Understanding Variability in Older Adults using Inertial Sensors

Rahul Soangra

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Thurmon E. Lockhart
Stefan M. Duma
Karen A. Roberto
Joel D. Stitzel
Andrew R. Kemper

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ABSTRACT

Falls are the most frequent cause of unintentional injuries among older adults; afflicting 30 percent of persons aged 65 and older and more than 50 percent of persons aged 85 and older. There is a serious need for strategies to prevent falls in elderly individuals, but an important challenge in fall prevention is the paucity of objective evidence regarding the mechanisms that lead directly to falls. There exists no mechanisms about how to predict and manage elderly falls, which has multifactorial risk factors associated with its occurrence in the elderly. As the U.S. population continues to age, both the number of falls as well as the cost of treatment of fall injuries will continue to grow. Decades of research in fall prevention has not led to a decrease in the fall incidence; thus new strategies need to be introduced to understand and prevent falls.

Aging reduces the adaptability of various physical and environmental stressors that hinder stability and balance maintenance and may therefore result in a fall. Movement variability in an individual's task performance can be used to assess the limitations of the movement control system. Maintaining variation in movement engenders flexible and adaptable modalities for elderly individuals to prevent falls in an unpredictable and ever changing external environment. Conversely, excessive variability of movement may drive the control system closer to its stability limits during balance and walking tasks.

Accordingly, inertial sensors are an emerging wearable technology that can facilitate noninvasive monitoring of fall prone individuals in clinical settings. This research examined the potential of inertial sensors for use in clinical settings, and evaluated their effectiveness in comparison to mature laboratory systems (i.e., force platform and camera system). Study findings showed a relationship between movement variability and fall risk among healthy young and older adults. Further, the outcomes of this work translates to the clinical environment to better understand the health status (leading to frailty) of cardiac patients; reflected by the underlying adaptability of the control system, but requires further improvements if to be used as robust clinical tool.

This research provides the groundwork for rapid clinical assessments in which its validity and robustness should be investigated in future efforts.

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CHAPTER 1: Overview

Rationale

Falls occur in one of three older adults, with its frequency increasing with age (Kenny, Romero-Ortuno, & Cogan, 2009). Falls are one of the leading causes for admissions to nursing homes (Masud & Morris, 2001) and the mean cost of injuries range from $3,476-10,749 per faller (Davis et al., 2010a, 2010b). Frequency of falls are projected to double by 2030 (Kannus, Palvanen, Niemi, & Parkkari, 2007) with the cost of direct and indirect injuries associated with falls projected to be $54.9 billion by the year 2020.

The most common manifestation of aging is the progressive degeneration of sensory, musculoskeletal (Calne, Eisen, & Meneilly, 1991) and cognitive systems. With aging, there is a degeneration of all sensory systems such as diminution of the axon population in optic nerves (Johnson, Miao, & Sadun, 1987), loss of spatial contrast sensitivity (Sekuler, Hutman, & Owsley, 1980), reduction in hair cells of semicircular canals (Rosenhall & Rubin, 1975), higher proprioceptive thresholds and decreased positional sense (Meeuwsen, Sawicki, & Stelmach, 1993; Skinner, Barrack, & Cook, 1984; Whanger & Wang, 1974), which can affect gait and biomechanics of slips leading to falls (T.E. Lockhart, Smith, & Woldstad, 2005; T. E. Lockhart, Woldstad, & Smith, 2003). Analogous declinations occur in muscle properties that influence movement execution including a reduced muscle cross-sectional area (Kent-Braun & Ng, 1999), with a reduction in strength and fiber type distribution (Lexell, 1995) that diminishes movement speed; and a coincident decrease in motor units (Galea, 1996). Likewise, the cognitive capacity of brain degrades attenuating

cognitive functions such as attention, executive function and, memory. These age-related deficits in sensory, muscular and cognitive systems are pronounced during dual-tasking (Hausdorff, Schweiger, Herman, Yogev-Seligmann, & Giladi, 2008; Plummer-D'Amato, Altmann, & Reilly, 2011) and are associated with increased fall risk in the elderly. Given there is no consensus in the approach to predict and manage elderly falls, this study focuses on variability associated with increased fall risk in the elderly. A review of the literature suggests that the parameters used in earlier studies to identify fall risk are seemingly inadequate to evince cogent fall-related characteristics.

The homogeneity of fall risk identification tools reinforces the need for novel linear and nonlinear tools based on movement variability associated with fall risk. Accordingly, dual-task interference could help identify potential predictors of gait related cognitive-motor interference through adaptive changes in variability which may adversely affect slip and fall risk.

Similarly, diagnostic capabilities in clinical setting require robust, objective methods to identify fall risk as well. Indeed, inertial measurement units (IMUs) provide an avenue to assess fall risk in a more quantitative manner in clinics. Furthermore, fall risk is associated with cardiovascular disorders in cardiovascular disease (CVD) patients (Cronin & Kenny, 2010; Dey, Stout, & Kenny, 1997), particularly frail individuals, and is reportedly, an apt predictor of disability and adverse outcomes in older adults undergoing cardiac surgery (Sundermann et al., 2011). Early identification of fall risk in this patient population has the potential to significantly reduce disability, morbidity and mortality.

In summary, falls in older adults continue to rise and are the leading cause of fatal and non-fatal injuries in the US. Akin to this, aging is coincident with increased fall risks. Apropos to movement variability in the high fall risk elderly, the dual task

paradigm (or multi-tasking) may be one method to evoke inherent fall-related characteristics. Moreover, nonlinear signal analysis methods have the potential to quantify loss of balance, stability, movement control and variability in humans. Since, falls that are experienced by the elderly have a multitude of risk factors associated with it: new insights relating variability of movements are needed to overcome the clinical diagnostic limitations of fall risk assessments using inertial sensors.

Specific Aims and Hypotheses

The primary aim of this study was to investigate the relationship between dual-tasking and fall risks. The study involves two groups of individuals with (known) different slip and fall risks. The effects of dual-tasking on gait parameters and slip initiation parameters were evaluated. This work will help establish an objective link between movement variability and fall risk. This study has evaluated the efficacy of inertial sensors in the objective assessment of gait and posture in laboratory and clinical environments.

Study I: Laboratory Based Slip Study

Specific Aim 1: To determine the effects of a dual-task condition on slip and fall risk in older individuals (illustrated as study I in Figure 1).

Hypothesis 1: Dual-task scenarios while walking will affect gait characteristics and may increase the slip initiation (such as RCOF, HCV), recovery (SDI, SDII) and other fall related characteristics in elderly individuals

Study II: Laboratory Based Movement Variability Study using Inertial Sensors

Specific Aim 2: To investigate if inertial sensors could be a useful tool in assessing linear and non-linear movement variability in young and old adults (illustrated as study II in Figure 1).

Hypothesis 2a: Linear variability in gait speed, postural transition times, and non-linear variability in postural stability, and dynamic stability will be significantly different in younger and older adults.

Hypothesis 2b: Higher variability will be seen in elderly adults irrespective of dual-task condition and the direction of progression on a treadmill.

Study III: Clinical Study

Utilizing the information acquired from the laboratory, the clinical study validated the link between variability and fall risk as measured by IMUs based on gait and posture measures (illustrated as study III in Figure 2).

Specific Aim 3: To determine if fall risk and movement variability is associated with post-operative adverse outcomes in cardiovascular disease patients.

Hypothesis 3: Gait and postural characteristics such as gait speed, postural transition times, and postural stability parameters in CVD patients derived from the IMU can

help identify a subset of high fall risk attributes that are associated with adverse health outcomes.

Hypothesis 3b: Linear and nonlinear postural variability will identify the subset of CVD patients with high fall risk and adverse post-operative outcomes.
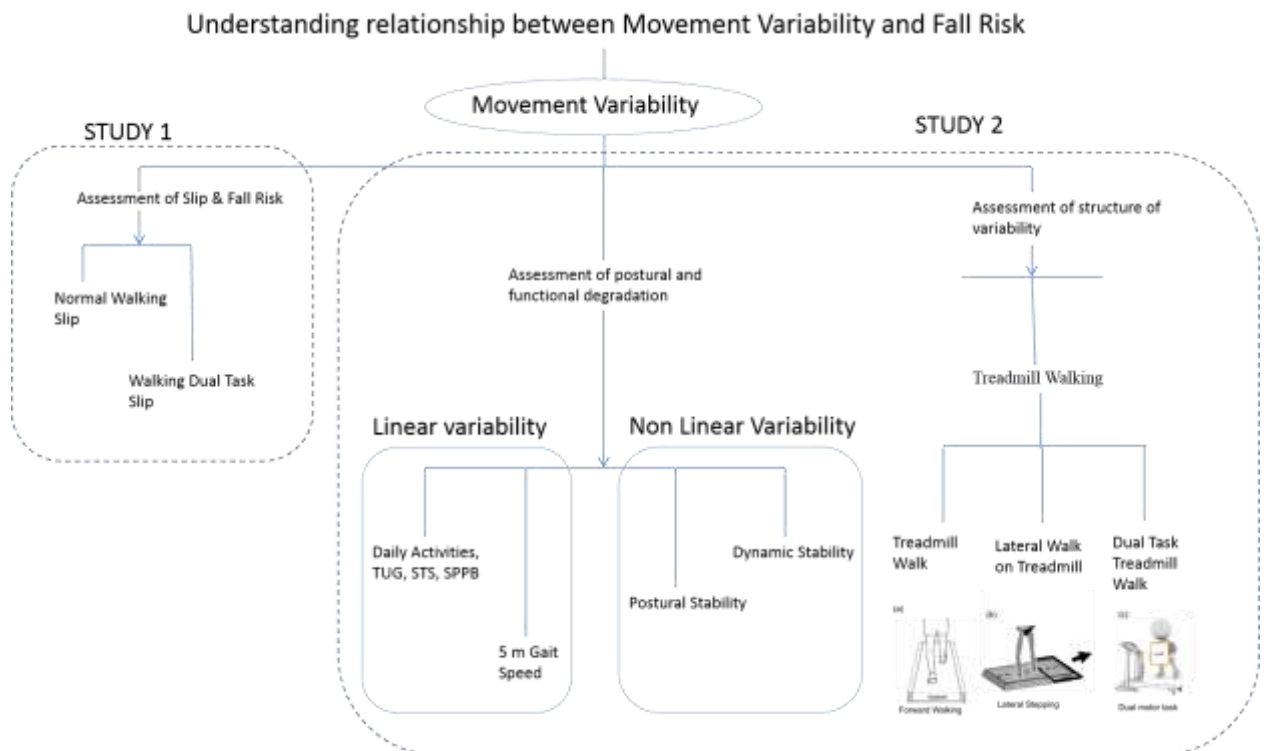
Figure 1. Schematic diagram of laboratory based studies Study I and Study II
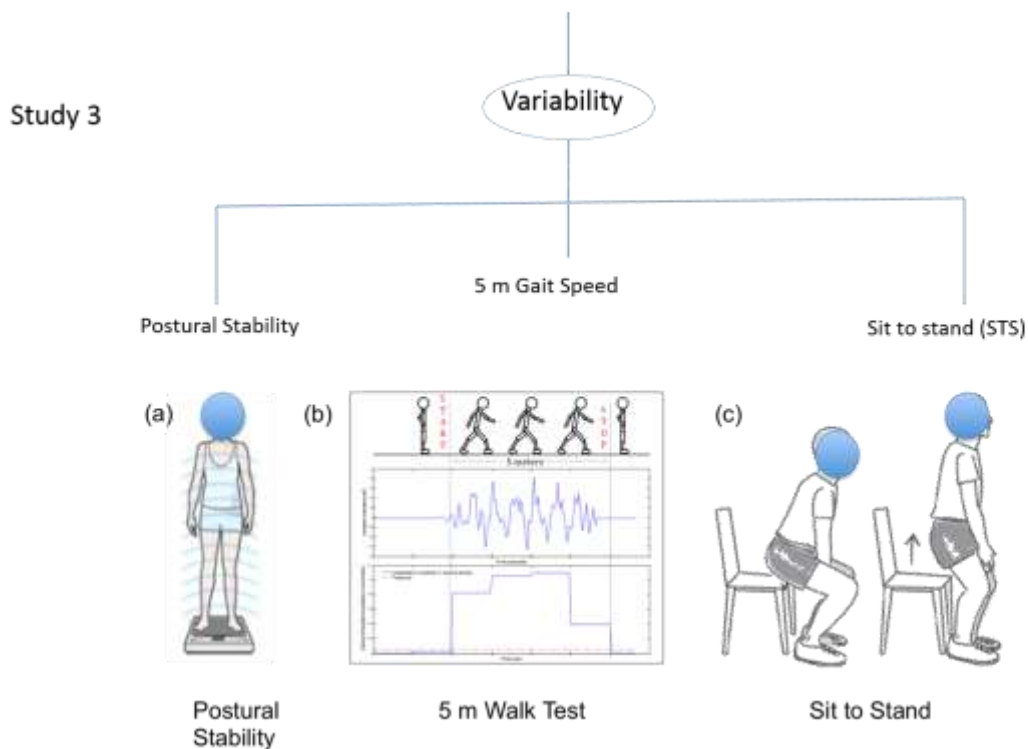
Figure 2. Schematic diagram of study performed in clinical setting

In summary, this research was intended to facilitate the understanding of movement variability associated with one's fall risk. The primary focus was to differentiate fall-related characteristics of elderly individuals via dual-task induced variability; the development of portable, non-invasive variability assessment clinical tools; and to further validate these variability tools in the assessment of fall risk and post-operative adverse outcomes in cardiovascular disease patients undergoing surgery. Specifically, two laboratory-based studies were designed (Study I and Study II) to achieve Specific Aims 1 and 2. Outcomes of this work are considered a critical step toward more efficient and effective point of care test systems and more targeted interventions, while future research will need to expand and validate these outcomes. The document is organized as follows: Chapter 2 presents a literature review; Chapter 3 describes the effects of dual task on slip and fall risk in healthy young and old adults (Specific Aim

1); Chapter 4 reports on the use of linear and nonlinear tools to assess variability in healthy subjects using inertial sensors; Chapter 5 describes the use of clinical variability assessment tools to identify fall risk and post-operative adverse outcomes in cardiovascular disease patients in clinical environments; Chapter 6 summarizes the major findings, discusses their practical implications, and provides suggestions for future research.

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CHAPTER 2: Literature Review

An overview of falls in elderly persons

Currently, there are 40 million people in the US aged 65 and above, and it is expected to double and reach 89 million by the year 2050 ("2012 National Population Projections,"). Falls occur in one of three older adults and its frequency increases with aging (Rose A. Kenny, Romero-Ortuno, & Cogan, 2009). Indeed, falls are the leading cause for admissions to nursing homes (Masud & Morris, 2001) and the mean cost of injuries caused by falls ranges from USD 3,476-10,749 per faller (Davis et al., 2010a, 2010b). Approximately, 2.2 million elderly individuals underwent treatment in emergency departments in 2009 for nonfatal fall injuries and around 581,000 were hospitalized (*National Center for Injury Prevention and Control*, 2010). In addition to financial losses, the fallers have to cope with physical and psychological trauma, loss of independence, disability, nursing home admissions, medical expenses and, death (Stevens, Corso, Finkelstein, & Miller, 2006). Thus, these injuries reduce the mobility of elderly individuals (B. H. Alexander, Rivara, & Wolf, 1992; Sterling, O'Connor, & Bonadies, 2001) and pose a significant social and economic cost to society as a whole. Fall risk increases with ageing. About 82% of fall related fatalities in 2008 were reported in the age group of 65 years and above (Stevens & Anne Rudd, 2010). With increasing falls the cost associated with falls is rising. The direct medical costs alone for falls was around $19 billion in 2000 (Stevens et al., 2006) and is projected to be about $ 54.9 billion by the year 2020 (*National Center for Injury Prevention and Control*, 2010). Thus, it becomes imperative to determine fall risk in elderly individuals and introduce early fall interventions among them. Fall may result from multitude of interacting factors and yet sometimes may result from a single cause

such as syncope (Campbell, Borrie, & Spears, 1989). Impairments in gait and balance being the most common risk factor leading to falls (Davis et al., 2010a, 2010b).

Fall Risk Assessments

Fall occurrence in elderly have devastating consequences in terms of morbidity, mortality, and loss of independence. Fall risk assessment is a widely used clinical practice for screening patients upon admission for regular examination in various clinical settings (Perell et al., 2001). The standard test involves collection of health and medical status information from patients self-report or by the judgments of medical staff. These subjective and visual fall risk assessments are typically preferred because they are short, fast, non-intrusive and require less skill to execute. However, as these assessments are dependent on observation and judgments of clinical staff or on the self-report of the patient, the reliability and objectivity of these methods is questionable (Dowding & Thompson, 2003). Various clinical fall risk assessment methods adopted are Morse Fall Scale (MFS) (Morse, Morse, & Tylko, 2010), Hendric II (Hendrich, 1988), St Thomas Risk Assessment Tool in Falling Elderly Inpatients (STRATIFY) (Oliver, Britton, Seed, Martin, & Hopper, 1997), Downton Fall Risk Index (DFRI) (Nyberg & Gustafson, 1996), Fall Risk Assessment tool (FRAT) (MacAvoy, Skinner, & Hines, 1996), Elderly Fall Screening Test (EFST) (Cwikel, Fried, Biderman, & Galinsky, 1998), Nursing care dependency Scale (NCD) (Dijkstra, Buist, Moorer, & Dassen, 1999), Spartanburg Fall Risk Assessment Tool (SFRAT) (Robey-Williams et al., 2007), Fall Efficacy Scale (FES) (M. E. Tinetti et al., 1994; M. E. Tinetti, Richman, & Powell, 1990), Fall Efficacy Scale International (FES-I) (Yardley, 2005), Activity Specific Balance Confidence (ABC) (Powell & Myers, 1995).

Some clinics use functional fall risk assessment methods, which reflect capability of patients to perform daily activities and also utilizes mechanisms of balance control in them. Timed Get-up and Go (GUG) (Mathias, Nayak, & Isaacs, 1986), Expanded Timed Get-up and Go (ETGUG) (Wall, Bell, Campbell, & Davis, 2000), Performance Oriented Mobility Assessment (POMA) (M. E. Tinetti, 1986), Guralnik Test Battery (GTB) (Guralnik et al., 1994), Berg Balance Test(Berg, 1989), Dynamic Gait Index (DGI) (Shumway-Cook & Woollacott, 1995), Physical Performance Test (PPT) (Reuben & Siu, 1990), Gait Abnormality Rating Scale (GARS) (Wolfson, Whipple, Amerman, & Tobin, 1990), modified GARS (GARS-M) (VanSwearingen, Paschal, Bonino, & Yang, 1996), and Physiologic Profile Assessment (PPA) (Lord, Menz, & Tiedemann, 2003) are some of the functional fall risk assessment tools used by the clinicians.

Locomotion and attention

In normal walking condition with highly predictable surroundings, locomotor synergies involving several lower extremity muscles may take place at lower spinal levels with neural circuitry tuned by local loops of assistance or self-organizing processes generated in coordinative networks. These models of network of interneurons (central pattern generators) produce rhythmical, patterned locomotor movements (S Grillner, 1975; Sten Grillner, 1981; Pearson, 1976). Additionally, afferent information is used when ongoing regulation is required or modification of stereotypical pattern is required (such as obstacles and change in direction) supraspinal involvement is necessary to perform movements adapted to the environmental context. If higher cognitive inputs are necessary for controlling and regulating gait, less stable walking movements could require more attention than a stable walking movements. When an unstable movement is identified or achieved (here walking condition), a corrective response needs to be subsequently organized at

supraspinal level (Stelmach & Worringham, 1985; Teasdale et al., 1992; Teasdale, Bard, LaRue, & Fleury, 1993).

Locomotion is essential to meet individual's goal and commonly involves performing of concurrent cognitive tasks while walking, such as recalling what to shop, where to go next, or attending a conversation with someone walking or on phone. Gait performance places continuous demands on sensory and cognitive systems (Sheridan & Hausdorff, 2007). Brain imaging studies have revealed that there is activation in higher cognitive control areas of brain during walking (Bakker et al., 2008; Harada, Miyai, Suzuki, & Kubota, 2009; Iseki, Hanakawa, Shinozaki, Nankaku, & Fukuyama, 2008; Miyai et al., 2001; Suzuki, Miyai, Ono, & Kubota, 2008). Thus it is seen that many falls in older persons occur not during normal walking, but rather when they are walking and performing a secondary task as well as talking. Fall risk are higher in older adults with impaired cognitive function ("Guideline for the prevention of falls in older persons. American Geriatrics Society, British Geriatrics Society, and American Academy of Orthopaedic Surgeons Panel on Falls Prevention," 2001) along with decline in balance and stability, when subjects are asked to walk with dual-tasking (Bloem, Valkenburg, Slabbekoorn, & van Dijk, 2001; Sheridan, Solomont, Kowall, & Hausdorff, 2003; Woollacott & Shumway-Cook, 2002).

Attention: Attention is an ability to focus one's cognitive resources to selectively process certain information from the environment while ignoring the irrelevant input (McDowd, 2007). It is dynamic executive function which is driven by sensory perception and the need to select a preferred stimulus for a particular action while ignoring the unnecessary and irrelevant information. There are three types of attention: 1. Selective (the ability to focus on single relevant stimulus while ignoring

the irrelevant stimuli), 2. Sustained (the maintenance of focused attention over an extended period of time); 3. Divided attention (the ability to focus on several relevant stimuli simultaneously) (Baddeley, Baddeley, Bucks, & Wilcock, 2001; Perry & Hodges, 1999). Attentional capacity is a network function that requires intact perception in humans. Damage or loss of posterior higher cortical sensory integration may disrupt executive capacity in the brain (Sheridan et al., 2003). Attention allocation in humans has underlying assumptions: 1) there is limited central processing capacity, 2) performing a task requires certain part (attention) of the limited processing capacity within the central nervous system, 3) in dual-tasking when both task share the processing capacity, the performance in one or both task gets disturbed if limited central processing capacity is exceeded (Kahneman, 1973; Parasuraman, 1985).

Executive Function (EF) refers to cognitive processes dealing with novelty, planning and implementation of strategies for motor performance such as initiation of gait, or intention of action. Actions are defined as goal directed behaviors involving movement (Jahanshahi, 1998). Dual tasking relies upon executive function and the ability to divide attention (Della Sala, Baddeley, Papagno, & Spinnler, 1995). EF orchestrate goal-directed activities and allocate attention among competing tasks (Royall et al., 2002). An ability to divide attention (e.g. dual-tasking) is an example of EF task. Some researchers have used neuroimaging techniques and reported that previously learned actions do not use executive function such as attention (Jahanshahi, 1998; Shallice & Burgess, 1996). Conversely, gait is a complex motor task; which is a learned action utilizing cognitive processes (especially executive function) (Springer et al., 2006). There are supporting evidences that executive function moderates the influence of motor and sensory impairments leading to falls

(Rapport, Hanks, Millis, & Deshpande, 1998). Some studies have reported decreased EF in fallers (Di Fabio, Kurszewski, Jorgenson, & Kunz, 2004; Lord & Fitzpatrick, 2001; Rapport et al., 1998). Existing association of EF and movement variability with evidences of association of cognitive loading increased in the fallers suggest further that EF changes play crucial role in the causal pathways of falls. Impairment in cognitive domains may prevent fallers in allocating attentional resources appropriately to their balance and gait, and further reduce their ability to confront and adapt to challenging environments (e.g., obstacles, turning gait, uneven path), and may increase the risk of fall. Interestingly, fallers have been found to have problems with EF and dual-tasking in gait (Cocchini et al., 2004; Hausdorff, Balash, & Giladi, 2003; Sheridan et al., 2003; Yogev et al., 2005). Apparently, routine walking and walking derived markers of fall risk are significantly associated with performance of executive function (Cocchini et al., 2004; Hausdorff, Yogev, Springer, Simon, & Giladi, 2005; Sheridan et al., 2003). In idiopathic elderly fallers such as Parkinson's disease patients (primarily a motor disorder), and Alzheimer's disease patients (primarily a cognitive disorder), it is seen that there is decrement in walking ability with dual tasking (Bloem et al., 2001; Sheridan et al., 2003; Woollacott & Shumway-Cook, 2002). Probably detrimental changes in executive function capabilities in them impairs juggling of multiple tasks and the regulation of complex cognitive responses during dual-tasking (Cocchini et al., 2004; Sheridan et al., 2003). Executive function measures are also found to be significantly correlated with gait variability during dual tasking, but not when performing usual walking (Yogev et al., 2005). This demonstrates that gait variability and rhythmicity are automatic to some extent and do not require attention. Accordingly, there is decline in EF in PD patients, which exacerbates effects of dual tasking on gait and potentially increases variability and fall risk.

Dual Task and Gait Variability

Dual task methods can help to determine the cognitive demand of gait control and has been used by many researchers (Woollacott & Shumway-Cook, 2002; Yogev-Seligmann, Hausdorff, & Giladi, 2008). Dual-task related gait changes have been reported amongst several populations however firm conclusions are lacking.

Dual tasking in Alzheimer Disease Patients:

Gait speed is significantly reduced and, variability increased with dual-task walking in Alzheimer Disease (AD) patients (Sheridan et al., 2003). Divided attention during dual-tasking markedly impairs the ability of AD patients to regulate the stride-to-stride variations in gait timing, which explains predilection of AD patients to falling (Sheridan et al., 2003). During walking, regulation of stride-to-stride timing is an important feature that reflects fall risk. This gait cycle timing is regulated by basal ganglia and prefrontal and frontal cortices to some extent (G. E. Alexander, DeLong, & Strick, 1986; Jahanshahi, 1998; Richer, Chouinard, & Rouleau, 1999). Particularly, executive dysfunction and inattention, pervasive in AD, is reported to be responsible for detrimental timing and sequencing of motor activity. Accordingly, gait variability, a measure of unsteadiness, is significantly increased when attention is divided in AD patients (Sheridan et al., 2003). Thus stride-to-stride variability is considered a measure of instability associated with fall risk (Blin, Ferrandez, & Serratrice, 1990; Hausdorff, Cudkowicz, Firtion, Wei, & Goldberger, 1998; Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Maki, 1997; H. Visser, 1983). Hausdroff and coworkers showed that coefficient of variation (CV) of stride time is easy to implement in clinical routines for geriatric outpatients and was a dependable marker of gait control (Newell & Corcos, 1993) and is a useful clinical index of gait

steadiness (Hausdorff et al., 1997; Hausdorff, Rios, & Edelberg, 2001). Furthermore Hausdroff reported that stride time variability undoubtedly reflects the control of rhythmic stepping mechanism (Gabell & Nayak, 1984).

Dual Task in Older Fallers:

The inability to perform in dual tasking has also been reported to increase risk of falls in older adults (Lundin-Olsson, Nyberg, & Gustafson, 1997). Age-related deficits are pronounced during dual-tasking (Hausdorff, Schweiger, Herman, Yogev-Seligmann, & Giladi, 2008; Plummer-D'Amato, Altmann, & Reilly, 2011). In older adults impaired walking performance is associated with impaired cognitive performance in dual task walking (Hollman, Kovash, Kubik, & Linbo, 2007). It is reported that frail older adults have higher variability of stride time while backwards counting (Beauchet, Dubost, Aminian, Gonthier, & Kressig, 2005), which has been reported to be related to its rhythmic character and not wholly attributable to attentional load. Although numerous studies have reported walking speed negatively influences stride time variability (Brach, Berthold, Craik, VanSwearingen, & Newman, 2001; Danion, Varraine, Bonnard, & Pailhous, 2003; Heiderscheit, 2000; Van Emmerik, Wagenaar, Winogrodzka, & Wolters, 1999), despite suggesting that stride time is independent of walking speed (Heiderscheit, 2000; Van Emmerik et al., 1999). Hollman and colleagues (Hollman et al., 2007) reported that older adults reduced their gait velocity by 20% and young adults from 7-8% in dual task walking condition. They also found increase in 1.3-1.5% increase in stride-to-stride variability in young adults and 2.9% in older adults, demonstrating that attention demanding tasks have a destabilizing effect on gait and that attentional processes are involved in walking. Dual-task is associated with slow speed may be because control of gait speed may involve higher order cognitive systems. Some researchers have shown that gait speed is dependent on

Prefrontal Cortex (PFC) activation (Harada et al., 2009; Suzuki et al., 2004) and others have linked gait speed with executive function (Alvarez & Emory, 2006). Thus walking and secondary task, compete for these shared neural networks, this leads to cognitive motor interference (CMI) (Klingberg, 2000). Gait speed and stride length are probably controlled by cortico-basal ganglia circuit through thalamus (Drew, Prentice, & Schepens, 2004; Takakusaki, Tomita, & Yano, 2008), whereas cadence is controlled by brainstem and spinal cord (Cho et al., 2010; Morris, Iansek, Matyas, & Summers, 1994).

Dual tasking in Parkinson's Disease Patients: Dual tasking heightens deficits in maintaining steady gait rhythm in PD patients, but not in healthy controls (Yogev et al., 2005). Yogev et al. (Yogev et al., 2005) found that both control and parkinsonian group had decline in gait velocity in dual-task condition but gait variability increased only in subjects with Parkinson's disease.

Energy Cost of walking

At preferred walking speed, the energy consumption per unit of distance of walking is minimum (Ralston, 1958). It is also known that freely chosen step rate requires the least oxygen consumption at a given speed (Holt, Jeng, Rr, & Hamill, 1995; Zarrugh & Radcliffe, 1978). Kurosawa found that walking at a preferred speed calls minimum attentional demand compared to that while walking at higher or lower speeds, which require more attention (Kurosawa, 1994). Sekiya et. al. (Sekiya, Nagasaki, Ito, & Furuna, 1997) suggested optimal method of walking with optimal criteria in terms of energy efficiency, temporal and spatial variability, and attention (Figure 3). They also found that variability in step length was minimum at preferred walking speed when one keeps the ratio of step length to step rate to about 0.006m/steps/min (Sekiya et al., 1997). Dual-tasking influences gait in healthy participants by reducing gait speed. A

reduced gait speed in response to dual tasking may be a protective response. Apparently the choice of gait speed is related to attentional demands.
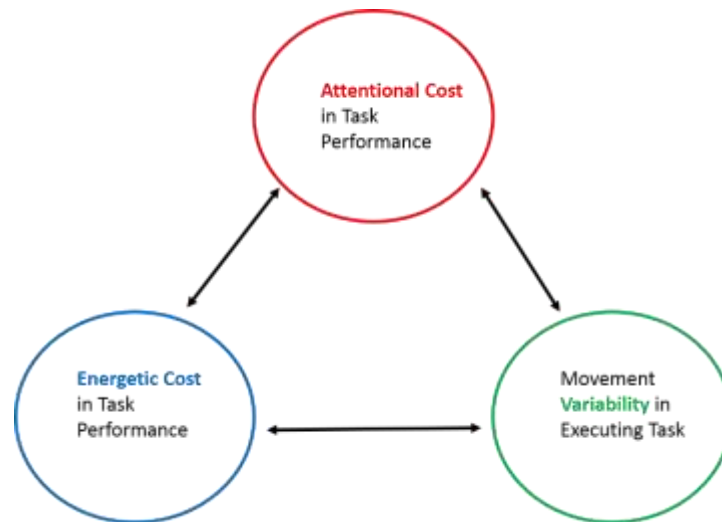
Figure 3. Inter-relationship of movement variability with attention and energy expenditure while performing task

Falls in Cardiovascular Disease Patients

Elderly patients admitted to emergency departments with unexplained or recurrent falls including falls associated with unexplained loss of consciousness constitute 77% of cardiovascular disease disorders (Daccarett et al., 2011; Davies & Kenny, 1996; Montero-Odasso et al., 2005). The association between unexplained falls and cardiovascular disorders in cardiovascular disease (CVD) patients is well known (Cronin & Kenny, 2010; Dey, Stout, & Kenny, 1997), but CVD relationship with gait disorders has yet to be established. Cardiovascular disorders (leading to frailty) has been investigated as a predictor of disability and adverse outcomes in older adults, and it has recently also been investigated as a risk factor for patients undergoing cardiac surgery (Sundermann et al., 2011). Cardiac surgeries are performed in nearly

half of the elderly patients in North America and 78% of these surgeries result in major complications and deaths (Afilalo et al., 2010; Shahian et al., 2009). Afilalo reported about 62% of cardiovascular disease (CVD) patients as frail in a population of 54,250 patients (Afilalo, Karunananthan, Eisenberg, Alexander, & Bergman, 2009) and emphasized on relationship between frailty and CVD; since CVD may lead to frailty as they share common biologic pathways. CVD is also associated with threefold increase in frailty. The 6-month mortality is around 14.1%, which is around fourfold greater than in non-frail persons. Also these fallers with intrinsic cardiovascular cases have a higher mortality rate (Kapoor, Karpf, Wieand, Peterson, & Levey, 1983). Recently several studies have reported significant overlap between syncope and falls (Dey et al., 1997; Gordon, Huang, & Gryfe, 1982; McIntosh, Da Costa, & Kenny, 1993). Given the prevalence of falls among CVD patients, syncopal events account for 40-60% of CVD patients at a syncope clinic (McIntosh et al., 1993). It has been reported that 20-30% of syncope events are due to orthostatic hypotension (Allcock & O'Shea, 2000). Cardioinhibitory carotid sinus hypersensitivity (CSH) is present in more than 20%, vasovagal syncope in 15% and cardiac arrhythmias in up to 20% of syncope cases (Kurbaan et al., 2003). Patients with syncope do not remember the loss of consciousness due to amnesia, and transient cerebral hypoperfusion causes loss of balance without true syncope (Daccarett et al., 2011).

