Age Related Changes in Balance and Gait

by

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ABSTRACT

Gait and balance disorders are the second leading cause of falls in the elderly. Investigating the changes in static and dynamic balance due to aging may provide a better understanding of the effects of aging on postural control system. Static and dynamic balances were evaluated in a total of 21 young (21-35 years) and 22 elderly (50-75 years) healthy subjects while they performed three different tasks: quiet standing, dynamic weight shifts, and over ground walking. During the quiet standing task, the subjects stood with their eyes open and eyes closed. When performing dynamic weight shifts task, subjects shifted their Center of Pressure (CoP) from the center target to outward targets and vice versa while following real-time feedback of their CoP. For over ground walking tasks, subjects performed Timed Up and Go test, tandem walking, and regular walking at their self-selected speed. Various quantitative balance and gait measures were obtained to evaluate the above respective balance and walking tasks. Total excursion, sway area, and mean frequency of CoP during quiet standing were found to be the most reliable and showed significant increase with age and absence of visual input. During dynamic shifts, elderly subjects exhibited higher initiation time, initiation path length, movement time, movement path length, and inaccuracy indicating deterioration in performance. Furthermore, the elderly walked with a shorter stride length, increased stride variability, with a greater turn and turn-to-sit durations. Significant correlations were also observed between measures derived from the different balance and gait tasks. Thus, it can be concluded that aging deteriorates the postural control system affecting static and dynamic balance and some of the alterations in CoP and gait measures may be considered as protective mechanisms to prevent loss of balance.

DEDICATION

I would like to dedicate this thesis to my mother, Latha and father, Balasubramanian for their unconditional love, support, and encouragement throughout my journey. I would also like to dedicate it to my sister Cheentu whose endless phone conversations have always lightened me up. I am grateful to my aunts and uncles in the US Prema, Venkat, Ramesh, and Shanthi, for providing valuable advice and making my transition to the US simpler.

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CHAPTER 1

INTRODUCTION

SIGNIFICANCE OF THE STUDY

According to the World Health Organization, 424,000 falls that occurred globally every year were fatal and more than 37.3 million falls led to injuries that required medical attention. Reports also suggest that the risk of falling increases with age and approximately one out of three persons older than 65 fall every year (Tinetti, Speechley, & Ginter, 1988).

Apart from reducing the comfort and quality of life, falls are also a big burden in healthcare. Injuries related to falls are one of the leading causes of hospitalizations among the elderly, leading to higher healthcare costs. The most common injuries related to falls include hip fractures, traumatic brain injuries and pneumonia (Hartholt et al., 2010). Even a mild fall can have serious consequences in the elderly due to their susceptibility to injury and age related physiological changes.

The causes of falls in the elderly are multi-fold and include visual disorders, lower extremity weakness, gait and balance disorders, confusion, dizziness, syncope etc. Amongst them, gait and balance disorders are the second major cause of falls, resulting in threefold increase in fall risk (Jeffrey M.Hausdorff, 2005). Since multiple causes lead to falling, there cannot be a single measure that is capable of detecting its risk. Thus studying effects of aging on gait and balance disorders, the most common cause of falls, is of primary importance.

In order to get a better insight into effect of aging on gait and balance, it is important to study the response of various physiological systems and mechanisms in controlling posture (Horak, 2006). A thorough understanding of such mechanisms and compensatory strategies can further improve the treatment for fall prevention.

POSTURAL CONTROL SYSTEM - OVERVIEW

Bipedal locomotion, one of the unique key traits of the human species is made possible by the highly evolved postural control system. The complex anatomical structure of humans, including the precise shape and positioning of the vertebral column enable us to expend relatively low levels of energy in musculoskeletal activities such as standing and walking (Skoyles, 2006). Posture is defined as the position of a body segment with respect to gravity. The two main goals of the control system are to achieve erect standing and locomotion, and to maintain it during activities of daily living (Skoyles, 2006).

The control of posture is termed as balance. Static balance is achieved when the projection of the Center of Mass (CoM) lies within the base of support. It exists only during quiet unperturbed standing. As a consequence of the accurate curves in the lumbar and thoracic bones, the CoM lies within the base of support formed by the feet ensuring static balance, during erect stance (Skoyles, 2006). However, any perturbation or voluntary movement shifts the CoM outside the base of support (Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). More complex mechanisms are adopted to achieve dynamic balance control. It is observed that the amount of time required to regain balance through feedback control from the brain is longer than the actual time for fall after CoM moves out of the base of support. This has led to the assumption that the postural control system is continually at play, to initiate control mechanisms well in advance (Morasso, Baratto, Capra, & Spada, 1999).

The postural control system is considered as a conglomeration of numerous complex sensorimotor processes. The deterioration of any one of the sensory or motor systems will have drastic effects on balance control. However, studies have shown the presence of redundancy in the posture control system, such that the loss of one of the sensory systems is compensated by the other systems (Winter, Patla, & Frank, 1990). The three main sensory processes responsible for balance are:

Visual System: Provides information about the location of surroundings and object with respect to the body.

Vestibular System: Senses position, linear accelerations, and rotational movements of the head.

Proprioception: Provides information about self, or the relative position of different parts of the body.

The sensory information from each of these systems is suitably weighted based on the environmental and physiological factors; and subsequently integrated to achieve balance control (Horak, 2006). The brain stem, cerebellum, and the cerebral cortex are the primary neuroanatomical structures that form the posture control system. Research on people suffering from balance and gait disorders due to Parkinson’s disease suggests that the dopaminergic pathway of the basal ganglia plays an important role in the integration and re-weighting of the sensory processes (Cham, Perera, Studenski, & Bohnen, 2007). Further, studies on animals and mammals indicate the importance of the cerebellum in coordination of the limb movements to achieve balance (Morton & Bastian, 2004).

Epidemiological evidence shows that more than 50% of the falls in the elderly occur during walking and activities of daily living (Barak, Wagenaar, & Holt, 2006). Thus, it is important to study the postural control system during standing as well as walking.

Usually, quantitative assessment of balance during quiet standing is performed using force platform which measures the forces and moments in three dimensions. Fall risk assessment in a clinical setting is carried out using the Limits of Stability (LoS) test and the timed-up and go test (Hirase, Inokuchi, Matsusaka, Nakahara, & Okita, 2014; Lemay et al., 2014; Salarian et al., 2010). Limits of stability characterizes the maximum distance to which the subject can lean (sway) without losing balance and the time taken to complete the task. On the other hand, Timed up and go task involves three phases of movement: sit-to-stand, walking and stand-to-sit. The total duration for completion of the task is taken as a balance assessment measure. Even though these tests assess the risk of falling, they provide limited information and do not quantify the cause or the underlying mechanism that might be affecting postural control.

PURPOSE OF THIS STUDY

The specific aims of this thesis are three fold.

1. To study the effects of age and visual input on quiet standing.
2. To study the effects of age in performing dynamic postural weight shifts.
3. To investigate the age-related changes in over ground gait patterns.

The purpose of the study is to determine the effects of aging on balance control. Many studies have looked at quiet standing and gait measures separately, in the elderly. However, there haven't been studies that have compared and correlated the effects of age on static and dynamic balance measures.

Studying the response of the body to absence of visual input during quiet standing will provide useful information on the type of compensatory mechanisms used by young and elderly during loss of visual information. Analysis of gait patterns during normal and tandem walking in the elderly will give insights into the effect of aging on locomotion and dynamic balance control.

The dynamic shift paradigm (dynamic postural weight shifts) utilized in this study is used to derive novel measures for improved balance evaluation. The task consists of a series of weight shifts in different directions and mimics posture shifts performed during some activities of daily living (ADL) such as reaching for an object on a shelf. This task challenges the balance control system and attempts to extract measures that will help in improved understanding of the response to such changes in balance and efficient characterization of the effects of aging on balance control. Correlations between the responses obtained from quiet standing, weight shifts, and gait will help in better interpretation of the overall changes in balance control with age.

In summary, this thesis will contribute additional information on age-related balance and gait deficits in able-bodied adults, to the already existing literature. This information will be useful in selecting the measures for fall-risk assessment in the elderly and in the design of rehabilitation procedures for fall prevention.

THESIS OUTLINE

This thesis is organized based on the different tasks performed by the subjects. Chapter 2 gives a detailed explanation about effect of age and visual input on quiet standing. Chapter 3 explains the effect of aging on posture shifts and Chapter 4 deals with effects of aging on gait. Chapter 5 discusses the correlations between the quiet standing, gait and posture shift measures, and their significance. Chapter 6 includes a discussion of the conclusions and potential applications.

CHAPTER 2

EFFECTS OF AGING ON QUIET STANDING

BACKGROUND

Quiet standing is a task that requires the person to hold the body upright in the absence of external perturbations. The position of the whole body Center of Mass (CoM) is a direct measure of balance. Since, determination of CoM requires anthropometric information about individual body segments and their positions; it is most often not used as a measure of balance. Instead, Center of pressure (CoP) is used as an alternative (Winter et al., 1990). CoP is defined as the position of the net ground reaction force derived from the two feet. The preference of CoP over CoM is due to its ease of measurement.

Currently, the standard instrument used for assessing balance is the force platform. The force plates provide information about the forces and moments exerted by the body in three dimensions. The position the CoP in the Anterior-Posterior (AP) and the Medio-Lateral (ML) planes can further be derived by calculating the moment arm.

Modeling the Postural Control System during Quiet stance. Various models of the postural control system have been proposed which include the contributions of the Central Nervous System (CNS), sensory, and motor systems.

Inverted Pendulum Model. Since the emergence of CoP as a measure for balance, scientists have worked extensively in deriving the relationship between CoP and CoM. The most simple and primitive model was demonstrated by Gage and Colleagues (1980). The main assumption of this model is that the entire body sways about the ankle as a single segment, in the AP direction. Thus, the body can be modeled as an inverted pendulum, with the ankle acting as a pivot. This model helps in deriving the relationship between whole body CoP and CoM (Gage, Winter, Frank, & Adkin, 2004).

In order to illustrate the model, consider the body to be swaying back and forth in the AP direction. When the CoP is ahead of the CoM, there is an increased forward sway due to a clockwise angular velocity ω. Similarly when the CoP is behind the CoM, there is an increased backward sway due to anticlockwise angular velocity. Assuming the inverted pendulum model, the difference between the clockwise and anticlockwise moments will equal the acceleration of the body CoM.

$$Rd - Wp = I\alpha \quad (1)$$

Where,

I is the moment of inertia

R is the vertical ground reaction force

W is the weight of the body

d and p represent the displacement of CoP and CoM, respectively

α is the angular acceleration of the inverted pendulum

The above equation can be used to deduce the angular acceleration of the body in response to the sway. If $Wp > Rd$, the body will accelerate in the forward direction and if $Wp < Rd$, the body accelerates in the backward direction.

Since R=W,

$$CoP - CoM = Kx \quad (2)$$

Where, x is the horizontal acceleration of CoM and K is proportionality constant.

Thus, according to the inverted pendulum model, the difference between CoP and CoM is directly proportional to the horizontal CoM acceleration, and both the parameters are negatively correlated. Hence, CoP-CoM can be considered as an error signal that is used to minimize the CoM acceleration by adjusting the position of CoP. From a controls system point of view, CoP is the controlling variable and CoM is the controlled variable. The range of sway of CoP is larger than CoM.

Although the inverted pendulum model is simple and easy to interpret, it does not consider the effect of movement in the other joints such as the hip and the trunk that might alter the CoP location. Furthermore, the validity of the inverted pendulum model holds good only in the AP direction, however, ML movements also exist. The ML CoP is found to be controlled majorly by the hip abductor/adductor muscles (Winter, Patla, Ishac, & Gage, 2003).

Various other models have since been proposed to relate the whole body CoM and CoP, and to characterize the CoP path during quiet standing. One such model is the two-segment double inverted pendulum model, which considers the contribution of the hip in CoP and CoM movements (Breniere & Ribreau, 1998). The study showed positive correlation between the CoP and CoM in the AP direction, indicating the two measures vary in phase with each other, whereas no such correlation was found in the ML direction.

Internal models of Quiet Stance. More complex models have emerged, that consider the amount of time required by the brain to process the sensory information and provide compensatory mechanisms to correct any internal or external fluctuations. A three linked model of standing was proposed, to provide the best possible estimate of the body's orientation in space, with the delayed information obtained from the sensory systems, based on optimal estimation theory (van der Kooij, Jacobs, Koopman, & Grootenboer, 1999).

Researchers have also proposed an internal model for balance control system, similar to the internal models proposed for explaining motor control tasks. Since the act of standing is controlled by sensory systems that have an inherent delay associated with them, the central nervous system must possess an internal model to predict the anticipatory actions required to maintain balance (Morasso et al., 1999). This was proven by illustrating the phase lock between CoP and CoM.

According to these models, the postural control system acts in two different modes: Reflex and Anticipatory. The anticipatory mode is functional during unperturbed quiet standing, where in the control system stabilizes the postural system well in advance, through anticipation of external or voluntary disturbances. The reflex mode, also known as the feedback mode, controls balance in response to perturbations (Deliagina, Orlovsky, Zelenin, & Beloozerova, 2006).

All the aforementioned models indicate the presence of an inbuilt anticipatory system in play that helps to maintain balance control even in the absence of external disturbances.

Center of Pressure Analysis. Center of Pressure obtained from the force platforms is one of the most widely used parameters for studying quiet standing. The normal posturographic technique involves the subject to stand quietly on the force platform with their eyes open, maintaining an erect position, for a defined period of time. The resultant plot obtained depicts the CoP trajectory in the AP and ML directions. This plot is known as the stabilogram.

The analysis of CoP has been carried out either using the AP and the ML time series separately, or by combining the AP and ML displacements, to obtain a planar time series signal of the resultant CoP.

The analysis of CoP time series can be broadly classified into two (Norris, Marsh, Smith, Kohut, & Miller, 2005):

1. Traditional Analysis
2. Statistical mechanics

Traditional Analysis. Traditional CoP analysis involves deriving measures from the CoP trajectory such as the total excursion, mean velocity, mean frequency, etc. For such methods, CoP is assumed to be a stationary time series (Norris et al., 2005). Several studies used traditional analysis to detect differences in quiet standing between young, elderly, and in neurodegenerative disorders like Parkinson's disease (Ickenstein et al., 2012; Prieto, Myklebust, Hoffmann, Lovett, & Myklebust, 1996).

Some of the main parameters extracted from the CoP trajectory are (Ickenstein et al., 2012):

1. Total Path length: Total length covered by the CoP path in the AP and ML directions.

2. Sway Area: Total area enclosed by the CoP trajectory per unit time

3. Mean Velocity: Total distance covered by CoP in the AP and ML directions per unit time

4. Mean distance: Distance vector from the mean CoP position in AP and ML directions.

Traditional methods of analysis of posturography are the simplest ways to study balance during quiet stance. These methods have also been used to study the effects of the three sensory mechanisms separately. For example, the differences in measures obtained from CoP trajectories during eyes open and eyes closed conditions provide insight into the effect of visual input on static balance..

Several studies have used traditional methods to study the effects of loss of sensory feedback. Alahmari et.al. (2014) studied the differences in CoP sway area and velocity in subjects with vestibular disorders. Moghadam et al. (2011) group compared CoP sway when the subject stood on a foam pad with eyes open and eyes closed. Romberg ratio is another important parameter used in traditional analysis to quantify the effect of visual information. Romberg ratio is defined as the ratio of a particular measure during eyes closed condition to the measure during eyes open condition (Fujita et al., 2005).

Many research groups questioned the reliability of measures obtained from CoP trajectory, quoting that such parameters vary across subjects and are also dependent on the time and frequency of acquisition (Doyle, Hsiao-Wecksler, Ragan, & Rosengren, 2007; Lacour, Bernard-Demanze, & Dumitrescu, 2008; Panzer, Bandinelli, & Hallett, 1995). On the other hand, there have been studies validating the test-retest reliability of parameters such as mean velocity and total path length, with very high intra class correlations (Scoppa, Capra, Gallamini, & Shiffer, 2013; Swanenburg, de Bruin, Favero, Uebelhart, & Mulder, 2008).

Statistical Mechanics. Statistical mechanics techniques analyze the fractal and evolutionary properties of the CoP time series. Such techniques can be used even if the CoP time series is non-stationary. The main reason to apply statistical mechanics techniques is the difficulty to interpret the underlying neural mechanisms from the results of traditional analysis (Slomka, Juras, Sobota, & Bacik, 2013).

One of the more popular methods is the Detrended Fluctuation Analysis (DFA). DFA attempts to detect any long term correlations in the CoP time series, that might indicate the presence of a memory component in CoP (Doyle et al., 2007). Briefly, DFA involves calculation of root mean square fluctuations of integrated and detrended time series of different time scales. The slope of logarithmic plot of fluctuations vs. time scale is alpha. The value of alpha indicates whether the correlations are positive or negative. Generally, DFA analysis of CoP time series during quiet standing consists of two distinct parts: A persistent high frequency region in the short-range time scale and an anti-persistent low frequency region in long-range time scale (Teresa Blázquez, Anguiano, de Saavedra, Lallena, & Carpena, 2009)

The results of DFA have also been supported by random walk analysis, commonly known as stabilogram diffusion analysis. The CoP trajectory is assumed to mimic Brownian motion. The random walk analysis also showed short term correlations in the CoP time series. These short term correlations have been attributed to an open loop control mechanism and the long term anti correlations to closed loop mechanisms of balance control (Collins & De Luca, 1994).

Quiet Standing Balance Control in the Elderly. Studies have shown that a decline of the dopaminergic system affects the integration and re-weighting of the sensory inputs necessary for balance in the basal ganglia. Dopaminergic depletion is found with aging, although not as severe as in Parkinson's disease (Cham, Perera, Studenski, & Bohnen, 2007).

This degeneration of the postural control system with age has inspired researchers to study quiet standing in elderly. Aging has a profound impact on balance control, making the elderly more prone to falls. It is characterized by deterioration of sensory systems, loss of motor units and decreased muscle strength (Abrahamova & Hlavacka, 2008).

