A LOOK AT AGING:
BALANCE ABILITY AND FALL PREVENTION INTERVENTIONS

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ABSTRACT

The main objective of this work is to address the growing concern of balance loss and falls in the aging population. The initial aspect looks at balance control in a dynamic environment. Observation of age and gender influence on motor control will be made related to a new dynamic balance testing platform (DBTP). The topic of focus in the second portion relates to reaction time in an unstable environment. Research has found that balance improves when physical activity is a part of daily life for seniors. Physical activity influence on reaction time will be investigated with a new approach to exercise classes for seniors. Finally, an understanding of motor control and balance may be acquired and physical activity incorporated into the life of an elderly individual, however this will never fully prevent falls from taking place. A novel approach to injury prevention due to falls is explored in the final portion of this thesis.

<u>Study One</u>

Using a newly constructed dynamic balance testing platform (DBTP), balance ability of three age groups was observed in two visual conditions and in relation to gender. Center of Gravity excursion (COGex) was observed to determine the differences between age groups and gender. Platform response patterns were also observed to asses the functionality of the DBTP as a new tool for balance testing. Three things were found: 1) Age differences related to platform movement suggested that balance decreased with age in both visual conditions. 2) Gender differences between COGex found that males covered the most distance in both visual conditions when compared to females. 3) Gender differences between platform characteristics showed that females balanced longer and had lower platform

movement rate than males, in both visual conditions. In order to consider the DBTP as a new tool for determination of balance ability, more refined tests are necessary.

Study Two

Using pre- and post-training tests, the effects of a Fitball® exercise program on performance in eight subjects was documented. The exercise program focused on improving dynamic balance and postural stability of seniors. To evaluate progress-related changes, pre- and post-tests in a dynamic environment were applied. Center of gravity (COG) excursion, catch success rate, and balance success rate were quantified, and synchronized data collection of 3D motion capture (VICON v8i) and ground reaction force (2 KISTLER platforms) was analyzed. During pre- and post-tests, participants stood in a walk-like stance and were asked to catch a weighted ball, which dropped unexpectedly. Results showed no significant changes in balance success rate. Significant improvements were found, however, in both COG control and catch success rate following training ($p<0.05$).

Study Three

Falls in the elderly are inevitable so it is necessary to take precautions. This study looks at falls in relation to velocity characteristics of various locations on the trunk, and contrasts them to activities of daily living (ADL) in 13 individuals. A threshold level was established to be 2.0m/s, a value that exceeded all maximum resultant velocities for ADL, but was superseded by all fall activity resultant velocities. This suggests that a life vest, which responds similar to a vehicle airbag, may be created and worn that will deploy past a threshold of 2.0m/s with the incidence of a fall.

ACKNOWLEDGMENTS

It has been my pleasure and an honour to use the resources and facilities provided by the University of Lethbridge and staff in completing my research towards a degree of Master in Kinesiology- Biomechanics. Equipment that boasts state-of-the-art status with accompanying software has facilitated ease in data collection, analysis and interpretation. Without the equipment, staff and the financial support afforded me, my voyage to further education and contribution to society would not have been easy to pursue.

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Finally, my deepest appreciation is expressed towards my husband, James. His encouragement for me to become involved in such a challenge as this, as well as his support and understanding throughout the past few years, was a great act of love and sacrifice that will ever be cherished. Thank you my love.

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LIST OF ABBREVIATIONS

ADL – Activities of Daily Living

ANOVA – Analysis of Variance

AP – Anterioposterior

BOS – Base of Support

BRD – Balance Response Data

CGRF – Center of Ground Reaction Force

CNS – Central Nervous System

COG – Centre of Gravity

COGex – Centre of Gravity Excursion

COM – Centre of Mass

COP – Centre of Pressure

DBTP – Dynamic Balance Testing Platform

DurB – Balance Duration

EC – Eyes Closed

EO – Eyes Open

ED – External Dynamics

L – Left

ML – Mediolateral

P-F – Push-Fall

PM – Platform Motion

PSI – Pounds/Square Inch

R – Right

ROM – Range of Motion

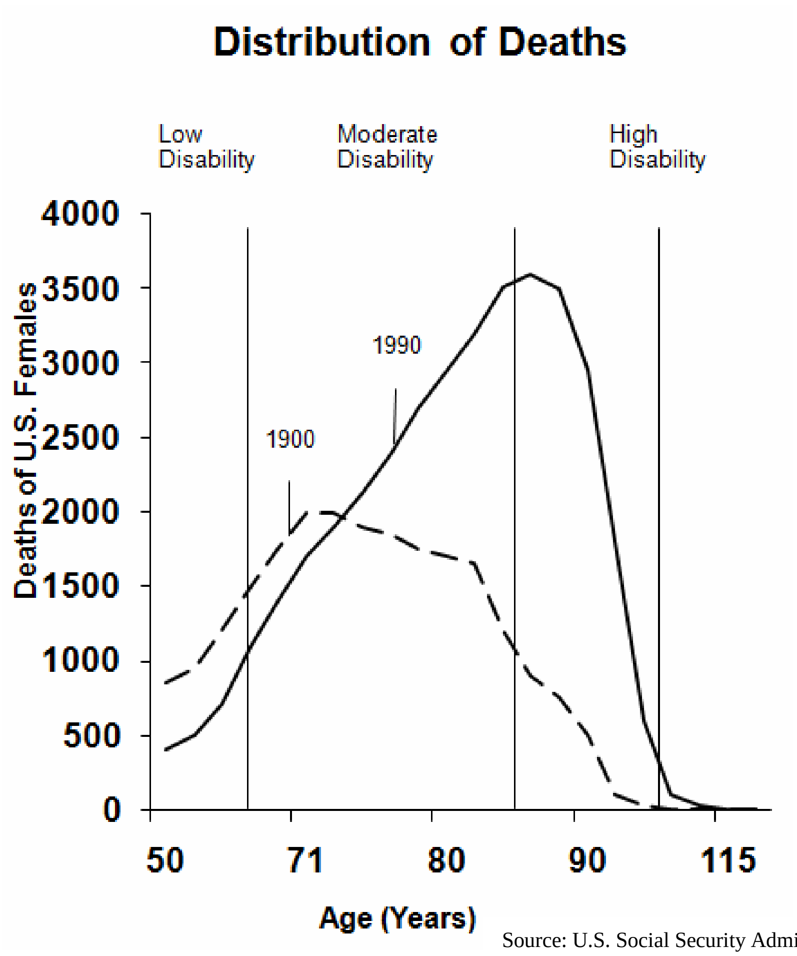
StDev – Standard Deviation

T-F – Trip-Fall

V – Vertical

GENERAL INTRODUCTION

As the medical community progresses and the standard of living among individuals increase, the likelihood of persons living to the age of old-old (≥85 for women, ≥80 for men) will also be on the rise (Chappell & Havens, 1980). With the increased life expectancy comes a greater population of adults over the age of 65 and a high probability of increased disability in older life. Figure 1.1 shows that as life expectancy increases, the amount of individuals with disability also escalates (Gerontology Lectures).



Source: U.S. Social Security Administration

Figure 1.1. Distribution of deaths in females and their level of disability -comparing 1900 to 1990.

Disabilities are often a result of serious falls in the elderly. Due to sensori-motor degradation related to age, ⅓ of people 65 years and older experience at least one fall each year (Edelberg, 2001). Experiences in the earlier years of life tend to compound, and by the age of 65 our overall health and way of life are influenced by these changes.

Balance and postural control, or body sway, is one area that is affected by the previous years of life (Christou, Moss, Boule, Yoon, Evans & Rosengren, 2000) and the general influence of aging. Loss of balance resulting in a fall can be damaging to the physical ability as well as the self-confidence of the older adult. Loss of stability in an upright stance can have a ripple effect in initially influencing the individual experiencing the loss as well as family, friends and society.

This severe problem has a significant impact on personal health and health-costs in older adults. As the leading edge of the baby-boom generation crosses into its fifties, falls, and the increased health care costs associated with this problem, will become a major problem in our society.

Postural stability is often taken for granted because of the natural ability of remaining in equilibrium or an upright stance, but as with anything "we miss what we don't have". Many systems work together to produce stability so the deterioration of one physical system such as vision may likely start a domino effect causing stability to teeter or fall. Decay of physical abilities hinders older adults in performance of daily activities. There is a need to prevent falls and resulting injury in order to add to the quality of life of the aged, and decrease the expenditures on preventable disabilities.

1. Ability of Postural Control / Characteristics of Balance

Postural control was defined by Maki and McIlroy (1996) as the Central Nervous System's (CNS) ability to keep the center of mass (COM) positioned properly over the base of support (BOS) by generating patterns of muscle activity for required regulation of movement. It was also stated that in general, sensory information about body-orientation is also required for postural control. Limb joints act as pivots or links to the system and create an inverse pendulum with a small BOS and dynamic upper body motion.

A feedback loop is the means of maintaining stability in this unstable stance. This feedback loop (Figure 1.2), as suggested by Maki et al. (1996) and Downton (1993), requires aspects of body function to run smoothly and remain in check. Redundancy of systems allows one to remain relatively balanced if one system's input is reduced or removed. Ageing brings about a reduction of sensitivity in aspects of input and responses leading to a decrease in postural control (Downton, 1993). Interruption in one of these three systems and a perturbation occurrence could easily result in a fall.

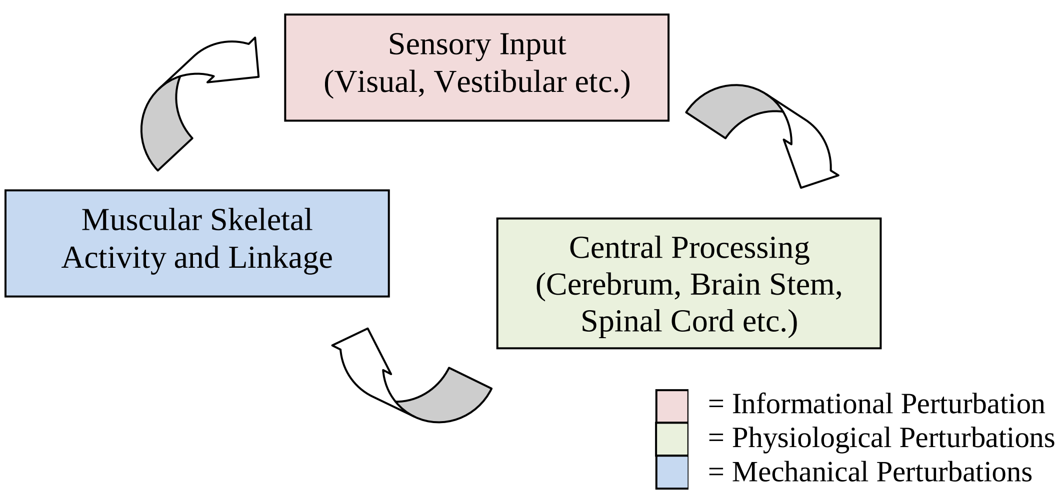


Figure 1.2. Feedback loop for postural control and corresponding perturbation classifications.

2. Epidemiology of Falls

Many definitions exist in terms of what a fall is. Some classify it as 'the location of COM in relation to BOS with no correction' (Isaacs, 1985) with others suggesting it is 'unintentionally coming to rest on the ground/floor/lower level' (Ory, Schechtman & Miller, 1993). These are just a couple examples of the definition of a fall, but in general two conditions must exist for a fall to occur (Maki et al, 1996). First, perturbation acting on the individual must take place. And second, failure of the posture control system to compensate for perturbation. At times falls may be an intrinsic disturbance, but more often are due to external disturbances. These external perturbations are classified as either mechanical or informational.

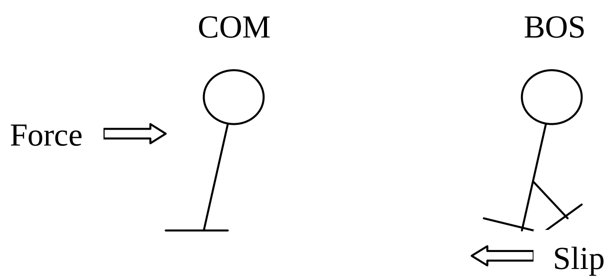


Figure 1.3. Two types of mechanical perturbations:
Center of Mass (COM) and Base of Support (BOS).

Mechanical perturbation can take place in two forms. A disturbance at the BOS such as a trip or slip prevents the COM from remaining in alignment and therefore results in a fall. A second disturbance or COM perturbation takes place in situations such as a push or other upper extremity movement (Figure 1.3). These two type of perturbations accounted for 86% of falls in a study done by Topper and colleagues (1993). Oft times these perturbations take place in dynamic circumstances. At least 77% of 314 falls in a residential care setting took place in situations other than a static position (Jensen, Lundin-Olsson, Nyberg & Gustafson, 2002).

The final cause of falls not obviously related to mechanical perturbation was classified as intrinsic factors or incongruence of sensory information to reality. An example of this may be a feeling of swaying left when in actuality you are swaying right. This can be influenced by many factors such as vision inconsistency or lack of proprioception (Petrella, Lattanzio & Nelson, 1997).

3. Factors Influencing Falls

3.1 Age

With changes in body systems due to advancing years it is natural to assume that postural control will be influenced. This assumption was confirmed by numerous researchers, concluding that the elderly are less stable than young or middle adults (Hellebrandt & Braun, 1939; Sheldon, 1963; Hasselkus & Shambes, 1975; Shan and Wilde, 2003). The changes that take place so as to decrease postural stability will be labeled the ‘age-effect’.

3.2 Gender

In general it has been found that women sway and lose their balance more than men (Wolfson, Whipple, Derby, Amerman & Nashner, 1994) resulting in more falls than men (Alexander, Rivara & Wolf, 1992). A reason for this was offered by Downton (1993) when she stated that the tendency of women to have a narrower walking and standing base than men, due to the configuration of the pelvis, was an aspect for consideration.

Though many studies find that women have poorer postural control, other studies have found that gender did not affect sway (Colledge, Cantley, Peaston, Brash, Lewis, & Wilson, 1994; Bryant, Trew, Bruce, Kuisma & Smith, 2005; Era, Schroll & Ytting, 1996). This may have been the result of differing test procedures or manner of analysis. Posturography tests were done by two conflicting studies, however Wolfson and colleagues (1994) employed a dynamic balance test procedure as opposed to the static approach taken by Colledge et al (1994). When gender differences were noted, but normalized to height (Bryant et al, 2005) or was used as a covariate (Era, 1996), the differences disappeared.

3.3 Musculature

In 1989 Rosenberg coined the term sarcopenia to denote the decline of muscle mass and strength common with healthy ageing. A study designed by Iannuzzi-Sucich and colleagues (2002) found that sarcopenia was prevalent in 22.6% of 195 females and 26.8% of 142 males ages 64-93. A sub-group of adults aged 80 and older revealed that 31.0% and 52.9%, of females and males respectively, experienced this phenomenon suggesting that this is common in adults over age 65 and increases with age. A loss of muscle mass and strength is observable in physical function ability and therefore is influential in postural control and balance ability (de Bruin, 2007).

3.4 Visual System

It is a known fact that vision influences balance ability. Previous research has looked at vision to determine the role it plays on balance ability and fall prevention. Bergland and Wyller (2004) found that vision impairment was a significant predictor for indoor falls. However, regular physical activity, even if it was begun late in life, improves the use of vestibular and somatosensory inputs thus decreasing the degree of reliance placed on vision (Buatois, Gauchard, Aubry, Benetos & Perrin, 2007).

4. Overview

In order to address such a broad topic as aging and its influence on balance and falls, this thesis has been divided into three studies:

Study 1 addressed the influence of age and gender in relation to postural control in a dynamic environment. As noted, various changes take place in the body. Few studies, if any, have attempted to use a dynamic platform to quantify and characterize the age effect across the lifeline and observe visual, age and gender characteristics in an attempt to creating a balance analysis tool.

Study 2 was focused towards the improvement of balance. Musculature or fat-free mass declines with age and past studies have shown that balance improves when physical activity is a part of daily living in the older adult. This study looked at the influence of a specific activity type aimed at improving core stability to improve balance and reaction time in a dynamic environment.