In the past decade, international guidelines had started citing the importance of cardiovascular assessments for prevention of falls. In the year 2001, evidence based guideline was developed by the American Geriatrics Society panel on fall prevention and which was later updated in the year 2009. This guideline recommended assessment of basic cardiovascular status, including postural pulse, heart rate, rhythm, and blood pressure (BP) and, if appropriate, responses of heart rate and BP to carotid

sinus massage ("Guideline for the prevention of falls in older persons. American Geriatrics Society, British Geriatrics Society, and American Academy of Orthopaedic Surgeons Panel on Falls Prevention," 2001). Furthermore these guidelines recommend targeted multifactorial intervention, including treatment of postural hypotension and cardiovascular disorders including cardiac arrhythmias. Cardiovascular examination has essentially become a part of multifactorial fall assessment and is recommended by National Institute of Health (National Guideline). Falls in CVD patients has serious consequences, including injury (Nevitt et al., 1989; Tan & Kenny, 2006; M. E. Tinetti et al., 1988), fear of falling again and depression (Arfken et al., 1994; Welmerink et al., 2010), admission to long-term care facility (Mary E. Tinetti & Williams, 1997), and death (Centers for Disease & Prevention, 2006). Early identification and prevention efforts in this population have the potential to significantly reduce pain, suffering, medical costs, and comorbidities.

Factors Leading to Falls in Cardiovascular Disease Patients

Cardiovascular causes of falls in patients can be neutrally mediated or intrinsic such as rhythm disorders and structural heart disease. More than one-third of patients have more than one attributable cause (Kurbaan et al., 2003) for syncope resulting in falls. The association between orthostatic hypotension, vasovagal syncope, and carotid sinus syndrome is well established (McIntosh et al., 1993) in the patients with high fall risk. In elderly patients; both falls and syncope are often indistinguishable and could result from identical pathophysiological processes. ***Fall*** is defined as an event whereby an individual unexpectedly comes to rest on ground without known loss of consciousness ("Guideline for the prevention of falls in older persons. American Geriatrics Society, British Geriatrics Society, and American Academy of Orthopaedic Surgeons Panel on Falls Prevention," 2001). Whereas ***syncope*** is defined as a

transient loss of consciousness due to transient global cerebral hypoperfusion characterized by rapid onset, short duration, and spontaneous complete recovery (with et al., 2009). It is apparent that insertion of dual-chamber pacemakers in non-accidental fallers with cardioinhibitory carotid sinus syndrome significantly reduce falls (Crilley et al., 1997; R. A. Kenny et al., 2001). Neurally mediated syncope do not increase cardiovascular morbidity or mortality(Kapoor, 1994). Whereas structural heart disease is looked upon as the most important predictor of mortality(Kapoor & Hanusa, 1996). Orthostatic hypotension, neutrally mediated syncope, cardiac arrhythmias, and structural heart disease may present as unexplained falls as well as syncope in older CVD patients. Cardiovascular causes of falls in CVD patients are provided in Table 1.

Vasovagal syncope: Vasovagal syncope (VVS) is a common neutrally mediated disorder. VVS may be precipitated by pain or emotional trauma during standing but is equally likely to occur while sitting and exceptionally when supine. Pre-syncopal symptoms usually include nausea, dizziness, palpitations, diaphoresis and chest pain. These symptoms may last in loss of consciousness or fall if evasive action is not taken. Sometimes premonitory symptoms may be attenuated in patients (Grubb, 2005) and falls due to recurrent VVS may carry substantial risks in them. The underlying pathophysiology of VVS is complex and may involve venous compliance, intravascular volume, respiration, serotonin homeostasis and endothelial, beta-adrenergic receptor and baroreceptor function (Cronin & Kenny, 2010).

Table 1 Cardiovascular causes of falls in CVD patients

Cardiovascular causes of falls in CVD patients
Neurally mediated syndromes
• Orthostatic hypotension
• Postprandial hypotension
• Carotid sinus syndrome
• Vasovagal syncope
• Situational syncope (cough, sneeze, micturition, deglutition, defecation, gastrointestinal stimulation, Valsalva, diving reflex)
Cardiac abnormalities
• Arrhythmias
• Bradycardias
• Supraventricular tachycardia
• Ventricular tachycardia
Structural disease
• Aortic stenosis
• Atrial myxoma
• Left ventricular outflow tract obstruction
• Subclavian steal syndrome
Miscellaneous
• Pulmonary embolism
• Subclavian steal syndrome
• Cerebral syncope
• Transient ischaemic attacks
• Migraine

Carotid Sinus Syndrome: Carotid sinus syndrome (CSS) is a disease usually found among older people 50 years of age and older. It is common cause of drop attacks (Dey et al., 1997) and accounts for unexplained non-accidental falls in about one-third of older people (Richardson, Bexton, Shaw, & Kenny, 1997). In CSS, amnesia for loss of consciousness is common and males are highly affected and the majority have existing cardiovascular co-morbidity in the form of hypertension, cerebrovascular disease or ischaemic heart disease (Draper, 1950). The pathophysiology of CSS is

complex and unclear but central abnormality of baroreflex gain may cause atherosclerosis-induced ischaemia at myocardial or brainstem level. CSS is associated with morbidity but not mortality and syncope is seen to reoccur in CSS patients (Brignole, Menozzi, Lolli, Bottoni, & Gaggioli, 1992).

Orthostatic Hypotension: Orthostatic Hypotension (OH) is seen to increase with advancing age and is associated with increased morbidity and mortality and partly due to increased incidence of falls and vascular death, such as stroke. Older patients who are hypertensive have OH frequently (Luukinen, Koski, Laippala, & Kivela, 1999).

Postprandial hypotension: Postprandial hypotension (PPH) is exacerbated with postural change but is a distinct entity that differs from OH. PPH is present in one-half of patients with unexplained syncope (Jansen, Connelly, Kelley-Gagnon, Parker, & Lipsitz, 1995) and is more common than OH (Jansen & Lipsitz, 1995). PPH occurs in both sitting and supine position (Haigh, Harper, Burton, Macdonald, & Potter, 1991), and has symptoms similar to those produced by OH.

Frailty in CVD patients: Gobbens and coworkers suggested that life-course determinants affect the occurrence of frailty disease (Gobbens, Luijkx, Wijnen-Sponselee, & Schols, 2010) in patients with cardiovascular disorders. Mediation is believed to occur if an independent variable (life-course determinants) has an effect on another independent variable (disease which is a mediator variable), which may have effect on the dependent variable (frailty). Frailty is a multidimensional entity and comprises of physical, psychological and sociological components as well. Fried and coworkers suggested that markers of old age such as lean body mass, strength, endurance, balance, walking performance and low physical activity (A Paw, Dekker, Feskens, Schouten, & Kromhout, 1999; Bortz, 1993; Buchner & Wagner, 1992; Campbell & Buchner, 1997; Lipsitz & Goldberger, 1992) are related clinically to

frailty and with declining energetics and reserve. Whitson and coworkers emphasized that frailty spectrum is composed of "phrailty" (physiologic frailty) and "F-frailty" (functional frailty) and clinicians recognize frailty as a multifactorial state brought about by accumulated conditions and age-related process (Whitson, Purser, & Cohen, 2007). Frailty may be seen as a continuum stretching from early stages to late stages. In early stages, it is difficult to identify frailty clinically under circumstances of everyday working life but symptoms are obvious when individual faces external stressors, where as in late stages with full-blown frailty, it is easy to recognize frailty because the symptoms interfere with daily routine and come close to a state of disability (Bauer & Sieber, 2008; Whitson et al., 2007).

Mechanistic link between CVD and Frailty

Afilalo highlighted in his studies that CVD and frailty share a common mechanistic link and are strongly tied to chronic low-grade inflammation (Afilalo, 2011). This inflammation may have been caused by lifelong antigenic exposure, angiotensin 1R activation, obesity, insulin resistance and redox imbalance (De Martinis, Franceschi, Monti, & Ginaldi, 2006), which are found in greater amounts in CVD as well as in frailty. Some other circulating inflammatory markers found in relatively greater amounts are neutrophils, monocytes, high sensitivity C-reactive protein, and interleukin-6 (IL-6) (Cesari et al., 2004; Leng et al., 2009; Leng, Xue, Tian, Walston, & Fried, 2007; M. Visser et al., 2002). Frailty is also associated with increased C Reactive Protein values and pro-inflammatory cytokines (Hubbard, O'Mahony, Calver, & Woodhouse, 2008; Leng et al., 2007; Walston et al., 2002). CVD is known to contribute to the development of frailty (Dumurgier et al., 2009; Newman et al., 2006). The inclusion of functional evaluation to the traditional clinical examination provides information that is critical in the comprehensive assessment of frailty in the elderly CVD patients (Applegate, Blass, & Williams, 1990; M. E. Tinetti & Ginter,

1988). Objective tests of subjects' performance of standardized tasks, evaluation through predetermined criteria such as repetition counting and timing of an activity (Guralnik, Branch, Cummings, & Curb, 1989) are helpful to predict outcomes of falls and death (Duncan, Studenski, Chandler, & Prescott, 1992; Nevitt et al., 1989; M. E. Tinetti et al., 1988).

Physical Performance Measures and Frailty

Physical frailty may cause decline in strength, poor balance, reduction in muscular endurance and paucity of movement (Era et al., 1997; Schenkman, Hughes, Samsa, & Studenski, 1996; Whipple, Wolfson, & Amerman, 1987). Frailty is multidimensional and affects multiple physical factors such as static and dynamic balance, decline in strength, range of motion, peripheral sensory information, gait speed and movement speed (Brown, Sinacore, Binder, & Kohrt, 2000). Thus frailty can be explained with accumulation of deficits across multiple domains rather than by any specific common deficit (Duncan, Chandler, Studenski, Hughes, & Prescott, 1993). Physical performance measures such as chair stands and timed walks (Guralnik et al., 1989), are important so to objectively assess the functional capacity of a person to interact with environment and are helpful in predicting disability (Guralnik, Ferrucci, Simonsick, Salive, & Wallace, 1995) and death (Williams, Gaylord, & Gerritty, 1994).

Frailty and Nutrition Status

Frailty is significantly associated with a daily energy intake below 21 kcal/kg bodyweight and low protein intake (Bartali et al., 2006). It is also seen that pre-frail and frail individuals have a higher prevalence of being deficient for vitamin B12,

vitamin D and alpha-tocopherol in contrast to non-frail female individuals (Michelon et al., 2006).

<u>Assessment of Frailty</u>

To date, there exists no standardized and valid assessment method to identify truly frail older adults. Campbell and Buchner suggested that frailty can be diagnosed clinically by measuring four key components required for successful interaction with the environment: musculoskeletal function, aerobic capacity, cognitive and integrative neurological function and nutritional state (Campbell & Buchner, 1997). Frailty is a complex and multidimensional disease and evaluation in limited domains such as strength or cognitive state will not suffice to gain insights into this complex phenomenon. Duncan and coworker suggested that frailty can better be explained by accumulation of deficits across multiple domains rather than by any specific common deficit (Duncan et al., 1993). An important determinant of frailty is multi-morbidity (Levers, Estabrooks, & Ross Kerr, 2006). Bortz explained frailty as a breakdown of essential maintenance capability of individuals function induced by precipitation of external environmental stresses (Bortz, 1993).

Fried et al. (2001), proposed an operational screening of frailty characterized by the following criteria: 1) **slow gait speed**, (<0.58 or 6 min walking distance ≤ 210m) 2) **muscle weakness** (weak grip strength ≤16 kg), 3) **unintentional weight loss** (>5 kg in preceding year), 4) **low physical activity** (lowest 20% of kcals expended per week by gender), and 5) **self-reported exhaustion** (Fried et al., 2001). Individuals that meet three or more of the criteria are diagnosed as frail, and the individuals diagnosed with one or two are categorized as prefrail and with none are considered nonfrail. Prefrail category is disregarded by some researchers because its prognostic utility is not that well established. They validated this clinical criteria and

explained its capability to predict disability, hospitalization and mortality among participants of Cardiovascular Health Study (CHS), a cohort study of 5888 community-dwelling elders (Fried et al., 1991; Fried et al., 2001). These 5 screening components are part of broader functional assessment and concentrate on the physical aspects of frailty and are easy as well as quick to implement in clinical environment. Fried's criteria has been most often adopted in frailty studies, may be because of its ease of application (Drey, Pfeifer, Sieber, & Bauer, 2011).

No standard clinical or diagnostic criteria has been established to assess frailty through gait and postural differences but many authors have reported weight loss, loss of lean body mass (sarcopenia), fatigue, and loss of strength or endurance (Fried, 1992; Palmer, 1990; Vandervoot & Symons, 2001; Wallace, Schwartz, LaCroix, Uhlmann, & Pearlman, 1995) as some physical criteria. Frail older people are seen as older adults who are lacking in general strength and who are unusually susceptible to disease or other infirmities. Physical frailty depends on various factors such as decline in strength, loss in range of motion, slowness of movement, paucity of movement, poor balance, and reduced muscular and cardiovascular endurance (Brown et al., 2000; Era et al., 1997; Schenkman et al., 1996; Whipple et al., 1987).

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CHAPTER 3: Dual-task does not increase slip and fall risk in healthy young and older adults

Abstract

Dual-task tests can identify gait characteristics peculiar to fallers and non-fallers. Understanding relationship between gait performance and dual-task interference is warranted to help identify potential predictors of gait related cognitive-motor interference through adapted changes in gait instability/variability which could adversely affect slip and fall risk. The influence of attention on gait stability has been investigated in numerous patient populations and results consistently show decreased gait velocity and increased gait variability in dual task conditions. The primary aim of this study is to answer the questions (1) does dual-task influence young and older adult gait performance and associated variability? (2) Does dual-tasking influence slip and fall risk in healthy participants? Seven healthy young and seven healthy old adults participated in this study. The results indicate that the gait changes in dual task walking characterize increased gait stability and indicate that the attention demanding tasks during walking have no destabilizing effect on gait in healthy people. This study found that during dual-tasking healthy humans adopt cautious gait mode (CGM) characterized by reduced walking speed, shorter step length, increased step width, reduced heel contact velocity, and is likely to be an adaptation to minimize attentional demand and decrease slip and fall risk during limited available attentional resources. Exploring interaction between gait variability and cognitive functions may lead to designing appropriate fall interventions for patients.

Introduction

Walking is a somewhat complex task associated with higher level cognitive processing such as estimation, planning and real-time adjustments, specifically, executive function is involved. During walking, Central Pattern Generators (CPG) create rhythm (Burke, Degtyarenko, & Simon, 2001; Lamb & Yang, 2000; Shefchyk & Jordan, 1985) and gait proceeds under the stewardship of the motor control system utilizing cognitive resources. While walking, numerous sensory (visual, proprioception and vestibular), conscious inputs and competing objectives (e.g., upright posture versus locomotion) are seamlessly integrated across hierarchical systems, with subtle real-time decisions and adjustments are made using cognitive capabilities (Hausdorff, Yogev, Springer, Simon, & Giladi, 2005).

Gait performance is affected by the simultaneous performance of dual tasks (Beauchet, Dubost, Aminian, Gonthier, & Kressig, 2005; Beauchet, Dubost, Herrmann, et al., 2005; Beauchet et al., 2003; Dubost et al., 2006; Lundin-Olsson, Nyberg, & Gustafson, 1997; Woollacott & Shumway-Cook, 2002). The dual task paradigm, is commonly used to assess multitasking capabilities. It is presumed that multitasking is influenced by age related changes in attentional capacities (Kerr, Condon, & McDonald, 1985) and reduce the abilities to shared processing domains for two concurrent tasks (Bowen et al., 2001).

OShea and colleagues (O'Shea, Morris, & Iansek, 2002) suggested that detrimental effects of physical task in the presence of competing attentional task supports a "**capacity sharing model**" of dual tasking. According to capacity sharing model, performing two attention demanding tasks reduces the performance of one or both tasks when attentional capacity limit is exceeded (O'Shea et al., 2002). Previous research has demonstrated that dual task interference results in slower gait speeds (Bowen et al., 2001; Hyndman, Ashburn, Yardley, & Stack, 2006; Plummer-D'Amato

et al., 2008), reduced cadence (Kemper, McDowd, Pohl, Herman, & Jackson, 2006; Plummer-D'Amato et al., 2008), shorter stride length (Hyndman et al., 2006; Plummer-D'Amato et al., 2008), increased stride duration (Plummer-D'Amato et al., 2008), and longer double support time (Bowen et al., 2001; Plummer-D'Amato, Altmann, Behrman, & Marsiske, 2010). Unwittingly, dual task performances can potentially identify gait characteristics peculiar to fallers and non-fallers (Zijlstra, Ufkes, Skelton, Lundin-Olsson, & Zijlstra, 2008).

The influence of attention on gait stability has been studied in numerous patient populations and results consistently show decreased gait velocity and increased gait variability in dual task conditions (Beauchet, Dubost, Aminian, et al., 2005; Hausdorff, Balash, & Giladi, 2003; Hausdorff, Schaafsma, et al., 2003; Sheridan, Solomont, Kowall, & Hausdorff, 2003; Yogev et al., 2005). Apparently, persons with history of falls have more significant gait changes while performing dual task than that of non-fallers (Faulkner et al., 2007; Toulotte, Thevenon, Watelain, & Fabre, 2006; Vaillant et al., 2006; Woollacott & Shumway-Cook, 2002). Despite of their report, dual task and its association with falls continues to be debated. Some studies showed worsening of gait performance while concluding dual tasking as a predictive of falls (Lundin-Olsson et al., 1997; Lundin-Olsson, Nyberg, & Gustafson, 1998; Verghese et al., 2002) while others have failed to establish any relationship (Bloem, Steijns, & Smits-Engelsman, 2003; Stalenhoef, Diederiks, Knottnerus, Kester, & Crebolder, 2002). One of the study reported that dual task related gait changes did not provide any additional information than performance under single task conditions (Bootsma-van der Wiel et al., 2003). But these differences may be due to several confounds such as age (Beauchet et al., 2003; Woollacott & Shumway-Cook, 2002) and comorbidities (Beauchet, Dubost, Herrmann, et al., 2005; Lundin-Olsson et al., 1998), and kind of attentional demanding task (Allali et al., 2007;

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Beauchet, Dubost, Aminian, et al., 2005). Indeed, dual task related gait changes are associated with intrinsic (subject-related) risk factors of falling similar to those of recurrent falls (Beauchet et al., 2008).

Some findings corroborate well with previous investigations that poorer ability of subjects to perform a basic mobility task while carrying a cup of water (Lundin-Olsson et al., 1997) and the cessation of walking when engaged in conversation (Lundin-Olsson et al., 1998) are both associated with a four times risk of fall. It is expected that old and frail people may have difficulty coping with many minor threats, or may be distracted by unexpected stimuli in the same environment than the younger counterparts. Falls are multifactorial in origin (Myers, Young, & Langlois, 1996), however considering cognitive perspective, elderly frail may require more attentional resources than non-frail elderly while walking and standing, the paucity of attentional resources may further emerge balance problems in them.

Nevertheless, common daily activity tasks require attention, rapid motor planning process, and effective inhibition of irrelevant or inappropriate details. In such scenarios, the inability to perform the dual task has been reported to increase risk of falls in older adults (Lundin-Olsson et al., 1997). Lundin-Olsson reported that dual motor task can differentiate fall prone frail elderly from healthy older adults (Lundin-Olsson et al., 1998). In essence, dual task paradigm is considered more sensitive to identify fall risk since it widens the gap between fallers and non-fallers (Bloem, Valkenburg, Slabbekoorn, & Willemsen, 2001; de Hoon et al., 2003; Springer et al., 2006).

Dual task tests can identify gait characteristics peculiar to fallers and non-fallers (Zijlstra et al., 2008). The influence of attention on gait stability has been studied in numerous patient populations and results consistently show decreased gait velocity and increased gait variability in dual task conditions (Beauchet, Dubost,

Aminian, et al., 2005; Hausdorff, Balash, et al., 2003; Hausdorff, Schaafsma, et al., 2003; Sheridan et al., 2003; Yogev et al., 2005). Fall risk is independent of gait speed but is modulated instead by gait variability (Maki, 1997). Previous studies demonstrate that gait speed and gait unsteadiness may be dissociated (Hausdorff, Edelberg, Mitchell, Goldberger, & Wei, 1997; Hausdorff et al., 2000; Maki, 1997). Healthy older adults walk with the same, small amount of variability as healthy young adults, even though they walk slower than healthy young adults (Hausdorff, Mitchell, et al., 1997). However, Springer et al. (Springer et al., 2006) reported that gait variability increased in older fallers and not in young adults and older non-fallers. They reported that healthy young adults and non-fallers maintain their stable gait in dual task walking and that there is no evidence of detrimental effects of dual task activities on gait variability associated with aging.

To stabilize, healthy people are found to decrease their gait speed (Springer et al., 2006). Accordingly, elderly non-fallers were also found to decrease their swing times and their gait speed (Springer et al., 2006). This dual task related decline in walking speed is interpreted as an implicit strategy to avoid loss of balance (Dubost et al., 2006). Reduction of gait speed among groups represents a coping mechanism to handle the attention demanding challenge of the dual task activity.

The finding of decrease of gait velocity in dual task walking is undebated and consistent with most of the studies (Hausdorff, Balash, et al., 2003; Hollman, Kovash, Kubik, & Linbo, 2007; Sheridan et al., 2003; Springer et al., 2006; Yogev et al., 2005). Although dual-task related decline in walking speed is not specific for increased risk of falling, increase of stride time variability is closely associated with the occurrence of falls (Hausdorff, Rios, & Edelberg, 2001; Maki, 1997). There exists association between low stride time variability and efficient executive function in healthy older adults (Hausdorff et al., 2005), and high stride time variability and

impaired executive function in demented older adults(Sheridan et al., 2003). Low stride time variability in healthy older adults is associated with minor involvement of attention in the control of the rhythmic stepping mechanism (Beauchet, Dubost, Aminian, et al., 2005).

Previous research has shown that dual-task related gait changes consisted of increase in number of stops, lateral deviations, steps and walking time (Beauchet, Dubost, Aminian, et al., 2005; Beauchet et al., 2003; Lundin-Olsson et al., 1997), and increase in stride width, stride length, stride time variabilites (Beauchet et al., 2003; Grabiner, Biswas, & Grabiner, 2001). Intrasubject variability of kinematic variables is an index of movement consistency or stability of gait performance. However, there exists negative correlation between variability in step width and balance performance of the elderly women (Heitmann, Gossman, Shaddeau, & Jackson, 1989) and also an increased variability in step length for hospitalized fallers compared with non-fallers (Guimaraes & Isaacs, 1980). Gabell and Nayak (Gabell & Nayak, 1984) could not find any effects of age on variability in step length and step width while walking. Maruyama and Nagasaki (Maruyama & Nagasaki, 1992) reported that temporal variability in stance phase durations in gait cycle was decreasing function of speed. Increasing the walking speed produced linear increment in step width variability in contrast to step length variability in healthy adults (Sekiya, Nagasaki, Ito, & Furuna, 1997). Gabell and Nayak (Gabell & Nayak, 1984) suggested that variability in step length is determined predominantly by gait patterning mechanism, on the contrary step width variability is largely determined by balance control mechanism. Similarly, Heitmann et. al. found negative correlation between balance performance and variability in step width but not the same for balance performance and step length.

Objective of this study

Performance of secondary task i.e. dual task affects certain aspects of gait, but the relationship between gait variability, dual tasking and slip and fall risk is not well understood. This study was conducted to better understand the motor control of gait and the relationship between an individual's motor variability and fall risk. This relationship in one's variability and associated fall risk has potential clinical implications in assessing fall risk in patients. Exploring dual-task related gait changes is of particular interest in understanding variability because a strong relationship exists between dual-task related gait changes and the risk of falling in older adults (Bloem et al., 2003; Lundin-Olsson et al., 1997; Verghese et al., 2002).

The primary objective of this study was to investigate the relationship between dual-task and fall risk. The study involves two groups of individuals with (known) different slip and fall risk, and to analyze their differences in gait parameters and slip severity. It was hypothesized that dual-tasking while walking would affect gait characteristics and may increase the slip initiation characteristics in the elderly individuals and will negatively influence slip-induced risk. It was also hypothesized that friction demand and, trip risk will be significantly different for normal walking and dual-task walking.

Materials and Methods

Subjects

Sample Size Estimation

To estimate sample size, power analysis was performed on results of the pilot study by focusing on sample sizes that are large enough to determine differences between the velocities during normal walking. The general test statistic for two populations

would be the standard two-sided t test, for which the power of the test (Neter, 1996) is given by:

$$Power = P\left\{|t*| > t(1 - \frac{\alpha}{2}; n - 2|\delta)\right\} \qquad 1$$

$$\delta = |A - B|/\sigma\sqrt{2/n} \qquad 2$$

δ is the noncentrality parameter, or a measure of the difference between the means of A and B (Velocity) σ is the standard deviation of the distribution of velocity and n is the number of participants.

Palombaro et. al, have determined that minimal clinically important difference (MCID) for habitual gait speed is 0.10m/s (Palombaro, Craik, Mangione, & Tomlinson, 2006). The standard deviation of measurement is 0.10m/s (Palombaro et al., 2006). Therefore, means and standard deviations of velocity in this study were used to compute the required sample size (using JMP, Version 7. SAS Institute Inc., Cary, NC, 1989-2007). The required sample size for detecting significant differences in velocity, given $\alpha = 0.05$ and power=0.80 was determined as $n = 14$.

Participants:

Seven young and seven old participants were recruited for this study. The younger population consisted college students of Virginia Tech campus, and older adults were retired people in Blacksburg area. The recruited participants were in a general good health condition, with no recent cardiovascular, respiratory, neurological, and musculoskeletal abnormalities. Only one of the elderly participants (O02) was suffering from chronic obstructive pulmonary disease (COPD). All participants were recruited based on criteria of complete ambulation, without the use of any assistive devices, and ability to rise from a chair without assistance and free of orthopedic

injury. This study was approved by the Institutional Review Board (IRB) of Virginia

Tech. All participants who participated in this study provided written consent prior to

the beginning of data collection. Demographic information for the participants is

provided in Table 2.

Table 2 Background characteristics of study participants

	Age Group			
	Old		Young	
	Mean	SD	Mean	SD
Age	71.143	6.5174	22.643	2.5603
Height [cm]	174.571	10.2446	170.376	9.3302
Weight [Kg]	78.559	18.2576	69.651	15.5270
BMI	25.529	4.2731	23.786	4.0004

Instrumentation

The experiments were conducted on a 15-meter linear walking track, embedded with

two force plates (BERTEC #K80102, Type 45550-08, Bertec Corporation, OH 43212,

USA and AMTI BP400600 SN6780, Advances Mechanical Technology Inc.,

Watertown, MA 02472, USA). A six-camera ProReflex system (Qualysis

Gothenburg, Sweden) was used to collect three-dimensional kinematics of posture

and gait data in participants. Kinematic data were sampled and recorded at 120 Hz.

Ground reaction forces of participants walking over the test surfaces were measured

using two force plates and sampled at a rate of 1200 Hz. A sixteen channel surface

electromyography (s-EMG) DTS Telemyo system (Noraxon 15770N Greenway-

Hayden Loop, #100 Scottsdale, AZ USA), was used to record the temporal activations

of two ankle muscles (gastrocnemius, and tibialis anterior) in the both lower

extremities during walking.

Experimental Preparations and Practices

All participants were first familiarized with laboratory equipment's and were provided a verbal explanation of the experimental procedure. Participants were requested to wear laboratory clothes and shoes, fitting to their sizes. Height and weight of participants were noted below the ID numbers assigned to the subject. Surface Electromyogram (s-EMG) electrodes were affixed by asking participants to plantarflex and dorsiflex their ankle for gastrocnemius and tibialis anterior muscles. Twenty-six reflective markers were attached to various bony landmarks of the body such as head, both ears, both acromio-clavicular joints, acromions, humoral condyles, ulnar stylus, knuckles, both right and left anterior superior iliac spine (ASIS), greater trochanters, both medial and lateral condyle of both limbs, malleolus (medial and lateral) and heel and toes of both feet (shown in Figure 4). The marker configuration was similar to that defined by Lockhart et. al 2003 and was used to derive a whole body biomechanical model (T. E. Lockhart, Woldstad, & Smith, 2003). Kinetic data was acquired using two forceplates such that two consecutive steps would fall on them. The slippery surface (which was at top of second forceplate) was covered with 1:1 water and jelly mixture to reduce the coefficient of friction (COF) of the floor surface (dynamic COF ~0.12). Participants were kept unaware of the position of this surface as the embedded forceplates are also covered with similar vinyl texture as the walkway. This is well standardized protocol used in several previous slip and fall research at Locomotion Research Laboratory (LRL).

Protocol

The experiment was divided into two sessions: normal session and dual-task session (Figure 5). Each session was separated by 4 days and each participant was randomly assigned to either normal or dual-task as his/her first session.

Sessions:

Normal Walking and Slip:

After attaching s-EMG's and markers, participants were instructed to walk on the walkway for 15-20 minutes at their self-selected pace. Participant's gait data were acquired using motion capture, IMU's, forceplates and EMG system. The starting point during the walk was adjusted such that their non-slipping foot (non-dominant) landed on the first forceplate and dominant foot landed on the second platform. The participants were told in the session that they "may or may not slip," and they should look forward while walking. Additionally, participants were unaware of the placement of slippery surface. Once five walking trials with complete foot fall on the forceplate were obtained, the slippery surface was introduced above the forceplate where dominant foot was expected to strike.

Figure 4. The picture of placement of reflective marker, inertial sensors and s-EMG
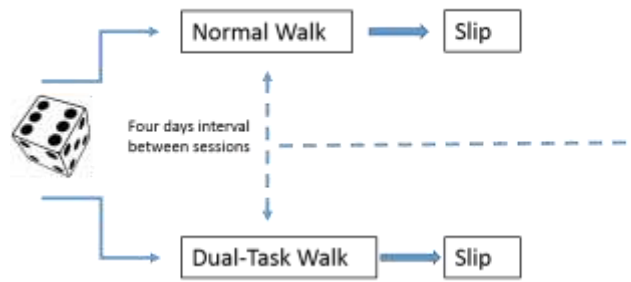
Figure 5. Participants were assigned to normal or dual-task session randomly and the listed tests were conducted

Dual task walking and slip

This study used a clear and standardized cognitive task, such as serial subtraction (Lezak, 1995; Strauss, Sherman, & Spreen, 2006). This session was similar to normal walking session described above, except that the participants were counting backwards when walking. The investigator told a random number before the walking trial and participants had to subtract the number by three continuously until he/she reached the other end of walkway. The investigator corrected the participants, if error was made in counting backwards.

Data Processing

Kinematic and Kinetic data processing

Normal and dual-task walking trials provided kinematic and kinetic data that was filtered using low-pass Butterworth filter at cut-off frequency of 6Hz. The EMG data was digitally band pass filtered at 20-500Hz. The EMG signals were then rectified and low-pass filtered using Butterworth filter with a 6Hz cut-off frequency to create a linear envelope. Heel contact (HC) and Toe-off (TO) events were identified from the ground reaction forces with threshold set at 11 Newton. The analysis was performed from stance phase (HC to TO) of non-slipping foot.

Dependent Variables

Gait Parameters

1. **Step Length (SL)**: The distance travelled by the participant in one step, it is computed as the anterior-posterior directed distance between ipsilateral limb heel and contralateral limb heel markers at one step (Figure 6).

2. **Step Width (SW)**: It is the mediolateral distance travelled in one step, it is computed as the mediolateral distance between the feet during a step (Figure 6).

3. **Walking speed**: Walking speed is computed from the time taken by participants to cover a distance of 5 meters.

$$Walking\ speed\ (m/s) = \frac{5\ (m)}{Time\ (\text{sec})} \qquad\qquad 3$$

4. **Double support time (DST)**: Double support is the time when both the feet are on the ground. In one stride there are two double support time intervals. It starts with initial contact of one foot until the toe off of the other foot.

