independently and complete 8 steps in > 20 seconds
( ) 2 able to complete 4 steps without aid with supervision
( ) 1 able to complete > 2 steps needs minimal assist
( ) 0 needs assistance to keep from falling/unable to try

STANDING UNSUPPORTED ONE FOOT IN FRONT
INSTRUCTIONS: (DEMONSTRATE TO SUBJECT) Place one foot directly in front of the other. If you feel that you cannot place your foot directly in front, try to step far enough ahead that the heel of your forward foot is ahead of the toes of the other foot. (To score 3 points, the length of the step should exceed the length of the other foot and the width of the stance should approximate the subject's normal stride width.)
( ) 4 able to place foot tandem independently and hold 30 seconds
( ) 3 able to place foot ahead independently and hold 30 seconds
( ) 2 able to take small step independently and hold 30 seconds
( ) 1 needs help to step but can hold 15 seconds
( ) 0 loses balance while stepping or standing

STANDING ON ONE LEG
INSTRUCTIONS: Stand on one leg as long as you can without holding on.
( ) 4 able to lift leg independently and hold > 10 seconds
( ) 3 able to lift leg independently and hold 5-10 seconds
( ) 2 able to lift leg independently and hold ≥ 3 seconds
( ) 1 tries to lift leg unable to hold 3 seconds but remains standing independently.
( ) 0 unable to try of needs assist to prevent fall

( ) TOTAL SCORE (Maximum = 56)

Appendix II

Appendix.
Scoring Form for the Brief Balance Evaluation Systems Test (Brief-BESTest).*

Brief Balance Evaluation Systems Test

Patient/Subject: _____
Examiner: _____
Date: _____

General Note: "instability" is defined as using more than an ankle strategy to maintain balance (eg, a hip strategy is used).

Section I. Biomechanical Constraints			
Item 1: Hip/Trunk Lateral Strength "Rest fingertips in my hands while you lift your leg to the side and hold, keep trunk vertical. You will hold for 10 s." Count 10 s, watch for straight knee, if they use moderate force on your hands, score as "without keeping trunk vertical."	(3) Normal (10 s with trunk vertical) (2) Mild (10 s without trunk vertical) (1) Moderate (1 hip abducts with trunk vertical) (0) Severe (neither hip, 10 s and vertical or not vertical)		
Section II. Stability Limits			
Item 2: Functional Reach Forward "Stand normally; lift both arms straight in front of you; reach as far forward as you can with arms parallel to the ruler without lifting your heels." 2 attempts Observe that patient does not lift heels, rotate trunk, or protract scapula. Watch for vertical initial alignment. Record best reach.	(3) >32 cm (12.5 in) (2) 16.5–32 cm (6.5–12.5 in) (1) <16.5 cm (6.5 in) (0) No measurable lean (or must be caught)	Trial 1 (cm or in) Trial 2 (cm or in)	
Section III. Transitions–Anticipatory Postural Adjustment			
Items 3 and 4: Stand on One Leg–Left and Right "Look ahead; hands must stay on hips; bend one leg behind you; stand on 1 leg as long as you can for up to 30 s. Do not let your lifted leg touch the other leg." Allow 2 attempts, record best attempt; record time up to 30 s (stop time if hands off hips or leg on floor or leg touches supporting leg).	(3) Normal (stable >20 s) (2) Trunk motion OR 10–20 s (1) Stand 2–10 s (0) Unable	Left Seconds Right Seconds	
Section IV. Reactive Postural Response			
Items 5 and 6: Compensatory Stepping–Lateral, Left and Right "Stand with feet nearly together; lean into my hands; I will remove my hands; do whatever necessary to keep balance, trying to take 1 step." Note: Stand next to and behind participant. Place hand on greater trochanter and brace yourself to hold the person's weight shifted to supported leg.	(3) Recovers with 1 side/crossover step (2) Several steps to recover independently (1) Steps but needs assist to prevent fall (0) No step OR falls	Left Right	
Section V. Sensory Orientation			
Item 7: Stance With Eyes Closed, on Foam Surface "Stand on foam with your eyes closed, your hands on your hips, and your feet close but not touching. Start by looking straight ahead, and I will start timing when you close your eyes. Stay as stable as possible and try to keep your eyes closed for the entire time. The goal is 30 s." Two trials, if necessary. Patient must step off foam between trials.	(3) 30 s stable (2) 30 s unstable (1) <30 s (0) Unable	Trial 1 (s) Trial 2 (s)	
Section VI. Stability in Gait			
Item 8: Timed "Up & Go" Test "When I say 'go,' stand up and walk quickly but safely to the tape, turn, and walk back and sit in chair." Start with back against chair, stop timing when buttocks hit the chair; chair should have arms to push from, if necessary. Imbalance might include trips or lateral/backward stumbles or crossovers.	(3) Fast, <11 s, good balance (2) Slow, >11 s, good balance (1) Fast, <11 s, imbalance (0) Slow, >11 s, imbalance	Time (s)	
	TOTAL:		

* The scoring form for the Brief-BESTest examination may not be used or reproduced without written permission of the authors.

Appendix III

As mentioned previously, COP can be extracted from both pressure mats and force plates. Even though both systems use different approaches to calculate the COP the results of the Cop obtained from both devices should be similar. In order to compare the COP obtained from the pressure mats and force plates the cop of both device were calculated during a walking trial. The COP obtained from the pressure mats and the force plates during a gait trial were compared. Figures 8 & 9 show the x and y coordinates obtained from force plate and pressure mat while an individual steps on both devices during a walking trial.