Study 3 addressed the inevitability of fall occurrence in older adults. Steps can be taken to advance the tools used for fall prediction or improve balance ability individually, but falls will always occur. Characteristics of falls were observed in order to lessen the damage of impact.

The combination of these studies address:

1) Characteristics of the age-effect in balance,
2) Gender and visual influence on postural control,
3) Improvement of balance ability in dynamic situations by way of core-strengthening,
4) Characteristics of falls to prevent injury from impact of the lower body upon surface contact resulting from a fall.

With a rate of one in every three individuals experiencing a fall each year after reaching age 65 (Dargent-Molina & Bréart, 1995) it is important to understand why falls occur and how to prevent them. It was hoped that results of these three individual studies will aide in improving knowledge and practices relating to aging, postural control and injury prevention.

STUDY ONE

Influence of Age and Gender in Relation to Postural Control in a Dynamic Environment

1. Introduction

Balance is the outcome of various systems working in sync and contributing to produce results of an upright stance. Barin (1989) estimated that upright posture in bipeds requires over 700 muscles in a multi-link system including more than 200 degrees of freedom. Unfortunately, increasing age introduces new challenges to balance as individuals are confronted with diminishing physical abilities and depleting sensory acuity. Older people experience muscle weakness, vision impairment and morphologic changes in body systems, such as the vestibular and sensory system (Hobeika, 1999), which decrease their ability to perform activities of daily living (ADL). Thus, it is not surprising that falls are common issues for members of the older age cohort. Each system could be studied individually to determine its separate influence on balance ability and the aging process, but the fact remains that aging alone brings a general decline to upright stance and balance (Camicioli, Panzer & Kaye, 1997).

Researchers have devoted immense time and resources to understanding balance in older persons, as well as predict older individuals who may be more likely to fall. The common objective has been driven by the hope of implementing interventions to decrease the number of injuries, which are noticed each year within this demographic group (Boulgarides, McGinty, Willett & Barnes, 2003; Scott, Votova, Scanlan & Close, 2007; Topper, Maki & Holliday, 1993). Previous studies on fall prevention and prediction have utilized various assessment strategies including posturography tests (Girardi, Konrad, Amin & Hughes, 2001) and platforms (Piirtola & Era, 2006). Posturography tests focus on determining underlying sensory and motor impairments, as well as improving diagnosis ability and the capacity to treat problems of imbalance and instability (NeuroCom, n.d.). *Platforms* can be divided into two categories of static and dynamic, which may or may not include use of load cells to determine pressure distribution. Dynamic tests tend to focus on control strategy in balance loss conditions, whereas static tests often observe centre of pressure (COP) displacement in attempts to assess fall-risk.

Questionnaires have also been used as measures of fall prediction. These include, for example, functional mobility assessments (Scott et al., 2007) such as the Tinetti and Berg Balance Tools (The Society of General Internal Medicine, www.sgim.org/TinettiTool.PDF; The Internet Stroke Center, http://64.37.123/trials/scales/berg.html) or multifactorial assessment tools (Scott et al., 2007) such as the Downton Index (Nyberg & Gustafson, 2003). Functional mobility or multifactorial assessments are designed to rate or score individuals as to the likelihood of experiencing a fall. Here in, the functional mobility tools focus on physical abilities such as walking, sitting or bending, where as the multifactorial tools are based on the cumulative effect of known risk factors -such as fall history, muscle strength, medication usage. Scores of these tests, based on performance and issued by clinicians' analysis, reflect the inclination towards a fall. Throughout these questionnaire based tests, gender and vision conditions have also been observed to determine their influence on balance.

These tests - performance and questionnaire based - are often contradictory in their suggestions of 'best-measures' for fall prediction (Brauer, Burns & Galley, 2000).

Moreover, some tools have been found impractical due to the complexities and diversity in the measurement techniques (Maki, 1993). Development of a reliable instrument, which will allow for the prediction of falls, is necessary first step to assist in reducing falls and injuries in older persons. Therefore, it was a primary objective to develop and test a reliable platform for testing dynamic balance performance.

The Dynamic Balance Testing Platform (DBTP) (Funded by NSERC) was created to observe an individual's balance performance in a dynamic environment. Most, if not all research utilizing platforms have neglected the dynamic aspect of balance such that the focus has been assessing balance in a static environment, or only allowed motion controlled by the investigator. The platform's upper layer moved in response to the participant's body sway, similar to that of the EquiTest system (NeuroCom International Inc.). In the case of the EquiTest, allowance was given to the platform to move in relation to body sway; however this was only in the anterioposterior (AP) direction. The DBTP differed in the fact that it responded to both AP and mediolateral (ML) motion resulting in platform changes as opposed to simply AP movement. The DBTP was constructed with five support points, four at each corner and one being centered to the platform, with the task of each participant being to remain centered on a single-point support. Thus, the current research utilized the DBTP to characterize body sway in a more dynamic condition, related closer to fall circumstances.

The comparison of balance ability of various age groups, as well as gender, will be observed to show differences and validate the functionality of the DBTP. In the complex motor control of maintaining upright posture during bipedal standing, the role of supraspinal mechanisms is crucial (Era et al, 1996). Aging brings about a general reduction in processing speed of the central nervous system and may thus be a factor in poor postural stability. Gender influence on stability, on the other hand, is still somewhat up for debate. Wolfson and colleagues (1994) suggest females have less stability, but Era and colleagues (1996) suggested that better performance of women in regard to balance tests may be due to anthropometric factors. Other research states that sway was not affected by gender at any age (Colledge et al, 1994). Balance ability of each gender will be looked at in this study to assist in determining if differences exist among males and females.

The current research attempted to answer the following questions:

1) *Are there differences in center of gravity excursion (COGex) in the anterioposterior (AP) mediolateral (ML) and vertical (V) directions depending on people's age?*

2) *Are there differences in balance response data (BRD) (duration of balance (DurB) and platform motion (PM)) depending on people's age?*

3) *Are there differences in COGex (AP, ML, V) depending on people's gender?*

4) *Are there differences in BRD (DurB & PM) depending on people's gender?*

It is thought that as the age increases the COGex will as well. By looking at the BRD perhaps the same inferences can be drawn without utilizing the COGex, thus creating a new way to observe balance ability in age or gender. These questions will be looked at to determine the feasibility of using the DBTP to establish balance differences as a factor of age or gender. The DBTP's sensitivity to gender and age may qualify it as a new tool for quantifying balance ability resulting in steps towards fall prevention.

2. Method

2.1 Structure and Principles of the DBTP

The first step to acquiring data was the construction of the DBTP. The base of the platform was a 3/16 of an inch sheet of steel measuring 80cm^2 (31.5 inches^2) with the perturbation layer made of a 3/8 steel sheet measuring 60cm^2 (23.5 inches^2). The perturbation layer of the platform was designed to supply vertical oscillating movement depending on the position of subject's COG, in order to observe an individual's motor control response to the oscillation. The perturbation and base layers were joined together by a center pivot (Figure 2.1) allowing 2D oscillation of the upper layer around the pivot. This facilitated a maximum vertical movement of ±2.5 cm (1 inch) at each corner.

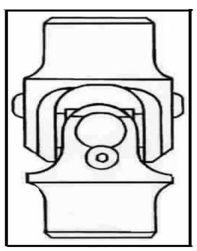


Figure 2.1. U-joint connector

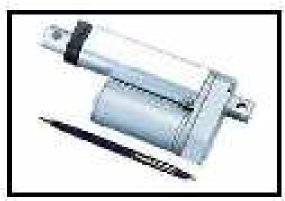


Figure 2.2. Linear Actuator

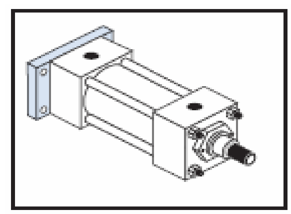


Figure 2.3. Air Cylinder

Three linear actuators (Firgelli Automations) were used (Figure 2.2) in order to stabilize the platform for the mounting and dismounting of participants. Each actuator had a 150lb push/pull force with a 5 cm (2 inch) stroke, and speed of 1.27cm (0.5 inches) per second, which was powered by 12vdc nominal voltage. Two of the three actuators were placed at the front of the platform where the participants mounted the perturbation layer, giving it extra stability, with the third placement at the back. An air cylinder (Bimba Manufacturing, Figure 2.3) was located at each corner of the perturbation layer. The perturbation layer was placed at a height that permitted the air cylinders to only reach a height of 2.5cm (1 inch) of their maximum stroke volume at the neutral stage, with the aim of allowing a range of motion (ROM) to be ±2.5cm (1 inch) in relation to the movement of the participant. The 250psi air cylinders had a rear flange, for attachment to the platform base, and single-rod stroke of 6.35cm (2.5 inches). Once in contact with the perturbation layer, the air cylinder's rod was a 2.5cm distance from the edges (Figure 2.4), sustaining the motion range for each corner at 5cm (2 inches). Two 12-volt batteries drove the actuators and air cylinders.

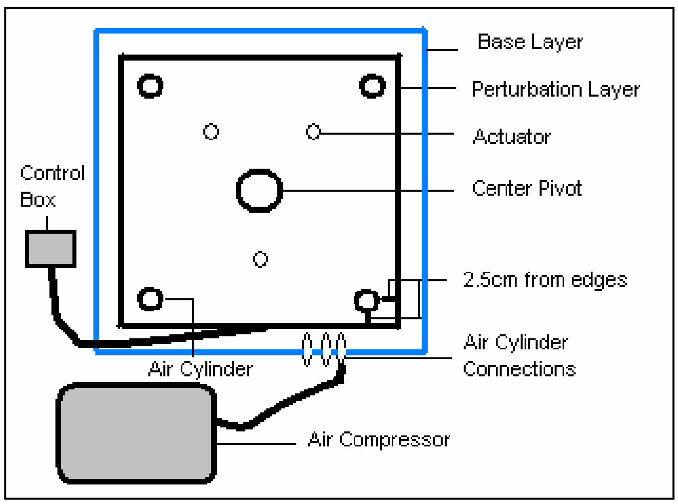


Figure 2.4. Layout of perturbation platform

Figure 2.5 depicts the platform from various angles. The air compressor used to pressure the air cylinders, ran at 3,900rpm (revolutions per minute). It had an oilless direct drive single-stage pump head with a maximum psi (pounds per square inch) of 100 and a peak power of 1/3hp (horse power). The air pressure setting was adjusted in relation to the participant's weight (Appendix VIII).

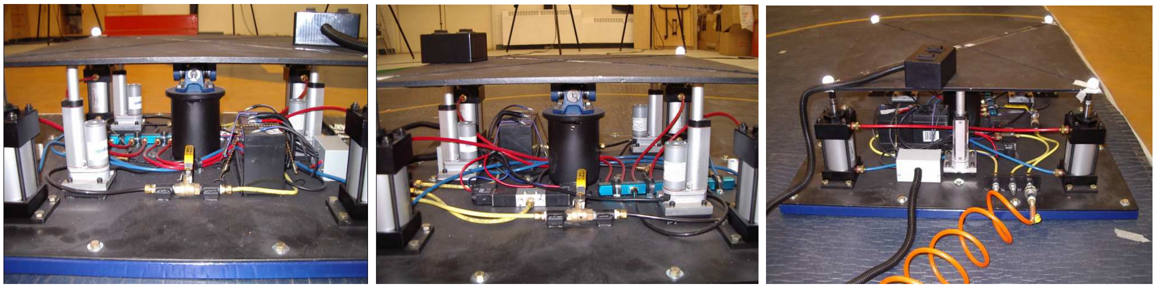


Figure 2.5. Depictions of DBTP (from left to right: view of right side, left side & back).

2.2 Subjects

The subjects represented three age cohorts, younger (18-29), middle-aged (30-59) and older (≥60), for both male and female. Student subjects were recruited from the community as well as the student population enrolled in a Biomechanics class at the University of Lethbridge. The 20 males and 27 females had a mean age (standard deviation) of 23.3 (±2.5) and 22.3 (±1.9) years respectively. Middle-aged participants (23 males and 21 females) were contacted by way of church and local community organizations in the Lethbridge area. Seniors (19 males and 27 females) were also recruited from these organizations, as well as through the seniors' centres and lodges in the area. The mean ages for middle age males and females were 40.7 (±10.1) and 42.9 (±10.3) respectively. The mean age for male senior participants was 74.4 (±7.5) and females' mean age 72.7 (±8.2).

To qualify as prospective subjects, individuals had to be able to stand unaided for 90 seconds, walk 10 meters and be able to understand verbal instructions (Maki, Holliday & Topper, 1994). These abilities were prerequisites due to the nature of the platform and test procedure. All subjects in the study were informed of the testing procedures and signed an informed consent form (See Appendix I – Consent Form Sample) for their voluntary participation.

The test protocol was approved by the Human Subjects Research Committee of the University of Lethbridge in view of the criteria from the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans, from the Natural Sciences & Engineering Research Council.

2.3 Test Procedure

Each subject was measured for body weight and shoulder width prior to testing in order to promote standardization of the test procedure. The body weight was used to establish the air pressure required for the DBTP with shoulder width acting as the foot location upon the platform. Participants were then informed of the platform characteristics once the stabilizers (actuators) were lowered. They were instructed that they were to keep the platform in a neutral position upon removal of the stabilization. The participants had to adjust their center of gravity (COG), to prevent the perturbation layer from resting on any of the four corners.

The participants were asked to stand with arms by their side, legs straight and eyes forward during the tests. The pressure in the air cylinders was adjusted to the participants body weight (APPENDIX VI: Pressure Table (PSI)) prior to removal of perturbation layer stabilization. Subjects were asked to stand with feet at shoulder width and in a diagonal position relative to the edges of the platform (Figure 2.6). It was thought that this pose would increase sensitivity of AP swing, which made the dynamic balancing control more challenging.

Figure 2.6. Participant in position upon perturbation platform

The subjects were allowed to warm-up on the platform executing two pre-test trials. This was considered adequate in reducing the influence of learning through the experiment. Such warm-up trials allowed for a consistent pre-test state and would increase the validity of the study (Shan & Wilde, 2002). Immediately following warm-up, four tests were completed, with the first two tests being executed with an eyes open (EO) condition and the second set with eyes closed (EC). Tests began with EO to allow participants to feel more comfortable on the platform prior to removal of vision. Each test lasted 20 seconds (See APPENDIX II: Protocol for – Platform). Air pressure was checked between the EO and EC tests to assure keeping of test standards. Each participant was given the option of wearing a harness (Figure 2.7) for safety; spotters (individuals placed around the platform) were also used for those not wanting to utilize the harness. A majority of the participants (98%) preferred spotters over the harness.

Figure 2.7. Optional safety harness used by participants

Anthropometrical data including the thickness of the carpometacarpal joint, ulnar side of the radiocarpal joint, and medial to lateral epicondyle of the humerus were measured with calipers for each participant. Measurements of the leg length, from the greater trochanter to the bottom of the foot, as well as foot thickness at the tarsometatarsal joint were also recorded for future use with kinematic calculations. This was recorded and utilized for reconstructing the 3D model and calculation of COG excursion.

2.4 3D Motion Capture and Biomechanical Modeling

In order to quantitatively determine the COG excursion during the dynamic balancing test, 3D motion capture was used. Specifically, a twelve-camera VICON v8i motion capture system (Oxford Metrics Ltd., Oxford, England) gathered kinematic data from the subjects. Capture occurred at a rate of 120 frames/second with the VICON software (Science & Engineering Software Suite, 2002) triangulating positions of each marker and rendering them to a three-dimensional computer space. Calibration residuals were determined in accordance with VICON's guidelines and yielded positional data accurate within 1.5 mm.

During data collection, subjects wore a comfortable black garment made of stretchable material, which covered the upper and lower body. Affixed to the garment were 42 reflective markers (See APPENDIX V: Anatomical Locations for Marker Placement), each with a diameter of 25 mm. The markers reflect infrared light to the cameras positioned around the subject.

From these 42 markers, a full body biomechanical model with 15 segments was built to determine COG excursion. The model worked as follows: from motion capture, anatomical positions were established, which then allowed the construction of a 15-segment full-body model. Using the fundamental precepts of physics, simple positional data were translated into the movement of the multiple-segment model, which was easily facilitated with usage of the Vicon system. In such individualized biomechanical modeling, the inertial characteristics of the body were established using anthropometric regression equations determined by Shan and Bohn (2002). The fifteen segments were head and neck, upper trunk, lower trunk, two upper arms, two lower arms, two hands, two thighs, two shanks and two feet.