5. **Heel Contact Velocity (HCV)**: It is the instantaneous horizontal velocity of the heel at the moment of heel contact. Heel contact is defined as the instant at which the vertical force on the forceplate exceeds 11N.
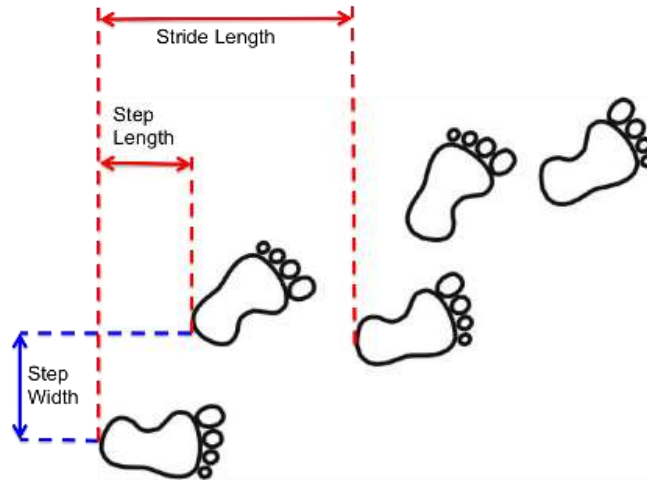
Figure 6 Step length, stride length and step width computation using coordinates of heel marker

After processing the heel marker data, HCV was extracted by horizontal heel position at 1/120 sec before and after heel contact.

$$\frac{X_{(i+1)} - X_{(i-1)}}{2\Delta t} \qquad 4$$

where, i is the frame index at the moment of heel contact. The variables $X_{(i+1)}$ and $X_{(i-1)}$ represent the horizontal heel positions at the frames occurring 1/120 sec before and after the instant of heel contact, respectively. The time variable Δt is 1/120 sec.

Slip propensity parameter

Required Coefficient of Friction (RCOF): The RCOF is the minimum coefficient of friction which is required between the shoe sole and floor interface to prevent slipping. Thus if the floor surface and shoe tribology can meet RCOF, walking is possible, whereas if the RCOF is greater than the available friction between the shoe and floor surface, then a slip occurs. The RCOF is defined as the ratio of forward horizontal ground reaction force to vertical ground reaction force, F_X/F_Z.

Transverse Coefficient of Friction (TCOF): The TCOF was defined as the ratio of lateral ground reaction force component to vertical ground reaction force, F_Y/F_Z.

Trip propensity parameter

Toe clearance: Toe clearance is a critical event during mid-swing of the foot when the foot clearance is achieved with minimum height from the ground surface (Figure 7).
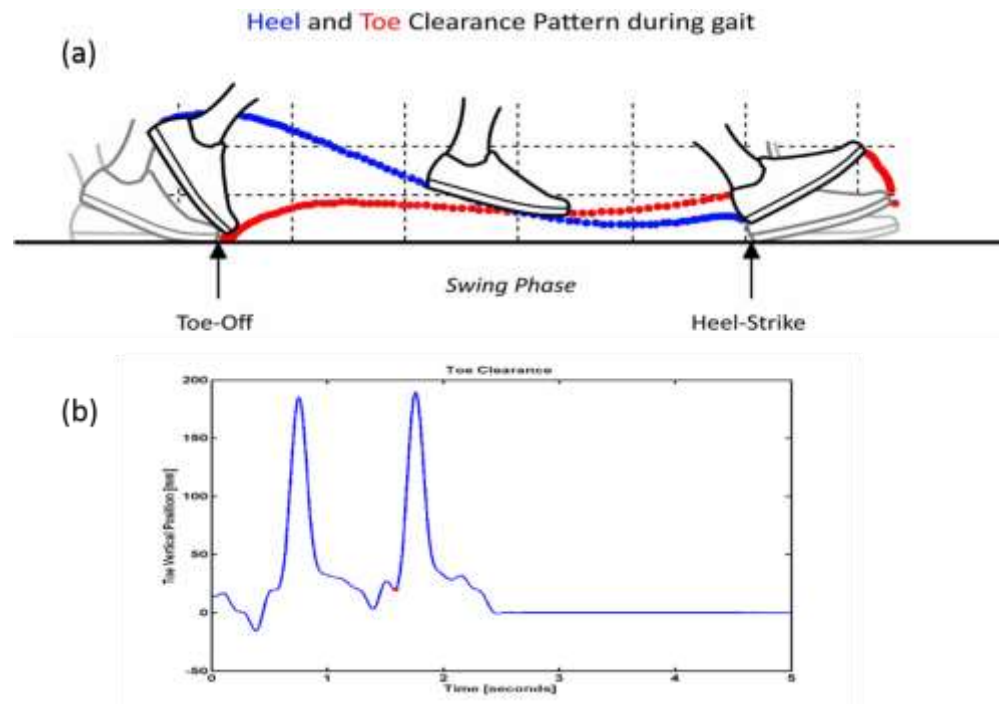
Figure 7. (a) Red colored trajectory of toe (b) toe marker vertical trajectory where red dot represents toe clearance value

Slip-severity parameters

Initial slip distance (SDI): Initial slip distance begins after heel contact when the first non-rearward positive acceleration of the foot is identified (Figure 8). This SDI is the distance travelled by the heel from this point of no-rearward positive acceleration (minimum velocity) to the time of the first peak in heel acceleration (T. E. Lockhart et al., 2003).

$$SDI = \sqrt{(X_2 - X_1)^2 + (Y_2 - Y_1)^2} \qquad 5$$

Slip distance II (SDII): Slip distance II begins at the slip-point of SDI. Slip-stop for SDII is the point at which the first maximum in horizontal heel velocity occurs after the start of SDII. SDI and SDII are used as indices for comparing the severity of slips (Figure 8).

Peak sliding heel velocity (PSHV): It is the maximum forward speed of the heel during slipping. This parameter is calculated using the time derivative of heel marker position during the slip.

Time2SDI: It was defined as the time to reach mid-slip from slip start or time to cover SDI.

TimeSDI2SDII: It was defined as the time taken from mid-slip to slip stop.

Time SD Total: It was defined as the time taken from slip start to slip stop events.

RFoot Mean CCI value: It was defined as the mean CCI in slipping foot during slip start to slip stop.

RFoot Peak CCI value: It was defined as the peak CCI value during slip start to slip stop.

RFoot Time2Peak CCI from LHC: It was defined as the time to generate peak ankle co-contraction from the heel contact of unperturbed foot.

LFoot Mean CCIvalue: It was defined as the mean CCI in non-slipping foot during slip start to slip stop.
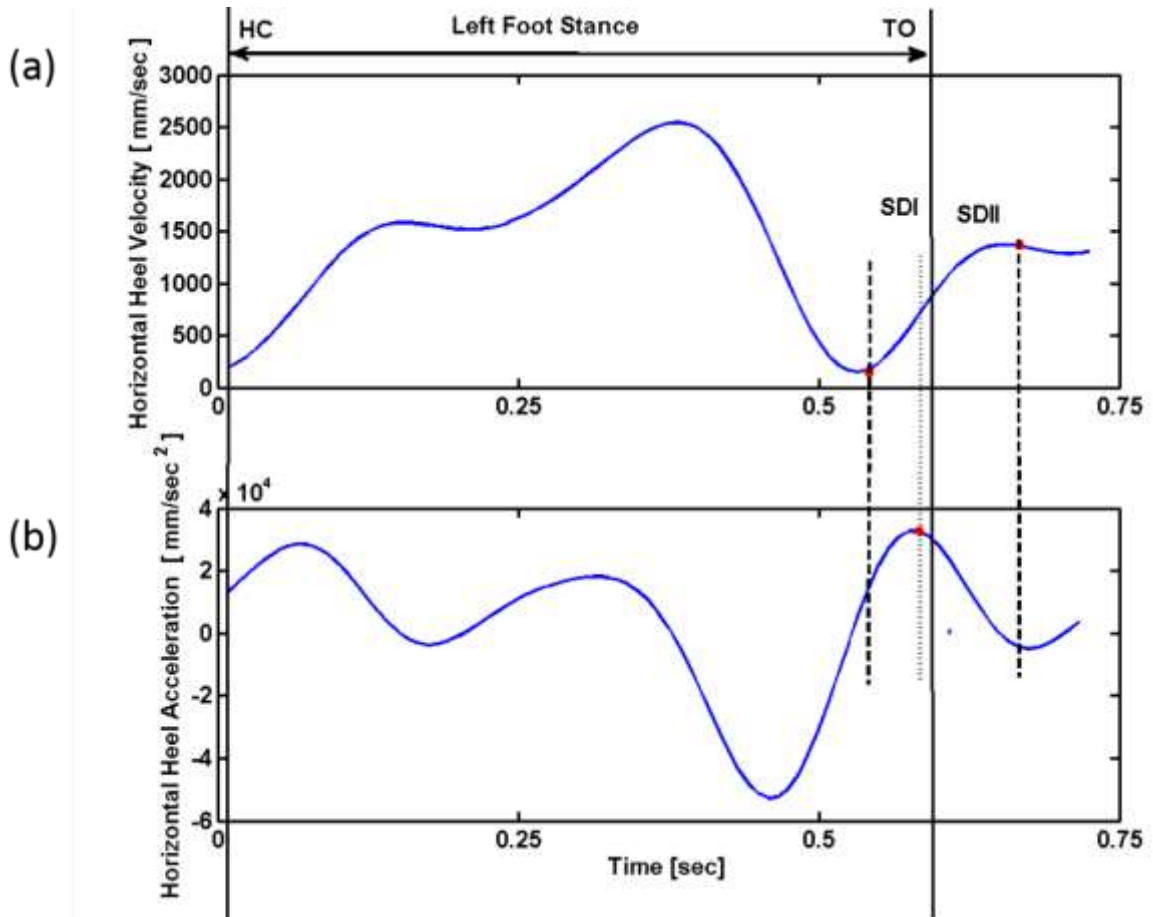
Figure 8. (a) Horizontal heel velocity of slipping foot (b) horizontal heel acceleration of the slipping foot

Plantar flexion Muscle co-contraction : EMG activity was peak normalized within

each subject using the ensemble average method during the complete gait cycle

(Kadaba et al., 1989). Then, co-contraction index (CCI) was calculated by the

following equation (Rudolph, Axe, Buchanan, Scholz, & Snyder-Mackler, 2001):

$$CCI = \frac{LowerEMG_i}{HigherEMG_i} \times (LowerEMG_i + HigherEMG_i) \qquad 6$$

where $LowerEMG_i$ refers to the less active muscle at time i

$HigherEMG$ refers to the more active muscle at time i

The ratio of the EMG activity of Tibialis Anterior/Gastrocnemius was considered for

this study (Figure 9). The ratio is multiplied by the sum of activity found in the two

muscles. The Co-Contraction Index was defined as the event when bursts of the

muscle activity of the agonist and antagonist muscles overlapped for at least 5 ms

(Tang & Woollacott, 1998).
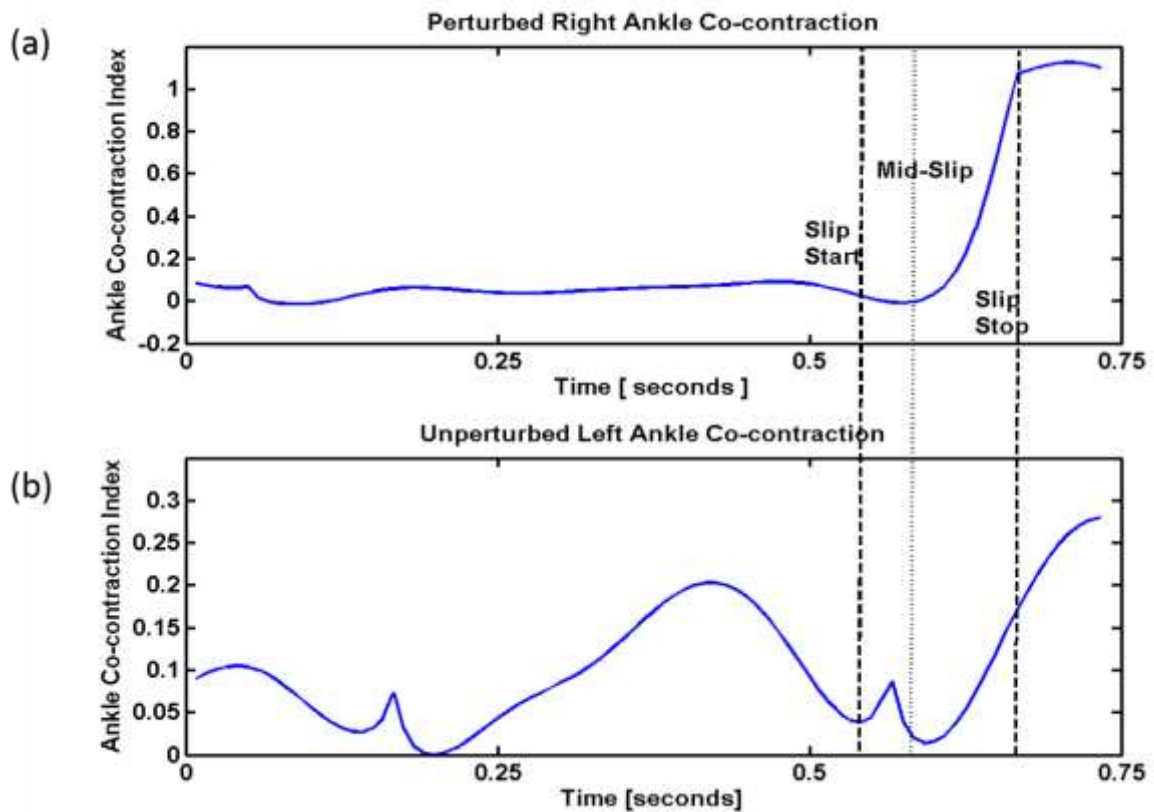
Figure 9. (a) Ankle co-contraction values in the slipping foot (b) ankle co-contraction
values in non-slipping foot

L Stance Time: It is the single stance duration in non-slipping foot right before the

perturbation event.

Mini-Mental state Examination (MMSE): The MMSE examines multiple areas of

cognition in human brain. The highest possible score is 30; a score of less than 24

denotes cognitive impairment. Mild cognitive impairment is reflected in scores of 18

to 23, moderate cognitive impairment is suggested by scores of 17 to 10, and severe

cognitive impairment is denoted by scores of less than 10 (Folstein, Folstein, &

McHugh, 1975).

Statistical Design

There were two independent variables; Age group (Young vs. Old) and Condition (Normal vs. Dual-task). The statistical model is mixed factor multivariate analysis of variance (MANOVA) was conducted where age group was a between-subjects factor and dual-task/normal conditions was within-subject factor. Using the Wilks' Lambda test, the MANOVA allowed for determination of which factors had significant effects on the multiple dependent variables as a whole (i.e., gait parameters, muscle co-contraction, slip parameters). Following MANOVA test, subsequent univariate ANOVA (mixed factor design) were conducted separately for each dependent variable. The statistical model for two-way mixed factor ANOVA with A-between subjects and B-within subject

$$Y_{ijkl} = \mu + \alpha_i + \beta_j + \gamma_{k(i)} + \alpha\beta_{ij} + \beta\gamma_{jk(i)} + \varepsilon_{l(ijk)} \qquad 7$$

Where Y is observation, μ is population mean, A is between subject variable (age group: i=2), B is within subject variable (dual-task/normal condition: j=2), S is the number of subjects in the group (k=7) and ε is the random error.

To determine if the groups had similar gait and slipping characteristics, a between subject one-way ANOVA was performed on slip distances (SDI & SDII) and PSHV. All statistical analyses were conducted using JMP (Pro 10.0.2, SAS Institute Inc.) with significance level of $\alpha < 0.05$ for all the statistical tests. All dependent variables were evaluated for normality (using Shapiro-Wilk W test) and residual analysis. The results did not indicate any violation of normality assumptions.

Results

Gait changes due to dual-task performance

The results indicated that both age groups (young/old) were affected by dual-task and their step length (df=1,p<0.0046) decreased significantly. Double support time (DST) (df=1,p<0.0048) and mean single stance time (SST) (df=1,p<0.013) increased as well in both young and elderly subjects. It was also found that RCOF and TCOF values decreased slightly due to dual-tasking in both younger and older individuals but the effects were not statistically significant. Older adults also have higher linear variability in some gait parameters as seen by standard deviation and coefficient of variation of step width, HCV, DST, SST and gait cycle time due to dual tasking (as seen in Table 3).

Table 3 General gait parameters for younger and older population for normal and dual-tasking types of walking

	Age Group											
	Old						Young					
	Condition						Condition					
	DTW			NW			DTW			NW		
	Mean	Std Dev	CV	Mean	Std Dev	CV	Mean	Std Dev	CV	Mean	Std Dev	CV
Step length [mm]	702.923	48.751	6.935	739.100	57.885	7.832	704.264	35.363	5.021	757.244	42.763	5.647
Step Width [mm]	117.891	21.290	18.059	113.163	20.246	17.891	119.744	31.369	26.196	117.407	28.529	24.300
HCV [mm/s]	993.758	516.513	51.976	1048.442	195.338	18.631	1297.521	870.848	67.116	1019.164	198.942	19.520
GCT [s]	1.106	0.072	6.472	1.076	0.066	6.169	1.129	0.058	5.124	1.082	0.064	5.879
DST [s]	0.266	0.028	10.598	0.221	0.021	9.484	0.256	0.030	11.838	0.244	0.029	11.986
SST [s]	0.680	0.048	7.105	0.647	0.037	5.715	0.707	0.041	5.854	0.662	0.050	7.622
Step Time [s]	0.545	0.044	8.091	0.526	0.041	7.798	0.558	0.035	6.220	0.531	0.035	6.593
Swing Time [s]	0.420	0.024	5.709	0.427	0.025	5.902	0.437	0.029	6.603	0.419	0.023	5.536
Toe Clearance [mm]	18.664	12.433	66.616	19.582	5.936	30.314	15.183	5.190	34.183	14.518	8.031	55.318
RCOF	0.173	0.022	12.916	0.194	0.018	9.054	0.194	0.028	14.253	0.205	0.031	15.236
TCOF	0.069	0.012	17.589	0.070	0.012	17.572	0.074	0.018	23.699	0.075	0.015	19.426

Changes in slip characteristics due to dual-tasking: It was seen from the parameter *RFootTime2PeakCCIFromLHC* that elderly people generated peak ankle muscle co-contractions in half of the time taken by younger adults (p<0.001) (Table 4 and Table 5).

Also in the same line, the results of LFoot Mean CCI value also depict that mean co-activity in non-slipping foot during the time of slip start to slip stop is significantly higher in older adults.

Table 4 Dual task changes in gait parameters

	Dual Task Walk		Normal Walk		
	Mean	SD	Mean	SD	p-Value
Step length [mm]*	703.79	39.25	750.89	47.87	0.004
Step Width [mm]	119.10	27.67	115.92	25.45	0.69
HCV [mm/s]	1191.20	765.06	1029.41	193.01	0.55
GCT [s]	1.12	0.06	1.08	0.06	0.075
DST [s]*	0.26	0.03	0.24	0.03	0.004
SST [s]*	0.70	0.04	0.66	0.05	0.013
Step Time [s]	0.55	0.04	0.53	0.04	0.067
Swing Time [s]	0.43	0.03	0.42	0.02	0.54
Toe Clearance	16.40	8.29	16.29	7.62	0.96
RCOF	0.19	0.03	0.20	0.03	0.08
TCOF	0.07	0.02	0.07	0.01	0.90

It was also found that Time2SDI was significantly increased in dual task walking trials (p=0.04) although there was no significant differences in SDI (Table 5). Interaction effects were seen for the Lfoot Mean CCI value (p=0.02) for the two independent variables age group and slipping condition (normal vs dual-task).

Table 5 Normal Slip parameters in young and older individuals

	Age Group					
	Old			Young		
	Mean	SD	CV	Mean	SD	CV
Time2SDI [s]	0.046	0.014	31.492	0.051	0.008	16.180
TimeSDI2SDII [s]	0.069	0.013	18.182	0.081	0.032	39.817
TimeSDTotal [s]	0.115	0.024	20.889	0.132	0.034	25.740
SDI [mm]	38.867	23.605	60.733	33.063	13.556	41.001
SDII [mm]	116.722	43.147	36.966	154.175	88.275	57.256
SDTotal [mm]	154.713	64.122	41.446	186.863	94.645	50.650
RFoot MeanCCIvalue	0.081	0.043	53.626	0.093	0.111	119.722
RFoot PeakCCIvalue	0.938	0.809	86.261	0.828	0.396	47.874
RFoot Time2PeakCCIFromLHC* [s]	0.383	0.412	107.398	0.752	0.081	10.722
LFoot_MeanCCIvalue*	0.052	0.030	57.476	0.022	0.013	60.464
PHSV	1033.688	404.843	39.165	1062.313	321.351	30.250
L StanceTime [s]	0.646	0.060	9.215	0.649	0.056	8.612

Table 6 Slip parameters for normal and dual-task conditions

	Dual Task Slip		Normal Walk Slip		
	Mean	SD	Mean	SD	p-Value
Time2SDI [s]*	0.07	0.00	0.05	0.01	0.047
TimeSDI2SDII [s]	0.08	0.01	0.08	0.03	1
TimeSDTotal [s]	0.14	0.01	0.13	0.03	0.46
SDI [mm]	17.65	3.89	35.00	16.65	0.19
SDII [mm]	87.09	20.68	141.69	76.20	0.43
SDTotal [mm]	104.20	16.38	176.15	84.10	0.34
RFoot_MeanCCIvalue	0.06	0.05	0.09	0.09	0.69
RFoot_PeakCCIvalue	0.36	0.01	0.86	0.53	0.24
Rfoot Time2PeakCCIFromLHC [s]	0.87	0.07	0.63	0.29	0.13
Lfoot MeanCCIvalue	0.03	0.03	0.03	0.02	0.90
PHSV [mm/s]	699.23	138.99	1052.77	332.59	0.22
L StanceTime [s]*	0.75	0.02	0.65	0.05	0.03

The L Stance time also increased in dual-task slipping (p=0.03), which was apparent from dual-task normal gait parameters.

Table 7 Slip parameters in young and older individuals in normal and dual task walking condition

	Age Group			
	Old		Young	
	Slip Condition		Slip Condition	
	DS	NS	DS	NS
	Mean	Mean	Mean	Mean
Time2SDI [s] *	0.067	0.046	0.067	0.051
TimeSDI2SDII [s]	0.083	0.069	0.067	0.081
TimeSDTotal [s]	0.150	0.115	0.133	0.132
SDI [mm]	14.903	38.867	20.406	33.063
SDII [mm]	101.71	116.722	72.463	154.175
SDTotal [mm]	115.78	154.713	92.622	186.863
RFoot_MeanCCIvalue	0.093	0.081	0.021	0.093
RFoot_PeakCCIvalue	0.368	0.938	0.355	0.828
Rfoot Time2PeakCCIFromLHC [s]	0.917	0.383	0.817	0.752
Lfoot MeanCCIvalue	0.011	0.052	0.059	0.022
PHSV [mm/s]	797.51	1033.688	600.950	1062.313
L StanceTime [s] *	0.767	0.646	0.742	0.649

Discussion

This study examined the effects of dual-task on older adults and established a relationship between dual-task adaptations in gait and fall risk. Major findings were that the dual-task paradigm influenced slip initiation characteristics by modulating to "safer" or "cautious" gait. Dual task related gait changes are associated with intrinsic (one's health related) risk factors for falls. As we did not have frail individuals in this study, we found that healthy young and older individuals adapted to dual-task scenarios by shifting to more "cautious" gait. This was well evidenced by a decrease in step length, heel contact velocity, and an increase in step width, single and double support time during gait.

The results suggest that attentional capacity limit for healthy young and old adults is perhaps exceeded during dual task walking but did not result in instability or increased fall risk. Collectively, the study findings argue in favor of a critical gait behavior "***Preferred speed of walking in healthy human beings requires less allocation of attentional resources for safe transitioning.***" These findings support previous investigations

1. Previously, Lajoie and coworkers found that reaction times when participants were in single support phase were significantly longer than those in double-support phase, suggesting that attentional demands increased with an increase in balance requirement tasks (Lajoie, Teasdale, Bard, & Fleury, 1993). Thus, attentional demands varied within a gait cycle. Dual-task walking resulted in higher double stance times, thus, it could be inferred that healthy young and old adapt their gait to reduce attentional demand.

2. Previous research has also reported that dual task interference results in slower gait speeds (Bowen et al., 2001; Hyndman et al., 2006; Plummer-

D'Amato et al., 2008), reduced cadence (Kemper et al., 2006; Plummer-D'Amato et al., 2008), shorter stride length (Hyndman et al., 2006; Plummer-D'Amato et al., 2008), increased stride duration (Plummer-D'Amato et al., 2008), and longer double support time (Bowen et al., 2001; Plummer-D'Amato et al., 2010). These cautious gait pattern adopted by healthy adults during dual-tasking , characterized by reduced speed, shorter step length, increased step width, is likely a consequence of adaptations to minimize perturbations to the body and reduce the risk of falls (Maki, 1997) during reduced attentional demands of walking. We also found that heel contact velocity and required coefficient of friction decreased during dual tasking. This unwittingly indicates that several mechanisms contribute to reduce risk of falls and adapt body movements to cautious gait mode, when less attentional resources are available for gait.

Because walking has greater attentional demands, from an information processing viewpoint, walking is not considered an automated task requiring no cognitive processing (Lajoie et al., 1993). Overall analysis of this study suggests that gait in healthy adults was affected by concurrent cognitive tasks and the evidence is sufficiently robust to support the notion of **cautious gait.** Even in healthy individuals, age-related changes have been reported in cognitive and motor systems, thus aging may attribute to higher cognitive-motor interference (Judge, Ounpuu, & Davis, 1996; Seidler et al., 2010). We believe that the dual-task changes observed are compensatory mechanisms to stabilize and allow safe locomotion, in a condition when less attention is available.

Intuitively, the concept of cautious gait adaptations observed in healthy younger and older adults while walking in a dual-task paradigm, draws intriguing

interests in understanding "**Why humans threaten their safety when walking at their preferred speed without dual-tasking?**"

Findings of this study elucidate that dual-task related changes in gait do not predispose healthy young and older adults to falls. Healthy people walk at their preferred walking speed, step length and cadence which is selected to optimize the stability of their gait pattern ("2012 National Population Projections," ; Latt, Menz, Fung, & Lord, 2008), this has been addressed in several studies in context of spatial variability (Brach, Berthold, Craik, VanSwearingen, & Newman, 2001; Sekiya et al., 1997; Yamasaki, Sasaki, & Torii, 1991), and temporal variability(Frenkel-Toledo et al., 2005). It is reported that shorter steps and longer double support times are associated with small sensorimotor and fronto-parietal regions, whereas cognitive processing speed is linked to individual differences in gait (Rosano et al., 2008). Accordingly, Sekiya et. al. (Sekiya et al., 1997) suggested an optimal method of walking that consists of optimal criteria in terms of energy efficiency, temporal and spatial variability, and attention (Figure 3). During dual task walking, less attentional resources are allocated for gait; thus, there is compromise with energy expenditure and variability of kinematic parameters. The present study determined that older adults redressed the diminished attentional investment through differing variability in selective gait parameters including standard deviation and coefficient of variation of step width, HCV, DST, SST and gait cycle time. Therefore, future studies in energy expenditure with a dual-task criteria would further strengthen intuitive understanding on relationship of energy expenditure, variability and attention.

This study also evaluated the effects of the dual-task paradigm on slip severity, and on linear and nonlinear measures of variability changes in two known healthy age groups. The results suggest that single stance duration is increased in dual-task walking trials which may elicit a congruent adaptation in both young and

elderly individuals to maintain "stable gait." During the stance phase of a gait cycle, proprioceptive input from extensor muscles and mechanoreceptors in the sole of the foot provide loading information (Dietz & Duysens, 2000) to the central nervous system. Thus, the increased stance duration increases foot-loading information through afferent sensory and proprioceptive mechanoreceptors, such as Golgi-tendon units, muscle spindles, and joint receptors, and may facilitate motor control of the lower extremity during walking (Zhang, Lockhart, & Soangra, 2013).

Additionally, the dual-task condition shortened step length significantly, which may be associated with modulation of self-selected pace in order to continue counting rhythmically and need of longer (longer single and double stance) and frequent (shorter steps) proprioception due to dual-task (Smith, Kubo, & Ulrich, 2012); or perhaps due to changes in the motor control schema with the adoption of alternative compensatory strategies to increase stability while walking with another task being given primary importance and walking being innately automatic to some extent.

Furthermore, dual-task trials did not significantly affect heel contact velocity (HCV), but slightly decreased RCOF and TCOF in walking trials although these effects were not statistically significant. Considering HCV as a kinematic gait parameter that can drastically alter the friction demand (by change in required coefficient of friction) [51] and influence the likelihood of slip-induced falls (Karst, Hageman, Jones, & Bunner, 1999; Mills & Barrett, 2001; Parijat & Lockhart, 2008), dual-task conditions ultimately decrease slip-induced fall risks. Likewise, considering dual-task events had no deleterious effects on toe clearance, therefore, it is inferred that participants delineated reduced trip risk as well.

Through the parameter RFoot Time2PeakCCIFromLHC, the results suggest that elderly people generated peak ankle muscle co-contractions in half of the time

taken by younger adults i.e., they may be quick to introduce ankle muscle coactivity in the slipping foot. Further, coactivity during slipping limits the ankle joint's degrees of freedom, thus reducing requisite motor control adaptability to recover balance post-slip. In turn, this affirms that the health status of older age group participating in this study was non-frail. It should be noted that the subject population recruited for the study was healthy with intact cognitive function (or EF) (MMSE=30).

The LFoot Mean CCI value which depicted that the mean co-activity in the non-slipping foot during the time of slip start to slip stop was significantly higher in older adults. This phenomena also lowers the degrees of freedom in the non-slipping foot. The reduction in degrees of freedom in both the right and left foot may influence slip severity amongst older adults. Although greater SDI was reported in the older population, in accordance with previous studies (T.E. Lockhart, Smith, & Woldstad, 2005; T. E. Lockhart et al., 2003), they were not significantly different for the current population. It was found that the Time2SDI in the dual task paradigm was significantly increased, thus if elderly individuals have a higher SDI, they require lower heel velocities to cover SDI, thus shows slower movement of heel from slip start to mid-slip is seen as an effect of dual tasking. Probably, this could also be partially explained by higher transitional acceleration of center of mass, in these individuals. Interaction effects were seen for the Lfoot Mean CCI value for the two independent variables age group and slipping condition (normal vs dual-task). Which is interesting because older people have lower non-slipping ankle co-activation (or stiffness reduced) during dual-tasking when compared to normal slipping. On the contrary, younger individuals have higher co-activity in ankle muscles of non-slipping foot during dual-activity. This might be influenced by age related involvement of attentional resources for the dual-task, perhaps this dual-task (counting backwards by

subtracting 3) may not be challenging enough to involve higher attentional resources for younger participants.

In sum, this study investigated the effects of attentional interference (induced by dual task) on gait variability and associated fall risk, particularly to understand: (a) what is the effect of dual task on spatio-temporal gait parameters (b) does dual task deteriorate or modify to unsafe gait by predisposing to falls. The findings suggest that in everyday walking tasks with increased attention demands would certainly reduce the resources available for other tasks which may be secondary. But the slow speed, wider step width and longer double support time adopted by participants ("2012 National Population Projections,") may serve to produce a more safer and stable gait (Ferrandez, Pailhous, & Durup, 1990; Winter, Patla, Frank, & Walt, 1990), energy-efficient speed of progression (Ferrandez et al., 1990; Larish, Martin, & Mungiole, 1988), or probably to maintain certain amount of variability in its kinematics. A cautious gait can be typically marked by moderate slowing, reduced stride length, mild widening of base-of-support characterized by step width (Nutt, 2001). It is also possible that the kinematic adaptations adopted may serve to reduce the cognitive demand necessary to control the continuous disequilibrium inherent to walking (Lajoie et al., 1993; Lajoie, Teasdale, Bard, & Fleury, 1996).

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Chapter 4: Can inertial sensors measure movement variability in young and older subjects

Abstract

The study of kinematic variability in human movements, which are performed commonly in daily life such as walking, standing, and sit-to-stand could offer complementary way of quantifying fall risk associated with disease as well as means of monitoring patients for effects of therapeutic interventions and rehabilitation. Previous work has suggested that variability in inter-stride time, step width are closely related to falls. Inertial sensors are appealing for unconstrained and non-invasive ambulatory measurements with heuristic approach to summarizing variability measures. Previous studies on dual task suggest that stride-to-stride fluctuations, step width and stride time variability are influenced by attention loading and are related to fall risk. Furthermore lateral walking is also associated with more attentional demand. This study uses inertial sensors to provide new insights into the factors that regulate gait variability in healthy young and older adults and the practical application of measures of the variability in the clinical settings.

