

Vibrotactile Sensory Augmentation and Machine Learning Based Approaches for Balance Rehabilitation

by

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To my family

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Abstract

Vestibular disorders and aging can negatively impact balance performance. Currently, the most effective approach for improving balance is exercise-based balance rehabilitation. Despite its effectiveness, balance rehabilitation does not always result in a full recovery of balance function. In this dissertation, vibrotactile sensory augmentation (SA) and machine learning (ML) were studied as approaches for further improving balance rehabilitation outcomes.

Vibrotactile SA provides a form of haptic cues to complement and/or replace sensory information from the somatosensory, visual and vestibular sensory systems. Previous studies have shown that people can reduce their body sway when vibrotactile SA is provided; however, limited controlled studies have investigated the retention of balance improvements after training with SA has ceased. The primary aim of this research was to examine the effects of supervised balance rehabilitation with vibrotactile SA. Two studies were conducted among people with unilateral vestibular disorders and healthy older adults to explore the use of vibrotactile SA for therapeutic and preventative purposes, respectively. The study among people with unilateral vestibular disorders provided six weeks of supervised in-clinic balance training. The findings indicated that training with vibrotactile SA led to additional body sway reduction for balance exercises with head movements, and the improvements were retained for up to six months. Training with vibrotactile SA did not lead to significant additional improvements in the majority

of the clinical outcomes except for the Activities-specific Balance Confidence scale. The study among older adults provided semi-supervised in-home balance rehabilitation training using a novel smartphone balance trainer. After completing eight weeks of balance training, participants who trained with vibrotactile SA showed significantly greater improvements in standing-related clinical outcomes, but not in gait-related clinical outcomes, compared with those who trained without SA.

In addition to investigating the effects of long-term balance training with SA, we sought to study the effects of vibrotactile display design on people's reaction times to vibrational cues. Among the various factors tested, the vibration frequency and tactor type had relatively small effects on reaction times, while stimulus location and secondary cognitive task had relatively large effects. Factors affected young and older adults' reaction times in a similar manner, but with different magnitudes.

Lastly, we explored the potential for ML to inform balance exercise progression for future applications of unsupervised balance training. We mapped body motion data measured by wearable inertial measurement units to balance assessment ratings provided by physical therapists. By training a multi-class classifier using the leave-one-participant-out cross-validation method, we found approximately 82% agreement among trained classifier and physical therapist assessments.

The findings of this dissertation suggest that vibrotactile SA can be used as a rehabilitation tool to further improve a subset of clinical outcomes resulting from supervised balance rehabilitation training. Specifically, individuals who train with a SA device may have additional confidence in performing balance activities and greater postural stability, which could decrease their fear of falling and fall risk, and subsequently increase their quality of life. This

research provides preliminary support for the hypothesized mechanism that SA promotes the central nervous system to reweight sensory inputs. The preliminary outcomes of this research also provide novel insights for unsupervised balance training that leverage wearable technology and ML techniques. By providing both SA and ML-based balance assessment ratings, the smart wearable device has the potential to improve individuals' compliance and motivation for in-home balance training.

Chapter 1 Introduction and Background

1.1. Human Balance

Balance describes the dynamics of body posture to prevent falls, standing balance describes the dynamics to maintain the body's center of mass (COM) over its base of support and to prevent falls, and gait balance describes the dynamics to move outside the base of support without falling during locomotor activities [1], [2]. Sensory inputs from the three major sensory systems, including somatosensory, visual, and vestibular systems, are conveyed to the central nervous system (CNS) to control balance during standing and gait [1], [3], [4]. The somatosensory system uses receptors in the skin, muscles, and joints to collect information about the position and velocity of body segments and their contacts with surrounding objects [1]. The visual system uses receptors in the retina to collect information about the body's orientation relative to surrounding objects and to plan locomotion [1], [5]. The vestibular system uses two otolith organs and three semicircular canals in the inner ears to collect information about the head's translational and rotational movements [1], [6]. The two small, sack-shaped otolith organs known as the utricle and saccule sense the horizontal and vertical accelerations, respectively [6]. The three semicircular canals (two vertical, one horizontal), positioned approximately on three orthogonal planes, sense the angular velocities of the head along three axes [6]. These signals are conveyed to the CNS via the eighth cranial nerve.

In addition to some types of injuries and infections, vestibular, neurological, orthopedic, and vascular disorders and aging can also disrupt or permanently damage the body's ability to maintain balance. Falls caused by balance disorders lead to loss of mobility, high levels of anxiety and depression, reduced quality of life, and even death [7]–[15]. This dissertation focuses on two populations: people with unilateral vestibular disorders and community-dwelling healthy older adults.

1.1.1. Vestibular Disorders

Vestibular disorders can result from head trauma (vehicle accidents, tripping and falling on uneven ground), drug toxicity, illness (meningitis), and aging. Common causes of vestibular disorders include benign paroxysmal positional vertigo (BPPV), labyrinthitis or vestibular neuritis, Ménière's disease, and abnormalities of the eighth cranial nerve [16]. Based on the National Health and Nutrition Examination Survey for 2001 to 2004, around 35% of American adults aged 40 years or older experienced some type of vestibular dysfunctions, and approximately 16 million new vestibular patients were diagnosed annually [17]. Vestibular disorders also affect children and younger adults [18]. Approximately 50% of older adults experience dizziness due to BPPV [19]. People with vestibular disorders usually have trouble maintaining straight posture, walking straight, walking with head turns, and walking in the dark [16]. People with unilateral vestibular disorders tend to fall toward the affected side during balance and gait exercises [7]. While falling is the primary concern expressed by people with vestibular disorders, they also report disruption of daily activities, the need for sick leave or medical consultation, and greater than expected anxiety, discouragement, and depression [8], [20].

1.1.2. Aging

Aging reduces sensory acuity, causes abnormal sensory weighting, and alters an individual's ability to quickly modify relative sensory weightings [21]–[23]. Older adults can experience decreased balance performance due to the combination of deteriorated sensory functions and reduced muscle strength [24]–[26]. Balance disorders increase an individual's risk of falling [10]–[12], [27] and fear of falling [13], and decrease mobility and confidence in performing daily activities, which in turn affect quality of life [14], [15]. Balance disorders are a major cause of falls [28], [29] for adults aged 65 and older. More than 30% of people aged 65 and older fall, and more than 50% of people aged 80 years and older fall at least once per year [12]. Older adults who have suffered recent falls tend to “sway” more than those who have not fallen in the past year [30]. Each year, emergency departments treat 2.5 million older people for fall injuries. The US Centers for Disease Control and Prevention estimated that the direct medical costs related to falls totaled \$19 billion in 2000 [31], [32].

1.1.3. Balance Models

Researchers have built inverted pendulum models to analyze how the central nervous system (CNS) controls postural balance during standing [1]. The simplest model of quiet standing stance assumes the body is a single inverted pendulum pivoting at the ankles [1]. A more general method models the body as multiple segments, e.g., the model assumes that in the frontal plane, the body is a parallelogram pendulum pivoting about two ankle joints and the hip joint, and that in the sagittal plane, the body is a pendulum pivoting about two ankle joints and the hip joint [1]. This model has been used to study ankle and hip strategies employed during perturbed standing [33]. Researchers have also incorporated sensory inputs and muscle

mechanics into the multiple-segmented models to study the relationship between the use of sensory inputs and change of stance width [34]. The studies of gait dynamics are more complicated. In general, for the upper body, researchers use the inverted pendulum model to study trunk motion, and for the lower body, they analyze the ground reaction force, moments of ankle, knee, hip, and joint angles within each gait/stride cycle.

1.1.4. Balance Performance Evaluation

To evaluate balance performance, researchers have developed the following common set of clinical outcome tests. For some clinical tests, researchers have developed minimal detectable change (MDC) and minimally clinically important difference (MCID) metrics to measure the effectiveness of balance training. The MDC is a statistical estimate of the smallest amount of change that is due to the rehabilitation training instead of a measurement error [35]. The MCID is the smallest amount of change considered important by the patient or the clinician [35]. The most common clinical tests used are listed below.

Activity-specific Balance Confidence (ABC) Scale: Identifies an individual's subjective measure of confidence when performing various ambulatory activities [36]. It contains 16 items, and the total score is 100 points. A higher score indicates stronger confidence, and scores less than 67 points indicate risk of falling [37].

Dizziness Handicap Inventory (DHI): Evaluates the self-perceived handicapping effects imposed by dizziness. A 25-item self-assessment inventory covers functional, emotional and physical domains (36, 36, 28 points, respectively). The total score is 100 points, and lower scores indicate less dizziness. For people with vestibular disorders, scores higher than 60 points indicate severe dizziness, and scores between 30 points and 61 points indicate moderate dizziness [38].

Computerized Dynamic Posturography (CDP): Quantitatively assesses the sensory, motor, and central adaptive impairments to standing balance control based on measured body sway responses to visual and mechanical (moving platform) perturbations. Within CDP, the *Sensory Organization Test (SOT)* protocol assesses the individual's ability to use somatosensory, visual, and vestibular systems to maintain postural stability during static standing as measured by SOT composite scores [39]. The six conditions for SOT are: 1) eyes open while standing on firm surface, 2) eyes closed while standing on firm surface, 3) eyes open with sway referenced visual surround, 4) eyes open while standing on sway referenced support surface, 5) eyes closed while standing on sway referenced support surface, and 6) eyes open while standing on sway referenced support surface and with sway referenced visual surround. For healthy young adults, the MDC of SOT is 8 points [40]. Somatosensory, visual, and vestibular reliance are calculated based on the SOT conditions shown in Equations 1-1, 1-2, and 1-3.

$$\text{Somatosensory Reliance} = \frac{\text{SOT condition 2}}{\text{SOT condition 1}} \quad \text{Equation 1-1}$$

$$\text{Visual Reliance} = \frac{\text{SOT condition 4}}{\text{SOT condition 1}} \quad \text{Equation 1-2}$$

$$\text{Vestibular Reliance} = \frac{\text{SOT condition 5}}{\text{SOT condition 1}} \quad \text{Equation 1-3}$$

Mini Balance Evaluation Systems Test (Mini-BESTest): Captures anticipatory postural adjustments, reactive postural control, sensory orientation, and dynamic gait performance based on 14 items [41]. There are two scoring systems: a total score of 28 points uses the lower score of the left and right sides of the body, and a total score of 32 points uses the total score of both sides. For people with balance disorders, the MDC is 3.5 points, and the MCID is 4 points [42].

Five Times Sit to Stand Test (5xSST): Evaluates the dynamic activity of functional lower limb muscle strength and change of transitional movements by measuring the time needed by the individual to complete the sit-to-stand task five times [43]. If the 5xSST duration is equal to or

longer than 12 seconds for older adults [44], further assessment for fall risk is needed. For people with vestibular disorders, the MCID is 2.3 seconds.

Four Square Step Test (FSST): Measures the individual's ability to step over objects forward, sideways, and backwards in seconds [45]. If the FSST duration is longer than 15 seconds for older adults, or 12 seconds for people with vestibular disorders, they are at risk for falls [45], [46].

Functional Reach Test (FRT): Assesses the individual's stability while standing in a fixed position and reaching forward. The maximum distance that an individual can reach forward is recorded in centimeters as the measure outcome [47]. An FRT less than 18 cm indicates limited mobility skills for older adults [48].

Ten-meter Walk Test: Measures gait speed in meters per second. For older adults, the MCID is 0.13 m/s [49].

Timed Up and Go (TUG): Measures the time (in seconds) needed by the individual to stand up from a chair and walk three meters quickly and safely [50]. The TUG with dual-task (cognitive task or motor task) adds a secondary task to the regular TUG to identify the individuals most at risk for falls. For older adults, a TUG-DT score above 15 seconds indicates a risk of fall [51].

Dynamic Gait Index (DGI): Assesses the individual's ability to modify balance while walking without external demands [2]. There are six walking tasks: steady state walking, walking with changing speeds, walking with head turns both horizontally and vertically, walking while stepping over and around obstacles, pivoting while walking, and stair climbing. The total score is 24 points (four points per task). For people with vestibular disorders, the MDC is 3.2 points [52]. Individuals with scores lower than 19 seconds are 2.6 times more likely to have

reported a fall than individuals with scores above 19 seconds [53]. For older adults, the MDC is 2.9 points [54] and the MCID is 1.9 points [55].

Functional Gait Assessment (FGA): Assesses the individual's stability of postural balance during walking [56]. FGA is a modification of the DGI to improve reliability. There are ten items measured using an ordinal scale from zero to three, and the total score is 30. For people with vestibular disorders, the MCID is 8 points.

Besides clinical tests, body sway while performing standing balance tasks is also used to assess balance performance. COP measured by a force plate, or COM measured by inertial measurement units (IMU) can be used to capture body sway. Force plates directly measure COP in the forward/backward and left/right directions, and IMUs measure body or body segment accelerations in the anterior/posterior (A/P) and medio-lateral (M/L) directions. Some common metrics used to quantify COP and IMU based balance performance include the Root-Mean-Square (RMS) of body sway, Elliptical fit Area (EA) of the body sway trajectory, and Percentage time in a one-degree Zone (PZ) [57].

1.2. Balance Rehabilitation

To retrain the individual's postural balance, balance rehabilitation is commonly prescribed or recommended. The hypothesized mechanism underlying balance rehabilitation is to leverage the ability of CNS to reweight the sensory inputs from intact sensory functions (i.e., somatosensory, visual, and vestibular sensory inputs) [58]. Generally, balance rehabilitation consists of various exercises designed to recover, retrain, or develop new sensorimotor strategies and facilitate the individual's functional mobility, decrease dizziness, and re-establish effective coordination [59]–[61]. Physical therapists can customize or adapt the exercises to the individual

(e.g., individuals with vestibular disorders, older adults) to optimize the efficacy [62]. By incorporating motor and sensory systems as well as cognitive and psychological processes, balance rehabilitation programs have proven more effective for reducing balance impairment and improving coordination than strength training alone [27], [63].

1.2.1. Vestibular Rehabilitation Therapy for People with Vestibular Disorders

Vestibular rehabilitation therapy (VRT) is commonly prescribed or recommended for individuals whose impairments are caused by vestibular disorders [64]–[66]. The therapy is designed to reduce the individual’s degree of handicap and improve the individual’s ability to perform daily activities [67]. VRT is most effective when vestibular symptoms first manifest themselves [68]. Researchers have assigned the exercises in VRT to various categories [69], e.g., Alsalaheen et al. use the categories of eye-head coordination, sitting balance, standing static balance, standing dynamic balance, and ambulation [70], and Hall et al. use the categories of gaze stability exercises, habituation exercises, standing and gait balance training, and walking exercises for endurance [71]. Klatt et al. categorized exercises into static standing, compliant surface standing, weight shifting, modified center of gravity, gait, and vestibulo-ocular reflex exercises [72].

Evidence of VRT’s effectiveness has increased in the past two decades [66], [73]. Vestibular rehabilitation improves both subjective measures (e.g., DHI, ABC) and objective measures (e.g., SOT, DGI) [66], [67], [74], [75]. Symptoms that have been reduced include dysregulated gait, fear of falling, falls, body sway in standing, blurred vision, and muscle weakness [73].

1.2.2. Exercise-based Rehabilitation for Older Adults

For community-dwelling older adults, typical interventions for improving balance performance include mobility aids, environmental modifications, medication modifications, and exercise programs [27]. Exercise-based rehabilitation programs, which have proven effective for improving balance performance [76], typically include strength and stamina training to counteract musculoskeletal degeneration, and balance training to isolate and challenge the individual's somatosensory, visual, and vestibular systems to improve overall balance [76], [77]. Clinicians or physical therapists demonstrate the exercise, give verbal feedback to the individual, and assess the short-term and long-term changes in performance throughout rehabilitation. Balance training lasting four to twelve weeks can yield significant improvements in clinical or functional outcome measures (e.g., SOT, TUG) corresponding to the reductions in fall rate, occurrence, and risk, and the prevention of declines in social and physical activities [27], [78]–[80]. A comparison of Tai Chi and balance training as a means to improve postural stability in older adults found that while Tai Chi can delay the onset of falling, it does not reduce body sway [81].

The exercises prescribed in clinical treatment can be performed at home. Campbell et al. reported that a six-month physiotherapist-recommended, customized strength and balance program reduced falls and improved functional performance in a group of women 80 years of age and older [82]. Miller et al., who conducted a pilot study with 14 older adults, found that debilitated, ambulatory, community-dwelling older adults significantly improved balance confidence, balance performance, and gait following participation in an in-home balance training program supervised by a therapist and family members [83].

1.2.3. Balance Rehabilitation Research Gaps

Although balance rehabilitation can improve balance performance, VRT does not necessarily result in the individual's full recovery of function or resolution of symptoms. In fact, many individuals reach a functional plateau or fail to achieve complete compensation [84]–[89]. Researchers have also noted that in-home balance training without direct clinical guidance is not as effective as in-clinic balance training due to the lack of supervision and consequent loss of motivation [63], [90]. For instance, Kao et al. reported that more participants in a supervised exercise program showed significantly improved clinical outcomes compared to participants who performed the same program at home without supervision [90].

1.3. Sensory Augmentation in Balance Training

Sensory augmentation (SA) provides additional information to complement and/or replace native sensory information from somatosensory, visual, and/or vestibular systems [91]. A typical SA system comprises a sensor or network of sensors (motion capture, force plate, inertial measurement unit (IMU), goniometer, etc.) to measure body motion, and a display to show body motions and/or provide instructional cues. Typical passive motion capture systems (e.g., Vicon [92]) comprise a set of cameras that measure the relative locations of a set of reflective markers in three-dimensional space. The measures are accurate, and the errors are within millimeters. After the markers are attached to the body, the motion capture system measures the biomechanics of whole body motion such as joint angles and walking speed using a model (e.g., Plug-in Gait model). However, the system only works in a space covered by the cameras and in lighting conditions that allow the cameras to capture the reflective markers. Force plates (e.g., AMTI [93]) measure the three-axis forces applied to them. In balance-related applications, force

plates measure the COP of body movements. When mounted on the ground, they measure standing balance, and when mounted on treadmills, they measure gait balance. Compared to the Vicon system, AMTI has a more flexible space requirement, but it only measures the pressures projected on the ground in two-dimensional space. Force plates are difficult to use during over-ground walking, and their weight and installation requirements impede their use outside the laboratory. IMUs (e.g., Xsens [94], APDM [95]) use accelerometers to measure three-axis acceleration, gyroscopes to measure three-axis angular velocity, and magnetometers to measure three-axis magnetism. In balance-related applications, an IMU can measure the angular movements of one body segment and frequently employ an extended Kalman Filter [57]; multiple IMUs are used to measure multiple body segments. IMUs are lightweight, easily transported, and have fewer space requirements than motion capture systems and force plates. However, the sample frequencies of IMUs are usually less than 150 Hz, whereas both motion capture systems and force plates can collect data at rates exceeding 500 Hz. The wires used to connect computers and IMUs limit applications to studies involving outdoor measures or multiple IMUs. Wireless IMUs have been developed, but the sampling frequency declines when multiple wireless IMUs are connected. This dissertation used a wired IMU (Xsens [94]) and an IMU embedded in the iPod Touch (Apple) to measure body sway.

Different SA modalities provide auditory, visual, electrotactile, vibrotactile, and multi-modal cues about the movements measured by the sensor(s). The correspondences between the cues and movements are predefined, and easily understood by the exerciser following a short training period for standing balance-related applications [96], [97]. Auditory SA provides cues by varying the volumes and the frequencies of the tones delivered to each ear [98], [99]. Visual SA provides cues by showing a virtual object on a monitor, a projection canvas, or a head-

mounted monitor in two-dimensional space [100]. Electrotactile SA delivers tactile sensations to the tongue by passing a local electric current through it (e.g., BrainPort®) [101]–[104].

Vibrotactile SA delivers tactile sensations to the skin by using vibrations generated by small motors (tactors) typically attached to the head, hand, torso, leg, or foot [105]. Multi-modal SA combines two or more modalities to further enhance SA [106]. This dissertation used vibrotactile SA to provide cues about body position and/or sway.

1.3.1. Vibrotactile Sensory Augmentation

Vibrotactile SA is widely used to deliver spatial and temporal information [107], [108]. In driving, vibrotactile displays provide drivers with navigation information [109] and serve as a warning to help prevent collisions [110], [111]. In flying, they provide pilots with altitude information and warning signals, and replace or reinforce visual and auditory cues [112], [113]. Vibrotactile displays have also been used to enhance physical sensations in the virtual environment [114], [115]. Vibrotactile displays have recently been used to provide real-time cues for specific body movements to individuals performing arm motion, balance, and gait rehabilitation training [96], [105], [116]–[122].

Vibrotactile stimuli are sensed by mechanoreceptors in the skin. One established theory is the four-channel theory: glabrous skin has four main types of mechanoreceptors (rapidly adapting (RA), Pacinian corpuscle (PC), slowly adapting I and slowly adapting II) [123]. RA and PC channels govern the vibrotactile sensations. Skin sensitivity changes based on vibration frequency, contact area, temperature, stimulus location, and age. Mechanoreceptors in the skin are frequency dependent. The PC, which is the dominant mechanoreceptor for sensing vibration, is most sensitive to stimuli around 250 Hz [124]–[126]. Skin sensitivity linearly increases when

the contact area is under 5 cm² [125]. Skin is most sensitive to temperatures around 35 °C at the vibration frequency of 250 Hz [127]. The anatomical locations of vibrotactile stimuli are also important (e.g., hands, fingers, and feet are more sensitive to vibration than the forehead and torso) [128]. Skin sensitivity decreases with aging at the vibration frequency of 250 Hz [129], [130].

The different vibrotactile actuators that generate vibrotactile stimuli have been categorized as [107]: linear electromagnetic actuators (e.g., C-2 tactor by EAI [131]), rotary electromagnetic actuators (e.g., coin-style motor by Precision MicrodrivesTM [132]), and non-electromagnetic actuators. In the first two categories, an electromagnet creates the vibration. The linear actuators contain a voice coil which is a moveable permanent magnet enclosed by an electrically conductive wire. The electrical signal applied to the voice coil varies over time to create vibrations. The movement of the voice coil itself is linear. The rotary actuators contain an off-center mass affixed to the output shaft. This mass continuously rotates with a constant voltage or current. In the third category, nonelectromagnetic actuators exploit the piezoelectric effect, where particular materials change their shapes when electrical voltage is provided.

Humans have a limited capacity to process information via each sensory system. The use of SA to convey additional information can interfere with or become difficult to discern if a given sensory system is “overloaded” [108]. When an overload occurs, the incidence of errors increases [108], [133]. Since humans depend heavily on auditory and visual modalities to perform daily activities, the use of vibrotactile SA to convey additional information is an appealing modality. In driving, navigational instructions conveyed via tactile information are effective because drivers depend heavily on their auditory and visual modalities to understand the surroundings [109]. Vibrotactile displays are particularly well suited for balance and gait

rehabilitation applications because some balance exercises require closed eyes or head movements. For these balance exercises, the use of the visual or auditory SA may not be applicable.

The major disadvantage of vibrotactile SA, however, is the amount of information it provides, which is usually discrete and only triggered if certain conditions are met. Moreover, vibrotactile SA can also be less intuitive and require more effort by the individual to learn and interpret. For example, researchers have explored attractive and repulsive directional cuing strategies for standing balance applications [134]. For attractive cues, individuals are instructed to move in the direction of the vibration, and for repulsive cues, they are instructed to move in the direction opposite from the vibration. Studies have shown that in the absence of instruction, vibrotactile stimulation induces small ($\sim 1^\circ$) non-volitional responses in the direction of its application location (i.e., compatible with an attractive cuing strategy) [135]–[137]. However, another study has shown that repulsive cues yield better real-time improvements in balance performance compared to attractive cues during their initial use; however, with additional training, the rate of balance improvement when using attractive cues may be better than repulsive cues [138].

1.3.2. Real-time Aid

Various SA modalities (e.g., vibrotactile [96], [119], [139]–[143], visual [100], auditory [98], [99], electrotactile [91], [144], and multi-modal [106], [145], [146]) have been shown to improve real-time balance performance by reducing body sway during static, dynamic, and perturbed standing tasks for healthy young adults, healthy older adults, and in individuals with balance disorders. Typical standing tasks include varying eye status (open/close), stance (feet

apart/Romberg/Semi-Romberg/Tandem/Single leg), head movement (none/pitch/yaw) and supporting surface (firm/foam/perturbed). While use of SA during gait tasks has been shown to reduce trunk sway, it is less successful compared with the use of SA during standing balance tasks [99], [146]–[148]. Additionally, stiffening in the coronal plane (i.e., rigid and awkward gait) has been observed [147].