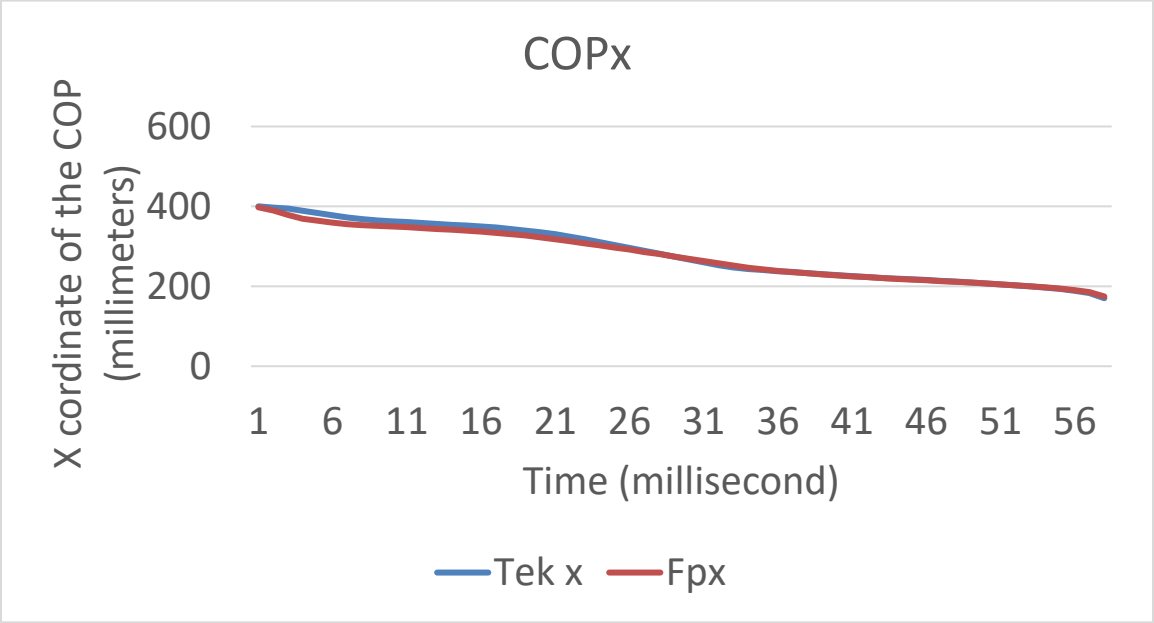

Figure 23: The x coordinate of the COP for the force plate (FPx) and the mat (Tek x).

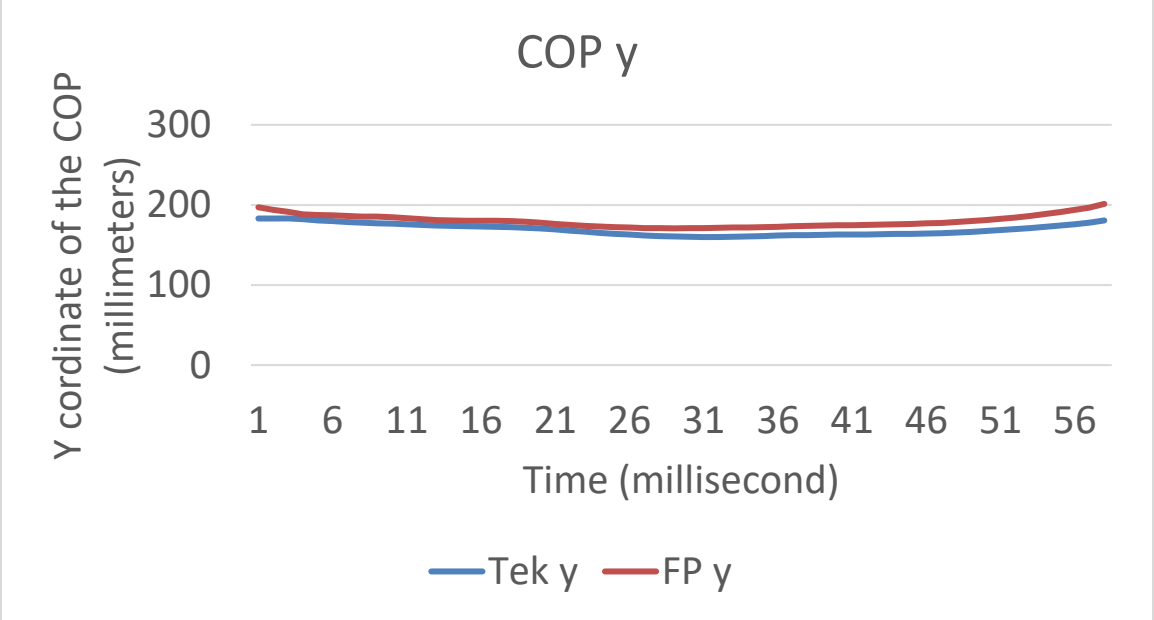

Figure 24: The x coordinate of the COP for the force plate (FPy) and the mat (Tek y).