Of the two kinds of strategies (ankle strategy and hip strategy) adopted to maintain stance when perturbed, the able-bodied elderly are found to use more of the hip strategy due to inadequate torque production in the ankle muscles. Able-bodied young adults adopt the ankle strategy which involves the swaying of the body as a single segment about the ankle. The hip strategy on the other hand involves movement around the hip. Elderly show higher hip EMG activity and joint displacements compared to the young (Amiridis, Hatzitaki, & Arabatzi, 2003).

Numerous studies have been carried out in the past, mostly using traditional methods of analysis of the CoP to compare young and healthy elderly. Apart from providing insights into aging and falls, these studies are also useful in separating the effects of aging and neurodegenerative disorders.

The results of the traditional methods show that mean displacement and velocity of CoP are consistent within an age group and different between young and elderly (Ickenstein et al., 2012; Prieto et al., 1996; Raymakers, Samson, & Verhaar, 2005); While other studies show increased sway frequencies (Vieira Tde, de Oliveira, & Nadal, 2009). Moghadam et. al. (2011) also found attention to be an important factor affecting the sway in elderly. This was evident from dual task experiments during quiet standing.

The effect of different sensory systems has also been studied in the elderly. The reliance on visual information for balance control tends to increase with age. Hip proprioception was also reduced in elderly and mid-aged adults, although this did not impact the CoP sway measures (Wingert, Welder, & Foo, 2014). Greater trunk sway, especially in the AP direction was found when the subjects stood on foam (Alahmari et al., 2014). All the above studies support the notion that elderly people rely more on sensory feedback and there is a loss of effective re-weighting of such inputs (Eikema, Hatzitaki, Tzovaras, & Papaxanthis, 2012).

The increase in CoP sway under different sensory conditions and during quiet standing in the elderly is also ascribed to increase in ankle stiffness (Cenciarini, Loughlin, Sparto, & Redfern, 2010; Lauk et al., 1998; Winter, Patla, Rietdyk, & Ishac, 2001). Elderly tend to increase co-contraction of the muscles. One study on CoP sway in the elderly during floor tilts showed increases in ankle stiffness (Cenciarini, Loughlin, Sparto, & Redfern, 2009). It is hypothesized that unreliable sensory information is compensated by increases in ankle stiffness. However, there has been controversy regarding the methodology followed in deriving ankle stiffness (Loram & Lakie, 2002; Morasso & Sanguineti, 2002).

Stabilogram diffusion analysis revealed that elderly adults used closed loop control mechanisms, but with a larger delay than the young adults. This is accounted for by increased reflex time, reduced muscular strength, and sensory perception (Lacour et al., 2008).

Thus, a wide range of literature is available on the effect of aging on balance control during quiet standing. The traditional measures are validated and reliable. The statistical mechanics techniques on the other hand provide additional information on postural control mechanisms.

METHODOLOGY

Subjects. Recruitment for the study was facilitated by displaying flyers in public bulletins, campus bulletins, and senior centers and was carried out under the Arizona State University Institutional Review Board (ASU IRB) approved study titled "Control of Posture and Walking in Able-Bodied Adults".

Recruitment for the study was based on the following inclusion and exclusion criteria: Inclusion criteria: Subjects between the ages 18 and 75, who were able to understand the instructions and willing to sign the informed consent to participate in the study were included.

Exclusion criteria: Subjects with neurological or orthopedic disorders that would affect their balance and walking were excluded. Subjects were also excluded if they had any of the following conditions: Congenital heart disorders, implanted device such as an orthopedic device or pacemaker, heart attack or stroke, heart palpitations, psychological disorders, respiratory problems such as asthma, arthritis or excessive soreness of joints, injuries related to fractures, or joint dislocation or torn ligaments.

A total of 43 subjects met the criteria and participated in the study. To investigate the effects of aging, subjects who were less than 30 years old were assigned to young group and subjects who were older than 50 are assigned to elderly group. 21 of the subjects (12 female, 9 male) fell in the young category (21-

35 years) and had a mean age of 23.0 ± 3.8 years. 22 of the subjects (12 female, 10 male) fell in the elderly category (50-75 years) with a mean age of 62.7 ± 8.5 years at the time of enrollment.

Experiment Protocol. All data were collected in a single session at the Center for Adaptive Neural Systems, Arizona State University (ASU), Tempe, AZ. The subjects were initially briefed about the study and the contents of the informed consent. The subjects then expressed their willingness to participate in the study by signing the informed consent. The subject's blood pressure and pulse rate were tested to confirm there was no risk associated with them participating in the study.

The force platform was warmed up by switching it on 30 minutes prior to data collection. During the quiet standing task, the subjects were instructed to stand on the force platform with their arms by their side, and their feet separated by hip-width. A trace of the subject's feet was then taken to ensure that the same position is maintained for all the subsequent trials. The quiet standing task involved two different conditions: standing with eyes open and standing with eyes closed. During the eyes open trials, the subject was instructed to stand as still as possible, concentrating on a point ahead of them. In case of any disturbance such as coughing or talking, the trial was repeated. During the eyes closed trials, subjects stood as still as possible, with their eyes closed.

A total of eight trials were recorded: five trials with eyes open followed by three trials with eyes closed. Each trial was for 60 seconds, with sufficient rest periods between the trials.

Force Platform and Data Collection Setup. Data was collected with a sampling frequency of 100 Hz, using a Bertec force plate. The plate had a dimension of 600 X 400 mm, with a resolution of 0.2 % of full scale. The plate consists of load transducers that are capable of measuring three components of forces and three components of moments in the orthogonal coordinate system. The 16 bit digital signal from the force plate was passed through AM6501, an A/D converter which has built-in amplification. The gain was set to 1. The analog signal from the amplifier was then fed into LabVIEW 8.0 using BNC 2115 for calculating CoP.

Center of Pressure Calculation. The forces and moments in three dimensions (Fx, Fy, Fz, Mx, My, Mz) were determined using the calibration matrix provided with the Force plate using the formula

$$Fx = C_1S_1 \quad (3)$$

$$Fy = C_2S_2$$

$$Fz = C_3S_3$$

$$Mx = C_4S_4$$

$$My = C_5S_5$$

$$Mz = C_6S_6$$

Where C represents the calibration matrix and S the scale factor for unity gain.

The coordinate system for the force plate is defined as: Positive Y axis directing forward, X-Axis to the left and Z axis downwards according to right hand rule.

CoP represents the X and Y coordinates of the point of application of the net ground reaction force. Using the relationship between the force and moment arm,

$$X_p = \frac{-Fx-My}{Fz} \quad (4)$$

$$Y_p = \frac{-Fy-Mx}{Fz} \quad (5)$$

Where X_p and Y_p are the ML and AP coordinates of the CoP with respect to the force plate coordinates, in meters. All the above calculations were performed in LabVIEW 8.0 and the resultant outputs include Fx, Fy, Fz, Mx, My, Mz ; X_p and Y_p.

Data Analysis. CoP preprocessing and analysis was performed using MATLAB 2013. Since most of the CoP frequencies lie within 3 Hz, the AP and ML CoP were filtered using a 4^{th} order Butterworth low-pass filter with zero phase shift, with a cut-off of 10 Hz.

Analysis was performed on the CoP data to derive various measures that would reflect the features of the stabilogram. CoP AP, ML and planar data were used to calculate the following measures.

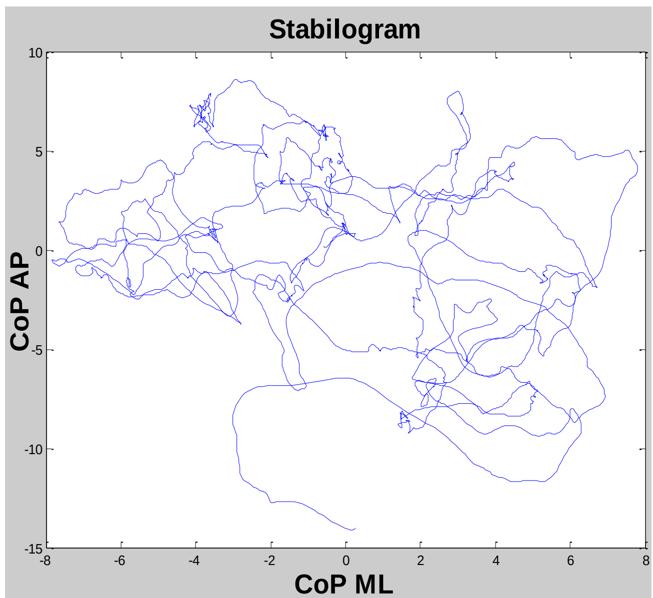

Figure 2.1. Stabilogram shows the anterio-posterior (AP) and medio-lateral (ML) displacements of CoP. All measurements are in millimeters. The distance and area measures are derived from the stabilogram.

Sway Area. Sway Area is a hybrid measure and is defined as the area enclosed by the CoP path per unit time. It is dependent on distance of the current position of CoP from the mean CoP and the distance travelled by the CoP.

$$\text{Area of Sway} = \frac{1}{2T}\sum_{n=1}^{N-1} |AP[n+1]ML[n] - AP[n]ML[n+1]| \quad (6)$$

Where N is the total number of data points

T is the time period of analysis (60 sec);

AP[n] =AP (n)-meanAP;

ML[n] =ML (n)-meanML

Resultant Distance (RD). The resultant distance is a time series representing the distance of the current points of AP and ML from the mean CoP Position.

$$RD[n] = [AP[n]^2 + ML[n]^2]^{1/2} \tag{7}$$

Mean Distance. Mean distance is defined as the mean of the RD time series. In other words, it represents the average distance from the mean CoP

$$MDIST = 1/N \sum_{n=1}^{N} RD[n] \tag{8}$$

Mean distance AP is the average distance of AP from mean CoP.

$$MDIST_{AP}= 1/N \sum_{n=1}^{N} AP[n] \tag{9}$$

RMS Distance. RDIST is the root mean square (RMS) distance of the resultant distance time series.

$$RDIST= [1/N \sum_{n=1}^{N} RD[n]^2]^{1/2} \tag{10}$$

Similarly $RDIST_{AP}$ is the RMS distance of AP from mean CoP (Standard Deviation)

$$RDIST_{AP} = [1/N \sum_{n=1}^{N} AP[n]^2]^{1/2} \tag{11}$$

Total Excursion. The total length of the CoP path approximated as sum of distances between consecutive points in the CoP time series

$$TOTEX = \sum_{n=1}^{N-1} \sqrt{[(AP(n+1)-AP(n))^2 + (ML(n+1)-ML(n))^2]} \tag{12}$$

Similarly the total excursion in the AP direction is the total CoP path covered in the AP directions as an approximation of sum of distances between consecutive points

$$TOTEX_{AP} = \sum_{n=1}^{N-1} | AP[n+1]-AP[n] | \tag{13}$$

Mean Velocity. The mean velocity of CoP is defined as the total distance covered, TOTEX over total time.

$$MVELO = TOTEX/T \quad (14)$$

Similarly the velocity in the AP is the total path length covered in AP direction over time

$$MVELO_{AP} = TOTEX/T \quad (15)$$

***95% Confidence Circle Area*.** The 95% confidence area is a circle of radius equivalent to 95% confidence limit of the resultant distance time series, assuming it is a normal distribution

$$AREA\text{-}CC = \pi\ (MDIST + z_{0.5}S_{RD})^2 \quad (16)$$

Where, S_{RD} is the standard deviation of the RD time series.

$z_{0.5}$ is the z statistic of the 95% confidence limit

95% Confidence Ellipse Area. Similar to AREA-CC, AREA-CE is the area of the bivariate ellipse which encloses 95% of the points of the CoP path. Let a and b be the major and minor axes of the ellipse.

Assuming the sample size is large,

$$a= [F_{0.05[2,n-2]}(S_{AP}^2+S_{ML}^2+D)]^{1/2} \quad (17)$$

$$b= [F_{0.05[2,n-2]}(S_{AP}^2+S_{ML}^2-D)]^{1/2}$$

F is the F statistic of 95% confidence level. For large sample size, F is approximated to 3. S is the standard deviation of AP and ML time series.

$D= [(S_{AP}^2+S_{ML}^2) - 4(S_{AP}^2S_{ML}^2 - S^2_{APML}]^{1/2}$

And $S_{APML} = 1/N\ \Sigma\ AP[n]\ ML[n]$

Finally, $$AREA\text{-}CE = 2\pi F_{0.05\ [2,\ n-2]}[\ S_{AP}^2S_{ML}^2 - S^2_{APML}]^{1/2} \quad (18)$$

Mean Frequency. Mean frequency is defined as the rotational frequency that the CoP travelled around a circle, with a radius equal to the mean distance

$$MFREQ = \frac{TOTEX}{2\pi MDIST} \quad (19)$$

Similarly the mean frequency in AP is calculated as

$$MFREQ_{AP} = \frac{TOTEX\ AP}{4\sqrt{2}MDIST\ AP*T} \tag{20}$$

Similar measures were derived for ML direction also.

Statistical Analysis. Statistical analyses were performed in IBM SPSS 22 (SPSS Inc., Chicago,IL) . The test chosen was based on the two main questions to be answered from the quiet standing measures: to determine the effect of age on quiet standing and to find the effect of absence of visual input on quiet standing.

Repeated measures Analysis of Variance. A simple independent sample t-test to find effect of aging and a paired sample t-test to find the effect of visual input are insufficient. This is because; such tests do not take the inter-trial and inter-subject variability into account. Thus, a general linear model was created using repeated measures Analysis of Variance (ANOVA). The advantage of this analysis is that it also considers the within-subject factor variability when calculating the ANOVA measure for age.

Each of the measures calculated from the CoP time series were considered as independent response variables. So, the repeated measures ANOVA was performed on each response separately. Since it is necessary to have equal number of trials for the two visual conditions, only three trials of the eyes open and eyes closed conditions were taken into consideration. The two within-subject factors were trials (3 levels) and visual conditions (2 levels-eyes open, eyes closed). The between subject factor was age (2 levels-young, elderly). Thus, this algorithm will take the effects of trial, visual condition, and the interaction between the two factors into account, in addition to age. A p-value of less than 0.05 was considered significant. Although repeated measures ANOVA with age as the between subject factor gives the effects of visual input, it does not tell us which if either one of the groups showed significance for vision or both the groups showed significance.

In order to answer the second question, the effect of change in visual conditions within each group, repeated measures ANOVA was performed on each age group separately. In this case, age is a constant and vision is the between subject factor and trial is the within subject factor. The normality of the response was determined by looking at the Quantile-Quantile (Q-Q) plot. In case of a right skewed distribution, log transformation was performed on the response distribution to make it normal. The repeated measures ANOVA was performed after the transformation.

Test-retest reliability. The test-retest reliability is very important to get a better understanding of the consistency of the data across trials. Test-retest reliability analysis was performed by calculating the Cronbach's Alpha and Intra-Class Correlation (ICC). The Cronbach's alpha value tells us if the data is internally consistent and reliable. The alpha value ranges between 0 and 1. Values of alpha greater than 0.8 indicate good consistency. ICC is a measure of reproducibility of the data. Unlike other correlation measures, ICC determines the correlation within a group instead of pairwise comparison. ICC is often used to determine if a single trial is sufficient to get a consistent result. ICC values greater than 0.8 indicate good reproducibility and correlations between the responses within each subject.

RESULTS

Effects of age on quiet standing. Repeated measures ANOVA was performed on the quiet standing data for all the 19 responses that were derived. The QQ plots indicated that the distributions of all the parameters were right skewed. Thus a log transformation was performed prior to the ANOVA test. Young subjects 1 and 3 were eliminated from the quiet standing analysis since the quality of the data was poor due to technical difficulties.

Out of the 19 quiet standing measures, nine of them showed significant differences between young and elderly groups: Sway area, Total excursion AP, Total excursion ML, Mean velocity, Mean velocity AP, Mean velocity ML, Mean frequency, and Mean frequency ML. Note that mean velocity is derived from total excursion, and thus both of them gave the same statistical results. Table 2.1 shows the F and p values for each of the significant parameters. Refer to appendix B for mean and standard deviation values of all the measures derived from CoP. Sway area and mean frequency showed significant trial to trial variations within subjects with F values of 5.552 (p=0.006) and 13.565 (p=0.001) respectively.

Table 2.1

Quiet standing measures that showed significant difference between young and elderly. F and p values from one way ANOVA are provided. p< 0.05 is considered significant.

PARAMETER	F VALUE	p -VALUE
Sway Area	10.329	0.003
Total excursion	17.670	< 0.001
Total excursion ML	6.811	< 0.001
Total excursion AP	20.894	< 0.001
Mean frequency	12.808	0.001
Mean Frequency AP	20.894	< 0.001
Mean Velocity	17.670	< 0.001
Mean Velocity ML	6.811	< 0.001
Mean Velocity AP	20.894	< 0.001

Effects of visual information on quiet standing. Repeated measures ANOVA was performed on the responses of each of the age groups to determine the effect of absence of visual input on quiet standing. The young and the elderly groups showed significant differences in the following measures: Sway area, Total excursion, Total excursion ML, Mean velocity, Mean velocity AP, Mean velocity ML, Mean distance ML, RMS distance ML , 95% confidence area circle, and 95% confidence area ellipse. In addition, the elderly showed significant differences in Total excursion AP. The p-values for each of these measures are provided in table 2.2. It is observed that absence of visual input tends to affect the time domain CoP measures specifically in the AP direction. Thus, both the young and elderly increase the excursion in the AP direction in the absence of visual feedback. In addition, the elderly also showed increase in total excursion in the ML direction.

Table 2.2

Quiet standing measures that showed significant difference between eyes open and eyes closed conditions within young and elderly groups. F and p-values from one way ANOVA are provided. p-value < 0.05 is considered significant. Blank table cells indicate that the measure was not significant.

PARAMETER	F value young	p-value young	F value Elderly	p-value Elderly
Sway area	7.882	0.008	10.355	0.003
Total excursion	14.418	0.001	11.518	0.002
Total excursion ML			6.662	0.014
Total excursion AP	20.087	< 0.001	12.377	0.001
Distance AP	4.633	0.038	7.223	0.010
Area CE	4.538	0.040	5.083	0.031
Area CC	6.205	0.017	7.701	0.008
RMS distance AP	5.591	0.024	7.284	0.010

The interaction between visual input and age was significant in total excursion ML. This indicates that young and elderly respond differently in the absence of visual input. It was observed that the there is a larger increase in the CoP parameters upon closing the eyes in the elderly when compared to young. This shows that the elderly rely more on visual information for controlling CoP. Romberg ratio, did not show any significant differences between young and the elderly. However, the Romberg ratio was higher in the elderly.