Introduction

Human movement variability may be defined as normal variations that occur while performing motor task across multiple repetitions (Stergiou, Harbourne, & Cavanaugh, 2006). A person being persistent and lacking variability in movement may indicate rigidness or inflexible motor behaviors and may have limited adaptability towards changing task and environmental demands, whereas greater than optimal variability characterizes human movement as noisy and unstable. An optimal amount of variability may be defined as the amount of variability necessary for healthy biological systems to be adaptable and flexible in unpredictable and ever changing environments.

Multiple degrees of freedom of the human body parts, including muscles, joints and nervous system along with external forces produce countless patterns, forms, and strategies during the movement (Bernshteĭn, 1967). With aging there are several changes in muscle properties that, may influence movement execution. Some of these changes include reduced muscle cross section area(Kent-Braun & Ng, 1999) with reduction in strength and fiber type distribution (Lexell, 1995) which reduces the movement speed. There is also decrease in motor units (Galea, 1996) and all these changes have implications for execution of movement in old age(Kurz & Stergiou, 2003). Old age and frailty may result in reduction of muscle strength and force production, which has a significant influence on movement trajectories and final position accuracy for rapid movements (van Galen & van Huygevoort, 2000).

Aging and disease reduce the capacity of the frail people to adapt to various forms of stressors that can potentially be hazardous in maintaining survival. This reduced capacity to adapt to stress may be due to loss of complexity with aging and disease (Lipsitz & Goldberger, 1992). This reduced complexity (Lipsitz & Goldberger, 1992), is dependent on the nature of the intrinsic dynamics of the system

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and one's ability for short time adaptive change, which is required to meet an immediate task demand is reduced (Vaillancourt & Newell, 2002).

Nonlinear tools help in describing structure of the variability (as opposed to amount of variability described by linear statistics such as standard deviation, coefficient of variation, range etc.). The linear measures of variability do not accurately define constructs important in movement, such as stability, because they only provide insights into the amount of variability (Harbourne & Stergiou, 2009). Nonlinear tools are used to assess variation in evolution of movement with time. Structure of variability is actually temporally organized and is quantified by the degree to which values emerge in an orderly (i.e., predictable) manner (Harbourne & Stergiou, 2009). The structure of variability pertains to the time ordered variance within the human movement (Wurdeman & Stergiou, 2012). Nonlinear variability will be expressed using the largest Lyapunov exponent.

Although traditionally, random error or noise within neuromuscular system was deemed responsible for movement variability (Glass & MacKey, 1988), nonlinear theories emphasize disequilibrium as an indicator of health. Thus according to nonlinear theories constant fluctuations characterize the healthy variability that allows for adaptation to environmental change (Rickles, Hawe, & Shiell, 2007). Instead of being a negative feature, variability exhibits important information for maintenance in healthy systems. Lack of variability in human movement may lead to abnormal mapping of the sensory cortex, which may subsequently disrupt motor function (Harbourne & Stergiou, 2009). The sensory and motor neural maps are more complex when movement variability is present and less complex when variability gets reduced (Byl, Nagarajan, Merzenich, Roberts, & McKenzie, 2002; Harbourne & Stergiou, 2009; Merzenich & Jenkins, 1993; Nudo, Milliken, Jenkins, & Merzenich, 1996). Variability allows different choices among available options, selection of

appropriate strategies, and flexibility to adapt to variations in the environment

(Harbourne & Stergiou, 2009). Variability may be viewed as increased flexibility of

skill to allow adaptation to external perturbations. Thus it is important to understand

how much is variability regulated in healthy young and older adults and this

information can be practically applied in assessing fall risk and post-operative clinical

outcomes.

The objective of this study was to evaluate linear and nonlinear variability in

healthy young and older adults using inertial sensors. Differences in gait speed,

postural transition times, postural stability, and dynamic stability between younger

and older adults were determined using inertial sensors. Temporal and intensity of

movement was also evaluated for both age groups.

Materials and Methods

Participants:

Seven young and seven old participants were recruited in this study. The recruited

participants were in a general good health condition, with no recent cardiovascular,

respiratory, neurological, and musculoskeletal abnormalities. Only one of the elderly

participant (O02) was suffering from chronic obstructive pulmonary disease (COPD).

All patients were recruited based on criteria of complete ambulation, without the use

of any assistive devices, and ability to rise from a chair without assistance and free of

orthopedic injury. This study was approved by the Institutional Review Board (IRB)

of Virginia Tech. All participants who participated in this study provided written

consent prior to the beginning of data collection. Demographic information is

provided in Table 8 below.

Instrumentation:

The experiments were conducted on a 15-meter linear walking track, embedded with two force plates (BERTEC #K80102, Type 45550-08, Bertec Corporation, OH 43212, USA and AMTI BP400600 SN6780, Advances Mechanical Technology Inc., Watertown, MA 02472, USA). A six-camera ProReflex system (Qualysis Gothenburg, Sweden) was used to collect three-dimensional kinematics of posture and gait data in participants. Kinematic data were sampled and recorded at 120 Hz. Ground reaction forces of participants walking over the test surfaces were measured using two force plates and sampled at a rate of 1200 Hz. Inertial measurement unit (IMU) used is TEMPO (Technology- Enabled Medical Precision Observation) 3.1 which is manufactured in collaborative research with inertia team in UVA (Barth, Hanson, Powell Jr, & Lach, 2009). It consists of MMA7261QT tri-axial accelerometers and IDG-300 (x and y plane gyroscope) and ADXRS300 as z-plane uniaxial gyroscope. The data acquisition was carried using a bluetooth adapter and laptop through a custom built LabView VI(Barth et al., 2009).

Table 8 Means and standard deviation of subjects' demographic information

	Age Group			
	Old		Young	
	Mean	SD	Mean	SD
Age	71.143	6.5174	22.643	2.5603
Height [cm]	174.571	10.2446	170.376	9.3302
Weight [Kg]	78.559	18.2576	69.651	15.5270
BMI	25.529	4.2731	23.786	4.0004

Another data acquisition IMU device used in this study was a smartphone (iPhone 5, Apple Inc., Cupertino, CA), which contains an ultra-compact low-power high performance 3-axis "nano" MEMS accelerometer, LIS331DLH. The LIS331DLH has user selectable full scales of ±2g/±4g/±8g and it is capable of measuring accelerations with output data rates 0.5Hz to 1kHz. It is capable of measuring acceleration data

with data sampling rate of 1000Hz. It also contains low-power 3-axis angular rate sensor, L3G4200D. The L3G4200D has a full scale of $\pm250/\pm500/\pm2000$ degrees per second and is capable of measuring angular rates at user-selectable bandwidth. An iOS 6 based App, named as "Lockhart Monitor" was designed to collect data at sampling frequency of 50 Hz. The App was programmed in objective C language using Xcode 4 IDE (Integrated Development Environment). The data was collected from inbuilt sensors, accelerometers and gyroscopes in the smartphone and stored in it. The collected data was either transferred using cloud service/ Email or by a USB cable to the computer for data analyses.

Experimental Preparations and Practices:

All participants were first familiarized with laboratory equipment's and were explained verbally about the experimental procedure. Participants were requested to sign a written consent form as per regulations of Virginia Tech Institutional Review Board (VT IRB). Participants were then requested to wear laboratory clothes and shoes, fitting to their sizes. Height and weight of participants were noted below the ID numbers assigned to the subject. IMU's were then affixed at sternum, right wrist, sacrum, right thigh and right shank. Smartphone was affixed at the right side of pelvis using smartphone holster and belt clip.

Protocol

The experiment was divided into two sessions: normal session and dual-task session (Figure 10). Each session was separated by 4 days and each participant was randomly assigned to either normal or dual-task as his/her first session.

IMU data processing:

The IMU data was post processed using custom software written in Matlab (MATLAB version 6.5.1, 2003, computer software, The MathWorks Inc., Natick, Massachusetts). Empirical Mode Decomposition (EMD) (Huang, Shen, & Long, 1999; Huang et al., 1998) is an adaptive time-frequency data analysis method and can adaptively divide the IMU signals into different intrinsic mode function (IMF) components according to different time scale, and noise which mainly concentrates in the high-frequency component. IMU signals are decomposed adaptively into oscillatory components called intrinsic mode functions (IMFs) by means of a process called *sifting*. The sifting process has two effects: (a) to eliminate riding waves and (b) to smooth uneven amplitudes. The traditional EMD involves the decomposition of a given signal x(t) into a series of IMFs, through the sifting process, each with distinct time scale (Huang et al., 1998).
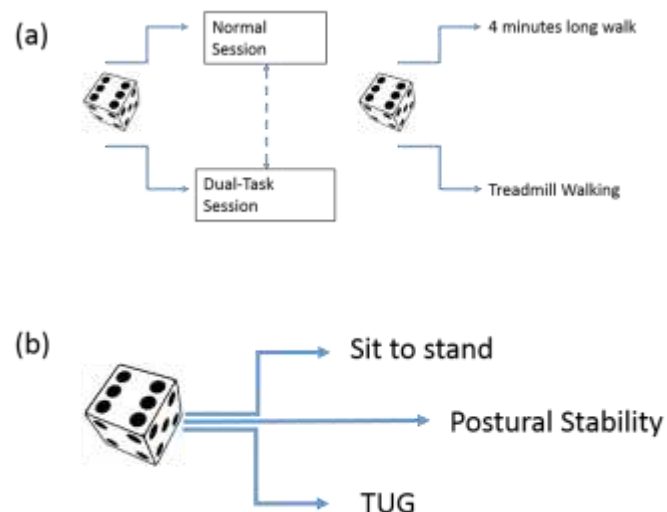
Figure 10. Participants were assigned to normal or dual-task session randomly and the listed tests were conducted

Unlike wavelet decompositions the major advantage of the EMD is that the basis functions (mother wavelet; in case of wavelet transform) are derived from the signal x(t) itself. Each IMF replaces then the detail signals of at a certain scale or frequency

band (Flandrin, Rilling, & Goncalves, 2004). The EMD picks out the highest frequency of oscillation that remains in the signal. An IMF has to follow two requirements: (R1) the number of extrema and the number of zero crossings are either equal or differ at most by one; (R2) at any point, the mean value of the envelope defined by the local maxima and the envelope defined by the local minima is zero. The sifting is repeated several times in order to get to be a true IMF that fulfills the requirements (R1) and (R2) mentioned above. The result of the sifting procedure is that x(t) is decomposed into IMFs, $IM_j(t)=j=1,2,....N$ and residual $r_N(t)$.

$$x(t) = \sum_{j=1}^{N} IM_j(t) + r_N(t) \qquad\qquad 8$$

Let $C_j(t)$ be a deterministic IMF with the finite length L and IM_j the corrupted IMF with additive noise $b_j(t)$ with variance $\sigma_j^2(t)$:

$$IM_j(t) = C_j(t) + b_j(t) \qquad\qquad 9$$

Ensemble Empirical Mode Decomposition (EEMD), is another approach which consists of sifting an ensemble of white noise-added signal and treats its mean as the true result (Wu & Huang, 2009). We used EEMD-Golay denoising on IMU signals. The number of ensembles chosen is 100 with ratio of standard deviation of the added noise to that of signal as 0.2. First half of the IMF's containing high frequency noise are filtered using savitzky-golay filter (polynomial order 3 and number of frames as 41) (Lockhart & Shi, 2010) and then reconstructed to get the denoised signals.

Dependent Variables

Sit-to-Stand (STS) Maneuver: The participants sat comfortably on chair with backrest and arm-rest with their thighs and feet parallel and were instructed to use arm-rest/ knee for support while performing STS task. The chair was kept on one force plate and the feet of participants rested on another. The spacing between feet

was maintained at 15 cm. Chair popliteal height was 45 cm and knee angle was maintained from 85°-90° using Styrofoam. Participants were instructed to sit such that thigh did not rest on seat and only buttocks rested on it. Participants were asked to wait for an auditory signal before initiating movement. The data was recorded for 12 seconds in total. The participants were given the auditory signal to stand after at least 3 seconds of data collection in order to ensure sitting, postural transition and stabilized standing is collected in all trials. Each participant performed three STS task using arm rest (STSA) and three STS using knee support (STSK) for rising. The investigator demonstrated the STS task prior to data collection.

   Timed Up and Go (TUG) Task: TUG included a series of daily life functional maneuvers, such as sit-to-walk, turning and stand to sit. Participants were given verbal instructions and also demonstrated how the test should be performed. They were instructed to rise from the chair (either knee or arm support), walk 3 meters as quick as possible, cross a line that contrasted with color of the floor, walk back, and sit down again. The participants were provided with time to familiarize with the task, by practicing one or two times, before timing them. The timing was started when the participants back left the back rest of chair and stopped when his/her buttocks touched the seat again.

Long Walk for 4 minutes: Participants were asked to walk continuously for 4 minutes around a square hallway with 20 meters straightway inside the building. Participants were also instructed that they have to walk uninterrupted and keep walking steadily in their self-selected pace. The participants were randomly assigned either normal walk or dual-task walk as their first trial of walking.

Postural Stability: Participants were asked to wear smartphone holster on the right side at pelvic level. The participants were further instructed to stand on force platform. Participants were tested for postural stability with **eyes open**, **eyes closed**

condition. The total data collection was two minutes for each condition and appropriate rest was provided in order to ensure fatigue did not develop in lower extremity muscles.

Five and ten meter gait speed: Participants were asked to wear smartphone holster on the right side at pelvic level. Participants were instructed to walk at their own preferred pace to the other end of the course (5-meter or 10-meter distance). Participants were allowed to use their walking aid or cane. Before testing, the participant's dominant foot was determined, and the participant was instructed to lift that foot first in each trial. Also before testing, the participant was asked to stand comfortably on the floor and the position of the feet was marked so that initial foot position remained constant across all trials.

Treadmill walking: Participants were asked to walk with four different conditions on a treadmill for 4 minutes, 1) Forward Walking- participants had to face forward along the direction of progression while walking. 2) Forward Walking with counting backwards- Participants had to face forward along the direction of progression while walking and had to count backwards. 3) Lateral stepping- participants had to walk sideways on a treadmill and had to face orthogonal to the direction of progression while walking. 4) Lateral stepping with dual tasking: participants had to walk sideways on a treadmill and had to face orthogonal to the direction of progression while walking along with counting backwards (Figure 11). Participants were corrected if they made an error in counting.

Participants were asked to face to their left, such that their right leg is leading leg and left leg is lagging leg. They were also asked to keep their head up while stepping, and not to cross their legs at any time, and not to have both of their feet off the ground at any time. Participants were asked to walk at their own preferred speed and this speed was determined by incrementally increasing the treadmill speed at

0.45m/s (Myers et al., 2009), until the participant indicated that his/her preferred speed was reached. Speed was incremented again to get a reconfirmation from the participant that it was fast and was reduced back to preferred speed. If it was not confirmed by the participant that the speed is preferred walking speed, then the process was repeated until the preferred speed was reached. After selection of preferred speed by the participants, they were given 10 minutes to get used to the treadmill; this amount of time has been previously found to be adequate to achieve a proficient treadmill walking pattern (Matsas, Taylor, & McBurney, 2000).This was followed by a minimum of 3 minute rest. Then the participants were requested to perform all four conditions of walking. The order was random for all four conditions.
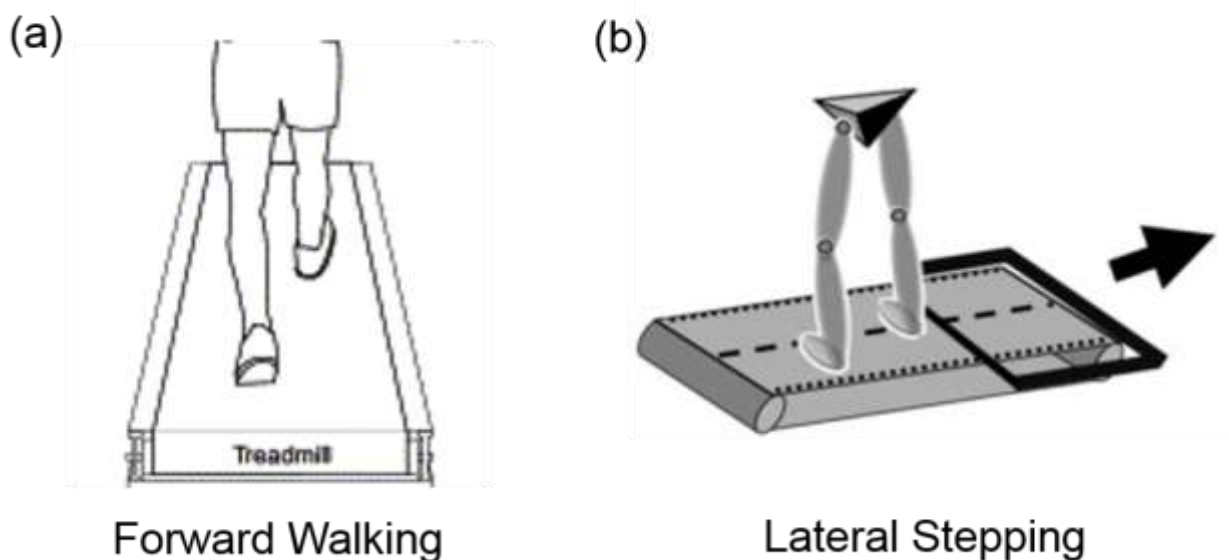
Figure 11. (a) Forward walking and (b) lateral walking on treadmill

Postural Stability

Postural control is the foundation of person's ability to walk and stand independently. Postural stability is defined as an ability of a person to maintain the position of body within specific boundaries of space without moving out from base of support. This

kind of postural control requires complex integration of sensory information regarding position of the body relative to the surrounding (proprioception, visual and vestibular), and the ability to generate appropriate forces to control the body movement.

It is well documented in previous studies that difficulties in maintaining balance and postural control pose a serious consequence for the older adults. Falls due to loss of balance is a serious problem for older people. This loss of balance and fall risk can be assessed by measuring postural sway during quiet and perturbed standing (such as eyes closed) (Campbell, Borrie, & Spears, 1989; Lord & Clark, 1996; Sherrington & Lord, 1998). Deterioration of postural stability in elderly persons may lead to falls and impaired balance has been well correlated with fall risk (Tinetti, Speechley, & Ginter, 1988). IMU can be a reliable method for measurement of balance during standing and walking, with high absolute test-retest reliability(Moe-Nilssen, 1998). Some researchers have used a trunk-mounted accelerometer to distinguish between balance conditions including standing on rigid, compliant surfaces, standing with feet together, feet apart, eyes open and eyes closed.

Postural Stability Assessment using IMU

Postural stability can be assessed using tri-axial accelerometers and computing angles made by sensitive axes along the direction of gravity (Figure 12).

$$a = \cos^{-1}\frac{a_x}{A} \qquad\qquad\qquad 10$$

$$b = \cos^{-1}\frac{a_y}{A} \qquad\qquad\qquad 11$$

$$g = \cos^{-1}\frac{a_z}{A} \qquad\qquad\qquad 12$$

Considering dz = height of IMU

$$D = d_z / \cos g \qquad\qquad 13$$

$$d_x = D * \cos a \qquad\qquad 14$$

$$d_y = D * \cos b \qquad\qquad 15$$

where d_x and d_y are representative of center of pressures x and y coordinates.

Sway Area: Sway area is computed by first calculating the mean radius of sway and then using area formula for a circle.

$$Sway\, Area = pr^2 \qquad\qquad 16$$

$$Where \quad r = \sqrt{d_x^2 + d_y^2} \qquad\qquad 17$$
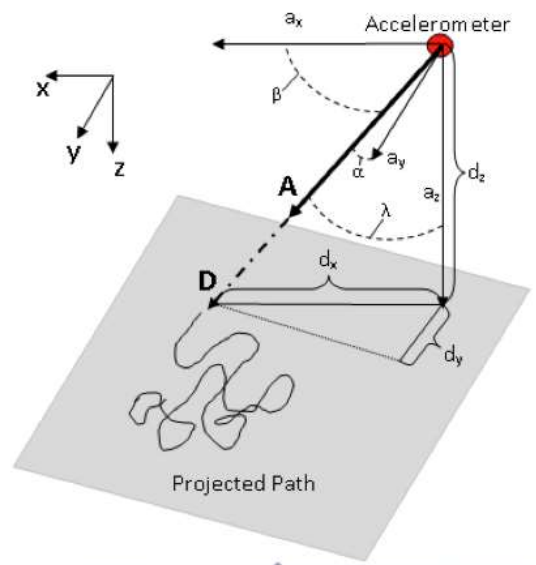
Figure 12. Schematic diagram of projection of sway

Sway Velocity: Sway Velocity is computed using total path length of sway over total time of data collection.

$$Sway\ Velocity = \frac{Total\ Sway\ Path\ Length}{Total\ Time\ Taken}$$

$$Total\ sway\ path\ length = \sum_{i=1}^{n-1} r_{i+1} - r_i$$

Five and Ten Meter Gait Speed:

Gait velocity is determined by computing time taken to walk a specified distance (5m or 10m).

$$GaitVelocity = \frac{Dis\tan ce\ Walked}{TimeTaken}$$

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In order to determine **start** and **stop event** of walking. Resultant acceleration is computed using the equation:

$$A = \sqrt{a_x^2 + a_y^2 + a_z^2}$$

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where a_x , a_y and a_z are accelerations in X , Y and Z sensitive axes of smartphone based inertial sensor. Then variance is computed over a window of 0.5 seconds. Figure 13 below illustrates the method of computation of walking time.

Timed Up & Go Test (TUG): The timed "Up and Go" (TUG) test, designed and developed by Podsiadlo and Richardson (Podsiadlo & Richardson, 1991) has been used to investigate functional mobility among the elderly individuals. The reliability and validity of TUG is high and healthy older people perform the TUG in 10 seconds or less (Podsiadlo & Richardson, 1991). TUG time was noted down using a stopwatch by the investigator.
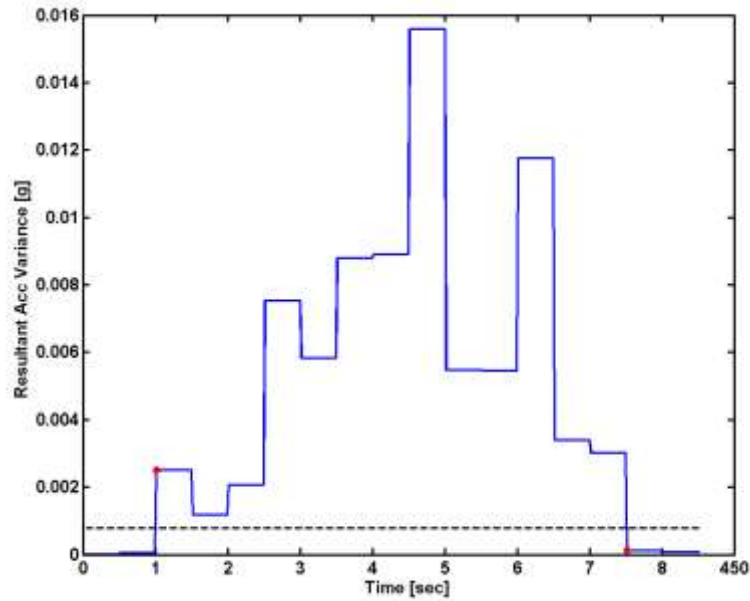
Figure 13. Resultant acceleration variance computed using moving window of 0.5 sec

Local dynamic stability computation: According to Taken's theorem (Takens, 1981), any single dimensional time series can be used to reconstruct a multi-dimensional state space via time-delayed coordinate approach and this phase plot created contains information for the underlying dynamics of the system. Two parameters, minimum embedding dimension (d_E) and the time delay (T), can be determined via the auto mutual information approach (Cao, 1997) and nearest false neighbors approach (Abarbanel, Brown, Sidorowich, & Tsimring, 1993). For each participant 50 continuous gait cycles was extracted from shank medio-lateral angular velocities and resampled to 5000 frames (Figure 14, 15 & 16).
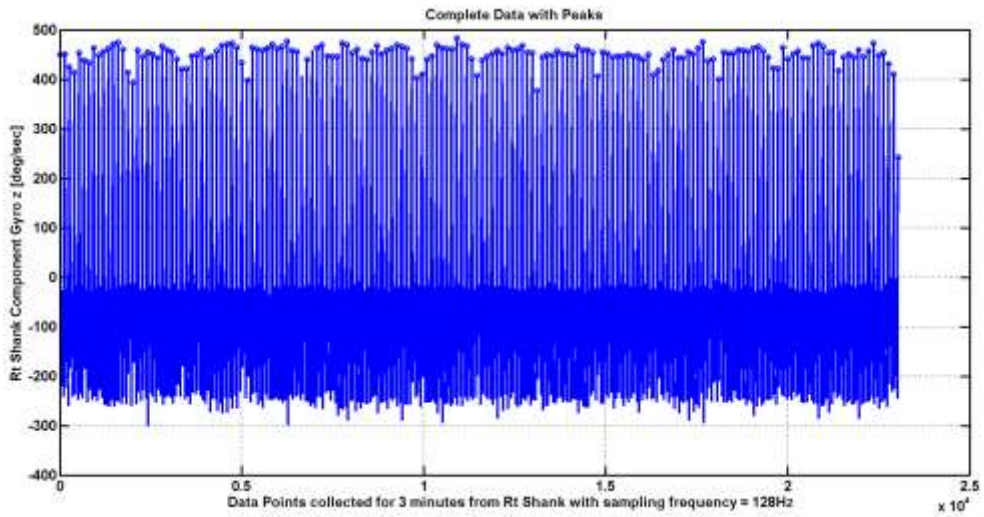
Figure 14. Three minutes of normal walking data from right shank gyroscope
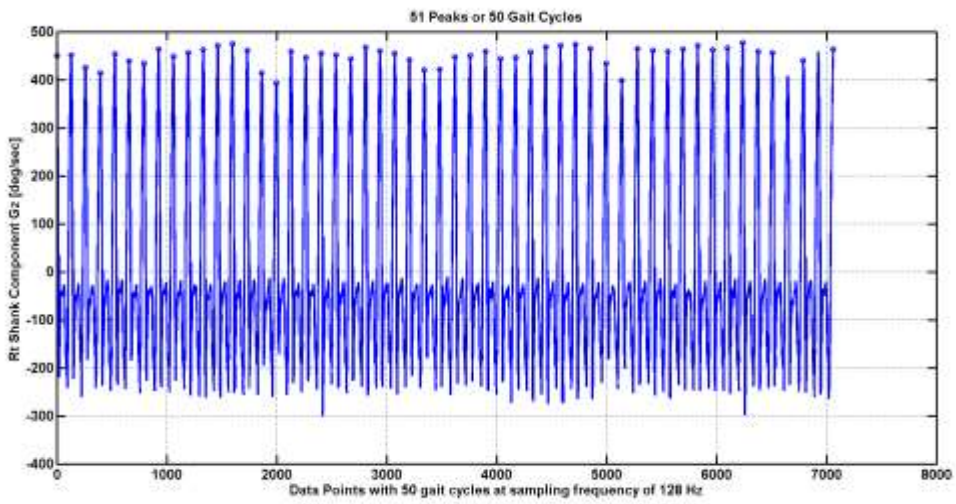
Figure 15. Fifty complete gait strides were identified from 51 peaks
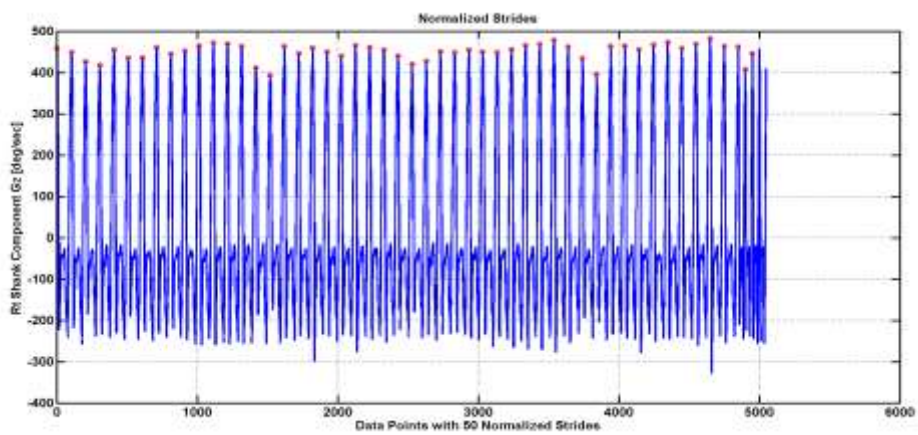
Figure 16. Each stride was normalized to 100 data points

Thus, each gait cycle was normalized to 100 frames. This re-sampling approach was done to ensure that between-subject comparison could be made on the same timescale without removing the cycle-to-cycle temporal variability information (England & Granata, 2007). The time-delayed coordinate approach (Packard, Crutchfield, Farmer, & Shaw, 1980) was then used to reconstruct the state space with embedding dimension 5 (Figure 18) and the time delay of 15 frames (Figure 17).
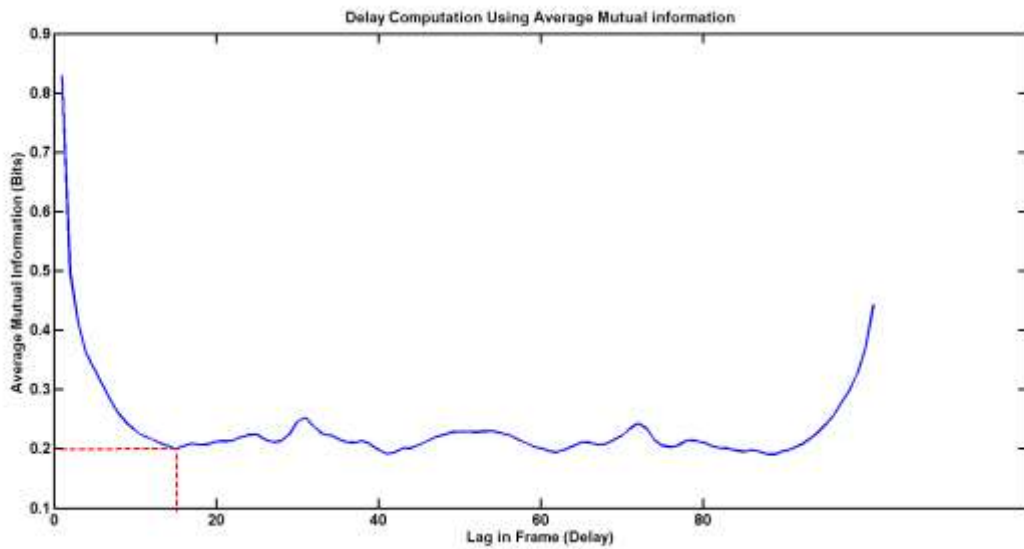
Figure 17. Delay computation by minimum average mutual information
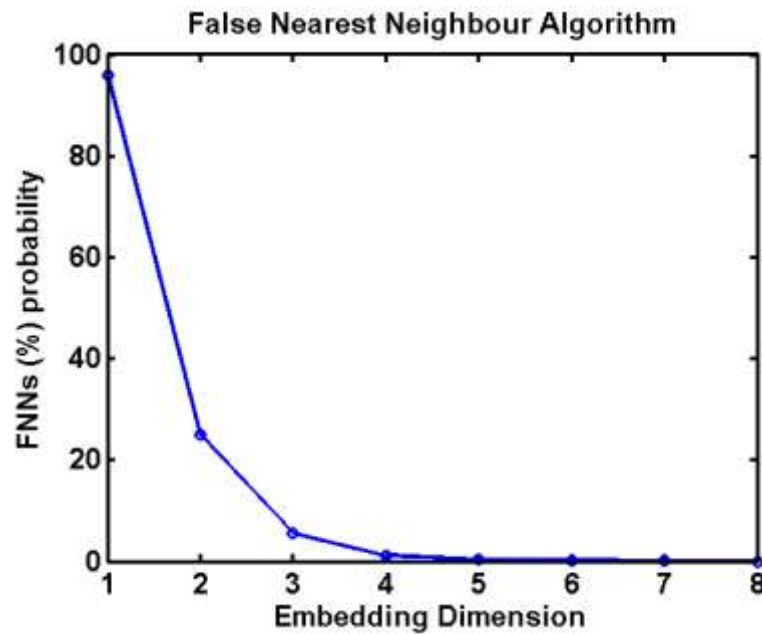
Figure 18. Embedding dimension selection using false nearest neighbor

With an initial single dimension time series data x(t), embedding dimension d_E, and delay T, the state space X(t) can be reconstructed (Figure 19) as

$$X(t) = [x(t), x(t + T), \dots, x(t + (d_E - 1)T)] \qquad 22$$
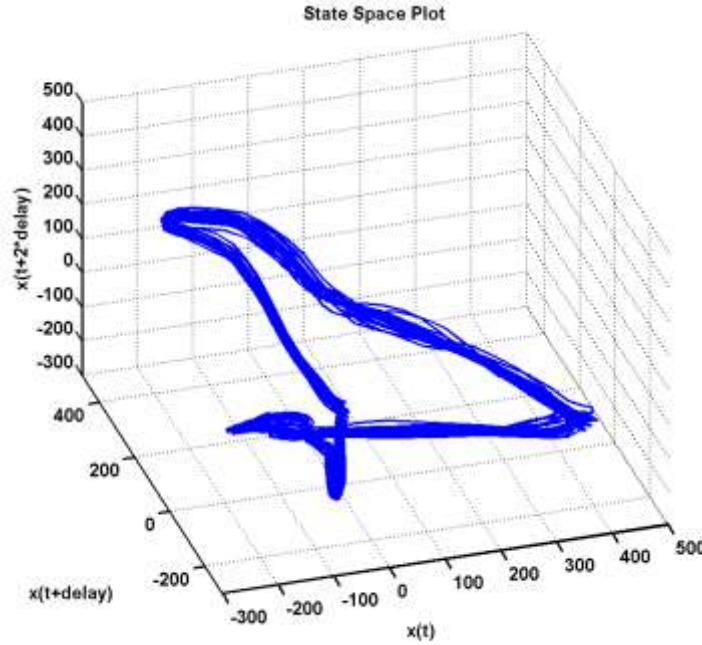
Figure 19. Representative state space plot

Rosenstein's algorithm was applied to compute average divergences between neighboring trajectories in the reconstructed state space (Rosenstein, Collins, & De Luca, 1993). The nearest neighbor points on separate strides diverge at a rate given by the max LE (Lockhart & Liu, 2008).

$$\lambda(i) = \langle \ln[D_j(i)] \rangle / \Delta t \qquad 23$$

Where $D_j(i)$ is the Euclidean distance between the jth pair of nearest neighbors after i discrete time steps, Δt is the sampling period of the time series data and <…> denotes average over all values of j.
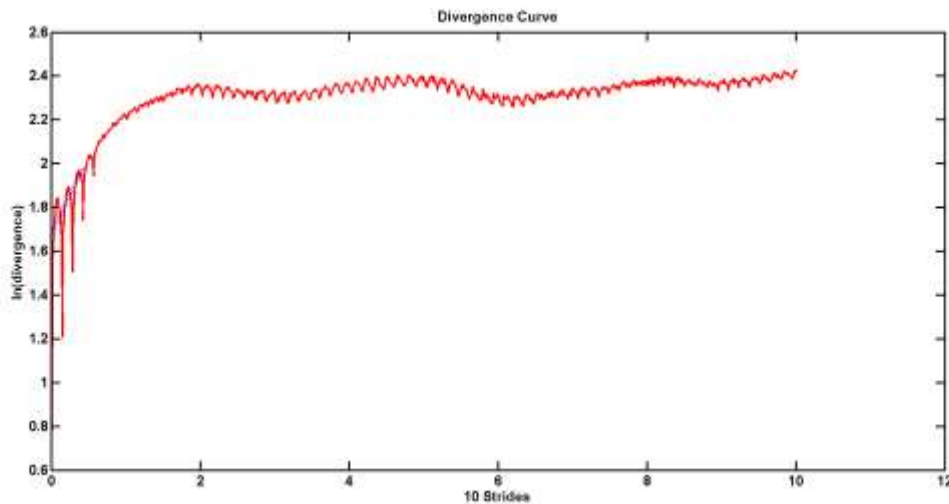
Figure 20. Divergence curve and slope for maxLE

The max LE was then calculated as the logarithmic rate of average divergence with

regard to the time duration of 0-50 frames (which corresponds to first gait step) and 0-

100 frames (which corresponds to first gait stride) (Figure 20).