Appendix IV

A transformation matrix was created on excel to transform the coordinates of the mats to the coordinates of the force plates. In order to create the transformation matrix of the pressure mats and the force plates a metal bar with an area of 0.5 cm^2 that covered two sensels (each sensel area is 0.25 cm^2) was pressed on the mat on three points while pressure mats were mounted on top of the force plates. In order to apply the pressure on three points that are 90 degrees apart a template was used. As it can be seen from figures below the angles between the three points were roughly 90 degrees:

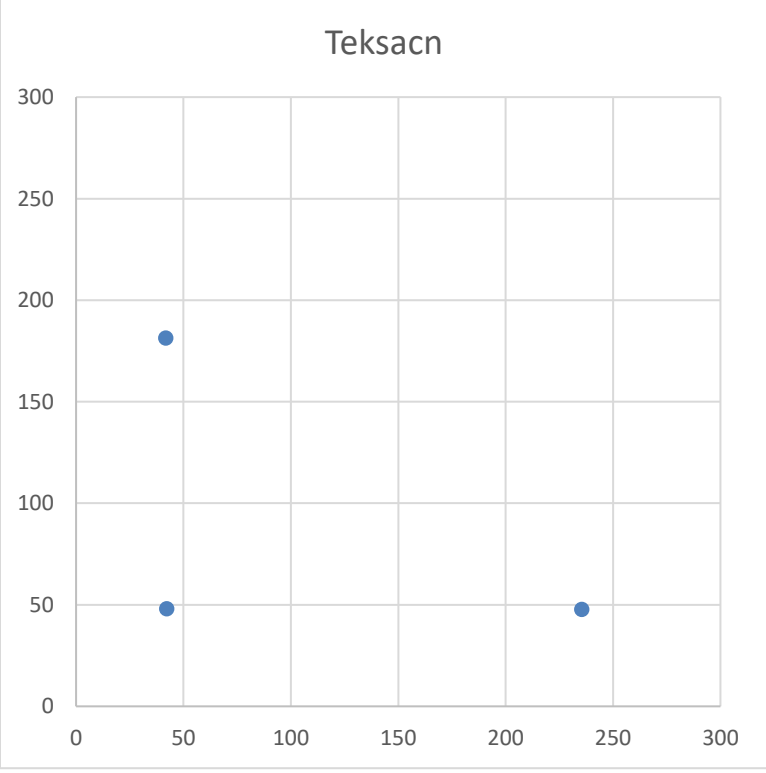

Figure 25: The three points of O, A, and B demonstrated by Tekscan.

Vectors OA and OB were created using these three points: O (42.208, 48.11) A (41.73, 181.31) B (235.47, 47.82). The angle between OA and OB was -90.29 degrees.

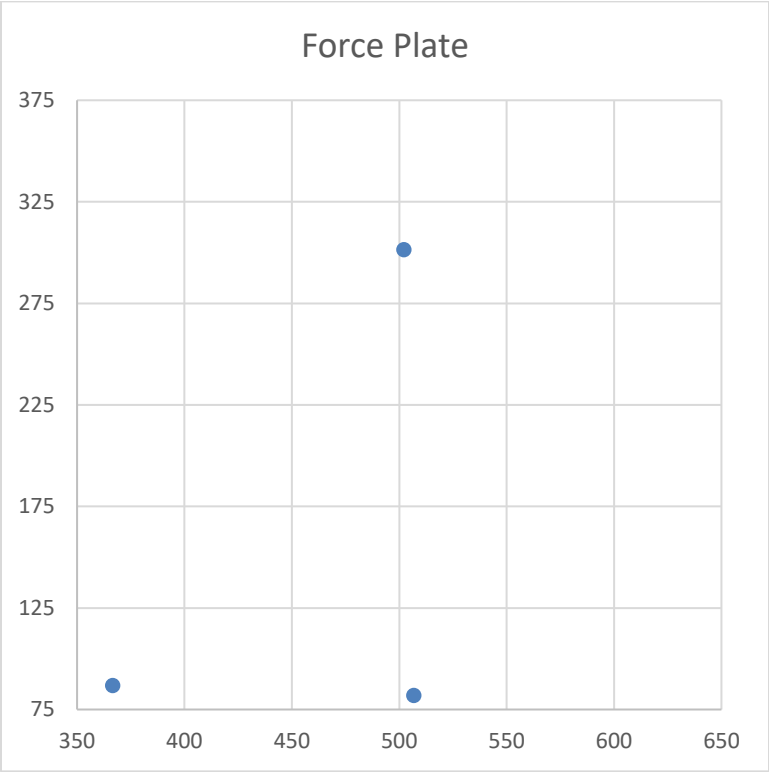

Figure 26: The three points of O, A, and B demonstrated by Force plate.

Vectors OA and OB were created using these three points: O (506.72, 81.88), A (366.64, 86.76), B (502.17, 301.40). The angle between OA and OB was 86.81 degrees.

The angles between vectors OA from Tekscan and vector OA from force plate and the angle between vector OB from force plate and vector OB from Tekscan should be similar values. The angle between vectors OA from Tekscan and vector OA from force plate was 87.80 degrees and the angle the angle between vector OB

from force plate and vector OB from Tekscan was 91.27 degrees. These values are not exactly the same but they were close enough to continue with creating a transformation matrix.