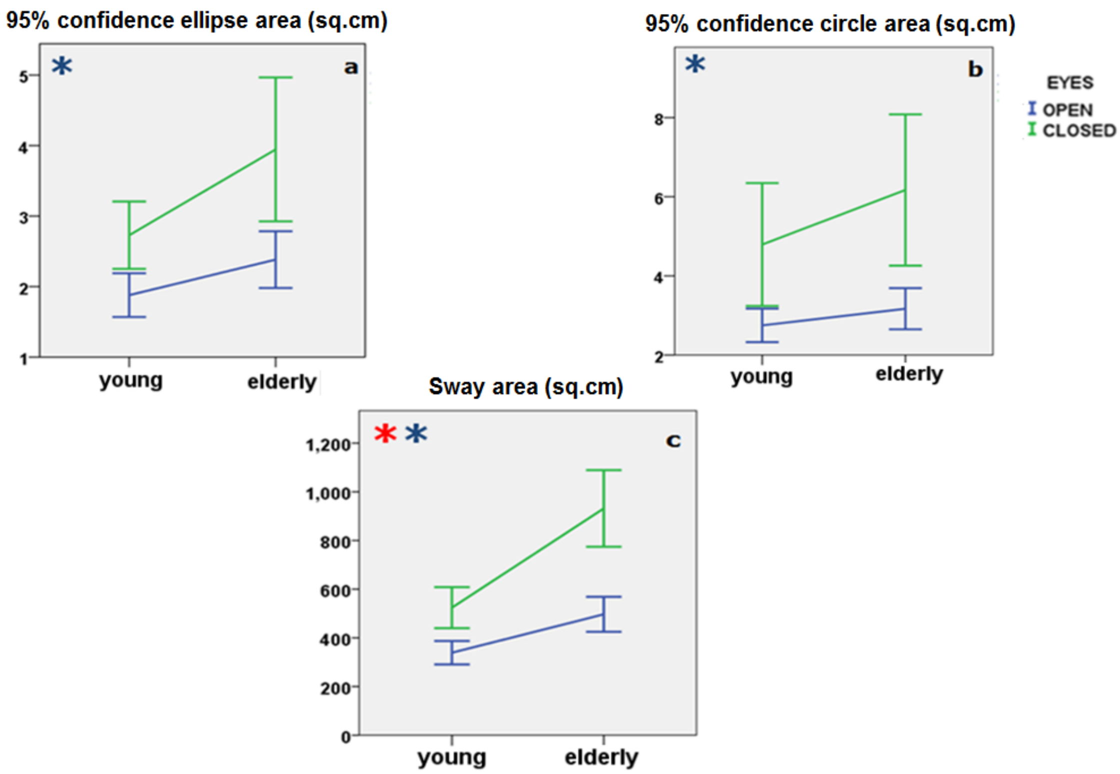

Figure 2.2. Change in time domain area measures with increase in age and absence of visual input: (a) change in 95% confidence area ellipse, (b) change in 95% confidence area circle, and (c) change in sway area. Blue color asterisk denotes significant difference due to visual input and red color asterisk denotes significant difference due to aging at $p < 0.05$. The vertical lines denote the 95% confidence intervals for the mean values.

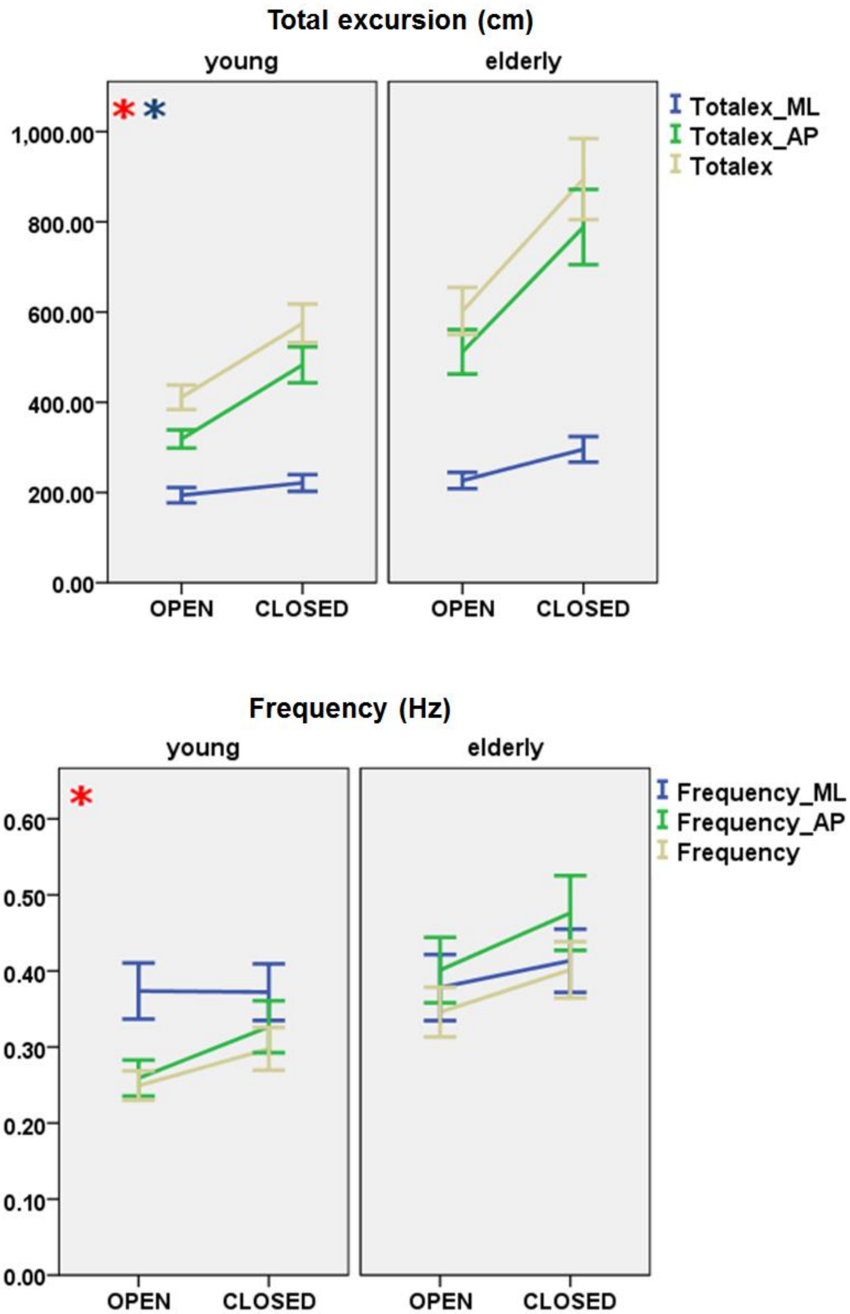

Figure 2.3. Change in time domain distance measures in the AP, ML, and planar directions, with increase in age and absence of visual input: change in total excursions due to visual input in the (a) young group and (b) elderly group. The changes in mean frequency due to visual input are shown for (c) young group and (d) elderly group. Red asterisk denotes significant difference between young and elderly groups ($p < 0.05$) and blue asterisk denotes significant difference due to visual input. The vertical lines denote the 95% confidence intervals for the mean values.

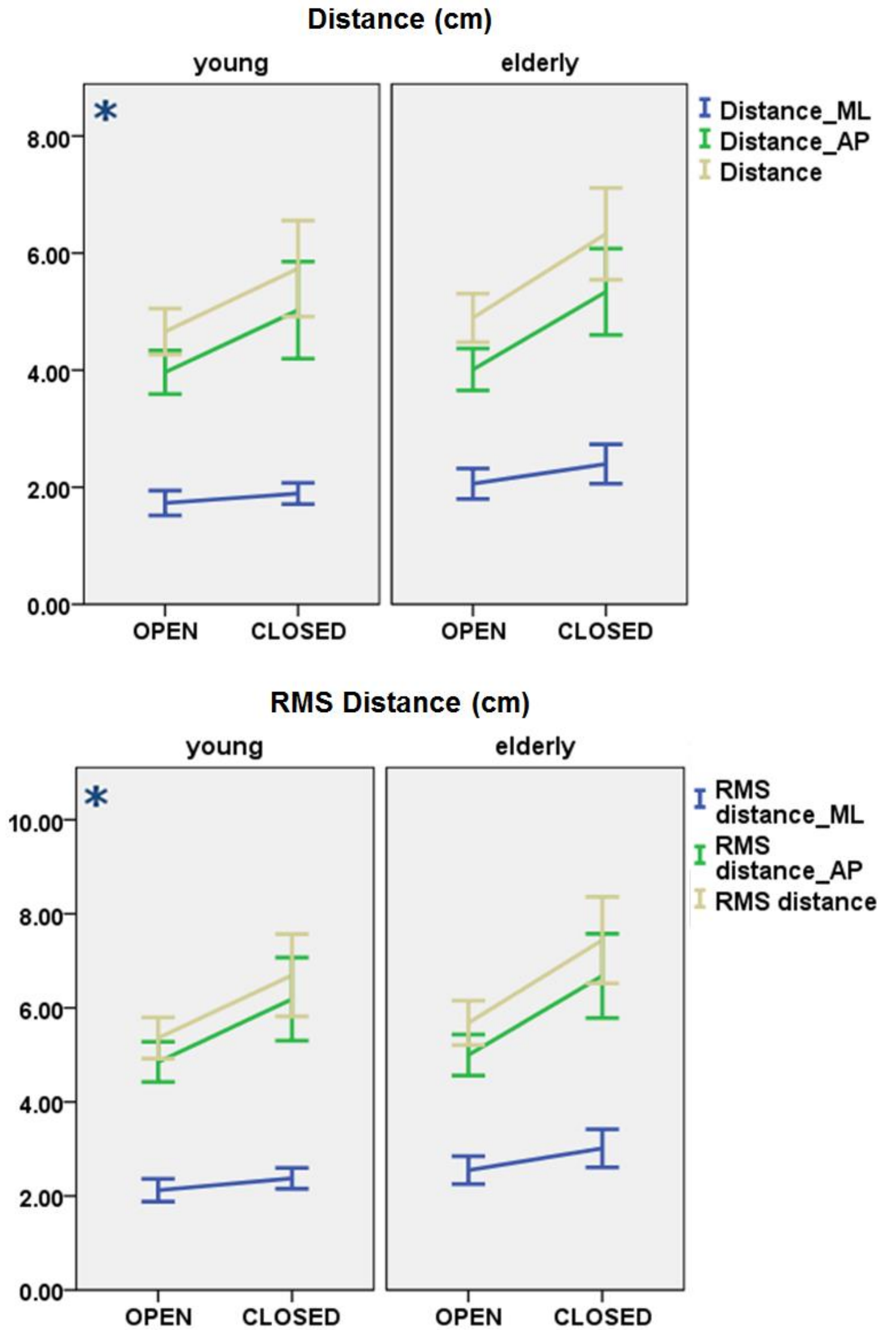

Figure 2.4. Change in time domain distance measures in the AP, ML, and planar directions, with increase in age and absence of visual input: (a) change in RMS distance due to visual input in the (a) young group and (b) elderly group. The changes in mean distance due to visual input are shown for (c) young group and (d) elderly group. Blue asterisk denotes significant difference ($p < 0.05$) due to visual input. The vertical lines denote the 95% confidence intervals for the mean values.

Test-retest reliability. Intraclass correlation coefficient and Cronbach's alpha were calculated for all the responses (Table 2.3, 2.4). The alpha values for all the parameters except mean distance and 95% confidence area ellipse were greater than 0.8. However, the ICC values were greater than 0.8 only for total excursion and mean velocity. The value of alpha indicates that all the measures except area ellipse and mean distance are internally consistent across trials. ICC shows that a single trial is sufficient to determine the differences across age groups for total excursion and mean velocity.

Table 2.3

Test-retest reliability measures Cronbach's alpha and Intra-class correlation coefficient (ICC) for different quiet standing balance measures during eyes open and eyes closed conditions in the young group. Alpha and ICC values greater than 0.8 indicate good consistency across trials. Abbreviations of the measures are explained under the section 'Data Analysis'.

YOUNG EYES OPEN			**YOUNG EYES CLOSED**		
Measure	**Alpha**	**ICC**	**Measure**	**Alpha**	**ICC**
SWAY	**0.870**	0.557	SWAY	0.627	0.359
TOTALEX	**0.950**	0.786	TOTALEX	**0.926**	**0.805**
RMSRD	**0.820**	0.445	RMSRD	0.344	0.154
VEL	**0.950**	0.786	VEL	**0.926**	**0.805**
AREACE	**0.808**	0.436	AREACE	0.610	0.346
AREA CC	0.791	0.401	AREA CC	0.129	0.050
DIST	0.796	0.411	DIST	0.328	0.146
FREQ	**0.807**	0.45	FREQ	**0.853**	0.654

Table 2.4

Test-retest reliability measures Cronbach's alpha and Intra-class correlation coefficient (ICC) for different quiet standing balance indices during eyes open and eyes closed conditions in the elderly group. Alpha and ICC values greater than 0.8 indicate good consistency across trials. Abbreviations of the measures are explained under the section 'Data Analysis'.

ELDERLY EYES OPEN			**ELDERLY EYES CLOSED**		
Measure	**Alpha**	**ICC**	**Measure**	**Alpha**	**ICC**
SWAY	**0.873**	0.546	SWAY	**0.919**	0.775
TOTALEX	**0.960**	**0.822**	TOTALEX	**0.971**	**0.896**
RMSRD	**0.861**	0.520	RMSRD	**0.819**	0.605
VEL	**0.960**	**0.822**	VEL	**0.971**	**0.896**
AREACE	**0.888**	0.578	AREACE	**0.843**	0.646
AREA CC	**0.804**	0.412	AREA CC	0.692	0.440
DIST	**0.864**	0.531	DIST	**0.814**	0.593
FREQ	**0.939**	0.715	FREQ	**0.879**	0.701

DISCUSSION

The effect of aging and visual input on balance control was studied. The results of the quiet standing task suggest that both age and visual information are important factors that alter balance control. Sway area, mean velocity and mean frequency are the three main measures affected by age. Although sway area and mean frequency showed significant inter-trial variability within the groups, this variability was taken into consideration for the ANOVA. Despite the variability, the effect of age was still significant on sway area and mean frequency. All the above parameters that increased with age are time domain measures. This indicates that the elderly sway their center of pressure to a larger extent in order to keep the CoM within the base of support. This could also be due to muscle weakness and increased ankle stiffness in the elderly as suggested by previous studies (Halliday, Winter, Frank, Patla, & Prince, 1998, Cenciarini, Loughlin, Sparto, & Redfern, 2010; Lauk et al., 1998; Winter, Patla, Rietdyk, & Ishac, 2001)

The quiet standing trials during eyes open and eyes closed conditions were not randomized and the eyes closed trials always followed the eyes open trials. There was a possibility that fatigue might have some effect on the measures obtained from eyes closed trials. Sufficient rest periods were provided between trials of each type and we believe that this might have minimized the effect of fatigue. Sway, mean velocity, mean frequency, and mean distance were affected by absence of visual input, especially in the AP direction, in both the young and elderly groups. In addition, the Romberg ratio for elderly was higher, implying that elderly rely more on visual input. One postulation is that the young compensate for the loss of visual input using the redundancy in the postural control system. However, the elderly population seems to rely more on visual input. This might also indicate abnormal re-weighting of the sensory inputs or deterioration in other sensory systems with age.

There have been many studies on quiet standing specifically on the effects of aging. Prieto et al (1996) performed a similar experiment to determine the effect of aging and visual input on quiet standing. They derived the CoP parameters from 20 young and 20 elderly subjects during eyes open and eyes closed conditions. According to their results, significant differences were found between age groups in mean velocity, mean velocity AP and mean frequency, mean frequency AP. The present study showed age related differences in Sway area and mean velocity in the ML direction in addition to the results obtained by Prieto. As in Prieto's study, the differences between eyes open and eyes closed conditions were statistically stronger in elderly. Also, the Romberg ratio although higher in the elderly, was not statistically significant. Overall, their study found mean velocity to be the only measure to show significant differences for changes in visual input and age. The current study found mean velocity and sway area to show differences with visual input and age. Since the mean velocity is derived from total excursion, both the measures show similar statistical differences between young and elderly.

Another study by Ickenstein and colleagues (2012) looked at the effects of aging and Parkinson's disease on quiet standing. The authors analyzed quiet standing during eyes open and eyes closed conditions in 10 elderly and 21 young subjects. Mean radius, sway area, and mean velocity were calculated. The current study agrees with the results of Ickenstein's study that the mean speed shows significant differences across age. However, aging effects were only evident during eyes closed conditions. It is important to note that only two trials were conducted in their study. Moreover, the number of subjects in the elderly group was smaller than the number of subjects in the young group and the subjects stood with their arms extended outwards, for 30 seconds.

Alahmari et.al (2014) studied 30 young and 30 elderly subjects with different visual conditions such as eyes open, eyes closed, and variations in visual surround. A balance rehabilitation Unit (BRU) was used to determine the 95 % confidence area ellipse and mean velocity from the CoP data. The result of the present study is in accordance with the results obtained by them. The authors showed significant differences in sway area and mean velocity across age groups. ICC for area and velocity was at least 0.76. Our study showed high ICC only for mean velocity and total excursion.

The study by Abrahamova et. al (2008) showed that CoP amplitude and velocity were the two most reliable measures to study age related differences. CoP amplitude is derived from the standard deviation of the CoP. They compared CoP parameters across three age groups- Junior, middle-aged and senior and performed a regression analysis. They also found noticeable differences in CoP sway, velocity, and amplitude in people above the age of 60.

Seigle and colleagues (2009) calculated the CoP total excursion and 95% confidence area ellipse for the young and elderly during eyes open and closed conditions. 11 young and 12 elderly participants were recruited. The results showed differences in total excursion and sway area between age groups only in the eyes closed condition. The difference in result compared to this study may be due to a smaller number of subjects and a shorter data acquisition time (30 seconds).

