AGING-RELATED DECREMENTS DURING THE ACTIVTIES OF THE TIMED UP AND GO TEST WHEN COMBINED WITH MOTOR TASK AND VISUAL STIMULATION

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ABSTRACT

Falls in older adults are linked with increased morbidity and mortality, and remain a major public health concern. Aging is associated with increased reliance on the visual system for postural control, termed “visual dependence”. Discordance between visual, vestibular and proprioceptive sensory information can lead to balance loss and falls. In addition to increased visual dependence (VD), older adults have more difficulty than younger adults in multi-tasking, performing simultaneous tasks (e.g. walk and talk), which may further increase fall risk. A common clinical test of physical function, the Timed Up and Go (TUG), requires the individual to stand from a seated position, walk forward, turn, walk back to the seat, turn and sit back down. Previous studies have explored the effect of multi-tasking during the TUG; however, the role of visual dependence and its interaction with multi-tasking on specific movement components of the TUG has not been deeply explored in the existing literature. The primary goal of this project was to understand the effects of aging and visual dependence on physical function as measured by the TUG. The three aims that guided this work are 1) to examine how aging affects performance metrics and movement components of the TUG (e.g. sit-to-stand, walking, turning, stand-to-sit) when simultaneously performing a motor task with or without with visual stimulation; 2) to examine how visual dependence affects performance metrics and movement components of the TUG when simultaneously performing a motor task with or without visual stimulation; and 3) to determine the effect of wearing a head mounted display (HMD) on performance metrics and movement components of the TUG in older and younger adults.

Twelve younger adults (6 males) and sixteen older adults (8 males) participated in this work. They were further classified as sixteen visually independent adults (VI) (9 younger adults, 7 older adults, 9 males) and 12 visually dependent adults (VD) (3 young adults, 9 older adults, 5 males). Participants completed eight conditions:

The dependent variables, measured using 6 inertial measurement unit sensors, included spatiotemporal variables of the TUG (total time, sub-component movement times, gait speed, step cadence during turning); three-dimensional peak trunk velocity (PTV) (i.e. around the mediolateral, vertical, and anteroposterior axes); acceleration range and jerk of sit-to-stand and stand-to-sit; and multitask cost. Multitask cost reflects the change in the motor behavior that occurs due to high attentional demanding conditions, with the lowest multitask cost reflecting poorer motor performance. The multitask cost was calculated as the percent change of each dependent variable in relation to the single task performance (i.e. TUG with no secondary task) in the third chapter.

Our results demonstrated that the wear of HMD has an impact on TUG kinematics, regardless of adding a visual stimulus, more than holding a cup of water. Providing a visual stimulus decreased the PTV in walking and acceleration range in sit-to-stand. Particularly, presenting a visual stimulus in a pitch up rotation decreased the PTV in turning and increased the peak trunk velocity in stand-to-sit when compared to standard TUG. Older adults showed a decrease in the multitask cost (i.e. poorer performance) of turn and sit-to-stand time and the PTV in turning and a lower variability in trunk velocity in turning and sit-to-stand and the acceleration jerk in sit-to-stand and stand-to-sit compared to younger adults. Older adults who were visually dependent showed a lower mean and variability in the mediolateral and vertical acceleration range of sit-to-stand than older adults who were visually independent.

Our results indicate that the wear of HMD has an impact on posture that should be taken into account in clinical research. Assessing the kinematics in turning and sitting-to-standing could be of a great interest for future studies that would include older adults with functional limitations (e.g. fallers versus non-fallers). Sit-to-stand motion, in particular, can differentiate older adults who are more sensitive to visual stimulation.

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CHAPTER 1

INTRODUCTION AND REVIEW OF THE LITERATURE

Falls are one of the leading causes of disabilities and mortality in older adults worldwide[1]. Approximately 35-40% of community-dwelling older adults over 65 years old and 50% of adults over 80 years old fall annually [2,3]. Although only 1% of falls cause hip fractures, more than 95% of hip fractures are caused by falls [4]. The risk of mortality with falls is three times greater in someone who experienced a hip fracture compared to someone who did not [5]. Given the high-risk impact of falls on the health of older adults, it's surprising that falls are under-reported with an estimated 75-80% of falls without injury not reported [6]. There are many factors that can increase the risk of falls in older adults including and not limited to: physiological decline in sensory function [7,8] such as diminished sensitivity of the vestibular system [9], diminished vision [10,11] and performing tasks that divide attentional demands while walking, i.e. multitasking [12].

The commonly-used clinical fall-risk assessment tools have been criticized for being subjective, poor sensitivity or specificity, high cost, length of administration time of testing, and poor identification of balance problems [13]. An improved characterization of the underlying kinematic differences in functional movements between young and old adults could advance our understanding of fall mechanisms, which may support improved preventative fall programs. The Timed Up and Go test (TUG) is a clinical examination frequently used to evaluate fall risk in older adults that is considered to be simple, short, and reliable [14]. The TUG is performed by asking the adult to rise from a chair, walk for meters, turn 180°, walk back, and sit on the chair [15] (Figure 1). Risk of falling is determined by multitasking paradigm. Multitasking requires dividing attention, which could lead to interference in performance if it exceeds the information processing capacity of the individuals. It has been documented that advancing age leads to difficulties in processing multiple simultaneous demands on attention [16] due to lesions of the diffuse white matter that affect the fronto-striatal circuits [17]. This is evidenced in the multitask walking literature where healthy older adults exhibit more postural instabilities under these conditions than without them[16], thus resulting in an increased risk of falling [12]. In addition, older adults are more *visually dependent* for postural stability than younger adults, which infers a higher reliance on visual information than other sensory systems to maintain balance[18]. Older adults are also more susceptible to postural instabilities under visually demanding conditions [19]. Given these two factors, it is possible to speculate that older adults may exhibit specific postural responses under visually demanding conditions that could be related to an increased risk of falling. Thus, this project is directed at exploring the role of aging and visual dependency on the motor behavior of younger and older adults.

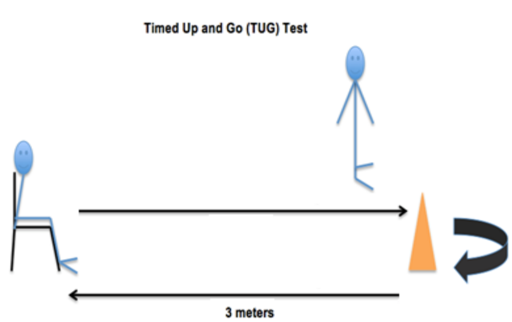

Figure 1 The Timed Up and Go (TUG) Tests

This dissertation explored the impact of wearing an HMD, and multi-tasking (i.e. by adding a motor task and/or visual stimulation to the TUG) on TUG performance both in full and in parts (i.e. sit-to-stand, walking, turning and stand-to-sit) in older and younger adults. We are aware of only one study that has explored the effects of adding a motor task to the TUG test activities on the kinematics of healthy younger and older adults [17]. This study examined the multitask cost of the kinematics defined as the percent change of each dependent variable in relation to single task performance (i.e. TUG with no secondary task). The multitask cost can reflect the change in the motor behavior that occurs due to high attentional demand conditions, with the lowest multitask cost reflecting poorer motor performance [20]. Visual stimulation during walking and sitting to standing has been reported in the literature [21,22]; however, no report of visual stimulation during other TUG activities is found. This study will determine how aging and visual dependency impacts movement patterns while completing the TUG common motor task, attending to a dynamic visual stimulation, and when simultaneously performing the different movement subtasks of the TUG test.

Sensory Contributions to Postural Control

Postural control is the ability to maintain balance and orientation of the upright position in space [23]. Sensory inputs from the somatosensory, vestibular, and visual systems to the central nervous system (CNS) contribute to postural control. The visual system is the predominant system for maintaining postural control compared to other sensory systems [24,25], and provides information regarding the movement and position of the head and body relative to the surrounding environment. The somatosensory inputs provide information about limb position and position of body segments in relation to each other. Additionally, the otolith organs and three semicircular canals of the vestibular system sense the linear acceleration, gravity, and the angular acceleration of the head. Feedback inputs are important to achieve the intended motion control [26]. Multisensory integration, which occurs in subcortical and cortical areas, is a process in which the nervous system integrates the inputs from sensory systems [27]. Under conditions of sensory manipulation, adults need to adjust their dependency level on each sensory system. For example, when the somatosensory system is perturbed, individuals typically decrease their reliance on, or down-weight, the somatosensory system information, and increase the weighting of the vestibular and visual inputs to maintain balance [28].

The somatosensory system undergoes changes in function in response to aging and illness which may lead to falls[26]. Evidence suggests that older adults are less efficient in this sensory integration and re-weighting process and exhibit more postural instabilities than younger adults [29,30]. With aging, there is degradation in the somatosensory system function due to anatomical and physiological changes [31]. Older adults exhibit a decrease in vibration perception threshold and plantar mechanoreceptor sensitivity [32]. Equally important to note, an increased risk of falls in elders is linked with impaired limb proprioception and decreased muscle strength [33].

Vestibular function also declines with aging. In the US, 35.4% of adults older than 40 years of age have a vestibular impairment [34]. In particular, older adults show a loss or death in inner ear cells [35], a decrease in the vestibular ocular reflex (VOR) amplitude [9], neuronal loss in the medial and lateral vestibular nuclei [36], and degeneration of neuronal structures such as the spiral ganglion cells in human ears [35]. These structural alterations often lead to functional deterioration in the vestibulospinal reflex, unilateral vestibular loss, balance problems, and dizziness [36]. Such vestibular impairments also affect posture control and may increase the risk for falls.

Vision also contributes to balance and can degrade with aging [35,36]. The visual field extends to approximately 200 degrees horizontally and 160 degrees vertically. Central vision detects information in the middle 2 to 5 degrees within the visual field and is used to recognize objects and assist in maintaining ambulatory direction. Peripheral vision detects information outside the limit of the central vision but within the visual field, and it updates spatial information. It is also used to update the spatial information in the surrounding environment, such as a potential structural impediment, which could potentiate fall-risk. Through peripheral vision, one uses the optical flow to control movement. Optical flow is the dynamical visual input from the environment that impacts the retina. Cues from the optic flow are used to maintain self-movement and upright control while producing the sufficient motor responses when interacting in dynamic environment [37]. Perceiving a dynamic visual scene through the optic flow showed to affect postural stability [19]. Specifically, when experiencing a moving visual scene, the visual stimulation related to self-motion is integrated with other sensory inputs, i.e. vestibular and somatosensory, to determine the direction and speed of self-motion and elicit the necessary responses to maintain postural control [38]. Further, it is important to attend to the changes in the optical flow that could happen naturally, such as changing the head position in turning or encountering a moving object or a person to maintain postural balance and avoid falls [37,38].

With aging, visual pathologies such as glaucoma and macular degeneration can occur, which affect the peripheral and central vision, respectively [35,36]. These disorders are associated with postural instability [39]. Under multitasking conditions, older adults with macular degeneration showed more instability when standing on firm and foam surfaces, while older adults with glaucoma showed difficulty when standing on a foam surface [40]. This suggests that individuals with these visual impairments have to up-weight their somatosensory inputs to maintain postural control. In addition, older adults exhibit a decrease in visual acuity and contrast sensitivity, which are conditions correlated with the increased risk of fall [8,9]. It is easy to understand how these age-associated changes in the visual system may lead to an increased risk of falls and impairments in posture control.

To maintain upright balance, it is essential to make an accurate temporal and spatial estimation of the body within the environment. Adults who make errors in the temporal and spatial estimation of the body within the environment are usually considered to be visually dependent. Visual dependency is a term used to describe individuals who rely more on the visual sensory information compared to somatosensory or vestibular information and are less efficient in sensory reweighting under faulty visual inputs [41–43]. Visual dependence is usually determined by using a test called the Rod and Frame test in which participants are asked to align a rod inside a frame to true vertical and horizontal position despite changes in alignment of the frame with true vertical and horizontal. Individuals that accurately estimate the vertical and horizontal alignment of the rod are considered visually independent (VI) and those who cannot are considered visually dependent (VD) [42]. Visually dependent adults experience more postural instabilities under visual manipulated inputs [19,43]. For example, adults who were VD showed larger center of mass (COM) and center of pressure (COP) responses in the direction of visual stimulus in standing compared to VI adults [15]. In the older adults, VD occurs with a higher frequency than in younger adults [18], and the elderly more often exhibit an increase in postural sway in the case of absent or misleading visual information than younger adults[44,45]. Given all of these age-related impairments in visual function and processing, an improved understanding of the impact of vision on motor behavior and activities of daily living would be helpful to reduce fall risk.

Given the impact of the visual system in maintaining postural control among older adults a careful examination of the vision on posture is warranted. By manipulating the visual system information during clinical functions tests such as the TUG, we may better be able to determine the function performance of older adults in a visually busy environment.

Virtual Reality

Virtual Reality (VR) is a computer-simulated environment that allows the user to interact with sensory inputs and induces postural sway [43]. The level of immersion is classified based on characteristic of the visual stimuli: non-immersive, semi-immersive, and fully immersive. Examples of a fully immersive VR system are the CAVE™ (University of Illinois at Chicago) [46] and the head mounted display (HMD) [47]. The HMD is a device that is worn on the head that has an optic display for the eyes (Figure 2).

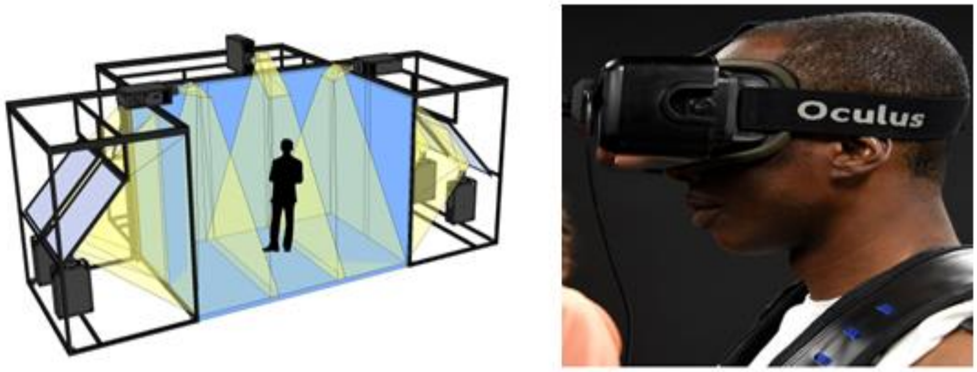

Figure 2 An example of the CAVE virtual reality system 48 (on the left) and the head mounted display (on the right)

Introducing a visual flow scene with rotations around the anteroposterior (roll), mediolateral (pitch), and vertical (yaw) axes (Figure 3) affects the motor behavior of adults, with the greatest on behavior observed when the visual stimulus was presented about the roll axis [48]. As noted before, older adults tend to be more visually dependent compared to younger adults [18,19]. The reliance on visual information, specifically the peripheral vision, increases under faulty somatosensory system inputs [49]. Visual information manipulation using VR can be used to systematically determine the impact of visual status on postural control. This study used VR combined with the TUG to determine the effects of visual environment on motor performance in younger and older adults.

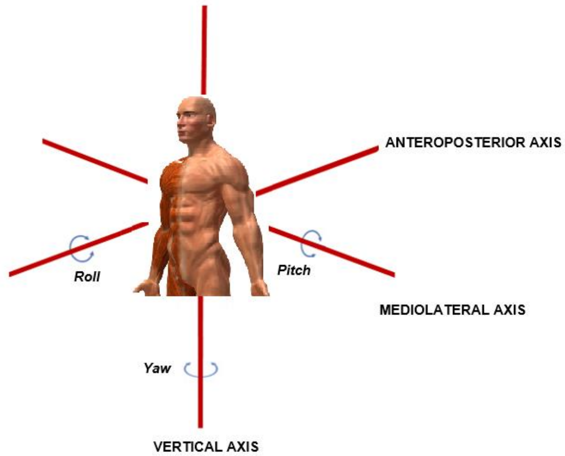

Figure 3 Pitch, yaw, and roll rotations

Virtual Reality and Motor Behavior

Motor tasks, such as holding a glass of water, require accurate eye-hand coordination, especially when performed during locomotion. Eye-hand coordination is slowed with advanced aging [50]. In studies that explored the effects of aging on eye-hand coordination, older adults showed a delay in saccades initiation, increase in the duration of the gaze on the target, and decrease in eye and hand movement coordination under various manual task constraints in comparison to younger adults [51–53]. In an adaptive locomotor task where the adults were asked to walk in a nine-meter pathway while stepping within target locations placed three steps apart, older adults looked sooner and fixated longer on the target when compared to younger adults. Adult fallers prioritize planning movement over execution; thus, they demonstrated an increase in errors and variability in foot placement [50,54]. Given the difficulty that older adults can experience while performing attention-demanding tasks that require visual-motor coordination, it is reasonable to assume that performing a manual task, e.g., holding a cup of water, may create more challenges to plan and execute the movement in older versus younger adults.

When an individual performs an automatic motor task such as reaching or holding a cup, the nervous system uses an internal representation of the body to compensate for errors to enhance the success of the motor task. For example in reaching, the representation model of the body includes internal model of the load, e.g. influence of gravity, forces, and internal factors such as Coriolis. Using the inputs from sensory systems, the body creates a sensed motion of the reach, which helps in creating a desired body and arm trajectory. Then, the body generates a Coriolis force based on the sensed motion of the reach and the planned motion of the reach. Together, the desired body and arm trajectory and the generated Coriolis force help in executing reaching through muscle activation [55].

Furthermore, the automatic motor behavior can be affected in immersive dynamic visual environments. In reaching tasks, exposure to a rotating virtual visual scene through an HMD increased the path deviation and endpoints [55]. Wright and Glasauer [56] and Wright et al. [57] showed that the automaticity of holding a glass of water can be influenced by the perception of self-orientation and motion in a VR environment. Perception of verticality of the person, which was determined by aligning the glass of water or a joystick with gravity, automatically improves if the task is more often used in real life circumstances. This was evident in a study by Wright and Glasauer that showed an improvement in performance when holding a glass of water, which is a task often used in daily activity, compared to holding a metal joystick. The same group showed similar results in two experiments where the visual manipulation was coupled with vestibular manipulation using a tilting chair. Holding a glass of water that has similar inertial properties to a metal joystick has been shown to improve the ability to estimate the subjective vertical of the adults [56,57]. However, when there was no manipulation of the vestibular and somatosensory systems, the perception of verticality did not affect the motor responses similarly independent of object type (glass of water or a metal cylinder) [58].

Exploring the motor responses of older adults while performing a motor task in a VR environment is an important adjunct to understanding the functionality of those who are visually dependent under faulty visual inputs. Older adults with impairments in sensory systems and multi-sensory integration may exhibit deterioration in movement when performing this common task in visually busy environments. This project will address this question, which may lead to an early and better assessment and intervention in rehabilitation setting. The following section will focus on discussing the biomechanics of each TUG subcomponent.

TUG Sub-components Biomechanics

Sit-to-stand

Rising up from a chair, i.e. sit-to-stand, is a common daily activity that depends heavily on concentric contraction of the knee and hip extensors, and the ankle dorsiflexors to accelerate the body upward. In addition to lower limb strength, sit-to-stand relies on multiple physiological and psychological factors such as lower limb proprioception, postural sway, and visual contrast sensitivity [59]. The biomechanics of sit-to-stand consist of three phases [60]: (1) Phase I: start of lift off, (2) Phase II: during lift off, and (3) Phase III: end of lift off. Phase I includes generation of the flexion momentum of the trunk and creation of upper body momentum to initiate the sit-to-stand motion. Some of scholars further analyzed Phase II, i.e. during lift-off, as two phases: (a) the momentum transfer phase where the individual relies on the dynamic stability to translate the body and center of mass (COM) anteriorly and upward and (b) the extension phase, the lift up from the chair seat begins. In Phase III, the end of lift off, the hip extension ends as the body moves into a typical standing position [61] (Figure 4).

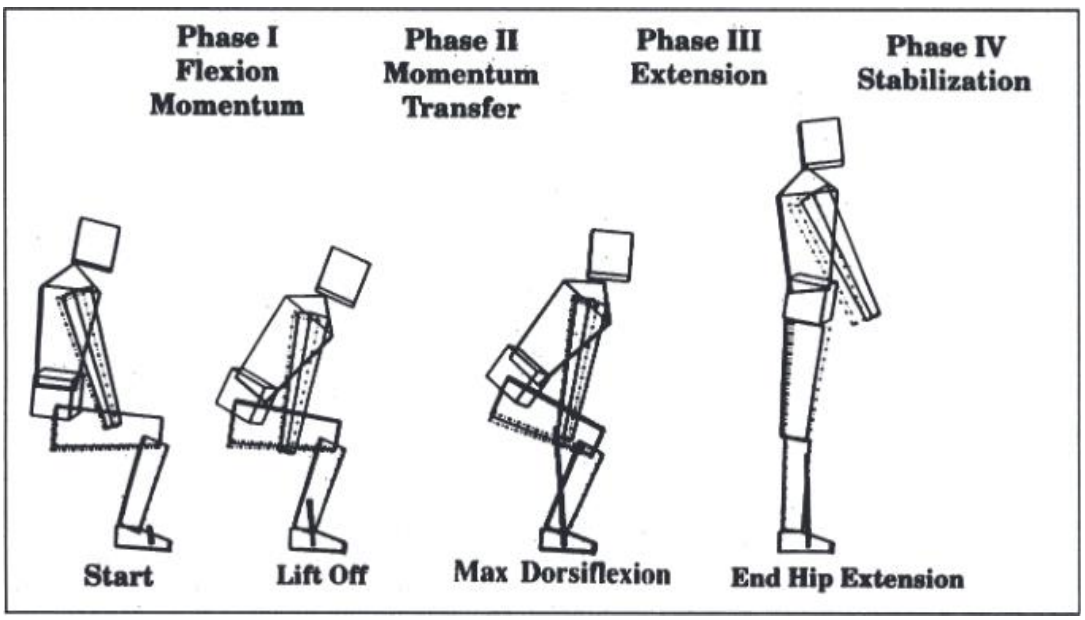

Figure 4 The four phases of sit-to-stand as described by Schenkman et al 62

Sit-to-stand is a complex activity that can be influenced by many factors, such as the chair properties and feet position. Using an armchair reduces the knee and hip moments [62], and pushing on the armrest decreases the knee joint forces to less than three times the body weight in comparison to seven times of the body weight that are produced when the armrest is not used [63]. Additionally, the motor behavior in sit-to-stand differs depending on the type of the chair. Wheeler et al. [64] used two types of chairs: a standard chair and a special chair, which were similar in the width between the armrest, height, and width; however, the special chair was wider, had a longer posterior slant, shorter armrests, more backrest incline, and was lower than the standard chair. Older adults showed increased activation in the vastus lateralis muscle when rising from the special chair, which reflects the difficulty of the task compared to the standard chair [64]. Additionally, using a lower chair creates greater biomechanical challenges in sit-to-stand than a higher chair especially for older adults [63,65]. A variation of this pattern has been observed in stroke survivors, in which the height of the chair did not affect the performance [66], which could be attributed to the high variability in the functional levels of their small sample size (n=12). These findings suggest that the type of chair should be taken into consideration when testing older adults in the TUG test. This study will use an armless chair to accurately capture the challenges that the elderly may face in rising from a chair during daily activities, especially under misleading visual inputs.