The following transformation matrix was calculated on an excel sheet. The average of the two angles were used. The matrix has been successful in transforming the coordinates of the Tekscan to the coordinates of the force plate.

$$\begin{bmatrix} F_x \\ F_y \end{bmatrix} = \begin{bmatrix} cos & -sin \\ sin & cos \end{bmatrix} \begin{bmatrix} Tx - T_{x0} \\ Ty - T_{y0} \end{bmatrix} + \begin{bmatrix} Fx_0 \\ Fy_0 \end{bmatrix}$$

Appendix V

To assess the effect of the walkway on the forces recorded from the force plates, an experiment with 10 stationary objects with different weights was conducted. The stationary objects weights were as following: 11.33 kg, 22.6kg, 34.01 kg, 45.37 kg, 56.69kg, 68.03kg, 79.37 kg, 90.71 kg, 102.5 kg, and 113.39 kg. The weights were placed on top of the force plates in two conditions. In the "force plate-alone" condition the weights were mounted on top of the two force plates (force plate 2 & force plate 3) that will be used for recording in the research project. After the weights were placed on top of each force plate, three trials with the duration of 10 seconds were recorded. In the "force plate–walkway" condition, the walkway was mounted on top of the force plates and the same procedure was repeated for each weight.

The mean and standard deviation of the 3 trials were calculated for all the forces obtained from both force plates during each condition. Table 1 and 2 show the average forces obtained from force plates 2 and 3 in both the "force plate-alone" condition and "force plate- walkway" condition. As it can be seen from the tables, when the walkway is mounted on top of the faceplates the recorded forces from the plates are mainly overestimated by the force plates. The only exception is with the 11.28 kg, and 22.6 kg weights; which mounting the walkway results in an underestimation of the force.

To calculate the error the differences between the two conditions was calculated. The average of the forces of the 3 trials for the "force plate-alone" condition was subtracted from average of the forces of the 3 trials for the "force plate –walkway" condition. Then, the absolute error percentage of both conditions were calculated by dividing the calculated error from the average force obtained from the "force plate-alone" condition. Table 3 and figure 1 &2 show the absolute error percentage for each weight for force plates 2 and 3 respectively. As it can be seen from table 1 the average of the absolute error between the two conditions were ± 0.76% for force plate 2 and ±0.61% for force plate 3.

Table 21: The Average and standard deviation of the force obtained from force plate 2 during the "force plate-alone" condition and the "force plate –walkway" condition.

Weights (Kg)	Avg F_z for "force plate - alone" (N)	Avg F_z for "force plate - walkway" (N)
11.33 kg	-116.593 ± 0.593	-114.474 ± 0.596
22.6 kg	-230.486 ± 0.597	-229.027 ± 0.592
34.01 kg	-345.108 ± 0.594	-346.764 ± 0.599
45.37 kg	-457.351 ± 0.593	-461.287 ± 0.605
56.69 kg	-571.024 ± 0.592	-577.688 ±0.600
68.03 kg	-685.709 ± 0.598	-689.727 ±0.609
79.37 kg	-800.529 ± 0.595	-806.283 ± 0.595
90.71 kg	-913.166 ± 0.593	-919.127 ± 0.600
102.5 kg	-1028.62 ±0.593	-1031.94 ± 0.598
113.39 kg	-1141.76 ± 0.596	-1145.63 ± 0.60

Table 22: The Average and standard deviation of the force obtained from force plate 3 during the "force plate-alone" condition and the "force plate –walkway" condition.

Weights (Kg)	Avg F_z for "force plate -alone" (N)	Avg F_z for "force plate -walkway" (N)
11.33 kg	-118.58 ± 0.514	-114.462 ± 0.518

22.6 kg	-231.59 ± 0.520	-230.23 ± 0.527
34.01 kg	-346.88 ± 0.529	-347.4 ±0.525
45.37 kg	-458.27 ± 0.536	-461.34 ± 0.541
56.69 kg	-573.4 ± 0.518	-576.64 ± 0.525
68.03 kg	-688.01 ± 0.527	-691.16 ± 0.527
79.37 kg	-804.04 ± 0.518	-804.99 ± 0.523
90.71 kg	-917.57 ± 0.528	-919.04 ± 0.516
102.5 kg	-1032.2 ± 0.523	-1032.9 ± 0.526
113.39 kg	-1145.4 ± 0.520	-1145.4 ± 0.521

Table 5: The absolute error between the two conditions for force plates 2 and 3

Weights (Kg)	Force plate 2 absolute error (N)	Force plate 3 absolute error (N)
11.33	1.82%	3.34%
22.6	0.63%	0.59%
34.01	0.48%	0.15%
45.37	0.86%	0.67%
56.69	1.17%	0.56%
68.03	0.59%	0.46%
79.37	0.72%	0.12%
90.71	0.65%	0.16%
102.5	0.32%	0.07%
113.39	0.34%	0.01%
Avg error	0.76%	0.61%