There have been many studies that have looked at the test-retest reliability of the CoP measures (Lafond, Corriveau, Hebert, & Prince, 2004; Lin, Seol, Nussbaum, & Madigan, 2008; Raymakers et al., 2005). All the studies come to the same conclusion that mean velocity is the most consistent measure to determine differences between age groups and for different vision conditions. The results of the present study was consistent with theirs

Thus to summarize, age has significant effects on balance control during quiet standing. This is reflected in three important measures derived from CoP, sway area, mean frequency and total excursion. The young and the elderly rely on visual input for quiet standing. This was evident from changes in distance and area measures with change in visual input, especially in the AP direction.

CHAPTER 3

EFFECTS OF AGING ON POSTURE SHIFTS

BACKGROUND

Posture weight shifts has not been studied as extensively as quiet standing. It involves shifting ones weight between the two feet, in order to lean to different directions without lifting ones foot and losing balance. In clinics, the ability to perform posture weight shifts is measured in terms of Limits of Stability (LoS) that quantifies the maximum distance up to which a person can move his/her CoP with a stable base of support without losing their balance.

During the LoS test, the participant is asked to lean as far as possible from his/her initial erect position in specific directions based on the visual feedback from the monitor placed straight ahead. The participants have their hands by the side and are instructed to use mostly their ankles and not their hips, while leaning. Most commonly derived measures from LoS test are: maximum distance, movement time and velocity. Fallers move a shorter distance with much smaller velocity than age matched non-fallers (Pickerill & Harter, 2011). The LoS measure is validated and is being used routinely in clinical settings for fall risk assessment.

Postural shifts have been studied in people with stroke. Chern et al. (2010) measured CoP measures during postural shifts to six different target locations. CoP excursion, mean velocity and bilateral limb ratios were assessed. The results showed significant differences between people with stroke and age matched controls in all measures except CoP velocity. Target preferences were significant in stroke patients alone. This study showed that larger displacements and slower velocity of CoP in stroke subjects indicate adoption of a compensatory postural mechanism.

Lemay and colleagues (2014) studied absolute maximum distance and total CoP path length during posture shifts while standing, in people affected by incomplete traumatic Spinal Cord Injury (SCI). Results showed people with SCI had significantly greater CoP path length in all target directions. The path length when progressing towards the center target was higher than when moving away from the center target in the anterior direction.

A study on voluntary shifts of CoP to different directions at different frequencies revealed that the voluntary shifts and background CoP sway are independent processes. The shifts required an internal command to initiate whereas the CoP sway was inherent and did not require an internal command (Latash, Ferreira, Wieczorek, & Duarte, 2003).

Another method for assessing dynamic balance is functional reach. Functional reach is similar to the LoS test, but the subject reaches to an object within their LoS. Wallmann et al. (2001) compared non-fallers and fallers over the age of 60 for differences in sway during functional reaching task. The results showed moderate correlation of functional reach and CoP path length in the AP direction for fallers.

A study on the effects of knee pain on functional reach and gait aimed at correlating the parameters from gait, quiet standing, and functional reach. No correlation was found between knee pain, timed up and go test, and functional reach test. This study did not consider the effects of age or risk of falls, but the only factor taken into consideration was knee pain (Takahashi et al., 2004).

A postural shift paradigm very similar to what is used in this study was extended to studying the effects of deep brain stimulation (DBS) on Parkinson's disease patients. The study focused on finding out how postural instability improves with DBS. The postural shift paradigm was performed during four stimulation conditions, and several parameters such as movement time, velocity, and path length were calculated during the initiation, movement, and hold phases of target reach. Results showed a reduction in peak velocity and velocity during the initiation and movement phases during deep brain stimulation-off condition compared to deep brain stimulation-on condition (Krishnamurthi, Mulligan, Mahant, Samanta, & Abbas, 2012).

After reviewing the existing literature on limits of stability and functional reach, it is evident that LoS is a standardized test for fall risk assessment; however, in most of the studies, the only parameters considered are time of reach and maximum distance. Moreover, the effects and correlation of CoP during quiet stance and posture shift has only been studied in people with stroke, SCI, or knee pain. The present study compares more detailed parameters extracted from CoP during posture shifts and compares them between the young and elderly.

METHODOLOGY

Subjects. The same subjects who were recruited for the quiet standing study also took part in the dynamic shift task. The recruitment criteria and other subject recruitment related information is provided in Chapter 2.

Experimental Protocol. Subjects participated in the posture shift task after completing the quiet standing task, during the same session. The setup requires the subject to stand on the force platform with their hands by their side and feet separated by hip-width. Once the subjects stood comfortably, a trace of their feet was taken to ensure consistent placement of the feet across trials. All subjects wore comfortable shoes.

Previously developed LabVIEW-based graphical user interface was utilized to provide real-time visual feedback of the position of the subject's CoP. At the start of the trial, the CoP of the subject was taken as the center of the center target (Figure 3.1). The subject viewed his/her CoP on the monitor placed in the front of the subject at eye level which provided real-time visual feedback. During the course of the trials, the outward targets were displayed in different positions, each separated by an angle of 45°. To facilitate comparison across subjects, the distance of the target circle from the center was set to 30% of the distance between the hip and the ankle, which has been demonstrated to be related to the LoS (Pickerill & Harter, 2011). The radius of the center and target circles was set at 10% of the distance between the hip and the ankle. The subject was instructed to move their CoP, displayed in a form of red circular cursor, to the target circle position by leaning without lifting their feet off the ground. Once his/her CoP entered the target circle, they were asked to hold their position as close as possible to the center of the target circle within the target for at least 2 seconds. After that, the current target circle disappeared and the center target appeared which became the new target. If the subject was unable to stay within the target for at least 2 seconds, then the new target appeared automatically in 10 seconds. If the subject stayed inside the target for at least 2 seconds, the target was considered successfully achieved.

The five different angles at which the targets presented were 0, 45, 90, 135, and 180 degrees. After reaching towards each target, the subject came back to the center target position before moving towards the next outward target.

Thus, a total of ten targets were provided during the trial- O-0°, 0°-O, O-45°, 45°-O, O-90°, 90°-O, O-135°, 135°-O, O-180°, 180°-O, where O represents the origin or center target. During a single trial, 20 targets were presented, i.e. each of the ten targets were presented twice. A total of five trials were performed, with sufficient rest periods in-between. The sequence of outward targets was randomly presented within and across trials to minimize learning effects or anticipation of the target.

Data Analysis. All the measures extracted from the dynamic shift data were based on the stabilogram obtained for the different trials and were derived using customized analysis programs developed in MATLAB 2013.

The entire trial period was divided into three phases: (a) Initiation phase (b) Movement Phase and (c) Hold Phase. The initiation phase spans the time period starting from the beginning of the presentation until the CoP cursor moved out of the start circle. The movement phase starts from the time the CoP moved out of the start circle until the last time point before CoP moved in to the target. During this phase, the subject's CoP lies in between the start circle and the target circle. The final hold phase covers the time period when the CoP cursor was held within the target circle for at least 2 seconds.

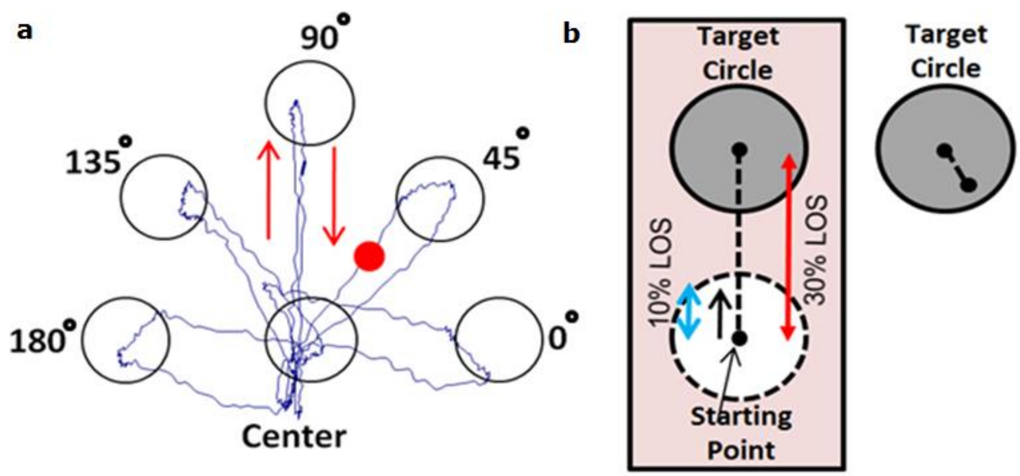

*Figure 3.1.*Posture shift paradigm (Pic courtesy: Dr. Krishnamurthi)

(a) The five outer circles represent the targets to be reached in different angles, 0, 45, 90, 135, and 180 degrees. The red dot denotes the position of the CoP of the subject at a particular time instance during target reach. The blue trace is the path traversed by the subject during different outward and center target reaches. (b) This schematic diagram explains the three phases during target reach: Initiation phase, movement phase, and hold phase. The radius of the target circles was set to 10% of the subjects' limits of stability and the distance between the starting point and the center of the target circle was set at 30% of limits of stability of the subjects to facilitate comparison across subjects.

Initiation Phase. (a) Initiation time: Total time spent within the initial start circle.

(b) Initiation pathlength: Total pathlength covered by the CoP in the AP and ML directions when inside the initial start circle.

(C) Initiation Velocity: The mean velocity of the CoP inside the start circle.

Movement Phase. (a) Movement time: Time spent in the movement phase as defined above.

(b) Movement path length: Total path length of the CoP travelled in AP and ML directions during the movement phase

(c) Movement velocity: Mean velocity of CoP when traversing the movement phase.

Hold Phase. (a) Number of re-entries: The total number of times the subject reentered the target circle (the first entry into the target was not considered).

(b) Inaccuracy: The mean of the distances between the center of the target circle and the position of the CoP during the hold phase.

(c) Unsteadiness: The standard deviation of the distances between the target circle and position of CoP during the hold phase.

(a) Peak Velocity: Maximum velocity of the CoP during the entire presentation of a target

Statistical Analysis.

Repeated Measures Analysis of Variance. Similar to the statistical analysis performed for quiet standing, as described in chapter 2, a general linear model with repeated measures ANOVA was implemented on each of the response variables separately. The main questions to answer were (a) Does age affects any of the response variables during the three phases of posture shifts? and (b) Do the responses vary significantly for different target angles in the young or elderly groups?   Each subject performed shifts to 10 different targets. Each of the targets were presented twice during each trial, and there were five trials. Thus, each target was reached 10 times by a single subject, and there were a total of 5 X 10 X 2= 100 reaches per subject.

In order to answer the first question, a repeated measures ANOVA was performed with age as the between subject factor. The within subject factors were the target angles (10 levels), trials (5 levels), replicates (2 levels). A p-value less than 0.05 was considered significant. The number of re-entries alone was compared across age groups using the generalized linear model using a Poisson distribution, since the response variables is a count data. All statistical analysis was performed using SPSS 22.

In order to answer the second question, a repeated measures ANOVA was performed for each age group separately, with the targets (10 levels) as the between-subject factor, and the trials and replicates as the within-subject factor. A p-value of less than 0.05 was considered significant.

All the main effects and the interactions were calculated and the bonferroni method was used to adjust the p-values accordingly, to take into account the subject-subject variability.

Test-retest reliability. Cronbach's alpha and ICC were calculated as explained in chapter 2, for each of the responses, to determine the measures that show high consistency and could possibly be used for balance assessment.

RESULTS

Effect of age. Repeated measures ANOVA was used to find the effects of age on dynamic shift task performance, taking the effects of inter-trial variability into consideration. Subjects 3 and 17, who belonged to the young group, were eliminated from the study due to poor data quality because of technical difficulties. Initiation time, initiation path length and movement path length, inaccuracy, and number of reentries increased significantly with age. Thus, all the three phases, initiation, movement, and hold phases were affected by age. The p-value for each of these measures is provided in Table 3.1. Initiation time, Initiation path length, movement time, movement path length had a right skewed distribution. Hence a log transformation was used on the data prior to analysis. There were significant trial-to-trial variations within subjects, for initiation path length and initiation time. However, none of them showed significant interaction between age and trials. This implies that the young and elderly respond similarly across trials and there is no trial-to-trial adaptation. Figures 3.2,3.3 and 3.4 show the line plots for the initiation phase, movement phase and hold phase measures respectively.

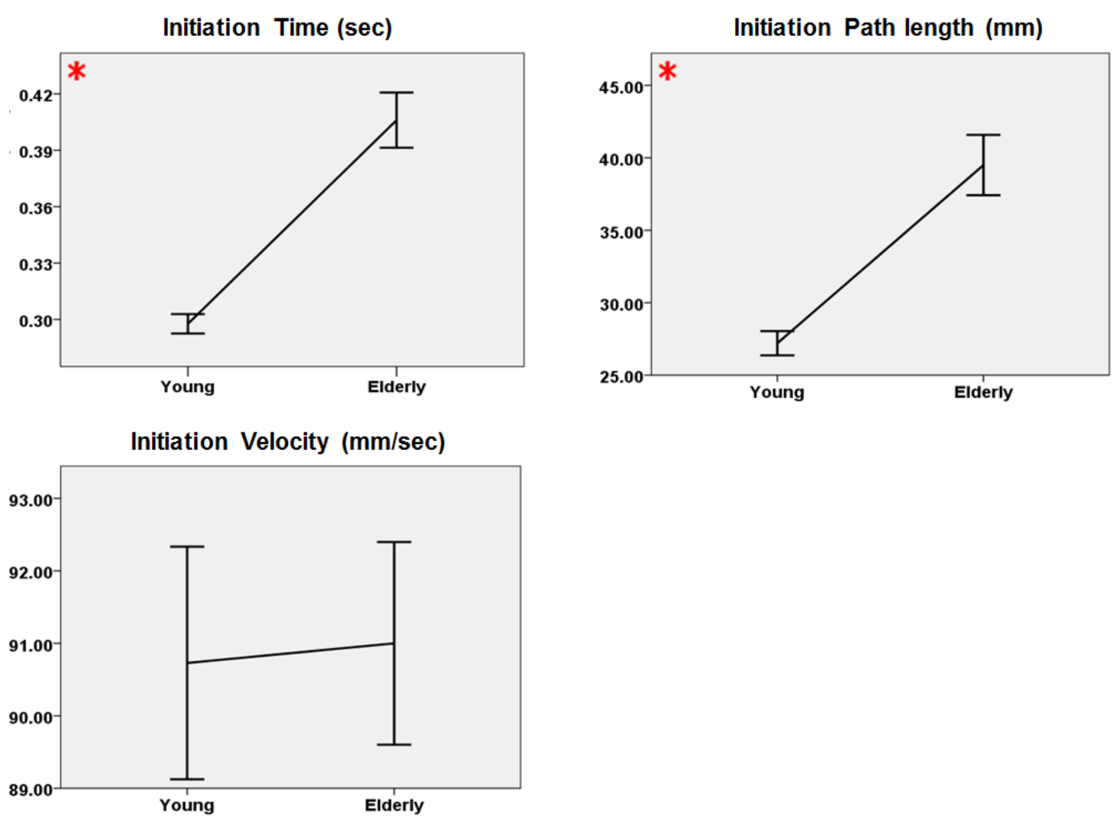

Figure 3.2. Change in initiation phase parameters between the young and the elderly groups. Red asterisk denotes significant difference between young and elderly groups ($p<0.05$) The vertical lines denote the 95% confidence intervals for the mean values.

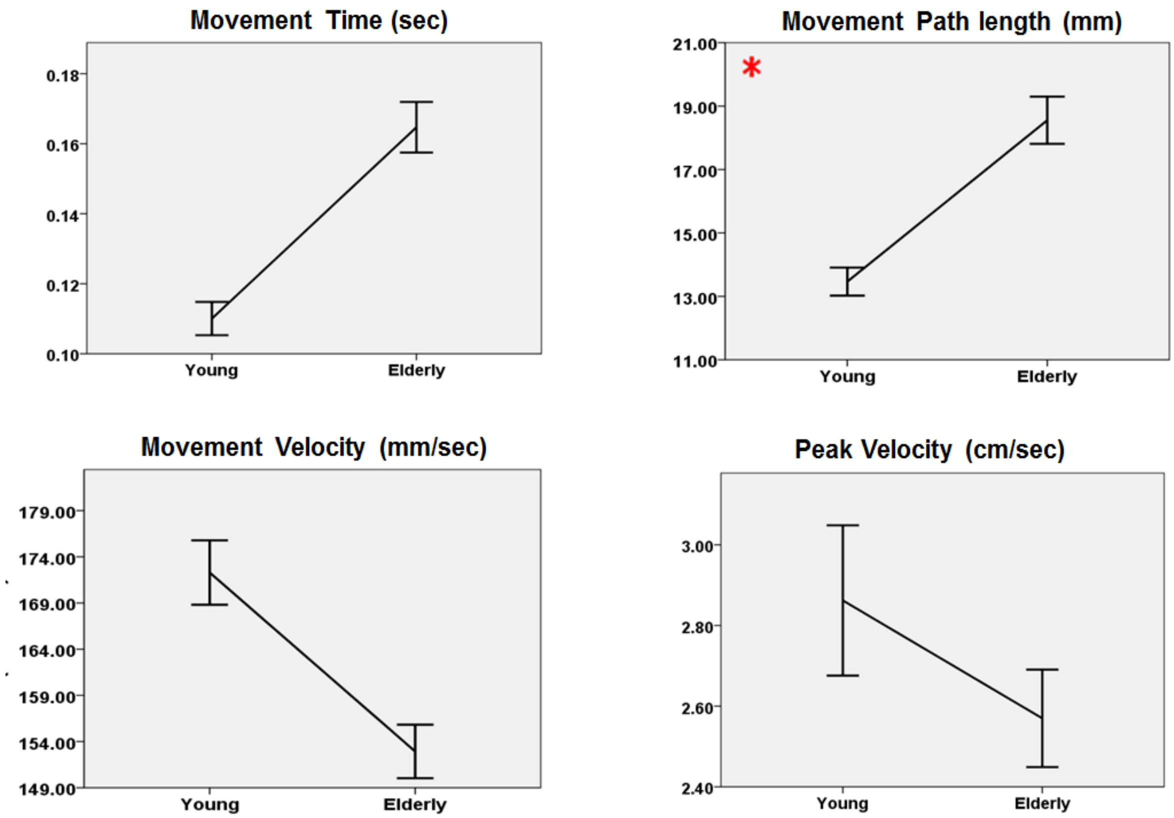

Figure 3.3. Change in movement phase parameters between the young and the elderly groups. Red asterisk denotes significant difference between young and elderly groups ($p<0.05$). The vertical lines denote the 95% confidence intervals for the mean values.