Another factor that affects sit-to-stand motion is the foot position. Positioning the feet forward creates more postural challenges than backward positioning [67,68]. When positioning the feet forward in phase I, the activities of the body start at trunk extension, followed by hip extension in phase II then ending up with knee extension in phase III; whereas in backward feet positioning in phase I, the body starts at knee extension in phase II followed by trunk then hip extension in phase III[67]. The foot forward position also requires an increase in the amplitude and peak velocities of angular velocities in the hip, muscular activation of the knee extensors and ankle dorsiflexors, forward and backward acceleration ground reaction force (GRF), and maximum vertical GRF [67,68]. Thus, a forward foot position exerts high biomechanical demands on postural stability in sit-to-stand transition.

Older adults use distinct movement strategies to attain sit-to-stand movement. In a sample of twenty-two elderly people [65], three strategies in sit-to-stand were identified: momentum transfer, stabilization, and a combination of both. Eleven older adults used a momentum strategy, which is considered a control strategy. This momentum strategy group took a shorter time (4 sec) and stood in one continuous motion by using a horizontal momentum of the trunk and knee extensor muscles during the phase II and III. Four older adults used a stabilization strategy, which is considered an efficiency strategy. This group took a longer time (8.5 sec) and moved their foot backward and buttock forward to attain a slow forward motion of the trunk, thus, relying less on the horizontal trunk momentum and more on knee extensor muscles. Lastly, five adults used a combination of both strategies at one time. Interestingly, even when changing the height of the chair, the elderly utilized the same strategy in completing the sit-to-stand movement.

In addition, differences are noted in how older adults execute sit-to-stand in comparison to younger adults due to possible strength and stability impairments. When standing up from an armless chair, older adults rotate their upper body, legs, and thighs more than the younger adults do [69]. During phase I and II, an increased hip and lumbar flexion and thoracic extension are observed [60,70]. However, compared to younger adults, elders who positioned their feet backward, were slower, showed increased trunk flexion, and increased the peak and average activities of the vastus lateralis muscle [64,71,72]. Based on these earlier studies, older adults start sit-to-stand movement with more thoracic kyphosis and a tighter lumbar spine than the younger adults. At Phase I, older adults positioned their COM posteriorly to establish a more stabilized posture. Thus, older adults increased the trunk to pelvis flexion, which can create a difficulty in getting up from a chair. Also, the elderly exhibited an increase in head to trunk maximum angles due to a decrease in their ROM, which may further compromise their postural stability by limiting their ability to process the visual inputs and decrease the cervical proprioception [73]. In Phase II, a decrease in hip and lumbar flexion and an increase in thoracic extension were observed [70]. To create greater stabilization during Phase II, older adults start moving their COM over their BOS by rotating the body forward [71], which is similar to previously reported findings [69]. Also, they show a decrease in head tilt and lumbar ROM [71]. These findings suggest that older adults prefer to use different movement patterns than younger adults when executing sit-to-stand transition, possibly due to insufficient proximal muscular strength, limited knee ROM, or joint pain which are common in older adults and linked to increased challenges in sit-to-stand motion [74,75].

Stand-to-sit

The ability to move from standing to sitting down onto a chair requires different muscle activation and biomechanics than sit-to-stand. Generally, stand-to-sit begins with a vertical deceleration of the trunk followed by a stooping posture. While the body weight transfers backward to the chair, knee and hip joints move into a flexion position until the hip reaches the maximum flexion angle, which marks the end of stand-to-sit [76,77].

The time taken to complete the stand-to-sit motion does not necessarily reflect the functional levels of the performer. In one study that involved twenty healthy adults (mean age= 32.6 years), it was noted that the participants took a longer time to complete stand-to-sit (4.62 sec) than sit-to-stand (3.33 sec) [78]. One possible reason is the need to use the visual inputs in stand-to-sit to locate the endpoint, the chair, before sitting down. However, this was not consistent with other findings. In healthy adults, Ker et al. did not find any significant difference in the completion time between sit-to-stand and stand-to-sit. In terms of functional level, stand-to-sit did not characterize the quality of performance in stroke survivors, which may suggest that this transition reflects the muscle coordination more than muscle strength, due to the higher dependency on visual inputs to successfully locate the chair prior to sitting down [78].

Compared to sit-to-stand literature, studies that examine the effects of aging on stand-to-sit transition are limited. In a sample of frail older adults, it has been shown that sitting back on a chair is more challenging than standing up, due to increased dependency on visual information and estimation of the end point, with the more frail adults demonstrating a decrease in trunk angle while sitting down [79]. Additionally, Dubost et al.[80] showed that older adults exhibit a decrease in maximal trunk orientation angle when sitting down in stand-to-sit, but not in maximal shank orientation angle compared to healthy young adults. Few studies have focused on the biomechanical differences in older and younger adults when completing stand-to-sit [79,80]. Therefore, studying the biomechanics and postural responses in older adults performing stand-to-sit under misleading visual information will help in understanding multisensory integration mechanisms. The improved understanding of multi-sensory integration during more challenging postural tasks could be used to develop a better assessment and training programs to adults with balance dysfunction.

Walking

Walking is a common motor act that involves rhythmic muscular activation of the legs and arms, and the efficiency and coordination of walking strategies depend on the environmental characteristics. Generally, aging leads to a decrease in step length, gait speed, symmetry, ankle power generation, ROM, and mechanical joint power and work in the lower limbs. Older adults show an increase in the step time, step width, anterior pelvic tilt, hip extension moment during swing phase, and mechanical energy demands of lower extremity muscles [81–83]. Older adults generate less ankle power than younger adults walking at both slow and fast speeds. When walking at comfortable, self-selected speed, older adults exhibit a lower peak knee flexion moment at midstance and increased peak knee power absorption at midstance than younger adults [82]. Strength limitations in older adults has been linked to loss of balance and falls [33]. In addition, elderly exhibited alteration in sensory functions [10,11,34], thus, a better understanding of how the functional decline and multisensory integration in old age may lead to postural instabilities during locomotion is needed.

Turning

Turning is a complex activity that requires a change in the step length and ground reaction force (GRF) to direct the COM in a new direction. It involves modifications in the anteroposterior (AP) and mediolateral (ML) postural reactions to slow down walking in the sagittal plane, which allows a successful turn [84]. In straight walking, the COM oscillates between the feet in a sinusoidal pattern as the ML impulse accelerates the body toward the opposite leg. In curved walking, individuals tend to decrease their walking velocity to achieve successful motion.

Older adults utilize different strategies in walking turns than younger adults. A comparison of gait velocity in a sample of 16 young adults (21-25 years) and 17 old adults (60-80 years) while circular walking demonstrated minimal differences between the two groups; however, differences were noted in patterns of muscular activation. Specifically, there were decreases in EMG amplitude in the biceps femoris and rectus femoris in the outer leg and tibialis anterior of the inner leg in younger adults, whereas older adults showed an increase in the EMG amplitude of the tibialis anterior in the outer leg and the biceps femoris in the inner leg [85]. Thigpen et al.[86] studied 20 healthy young adults, 15 older adults with no difficulty in turning, and 15 older adults with difficulty in turning with no reported musculoskeletal or neurological disorders. They found that young adults and most (58%) older adults who do not have difficulty in turning used a spin turn to complete the 180° turning subtask in the TUG test. The remaining older adults who did not report a difficulty in turning (42%) achieved this task by using a mixed strategy of a spin turn and increased step frequency. Older adults who reported having a difficulty did not use a spin turn in turning rather they either took five or more steps or used a weight shift over a longer time (three seconds or more) to achieve a successful turn [86]. This suggests that older adults may be more impaired in walking turns and face greater challenges with strength, coordination, and balance to successfully navigate their environment.

TUG Activities, Multitasking, and Aging

Although multitasking, the skill of maintaining postural balance and performing secondary tasks at the same time, seems to be automatic, studies have shown that attention impacts the biomechanics in static and dynamic postural control during dual-task performance [87], and may increase fall risk [12]. Multitasking requires diverting attention to a secondary task, which often interferes with primary task performance when the information processing capacity of the individual is exceeded [88].

Adding a secondary task to sit-to-stand and stand-to-sit transitions can create more challenges to the performance of adults. This was evident in a study by Teasdale et al. [89] that tested a sample of adults in standing position after adding an auditory reaction time task while changing the surface and/or vision. They found that as the postural task complexity increases, both young and old adults become more affected, with older adults requiring more attentional resources, as they showed greater reaction times in the secondary task. Similarly, Maylor and Wing [90] found an aging effect on the postural stability in the standing position on a sample of younger and older adults. In general, DT standing is a more challenging posture than DT sitting. This was shown in the findings from Lajoie et al. [91], which concluded that as the postural task complexity increases from sitting to standing to walking, the attentional resources of the older adults decreased and it became evident in their motor acts. Older adults showed a decrease in the reaction time task when they stood with a narrow BOS and walked slower with a shorter stride length compared to younger adults. Additionally, it appears that the types of the cognitive tasks matter when testing the standing postural balance. For instance, the auditory reaction time tasks had a similar effect on younger and older adults due to its simplicity [91], whereas tasks that require visual-spatial ability and working memory have varied effects on both age groups [87,90], with the older adults being more affected than younger adults. It is not surprising that the visual tasks are more demanding than the auditory tasks since the postural balance depends heavily on the visual system [24,25].

Sit-to-Stand and Stand-to-Sit Transitions

Sit-to-stand and stand-to-sit transitions can be impacted during visually demanding tasks. [22] Slaboda et al. tested young adults in a virtual environment of a rotating visual scene around the pitch or roll axes while standing from a seated position. [19] The direction of the visual scene rotation affected the motor responses similarly, whereas the timing of the optic field showed a different effect on these responses. Specifically, adults tend to show a poorer performance in sitting to standing after a period of 10 sec of immersion in the visual environment prior to transitioning to stand in comparison to the condition when the scene rotates at the onset and continues concurrently with the sit-to-stand transition. To my knowledge, there is a gap in the literature exploring the impact of visual stimulation in neurologically intact and impaired older adults while performing sit-to-stand and stand-to-sit transitions.

Walking

Changes in gait parameters are documented in multitasking conditions. Specifically, gait speed and cadence are found to be two of the most sensitive measures when completing a secondary task while walking [16]. Under multitasking conditions, older adults exhibit a slow gait speed, increase in stride time variability, and a decrease of the ML distance between the sacral marker and enter of pressure (COP) compared to younger adults [92,93]. Similarly, when performing a TUG test, older fallers were slower compared to older non-fallers and younger adults [94]. Srygly et al. [95] found an increase in the reaction time and the number of errors in a mental calculation task when combined with locomotion in older adults in comparison to younger adults. Conversely, another author found no significant difference between single and dual tasking conditions when the healthy elders walked while stepping over an obstacle [96]. Results from these studies suggest that adding a secondary task to the locomotion can greatly influence the postural stability, and thus may contribute to loss of balance and the increased risk of fall of older adults.

Visual flow manipulation can affect walking strategies. Presenting an immersive dynamic visual flow at constant velocity influences the locomotion patterns [21]. A tendency to shorten step lengths and increase ankle plantar flexion is observed in walking in an immersive dynamic visual flow[21]. The speed of the visual flow manipulation could affect the gait speed in stroke survivors. Using a faster optical flow speed resulted in slower gait speed and vice versa. However, no study has examined the biomechanical differences between young and old adults who are visually dependent under visual manipulation conditions in a common motor act such as walking. Examination of gait changes in healthy older adults coupled with functional task performance in a visually busy environment could improve our understanding of motor behaviors in real life situations, e.g. walking through a crowded room with a drink.

Turning During Walking

Adding a secondary task to turning during walking requires more attentional resources to maintain postural control [12]. Given that turning during ambulation is responsible for eight times the number of hip fractures than occur during straight ambulation [97], it is imperative to explore the motor behaviors of older adults to better understand increased falls are associated with turning during walking. In a cross-sectional study with a total of 370 subjects (n=27 fallers (self-reported 2 or more falls), n=68 who self-reported one fall in the previous year, and n=265 non-fallers older adults), Faulkner et al. [98] found decrements in performance during turning that was associated with higher odds of recurrent history of fall. Specifically, older adults showed a poorer performance while dual tasking during turning more than while straight ambulation, suggesting that turning may involve more cognitive resources to maintain postural balance than regular walking. In another study by Bootsma-van der Wiel [99], a sample of older adults with high and low risk of fall were asked to complete a walking task that included three 180° turns while performing a secondary task. The authors reported that the occurrence of falling did not significantly differ from those who were able to dual-task successfully (42% of the participants) from those who did not (40% of the participants).

Vision plays an important role in turning activity with impairments of central and peripheral visual field having differential impacts on turning. Few studies have looked into the contribution of the visual inputs in turning. One study explored steering in walking when the eyes were open and closed [100]. They found that the eye and head control did not significantly differ in healthy young adults with eyes were open and closed [100]. When the visual inputs were not available peripherally, adults showed a motor incoordination in the form of increasing the variability of segment rotation onset duration [101]. Under the occlusion of the central visual information, there was a tendency to delay the segment orientation in the rotation [101]. The onset of rotation occurred approximately 200 ms later than the condition when the visual information was available [101].

The present study used a head mounted display (HMD) — the Oculus Rift development Kit 2 (Oculus VR, 2014 ™) to present a visual stimulus of bright dots (i.e. snowflakes) throughout the activities of the TUG test in groups of older and young adults further classified as visually dependent (VD) or visually independent (VI). Visual dependency is a term used to describe those who make an inaccurate estimation of verticality within their environment. A system of inertial measurement units (IMU) — Trigno™ wireless sensors (Delsys Inc.) were used to define the motor sub-tasks of the TUG, and to measure the trunk biomechanics of the movement during the TUG during a single session. The general hypotheses guiding this work were that older adults were expected to show poorer performance when compared to younger adults with the VD adult being more affected negatively by the visual stimulation more than VI adults.

Instrumentation

Head Mounted Display (HMD) - The Oculus Rift Development Kit 2

Head mounted displays (HMDs) are more portable and less expensive, but have a limited field of view compared to large, earth-fixed screen immersive VR systems. The Oculus Rift Development Kit 2 has a visual scene refresh rate of 75 Hz and a head tracking system composed of a 3-axis accelerometer, gyroscope, and magnetometer sensors with 1000 Hz update rate. This HMD has a latency of 20ms-40ms, between the actual head motion of the user and presentation of the updated scene to the subject (Oculus Rift Specs-DK1 vs DK2 comparison, 1 June 2016)[102]. Compared to humans' wide visual field (i.e. 200° horizontal by 130° vertical with the binocular vision occupying the overlapping central area of 120°), the Oculus Rift 2 HMD offers a limited diagonal visual field of 100° (Oculus Rift Specs-DK1 vs DK2 comparison, 1 June 2016) [102].. This visual field limitation in the Oculus Rift restricts peripheral vision and consequently users rely more on their central vision, looking more in a forward direction and that may challenge navigation during locomotion [103]. Though HMDs have been used by several studies to examine the effects of visual attention on gait kinematics, there are no reports to my knowledge, of the effects of simply wearing the HMD without any VR displayed.

Inertial Measurement Unit Devices

Inertial measurement unit devices are typically body mounted sensors used to capture body kinematics, specifically linear acceleration, angular velocities, and spatial orientations [104]. Specifically, IMUs are consist of an accelerometer and gyroscope, and may include a magnetometer. When compared to optical and camera-based motion capture systems, IMUs can provide similar types of kinematic data and are less expensive, and more portable [105]. However, IMUs may provide less accurate position data, and are more prone to measurement errors and drift [106–108].

Accelerometers are electromechanical devices that measure the acceleration of the body segments. A variety of transducers are used to measure the acceleration, e.g. piezoresistive sensors or electronic piezoelectric sensors; however, in most cases, the accelerometer uses a spring-mass system. [109]. Piezoresistive accelerometers are commonly used to analyze human movements. The resistive and capacitive accelerometers take into account the inertial (dynamic) and gravitational (static) accelerations. Therefore, when a body segment is motionless, the acceleration signal represents the gravitational acceleration that depends on the orientation of the sensors with respect to the gravitational field, whereas when the body moves, the signal combines the gravitational acceleration with the actual body segment movement [106,107]. The acceleration signal is subject to errors and noise, especially at the highest speeds. Measuring inertial and gravitational accelerations has its advantages and disadvantages. The inclination of the motionless body segments, e.g. trunk or pelvic, can be obtained [106,107]; however, a special consideration should be taken in computation to separate both types of acceleration when movement is the primary interest [108].

Gyroscopes measure the angular velocity of body motion. Gyroscope measures are based on the Coriolis force that is proportional to the rate of the angular rotation of the object and converts the rotatory motion of the object into a measurable linear motion. Gyroscopes may include transducers such as the piezoelectric, capacitive, and resistive sensors [110]. The gyroscope is considered a newer tool than the accelerometer in analyzing human movements and has several advantages. First, forces due to gravitational acceleration are not measured; thus, the interpretation of the signal is clearer. Second, the gyroscope signal has less noise compared to the accelerometer signal, which is the derivative of velocity that has high-frequency elements [108]. Third, the output signal from the gyroscope is almost identical, regardless of sensor's placement on the body [111]. However, the accelerometer has some advantages compared to the gyroscope such as the lower price, less power consumption, and less sensitivity to shock and drift. The drift of the gyroscope can be decreased when the signals from the accelerometer, gyroscope, and magnetometer are merged as done in IMUs[108].

The magnetometer is used to measure the orientation of the body segment relative to the direction of motion and provides correction information for the drift in the gyroscope signal [110]. The obtained angles from the magnetometer can be useful to separate the kinematics from the gravitational acceleration signal; thus, it can help in a better understanding of human motion [112]. However, the magnetometer has some disadvantages. For instance, the inaccurate measurements of the angles can occur when a magnetic source influences the Earth's magnetic field, which can be categorized into soft and hard iron effects [112]. The hard iron effect occurs due to the permanent magnetic objects that exist in a fixed location in relation to the sensors and can be adjusted with autocalibration methods [113], whereas the soft iron effect occurs due to the interaction between the earth's magnetic field and any soft magnetic object that exists in the same location of the sensors.

In this study, the auto calibrated Trigno™ wireless motion sensors (Delsys Inc.) were used to capture the movement of the participants. The sensors include a tri-axial accelerometer (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 3.5 mg); a tri-axial gyroscope (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 0.05°/sec); and a tri-axial magnetometer (range 40 m, resolution 16 bit, sampling frequency 74 s/sec, Noise < 0.4 uT). Each sensor weight was 14.7 g. The signals obtained from the gyroscope will be of particular interest to this study.

IMU and Motor Behavior

Body mounted sensors have been used extensively in the literature to characterize the motor behavior of older adults. The analysis of the accelerometer and gyroscope signals was helpful in identifying the kinematic differences of neurologically impaired and intact populations (e.g. [113,114]), especially when compared to the standard TUG test score based only on the total time [12].

Body-mounted sensors can be useful in characterizing the functional levels of neurologically impaired populations in research settings. Salarian et al.[114] and Zampieri et al. [115] used several sensors (five or seven sensors) to the forearms, shanks, thighs, and sternum to obtain kinematic data during the TUG test to detect additional movement information in adults with Parkinson's disease (PD). Both groups, healthy controls and adults with PD, varied significantly in specific TUG sub-components such as gait, turn, and turn to sit. In particular, peak arm swing and turning velocity, 180° turning and turning to sitting time, cadence, angular velocity of arm swing, and peak trunk rotation velocity were significantly different between people with PD and healthy older adults [114,115]. In addition, another study used six IMUs (placed above both left/right malleoli, left/right wrists, sternum, and lumbar region), and found adults with multiple sclerosis had increased turning time and trunk angular ROM in the roll and yaw axes compared to healthy adults [116]. A variety of studies used the acceleration and the angular velocity signals in identifying the functional levels of people with neurological injuries [114–116]; while others only relied on the acceleration signals [117–120]. For instance, Weiss et al. [120] found that the signal from a 3D accelerometer placed the lower back sensor was useful in revealing some variables that are more sensitive to change in people with PD such as the range of the sit-to-stand, stand-to-sit and the jerk of the sit-to-stand and the acceleration standard deviation[120]. Another study used acceleration measures from two accelerometers (lumbar and leg) during the TUG to differentiate functional gait levels of stroke survivors[118]. Specifically, those who could walk independently differed from those who required supervision while walking in the acceleration signal in the walking phase of the TUG test[118]. In people with neuropathy, an accelerometer (sternum) was able to differentiate fallers from non-fallers (based on Tinetti scores) as fallers took longer time in the stand to sit TUG phase [119]. Gillian et al. [117] determined group difference between adults with and without mild cognitive impairments. The acceleration signal was sufficient to differentiate between healthy adults, people with mild cognitive impairment, and those with Alzheimer disease. These studies suggest that an accurate assessment of the functional level of adults with neurologically impairment can be obtained by using the acceleration and angular velocity signals of the IMU sensors during the TUG.