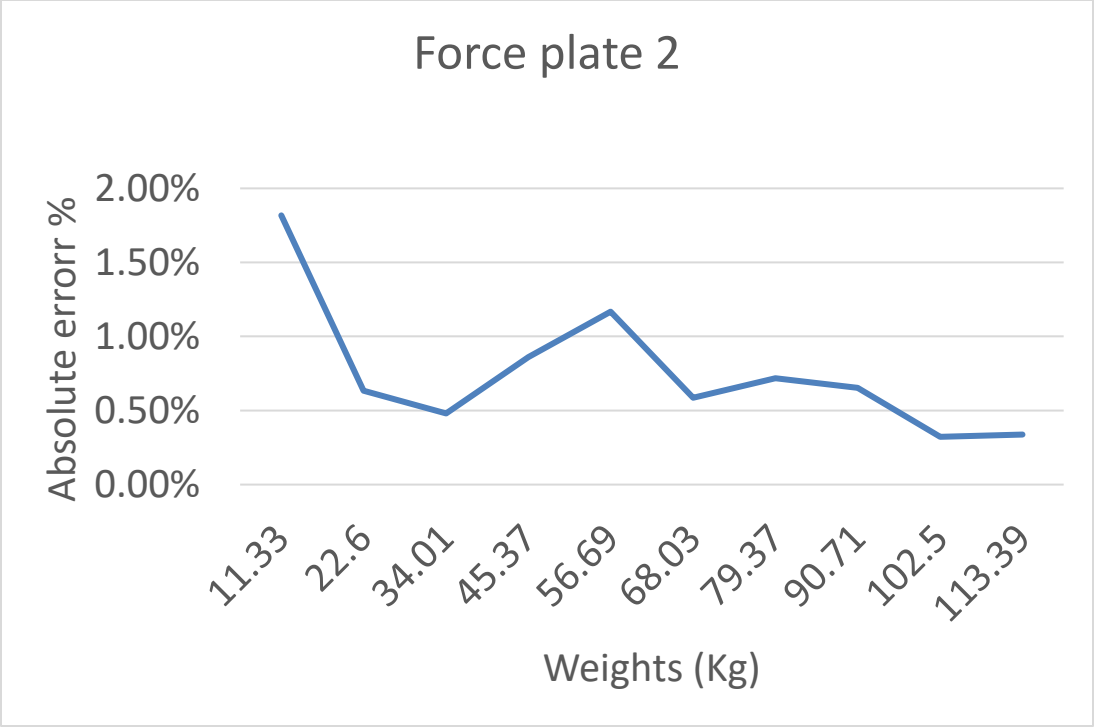

Figure 27:The absolute error percentage between the two conditions for force plate 2.

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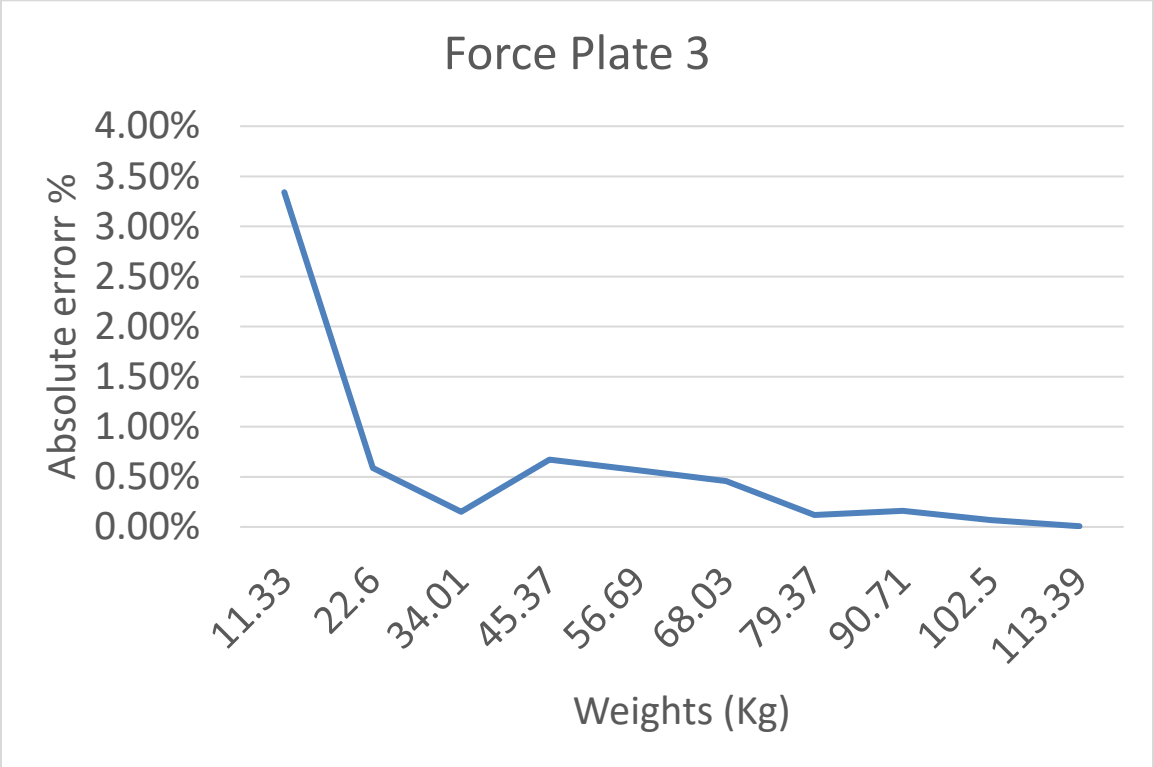

Figure 28:The absolute error percentage between the two conditions for force plate 3

Appendix VI[4]

Input: one sigma uncertainty in the x-direction: σx

one-sigma uncertainty in the y-direction: σy

correlation coefficient ρxy

center of the ellipse: (x,y)

aspect ratio of the plot: (Δy/vertical plot length)/(Δx/horizontal plot length)

1) Calculate the covariance between x and y:

$$\sigma xy = \rho xy \cdot \sigma x \cdot \sigma y$$

2) Construct the covariance matrix:

$$covmat = \begin{pmatrix} (\sigma x)^2 & \sigma xy \\ \sigma xy & (\sigma y)^2 \end{pmatrix}$$

3) Calculate the lengths of the ellipse axes, which are the square root of the eigenvalues of the covariance matrix:

$$eigval = eigenvalues(c)$$

4) Calculate the counter-clockwise rotation (θ) of the ellipse:

$$\theta = \frac{1}{2} \cdot \mathrm{Tan}^{-1}\left[\left(\frac{1}{aspectratio}\right) \cdot \left(\frac{2 \cdot \sigma xy}{(\sigma x)^2 - (\sigma y)^2}\right)\right]$$

5) To create a 95% confidence ellipse from the 1σ error ellipse, we must enlarge it by a factor of $scalefactor = 2.4477$.