1.3.3. Rehabilitation Tool

Balance-related SA is considered an effective rehabilitation tool if reduced body sway resulting from balance training with SA can be carried over to improved clinical outcome measures (carry-over effects) and the balance improvements (i.e., reduced postural sway and potential improved clinical outcome measures) can be retained when SA is no longer provided (retention effects). Studies of individuals with balance disorders have shown that retention effects lasted hours to days following short-term (i.e., less than 1 week) training with SA [96], [106], [149]. Retention effects lasting weeks to months have also been demonstrated following multi-session (i.e., more than 1 week) training with SA [85], [150], [151]. Rossi-Izquierdo et al. found that after training with vibrotactile SA for two weeks, participants with Parkinson's disease had improved clinical outcomes measures (e.g., DHI, SOT) and reduced their number of fall occurrences as carry-over effects [150]. Basta et al. found that reduced body sway was retained in a group of people with various balance disorders who trained with vibrotactile SA, but no such effect was observed in a group trained with erroneous SA signals [85]. Brugnera et al. noted that after a two-week balance training program, people with vestibular disorders trained with vibrotactile SA showed improved clinical measures (e.g., SOT, ABC) as carry-over effects, but the participants trained with standard rehabilitation practices showed no improvement [151].

However, most of these studies were conducted without a control group or lacked follow-up assessments to ascertain long-term carry-over or retention effects. In Chapter 2 of this dissertation, we studied the long-term training and retention effects for people with unilateral vestibular disorders who trained with vibrotactile SA.

A few studies of healthy older adults have examined the retention effects of longer-term training with SA. Video game-based in-home balance training has been shown to improve clinical measures (e.g., maximal muscle strength, Activity-specific Balance Confidence, risk of falling) after a minimum of five weeks of training [152]–[155]. Video game-based balance training uses balance platforms (e.g., Wii Fit balance board) and a display screen to provide visual cues of balance. However, the utility can be limited during balance exercises that require closed eyes, head movements, and altered stances. Lim et al. used multi-modal SA to investigate the effects of a two-week balance training program on 36 healthy older adults [156]. Both the augmented group and the control group wore an SA device (SwayStar™), but only the augmented group received SA. Participants trained on the same seven standing and gait tasks for two consecutive weeks (3 sessions per week) and trunk sway was monitored. Both groups showed reduced body sway during the final training session, but training with SA provided little benefit compared to training without SA. Moreover, neither group retained sway reduction in the post-training assessments for most tasks. However, in Lim’s study, all participants performed a fixed set of exercises and received SA through the training, which may have limited effectiveness. In Chapter 3 of this dissertation, we investigated the effects of long-term training with vibrotactile SA among community-dwelling healthy older adults.

1.3.4. Mechanism

Different hypotheses concerning the potential mechanisms of SA for balance training applications have been proposed [97]. The dominant hypothesis holds that SA assists the CNS to reweight the remaining, intact somatosensory, visual, and vestibular system inputs [157]. This sensory-reweighting hypothesis is in line with the concept underlying traditional balance rehabilitation [62], [71], [74]. The “sixth sense” hypothesis holds that the CNS incorporates SA as a new additive input to supplement other sensory information [97]. With continued training, the CNS learns to use the information provided by SA to control body posture. The “cognition” hypothesis holds that the CNS processes SA as a cognitive task, which makes the individual concentrate more on balance, but that the CNS does not adjust the sensory weights from the intact sensory inputs [97], [142]. The “context-specific adaptation” hypothesis holds that a new sensorimotor function develops only when SA is presented [97], [158], [159]; the body sway reduction is a combination of volitional and non-volitional response [160], [161]. In Chapters 2 and 3 of this dissertation, we measured vestibular reliance to preliminarily investigate the sensory reweighting mechanism.

1.4. Machine Learning

Prior work has shown that the amount of supervision can affect the efficacy of in-clinic balance rehabilitation and in-home balance rehabilitation [63], [90]. During supervised in-clinic balance rehabilitation sessions, exercise performance is typically assessed based on the experience of the physical therapist [62] and/or the clinical outcome tests described in Section 1.1.4. Rating scales such as the ones developed by Keith et al. and Espy et al. can also be used by physical therapists and patients, respectively, to assess balance performance [162], [163]. For

example, the Espy et al.'s visual analog scale defines performance characteristics using five levels and patients self-rate their own balance [163]. In research studies (and sometimes in clinical treatment settings), balance exercise performance is measured by force plates/pressure sensitive mats or wearable IMUs [57], [97]. Commonly used kinetic and kinematic metrics include COP, RMS of body sway, EA of body sway trajectory, and PZ as mentioned in Section 1.1.4 [57], but these metrics can be difficult to interpret with respect to clinical outcome tests and physical therapist's subjective ratings. Machine learning (ML) techniques offer the possibility of mapping quantitative sway parameters to the subjective ratings provided by physical therapists for the purpose of estimating and providing expert-like assessments of patients' performance in the absence of a physical therapist.

ML, an evolving and promising multi-disciplinary field, develops algorithms that can learn from and make predictions on data [164]. ML theory and methodology arose from the fields of artificial intelligence, signal processing, and statistics. ML is widely used in computer vision, natural language processing, economics, investing, marketing, search engines, fraud detection, and bioinformatics [165], [166]. The techniques can be categorized as supervised learning, unsupervised learning, and reinforcement learning [167], [168].

The recent growth in the availability of clinically relevant datasets has prompted the use of ML techniques for clinical tasks [169]–[178]. The idea is to capture the common patterns (i.e., trends) in the data that correlate with the clinically relevant outcomes or disease categories. Compared to traditional low-dimensional statistical approaches, ML uses optimization methods to understand the complex relationships in high-dimensional settings. For example, physiological waveform data (e.g., ECG) contain information that potentially can accurately identify an individual's current and future pathological states [169]. Beyond physiological signals,

biomechanical signals captured from wearable sensors (e.g., FitBit) have used ML techniques for applications ranging from predicting the patient's mood to determining the types of exercise performed [179].

1.4.1. Supervised Learning

Supervised learning, one of the most frequently studied ML frameworks, develops functions/algorithms based on a labeled training dataset to correctly predict or classify the unseen instances [167]. The training dataset is a set of training pairs. Each pair consists of an input vector, an object, or a 3D volume to describe the object and assign it a unique label. Generally, supervised learning includes classification and regression. In classification, labels are categorical values (e.g., species of flower, classes of animal, and whether an email is a spam). Based on the number of unique labels, the classification problems are divided into binary or multi-class classification problems. Multi-class classification is a generalization of a binary classification setting and as such, multi-class solutions usually use a combination of multiple binary classifiers. Examples of classifiers include logistic regression, Naïve Bayes, Fisher's linear discriminant, support vector machines, k-nearest neighbor, decision trees, random forests, and neural networks. If the categorical value is derived from levels (e.g., ratings of movies on a scale of 1-5), the labels themselves are sometimes treated as ordinal numbers and an ordinal transformation is used. When the labels are real numbers, the supervised learning is considered as a regression problem. The real numbers can be the size of an object, the coordination of an object, or the miles per gallon of a vehicle. Some categorical values, like ratings, can also be considered as real values and if so, specific rules apply to the output.

1.4.2. Applications of Machine Learning in Balance and Gait

The most common applications of ML techniques in the posture and gait fields include activity identification, gait pattern mining, and gait disorder detection. Given the increasing availability and low cost of wearable sensors, activity classification (e.g., walking, running, sitting, and lying down) has become a popular research area. Studies have achieved high accuracy (>80%) with various supervised algorithms (e.g., support vector machines, hidden Markov models) [179], [180]. Clustering approaches have been applied to group gait patterns into meaningful clinical categories. Toro et al. applied hierarchical clustering to kinematic gait data from children with cerebral palsy, and identified 13 different gait clusters organized into three gait types [181]. Subgroups have also been identified among healthy subjects. Phinyomark et al., who identified two distinct running gait patterns among healthy runners, suggested that care must be taken when investigating gait pathomechanics even in healthy individuals [182]. Using IMUs as inputs to the algorithms, researchers have successfully classified gait patterns for older adults versus people with stroke and Huntington's disease [183], and older adults versus people with Friedreich's ataxia [184], thus establishing a methodological foundation for mobile-based gait assessment tools and enhancing diagnostic acuity and the detection of gait alterations at earlier stages of the disease. Using kinetic and kinematic data from force plates and motion capture systems, respectively, researchers have successfully distinguished people with vestibular deficits from healthy adults during coordination and stretching movements [185], older adults from young adults during gait [186], and fall events ~350 ms in advance of the fall [187].

1.5. Dissertation Aims

In this dissertation, vibrotactile SA and ML were studied as approaches for further improving balance rehabilitation outcomes (Figure 1.1). Specifically, SA was studied to determine whether the addition of cues regarding body orientation with respect to the gravito-inertial vector during balance training could further enhance balance rehabilitation outcomes compared to balance training alone among people with unilateral vestibular disorders and community-dwelling healthy older adults. This approach was used to address the first major research gap of incomplete recovery of balance function following balance rehabilitation. ML was investigated as a means of providing supervision for potential in-home balance rehabilitation in the absence of physical therapists. This approach was used to address the second major research gap regarding the relative low efficacy of unsupervised in-home balance rehabilitation.

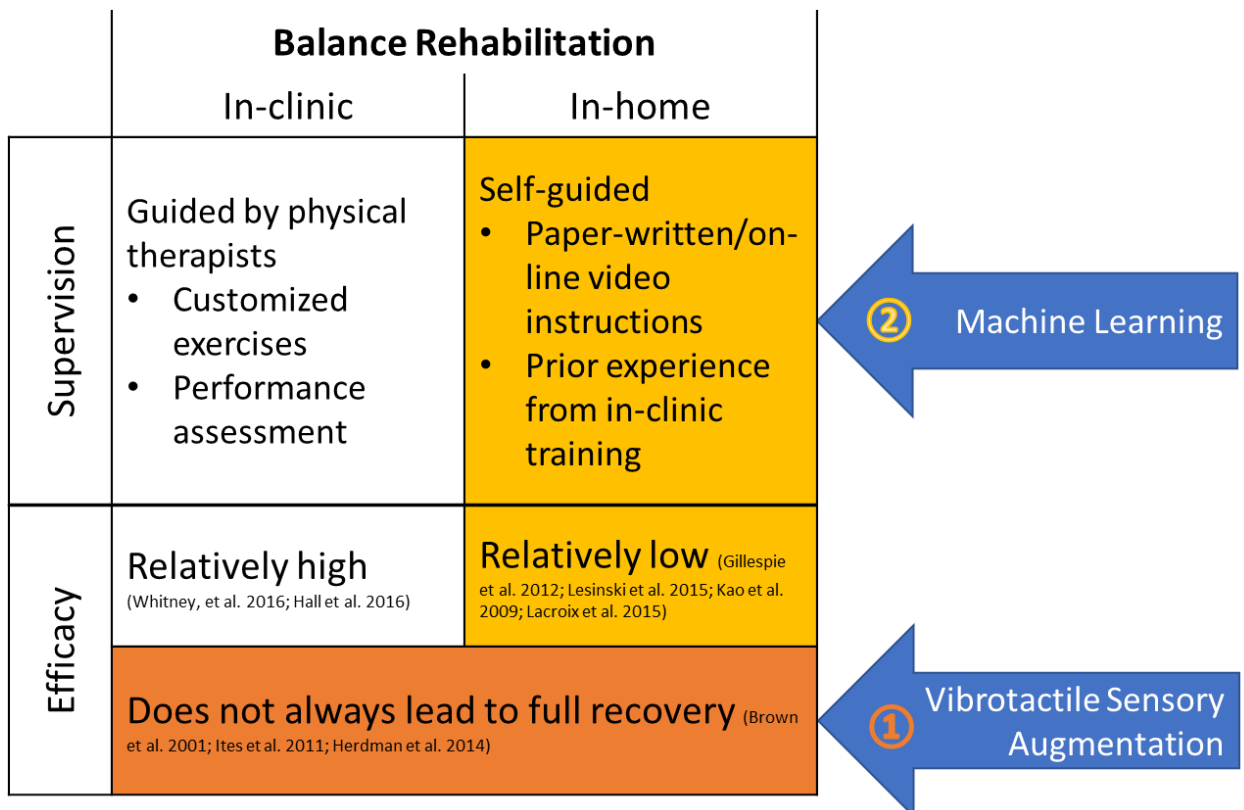


Figure 1.1: The two research gaps are shown in orange and khaki. The two proposed approaches are indicated with the blue arrows.

Based on the two proposed approaches, this dissertation developed four aims.

Aim 1: To study whether vestibular rehabilitation therapy (VRT) with vibrotactile SA produces added benefits for people with unilateral vestibular disorders compared with VRT alone.

Hypothesis 1.1: All participants, regardless of whether they receive vibrotactile SA or not, will show significant postural sway reductions in standing exercises and improvements in clinical outcome measures after completing balance training.

Hypothesis 1.2: Participants who train with vibrotactile SA will show significantly greater postural sway reduction in standing exercises compared to those who train without vibrotactile SA in the first week after completing training.

Hypothesis 1.3: Participants who train with vibrotactile SA will show significantly greater improvements in clinical outcome measures compared to those who train without vibrotactile SA in the first week after completing training.

Hypothesis 1.4: Participants who train with vibrotactile SA will continue to show significantly greater improvements in clinical outcome measures compared to those who train without vibrotactile SA for up to six months after completing training.

Aim 2: To study the effectiveness of long-term, home-based balance training with vibrotactile SA compared to training without vibrotactile SA among community-dwelling healthy older adults.

Hypothesis 2.1: Participants can use a smartphone balance trainer independently in their homes without pain, injury, or falls.

Hypothesis 2.2: Participants can reduce real-time trunk sway by using the vibrotactile SA provided by the smartphone balance trainer without supervision.

Hypothesis 2.3: All participants, regardless of whether they receive vibrotactile SA or not, will show significant improvements in clinical outcome measures after completing training.

Hypothesis 2.4: Participants who use the vibrotactile SA during home-based balance training will show significantly greater improvements in clinical outcome measures compared to those who train without vibrotactile SA.

Aim 3: To quantify how various factors affect participants' reaction times to vibrotactile cues.

Hypothesis 3.1: Young adults will react faster than older adults for the same condition (i.e., same experimental setup).

Hypothesis 3.2: Reaction times will be faster for vibrotactile stimuli at locations closest to the cortical area; for stimuli provided by linear actuators than coin-style motors; for stimuli with ACV generated by the vibration; for stimuli at 250 Hz than 200 Hz stimuli; for larger contact areas; and for single-task than dual-tasks.

Aim 4: To study whether physical therapists' assessments of balance performance can be automatically generated by ML algorithms based on trunk sway.

Hypothesis 4.1: The agreement between predictions by the ML algorithms and physical therapists' assessments will be significantly higher than the agreement between participants' self-assessments and physical therapists' assessments.

Hypothesis 4.2: A subset of features (e.g., RMS of body sway) extracted from the trunk sway will best predict the balance performance assessments by physical therapist.

Hypothesis 4.3: Predictions by the ML algorithms can provide accurate balance exercise assessments in the absence of physical therapists.

1.6. Chapter Overview

Chapter 2, *Vestibular rehabilitation therapy with vibrotactile sensory augmentation for people with unilateral vestibular disorders*, describes a randomized controlled experiment to understand whether the long-term training with vibrotactile SA produces additional benefits compared to training without vibrotactile SA. The experiment consists of a six-week training regimen (3 sessions per week), and five balance performance assessments (prior to the start of the training, midway through the training, and one week, one month, and six months after the end of the training). The results suggest that training with vibrotactile SA led to additional body sway reduction for balance exercises with head movements, and retention of the improvements for up to six months. Training with vibrotactile SA did not lead to significant additional improvements in most of the clinical outcomes except for the Activities-specific Balance Confidence scale.

Chapter 3, *Home-based balance training with vibrotactile sensory augmentation for community-dwelling healthy older adults*, describes a randomized controlled experiment to understand whether long-term, in-home training with vibrotactile SA produces additional benefits compared to training without vibrotactile SA. This chapter introduces the smartphone balance trainer developed to support in-home balance training with vibrotactile SA. The experiment consists of an eight-week training regimen (3 sessions per week) and three balance performance assessments (prior to the start of the training, midway through the training, and approximately one week following the end of the training). The results suggest that a subset of the clinical outcome tests including SOT and Mini-BESTest were further improved by the use of SA and indicated the feasibility of telerehabilitation therapy with a smartphone balance trainer.

Chapter 4, *Quantitative effects of various factors on reaction time to the vibrotactile stimuli*, describes a study focused on quantifying participants' reaction times to typical vibrotactile stimuli parameters and assessing the relative importance of the factors on reaction time. The results showed that auditory cues generated by the factors, vibration frequency, number of factors in the same location, and factor type had relatively small effects on reaction times (<50 ms), whereas stimulus location aging, and secondary cognitive task had relatively large effects. The results can be used to inform the design of vibrotactile displays.

Chapter 5, *Machine learning approach to automatically evaluate the performance of balance rehabilitation exercises*, describes a preliminary study to develop a machine learning algorithm-based system that maps body sway data to the balance assessment ratings provided by physical therapists during balance rehabilitation training. The results showed that agreement attained between the proposed support vector machine classifier and physical therapist ratings was superior to the agreement attained between the participants' and physical therapist' ratings. The results suggest that ML algorithms may be viable for supporting the exercise progression of in-home balance training in the absence of a physical therapist.

Chapter 6, *Discussion*, summarizes the findings, discusses the contributions, implications, limitations of this research, and recommends future work.

1.7. References

- [1] D. Winter, "Human balance and posture control during standing and walking," *Gait Posture*, vol. 3, no. 4, pp. 193–214, Dec. 1995.
- [2] A. Shumway-Cook and M. H. Woollacott, *Motor control: theory and practical applications*. Lippincott Williams & Wilkins, 1995.

- [3] A. Shumway-Cook and F. B. Horak, “Assessing the influence of sensory interaction on balance: suggestion from the field,” *Phys. Ther.*, vol. 66, no. 10, pp. 1548–1550, 1986.
- [4] R. J. Peterka, “Sensorimotor Integration in Human Postural Control,” *J. Neurophysiol.*, vol. 88, no. 3, pp. 1097–1118, 2002.
- [5] D. N. Lee and J. R. Lishman, “Visual proprioceptive control of stance,” *J. Hum. Mov. Stud.*, vol. 1, no. 2, pp. 87–95, 1975.
- [6] S. M. Highstein, R. R. Fay, and A. N. Popper, *The Vestibular System*, vol. 19. 2004.
- [7] J. H. J. Allum and A. L. Adkin, “Improvements in Trunk Sway Observed for Stance and Gait Tasks during Recovery from an Acute Unilateral Peripheral Vestibular Deficit,” *Audiol. Neuro-Otology*, vol. 8, no. 5, pp. 286–302, 2003.
- [8] H. K. Neuhauser, “Epidemiology of vertigo,” *Curr. Opin. Neurol.*, vol. 20, no. 1, pp. 40–46, 2007.
- [9] S. J. Herdman, P. Blatt, M. C. Schubert, and R. J. Tusa, “Falls in patients with vestibular deficits,” *Am. J. Otol.*, vol. 21, no. 6, pp. 847–851, 2000.
- [10] K. Delbaere, J. C. T. Close, J. Heim, P. S. Sachdev, H. Brodaty, M. J. Slavin, N. A. Kochan, and S. R. Lord, “A multifactorial approach to understanding fall risk in older people,” *J. Am. Geriatr. Soc.*, vol. 58, no. 9, pp. 1679–1685, 2010.
- [11] F. B. Horak, “Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls?,” *Age Ageing*, vol. 35, no. SUPPL.2, pp. 7–11, Sep. 2006.
- [12] M. E. Tinetti, “Preventing Falls in Elderly Persons,” *N. Engl. J. Med.*, vol. 348, pp. 42–49, 2003.
- [13] A. C. Scheffer, M. J. Schuurmans, N. Van dijk, T. Van der hooft, and S. E. De rooij, “Fear

- of falling: Measurement strategy, prevalence, risk factors and consequences among older persons,” *Age Ageing*, vol. 37, no. 1, pp. 19–24, 2008.
- [14] O. A. Donoghue, H. Cronin, G. M. Savva, C. O’Regan, and R. A. Kenny, “Effects of fear of falling and activity restriction on normal and dual task walking in community dwelling older adults,” *Gait Posture*, vol. 38, no. 1, pp. 120–124, 2013.
- [15] W. J. Rejeski and S. L. Mihalko, “Physical Activity and Quality of Life in Older Adults,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 56, no. Supplement 2, pp. 23–35, Oct. 2001.
- [16] M. C. Schubert and L. B. Minor, “Vestibulo-ocular Physiology Underlying Vestibular Hypofunction,” *PHYS THER. Phys. Ther.*, vol. 84, no. 4, pp. 373–385, 2004.
- [17] Y. Agrawal, J. P. Carey, C. C. Della Santina, M. C. Schubert, L. B. Minor, C. C. Della Santina, M. C. Schubert, and L. B. Minor, “Disorders of balance and vestibular function in US adults: data from the National Health and Nutrition Examination Survey, 2001-2004,” *Arch. Intern. Med.*, vol. 169, no. 10, pp. 938–944, May 2009.
- [18] R. M. Rine, “Growing evidence for balance and vestibular problems in children,” *Audiol. Med.*, vol. 7, no. 3, pp. 138–142, 2009.
- [19] T. D. Fife, D. J. Iverson, and T. Lempert, “vertigo (an evidence-based review) : Report of the Quality Standards Subcommittee of the American Academy of Neurology Practice Parameter : Therapies for benign paroxysmal positional vertigo (an evidence-based review),” *Neurology*, vol. 70, no. July 2007, pp. 2067–2074, 2008.
- [20] D. Monzani, L. Casolari, G. Guidetti, and M. Rigatelli, “Psychological distress and disability in patients with vertigo.,” *J. Psychosom. Res.*, vol. 50, no. 6, pp. 319–323, 2001.
- [21] M. H. Woollacott, A. Shumway-Cook, and L. M. Nashner, “Aging and posture control:

- changes in sensory organization and muscular coordination,” *Int. J. aging Hum. Dev.*, vol. 23, no. 2, pp. 97–114, 1986.
- [22] M. H. Woollacott and A. Shumway-Cook, “Changes in posture control across the life span--a systems approach.,” *Phys. Ther.*, vol. 70, no. 12, pp. 799–807, 1990.
- [23] C. Maurer, T. Mergner, and R. J. Peterka, “Multisensory control of human upright stance,” *Exp. Brain Res.*, vol. 171, no. 2, pp. 231–250, 2006.
- [24] B. E. Maki and W. E. McIlroy, “Postural control in the older adult.,” *Clin. Geriatr. Med.*, vol. 12, no. 4, pp. 635–58, Nov. 1996.
- [25] N. Benjuya, I. Melzer, and J. Kaplanski, “Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing,” *J. Gerontol. Med. Sci.*, vol. 59A, no. 2, pp. 166–171, 2004.
- [26] S. W. Shaffer and A. L. Harrison, “Aging of the somatosensory system: a translational perspective.,” *Phys. Ther.*, vol. 87, no. 2, pp. 193–207, 2007.
- [27] L. D. Gillespie, M. C. Robertson, W. J. Gillespie, C. Sherrington, S. Gates, L. M. Clemson, and S. E. Lamb, “Interventions for preventing falls in older people living in the community,” in *Cochrane Database of Systematic Reviews*, no. 9, L. D. Gillespie, Ed. Chichester, UK: John Wiley & Sons, Ltd, 2012.
- [28] A. J. Campbell, M. J. Borrie, G. F. Spears, S. L. Jackson, J. S. Brown, and J. L. Fitzgerald, “Circumstances and consequences of falls experienced by a community population 70 years and over during a prospective study,” *Age Ageing*, vol. 19, no. 2, pp. 136–141, 1990.
- [29] M. E. Tinetti, D. I. Baker, G. McAvay, E. B. Claus, P. Garrett, M. Gottschalk, M. L. Koch, K. Trainor, and R. I. Horwitz, “A multifactorial intervention to reduce the risk of falling among elderly people living in the community,” *N. Engl. J. Med.*, vol. 331, no. 13,

- pp. 821–827, 1994.
- [30] C. Ring, U. S. Nayak, and B. Isaacs, “Balance function in elderly people who have and who have not fallen.,” *Arch. Phys. Med. Rehabil.*, vol. 69, no. 4, pp. 261–264, 1988.
- [31] “Centers for Disease Control and Prevention, National Center for Injury Prevention and Control,” *Web-Based Injury Statistics Query and Reporting System (WISQARS)*, 2013. [Online]. Available: <http://www.cdc.gov/injury/wisqars/>.
- [32] J. A. Stevens, P. S. Corso, E. A. Finkelstein, and T. R. Miller, “The costs of fatal and non-fatal falls among older adults.,” *Inj. Prev.*, vol. 12, no. 5, pp. 290–5, 2006.
- [33] T. Ersal, J. L. McCrory, and K. H. Sienko, “Theoretical and experimental indicators of falls during pregnancy as assessed by postural perturbations.,” *Gait Posture*, vol. 39, no. 1, pp. 218–23, Jan. 2014.
- [34] A. D. Goodworth, P. Mellodge, and R. J. Peterka, “Stance width changes how sensory feedback is used for multisegmental balance control.,” *J. Neurophysiol.*, vol. 112, no. 3, pp. 525–42, Aug. 2014.
- [35] “Rehabilitation Measures Database.” [Online]. Available: <https://www.sralab.org/statistical-terms-use>. [Accessed: 09-Oct-2017].
- [36] L. E. Powell and A. M. Myers, “The activities-specific balance confidence (ABC) scale,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 50, no. 1, pp. M28--M34, 1995.
- [37] Y. Lajoie and S. P. Gallagher, “Predicting falls within the elderly community: Comparison of postural sway, reaction time, the Berg balance scale and the Activities-specific Balance Confidence (ABC) scale for comparing fallers and non-fallers,” *Arch. Gerontol. Geriatr.*, vol. 38, no. 1, pp. 11–26, 2004.
- [38] S. L. Whitney, D. M. Wrisley, K. E. Brown, and J. M. Furman, “Is perception of handicap