2.5 Analytic Procedures

2.5a Definition of variable. Analysis of variance (ANOVA) and two-tailed t-tests determined the differences between the dependent variables (centre of gravity and balance response data) in relation to the independent variables (age, gender and visual condition). Analysis of the data took five parameters into account. Dependent variables were centre of gravity excursion (COGex) in the AP, ML and V directions; and balance response data (BRD) such as duration of balance ability (DurB) and quantification of normalized platform movement (PM) (See Figure 2.8). This was a 3X2X2 repeated measures test.

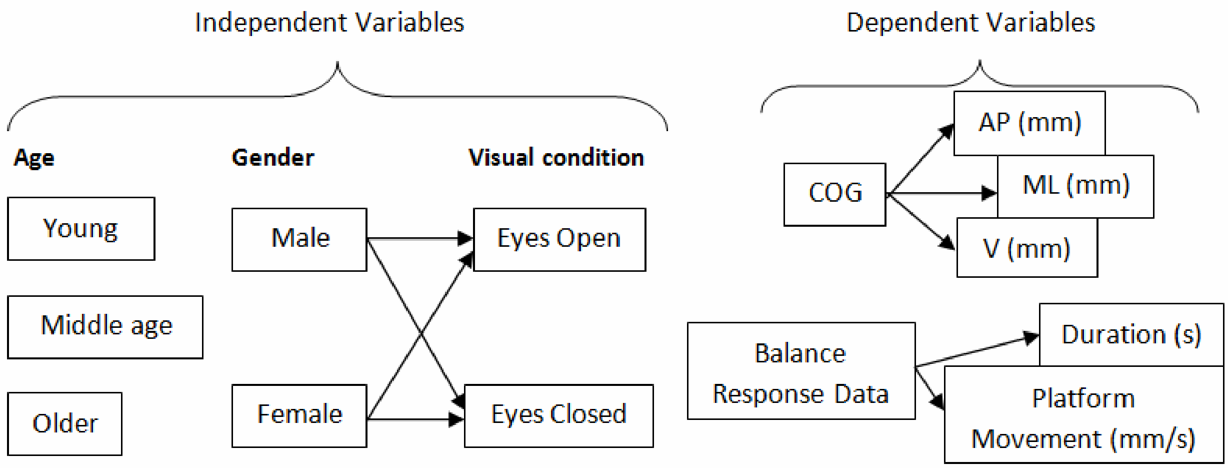


Figure 2.8. Detailed breakdown of in/dependent variables with age ranging from 18 to 87. X=anterior-posterior, Y=medial-lateral, Z=vertical direction, mm=millimeters, s= seconds

Center of gravity excursion was regarded as the range of motion (ROM) travelled by the subjects in each of the three directions. COGex of each trial utilized the time frame in which subjects were able to remain balanced upon the single-point support (or duration of balance ability -DurB). Therefore, the DurB was defined as the maximum length of time the subject was able to remain on the platform without the perturbation layer reaching minimum stroke length at any of the four corners (i.e. resting on a corner resulting in two or more points of support). Subjects' trial data were not useable if they were unable to balance on the center support. The platform movement (PM) was defined as the sum of distance travelled by the four platform corners during the longest balance performance in each 20-second trial (DurB). This distance was then divided by the length of balance time (seconds) on a single-point support for each trial. This resulted in a variable which considered the sum of distance travelled by all corners (mm) per second. An example of this would be Subject A balancing on a single-point support for 12seconds of the 20-second trial. Corners one, two, three and four travelled 32mm, 69mm, 49mm, and 73mm respectively for a total of 223mm travelled during the 12 second period. Normalization of these results entailed dividing the distance travelled (223mm) by the duration of balance (12sec) for an average of 18.58mm/sec.

Each subject underwent two trials for each visual condition Eyes Open/Eyes Closed (EO/EC). The completion of two tests was required for each visual condition, which were averaged to provide a mean score per condition. Trials in which the subject was unable to remain on a single point-of-support for both trials were not used in analysis, as the data was invalid. Thirteen participants in the eyes closed condition and 3 participants in the eyes open condition failed at both trials. Data was not transformed for skewness or kurtosis to maintain the meaningfulness of the results.

2.5b Analysis of question 1: are there differences in COG depending on people's age?

A one-way analysis of variance (ANOVA) was conducted to examine the differences among the three age groups (young, middle, old) for the outcome measures of COGex in the AP, ML and V directions for a total of six tests in two visual conditions (i.e. young, middle age and old in *EO* condition for *X*). Variables manifesting differences between age groups underwent post hoc tests by way of two-tailed t-tests to determine where the differences were.

2.5c Analysis of question 2: are there differences in balance response data (BRD) depending on people's age? A one-way analysis of variance (ANOVA) was conducted to examine the difference between three age cohorts (young, middle, old) in the DurB and PM parameters for a total of four tests in two visual conditions. Conditions showing significant differences underwent post hoc tests.

2.5d Analysis of question 3: are there differences in COG depending on people's gender? Six two-tailed independent samples *t*-tests were conducted to examine the differences between males and females for the outcome measures of COGex in the AP, ML and V directions in two visual conditions.

2.5e Analysis of question 4: are there differences in balance response data (BRD) depending on people's gender? Four two-tailed independent samples *t*-tests were conducted to examine the differences between males and females for the outcome measures of DurB and PM in two visual conditions (i.e. young, middle age and old in *EO* condition for *DurB*).

Due to the large amount of tests for each of the four questions, Bonferroni corrections were also made to reduce the chance of type one errors.

3. Results

3.1 Question 1: Are there differences in COGex depending on people's age?

Table 2.1 and Table 2.2 show the means and standard deviations of the movements in COGex (X3) with eyes open and eyes closed. From the descriptive statistics, in the EO condition there is less movement in the AP and ML directions for all three age groups in comparison to the EC results. The V direction, however, shows less motion in the EC condition (Tables 2.1 and 2.2) as opposed to EO, for all groups. In comparing age groups, the young participants had the greatest movement in the AP in both visual conditions. The middle age group had the least amount of motion in the V direction for both visual conditions. Also, in both visual conditions there was, in general, a larger degree of motion in the AP direction followed by ML and V movement respectively.

Table 2.1. *Means (M) and Standard Deviations (StDev) comparing three age groups' COG excursion (in millimeters) with eyes open.*

		Eyes Open					
		Anteroposterior (AP)		Mediolateral (ML)		Vertical (V)	
Age Group	n	M (mm)	St Dev	M (mm)	St Dev	M (mm)	St Dev
Young	45	47.98	16.50	27.02	9.63	7.59	5.56
Middle age	38	47.21	19.32	30.06	17.25	6.35	3.07
Older	43	44.35	20.96	25.73	13.70	7.02	4.82

Table 2.2. *Means (M) and Standard Deviations (StDev) comparing three age groups' COG excursion (in millimeters) with eyes closed.*

		Eyes Closed					
		Anteroposterior (AP)		Mediolateral (ML)		Vertical (V)	
Age Group	n	M (mm)	St Dev	M (mm)	St Dev	M (mm)	St Dev
Young	43	53.75	20.13	30.15	12.07	6.56	3.55
Middle age	37	49.21	19.59	31.57	17.18	6.00	3.63
Older	36	53.40	23.15	31.37	14.03	6.83	5.24

The ANOVA did not reveal significant results, $p>0.05$, for any of the three age groups in either the EO ($F_{AP/EO}=0.44$, $p=0.645$; $F_{ML/EO}=1.06$, $p=0.351$; $F_{V/EO}=0.74$, $p=0.482$) or the EC ($F_{AP/EC}=0.55$, $p=0.576$; $F_{ML/EC}=0.11$, $p=0.892$; $F_{V/EC}=0.39$, $p=0.682$) condition (See Tables 2.3 and 2.4). The p-values were all greater than the set alpha of 0.05 in each of the six tests indicating no difference in the AP, ML and V directions in relation to age.

Table 2.3. *ANOVA test for anteroposterior (AP), mediolateral (ML) and vertical (V) excursion in eyes open condition. df =degree of freedom, SS =sum of Squares, MS =mean squares*

		Eyes Open											
		Anteroposterior (AP)				Mediolateral (ML)				Vertical (V)			
Age Group	*df*	SS	MS	F	p	SS	MS	F	p	SS	MS	F	p
Between Groups	2	316.47	158.23	0.44	0.645	394.91	197.46	1.06	0.351	32.07	16.04	0.74	0.482
Within Groups	123	44238.17	359.66			22973.60	186.78			2684.09	21.82		
Total	125	44554.63				23368.52				2716.16			

Table 2.4. *ANOVA test for anteroposterior (AP), mediolateral (ML) and vertical (V) excursion in eyes closed condition. df =degree of freedom, SS =sum of Squares, MS =mean squares*

		Eyes Closed											
		Anteroposterior (AP)				Mediolateral (ML)				Vertical (V)			
Age Group	*df*	SS	MS	F	p	SS	MS	F	p	SS	MS	F	p
Between Groups	2	486.05	243.03	0.55	0.576	47.64	23.82	0.11	0.892	13.37	6.69	0.39	0.682
Within Groups	113	49591.33	438.86			23636.18	209.17			1963.69	17.38		
Total	115	50077.39				23683.82				1977.06			

3.2 Question 2: Are there differences in balance response data (BRD) depending on people's age?

Tables 2.5 and 2.6 show the means and standard deviations of the platform characteristics in relation to DurB and PM for eyes open and closed. From the descriptive statistics, it was observed that as age increased the length of balancing time tended to decrease in both EO and EC conditions (Table 2.5 & 2.6). Platform motion, however, increased as the age increased which might suggest more extreme movement per second in the older age individuals. Also, the descriptive statistics showed a decrease from EO to EC in the DurB means and standard deviations condition. The PM, however, increased in these calculations (Table 2.6).

Table 2.5. *Means (M) and Standard Deviations (StDev) comparing three age groups' Duration of Balance (DurB) and Platform Motion (PM) with Eyes Open*

		Eyes Open			
		Duration of Balance (DurB)		Platform Motion (PM)	
Age Group	n	M (sec)	St Dev	M (mm/sec)	St Dev
Young	46	16.20	3.89	15.67	7.47
Middle age	37	15.54	6.24	25.09	22.05
Older	43	10.23	6.76	29.11	18.47

Table 2.6. *Means (M) and Standard Deviations (StDev) comparing three age groups' Duration of Balance (DurB) and Platform Motion (PM) with Eyes Closed*

		Eyes Closed			
		Duration of Balance (DurB)		Platform Motion (PM)	
Age Group	n	M (sec)	St Dev	M (mm/sec)	St Dev
Young	43	12.66	6.57	30.23	20.92
Middle age	37	11.77	7.67	33.88	29.30
Older	34	8.62	7.43	35.53	25.93

The ANOVA results represented in Table 2.7 revealed a statistically significant difference among the three age groups in the EO condition for both DurB (F=12.77, p=0.000) and PM (F=7.55, p=0.001). This suggests that there is a significant difference in the DurB and PM based on different age groups in the EO condition.

Table 2.7. *ANOVA test results for Duration of Balance (DurB) and Platform Motion (PM) in eyes open condition. df =degree of freedom, SS =sum of Squares, MS =mean squares*

		Eyes Open							
		Duration of Balance (DurB)				Platform Motion (PM)			
Age Group	*df*	SS	MS	F	p	SS	MS	F	p
Between Groups	2	829.89	414.94	12.77	**0.000**	4206.31	2103.15	7.55	**0.001**
Within Groups	123	3997.60	32.50			34004.01	278.72		
Total	125	4827.48				38210.31			

The EC condition also revealed a significant difference in the DurB, F=3.186, p=0.045 (Table 2.8). This result reveals that significant differences in the DurB existed between the different age groups for the EC condition.

Table 2.8. *ANOVA test results for Duration of Balance (DurB) and Platform Motion (PM) in eyes closed condition. df =degree of freedom, SS =sum of Squares, MS =mean squares*

		Eyes Closed							
		Duration of Balance (DurB)				Platform Motion (PM)			
Age Group	*df*	SS	MS	F	p	SS	MS	F	p
Between Groups	2	329.84	164.92	3.19	**0.045**	625.12	312.56	0.49	0.617
Within Groups	111	5746.50	51.77			714.29	644.06		
Total	113	6076.34				72115.41			

Post hoc tests were completed to determine where the differences lie within the three age groups in relation to DurB and PM in the EO condition as well as the DurB with EC. With the assumption of equal variances being violated, a Games-Howell test for DurB with EO (Figure 2.9) was used (Morgan, Leech, Gloeckner & Barett, 2004). This showed significant differences in the DurB between the middle age and older individuals (p=0.011) as well as the young and older (p=0.000) in the EO condition. This suggests that the young and middle age individuals were both able to balance significantly longer than the older adults.

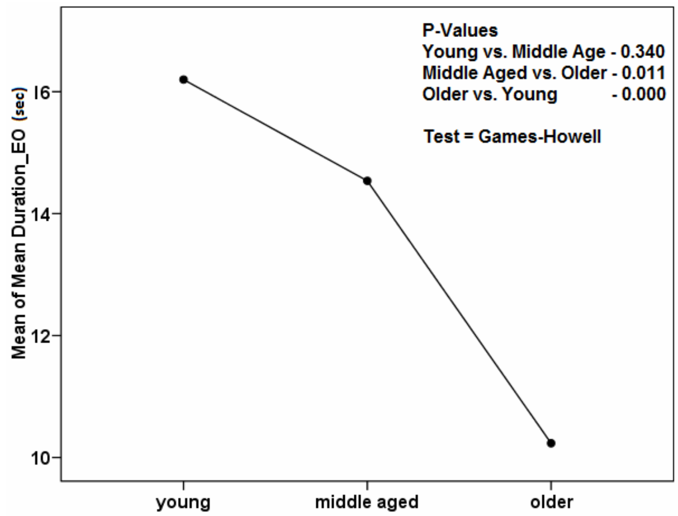


Figure 2.9. Mean duration with Eyes Open (EO) of each age group.

A second Games-Howell post hoc test was conducted to text for significant differences between the different age groups in the PM EO condition (Figure 2.10). Results suggested that there were balance differences between the young and older age group (p=0.000) as well as the young and middle age (p=0.044). It can be presumed that younger participants initiated significantly less motion in the DBTP than the middle age and older subjects. No difference in PM was noted between the middle age and older participants (p=0.659).

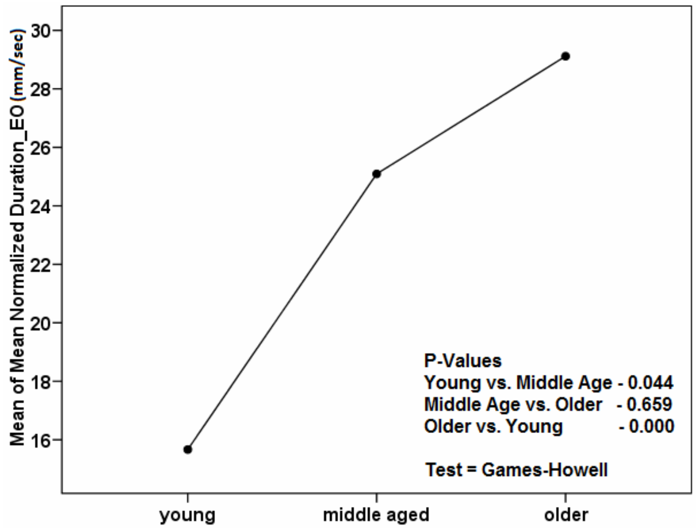


Figure 2.10. Mean normalized duration with eyes open (EO) of each age group.

A Tukey post-hoc test was performed for the final post hoc test with the assumption of equal variances being accepted (Morgan et al., 2004). The results of DurB data in the EC condition (Figure 2.11) showed a significant difference between only the young and old individuals (p=0.042).