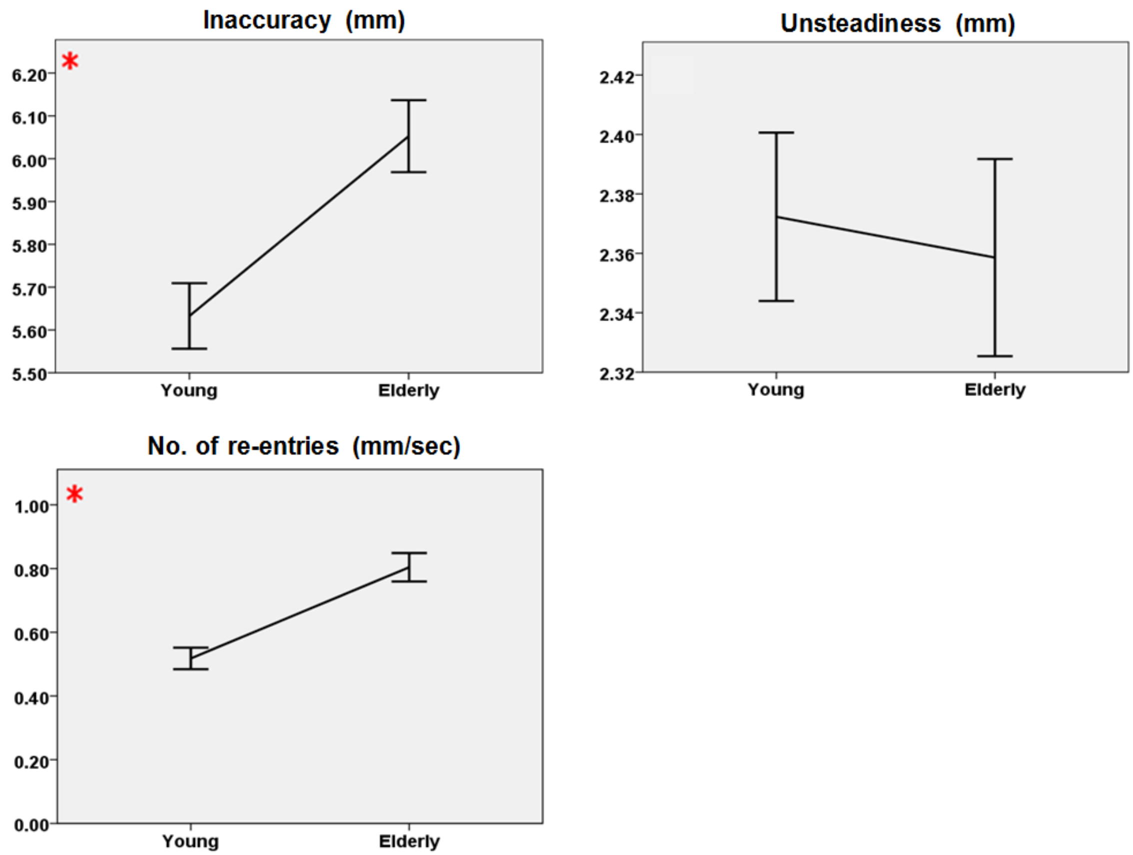

Figure 3.4. Change in hold phase parameters, inaccuracy, unsteadiness, and mean number of re-entries between the young and the elderly groups. Red asterisk denotes significant difference between young and elderly groups ($p<0.05$). The vertical lines denote the 95% confidence intervals for the mean values.

Table 3.1

Posture shift parameters that showed significant difference between young and elderly groups. The F and p-values obtained from repeated measures ANOVA are provided. P < 0.05 is considered significant. The F value for the re-entries refers to the Wald chi square value from the Poisson regression analysis.

PARAMETER	F VALUE	p-VALUE
Initiation Time	6.799	0.013
Initiation path length	4.568	0.039
Movement path length	7.087	0.011
Inaccuracy	5.072	0.030
Re-entries	123.718	< 0.001

Effect of direction of targets. Repeated measures ANOVA was performed on both the young and elderly groups separately to determine if there were any preferences to certain targets when compared to others. In the young group, initiation time, movement time, inaccuracy and movement path length were significantly different for at least one of the targets. *Post-hoc* analysis revealed the specific targets showed differences in the dynamic shift measures. Initiation time was significantly lower for all the targets when the subject returned from the outward circle to center target. Movement time, inaccuracy and movement path length showed differences between 135 degree target and 180 degree target. In the elderly population, significant differences between at least two of the targets was observed in initiation time, initiation path length, movement path length, and inaccuracy. Similar to the young group, initiation time was lower for movement towards the origin when compared to moving towards the outward targets. Movement path length was significantly lower for 0 and 180 degree targets when compared to 45 and 135 targets. This indicates that all the subjects move in a more direct path towards the 0 and 180 degree targets. However, inaccuracy was highest for the 0 and 180 degree targets.

On the other hand, Inaccuracy was significantly lower for the 90 degree target. Also, the movement time for the 90 degree target is lesser in the elderly group. This is accompanied by lower number of re-entries. Figures 3.14 to 3.22 depict these differences in responses for various targets.

To summarize, initiation time for all the outward going targets was higher compared to targets towards the origin. The elderly tend to take more time to initiate the movement towards the 90 degree target. However, they tend to move at higher velocities and with decreased inaccuracy towards the 90 degree target. In contrast, 0 and 180 degree targets have shorter movement path lengths and higher inaccuracy.

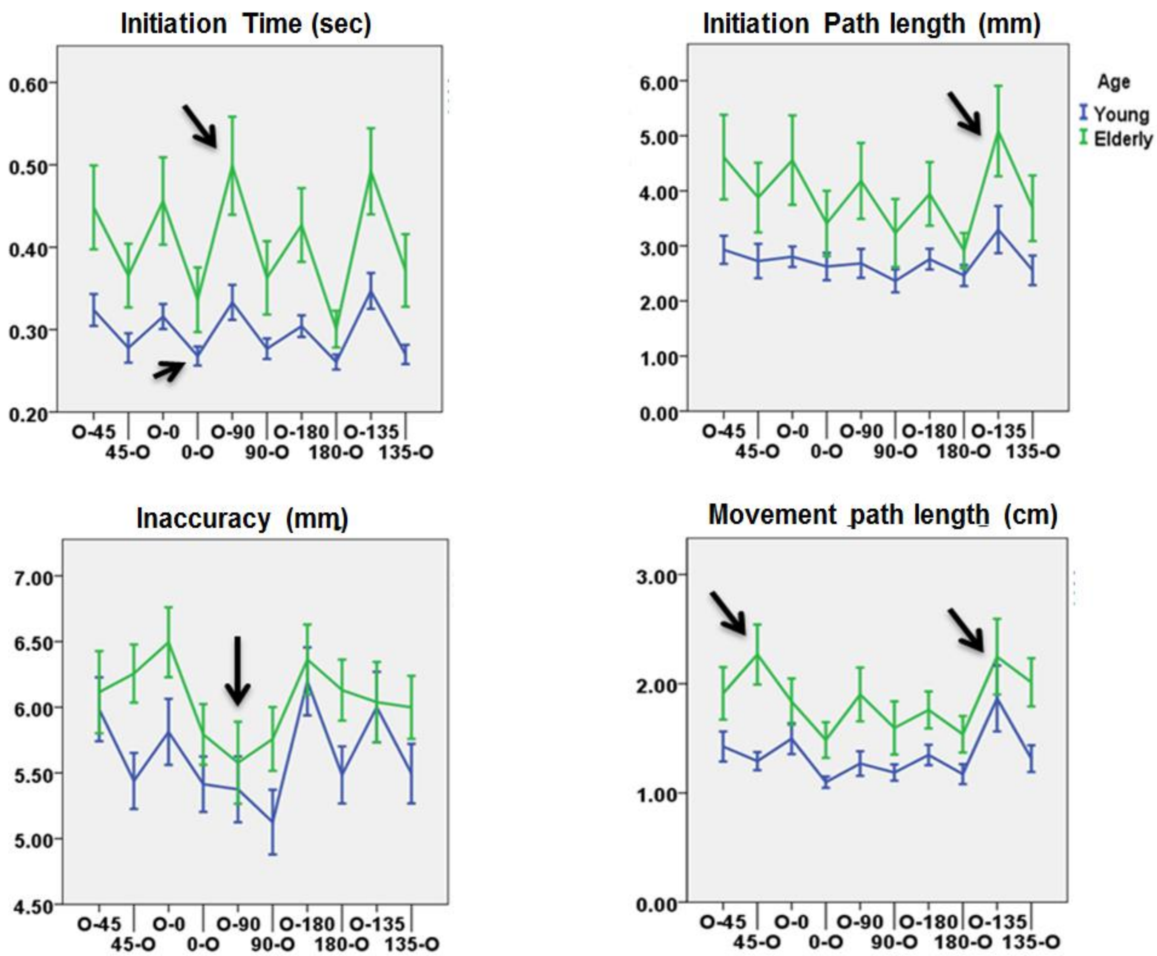

Figure 3.5. Change in dynamic shifts measures that show significant difference ($p < 0.05$) in at least one of the target directions in the young and elderly groups. In each subplot, x-axis indicates target directions. The target labels that start with 'O' indicate reaching towards outward targets and the target labels that end with 'O' indicate center target reach. Black arrows denote significant change in the corresponding parameters within young and elderly groups.

Test-retest reliability. The Cronbach's alpha values for all the dynamic shift measures were greater than 0.8 indicating good consistency across trials. ICC was greater than 0.8 for all the velocity measures - initiation, movement, and peak velocity. In addition, the elderly showed ICC greater than 0.8 for initiation time and path length. This indicates that the parameters were consistent as well as robust and there was any adaptation across trials.

Table 3.2

Test-retest reliability measures Cronbach's alpha and Intra-class correlation coefficient (ICC) for different dynamic shifts measures in the young and elderly groups. Alpha and ICC values greater than 0.8 mentioned in bold and indicate good consistency across trials.

YOUNG			**ELDERLY**		
MEASURE	**ALPHA**	**ICC**	**MEASURE**	**ALPHA**	**ICC**
Initiation Path length	**0.891**	0.602	Initiation Path length	**0.981**	**0.898**
Initiation Time	0.722	0.328	Initiation Time	**0.981**	**0.904**
Initiation Velocity	**0.962**	**0.836**	Initiation Velocity	**0.968**	**0.838**
Movement Path length	0.763	0.395	Movement Path length	**0.942**	0.745
Movement Time	**0.923**	0.686	Movement Time	**0.930**	0.704
Movement Velocity	**0.969**	**0.863**	Movement Velocity	**0.957**	**0.816**
Peak Velocity	**0.972**	**0.879**	Peak Velocity	**0.968**	**0.851**
Reentry	**0.949**	0.777	Reentry	**0.945**	0.748
Unsteadiness	0.791	0.437	Unsteadiness	**0.898**	0.643
Inaccuracy	**0.931**	0.715	Inaccuracy	**0.944**	0.768

DISCUSSION

A posture shift paradigm was tested to study the effect of age on dynamic balance. It is important to study the effects of age on dynamic balance since most of the falls occur during locomotion and activities of daily living such as reaching.

The results demonstrate that age affects initiation time, initiation path length, movement path length, inaccuracy, and number of reentries. It was observed that initiation time and initiation path length increase with age. The increase in initiation time could be due to various reasons such as delayed torque production in the ankles, muscle weakness, or reduced reaction time once the target is presented. During the movement phase, the subject voluntarily moves his/her CoP outside the base of support. The hold phase requires the subject to maintain the CoP in a leaning position. Increase in inaccuracy with age shows that the elderly find it difficult to maintain the CoP away from the rest position and as close as possible to the center of the target circle for a prolonged period of time. An increase in movement path length and number of reentries in the elderly might indicate the use of hip strategy during the movement. The control of CoP by the ankle muscles is found to be insufficient in elderly population.

From the results of the effects of target directions in the young and elderly, it can be seen that elderly hesitated to shift in the AP direction. They began with a slower initiation velocity and subsequently increased their movement velocity. However, inaccuracy was found to be least in the AP direction indicating better stability in the AP direction. From the results of the quiet standing task, it was observed that visual input improves the stability in the AP direction rather than ML direction. This could be the reason for better accuracy in the AP direction. Also, slower velocity during initiation might have helped to reduce the inaccuracy during hold phase. On the other hand, inaccuracy increased and movement path length decreased in the ML direction for the elderly indicating poorer performance for horizontal targets. Initiation velocity for the horizontal targets was also higher than for the vertical targets. Thus, inaccuracy in the horizontal targets could be due to the increase in initiation velocity. This implies that elderly find it more difficult to hold their CoP near the target in the ML direction. Initiation time for outward targets was higher compared to targets towards the origin. This indicates that subjects were more comfortable coming back to the center target (equilibrium position) than moving outside the base of support to reach outward targets. This might also be due to the subjects' anticipation to come back to the center target after every target presentation. The calculation of error or inaccuracy separately for AP and ML directions may help to better understand and explain the differences in the performances for different target directions. The direction information will indicate whether the subject overshot the target or if the subject hesitated to move towards the target in a particular direction (AP or ML) more than the other.

Many studies have looked at LoS and functional reach, but none of the studies have derived such detailed measures during the three phases of the target reach task. Limits of stability and functional reach tests have been studied on people with stroke, risk of falls and spinal cord injuries. The study on stroke patients showed increase in CoP excursion and no difference in CoP velocity (Chern et al., 2010). They also showed a preference towards certain targets depending on the affected side. Spinal cord injury patients showed increased path length during posture shifts, especially in the AP direction (Latash, Ferreira, Wieczorek, & Duarte, 2003). This is the first time posture shifts have been studied thoroughly in healthy elderly. Novel measures derived in this study are inaccuracy and number of re-entries, which could potentially be used to also study effects of neurodegenerative disorders and falls.

Based on the above observed results, it is clear that aging affects balance control involved during tasks such as reaching and the proposed dynamic shift paradigm can be effectively utilized to characterize the effects of aging on balance control. Further, correlation between measures of dynamic shifts with quiet standing would be useful in interpreting the alterations in physiological mechanisms associated with balance control.

CHAPTER 4

EFFECTS OF AGING ON GAIT

BACKGROUND

Bipedal locomotion is one of the most complex mechanisms inherited as a result of evolution. Gait results from a series of intercepted falls from a single limb stance. Human gait is achieved by successful integration of the visual, vestibular, and somatosensory information along with motor variables such as muscle strength, time of activation of the muscles, and joint mobility (Halliday, Winter, Frank, Patla, & Prince, 1998)

A study proposed by Grillner and Wallen (1985) suggests that there are three main functions performed by the Central Nervous System (CNS) to deliver normal gait patterns. Firstly, the rhythmical gait patterns have to be generated for proper coordination of the different muscle groups. Secondly, the CoM must be controlled such that it doesn’t make the system unstable. Thirdly, CNS must possess adaptive capabilities to correct gait during perturbations. These functions are found to be performed by the motor cortex, basal ganglia, and the cerebellum. Basal ganglia is involved mainly in the initiation and regulation of gait by integrating the information from the sensory systems. The cerebellum is responsible for gait coordination and generation of patterns for limb movements. Cerebellar dysfunction is found to cause difficulty in walking including, variable foot placement, wider base of support, and abnormal joint coordination (Jeffrey M.Hausdorff, 2005). Loss of function in the basal ganglia is found to cause difficulty in gait initiation and larger stride to stride variability (Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998).

Quiet standing involves maintaining the body’s CoM within the base of support. Walking on the other hand involves stabilizing the body even when the CoM is outside the base of support (Kirtley, 2006). This type of balance, known as dynamic balance is achieved by a process known as re-stabilization. During this process, the human body averts a fall after every step, by coordinated positioning of the stance and swing limb. Thus, walking involves moving the CoM voluntarily outside the base of support, and further moving the swing limb in such a way as to re-stabilize the system (Kirtley, 2006). Unlike standing, gait cannot be controlled majorly by the ankle muscles. It requires the complex coordination of multiple joints to achieve stability.

Gait Phases. Normal human gait can broadly be divided into two phases: Stance and Swing phases. Typical gait cycle starts with one heel strike and continues to the next heel strike by the same foot. Almost 60 % of the gait cycle is the stance phase, which can be further divided into sub-phases, namely, double support and single support phases based on the number of limbs on the ground. The single support phase can further be divided into heel strike, mid-stance, and toe off events. The double support time acts to stabilize the act of locomotion. Longer double support times occur during slower movement velocities. The swing phase occupies 40% of the gait cycle and is divided into three sub-phases: initial swing, mid swing, and terminal swing. Initial swing occurs as soon as the foot is lifted from the ground and is a period of acceleration of the limb. Mid swing is the period when the stance and swing limbs align with each other. The terminal swing phase occurs prior to landing of the foot in the ground and is a phase of deceleration (Kirtley, 2006).

Basic Gait Parameters. Various parameters are extracted from gait patterns to better understand the differences between normal and pathological gait. A brief overview of some of the definitions of gait parameters is provided:

Gait cycle: The time period between one heel strike and the subsequent heel strike by the same foot.

Step Length: Distance between successive heel contacts of the two feet.

Stride Length: Distance between heel contacts of the same feet at successive times.

Cadence: Number of steps covered per unit time

Stride velocity: Stride length divided by gait cycle.

Gait cycle time: Time taken to complete one gait cycle, starting from heel strike on one foot to next subsequent heel strike of the same foot.