- related to functional performance in persons with vestibular dysfunction?,” *Otol. Neurotol.*, vol. 25, no. 2, pp. 139–143, 2004.
- [39] R. A. Clendaniel, “Outcome measures for assessment of treatment of the dizzy and balance disorder patient,” *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 519–533, 2000.
- [40] D. M. Wrisley, M. J. Stephens, S. Mosley, A. Wojnowski, J. Duffy, and R. Burkard, “Learning Effects of Repetitive Administrations of the Sensory Organization Test in Healthy Young Adults,” *Arch. Phys. Med. Rehabil.*, vol. 88, no. 8, pp. 1049–1054, 2007.
- [41] F. Franchignoni, F. Horak, M. Godi, A. Nardone, and A. Giordano, “Using psychometric techniques to improve the balance evaluation systems test: The mini-bestest,” *J. Rehabil. Med.*, vol. 42, no. 4, pp. 323–331, 2010.
- [42] M. Godi, F. Franchignoni, M. Caligari, A. Giordano, A. M. Turcato, and A. Nardone, “Comparison of Reliability, Validity, and Responsiveness of the Mini- BESTest and Berg Balance Scale in Patients With Balance Disorders,” *Phys. Ther.*, vol. 93, no. 2, pp. 158–167, 2013.
- [43] M. Csuka and D. J. McCarty, “Simple method for measurement of lower extremity muscle strength,” *Am. J. Med.*, vol. 78, no. 1, pp. 77–81, 1985.
- [44] A. Tiedemann, H. Shimada, C. Sherrington, S. Murray, and S. Lord, “The comparative ability of eight functional mobility tests for predicting falls in community-dwelling older people,” *Age Ageing*, vol. 37, no. 4, pp. 430–435, 2008.
- [45] W. Dite and V. A. Temple, “A clinical test of stepping and change of direction to identify multiple falling older adults,” *Arch. Phys. Med. Rehabil.*, vol. 83, no. 11, pp. 1566–1571, 2002.
- [46] S. L. Whitney, G. F. Marchetti, L. O. Morris, and P. J. Sparto, “The Reliability and

- Validity of the Four Square Step Test for People With Balance Deficits Secondary to a Vestibular Disorder,” *Arch. Phys. Med. Rehabil.*, vol. 88, no. 1, pp. 99–104, 2007.
- [47] P. W. Duncan, D. K. Weiner, J. Chandler, and S. Studenski, “Functional reach: a new clinical measure of balance,” *J. Gerontol.*, vol. 45, no. 6, pp. M192–M197, 1990.
- [48] D. K. Weiner, P. W. Duncan, J. Chandler, and S. a Studenski, “Functional reach: a marker of physical frailty,” *J. Am. Geriatr. Soc.*, vol. 40, no. 3, pp. 203–207, 1992.
- [49] S. Perera, S. H. Mody, R. C. Woodman, and S. A. Studenski, “Meaningful change and responsiveness in common physical performance measures in older adults,” *J. Am. Geriatr. Soc.*, vol. 54, no. 5, pp. 743–749, 2006.
- [50] D. Podsiadlo and S. Richardson, “The Timed ‘Up & Go’: A Test of Basic Functional Mobility for Frail Elderly Persons,” *J. Am. Geriatr. Soc.*, vol. 39, no. 2, pp. 142–148, 1991.
- [51] A. Shumey-Cook, S. Brauer, and M. Woollacott, “Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test,” *Phys. Ther.*, vol. 80, no. 9, pp. 896–903, 2000.
- [52] C. D. Hall and S. J. Herdman, “Reliability of Clinical Measures Used to with Peripheral Vestibular Disorders,” *J. Neurol. Phys. Ther.*, vol. 30, no. 2, pp. 74–81, 2006.
- [53] S. L. Whitney, M. T. Hudak, and G. F. Marchetti, “The dynamic gait index relates to self-reported fall history in individuals with vestibular dysfunction,” *J. Vestib. Res.*, vol. 10, no. 2, pp. 99–105, 2000.
- [54] S. Romero, M. D. Bishop, C. A. Velozo, and K. Light, “Minimum detectable change of the Berg Balance Scale and Dynamic Gait Index in older persons at risk for falling,” *J. Geriatr. Phys. Ther.*, vol. 34, no. 3, pp. 131–137, 2011.

- [55] P. K. Pardasaney, N. K. Latham, A. M. Jette, R. C. Wagenaar, P. Ni, M. D. Slavin, and J. F. Bean, "Sensitivity to change and responsiveness of four balance measures for community-dwelling older adults," *Phys. Ther.*, vol. 92, no. 3, pp. 388–397, 2012.
- [56] D. M. Wrisley, G. F. Marchetti, D. K. Kuharsky, and S. L. Whitney, "Reliability, internal consistency, and validity of data obtained with the functional gait assessment.," *Phys. Ther.*, vol. 84, no. 10, pp. 906–918, 2004.
- [57] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, "Cell phone based balance trainer," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, Jan. 2012.
- [58] B. I. Han, H. S. Song, and J. S. Kim, "Vestibular rehabilitation therapy: Review of indications, mechanisms, and key exercises," *J. Clin. Neurol.*, vol. 7, no. 4, pp. 184–196, 2011.
- [59] E. E. Hansson, N.-O. Månsson, K. A. Ringsberg, and A. Håkansson, "Falls among dizzy patients in primary healthcare: an intervention study with control group.," *Int. J. Rehabil. Res.*, vol. 31, no. 1, pp. 51–7, Mar. 2008.
- [60] J. Y. Jung, J.-S. Kim, P. S. Chung, S. H. Woo, and C. K. Rhee, "Effect of vestibular rehabilitation on dizziness in the elderly.," *Am. J. Otolaryngol.*, vol. 30, no. 5, pp. 295–9, Jan. 2009.
- [61] A. Shumway-Cook, F. B. Horak, L. Yardley, and A. M. Bronstein, "Rehabilitation of balance disorders in the patient with vestibular pathology," *Clin. Disord. Balanc. Posture, Gait. London, United Kingdom Arnold, Div Hodder Headl. PLC*, pp. 213–220, 1996.
- [62] B. N. Klatt, W. J. Carender, C. Lin, S. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, "A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular Disorders," *Phys. Med. Rehabil. - Int.*, vol. 2, no. 4, p. 1044,

- 2015.
- [63] A. Lacroix, R. W. Kressig, T. Muehlbauer, Y. J. Gschwind, B. Pfenninger, O. Bruegger, and U. Granacher, “Effects of a Supervised versus an Unsupervised Combined Balance and Strength Training Program on Balance and Muscle Power in Healthy Older Adults: A Randomized Controlled Trial,” *Gerontology*, vol. 62, no. 3, pp. 275–288, 2015.
- [64] S. L. Whitney and M. M. Rossi, “Efficacy of vestibular rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 659–672, Jun. 2000.
- [65] J. L. Cowand, D. M. Wrisley, M. Walker, B. Strasnick, and J. T. Jacobson, “Efficacy of Vestibular Rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 118, pp. 49–54, 1998.
- [66] S. L. Whitney, A. H. Alghadir, and S. Anwer, “Recent Evidence About the Effectiveness of Vestibular Rehabilitation,” *Curr. Treat. Options Neurol.*, vol. 18, no. 3, pp. 1–15, 2016.
- [67] A. Meli, G. Zimatore, C. Badaracco, E. De Angelis, and D. Tufarelli, “Vestibular rehabilitation and 6-month follow-up using objective and subjective measures.,” *Acta Otolaryngol.*, vol. 126, no. 3, pp. 259–66, 2006.
- [68] M. Lacour, “Restoration of vestibular function: basic aspects and practical advances for rehabilitation.,” *Curr. Med. Res. Opin.*, vol. 22, no. 9, pp. 1651–9, Sep. 2006.
- [69] S. J. Herdman, *Vestibular Rehabilitation*, Second Ed. Philadelphia: F.A. Davis Company, 2000.
- [70] B. A. Alsalaheen, S. L. Whitney, A. Mucha, L. O. Morris, J. M. Furman, and P. J. Sparto, “Exercise prescription patterns in patients treated with vestibular rehabilitation after concussion,” *Physiother. Res. Int.*, vol. 18, no. 2, pp. 100–108, 2013.
- [71] C. D. Hall, S. J. Herdman, S. L. Whitney, S. P. Cass, R. A. Clendaniel, T. D. Fife, J. M. Furman, T. S. D. Getchius, J. A. Goebel, N. T. Shepard, and S. N. Woodhouse, *Vestibular*

Rehabilitation for Peripheral Vestibular Hypofunction: An Evidence-Based Clinical Practice Guideline, vol. 40, no. April. 2016.

- [72] T. Bao, B. Klatt, W. Carender, C. Kinnaird, S. Whitney, and K. Sienko, “Evaluating a conceptual progression framework by grouping balance exercises based on difficulty,” in *International Society for Posture and Gait Research*, 2015.
- [73] S. L. Hillier and M. McDonnell, “Vestibular rehabilitation for unilateral peripheral vestibular dysfunction.,” *Clin. Otolaryngol.*, vol. 36, no. 3, pp. 248–249, 2011.
- [74] M. B. Badke, J. A. Miedaner, C. R. Grove, T. A. Shea, and G. M. Pyle, “Effects of vestibular and balance rehabilitation on sensory organization and dizziness handicap,” *Ann. Otol. Rhinol. Laryngol.*, vol. 114, no. 1 I, pp. 48–54, 2005.
- [75] M. Mruzek, K. Barin, D. S. Nichols, C. N. Burnett, and D. B. Welling, “Effects of vestibular rehabilitation and social reinforcement on recovery following ablative vestibular surgery.,” *Laryngoscope*, vol. 105, no. 7 Pt 1, pp. 686–92, Jul. 1995.
- [76] A. Shumway-Cook, W. Gruber, M. Baldwin, and S. Liao, “The effect of multidimensional exercises on balance, mobility, and fall risk in community-dwelling older adults.,” *Phys. Ther.*, vol. 77, no. 1, pp. 46–57, 1997.
- [77] M. H. Hu and M. H. Woollacott, “Multisensory Training of Standing Balance in Older Adults: I. Postural Stability and One-Leg Stance Balance,” *J. Gerontol.*, vol. 49, no. 2, pp. M52–M61, Mar. 1994.
- [78] M. E. Tinetti and C. S. Williams, “The Effect of Falls and Fall Injuries on Functioning in Community-Dwelling Older Persons,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 53A, no. 2, pp. M112–M119, Mar. 1998.
- [79] K. Hauer, B. Rost, K. Rüttschle, O. Hedda, N. Specht, P. Bärtsch, P. Oster, and G. Schlierf,

- “Exercise Training for Rehabilitation and Secondary Prevention of Falls in Geriatric Patients With A History of Injurious Falls,” *J. Am. Geriatr. Soc.*, vol. 49, no. 1, pp. 10–20, 2001.
- [80] M. Lesinski, T. Hortobágyi, T. Muehlbauer, A. Gollhofer, and U. Granacher, “Effects of Balance Training on Balance Performance in Healthy Older Adults : A Systematic Review and Meta-analysis,” *Sport. Med.*, vol. 45, no. 12, pp. 1721–38, Sep. 2015.
- [81] S. L. Wolf, H. X. Barnhart, G. L. Ellison, and C. E. Coogler, “The effect of Tai Chi Quan and computerized balance training on postural stability in older subjects. Atlanta FICSIT Group. Frailty and Injuries: Cooperative Studies on Intervention Techniques.,” *Phys. Ther.*, vol. 77, no. 4, pp. 371-381-384, 1997.
- [82] A. J. Campbell, M. C. Robertson, M. M. Gardner, R. N. Norton, M. W. Tilyard, and D. M. Buchner, “Randomised controlled trial of a general practice programme of home based exercise to prevent falls in elderly women,” *Br. Med. J.*, vol. 315, no. 7115, pp. 1065–1069, 1997.
- [83] K. L. Miller, J. R. Magel, and J. G. Hayes, “The effects of a home-based exercise program on balance confidence, balance performance, and gait in debilitated, ambulatory community-dwelling older adults: a pilot study.,” *J. Geriatr. Phys. Ther.*, vol. 33, no. 2, pp. 85–91, 2010.
- [84] S. J. Herdman and R. A. Clendaniel, *Vestibular Rehabilitation*, Fourth. 2014.
- [85] D. Basta, M. Rossi-Izquierdo, A. Soto-Varela, M. E. Greters, R. S. Bittar, E. Steinhagen-Thiessen, R. Eckardt, T. Harada, F. Goto, K. Ogawa, and A. Ernst, “Efficacy of a vibrotactile neurofeedback training in stance and gait conditions for the treatment of balance deficits: a double-blind, placebo-controlled multicenter study.,” *Otol. Neurotol.*,

- vol. 32, no. 9, pp. 1492–9, Dec. 2011.
- [86] B. S. Robinson, J. L. Cook, C. M. Richburg, and S. E. Price, “Use of an electrotactile vestibular substitution system to facilitate balance and gait of an individual with gentamicin-induced bilateral vestibular hypofunction and bilateral transtibial amputation.” *J. Neurol. Phys. Ther.*, vol. 33, no. 3, pp. 150–159, 2009.
- [87] K. E. Brown, S. L. Whitney, D. M. Wrisley, and J. M. Furman, “Physical therapy outcomes for persons with bilateral vestibular loss.” *Laryngoscope*, vol. 111, no. 10, pp. 1812–1817, 2001.
- [88] K. I. Ites, E. J. Anderson, M. L. Cahill, J. A. Kearney, E. C. Post, and L. S. Gilchrist, “Balance Interventions for Diabetic Peripheral Neuropathy: A Systematic Review,” *J. Geriatr. Phys. Ther.*, vol. 34, pp. 109–116, 2011.
- [89] C. G. C. Barros, R. S. M. Bittar, and Y. Danilov, “Effects of electrotactile vestibular substitution on rehabilitation of patients with bilateral vestibular loss.” *Neurosci. Lett.*, vol. 476, no. 3, pp. 123–6, Jun. 2010.
- [90] C. L. Kao, L. K. Chen, C. M. Chern, L. C. Hsu, C. C. Chen, and S. J. Hwang, “Rehabilitation outcome in home-based versus supervised exercise programs for chronically dizzy patients,” *Arch. Gerontol. Geriatr.*, vol. 51, no. 3, pp. 264–267, 2009.
- [91] P. Bach-y-Rita, “Tactile sensory substitution studies.” *Ann. N. Y. Acad. Sci.*, vol. 1013, pp. 83–91, May 2004.
- [92] “Motion Capture Systems | VICON.” [Online]. Available: <https://www.vicon.com/>. [Accessed: 13-Mar-2017].
- [93] “AMTI Home Page.” [Online]. Available: <http://www.amti.biz/index.aspx>. [Accessed: 13-Mar-2017].

- [94] “Home - Xsens 3D motion tracking.” [Online]. Available: <https://www.xsens.com/>.
[Accessed: 13-Mar-2017].
- [95] “APDM Wearable Technologies.” [Online]. Available: <https://www.apdm.com/>.
[Accessed: 13-Mar-2017].
- [96] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.
- [97] K. H. Sienko, S. L. Whitney, W. J. Carender, and C. Wall III, “The role of sensory augmentation for people with vestibular deficits: real-time balance aid and/or rehabilitation device?,” *J. Vestib. Res.*, vol. 27, no. 1, pp. 63–76, 2017.
- [98] M. Dozza, F. B. Horak, and L. Chiari, “Auditory biofeedback substitutes for loss of sensory information in maintaining stance.,” *Exp. brain Res.*, vol. 178, no. 1, pp. 37–48, Mar. 2007.
- [99] J. Hegeman, F. Honegger, M. Kupper, and J. H. J. Allum, “The balance control of bilateral peripheral vestibular loss subjects and its improvement with auditory prosthetic feedback.,” *J. Vestib. Res.*, vol. 15, no. 2, pp. 109–17, Jan. 2005.
- [100] Z. Halická, J. Lobotková, K. Bučková, and F. Hlavačka, “Effectiveness of different visual biofeedback signals for human balance improvement,” *Gait Posture*, vol. 39, no. 1, pp. 410–414, 2014.
- [101] K. A. Kaczmarek, J. G. Webster, P. Bach-y-rita, W. J. Tompkins, and S. Member, “Electrotactile and Vibrotactile Displays for Sensory Substitution Systems,” vol. 38, no. 1, 1991.

- [102] “BrainPort V100 Vision Aid.” [Online]. Available: <http://www.wicab.com/>. [Accessed: 13-Mar-2017].
- [103] P. Bach-y-Rita, K. A. Kaczmarek, M. E. Tyler, and J. Garcia-Lara, “Form perception with a 49-point electrotactile stimulus array on the tongue: a technical note.,” *J. Rehabil. Res. Dev.*, vol. 35, no. 4, pp. 427–430, 1998.
- [104] P. Bach-y-Rita, Y. Danilov, M. Tyler, and R. J. Grimm, “Late human brain plasticity: vestibular substitution with a tongue BrainPort human-machine interface,” *Intellectica*, vol. 1, no. 40, pp. 115–122, 2005.
- [105] P. B. Shull, W. Jirattigalachote, M. a Hunt, M. R. Cutkosky, and S. L. Delp, “Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention.,” *Gait Posture*, vol. 40, no. 1, pp. 11–19, May 2014.
- [106] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, “Trunk sway reductions in young and older adults using multi-modal biofeedback.,” *Gait Posture*, vol. 31, no. 4, pp. 465–72, Apr. 2010.
- [107] S. Choi and K. J. Kuchenbecker, “Vibrotactile display: Perception, technology, and applications,” *Proc. IEEE*, vol. 101, no. 9, pp. 2093–2104, 2013.
- [108] K. Myles and M. S. Binseel, “The Tactile Modality : A Review of Tactile Sensitivity and Human Tactile Interfaces,” *Army Res. Lab.*, no. May, pp. 1–27, 2007.
- [109] J. B. F. Van Erp and H. a H. C. Van Veen, “Vibrotactile in-vehicle navigation system,” *Transp. Res. Part F Traffic Psychol. Behav.*, vol. 7, pp. 247–256, 2004.
- [110] F. Meng, C. Ho, R. Gray, and C. Spence, “Dynamic vibrotactile warning signals for frontal collision avoidance: towards the torso versus towards the head,” *Ergonomics*, vol.

- 58, no. March 2015, pp. 411–425, 2014.
- [111] C. Ho, H. Z. Tan, and C. Spence, “Using spatial vibrotactile cues to direct visual attention in driving scenes,” *Transp. Res. Part F Traffic Psychol. Behav.*, vol. 8, pp. 397–412, 2005.
- [112] Y. Salzer, T. Oron-Gilad, A. Ronen, and Y. Parmet, “Vibrotactile ‘on-thigh’ alerting system in the cockpit,” *Hum. Factors*, vol. 53, no. 2, pp. 118–131, 2011.
- [113] H. a H. C. van Veen and J. van Erp, “Tactile information presentation in the cockpit,” *Haptic Human-Computer Interact.*, vol. 2058, pp. 174–181, 2001.
- [114] A. M. Okamura, M. R. Cutkosky, and J. T. Dennerlein, “Reality-based models for vibration feedback in virtual environments,” *IEEE/ASME Trans. Mechatronics*, vol. 6, no. 3, pp. 245–252, 2001.
- [115] A. Bloomfield and N. I. Badler, “Virtual Training via Vibrotactile Arrays,” *Presence Teleoperators Virtual Environ.*, vol. 17, no. 2, pp. 103–120, 2008.
- [116] K. Bark, P. Khanna, R. Irwin, P. Kapur, S. A. Jax, L. J. Buxbaum, and K. J. Kuchenbecker, “Lessons in using vibrotactile feedback to guide fast arm motions,” *2011 IEEE World Haptics Conf. WHC 2011*, pp. 355–360, 2011.
- [117] W. Nanhoe-Mahabier, J. H. Allum, E. P. Pasman, S. Overeem, and B. R. Bloem, “The effects of vibrotactile biofeedback training on trunk sway in Parkinson’s disease patients,” *Parkinsonism Relat. Disord.*, vol. 18, no. 9, pp. 1017–21, Nov. 2012.
- [118] P. B. Shull, K. L. Lurie, M. R. Cutkosky, and T. F. Besier, “Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback,” *J. Biomech.*, vol. 44, no. 8, pp. 1605–9, May 2011.
- [119] C. Wall III and E. Kentala, “Effect of displacement, velocity, and combined vibrotactile tilt feedback on postural control of vestibulopathic subjects,” *J. Vestib. Res.*, vol. 20, no.

- 1, pp. 61–9, Jan. 2010.
- [120] C.-C. Lin, S. L. Whitney, P. J. Loughlin, J. M. Furman, M. S. Redfern, K. H. Sienko, and P. J. Sparto, “The effect of age on postural and cognitive task performance while using vibrotactile feedback,” *J. Neurophysiol.*, vol. 113, no. 7, pp. 2127–2136, 2015.
- [121] J. A. Goebel, B. C. Sinks, B. E. Parker, N. T. Richardson, A. B. Olowin, and R. W. Cholewiak, “Effectiveness of head-mounted vibrotactile stimulation in subjects with bilateral vestibular loss: a phase 1 clinical trial,” *Otol. Neurotol.*, vol. 30, no. 2, pp. 210–216, 2009.
- [122] F. Asseman, A. M. Bronstein, and M. A. Gresty, “Using vibrotactile feedback of instability to trigger a forward compensatory stepping response,” *J. Neurol.*, vol. 254, no. 11, pp. 1555–1561, 2007.
- [123] R. S. Johansson and A. B. Vallbo, “Tactile sensory coding in the glabrous skin of the human hand,” *Trends Neurosci.*, vol. 6, no. 1, pp. 27–32, 1983.
- [124] S. J. Bolanowski, G. A. Gescheider, R. T. Verrillo, and C. M. Checkosky, “Four channels mediate the mechanical aspects of touch,” *J. Acoust. Soc. Am.*, vol. 84, no. 5, pp. 1680–1694, 1988.
- [125] R. T. Verrillo, “Effect of Contactor Area on the Vibrotactile Threshold,” *J. Acoust. Soc. Am.*, vol. 35, no. 12, p. 1962, Jul. 1963.
- [126] J. C. Craig and C. E. Sherrick, “The role of skin coupling in the determination of vibrotactile spatial summation,” *Percept. Psychophys.*, vol. 6, no. 2, pp. 97–101, 1969.
- [127] B. G. Green, “The effect of skin temperature on vibrotactile sensitivity,” *Percept. Psychophys.*, vol. 21, no. 3, pp. 243–248, May 1977.
- [128] A. Wilska, “On the vibrational sensitivity in different regions of the body surface,” *Acta*

- Physiol. Scand.*, vol. 31, no. 1952, pp. 284–289, 1954.
- [129] R. T. Verrillo, “Age related changes in the sensitivity to vibration.,” *J. Gerontol.*, vol. 35, no. 2, pp. 185–193, 1980.
- [130] G. A. Gescheider, S. J. Bolanowski, K. L. Hall, K. E. Hoffman, and R. T. Verrillo, “The Effects of Aging on Information-Processing Channels in the Sense of Touch: I. Absolute Sensitivity,” *Somatosens. Mot. Res.*, Jul. 2009.
- [131] “ATAC | Engineering Acoustics, Inc.” [Online]. Available: <https://www.eaiinfo.com/tactor-landing/>. [Accessed: 14-Mar-2017].
- [132] “Coin Vibration Motors | Precision Microdrives.” [Online]. Available: <https://www.precisionmicrodrives.com/vibration-motors/coin-vibration-motors>. [Accessed: 14-Mar-2017].
- [133] S. Oviatt, “Ten myths of multimodal interaction,” *Commun. ACM*, vol. 42, no. 11, pp. 74–81, 1999.
- [134] B. Lee and K. H. Sienko, “Effects of attractive versus repulsive vibrotactile instructional cues during motion replication tasks,” pp. 3533–3536, 2011.
- [135] B. C. Lee, B. J. Martin, and K. H. Sienko, “Directional postural responses induced by vibrotactile stimulations applied to the torso,” *Exp. Brain Res.*, vol. 222, no. 4, pp. 471–482, 2012.
- [136] B.-C. Lee, B. J. Martin, and K. H. Sienko, “The effects of actuator selection on non-volitional postural responses to torso-based vibrotactile stimulation.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 21, 2013.
- [137] B.-C. Lee, B. J. Martin, A. Ho, and K. H. Sienko, “Postural Reorganization Induced by Torso Cutaneous Covibration,” *J. Neurosci.*, vol. 33, no. 18, pp. 7870–7876, 2013.