6) Plot the ellipse:

7a) Plot an ellipse with semi-major and semi-monor axes parallel to the x- and y-axes of the graph, centered at (x,y). These axis lengths are the square roots of the eigenvalues. The larger eigenvalue belongs to the axis with the larger uncertainty:

If σx > σy semi-major axis length parallel to x = sqrt(maximum(eigenvalues))*scalefactor

semi-minor axis length parallel to y = sqrt(minimum(eigenvalues))*scalefactor

If σy > σx semi-minor axis length parallel to x = sqrt(minimum(eigenvalues))*scalefactor

semi-major axis length parallel to y = sqrt(maximum(eigenvalues))*scalefactor

7b) Rotate the ellipse θ radians counter-clockwise from its original orientation.

In order to validate the Matlab code that was utilized for calculating the 95% confidence ellipse area the following calculations were performed:

First values of the following points were plugged into the MATLB code V1=[(-1, -1), (1,1), (1,1), (-1, -1)] and V2=[(-1, -1), (0,0), (1, 1)] . As expected the calculated 95% ellipse area surrounding these points were equal to zero in both cases

Afterwards, multiple points with standard deviation of zero (StDev=0) and covariant (Cov=0) of zero were created. As expected the 95% ellipse area was shaped as a circle. This is due to the fact that a StDev and Cov of zero results into the two radius of the ellipse being equal to zero, which creates a circle. Figure 7 shows the randomly created points and the ellipse (in this case circle) that contained 95% of the data.

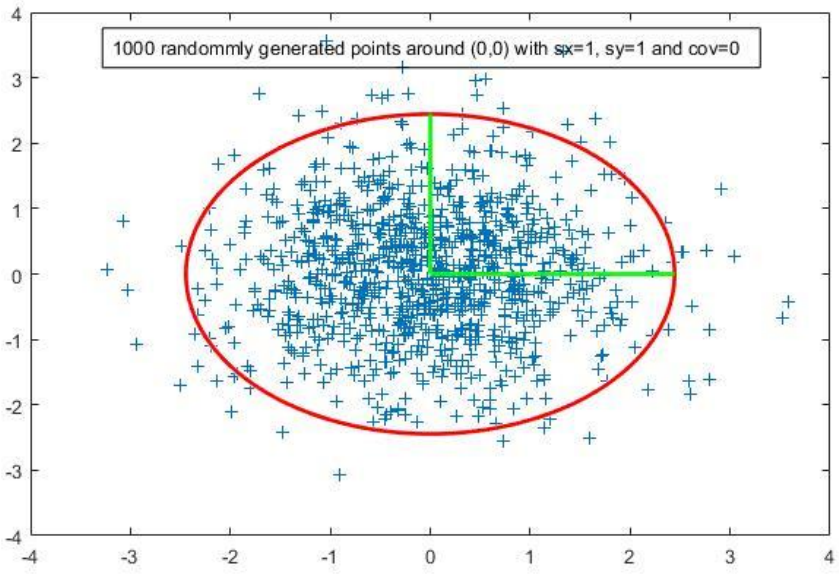

Figure 29: 95% confidence ellipse area surrounding 1000 randomly created points with StDev and Cov of zero.

Appendix VII

The sample entropy of the ΔΦ was calculated in MATLB. The code for calculating the sample entropy was taken from an online source (https://physionet.org/physiotools/sampen/). The code was assessed by using the sample entropy definition provided by Richman & Moorman (2000). According to the authors "SampEnt is precisely the negative natural logarithm of the conditional probability that two sequences similar for "m" points remain similar at the next point, where self-matches are not included in calculating the probability" (Richman & Moorman, 2000, p. H2039). In order to validate the MATLAB code the following example hand calculation was made:

For the following series (S) if we set the window to m=1 the probability of having a matched[5] value will be $P = \frac{6}{8}$ (there were 8 times that the windows matched and among those 8 times there were 6 times that that values after the windows were the same)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S=1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S=1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

[5] According to the MATLAB code if the distance between the vectors are less than 0.2 than it would be considered a match. However, in this context since we are using integers this means an exact match.