Several studies show that analyses of kinematic variables from IMUs during the performance of TUG sub-components were sufficient to differentiate between fallers and non-fallers. For example, analyses of kinematic data from the accelerometer and gyroscope was able to increase the sensitivity and specificity of the TUG test (sensitivity 77.3%, specificity 75.9%) when compared to the standard TUG and Berg Balance Scale (BBS) retrospectively (sensitivity 58.0% and 57.8%, specificity 64.8% and 64.2% respectively) [121], and increased the predictive accuracy of the test to 80% in identifying the risk of fall [122] in comparison to 57%, which is the predictive accuracy of the standard TUG [123]. Similarly, Tanaka et al. [124] have used one inertial sensor on the waist to analyze the acceleration and angular velocity signals during the TUG on elders who are at high and low risk of falling. They identified the risk of falling based on the cut-off point (13.5 seconds) that was suggested by Shumway-Cook et al. [88] and concluded that the most differentiating measure between both groups was the turning angular velocity signal followed by the cadence of turning. While some studies identified the fall risk in older adults by integrating both the acceleration and angular velocity signals [121,122], Weiss et al. [125]analyzed the acceleration signal strictly for the purpose of characterizing the fallers from non-fallers. Using accelerometer derived parameters, the jerk of sit-to-stand, the standard deviation, and the average step duration, increased the clinical utility of the TUG test in identifying fallers retrospectively to 87% compared to the standard TUG, which detected 63% of these individuals. Older adults who are at high risk of falls took a longer time to complete the TUG, had a lower acceleration range, and increased jerk in the sit to stand and stand to sit transitions compared to non-fallers [125]. Thus, ample evidence supports the use of body mounted sensors, particularly IMUs, as a useful tool that can capture the biomechanical differences in identifying older adults who are at high risk of fall in comparison to the standard TUG test.

SUMMARY

Older adults use different movement strategies when performing common daily activities when compared to younger adults. Activities such as sitting to standing and turning are linked to an increased risk of fall in older adults [126,127]. Furthermore, older adults react poorly to multitasking conditions and are more visually dependent than younger adults [18,19,87].

TUG test is composed of several functional activities and is a commonly used clinical test of functional balance. Performance on the TUG in a virtual environment with visual stimulation in combination with dual-tasking (motor) could provide additional challenges and better elucidate changes in motor behavior between younger and older adults with and without visual dependence.

PROJECT OBJECTIVES

The TUG test is composed of a series of functional skills and is used to assess the risk of fall in older adults. To date, few studies have focused on exploring the effects of adding attentional demands to the TUG sub-components in young and old adults [20,117,128]. In addition, the role of processing the visual inputs in visually dependent individuals in various types of daily activities is not well-specified in the literature. It is important to understand how older adults with increased reliance on vision more than other sensory systems [18,19], less efficiency in sensory weighting under visually manipulated conditions [41], and difficulty to multitask [87] will perform under high motor and visually-demanding conditions.

Therefore, the overall goal of this project is to understand the effects of aging and visual dependency on the postural responses that occur with normal functional activities that are part of the TUG test and that could be linked to loss of balance and falls. This project will examine kinematic properties of the motor behavior of younger and older adults who are visually dependent and independent during attentionally demanding conditions while performing the TUG, a clinical test that includes functional skills. This study will be the first to consider how dynamic visual inputs affect the movement strategies during the TUG test.

SPECIFIC AIMS

Specific Aim I. To determine the influence of wearing an HMD with and without presentation of augmented virtual stimulation during the TUG activities in old and young adults.

Hypotheses 1.1: Visual stimulation: Adults will show a decrease in the turning cadence, gait speed, peak trunk velocity (PTV) in all TUG activities, acceleration range and jerk of sit-to-stand and to-sand-to-sit when wearing HMD with visual stimulation compared to wearing HMD without visual stimulus.

Hypotheses 1.2: Aging: Older adults will exhibit a decrease in all dependent variables (mentioned in hypothesis 1.1) compared to younger adults

Hypotheses 1.3: Interaction: When compared to younger adults, older adults will show a decrease in all the dependent variables (mentioned in hypothesis 1.1) when wearing the HMD with visual stimulus compared to wearing the HMD without visual stimulus.

Specific Aim II. To explore how aging affects the multitask cost of the time of TUG and its activities when combined with a motor task and/or attending to a visual stimulus through an HMD.

Hypothesis 2.1: Tasks: all adults will exhibit a decrease in the multitask cost of the time of the TUG and its activities when wearing a head mounted display (HMD) compared to *the motor task*

Hypothesis 2.2: Aging: *older adults* will exhibit a decrease in the multitask cost of the time of the TUG and its activities more than *younger adults.*

Hypothesis 2.3: interaction: when compared to younger adults, older adults will show a decrease in the multitask cost of the timing of the TUG and its activities when wearing a head mounted display (HMD) compared to *the motor task.*

Specific Aim III. To explore the effects of aging and visual dependency on motor performance under increased stabilization demands by adding a motor task and a visual stimulus to TUG sub-components.

Hypotheses 3.1a: Tasks: All adults will exhibit a decrease in the turning cadence, gait speed, PTV in all TUG activities, acceleration range and jerk of sit to stand when wearing the HMD compared to the motor task.

3.1 b. Tasks: All adults will exhibit a decrease in the acceleration jerk of the cup of water used in the motor task when wearing the HMD compared to not wearing the HMD.

Hypotheses 3.2: Aging: older adults will exhibit a decrease in all dependent variables (mentioned in hypothesis 3.2a and 3.2) compared to younger adults.

Hypotheses 3.3: Visual dependency: VD older adults will exhibit a decrease in the dependent variables (mentioned in hypothesis 3.2a and 3.2) compared to VI older adults.

CHAPTER 2

EFFECTS OF HEAD MOUNTED DISPLAY ON KINEMATICS OF OLDER AND YOUNGER ADULTS: DOES THE ADDITION OF A VISUAL STIMULUS MATTER?

This chapter will discuss the effects of wearing a head mounted display (HMD) on the performance metric of the TUG and its components with and without the addition of a rotated visual stimulus in a group of young and old adults.

Introduction

Vision is the predominant system in maintaining balance [24,25], especially with advanced aging [18], and has a tremendous impact on postural control. To understand how vision contributes to posture, virtual reality (VR) was introduced to rehabilitation research. In particular, HMDs showed to be promising tools for clinical use because of their portability and programming ease that offers a great immersion experience. These tools can offer a variable, motivating, yet safe environment to assess and treat different populations [47,129]. Some concerns have been reported on the effects of HMDs on vision and posture. These include, but are not limited to, subjective discomfort of the users [1301] and modifiable neck and trunk postures [131], which could occur due to several reasons such as the added weight on the head, field of view restriction, and unfamiliarity in using the device [132]. Given all of these factors, a question that is usually ignored in the literature is whether the postural reactions are caused by the physical properties of the HMDs or the visual imposing stimulation that is introduced by the HMDs. This is specifically important to take into account when the HMDs are used with older adults, who show a decrease in visual function [7,8,10,11] and postural control [19,44].

The questions that this work is going to address are 1) does adding the augmented visual stimulus to the HMD affect the motor performance more than wearing the HMD without a visual stimulus, and 2) is there an age effect or interaction. The hypotheses are: 1) wearing the HMD with visual stimulation will cause decrements in the dependent variables which are: turning cadence, gait speed, and peak trunk velocity (PTV), acceleration range and jerk of sit-to-stand and stand-to-sit, 2) older adults will show decrement in the dependent variables more than younger adults especially when wearing the HMD with a visual stimulus.

Methods

Participants

Twelve younger adults (6 males) and sixteen older adults (8 males) gave informed consent to participate in this study approved by Temple University's Institutional Review Board (IRB). Inclusion criteria were as follows: (a) Age between 21-40 (young adult) or 60-90 years old (older adult), (b) Ability to walk independently, and (c) No neurological or cognitive dysfunction.

To assess the balance and cognitive abilities of the participants, they underwent the following functional and balance clinical measures prior to running the protocol: 1) Activity-specific Balance Confidence (ABC) scale, a self-reported measure of the participants' confidence in performing activities [133]; 2) the Berg Balance Scale (BBS), an assessment of static balance and fall risk[133]; 3) Functional Reach Test (FRT), a measure of postural stability[134]; and 4) the Timed Up and Go test (TUG) to assess the dynamic balance and mobility in the elderly[135]. The Mini-Mental State Examination (MMSE) was administered to determine cognitive function[136].

Experimental Protocol

All adults wore comfortable shoes during the TUG. Each participant had a practice trial of the TUG followed by the experimental conditions. In accordance with the standard TUG test instructions, when the command “go” was given, participants were asked to rise up from a chair, walk 3 meters at a comfortable speed, turn around 180°, walk back to the chair, and sit down [15]. The chair was armless and had a dimension of 44 cm (height) by 46 cm (width). A line of tape and a cone were placed on the floor to indicate where the adults should turn. (Figure 5)

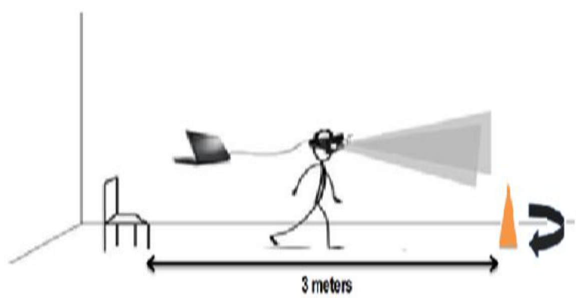

Figure 5 The Timed Up and Go Test Experimental Setting

The HMD that was used in this study was the Oculus Rift Development Kit 2 (Oculus VR, 2014 ™). The augmented reality scene included random dots of virtual bright dots (i.e. snowflakes) (Figure 6) that rotated at a constant speed (5 °/sec) in the pitch up (PU) or down (PD) directions. Participants were asked to start the TUG test after 10 seconds of visual scene motion. We choose to start the TUG after approximately 10 seconds because it was shown that the motor responses' of adults were more affected by the virtual environment rotation when they were asked to perform the movement after an exposure to the scene motion of 10 seconds [19]. The Oculus Rift consists of two displays, one for each eye, with a resolution of 960 x1080 pixels per eye, a maximum refresh of 75 Hz, and a weight of 440 grams. Those with eyeglasses could wear them while wearing the HMD. The viewing optics allows for a 100° viewing angle. An Ovrvision mount was used, which is a high-performance USB stereo camera that does not affect the size of the visual world, customized for the Oculus Rift, which allows users to replace their current display with a view of the real world. The resolution for the Ovrvision is 640 x 480 per eye (1280 x 480), the frame rate is 60 FPS, the angle of view is H90 °, V75 °, the latency is 50 msec, the pixel number is 0.6 MP, and its weight is 55 grams.

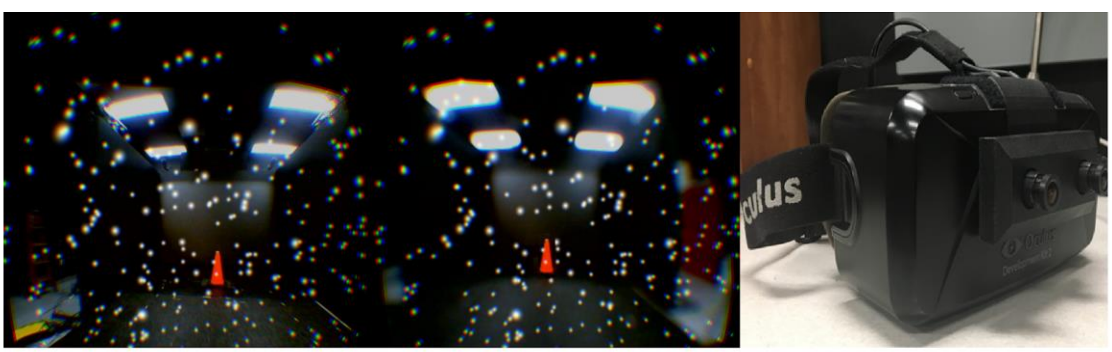

Figure 6 The Oculus Rift (on the right side) and the presented augmented virtual scene through it (on the left side)

Motor performance was measured using Trigno™ wireless sensors (Delsys Inc.). Sensors were placed on the participants' sternum, lumbar, both wrists, and both shanks. The sensors include a tri-axial accelerometer (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 3.5 mg); a tri-axial gyroscope (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 0.05°/sec); and a tri-axial magnetometer (range 40 m, resolution 16 bit, sampling frequency 74 s/sec, Noise < 0.4 uT). The obtained signals from body motion during the TUG test include three acceleration axes: anterior-posterior (AP) acceleration, vertical acceleration, medio-lateral (ML) acceleration, and three angular velocity axes: pitch (rotation around the ML axis), yaw (rotation around the vertical axis), and roll (rotation around the AP axis). Each sensor weight was 14.7 g.

Participants were asked to complete two trials of four conditions in the following order: (1) TUG without a secondary task (TUG), (2) TUG while wearing the HMD without a visual stimulus (TUG_{HMD}), (3) TUG with a visual stimulus dynamically pitching-up (TUG_{PU}), (4) TUG with a visual stimulus dynamically pitching-down (TUG_{PD}).

Data Analysis

All acceleration and angular velocity signals were analyzed using custom-written Matlab Codes (MathWorks, Natick, MA, USA). Each experimental condition was collected twice and averaged and expressed as the averaged mean [137] and standard deviation f both conditions. To minimize signal distortion and noise, we filtered the accelerometer and gyroscope signals using a fourth order Butterworth low pass filter with a cut-off frequency of 2 Hz.[138] Consistent with previous studies, pitch angular velocity signal from the lumbar sensor was used when identifying sit-to-stand, stand-to-sit, and turning [118]. The transitions were determined as follows: 1) Sit-to-stand: first 10 °/sec and the second -10 °/sec were considered as the start and end of the sit-to-stand, respectively, 2) Stand-to-sit: last 10 °/sec and the last -10 °/sec were considered as the start and end of stand-to-sit, respectively [118], 3) Turning, both Turn 1 (around the cone) and Turn 2 (prior to sitting back down) was determined based on the yaw angular velocity signal at the lumbar sensor by first identifying the maximum angular velocity and its corresponding time [118,139] . The 20% of the maximum angular velocity points were then determined in order to demarcate the start and end of the turns[118,139], and 4) Walk 1 (from chair to cone) was determined to start the end of sit-to-stand period to the beginning of turn 1, respectively. The start and end of Walk 2 (from cone to chair) were identified as the end of turn 1 and the start of stand-to-sit, respectively[139]. We choose to include all steps taken by the participant in walking and to not exclude the first and last steps due to two reasons: (1) limited walking space when completing the TUG test (around 6 meters going back and forth from the chair) and (2) the differences in motor behavior that older adults may exhibit as they may encounter more challenges in gait termination and initiation compared to the younger adults [140,141]. All Matlab algorithm results were verified using a video analysis. For example, if the Matlab code failed to identify the start and end of turning in the peak yaw angular velocity, a confirmation would be obtained using the video analysis. This analysis involves counting the number of steps taken by the participant, which followed by visualizing the shank sensors pitch angular velocity signals to identify the correct indexes.

We analyzed the mean and standard deviation of the values of two trials/condition that was completed by each participant for all dependent variables. For example, if the first trial of TUG condition for turning cadence was $TUG_{FirstTrial}$=90 step/min and the second trial was $TUG_{SecondTrial}$ =80 step/min, then we calculated the mean and standard deviation of $TUG_{FirstTrial}$ and $TUG_{SecondTrial}$ values to represent TUG condition. The dependent variables were defined as follows:

a. Turning cadence (step/min) as the number of steps per minutes[115].

b. Gait speed in walking (m/sec) as the time it takes the individual to travel the distance.

c. PTV in sit-to-stand, turn, walk, and stand-to-sit (°/sec) as the maximum angular velocity of the trunk around the ML (pitch), vertical (yaw), and AP (roll) axes [114].

d. ML, vertical, and AP acceleration amplitude range (g) and jerk (g s^{-1}) of sit-to-stand and stand-to-sit: The range was calculated as the (maximum-minimum acceleration value) and the jerk was calculated the derivative of *acceleration* with respect to time[125].

A sample size of 28 participants was determined to be adequate through a power analysis using G*Power 3.1.9.2 based ANOVA model with repeated measures, within-between interaction, with a moderate effect size of 0.25, alpha level of 0.05, and 80% power. A two factor (2 groups "younger and older adults" x 4 conditions (TUG, TUG_{HMD}, TUG_{PU}, and TUG_{PD})) repeated measures (within-between subjects) ANOVA was conducted on the dependent variables (significance level $p<0.05$). Bonferroni post hoc analysis was followed in case of main-effect significance with a corrected level of significance ($p<0.025$)[142].

Results

Older and younger adults' scores on the clinical balance measures and cognitive function were comparable except in BBS where older adults showed poorer performance compared to younger adults (Table 1). (Table 2) summarized the raw data for all the dependent variables in the four conditions.

Table 1 Participants demographics and clinical characteristics

	Young adults (n=12, 6M, 6F)	**Old adults (n=16, 8M, 8F)**	**p-value**
Age (years)	25.9 ± 3.9	69.0 ± 4.4	p<0.001 **
BMI	24.9 ± 4.6	24.0 ± 5.9	p=0.65
MMSE	29.1 ± 0.8	29.7± 1.0	p = 0.11
BBS	56±0.00	54.6 ±1.3	p =0.001 **
ABC	97.5 ± 4.7	94.1 ± 8.1	p =0.2
TUG (sec)	10.6 ± 1.5	10.4± 2.7	p=0.8
FRT (inch)	12.8 ± 2.0	11.9 ± 5.7	p=0.6
Bucket test	0.6 ± 0.48	1.5 ± 1.2	p=0.01 **
RFT	2.5 ± 14.8	3.7±8.1	p=0.78
Note: Values are represented in mean ± SD. BMI (Body Mass Index), MMSE (Mini Mental State Examination), BBS (Berg Balance Scale), ABC (Activity-specific Balance Confidence), TUG (Timed Up and Go Test), FRT (Forward Reach Test), and RFT (Rod and Frame test)			

Table 2, continues Raw data for the dependent variables represented as in mean±SD. OA- older adults, YA- younger adults.

	TUG	**TUG_{HMD}**	**TUG_{PU}**	**TUG_{PD}**
Turning Cadence (step/min)	172.7±34.8	147.1±47.1	142.1±35.9	144.6±35.7
Gait speed (m/sec)	1.1±0.3	0.8±0.3	0.78±0.28	0.75±0.27
PTV around ML in sit-to-stand (°/sec)	96.7±18.4	79.2±15.1	79.4±15.2	76.03±15.2
PTV around vertical in sit-to-stand (°/sec)	48.1±15.7	40.7±16.1	39.1±11.5	44.2±15.9
PTV around AP in sit-to-stand (°/sec)	33.1±15.1	25.6±10.3	25.2±9.2	25.4±9.1
PTV around ML in turn (°/sec)	56.1±20.1	47.3±16.02	47.5±15.6	46.1±16.1
PTV around vertical in turn (°/sec)	157.1±33.1	121.1±42.4	177.8±33.3	114.6±34.02
PTV around AP in turn (°/sec)	51.1±25.4	41.9±18.9	42.3±16.9	43.2±20.1
PTV around ML in walk (°/sec)	46.7±16.1	41.8±14.9	40.8±13.1	39.3±14.8
PTV around vertical in walk (°/sec)	78.9±39.4	62.1±34.1	65.8±39.0	62.1±33.0
PTV around AP in walk (°/sec)	34.7±15.7	31.1±17.4	30.2±15.8	30.2±13.3
PTV around ML in stand-to-sit (°/sec)	87.8±22.5	67.8±22.6	70.8±16.5	66.3±12.8

PTV around vertical in stand-to-sit	43.3±18.5	52.3±22.1	58.5±26.8	48.1±23.1
PTV around AP in stand-to-sit (°/sec)	31.6±12.4	35.8±18.4	37.9±16.5	34.3±17.6
ML acceleration range in sit-to-stand (g)	0.45± 0.2	0.53± 0.3	0.37± 0.1	0.37± 0.1
Vertical acceleration range in sit-to-stand (g)	0.92±0.3	0.87±0.3	0.72±0.22	0.76±0.3
AP acceleration range in sit-to-stand (g)	1.0±0.23	1.0±0.36	0.87±0.16	0.88±0.14
ML acceleration jerk in sit-to-stand (g s^{-1})	0.04±0.02	0.03±0.03	0.03±0.03	0.12±0.3
Vertical acceleration jerk in sit-to-stand (g s^{-1})	0.1±0.05	0.08±0.09	0.08±0.09	0.09±0.07
AP acceleration jerk in sit-to-stand (g s^{-1})	0.36±0.12	0.31±0.1	0.32±0.1	0.3±0.1
ML acceleration range in stand-to-sit (g)	0.48±0.46	0.44±0.19	0.42±0.17	0.42±0.28
Vertical acceleration range in stand-to-sit (g)	0.89±0.52	0.84±0.45	0.72±0.3	0.67±0.3
AP acceleration range in stand-to-sit (g)	1.1±0.4	0.97±0.2	0.86±0.3	0.9±0.2
ML	0.08±0.08	0.06±0.07	0.06±0.08	0.05±0.05

acceleration jerk in stand-to-sit (g s^{-1})				
Vertical acceleration jerk in stand-to-sit (g s^{-1})	0.29±0.17	0.17±0.13	0.18±0.11	0.18±0.11
AP acceleration jerk in stand-to-sit (g s^{-1})	0.7±0.3	0.4±0.15	0.43±0.15	0.43±0.17

Adults showed decrements in performance when wearing the HMD, regardless of the addition of a visual stimulation, compared to viewing the normal view of the room in the standard TUG test. Specifically, all adults showed a decrease in turning cadence, gait speed, PTV around the ML axis in sit-to-stand and stand-to-sit, PTV around the vertical axis in TUG activities except sit-to-stand, PTV around the AP axis in all TUG activities except stand-to-sit, lower AP jerk in sit-to-stand and stand-to-sit, and vertical jerk of stand-to-sit (all $ps<0.05$) .

As illustrated in (Figure 7, 8, and 9), visual stimulation affected the trunk kinematics and the acceleration range in walking and sit-to-stand, respectively. While viewing a rotated visual scene in a pitch up or pitch down direction, adults significantly decreased their PTV around the ML axis in walking and their AP acceleration range in sit-to-stand compared to standard TUG conditions (all $ps<0.01$). Adults showed a lower vertical acceleration range in sit-to-stand in sit-to-stand as well while viewing a pitch up rotating visual scene only ($p=0.002$) compared to TUG.