- [138] C. Kinnaird, J. Lee, W. J. Carender, M. Kabeto, B. Martin, and K. H. Sienko, “The effects of attractive vs. repulsive instructional cuing on balance performance,” *J. Neuroeng. Rehabil.*, vol. 13, no. 1, p. 1, 2016.
- [139] K. H. Sienko, M. D. Balkwill, and C. Wall, “Biofeedback improves postural control recovery from multi-axis discrete perturbations.,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 53, 2012.
- [140] K. E. Bechly, W. J. Carender, J. D. Myles, and K. H. Sienko, “Determining the preferred modality for real-time biofeedback during balance training,” *Gait Posture*, vol. 37, no. 3, pp. 391–6, Mar. 2013.
- [141] C. Wall, D. M. Wrisley, and K. D. Statler, “Vibrotactile tilt feedback improves dynamic gait index: a fall risk indicator in older adults.,” *Gait Posture*, vol. 30, no. 1, pp. 16–21, Jul. 2009.
- [142] M. Janssen, R. Stokroos, J. Aarts, R. van Lummel, and H. Kingma, “Salient and placebo vibrotactile feedback are equally effective in reducing sway in bilateral vestibular loss patients.,” *Gait Posture*, vol. 31, no. 2, pp. 213–7, Feb. 2010.
- [143] J. H. J. Allum, M. G. Carpenter, B. C. Horslen, J. R. Davis, F. Honegger, K.-S. Tang, and P. Kessler, “Improving impaired balance function: real-time versus carry-over effects of prosthetic feedback.,” *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2011, pp. 1314–8, Jan. 2011.
- [144] P. Bach-y-Rita and S. W. Kercel, “Sensory substitution and the human–machine interface,” *Trends Cogn. Sci.*, vol. 7, no. 12, pp. 541–546, Dec. 2003.
- [145] J. L. Huffman, L. E. Norton, A. L. Adkin, and J. H. J. Allum, “Directional effects of biofeedback on trunk sway during stance tasks in healthy young adults,” *Gait Posture*,

- vol. 32, no. 1, pp. 62–66, 2010.
- [146] L. J. F. Janssen, L. L. Verhoeff, C. G. C. Horlings, and J. H. J. Allum, “Directional effects of biofeedback on trunk sway during gait tasks in healthy young subjects.,” *Gait Posture*, vol. 29, no. 4, pp. 575–81, Jun. 2009.
- [147] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “The effect of vibrotactile feedback on postural sway during locomotor activities.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 93, Jan. 2013.
- [148] M. Dozza, C. Wall, R. J. Peterka, L. Chiari, and F. B. Horak, “Effects of practicing tandem gait with and without vibrotactile biofeedback in subjects with unilateral vestibular loss.,” *J. Vestib. Res.*, vol. 17, no. 4, pp. 195–204, Jan. 2007.
- [149] Y. P. Danilov, M. E. Tyler, K. L. Skinner, and P. Bach-y-Rita, “Efficacy of electrotactile vestibular substitution in patients with bilateral vestibular and central balance loss.,” in *IEEE EMBS Annual International Conference*, 2006, pp. 6605–9.
- [150] M. Rossi-Izquierdo, A. Ernst, A. Soto-Varela, S. Santos-Pérez, A. Faraldo-García, A. Sesar-Ignacio, and D. Basta, “Vibrotactile neurofeedback balance training in patients with Parkinson’s disease: reducing the number of falls.,” *Gait Posture*, vol. 37, no. 2, pp. 195–200, Feb. 2013.
- [151] C. Brugnera, R. S. M. Bittar, M. E. Greters, and D. Basta, “Effects of vibrotactile vestibular substitution on vestibular rehabilitation - preliminary study,” *Braz. J. Otorhinolaryngol.*, vol. 81, no. 6, pp. 616–621, 2015.
- [152] M. G. Jorgensen, U. Laessoe, C. Hendriksen, O. B. F. Nielsen, and P. Aagaard, “Efficacy of nintendo wii training on mechanical leg muscle function and postural balance in community-dwelling older adults: A randomized controlled trial,” *Journals Gerontol.* -

- Ser. A Biol. Sci. Med. Sci.*, vol. 68, no. 7, pp. 845–852, 2013.
- [153] C. Whyatt, N. A. Merriman, W. R. Young, F. N. Newell, and C. Craig, “A Wii bit of fun: A novel platform to deliver effective balance training to older adults,” *Games Health J.*, vol. 4, no. 6, pp. 1–11, 2015.
- [154] S. Roopchand-Martin, R. McLean, C. Gordon, and G. Nelson, “Balance Training with Wii Fit Plus for Community-Dwelling Persons 60 Years and Older.,” *Games Health J.*, vol. 4, no. 3, pp. 247–52, 2015.
- [155] A. S. Fu, K. L. Gao, A. K. Tung, W. W. Tsang, and M. M. Kwan, “Effectiveness of Exergaming Training in Reducing Risk and Incidence of Falls in Frail Older Adults with a History of Falls,” *Arch. Phys. Med. Rehabil.*, vol. 96, no. 12, pp. 2096–2102, 2015.
- [156] S. B. Lim, B. C. Horslen, J. R. Davis, J. H. J. Allum, and M. G. Carpenter, “Benefits of multi-session balance and gait training with multi-modal biofeedback in healthy older adults,” *Gait Posture*, vol. 47, pp. 10–17, 2016.
- [157] F. B. Horak, “Postural compensation for vestibular loss and implications for rehabilitation.,” *Restor. Neurol. Neurosci.*, vol. 28, no. 1, pp. 57–68, Jan. 2010.
- [158] A. D. Goodworth, C. Wall, and R. J. Peterka, “Influence of feedback parameters on performance of a vibrotactile balance prosthesis,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 17, no. 4, pp. 397–408, 2009.
- [159] A. D. Goodworth, C. Wall, and R. J. Peterka, “A balance control model predicts how vestibular loss subjects benefit from a vibrotactile balance prosthesis.,” in *Annual International Conference of the IEEE EMBS*, 2011, pp. 1306–9.
- [160] B.-C. Lee, A. Ho, B. J. Martin, and K. H. Sienko, “Effects of co-vibrotactile stimulations around the torso on non-volitional postural responses.,” *Conf. Proc. ... Annu. Int. Conf.*

- IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2012, pp. 6149–52, Jan. 2012.
- [161] B. Lee, B. J. Martin, and K. H. Sienko, “Comparison of Non-Volitional Postural Responses Induced by Two Types of Torso Based Vibrotactile Stimulations,” pp. 195–198, 2012.
- [162] R. A. Keith, C. V. Granger, B. B. Hamilton, and F. S. Sherwin, “The functional independence measure: a new tool for rehabilitation,” *Adv Clin Rehabil*, vol. 1, pp. 6–18, 1987.
- [163] D. Espy, C. Nathan, W. Tony, K. Natalie, and R. M., “Development of a rating scale for perceived stability during balance training,” in *International Society for Posture and Gait Research*, 2014.
- [164] S. J. Russell, P. Norvig, J. F. Canny, J. M. Malik, and D. D. Edwards, *Artificial intelligence: a modern approach*, vol. 2, no. 9. Prentice hall Upper Saddle River, 2003.
- [165] A. Khan, B. Baharum, L. H. Lee, and K. Khairullah, “Journal of Advances in Information Fusion,” *J. Adv. Inf. Technol.*, vol. 1, no. 1, pp. 4–20, 2010.
- [166] Y. Saeys, I. Inza, and P. Larrañaga, “A review of feature selection techniques in bioinformatics,” *Bioinformatics*, vol. 23, no. 19, pp. 2507–2517, 2007.
- [167] S. B. Kotsiantis, I. D. Zaharakis, and P. E. Pintelas, “Machine learning: A review of classification and combining techniques,” *Artif. Intell. Rev.*, vol. 26, no. 3, pp. 159–190, 2006.
- [168] M. Längkvist, L. Karlsson, and A. Loutfi, “A review of unsupervised feature learning and deep learning for time-series modeling,” *Pattern Recognit. Lett.*, vol. 42, no. 1, pp. 11–24, 2014.

- [169] I. Kononenko, “Machine learning for medical diagnosis: History, state of the art and perspective,” *Artif. Intell. Med.*, vol. 23, no. 1, pp. 89–109, 2001.
- [170] J. Wiens, J. Guttag, and E. Horvitz, “Patient Risk Stratification with Time-Varying Parameters : A Multitask Learning Approach,” *J. Mach. Learn. Res. LEHD*, vol. 17, pp. 1–23, 2016.
- [171] M. Ghassemi, M. A. F. Pimentel, T. Naumann, T. Brennan, D. A. Clifton, P. Szolovits, and M. Feng, “A multivariate timeseries modeling approach to severity of illness assessment and forecasting in ICU with sparse, heterogeneous clinical data,” in *Proceedings of the... AAAI Conference on Artificial Intelligence. AAAI Conference on Artificial Intelligence*, 2015, vol. 2015, p. 446.
- [172] D. C. Kale, D. Gong, Z. Che, Y. Liu, G. Medioni, R. Wetzel, P. Ross, P. Laura, and L. K. Whittier, “An Examination of Multivariate Time Series Hashing with Applications to Health Care,” *IEEE Int. Conf. Data Min.*, pp. 260–269, 2014.
- [173] R. G. Khemani, E. B. Celikkaya, C. R. Shelton, D. Kale, P. a Ross, R. C. Wetzel, and C. J. L. Newth, “Algorithms to estimate PaCO₂ and pH using noninvasive parameters for children with hypoxemic respiratory failure.,” *Respir. Care*, vol. 59, no. 8, pp. 1248–57, 2014.
- [174] X. Wang, D. Sontag, and F. Wang, “Unsupervised Learning of Disease Progression Models,” pp. 85–94, 2014.
- [175] B. C. . Wallace, I. J. . Dahabreh, T. A. . Trikalinos, M. B. . Laws, I. . Wilson, and E. . Charniak, “Identifying differences in physician communication styles with a log-linear transition component model,” *Proc. Natl. Conf. Artif. Intell.*, vol. 2, no. January, pp. 1314–1320, 2014.

- [176] B. M. Marlin, D. C. Kale, R. G. Khemani, and R. C. Wetzel, “Unsupervised pattern discovery in electronic health care data using probabilistic clustering models,” *Proc. 2nd ACM SIGHIT Symp. Int. Heal. informatics - IHI '12*, p. 389, 2012.
- [177] D. Page, V. S. Costa, S. Natarajan, A. Barnard, P. Peissig, and M. Caldwell, “Identifying adverse drug events by relational learning,” in *Proceedings of the... AAAI Conference on Artificial Intelligence. AAAI Conference on Artificial Intelligence*, 2012, vol. 2012, p. 790.
- [178] S. Saria, A. K. Rajani, J. Gould, D. Koller, and A. a Penn, “Integration of early physiological responses predicts later illness severity in preterm infants.,” *Sci. Transl. Med.*, vol. 2, no. 48, p. 48ra65, 2010.
- [179] L. Bao and S. S. Intille, “Activity Recognition from User-Annotated Acceleration Data,” in *Pervasive Computing*, 2004, pp. 1–17.
- [180] A. Mannini and A. M. Sabatini, “Machine learning methods for classifying human physical activity from on-body accelerometers.,” *Sensors (Basel)*, vol. 10, no. 2, pp. 1154–75, Jan. 2010.
- [181] B. Toro, C. J. Nester, and P. C. Farren, “Cluster analysis for the extraction of sagittal gait patterns in children with cerebral palsy,” *Gait Posture*, vol. 25, no. 2, pp. 157–165, 2007.
- [182] A. Phinyomark, S. Osis, B. A. Hettinga, and R. Ferber, “Kinematic gait patterns in healthy runners: A hierarchical cluster analysis,” *J. Biomech.*, vol. 48, no. 14, pp. 3897–3904, 2015.
- [183] A. Mannini, D. Trojaniello, A. Cereatti, and A. M. Sabatini, “A machine learning framework for gait classification using inertial sensors: Application to elderly, post-stroke and huntington’s disease patients,” *Sensors (Switzerland)*, vol. 16, no. 1, 2016.
- [184] R. LeMoyne, F. Heerinckx, T. Aranca, R. De Jager, T. Zesiewicz, and H. J. Saal,

- “Wearable body and wireless inertial sensors for machine learning classification of gait for people with Friedreich’s ataxia,” *BSN 2016 - 13th Annu. Body Sens. Networks Conf.*, pp. 147–151, 2016.
- [185] S. Yeh, M. Huang, and P. Wang, “Machine learning-based assessment tool for imbalance and vestibular dysfunction with virtual,” *Comput. Methods Programs Biomed.*, vol. 116, no. 3, pp. 311–318, 2014.
- [186] R. Begg and J. Kamruzzaman, “A machine learning approach for automated recognition of movement patterns using basic, kinetic and kinematic gait data.,” *J. Biomech.*, vol. 38, no. 3, pp. 401–8, Mar. 2005.
- [187] D. Martelli, F. Artoni, V. Monaco, A. M. Sabatini, and S. Micera, “Pre-impact fall detection: Optimal sensor positioning based on a machine learning paradigm,” *PLoS One*, vol. 9, no. 3, pp. 1–8, 2014.

Chapter 2 Effects of Long-term Vestibular Rehabilitation Therapy with Vibrotactile Sensory Augmentation for People with Unilateral Vestibular Disorders

2.1. Introduction

From 2001 through 2004, approximately 35% of Americans age 40 years and older experienced some form of vestibular dysfunction, which is equivalent to approximately 16 million new vestibular patients per year [1]. People with vestibular disorders have increased dizziness, reduced postural control, increased fall risk and fear of falling, increased interruptions of daily activities, and the need for additional sick leave or medical consultation [2]–[4]. They also experience more anxiety and depression than people without vestibular disorders [5]. People with unilateral vestibular disorders (UVD) tend to fall toward the affected side during balance and gait exercises [2]. They often experience dizziness and instability while moving their heads or walking within dark environments. Vestibular rehabilitation therapy (VRT) is widely prescribed to reduce dizziness and improve balance control [6]–[15]. VRT is an exercise-based program that includes four major types of exercises to facilitate central compensation: gaze stability exercises, balance exercises, habituation exercises and walking for endurance [13]. During VRT, physical therapists tailor the exercises to the individual to optimize the effectiveness of VRT. [16]. However, VRT does not always result in a full recovery of function

or complete resolution of symptoms; individuals may reach a plateau or fail to achieve complete compensation [17]–[22].

Sensory augmentation (SA), which has been widely explored in the last few decades, provides additional cues to augment/substitute with intact sensory inputs from somatosensory, visual and vestibular systems [23]. Typical SA devices have one or multiple sensors that measure body motion and a wearable display that provides instructional cues (e.g., vibrotactile [24]–[30], visual [31], auditory [32], [33], electrotactile [23], [34], and multi-modal [35]–[37]). It has been demonstrated that body sway during balance tasks can be reduced in a real-time manner when SA is provided [23]–[37]. It is important to better understand whether the reduction of body sway with the use of SA is carried over to improved clinical outcomes (carry-over effects) and whether postural sway reduction and potential clinical outcome improvements are retained during subsequent long-term assessments (retention effects).

Previous studies found that body sway reductions were retained for hours to days following short-term (i.e., less than one week) balance training with SA [24], [34], [37], but a few studies investigated changes in body sway or clinical outcomes following multi-session (i.e., more than 1 week) training with SA [18], [19], [22], [38]. Basta et al. demonstrated reduced body sway and improved clinical outcome measures (e.g., Sensory Organization Test (SOT) and Dizziness Handicap Inventory (DHI)) in people with vestibular disorder who trained with vibrotactile SA over a two-week (i.e., ten-session) training program (carry-over effects), but observed no such effect in their placebo group training with a sham device which generated random SA cues [18]. Further, the balance improvements were retained at the three-month follow-up assessment (retention effects) [18]. Similarly, Barros et al. showed improved SOT scores among people with bilateral vestibular loss following a two-week (i.e., six-session)

balance training program with electrotactile SA (carry-over effects) [22]. Robinson et al. found that after 12 months of balance training with electrotactile SA, a participant with bilateral vestibular disorders improved balance-related outcome measures (SOT, Dynamic Gait Index (DGI)) after previously completing VRT (carry-over effects) [19]. However, none of these three studies included a control group that trained without SA. Brugnera et al. found that after a two-week VRT program people with vestibular disorders who trained with vibrotactile SA had improved clinical measures (e.g., SOT and Activities-specific Balance Confidence (ABC) scale) (carry-over effects), but participants who trained without vibrotactile SA showed no significant improvements [38]. No further assessments were conducted to investigate potential retention effects after VRT. In summary, previous studies found balance improvements when balance training was supplemented with SA, but they did not include a control group or conduct follow-up assessments.

The aims of this study were to investigate whether a course of six-week VRT training program (including standing, gait and Vestibular Ocular Reflex (VOR) gaze stabilization exercises) with vibrotactile SA leads to additional body sway reduction and clinical outcome improvements (carry-over effects) compared with VRT alone among participants with UVD who had previously completed the VRT program, and whether balance improvements are retained for up to six months after VRT (retention effects).

2.2. Methods

2.2.1. Participants

Sixteen people with UVD were recruited for study eligibility assessment by physical therapists and via flyers at the University of Pittsburgh Medical Center. A neurologist diagnosed

the participants with UVD based on presence of a reduced vestibular response greater than or equal to 24% on caloric testing. Participants were excluded if they had confounding neurologic or neuromuscular disorders, known pregnancy, recent lower extremity fractures/severe sprains (within the last six months), previous lower extremity joint replacement, incapacitating back or lower extremity pain, were unable to stand for three minutes without rest, if their bodies were too large for experimental equipment (waist circumference of > 50 inches; 290 pounds), or if they had a Montreal Cognitive Assessment score of less than 26 points. Seven referrals were excluded because of low Montreal Cognitive Assessment scores. All participants completed a standard VRT prior to enrollment. After enrollment, participants were randomly assigned to an experimental group (EG) or a control group (CG). Participants in the EG received supervised VRT combined with vibrotactile SA (except for gait exercises), and participants in the CG received supervised VRT without vibrotactile SA. The study was conducted at the University of Pittsburgh Medical Center. One participant in the CG dropped out because of an orthopedic injury unrelated to the study. In total, four participants in the EG (68.1 ± 7.5 yrs, one male) and four participants in the CG (63.1 ± 11.3 yrs, one male) completed the study as shown in Table 2.1.

Table 2.1: Demographic information of the participants with UVD who completed the study

ID	Age	Gender	Time enrolled post 1st course of VRT	Group Designation
1	64	M	56 months	Control
2	47	F	8 months	Control
3	67	F	3 months	Control
4	68	F	23 months	Experimental
5	63	F	144 months	Experimental
6	79	M	3 months	Experimental
7	74	F	54 months	Control
8	63	F	37 months	Experimental

All participants gave written informed consent and the study was conducted in accordance with the Declaration of Helsinki. The study protocol was reviewed and approved by the University of Pittsburgh Institutional Review Board (PRO13020399).

2.2.2. Protocol

Participants completed a 6-week, 18-session VRT training program with five assessment sessions as shown in Figure 2.1. The assessments were performed before training, midway through training, one week after the training, one month after the training, and six months after the training.

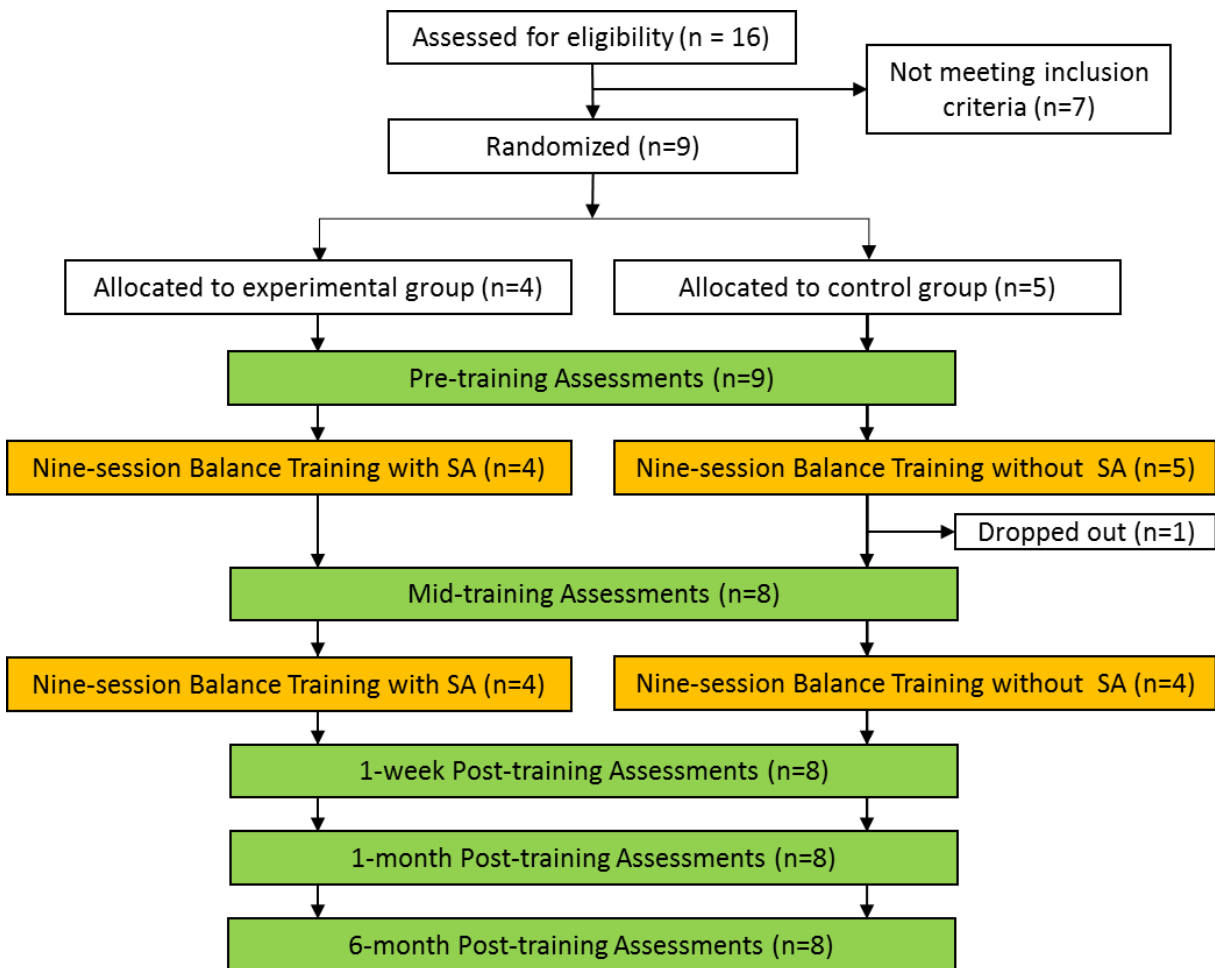


Figure 2.1: Study protocol consists of five assessment sessions (highlighted in green) and two nine-session rehabilitation balance training blocks (highlighted in orange).

A physical therapist, who was blinded to participants' group assignments, assessed each participant's balance. The assessment included a battery of clinical tests and five standing balance exercises. Clinical tests included 1) ABC Scale [39], [40], 2) DHI [41], 3) Computerized Dynamic Posturography: SOT [42], 4) Mini Balance Evaluations Systems Test (Mini-BESTest) [43], 5) Functional Reach Test (FRT) [44], 6) Gait Speed Test [45], 7) Timed Up and Go (TUG) [46], 8) DGI [47], and 9) Functional Gait Assessment (FGA) [48]. Using the results from the six SOT conditions, we calculated the individuals' reliance on the somatosensory input (ratio of SOT conditions 2 score to the SOT condition 1 score), visual input (ratio of SOT conditions 4 score to the SOT condition 1 score), and vestibular input (ratio of SOT conditions 5 score to the SOT condition 1 score) to maintain postural stability [15], [49].

The five standing balance exercises, shown in Table 2.2, were selected from a recently published conceptual progression framework at the easy to moderate difficulty level [16]. Specifically, Exercises 1-4 disturbed or removed one or more of the sensory inputs by varying visual inputs, stance, head movements and standing surface. Exercise 5 is a VOR gaze stabilization exercise. Each participant performed each exercise three times (i.e., three trials), and each trial lasted for 30 seconds or to the point of loss of balance. Trunk sway in anterior/posterior and medial/lateral directions was recorded. Balance performance was evaluated by three kinematic metrics including the root-mean-square (RMS) of trunk sway, percentage time within a one-degree zone (PZ), and the elliptical area fit of trunk sway (EA). Lower RMS, lower EA, and higher PZ indicate better balance performance [24], [50], [51].

Table 2.2: Five standing balance exercises were selected to measure postural stability [16].

Exercise 1	Feet apart stance on firm surface with eyes closed
Exercise 2	Romberg stance on firm surface with eyes closed and pitch head movements
Exercise 3	Romberg stance on foam surface with eyes open
Exercise 4	Semi-tandem Romberg stance on foam surface with eyes open
Exercise 5	Feet apart stance on firm surface with VOR gaze stabilization (maintain a clear and steady gaze on a fixed target while moving their head horizontally)

Before and after each balance training session, participants performed two normalization exercises without SA. Each exercise was performed twice. During the training session, participants performed exercises from the six categories (firm surface standing, compliant surface standing, weight shifting, modified center of gravity by raising arms, gait, and VOR exercises) as shown in Table 2.3. For the weight shifting category, all participants were asked to shift their weight to the target tilt value in either the forward or rightward direction and hold their position for five seconds. Then, they were asked to shift their weight to the target tilt value in either the backward or leftward direction and hold their position for five seconds. For the VOR exercise category, participants completed either VOR x1 or VOR x2 gaze stabilization [13], [16]. For VOR x1, participants were asked to maintain a clear and steady gaze on a target while moving their head 30 degrees horizontally or vertically at the fastest speed possible. For VOR x2, they were asked to maintain a clear and steady gaze on a target while moving their head and the target in opposite directions (30 degrees vertically or horizontally).