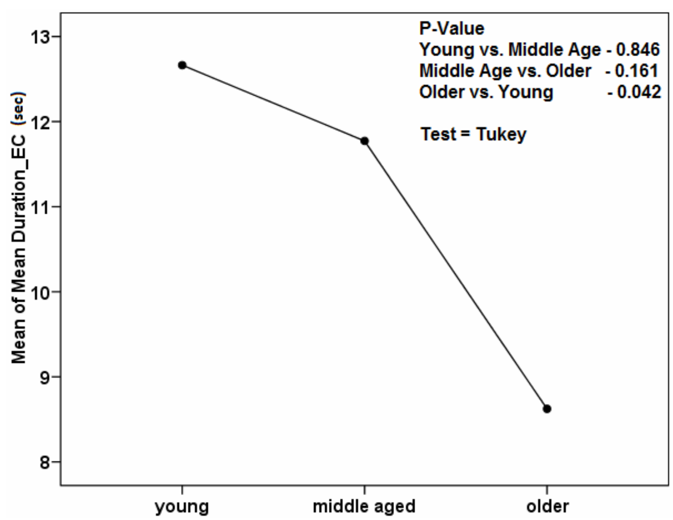


Figure 2.11. Mean duration with eyes closed (EC) of each age group.

Due to the number of post-hoc comparisons, Bonferroni corrections were made to the p-value representing significant difference ($p<0.05$). The corrected p-value = 0.00555 was given as the significance level to control for type one errors. This resulted in two, as opposed to five, variables which showed significant difference. These differences were between young and older participants in DurB as well as PM in the EO condition.

3.3 Question 3: Are there differences in COG depending on people's gender?

The gender comparison by means of two-tailed independent *t*-tests shows that males had a higher COG excursion than females in both visual conditions (Tables 2.9 and 2.10). As the visual condition changed from eyes open to closed, body sway increased for both genders. The largest distance travelled in both visual conditions, for males and females, was in the AP direction followed by the ML and V directions respectively.

Table 2.9. *Means (M) and Standard Deviations (StDev) comparing gender COG excursion (in millimeters) with eyes open.*

		Eyes Open					
		Anteroposterior (AP)		Mediolateral (ML)		Vertical (V)	
Gender	n	M (mm)	St Dev	M (mm)	St Dev	M (mm)	St Dev
Male	56	49.08	18.88	28.71	11.45	8.47	5.89
Female	64	44.45	19.50	26.52	15.23	5.94	3.77

Table 2.10. *Means (M) and Standard Deviations (StDev) comparing gender COG excursion (in millimeters) with eyes closed.*

		Eyes Closed					
		Anteroposterior (AP)		Mediolateral (ML)		Vertical (V)	
Gender	n	M (mm)	St Dev	M (mm)	St Dev	M (mm)	St Dev
Male	51	55.19	21.16	32.77	13.91	7.33	4.81
Female	59	49.84	20.49	29.58	14.64	5.79	3.43

The t-tests for the gender comparisons in the EO condition (Figure 2.12) were statistically significant only in the V direction $t(77) = -3.021$, $p = .003$ (two-tailed, unequal variance assumed). Results suggest that differences existed between males and females in terms of their vertical direction movement with eyes open.

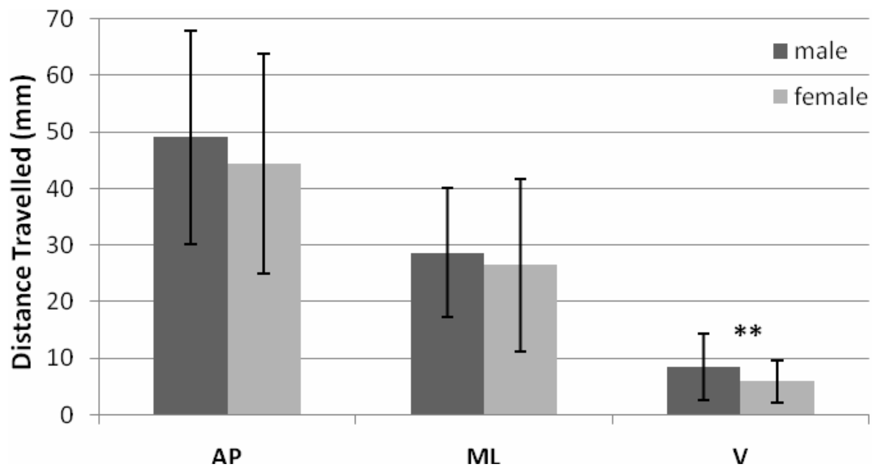


Figure 2.12. Comparison of COG excursion with standard deviations for male and female subjects in three directions in the Eyes Open (EO) condition. AP-Anterioposterior, ML-Mediolateral, V-Vertical. ******=$p<0.01$

The EC comparison between genders (Figure 2.13) shows similar results. Again the t-tests were only statistically significant in the V direction, $t(114) = -2.010$, $p = .047$ (two-tailed, equal variance assumed), indicating that in the EC condition males move more in the vertical direction than females. Bonferroni corrections were calculated for these t-test to determine a more appropriate significance level to account for type one errors. The resulting significance number ($p<0.00833$) suggested that differences were only noteworthy in the V direction with EO.

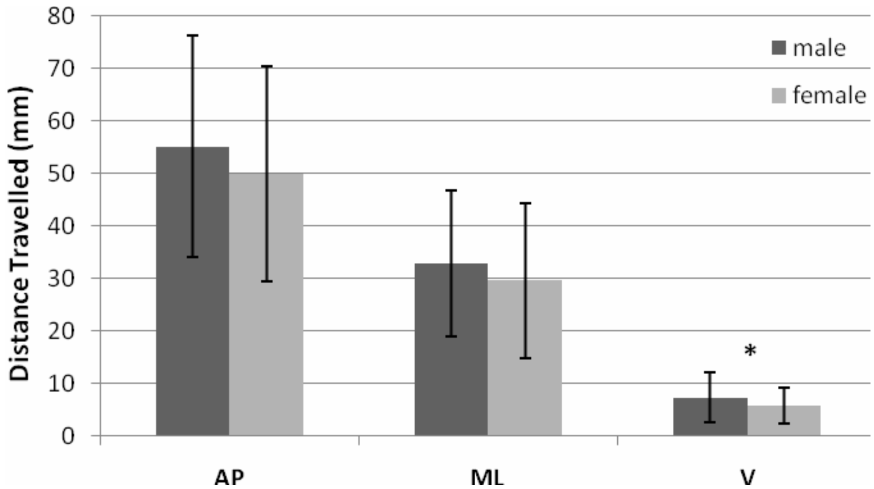


Figure 2.13. Comparison of COG excursion with standard deviations for male and female subjects in three directions with Eyes Closed (EC) condition. AP-Anterioposterior, ML-Mediolateral, V-Vertical. *****=p<0.05

3.4 Question 4: Are there differences in balance response data (BRD) depending on people's gender?

The evaluation of the gender effect on BRD by means of two-tailed independent *t*-tests revealed significant differences in balance duration between males and females in both visual conditions (Tables 2.11 and 2.12). The time length of balancing decreased for both genders when eyes were closed. Results showed also that males had the largest PM, or distance travelled by the corners of the platform per second, in both visual conditions. This may lead to a conclusion that a greater amount of motion results in reduced balance ability as seen in the male calculations. Platform motion increased for both genders when sight was removed.

Table 2.11. *Means (M) and Standard Deviations (StDev) comparing gender Duration of Balance (DurB) and Platform Motion (PM) with Eyes Open.*

		Eyes Open			
		Duration of Balance (DurB)		Platform Motion (PM)	
Gender	n	M (sec)	St Dev	M (mm/sec)	St Dev
Male	56	12.32	5.86	25.38	15.33
Female	64	14.72	6.32	21.15	18.97

Table 2.12. *Means (M) and Standard Deviations (StDev) comparing gender Duration of Balance (DurB) and Platform Motion (PM) with Eyes Closed.*

		Eyes Closed			
		Duration of Balance (DurB)		Platform Motion (PM)	
Gender	n	M (sec)	St Dev	M (mm/sec)	St Dev
Male	51	8.17	6.23	41.90	25.41
Female	59	13.43	7.34	26.15	23.11

The independent *t*-test for BRD and gender was statistically significant, $t(124)= 2.184$, $p = .031$ (two-tailed *t*-test, equal variances assumed), indicating that females had a significantly higher ability to balance in the EO condition than the males (Figure 2.14). In the EC condition, the ability of females to balance longer than males was also statistically significant, $t(112)= 4.034$, $p = .000$ (two-tailed *t*-test, equal variances assumed).

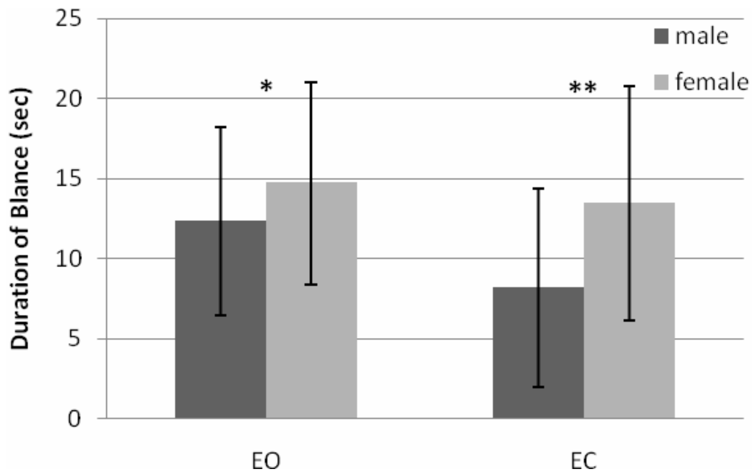


Figure 2.14. Duration of Balance in the Eyes Open (EO) and Eyes Closed (EC) conditions comparing males and females. *****=p<0.05, ******=p<0.01

No statistically significant differences were found between the males and females in the EO condition in relation to PM (Figure 2.15), $t(123) = -1.341$, $p = .182$ (two-tailed, equal variances assumed). The final *t*-test showed a statistically significant difference between males and females in the EC condition, $t(112) = -3.452$, $p = .001$ (two-tailed, equal variance assumed), indicating that males had a higher PM value than did the females in this condition. Bonferroni corrections in this data set again suggested a p-value of 0.00833, thus cancelling out significant differences in the EO condition for DurB.

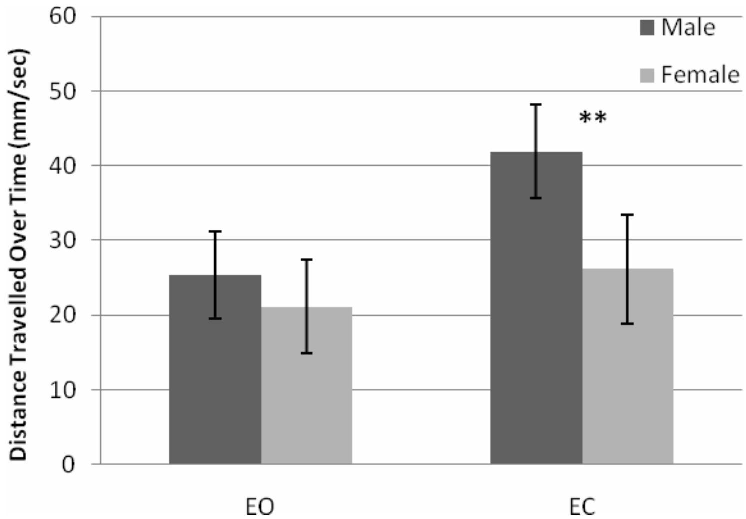


Figure 2.15. Platform Motion (PM) in the Eyes Open (EO) and Eyes Closed (EC)conditions comparing males and females. *****=p<0.01

4. Discussion

4.1 Use of platforms and measurements to determine difference in stability

The initial stage of the research was to create a platform that would respond to AP and ML sway of the subjects, without investigator intervention, in order to produce a dynamic environment requiring balance ability. Maki and McIlroy (1996) stated that 54% of the falls experienced by residents of a self-care facility, during a one year monitoring period, were due to base of support (BOS) perturbation, so studies related to this aspect would improve fundamental knowledge.

Previous studies have utilized perturbations to disturb BOS. These have used numerous approaches and objectives such as invoking stepping motions for observation of the response strategy (McIlroy & Maki, 1996), determine body segmental movements of head, trunk, thigh, shank and foot in response to the motion (Wu, 1997), and other control strategies (Nardone, Grasso, Tarantola, Corna & Schieppati, 2000; Shimada, Obuchi, Kamide, Shiba, Okamoto & Kakurai, 2003). These studies employed only platforms that move anteroposteriorly, lacking exploration of the impact of ML BOS motion.

Two studies were found which related to platform motion in the ML direction. Kim & Robinson (2005) used the Sliding Linear Investigative Platform for Analyzing Lower Limb Stability (SLIP-FALLS), to investigate either AP or ML direction, in response to investigator initiated perturbation. The focus was to explore control strategy during balance loss among diabetic and non-diabetic seniors. The second study employed small pseudorandom platform motions to perturb balance in the mediolateral direction (Maki, Holliday & Topper, 1994). Results of this test found that control of lateral stability had the most pronounced differences and was the single best predictor for future falling risk. Maki and colleagues conclude by stating that lateral stability may be an important area of study for fall-prevention intervention.

4.2 Question 1: Age differences in COGex

The primary interest was the exploration of age differences in COGex, under two visual conditions (EO and EC). Past research has shown centre of pressure (COP) to vary across age groups (Cohen, Heaton, Congdon & Jenkins, 1996; Shimada et al., 2003), but few have looked at the characteristics of COG. Previous research, which did analyze COGex, used the factor of total path length and found an age effect such that as age increased so did path length (Colledge, Cantley, Peaston, Brash, Lewis & Wilson, 1994). In the current study, no significant differences in COGex during the single point support were found for the age groups. This result may be due to the limited range of motion of ±2.5cm at the platform corners.

The current research results did not confirm previous research findings relating to COGex. This may have also been due to the number of "unsuccessful performances" that were experienced in each age group. A trial that showed no balance ability, such that no length of time was spent on a single-point support, was designated as unsuccessful. Table 2.13 shows the number of futile attempts (out of two trials per subject) that were experienced demonstrating the dramatic difference between completed trials among age groups. This may account for the lack of difference between young, middle-aged and older subjects in COGex. In order to remain stable, it required a certain ability that was likely shared among those that were able to balance on the platform.

Table 2.13. *Fail grades given in each cohort.*

	Fail Grades/Total Trial		Percentages	
	Eyes Open	Eyes Closed	Eyes Open	Eyes Closed
Young Females	00/56	05/56	0.00%	8.90%
Young Males	00/42	06/42	0.00%	14.30%
Middle age Females	00/33	03/33	0.00%	9.10%
Middle age Males	02/39	00/35	5.10%	0.00%
Older Females	06/55	20/55	10.90%	36.40%
Older Males	07/36	13/35	19.40%	37.10%

Visual differences suggested that in the eyes closed condition, all age groups increased their body sway in the anteroposterior and mediolateral directions. However in the vertical direction, the eyes closed condition had a reduced COGex (Tables 2.1 & 2.2). This may suggest that compensation strategies of the participants utilize direction change in the horizontal plane when they have an inability to use sight, while neglecting or minimizing their vertical changes. This supports previous research, in that vision is an important aspect of balance ability (Hobeika, 1999).

4.3 Question 2: Age differences in balance response data

The DBTP was able to move in both AP and ML directions in relation to body sway of an individual. The fact that it was created to be subject induced as opposed to investigator controlled motion may show to be important. This may be a good future tool for clinical fall-screening application due to the ability of the DBTP to amplify lateral as well as AP instability directed by the subject, as Maki et al. (1994) and Rogers and Mille (2003) state. The platform amplified dynamic balance ability in relation to body sway of the participants and demonstrated differences in balance ability between age groups as well as gender.

Results related to duration of balance upon the one-point support showed that in all instances the young were able to balance longer than both middle aged and older individuals. There was significant difference in DurB between the young and old as well as the middle age and old in the eyes open condition (Figure 2.9 & 2.11). Interestingly, significant differences in the eyes closed condition were only found between the young and old. Perhaps this shows that middle age individuals are in the process of proprioception loss and therefore have a more difficult time balancing than the younger cohort. Vision for the middle age and older individuals are therefore more necessary than for the younger generations.