Non Linear Measures of Postural Complexity:

Nonlinear dynamics is a powerful tool to understand neuromuscular control

mechanisms involved in biological time series such as COPx (Anterio-posterior axis)

and COPy (Mediolateral axis). Approximate entropy (ApEn) is a recently developed

statistic quantifying regularity and complexity that appears to have potential

application to a wide variety of physiological time-series data. Approximate entropy

(ApEn), is an approach to quantify the complexity and regularity of a system, which

was introduced by Pincus (S. M. Pincus & Goldberger, 1994). It suggests that

postural stability arises from the combination of specific feedback mechanisms and

spontaneous properties of interconnected neurons, thus a weak or degraded

neuromuscular mechanism may be characterized by an increased irregularity in the

physiological time series (S. Pincus, 1995; S. M. Pincus, 2006).

ApEn is considered to provide a direct measurement of feedback among neuro-muscular connections, and low ApEn would indicate high predictability and regularity of time series data, where as high ApEn values would indicate unpredictability and random variation (Khandoker, Palaniswami, & Begg, 2008; S. Pincus, 1995).

Approximate Entropy: The algorithm for estimating ApEn was first reported by Pincus (S. M. Pincus & Goldberger, 1994). We here explain that approach as applied to Center of pressure (COP) data. ApEn is defined as the logarithmic likelihood that the patterns of the data that are close to each other will remain close for the next comparison within a longer pattern. Given a sequence of total **N** numbers of COP (x or y coordinate) e.g.. COPx(1), COPx (2),........., COPx (N), similarly for COPy(1), COPy (2),........., COPy (N). To compute ApEn of each COPx & COPy data set, m-dimensional vector sequences pm (i) were constructed from the COP time series similar to [pm (1), pm (2),................, pm (N-m+1)], where the index i can take values ranging from 1 to N-m+1. Where the distance between two vectors pm (i) and pm (j) is defined as |pm (j) - pm (i)|,

$$C_i^m(d) = \frac{1}{N-m+1} \quad \text{such that } |P_m(j) - P_m(i)| < d$$

24

Where **m** specifies the pattern length which is 2 in our study, **d** defines the similarity coefficient which has been set at 0.2% of the standard deviation of 7200 COP data (collected for 1 minute at 120Hz sampling frequency) which can produce reasonable statistical validity of ApEn (S. Pincus, 1995). These constants yielded statistically reliable and reproducible results. Cim(d) is considered as the mean of the fraction of patterns of length *m* that resemble the pattern of the same length that begins at index i. ApEn is computed as

$$ApEn(N, m, d)$$

$$= (N - m$$

$$+ 1)^{-1} \sum_{i=1}^{N-(m-1)} ln C_i^m(d) - (N - m)^{-1} \sum_{i=1}^{N-m} ln C_i^{m+1}(d)$$

25

Detrended Fluctuation Analysis

Detrended fluctuation analysis (DFA) is a tool used to detect long range correlations in time series with non-stationarities(Peng et al., 1994). To detect long-range correlations, intrinsic trends from long-range fluctuations are firstly distinguished, as strong trends in data can mislead for long range correlations. DFA can systematically eliminate trends and provide insights into scaling behavior of natural variability and trends in time series. DFA is computed in two steps

1. The time series B(k) is shifted by the mean <B> and integrated (cumulatively summed),

$$y(k) = \sum_{i=1}^{k} [B(i)- < B >]$$

26

then segmented into windows of various sizes Δn; and

2. In each segmentation the integrated data is locally fit to a polynomial

$$y_{\Delta n}(k)$$

27

and mean-squared residual F(Δn) (fluctuations) with N as total number of data points

$$F(\Delta n) = \sqrt{\frac{1}{N} \sum_{k=1}^{N} [y(k) - y_{\Delta n}(k)]^2}$$

28

$F^2(\Delta n)$ is the average of the summed squares of the residual in windows. DFA procedure tests for self-similarity or fractal properties at different resolutions (windows sizes).

$$F(\Delta n) = C(\Delta n)^{\propto} \qquad\qquad 29$$

where C is a constant and α is estimated from a least-square fit.

$$\ln\big(F(\Delta n)\big) = \propto \ln(\Delta n) + \ln(C) \qquad\qquad 30$$

This scaling coefficient α is a measure of correlation in the noise and is an estimate of the Hurst exponent H.

Sit-to-Walk Analysis Using Inertial Sensors:

There are in total eight postural transition and gait events which can be easily identified from denoised STW component data from trunk, right and left shank IMUs'. They are (E1) initiation of STW, (E2) peak flexion angular velocity, (E3) seat-off event, (E4) peak extension angular velocity, (E5) swing toe-off, (E6) swing heel strike, (E7) stance toe-off and (E8) stance heel strike (Figure 21 and Figure 22). Previously based on similar postural transition events and gait events, STW phases have been defined and validated by Kerr et. al.(A. Kerr, Durward, & Kerr, 2004; Andy Kerr, Rafferty, Kerr, & Durward, 2007) and Buckley et.al.(T. Buckley, Pitsikoulis, Barthelemy, & Hass, 2009; T. A. Buckley, Pitsikoulis, & Hass, 2008). In absence of kinetic information from force plates and conflicts among previous studies (T. A. Buckley et al., 2008; A. Kerr et al., 2004) for gait initiation (GI) event detection, we have combined phase 2 (extension phase) & phase 3 (Unloading phase) of the four phases defined by Kerr et. al. (A. Kerr et al., 2004; Andy Kerr et al., 2007) and Buckley et. al (T. Buckley et al., 2009; T. A. Buckley et al., 2008).
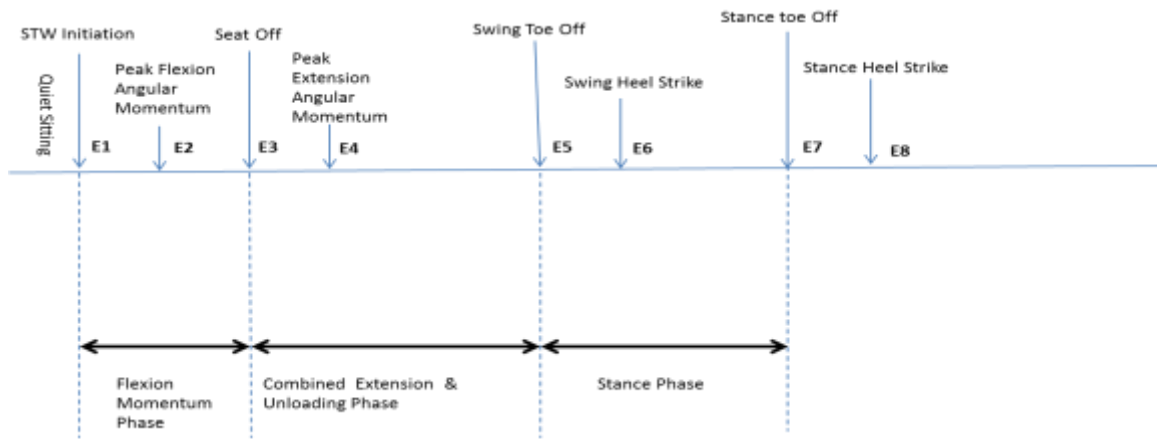
Figure 21. Three phases of STW 1) Flexion momentum phase 2) Combined extension and unloading phase 3) Stance phase

All previous studies on STW are laboratory based and required camera system and force plate in order to evaluate center of mass (COM) vertical velocity and gait initiation event. In order to develop a more robust detection algorithms, we have divided STW movement into three phases (Figure 21) as flexion momentum phase (phase 1), combined extension and unloading phase (phase 2), and stance phase (phase 3). The first phase of STW is flexion momentum phase (phase 1), which encompasses the beginning of the movement (E1) until seat-off (E3) (Figure 22).
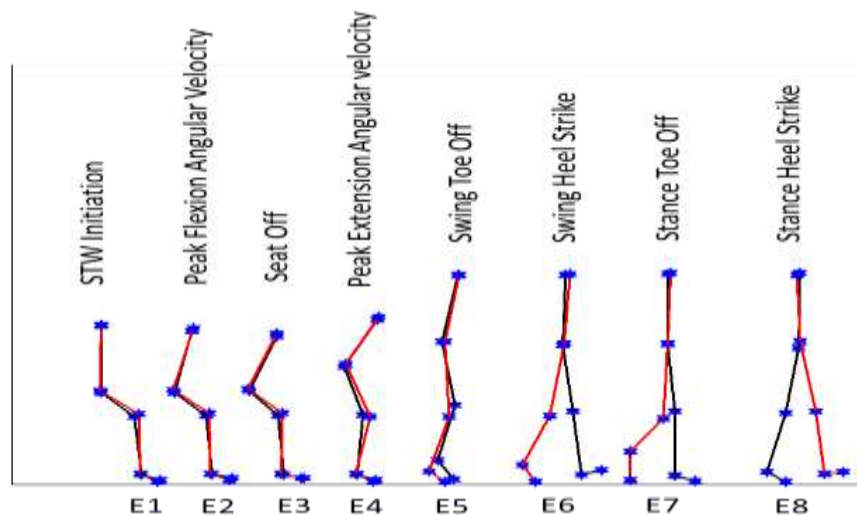
Figure 22. Eight postural transition and gait events detected using denoised IMU signals (E1-E8)

In this phase high flexion velocity is generated and is later followed by seat unloading. Initiation of STW event (E1) is defined by denoised signals of IMU situated at trunk to be as first local maxima before the peak flexion angular velocity (global minima) (E2) in denoised Gyro X signals (Figure 23a). Seat-off event (E3) is detected as maximum acceleration in denoised Acc Z signals when STW signals are truncated to half of their total length (neglecting return data of TUG test) (Figure 23b). Also, denoised signals from trunk Gyro-X (across medio-lateral axis) were used to acquire trunk peak flexion and peak extension angular velocities. Initial flexion angular acceleration of trunk is computed by fitting a line from STW initiation event (E1) to peak flexion angular velocity event (E2) and finding its slope. The remaining two phases of STW are the combined extension and unloading phase (phase 2) and stance phase (phase 3). Combined extension and unloading phase (phase 2) starts with development of extension velocity and is later followed by momentary stabilization, gait initiation adjustments and unloading. It encompasses from seat-off event (E3) until swing toe-off event (E5). Swing toe-off event (E5) is detected from swing foot IMU situated at shank (Figure 24a and 24b).
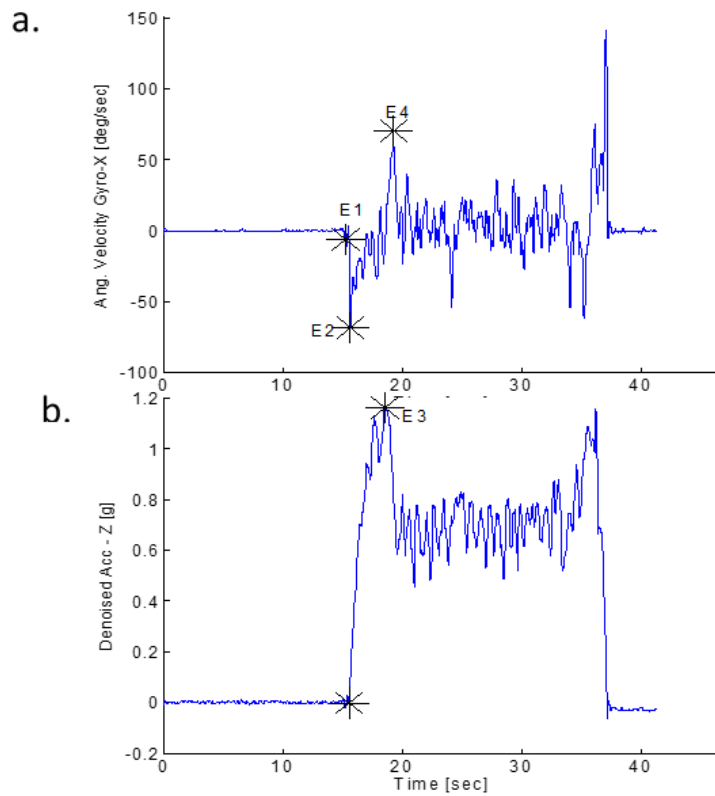
Figure 23. Event detection: (a) STW Initiation (E1), Peak flexion angular momentum (E2), Peak extension angular momentum (E4) (b) seat-off event (E3)
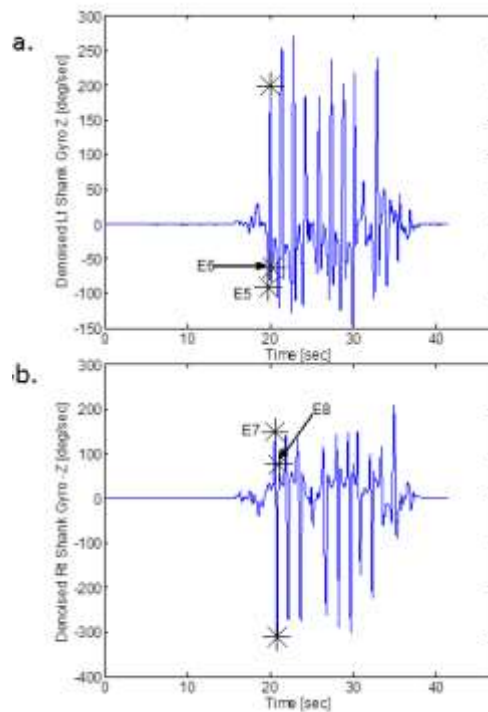
Figure 24. Event detection from swing (left) and stance (left) foot sensors. (a) swing foot and corresponding toe off event (E5) and heel strike event (E6). (b) stance foot and corresponding toe off event (E7) and heel strike event (E8)

The denoised Gyro Z signals (across mediolateral axis) is used and its first peak is maximum mid-swing angular velocity and the local minima to the left and right are swing toe-off (E5) and swing heel-strike (E6) events respectively. Similarly, stance toe-off (E7) and stance heel-strike (E8) can be computed.

Sit to Stand Analysis Using Inertial Sensors:

Gyroscope data from the axis perpendicular to sagittal plane can be used to determine trunk flexion and extension angular velocities. Accelerometer data from the axis perpendicular to the coronal plane determines anterio-posterior accelerations. In total, five events are identified, a) STS Initiation, b) Peak flexion angular velocity, c) Peak extension angular velocity and d) STS termination from gyroscope data (Figure 25b upper section) and seat-off (event f) is determined by accelerometer data (Figure 25b lower section). Also peak-to-peak (P2P) acceleration is determined as the absolute acceleration in STS, which is either from anterio-posterior (AP) acceleration initiation (event e) to seat-off (event f)(Figure 25b lower section) or from seat-off (event f) to AP acceleration termination (event g). P2P flex./extn. Angular velocity is absolute value from (event b) to (event c). Peak flexion angular velocity (event b) is the maxima, and peak extension angular velocity (event c) is the minima as shown in Figure 25b upper section. Initial flexion angular acceleration (slope from event a to event b), late extension angular acceleration (slope from event c to event d) can be computed by fitting the section in-between events of gyroscope signal by a straight line. Similarly, Pre seat-off flexion angular deceleration (slope from event b to the time at which seat-off event f occurs), post seat-off extension angular acceleration (slope from event c to the time at which seat-off event f had occurred) can be computed.
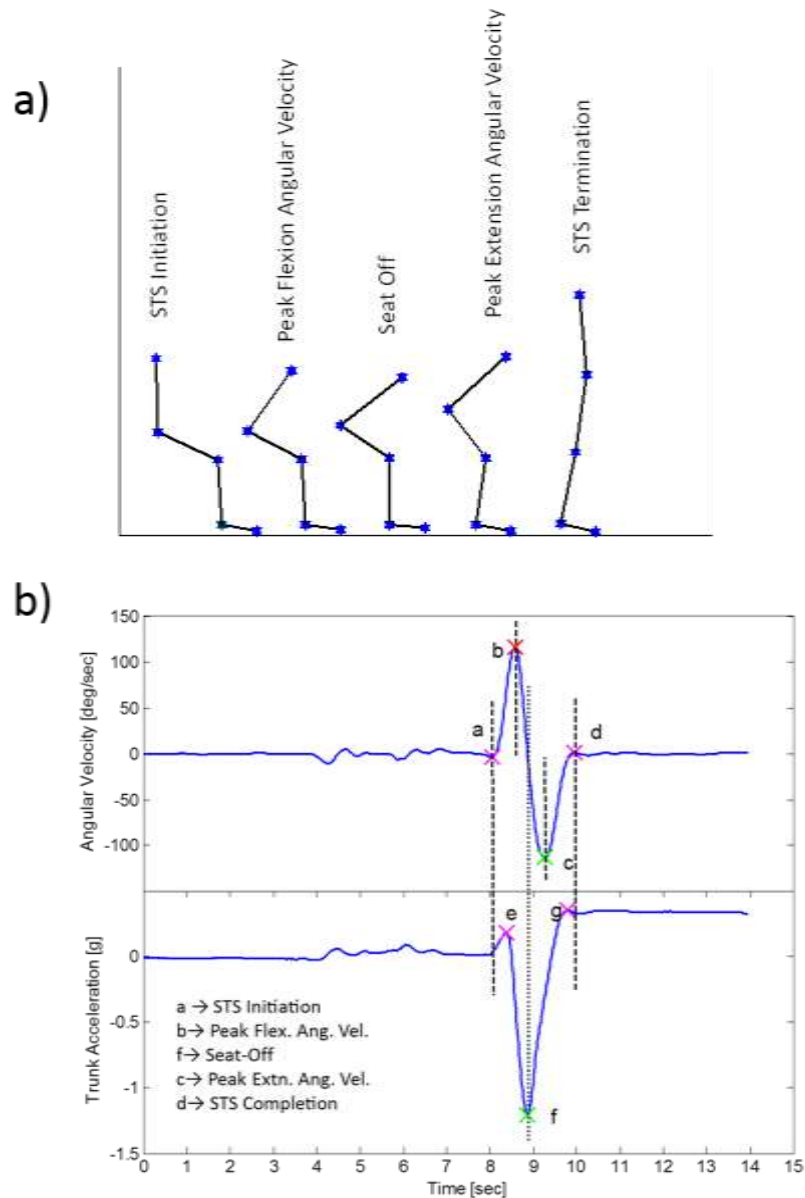
Figure 25. a) STS Events identified by IMU data. 13b) upper section of the figure is trunk angular velocity with events (a. STS Initiation, b. Peak flexion angular velocity, c. Peak extension angular velocity, d. STS termination) and lower section of 3b is trunk

Sit to Stand Analysis Using Forceplate:

STS standardization method proposed by Ethyre and Thomas (Etnyre & Thomas, 2007) was used for comparison in this laboratory and results are compared with those of wireless IMU. The seat-off event is defined as the instant where the rate of change in vertical force is maximum for the force plate below the chair. The six events

identified are 1) Initiation, 2) Counter, 3) Seat-Off, 4) Peak Vertical Force, 5)

Rebound, 6) Steady standing and are shown in figure 26 below:
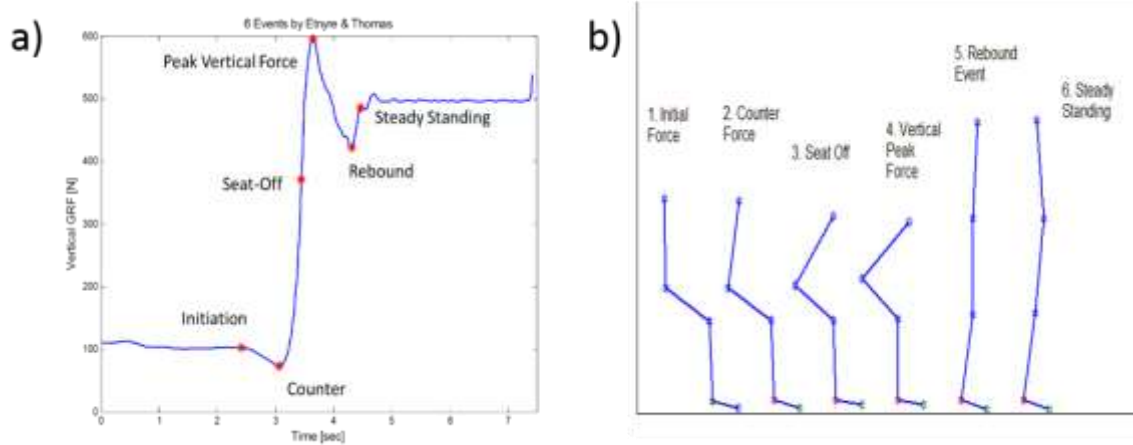
Figure 26. a) Identification of six events using vertical reaction force (Etnyre &
Thomas, 2007). b) Five reflective markers on right heel, right toe, right lateral
condyle, right trochanter and right acromio-clavicular positions to produce stick
figure

Results

The Short Physical Performance Battery scores could not detect any difference

between young and older individuals (Figure 27).
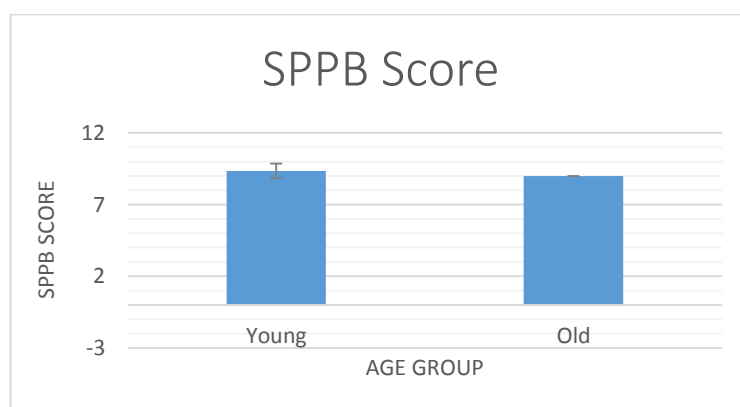
Figure 27. Short Physical Performance Battery (SPPB) for young and elderly
participants

TUG Times in young and older adults

One way ANOVA analysis was performed on TUG time and it was found that older adults had significantly higher TUG times than younger adults (p=0.01) (Figure 28).
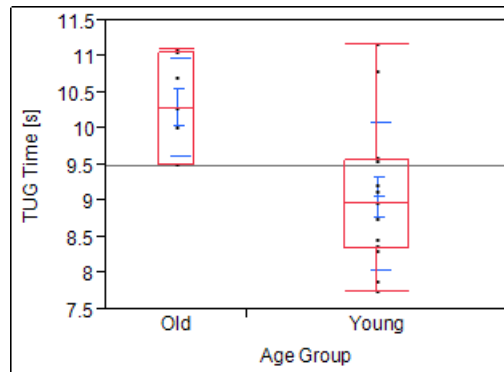
Figure 28. Timed Up and Go (TUG) time for young and older adults

Non-linear Measures of Postural Stability from Forceplate

Table 9. Approximate entropy and scaling exponent from COP signals

	Age Group															
	Old								Young							
	Visual Condition								Visual Condition							
	EC				EO				EC				EO			
	Direction				Direction				Direction				Direction			
	AP		ML		AP		ML		AP		ML		AP		ML	
	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev
Alpha	1.741	0.063	1.751	0.104	1.743	0.086	1.782	0.070	1.807	0.072	1.801	0.076	1.797	0.083	1.786	0.082
ApEn	0.051	0.018	0.104	0.070	0.045	0.021	0.153	0.116	0.036	0.014	0.103	0.042	0.033	0.012	0.162	0.092

From postural stability signals from forceplate, it was found that scaling exponent (alpha) in elderly people is significantly lower than that of younger counterpart (p=0.03) and approximate entropy in anterior-posterior direction was significantly higher in older adults (p<0.001) (Table 9). It was also found that variability (SD) of alpha was lower in AP direction in eyes closed condition than that in eyes open condition for both young and old adults. Whereas variability (SD) of alpha was lower in ML direction in eyes closed condition than that in eyes open condition for young adults but variability increased for old adults in eyes closed condition.

Variability analysis for 4 minute walk

Various nonlinear measures such as maximum Lyapunov exponent, approximate entropy and detrended fluctuation analysis (DFA) were evaluated for 3 minutes of normal walking and dual-task walking data. It was also found that dynamic stability of older adults was significantly higher than the younger counterparts (p=0.008) (Figure 29). But dual-tasking did not affect dynamic stability while walking. The ANOVA indicated no significant differences in scaling exponent alpha value and approximate entropy in mediolateral direction of walking but when paired t-test was conducted for normal walking and dual task walking, it was found that alpha values increased significantly in anterior posterior direction while walking (p=0.03) (Table 10).

Table 10 Non-linear measures of variability: max LE, scaling exponent (alpha) and approximate entropy in both anterior-posterior (AP) and medio-lateral direction (ML)

	Old				Young			
	DT		NW		DT		NW	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD
maxLE*	1.396	0.178	1.247	0.188	1.198	0.180	1.124	0.153
Alpha AP	0.290	0.135	0.261	0.095	0.327	0.121	0.266	0.094
Alpha ML	0.330	0.058	0.342	0.054	0.371	0.088	0.370	0.076
ApEn AP	0.790	0.071	0.776	0.097	0.809	0.069	0.808	0.066
ApEn ML	0.800	0.044	0.840	0.050	0.839	0.064	0.826	0.050
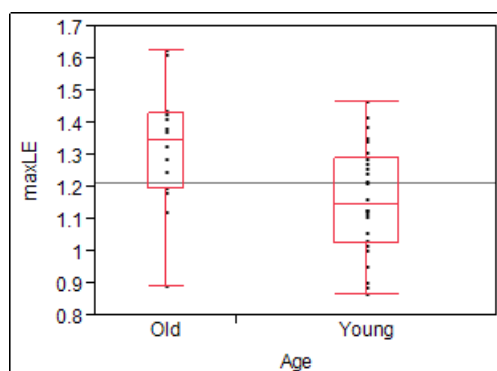
Figure 29. max LE for young and older individuals

Variability in sit to stand parameters:

Both young and older adults exhibited higher linear variability in counter force event for STSK (STS with knee support) and rebound event (Table 11 and 12).

Table 11 Vertical forces in STSA and STSK

	Age Group											
	Old						Young					
	STS Type						STS Type					
	STSA			STSK			STSA			STSK		
	Mean	Std Dev	CV	Mean	Std Dev	CV	Mean	Std Dev	CV	Mean	Std Dev	CV
Initial Force	0.249	0.074	29.497	0.368	0.095	25.835	0.213	0.043	19.961	0.282	0.070	24.780
Counter Force	0.139	0.048	34.778	0.168	0.098	58.358	0.136	0.052	38.296	0.122	0.083	68.239
Seat-Off Force	0.579	0.140	24.256	0.919	0.116	12.603	0.619	0.187	30.203	0.940	0.127	13.476
Vertical Peak Force	1.310	0.066	5.027	1.660	0.104	6.268	1.301	0.052	3.990	1.599	0.110	6.890
Rebound Force	1.056	0.055	5.221	1.005	0.128	12.769	1.009	0.087	8.628	1.015	0.076	7.535
Steady Standing Force	1.202	0.083	6.938	1.331	0.119	8.940	1.186	0.032	2.690	1.274	0.063	4.969
Initiation to Counter	0.246	0.037	15.082	0.273	0.128	46.769	0.248	0.061	24.521	0.293	0.055	18.784
Initiation to Seat-Off	0.411	0.051	12.323	0.431	0.064	14.763	0.479	0.092	19.268	0.437	0.037	8.398
Initiation to Peak Force	0.709	0.006	0.866	0.556	0.099	17.739	0.688	0.062	9.079	0.512	0.046	9.022
Initiation to Rebound	0.900	0.031	3.492	0.889	0.039	4.341	0.851	0.041	4.772	0.856	0.034	3.927
Initiation to Stand Still	1.000	0.000	0.000	1.000	0.000	0.000	1.000	0.000	0.000	1.000	0.000	0.000

Table 12 Various events and phases for STSA and STSK

| | STSA | | STSK | | |
	Mean	SD	Mean	SD	p-value
P2P AP Acceleration (@SeatOff) *	0.903	0.175	1.325	0.122	<0.001
P2P Flex./Ext. Ang. Vel. *	151.429	21.587	193.273	18.876	0.001
Peak Flex. Ang. Vel. *	70.443	13.566	106.378	13.741	0.0003
Peak Ext. Ang. Vel. *	75.714	13.500	86.400	13.767	0.009
Init. Flex. Ang. Acc. *	2.180	0.586	3.213	0.579	0.01
Late Ext. Ang. Acc.	2.129	0.731	2.214	0.535	0.32
Pre_SeatOff Flex. Ang. Dece. *	4.163	1.170	5.381	1.111	0.009
Post_SeatOff Extn. Ang. Acc. *	3.073	1.110	3.786	1.295	0.01
Trunk Forward Jerk *	0.020	0.007	0.044	0.009	<0.001
Trunk Backward Jerk *	0.017	0.004	0.020	0.006	0.008
t_a-b	0.654	0.141	0.667	0.132	0.44
t_b-f *	0.387	0.046	0.300	0.036	0.001
t_f-c *	0.359	0.099	0.473	0.114	0.002
t_c-d	0.819	0.196	0.839	0.108	0.34
t1_PeakFlexion	0.654	0.141	0.667	0.132	0.44
t2_SeatOff	1.042	0.123	0.968	0.148	0.79
t3_PeakExtension	1.402	0.207	1.441	0.213	0.35
t4_CompleteSTS	2.221	0.383	2.281	0.263	0.34
t5_Initiation_to_SeatOff	1.042	0.123	0.968	0.148	0.79
t6_SeatOff_to_STS_Termination *	1.178	0.283	1.312	0.196	0.04
t7_Time in between Peak Flex./Ext. Momentums	0.747	0.110	0.773	0.143	0.19

It was seen that peak-to-peak acceleration at seat-off were significantly higher for STSK (p<0.001) (Table 12) and also younger adults were found to have exerted higher

accelerations at seat-off (p=0.005) (Table 13). The variability in sit-to-walk parameters is reported below in Table 14.