Functional assessment of gait: The assessment of gait is important to detect alterations in dynamic balance due to aging and pathological conditions which may increase risk of falls (Hausdorff et al., 1998). Different gait assessment techniques have emerged over the years, from analyzing the center of pressure to determining joint kinematics and range of motion (Jeffrey M.Hausdorff, 2005). The most prevalent test undertaken in the clinical setting is the timed-up and go test (TUG). TUG consists of the subject to raise from a chair, walk 3 m, and return back to the chair. Most physical therapists carefully observe the subject for any abnormalities and measure the cadence and gait duration, both of which have been proven to be valid measures for dynamic balance assessment (Salarian et al., 2010). Other characteristics of gait that are derived include double support time, stride length, symmetry and trunk sway (Jeffrey M.Hausdorff, 2005).

Physiological effects of aging on gait: Aging has a profound impact on gait. It is found that 20% of the elderly adults require assistance during walking and have difficulty in performing activities of daily living. This has increased the risk of falls during locomotion, bending over, and turning (Woollacott & Tang, 1997). Some of the main reasons for gait impairments due to aging include loss of muscle volume, weakening of muscles, and loss of sensory acuity. Elderly population rely more on sensory systems such as visual input during walking (Snijders, van de Warrenburg, Giladi, & Bloem, 2007). This increases the cognitive load and attention required to maintain normal gait. This might be due to the reduction in dopamine producing neurons by 30-50 % by the age of 65 (Halliday et al., 1998)

The effect of aging can be observed in all stages of the gait cycle. Halliday et. al (1998) compared gait initiation between young and elderly adults. Results show that, although there is no significant difference in gait initiation time, the velocity of the whole body CoM is significantly higher for young than the elderly (Snijders et al., 2007). This was reflected in the gait patterns as increased stride length of the first step in young subjects.

Winter and colleagues (1990) compared gait patterns of healthy elderly adults with young adults. The elderly adults seemed to have larger step widths compared to young group. This was attributed to reduced moment of CoM in the ML direction to gain lateral stability. Also, the hip abductors that play a major role in step width regulation were found to be less effective with age (Woollacott & Tang, 1997). Other studies have shown significant decrease in stride velocity and stride length in the elderly, but no variations in gait cycle duration (Kimura, Kobayashi, Nakayama, & Hanaoka, 2007).

Extensive research has been carried out to study the variability in stride length and stride time as a consequence of age. Increased stride length variability (Hausdorff, 2005; Kang & Dingwell, 2008; Woollacott & Tang, 1997) and stride time variability (Hausdorff, 2005, 2007; Kang & Dingwell, 2008) have been reported in the elderly population. Most of the studies have related this variability to decrease in gait speed (Hausdorff et al., 1998). However, others have attributed it to loss of muscle strength and decreased range of motion (Hausdorff, 2005).

Cao and colleagues (2007) studied the effects of age on gait termination. They found elderly people took longer time for termination compared to young adults. This is due to use of two step termination in the elderly as opposed to a one step termination in the young. This is also ascribed to earlier activation of the ankle muscles in the younger group (Sparrow & Tirosh, 2005).

The measures obtained from timed-up and go tests have been compared between young, healthy elderly, and elderly fallers. Most of the studies have focused on the total duration of the task and the sit to walk measures. Kerr and colleagues (2007) measured the time for sat off, swing off, and stance off during routing TUG task. They found the sit to walk duration to be significantly longer in elderly fallers, but not in the two control groups. On the contrary, Buckley and colleagues (2009) showed significant differences in initiation of gait during sit to walk in healthy elderly compared to young.

Apart from studying gait during normal walking, studies have looked at performance of young, elderly, and fallers in a tandem gait task. Difficulty in performing the tandem gait task was attributed to disturbances in cerebellar regulation of gait (Lark & Pasupuleti, 2009). One study tested elderly fallers vs. non fallers in a reduced base of support walking and tandem walking. The results showed that only some of the subjects in both fallers and non-fallers could not complete the tandem walk test and fallers took a longer time of completion (Hiura, Nemoto, Nishisaka, Higashi, & Katoh, 2012; Lark & Pasupuleti, 2009; Louis, Rios, & Rao, 2010; Speers, Ashton-Miller, Schultz, & Alexander, 1998)

METHODOLOGY

Subjects. Subjects who participated in the quiet standing and posture shift task, also took part in the gait task. Please refer to the methodology provided in chapter 2 for a detailed description of the recruitment criteria and general information about the subjects.

Experimental Setup. Gait data collection and analysis was performed using APDM Mobility Lab gait system (APDM Inc, Portland, OR). Mobility Lab is a portable gait and balance system used to collect, store, and analyze data from inertial measurement units (IMUs) and calculation of gait related measures. The setup consists of 6 IMUs, which were attached to the subject non-invasively, over clothing. The data from the IMUs were wirelessly transmitted to a receiver. The receiver then synchronized all of the IMU data and fed it into the Mobility Lab software to calculate measures for different gait tasks. The software consisted of test protocols such as Instrumented Timed Up and Go (ITUG) and Instrumented long walk (IWalk) for easy gait data collection and analysis.

Experimental Protocol. The subjects were asked to wear comfortable shoes. Six inertial sensors were worn using Velcro straps: Two sensors on the ankles, two sensors on the wrists, one sensor positioned in the lumbar region and one on the trunk/sternum. They performed three types of over-ground walking tasks.

ITUG test. During this task, the subject was asked to sit on a chair. Once the test starts, the subject got up from the chair without using the arm supports, walked 7 meters, turned and returned to the chair, and sat, again without the support of the arms of the chair. All the subjects walked in their comfortable pace. A total of five trials were conducted for each subject.

Tandem walking test. For the tandem walk, the trial started from the standing position. The subjects were instructed to walk with the heel of one foot touching the toe of the other, following a straight path as much as possible, for a distance of 7 meters. Three trials were performed. In case the subject had any difficulty performing the task or did not volunteer to complete the task, the test was not conducted.

Table 4.1

The list of gait measures obtained from different walking tasks (Timed-up and Go, normal walk, and tandem walk) and that were compared across age groups is provided. COV and ROM stand for Coefficient of variation and Range of Motion, respectively.

ITUG	**NORMAL WALK (6-sensor setup)**	**TANDEM WALK**	**NORMAL WALK (2- Sensor Setup)**
Total duration	Stride length, COV	Cadence	Cadence
Sit-to-Stand Duration	Stride velocity, COV	Peak arm swing velocity	Gait cycle duration
Sit-to-Stand Peak velocity	Cadence, COV	ROM arm swing	Gait speed
Sit-to-Stand ROM trunk	Gait time, COV	ROM trunk horizontal	Initial double support time
Turn Number of Steps	ROM arm swing, COV	ROM trunk sagittal	Terminal double support time
Turn Peak velocity	% Double support time, COV	ROM trunk frontal	Single limb support time
Turn Step time	%Swing time, COV	Peak trunk velocity horizontal	Step duration
Turn Duration	% Stance time, COV	Peak trunk velocity frontal	Stride length
Turn-to-Sit Duration		Peak trunk velocity sagittal	Stride length variability
Turn-to-Sit Peak velocity			
Turn-to-Sit ROM trunk			
Step time before turn			

Long walking test. During this task, the subjects were asked to walk normally in their comfortable pace for a distance of about 40 meters. Five trials were performed. In addition to these trials, in the last 22 subjects tested (6 young, 16 elderly), long walking trials were performed with 2-sensor placement on the top of each foot for more reliable calculation of stride length as recommended by the manufacturer of the Mobility Lab gait system, APDM.

Data Analysis. Various measures of gait were obtained from each of the three tasks of over ground walking. The processing of the data from inertial sensors and algorithm to calculate the gait measures were carried out using Mobility Lab software. The following measures were used for comparison between young and elderly for each of the tasks:

TUG measures. TUG task can be classified into three phases: Sit-to-stand, walk, and turn-to-sit. Measures of sit-to-stand include (a) sit-to-stand duration, (b) sit-to-stand peak velocity, and (c) sit-to-stand range of motion (ROM) of the trunk.

The turning parameters obtained from walking phase are (a) Turn duration, (b) Turn number of steps, (c) Turn peak velocity, (d) Turn step time, and (e) step-time before turn.

The turn-to-sit parameters are (a) turn-to-sit duration, (b) turn-to-sit peak velocity, and (c) turn-to-sit range of motion of the trunk.

Tandem walking. The following measures were compared for differences between young and elderly during tandem walking: (a) Cadence, (b) Peak arm swing velocity, (c) Coefficient of variation of peak arm swing velocity, (d) ROM of arm swing, (e) ROM of the trunk in sagittal, frontal, and horizontal planes, and (f) Peak velocity of the trunk in sagittal, frontal, and horizontal planes.

Long walking: The following measures were derived from the long walking tasks: (a) stride length, (b) stride length Cofficient of Variation (COV), (c) stride velocity, (d) stride velocity COV, (e) cadence, (f) cadence COV, (g) gait cycle time, (h) gait cycle time COV, (i) double support percent, (j) swing percent, (k) stance percent, (l) ROM of arm swing, and (m) ROM of arm swing COV.

Statistical Analysis.

Repeated measures ANOVA. Similar to the statistical analysis performed in chapters 2 and 3, repeated measures ANOVA was used to investigate the gait measures affected by age. The between-subject factor was age (2 levels-young, elderly) and the within-subject factor was trials (5 levels for TUG and walk and 3 levels for tandem walk). The main effects of trials and age and the trials-age interaction effects were calculated. A p-value less than 0.05 was considered significant.

RESULTS

Three different tasks were performed by the subjects, TUG, normal walking and tandem walking. Table 4.2 shows all the measures that showed significant differences in all three tasks.

Timed-up and go. Repeated measures ANOVA was performed to determine the effect of age. The Q-Q plots indicated that the distributions of some of the parameters were right skewed. Thus a log transformation was performed on them prior to the ANOVA test. The main parameters analyzed from TUG tests are turning parameters, sit-to-stand, and turn-to-sit parameters. The total duration to perform TUG increased in the elderly. This was due to increase in the turn duration and turn-to-sit duration. Turn velocity and turn-to-sit velocity decrease in the elderly. Total duration also showed increased trial to trial variability, although this effect is taken into consideration during the statistical analysis. Figure 4.1 shows the changes in TUG measures that showed significant differences with age.

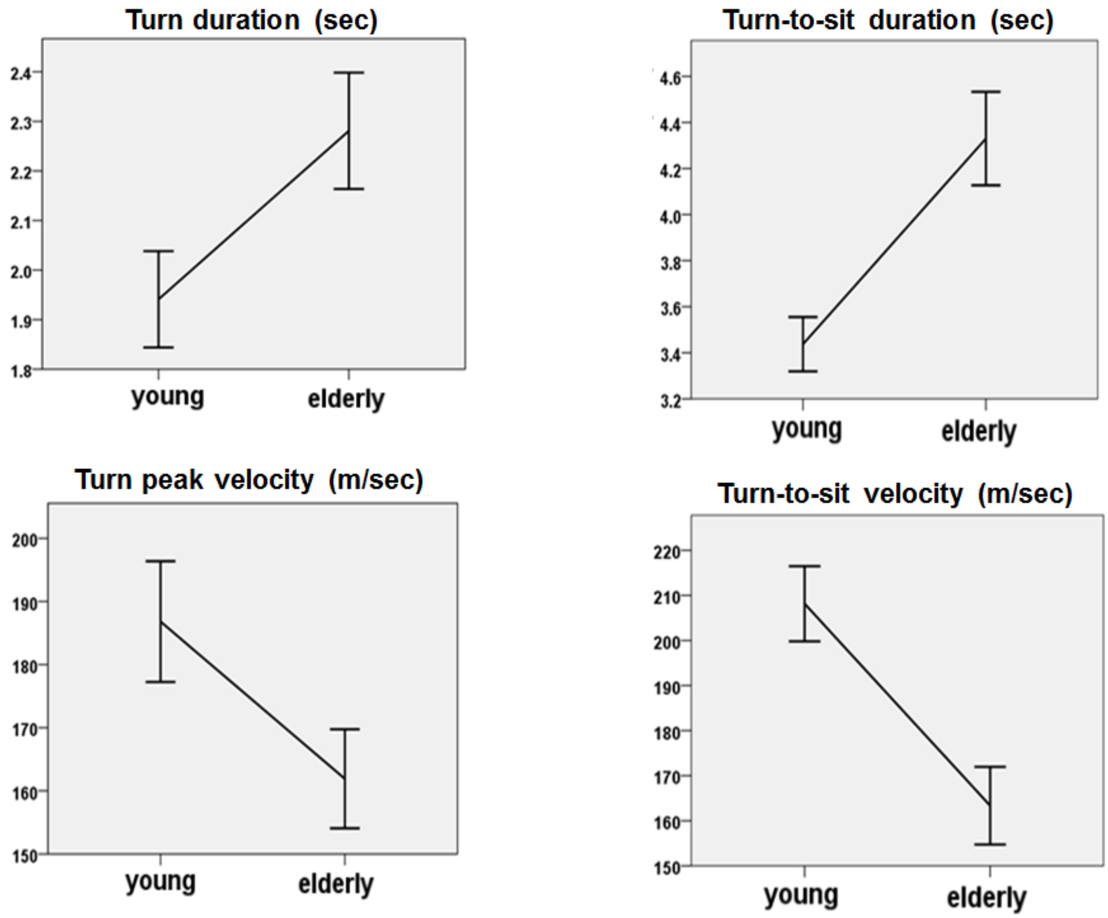

Figure 4.1. The turn duration, turn-to-sit duration, turn peak velocity, turn-to-sit peak velocity, and total duration for TUG test that were significantly ($p < 0.05$) changed due to aging are shown. The vertical lines denote the 95% confidence intervals for the mean values.

Normal walk. Similar to the TUG test, ANOVA was performed to derive the differences in stride related parameters. Stride length, stride length variability, stride velocity variability, and ROM of arm swing increased significantly with age. Stride length variability and stride velocity variability show trial to trial variability. Figure 4.2 shows the changes in measurs from normal walking that showed significant effects of age.

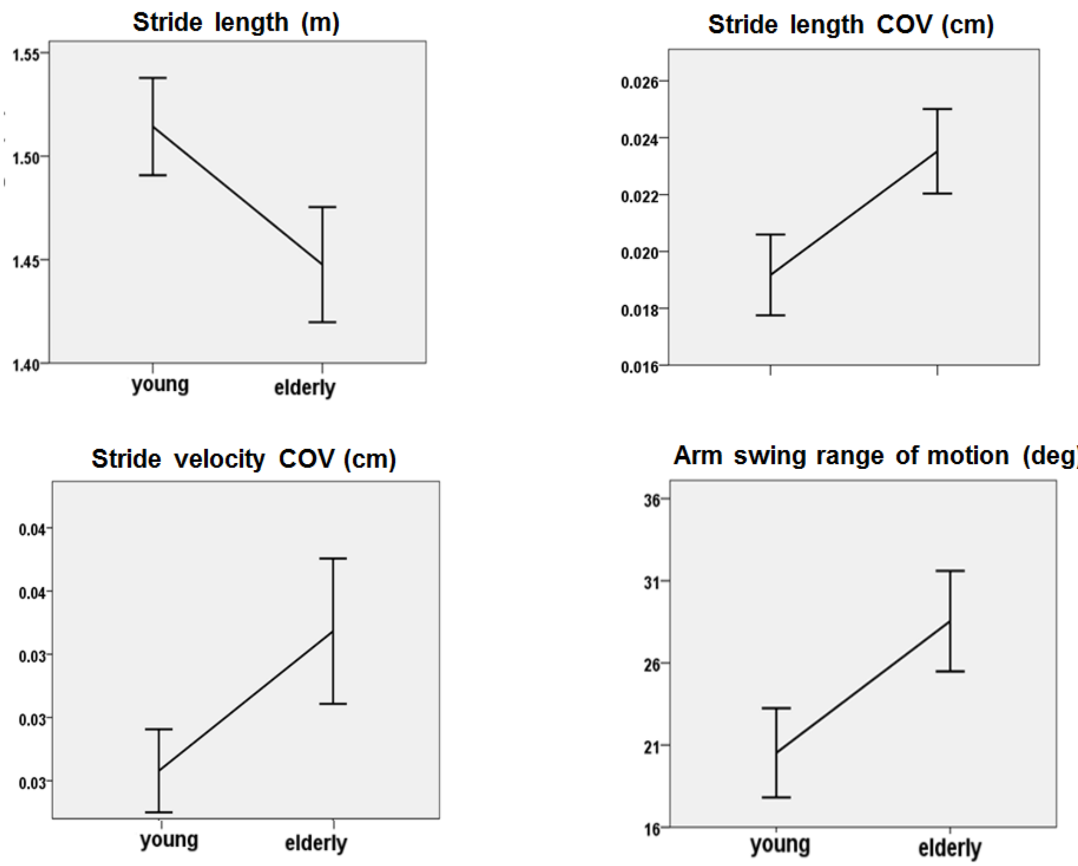

Figure 4.2. The statistical significant ($p < 0.05$) changes in stride length, stride length CoV, arm swing RoM, and stride velocity COV of arm wing, and CoV due to aging are presented. The vertical lines denote the 95% confidence intervals for the mean values.

Tandem walk. 19 elderly and 20 young subjects completed the task successfully. In general, the elderly found it more difficult to perform the tandem walk without losing balance. RoM arm swing showed significant increase with age (figure 4.3).

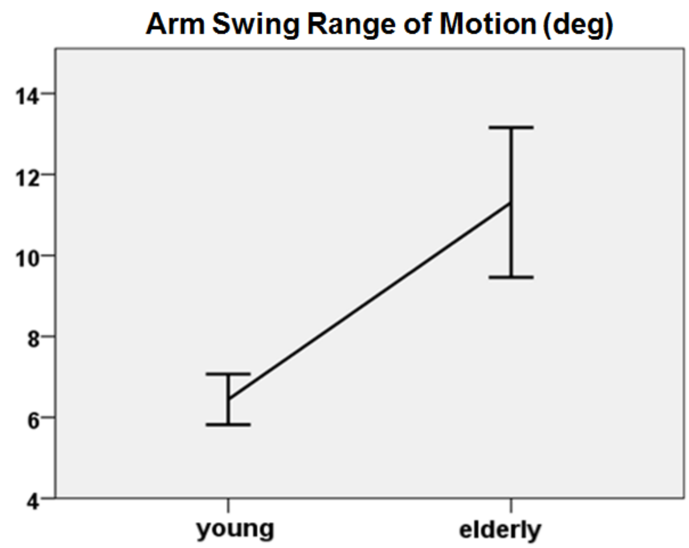

Figure 4.3. Change in range of motion of arm swing derived from tandem walking. It showed significant differences ($p < 0.05$) between the young and the elderly group. The vertical lines denote the 95% confidence intervals for the mean values.