4.4 Question 3: gender differences in COGex

Previous studies are inconclusive regarding the influence of gender. Some have found females to have less stability (Wolfson, Whipple, Derby, Amerman & Nashner, 1994). These differences, however, were negligible when results used body height as a covariate (Era, Schroll, Ytting, Gause-Nilsson, Heikkinen & Steen, 1996). Other research states that sway was not affected by gender at any age (Colledge et al, 1994). This may have been due to the diversity of tests procedures.

Though the vertical direction was the only comparison with significant difference between genders, all three directions showed that females had a smaller COGex. Perhaps this would have shown negligible when using body height as a covariate.

The only differences found in relation to COGex in our study were in the V motion in both the EO and EC conditions. No other located research has used or discussed the vertical motion as a parameter. Though this shows difference between male and female, height change would likely be manifest due to balance control strategies (i.e. using ankle versus hip to correct for disturbance of BOS), as opposed to balance ability. This parameter was thought to be of limited importance.

Again, as the visual condition changed from eyes open to eyes closed, an increase in COGex was noticed for both gender categories.

4.5 Question 4: Gender differences in balance response data

As with age, the largest differences between genders were found in BRD.

4.6 Summary

A pattern was noticed in three of the question's results. 1) Age differences between BRD found that DurB increased with age in both visual conditions the, resulting in a higher PM score as age increased. 2) Gender differences between COGex found that males covered the most distance in both visual conditions when compared to females. 3) Gender differences between BRD showed that females balanced longer and had lower PM than males, in both visual conditions.

4.7 Limitations

Limitations could be divided in to three areas: DBTP, investigator and analysis. The DBTP created motion in the vertical as opposed to the transverse plane. Many investigators of elderly fall prevention would look unfavorable upon this since falls tend to occur in a slip or trip fashion. Therefore it could be argued that the data collected is limited in applicable knowledge. A second factor that didn't work according to plan was the use of air cylinders. The perturbation layer of the DBTP was expected to move in a fluid motion and be responsive to sight changes in COG. Instead, due to the functionality of the air cylinders the fluidity was lost.

The main challenge presented on behalf of the investigator was the lack of subject screening. A very general qualification was accepted when allowing participation. This may have skewed the results since many older individuals use medication or are challenged by other factors that healthy youth are not limited by. Results may have been different had the populations selected been more particular to choose the more active, healthy individuals as the initial population of each cohort.

Finally, the analysis of results could also be disputed. Majority of previous research used COG to determine balance ability, which is the angle of approach here. The characteristics of the DBTP may suggest a different approach to analysis such as the number of times that the participant changes direction upon the platform, or observing the COGex throughout the entire 20 seconds as opposed to the limited time that the participants were able to balance. With the new tool come uncharted waters to be mapped out over subsequent projects.

4.8 Future research

The focus of the current research was to develop a novel approach to quantifying balance ability by way of COGex and BRD. The variables used were effective in suggesting that the platform has potential in adding new understanding to the challenge of fall prediction and prevention in the elderly, but further research may have to look at other variables such as speed and acceleration of COGex and/or the platform corners as well as the correlation between age and platform motion. It may also be beneficial to look at the number of times body sway direction change takes place for each subject. Young subjects tend to change directions more frequently and cover larger distance than seniors when comparing body sway characteristics (Shan & Wilde, 2002). Behavioral compensation strategies, such as emphasizing hip versus knee, would also be a future aspect to study to see response to balance loss in each age group (Wu, 1997).

5. Conclusion

A new platform, the DBTP, was constructed to describe individuals' balance ability while facilitating AP and ML motion. The equipment was used with three age cohorts, men and women, under two visual conditions. Differences in COGex were not found in any circumstance except in the vertical motion between males and females in one visual condition following Bonferroni corrections. The DBTP characteristics (DurB and PM) in relation to body sway were the best determinant for age differences and in showing gender variability. It was determined that females tend to be able to balance longer than males in both visual conditions. Age cohort related test results showed that older individuals are less able to stabilize themselves in a dynamic condition. In order to consider the DBTP as a new tool for determination of balance ability, more refined tests are necessary.

STUDY TWO

Senior Balance: Exercise Influence on Motor Control Response in a Dynamic Environment

1. Introduction

Age related sensori-motor degradation in the human system makes daily living more challenging for seniors when they are required to react to external dynamics (ED) due to situational surroundings. Previous studies have shown that aging is associated with an increased reaction time (Haier, Jung, Yeo, Head & Alkire, 2005; Melzer & Oddsson, 2004) as well as decreased lateral balance abilities (Mille, Johnson, Martinez & Rogers, 2005), which may be attributed to a loss of lean body mass (Prothro & Rosenbloom, 1995). Reasons for increased falling susceptibility in the aged include sudden obstacles, material defects, and unintentional contacts by surrounding individuals; all of which require quick reaction of the sensori-motor system (Kallin, Jensen, Olsson, Nyberg & Gustafson, 2004). These external factors, coupled with the degradation of balance dysfunction, reflect a decline in motor abilities and act as a known trigger of falls and injury (Rogers, Hedman, Johnson, Cain & Hanke, 2001). To minimize the deleterious effect, elderly individuals should attempt to improve or, at least, prevent the decline in their reaction abilities and balance stability to avoid a fall from taking place.

To reduce falls and injuries, Campbell et al. (1997) reported that a personal exercise program was effective in retaining strength and balance. They demonstrated that, when compared against the non-exercising control group, the mean rate of falls decreased for the individuals undergoing exercises. Rooks et al. (1997) also showed that exercising has a positive effect on slowing sensory-motor degradation related to age, suggesting that motor control is better retained in physical activity participants. Self-paced resistance training as well as walking exercises in community dwelling older adults were able to make improvements on tandem stance balance ability as well as other neuromotor performances and functional capacity. The importance of exercise is further supported by studies identifying that improvements are also seen in those with a history of falling (Shumway-Cook, Gruber & Baldwin, 1997) as well as individuals who began participating in physical activity following retirement (Buatos et al., 2007). Buatos et al. (2007) stated that regular physical activity, even when started late in life, allows appropriate reorganization of the different components of postural control and are able to adopt a more appropriate balance strategy.

Since exercise has been shown to be an effective means of improving balance ability and a good fall prevention strategy (Gu, Jeon & Eun, 2006; DiBrezzo, Shadden, Raybon & Powers, 2005), it is expected that subjects will show changes in postural stability, center of gravity (COG) excursion and/or response to dynamic factors following exercise sessions for beginners. A five-week exercise program has been shown to reduce falls and improve avoidance of obstacles in the elderly (Weerdesteyn et al., 2006); therefore, an extended four-month exercise session was chosen for the current research in order to observe effectiveness of the training style. A session exceeding four months may combine physical training effects with the aging effects and cause difficulties in analysis when determining the origin of results.

Fitball® exercise is becoming more and more popular in health-oriented fitness focused towards core strength/stability and was therefore our exercise style of choice. Core strength is a description of the muscular control required in the lumbar spine region to maintain functional stability (Akuthota & Nadler, 2004). To the best of our knowledge, biomechanical evaluations on senior exercise using Fitball® interventions are currently not available and core-strengthening programs are not well researched (Aduthota & Nalder, 2004). The current study tried to quantify the effects of the training session of a senior's Fitball® exercise program, by a pre- and post-exercise session strategy, in a simulated dynamic environment.

A balance test simulating a dynamic environment requires a specific stance of the participant as well as a reaction task to take place. The large number of previous studies related to balancing ability in seniors have examined *quiet* stance with feet at shoulder-width (Shan & Wilde, 2003; Brauer, Burns & Galley, 2000; Topper, Maki & Holiday, 1993) and its alternations, e.g. perturbations added to a standing platform (Henry, Fung, & Horak, 1998; Pai, et al., 2000). Jonsson et al. (2005) and Jensen et al. (2002) however, stated that a *narrow/tandem* stance during balance trials offers a close relation to a dynamic circumstance or situations in which falls often occur. Past studies have required a tandem stance position when testing participants' balance characteristics and ability (Era et. al, 2006; Melzer, Benjuya & Kaplanski, 2001; Era et. al, 1996; Rooks et al, 1997), but they failed to acknowledge the compounding impact of ED on loss of balance. ED can be described as the effects of forces on the motion of a body that do not originate within the body itself.

Factors that surround and impact an individual's equilibrium, while his/her attention may be projected elsewhere, can precipitate COG displacement to the point of imbalance resulting in a fall. A real life scenario may include an elderly individual who just completed grocery shopping and, upon searching for keys while walking to the car, stumbles over a speed bump. A tragic fall may be the result due to COG displacement from the external trigger, lack of motor control and slow reaction time. With falls occurring often in dynamic circumstances, such as walking, it is necessary to understand balance in a dynamic environment, which reflects real-life situations, as well as to develop an understanding of what can be done to improve postural control adaptation in relation to environmental changes. This study addresses this issue by introducing ED into the balance test and observing the influence of exercise on the participant's motor control response.

One gap of previous studies on postural adjustment is neglect of other attention-diverting tasks during balance control. Such multiple-reaction processes are much more complex than the single-reaction process that is balance recovery. This study tried to close the gap by mimicking such a process using a random weight-drop.

In summary, the aims of this study were 1) to explore the efficiency of new test method to quantify seniors' balancing ability by applying a narrow, walk-like pose in combination with ED and 2) to evaluate the effectiveness of a senior's Fitball® training program in improving motor control using an innovative and improved protocol.

2. Method

2.1 Subjects

Eight community-dwelling subjects, mainly of Caucasian descent and without co-morbidities, (mean age 63.6yrs±9.1, average weight of 69.4kg±17.4kg and average height 1.64 meters±6.4cm) participated in this study. . They were selected from the only two Fitball® classes in the city, which had a total enrolment of 80 participants. Only those new to the program and not involved with another exercise regime were asked to participate. This was done to eliminate any possible contamination effects resulting from preceding Fitball®

experience or other physical training. Measurements of each subject were taken at the beginning of the training session, September 2005. The Human Subjects Research Committee of the University of Lethbridge scrutinized and approved this protocol as meeting the criteria from the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans, from the Natural Sciences & Engineering Research Council. All subjects in the study were informed of the testing procedures, signed an approved consent form and voluntarily participated in the data collection.

2.2 Exercise intervention

The exercise intervention employed Fitballs®, large air-filled balls, to be used in various postures and movements with focus placed on strengthening core muscles through both static and dynamic exercise activities. Akuthota and Nadler (2004) stated that the musculature composing the core serves as the centre of the functional kinetic chain to stabilize the body and spine. Table 3.1 shows an outline of an exercise class with examples of exercises performed. The initial portion of the class consisted of a short warm-up followed by exercises focused towards static stiffening. Static exercises required participants to push their COG to the outer limits of stability and hold the position until the exercise coordinator instructed them to release.

Table 3.1. *Fitball® exercise class structure. L=left, R=Right. *base of support was single, two leg or tandem base of support during defined exercise.*

Interval	Time (min)	Short Description	Specifics (According to Exercise Coordinator)
Warm-up	5+	Laps with ball	Vary walking speed while carrying or bouncing ball
			Walk while tossing ball up and catching
			Walk while passing ball quickly from hand to hand
			Walk while passing ball around torso or rolling it in-front
Static	20-25	Stand (on the spot)	Roll ball on ground or torso height around the body*
			Lift ball overhead, maximal reach & hold in all directions*
			Lean forward, stretch ball forward, toss & catch*
			Toss up and catch behind back*
		Sit (on ball)	In front, using a tennis ball: bounce L, catch R & vice versa*
			Bounce tennis ball (counter) clockwise around body
		Lay (on ball)	Back to ball: Arms out from sides, roll on ball as far L or R as possible & hold.*
			Back to ball: With tennis ball in each hand, arms out from sides, use max voluntary contraction to squeeze L tennis ball. Hold until instructed to release, then R side.*
			Back to ball: Extend L/R leg, make little/big (counter) clockwise circular motions with leg.
			Belly to ball: Roll tennis ball from far L to far R across front.*
			Belly to ball: Extend one leg & opposite arm, lift high, hold & alternate sides.
Break (5-10 min)			
Dynamic	15-20	Stand	Bounce ball around body or under raised leg
			Facing partner: Bounce ball to partner, partner tosses ball at same time. Catch and repeat, reverse roles.
		Sit (on ball)	Arms out to sides, slowly lean back while walking feet out from ball. End up in a horizontal position on ball. Reverse back to a sitting position.
			Facing partner: medial aspect of one foot in contact with partner's medial aspect of foot. Push against one another to knock off balance.
		Lay (on ball)	Belly to ball: Roll body forward & back with hands (no foot to ground contact).
			Belly to ball facing partner: Push/pull one another to knock off balance.*
Cool-down	5+		Stretches utilizing or without ball.

Dynamic exercises and a cool-down made up the latter portion of the class. Dynamic exercises worked on improving active stiffness of core muscles by use of the balls with or without a partner or other equipment (i.e. tennis balls). In all exercises, participants' positions to the ball varied from vertical to horizontal in either a tandem stance or with one or two limbs in contact with the floor. The exercisers sat or laid upon the equipment, tossed, bounced or rolled it. Supplemental equipment, such as tennis balls, was also incorporated to improve fine motor control activities and to relate the exercises to daily activities. These classes took place three days a week for one hour each day lasting four months. A four month session was chosen to give participants sufficient time to adapt the training strategies and increase in strength, but be short enough that the ageing effect would be negligible. Adherence to program attendance was 65% or 26.5 of 41 classes. The Figure 3.1 shows a few of the exercises performed during a session.

Figure 3.1. Fitball® exercises. Left to Right: tennis ball squeeze with one limb support, ball catch, leg adduction against pressure with one limb support, bounce on Fitball® and tennis ball pass around torso.

2.3 Test procedure

Since sensori-motor function of seniors do not improve in the short period of four months without physical intervention, it was deemed sufficient to determine the training effects via comparison of post-test to pre-test.

Prior to beginning the exercise sessions, the participants were asked to undergo the pre-test. The test protocol required the subjects to stand with arms by their side in a walk-like stance, one foot placed in front of the other for a narrow base of support in the medial-lateral (ML) direction, with one foot per platform (Figure 3.2). The subjects were required to remain in the walk-like stance during all trials. The lead foot was self-determined by the subjects for the pre-exercise test sessions, but remained the same for the post-exercise test sessions – four months after the first assessment. To ensure safety, a spotter was situated next to the subject to pull or support individuals should they lose balance. However, no incidents occurred. A ball weight (4.5 or 8.5lbs) was positioned 50cm in front and 1m above each subject's head. The subjects were required to read aloud various letters, numbers and short words (5cm in height), which were projected low on a wall 8m away, 40cm from ground level. This was done to interfere with stability (Raymakers, Samson & Verhaar, 2005), keep their attention focused away from the weighted ball overhead and prevent use of peripheral vision for an early reaction to the weight drop. Projected characters appeared for 1.5 to 4.0 seconds. The weight was dropped at random and subjects were required to react to the falling object. The first weight of 4.5 lbs was released four times followed by four releases of the heavier weight, 8.5 lbs, for a total of eight trials. Analysis of four trials was chosen based on previous studies (Boulgarides et al.2003; Baloh et al., 1998; Thapa et al., 1996). Each participant was allotted sufficient warm-up time and familiarization tests prior to data collection in order to prevent influence of a learning curve. Drop velocity was controlled at 3m/s for all tests with differing weights used to determine the sensori-motor reaction on catch-response to various objects.

Figure 3.2. Walk-like stance of subject with spotter.

2.4 3D motion capture, ground reaction measurement and biomechanical modeling

Force Platforms are often used to characterize balance (Gatts & Woollacott, 2006; Papa & Cappozzo, 2000); they determine body sway characteristics in the anterior-posterior (AP) and ML directions. However, force platform data fails to supply such information for the vertical direction, which is essential in understanding the full body reaction taking place in response to ED. To remedy this concern, 3D motion capture with biomechanical modeling was applied for obtaining COG and determine dynamic reaction during the balance tests. For verifying the validity of COG calculations using biomechanical modeling, two force platforms (KISTLER AG, Switzerland) were used to acquire center of ground reaction force (CGRF) in order to compare the calculated COG and measured CGRF excursion in the horizontal plane.