Table 2.3: Exercise pool adapted from a recently published conceptual progression framework [16]. (* exercises where EG received vibrotactile SA; † weight shifting limits were pre-determined, the maximum limits were 6 deg., 3 deg., 3.5 deg. and 3.5 deg. in the forward, backward, rightward, and leftward directions, respectively; the medium limit was half of maximum limit)

Category	Variables
1 Standing on firm surface*	Eyes (open/closed), stance (feet apart/Romberg/semi-tandem-Romberg/tandem/single leg), head movement (none/yaw/pitch)
2 Standing on foam surface*	Eyes(open/closed), stance (feet apart/Romberg/semi-tandem-Romberg/tandem), head movement (none/yaw/pitch)
3 Weight shifting*	Eyes (open/closed), standing surface (firm/foam), shifting limit (medium/max)†, shifting speed (fast/slow), shifting direction (forward→backward/right→left)
4 Modified center of gravity* (arm raises to 90°)	Eyes(open/closed), stance (feet apart/Romberg/semi-tandem Romberg), weight in hand (none/light/heavy), arm raising speed (fast/slow), surface (firm/foam/ramp inclined 10 degrees/ramp declined 10 degrees)
5 Gait	Eyes (open/closed), type of walking (normal, tandem, backward), head movement (none/yaw/pitch), walking speed (self-selected, fast slow)
6 Gaze Stabilization VOR*	Type of VOR (x1#/x2#), stance (feet apart/Romberg/semi-tandem Romberg/tandem), standing surface (firm/foam), distance to target (1m/3m), background of target (white/complex)

During each balance training session, participants performed one exercise from each of the six categories. Each exercise was performed six times (i.e., six trials). Trunk sway data was monitored by a customized SA device, which is detailed in the next section. Participants performed each exercise trial for 30 seconds, except for those in the weight shifting category. The physical therapist manually stopped the exercise if a participant needed to step out of position to maintain balance. For all exercise categories except for the gait category, vibrotactile SA was provided to the EG. Vibrotactile SA was provided in four randomly selected trials out of the six trials per exercise to enhance motor learning [52]. For all exercise categories, vibrotactile SA was not provided to the CG. Details of the SA device are described in the next section.

After each exercise, the physical therapist rated the participants' balance performance on a scale of one to five adapted from the Functional Independence Measure [53] as described in

Table 2.4. The physical therapist determined the set of exercises in the next training session using a recently published conceptual progression framework [16], participants' ratings of the performed exercises, and clinical judgment. The treating physical therapist was blinded to participants' pre-training and mid-training balance assessments.

Table 2.4: Balance performance scale adapted from the Functional Independence Measure [53]

Scale	Description
1	Independent with no sway
2	Supervision with minimal sway
3	Close supervision with moderate sway
4	Require physical therapist's assist (or step out) after 15 seconds
5	Unable/falls with immediate assist/step out

2.2.3. Instrumentation

The customized SA device shown in Figure 2.2 included an inertial measurement unit (IMU, MTx by Xsens), four C-2 tactors (EAI Inc), a tactor controller unit (EAI Inc), a belt, a laptop (Dell Inc), and customized software written in C++. The IMU was placed on the back and the four tactors were placed on the front, back, left, and right sides of the torso at the L4/L5 level. The IMU's three-axis accelerometer and gyroscope were used to measure the acceleration and angular velocity with a sampling frequency of 100 Hz. All data were sent to the laptop to calculate trunk tilt and tilt rate in anterior/posterior and medial/lateral directions in real time. Trunk tilt plus half of the tilt rate was compared with a pre-set threshold to determine when to provide vibrotactile SA for firm surface standing, compliant surface standing, modified center of gravity by raising arms, and VOR exercises [24], [25], [54]. For the weight shifting exercises, only trunk tilt was used to determine when to provide vibrotactile SA. The pre-set thresholds were determined in pilot testing by experienced physical therapists who had completed multiple pilot trials for each exercise. Customized software displayed the trunk motion and tactor

activation information on the laptop screen. It also enabled the physical therapist to select individual exercises for each participant. The software was loaded with the preset factor activation thresholds and randomized the vibrotactile SA trials for the EG participants. The participants did not see the screen.

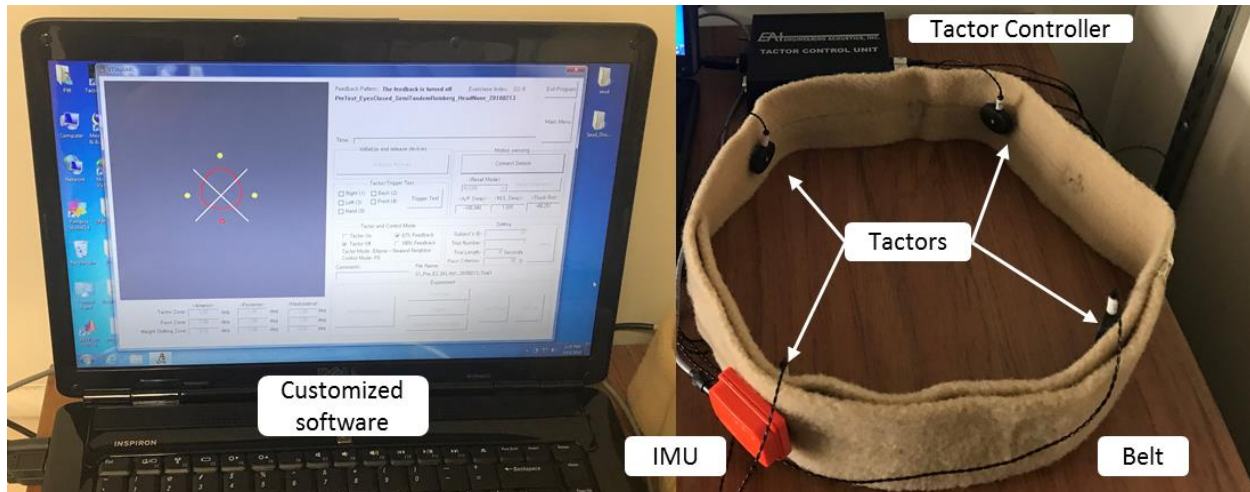


Figure 2.2: The customized sensory augmentation device included an inertial measurement unit (IMU), four C-2 tactors, a tactor controller unit, a belt, a laptop, and customized software.

2.2.4. Statistical Analysis

The differences between the EG and CG at the pre-training assessment were tested using an independent samples two-tailed student's t-test. Linear mixed effect models were used to analyze balance improvements and their persistence after the VRT training. The fixed factors were the group (EG, CG), assessment session (pre-, mid-, one-week post-, one-month post- and six-month post-training) and their interactions. The random effect was the difference among individual participants. For the kinematic metrics (RMS, PZ, and EA), three repeated exercise trials were averaged since there is no significant difference among the trials by using linear mixed effect models with trial as a fixed factor. The significance level was set at 0.05. All statistical analyses were performed in R (<https://www.r-project.org>). Due to the relatively small

sample size, the minimal detectable change (MDC) for some clinical outcome measures was used to determine whether the observed improvements achieved clinical significance [55].

2.3. Results

There were no statistical differences between the EG and CG in gender, age, clinical outcome measures, and standing balance exercise performance at pre-assessment.

The results of the clinical outcome measures are shown in Figure 2.3. There were significant main session effects (regardless of group) for SOT, FRT, and gait speed at the mid-training assessment ($p < 0.05$, $p < 0.05$, $p < 0.01$); for SOT, FRT, and FGA at the 1-week post-training assessment ($p < 0.05$); for SOT, gait speed, DGI, FGA and Mini-BESTest at the 1-month post-training assessment ($p < 0.01$, $p < 0.05$, $p < 0.001$, $p < 0.001$, $p < 0.05$); and for SOT, DGI, Mini-BESTest at the 6-month post-training assessment ($p < 0.01$, $p < 0.05$, $p < 0.05$). For the ABC scale, there was a significant interaction effect during the 1-week and 6-month post-training assessments ($p < 0.05$); specifically, the ABC scores for the EG increased and the ABC scores for the CG decreased. For the SOT, one participant in both the EG and CG achieved an MDC of 8 points [56] in the follow-up assessments. For the Mini-BESTest, one participant in the EG achieved an MDC of 3.5 points [57], but no participants in the CG achieved this MDC in the follow-up assessments.

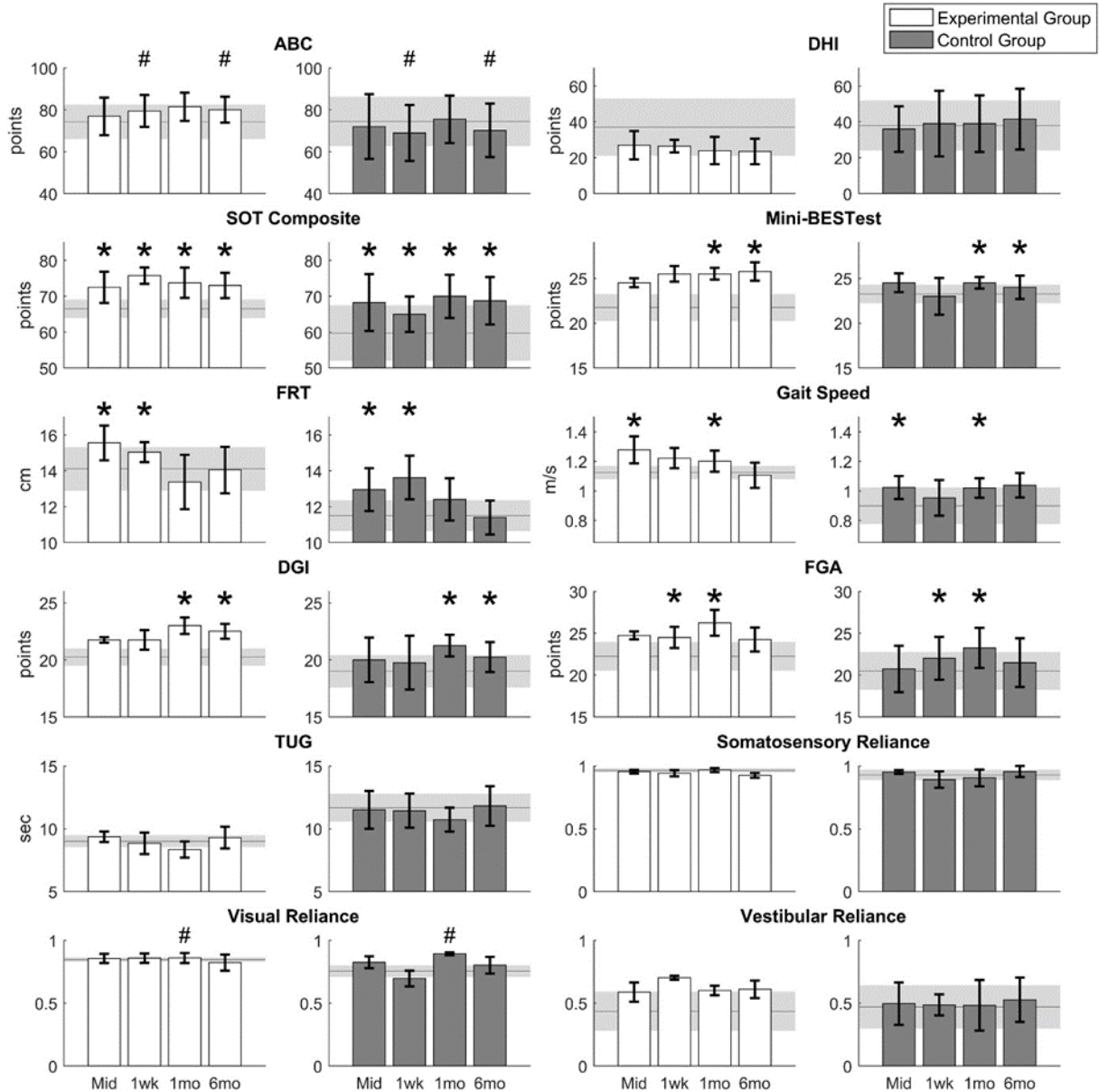


Figure 2.3: Results of clinical outcome measures for eight participants with UVD. Horizontal lines and shaded areas denote the mean pre-training assessment values and standard errors of the means, respectively. Columns and error bars denote the mean mid-training (Mid), 1-week (1wk), 1-month (1mo), and 6-month (6mo) post-training assessment values and standard error of mean. An asterisk denotes the significant main session effects, and a hash mark denotes the significant interaction effects between groups and sessions.

Balance performance outcomes for the five standing balance exercises are shown in Figure 2.4. Due to time constraints, Participant # 8 in the CG did not perform the balance exercises at the 6-month post-training assessment. For Exercises 1, 3, and 4, there were no

significant main session effects or interaction effects. For Exercise 2, there were significant increases (regardless of group) for the PZ metric at the mid-training assessment regardless of group ($p < 0.05$). At the 1-week and 6-month post-assessments, there were significant interaction effects for the RMS and EA metrics ($p < 0.05$); the EG showed improvements while the CG performed worse compared to the pre-training assessment. For Exercise 5, there were significant interaction effects for the RMS and EA metrics at the 1-week post-training assessments ($p < 0.05$); the EG showed improvements while the CG performed worse compared to the pre-training assessment.

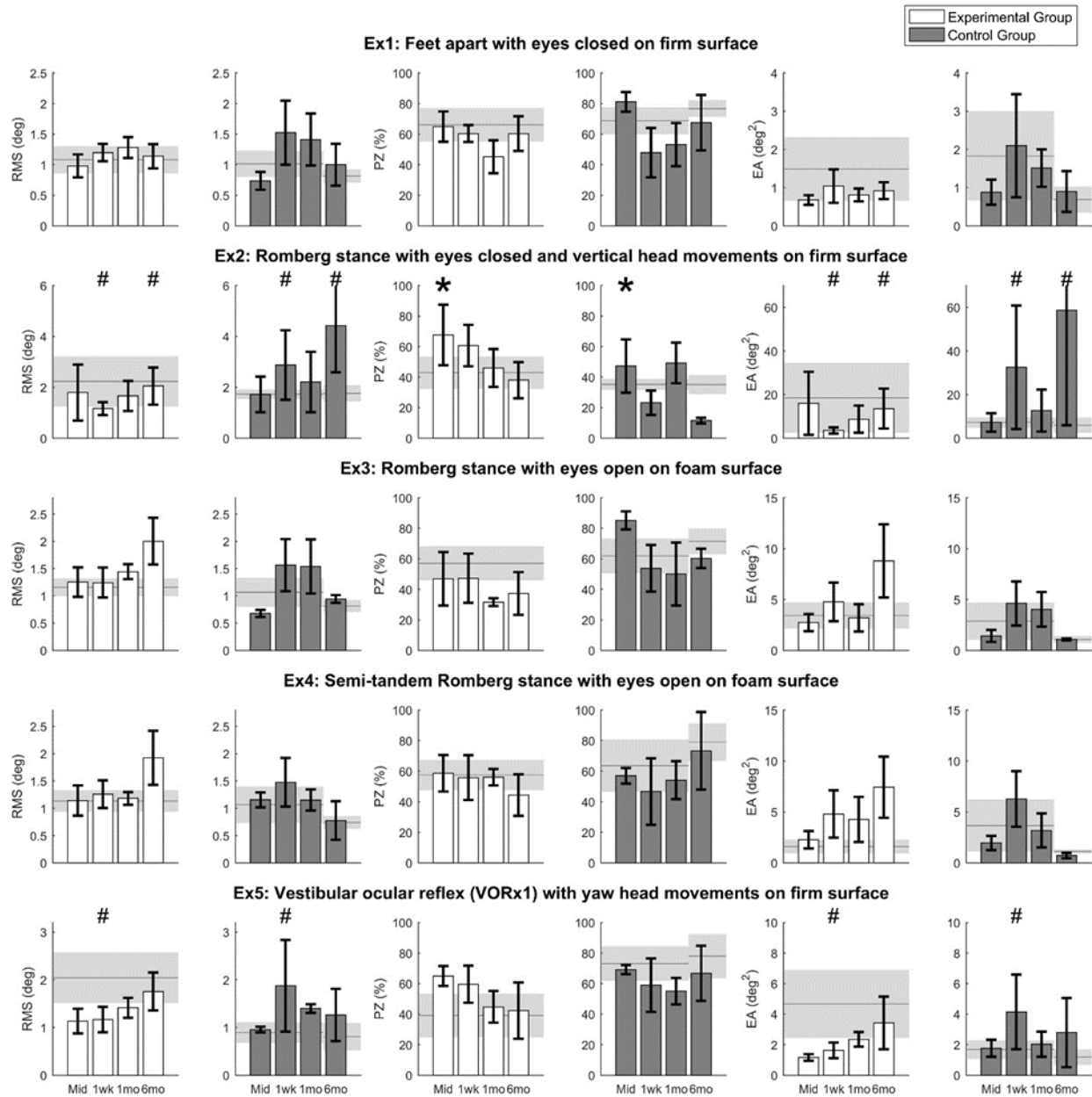


Figure 2.4: Results of five standing balance exercise performance for eight participants with UVD. Horizontal lines and shaded areas denote the mean pre-training assessment values and standard errors of the means, respectively. Columns and error bars denote the mean mid-training (Mid), 1-week (1wk), 1-month (1mo), and 6-month (6mo) post-training assessment values and standard error of mean, respectively. An asterisk denotes the significant main session effects, and a hash mark denotes the significant interaction effects between groups and sessions. Lower RMS, lower EA, and higher PZ indicate better balance performance.

2.4. Discussion

This study investigated balance improvements after an 18-session VRT training with and without vibrotactile SA, and the persistence of balance improvements for up to six months after completion of training. Compared to previous studies with SA, this study's unique methodological procedures, including reduced feedback frequency, customized exercise programs, customized thresholds, and a longer training period were designed to optimize functional recovery [18], [52].

Notably, the training with SA led to significantly greater improvements than training alone for the ABC scale and two standing balance exercises (i.e., Exercise 2: Romberg on firm surface with eyes closed and pitch head movements, and Exercise 5: feet apart on firm surface with VOR gaze stabilization) at the one-week post-training assessment. Moreover, the difference between the two groups was still observed at the one-month and six-month post-training assessments, but the significance was only observed for the ABC scale and Exercise 2 at the six-month post-training assessment.

For the ABC at the one-week post-training assessment, the EG's ABC score improved by 5.3 points and the CG's ABC score got worse by 5.5 points. Our findings are similar to a preliminary study by Brugnera et al., which reported that the EG and CG gained balance confidence by approximately 20 points and 11 points, respectively [38]. However, the extent of the improvements and changes in balance confidence were not equivalent. The differences could be due to the diagnoses of participants and their pre-training performance. In the study by Brugnera et al., most of the participants were diagnosed with bilateral vestibular disorders, and the pre-training measures of ABC were lower than our study (69 points vs 74 points for the EG and 59 points vs 74 points for the CG).

The primary differentiator between the standing balance exercises where there was a significant group difference (i.e., Exercises 2 and 5) and the standing balance exercises where there was no significant group difference (i.e., Exercises 1, 3 and 4) is the presence of head movements. This suggests that training with SA may be more effective at improving postural stability during activities that incorporate dynamic head movements. Given that the EG exhibited larger increases in vestibular reliance values, its participants may have been able to maintain better balance during the balance exercises that challenge the vestibular system. In addition, Exercise 5 was a VOR-based exercise; training with SA may have helped the participants to utilize vestibular inputs to control balance during the dynamic tasks [58].

We found that training with SA led to larger increases than training alone for the SOT, Mini-BESTest, gait speed, DGI, and FGA at the one-week post-training assessment, although there was no significant difference for these tests ($p = 0.07$ for the Mini-BESTest). The differences between the two groups were still observable for the Mini-BESTest, DGI and FGA at one-month and six-month post-training assessments. The small sample size in this study may have caused the lack of significance. Furthermore, for the gait-related measures (i.e., gait speed, DGI, FGA, and TUG), the lack of significant differences between the CG and EG may also be attributed to the lack of vibrotactile SA provided during gait exercises for the EG, i.e., neither group received SA during gait exercises. The difference between the two groups may have resulted from the carry-over effects of the standing balance training, since standing stability and locomotor performance can be highly related [59].

For the SOT, Basta et al. found that the average change after training with vibrotactile SA was approximately 7 points for the SOT for people with UVD [18], which is slightly smaller than the improvement in our study (9.3 ± 3.3 points). There are differences between the training

period and training exercises between our study and the study by Basta et al., i.e., their study had 10 training sessions and six exercises, whereas our study included 18 training sessions and a customized exercise progression prescribed by a physical therapist using a progression protocol [18]. Based on the results of earlier studies, a longer training period and customized exercise progression might lead to greater improvements [18], [60].

When the sensory systems are intact, balance is properly maintained by the feedback mechanism that the central nervous system weights the intact sensory inputs from somatosensory, visual and vestibular systems [61]. During VRT, the gaze stability and balance exercises are specifically designed to disturb or remove one or more of three sensory inputs and promote their sensory reweights [15], [13], [16]. Aligned with this, one of the hypotheses for balance improvements after training with SA is that the use of an additional channel of information may facilitate the ability to reweight the intact sensory inputs and achieve better balance performance [62], [63]. In our study, calculating the vestibular reliance values from the SOT protocol showed that the training with SA led to a larger increase for the vestibular reliance than training without SA. Badke et al. demonstrated a significant improvement (~ 0.19 on average) for the vestibular reliance among people with peripheral vestibular dysfunction after the completion of VRT [15], but in our study the CG who received VRT alone showed no such improvements. A possible explanation is the participants in our CG had already completed a standard VRT beforehand, thus may have reached a functional plateau. On the other hand, the participants in our EG were still able to achieve an increase with the use of SA despite completing the standard VRT prior to our study. This latter finding suggests that training with SA may induce the central nervous system to weigh the remaining functional vestibular inputs

higher compared to training without SA, which may have provided support for the sensory-reweighting mechanism hypothesis.

Overall, the VRT training with vibrotactile SA may provide added benefits (i.e., body sway reduction and carry-over functional improvements) than training alone, which may be retained up to six months after training. These findings suggest that the use of vibrotactile SA may also be used as a rehabilitation tool to enhance the VRT for people with UVD. The participants in our study were limited to those who had normal motor and cognitive function (e.g., could stand for three minutes without rest; no lower extremity joint replacement; no back or lower pain). However, it is not clear whether our findings can be generalized to all people with UVD.

This study has two limitations. First, the sample size was small. The required 18 visits spread across six consecutive weeks and the post-training assessments for up to six months after completing the program made it difficult to recruit more participants. Several studies investigated home-based balance training with SA [64]–[68]. Jorgensen et al. and Whyatt et al. demonstrated that video game-based training with the Nintendo Wii board improved ABC scores and reduced body sway among older adults [64], [65]. In these studies, participants performed balance exercises while playing video games. Recently, Bao et al. demonstrated that home-based balance training with vibrotactile SA led to greater improvements in the SOT and Mini-BESTest scores than training without SA among older adults [68]. The smart phone balance trainer developed by Bao et al. provided vibrotactile SA to the exerciser and allowed the physical therapist to prescribe and monitor the exercises remotely [68]. Developing VRT with SA and built-in real time therapeutic support for in-home use is an attractive alternative for people with UVD who cannot attend training sessions in clinics. Second, vibrotactile SA was not provided to

the EG during gait exercises. To date, only a few studies have investigated the use of vibrotactile SA during gait exercises [54], [63], [69]; therefore, we suggest developing useful feedback strategies to support locomotor-based balance training as a logical extension of our research.

2.5. Conclusion

VRT with vibrotactile SA introduced additional benefits for ABC scale and balance exercises with head movements than VRT without SA. Real-time body sway reduction during training with SA may carry over to improved clinical outcome measures, and the improvements may be retained for up to six months after the completion of VRT with SA. Vibrotactile SA may further rehabilitation outcomes when combined with standard VRT.

2.6. References

- [1] Y. Agrawal, J. P. Carey, C. C. Della Santina, M. C. Schubert, L. B. Minor, C. C. Della Santina, M. C. Schubert, and L. B. Minor, “Disorders of balance and vestibular function in US adults: data from the National Health and Nutrition Examination Survey, 2001-2004,” *Arch. Intern. Med.*, vol. 169, no. 10, pp. 938–944, May 2009.
- [2] J. H. J. Allum and A. L. Adkin, “Improvements in Trunk Sway Observed for Stance and Gait Tasks during Recovery from an Acute Unilateral Peripheral Vestibular Deficit,” *Audiol. Neuro-Otology*, vol. 8, no. 5, pp. 286–302, 2003.
- [3] H. K. Neuhauser, “Epidemiology of vertigo,” *Curr. Opin. Neurol.*, vol. 20, no. 1, pp. 40–46, 2007.
- [4] S. J. Herdman, P. Blatt, M. C. Schubert, and R. J. Tusa, “Falls in patients with vestibular deficits,” *Am. J. Otol.*, vol. 21, no. 6, pp. 847–851, 2000.