DISCUSSION

The effect of aging on gait was studied using three different tasks: TUG, tandem walking, and normal walking. The results show that walking, turning, and RoM measures significantly differ across age groups. Several studies have been performed to study the effect of aging on gait. However, some of the parameters such as turn-to-sit and turn velocity are unique to this study.

TUG test is a routine clinical test used for balance assessment. The main parameter used is total duration of TUG (Kerr et al., 2007). In addition, research has been done on analysing the sit-to-walk performance across age groups and in people with increased fall risks. Kerr and colleagues (2007) studied 20 young, 18 healthy elderly and 18 elderly with increased fall risk on sit-to-stand performance. They showed that the duration of sit-to-walk was statistically different in people with fall risk. This is in accordance with the current study where no sit-to-stand measures showed differences between healthy young and elderly.

In contrast, a study by Buckley et.al (2009) showed significant differences in sit-to-walk duration in the elderly. These differences could be due to the definition of sit-to-walk. The measure calculated in this study is sit-to-stand ; but their study takes into account the gait initiation time also. Their results also showed overall longer duration to complete the task, which is in accordance to the present study. However, in this study, the total increase in duration of TUG was mainly due to increased turn duration and turn-to-sit duration.

None of the previous studies have looked at turning and turn-to-sit duration. It was observed that elderly people took a longer turn-to-sit and turning duration. Turning is an important event to study since it involves sudden change in the position of the COM. Turning includes moving one foot using the other foot as a pivot so as to rotate the body. Elderly tend to take more steps and an increased U-turn, which results in a longer turn duration.

Results of normal walk showed differences in stride length variability, stride velocity variability, stride length, RoM of arm swing. Gait variability has been studied extensively in the elderly and in neurodegenrative disorders (Espy, Yang, Bhatt, & Pai, 2010; Hausdorff et al., 1998; Kang & Dingwell, 2008). Kang studied gait variability in 8 old and 18 young healthy adults. The subjects walked on the treadmill in a self selected for 5 mins. It was found that there was no preference of walking speed in young and elderly. The results showed significant variability in stride time and stride length. The main limitation of the study is the subjects walked on a treadmill in a constant speed that must've influenced the variability in stride time and stride length. Also, Stride velocity variability couldn't be determined. On the other hand, Our study involved over-ground walking and showed significant stride velocity and stride length variability but not stride time variability. The variability in time increased in elderly but did not show statistical significance.

Shorter step length and larger gait velocity is associated with prevention of falls (Espy et al., 2010). Kimura et al studied 20 young and 52 elderly participants for differences in stride length and velocity. He found reduced stride lengths and speed in elderly compared to young, and no difference in gait cycle duration. This agrees with the results of the present study. Although not signficant, both studies show increased double support and stance phase duration in the elderly. None of the previous gait studied looked at range of motion of the arm swing in aging. Brujin et al. studied the use of arm swing in gait. They found that arm swing reduces the angular momentum and energy expenditure during walking. Although arm swings do not improve stability, they help to recover from a perturbation. Significant increase in RoM of arm swing was observed in the elderly indicating prevention mechanism for falls during locomotion. Arm swings are also seen to reduce the lateral movement of the CoM and lower the momentum required to counteract the swinging legs. (Arellano & Kram, 2011). Thus, reduced stride lengths and increased RoM arm swing are indicators of compensatory mechanisms used by the elderly, to prevent loss of balance during locomotion.

Tandem walking has not been studied as much as normal walking to determine age related differences. Verlinden and collagues (2013) assessed the risk of falls in the elderly by looking at tandem gait, turning, and normal walking. 1,500 adults over the age of 50 were recruited to perform tandem and normal walking. The results showed that measures associated with tandem walking such as number of steps and duration were different across age groups and is a marker for risk of falling. Speers et al (1998) studied tandem standing and walking to find age related differences. The success rate of completion of tandem walk was higher in the young group. Range of motion of the arm during tandem walking has never been studied before. It was observed that elderly extended their arms outward to stabilize themselves during tandem walk. ROM was significantly increased in elderly during tandem walking indicating that they use more of their arms to recover from perturbations by reducing the angular momentum. It was also observed that step width and arm swings play important roles in lateral stabilization of the CoM, apart from reducing the energy cost. Since tandem walking reduces the step width, increased RoM of the arm could have been used to compensate for it and to improve the lateral stabilization of CoM (Arellano & Kram, 2011).

CHAPTER 5

CORRELATION BETWEEN STANDING, POSTURE SHIFT AND GAIT

The various measures obtained from quiet standing, posture shifts, and gait were analyzed to see if they were correlated, which may help to better interpret overall differences in postural control observed across age groups. This may also help to understand how the novel measures derived from dynamic shifts are related to widely used tests for balance assessment.

CORRELATION BETWEEN QUIET STANDING AND POSTURE SHIFT

Each of the parameters derived from quiet standing in the eyes open condition were compared with every other measures obtained from posture shift. The pearson's correlation coefficient was determined for the young and the elderly group separately. A p-value less than 0.05 was considered to be significant.

Results. Some of the parameters of quiet standing in the eyes open condition and dynamic shifts showed significant correlations in both the young and elderly groups. The strength of the correlation was determined according to the classification given by Dancey and Reidy (2004). Pearson correlation coefficient between 0.6-0.7 was considered moderate correlation and 0.7-0.9 was considered strong correlation. Table 5.1 shows the correlations in the young and elderly along with the correlation strength.

In the young group, movement pathlength during dynamic shifts was moderately correlated with sway area, total excursion and total excursion AP during quiet standing. This indicates that higher total excursion observed during quiet standing reflects as increase in movement length during posture shifts. The subjects swayed more before achieving the target. The total excursion in ML direction was moderately correlated with unsteadiness.

In the elderly group, mean frequency was moderately correlated with initiation length, initiation time and movement length. The correlation was stronger in the AP direction. Similar to the results obtained for the younger group, total excursion in the ML direction was moderately correlated with unsteadiness.

From the results, it can be said that an increase in fluctuations of CoP delays the initiation time for movement. Unsteadiness is defined as the deviation from the target center over time during the hold phase. The results indicate that an increase in excursion and frequency during standing is indicative of difficulty in maintaining the CoP in a particular position during hold phase of the posture weight shifts.

Table 5.1

Strength of correlations between quiet standing (eyes open) and dynamic shifts measures. A correlation is considered moderate when Pearson's correlation coefficient r=0.6-0.7 or strong if r=0.7-0.9.

AGE GROUP	QUIET STANDING	DYNAMIC SHIFTS	CORRELATION STRENGTH
YOUNG	Sway area Total excursion Total excursion AP	Movement Length	Moderate
	Total excursion ML	Unsteadiness	Moderate
ELDERLY	Mean frequency	Initiation length Initiation time Movement length	Moderate
	Mean frequency	Initiation length Initiation time Movement length	Strong
	Total excursion Total excursion ML	Unsteadiness	Moderate

CORRELATION BETWEEN QUIET STANDING AND GAIT PARAMETERS

The measures obtained from quiet standing during eyes open condition was correlated with the gait measures obtained from Timed-up and go, tandem and normal walking tasks. Pearson's correlation coefficient was calculated and a p value of less than 0.05 was considered as significant correlation.

Results. In the young group, moderate correlations were observed between total excursion, total excursion AP and sway area of quiet standing and double support and stance time during gait (Table 5.2). An increase in sway area indicates weaker balance control. Thus subjects with an increased sway had increased double support time to stabilize the position of the COM. In addition, the young showed moderate correlation between turn-to-sit duration and sway area, total excursion and total excursion AP.

Table 5.2

Strength of correlations between quiet standing (eyes open) and gait measures (TUG, normal walk, and tandem walk). A correlation is considered moderate when Pearson's correlation coefficient r=0.6-0.7 or strong if r=0.7-0.9.

AGE GROUP	QUIET STANDING	GAIT	CORRELATION RANGE
YOUNG	Total excursion AP Total excursion Sway area	Double support time Turn-to-sit duration	Moderate
ELDERLY	Total excursion Total excursion AP	Stride length Peak horizontal trunk velocity (tandem)	Moderate
	Mean frequency Mean frequency AP	Cadence (Tandem) Peak trunk velocities	Strong
		Peak arm swing velocity	Strong

The elderly group had more measures showing significant correlations when compared to the young. Total excursion and total excursion AP showed moderate negative correlation with stride length. Increased CoP path length and decreased stride length are characteristics of deterioration of balance control. Strong correlations were observed between the measures from tandem walking and mean frequency, mean frequency AP. Increase in frequency escpecially in the AP direction was associated with increase in peak velocity of the arm swing and the trunk in horizontal, sagittal and frontal directions. This could be to compensate for the increased fluctuations of CoP by using a different kind of strategy using the trunk and arms by the elderly, to reduce the overall angular momentum of the body.

CORRELATION BETWEEN POSTURE SHIFT AND GAIT MEASURES

Similar to the correlations performed previously, Pearson's correlation coefficient was determined for gait and postural shift measures. A p-value less than 0.05 was considered significant correlation.

Result. Similar to the previous results, the younger group showed lesser number of correlations. In the young group, moderate correlations were observed between movement time and double support time. Increased double support time indicates reduced stability. Increase in movement time during shifts indicates the subject is more cautious in moving towards the targets.

In the elderly group, strong correlations were observed between initiation time, pathlength, Initiation velocity , movement pathlength; and peak arm swing velocity. The results show that arm swing plays an important role in dynamic stability. Also, the range of motion of arm swing moderately correlated with unsteadiness. This shows that people who show increased unsteadiness in dynamic shifts increase their arm swing during walking. This could possibly be a stabilizing mechanism to reduce the lateral CoM movement. Peak velocity of the trunk in the sagittal plane also showed strong correlation with initiation pathlength and movement pathlength (Table 5.3).

Table 5.3

Strength of correlations between dynamic shifts and gait measures (TUG, normal walk, and tandem walk). A correlation is considered moderate when Pearson's correlation coefficient r=0.6-0.7 or strong if r=0.7-0.9.

AGE GROUP	POSTURE SHIFT	GAIT	CORRELATION RANGE
YOUNG	Initiation time	Cadence	Moderate
	Movement time	Double support time	Moderate
ELDERLY	Initiation length Movement length Initiation time	Peak arm swing velocity	Strong
		Cadence	Strong
		Peak sagittal trunk velocity	Strong
	Unsteadiness	Range of motion of arm swing	Moderate

CHAPTER 6

CONCLUSIONS

The main aim of this thesis was to determine the effects of aging on balance control during standing and walking. Three different experimental protocols were designed to investigate different aspects of balance control to efficiently characterize changes in balance control due to aging. Furthermore, correlations between various balance measures obtained during different static and dynamic balance tasks were studied.

The first experiment based on quiet standing was used to determine the effects of age and visual input on balance control. The results of the task provided useful insights on the static balance control system and its deterioration with age. Mean velocity and total excursion were found to be the most reliable measures to assess balance during standing. Other important measures include sway area and mean frequency. One important conclusion from the study was that the control of balance in the AP direction heavily relies on visual information. Also, the eldery population tend to rely more on visual input for better balance control.

The second experiment required the subjects to reach for targets in different directions that required them to voluntarily shift their CoP outside the base of support. The results showed increased initiation time and pathlength, movement length, inaccuracy, and number of reentries in elderly participants. This increase could be associated with increased use of hip strategy when compared to the ankle strategy in the elderly group.

The third experiment tested balance control during walking. Three tasks: TUG, normal, and tandem walking were performed. Age affected turn duration, turn velocity, turn-to-sit duration and velocity during TUG. Shorter stride length and increased variabilties in stride length and stride velocity with increased age were observed during normal walking. Most elderly found the tandem walking difficult to perform. However, among the ones who completed the task, the ROM arm were significantly higher compared to the young adults. In the elderly group, ROM arm swing was higher during normal walking too. It can be concluded that arm swing and turn duration can be utilized to assess balance control, apart from stride related parameters.

Some interesting correlations were observed between the measures from all three different experimental tasks. The correlation between the measures were higher in elderly compared to that of the young group. The dynamic shift measures initiation pathlength and time, movement pathlength, and unsteadiness showed good correlations with mean velocity and sway area in the young; and mean frequency and mean frequency AP in the elderly. Very high correlations were observed in the elderly, between dynamic shift and gait measures, specifically for tandem walking. RoM of arm swing and peak arm swing velocity were positively correlated with initiation and movement parameters. This may imply that dynamic posture shifts paradigm has a potential use in assessing balance control.

However, very high confidence intervals were observed for the elderly compared to the young in all the measures tested. This may suggest that balance control in elderly is more affected in a manner that alters both static and dynamic balance.

Overall, this study emphasized on gait and balance measures that are often ignored, such as turning and arm swing measures. This study also found new measures, inaccuracy and number of re-entries, that were significantly different between the young and eldlerly. The results obtained from this study can further be used to compare people with increased fall risks with healthy elderly and young adults. Also, the experimental protocols and various measures used in the study may be helpful for better characterization of balance control in neurodegenerative diseases such as Parkinson’s and Hungtington’s disease.

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APPENDIX A

IRB APPROVAL LETTER

Knowledge Enterprise Development

Office of Research Integrity and Assurance

To:	James Abbas ISTB1
From:	Carol Johnston, Chair Biosci IRB
Date:	02/11/2013
Committee Action:	**Expedited Approval**
Approval Date:	02/11/2013
Review Type:	Expedited F4 F7
IRB Protocol #:	1302008793
Study Title:	Control of Posture and Walking in Able-Bodies Adults
Expiration Date:	02/10/2014

The above-referenced protocol was approved following expedited review by the Institutional Review Board.

It is the Principal Investigator's responsibility to obtain review and continued approval before the expiration date. You may not continue any research activity beyond the expiration date without approval by the Institutional Review Board.

Adverse Reactions: If any untoward incidents or severe reactions should develop as a result of this study, you are required to notify the Biosci IRB immediately. If necessary a member of the IRB will be assigned to look into the matter. If the problem is serious, approval may be withdrawn pending IRB review.

Amendments: If you wish to change any aspect of this study, such as the procedures, the consent forms, or the investigators, please communicate your requested changes to the Biosci IRB. The new procedure is not to be initiated until the IRB approval has been given.

Please retain a copy of this letter with your approved protocol.

APPENDIX B

DESCRIPTIVE STATISTICS

Table 1

Descriptive statistics for the quiet standing measures are given. Mean and standard deviation values are provided for four different conditions, young group during eyes open and eyes closed conditions and elderly group during eyes open and eyes closed conditions.

MEASURE	YOUNG EYES OPEN	YOUNG EYES CLOSED	ELDERLY EYES OPEN	ELDERLY EYES CLOSED
Sway area (sq. cm)	338.91±181.8	524.01±317.32	496.63±273	931±592.29
Total excursion (cm)	4.11±1.02	5.74±1.61	6.02±1.98	8.94±3.38
Total excursion AP (cm)	3.18±7.57	4.83±1.50	5.11±1.87	7.88±3.14
Total excursion ML (cm)	1.94±0.64	2.21±0.69	2.26±0.68	2.95±1.06
Mean velocity (cm/sec)	0.68±0.17	0.95±0.26	0.10±0.33	0.14±0.05
Mean velocity AP (cm/sec)	0.53±0.12	0.80±0.25	0.85±0.03	13.1±0.05
Mean velocity ML (cm/sec)	0.32±0.10	0.36±0.11	0.37±0.11	0.49±0.17
Mean frequency (Hz)	0.24±0.07	0.29±0.10	0.34±0.12	0.40±0.13
Mean frequency AP (Hz)	0.25±0.08	0.32±0.12	0.40±0.16	0.47±0.18
Mean frequency ML (Hz)	0.37±0.13	0.37±0.14	0.37±0.16	0.41±0.15
Mean distance (cm)	0.46±1.48	0.57±0.30	0.48±0.15	0.63±0.29
Mean distance AP (cm)	0.39±0.14	0.50±0.31	0.40±0.13	0.53±0.27
Mean distance ML (cm)	0.17±0.07	0.18±0.06	0.20±0.09	0.23±0.12
RMS distance (cm)	0.53±0.16	0.66±0.32	0.56±0.17	0.74±0.34
RMS distance AP (cm)	0.48±0.06	0.61±0.33	0.49±0.16	0.66±0.33
RMS distance ML(cm)	0.21±0.09	0.23±0.08	0.25±0.12	0.30±0.15
Area CE(sq. cm)	1.87±1.16	2.72±1.80	2.38±1.53	3.94±3.84
Area CC(sq. cm)	2.74±1.59	4.79±5.85	3.17±1.97	6.16±7.20

Table 2

Descriptive statistics for the dynamic shifts measures. Mean and standard deviation values are provided for young and elderly groups.

MEASURE	YOUNG	ELDERLY
Initiation time (sec)	0.29 ± 0.11	0.40 ± 0.35
Initiation path length (cm)	2.72 ± 1.85	3.95 ± 4.97
Initiation velocity (cm/s)	9.07 ± 3.56	9.09 ± 3.34
Movement time (sec)	0.11 ± 0.10	0.16 ± 0.17
Movement path length (cm)	1.34 ± 0.98	1.85 ± 1.78
Movement velocity (cm/s)	17.2 ± 7.70	15.2 ± 6.9
Peak velocity (cm/s)	27.8 ± 11.3	26.3 ± 10.4
Inaccuracy (cm)	0.56 ± 0.17	0.60 ± 0.20
Unsteadiness (cm)	0.23 ± 0.06	0.23 ± 0.79
Re-entries	0.59 ± 0.75	0.80 ± 1.06

Table 3

Descriptive statistics for the gait measures obtained from timed-up and go test. Mean and standard deviation values are provided for young and elderly groups.