A twelve-camera VICON V8i motion capture system (Oxford metrics Ltd., Oxford, England) was used to capture and synchronize body kinematics with force plate data by tracking 42 reflective markers (See APPENDIX V: Anatomical Locations for Marker Placement). Calibration of the system was based on Vicon's technical manual for camera and DataStation operation. Motion capture occurred at a rate of 120 frames per second, with the VICON software triangulating positions of each marker and rendering them in three-dimensional computer space. Accuracy was within 1.5 mm. Positional data of the markers received a 5 point weighted filter within the BodyBuilder supporting software then exported in ASCII format. A Sony Digital camera (30fps) was also synchronized to 3D and force capture to obtain video/visual reference data.

From the motion capture data, a full body biomechanical model with 15 segments (Figure 3.3) was built to determine COG excursion in 3D space. The model worked by establishing anatomical positions through motion capture, which allowed a 15-segment full-body model to be built. Using the fundamental precepts of physics, simple X/Y/Z positional data were translated into the movement of the multiple-segment model. In such individualized biomechanical modeling, the inertial characteristics of the body were estimated using anthropometric "norms" found through statistical studies (Shan & Bohn, 2003). The fifteen segments were head & neck, upper trunk, lower trunk, and two of each segment: upper arms, lower arms, hands, thighs, shanks and feet. The quantitative determination of COG enabled researchers to postulate balance control patterns employed by seniors under various conditions.

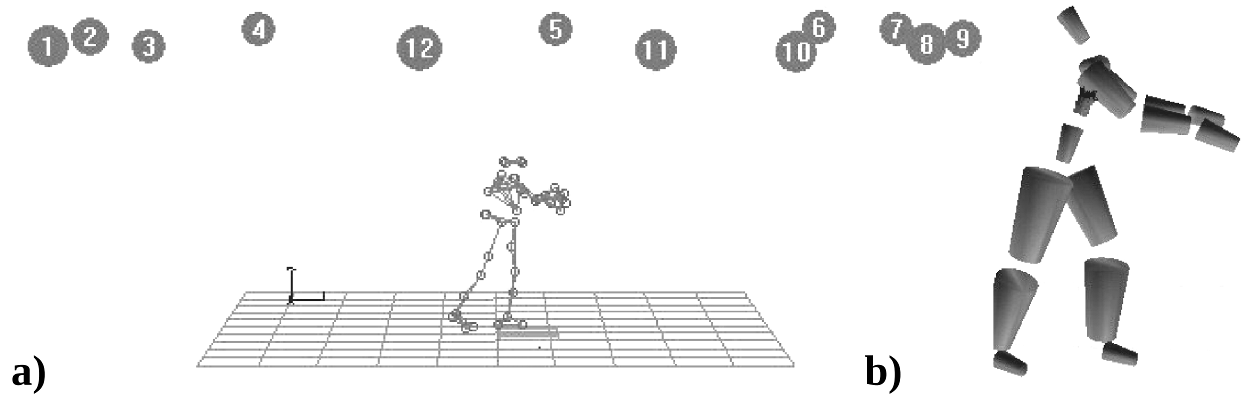


Figure 3.3. a) example of 3D motion with marker placement and camera orientation, b) The 15-segment model used in biomechanical analysis.

2.5 Data and statistical analysis

Three sequential phases were examined as participants stood in a walk-like position. The three phases were *quiet stance*, or ready-stance prior to visualization of ball drop; *reaction*, time from which the lower end of the measuring string was at subject's eye level to the position of weight-catch; and *follow-up*, from catch attempt to regaining of stability or both feet returned to ground contact. The objective was to test the balancing ability of seniors in response to sudden environment change pre, during and post-catch in relation to pre and post exercise sessions.

Quantification focused on *catch success rate* (number of successful weight catches), *balance success rate* (subject's ability to remain on the platform with both feet planted throughout the trial) and *COG excursion* during the three phases. A participant failing to catch the ball or keep both feet on the platform, were considered non-successful for that trial. Results were tabulated and analyzed by viewing video data.

A decrease of COG excursion for the ML and vertical directions was expected as core stability and strength developed since tandem stance requires greater use of hip ab/adductor or core control (Winter et al., 1996). Basic descriptive statistics were used, such as mean and standard deviation of selected parameters, to determine COG characteristics. . Paired t-tests were also used in comparing pre- and post-measurements of the training session. Significant level was set at $p<0.05$.

3. Results

The comparison between force platform data and 3D modeling calculation in AP and ML sway showed no significant difference, which validates the 3D modeling approach resulting in reliable results. Figure 3.4 shows an exemplary set of data of both COG and CGRF calculations in the AP direction. The two excursions, with a less than 7% error, confirm the effectiveness of using biomechanical modeling based on 3D motion capture to obtain a third parameter not attainable through use of force platforms.

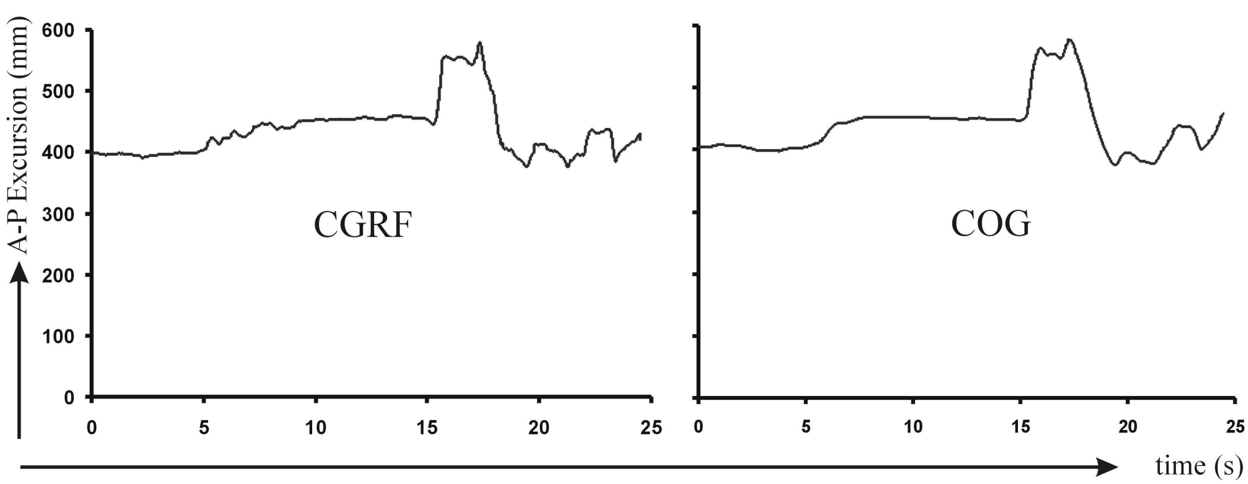


Figure 3.4. Sample of CGRF and COG excursion (mm) in the Anterior-Posterior (AP) direction during test conditions. The difference is <7%.

A fail grade for *balance success rate* was issued when removal of one or both feet from the platform was observed through video analysis. The results confirmed that between pre- and post-exercise session trials, the four-month training program significantly increased the *catch success rate* for the 4.5lb weight ($p<0.05$, Table 3.2). There were, however, no significant differences when the 8.5lb weight ball was used ($p>0.05$). *Balance success rate* showed no significant changes between pre and post exercise session for either the 4.5lb or 8.5lb weight drop ($p>0.05$).

Table 3.2. *Balance and Catch Success Rate comparing Pre- and Post- situations. bold = improvement.*

	Catch/Balance Success							
	4.5 lbs		4.5 lbs		8.5 lbs		8.5 lbs	
	Catch		Balance		Catch		Balance	
Subject	**Pre**	**Post**	**Pre**	**Post**	**Pre**	**Post**	**Pre**	**Post**
1	**0**	**2**	**2**	**4**	**0**	**2**	**2**	**4**
2	2	2	**0**	**3**	**2**	**3**	**2**	**4**
3	**0**	**3**	4	3	3	2	**1**	**4**
4	**0**	**2**	4	3	**3**	**4**	**3**	**4**
5	2	2	4	4	3	2	**3**	**4**
6	2	2	4	3	2	1	4	2
7	2	2	**3**	**4**	2	2	4	4
8	**0**	**2**	3	3	3	3	4	3
P-Values (pre vs. post)	**p=0.038**		p=0.504		p=0.763		p=0.195	

The COG excursion was evaluated by the range of motion (ROM), defined by maximum COG minus minimum COG. The examination of the training effects was done by comparing COG ROMs of pre- and post-test in 3 directions and in the three catch phases separately. The results show no significant influences of the training session on the *quiet stance phase*. Details revealed that quiet stance comparison of the pre- to post- exercise showed all p-values were larger than 0.05 in the AP, ML and vertical directions (Table 3.3).

Table 3.3. *Average ROM of COG in millimeters with standard deviation (St.Dev.) and comparison between pre- and post- tests. *: p <0.05, **: p<0.01, AP: Anterior-Posterior, ML: Medial-Lateral, 4.5/8.5 Lbs: Weight of ball.*

	Average COG Displacement and Standard Deviation (mm)											
	Anteroposterior (AP)				Mediolateral (ML)				Verticle (V)			
	Pre		Post		Pre		Post		Pre		Post	
Quiet Phase	**Average**	**St Dev**	**Average**	**St Dev**	**Average**	**St Dev**	**Average**	**St Dev**	**Average**	**St Dev**	**Average**	**St Dev**
4.5 lbs	19.3	5.7	22.6	7.5	21.7	3.4	20.3	8.5	6.3	2.2	4.9	1.9
8.5 lbs	18.8	8.3	19.9	10.5	16.8	1.9	23.6	7.0	8.1	3.8	8.0	3.8
Reaction Phase												
4.5 lbs	91.8	22.0	99.8	23.5	13.7	1.9	18.7	4.8	55.6	16.7	61.0	10.6
8.5 lbs	112.3	26.6	85.4	29.8	17.4	3.7	19.0	5.9	111.4	36.6	**51.9****	**8.3**
Follow-Up Phase												
4.5 lbs	105.4	32.2	95.4	32.9	59.1	26.1	38.8	22.6	93.1	27.7	75.5	17.7
8.5 lbs	128.3	29.3	82.9	35.1	74.1	31.2	35.0	14.8	**136.4**	**46.5**	**79.2***	**21.6**

During the *reaction phase* and the *follow-up phase,* a highly significant ($p<0.01$) or significant ($p<0.05$) difference was found in the vertical direction, with a decrease in ROM excursion when comparing pre- and post-exercise with the 8.5lb ball weight condition. Significant difference was also found in the ML direction during *follow-up phase* when the 8.5lb ball was used ($p<0.05$). However, no significant differences were found in either phase for both ball weight conditions in AP direction or for the 4.5lb ball in ML direction ($p>0.05$). The average training effects revealed by the COG excursion in millimeters can be found in Table 3.3. The majority of the measurements showed an improvement by reducing the ROMs between pre and post measurements, though the difference was not significant.

4. Discussion

Physical activity has been used as an intervention and shown to be an effective fall prevention technique in a number of studies (Buatos et al., 2007; Gu et al., 2006; Rook et al., 1997). A large aspect of the present research was to determine the usefulness of a four-month Fitball® intervention session in relation to a dynamic environment. The Fitball® exercise session focused on strengthening core musculature with the intent to improve balance ability and motor control, an area that has limited research results. 3D motion capture, ground reaction force measurement, biomechanical modeling and video (for observing *catch* & *balance success rates*) were used to determine the effectiveness of a Fitball® exercise intervention. Improvements were observed in COG control strategy among the subjects. The ROM of the COG decreased in the *reaction phase* as well as the *follow-up phase* in both the ML and vertical direction thus showing a more focused command and higher efficiency in muscle control (Hue et al., 2004). These changes showed a form of progress on motor control response in a dynamic environment; however no change was apparent in the AP direction. It is possible that AP movement improvements were disguised by the different forms of response strategies/techniques that each person uses (Winter, 1995). ML stability, due to the narrower base of support used in this test, relied more so on motor control and strength than did the AP stability. A narrow base of support requires hip abd/adductors as the dominant defense in ML direction, whereas the AP direction is more dependent upon reaction strategy in the hip, knee and ankle joints.

Effectiveness of the intervention was also determined by observing reaction time and balance ability by way of *catch success rate* and *balance success rate*. The first improvement revealed by this study through one beginner training session was *catch success rate*. This was seen to progress in accordance with previous studies on the elderly, suggesting that balance and strength improve with engagement in physical activity (Ashmead & Bocksnick, 2002). However, the improvement was only found during the trials employing the lighter weight. Based on the study by Proteau and colleagues (1994), such improvement can not be the effects of learning when activities have a blocked schedule of practice. The study found that large errors were observed in retention when acquisition trials were in block format. The current test schedule was established in a block style with the light weight being used initially followed by the heavier weight. Therefore, the improvement found by employing the lighter weight was the result of Fitball® exercise. The lack of visible improvements in catching the heavier weight could be due to the fact that the heavy weights are more demanding physically and thus more training could be needed before improvements can be seen.

Often in daily circumstances situations occur that require a rapid physical adjustment to regain equilibrium in a dynamic environment. Melzer and colleagues (2001) found that postural adjustments required cognitive processing, so distractions are likely to influence balance and rapid physical adjustment. Quick response time is often difficult when attention is directed elsewhere, which is the reason ED was added to a walk-like pose in order to explore a different way of quantifying seniors' balancing ability. As established in the introduction, this test posture is closely related to real-life situations where seniors lose their balance. Panzer et al., (1995) found that aging provided no evidence of postural instability in quiet stance when compared to younger individuals, but an altered postural control strategy was witnessed in the older participants. It was suggested that due to the altered control strategy, balance might be lost more easily in the case of sudden or severe equilibrium changes. The combined use of a walk-like stance and ED magnified these effects to allow for analysis of motion in three different directions to show the influence of the exercise intervention. In the future, it may be worthwhile to use this protocol to identify aging effects by contrasting seniors with young adults.

Including ED to a balance test using the tandem posture yielded results that are more pertinent to daily activities. Research in the past has often focused on balance ability pre- and post-exercise (Helbostad, et al, 2004; Melzer et al, 2001 & 2005), but few have addressed the issues of visual distractions requiring a reaction and the influence of physical activity involvement. By showing improvement or retention of balance in somewhat true-to-life conditions, the practicality of exercise programs that focus on improving or retaining balance ability was shown to have merit. This study showed that participation in physical activity on a regular basis results in a decreased response time (increased catch success rate, Table 3.2) to ED, as well as improved motor control ability in some aspects.

The results of this research were in accordance with previous studies that state exercise improves posture control as well as strength (Ashmead et al, 2002; Helbostad et al, 2005; Melzer et al, 2005). However, balance and strength tests are recommended for future studies in conjunction with ED to understand why observed changes occurred from the perspective of other test procedures. This would allow for a more detailed understanding of how balance and strength improved and what was the mechanism that promoted improvements in ED responses.

5. Conclusion

The comparison between pre-exercise and post-exercise conditions shows that a static walk-like stance with a random ED is an effective way to study balance ability in seniors. The *catch success rate*, *balance success rate* and the ROM of COG could be used jointly to quantify the dynamic balancing ability of seniors as well as to evaluate the training effects of senior training programs. The results of this study reveal that a four-month Fitball® exercise session has potential in improving the dynamic balancing ability of seniors who have no regular physical training, but further research is warranted.

Note - this study has been accepted for publication:

Dunn, B., Bocksnick, J., Hagen, B., Fu, Y., Li, X. Yuan, J., & Shan, G.B. (2008). Impact of Exercise on Seniors' Motor Control Response to External Dynamics. *Research in Sports Medicine*, 16 (1); 39-55.