- [5] D. Monzani, L. Casolari, G. Guidetti, and M. Rigatelli, “Psychological distress and disability in patients with vertigo.,” *J. Psychosom. Res.*, vol. 50, no. 6, pp. 319–323, 2001.
- [6] A. Meli, G. Zimatore, C. Badaracco, E. De Angelis, and D. Tufarelli, “Vestibular rehabilitation and 6-month follow-up using objective and subjective measures.,” *Acta Otolaryngol.*, vol. 126, no. 3, pp. 259–66, 2006.
- [7] S. L. Whitney and M. M. Rossi, “Efficacy of vestibular rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 659–672, Jun. 2000.
- [8] J. L. Cowand, D. M. Wrisley, M. Walker, B. Strasnick, and J. T. Jacobson, “Efficacy of Vestibular Rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 118, pp. 49–54, 1998.
- [9] S. L. Whitney, A. H. Alghadir, and S. Anwer, “Recent Evidence About the Effectiveness of Vestibular Rehabilitation,” *Curr. Treat. Options Neurol.*, vol. 18, no. 3, pp. 1–15, 2016.
- [10] E. E. Hansson, N.-O. Månsson, K. A. Ringsberg, and A. Håkansson, “Falls among dizzy patients in primary healthcare: an intervention study with control group.,” *Int. J. Rehabil. Res.*, vol. 31, no. 1, pp. 51–7, Mar. 2008.
- [11] J. Y. Jung, J.-S. Kim, P. S. Chung, S. H. Woo, and C. K. Rhee, “Effect of vestibular rehabilitation on dizziness in the elderly.,” *Am. J. Otolaryngol.*, vol. 30, no. 5, pp. 295–9, Jan. 2009.
- [12] A. Shumway-Cook, F. B. Horak, L. Yardley, and A. M. Bronstein, “Rehabilitation of balance disorders in the patient with vestibular pathology,” *Clin. Disord. Balanc. Posture, Gait. London, United Kingdom Arnold, Div Hodder Headl. PLC*, pp. 213–220, 1996.
- [13] C. D. Hall, S. J. Herdman, S. L. Whitney, S. P. Cass, R. A. Clendaniel, T. D. Fife, J. M. Furman, T. S. D. Getchius, J. A. Goebel, N. T. Shepard, and S. N. Woodhouse, *Vestibular Rehabilitation for Peripheral Vestibular Hypofunction: An Evidence-Based Clinical*

Practice Guideline, vol. 40, no. April. 2016.

- [14] S. L. Hillier and M. McDonnell, “Vestibular rehabilitation for unilateral peripheral vestibular dysfunction.” *Clin. Otolaryngol.*, vol. 36, no. 3, pp. 248–249, 2011.
- [15] M. B. Badke, J. A. Miedaner, C. R. Grove, T. A. Shea, and G. M. Pyle, “Effects of vestibular and balance rehabilitation on sensory organization and dizziness handicap,” *Ann. Otol. Rhinol. Laryngol.*, vol. 114, no. 1 I, pp. 48–54, 2005.
- [16] B. N. Klatt, W. J. Carender, C. Lin, S. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, “A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular Disorders,” *Phys. Med. Rehabil. - Int.*, vol. 2, no. 4, p. 1044, 2015.
- [17] S. J. Herdman and R. A. Clendaniel, *Vestibular Rehabilitation*, Fourth. 2014.
- [18] D. Basta, M. Rossi-Izquierdo, A. Soto-Varela, M. E. Greters, R. S. Bittar, E. Steinhagen-Thiessen, R. Eckardt, T. Harada, F. Goto, K. Ogawa, and A. Ernst, “Efficacy of a vibrotactile neurofeedback training in stance and gait conditions for the treatment of balance deficits: a double-blind, placebo-controlled multicenter study.” *Otol. Neurotol.*, vol. 32, no. 9, pp. 1492–9, Dec. 2011.
- [19] B. S. Robinson, J. L. Cook, C. M. Richburg, and S. E. Price, “Use of an electrotactile vestibular substitution system to facilitate balance and gait of an individual with gentamicin-induced bilateral vestibular hypofunction and bilateral transtibial amputation.” *J. Neurol. Phys. Ther.*, vol. 33, no. 3, pp. 150–159, 2009.
- [20] K. E. Brown, S. L. Whitney, D. M. Wrisley, and J. M. Furman, “Physical therapy outcomes for persons with bilateral vestibular loss.” *Laryngoscope*, vol. 111, no. 10, pp. 1812–1817, 2001.

- [21] K. I. Ites, E. J. Anderson, M. L. Cahill, J. A. Kearney, E. C. Post, and L. S. Gilchrist, "Balance Interventions for Diabetic Peripheral Neuropathy: A Systematic Review," *J. Geriatr. Phys. Ther.*, vol. 34, pp. 109–116, 2011.
- [22] C. G. C. Barros, R. S. M. Bittar, and Y. Danilov, "Effects of electrotactile vestibular substitution on rehabilitation of patients with bilateral vestibular loss.," *Neurosci. Lett.*, vol. 476, no. 3, pp. 123–6, Jun. 2010.
- [23] P. Bach-y-Rita, "Tactile sensory substitution studies.," *Ann. N. Y. Acad. Sci.*, vol. 1013, pp. 83–91, May 2004.
- [24] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, "Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations," *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.
- [25] C. Wall III and E. Kentala, "Effect of displacement, velocity, and combined vibrotactile tilt feedback on postural control of vestibulopathic subjects.," *J. Vestib. Res.*, vol. 20, no. 1, pp. 61–9, Jan. 2010.
- [26] K. H. Sienko, M. D. Balkwill, and C. Wall, "Biofeedback improves postural control recovery from multi-axis discrete perturbations.," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 53, 2012.
- [27] K. E. Bechly, W. J. Carender, J. D. Myles, and K. H. Sienko, "Determining the preferred modality for real-time biofeedback during balance training.," *Gait Posture*, vol. 37, no. 3, pp. 391–6, Mar. 2013.
- [28] C. Wall, D. M. Wrisley, and K. D. Statler, "Vibrotactile tilt feedback improves dynamic gait index: a fall risk indicator in older adults.," *Gait Posture*, vol. 30, no. 1, pp. 16–21,

- Jul. 2009.
- [29] M. Janssen, R. Stokroos, J. Aarts, R. van Lummel, and H. Kingma, “Salient and placebo vibrotactile feedback are equally effective in reducing sway in bilateral vestibular loss patients.,” *Gait Posture*, vol. 31, no. 2, pp. 213–7, Feb. 2010.
- [30] J. H. J. Allum, M. G. Carpenter, B. C. Horslen, J. R. Davis, F. Honegger, K.-S. Tang, and P. Kessler, “Improving impaired balance function: real-time versus carry-over effects of prosthetic feedback.,” *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2011, pp. 1314–8, Jan. 2011.
- [31] Z. Halická, J. Lobotková, K. Bučková, and F. Hlavačka, “Effectiveness of different visual biofeedback signals for human balance improvement,” *Gait Posture*, vol. 39, no. 1, pp. 410–414, 2014.
- [32] M. Dozza, F. B. Horak, and L. Chiari, “Auditory biofeedback substitutes for loss of sensory information in maintaining stance.,” *Exp. brain Res.*, vol. 178, no. 1, pp. 37–48, Mar. 2007.
- [33] J. Hegeman, F. Honegger, M. Kupper, and J. H. J. Allum, “The balance control of bilateral peripheral vestibular loss subjects and its improvement with auditory prosthetic feedback.,” *J. Vestib. Res.*, vol. 15, no. 2, pp. 109–17, Jan. 2005.
- [34] Y. P. Danilov, M. E. Tyler, K. L. Skinner, and P. Bach-y-Rita, “Efficacy of electrotactile vestibular substitution in patients with bilateral vestibular and central balance loss.,” in *IEEE EMBS Annual International Conference*, 2006, pp. 6605–9.
- [35] J. L. Huffman, L. E. Norton, A. L. Adkin, and J. H. J. Allum, “Directional effects of biofeedback on trunk sway during stance tasks in healthy young adults,” *Gait Posture*, vol. 32, no. 1, pp. 62–66, 2010.

- [36] L. J. F. Janssen, L. L. Verhoeff, C. G. C. Horlings, and J. H. J. Allum, "Directional effects of biofeedback on trunk sway during gait tasks in healthy young subjects.," *Gait Posture*, vol. 29, no. 4, pp. 575–81, Jun. 2009.
- [37] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, "Trunk sway reductions in young and older adults using multi-modal biofeedback.," *Gait Posture*, vol. 31, no. 4, pp. 465–72, Apr. 2010.
- [38] C. Brugnera, R. S. M. Bittar, M. E. Greters, and D. Basta, "Effects of vibrotactile vestibular substitution on vestibular rehabilitation - preliminary study," *Braz. J. Otorhinolaryngol.*, vol. 81, no. 6, pp. 616–621, 2015.
- [39] L. E. Powell and A. M. Myers, "The activities-specific balance confidence (ABC) scale," *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 50, no. 1, pp. M28--M34, 1995.
- [40] Y. Lajoie and S. P. Gallagher, "Predicting falls within the elderly community: Comparison of postural sway, reaction time, the Berg balance scale and the Activities-specific Balance Confidence (ABC) scale for comparing fallers and non-fallers," *Arch. Gerontol. Geriatr.*, vol. 38, no. 1, pp. 11–26, 2004.
- [41] G. P. Jacobson and C. W. Newman, "The development of the Dizziness Handicap Inventory.," *Arch. Otolaryngol. - Head Neck Surg.*, vol. 116, no. 4, pp. 424–427, 1990.
- [42] R. A. Clendaniel, "Outcome measures for assessment of treatment of the dizzy and balance disorder patient," *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 519–533, 2000.
- [43] F. Franchignoni, F. Horak, M. Godi, A. Nardone, and A. Giordano, "Using psychometric techniques to improve the balance evaluation systems test: The mini-bestest," *J. Rehabil. Med.*, vol. 42, no. 4, pp. 323–331, 2010.
- [44] P. W. Duncan, D. K. Weiner, J. Chandler, and S. Studenski, "Functional reach: a new

- clinical measure of balance.,” *J. Gerontol.*, vol. 45, no. 6, pp. M192–M197, 1990.
- [45] S. Perera, S. H. Mody, R. C. Woodman, and S. A. Studenski, “Meaningful change and responsiveness in common physical performance measures in older adults,” *J. Am. Geriatr. Soc.*, vol. 54, no. 5, pp. 743–749, 2006.
- [46] D. Podsiadlo and S. Richardson, “The Timed ‘Up & Go’: A Test of Basic Functional Mobility for Frail Elderly Persons,” *J. Am. Geriatr. Soc.*, vol. 39, no. 2, pp. 142–148, 1991.
- [47] A. Shumway-Cook and M. H. Woollacott, *Motor control: theory and practical applications*. Lippincott Williams & Wilkins, 1995.
- [48] D. M. Wrisley, G. F. Marchetti, D. K. Kuharsky, and S. L. Whitney, “Reliability, internal consistency, and validity of data obtained with the functional gait assessment.,” *Phys. Ther.*, vol. 84, no. 10, pp. 906–918, 2004.
- [49] L. M. Nashner, F. O. Black, and C. Wall, “Adaptation to altered support and visual conditions during stance: patients with vestibular deficits.,” *J. Neurosci.*, vol. 2, no. 5, pp. 536–44, 1982.
- [50] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, “Cell phone based balance trainer,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, Jan. 2012.
- [51] D. Basta, F. Singbartl, I. Todt, A. Clarke, and A. Ernst, “Vestibular rehabilitation by auditory feedback in otolith disorders.,” *Gait Posture*, vol. 28, no. 3, pp. 397–404, Oct. 2008.
- [52] S. K. Subramanian, C. L. Massie, M. P. Malcolm, and M. F. Levin, “Does provision of extrinsic feedback result in improved motor learning in the upper limb poststroke? A systematic review of the evidence.,” *Neurorehabil. Neural Repair*, vol. 24, no. 2, pp. 113–

- 124, 2010.
- [53] R. A. Keith, “The functional independence measure: a new tool for rehabilitation,” *Adv Clin Rehabil*, vol. 1, pp. 6–18, 1987.
- [54] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “The effect of vibrotactile feedback on postural sway during locomotor activities.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 93, Jan. 2013.
- [55] “Rehabilitation Measures Database.” [Online]. Available: <https://www.sralab.org/statistical-terms-use>. [Accessed: 09-Oct-2017].
- [56] D. M. Wrisley, M. J. Stephens, S. Mosley, A. Wojnowski, J. Duffy, and R. Burkard, “Learning Effects of Repetitive Administrations of the Sensory Organization Test in Healthy Young Adults,” *Arch. Phys. Med. Rehabil.*, vol. 88, no. 8, pp. 1049–1054, 2007.
- [57] M. Godi, F. Franchignoni, M. Caligari, A. Giordano, A. M. Turcato, and A. Nardone, “Comparison of Reliability, Validity, and Responsiveness of the Mini- BESTest and Berg Balance Scale in Patients With Balance Disorders,” *Phys. Ther.*, vol. 93, no. 2, pp. 158–167, 2013.
- [58] M. C. Schubert and L. B. Minor, “Vestibulo-ocular Physiology Underlying Vestibular Hypofunction,” *PHYS THER. Phys. Ther.*, vol. 84, no. 4, pp. 373–385, 2004.
- [59] M. A. Dettmann, M. T. Linder, and S. B. Sepic, “Relationships among walking performance, postural stability, and functional assessments of the hemiplegic patient.,” *Am. J. Phys. Med. Rehabil.*, vol. 66, no. 2, pp. 77–90, 1987.
- [60] M. Lesinski, T. Hortobágyi, T. Muehlbauer, A. Gollhofer, and U. Granacher, “Effects of Balance Training on Balance Performance in Healthy Older Adults : A Systematic Review and Meta-analysis,” *Sport. Med.*, vol. 45, no. 12, pp. 1721–38, Sep. 2015.

- [61] R. J. Peterka, "Sensorimotor Integration in Human Postural Control," *J. Neurophysiol.*, vol. 88, no. 3, pp. 1097–1118, 2002.
- [62] F. B. Horak, "Postural compensation for vestibular loss and implications for rehabilitation.," *Restor. Neurol. Neurosci.*, vol. 28, no. 1, pp. 57–68, Jan. 2010.
- [63] K. H. Sienko, S. L. Whitney, W. J. Carender, and C. Wall III, "The role of sensory augmentation for people with vestibular deficits: real-time balance aid and/or rehabilitation device?," *J. Vestib. Res.*, vol. 27, no. 1, pp. 63–76, 2017.
- [64] M. G. Jorgensen, U. Laessoe, C. Hendriksen, O. B. F. Nielsen, and P. Aagaard, "Efficacy of nintendo wii training on mechanical leg muscle function and postural balance in community-dwelling older adults: A randomized controlled trial," *Journals Gerontol. - Ser. A Biol. Sci. Med. Sci.*, vol. 68, no. 7, pp. 845–852, 2013.
- [65] C. Whyatt, N. A. Merriman, W. R. Young, F. N. Newell, and C. Craig, "A Wii bit of fun: A novel platform to deliver effective balance training to older adults," *Games Health J.*, vol. 4, no. 6, pp. 1–11, 2015.
- [66] S. Roopchand-Martin, R. McLean, C. Gordon, and G. Nelson, "Balance Training with Wii Fit Plus for Community-Dwelling Persons 60 Years and Older.," *Games Health J.*, vol. 4, no. 3, pp. 247–52, 2015.
- [67] A. S. Fu, K. L. Gao, A. K. Tung, W. W. Tsang, and M. M. Kwan, "Effectiveness of Exergaming Training in Reducing Risk and Incidence of Falls in Frail Older Adults with a History of Falls," *Arch. Phys. Med. Rehabil.*, vol. 96, no. 12, pp. 2096–2102, 2015.
- [68] T. Bao, W. J. Carender, C. Kinnaird, V. J. Barone, G. Peethambaran, S. L. Whitney, M. Kabeti, R. D. Seidler, and K. H. Sienko, "Effects of long-term balance training with vibrotactile sensory augmentation among community-dwelling healthy older adults," *J.*

Neuroeng. Rehabil., vol. 15, no. 5, pp. 1–13, 2018.

- [69] L. L. Verhoeff, C. G. C. Horlings, L. J. F. Janssen, S. a Bridenbaugh, and J. H. J. Allum, “Effects of biofeedback on trunk sway during dual tasking in the healthy young and elderly.,” *Gait Posture*, vol. 30, no. 1, pp. 76–81, Jul. 2009.

**Chapter 3 Effects of Long-term Balance Training with Vibrotactile Sensory
Augmentation among Community-dwelling Healthy Older Adults: A Randomized
Preliminary Study**

3.1. Background

Age-related deterioration of sensory function, inefficient integration of sensory systems, and reduced muscle strength contribute to decreased balance performance in older adults [1]–[3]. Degradation of balance performance increases fall risk [4]–[6] and fear of falling [7], and inhibits mobility, thereby reducing independence and quality of life [8], [9].

Exercise-based rehabilitation programs are effective for improving balance performance in community-dwelling older adults [10]. Typical regimens emphasize building strength and stamina to counteract musculoskeletal degeneration, and balance training to isolate and challenge the somatosensory, visual, and vestibular systems [10], [11]. In particular, balance training leverages the ability of the central nervous system to “reweight” functioning sensory inputs [12]. Clinic-, group-, and home-based balance training programs guided by clinicians and lasting four to twelve weeks have yielded significant improvements in clinical outcome measures (e.g., Sensory Organization Test, Timed Up-and-Go) corresponding directly to reductions in fall rate, occurrence, and risk, and maintenance of social and physical activity [13]–[16]. Training programs without direct clinical guidance, however, are less effective [17], [18]. Kao et al.

reported that more participants in a supervised exercise program group improved clinically significant outcomes than those performing the same exercises at home without supervision [17]. Lacroix et al. showed that a 12-week supervised balance and strength training program improved static, dynamic, proactive, and reactive measures of balance more than an unsupervised program in healthy older adults [18]. These studies suggest that monitoring performance and providing feedback during balance and strength training may improve program efficacy.

Sensory augmentation (SA) provides additional information to complement and/or replace native sensory input from the somatosensory, visual, and/or vestibular systems [19]. SA systems for balance applications typically employ one sensor or a network of sensors (e.g., motion capture, force plate, inertial measurement unit, goniometer) to measure body motion, and a display to communicate body motion and/or provide instructional cues. Most studies of balance-related SA have explored real-time usage applications as an assistive device in lieu of the somatosensory contact cues provided by a cane or walker to the fingertips [19]–[34]. During real-time use, it is hypothesized that the central nervous system incorporates SA as an additive input, supplementing other sensory information [22], [35], [36]. In individuals with balance deficits, healthy young adults, and healthy older adults, various SA modalities (e.g., vibrotactile [23]–[29], visual [30], auditory [31], [32], electrotactile [19], [37], and multi-modal [33]) have been shown to improve real-time balance performance by reducing body sway during static, dynamic, and perturbed standing tasks. While use of SA based on trunk sway during gait tasks (e.g., tandem walk, straight walk) has been shown to reduce trunk sway, the effects are limited [32], [38]–[40].

In addition to real-time assistive device applications, SA may be useful as a rehabilitation tool during traditional balance training. It is hypothesized that cues from SA may facilitate the

central nervous system in “reweighting” sensory inputs during training to improve balance ability [22], [41]. To evaluate whether SA devices can be used as rehabilitation tools, it is important to analyze whether balance improvements observed during training persist after training is completed and SA is not provided.

Prior studies have shown that post-training improvements persist hours to days following short-term (i.e., less than 1 week) training with SA [23], [33], [42] and on the order of weeks to months following multi-session (i.e., more than 1 week) training with SA for people with balance deficits [43]–[45]. Rossi-Izquierdo et al. showed that after two weeks of exercise training with vibrotactile SA, people with Parkinson’s disease reduced trunk sway on trained exercises and demonstrated improved clinical outcome measures (e.g., Dizziness Handicap Inventory, Sensory Organization Test) [43]. Furthermore, these improvements persisted three months after training [43]. Basta et al. reported similar improvements in a group of people with various balance disorders trained with vibrotactile SA, but found no such effect in a group trained with erroneous SA signals [44]. Brugnera et al. found improved clinical balance measures (e.g., Sensory Organization Test, Activities-specific Balance Confidence) among people with vestibular disorders following two weeks of balance training with vibrotactile SA, but found no improvements among participants trained following standard rehabilitation practices [45].

Limited studies have examined balance improvements after longer-term training with SA among healthy older adults. Video game–based in-home balance training was shown to improve clinical measures (e.g., maximal muscle strength, Activity-specific Balance Confidence, risk of falling) shortly after a minimum of five weeks of training [46]–[49]. However, video game–based balance training typically requires a balance platform (e.g., Wii Fit balance board) and a

display screen to provide visual cues, which can limit its utility during balance exercises that require closed eyes, head movements, and altered stances. Lim et al. used multi-modal SA to investigate balance improvements after a two-week balance training program involving 36 healthy older adults [50]. All participants wore a SA device (SwayStar™), but only the experimental group received SA. Participants' trunk sway was monitored as they trained on the same seven standing and gait tasks for two consecutive weeks (3x/week). Both experimental and control groups showed reduced body sway during the final training session, but training with SA provided little benefit over training alone. For most tasks, sway reductions did not persist in either group to immediate and one-month post-training assessments. However, given that balance training is most effective following longer training periods (i.e., up to 12 weeks [16]), balance improvements and the persistence of the potential improvements between the groups may not have been realized given the relatively short training period.

To understand the efficacy of SA as a rehabilitation tool among community-dwelling older adults, this preliminary study investigated balance improvements after long-term (eight weeks) balance training with and without SA. We hypothesized that all participants would show improved clinical outcome scores after training, but that participants receiving SA would show greater improvements.

3.2. Methods

3.2.1. Participants

Twelve community-dwelling healthy older adults were recruited to participate following a screening session. The sample size was partially informed by single day sensory augmentation study findings [23], [33], [51]. Participants in the greater Ann Arbor, MI area were recruited via

flyers and online advertisements on the website umhealthresearch.org. The recruiting period started in 2014 and ended in 2016. Participants were eligible for inclusion if they were 65-85 years of age; medically stable; scored more than 26 points on the Montreal Cognitive Assessment; could stand unassisted for ten minutes; reported balance concerns (≥ 1 confirmative answer to balance perception questions, e.g., fear of falling, falls in the past year, losses of balance in the past 12 months, balance ratings ≥ 2 on a five-point scale, Figure 3.2); and could walk the distance of a city block without using an assistive device. Participants who had sustained a fall that required hospitalization or serious injury, had severe uncorrected vision or hearing loss, had a lower extremity fracture or sprain in the last six months or previous lower extremity joint replacement, had a history of a neurological condition (e.g., Parkinson's disease, multiple sclerosis, stroke), had motion-provoked vertigo or diagnosed vestibular deficit, or had a body mass index larger than 30 kg/m^2 were excluded.

The twelve participants were randomly assigned to the experimental group (EG) or control group (CG) before pre-training assessments with a one-to-one allocation ratio. The EG ($n = 6$, 76.2 ± 5.5 yrs, 1 male/5 females) received vibrotactile SA during the training, while the CG completed the training without vibrotactile SA ($n = 6$, 75.0 ± 4.7 yrs, 3 males/3 females). The study team randomized the participant assignments by blindly drawing sealed slips of paper with group designations. The first two participants were randomized in one block and the following ten participants were randomized in a second block. All participants gave written informed consent and the study was conducted in accordance with the Declaration of Helsinki. The study was reviewed and approved by the University of Michigan Institutional Review Board (HUM00086479).

3.2.2. Protocol

The experimental protocol, as shown in Figure 3.1, comprised pre-training assessment with clinical balance testing (CBT), eight-week in-home balance training, mid-training assessment with CBT after four-week training, and post-training assessment with CBT. In-home balance training started within a week of the pre-training assessment and the post-assessment was completed within one week after training.

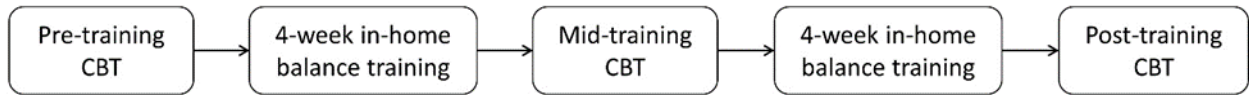


Figure 3.1: Study protocol includes three clinical balance testing (CBT) sessions and eight weeks of in-home balance training.