Table 13 Various events and phases of STS in young and older adults

	Young		Old		p-value
	Mean	SD	Mean	SD	
P2P AP Acceleration (@SeatOff) *	1.141	0.222	0.932	0.141	0.005
P2P Flex./Ext. Ang. Vel. *	190.598	49.792	147.185	24.896	0.006
Peak Flex. Ang. Vel. *	105.079	35.425	78.235	15.462	0.012
Peak Ext. Ang. Vel. *	85.862	25.791	70.280	11.616	0.033
Init. Flex. Ang. Acc.	3.301	0.910	2.784	0.932	0.092
Late Ext. Ang. Acc.	2.621	1.207	2.142	0.845	0.133
Pre_SeatOff Flex. Ang. Dece. *	5.323	1.767	3.952	1.067	0.014
Post_SeatOff Extn. Ang. Acc.	4.278	1.561	3.968	1.142	0.290
Trunk Forward Jerk *	0.039	0.008	0.032	0.008	0.049
Trunk Backward Jerk	0.019	0.005	0.018	0.004	0.204
t_a-b	0.622	0.115	0.584	0.140	0.241
t_b-f	0.297	0.051	0.299	0.060	0.542
t_f-c	0.421	0.131	0.363	0.073	0.094
t_c-d	0.793	0.206	0.765	0.237	0.382
t1_PeakFlexion	0.622	0.115	0.584	0.140	0.241
t2_SeatOff	0.919	0.123	0.883	0.176	0.289
t3_PeakExtension	1.340	0.223	1.246	0.228	0.161
t4_CompleteSTS	2.133	0.325	2.012	0.425	0.224
t5_Initiation_to_SeatOff	0.919	0.123	0.883	0.176	0.289
t6_SeatOff_to_STS_Termination	1.214	0.244	1.128	0.258	0.207
t7_Time inbetween Peak Flex./Ext. Momentums	0.718	0.116	0.662	0.095	0.106

Variability in sit to walk parameters

Effects of dual task on forward and lateral walking on treadmill:

Using paired t-test for forward walking condition, it was found that dual task affected forward walking and approximate entropies while walking forward on treadmill were significantly higher than that when simply walking on treadmill (p<0.001) for both young and older adults (Table 15). No significant differences were found for scaling exponent alpha or in dynamic stability due to dual tasking in both young and older participants and in both forward and lateral directions. However, when compared to younger individuals the scaling exponent alpha value was significantly higher among

older adults (Figure 30) in all kinds of walking condition (normal/dual-task) and

directions (forward/lateral) (p=0.02).

Table 14 STW parameters using inertial sensors

	Old			Young		
	Mean	**SD**	**CV**	**Mean**	**SD**	**CV**
P2P AP Acceleration(@SeatOff) [g]	0.14	0.06	42.92	0.16	0.11	69.93
P2P Flex./Ext. Ang. Vel. [deg/sec]	188.49	29.13	15.45	184.43	47.75	25.89
Peak Flex. Ang. Vel. [deg/sec]	56.47	37.46	66.33	77.52	34.34	44.30
Peak Ext. Ang. Vel. [deg/sec]	132.02	12.91	9.78	106.91	14.77	13.81
Init. Flex. Ang. Acc [deg/sec^2]	1.56	0.29	18.93	1.62	1.05	64.77
Pre_SeatOff Flex. Ang. Dece. [deg/sec^2]	1.90	1.17	61.86	2.08	1.01	48.62
Post_SeatOff Extn. Ang. Acc. [deg/sec^2]	1.70	0.44	26.38	2.86	3.64	127.26
Trunk Forward Jerk [g/sec]	0.01	0.002	15.32	0.01	0.009	50.16
Trunk_Transition_Dece_Acc Rate [deg/sec^2]	1.74	0.14	8.009	2.36	1.82	77.42
t1_PeakFlexion [sec]	0.71	0.06	8.85	0.67	0.21	32.21
t2_SeatOff [sec]	1.03	0.11	11.34	0.99	0.24	25.04
t3_PeakExtension [sec]	1.45	0.07	5.09	1.51	0.49	32.42
t4_Swing_TO [sec]	2.06	0.95	46.49	2.17	1.15	52.97
t5_Swing_HS [sec]	2.48	1.00	40.21	2.85	1.19	41.94
t6_Stance_TO [sec]	2.59	1.07	41.35	2.70	1.22	45.05
t7_Stance_HS [sec]	3.09	1.09	35.27	3.20	1.19	37.20
t8_for_Initital_Gait_Cycle [sec]	1.03	0.16	16.19	1.03	0.04	4.24
t9_for_TotalSingleStance [sec]	0.91	0.07	8.03	1.17	0.33	28.43
t10_for_GI_PartialDoubleSupport [sec]	0.11	0.09	83.23	0.14	0.34	242.24
t11_Time inbetween Peak Flex./Ext. Momentums [sec]	0.74	0.076	10.188	0.83	0.42	51.01
t12_PeakFlexion to Seat-Off [sec]	0.42	0.072	16.972	0.52	0.29	57.10
t13_Seat-Off to Peak Extension [sec]	0.32	0.145	44.763	0.31	0.13	43.14
t14_Time_to_STW_Completion [sec]	3.09	1.091	35.270	3.20	1.19	37.20
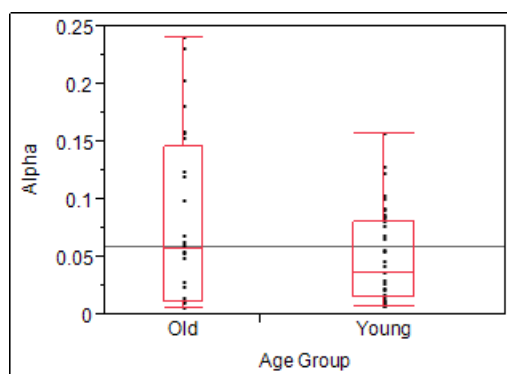
Figure 30. Scaling exponent alpha value for older and young participants

Table 15 Nonlinear measures for normal and dual-task walk on treadmill in forward and lateral direction where FNW=Forward normal walk; FDTW=Forward dual-task walk; LNW=lateral Normal walk; LDTW=lateral dual task walk

	Age Group															
	Old								Young							
	Walk Type								Walk Type							
	FDTW		FNW		LDTW		LNW		FDTW		FNW		LDTW		LNW	
	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev	Mean	Std Dev
SD	0.109	0.031	0.118	0.034	0.052	0.020	0.045	0.008	0.118	0.034	0.129	0.042	0.068	0.021	0.068	0.019
Alpha	0.022	0.019	0.026	0.027	0.129	0.076	0.140	0.057	0.018	0.009	0.017	0.014	0.072	0.023	0.084	0.039
ApEn	1.424	0.277	1.424	0.193	1.651	0.075	1.705	0.025	1.385	0.146	1.344	0.141	1.650	0.111	1.640	0.099
maxLE	1.597	0.052	1.613	0.088	1.593	0.050	1.593	0.024	1.706	0.099	1.709	0.082	1.653	0.056	1.668	0.074

Smartphone based gait and posture assessments in laboratory

Inertial sensor inside smartphone was used to assess postural parameters in young and elderly participants. It was found that mean radius of Center of Pressure (COP) trajectories (p=0.005), circular area of COP (p=0.01), elliptical area (p<0.001) were significantly higher in older adults than to that in younger counterparts (Table 16). The mean power frequency in medio-lateral direction was significantly higher in eyes closed condition than to that in eyes open condition for both age groups. Significantly higher standard deviation in COP along anterior-posterior direction (p<0.001) were found in older adults.

Table 16 Smartphone derived postural sway linear parameters

		Old		Young	
		Eyes Closed	**Eyes Open**	**Eyes Closed**	**Eyes Open**
Mean Radius [mm]	Mean	7.870	8.223	6.125	5.270
	SD	3.099	3.062	1.756	2.198
Circular Area [mm²]	Mean	223.321	258.708	134.393	104.427
	SD	161.583	238.202	86.231	109.773
Elliptical Area [mm²]	Mean	1377.811	992.419	604.724	479.908
	SD	914.914	338.020	448.749	305.359
AP MPF [Hz]	Mean	0.709	0.701	0.696	0.681
	SD	0.088	0.116	0.081	0.086
ML MPF [Hz]	Mean	0.774	0.620	0.663	0.647
	SD	0.123	0.076	0.087	0.074

There was strong correlation between 5m and 10 m gait speed when compared with that of stop watch. At 5m walk distance the correlation $R^2=0.70$ and for 10m walk speed, $R^2=0.60$.
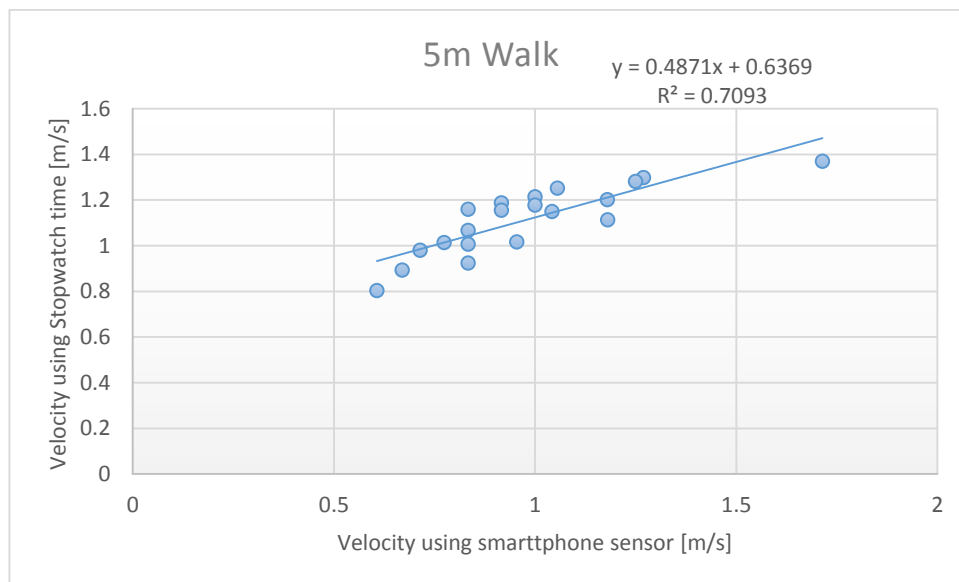
Figure 31. Relationship between velocities from the two system a) stopwatch timing and b) smartphone sensor for 5m long walk
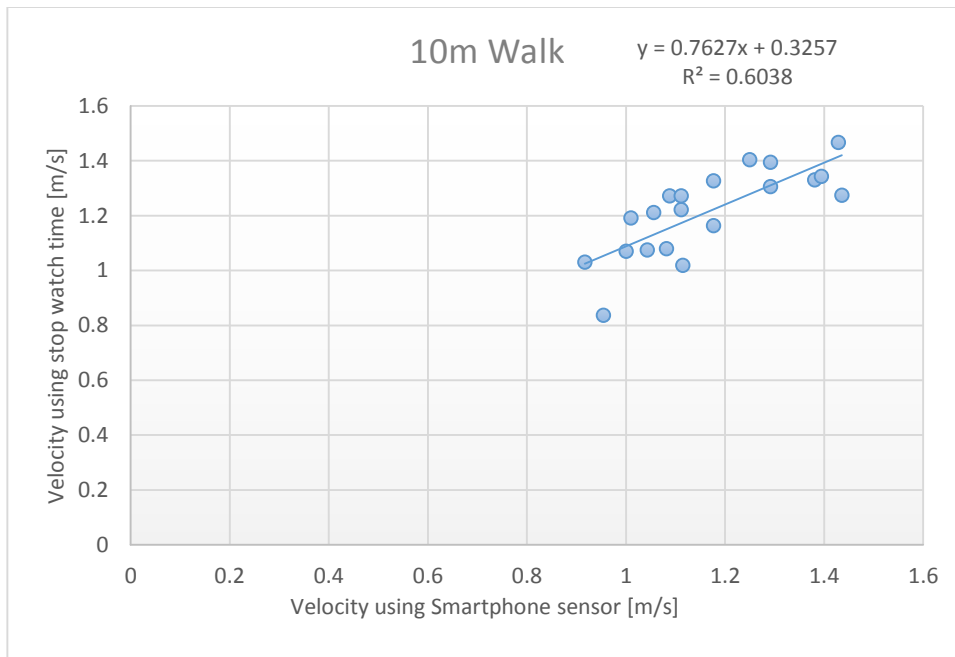
Figure 32. Relationship between velocities from the two system a) stopwatch timing and b) smartphone sensor for 10m long walk

Discussion

The findings support the use of inertial sensors as a tool for understanding variability in healthy young and older adults and augments preexisting knowledge of variability structure in elderly and their magnitudes. This study provides greater insights into postural transitions while performing functional daily living activities (like STS and STW) and furthers investigation into the changes in structure of variability while walking on a treadmill with four conditions (forward, lateral and with/without dual-tasking).

TUG times for older adults were significantly greater than younger counterparts (Figure 28). Nonlinear center of pressure (COP) information revealed the scaling exponent, alpha, was >0.5 for both the young and elderly; signifying that both populations elicited persistent COP signals. This was expected as both young and older group participants are sampled from healthy populations. However, the older participants were found to be less persistent with significantly lower values, indicating that older adults utilize more robust control mechanisms. Regarding the regularity of COP fluctuations, the results reveal greater approximate entropy values in the anterior-posterior direction for the elderly, in accordance with previous studies(Borg & Laxaback, 2010). Similarly, dynamic stability readily differentiated younger and older adults. Older individuals had significantly higher maxLE exponents than younger adults (Figure 29). During dual task forward walking both younger and older adults were more anti-persistent than normal walking (Table 10).

It was apparent that linear variability existed in movements such as sit-to-stand, and participants had higher peak-to-peak acceleration at seat-off was significantly higher for STSK (Table 12). Younger participants were found to have exerted higher accelerations at seat-off than older study participants (Table 13).

Treadmill walking revealed that the dual-task condition affected forward walking and approximate entropies while walking forward on treadmill (significantly higher than that when simply walking on treadmill) for both young and older adults (Table 15). These results conform to the "cautious gait" concept suggested by Maki (Maki, 1997), implying that both older and young adults utilize higher complex neuromuscular control while walking in dual task scenario. Along with this, it was also found that older individuals had higher scaling exponent alpha values (or fractal dimension) (Figure 30) in all kinds of walking conditions (normal/dual-task) and directions (forward/lateral), which implies less anti-persistence or less neuronal-control involved in the movement. This is suggestive of higher persistence in older adults while walking on treadmill.

Accordingly, COP measures from inertial sensors successfully differentiated physiological differences among disparate groups. Mean radius of COP trajectories, circular area of COP, and elliptical area were significantly greater in older adults than to their younger counterparts. In addition, the standard deviation of the COP in the anterior posterior direction was significantly higher in the elderly. Interestingly, the mean power frequency in the medio-lateral direction was significantly higher in eyes-closed condition than the eyes open condition for both age groups. Thus, COP measures from inertial sensors may provide body sway information similar to that of the forceplates – the "gold standard" for extracting COP information. As a result, the present study reveals that IMU's exhibit a robust capacity to measure physiological information outside laboratory environments.

Gait speed was computed using smartphone inertial sensors for a 5m walk (R^2=0.71) and for a 10m walk (R^2=0.60), and was found to be highly correlated with the velocities using a standard stopwatch (Figure 31 and Figure 32). These measures are further validated in clinical environments in the next chapter. The measures

derived in this study serve as ground work for future research and will provide an understanding of movement variability to reduce falls in frail older individuals, and in designing effective interventions to reduce fall risk and establish a link between fall risk and variability in human movement.

Conclusion

This study has contributed in measuring linear and nonlinear variability using portable inertial sensors. This study is an important ground work for launching inertial sensors in clinical setting for measuring patients with pathologies. For healthy young and old adults various linear and non-linear variability parameters were determined using inertial sensors which showed promising results.

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Chapter 5: Movement Variability is a risk indicator in assessment of falls and adverse post-operative outcomes in cardiovascular disease patients

Abstract

Prevalence of falls among older adults due to cardiovascular disorders remains largely unknown. About 20% of patients with unexplained falls are reported to have neurally mediated cardiovascular disorders and hypotensive syndromes. As the number of falls in elderly population continues to rise and an analogous increase in healthcare utilization, the assessment and prevention of falls in frail elderly cardiovascular disease (CVD) patients remains a high priority for the health care professionals. Patients with intrinsic cardiac cause for falling have been found to have higher mortality. In elderly CVD patients, gait speed is conjectured by Society of Thoracic Surgeons (STS) as an independent predictor of post-operative morbidity and mortality. However, this guideline by STS has not been studied adequately with large sample size; rather it is based largely based on expert opinions. Although it is well recognized that gait speed is not a fall risk factor, gait speed is a quick robust measure to classify frail/non-frail CVD patients and undoubtedly frail individuals are more prone to falls. Thus, it is important to examine the effects of movement variability in assessing of fall risk and identifying patients likely to have adverse post-operative outcome. One approach understanding these relationship is to analyze gait and postural predictor variables that describe the fall risk and associated post-operative adverse outcomes. Postural assessments are promising for patients who are unable to walk for 5m distance. Accordingly, inertial sensors are indispensable for the assessment of elderly patients in clinical environments and may be necessary for objective assessment. Sixteen elderly CVD patients (Age 76.1±3.6 years) who were scheduled for cardiac surgery the next day were recruited for this study. Based on

STS recommendation guideline, eight of the CVD patients were classified as frail (prone to adverse outcomes with gait speed ≤0.833 m/s) and other eight patients as non-frail (gait speed > 0.833 m/s). All data collection took place in Carilion Roanoke Memorial Hospital (CRMH), Department of Cardiac Surgery in presence of skilled clinical research staff. This study identifies various postural and movement variability parameters that could offer insights into associated fall risk as well as post-operative adverse outcomes of CVD patients. The smartphone based measurements could serve as clinical decision support system for assessing patients quickly in perioperative period.

Introduction

Falls in elderly patients is multifactorial (Fleming & Pendergast, 1993), attributed to complex interaction of intrinsic and extrinsic risk factors superimposed on normal aging process (Campbell, Borrie, & Spears, 1989; Nevitt, Cummings, Kidd, & Black, 1989; Tinetti, Speechley, & Ginter, 1988; Tinetti, Williams, & Mayewski, 1986). Patients with intrinsic cardiac cause for falling have been found to have higher mortality rate than those with non-cardiovascular or unknown causes of falls (Kapoor, Karpf, Wieand, Peterson, & Levey, 1983). Falls in cardiovascular disease (CVD) patients is reported to be caused by underlying cardiovascular disorders or because of amnesia or loss of consciousness during unwitnessed syncope. Indeed, there is significant overlap between falls and syncope in older CVD patients. CVD patients with syncope also experience unexplained and recurrent falls (Dey, Stout, & Kenny, 1997).

It remains unclear which factors are responsible for high fall risk in CVD patients, but some experts speculate that certain environments, medications, age-related changes, and diseases make a particular genotype of people vulnerable to

frailty in CVD patients (Morley, Perry, & Miller, 2002; J. F. Wilson, 2004). This frailty phenotype is independently predictive of falls (Fried et al., 2001). Some researchers have also linked functional limitation (Arozullah, Daley, Henderson, & Khuri, 2000; Arozullah, Khuri, Henderson, Daley, & Participants in the National Veterans Affairs Surgical Quality Improvement, 2001; Fukuse, Satoda, Hijiya, & Fujinaga, 2005; Polanczyk et al., 2001), poor nutritional status (Arozullah et al., 2000; Arozullah et al., 2001; Maurer, Luchsinger, Wellner, Kukuy, & Edwards, 2002), cognitive impairment (Arozullah et al., 2001; Galanakis, Bickel, Gradinger, Von Gumppenberg, & Forstl, 2001; Marcantonio et al., 1994) , depression (Berggren et al., 1987; Galanakis et al., 2001) and loneliness (Herlitz et al., 1998) with cardiovascular disorders and frailty.

Analysis of gait and postural predictor variables that describe the underlying neuromuscular function are indispensable for the diagnosis and treatment of elderly patients and may be necessary for objectively assessing fall risk in CVD patients. As the falls in elderly population continues to rise and an analogous increase in healthcare utilization, the assessment and prevention of falls in CVD patients also remains a high priority for the health care professionals. Bereft of multisystem reserves, the elderly CVD patients (particularly who are frail) are increasingly vulnerable to an array of adverse health outcomes, including sarcopenia, hospitalization, negative energy balance, exhaustion, falls, and loss of independence (Newman et al., 2001) and mortality. It is seen that 1 out of 10 falls result in serious injury and 1 out of 5 in a fracture in elderly (Tinetti, 2003). Falls are responsible for about 66% of deaths resulting from unintentional injuries and where unintentional injuries being the fifth leading cause of death in the elderly (Tan & Kenny, 2006). One of the major consequences of falls are hip fractures which result in 25% reduction in life expectancy and institutionalization rates of 8-34% for patients from

communities (Braithwaite, Col, & Wong, 2003). It is evident that the elderly with cardiovascular disorders along with history of falling have a two-third chance of falling over the next year (Nevitt et al., 1989). Recurrent falls have been cited as common reason for admission to long-term care institutions.

Frailty and Operative Risk

Older adults have heterogeneity of health status and this leads to increased risk of post-operative complications (Makary et al., 2010; Polanczyk et al., 2001), and thus surgical decision-making is challenging for clinicians. Preoperative risk assessment is essential but there is paucity of tools for predicting operative risk. Physiologic reserve in an older adult can determine his/her resilience to recover from an operation. However there is no standardized method of measuring physiologic reserve in older surgical patients (Makary et al., 2010). Frailty is a marker of decreased physiologic reserves and resistance to stressors (Boyd, Darer, et al., 2005; Fried et al., 1998; Fried et al., 2001; Woods et al., 2005) and predicts operative risk in older surgical patients (Makary et al., 2010). In clinical operative setting, clinicians have tried to link postoperative adverse outcomes with various components of frailty (Dasgupta, Rolfson, Stolee, Borrie, & Speechley, 2009).

Researchers have also reported that age remains an independent risk factor even after controlling for co-morbid illnesses and functional impairment for postoperative complications (Alibhai et al., 2005; Arozullah et al., 2000; Arozullah et al., 2001; Dasgupta et al., 2009; Marcantonio et al., 1994; Polanczyk et al., 2001). Chronological age of a patient does not reflect his/her biological age, and elderly patients have range of biological status that varies from robust to frail (Mitnitski, Graham, Mogilner, & Rockwood, 2002; J. F. Wilson, 2004).

Objective

The objective of this study is to utilize the information acquired in laboratory studies to assess movement variability using inertial sensors, and apply it to clinical setting situated in Roanoke for quick assessment of fall risk and post-operative adverse outcomes in cardiovascular patients. It was hypothesized that IMU can help identify subset of high fall risk. Gait speed, postural transitions, and postural stability measures have potential to identify frail patients with adverse post-operative health outcome.

Materials and Method

Sixteen CVD patients have been included in this study (Table 17). Patients were included in the study only if they: i) consented to participate and ii) were going to be operated next day for cardiovascular disorder. The patients were categorized into frail (F) group (walking velocity ≤0.833 m/s) and non-frail (NF) group (walking velocity>0.833 m/s). The sample population had five females (ID17, ID18, and ID20-22) and 11 males.

Inclusion Criteria for patients: Patients scheduled for any cardiac surgery, who spoke English, was at least 70 years of age, who was able to ambulate (not on bed rest, unstable, etc.) with either the need for cane or walker, was cognitively able to follow instructions and available for testing within 7 days of surgery was included in this study.

Exclusion Criteria: Patients who were unable to ambulate, had severe neuropsychiatric conditions, or their surgery is cancelled were excluded from the study.

Instrumentation:

An IMU (smartphone) was affixed at pelvic region using a smartphone holster and clip. The app had a start and stop button and voice commands were input into the app which instructed patients to get ready and perform activities in particular sequence (Table 18). The data was truncated using the temporal information of voice commands to the patient (Figure 34).

Table 17 Means and standard deviations of patients' anthropometric and age information

	Frail		Non Frail		All CVD Patients	
	Mean	SD	Mean	SD	Mean	SD
Age (years)	76.38	4.03	76.00	3.55	76.18	3.67
Height (cm)	172.34	12.92	171.30	6.43	171.81	9.87
Weight (kg)	87.41	20.32	77.41	15.89	82.40	18.36
BMI (kg/m^2)	29.69	7.46	26.26	4.73	27.97	6.28

And a portable forceplate (Bertec Corporation, FP4060-05-PT) was used to measure postural stability information. IMU signals were sampled at 50Hz using the customized smartphone app "Lockhart Monitor" and, further data processing was accomplished using custom-made Matlab (MATLAB version 6.5.1, 2003, computer software, The MathWorks Inc., Natick, Massachusetts) programs.

Table 18 Voice commands used in app for data collection in clinical environment

Time Interval [sec]	Sound & Voice Commands	Total Elapsed Time [sec]
0	Beep	0
10	Please Keep Sitting	10
10	Ready to Stand	20
10	Please stand	30
10	Very Good! Please Stand-Still	40
65	Relax ! Please Sit	105
60	Please Keep Sitting !	165
10	Ready to Stand	175
10	Very Good! Please Stand Still	185
65	Relax !	250

Experimental Procedure

Patients who had been scheduled for cardiac surgery and present to the Cardiac Surgery Pre-Surgical Testing (PST) area on 6 West at CRMH were screened by the PST nurse to determine whether all inclusion/exclusion criteria were met. If the patient was appropriate for inclusion in the study, the PST nurse had asked the patient if he/she were interested in talking with a study team member about the study. If the patient was interested, a consenter (Registered Nurse Specialist) would discuss the study with the patient, answer all questions, and obtain written consent.

Patients who meet all inclusion criteria and had consented were requested to wear a waist belt and smartphone (inside holster) was clipped to the waist belt. All the experiments were conducted in a well-lit room with an unobstructed walking area with clear floor markings at 0 and 5 meters. The patients were asked to rise from the chair to a standing position and follow instruction as per the voice commands of the app (Table 18). They were allowed to use a walking aid (cane, walker) if they needed. A standard digital stopwatch was used; the stopwatch was started with the first footfall after the 0-meter line and stopped with the first footfall after the 5-meter line. The walk was repeated 3 times, with sufficient time for subject recuperation between trials. Each 5-meter walk time (in seconds) was recorded on the data collection form (Figure 33). The average speed for the 3 trials was calculated and was also recorded on the data collection form. The participant's postural transition, static stability was measured using the forceplate and IMUs, whereas, the walking speed was measured using IMU and also recorded using stopwatch time.

Data Analyses:

The three trials of 5m gait data were acquired in a single file created by smartphone and postural standing and sit-to-stand transition were collected in another data file.

The data was resampled to 50Hz, using the time stamp. The continuously collected data was then truncated at intervals and the truncated signal was used for further analysis (Figure 34). The signals were filtered using low-pass Butterworth filter with zero lag at a cut-off frequency of 6Hz.
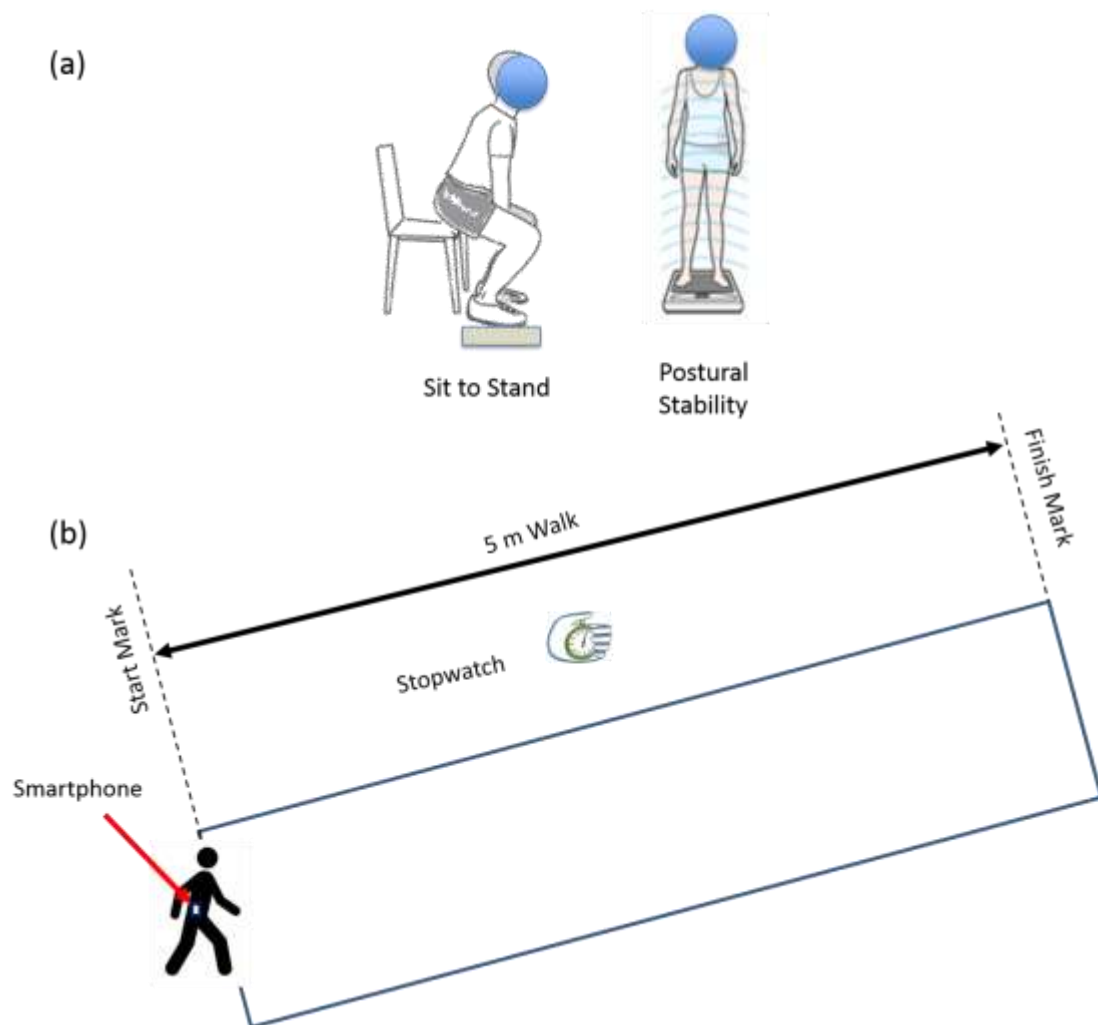
Figure 33. All patients (a) stand still for 60 seconds and perform sit-to-stand transitions, (b) walk a distance of 5 m

Dependent variables

To quantify the postural transition several parameters were derived from forceplate placed below the feet of patients while standing.