MEASURE	YOUNG	ELDERLY
Total duration (s)	16.70 ± 1.72	20.17 ± 2.88
Turn duration (s)	1.94 ± 0.38	2.28 ± 0.47
Turn number of steps	4.11 ± 1.04	4.48 ± 0.98
Turn peak velocity (m/s)	186.8 ± 37.97	161.90 ± 31.9
Turn step time before turn(s)	0.53 ± 0.04	0.56 ± 0.06
Sit-to-stand duration(s)	2.21 ± 0.39	2.31 ± 0.76
Sit-to-stand peak velocity (m/s)	129.8 ± 55	123.8 ± 62.01
Sit-to-stand trunk ROM(deg)	38.36 ± 10.89	36.69 ± 9.05
Turn-to-sit peak turn velocity (m/s)	208.14 ± 33.1	163.34 ± 35.05
Turn-to-sit duration (s)	3.43 ± 0.46	4.33 ± 0.82
Turn-to-sit trunk ROM (deg)	29.78 ± 12.69	24.64 ± 8.82

Table 4

Descriptive statistics for the gait measures derived from long walking. Mean and standard deviation values are provided for young and elderly groups.

MEASURE	YOUNG	ELDERLY
Stride length (m)	1.51 ± 0.09	1.44 ± 0.11
Stride length COV (m)	0.01 ± 0.00	0.02 ± 0.00
Stride velocity (m/s)	1.45 ± 0.12	1.35 ± 0.13
Stride velocity COV (m/s)	0.02 ± 0.00	0.03 ± 0.01
Cadence (steps/min)	115.49 ± 7.79	112.27 ± 9.42
Gait cycle time (s)	1.04 ± 0.07	1.07 ± 0.09
Double support %	19.8 ± 3.95	22.53 ± 4.93
Stance %	59.90 ± 1.97	61.25 ± 2.46
Swing %	40.09 ± 1.97	38.74 ± 2.46
ROM arm swing (deg)	20.52 ± 10.7	28.54 ± 12.44

Table 6

Descriptive statistics for the gait measures derived from tandem walking. Mean and standard deviation values are provided for young and elderly groups.

MEASURE	YOUNG	ELDERLY
Cadence (Steps/min)	62.73 ± 13.17	63.09 ± 17.00
Peak arm velocity(m/s)	57.7 ± 17.7	73.36 ± 42.63
Arm swing ROM(deg)	6.44 ± 2.41	11.30 ± 6.96
Trunk horizontal ROM(deg)	6.15 ± 2.61	7.33 ± 2.78
Trunk sagittal ROM(deg)	4.09 ± 1.30	4.55 ± 1.74
Trunk frontal ROM(deg)	7.21 ± 2.57	7.53 ± 2.40
Trunk horizontal velocity(m/s)	17.58 ± 6.03	21.30 ± 9.02
Trunk sagittal velocity(m/s)	15.81 ± 5.50	17.25 ± 7.62
Trunk frontal velocity(m/s)	22.08 ± 6.57	23.90 ± 7.89

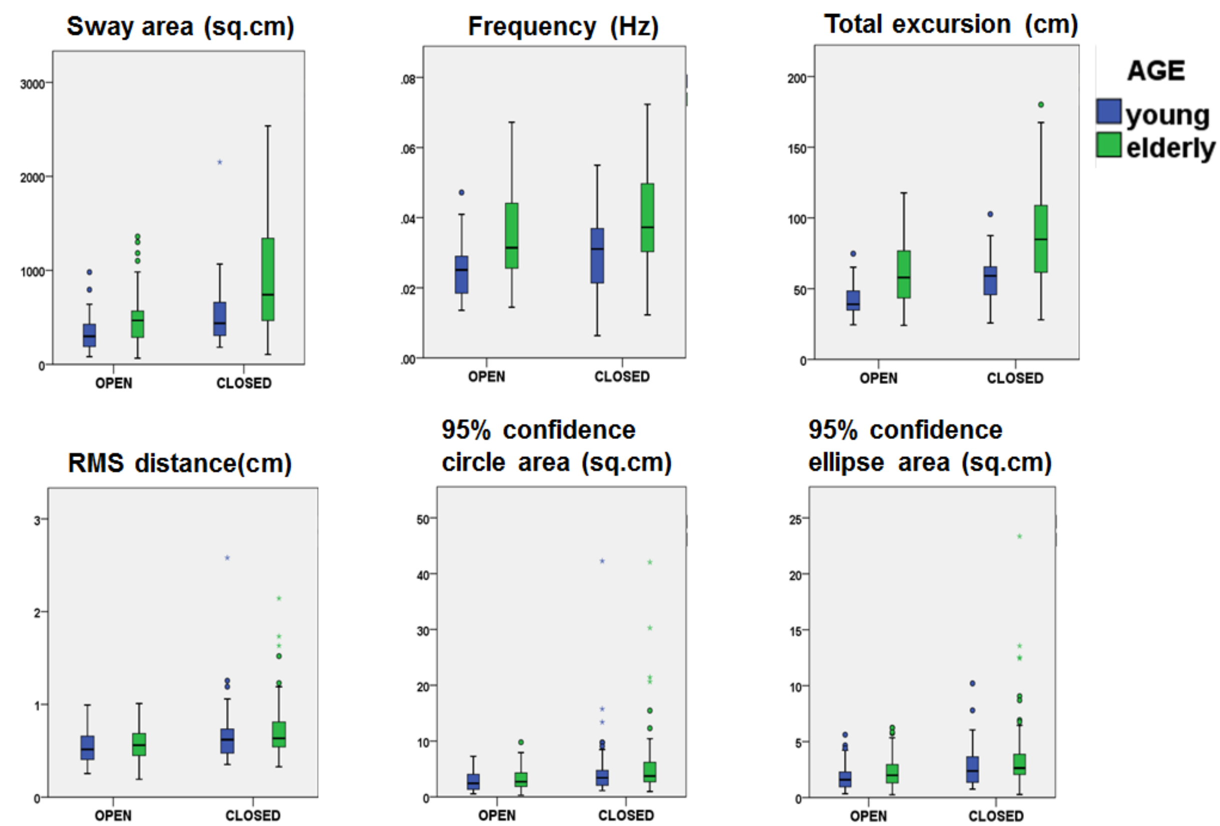

Figure 1. Box plots[1] showing the distribution of quiet standing measures with change in age and visual inputs. Blue boxes indicate eyes open condition and green boxes indicate eyes closed condition.

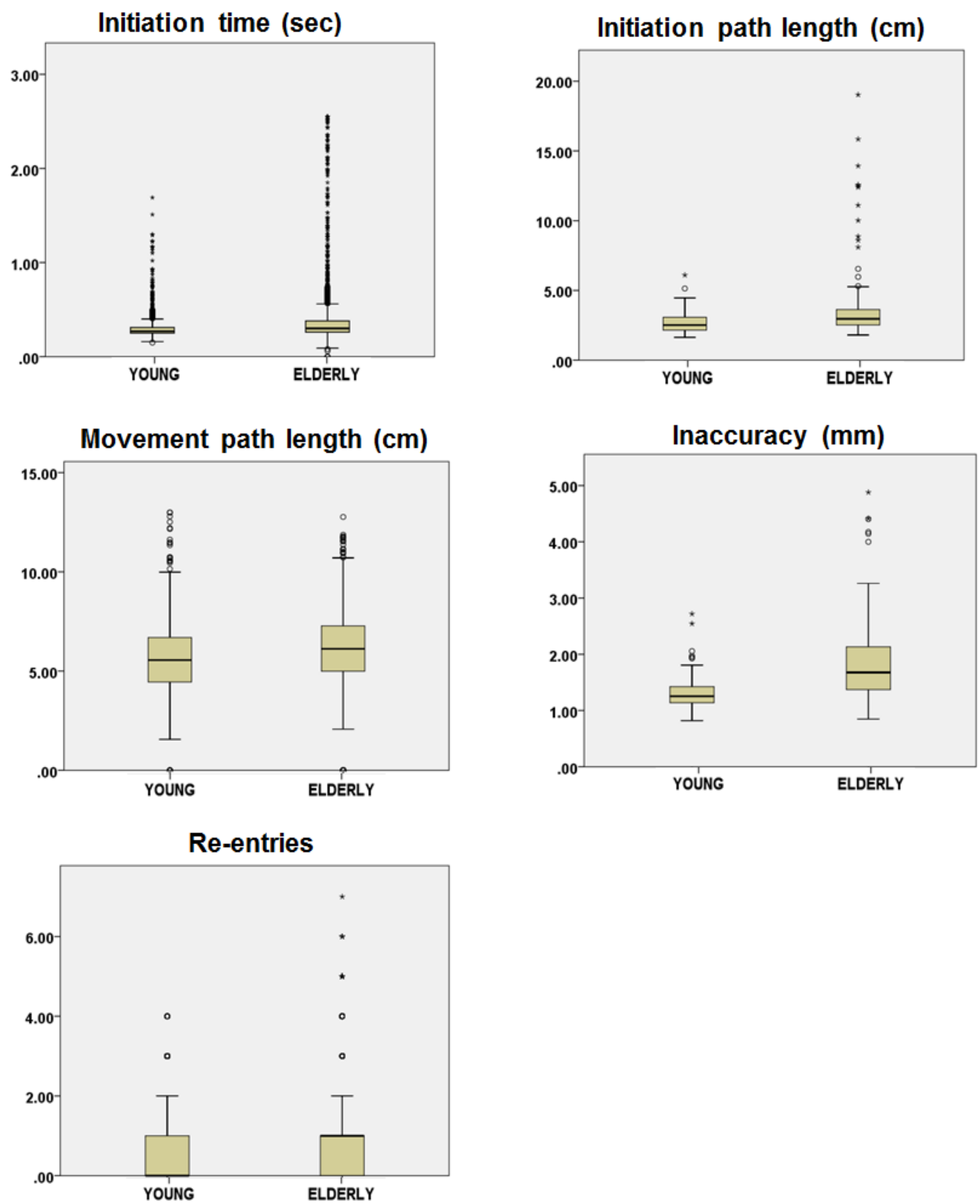

Figure 2. Box plots[1] showing the distribution of dynamic shifts measures that showed significant difference ($p < 0.05$) between the young and elderly groups. The distributions were right skewed. Hence a logarithmic transform was used prior to ANOVA.

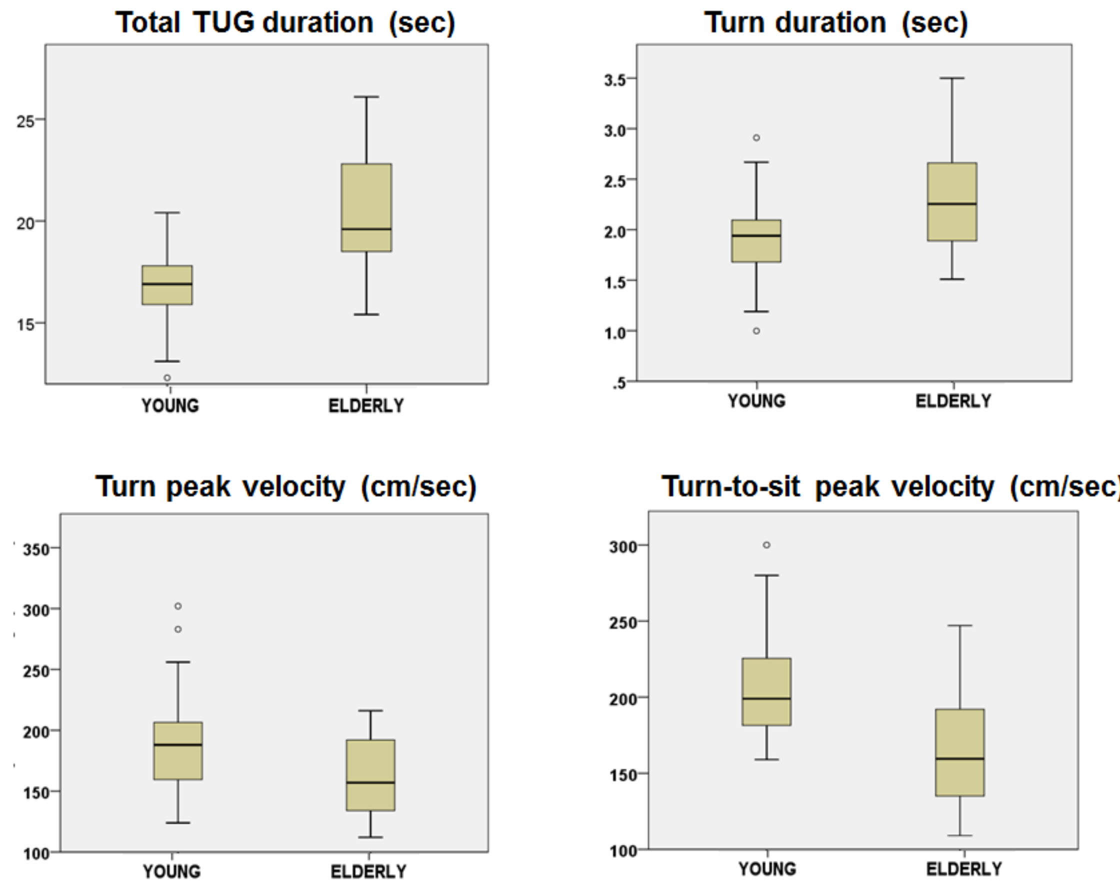

Figure 3. Box plots[1] showing the distribution of timed up and go measures that showed significant difference ($p < 0.05$) between the young and elderly groups. The distributions were right skewed. Hence a logarithmic transform was used prior to ANOVA.

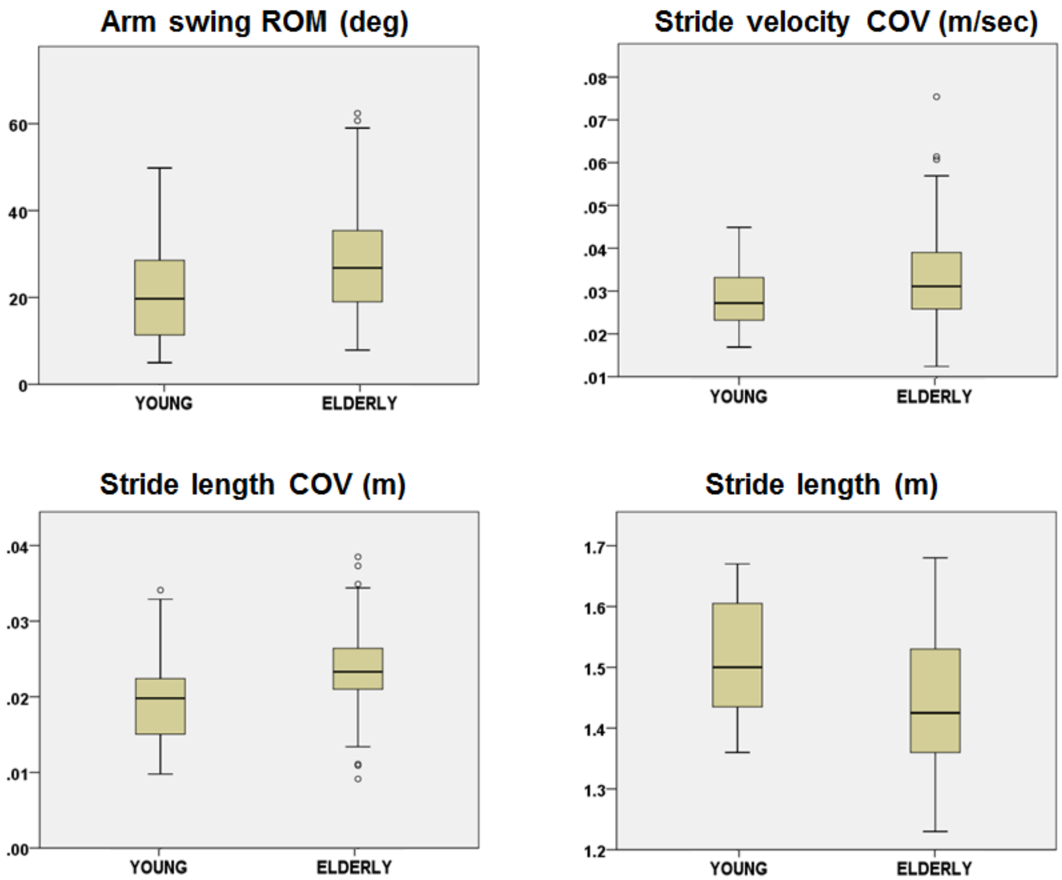

Figure 4. Box plots[1] showing the distribution of long walk measures that showed significant difference ($p < 0.05$) between the young and elderly groups. The distributions were right skewed. Hence a logarithmic transform was used prior to ANOVA.

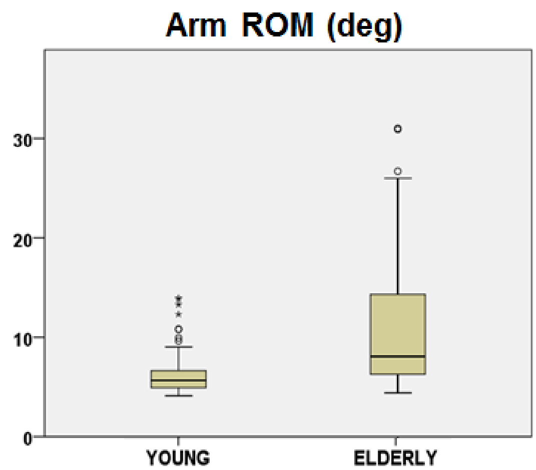

Figure 5. Box plots[1] showing the distribution of arm swing ROM during tandem walk that showed significant difference ($p < 0.05$) between the young and elderly groups. The distributions were right skewed. Hence a logarithmic transform was used prior to ANOVA.

Notes: [1]The circles represent outliers and the stars represent extreme outliers. The dark line at the center of each box represents the median of the distribution. The bottom of the box is the 25th percentile: 25% of the data below this line. The top of the box is the 75th percentile. The vertical lines that extend on either side of the box extend up to 1.5 times the length of the box. Any values outside these lines are considered as outliers.