CBT, which included eight clinical outcome measures to evaluate balance and gait performance, was completed in the clinical setting by a physical therapist blinded to the participants' study group assignment (vibrotactile SA was not provided during CBT):

- 1) Activity-specific Balance Confidence (ABC, out of 100) [52]: Identifies an individual's subjective measure of confidence in performing balance related activities of daily living. An ABC score of less than 67 indicates an increased risk for falling [53].
- 2) Computerized Dynamic Posturography: Sensory Organization Test (SOT) protocol [54]: Assesses an individual's ability to use their somatosensory, visual, and vestibular systems to maintain postural stability during static standing, measured by SOT composite score. Somatosensory, visual and vestibular reliance are calculated based on the SOT to evaluate the reliance on each sensory system as shown below.

$$\text{Somatosensory Reliance} = \frac{\text{SOT condition 2}}{\text{SOT condition 1}} \quad \text{Equation 3-1}$$

$$\text{Visual Reliance} = \frac{\text{SOT condition 4}}{\text{SOT condition 1}} \quad \text{Equation 3-2}$$

$$\text{Vestibular Reliance} = \frac{\text{SOT condition 5}}{\text{SOT condition 1}} \quad \text{Equation 3-3}$$

- 3) Mini Balance Evaluations Systems Test (Mini-BESTest) [55]: Uses 14 items to capture anticipatory postural adjustments, reactive postural control, sensory orientation, and dynamic gait performance. The Mini-BESTest was measured with two scoring systems: total score of 28 points (MiniBESTest28) uses the lower score of the left and right sides for unilateral balance tasks; total score of 32 points (MiniBESTest32) uses the cumulative score of both sides.
- 4) Five Times Sit to Stand Test (5xSST) [56]: Tests dynamic activity of functional lower limb muscle strength and change of transitional movements, measured in seconds. In older adults, a 5xSST duration equal to or greater than 12 seconds indicates a need for additional fall assessment [57].
- 5) Four Square Step Test (FSST) [58]: Assesses the ability to step over objects forward, sideways, and backwards, measured in seconds. FSST duration greater than 15 seconds for older adults indicates an increased risk for multiple falls [58].
- 6) Functional Reach Test (FRT) [59]: Assesses stability by measuring the maximum distance reached forward with feet in a fixed position, measured in centimeters. A FRT of less than 18 cm indicates limited mobility skills for older adults [60].
- 7) Ten-meter walk test: Assesses normal gait speed and fast gait speed, measured in meters per second. A substantial meaningful change in normal gait speed is 0.13 m/s for older adults [61].
- 8) Timed Up and Go (TUG) and Timed Up and Go with Cognitive Task (TUG-COG) [62]: Assesses mobility, balance, and fall risk assessment with and without a cognitive dual-task (count backwards by three), measured in seconds. For older adults, a TUG score

greater than 13.5 seconds or the TUG-COG score greater than 15 seconds indicates fall risk [63].

After CBT but prior to beginning training, the treating physical therapist (different from the blinded assessor) and study team made one initial home visit to teach the participants how to use the smart phone balance trainer (detailed in the next section) and how to correctly perform independent in-home balance training exercises. Participants performed exercises from five categories as shown in Table 3.1. For exercises in Categories 1 (static standing) and 2 (compliant surface standing), participants performed static balance exercises on firm and foam surfaces, respectively. For Category 3 (weight shifting) exercises, participants were instructed to shift their body to and maintain their body at a target angle for five seconds in four directions (i.e., forward, backward, left and right). Movement angle was measured on the trunk, and the target angle was determined by the research team’s physical therapists and was the same for all participants. For Category 4 (modified center of gravity) exercises, participants raised and lowered their arms from a resting position along the sides of their bodies with palms pronated and elbows locked, to 90° of shoulder flexion repeatedly. For Category 5 (gait) exercises, participants performed various overground locomotor tasks.

Table 3.1: Exercise pool modified from a recently published conceptual progression framework [64]. *indicates the exercises for which vibrotactile SA was provided for the EG.

Category	Variables
1. Static standing*	Eyes, stance, head movement (yaw and pitch), cognitive tasks
2. Compliant surface standing*	Eyes, stance, head movement
3. Weight shifting*	Shifting limit, shifting speed, shifting direction
4. Modified center of gravity*	Arm raising speed, surface (firm, compliant and ramps), head movement
5. Gait	Walk with different speed and head movements, high march*, step over shoe box*, sidestepping, walk on heels/toes*, backward walking*, figure-of-8 walk, tandem*; cognitive tasks

Participants were asked to exercise three times per week for eight weeks (24 sessions in total). For each session, participants were given a single exercise from each of the first four categories and two exercises from the fifth category as remotely recommended by the treating physical therapist. Each exercise was performed six times for 30 seconds (except Category 3 where the trial stopped after participants maintained the target positions for five seconds). The training duration for each session was about 45 min. Vibrotactile SA was provided to the EG via the smart phone balance trainer for all the exercises in the first four exercise categories and select exercises in the fifth category, as shown in Table 3.1. For these exercises, vibrotactile SA was provided during four randomly selected repetitions out of the six repetitions. The CG also wore the smart phone balance trainer, but never received vibrotactile SA. After each trial, participants were prompted by the smart phone to note any step-outs that occurred. A step-out is defined as taking a step to regain balance, touching wall or chair for support, or opening eyes (on eyes closed tasks). After six repetitions participants rated their perceived stability on a visual analog scale of 1–5 (see Figure 3.2) [65]. The treating physical therapist used the reported number of step-outs and perceived stability scores to prescribe exercises weekly (three sessions per week) based on clinical experience and an exercise framework modified from a recently published conceptual progression framework [64]. The goal was to assign exercises that provided a moderate level of challenge, which was characterized by a score of 3 on the VAS. If there were no step outs and the participant rated the exercise a 1 on the VAS, a more challenging exercise was chosen; for example, adding pitch head movements would increase difficulty. If the exercises appeared too challenging (i.e., multiple step outs) and the participant rated the exercise a 4 or 5 on the VAS scale, an easier exercise was attempted until a moderately difficult exercise was found. The first set of exercises was determined during the initial home visit. Participants

were asked to complete a weekly activity log to note pain that limited movement, falls, changes in medication, and any injuries from performing the exercises. MRIs were performed on a subset of the participants (n=5) pre- and post-training for future analysis.

Scale	Description
1	I feel completely steady
2	I feel a little unsteady or off-balance
3	I feel somewhat unsteady or like I may lose my balance
4	I feel very unsteady or like I will lose my balance
5	I lost my balance

Figure 3.2 Visual analog scale used by participants to rate their stability when performing the balance exercises.

3.2.3. Smart Phone Balance Trainer

A smart phone balance trainer was developed using design ethnography techniques during a co-creative design process involving engineers, physical therapists, and older adults [66]. The smart phone balance trainer comprised two Apple iPods (6th generation iPod touch, 2015), an elastic belt and a “tactor bud” accessory, as shown in Figure 3.3. The two iPods are referred to as the “sensing” unit and “user interface” unit, respectively. The tactor bud contained a PCB-designed controller board, a 3.7V battery, and four tactors (Precision MicrodrivesTM, 310-101 vibration motors encased in plastic housings [51]). The sensing unit was attached to an elastic belt and was worn around the torso at the L4/L5 level to measure trunk sway, and the user interface unit attached to a lanyard and was worn around the neck. The four tactors were aligned over the navel, lumbar spine, and right and left sides of the torso to provide directional vibrotactile cues.

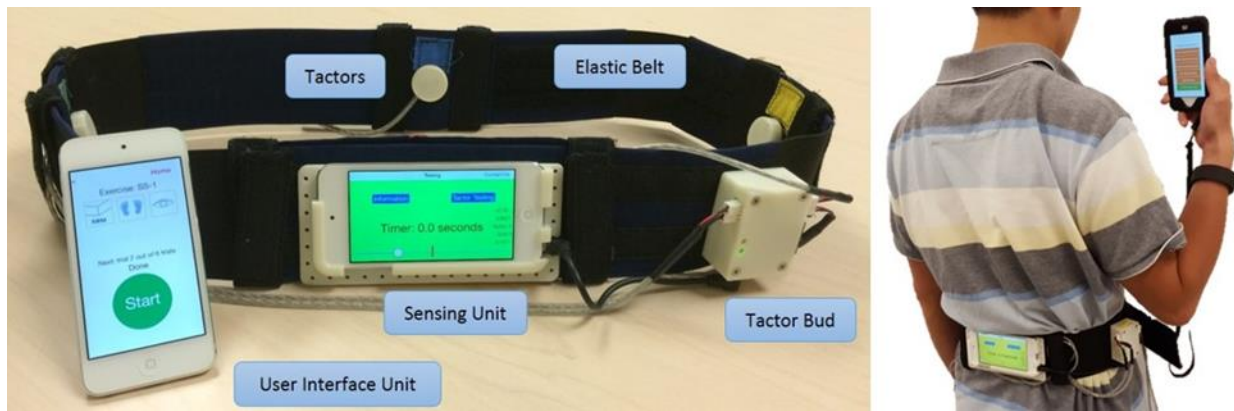


Figure 3.3: Smart phone balance trainer

Custom software (iOS application, Apple SDK) was developed to provide a semi-automated exercise progression routine with five exercise categories for in-home training, as shown in Figure 3.4. Upon the launch of the software in the user interface unit, participants were asked to select an exercise to perform. Written, graphic and additional video instructions were presented on screen once the exercise was selected.

During each repetition, the sensing unit used gravitational outputs (*Class CoreMotion*, Apple Inc.) to estimate angular displacements (tilt angles) in the anterior-posterior and medial-lateral directions, adopted from Lee et al.'s algorithm [51]. Angular velocities were measured by the gyroscopes. Both accelerometers and gyroscopes were sampled at 50 Hz. The user interface unit triggered the sensing unit to record trunk motion and the sensing unit informed the user interface unit of repetition completion via Bluetooth. The tactor activation signal was defined as the tilt angle plus one half of tilt angular rate for Categories 1, 2, 4 and 5, and as the tilt angle for Category 3 [24]. If the tactor activation signal exceeded a pre-set threshold [23], [38], the sensing unit sent audio output signals to the tactor bud accessory. The tactor bud accessory analyzed these audio signals and activated the corresponding tactor to provide vibrotactile SA. At the end of each repetition, the trunk motion data, number of step-outs, and visual analog scale ratings were uploaded to a secured server via Wi-Fi.

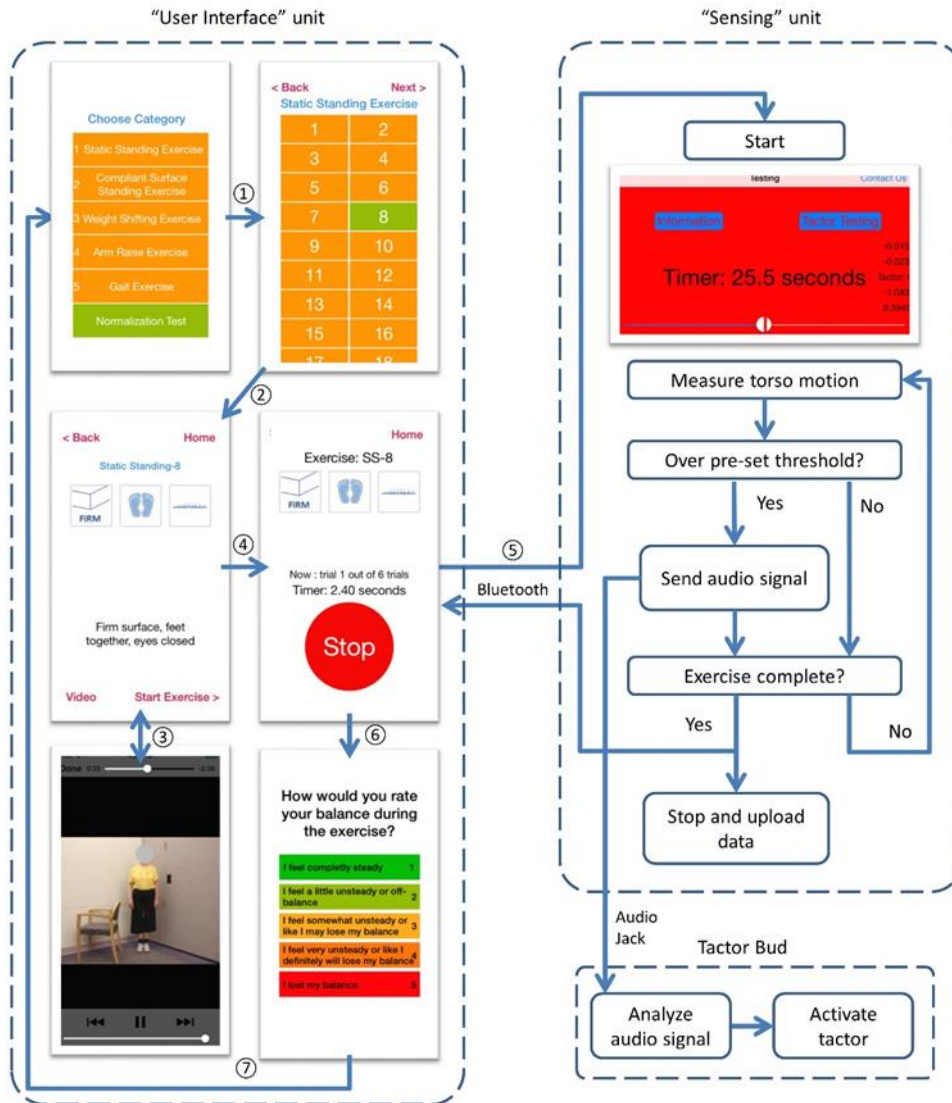


Figure 3.4: Software schematics for the user interface and sensing units.

3.2.4. Statistical Analysis

Data are presented as group mean values plus or minus (\pm) the standard deviation. Differences between the two groups at the pre-training assessment were tested using an independent samples two-tailed student's t-test. The effects of training with versus without SA on the clinical outcome measures were analyzed using a linear mixed model with group (experimental, control), time (pre-, mid-, post-training) and their interaction as fixed effects and the differences among individual participants as random effects. The measurements were

logarithmically transformed if they were not normally distributed (e.g., 5xSST duration, fast gait speed, and TUG-COG duration). To investigate the time effects within each group, two tailed paired samples t-tests within each group were performed to detect statistically significant improvement, comparing mid- and post-training assessment with pre-training assessment. The significance level was set at 0.05. Bonferroni corrections were used for the paired t-tests. Due to the relatively small sample size, the minimal detectable change (MDC) was also evaluated within groups. The MDC is defined as “a statistical estimate of the smallest amount of change that can be detected by a measure that corresponds to a noticeable change in ability” [67]. It reflects the minimal amount of change in a participant’s score that ensures the change is not the result of measurement error, but is due to rehabilitation.

3.3. Results

Both the EG and CG contained six participants. There were no significant differences in age or gender between the two groups. All participants completed the training and the three CBTs without complaints, pains, falls, or injuries, which demonstrates the feasibility of the smart phone balance trainer for in-home balance training applications [66].

The data collected by the sensing unit indicated that without supervision from the study team, the EG participants were able to successfully use the vibrotactile SA in their homes to reduce their trunk sway. As an example, Figure 3.5 shows illustrative data from Participant 6 performing an exercise (tandem Romberg stance on a firm surface with eyes open) in his home with and without vibrotactile SA provided by the smart phone balance trainer.

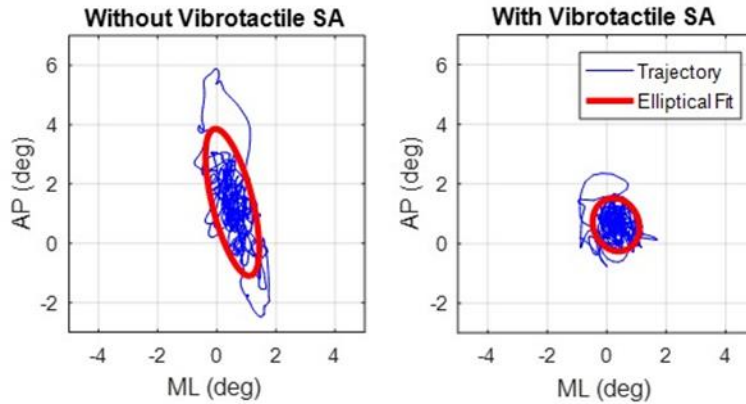


Figure 3.5: Bird's-eye view of the body tilt trajectory in the anterior-posterior (AP) and medio-lateral (ML) directions for a sample exercise (tandem Romberg stance on firm surface with eyes open) performed by participant 6 with and without vibrotactile SA.

The two tailed, independent samples t-test showed no significant differences for all clinical outcome measures between the two groups during the pre-training CBT ($p > 0.1$). Table 3.2 lists the results for a subset of the clinical outcome measures (SOT, MiniBESTest28, 5xSST) for all participants.

Table 3.2: Participants' demographic information and results of a subset of clinical outcomes measures.

Participant ID	Group	Age	Gender	SOT			Mini-BESTest28			5xSST duration (s)		
				Pre	Mid	Post	Pre	Mid	Post	Pre	Mid	Post
1	CG	83	M	71	81	81	21	22	25	11.0	12.5	14.6
2	EG	83	F	63	72	83	22	24	25	12.0	10.3	11.0
3	EG	70	F	78	81	76	20	26	27	10.1	9.7	8.7
4	CG	72	F	49	45	46	25	25	24	7.4	9.3	11.0
5	CG	70	M	76	79	73	26	27	22	9.7	8.4	11.4
6	EG	80	M	68	72	86	23	24	23	14.7	13.6	12.3
7	CG	73	M	60	68	65	24	25	25	8.8	10.0	8.3
8	EG	70	F	83	86	85	24	28	27	12.2	8.4	6.6
9	EG	82	F	74	78	79	25	25	26	17.8	14.2	15.0
10	CG	78	F	83	84	85	23	22	25	13.7	9.7	9.9
11	CG	74	F	68	77	76	24	24	24	17.7	9.7	9.5
12	EG	74	F	50	73	79	19	26	26	14.6	11.1	10.6

Table 3.3 shows the CBT assessment results for each group on average and the changes in clinical outcome measures from both pre-training to mid-training and pre-training to post-training. The linear mixed model showed significant main effects from pre-training assessments to post-training assessments in SOT composite score ($p < 0.001$), vestibular reliance ($p < 0.01$), Mini-BESTest28 ($p < 0.01$) and Mini-BESTest32 ($p < 0.01$) and TUG-COG duration ($p < 0.05$). The linear mixed model also showed significant interaction effects between groups from pre-training assessments to post-training assessments in SOT composite score ($p < 0.05$), Mini-BESTest28 ($p < 0.05$), Mini-BESTest32 ($p < 0.05$). These significant interaction effects indicate greater improvements for the EG than CG with average increases of 1.1, 0.40, and 0.58 points per week for the SOT composite scores, Mini-BESTest28, and Mini-BESTest32, respectively. There were no significant interaction effects for the other CBT outcomes. The within group paired t-test showed significant improvements for the EG in 5xSST duration during both mid- ($p < 0.01$) and post-training ($p < 0.01$). For the CG, there were no significant improvements on any of the CBT outcomes.

Table 3.3: Clinical outcome measure results for pre-, mid- and post-training CBT and the changes from pre-training CBT for the EG and CG. Average values with standard error of the mean are shown. Superscripts indicate ^asignificant main effects, ^bsignificant interaction effects from the linear mixed model ($p < 0.05$), and ^csignificant differences from the group-paired t-tests ($p < 0.017$).

	Experimental Group					Control Group				
	Pre	Mid	Post	Mid - Pre	Post - Pre	Pre	Mid	Post	Mid - Pre	Post - Pre
ABC score	90.9± 1.4	89.4± 2.6	91.3± 2.6	- 1.5±3 .3	0.5±3 .6	94.0± 1.6	91.8± 1.9	93.8± 1.4	- 2.2±1 .2	- 0.2±1 .0
SOT score ^{a, b}	69.3± 4.8	77.0± 2.3	81.3± 1.6	7.7±3 .2	12.0± 5.0	67.8± 4.9	72.3± 5.9	71.0± 5.7	4.5±2 .2	3.2±2 .2
Somatosensory reliance	0.98± 0.01	0.98± 0.01	0.98± 0.01	0.00± 0.01	0.01± 0.01	0.95± 0.01	0.96± 0.01	0.95± 0.01	0.01± 0.01	0.00± 0.01
Visual reliance ^a	0.82± 0.05	0.86± 0.03	0.89± 0.01	0.04± 0.04	0.07± 0.05	0.87± 0.02	0.87± 0.04	0.88± 0.04	0.01± 0.02	0.01± 0.03
Vestibular reliance ^a	0.53± 0.09	0.63± 0.08	0.74± 0.03	0.10± 0.03	0.21± 0.07	0.40± 0.11	0.57± 0.13	0.56± 0.13	0.17± 0.10	0.16± 0.10
Mini-BESTest28 ^{a, b}	22.2± 0.9	25.5± 0.6	25.7± 0.6	3.3±1 .2	3.5±1 .2	23.8± 0.7	24.2± 0.8	24.2± 0.5	0.3±0 .3	0.3±1 .1
Mini-BESTest32 ^{a, b}	25.0± 1.1	28.7± 0.9	29.3± 0.8	3.7±1 .4	4.3±1 .4	26.8± 0.8	27.5± 1.1	26.5± 0.6	0.7±0 .7	- 0.3±1 .3
5xSST duration (s) ^a	13.5± 1.1	11.2± 0.9	10.7± 1.2	- 2.4±0 .6 ^b	- 2.9±0 .7 ^b	11.4± 1.5	9.9±0 .6	10.8± 0.9	- 1.5±1 .6	- 0.6±1 .9
FSST duration (s)	9.9±0 .9	9.4±0 .9	9.5±0 .9	- 0.6±0 .3	- 0.6±0 .2	10.5± 0.7	10.8± 0.6	10.1± 0.9	0.3±0 .6	- 0.4±0 .5
FRT (cm)	35.0± 2.7	29.9± 2.6	31.9± 2.2	- 5.1±2 .0	- 3.0±1 .9	31.9± 2.0	28.9± 1.4	35.0± 1.1	- 3.0±1 .7	- 3.1±1 .5
Normal gait speed (m/s)	1.22± 0.03	1.28± 0.06	1.27± 0.08	0.06± 0.07	0.06± 0.08	1.29± 0.04	1.23± 0.06	1.26± 0.06	- 0.07± 0.03	- 0.04± 0.06
Fast gait speed (m/s)	1.62± 0.10	1.70± 0.11	1.69± 0.12	0.07± 0.08	0.06± 0.13	1.58± 0.05	1.57± 0.08	1.54± 0.09	- 0.01± 0.04	- 0.04± 0.07
TUG duration (s)	10.8± 0.9	9.6±0 .5	10.5± 0.6	- 1.1±0 .7	- 0.3±0 .6	10.0± 0.5	9.8±0 .7	9.8±0 .5	- 0.2±0 .4	- 0.2±0 .7
TUG-COG duration (s)	13.3± 1.2	11.6± 0.6	11.8± 0.9	- 1.7±0 .7	- 1.5±0 .6	10.9± 1.1	11.0± 0.9	9.7±0 .7	0.2±0 .7	- 1.1±1 .2

3.4. Discussion

This is the first study to investigate the effects of long-term (eight-week) balance training with and without vibrotactile SA on clinical outcome measures for community-dwelling older adults. Analysis of the twelve participants' scores showed that both the EG and CG had significant improvements in SOT composite scores, vestibular reliance, Mini-BESTest28, Mini-BESTest32 and TUG-COG duration; however, the EG improved significantly more than the CG in SOT composite scores, Mini-BESTest28, and Mini-BESTest32. In addition, significant improvements in 5xSST duration were found within the EG, whereas no significant improvements were found within the CG. However, no significant improvements were found in the ABC score, somatosensory reliance, visual reliance, FSST duration, FRT, gait speed, and TUG duration.

After training, both groups showed improvements in the SOT composite score; significantly greater improvements were found for participants trained with SA than without SA (8 points vs. 5 points at mid-training, 12 points vs. 3 points at post-training on average). Prior studies have shown that when SA was provided, real-time sway reductions were noted for exercises in the SOT protocol [68]. The results of the current study indicate that after long-term training with SA, balance improvements in SOT protocol can be retained even when SA was removed. SOT composite score improvements after training with SA (2-8 weeks) have been demonstrated in people with Parkinson's disease (~18 points) [43], people with bilateral vestibular disorders (~9 points) [45], [69], and people with other balance disorders (~8 points) [44]. Although the participants in our study reported no specific balance disorders, the EG exhibited similar improvements in SOT composite score (~12 points) after long-term training with SA. The MDC for the SOT composite score for young adults was previously determined to

be greater than 8.1 points [70]. In this study, three participants in the EG achieved MDCs in SOT composite scores, while only one participant in the CG achieved a MDC. Furthermore, these three EG participants improved by at least 15 points, while the CG participant improved by 10 points. These results indicate that training with SA may be more effective than training alone for achieving MDCs in SOT performance. From mid-training to post-training, the EG showed continuous improvements, while the CG showed a plateau effect, which suggests that training with SA could result in higher potential improvement than training without SA.