Variability in the PTV around the ML axis significantly decreased between conditions in turning ($p=0.047$) and walking ($p=0.01$). In turn, adults tend to decrease their variability in the trunk once they wore the HMD, however, this does not reach a significant level. In walking, a rotated visual stimulus in a pitch down direction decreased the variability of PTV around ML axis compared to standard TUG ($p=0.009$).

(Table 3) summarized repeated measures ANOVA results with the estimated marginal means between old and young adults. There was an age-related difference in the trunk kinematic ($p=0.03$). Older adults significantly decreased their PTV around the AP axis while turning compared to younger adults (Figure 10). There was no difference in the variability between younger and older adults in all dependent variables (all $p \geq 0.05$). Other predicted aging and interaction effects between aging and conditions in all other dependent variables were not significant (all $p \geq 0.05$).

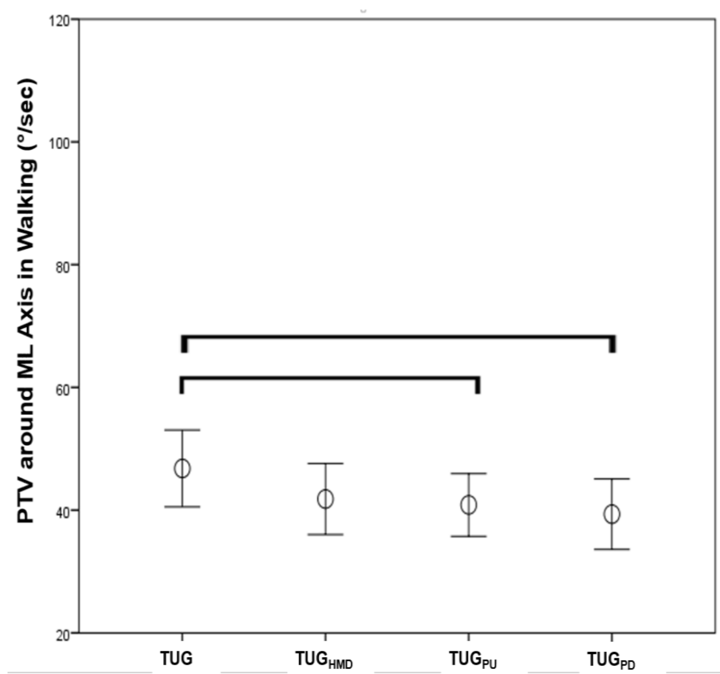

Figure 7 Peak trunk velocity around the mediolateral axis in walking between the four conditions of all adults

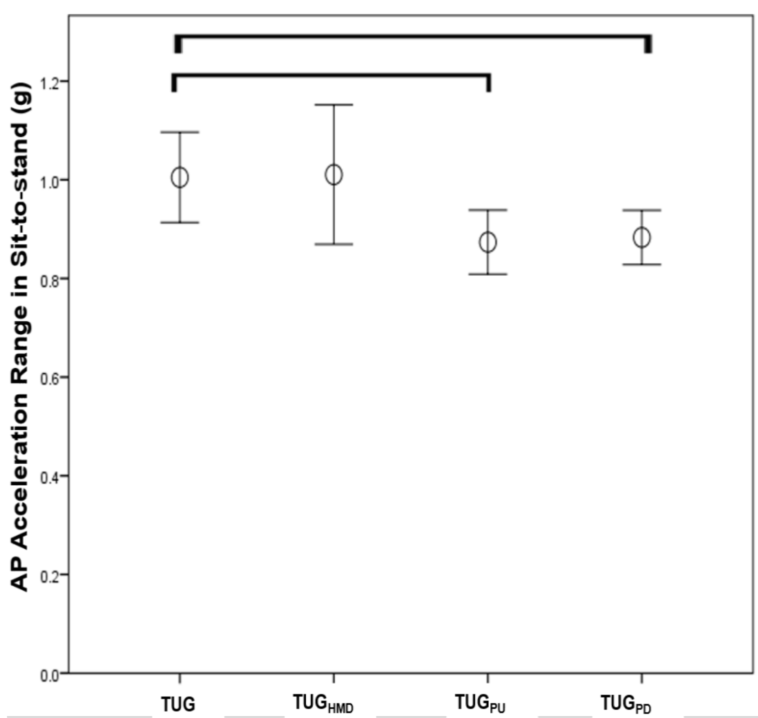

Figure 8 Anteroposterior acceleration range in sit-to-stand between the four conditions of all adults

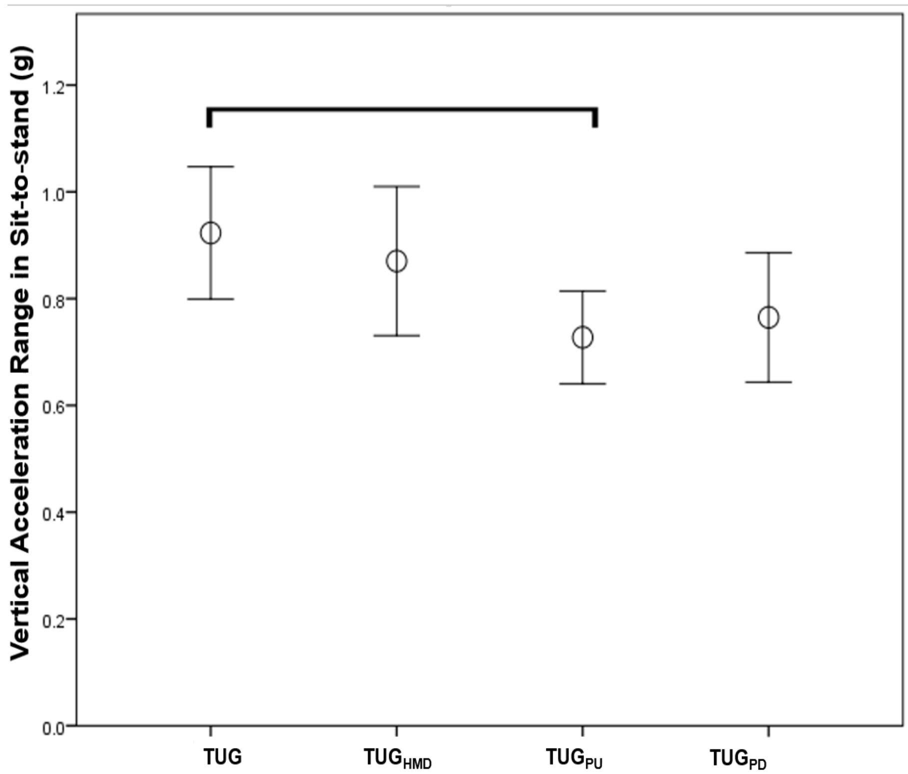

Figure 9. Vertical acceleration range in sit-to-stand between the four conditions of all adults

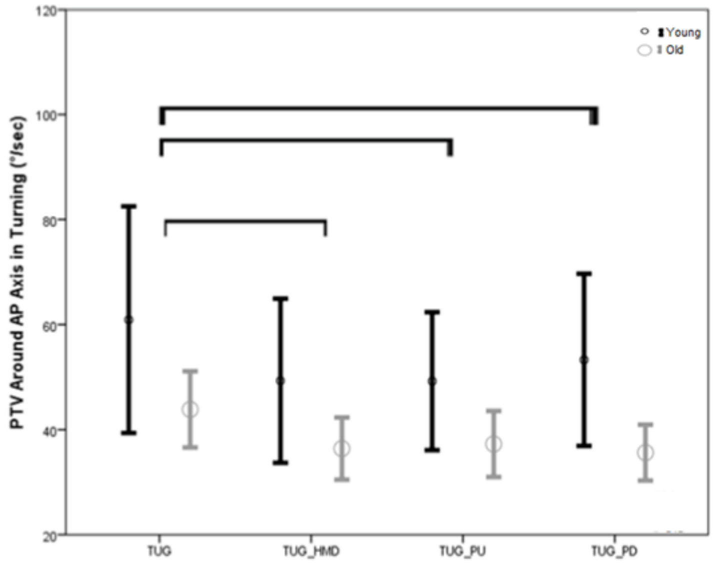

Figure 10 Peak trunk velocity around the anteroposterior axis in turning between the four conditions in younger (black) and older (gray) adults

Discussion and Conclusion

The objective of this work was to determine the influence of wearing an HMD with and without presenting a visual stimulus between old and young adults. Our findings showed that simply wearing the HMD without adding a visual stimulation had an impact on TUG kinematics. Adding a visual stimulation in both directions, i.e. PU or PD, decreased the PTV around the ML axis in walking and AP acceleration range in sit-to-stand. The addition of a visual stimulus that rotated in a pitch up direction decreased the vertical acceleration range in sit-to-stand. Whereas, presenting a visual stimulus in a pitch down rotation cause a decrease in the variability in the PTV around the ML axis in walking. In addition, we found that in comparison to younger adults, older adults exhibited a decline in the PTV around the AP axis in turn.

Table 3 Repeated measures univariate for the dependent variables with the estimated marginal means between younger and older adults expressed as mean (standard error).

Dependent variables	**Young Adults**	**Older Adults**	**Aging**
Turning Cadence (step/min)	152.5(9.8)	150.9(8.5)	P=0.9
Gait speed (m/sec)	0.92(0.07)	0.83(0.06)	P=0.4
PTV around ML in sit-to-stand (°/sec)	82.1(3.9)	83.4(3.4)	P=0.8
PTV around vertical in sit-to-stand (°/sec)	45.1(3.5)	41.5(3.04)	P=0.4
PTV around AP in sit-to-stand (°/sec)	29.6(2.4)	25.7(2.1)	P=0.2
PTV around ML in turn (°/sec)	54.1(4.2)	45.5(3.6)	P=0.1
PTV around vertical in turn (°/sec)	138.1(9.5)	119.8(8.2)	P=0.1
PTV around AP in turn (°/sec)	53.1(5.1)	38.2(4.4)	**P=0.03***
PTV around ML in walk (°/sec)	47.3(3.8)	38.3(3.2)	P=0.08
PTV around vertical in walk (°/sec)	78.8(9.8)	58.6(8.4)	P=0.1
PTV around AP in walk (°/sec)	36.4(4.1)	27.9(3.5)	P=0.1
PTV around ML in stand-to-sit (°/sec)	73.9(4.1)	72.6(3.5)	P=0.8
PTV around vertical in stand-to-sit	54.5(5.1)	47.5(4.4)	P=0.3
PTV around AP in stand-to-sit (°/sec)	36.7(3.6)	33.6(3.1)	P=0.5
ML acceleration range in sit-to-stand (g)	0.43(0.04)	0.44(0.4)	P=0.9
Vertical acceleration range in sit-to-stand (g)	0.8(0.06)	0.83(0.05)	P=0.6
AP acceleration range in sit-to-stand (g)	0.91(0.05)	0.96(0.4)	P=0.5
ML acceleration jerk in sit-to-stand (g s^{-1})	0.04(0.02)	0.07(0.02)	P=0.29
Vertical acceleration jerk in sit-to-stand (g s^{-1})	0.11(0.01)	0.07(0.01)	P=0.1
AP acceleration jerk in sit-to-stand (g s^{-1})	0.33(0.02)	0.31(0.02)	P=0.6
ML acceleration range in stand-to-sit (g)	0.43(0.05)	0.44(0.05)	P=0.9
Vertical acceleration range in stand-to-sit (g)	0.77(0.08)	0.79(0.07)	P=0.6
AP acceleration range in stand-to-sit (g)	0.93(0.07)	0.97(0.06)	P=0.68
ML acceleration jerk in stand-to-sit (g s^{-1})	0.08(0.01)	0.05(0.01)	P=0.1
Vertical acceleration jerk in stand-to-sit (g s^{-1})	0.2(0.02)	0.19(0.03)	P=0.38
AP acceleration jerk in stand-to-sit (g s^{-1})	0.5(0.04)	0.4(0.03)	P=0.6

Our finding that simply wearing the HMD without adding a visual scene impacted TUG kinematics may be related to several factors. First, the HMD can limit the peripheral vision, thereby increasing the displacement and decrease the velocity of the head[143]. Peripheral vision plays a role in the movement control and organization and uses optical flow to provide updates about the spatial environment characteristics [144–146] . For example, in turning, peripheral visual inputs can decrease the variability of the timing of body rotation[101], while in standing it is used for a viewer-centered frame of a reference controlled by the direction of the head and gaze[147]. Because peripheral vision offers a unique contribution toward postural control, the unavailability or diminishing of it can influence the movement negatively. Second, HMD wear can increase the rotation of the neck and the flexion of the neck and trunk of the users as they try to be more comfortable with the device, overcome the limitation in the visual field, and choose the best viewing angle to navigate in the environment[132].

The addition of a visual stimulation affected the trunk kinematic in walking and the AP and vertical displacement in sit-to-stand. Visual perturbation showed to increase the postural instabilities as it was demonstrated in an increased sway in quiet stance [148], shorten step length in walking [21], and decreased trunk and head velocities in sit-to-stand[22]. Adults decreased their vertical displacement in sit-to-stand when viewing a visual stimulus in a pitch up rotation and decreased their PTV variability around the ML axis in walking when viewing a visual stimulus in a pitch down rotation. Viewing the visual stimulus in a pitch up rotation affected the vertical displacement in sit-to-stand can create an illusion of going backward [149]; thus, the participants may up-weight the somatosensory inputs and down-weight the visual inputs, which translated into a smaller range in sit-to-stand motion. In contrast, viewing the room in a pitch down rotation can create the illusion of moving forward [149], and therefore, the adults showed decreased in flexibility in the PTV around the ML axis as they move in a forward direction in walking.

We found that older adults decreased their PTV around the AP axis in turning compared to younger adults. Turning is a challenging activity that requires changing the direction of the body from AP to ML direction. The difficulty to achieve a successful turn increases when the visual inputs are not available peripherally. It was found previously that the unavailability of peripheral visual inputs caused a motor incoordination in a form of increasing the variability of segment rotation onset duration[132]. Aging- related changes in turning were observed previously. To orient the body into ML direction, older adults showed a delay in the ML center of mass (COM) reorientation, slow gait velocity, and an increase in step width compared to younger adults [150]. Decreasing the flexibility of the trunk velocities might be a strategy used to maintain balance while turning with a limited peripheral vision and unusual visual inputs through the HMD.

Our results support the conclusion that wearing an HMD affects TUG kinematics. Aging-related decrements were found in the trunk velocity in turning compared to younger adults. These findings suggest that wearing the HMD can present perceptual challenges that need to be addressed when using VR in future for rehabilitation purposes.

CHAPTER 3

THE EFFECT OF A VISUAL STIMULATION ON TIMING OF THE TIMED UP AND GO TEST (TUG)

This chapter will discuss how aging can influence the duration of the TUG and its movement components (i.e. sit to stand, walk, turn, and stand-to-sit), when a motor task and/or a visual stimulus through a head mounted display (HMD) were added, in a group of young and old adults.

Introduction

Falling is one of the leading causes of disability and mortality. Approximately 1 in 3 adults falls annually [2,3] resulting in various types of injuries. A fall can lead to a fear of falling, which is a psychological state recognized as a predictor of experiencing a future fall[151]. In addition to the physical and psychological impacts of a fall, the financial impact from a fall cannot be underestimated. The fall-related injuries' expenditure in the US was more than 50 billion dollars in 2015 [152]. An occurrence of a fall is multifactorial in nature, thus a comprehensive evaluation of functional mobility is necessary when exploring the probability for a fall. The use of an assessment method that focuses only on task performance in a predictable environment is likely not sufficient for identifying the probability of falls among older adults.

The Timed Up and Go (TUG) test is a recommended and simple clinical tool that is frequently used to evaluate fall risk in older adults[14]. The TUG test is composed of several functional dynamic activities: sit-to-stand, turn, walk, turn, and stand-to-sit. Previous reports have supported the TUG is a tool that predicts the probability of a fall[135]; however, other studies suggest that the TUG is inadequate to predict falls in community-dwelling older adults[123]. One potential explanation for this limitation is that the causes of falls are multifactorial and the standard TUG does not account for testing all risk factors of falling such as cognitive and visual impairments[153]. More recent work has expanded the TUG test by adding a secondary task, cognitive [154] and motor[155], and assessed sensitivity for detecting the probability of fall and found that when a cognitive task was added the sensitivity of the TUG was increased to detect the fallers. However, manipulating sensory systems, such as visual perturbations, during the TUG test, has not yet been explored. Older adults exhibit postural instabilities [44] under misleading visual information that possibly increases the risk of fall. Thus, older adults may experience more difficulty than younger adults when completing the TUG in a noisy visual environment. In addition to the total time for TUG completion as the primary score for the TUG, characterizing the spatio-temporal, kinematics during each TUG sub-task (sit-to-stand, walk, turns, and stand-to-sit) may provide means to enhance scoring of the TUG providing additional clinical utility for the measure.

Multisensory integration is the process of integration of the inputs from the three sensory systems: vision, vestibular, and proprioception for posture and movement control [154]. Adults adjust reliance on a single stream of sensory information when it appears less reliable than others. For example, when there are inaccurate somatosensory inputs, adults tend to decrease the weighting of the somatosensory system and increase the weighting of vestibular and visual inputs to maintain balance[28]. Some evidence suggests that older adults are less efficient in this re-weighting of sensory system information[29]. For example, older adults showed more postural instability when under misleading somatosensory and visual cues in sensory organization test than younger adults, suggesting their decreased efficiency in choosing the most appropriate responses to avoid falls under sensory misleading conditions[29]. Aging is also associated with decreased visual discernment including diminished visual acuity, loss of visual fields due to senile ptosis, loss of ability to see in dim-lighted environments and contrast sensitivity[156,157]. These age-related impairments affect the ability to recognize and distinguish objects in a number of environmental concerns, and can increase the risk of fall[158]. In addition, older adults exhibit greater visual dependence than younger adults[18]. Functionally, this means that older adults exhibit high reliance on visual information and difficulty in integrating visual inputs under visually demanding conditions. An increased risk of falls in older adults could be associated with increased visual dependence, decreased efficiency in weighting sensory systems, and increased postural instabilities under visually manipulated conditions.

Virtual reality (VR) can be used to understand how adults react to a visually complex environment. Postural control studies have used immersive VR via large display systems or with head mounted displays (HMDs) that are portable and easy to use and, therefore, have a greater potential to be used in clinics. Studies have suggested that interventions using VR can decrease the fear of falling and improve cognition, mobility[159], and balance control[160]. Recognizing that HMDs usually restrict the peripheral visual field possibly decreasing the ability to accurately perceive body orientation and detect objects in dynamic environments[145], an improved understanding of how HMDs may impact motor behaviors without additional VR has clinical relevance. Therefore, one should consider the effects of the physical properties of the HMDs, especially when used with dynamic sequential activities such as turning while walking.

The goal of this work was to explore how aging affects the temporal components of the TUG test when performing a common secondary motor task and/or during a perturbing visual condition. We used the multitask cost of the time to determine the impact of the conditions. We also explored if wearing the HMD impacted motor performance. We hypothesized that: (1) all adults will exhibit an increase in the multitask cost of time in a negative direction of TUG activities when wearing the HMD more than when completing a motor task and (2) older adults will exhibit a greater decline in performance, particularly in turning, compared to younger adults.

Methods

Participants

Twelve younger adults and sixteen older adults gave informed consent to participate in this study approved by Temple University's Institutional Review Board (IRB). Inclusion criteria for the study were ages were between 21-40 (young adult) or 60-90 years old (older adult), ability to walk independently, and no neurological or cognitive dysfunction.

Procedure

Prior to experimental testing, they underwent the following functional and balance clinical measures: 1) Activity-specific Balance Confidence (ABC) scale, a self-reported measure of the participants' confidence in performing activities [133]; 2) The Berg Balance Scale (BBS), an assessment of static balance and fall risk[133]; 3) Functional Reach Test (FRT), a measure of postural stability[134]; and 4) The Timed Up and Go test (TUG) to assess the dynamic balance and mobility in the elderly[135]. The Mini-Mental State Examination (MMSE) was administered to determine cognitive function.[136]

Experimental Protocol

All adults wore comfortable shoes during the TUG. Each participant had a practice trial of the TUG followed by the experimental conditions. In accordance with the standard TUG test instructions, when the command "go" was given, participants were asked to rise up from a chair, walk 3 meters at a comfortable speed, turn around 180°, walk back to the chair, and sit down[15]. The chair was armless and had a dimension of 44 cm (height) by 46 cm (width) (Figure 11). A line of tape and a cone were placed on the floor to indicate where the adults should turn. Two conditions were used to augment the TUG. First, a motor task (m) that included holding a half-full glass of water covered with a plastic cover (used to reduce spills) (Figure 12), which was then covered by a thin piece of fabric (to hide the plastic cover). Second, a visual perturbation consisting of a virtual scene of random dots of bright dots (i.e. snowflakes) that rotated at a constant speed (5 °/sec) in the pitch up (PU) or down (PD) directions viewed through an HMD. For the visual conditions, participants viewed the scene of moving random dots for 10 seconds before the TUG test started and during the TUG. We choose to start the TUG after approximately 10 seconds because it was shown that the motor responses of adults were more affected by the virtual environment rotation when they were asked to perform the movement after an exposure to the scene motion of 10 seconds [22].

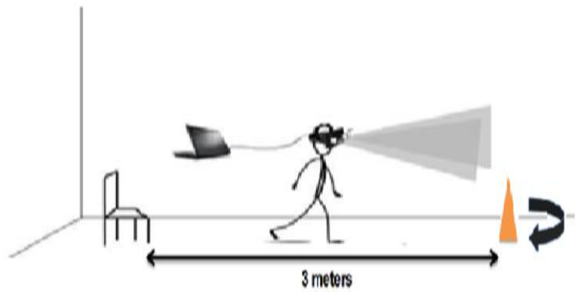

Figure 11 The Timed Up and Go test experimental setting


Figure 12. The visual scene presented to the adults via the HMD (left side of picture) and the cup of water held in the motor task (right side of picture)

The Oculus Rift Development Kit 2 HMD (Oculus VR, 2014 ™) was used to present visual scene to the participants (Figure 12). The Oculus Rift consists of two displays, one for each eye, with a resolution of 960 x1080 pixels per eye, a maximum refresh of 75 Hz, and a weight of 440 grams. Those with eyeglasses could wear them while wearing the HMD. The viewing optics allow a 100° viewing angle. An Ovrvision mount was used, which is a high-performance USB stereo camera customized for the

Oculus Rift, which allows users to replace their current display with a view of the real world. The resolution for the Ovrvision is 640 x 480 per eye (1280 x 480), the frame rate is 60 FPS, the angle of view is H90 °, V75 °, the latency is 50 msec, the pixel number is 0.6 MP, and its weight is 55 grams.