Body Jerk [Newton/sec] was defined as the rate of change of force during the transition. It was calculated as slope of line connecting highest force to the lowest vertical force for sit-to-stand and stand-to-sit transitions (Figure 35).

$$BJ = \frac{dF_V}{dt}$$

31

Where Fv is vertical force and Δt is change in time (transition time).
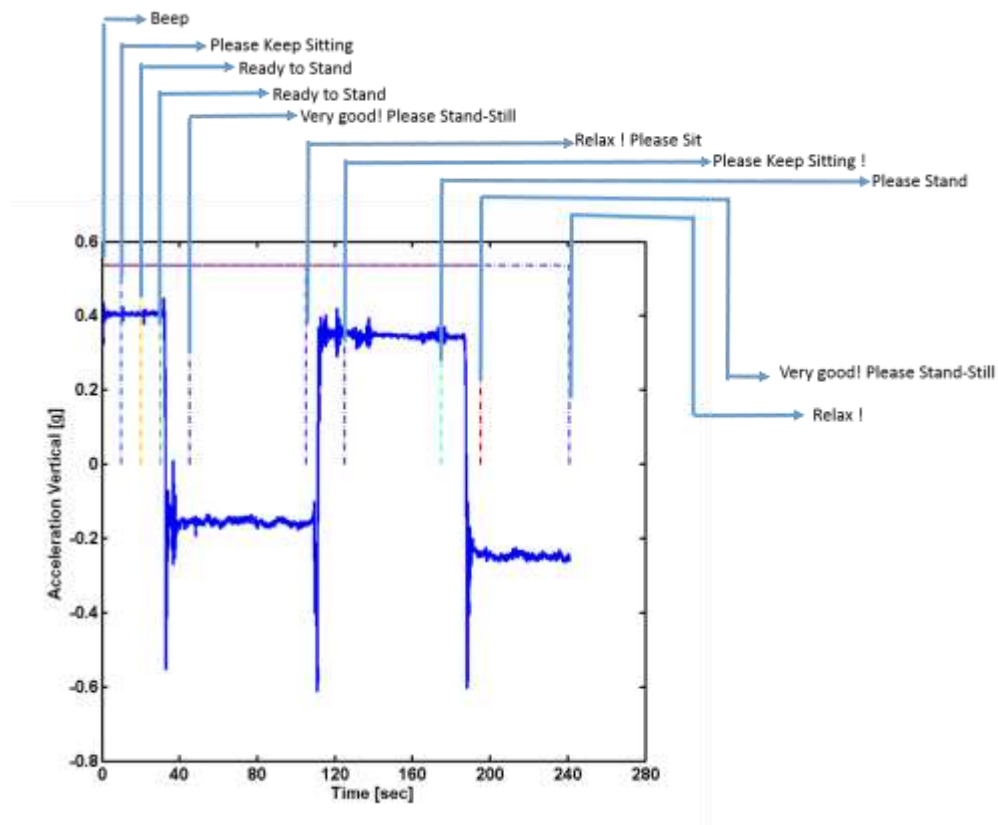
Figure 34. Truncation of smartphone IMU signals using temporal information of voice commands in app

123
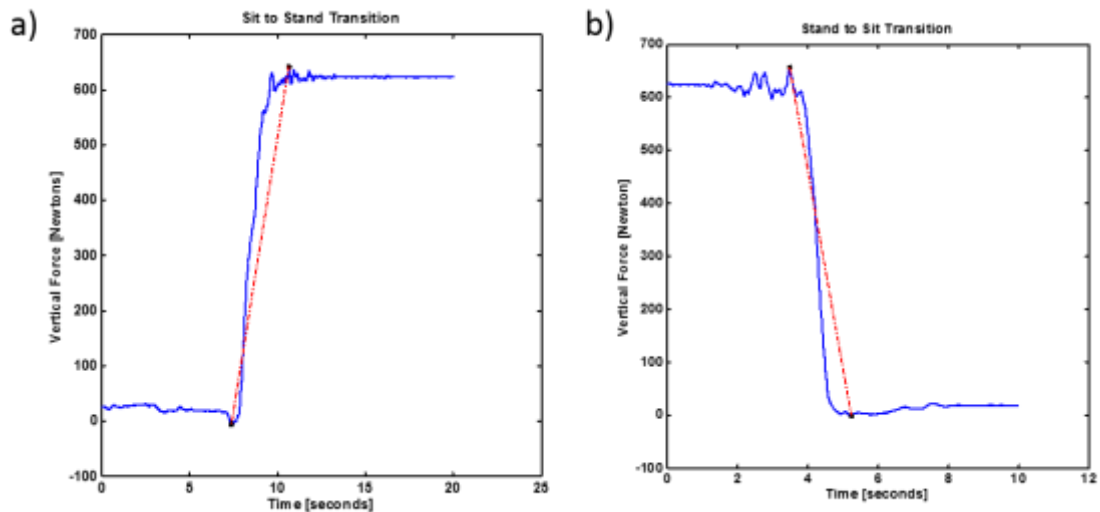
Figure 35. (a) sit-to-stand vertical force and body jerk (b) stand-to-sit vertical force and associated body jerk.

Jerk [m/s³]: Jerk was computed using accelerometer signals from steady state to maximum achieved acceleration during sit-to-stand or stand-to-sit movements. Figure 36 shows a typical signal of sit-to-stand from a smartphone.

Center of Pressure (COP) radius [mm]: It was calculated resultant of mean of COP AP and COP ML trajectories.

RMS COP [mm]: Root Mean Square value of COP trajectory in a particular direction (AP or ML).

Circular area of COP [mm²]: It was computed using mean radius of center of pressure.
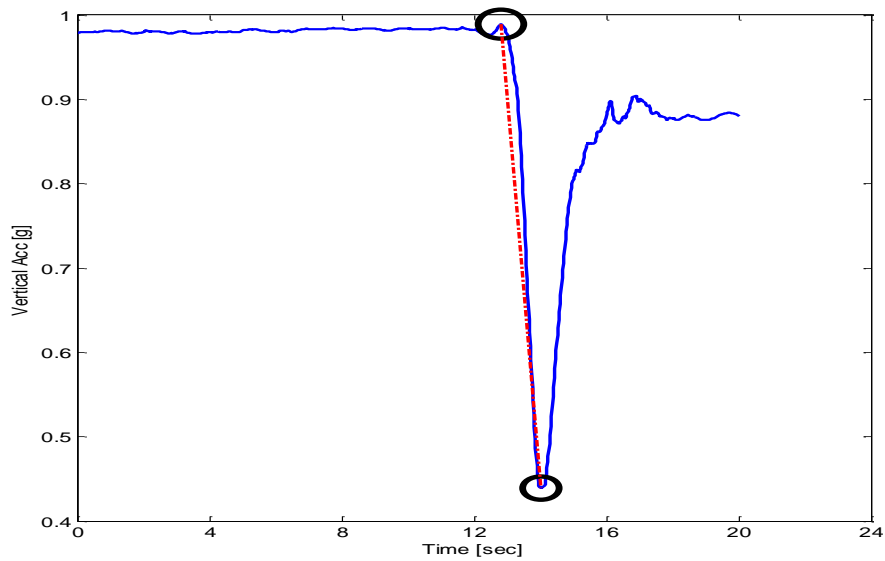
Figure 36. Peak flexion acceleration during sit-to-stand and stand-to-sit movements

Gait speed [m/s]: Gait speed was computed using stopwatch as well as IMU for a 5m long walk. Acceleration signals from three directions were used to compute resultant acceleration. The resultant acceleration signals were filtered using 4th order dual low-pass butterworth filter with cut-off frequency as 6 Hz. One half second moving window variance was computed and threshold was set using initial stand-still data (Figure 37). Once start and stop time are detected, average velocity is computed over 5 m distance walk.
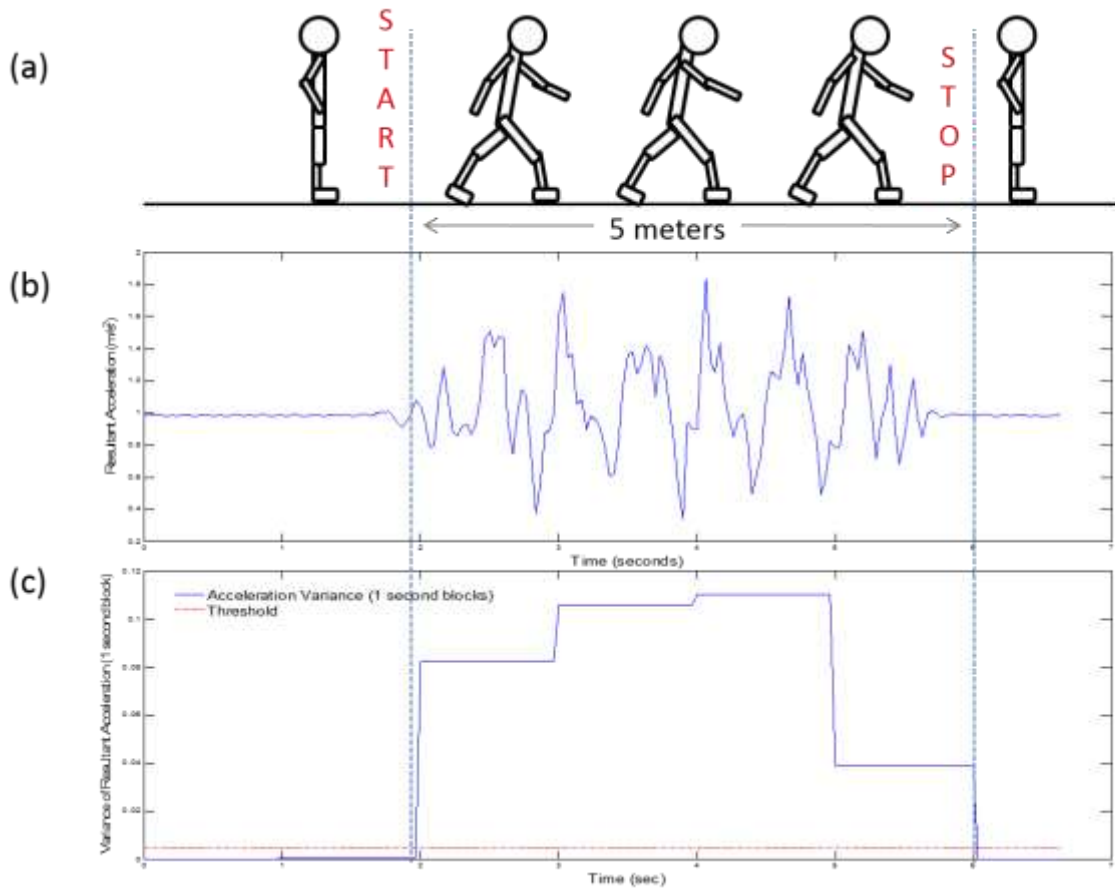
Figure 37. (a) the test starts from still-standing followed by 5m walk and stops at still-standing as well (b) resultant acceleration signals (in g-units) (c) moving window (0.5 sec) variance of low-pass filtered resultant acceleration.

Root Mean Square (RMS):- RMS is a measure of dispersion of the data relative to zero, whereas standard deviation is a measure of dispersion relative to mean. This value is an indication of average magnitude of accelerations in each direction during a complete walking trial. Six parameters were calculated using accelerations from three directions:

$$RMST = \sqrt{RMS\_AP^2 + RMS\_V^2 + RMS\_ML^2} \qquad 32$$

$$RMSR_{AP} = RMS_{AP}/RMST \qquad 33$$

$$RMSR_{ML} = \frac{RMS_{ML}}{RMST} \qquad\qquad 34$$

$$RMSR_V = \frac{RMS_V}{RMST} \qquad\qquad 35$$

Where RMS_AP, RMS_ML, and RMS_V represent all three directions. RMS is statistical measure of magnitude of acceleration in each direction. RMSR represents the ratio between RMS in each direction and the RMS vector magnitude (RMST). RMSR is apparently the RMS normalized by the RMST (Sekine et al., 2013).

Harmonic ratio: The harmonic ratio was described by Gage (Gage, 1965) and Smidt (Smidt, Arora, & Johnston, 1971), to provide an indication of smoothness and rhythm of acceleration patterns. The harmonic ratio proposed by Gage is based on premise that a stable rhythmic gait pattern should consist of acceleration patterns that repeat in multiples of two. Those which do not repeat in multiples of two are out of phase accelerations and therefore manifest as irregular accelerations during walking. The harmonic content of acceleration signal is evaluated in each direction using stride frequency as the fundamental frequency component. The acceleration signals that are in phase (even harmonics) are compared to components out of phase (odd harmonics) using finite fourier series (Figure 38). The harmonic ratio is calculated by dividing the sum of amplitudes of the first ten even harmonics by the first ten odd harmonics for AP and Vertical direction (since both AP and vertical directions are biphasic for any stride) and its inverse for medio-lateral direction (basic ML pattern is limb dependent and only repeated once for any given stride). Higher harmonic ratio represents smoother walking pattern.
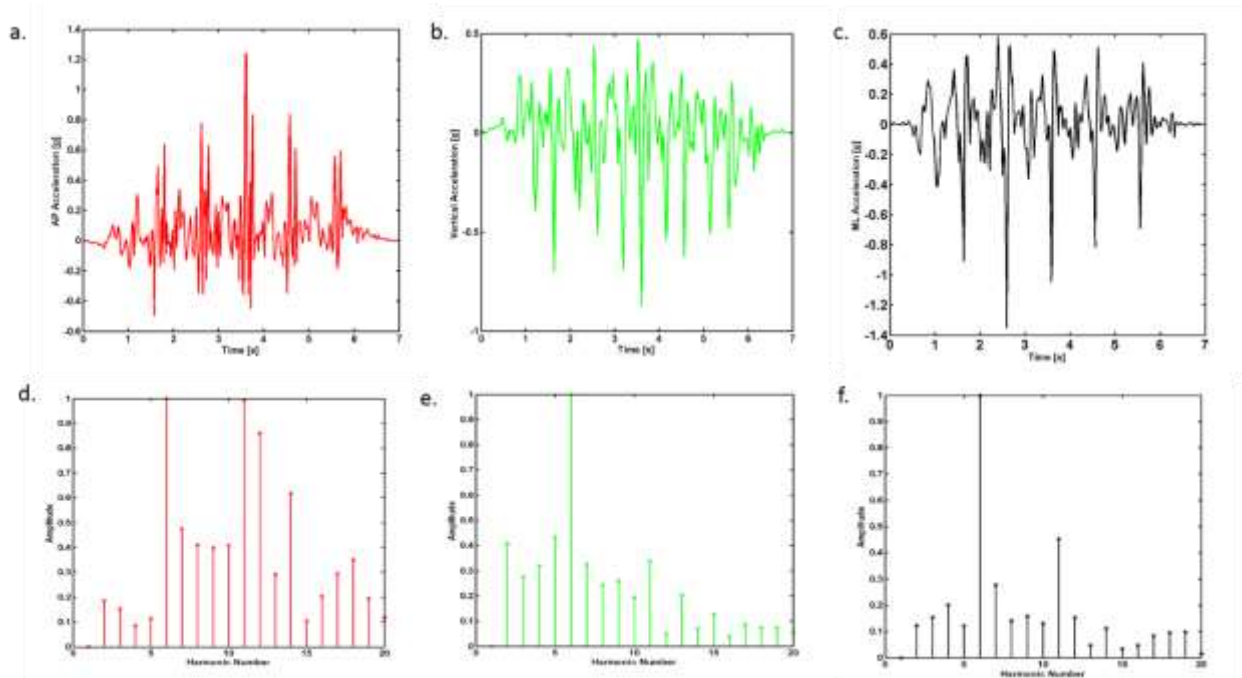
Figure 38. a) Acceleration in AP direction, b)acceleration in vertical direction, c) acceleration in ML direction, d) harmonics of AP acceleration, e)harmonics in vertical acceleration, f) harmonics in ML acceleration for 5m walk.

For Vertical and anterior posterior directions

$$Harmonic\ Ratio = \sum even\ harmonics / \sum odd\ harmonics \qquad 36$$

For Medio-lateral direction

$$Harmonic\ Ratio = \sum odd\ harmonics / \sum even\ harmonics \qquad 37$$

Results

Walking velocities computed using stopwatch time and smartphone time were found to be correlated with Pearson correlation coefficient=0.8154 and spearman's rho=0.8834(Figure 39). Eight participants were classified as frail and eight as non-frail using the velocities from stopwatch (with cut-off velocity =0.833 m/s).
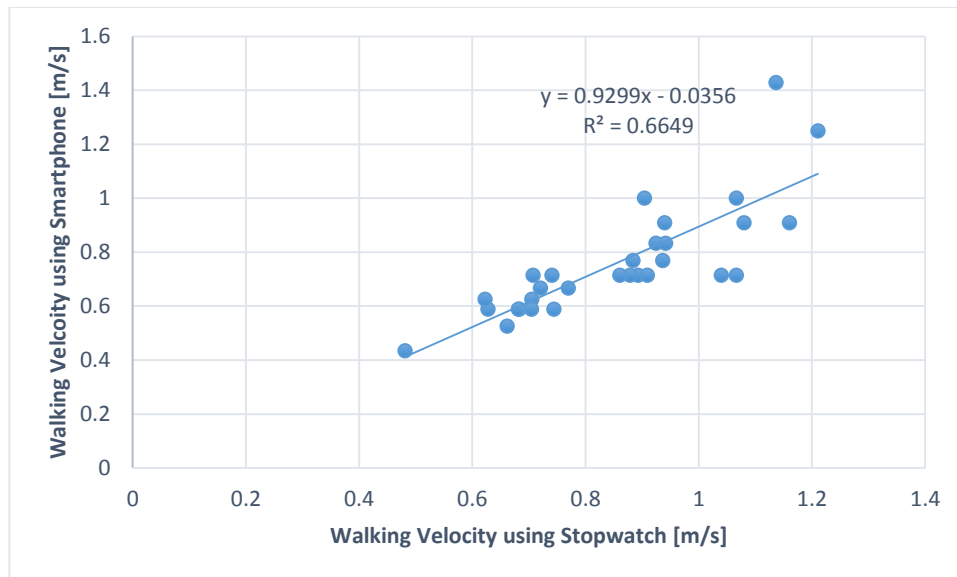
Figure 39. Relationship between the velocities of CVD patients computed from the two systems: stopwatch and smartphone

Table 19 lists velocities from two different systems (stopwatch vs smartphone).

Figure 40 shows interactive dot diagram of the data of the frail and non-frail groups

are displayed in dots on two vertical axes. The horizontal line indicates the cut-off

point with best separation (minimal false negative and false positive results) between

the two groups. The specificity=91.3% and sensitivity =79.2 %. The mean walking

velocity by stopwatch for frail was 0.67 m/s and for non-frail group it was 0.98 m/s.

Using smartphone IMU's the mean walking velocity for frail group was 0.75 m/s

(corrected using the regression equation in Figure 39) and for non-frail group 0.87

m/s (Table 20). Forceplate system detected significantly higher mean COP radius

(p<0.01), COP area (p<0.01), COP path length (p<0.01), mean COP Velocity

(p<0.01) and higher linear variability in parameters such as SD COP-AP (p<0.01), SD

COP-ML (p=0.01), SD COP-R (p=0.02). Using non-linear variability parameters such

as entropy, it was found that there was loss in complexity in AP direction in frail

group.

ApEn COP-AP was found to significantly lower in frail patients (p<0.01). Sample entropy was also found to be lower in AP direction (p<0.01). Mean power frequency was found to be lower in frail group than the non-frail group (p<0.01).

Smartphone based velocity was significantly lower in frail patients than that in non-frail patients (p<0.01). Mean COP radius (p<0.01), COP Area (p<0.01), COP path length (p<0.05) and mean COP velocity (p<0.05) were found to be significantly higher in frail patients than that in non-frail patient group.

Table 19 Classification of patients to frail and non-frail categories using velocities (cut-off = 0.833 m/s) from stopwatch. Velocities computed using smartphone walking signals are also listed below.

ID	Classification Frail(F), Non-Frail (NF)	Stopwatch Velocity			Smartphone Velocity			Classif-ication Errors
		Mean	SD	CV	Mean	SD	CV	
ID04	NF	0.92	0.02	2.44	0.81	0.04	4.56	X
ID06	F	0.59	0.07	12.25	0.61	0.03	4.29	
ID08	F	0.73	0.05	6.44	0.66	0.06	9.70	
ID09	NF	1.21	0.07	5.89	1.23	0.22	17.56	
ID10	NF	1.06	0.14	12.95	0.97	0.05	5.41	
ID11	F	0.73	0.03	3.45	0.67	0.04	6.68	
ID13	F	0.83	0.09	10.27	0.75	0.07	9.11	
ID14	NF	1.06	0.02	2.34	0.81	0.16	20.38	X
ID17	F	0.67	0.01	1.51	0.58	0.05	8.60	
ID18	NF	0.91	0.03	2.97	0.70	0.07	10.36	X
ID19	F	0.58	0.09	15.04	0.59	0.14	24.14	
ID20	NF	0.97	0.10	9.92	0.88	0.04	4.95	
ID21	F	0.83	0.06	7.02	0.72	0.05	7.00	
ID22	F	0.74	0.05	6.95	0.65	0.10	16.11	
ID23	NF	0.95	0.09	9.24	0.77	0.06	7.71	
D24	NF	1.00	0.06	6.33	1.01	0.10	10.04	
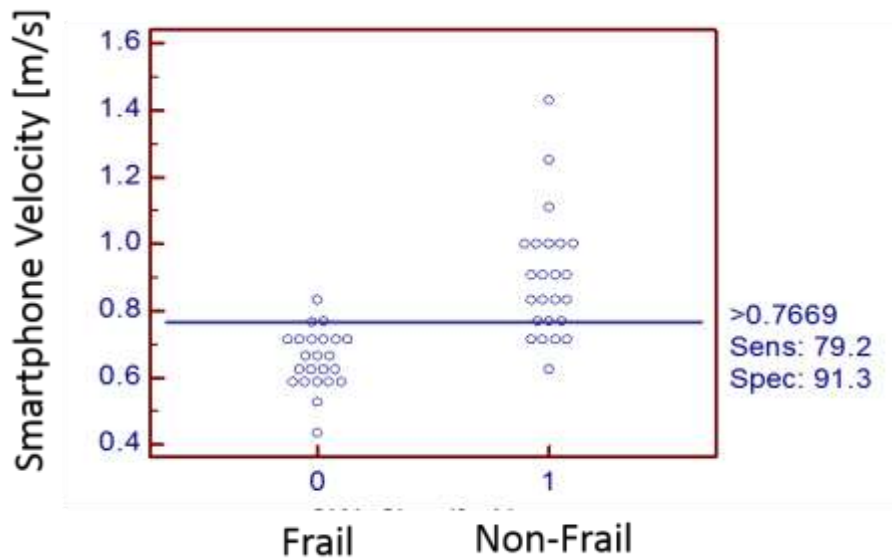
Figure 40. Integrative dot diagram suggesting specificity =91.3% and Sensitivity =79.2% for velocity derived from smartphone signals in classification of frail/non-frail patients

SD COP-AP (p<0.01), SD COP-ML (p<0.01), and SD COP-R (p<0.01) were found to be significantly higher in frail participants. Similarly, RMS COP-AP (p<0.01), RMS COP-ML (p<0.01), RMS COP-R (p<0.01) were also found to be significantly higher in frail participants (Figure 41). There was no significant change found in approximate entropy using smartphone signals. However, sample entropy was significantly lower in frail participants using the smartphone. Both SampEn AP (p<0.01) and SampEn R (p<0.01) were found to be significantly lower than in non-frail group.

Table 20 Linear and non-linear variability parameters from forceplate and smartphone
IMU's for frail and non-frail participants

Instrument		p-Value	Frail			Non-Frail		
			Mean	SD	CV	Mean	SD	CV
Stopwatch			**0.67**	**0.08**	**12.03**	**0.98**	**0.13**	**13.10**
Forceplate	Jerk [N/s]	0.83	0.01	0.00	72.96	0.01	0.00	46.55
	Peak2Peak Time[s]	0.09	1.81	0.86	47.33	1.38	0.69	49.71
	Mean COP Radius* [mm]	<0.01	7.38	2.02	27.30	4.88	1.59	32.60
	COP Area*	<0.01	183.37	114.36	62.36	82.44	53.97	65.46
	SD [COP-AP]*	<0.01	4.50	1.76	39.15	1.79	0.46	25.45
	SD [COP-ML]*	0.01	7.09	1.87	26.45	5.50	2.00	36.44
	SD [COP-R]*	0.02	6.63	2.20	33.17	4.98	2.33	46.87
	RMS [COP-AP]	0.92	28.61	25.42	88.87	27.62	36.61	132.51
	RMS [COP-ML]	0.08	55.13	32.29	58.57	40.29	18.68	46.36
	RMS [COP-R]	0.21	67.71	30.40	44.89	55.64	30.69	55.16
	Path Length [mm]*	<0.01	653.62	192.27	29.42	423.67	94.10	22.21
	Mean COP Velocity*[mm/s]	<0.01	21.79	6.41	29.42	14.12	3.14	22.21
	Alpha [COP-AP]	0.25	1.02	0.27	26.35	0.93	0.24	25.71
	Alpha [COP-ML]	0.13	0.93	0.20	21.40	1.03	0.24	22.85
	Alpha [COP-R]	0.20	0.92	0.21	22.31	1.01	0.24	23.47
	ApEn [COP-AP]*	<0.01	0.56	0.12	22.04	0.69	0.13	18.54
	ApEn [COP-ML]	0.25	0.74	0.15	20.58	0.69	0.14	19.83
	ApEn [COP-R]	0.28	0.71	0.17	23.49	0.66	0.12	18.56
	SampEn [COP-AP]*	<0.01	0.12	0.05	42.36	0.19	0.07	35.03
	SampEn [COP-ML]	0.30	0.19	0.09	47.49	0.16	0.07	41.49
	SampEn [COP-R]	0.34	0.18	0.09	51.19	0.16	0.05	34.15
	MPF [COP-AP]*	<0.01	0.03	0.01	35.61	0.04	0.02	35.92
	MPF [COP-ML]	0.10	0.04	0.02	44.30	0.04	0.01	32.56
Smartphone	Walk Velocity* [m/s]	<0.01	0.65	0.11	17.37	0.87	0.18	20.62
	Mean COP Radius* [mm]	<0.01	14.24	3.93	27.59	8.88	3.76	42.36
	COP Area* [mm^2]	<0.01	682.54	362.45	53.10	289.02	270.77	93.69
	Path Length* [mm]	<0.05	9217.19	2426.75	26.33	7108.34	2010.22	28.28
	Mean COP Velocity* [mm/s]	<0.05	307.19	80.80	26.30	236.94	67.01	28.28
	SD [COP-AP]*	<0.01	12.18	4.69	38.55	6.87	4.27	62.10
	SD [COP-ML] *	<0.01	10.88	3.36	30.89	7.33	2.78	37.87
	SD [COP-R]*	<0.01	8.57	2.75	32.08	5.16	2.64	51.13
	RMS [COP-AP]*	<0.01	12.17	4.69	38.55	6.87	4.27	62.10
	RMS [COP-ML]*	<0.01	10.87	3.36	30.89	7.33	2.78	37.87
	RMS [COP-R]*	<0.01	16.65	4.67	28.05	10.29	4.54	44.16
	Alpha [COP-AP]	0.30	1.03	0.32	31.11	1.15	0.28	24.45
	Alpha [COP-ML]	0.54	1.10	0.24	21.60	1.05	0.20	18.61
	Alpha [COP-R]	0.49	0.96	0.23	24.00	0.90	0.21	23.78
	ApEn [COP-AP]	0.26	1.07	0.10	8.91	1.10	0.04	3.86
	ApEn [COP-ML]	0.43	1.09	0.05	4.91	1.08	0.08	7.76
	ApEn [COP-R]	0.40	1.14	0.05	3.95	1.13	0.03	3.10
	SampEn [COP-AP]*	<0.01	1.36	0.27	20.07	1.69	0.28	16.85
	SampEn [COP-ML]	0.08	1.24	0.25	19.78	1.42	0.31	21.96
	SampEn [COP-R]*	<0.01	1.48	0.22	14.98	1.70	0.20	11.55

Table 21 Post-operative morbidity and mortality of CVD patients

| ID | Age | Sex | Frail/Non Frail | Morbidity | | | | | Mortality | | |
				Stroke	Renal Failure	Prolonged Ventilation	DSWI	Re-Operation >24 h	Death	Skilled Nursing Facility	Length of Stay >14 Days
ID04	77	M	NF	X	X	X	X	X	X	X	X
ID06	78	M	F	YES	X	X	X	X	X	YES	X
ID08	75	M	F	X	X	YES	X	X	X	X	X
ID09	72	M	NF	X	X	X	X	X	X	X	X
ID10	76	M	NF	X	X	X	X	X	X	X	X
ID11	75	M	F	X	YES	YES	X	X	X	YES	YES
ID13	70	M	F	X	X	X	X	X	X	X	X
ID14	77	M	NF	X	X	X	X	X	X	X	X
ID17	72	F	F	X	X	X	X	X	X	X	X
ID18	80	F	NF	X	X	X	X	X	X	X	X
ID19	80	M	F	X	X	X	X	X	X	X	X
ID20	74	F	NF	X	X	X	X	X	X	X	X
ID21	80	F	F	YES	X	YES	X	YES	YES	N/A	N/A
ID22	81	F	F	X	X	X	X	X	X	YES	X
ID23	81	M	NF	X	X	X	X	X	X	YES	X
ID24	71	M	NF	X	X	X	X	X	X	X	X
				F=2	F=1	F=3		F=1	F=1	F=3, NF=1	F=1

Post-operative outcomes consisted both morbidity and mortality. Two frail patients were diagnosed with stroke (ID06 and ID21), one frail patient (ID11) with renal failure, three frail patients (ID08, ID11 and ID21) were kept for prolonged ventilation, one frail patient (ID21) had to be re-operated, 3 frail (ID06, ID11 and ID22) and 1 non-frail (ID23) were sent to skilled nursing facility, only one frail patient (ID11) had length of stay more than 14 days, and one frail patient had mortality (ID21) (Table 21 and Table 22).

Table 22 Definitions of criteria for morbidity and mortality

Stroke*	Stroke (Central Neurological Deficit Persisting >72 hours)
RF*	Renal Failure (new requirement for dialysis or increase in serum creatinine >153 micromoles/liter [>2 mg/dl] and >2-fold the preoperative level)
↑ Vent*	Prolonged Ventilation (>24 hours)
DSWI*	Deep Sternal Wound Infection (requirement for operative intervention and antibiotic therapy, with positive culture)
Reop*	Need for reoperation (for any reason)
Death*	Death (all cause)
SNF*	Discharge to skilled Nursing Facility (health care facility) (rehabilitation , Convalescence, other hospital, nursing home) for on-going care or rehabilitation
↑LOS	Prolonged length of stay (>14 days postoperatively)

It was found that non-frail patients produced higher range of accelerations while performing sit-to-stand maneuver with lower overall variability (Table 23) whereas frail patients produced lower range of accelerations while performing sit-to-stand maneuver with higher variability (measured by Coefficient of variation). The variability in jerk produced during sit-to-stand was also found to be higher in frail patients than in non-frail patients. The mean time taken by frail patients in performing sit-to-stand and stand-to-sit was higher than non-frail patients (Table 23).

Table 23 Variability's in sit-to-stand and stand-to-sit movement parameters in frail and non-frail elderly CVD patients

	Frail						Non-Frail					
	Sit To Stand			Stand To Sit			Sit To Stand			Stand To Sit		
	Mean	SD	CV	Mean	SD	CV	Mean	SD	CV	Mean	SD	CV
Acc Range [m/s2]	44.6	23.8	53.4	55.3	23.0	41.6	52.4	25.0	47.7	72.6	31.3	43.0
Jerk [mm/s3]	15.8	13.9	87.8	17.3	12.4	71.5	15.9	9.2	58.1	21.6	11.1	51.4
Time [s]	1.6	0.6	41.8	1.5	0.7	48.0	1.4	0.4	32.4	1.3	0.3	23.1

There was no significant difference found in harmonic ratios of 5m walk acceleration signals in all 3 directions. However, root mean square (RMS) in all three directions were significantly different in frail and non-frail patients. Non-frail patients produced significantly higher RMS-AP ($p<0.02$), RMS-V ($p<0.01$) and RMS-ML ($p<0.02$) than frail patients (Table 24).

Interactive dot diagrams indicated that RMS in vertical direction could provide results with 100% sensitivity and 100% specificity. Variability measures such as SD and RMS from smartphone postural stability data provided specificity of 93.7% and sensitivity of 50%. The cut-off point being 6.4547 mm could classify frailty with specificity of 93.7% (Figure 41 and Figure 42).