STUDY THREE

Characteristics of the Silent Assassin–Ground Level Falls

1. Introduction

The Baby Boomer generation is steadily increasing the senior population and with it comes an increase in health concerns and costs related to injury and illness. One commonly occurring incident in the demographic of those over the age of 65 is falling. Over one third of this population experiences a fall, at least once a year, after they have reached age 65 (Dargent-Molina & Bréart, 1995). As the population of seniors increase, so will the number of injuries and costs involved in health care. Statistics have shown that falls often result in serious injury (Zhang, Wang, Xu & Liu, 2006) such as hip fractures, and estimates have been stated that world-wide fractures due to falls will increase from 1.7 million in 1990 to 6.3 million by 2050 (Cooper, Campion & Melton, 1992). Apple and Hays (1994) calculated that of the total hip fractures that occur, 90% to 96% (Norton, Campbell, Lee, Robinson & Butler, 1997) of them are associated with falls. Individuals that had a fall, resulting in serious injury, often experienced consequential complications (Davidson et al., 2001) or even death for 12-29% of the victims within the first year (Aharonoff, 1997; Jette et al, 1987). Vital Statistics for Ireland also attributed falls as the leading cause, at 83%, of injury deaths (2005). The seriousness of falls for individuals over the age of 65 is not a matter to be taken lightly. In order to prepare the aging generation for this *silent assassin* it is pertinent that a better understanding of falls be developed.

A number of interventions have been created to prevent injury in the case of a fall. Passive injury prevention devices such as hip protectors have been manufactured with energy shunting properties that displace collision forces by compressing upon impact, protecting fall victims. Van Schoor and colleagues (2003) suggest that in evaluating the effectiveness of these devices evidence advocates that there is no added benefit to the use of hip protectors for at-risk individuals. Minimal difference between numbers of hip fractures occurred in the hip protector group compared to the control group, whether due to lack of compliance and/or lower than expected effectiveness of the device was undetermined. Activating injury-prevention devices such as a Personal Emergency Response System (PERS) were also developed in order to prevent the length of time it takes for a senior to be discovered and assisted after a fall. The push-button PERS were found ineffective in the case of consciousness loss or fainting (Gurley, 1996). Porter (2005) also expressed concern about the lack of consistent use of such interventions even upon falling.

What can be done then to assist at-risk individuals in injury prevention in the case of a fall? Techniques have been developed to distinguish falls such as wearable sensors that monitor daily activity and detect falls by way of algorithms (Zhang et al., 2006; Zhang, Wang, Xu & Liu, 2006). At the moment, wearable sensor-based fall detection systems tend to collect information such as velocity, acceleration, vibration and tilt signals. Threshold values, which discriminate between normal and abnormal activities (i.e. falls), are then established for these signals, and type-of-activity decisions can be determined by comparing dynamic motion to threshold values (Zhang et al., 2006). Researchers have placed sensors on various anatomical locations to distinguish a fall; however the kinematic magnitude of various sensor locations has not been identified throughout ADL and fall activities.

Research tends to focus on balance ability of seniors (Shan & Wilde, 2002) and what causes a fall to occur (Norton et al., 1997), but what if the inevitable does take place? It is essential to determine fall characteristics and how they differ from that of normal activities, or if they do. Wu (2000) as well as Bourke and Lyons (2007) suggest that horizontal and vertical velocity of normal activities do not demonstrate the same signatures as found in the fall actions, which suggests that fall detection is achievable. Understanding the characteristics of the fall process will perhaps lead to further development of functional devices such as sensory-triggered airbag devices (Chase, 1993; Davidson, 2004; Lockhart, 2006; Ulert, 2002). A low-profile garment containing sensors would detect fall activity and inflate in the event of a fall. Establishing kinematic characteristics of falls will accommodate triggering of such a device to allow necessary protection. With a maximum velocity occurring too late in the event of a fall, minimal time would be allocated for airbag deployment; therefore prompt triggering of this mechanism would be essential for effective injury prevention. A fall detection sensor must be placed in a location on the body so as to trigger 'detonation' quick enough to be influential in high impact prevention.

The intent of this study was to demonstrate the differences in velocity found at various anatomical locations that will allow future studies employing sensors to use appropriate thoracic positioning. It is hypothesised that there will be differences between the various anatomical locations which will classify a location as best able to signal fall patterns prior to ground contact. Demonstrating maximum velocity occurrence as a percentage of time elapsed, as well as a suggesting minimum threshold value for velocity, were also objectives.

2. Method

2.1 Subjects

Thirteen subjects participated in the study with ages ranging from 17 to 54. Three males and ten females participated. This study was approved by the Human Subjects Research Committee of the University of Lethbridge in accordance with the Tri-Council Policy Statement: Ethical Conduct for Research Involving Humans.

2.2 Tasks

Subjects underwent one trip-fall (T-F) condition as well as five push-fall (P-F) conditions that required falls in various directions. The T-F required the subject to walk with eyes closed towards a bar, held at mid-tibia height. Upon contact with the bar they were to trip and fall forward, landing on 38 centimeters (15 inches) of foam and air mattresses (Figure 4.1). Though individuals walk more slowly with closed eyes, this was done to eliminate artificial preparation for the unexpected T-F. This condition was completed twice for each subject. P-F trials commenced by having the subject vertically rotate with closed eyes, in order to create disorientation. When the subject was asked to stop they were repositioned next to the cushioning (if necessary) by minimal contact from the tester. They were then pushed in one of five directions to promote a vertical drop, falling to their left side, right side, back or front (Figure 4.1). Subjects completed each of these five randomly occurring fall conditions twice for a total of ten P-F trials. Eyes were to be kept closed during *all* fall trials to simulate spontaneity and encourage an automatic response.

Fall conditions were compared to ADL (Figure 4.1) such as reaching for an object, sitting down, and ascending and descending stairs. A bottle of water placed on a waist-height table facilitated the reach trial. Subjects were requested to walk to the table, pick up the bottle, “take a drink”, replace the bottle on the table and walk away. Secondly, subjects walked up to a structure, sat down, got up and walked away. For the final common activity, a unit of four stairs was set up. Subjects climbed to the top of the staircase, turned around and descended. The reach, sit, and climb tests were completed twice for a total of six alternative motion trials to be used in comparison to the fall data. Each subject completed a total of 18 trials.

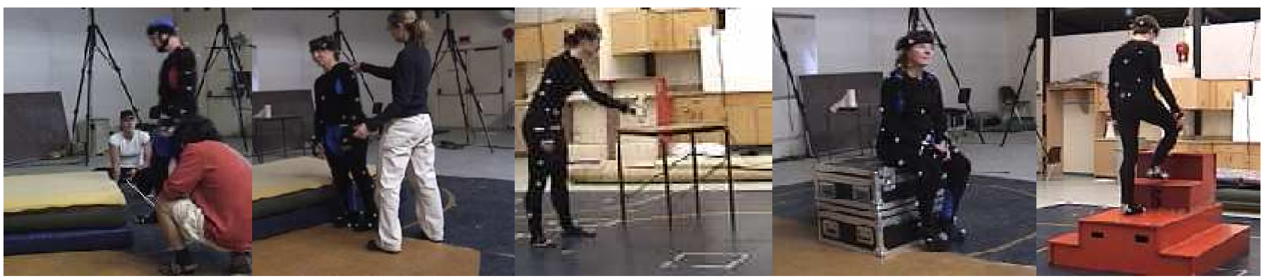


Figure 4.1. Subjects in the various test circumstances; trip-fall, push-fall, reach, sit and climb (left to right respectively).

2.3 Instrumentation

Distinguishing features such as maximum velocity and point-in-time of max occurrence of the eight anatomical landmarks were needed to determine the differences between fall events versus ADL such as sitting. This was accomplished by way of a twelve-camera VICON V8i motion capture system (Oxford metrics Ltd., Oxford, England), which tracked 42 reflective markers. Calibration of the motion capture equipment, for coordinate tracking, was completed prior to each of the test sessions. The 42 markers were placed on specific anatomical landmarks (See APPENDIX V: Anatomical Locations for Marker Placement) and allowed each subject to be represented by way of a 15-segment model (Figure 4.2) as in previous studies. BodyBuilder software was then used for smoothing the displacement in the trajectories, by way of a 5 point weighted filter, as well as filling trajectory gaps due to marker occlusion. Accuracy of marker location was calculated to be within 1.75 mm. A Sony Digital camera (30fps) was also synchronized to 3D data to obtain video/visual reference data.

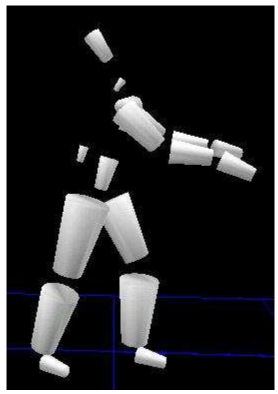


Figure 4.2. 15-Segment Model

2.4 Analysis

Analysis consisted of examining eight markers (C7 vertebrae, T10 vertebrae, sternal notch, xiphoid process, left superior & right inferior angles of the scapula, and left & right acromion process) to determine maximum resultant velocity throughout the various maneuvers. 3D-X/Y/Z positional data for the eight locations were exported from a .C3D file format to ASCII via BodyBuilder software and the following formula was used (Microsoft Excel 2007) to calculate marker position (P) in each frame (n): $P_n = \sqrt{((X_2-X_1)^2 + (Y_2-Y_1)^2 + (Z_2-Z_1)^2)}$

Velocity was then obtained through single-step difference, with each frame (step) lasting 1/120 of a second. Comparison of the averaged maximum values for each of the three ADL was done by repeated measures Analysis of Variance test (ANOVA) in SPSS (version 13.0). The same process was used for determining relationship among the six fall activities.

A one-tail independent t-test calculated P-values for comparing ADL to fall activities. For this calculation, all three maximum ADL values were averaged for each marker location to obtain eight values. This was repeated for the six fall activities, resulting in two variables (ADL max averages and fall max averages) for the t-test. In summary, a total of 24 values were obtained to classify the differences between and within activities and marker locations.

Time was calculated by determining the duration of the ADL from beginning of activity-motion to end of activity. To prevent the influence of the walking phase during the ADL trials, walking portions were removed prior to data processing. The walking phase during a trial was used only to promote natural motion upon engaging in the desired activities. In the case of the falls, ‘end’ was defined as initial hip contact of cushioning to prevent the damping effect from influencing calculations. Decision of end time was established from use of the .AVI files acquired by the Sony digital camera. The time-at-maximum factor for each of the two activity types was calculated as a percentage of maximum velocity occurrences relative to total time elapsed.

A threshold value was also established in order to determine fall activity early enough to detect a fall. This was done simply by observation of ADL maximum velocities and assigning a threshold value slightly above that of the largest maximum. Results from Subject A were used to demonstrate the relative time to velocity passing threshold and velocity reaching maximum for each sensor location.

3. Results

Table 4.1 is a synopsis of velocities in each of the eight marker locations in relation to the nine activities engaged in with corresponding standard deviation. The maximum resultant velocities for normal activities averaged 1.303 m/s whereas the maximum velocities for fall activities averaged 3.603m/s – almost three times that of normal motion. Results within the ADL category showed no highly significant differences ($p<0.01$) between seven of the eight locations with only three having significant difference ($p<0.05$). More difference was seen between the fall types in the eight marker locations with a p-value of less than 0.01 at all of the eight marker positions. Analysis of the ADL versus fall activities revealed highly significant differences with p-values dramatically lower than 0.01 (Table 4.2).

Table 4.1. *Maximum velocities (meters per second) with standard deviation at each anatomical location for the nine activities. TF=Trip-Fall, FL=Fall Left, FR=Fall Right, FD=Fall Down, FF=Fall Forward, FB=Fall Back*

	Maximum Velocity (m/s)															
	C7		T10		Sternal Notch		Xiphoid Process		Left Scapula		Right Scapula		Left Shoulder		Right Shoulder	
	Max	St Dev	Max	St Dev	Max	St Dev	Max	St Dev	Max	St Dev	Max	St Dev	Max	St Dev	Max	St Dev
Stairs	1.322	0.196	1.405	0.257	1.292	0.208	1.361	0.422	1.559	0.218	1.382	0.224	1.418	0.229	1.395	0.206
Reach	1.209	0.199	1.244	0.429	1.145	0.154	1.157	0.193	1.187	0.239	1.238	0.217	1.167	0.188	1.269	0.204
Sit	1.259	0.251	1.351	0.261	1.25	0.357	1.221	0.256	1.35	0.435	1.447	0.317	1.249	0.261	1.4	0.268
TF	4.134	0.673	3.875	0.518	4.029	0.544	3.43	0.506	4.051	0.61	3.675	1.688	4.295	0.823	4.236	0.634
FL	3.865	0.401	3.552	0.386	3.558	0.433	3.179	0.45	3.737	0.332	3.742	0.399	3.748	0.307	3.841	0.498
FR	3.951	0.447	3.65	0.461	3.527	0.425	3.057	0.474	3.789	0.353	3.692	0.366	3.821	0.417	3.688	0.439
FD	3.274	0.73	2.98	0.51	2.926	0.656	2.526	0.556	3.136	0.634	3.118	0.686	3.173	0.672	3.138	0.717
FF	3.931	0.796	3.625	0.446	3.725	0.484	3.352	0.388	3.708	0.364	3.707	0.494	3.82	0.364	3.912	0.552
FB	3.792	0.501	3.298	0.396	3.689	0.492	3.289	0.597	3.623	0.403	3.56	0.313	3.684	0.609	3.819	0.43

Table 4.2. *P-values from ANOVA calculations showing highly significant variation between fall and daily activities.*

	p-Values		
Location	**Normal**	**Fall**	**Normal + Fall**
C7	0.177	<0.001	<0.001
T10	0.184	<0.001	<0.001
Sternal Notch	0.096	<0.001	<0.001
Xiphoid Process	0.047	<0.001	<0.001
Left Scapula	0.029	<0.001	<0.001
Right Scapula	0.022	<0.001	<0.001
Left Shoulder	<0.001	<0.001	<0.001
Right Shoulder	0.064	<0.001	<0.001

The maximum velocity occurrence in relation to elapsed time can be seen in Table 4.3. It was concluded that a marker on the xiphoid process would reach its peak value prior to all other marker positions, when averaged across all fall types. The marker placed on the xiphoid process reached the maximum velocity of the six-falls combined at an average of 86.6% following fall start. A marker placed at C7 vertebrae typically took the greatest time to reach maximum velocity, at 90.1% of the fall time.

Table 4.3. *Percentage of time elapse at maximum velocity occurrence.*

	Percentage of Time Elapsed
C7	90.1
T10	88.9
Sternal Notch	89.8
Xiphoid Process	**86.6**
Left Scapula	89.8
Right Scapula	88.9
Left Shoulder	89.3
Right Shoulder	89.6

Threshold was established to be 2.0m/s, a value that exceeded all maximum resultant velocities for ADL, but was superseded by all fall activity resultant velocities. Subject A shows an average difference of 26% between the time that threshold was surpassed to the time maximum velocity was reached (Figure 4.3).

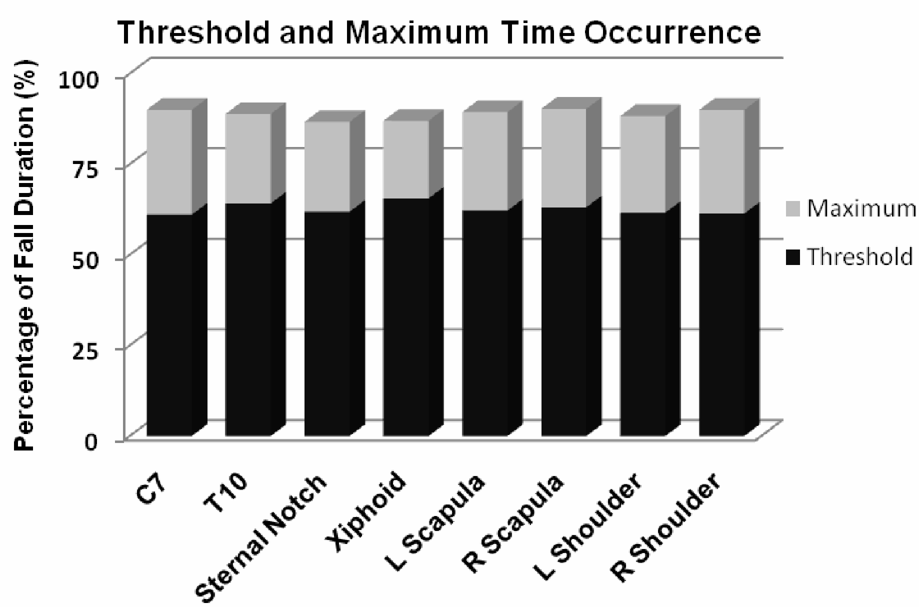


Figure 4.3. Time percentages described as factors of threshold (2.0m/s) and maximum velocity occurrence relative to total fall duration for Subject A.