Temporal components of the TUG test were measured using Trigno™ wireless sensors (Delsys Inc.). Sensors were placed on the participants' sternum, lumbar, both wrists, and both shanks. The sensors include tri-axial accelerometer (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 3.5 mg); a tri-axial gyroscope (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 0.05°/sec); and a tri-axial magnetometer (range 40 m, resolution 16 bit, sampling frequency 74 s/sec, Noise < 0.4 uT). Each sensor weight was 14.7g. Kinematic data derived from the IMUs included three acceleration axes: anterior-posterior acceleration, vertical acceleration, and medio-lateral acceleration, and three angular velocity axes: pitch (rotation around the medio-lateral axis), yaw (rotation around the vertical axis), and roll (rotation around the anterior-posterior axis).

Participants were asked to complete two trials for each of eight conditions in the following order: (1) TUG without a secondary task, (2) TUG with a motor task (TUG_{m}), (3) TUG while wearing the HMD without visual stimulus (TUG_{HMD}), (4) TUG while wearing the HMD without visual stimulus with the motor task (TUG_{HMD_m}), (5) TUG with visual stimulus dynamically pitching- up (TUG_{PU}), (6) TUG with visual stimulus dynamically pitching- down (TUG_{PD}), (7) TUG with visual stimulus dynamically pitching- up and motor tasks (TUG_{PU_m}), and (8) TUG with visual stimulus dynamically pitching- down and motor tasks (TUG_{PD_m}).

Data Analysis

All acceleration and angular velocity signals were analyzed using custom-written MatLab Codes (MathWorks, Natick, MA, USA). Each experimental condition was collected twice and averaged [137] and expressed as the mean and standard deviation f both conditions. To minimize signal distortion and noise, we filtered the accelerometer and gyroscope signals using a fourth order Butterworth low pass filter with a cut-off frequency of 2 Hz.[138]. Consistent with previous studies, pitch angular velocity signal from the lumbar sensor was used when identifying sit-to-stand, stand-to-sit, and turning [118]. The transitions were determined as follows: 1) Sit-to-stand: first 10 °/sec and the second -10 °/sec were considered as the start and end of the sit-to-stand, respectively, 2) Stand-to-sit: last 10 °/sec and the last -10 °/sec were considered as the start and end of stand-to-sit, respectively [118], 3) Turning, both Turn 1 (around the cone) and Turn 2 (prior to sitting back down) was determined based on the yaw angular velocity signal at the lumbar sensor by first identifying the maximum angular velocity and its corresponding time [118,139] . The 20% of the maximum angular velocity points were then determined in order to demarcate the start and end of the turns[118,139], and 4) Walk 1 (from chair to cone) was determined to start the end of sit-to-stand period to the beginning of turn 1, respectively. The start and end of Walk 2 (from cone to chair) were identified as the end of turn 1 and the start of stand-to-sit, respectively[139]. We calculated the total time taken to complete the TUG test and the times of each sub-component in seconds. All Matlab algorithm results were verified using a video analysis. For example, if the Matlab code failed to identify the start and end of turning in the peak yaw angular velocity, a confirmation would be obtained using the video analysis. This analysis involves counting the number of steps taken by the participant, which followed by visualizing the shank sensors pitch angular velocity signals to identify the correct indexes.

The multitask cost can reflect the change in the motor behavior that occurs due to high attentional demand conditions, with the lowest cost reflecting poorer motor performance [20]. The multitask cost was calculated as the percent change of the time (total TUG time and the time of each TUG activity) in relation to the single task performance (i.e. TUG with no secondary task) as:

$$-[\,(Dual\ Task - Single\ Task)/(Single\ Task)\times 100]$$ [20]

A sample size of 28 participants was calculated through a power analysis using G*Power 3.1.9.2 based ANOVA model with repeated measures, within-between interaction, with a moderate effect size of 0.4, alpha level of 0.01, and 95% power (which suggested using n=20).. A two factor (2 groups "younger and older adults" x 7 conditions) repeated measures (within subjects) ANOVA was conducted on the dependent variables, which were the multitask cost of the time of each TUG sub-component. Bonferroni post hoc analysis was followed in case of main-effect significance with a corrected level of significance ($p<0.025$)[142] .

Results

The demographic data for all participants are presented in Table 4 and the time in seconds for the TUG and its activities are presented in Table 5. Older adults' scores in the balance measures and cognitive function were comparable to younger adults except in the BBS, where older adults showed poorer performance compared to younger adults ($p=.001$).

Table 4 Participants demographics and clinical characteristics

	Young adults (n=12, 6M, 6F)	**Old adults (n=16, 8M, 8F)**	**P-value**
Age (years)	25.9 ± 3.9	69.0 ± 4.4	p<0.001 **
BMI	24.9 ± 4.6	24.0 ± 5.9	p=0.65
MMSE	29.1 ± 0.8	29.7± 1.0	p = 0.11
BBS	56±0.00	54.6 ±1.3	p =0.001 **
ABC	97.5 ± 4.7	94.1 ± 8.1	p =0.2
TUG (sec)	10.6 ± 1.5	10.4± 2.7	p=0.8
Note: Values are represented in mean ± SD. BMI (Body Mass Index), MMSE (Mini Mental State Examination), BBS (Berg Balance Scale), ABC (Activity-specific Balance Confidence), and TUG (Timed Up and Go Test)			

As illustrated in Table 6, the multitask cost on the total TUG time was significantly different across conditions (Figure. 13) [$p<0.001$, $\eta2= 0.47$]. More specifically in:

(a) Walking [$p<0.001$, $\eta2= 0.31$] between TUG_m and all the conditions ($p<0.03$) (Figure 14),

(b) Stand-to-sit [$p<0.001$, $\eta2= 0.19$] between TUG_m all conditions ($p<.005$) except TUG_{PD_m} (Figure 15) and

Table 5 The time of TUG and its sub-components (in seconds).

		TUG	TUG_m	TUG_{HMD}	TUG_{HMD_m}	TUG_{PU}	TUG_{PD}	TUG_{PU_m}	TUG_{PD_m}
Total TUG Time	**YA**	12.3±2.1	12.4±2.1	15.7±3.3	15.7±3.0	15.3±2.3	15.8±2.3	15.8±2.9	15.1±2.1
	OA	13.0±3.5	13.1±3.2	23.4±12.9	22.7±10.1	23.2±12.9	23.7±11.4	23.0±11.4	22.7±11.3
sit-to-stand Time	**YA**	1.6±0.3	1.7±0.4	1.8±0.5	1.7±0.4	1.7±0.5	1.7±0.4	1.7±0.5	1.5±0.4
	OA	1.5±0.2	1.6±0.3	1.8±0.4	1.8±0.4	1.8±0.3	1.9±0.5	1.8±0.4	1.8±0.3
Turn Time	**YA**	1.8±0.3	1.8±0.3	2.6±0.7	2.6±0.7	2.5±0.6	2.5±0.5	2.6±0.6	2.6±0.6
	OA	2.0±0.6	2.1±0.5	3.4±0.9	3.7±1.2	3.7±1.8	3.6±1.2	4.1±2.2	4.1±2.0
Walk Time	**YA**	2.8±0.7	2.7±0.7	3.4±1.0	3.4±0.6	3.5±0.8	3.7±0.9	3.5±0.6	3.3±0.4
	OA	2.9±1.0	2.7±0.9	6.3±5.6	5.6±3.8	6.0±4.6	6.2±4.4	5.5±3.4	5.3±3.7
Stand-to-sit Time	**YA**	1.0±0.2	1.1±0.3	1.5±0.5	1.8±0.8	1.6±0.5	1.5±0.4	1.7±0.6	1.6±0.8
	OA	1.2±0.4	1.2±0.4	2.0±0.7	1.7±0.5	1.7±0.7	1.6±0.5	1.6±0.5	1.6±0.6

Note: Values are represented in mean ± SD. OA – older adults, YA – younger adults

(c) Turning [$p<0.001$, $\eta2= 0.44$] between TUG_m all the conditions (p<.001) (Figure 16),

There was an age-related decrement of the multitask cost of the total TUG time (p<0.001). Particularly, in turning (p=0.001) and sitting-to-standing (p=0.02) (Figure 16). The time-related decrements were observed more in older adults compared to younger adults. There was an interaction effect of aging and tasks in total TUG time (p=0.01). An aging-related decline in time was shown with more complex conditions when the visual stimulus and motor tasks were simultaneously performed with the TUG.

Table 6 The p-values and partial eta-squared (η2) for all the multitask cost of the total TUG and its activities time

Multitask cost (%)	**Tasks (p-value** η2)	**Aging (p-value,** η2)	**Interaction (p-value,** η2)
Total TUG Time	(p<0.001, 0.4)	(p=.01, 0.2)	(p=0.01, 0.1)
Sit-to-stand	(p=0.3, 0.04)	(p=0.02, 0.1)	(p=0.1,0.07)
Turn	(p<0.001, 0.4)	(p=0.001,0.3)	(p=0.1,0.07)
Walk	(p<0.001,0.3)	(p=0.06,0.1)	(p=0.06,0.1)
Stand-to-sit	(p<.001,0.1)	(p=0.2,0.05)	(p=0.2,0.05)

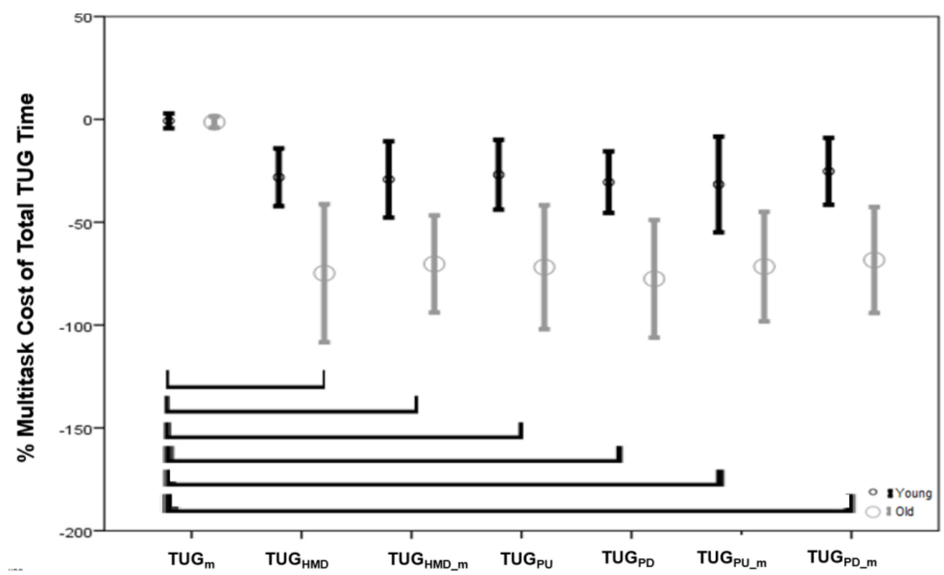

Figure 13. The multitask cost on the total TUG time is plotted for the different secondary tasks used simultaneously with TUG. P values indicate significance between the conditions in comparison to the motor task (m)

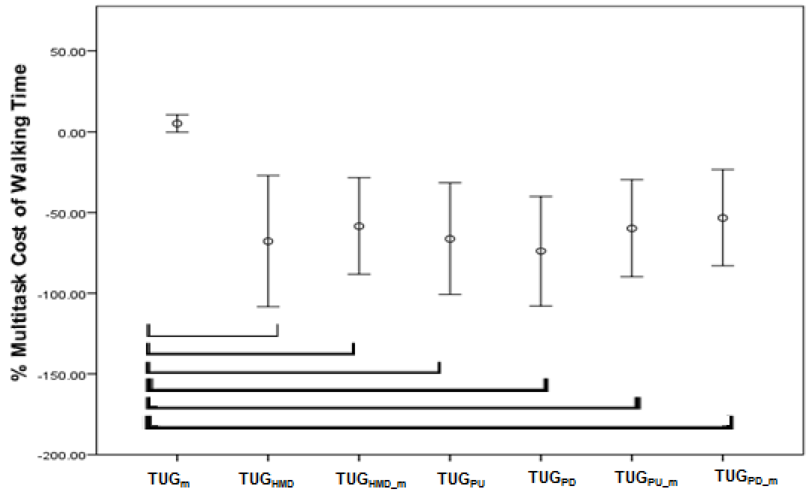

Figure 14 The multitask cost on the walking time is plotted for the different secondary tasks used simultaneously with TUG. P values indicate significance between the conditions in comparison to the motor task (m)

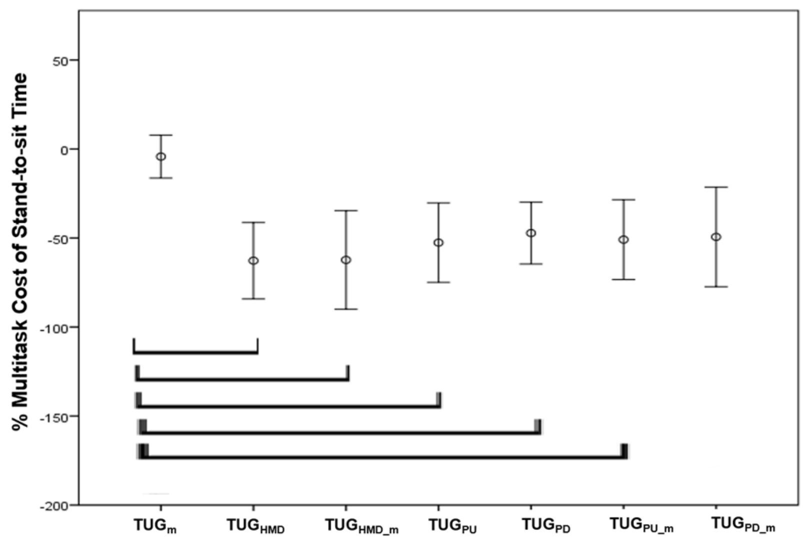

Figure 15 The multitask cost on the stand-to-sit time is plotted for the different secondary tasks used simultaneously with TUG. P values indicate significance between the conditions in comparison to the motor task (m)

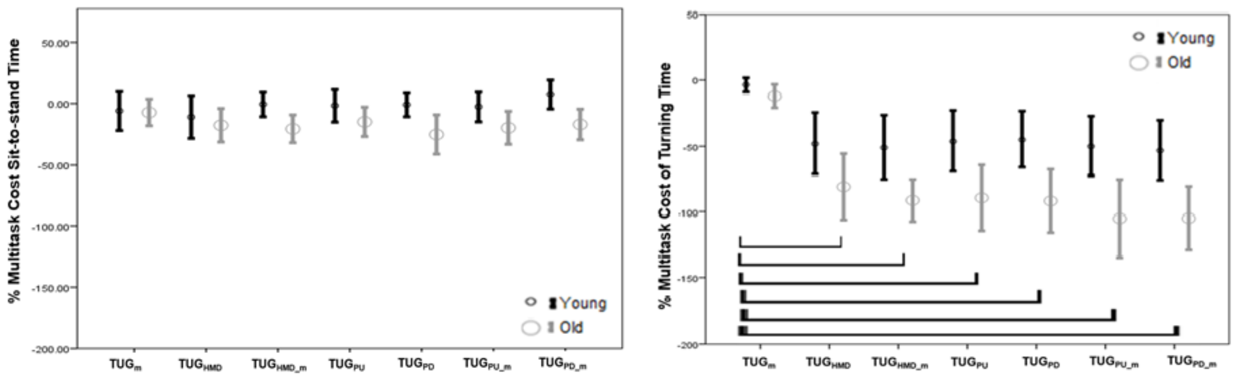

Figure 16 The multitask cost on the sit-to-stand (left side) and turning (right side) time is plotted for the different secondary tasks used simultaneously with TUG. P values indicate significance between the conditions in comparison to the motor task

Discussion and Conclusion

This study is the first to explore the effects of manipulating the visual system when performing the functional activities of the TUG test. We examined the differences in older and younger adults' motor behaviors in terms of the total time of the TUG, time of each TUG component, and the multitask costs on these times. We chose the multitask cost on time as our dependent variable because time is the standard and an easy measure used by clinicians. Also, time to complete TUG has been shown to be valid and reliable in differentiating population with various functional levels[161–163]. Our results indicate that, in this particular experimental activity, TUG components of turning, walking, and stand-to-sit are the activities that adults spent a longer time performing when combined with a perturbing visual stimulation, but not with a motor task. Additionally, older adults showed decrements in the total TUG time, particularly in turn and sit-to-stand, with the addition of a motor task and visual stimulus.

The addition of a motor task to the TUG did not cause a significant effect on the multitask cost of the TUG times compared to the conditions where the HMD were worn. A manual-motor behavior may require less attentional demands while completing TUG activities. Adults improved the ability to indicate verticality when they were asked to hold a glass of water compared to holding a joystick [56,58], which was linked to the functionality of holding a glass of water. Similarly, in this study, the performance of adults was not influenced by holding a cup of water, which could be linked to their familiarity level of holding a cup of water compared to wearing the HMD.

Wearing the HMD while completing the dynamic activities of the TUG increased the time taken to complete the test, particularly in turning, walking, and stand-to-sit activities for both young and older adults. Specifically, older adults were more impacted by turning and sitting-to-standing while multitasking more than younger adults. This could be due to unfamiliarity with navigating around with the HMD and the additional, though slight, weight on the head. The participants may also have experienced a possible visual updating delay (20 – 40 msec) due to computer processing time when viewing the room via the HMD, which has been shown to affect the performance of the users [164]. All of these factors can affect the visual ocular reflex (VOR) and optokinetic calibration, which may influence the motor responses of adults[101]. Most importantly, wearing HMD reduces the peripheral field of view, which is necessary for updating spatial information and using optical flow to control movement[145,146]. Compared to central vision, the peripheral vision has a greater impact in reducing postural sway in the direction of the presented visual stimuli[147].

The performance of younger and older adults did not significantly differ with the addition of a perturbing visual condition through the HMD. This result could be due to two reasons. First, wearing the HMD already resulted in prolonged time for TUG completion. A second reason may be that the simplicity of the visual scene used in this experiment was not impactful for the high functional level of our sample. Together, these findings suggest that visual field loss impacting the periphery, such as that occurs with normal aging, may be the cause of reduced postural stability in VR conditions using HMDs.

Further, transitioning from one body position to another (sit-to-stand or stand-to-sit) and turning while walking requires changes in body direction and momentum, and requires additional balance skills. This is consistent with previous findings showing that the duration of turning and walking during the TUG is sensitive to multitasking [20,128]. In standing-to-sitting, we observed an effect on duration in all conditions except one of the combined visual and motor task conditions (TUG_{PD_m}). It is possible that adults were able to improve their performance during the earlier condition trials, hence had more practice with the HMD conditions before the TUG_{PD_m} condition, which was due to the fixed order of the experimental conditions to the participants. Compared to sit-to-stand, a successful, controlled stand-to-sit transition requires motor coordination, eccentric control, and higher reliance on visual inputs to successfully locate the chair prior to sitting down[78]. Thus, if the visual input is diminished or unreliable, adults may need more time to ensure a safe and controlled landing on the chair.

The slowing of total TUG time in the older adult group when multitasking appeared to be associated with multitasking decrements in turning and sitting-to-standing time. It has been found previously that older adults show a poorer performance in a reaction time task while walking and turning than during walking in a straight line, suggesting that turning may involve more cognitive resources than walking. In our study, older adults were more challenged during turning especially when the visual field was compromised as a result of wearing the HMD. In dynamic activities, such as turning, peripheral visual field loss can produce motor incoordination in the form of increased variability of body-segments rotation[101]. In addition, sitting-to-standing requires achieving a vertical position while maintaining the center of mass over the base of support. Older adults have shown to have limitations in ankle strength and flexibility and decreased plantar tactile sensitivity [165], which are coupled with using of an armless chair and complex task (i.e. viewing the world through HMD and holding a cup of water). All of these factors can create challenges for older adults to complete sit-to-stand.

Limitations of this study included a fixed order of the experimental conditions that were presented to the participants, which may have contributed to an improved performance in the latest conditions compared to the earlier ones. Also, we did not screen for visual acuity, somatosensory loss, or vestibular dysfunction. Future studies should examine cohort with these impairments as it possible that adults with visual or vestibular disorders may exhibit different motor responses than those with intact visual and vestibular function.

To conclude, turning during the TUG revealed age-related decrements, but the time needed to perform the other TUG subcomponents did not. The wearing of an HMD appears to impact TUG kinematics in both younger and older adults, particularly in turning. This work provides insights into the impact of HMD wear, motor dual-task and perturbing visual conditions on TUG performance in healthy older and younger adults. These insights may support the development of TUG modifications that are more sensitive to age-related changes in balance control that could improve the measurement characteristics of the TUG.

CHAPTER 4

INFLUENCE OF VISUAL STIMULATION AND MOTOR MANIPULATION ON KINEMATICS OF YOUNGER AND OLDER ADULTS DURING THE ACTIVITIES OF THE TIMED UP AND GO TEST (TUG)

This chapter will examine how aging and visual dependency affects the kinematics of adults when they simultaneously perform a motor task and/or attend to a rotated visual stimulus.

Introduction

Successful navigation in the physical environment requires maintaining balance even when sensory information or one's attention is compromised. To maintain balance, one integrates and weighs inputs from sensory systems — the somatosensory, visual, and vestibular — based on the qualities and demands of the environment[28]. In humans, the visual system is usually the predominant system for maintaining balance [24,25]. Visual dependence (VD) is a term used to describe those who rely heavily on the visual system to maintain postural balance [151,152]. Advanced aging is linked with increased visual dependency [18] and postural instabilities under misleading visual inputs, i.e. moving virtual environments,[19,22,166] , which may lead to loss of balance. It is possible to speculate that aging and visual dependency can lead to specific motor changes when older adults are navigating in visually-demanding dynamic environments.