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

The sample Entropy of this times series with a window of m=2 is calculated using the following equation:

$$SampEnt = \frac{-\log(\frac{6}{8})}{\log(e)} = 0.28$$

In the same series (S) if we set the window to m=2 the probability of having matched value will be $P = \frac{4}{6}$ (there were 6 times that the windows matched and among those 6 times there were 4 times that that values after the windows were the same)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S=1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

The sample Entropy of this times series with a window of m=2 is calculated using the following equation:

$$SampEnt = \frac{-\log(\frac{4}{6})}{\log(e)} = 0.40$$

Now if we set the window to m=3 the probability of having matched value will be $P = \frac{2}{4}$ (there were 4 times that the windows matched and among those 4 times there were 2 times that that values after the windows were the same)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S=1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match and the following numbers match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

The sample Entropy of this times series with a window of m=2 is calculated using the following equation:

$$SampEnt = \frac{-\log(\frac{2}{4})}{\log(e)} = 0.69$$

Finally if we set the window to m=4 the probability of having the matched value will be $P = \frac{0}{2}$ (there were 2 times that the windows matched and among those 4 times there were 0 times that that values after the windows were the same)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match but the following numbers do not match)

S=1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (windows match but the following numbers do not match)

S= 1, 2, 3, 4, 1, 2, 3, 4, 5 (Windows do not match)

The sample Entropy of this times series with a window of m=2 is calculated using the following equation:

$$SampEnt = \frac{-\log(\frac{0}{2})}{\log(e)} = not\ defined$$

We can also consider a window for m=0 which in this case we will end up matching each number with every other number except for itself. For the aforementioned series the probibilaty will be $P = \frac{8}{72}$ which will give us the following Sample Entropy

$$SampEnt = \frac{-\log(\frac{8}{72})}{\log(e)} = 2.19$$

The S series was plugged in the Matlab code and the results for each different window were as the following:

For m=0, SampEnt =2.1972

For m=1, SampEnt = 0.2877

For m=2, SampEnt =0.4055

For m=3, SampEnt = 0.6931

For m=4, SampEnt =Inf

Finally, based on previous work done regarding sample entropy of ΔΦ the window of five (m=5) was chosen for this study (Rhea et al. 2014).

Furthermore, Richman & Moorman (2000) state that "a lower value of SampEn also indicates more self-similarity in the time series" (Richman & Moorman, 2000, p. H2039). To test this statement three data sets (modulus, sine, and line) were made or further evaluation of the MATLAB code. Each data set contained 1000 values that were repeated 8 times and randomly created noise multiplied by a factor (from 0 to 2 with 0.1 intervals). Since a smaller value of SampEnt indicates more self-similarity (or less noise) in data set, it is expected to see higher values of SampEnt as the noise increases. The following graphs indicate the average sample entropy for different noise/signal ratio (modulus, sine, and line).

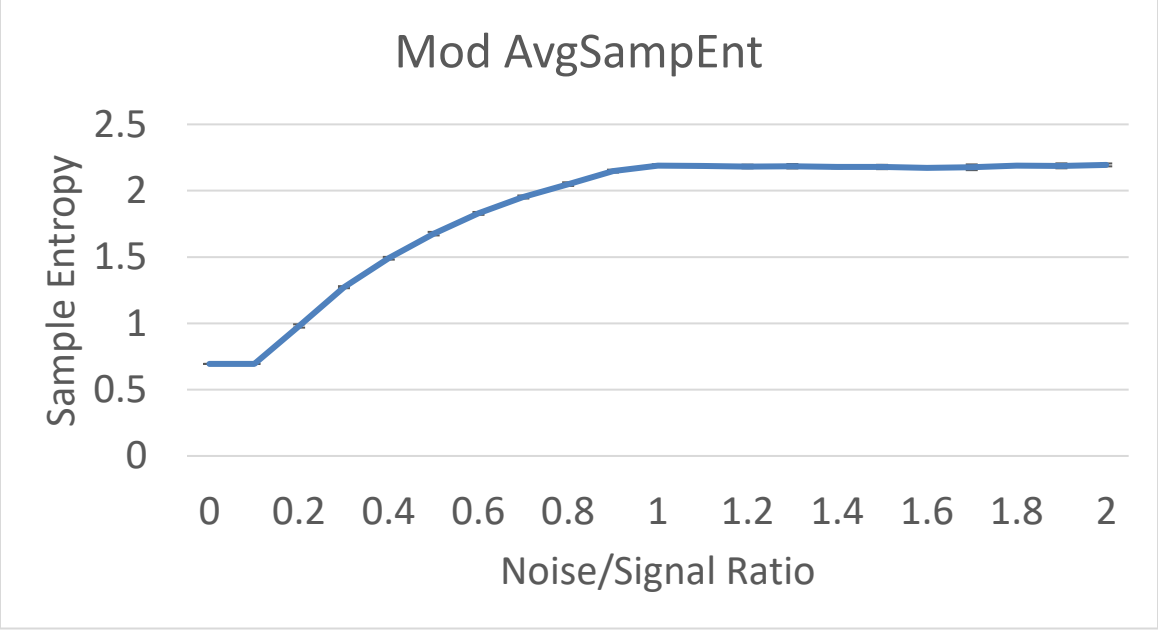

Figure 30: The sample entropy and noise/signal ratio for the modulus data set.