4. Discussion

As previous studies have shown, general activity types have distinguishing features allowing for differentiation between falls and common ADL (Wu, 2002; Zhang et al. 2006; Bourke et al, 2007). These studies typically established a threshold level that would differentiate between ADL and fall events. In this study, a threshold level could be established over 2.0 m/s at any of the eight trunk locations, which would indicate a high likelihood that a fall was taking place. The difference in time between reaching threshold compared to the maximum velocity value for Subject A is demonstrated in Figure 4.3. Were a threshold level to be established at 2.0m/s, more time could be allotted to fall detection and injury prevention. Interventions, such as an anatomical airbag, created to prevent injury upon impact in the case of a fall would benefit from use of this finding by triggering deployment as velocities surpass the defined threshold value. Not only would this assist the victim, but also the health costs related to resultant injury.

The location found to reach it's maximum velocity the quickest was positioned on the xiphoid process and would be the recommended placement for earliest fall detection if the maximum value was used for triggering. A threshold value of 2.0 m/s however, would suggest that a sensor be placed on C7 to allow the largest time frame to distinguish between falls and ADL. Airbag deployment for vehicles requires approximately 15-20 milliseconds for the sensors and control unit to determines severity of an accident and decide whether to deploy, by 25 milliseconds it begins to inflate and at 45 milliseconds it reaches full inflation (PC Police, retrieved May 6, 2007). Time is an issue with fall durations of approximately one second. Adequate foreknowledge of a fall enhances the ability to deploy the airbag mechanism with substantial time to be an effective intervention. Though a custom anatomical airbag device would need to be fabricated, it can be expected that such a device would not be created to deploy any quicker than that used for automobiles. This in mind, a sensor location that gives the largest amount of warning would be suggested.

Understanding the characteristics of the silent assassin may also contribute to psychological ease in the targeted age group. Hayes and colleagues (1996) found that striking the ground in a stiff state was actually found to increase forces upon impact. This seems counter-intuitive as one may suppose protection to be afforded to a joint at impact, which is encased by flexed muscles as opposed to a relaxed state. Marks et al (1997) suggests that a study by Luukinen, Koski, Laippala and Kivela (1997) may have a plausible explanation in that the fear of falling amplifies co-contraction and tenseness as opposed to graded contraction of musculature. The resultant stiffening may contribute to, rather than absorb, the impact experienced during a fall. Knowledge, as well as effective physical interventions, potentially reduces fears that could contribute to impact injury.

Future studies in determining fall characteristics should be conducted to further understand the characteristics of ground level falls. Included in these studies should be the condition of unexpected falls with eyes open. This would allow quantitative analysis of velocity or acceleration in more life-like fall conditions. Another study should focus on other signals such as trunk angle or acceleration. In this way interventions such as an anatomical airbag may be allotted even greater amounts of time to react to a fall situation. Further studies might also include marker characteristics on lower limb locations to determine differences in upper and lower body velocity or acceleration.

5. Conclusion

Discrimination between falls and ADL is possible using resultant velocity parameters of anatomical locations on the trunk. Placement of a sensor on the xiphoid process would be best if triggering at maximum velocity, but threshold values offer earlier triggering for fall activities. This early triggering may be critical for future interventions, such as the creation of inflatable devices for hip protection.

GENERAL CONCLUSION

The focus of this research was to investigate motor control and balance ability in an aging cohort. Balance is affected by various parameters such as movement speed, muscular strength (de Bruin, 2007), and proprioception (Petrella et al., 1997). These parameters typically decline with age. As a result fear and inactivity settle in and often become contributors to falls resulting in immediate consequences such as fractures, as well as remote complications such as dependent living.

In an attempt to investigate the aforementioned objectives, three studies were conducted, each observing a different aspect of the aging process in relation to motor control, balance ability and falls. The first study addressed postural control in a dynamic environment in attempts to gain insight into individuals' balance abilities. The second study attended to an intervention focused towards balance improvement in a dynamic environment. The third study examined an injury prevention approach to fall incidents.

This thesis looks at balance and fall situations by observing balance ability in a dynamic environment and examining two resulting approaches to injury prevention. Active intervention is a primary approach to injury prevention (Gu et al., 2006). Keeping the elderly population well and active will assist in improving their quality of life by supporting independent living and promoting healthy choices. Passive intervention should be secondary to activity involvement and used in the circumstances in which the individual is unable to perform physical activity or the primary approach (Figure 5.1).

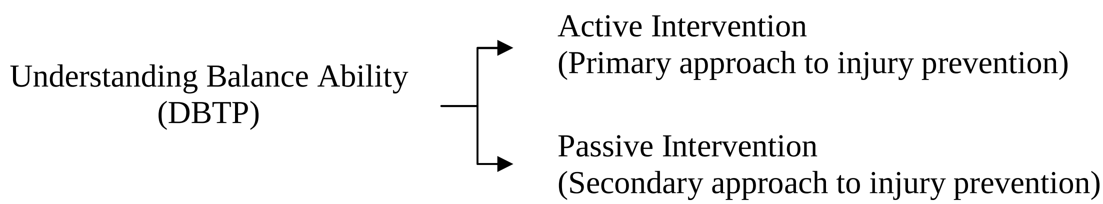


Figure 5.1. Overview of thesis view points

Study one used a moving platform, the Dynamic Balance Testing Platform (DBTP), in attempts to understand balance ability. Previous tests on balance ability often employed force platforms in a static environment, from which transferability to real life can be argued. Activities of daily living (ADL) are generally dynamic events that rely on the interaction of individual and environment. Use of the DBTP attempted to bring the dynamic environment connect into play while searching for an understanding of balance in individuals. This study replicated previous research findings (Maki & McIlroy, 1996); individuals' response to the environment is more rigid with increased age, in turn amplifying processing time and movement responses.

Study two was to evaluate the effects of an exercise intervention employing a novel exercise program and testing protocol. Exercise is often encouraged as a means of improving health and balance in the aging population (Campbell et al., 1997). Since exercise is recommended by numerous health professionals, it was expected that balance performance following the exercise program would be better when compared to those preceding the four month exercise regime. Results showed that reaction time and COG excursion improved as expected.

The third study was designed to test a passive approach to injury prevention. The premise of this investigation was that velocities of fall activities differ from those of activities of daily living (ADL). Steps can be taken to minimize the occurrence of falls, but an elimination of falls is unlikely. A passive approach to reduce fall injury is necessary for those who choose not to involve themselves with exercise programs or who are incapable of strong motor control. The observations of ADL and fall velocities unveiled differences between the two types of activities, which suggests the feasibility of an airbag injury prevention system. To confirm the probable application and efficacy of airbags, however, future research needs to address the number of sensors needed to prevent accidental deployment, speed of deployment, airbag size and airbag shape. Improper size and shape could prevent damage to the protected area, but contribute to injury at other locations on the body (i.e. head and neck).

The three interrelated projects have provided insight into balance ability, the use of physical activity as an active intervention, as well as the hypothetical feasibility of a passive injury prevention technique. This research has also given insight into further research opportunities.

The first study using the DBTP was a new approach to understanding balance characteristics. Because of the novel approach, a number of questions should be addressed to establish it as a viable measuring tool. First, researchers need to address the natural validity of the measuring device. Dynamic environments that are encountered in daily life rarely resemble collapsing floors. Second, researchers need to determine if balance on the device is a learned skill and weather the amount of time given to leaning the skill for the older adult is sufficient. Third, industrial air cylinders are made to withstand weight as opposed to responding with the force exerted on it. The lack of responsiveness in the cylinders may have been a contributing factor to the lack of COG differences between age or gender cohorts. Fourth, the perturbation layer was established to move within a small increment of 2.5 cm in both directions. This was chosen as a protection to those who were not very stable such that a small incremental drop would be less fearful than one of greater amplitude. It may be argued that the 2.5cm distance may have not been large enough to project the differences between age/gender. Finally, there is also a need to look at joint control strategy in the hips, knees and ankles of the three age groups in order to give further insight to the method of reaction to external dynamics (Nardone et al, 2000).

The second study also had a few qualities that could be improved upon. One major oversight was to neglect retesting the control. Though improvements in the participants' reaction time and balance ability were noted, it is undetermined how those results would compare to the control group participants. It is possible that the participants learned the test tasks as opposed to demonstrating an intervention based improvement. As well, a more regulated strategy for the ball drop, such as a mechanical drop device, would offer more consistency as opposed to a manual release of the rope.

Studying the interaction of aging and balance is a complex matter as numerous factors contribute to the outcomes. The series of studies conducted here provide some further insight into critical issues such as balance measurement, control strategies, acquisition of targeted behaviours for fall prevention, and even the use of secondary or passive protective devices. It is hoped that the investigations of the suggested modifications to the DBTP could possibly assist in predicting individuals' likeliness to fall. The basic assumption continues to be that physical activity assists in improving and/or retaining strength and balance as well as having a positive effect on slowing sensory-motor degradation (Campbell et al, 1997; Rooks et al, 1997). Consequently, developing an effective tool for identifying balance related short-comings is an asset. Those more likely to fall may be prescribed various activities such as Fitball® classes. If exercise is not plausible, the individual may need a passive form of protection such as a wearable airbag.

Growing old brings many challenges. Postural instability and potential falls are just two of these challenges, which compromise a higher quality of life. In order to offer our aging generation a higher quality of life, it is in our best interest to understand why falls occur and what we can do to prevent them from occurring so that resultant injury or fatalities can be reduced.

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APPENDIX I: Consent Form Sample

Informed Consent

Fall Characterization, Prediction and Prevention in the Aging

Biomechanics Laboratory
University of Lethbridge

You are invited to participate in a research study performed by Brandie Dunn, of the Department of Kinesiology at the University of Lethbridge. The purpose of this study is to determine the characteristics of body sway to predict balance/posture control ability. The experiment takes about ½ hour and requires that you perform upright standing tasks on a platform. The results from this research will help community practitioners in the health care industry and hospitals acquire a means by which they can identify high-risk groups and reduce the likelihood of falls. As the population ages, falls become one of the major health problems, not only for those with some degree of balance or mobility impairment, but also among healthy active seniors.

Should you consent to participate in this study, you will be asked to stand upright on a platform (10” high) with eyes open. The area of the surface is 2’ X 2’ (60 cm X 60 cm) and each of the four corners will deviate in a vertical direction up to 1.5” (4 cm) in relation to your body sway. The tests are rarely related to any danger.

The only potential risk is that you may lose your balance. However, to reduce this risk, a safety harness and spotters will be used as a precautionary measure, so you will be absolutely safe.

The information gathered from you during this study is considered confidential, similar to a doctor’s records. To maximize your confidentiality, you will be assigned a code, and this code will be used instead of your name at all times. All personal information will remain locked in a file cabinet that can only be accessed by researchers involved in this study and will not be disclosed without your permission. We may, however, wish to use your measurement for a research presentation or educational purposes in the future. Your identity will be kept confidential. If you would like to give your permission at this time for use of measurement data for research and educational purposes, please place your initials by "yes". If you do not wish to give permission at this time, please place your initials by "no".

Yes______________________ No______________________

APPENDIX I: Consent Form Sample (cont.)

Your participation in this study is entirely voluntary and you may withdraw from participating at any time. Should you decide not to participate in this study, your relationship with the biomechanics Laboratory or any other department of the University of Lethbridge will not be affected in any way. If you have any further questions about this research, please feel free to contact Brandie Dunn, at 329-2111 ext.7563. If you have any further questions regarding your rights as a participant please contact the University of Lethbridge Office of Research Services at 329-2747.

Your signature below indicates that you have read and understood the information provided above, and that any and all questions you might ask have been answered to your satisfaction. Your signature also indicates that you willingly agree to participate in this study, you understand that you may withdraw from this experiment at any time and will not hold the University of Lethbridge, Kinesiology Department, Biomechanics Lab and/or individuals performing the research, liable for any injury that may occur relating to the studies.

I have read the attached Informed Consent form and I consent to participate in the "Fall Characterization, Prediction and Prevention in the Aging" research study.

Printed Name: ______________________________ Date: __________________

Signature:__________________________________

Witnessed by: ______________________________ Date:__________________

APPENDIX II: Protocol for - Platform

Name: ____________________

DOB: _____________________ Date: __________________

Weight: ___________________ PSI: _____________________

Height: ____________________ Shoulder Width: ___________

Mean Residual: ___________ Instructions: No bend in knees
Hands at side

Marker Placement Complete:

Consent Form Complete:

Start **Capture** *then* Lower **Actuators**

MOTO Pose	#	#

20 Seconds Each	1	2
Test	#	#
Eyes Open	#	#
Eyes Open	#	#
Eyes Closed	#	#
Eyes Closed	#	#

Measurements (mm)

Hand Thickness	____________________
Wrist Thickness	____________________
Elbow Width	____________________
Foot Thickness	____________________
Leg Length	____________________

Notes

APPENDIX III: Protocol for -Impact of Exercise on Senior's Motor Control Response to External Dynamics

Posture Change with Random Weight Drop

Name:

Date:

Age:

Video Accuracy:

Camera Set-Up:

Force Plate Working:

Moto-Pose(s):

Trial #:

Trial #:

2Lbs

Trial #

Drop 1:

Drop 2:

Drop 3:

Drop 4 :

Drop 5 :

5Lbs

Trial #

Drop 1:

Drop 2:

Drop 3:

Drop 4 :

Drop 5 :

Measurements

Height:

Weight:

Leg Length:

Wrist Thickness:

Wrist Width:

Elbow Width:

Foot Thickness

Notes:

APPENDIX IV: Protocol for - Fall Characteristics

Name: ________________________ Date: ______________

Height: ________________________ DOB: _______________

Weight: _______________________ Mean Residual: ______

Marker Placement: Y N Consent Form: Y N

Trial # Trial #

Moto Pose:

Trip Fall:

Fall Forward:

Fall Backward:

Fall Left:

Fall Right:

Fall Down:

Stair Climb:

Reach and Grab:

Turn, Sit and Stand:

Measurements (mm) Notes

Hand Thickness	__________________
Wrist Thickness	__________________
Elbow Width	__________________
Foot Thickness	__________________
Leg Length	__________________

APPENDIX V: Anatomical Locations for Marker Placement

Location	Anatomical Position	Number of Markers
HEAD	Left and Right Sphenoid	2
	Left and Right Back Head (Near the Lambdoid Suture)	2
TORSO	Sternal Notch	1
	Xiphoid Process	1
	C7	1
	T10	1
	Left Superior Angle of the Scapula	1
	Right Inferior Angle of the Scapula	1
ARMS	Left and Right Acromion Process	2
	Left and Right Upper Arm	2
	Left and Right Leteral Epicondyle of the Humerous	2
	Left and right Lower Arm	2
	Left and Right Stloid Process of the Radius	2
	Left and Tight Styloid Process of the Ulna	2
	Third Metacarpophalangeal Joint (Left and Right Hands)	2
PELVIC GIRDLE	Left and Right Anterior Superior Iliac Crest	2
	Left and Right Posterior Superior Iliac Crest	2
LEGS AND FEET	Left and Right Thigh	2
	Left and Right Leteral Condyle of the Tibia	2
	Left and Right Shank	2
	Left and Right Lateral Malleolous of the Fibula	2
	Left and Right Posterior Calcaneous	2
	Left and Right Tuberosity fo the Fifth Metatarsal	2
	Left and Right head of the Hallicus	2
TOTAL MARKERS		42

APPENDIX VI: Pressure Table (PSI)

Weight (Lbs)	Weight (Kg)	Test Pressure (PSI)
100	45.5	6.5
110	50.0	7.0
120	54.5	8.0
130	59.1	8.5
140	63.6	9.0
150	68.2	10.0
160	72.7	10.5
170	77.3	11.0
180	81.8	12.0
190	86.4	12.5
200	90.9	13.0
210	95.5	14.0
220	100.0	14.5
230	104.5	15.0
240	109.1	16.0
250	113.6	16.5
260	118.2	17.0
270	122.7	18.0
280	127.3	18.5
290	131.8	19.0