Previous reports showed the effects of visual perturbation in standing, walking, or sitting-to-standing tasks [19,22,166] ; however, in everyday activities, these tasks are not usually performed discretely but as consecutive skills with movement transitions between each task. Thus, it is important to understand the biomechanical differences between young and older adults when they transition in a series of motions observed in daily living under multitasking and sensory conflicts conditions. In this research, we explore the effects of aging and visual dependency on motor behavior under increased stabilization demands by adding a motor task and a visual stimulus perturbation through a head mounted display (HMD) to different dynamic tasks. The dynamic tasks included sit-to-stand, turn, walk, and stand-to-sit, which are parts of a fall risk assessment test embedded within the clinical tool the Timed Up and Go (TUG) [14]. We hypothesized that: (1) increasing visual demands by the wear of HMD will cause decrements to the mean and standard deviation of dependent variables as follow: a decrease in the turning cadence, gait speed, peak trunk velocity (PTV), acceleration range and jerk of sit-to-stand and stand-to-sit more than adding a motor task. Additionally we anticipated to observe a lower jerk of the cup of water when wearing the HMD is added compared to not wearing the HMD (2) older adults will show decrements in the dependent variables more than younger adults, and (3) VD older adults will show decrements in the dependent variables more than VI older adults.

Methods

Participants

Twelve younger adults (6 males) and sixteen older adults (8 males) provided informed consent to participate in this study approved by Temple University's Institution Review Board (IRB). Participants were between 21-40 years of age (younger adults) or 60-90 years (older adults), and able to walk independently with no diagnosed neurological or cognitive condition interfering with activities of daily life.

Procedure

To assess the balance and cognitive abilities of the participants, they underwent the following functional and balance clinical measures prior to running the protocol: 1) Activity-specific Balance Confidence (ABC) scale, a self-reported measure of the participants' confidence in performing activities [133]; 2) The Berg Balance Scale (BBS), an assessment of static balance and fall risk[133]; 3) Functional Reach Test (FRT), a measure of postural stability[134]; and 4)The Timed Up and Go test (TUG) to assess the dynamic balance and mobility in the elderly[135]. The Mini-Mental State Examination (MMSE) was administered to determine cognitive function.[136]

Two tests were used to examine visual function in terms of visual dependency and subjective visual vertical. First, visual dependency was tested using the computerized Rod and Frame test (RFT). Participants were asked to align a virtual rod to true vertical and horizontal orientations inside a virtual square frame that was rotated relative to a true vertical. Those who have a score within 4.5 degrees were considered as VI and those who fell outside of that range were considered as VD [42]. Second, the bucket test was used to examine the subjective visual vertical. This test is used to determine the verticality perception of the adults, which can be affected by vestibular disorders, and the used cutoff score was 2 degrees [167].

Experimental Protocol

All adults wore comfortable shoes during the TUG. Each participant had a practice trial of the TUG followed by the experimental conditions. In accordance with the standard TUG test instructions, when the command "go" was given, participants were asked to rise up from a chair, walk 3 meters at a comfortable speed, turn around 180°, walk back to the chair, and sit down[15] (Figure 17). The chair was armless and had a dimension of 44 cm (height) by 46 cm (width). A line of tape and a cone were placed on the floor to indicate where the adults should turn. Two conditions were added to the TUG. First, a motor task (m) that included holding a half-full glass of water covered with a plastic cover (used to reduce spills), which was then covered by a thin piece of fabric (to hide the plastic cover) (Figure 18). All adults were right-handed except one participant and they held the cup of water using the dominant hand. Second, a visual perturbation consisting of a virtual scene of random dots of bright dots (i.e. snowflakes) that rotated at a constant speed (5 °/sec) in the pitch up (PU) or down (PD) directions viewed through an HMD. For the visual conditions, participants viewed the scene of moving random dots for 10 seconds before the TUG test started and during the TUG. We choose to start the TUG after approximately 10 seconds because it was shown that the motor responses' of adults were more affected by the virtual environment rotation when they were asked to perform the movement after an exposure to the scene motion of 10 seconds[22].

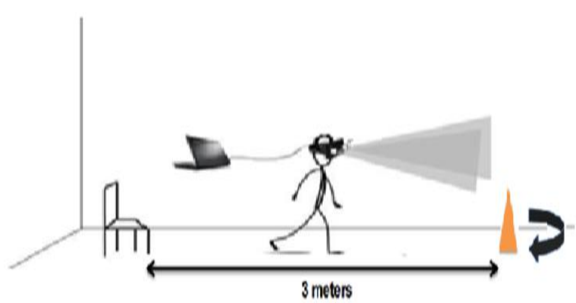

Figure 17 The Timed Up and Go test experimental setting

Figure 18 The visual scene presented to the adults via the HMD (left side of picture) and the cup of water held in the motor task (right side of picture)

The HMD consists of two displays, one for each eye, with a resolution of 960 x1080 pixels per eye, a maximum refresh of 75 Hz, and a weight of 440 grams. The viewing optics allows for a 100° viewing angle. An Ovrvision mount was used, which is a high-performance USB stereo camera customized for the Oculus Rift, which allows users to replace their current display with a view of the real world. The resolution for the Ovrvision is 640 x 480 per eye (1280 x 480), the frame rate is 60 FPS, the angle of view is H90 °, V75 °, the latency is 50 msec, the pixel number is 0.6 MP, and its weight is 55 grams.

To examine motor performance, seven Trigno™ wireless sensors (Delsys Inc.) were placed on the participants' sternum, lumbar, both wrists, and both shanks, and one sensor was placed on the cup of water. The sensors include a tri-axial accelerometer (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 3.5 mg); a tri-axial gyroscope (range 40 m, resolution 16 bit, sampling frequency 148 s/sec, noise < 0.05°/sec); and a tri-axial magnetometer (range 40 m, resolution 16 bit, sampling frequency 74 s/sec, Noise < 0.4 uT). Each sensor weight was 14.7 g. Kinematic data derived from the IMUs included three acceleration axes: anterior-posterior (AP) acceleration, vertical acceleration, and medio-lateral (ML) acceleration, and three angular velocity axes: pitch (rotation around the ML axis), yaw (rotation around the vertical axis), and roll (rotation around the AP axis).

Participants were asked to complete two trials of eight conditions in the following order: (1) TUG without a secondary task (TUG), (2) TUG with a motor task (TUG_{m}), (3) TUG while wearing the HMD without visual stimulus (TUG_{HMD}), (4) TUG while wearing the HMD without visual stimulus with the motor task (TUG_{HMD_m}), (5) TUG with visual stimulus dynamically pitching-up (TUG_{PU}), (6) TUG with visual stimulus dynamically pitching-down (TUG_{PD}), (7) TUG with visual stimulus dynamically pitching-up and motor task (TUG_{PU_m}), and (8) TUG with visual stimulus dynamically pitching-down and motor task (TUG_{PD_m}).

Data Analysis

All acceleration and angular velocity signals were analyzed using custom-written MatLab Codes (MathWorks, Natick, MA, USA). Each experimental condition was collected twice and averaged [137] and expressed as the mean and standard deviation f both conditions. To minimize signal distortion and noise, we filtered the accelerometer and gyroscope signals using a fourth order Butterworth low pass filter with a cut-off frequency of 2 Hz.[138] Consistent with previous studies, pitch angular velocity signal from the lumbar sensor was used when identifying sit-to-stand, stand-to-sit, and turning [118]. The transitions were determined as follows: 1) Sit-to-stand: first 10 °/sec and the second -10 °/sec were considered as the start and end of the sit-to-stand, respectively, 2) Stand-to-sit: last 10 °/sec and the last -10 °/sec were considered as the start and end of stand-to-sit, respectively [118], 3) Turning, both Turn 1 (around the cone) and Turn 2 (prior to sitting back down) was determined based on the yaw angular velocity signal at the lumbar sensor by first identifying the maximum angular velocity and its corresponding time [118,139] . The 20% of the maximum angular velocity points were then determined in order to demarcate the start and end of the turns[118,139], and 4) Walk 1 (from chair to cone) was determined to start the end of sit-to-stand period to the beginning of turn 1, respectively. The start and end of Walk 2 (from cone to chair) were identified as the end of turn 1 and the start of stand-to-sit, respectively[139]. We choose to include all steps taken by the participant in walking and to not exclude the first and last steps due to two reasons: (1) limited walking space when completing the TUG test (around 6 meters going back and forth from the chair) and (2) the differences in motor behavior that older adults may exhibit as they may encounter more challenges in gait termination and initiation compared to the younger adults [140,141]. All Matlab algorithm results were verified using a video analysis. For example, if the Matlab code failed to identify the start and end of turning in the peak yaw angular velocity, a confirmation would be obtained using the video analysis. This analysis involves counting the number of steps the participant took, which followed by visualizing the shank sensors pitch angular velocity signals to identify the correct indexes.

We analyzed the mean and standard deviation of the values of two trials/condition that was completed by each participant for all dependent variables. For example, if the first trial of TUG condition for turning cadence was $TUG_{FirstTrial}$=90 step/min and the second trial was $TUG_{SecondTrial}$ =80 step/min, then we calculated the mean and standard deviation of $TUG_{FirstTrial}$ and $TUG_{SecondTrial}$ values to represent TUG condition. The dependent variables were defined as follows:

a. Turning cadence (step/min) as the number of steps per minutes[115].
b. Gait speed in walking (m/sec) as the time it takes the individual to travel the distance.
c. Peak trunk velocity in sit-to-stand, turn, walk, and stand-to-sit (°/sec) (PTV) as the maximum angular velocity of the trunk around the mediolateral (ML) (pitch), vertical (yaw), and anteroposterior (AP) (roll) axes [114].
d. ML, vertical, and AP acceleration amplitude range (g) and jerk (g s^{-1}) of sit-to-stand and stand-to-sit: The range was calculated as the (maximum-minimum acceleration value) and the jerk was calculated the derivative of *acceleration* with respect to time[125].
e. The ML, vertical, and AP jerk of glass of water (g s^{-1}): as the derivative of *acceleration* with respect to time.

A mixed repeated measures of ANOVA (2 motor x 4 vision) with between factors as (aging and VD) was conducted on the mean and standard deviation of all dependent variables the jerk of glass of water, which was analyzed using repeated measures of ANOVA (4 motor for the jerk of glass of water) between (aging and VD). Bonferroni post hoc analysis was followed in case of main-effect significance with a corrected level of significance ($p<0.025$)[142].

Results

Older and younger adults' scores on the balance measures and cognitive function were comparable except in the Berg Balance Scale, in which older adults showed poorer performance compared to younger adults (p=.001). VI and VD adults were also comparable in the demographics and clinical measures (Table. 7). All raw data for dependent variables are presented in (Table 8 and 9)

Table 7, continues Participants demographics and clinical characteristics

	Young adults (n=12, 6M, 6F)	**Older adults (n=16, 8M, 8F)**	**P-value**	**VI Older adults (n=7, 4M, 3F)**	**VD Older adults (n=9, 5M, 4F)**	**P-value**
Age (years)	25.9 ± 3.9	69.0 ± 4.4	p<0.001 **	68.1 ± 4.3	69.6 ± 4.6	p=0.5
BMI	24.9 ± 4.6	24.0 ± 5.9	p=0.65	24.7 ± 4.7	23.3 ± 7.1	p=0.6
MMSE	29.1 ± 0.8	29.7± 1.0	p = 0.11	30.0 ± 0.0	29.5 ± 1.3	p=0.4
BBS	56±0.00	54.6 ±1.3	p =0.001 **	54.8 ± 1.3	54.4 ± 1.3	p=0.5
ABC	97.5 ± 4.7	94.1 ± 8.1	p =0.2	94.6 ± 4.3	93.6 ± 10.4	p=0.8
TUG (sec)	10.6 ± 1.5	10.4± 2.7	p=0.8	11.5 ± 3.6	9.6 ± 1.7	p= 0.1
FRT (inch)	12.8 ± 2.0	11.9 ± 5.7	p=0.6	14.1 ± 7.6	10.3 ± 3.3	p=0.2
Bucket test	0.6 ± 0.48	1.5 ± 1.2	p=0.01	1.5 ± 0.9	1.5 ± 1.5	p= 0.9

			**			
RFT	2.5 ± 14.8	3.7±8.1	p=0.78	0.9 ±1.3	5.9 ± 10.4	p=0.004
Note: Values are represented in mean ± SD. BMI (Body Mass Index), MMSE (Mini Mental State Examination), BBS (Berg Balance Scale), ABC (Activity-specific Balance Confidence), TUG (Timed Up and Go Test), FRT (Forward Reach Test), and RFT (Rod and Frame test)						

There were no significant differences in the mean and standard deviation of jerk of cup of water in all three axes (all ps ≥ 0.05).

Effects of adding a motor task

There was a main effect of adding a motor task. With the addition of a motor task, all adults walked slower [$p=0.04$, $\eta2= 0.15$] and showed a lower AP acceleration jerk in sit-to-stand ($p=0.001$, $\eta2=0.3$). In addition, they decreased their PTV around the ML axis in all TUG activities (all ps<0.003, $\eta2$ =0.3-0.6). There was a decrease in PTV around the vertical axis in all TUG activities except stand-to-sit (all ps<0.001, $\eta2$=0.3-0.7) and in the PTV around the AP axis in all TUG activities except sit-to-stand (all ps<0.006, $\eta2$=0.3-0.5).

In terms of variability, adults were less variable in turning cadence, PTV around the ML axis in sit-to-stand and stand-to-sit (ps <0.04, $\eta2$=0.1-0.3) and PTV around the

Table 8, continues Raw data for all dependent variables between younger (YA) and older adults (OA)

		TUG	TUG_m	TUG_{HMD}	TUG_{HMD_m}	TUG_{PU}	TUG_{PD}	TUG_{PU_m}	TUG_{PD_m}
Turning Cadence	**YA**	171.8±29.5	173.1±27.2	151.9±33.3	144.3±29.4	143.5±25.9	142.8±35.6	142.8±31.1	145.6±34.1
	OA	173.5±39.3	174.7±36.9	143.4±56.0	135.1±32.3	140.9±42.7	145.9±36.8	134.5±40.1	146.3±40.9
Gait Speed	**YA**	1.15±0.4	1.2±0.4	0.8±0.3	0.86±0.23	0.82±0.21	0.82±0.2	0.81±0.1	0.89±0.1
	OA	1.1±0.2	1.17±0.3	0.78±0.4	0.78±0.39	0.74±0.32	0.74±0.3	0.78±0.4	0.8±0.3
PTV around the ML axis in sit-to-stand	**YA**	90.3±21.9	72.6±13.1	81.4±15.9	69.9±14.8	80.9±14.1	75.7±10.8	73.2±12.9	75.5±15.1
	OA	101.5±14.1	83.1±18.6	77.6±14.6	68.8±17.6	78.2±16.3	76.2±18.1	70.2±18.2	71.2±16.1
PTV around the ML axis in turn	**YA**	59.8±25.7	53.1±17.9	53.6±16.5	47.7±14.3	53.4±13.1	49.5±13.8	47.7±14.3	49.3±9.5
	OA	53.3±15.1	45.0±12.9	42.5±14.3	36.5±13.9	43.1±16.1	43.5±17.7	39.1±12.6	38.6±13.1
PTV around the ML axis in walk	**YA**	52.1±18.1	46.1±18.0	47.8±17.5	42.9±16.5	45.3±14.8	44.3±15.2	46.7±13.9	48.6±15.6
	OA	42.7±13.8	40.1±14.7	37.2±11.1	33.8±11.8	37.4±11.1	35.6±13.7	34.1±12.0	35.2±11.8
PTV around the ML axis in stand-to-sit	**YA**	87.2±26.3	66.5±22.9	64.3±17.9	53.9±13.9	74.7±18.6	69.3±13.0	58.9±17.1	63.3±15.2
	OA	88.2±20.1	64.17.0	70.4±25.8	55.1±13.6	67.8±14.7	64.1±12.5	59.5±15.0	65.0±13.3
PTV around the vertical axis in sit-to-stand	**YA**	48.3±14.6	37.7±13.9	40.8±7.9	37.6±10.8	42.4±11.3	48.8±17.6	39.6±12.5	43.20.5
	OA	48.1±16.9	38.9±13.3	40.6±20.4	32.7±7.1	36.5±11.3	40.7±14.1	33.2±9.8	35.2±13.4

PTV around the vertical axis in turn	**YA**	168.2±31.8	156.1±35.2	134.8±50.5	123.3±49.3	127.1±38.3	122.3±42.4	118.6±37.6	120.8±34.7
	OA	148.9±32.4	134.6±26.5	110.8±33.3	101.9±24.1	110.9±28.1	108.8±26.0	100.3±26.4	101.0±26.3
PTV around the vertical axis in walk	**YA**	89.2±49.1	84.8±58.1	70.8±40.2	69.2±45.3	79.1±51.3	76.1±42.6	67.8±29.8	71.4±43.7
	OA	71.1±29.6	64.3±25.2	55.7±28.3	51.8±21.7	55.9±23.6	51.6±18.8	52.3±23.4	52.6±22.1
PTV around the vertical axis in stand-to-sit	**YA**	43.6±21.4	43.5±16.9	53.0±17.3	57.7±29.3	67.6±23.1	53.9±21.5	61.2±25.2	54.0±19.2
	OA	43.0±16.7	38.5±15.9	51.9±25.6	39.3±16.9	51.6±28.1	43.8±23.9	40.1±16.3	41.1±21.7
PTV around the AP axis in sit-to-stand	**YA**	37.5±20.3	26.5±7.1	27.7±11.1	25.3±7.3	25.8±8.2	27.4±8.6	27.1±10.4	28.9±15.7
	OA	29.8±9.1	31.5±16.5	24.1±9.8	26.1±18.8	24.8±10.1	24.0±9.4	27.9±17.6	27.7±18.3
PTV around the AP axis in turn	**YA**	60.9±33.9	51.7±26.1	49.3±24.5	47.4±24.4	49.2±20.6	53.2±25.7	45.3±17.6	48.7±22.1
	OA	43.8±13.6	40.1±13.1	36.4±11.1	32.8±9.6	37.2±11.7	35.6±9.9	32.1±8.9	33.3±10.3
PTV around the AP axis in walk	**YA**	39.0±18.5	37.1±22.5	35.2±23.3	31.9±21.7	35.8±21.1	35.6±16.8	32.1±14.4	36.0±20.1
	OA	31.6±12.9	26.1±10.9	28.1±10.9	25.1±11.1	26.1±8.8	26.1±8.3	24.1±8.5	24.5±8.3
PTV around the AP axis in stand-to-sit	**YA**	32.6±13.8	26.1±10.4	30.1±11.8	30.6±10.1	42.4±14.7	41.6±20.7	31.2±9.8	36.4±14.9
	OA	30.8±11.6	30.1±11.8	40.1±21.4	28.7±11.8	34.5±17.4	28.8±13.1	27.1±13.9	30.7±14.3
ML Acceleration Range of Sit-to-stand	**YA**	0.48±0.2	0.5±0.2	0.5±0.4	0.4±0.1	0.36±0.1	0.37±0.09	0.43±0.2	0.54±0.6
	OA	0.42±0.1	0.45±0.2	0.55±0.4	0.36±0.1	0.37±0.1	0.37±0.1	0.46±0.3	0.48±0.3
Vertical Acceleration Range of Sit-to-stand	**YA**	0.9±0.3	0.98±0.6	0.8±0.2	0.8±0.3	0.7±0.2	0.7±0.1	0.8±0.4	0.98±0.8
	OA	0.9±0.3	1.0±0.5	0.9±0.4	0.8±0.4	0.7±0.1	0.78±0.4	0.76±0.3	0.85±0.5

AP Acceleration Range of Sit-to-stand	**YA**	1.0±0.2	1.1±0.5	0.98±0.3	0.86±0.2	0.85±0.2	0.8±0.1	0.9±0.3	1.1±0.8
	OA	1.0±0.2	0.9±0.25	1.0±0.3	0.94±0.2	0.8±0.1	0.9±0.1	1.0±0.5	1.0±0.5
ML Acceleration Range of stand-to-sit	**YA**	0.36±0.1	0.54±0.4	0.42±0.1	0.36±0.1	0.47±0.1	0.49±0.3	0.45±0.1	0.4±0.1
	OA	0.57±0.6	0.46±0.4	0.45±0.2	0.5±0.7	0.38±0.1	0.37±0.2	0.36±0.1	0.4±0.3
Vertical Acceleration Range of stand-to-sit	**YA**	0.7±0.2	0.86±0.5	0.86±0.5	0.6±0.2	0.7±0.4	0.7±0.3	0.75±0.4	0.63±0.2
	OA	0.9±0.6	0.91±0.8	0.82±0.3	0.85±0.8	0.68±0.2	0.6±0.3	0.67±0.3	0.72±0.3
AP Acceleration Range of stand-to-sit	**YA**	0.9±0.3	1.2±0.8	0.95±0.3	0.8±0.2	0.9±0.36	0.89±0.3	0.9±0.5	0.91±0.3
	OA	1.2±0.5	1.0±0.4	0.99±0.3	1.1±0.7	0.83±0.3	0.9±0.2	0.9±0.4	0.95±0.3
ML Acceleration Jerk of Sit-to-stand	**YA**	0.04±0.02	0.04±0.03	0.029±0.03	0.07±0.06	0.04±0.03	0.04±0.03	0.03±0.05	0.06±0.04
	OA	0.04±0.02	0.03±0.03	0.04±0.04	0.05±0.05	0.03±0.03	0.03±0.03	0.05±0.03	0.038±0.03
Vertical Acceleration Jerk of Sit-to-stand	**YA**	0.1±0.06	0.12±0.1	0.1±0.12	0.11±0.1	0.11±0.12	0.13±0.09	0.12±0.09	0.12±0.12
	OA	0.1±0.05	0.09±0.08	0.06±0.06	0.07±0.06	0.07±0.07	0.06±0.04	0.06±0.04	0.07±0.06
AP Acceleration Jerk of Sit-to-stand	**YA**	0.37±0.07	0.27±0.1	0.3±0.08	0.32±0.08	0.32±0.07	0.33±0.07	0.3±0.07	0.3±0.1
	OA	0.36±0.1	0.34±0.1	0.3±0.1	0.26±0.1	0.31±0.12	0.28±0.1	0.25±0.1	0.26±0.1
ML Acceleration Jerk of stand-to-sit	**YA**	0.09±0.09	0.1±0.07	0.09±0.09	0.08±0.1	0.08±0.1	0.08±0.07	0.08±0.07	0.07±0.06
	OA	0.07±0.09	0.08±0.06	0.05±0.04	0.04±0.03	0.06±0.05	0.03±0.02	0.04±0.05	0.05±0.04
Vertical Acceleration Jerk of stand-to-sit	**YA**	0.3±0.2	0.28±0.1	0.21±0.1	0.2±0.1	0.18±0.1	0.2±0.1	0.17±0.1	0.24±0.1
	OA	0.3±0.1	0.26±0.2	0.14±0.1	0.15±0.1	0.18±0.09	0.16±0.09	0.17±0.1	0.2±0.1
AP Acceleration Jerk of stand-to-sit	**YA**	0.7±0.2	0.5±0.3	0.4±0.1	0.4±0.2	0.4±0.2	0.4±0.2	0.4±0.18	0.4±0.17
	OA	0.7±0.4	0.6±0.1	0.3±0.1	0.4±0.1	0.4±0.1	0.4±0.1	0.4±0.08	0.4±0.1
ML Acceleration Jerk of cup of water	**YA**	-	0.15±0.5	-	1.4±2.8	-	-	1.6±3.2	0.8±2.1
	OA	-	0.6±1.2	-	0.8±1.3	-	-	1.5±2.4	1.2±2.3
Vertical Acceleration Jerk of cup of water	**YA**	-	0.2±0.17	-	0.1±0.1	-	-	0.5±1.3	0.2±0.2
	OA	-	0.4±0.8	-	0.1±0.1	-	-	0.1±0.14	0.3±0.7

AP Acceleration Jerk of cup of water	**YA**	-	0.2±0.3	-	0.1±0.1	-	-	0.07±0.08	0.1±0.1
	OA	-	0.1±02	-	0.1±0.09	-	-	0.1±0.2	0.3±1.1
Note: Values are represented in mean ± SD. OA – older adults, YA – younger adults									

Table 9, continues Raw data for all dependent variables between older adults who are visually independent (VI) and visually dependent (VD).