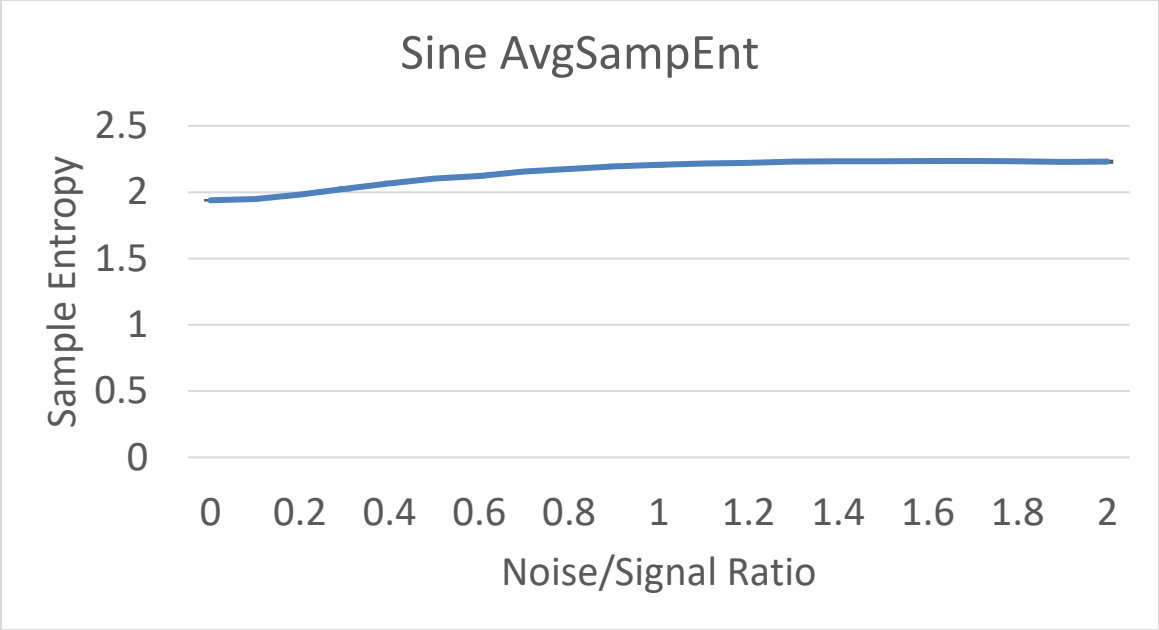

Figure 31: The sample entropy and noise/signal ratio for the Sine data set.

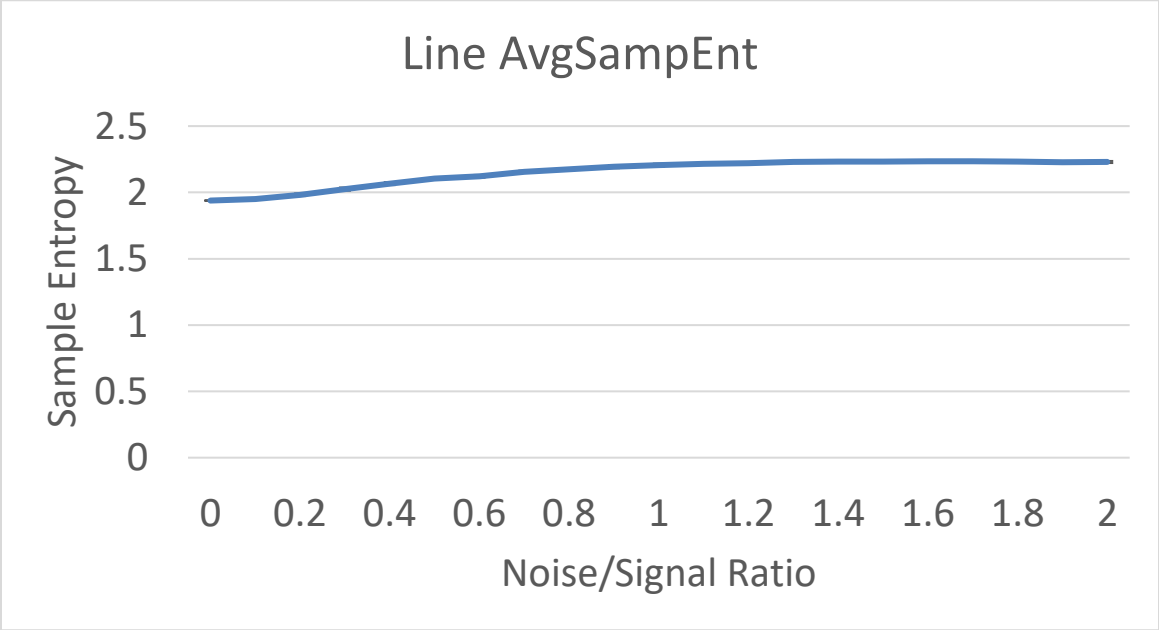

Figure 32: The sample entropy and noise/signal ratio for the line data set.

As it can be seen from the graphs (more obvious in the Modulus graph, fig14) when the noise reaches a certain threshold (>1) the sample entropy stays at a certain high level. This shows that the noise is so high that the data does not show any similarity anymore.

As for the line data set and graph (fig 10), as expected the sample entropy stays at a certain high level showing that regardless of the amount of noise there are no similarities (or repetition) between the values that form a line.

Curriculum Vitae

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Publications

None

Conference Presentations

S. Oladi and P. Patros, “Towards Biomechanical Gait Analytics on the Cloud with ARVIN II”, 27th Annual International Conference on Computer Science and Software Engineering (CASCON 2017).

S. Oladi and P. Patros, “ARVIN II: A pressure Analysis Tool”, 14th Annual Research Expo of the UNB Faculty of Computer Science, Fredericton, Canada Apr 7, 2017.

R. Oladi, M. Yassin, A. Ahmadi, and J. Sarrafzadeh, “The effect of Laser & Hydrocortisone Phonophoresis on Some clinical parameters of Trigger points”, First Annual Kinesiology Research day of the UNB Faculty of Kinesiology, Fredericton, Canada March 26th, 2015.