Table 24 Harmonic ratio's and root mean square (RMS –AP,ML,V) and normalized RMSR-AP,ML and V from 5m walk smartphone signals

Variables	Health Status					
	F			NF		
	Mean	Std Dev	CV	Mean	Std Dev	CV
Harmonic Ratio_AP	1.06	0.17	15.96	1.16	0.19	16.77
Harmonic Ratio_V	0.96	0.29	30.50	1.18	0.26	22.14
Harmonic Ratio_ML	1.09	0.37	34.36	0.98	0.25	25.19
RMS_AP*	0.12	0.03	24.50	0.15	0.02	15.23
RMS_V*	0.11	0.01	11.82	0.17	0.02	10.40
RMS_ML*	0.11	0.02	19.63	0.15	0.03	22.48
RMSR_AP	0.59	0.07	11.57	0.55	0.09	16.90
RMSR_V	0.56	0.07	12.89	0.62	0.05	8.47
RMSR_ML	0.58	0.05	9.41	0.55	0.09	15.50
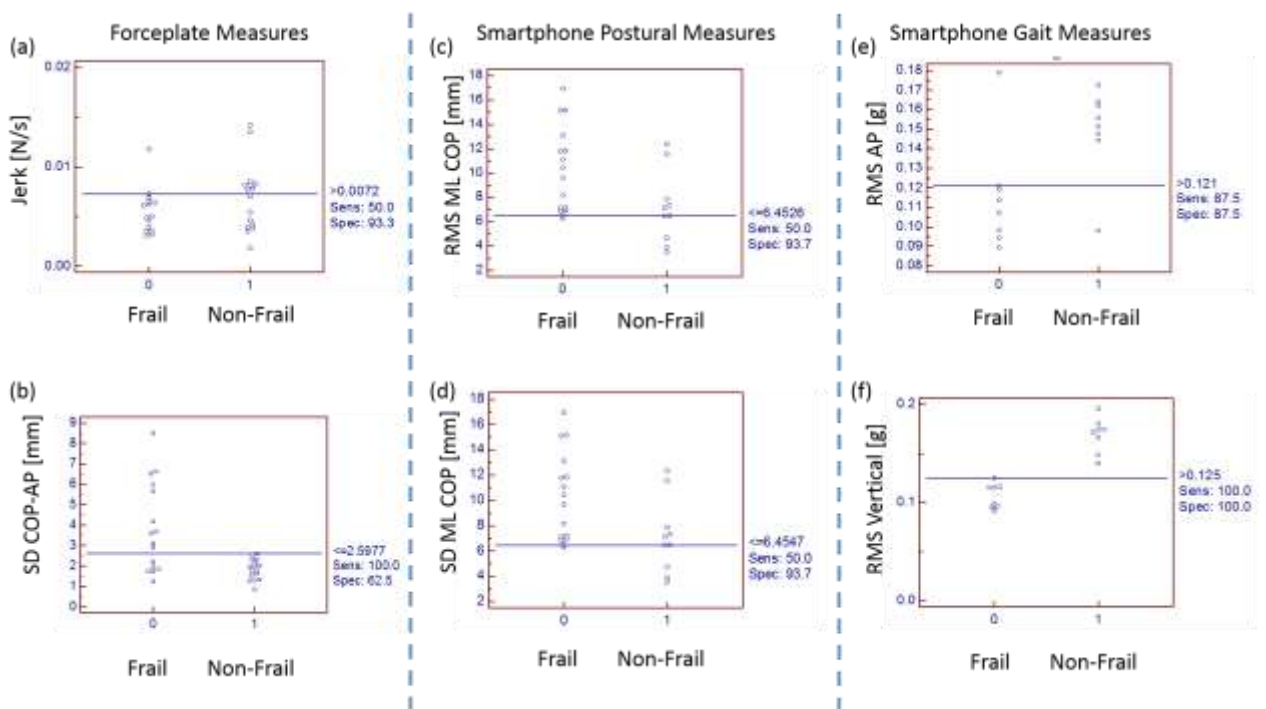
Figure 41. Interactive dot diagram of postural measures from forceplate a) jerk, b) SD of COP-AP, postural measures from smartphone c) RMS COP-ML d) SD of COP-ML and gait measures of smartphone e) RMS AP and f) RMS Vertical.
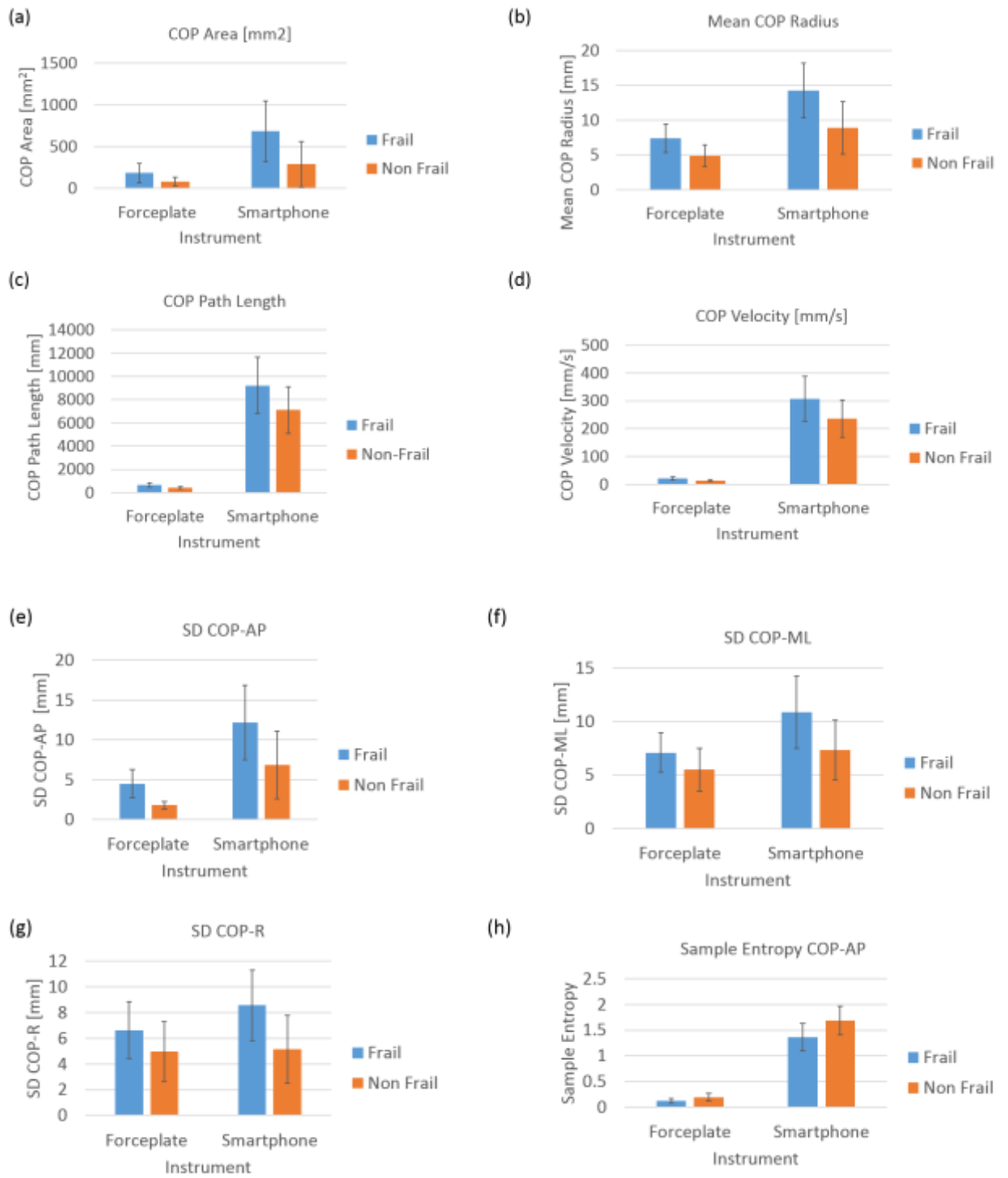
Figure 42. Postural parameters a) COP Area, b) Mean COP area, c) COP path length, d) COP Velocity, e) SD of COP-AP, f) SD of COP-ML, g) SD of COP-R, h) Sample Entropy of COP-AP from both systems i.e. forceplate and smartphone

Discussion

Armed with the above mentioned linear/nonlinear tools and inertial sensors for assessing movement variability, a trait of human movement performance, this study explored the fall risk and its association with variability in cardiovascular disease patients. This study was conducted in clinical environment using smartphone based inertial sensors and found that variability of postural and gait movements in CVD patients was associated with fall risk and adverse post-operative outcomes.

The Society of Thoracic Surgeons has recommended the use of quick tests such as gait speed for assessment of frailty among cardiovascular patients. The frailty status in CVD patients is predictive of adverse health outcome, including falls, institutionalization, hospitalization and mortality (Boyd, Xue, Simpson, Guralnik, & Fried, 2005; Fried et al., 2001; Rockwood et al., 1999). Frail individuals are also at extremely high risk for falls, fractures and hospitalizations leading to death compared with their age matched non-frail counterparts (Fried et al., 2001). Gait speed suggested by STS guideline is a robust measure in health care research, particularly among preoperative cardiac patients (Afilalo, 2011; Afilalo et al., 2010; C. M. Wilson, Kostsuca, & Boura, 2013). Therefore, all study patients were divided into frail and non-frail group using the 5m walk gait speed. This study has established a relationship between frail cardiac patients and their inherent variability. Some studies have previously used inertial sensors in fall risk assessment in hemodialysis clinics (Lockhart et al., 2010; Soangra, Lockhart, Lach, & Abdel-Rahman, 2013). Consistently, in this study, postural stability, postural transition times and gait speed (related to major health related outcomes in frail population), are measured feasibly using smartphone based methodology in clinical environments. There may be important losses of information when measurement of gait velocity is prone to human timing errors (use of stopwatch also requires experimenter's attention and reaction

time to press start or stop pushbuttons after visual verification of event). In clinical practice, where gait speed is an important predictive of severe health-outcomes such as mortality and a subsequent physical disability, an objective, accurate, and reliable way is required for gait speed measurement. For this study, we devised use of smartphone with embedded inertial sensors. Five meter gait speed certainly does not introduce fatigue in patients with cardiovascular impairments awaiting surgery (C. M. Wilson et al., 2013). But some patients who are awaiting their surgery may not be healthy enough to walk for 5 m distance also. It is worthwhile to examine the effects of postural control for balance and transitioning. It was found that in 5 m walk trials, that stopwatch time and smartphone time were highly correlated.

However, we found 3 errors in classification where non-frails were classified as frails. Thus, if non-frail being the desired outcome with smartphone based gait speed, the sensitivity =79.2% and specificity=91.3% when the cut-off velocity is chosen as 0.766 m/s rather than prescribed 0.833m/s. The feasibility and agreement of this smartphone app in estimation of 5 m gait speed in clinical environment has been reported earlier(Soangra & Lockhart, 2014). Frail CVD patients (0.67±0.08 m/s) walked slower than non-frail (0.98±0.13 m/s) counterparts.

It was hypothesized that frailty will increase fall risk in frail CVD patients by both linear and nonlinear measures of postural sway. In support of this hypothesis, we found significant increase in linear parameters such as mean COP radius, COP area, COP path length, mean COP Velocity for frail patients than non-frail patients. Coherently, it was also seen in linear variability measures that frail patients had significantly higher standard deviation (SD) in anterio-posterior, medio-lateral and resultant directions of COP. Frail patients had significantly lower complexity than the non-frail patients in anterior posterior direction. Statistical variability such as range, SD reflect overall magnitude of COP displacement, without considering the temporal

structure of COP time series. This fundamental difference may explain that nonlinear measures of postural signals reveal subtle temporal properties of signals which are not detected in obese through traditional linear approach (Cavanaugh et al., 2005; Handrigan, Corbeil, Simoneau, & Teasdale, 2010; Hue et al., 2007). Traditionally, higher COP displacements have been linked with less stability and consequently, pathology. However, biological systems are intrinsically complex and linear analysis does not account for the time-dependent evolution of the system, eschewing patterns within the time series and appreciable amount of information on system dynamics. Thus, an increased COP movement should not unwittingly indicate deficient postural stability, but rather an element of a healthy vigilant system able to adapt to unexpected perturbations in an attempt to maintain balance.

Entropy-based estimations of signal irregularity and concurrent organizational variability is associated with the adaptive capacity of frail/non-frail participants to maintain balance – frail participants were found to have significantly lower ApEn and SaEn values during prolonged quiet standing in the AP direction, indicating greater regularity and possibly decreased complexity. The findings coincide with previous investigations (Pincus, 1991), linked with the theory of decreased complexity attributed to pathology and aging (Lipsitz & Goldberger, 1992) . Probably, in frail patients the bodily degrees of freedom are constrained in the AP direction compared to the ML direction, whereby coordination of the physiological system, coupled with environmental interactions, lead behavioral processes into less complex, more stable response modes (i.e., more regular sway pattern and closed loop short term dependencies to restore balance). Hence, the motor system is unable to adjust to the demands inherent to frailty, therefore movements transition to a more rigid postural control behavior in the AP direction – delineated by repeated patterns and decreased complexity – diminishing both adaptability and stability. In this context, the decrease

in complexity may be due to impaired feedback control or impaired proprioception leading to reduced adaptive capacity of the underlying postural system (Manor et al., 2010).

Fractal analysis of the COP time series revealed relatively marginal differences in frail versus non-frail patients in all the AP and ML and resultant directions. Frail patients generally had higher α values in the AP direction and non-frail had higher α values in ML direction in eyes-open condition, without reaching significance. From a biomechanics perspective, it may also be due to inability of elderly people to control and accelerate center-of-mass (COM) over base of support, perhaps due to lack of strength and degradation of type II fibers in skeletal muscles in presence of sarcopenia or any other frailty disorder. While muscle strength was not objectively measured in this study, it has been documented that many older people have relatively weaker tibialis anterior and vastus lateralis muscle strength compared to that of healthy adults (Hurley, Rees, & Newham, 1998; Murray, Gardner, Mollinger, & Sepic, 1980).

Frailty is also found related with lower level of physical activity and impaired cardiorespiratory fitness and grip strength compared to lean counterparts (Fried et al., 2001), which could possibly impair an individual's ability to correct a shift in the body's COM and effectively prevent from falling. Probably increased postural sway could be an adaptive strategy to provide additional stability under conditions of weakness in muscles involved for postural control. Age-related deterioration of sensory and neuromuscular control mechanisms could have added to this problem(Lockhart, Woldstad, & Smith, 2003). Degradation of balance shows that fall risk is increased in frail CVD patients. Smartphone based variability information was found similar to that from forceplate, in addition, root mean square acceleration in AP, ML and resultant (R) directions were found to be significantly different among

the frail CVD patients than the non-frail counter parts (Table 20). Additionally, significantly higher sample entropy was found in AP direction in frail patients using the smartphone.

Most of the frail patients met with adverse post-operative outcome which included stroke (2 frail), renal failure (1 frail), prolonged ventilation (3 frail), reoperation (1 frail), longer length of stay (1 frail) and admissions to skilled nursing facility (3 frail and 1 non-frail). And there was one mortality of frail patient. The post-operative outcomes such stroke, renal failure, prolonged ventilation, reoperation longer length of stay in intensive care and admission to skilled nursing facility can be classified as morbidity. The one non-frail elderly participant requested discharge to a skilled nursing facility due to personal/social reasons (in absence of anyone at home to take care). Frail patients took longer time to perform daily activities such as sit-to-stand (1.62±0.68 seconds vs 1.45±0.47 seconds) and failed to produce accelerations comparable to that of non-frail, but these differences were not significant. The variability in jerk during sit-to-stand maneuver was found to be higher than non-frail.

Walking patterns and variability may be optimal from the perspective of energy expenditure (Zarrugh, Todd, & Ralston, 1974), temporal variability (Maruyama & Nagasaki, 1992), spatial variability ("Optimal Walking in Terms of Variability in Step Length," 1997), and attentional demands(Kurosawa, 1994). Stability while walking is an important since, as up to 70% of falls occur during locomotion (BERG, ALESSIO, MILLS, & TONG, 1997; Cali & Kiel, 1995). Moe-Nilssen evaluated walking stability using accelerometers at lumbar spine (Moe-Nilssen, 1998a, 1998b), and reported higher average accelerations in people with balance impairments (Moe-Nilssen, 1998c). During the course of locomotion humans respond to multiple irregular perturbations generated by walking. The task of maintaining stability while walking primarily requires controlling the motion of

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COM. It has been reported that normal subjects walking with plaster casts and crutches (Smidt et al., 1971), amputees walking with prosthetic limbs (Robinson, Smidt, & Arora, 1977) and older people with balance problems (Yack & Berger, 1993) have lower harmonic ratio. But we did not find any significant differences in harmonic ratios of frail and non-frail group for 5 m distance walk in all 3 directions.

In this study, acceleration patterns were measured at the pelvis when walking (5 m walk) to provide an indicator of whole body stability in response to multiple unpredictable perturbations during walking. All humans have a preferable walking speed that is combination of step length and step frequency and is important factor in control of balance since while walking considerable potential for imbalance exists due to inertia of the upper body and the small contact area provided by the foot during single limb stance (Latt, Menz, Fung, & Lord, 2008).This preferable or usual speed is selected to optimize the stability of gait pattern. In this study, acceleration patterns were measured at the pelvis when walking to provide an indicator of whole body stability in response to multiple unpredictable perturbations during walking. However, we found that root mean square (RMS) in all three directions were significantly different in frail and non-frail patients. Probably, RMS acceleration are correlated with walking speed. The frail patient's comfortable walking speed selected is slower when compared to non-frail patient to minimize acceleration variability but instead RMS values were found significantly higher and thus were unable to provide smooth and rhythmic movements of pelvis. Interactive dot diagram suggested that of postural measures from forceplate such as jerk, SD of COP-AP, postural measures from smartphone such as RMS COP-ML, SD of COP-ML and gait measures of smartphone such as RMS AP and RMS Vertical are predictive of frailty with high accuracy. These methods build on narrative descriptions of variability in fall risk assessment by quantifying qualities of postural control, postural transition and gait. In combination,

linear and nonlinear variability analysis quantified postural and gait control to provide a more complete understanding of the adaptive strategies used in neuromuscular control than either method could provide alone. Thus, fall risk assessment has high predictive validity to identify patients with adverse post-operative outcomes through an objective method of assessment using smartphone. These fall risk indicators of variability could be used as prescreening tools for many different kinds of surgical procedures and in turn help clinicians to identify frail patients who may need intensive rehab or preplan their stay in hospital with specialized nursing care before return to home.

Thus inertial sensors have potential to measure gait and posture in CVD patients, although a great deal of work is required in future research to make such research tools easy to use for clinicians. The data collection was mainly conducted by hospital staff (Registered nurse specialist) using smartphone and forceplate in the clinical setting and experimental protocols were modified as per clinical requirements. To meet the challenges of patient safety and point of care, new technologies are needed in future such that the patient data can be acquired without hindering medical routine for patients and hospital staff.

Limitations of the study

The strength of the conclusions of this study must be tempered by the study's limitations. The patients were aware that they were participating in a fall risk assessment protocol. This could be a bias in the population studied. They may be conscious of the environment and their performance may have been affected by the clinical environment. The hospital setting in which data were obtained for this study provided an unusual environment for the cardiac patients. At the same time, they might be stressed to some extent for their surgery allotted for next day and non-

laboratory setting limited the scope of this data. However, such analyses may provide insight as to the potential fall risk and chances of adverse post-operative outcomes were associated with frail condition of patients.

Another limitation of current study was that the smartphone based assessments required patients to stand-still before and after 5-m walk. As automatic algorithms developed in smartphone app determined velocity by evaluating start and stop times of movement. Automatic gait speed estimation by smartphone required strict following of protocol. If any other movement artifact is followed after or prior to the walking task, the movement time may get increased than the actual walking time and thus data had to be checked visually and truncated for correcting this.

Conclusion

The accurate measure of gait speed, as well as variability measures can improve the clinical evaluation of cardiac patients, providing an earlier detection of individuals at higher risk of major health-related events such as physical disability and mortality. This study demonstrated that 5-m gait speed measurement using smartphone is also a reliable objective measure, however adhering to certain protocol is suggested for using smartphone app. Although different methods have been used previously to measure gait speed and these have affected clinical interpretation and implementation of the gait speed (Graham, Ostir, Fisher, & Ottenbacher, 2008; Graham, Ostir, Kuo, Fisher, & Ottenbacher, 2008). By providing a smartphone based clinically useful gait speed assessment method with a well-defined protocol which is simple, quick, easy to perform in clinics, it is hoped that smartphone for gait speed assessment will be promoted and encouraged in clinical and research settings. In addition, nonlinear postural variability measures such as complexity can be easily implemented in patients who are unable to walk but can stand-still for at-least 30 seconds.

The study protocol and findings suggest that various variability parameters in walking and stand-still posture can be easily implemented in cardiovascular clinical practice with high acceptability by the patients and clinical research staff. Patients started with standing still posture and walked at their usual pace, as if they were walking in their own home, and given no further encouragement or instructions. This data can be readily collected in non-laboratory environments and can be used to help interpret the results for health related events.

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Chapter 6 Summary and Conclusions

Summary

Approximately one-third of adults (aged 65 and older) are estimated to fall every year (Campbell, Borrie, & Spears, 1989; Tinetti, Speechley, & Ginter, 1988). Likewise, frailty is highly prevalent with old age (Fried et al., 2001) and associated with assorted CVD outcomes. Recently, the Society of Thoracic Surgeons recommended quick frailty assessment be used in cardiac clinics, because frail CVD patients have adverse post-operative outcomes. In recent decades, the mitigation of fall accidents in the elderly population has been substandard and incidence of fall accidents remains high. Thus, the paucity in adequate information to prevent falls requires alternative and novel linear and nonlinear variability tools to assess fall risk among older adults. First, there is a need to investigate if movement variability in older adults is related to slip and fall risk. Second, can this movement variability be measured using inertial sensors. Third, can these inertial sensors be used to measure the gait and posture movement variability of frail patients in cardiovascular clinics.

Previous studies have shown that the dual-task paradigm influences age related changes in attentional capacities (Kerr, Condon, & McDonald, 1985), limiting the shared processing (Bowen et al., 2001) of two concurrent tasks. Further, prior investigations report that individuals with a history of falls have more significant gait changes while performing a dual task than that of non-fallers (Faulkner et al., 2007; Toulotte, Thevenon, Watelain, & Fabre, 2006; Vaillant et al., 2006; Woollacott & Shumway-Cook, 2002). However, the present study reveals contrasting results: dual task related gait changes are favoring "cautious gait" in both healthy young and older

adults; moreover, the dual-task shifted the slip initiation risk and slip severity parameters towards a safer gait.

Being a multifactorial problem in its origin (Myers, Young, & Langlois, 1996), falls may be associated with frailty. Frail elderly may therefore demand more attention in gait, postural, functional and mobility tasks. Frail subjects may struggle with dual-task conditions failing to shift into "**cautious gait mode,**" demonstrated by their healthy counterparts. These frailty characteristics were marginally apparent in sit-to-stand transition times but variability in postural parameters such as complexity and COP standard deviations were significantly different for frail and non-frail patients.

Thus, inertial measurement units may provide noninvasive clinical point of care testing in clinical settings. Considering the aforementioned society of thoracic surgeon's recommendations and the concomitant adverse outcomes of frailty, inertial sensors could be an additional tool of assessing frailty and associated underlying variability in the neuromuscular control system to evaluate elderly patients at mortality risk before cardiac surgical intervention. Understanding the relationship of multiple impairments along with balance and stability to one's fall risk will enable better understanding to clinicians to develop appropriate remedial strategy of frailty. Although this study is limited to 16 CVD patients, findings are promising and further analysis on a larger cardiac patient population is warranted.

Future Recommendations

While the outcome of this work contributes to the apperception of various movement variability parameters assessed using inertial sensors, there are areas that still require further investigation. This study was partially completed in the Locomotion Research Laboratory, Virginia Tech and partially in Carilion Roanoke Memorial Hospital, Cardiac Surgery Department, which limited comparison of variability into

populations at two different environments. In future, study need to be designed in larger population size in clinical environments with CVD patients (consisting both frail and non-frail population) and age matched healthy control group.

Conclusion

Overall, the current research has contributed knowledge about variability in healthy young and older adults and the effects that a dual-task paradigm has on movement variability, slip initiation characteristics and slip severity. The study results suggest that a dual-task elicits a "cautious gait mode" (CGM) which is an innate adaptive response to counter reduced attention while walking. Attention resources are appropriated for the relevant cognitive task (e.g. counting backwards), thereby the healthy human response is to adopt a cautious gait mode, which includes a shorter step length, longer stance duration - acquire more proprioceptive information from the ground (or use less attentional resources) - and inducement of higher complexity in the anterior-posterior direction of movement.

The information attained by understanding variability aspects of dual-task paradigm (in study I) and revelation of the common risk factors for falls and adverse post-operative outcome (in study III) has critical implication in the future research on motor variability in frail older adults. The response of CGM is innate for healthy human beings, but in the case of frail elderly persons, who require considerable attention for performing relatively perfunctory gait and postural movements, they may find it challenging to maintain stability.

In light of our findings, the next step is to test elderly individuals in their home, community or clinical environments using variability information relevant to fall risk. Future work is needed to confirm that inertial sensors embedded inside a

smartphone is an effective data collection tool to measure gait, posture stability,

postural transitions and gait stability among elderly individuals in their daily

environments.

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Appendix

Mini Mental State Examination

Patient: _____   **Date:** _____

Mini-Mental State Exam

Folstein MF et al, *J Psychiatr Res*. 1975; 12:189-98.

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ORIENTATION
Ask each of the following questions, and score 1 for each correct answer

1. What is the day of the week❏, month❏, date❏, year❏, season❏?   __ /5

2. Where are we? state❏, county❏, town❏, hospital❏, floor❏.   __ /5

REGISTRATION
Name 3 objects slowly and clearly: "**apple, penny and table**". Ask the patient to repeat them. Tell the patient to remember the objects because s/he will be asked to name them in a few minutes.

Score the first try: apple❏, penny❏, table❏.   __ /3

Repeat objects until all are learned, up to 6 trials.

ATTENTION AND CALCULATION
Ask the patient to perform serial 7 subtraction from 100.   __ /5
Stop after 5 numbers and score 1 for each number.

                                           score serial 7's, unless patient refuses or is unable to answer

If the patient scores less than 5, ask her/him to spell the word "WORLD" backward.   __ /5

Include the higher of these two scores in final score.

RECALL
Ask the patient to recall the names of the three objects which you asked her/him to repeat above and score 1 for each correct name: apple❏, penny❏, table❏.   __ /3

LANGUAGE
1. Naming: Point to two objects (e.g., watch and pen) and ask the patient to name them. Score 1 for each correct name.   __ /2

2. Repetition: Ask the patient to repeat, "No ifs, ands or buts." Allow only one trial..   __ /1

3. Three-stage command: Ask the patient to "Take a paper in your right hand, fold it in half, and put it on the floor." Score 1 for each part correctly executed.   __ /3

4. Reading: Point to the phrase "CLOSE YOUR EYES" on page 2. Ask the patient to read the sentence to do what it says. Score 1 if eyes are closed.   __ /1

5. Writing: Ask the patient to write a sentence on page 2. Do not dictate a sentence. The sentence must contain a patient and verb, and must make sense. Correct spelling and punctuation are not necessary. Score 0 or 1.   __ /1

6. Copying: Ask the patient to copy the figure on the page 2 exactly. All 10 angles and intersections must be present to score 1.   __ /1
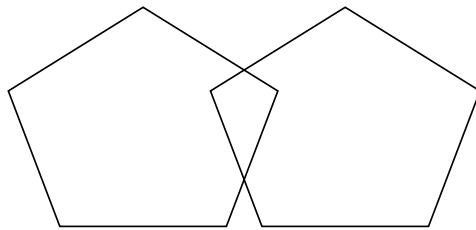   _____

                                    TOTAL: __ __ / __ __

CLOSE YOUR EYES

Write sentence here:

_____

_____

Copy the figure:

Short Physical Performance Battery

1. Repeated Chair Stands

Instructions: Do you think it is safe for you to try and stand up from a chair five times without using your arms? Please stand up straight as quickly as you can five times, without stopping in between. After standing up each time, sit down and then stand up again. Keep your arms folded across your chest. Please watch while I demonstrate. I'll be timing you with a stopwatch. Are you ready? Begin

Grading: Begin stop watch when subject begins to stand up. Count aloud each time subject arises. Stop the stopwatch when subject has straightened up completely for the fifth time. Also stop if the subject uses arms, or after 1 minute, if subject has not completed rises, and if concerned about the subject's safety.. Record the number of seconds and the presence of imbalance.. Then complete ordinal scoring.

Time: _____sec (if five stands are completed)

Number of Stands Completed: 1   2   3   4   5

Chair Stand Ordinal Score: _____

    0 = unable
    1 = > 16.7 sec
    2 = 16.6-13.7 sec
    3 = 13.6-11.2 sec
    4 = < 11.1 sec

2. Balance Testing

Begin with a semitandem stand (heel of one foot placed by the big toe of the other foot). Individuals unable to hold this position should try the side-by-side position. Those able to stand in the semitandem position should be tested in the full tandem position. Once you have completed time measures, complete ordinal scoring.

a. Semitandem Stand

Instructions: Now I want you to try to stand with the side of the heel of one foot touching the big toe of the other foot for about 10 seconds. You may put either foot in front, whichever is more comfortable for you. Please watch while I demonstrate.

Grading: Stand next to the participant to help him or her into semitandem position. Allow participant to hold onto your arms to get balance. Begin timing when participant has the feet in

position and lets go.

Circle one number

    2. Held for 10 sec

    1. Held for less than 10 sec; number of seconds held _____

    0. Not attempted

b. Side-by-Side stand

Instructions: I want you to try to stand with your feet together, side by side, for about 10 sec. Please watch while I demonstrate. You may use your arms, bend your knees, or move your body to maintain your balance, but try not to move your feet. Try to hold this position until I tell you to stop.

Grading: Stand next to the participant to help him or her into the side-by-side position. Allow participant to hold onto your arms to get balance. Begin timing when participant has feet together and lets go.

Grading

    2. Held of 10 sec

    1. Held for less than 10 sec; number of seconds held_____

    0. Not attempted

c. Tandem Stand

Instructions: Now I want you to try to stand with the heel of one foot in front of and touching the toes of the other foot for 10 sec. You may put either foot in front, whichever is more comfortable for you. Please watch while I demonstrate.

Grading: Stand next to the participant to help him or her into the side-by-side position. Allow participant to hold onto your arms to get balance. Begin timing when participant has feet together and lets go.

Grading

    2. Held of 10 sec

    1. Held for less than 10 sec; number of seconds held_____

    0. Not attempted

Balance Ordinal Score: _____

    0 = side by side 0-9 sec or unable

    1 = side by side 10, <10 sec semitandem

2 = semitandem 10 sec, tandem 0-2 sec 3 = semitandem 10 sec, tandem 3-9 sec 4 = tandem 10 sec

3. 8' Walk (2.44 meters)

Instructions: This is our walking course. If you use a cane or other walking aid when walking outside your home, please use it for this test. I want you to walk at your usual pace to the other end of this course (a distance of 8'). Walk all the way past the other end of the tape before you stop. I will walk with you. Are you ready?

Grading: Press the start button to start the stopwatch as the participant begins walking. Measure the time take to walk 8'. Then complete ordinal scoring.

Time: _____ **sec**
Gait Ordinal Score: _____

    0 = could not do
    1 = >5.7 sec (<0.43 m/sec)
    2 = 4.1-6.5 sec (0.44-0.60 m/sec)
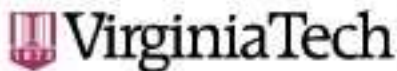   3 = 3.2-4.0 (0.61-0.77 m/sec)
    4 = <3.1 sec (>0.78 m/sec)

Summary Ordinal Score: _____
Range: 0 (worst performance) to 12 (best performance). Shown to have predictive validity showing a gradient of risk for mortality, nursing home admission, and disability.

Informed Consent Form

Office of Research Compliance
Institutational Review Board
North End Center, Suite 4120, Virginia Tech
300 Turner Street NW
Blacksburg,
Virginia 24061
540/231-4606 Fax
540/231-0959
email irb@vt.edu
website http://www.irb.vt.edu

VirginiaTech

MEMORANDUM

DATE: October 4, 2013

TO: Thurmon E Lockhart, Rahul Soangra, Jian Zhang

FROM: Virginia Tech Institutional Review Board (FWA00000572, expires April 25, 2018)

PROTOCOL TITLE: Non-Intrusive Multi-Patient Fall-Risk Monitoring in Health Care Facilies

IRB NUMBER: **11-1088**

Effective October 3, 2013, the Virginia Tech Institution Review Board (IRB) Chair, David M Moore, approved the New Application request for the above-mentioned research protocol.

This approval provides permission to begin the human subject activities outlined in the IRB-approved protocol and supporting documents.

Plans to deviate from the approved protocol and/or supporting documents must be submitted to the IRB as an amendment request and approved by the IRB prior to the implementation of any changes, regardless of how minor, except where necessary to eliminate apparent immediate hazards to the subjects. Report within 5 business days to the IRB any injuries or other unanticipated or adverse events involving risks or harms to human research subjects or others.

All investigators (listed above) are required to comply with the researcher requirements

outlined at: http://www.irb.vt.edu/pages/responsibilities.htm

(Please review responsibilities before the commencement of your research.)

PROTOCOL INFORMATION:

Approved As: **Expedited, under 45 CFR 46.110 category(ies) 4,7**
Protocol Approval Date: **October 3, 2013**
Protocol Expiration Date: **October 2, 2014**
Continuing Review Due Date*:   **September 18, 2014**
*Date a Continuing Review application is due to the IRB office if human subject activities covered under this protocol, including data analysis, are to continue beyond the Protocol Expiration Date.

FEDERALLY FUNDED RESEARCH REQUIREMENTS:

Per federal regulations, 45 CFR 46.103(f), the IRB is required to compare all federally funded grant proposals/work statements to the IRB protocol(s) which cover the human research activities included in the proposal / work statement before funds are released. Note that this requirement does not apply to Exempt and Interim IRB protocols, or grants for which VT is not the primary awardee.

The table on the following page indicates whether grant proposals are related to this IRB protocol, and which of the listed proposals, if any, have been compared to this IRB protocol, if required.

Date*	OSP Number	Sponsor	Grant Comparison Conducted?
09/26/2013	11058609	National Science Foundation	Compared on 09/26/2013

* Date this proposal number was compared, assessed as not requiring comparison, or comparison information was revised.

If this IRB protocol is to cover any other grant proposals, please contact the IRB office (irbadmin@vt.edu) immediately.