		TUG	**TUG_m**	**TUG_{HMD}**	**TUG_{HMD_m}**	**TUG_{PU}**	**TUG_{PD}**	**TUG_{PU_m}**	**TUG_{PD_m}**
Turning Cadence	**VI**	164.6±39.8	156.7±37.1	138.6±42.2	123.1±20.1	122.1±36.7	126.8±34.9	126.2±27.4	133.7±34.8
	VD	180.3±39.8	188.8±31.8	147.1±67.1	144.3±38.0	155.4±43.1	160.7±32.7	140.9±48.4	156.1±44.4
Gait Speed	**VI**	1.1±0.2	1.1±0.2	0.7±0.4	0.7±0.4	0.6±0.3	0.6±0.3	0.7±0.4	0.7±0.5
	VD	1.1±0.2	1.2±0.3	0.8±0.4	0.9±0.4	0.8±0.3	0.7±0.3	0.8±0.4	0.8±0.4
PTV around the ML axis in sit-to-stand	**VI**	106.3±14.0	82.3±17.3	78.8±13.8	68.6±17.2	80.2±14.6	83.1±22.0	71.1±15.0	69.8±16.4
	VD	97.7±13.7	83.7±20.5	76.6±15.9	68.9±18.9	76.6±18.3	70.9±13.4	69.6±21.3	72.3±16.8
PTV around the ML axis in turn	**VI**	60.7±17.3	53.5±14.8	45.5±17.5	42.7±17.3	49.6±18.5	49.6±22.3	43.6±12.8	40.6±15.6
	VD	47.7±10.7	38.4±5.8	40.2±11.7	31.6±8.9	38.0±12.9	38.7±12.5	35.6±12.0	37.1±11.4
PTV around the ML axis in walk	**VI**	48.9±15.7	43.3±15.3	39.3±11.7	37.7±15.6	42.7±13.8	40.0±18.2	37.9±15.5	39.8±14.3
	VD	37.9±10.5	37.6±14.6	35.6±11.1	30.8±7.5	33.4±6.6	32.2±8.5	31.2±8.2	31.6±8.6
PTV around the ML axis in stand-to-sit	**VI**	86.4±22.6	64.3±13.6	61.9±16.1	51.6±9.1	63.1±14.1	64.4±15.0	57.1±11.5	65.5±13.6
	VD	89.5±19.1	64.5±20.1	77.0±30.7	57.7±16.3	71.5±14.8	63.7±11.2	61.4±17.7	64.6±13.9
PTV around the vertical axis in sit-to-stand	**VI**	55.0±17.5	42.0±10.1	42.0±12.3	35.4±7.4	41.6±11.6	44.2±18.3	38.1±10.5	39.8±13.0
	VD	42.6±15.3	36.5±15.4	39.6±25.8	30.7±6.5	32.5±9.9	38.0±9.9	29.3±7.6	31.6±13.3
PTV around	**VI**	147.7±26.1	130.0±21.5	103.8±25.3	98.1±23.3	104.8±21.5	107.1±24.5	95.2±22.1	95.1±22.3

the vertical axis in turn	**VD**	149.8±38.2	138. ±30.6	116.2±39.1	104.9±25.6	115.7±32.8	110.1±28.5	104.3±30.1	105.5±29.5
PTV around the vertical axis in walk	**VI**	70.5±25.3	57.1±15.6	55.5±21.4	53.4±19.0	54.6±14.5	53.1±18.4	50.2±21.2	50.6±21.1
	VD	71.6±34.1	70.1±30.4	55.8±34.1	50.5±24.7	56.9±29.7	50.4±20.1	53.9±26.1	54.1±24.1
PTV around the vertical axis in stand-to-sit	**VI**	44.7±18.2	39.3±21.8	50.2±26.7	45.3±20.3	50.6±27.7	48.1±30.9	43.6±18.5	47.0±26.6
	VD	41.7±16.4	37.8±10.9	53.1±26.3	34.7±13.2	52.3±30.1	40.5±18.1	37.4±15.0	36.4±17.3
PTV around the AP axis in sit-to-stand	**VI**	33.4±8.3	35.8±19.1	26.6±9.6	31.5±24.8	23.3±6.5	25.9±12.9	32.9±21.8	34.6±24.9
	VD	27.1±9.1	28.1±14.1	22.2±10.1	21.8±12.5	25.9±12.5	22.4±10.1	24.1±13.7	22.3±9.7
PTV around the AP axis in turn	**VI**	44.2±15.1	43.3±15.2	37.2±12.3	33.9±10.3	39.7±11.2	36.6±11.6	32.9±9.1	33.3±12.5
	VD	43.5±13.2	37.6±11.6	35.7±10.7	32.0±9.6	35.3±12.5	34.8±9.1	31.4±9.3	33.4±9.0
PTV around the AP axis in walk	**VI**	33.3±15.8	27.6±12.3	27.3±7.5	25.1±9.3	26.9±8.7	27.3±9.0	23.9±8.9	26.4±8.2
	VD	30.2±10.9	25.1±10.4	28.6±13.5	25.2±12.8	25.4±9.3	25.1±8.1	24.3±8.8	23.1±8.6
PTV around the AP axis in stand-to-sit	**VI**	30.7±10.3	27.6±10.8	41.3±23.7	35.3±14.2	37.8±18.5	34.05±16.4	33.4±14.7	35.7±16.7
	VD	31.0±13.2	23.4±8.4	39.3±20.9	23.6±6.8	32.0±17.2	24.8±8.6	22.2±11.6	26.9±11.7
ML Acceleration Range of Sit-to-stand	**VI**	0.5±0.2	0.57±0.2	0.7±0.5	0.4±0.09	0.4±0.2	0.4±0.2	0.5±0.4	0.7±0.45
	VD	0.4±0.1	0.35±0.1	0.39±0.1	0.3±0.1	0.3±0.1	0.3±0.1	0.4±0.26	0.3±0.16
Vertical Acceleration Range of Sit-to-stand	**VI**	1.02±0.36	1.3±0.73	1.01±0.45	1.1±0.6	0.7±0.2	0.97±0.53	0.9±0.46	1.1±0.8
	VD	0.86±0.27	0.76±0.2	0.8±0.4	0.7±0.15	0.7±0.14	0.62±0.12	0.65±0.25	0.65±0.12
AP Acceleration Range of Sit-to-stand	**VI**	1.1±0.2	1.1±0.35	1.2±0.45	1.1±0.38	0.9±0.14	0.97±0.16	1.2±0.7	1.2±0.7
	VD	0.9±0.15	0.9±0.1	0.8±0.13	0.83±0.1	0.87±0.15	0.87±0.1	0.8±0.18	0.8±0.1
ML Acceleration Range of stand-to-sit	**VI**	0.6±0.7	0.3±0.1	0.39±0.2	0.7±1.1	0.3±0.15	0.3±0.2	0.37±0.1	0.45±0.32
	VD	0.5±0.46	0.57±0.59	0.5±0.24	0.3±0.15	0.4±0.18	0.4±0.24	0.35±0.19	0.43±0.35
Vertical Accelerati	**VI**	0.97±0.8	0.67±0.2	0.7±0.2	1.03±1.3	0.6±0.25	0.5±0.14	0.7±0.4	0.7±0.37

on Range of stand-to-sit	**VD**	1.01±0.5	1.1±1.1	0.9±0.4	0.7±0.17	0.7±0.3	0.7±0.45	0.66±0.25	0.73±0.3
AP Acceleration Range of stand-to-sit	**VI**	1.3±0.6	0.9±0.2	0.96±0.3	1.3±1.1	0.7±0.2	0.89±0.2	1.1±0.5	1.1±0.4
	VD	1.1±0.4	1.1±0.6	1.0±0.3	0.8±0.2	0.9±0.35	0.9±0.27	0.8±0.1	0.86±0.22
ML Acceleration Jerk of Sit-to-stand	**VI**	0.05±0.02	0.03±0.03	0.02±0.04	0.04±0.02	0.04±0.03	0.1±0.3	0.04±0.02	0.04±0.04
	VD	0.03±0.02	0.03±0.02	0.04±0.03	0.05±0.06	0.03±0.02	0.2±0.5	0.06±0.04	0.04±0.04
Vertical Acceleration Jerk of Sit-to-stand	**VI**	0.1±0.07	0.09±0.09	0.06±0.06	0.06±0.04	0.07±0.05	0.04±0.02	0.05±0.03	0.045±0.03
	VD	0.1±0.03	0.1±0.09	0.06±0.06	0.07±0.08	0.067±0.08	0.08±0.05	0.07±0.04	0.09±0.07
AP Acceleration Jerk of Sit-to-stand	**VI**	0.35±0.2	0.3±0.1	0.28±0.1	0.25±0.1	0.27±0.1	0.27±0.12	0.22±0.11	0.22±0.13
	VD	0.26±0.08	0.34±0.1	0.32±0.1	0.27±0.1	0.34±0.12	0.29±0.11	0.28±0.12	0.3±0.1
ML Acceleration Jerk of stand-to-sit	**VI**	0.06±0.08	0.08±0.07	0.04±0.04	0.02±0.01	0.06±0.07	0.03±0.01	0.02±0.02	0.06±0.03
	VD	0.07±0.09	0.09±0.06	0.05±0.05	0.05±0.04	0.05±0.05	0.04±0.03	0.06±0.06	0.04±0.05
Vertical Acceleration Jerk of stand-to-sit	**VI**	0.3±0.2	0.25±0.2	0.1±0.05	0.12±0.05	0.16±0.08	0.12±0.06	0.16±0.09	0.15±0.1
	VD	0.29±0.1	0.3±0.2	0.2±0.1	0.2±0.1	0.2±0.1	0.19±0.1	0.18±0.1	0.24±0.17
AP Acceleration Jerk of stand-to-sit	**VI**	0.8±0.58	0.6±0.16	0.37±0.16	0.36±0.13	0.39±0.1	0.47±0.06	0.43±0.07	0.39±0.1
	VD	0.6±0.18	0.6±0.17	0.36±0.13	0.45±0.09	0.46±0.1	0.39±0.2	0.4±0.1	0.46±0.17
ML Acceleration Jerk of cup of water	**VI**	-	0.87±1.2	-	1.1±1.8	-	-	0.9±1.6	1.7±2.1
	VD	-	0.47±1.2	-	0.7±1.01	-	-	2.1±2.9	0.8±2.5
Vertical Acceleration Jerk of cup of water	**VI**	-	0.6±1.3	-	0.1±0.1	-	-	0.08±0.08	0.1±0.1
	VD	-	0.1±0.2	-	0.1±0.2	-	-	0.1±0.18	0.4±1.0
AP Acceleration Jerk of cup of water	**VI**	-	0.1±0.2	-	0.1±0.1	-	-	0.1±0.08	0.1±0.1
	VD	-	0.1±0.1	-	0.08±0.08	-	-	0.15±0.2	0.6±1.5

Note: Values are represented in mean ± SD. OA – older adults, YA – younger adults vertical axis in turning ($p=0.003$, $\eta 2 = 0.2$) when holding a cup of water compared to the conditions where they were not holding the cup.

Effects of Wearing the HMD

An effect of wearing the HMD was found, independent of adding a visual stimulus. While wearing the HMD, adults turned with a lower turning cadence [$p<0.001$, $\eta 2= 0.4$], walked slower [$p<0.001$, $\eta 2= 0.5$, power=1.00], decreased the PTV around ML axis in all TUG activities (all ps <0.003, $\eta 2 = 0.3-0.4$), PTV around vertical axis in all TUG activities except sit-to-stand (all ps<0.002, $\eta 2=0.2-0.7$), and PTV in all TUG activities except stand-to-sit (all ps<0.4, $\eta 2=0.1-0.2$).

Sit-to-stand and stand-to-sit transitions were also influenced by the wearing of HMD. Specifically, participants showed a smaller vertical acceleration range in sit-to-stand ($p=0.02$, $\eta 2 = 0.1$) and AP acceleration range in stand-to-sit ($p=0.3$, $\eta 2=0.1$). In addition, there was a decrease in the AP and vertical acceleration jerk of stand-to-sit (ps<0.02, $\eta 2=0.1-0.3$).

There was a decrease in the AP acceleration jerk in sit-to-stand ($p=0.01$) and PTV around the AP axis in turning ($p=0.004$) when wearing the HMD compared to the standard TUG test. Compared to the standard TUG, adding a visual stimulus that rotates in a *pitch up direction* caused an increase in the PTV around the vertical axis in stand to sit ($p=0.01$) and a decrease in the PTV around the AP axis in turning ($p=0.001$) (Figure 19 and 20).

Adults were less variable when wearing a HMD than standard TUG in turning cadence (p=0.03, η2=0.1), PTV around the vertical axis in turning (p=0.049, η2=0.1), PTV around the ML axis in stand-to-sit (p=0.045, η2=0.1), and ML and AP acceleration jerk of stand-to-sit (ps<0.02, η2=0.1-0.3).

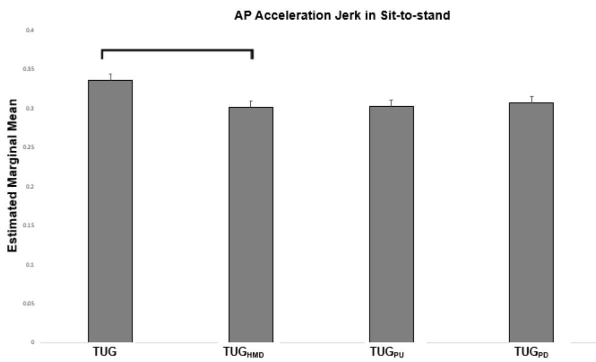

Figure 19 Anterioposterior Acceleration Jerk in sit-to-stand between the four visual conditions

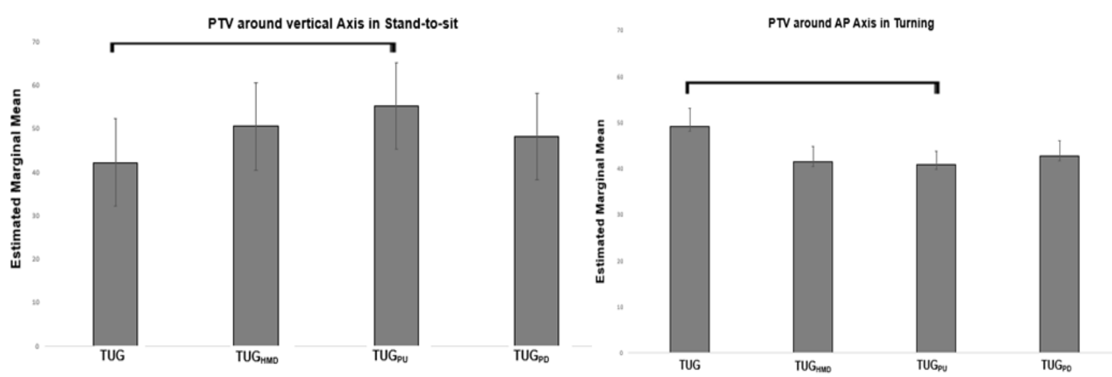

Figure 20 Peak trunk velocity around the vertical axis in stand-to-sit (left side) and around the AP axis in turning (right side) between the four visual conditions

Aging and VD

As shown in Figure 21, there was a main effect of aging on PTV in turning compared to younger adults. Older adults decreased their PTV around AP axis in turning [$p=0.03$, $\eta2= 0.16$,]. Additionally, compared to VI older adults, VD older adults showed significant decrease in the ML and vertical acceleration range in sit-to-stand [$p<0.049$, $\eta2= 0.25$-0.27] (Figure 22).

Compared to younger adults, older adults decreased their variability in turning cadence ($p=0.02$, $\eta2=0.1$), PTV around the vertical axis in turning ($p=0.04$, $\eta2=0.1$), and the ML and AP acceleration jerk in stand-to-sit ($p<0.03$, $\eta2=0.1$-0.3). Older adults who were visually dependent showed a less variability in the ML and vertical acceleration range of sit-to-stand ($p=0.01$, $\eta2=0.3$) compared to older adults who were visually independent.

No main effect of aging and visual dependency was found on other variables. (all $p \geq 0.06$).

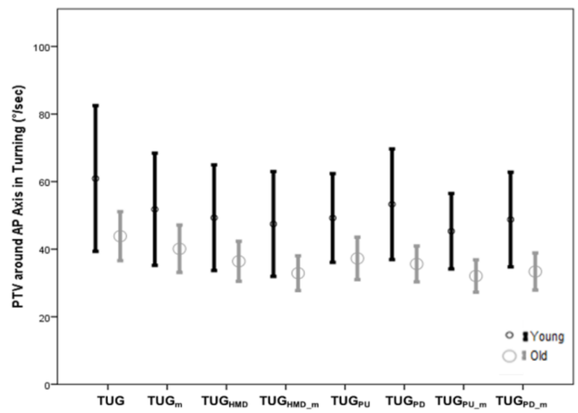

Figure 21 Peak trunk velocity around the AP axis “pitch” of younger and older adults in turning

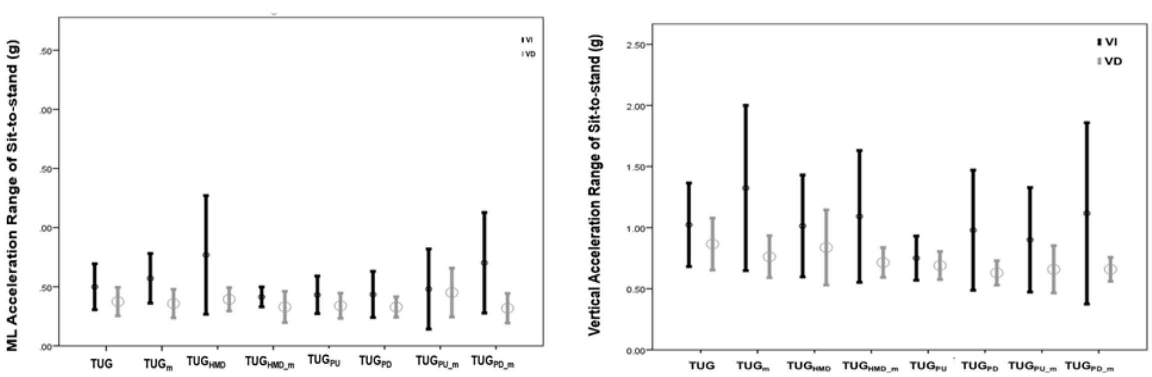

Figure 22 Mediolateral (left side) and vertical (right side) acceleration range of sit-to-stand of VI and VD older adults

Discussion and Conclusion

This work explored the effects of adding a motor task and a visual stimulation on the kinematics of adults while navigating through four dynamic tasks. The results indicated that: (1) Increasing the visual demands through wearing the HMD was associated with a decline in motor performance more than the motor task (2) Older adults exhibited a decrease in trunk velocity around the AP axis in turning compared to younger adults, and (3) VD older adults showed a decrease in the ML and vertical acceleration range of sit-to-stand compared to VI older adults.

Results showed that wearing the HMD with or without the visual stimulation can degrade the mobility of adults more than adding a motor task. Consistent with these results, Porciuncula et al. reported that a motor task of carrying a cup of water did not impact the motor behavior of older adults as much as does performing more complex cognitive-motor tasks [17]. Our results also indicate that holding the cup of water decreases the variability of turning cadence and trunk velocities in sit-to-stand, stand-to-sit, and turning, while wearing the HMD decreases the variability of turning cadence and the range and smoothness of sit-to-stand and stand-to-sit. It was not surprising to find that the motor task did not affect as many postural responses as wearing the HMD. Wearing the HMD presents perceptual issues to the users in terms of added weight on the head, limited peripheral vision, optokinetic calibration, and inaccurate estimation of subject-to-object and object-to-object distance [158,159]. The adults were exposed to an unrealistic visual environment while performing the TUG motor tasks.

Adding a visual perturbation with the HMD did not seem to affect mobility compared to wearing the HMD without a visual scene in most of the variables. However, adults did seem differentially affected by the direction of a pitch up rotation of the visual scene in turning and stand-to-sit motions. All participants showed a decrease in peak trunk velocity during turning around the AP axis with a pitch up visual scene in comparison to turning during standard TUG test. Adults choose to decrease the flexibility of their trunk as a strategy to ensure their safety while preventing the body from moving backward in the direction of the visual scene in a pitch up rotation. Similarly, an increase in the peak trunk velocity around the vertical axis was noted with the pitch up visual scene in stand-to-sit than during the standard TUG test. This result was not surprising as the individuals might choose to increase their trunk rotation to gain more visual feedback to where the chair was located before sitting down, which could be more challenging to achieve when the visual stimulus was added. Also, it is possible that this observation was due to conditions' fixed order that was presented to the sample. A dynamically visual stimulus pitching- up was presented before pitch down rotation; thus, an improvement in the performance may have occur accordingly.

Aging was associated with a decreased trunk velocity in turning, i.e. in roll axis. In addition, older adults showed lower variability in turning cadence, PTV around vertical axis in turning, and the jerk of stand-to-sit than younger adults.   The turning part of the TUG requires more navigational skills in older as compared to younger adults and with straight walking as compared with turns; thus, is translated into a poorer performance in the older adults as compared with younger adults. Turning is achieved using a top-down synergy that starts with a visual saccade [168,169]. Aging is associated with a delay in initiating visual saccade [53], which could affect motor planning while turning. Porciuncula et al. [20] found that younger and older adults were comparable in the peak trunk velocity but the elders showed more difficulty in the spatiotemporal measures in walking and turning when they added a cognitive-motor task but not a motor task only. In the current study, older adults showed a decline in the PTV in turning only, which could be due to the nature of the secondary tasks that were used that required motor-visual control. It is possible that the elderly were required to shift their attention more to the secondary tasks when they needed to change the direction of the body to turn from walking as compared to other TUG activities.

Increased reliance on the visual inputs in older adults was linked with a decrease in the ML and horizontal acceleration range and variability of sit-to-stand. Previous reports showed that VD adults tend to up-weight visual information and down-weight information from other sensory systems under conditions of visual perturbation[170], have high visual flow sensitivity, [171]and greater postural instabilities under visual manipulation conditions[19]. During the TUG, our VD older participants showed smaller ML and vertical acceleration range in sit-to-stand compared to VI individuals, which could be a preferred mechanism to ensure their safety in this activity. The primary objective from sit-to-stand activity is to vertically orient the body in space, and since VD adults showed more difficulty to estimate verticality in the environment [43,171], this could affect their range and limit their mobility in sit-to-stand. This result may indicate that VD older adults find it challenging to accelerate in sit-to-stand more than decelerating in stand-to-sit task.

In conclusion, these results provide an insight into the unique motor behavior for young and old adults who are VD when they navigating dynamically in a visually busy environment. The findings suggest that dividing visual-attention has a greater impact on the postural responses than adding a motor manipulation. Subtle differences in the trunk kinematics observed with aging in turning and sit-to-stand could point to a greater challenge to perform these activities more than other dynamic tasks of the TUG. Increased reliance on vision might be linked with postural instability when moving dynamically from sit-to-stand motion to walking.

CHAPTER 5

DISCUSSION

Review of Specific Aims

The overall purpose of this body of work was to examine the aging-related decrements that are associated with multitasking conditions when completing different dynamic activities that are part of the Timed Up and Go (TUG) test. This study used two forms of multitasking. First, a motor functional task that is used commonly in daily activities. Second, a visual stimulus that was presented by wearing a head mounted display (HMD) device, which overlays the normal view of the room. This work was guided by the following specific aims:

Specific Aim I. To determine the influence of wearing an HMD with and without presentation of augmented virtual stimulation during the TUG activities in old and young adults.

Specific Aim II. to explore how aging affects the multitask cost of the time of TUG and its activities when combined with a motor task and/or attending to a visual stimulus.

Specific Aim III. To explore the effects of aging and visual dependency on motor performance under increased stabilization demands by adding a motor task and a visual stimulus to TUG sub-components.

Summary of Results

A total of twelve younger adults (6 males) and sixteen older adults (8 males) were included in this work. Of the older participants, there were seven visually independent (VI) (4 males) and nine visually dependent (VD) (5 males). The demographic and clinical measures between younger and older adults were comparable except in the Berg Balance Scale (BBS) [140] and the bucket test [154]. Older adults showed significantly lower scores compared to younger adults (56±0.00 versus 54.6 ±1.3 in BBS and 0.6 ± 0.48 versus 1.5 ± 1.2 in bucket test). VI and VD older adults were comparable in all demographics and clinical measures. All participants were independent walkers and free from neurological and cognitive impairments at the time the study was conducted.

Aim I

To further explore the influence of the HMD wear on posture, I investigated the effects of wearing an HMD with and without attending to a rotating visual stimulus on TUG kinematics in a group of young and older adults in chapter 2. Adding a visual stimulus did not affect the TUG kinematics more than wearing the HMD without the visual stimulus, except in a few variables. This may suggest that the wearing of an HMD has an effect on posture regardless of adding any visual perturbation. It was found previously that wearing the HMD would limit the peripheral visual inputs, increase the weight on the head and thus change the neck and head posture, lead to inaccurate distance estimation [132]. This work adds to the body of literature by investigating the influence of HMD wear on other kinematic variables while transitioning dynamically within the environment.

The results from chapter 2 do not support that the addition of a visual stimulation to an HMD would cause a greater decline in motor responses compared to wearing the HMD without visual stimulation. Rather, it documents that wearing the HMD without a visual stimulus can decrease the turning cadence, slow the gait velocity, and decrease the trunk kinematics in dynamic activities. Previous research has viewed the HMD as a promising tool to be used in balance training for neurologically impaired individuals [172]. If we are looking to adopt these technological tools in clinical research, then, understanding their influence on the posture of users, especially in those who are fragile, becomes an important factor.

Although most of the motor response alteration occurred due to the wear of HMD, the addition of the visual stimulation influenced few variables in this study, particularly in walking and sit-to-stand. Viewing a rotated visual stimulus has been shown previously to increase postural sway, shorten step length, and decrease the trunk and head kinematics [19,22,43]. We found that adults tend to stiffen their posture in walking and limit their range in sit-to-stand when viewing a visual stimulation. This could be due to two reasons. First, in walking and sit-to-stand the participants were not challenged by the peripheral visual field limitation as in turning and stand-to-sit because they were not required to locate the cone or the chair, thus, they might be attending more to the visual stimulus effect, and hence, more affected by it. Second, sit-to-stand and walk were presented to all adults prior to turn and stand-to-sit, as per TUG protocol; therefore, an improvement in performance might occur in later activities. It will be interesting if future research answered this question while randomizing the order of the TUG activities. This will give a better insight into whether the effect seen in this study was due to the fixed order of TUG activities or inherent challenges in walk and sit-to-stand tasks when coupled with visual stimulation.

Compared to younger adults, older adults demonstrated a lower PTV around the AP axis in turn. It was shown previously that older adults exhibited more decrements in turning compared to older adults [86]. They used a multi-segmental top-down strategy that starts with orienting the head and gaze toward the direction of the turn to collect the essential visual feedforward information, which followed by orienting the trunk and medio-lateral foot deviation [173]. In this study, older adults were further challenged by the wear of HMD that limit their peripheral visual field and threaten their safety as they try to negotiate the cone or the chair while turning. Consequently, older adults may stiffen their trunk to ensure safe turning and prevent postural instabilities.

Aim II

In the second chapter of this dissertation, I focused on exploring the effects of multitasking in form of adding a visual stimulus through a head mounted display (HMD) and a motor task on the multitask cost of the TUG test and its activities in young and old adults. Previous works have looked into the effects of increasing the attentional demands in the TUG test using cognitive or motor tasks[154,155]. However, even with those modifications, TUG still has a poor identification of the balance problem that the individual may have [13], and therefore, I aimed to manipulate the visual system using this test. This study was the first to explore the influence of manipulating the visual system, in addition to a motor task, on the multitask cost of the TUG activities duration.

Wearing the HMD with or without an augmented visual stimulus increased the time taken to complete the TUG activities more than holding a cup of water, particularly in turning, walking, and stand-to-sit activities. Wearing the HMD can be more challenging than the motor task due to several factors such as limited peripheral vision, added weight on the head, and modified neck and trunk posture of the users[130–132]. The functionality of the used motor task could contribute to this result as it has been shown previously that similar motor task can improve the ability to indicate verticality in comparison to holding a joystick [56,58]. In addition, it was found that holding a pole with relaxed arms decreased the center of pressure dispersion while maintaining the balance in a tandem stance task [174]. It is possible to speculate then that due to the functionality of the used motor task in this study contribute to the finding in this work, i.e. fewer decrements associated with the motor task.

Older adults showed a decline in the multitask cost of the duration of TUG activities, particularly in turning and sitting-to-standing, compared to younger adults. Older adults have shown more difficulty in multitasking while turning more than straight walking[98]. Further, peripheral vision limitation in turning can produce motor incoordination of the body segment rotation[101], which may contribute to high attentional demands while turning by the elders, especially with an HMD, more than younger adults who might be more flexible to turn with this limitation. Aging was associated with an increased sit-to-stand time [67]. In this study, older adults stood up from an armless chair; thus, they had to ensure their safety while placing the center of mass within the base of support as they attain a vertical position. Older adults tend to place their feet backward and buttock forward as they increase their dependence on knee extensor muscle and decrease their dependence on horizontal trunk momentum[67]. Due to the limitations that older adults may encounter to complete a turn or sit-to-stand that were documented through this work, these activities could be of a great interest to study when assessing the risk of fall.

Aim III

In the fourth chapter, I explored the effects of aging and visual dependency on motor behavior under increased stabilization demands in the form of adding a motor task and/or visual stimulation to the different activities of the TUG test. This study, to my knowledge, is the first to explore the effects of adding a visual stimulus to the kinematics of older adults who are visually dependent.

Increasing the visual attentional demands in the form of wearing the HMD, regardless of the addition of a visual scene, was associated with a decline in TUG kinematics compared to the motor task. Users can experience several biomechanical and perceptual issues while wearing the HMD such as the added weight of the device, limited peripheral vision, optokinetic calibration, and inaccurate estimation of subject-to-object and object-to-object distances[132,175]. As a result, this can create challenges to the users' ability to use and switch between the spatial frames of reference.

Similar to chapter 2, I found that adding a rotating visual stimulus in a pitch up direction affected turn and stand-to-sit. Specifically, adults increased trunk rotation to find the chair before sitting (increase in PTV around vertical axis), and stiffen their trunk while turning (decrease in PTV around AP axis). These responses seem intuitive as the individuals used these postural compensations to *overcome* (1) the limitation in peripheral visual field, created by the HMD wear, to locate the chair in stand-to-sit and (2) the tendency to go backward with the pitch up rotated visual stimulus and losing their balance in turning, i.e. vection effect[149] (i.e. self-motion induced by the visual stimulus motion). Still, we cannot rule out the possibility of an order effect since a pitch up rotated scene was presented prior to pitch down. Previous results suggest that repeated exposure to a visual stimulus can result in adaptation in performance. For instance, in walking to a visible target at the ground, adults decreased their heading errors when exposed to a rotating visual stimulus within 10 trials[176]. Thus, a possibility of learning effect in the latest trials (with pitch down visual stimulus) may occur in this work.

In older adults, it seems that they faced greater challenges to complete turning and sitting to standing parts of the TUG. Aging was linked with a decrease in variability in turning and stand-to-sit and a lower trunk velocity in turning. Decrease in variability in dynamic movements is linked with functional level impairments and can reflect a restricted motion when transferring between activities. For example, a lower acceleration standard deviation in sit-to-stand was found in groups of individuals who are at high risk of fall and those with Parkinson's disease [120,125]. Although the functional level of older adults in our sample was comparable to younger adults, wearing the HMD that limit the peripheral visual inputs may play a role in restricting their movements while transferring from one dynamic activity to another. It was suggested that the peripheral visual inputs plays a role in decreasing the postural sway [147], thus, a limitation to these inputs while navigating dynamically in the environment could present more challenges to the user.

Further, older adults who were visually dependent showed smaller mediolateral and vertical acceleration range and variability in sit-to-stand compared to older adults who were visually independent. This suggest that individuals with increased reliance on vision may face some challenges to accelerate the body in sit-to-stand motion, especially when the HMD was worn. Sit-to-stand requires the individuals to produce an acceleration to attain a vertical position while aligning the center of mass over the base of support. VD older adults have inaccurate estimation of the verticality in their environment generally

[19,171] and exhibit more postural instabilities under perturbed visual inputs [19,43]. Due to these reasons, they may experience more difficulties and limit their displacement while attaining a vertical position in sit-to-stand. However, prior to generalizing this result, future research should investigate different cohorts who are visually dependent, including young adults.

Strength and limitations

As inherent in all research, the studies included in this dissertation have methodological strengths and limitations. An important strength to this work is that it is the first to explore the effects of manipulating the visual system while completing different dynamic tasks of the TUG test in younger and older adults. A novel method was implemented by presenting a virtual scene while completing the activities that can be used commonly in daily life during a clinical test.

As for limitations, the order of conditions was fixed throughout the experiment. Although this study allowed each participant to practice the HMD with the visual stimulus prior to data collection, presenting the conditions in a specific order for all the participants might introduce an order effect and allow them to perform better in the last conditions. Second, the small sample size of visually dependent younger adults (n=3) compared to the visually dependent older adults (n=9) limited our analysis of visual dependency in older adults only. Finally, we did not screen the participants for visual acuity, confounding musculoskeletal impairments, sensory/motor neuropathy or vestibular dysfunction. It is possible that the participants who have vestibular dysfunction can navigate differently with the HMD by fixing the gaze on the cone or the chair and limiting their head motion, especially in dynamic activities such as turning or sit-to-stand. Future work should consider these limitations to clarify a clearer conclusion.

Future Research

This is the first study to examine the effects of adding a functional motor task and/or wearing the HMD with an augmented virtual scene on the dynamic activities of the TUG test in younger and older adults. Future work should consider three factors. First, randomizing the order of the conditions of wearing the HMD, with and without the augmented virtual scene, would be beneficial in determining the effects of adding the visual flow without introducing an order effect in the experiment. Second, including an equal sample size of younger and older adults, who are visually dependent and independent, would be helpful to draw a more accurate conclusion about the effect of visual dependency on motor behavior under visual perturbation. Third, prior screening for sensory/motor neuropathy, visual acuity, and vestibular function would be useful to determine if certain individuals utilize different strategies to navigate using the HMD, due to specific visual or vestibular dysfunction. Four, exploring if the motor responses would be similar if the TUG was performed as one single continuous movements pattern versus performing the TUG activities as discrete movements. It will also be interesting to understand if individuals will exhibit the same motor responses if the TUG activities were presented in a different order (e.g. walk, turn, stand-to-sit, sit-to-stand versus the regular order of TUG test).

To conclude, older adults exhibit an increased reliance on the visual system [43], postural instabilities under misleading visual conditions [16], and difficulty in multitasking [131] that is linked to the increased risk of fall. All of these factors motivated this work, which builds upon previous literature by exploring the impact of multitasking in form of adding a motor task and attending to a rotating visual stimulus via an HMD device while completing the activities of the TUG test. This work suggests that the wearing of an HMD has an impact on TUG kinematics that should be accounted for in clinical research. Assessment of spatiotemporal measures in turning and sit-to-stand could be of a great interest to assess the risk of fall. The ability to perform sitting-to-standing could distinguish subgroups in older adults who are more sensitive to visual manipulation. The results of this work help to understand the sensory reweighting abilities of young and old adults under visually challenging conditions. This could help to better target interventions using sensory experience training to assess or improve balance.

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APPENDIX A

REPRESENTATIVE RAW DATA AND GENERAL METHODS

This appendix includes a description of the methods used in this work and a representative data from one young and one older participant. This includes the raw and filtered signals of the pitch and yaw angular velocity, which were obtained from the lumbar sensors and used to identify the TUG activities (Figure 23 and 24).

All signals were analyzed using custom-made Matlab Codes (MathWorks, Natick, MA, USA). A low pass Butterworth filter was applied with a cut off frequency of 2 Hz.

Like other previous studies, we elected to choose the lumbar sensor to identify sit-to-stand, stand-to-sit, and turning [118]. To identify sit-to-stand and stand-to-sit, we analyzed the pitch angular velocity signal at the lumbar sensor. The first 10°/sec in this signal and the second -10 °/sec were considered as the start and end of sit-to-stand, respectively. The last 10 °/sec and -10 °/sec in the signal were considered as the start and end of stand-to-sit, respectively [118].

We analyzed turning using the yaw angular velocity signal at the lumbar sensor. First, we identified the maximum two peaks in the yaw angular velocity and its corresponding time. Then we identified the two points that were 20% of the 1st peak as the start and end of turn 1. Turn 2 was identified similarly using the 2nd peak [118,144].

The start and end of walk 1 were selected as the end of sit-to-stand and the beginning of turn 1, respectively. The start and end of walk 2 were identified as the end of turn 1 and the start of stand-to-sit, respectively[144].

This study analyzed the following dependent variables:

a. Time of total TUG test and its activates (sec)

b. Turning cadence (step/min) as the number of steps per minutes [115].

c. Gait speed in walking (m/sec) as the time it takes the individual to travel the distance

d. Peak trunk velocity (°/sec) (PTV) as the maximum angular velocity of the trunk around the mediolateral (ML) (pitch), vertical (yaw), and anteroposterior (AP) (roll) axes [113].

e. Mediolateral (ML), horizontal, and Anteroposterior (AP) acceleration amplitude range (g) and jerk (g s^{-1}) of sit-to-stand and stand-to-sit: The range was calculated as the maximum-minimum acceleration value and the jerk was calculated from the derivative of *acceleration* with respect to time [114].

f. The ML, horizontal, and AP jerk of glass of water (g s^{-1}): as the derivative of *acceleration* with respect to time.

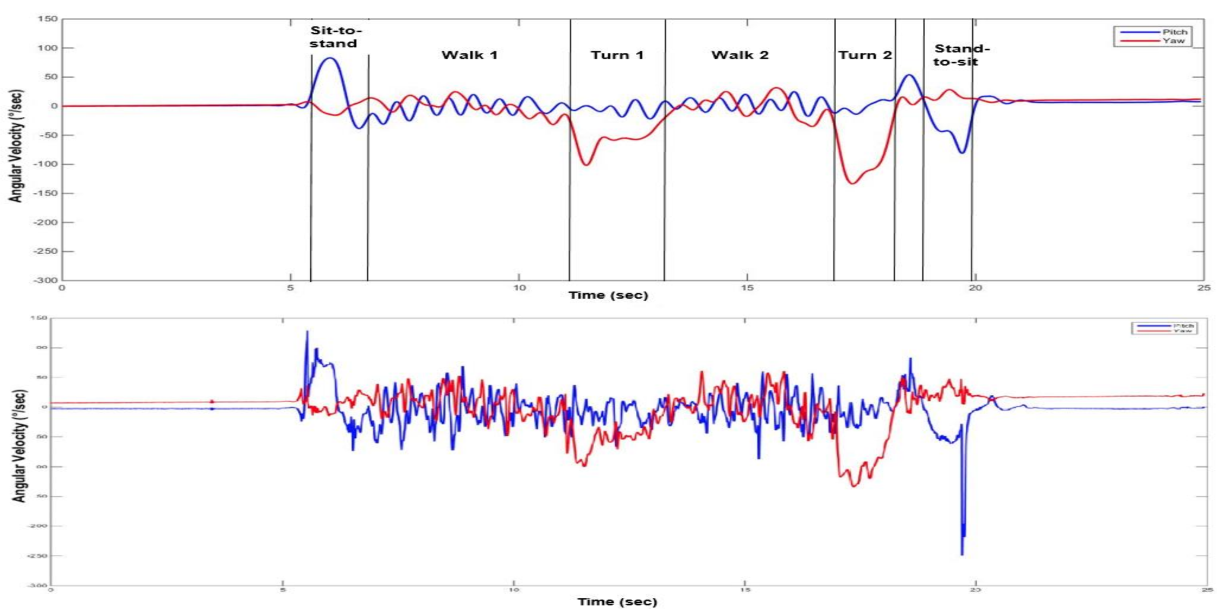

Figure 23 Raw and filtered signals of the pitch (blue) and yaw (red) angular velocity signals (°/sec) that are obtained from the lumbar sensor from one younger participant. The upper picture represents the filtered signals

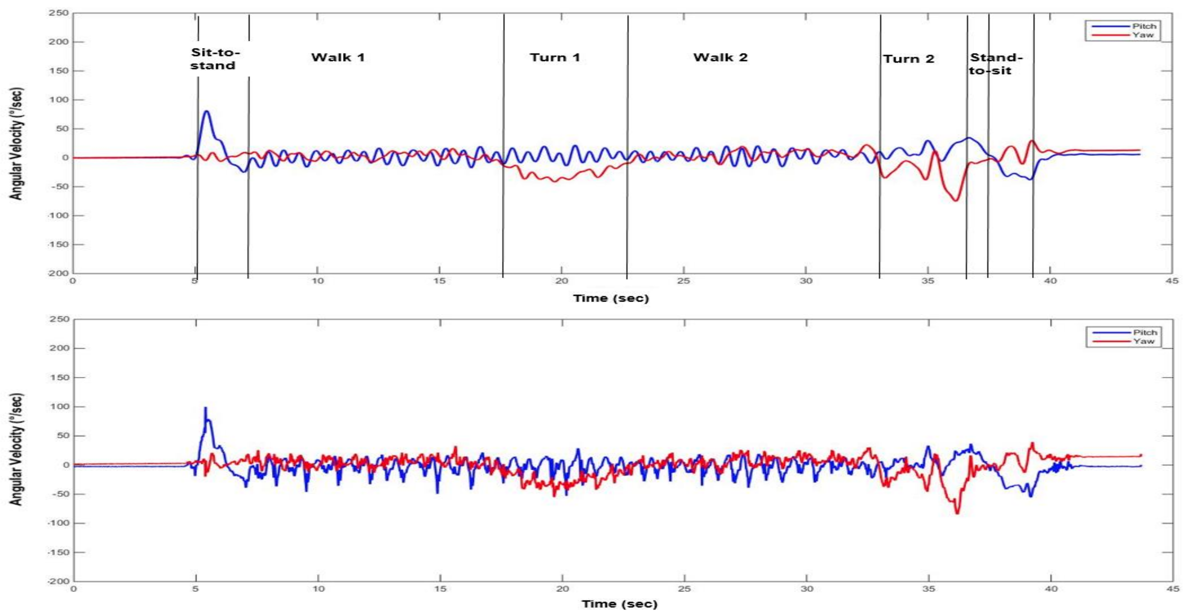

Figure 24 Raw and filtered signals of the pitch (blue) and yaw (red) angular velocity signals (°/sec) that are obtained from the lumbar sensor from one older participant. The lower picture represent the unfiltered signals