Somatosensory, visual, and vestibular reliance were calculated using SOT Conditions 1, 2, 4, and 5. Somatosensory reliance did not significantly improve following training, however, the margin for improvement was limited by high levels of somatosensory reliance prior to training. Participants in both the EG and CG relied more on visual and vestibular inputs for maintaining balance after training, although vestibular reliance showed a larger increase. These shifts in visual and vestibular reliance may support the “reweighting” hypothesis for balance training [41]. Increased vestibular reliance may be attributed to performing exercises with eyes closed and/or incorporating head movements [71]. Moreover, greater increases on vestibular reliance were observed in the EG, which may suggest that training with SA has a greater impact on reweighting vestibular inputs than training without SA.

The EG showed significantly greater improvements than the CG for both Mini-BESTest28 and Mini-BESTest32 scores. Training with SA while performing static and dynamic standing and gait exercises could explain this difference because the Mini-BESTest assesses dynamic balance [55]. Although no MDC data are available for older adults, Godi et al. reported a 3.5 point MDC for the Mini-BESTest28 among people with Parkinson’s disease (baseline Mini-BESTest28 value was 12.8 points) [72]. In our study, three participants in the EG

demonstrated a 3.5-point change during mid-training CBT versus no participants in the CG. The average baseline Mini-BESTest28 value for all participants was 22.2 points.

Within-group analysis of 5xSST performance showed significant improvement of test duration for the EG but not for the CG after training concluded. Additionally, all participants in the EG improved their 5xSST durations after training, but only three out of six participants in the CG showed improvements. Given that improvements in 5xSST duration are correlated with improved lower limb muscle strength and stability during transitional movements [56], training with SA may be more effective for improving functional mobility. It was also noted that two participants from each group reduced their 5xSST durations to less than 12 seconds (a fall risk indicator [57]) at the mid- and post-training CBTs. Finally, although the sit-to-stand task was not an exercise performed during balance training, training effects from dynamic standing tasks (especially in Category 3, Weight Shifting, and Category 4, Modified Center of Gravity) may have been transferred to the sit-to-stand task.

No significant changes in ABC scores were found in either the EG or the CG, although scores generally declined from pre-training to mid-training and increased from mid-training to post-training. As healthy older adults, all participants had relatively high ABC scores (>85) at the pre-training assessment, and were therefore unlikely to show further improvement in balance confidence [73]. Declines in scores from pre- to mid-training are consistent with an initial overconfidence in balance abilities and a shift in awareness of limitations [74]. Score increases from mid- to post-training may reflect improvements in overall balance performance, although the improvements were relatively small.

The EG and CG showed significant improvements in TUG-COG duration regardless of group. This might be due to the training with cognitive tasks. However, no significant

improvements for FRT distance, FSST duration, normal gait speed, fast gait speed, and TUG duration were noted. The lack of significant improvements may be due to ceiling effect compared to people with balance disorders at the pre-training assessment, or the difficulty level of the selected exercises may have been below the necessary level to elicit improvements for these outcomes. However, on average, the EG showed greater improvements in normal gait speed and fast gait speed than the CG. Additionally, one participant in the EG achieved a MDC for normal gait speed (0.18 m/s for Parkinson's Disease) and two participants in the EG achieved a MDC for fast gait speed (0.25 m/s for Parkinson's Disease)[75]. No participants achieved MDCs in the CG. Transfer effects from using SA during static and dynamic exercises may account for observed differences.

Our findings appear to contradict those of Lim et al., who found no significant difference in body sway between a group training with SA and a group training without SA after a two-week program [50]; however these two studies have several important differences. First, participants in Lim et al.'s study trained for two weeks (3x/week) with 6 training sessions in total, whereas in this study all participants trained for eight weeks (3x/week) with 24 training sessions in total. As shown by Lesinski et al., longer training periods result in larger improvements in balance performance [16]. Second, Lim et al.'s study provided SA to the experimental group for all exercise repetitions in all sessions (i.e., 100% feedback), whereas this study provided SA to the EG for four out of the six repetitions for each exercise (i.e., 67% feedback). Feedback can have negative effects if provided too frequently. Winstein and Schmidt found that providing feedback for the entire duration of motor skill training can improve short term performance but limit motor learning, while providing feedback for only portions of training produces poor initial performance results but improves motor skill retention [76], [77].

Therefore, training with reduced feedback frequency in this study may have improved skill retention after training concluded. Third, Lim et al. trained all participants using the same exercises throughout all sessions regardless of balance ability, whereas in this study a physical therapist selected the exercises performed by each participant based on their historical performance. Training with a constant set of exercises may limit the margin for improvement among high functioning participants, while those with poorer balance ability may experience larger improvements. In balance rehabilitation programs, experienced physical therapists progress balance exercises to achieve greater balance improvements [64]. Individualized exercise selection in this study allows participants to perform progressively challenging exercises throughout the entire training program to maximize improvement regardless of skill level. Fourth, Lim et al. quantified the effects of balance training by comparing measures of trunk motion, whereas this study used clinical outcome measures to evaluate improvements.

This study employed some elements of telerehabilitation to monitor performance and provide custom exercise regimens. Body motion, subjective ratings of balance and number of step outs were captured on the smart phone balance trainer during home-based balance training. This information was sent to a physical therapist via a wireless internet connection. Exercise programs were customized based on performance data and updated regimens were sent from the physical therapist to the participants via email. Conceptually, this aligns with telerehabilitation models, which deliver remote rehabilitation services, including assessments and intervention, via telecommunication networks [78]. However, our paradigm required less expert (i.e., physical therapist) engagement with participants as compared to traditional programs because we provided within-session vibrotactile SA as instructional balance cues. Physical therapist's time commitments were limited to less than thirty minutes per week per subject and focused on

analysis of previous balance performance and customization of the rehabilitation program, whereas typical telerehabilitation programs generally engage the expert and user remotely for the duration of the training session [79]. The findings of this study are consistent with prior work that has shown improved balance performance following telerehabilitation interventions for both people with balance deficits and community dwelling older adults [79]–[83]. Previously published research has also demonstrated potential economic benefits to using telerehabilitation approaches [79], [80]. While cost effectiveness was not explored as part of this study, the smart phone balance trainer (<\$1k) coupled with reduced patient-expert interaction could reduce the overall cost of providing rehabilitative care for a subset of people with balance deficits and simultaneously mitigate future costs stemming from injurious loss-of-balance events. Overall, improvements in clinical outcomes support the potential use of smart phone balance trainer as a telerehabilitation tool.

Our study is not without limitations. First, vibrotactile SA of trunk sway was only presented during a subset of exercises under the gait category because few studies have addressed the effectiveness of SA for improving stability during locomotor tasks [22]. Typical feedback strategies during gait activities include walking in step with auditory or visual cues or vibrotactile cues presented to a single body segment or joint to warn of extension beyond a desired angle [32], [38], [84]. Sienko et al. provided continuous vibrotactile SA based on trunk motion during overground locomotion, but slightly reduced trunk sway was observed in a subset of the trials and some participants demonstrated stiffening in the coronal plane [38]. Second, although this study employed an experienced physical therapist to instruct participants on correct exercise performance, provided handouts with instructional text and pictures, and provided exercise videos, correctness of exercise performance was not monitored during training because

training occurred in participants' homes. Third, despite the statistical and clinical significances found in this study, the sample size is relatively small. Finally, although this study used an experienced physical therapist to recommend the exercises remotely, the information provided to the physical therapist by the smart phone balance trainer was limited to the number of step-outs in the six repetitions and the stability perception ratings from the participants. A more sophisticated algorithm that captures exercise performance more comprehensively could help therapists make better recommendations in the future.

3.5. Conclusions

In-home balance training with vibrotactile SA for eight weeks improved balance performance of community-dwelling healthy older adults in this preliminary study. Participants trained with SA improved more than those trained without SA, particularly in SOT composite, Mini-BESTest, and 5xSST performance. Balance training with SA also increased visual and vestibular reliance, and improved static and dynamic balance, compared to training without SA. The lack of significant improvements in gait-related clinical outcome measures may be due to the lack of meaningful SA when performing gait exercises and the limited transfer effects from performance of standing exercises. All participants completed the eight-week training and reported no pain, injuries, or falls throughout, which suggests that healthy older adults are able to use the smart phone balance trainer safely and independently. Overall, this study supports a SA as a balance rehabilitation tool and potential telerehabilitation tool for use by community-dwelling healthy older adults.

3.6. References

- [1] B. E. Maki and W. E. McIlroy, "Postural control in the older adult.," *Clin. Geriatr. Med.*, vol. 12, no. 4, pp. 635–58, Nov. 1996.
- [2] N. Benjuya, I. Melzer, and J. Kaplanski, "Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing," *J. Gerontol. Med. Sci.*, vol. 59A, no. 2, pp. 166–171, 2004.
- [3] S. W. Shaffer and A. L. Harrison, "Aging of the somatosensory system: a translational perspective.," *Phys. Ther.*, vol. 87, no. 2, pp. 193–207, 2007.
- [4] K. Delbaere, J. C. T. Close, J. Heim, P. S. Sachdev, H. Brodaty, M. J. Slavin, N. A. Kochan, and S. R. Lord, "A multifactorial approach to understanding fall risk in older people," *J. Am. Geriatr. Soc.*, vol. 58, no. 9, pp. 1679–1685, 2010.
- [5] F. B. Horak, "Postural orientation and equilibrium: What do we need to know about neural control of balance to prevent falls?," *Age Ageing*, vol. 35, no. SUPPL.2, pp. 7–11, 2006.
- [6] M. E. Tinetti, "Preventing Falls in Elderly Persons," *N. Engl. J. Med.*, vol. 348, pp. 42–49, 2003.
- [7] A. C. Scheffer, M. J. Schuurmans, N. Van dijk, T. Van der hooft, and S. E. De rooij, "Fear of falling: Measurement strategy, prevalence, risk factors and consequences among older persons," *Age Ageing*, vol. 37, no. 1, pp. 19–24, 2008.
- [8] O. A. Donoghue, H. Cronin, G. M. Savva, C. O'Regan, and R. A. Kenny, "Effects of fear of falling and activity restriction on normal and dual task walking in community dwelling older adults," *Gait Posture*, vol. 38, no. 1, pp. 120–124, 2013.
- [9] W. J. Rejeski and S. L. Mihalko, "Physical Activity and Quality of Life in Older Adults,"

- Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 56, no. Supplement 2, pp. 23–35, Oct. 2001.
- [10] A. Shumway-Cook, W. Gruber, M. Baldwin, and S. Liao, “The effect of multidimensional exercises on balance, mobility, and fall risk in community-dwelling older adults.,” *Phys. Ther.*, vol. 77, no. 1, pp. 46–57, 1997.
- [11] M. H. Hu and M. H. Woollacott, “Multisensory Training of Standing Balance in Older Adults: I. Postural Stability and One-Leg Stance Balance,” *J. Gerontol.*, vol. 49, no. 2, pp. M52–M61, Mar. 1994.
- [12] S. J. Herdman, P. Blatt, M. C. Schubert, and R. J. Tusa, “Falls in patients with vestibular deficits.,” *Am. J. Otol.*, vol. 21, no. 6, pp. 847–851, 2000.
- [13] M. E. Tinetti and C. S. Williams, “The Effect of Falls and Fall Injuries on Functioning in Community-Dwelling Older Persons,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 53A, no. 2, pp. M112–M119, Mar. 1998.
- [14] K. Hauer, B. Rost, K. Rüttschle, O. Hedda, N. Specht, P. Bärtsch, P. Oster, and G. Schlierf, “Exercise Training for Rehabilitation and Secondary Prevention of Falls in Geriatric Patients With A History of Injurious Falls,” *J. Am. Geriatr. Soc.*, vol. 49, no. 1, pp. 10–20, 2001.
- [15] L. D. Gillespie, M. C. Robertson, W. J. Gillespie, C. Sherrington, S. Gates, L. M. Clemson, and S. E. Lamb, “Interventions for preventing falls in older people living in the community,” in *Cochrane Database of Systematic Reviews*, no. 9, L. D. Gillespie, Ed. Chichester, UK: John Wiley & Sons, Ltd, 2012.
- [16] M. Lesinski, T. Hortobágyi, T. Muehlbauer, A. Gollhofer, and U. Granacher, “Effects of Balance Training on Balance Performance in Healthy Older Adults : A Systematic Review

- and Meta-analysis,” *Sport. Med.*, vol. 45, no. 12, pp. 1721–38, Sep. 2015.
- [17] C. L. Kao, L. K. Chen, C. M. Chern, L. C. Hsu, C. C. Chen, and S. J. Hwang, “Rehabilitation outcome in home-based versus supervised exercise programs for chronically dizzy patients,” *Arch. Gerontol. Geriatr.*, vol. 51, no. 3, pp. 264–267, 2009.
- [18] A. Lacroix, R. W. Kressig, T. Muehlbauer, Y. J. Gschwind, B. Pfenninger, O. Bruegger, and U. Granacher, “Effects of a Supervised versus an Unsupervised Combined Balance and Strength Training Program on Balance and Muscle Power in Healthy Older Adults: A Randomized Controlled Trial,” *Gerontology*, vol. 62, no. 3, pp. 275–288, 2015.
- [19] P. Bach-y-Rita and S. W. Kercel, “Sensory substitution and the human–machine interface,” *Trends Cogn. Sci.*, vol. 7, no. 12, pp. 541–546, Dec. 2003.
- [20] C. Wall and M. S. Weinberg, “Balance Prostheses for Postural Control,” *IEEE Eng. Med. Biol. Mag.*, vol. 22, no. 2, pp. 84–90, 2003.
- [21] J. J. Jeka, “Light Touch Contact as a Balance Aid,” *Phys. Ther.*, vol. 77, no. 5, pp. 476–487, 1997.
- [22] K. H. Sienko, S. L. Whitney, W. J. Carender, and C. Wall III, “The role of sensory augmentation for people with vestibular deficits: real-time balance aid and/or rehabilitation device?,” *J. Vestib. Res.*, vol. 27, no. 1, pp. 63–76, 2017.
- [23] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.
- [24] C. Wall III and E. Kentala, “Effect of displacement, velocity, and combined vibrotactile tilt feedback on postural control of vestibulopathic subjects,” *J. Vestib. Res.*, vol. 20, no.

- 1, pp. 61–9, Jan. 2010.
- [25] K. H. Sienko, M. D. Balkwill, and C. Wall, “Biofeedback improves postural control recovery from multi-axis discrete perturbations.,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 53, 2012.
- [26] K. E. Bechly, W. J. Carender, J. D. Myles, and K. H. Sienko, “Determining the preferred modality for real-time biofeedback during balance training.,” *Gait Posture*, vol. 37, no. 3, pp. 391–6, Mar. 2013.
- [27] C. Wall, D. M. Wrisley, and K. D. Statler, “Vibrotactile tilt feedback improves dynamic gait index: a fall risk indicator in older adults.,” *Gait Posture*, vol. 30, no. 1, pp. 16–21, Jul. 2009.
- [28] M. Janssen, R. Stokroos, J. Aarts, R. van Lummel, and H. Kingma, “Salient and placebo vibrotactile feedback are equally effective in reducing sway in bilateral vestibular loss patients.,” *Gait Posture*, vol. 31, no. 2, pp. 213–7, Feb. 2010.
- [29] J. H. J. Allum, M. G. Carpenter, B. C. Horslen, J. R. Davis, F. Honegger, K.-S. Tang, and P. Kessler, “Improving impaired balance function: real-time versus carry-over effects of prosthetic feedback.,” *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2011, pp. 1314–8, Jan. 2011.
- [30] Z. Halická, J. Lobotková, K. Bučková, and F. Hlavačka, “Effectiveness of different visual biofeedback signals for human balance improvement,” *Gait Posture*, vol. 39, no. 1, pp. 410–414, 2014.
- [31] M. Dozza, F. B. Horak, and L. Chiari, “Auditory biofeedback substitutes for loss of sensory information in maintaining stance.,” *Exp. brain Res.*, vol. 178, no. 1, pp. 37–48, Mar. 2007.

- [32] J. Hegeman, F. Honegger, M. Kupper, and J. H. J. Allum, “The balance control of bilateral peripheral vestibular loss subjects and its improvement with auditory prosthetic feedback,” *J. Vestib. Res.*, vol. 15, no. 2, pp. 109–17, Jan. 2005.
- [33] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, “Trunk sway reductions in young and older adults using multi-modal biofeedback,” *Gait Posture*, vol. 31, no. 4, pp. 465–72, Apr. 2010.
- [34] P. Bach-y-Rita, Y. Danilov, M. Tyler, and R. J. Grimm, “Late human brain plasticity: vestibular substitution with a tongue BrainPort human-machine interface,” *Intellectica*, vol. 1, no. 40, pp. 115–122, 2005.
- [35] A. D. Goodworth, C. Wall, and R. J. Peterka, “Influence of feedback parameters on performance of a vibrotactile balance prosthesis,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 17, no. 4, pp. 397–408, 2009.
- [36] A. D. Goodworth, C. Wall, and R. J. Peterka, “A balance control model predicts how vestibular loss subjects benefit from a vibrotactile balance prosthesis,” in *Annual International Conference of the IEEE EMBS*, 2011, pp. 1306–9.
- [37] P. Bach-y-Rita, “Tactile sensory substitution studies,” *Ann. N. Y. Acad. Sci.*, vol. 1013, pp. 83–91, May 2004.
- [38] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “The effect of vibrotactile feedback on postural sway during locomotor activities,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 93, Jan. 2013.
- [39] M. Dozza, C. Wall, R. J. Peterka, L. Chiari, and F. B. Horak, “Effects of practicing tandem gait with and without vibrotactile biofeedback in subjects with unilateral vestibular loss,” *J. Vestib. Res.*, vol. 17, no. 4, pp. 195–204, Jan. 2007.

- [40] L. J. F. Janssen, L. L. Verhoeff, C. G. C. Horlings, and J. H. J. Allum, "Directional effects of biofeedback on trunk sway during gait tasks in healthy young subjects.," *Gait Posture*, vol. 29, no. 4, pp. 575–81, Jun. 2009.
- [41] F. B. Horak, "Postural compensation for vestibular loss and implications for rehabilitation.," *Restor. Neurol. Neurosci.*, vol. 28, no. 1, pp. 57–68, Jan. 2010.
- [42] Y. P. Danilov, M. E. Tyler, K. L. Skinner, and P. Bach-y-Rita, "Efficacy of electrotactile vestibular substitution in patients with bilateral vestibular and central balance loss.," in *IEEE EMBS Annual International Conference*, 2006, pp. 6605–9.
- [43] M. Rossi-Izquierdo, A. Ernst, A. Soto-Varela, S. Santos-Pérez, A. Faraldo-García, A. Sesar-Ignacio, and D. Basta, "Vibrotactile neurofeedback balance training in patients with Parkinson's disease: reducing the number of falls.," *Gait Posture*, vol. 37, no. 2, pp. 195–200, Feb. 2013.
- [44] D. Basta, M. Rossi-Izquierdo, A. Soto-Varela, M. E. Greters, R. S. Bittar, E. Steinhagen-Thiessen, R. Eckardt, T. Harada, F. Goto, K. Ogawa, and A. Ernst, "Efficacy of a vibrotactile neurofeedback training in stance and gait conditions for the treatment of balance deficits: a double-blind, placebo-controlled multicenter study.," *Otol. Neurotol.*, vol. 32, no. 9, pp. 1492–9, Dec. 2011.
- [45] C. Brugnera, R. S. M. Bittar, M. E. Greters, and D. Basta, "Effects of vibrotactile vestibular substitution on vestibular rehabilitation - preliminary study," *Braz. J. Otorhinolaryngol.*, vol. 81, no. 6, pp. 616–621, 2015.
- [46] M. G. Jorgensen, U. Laessoe, C. Hendriksen, O. B. F. Nielsen, and P. Aagaard, "Efficacy of nintendo wii training on mechanical leg muscle function and postural balance in community-dwelling older adults: A randomized controlled trial," *Journals Gerontol.* -

- Ser. A Biol. Sci. Med. Sci.*, vol. 68, no. 7, pp. 845–852, 2013.
- [47] C. Whyatt, N. A. Merriman, W. R. Young, F. N. Newell, and C. Craig, “A Wii bit of fun: A novel platform to deliver effective balance training to older adults,” *Games Health J.*, vol. 4, no. 6, pp. 1–11, 2015.
- [48] S. Roopchand-Martin, R. McLean, C. Gordon, and G. Nelson, “Balance Training with Wii Fit Plus for Community-Dwelling Persons 60 Years and Older.,” *Games Health J.*, vol. 4, no. 3, pp. 247–52, 2015.
- [49] A. S. Fu, K. L. Gao, A. K. Tung, W. W. Tsang, and M. M. Kwan, “Effectiveness of Exergaming Training in Reducing Risk and Incidence of Falls in Frail Older Adults with a History of Falls,” *Arch. Phys. Med. Rehabil.*, vol. 96, no. 12, pp. 2096–2102, 2015.
- [50] S. B. Lim, B. C. Horslen, J. R. Davis, J. H. J. Allum, and M. G. Carpenter, “Benefits of multi-session balance and gait training with multi-modal biofeedback in healthy older adults,” *Gait Posture*, vol. 47, pp. 10–17, 2016.
- [51] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, “Cell phone based balance trainer,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, Jan. 2012.
- [52] L. E. Powell and A. M. Myers, “The activities-specific balance confidence (ABC) scale,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 50, no. 1, pp. M28–M34, 1995.
- [53] Y. Lajoie and S. P. Gallagher, “Predicting falls within the elderly community: Comparison of postural sway, reaction time, the Berg balance scale and the Activities-specific Balance Confidence (ABC) scale for comparing fallers and non-fallers,” *Arch. Gerontol. Geriatr.*, vol. 38, no. 1, pp. 11–26, 2004.
- [54] R. A. Clendaniel, “Outcome measures for assessment of treatment of the dizzy and balance disorder patient,” *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 519–533, 2000.

- [55] F. Franchignoni, F. Horak, M. Godi, A. Nardone, and A. Giordano, "Using psychometric techniques to improve the balance evaluation systems test: The mini-betest," *J. Rehabil. Med.*, vol. 42, no. 4, pp. 323–331, 2010.
- [56] M. Csuka and D. J. McCarty, "Simple method for measurement of lower extremity muscle strength," *Am. J. Med.*, vol. 78, no. 1, pp. 77–81, 1985.
- [57] A. Tiedemann, H. Shimada, C. Sherrington, S. Murray, and S. Lord, "The comparative ability of eight functional mobility tests for predicting falls in community-dwelling older people," *Age Ageing*, vol. 37, no. 4, pp. 430–435, 2008.
- [58] W. Dite and V. A. Temple, "A clinical test of stepping and change of direction to identify multiple falling older adults," *Arch. Phys. Med. Rehabil.*, vol. 83, no. 11, pp. 1566–1571, 2002.
- [59] P. W. Duncan, D. K. Weiner, J. Chandler, and S. Studenski, "Functional reach: a new clinical measure of balance.," *J. Gerontol.*, vol. 45, no. 6, pp. M192–M197, 1990.
- [60] D. K. Weiner, P. W. Duncan, J. Chandler, and S. a Studenski, "Functional reach: a marker of physical frailty.," *J. Am. Geriatr. Soc.*, vol. 40, no. 3, pp. 203–207, 1992.
- [61] S. Perera, S. H. Mody, R. C. Woodman, and S. A. Studenski, "Meaningful change and responsiveness in common physical performance measures in older adults," *J. Am. Geriatr. Soc.*, vol. 54, no. 5, pp. 743–749, 2006.
- [62] D. Podsiadlo and S. Richardson, "The Timed 'Up & Go': A Test of Basic Functional Mobility for Frail Elderly Persons," *J. Am. Geriatr. Soc.*, vol. 39, no. 2, pp. 142–148, 1991.
- [63] A. Shumey-Cook, S. Brauer, and M. Woollacott, "Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test," *Phys. Ther.*, vol. 80,

- no. 9, pp. 896–903, 2000.
- [64] B. N. Klatt, W. J. Carender, C. Lin, S. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, “A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular Disorders,” *Phys. Med. Rehabil. - Int.*, vol. 2, no. 4, p. 1044, 2015.
- [65] D. Espy, C. Nathan, W. Tony, K. Natalie, and R. M., “Development of a rating scale for perceived stability during balance training,” in *International Society for Posture and Gait Research*, 2014.
- [66] W. Carender, C. Kinnaird, G. Peethambaran, G. Fantich, S. Whitney, and K. Sienko, “Balance telerehabilitation: considering user preferences for the design of a cell phone balance trainer,” in *International Society for Posture and Gait Research*, 2015.
- [67] “Rehabilitation Measures Database.” [Online]. Available: <https://www.sralab.org/statistical-terms-use>. [Accessed: 09-Oct-2017].
- [68] C. Wall III and E. Kentala, “Control of sway using vibrotactile feedback of body tilt in patients with moderate and severe postural control deficits.,” *J. Vestib. Res.*, vol. 15, no. 5–6, pp. 313–325, 2005.
- [69] T. Bao, B. Klatt, W. Carender, C. Kinnaird, S. Whitney, and K. Sienko, “Long-term balance training with vibrotactile biofeedback: Two case reports,” in *International Society for Posture and Gait Research*, 2015.
- [70] D. M. Wrisley, M. J. Stephens, S. Mosley, A. Wojnowski, J. Duffy, and R. Burkard, “Learning Effects of Repetitive Administrations of the Sensory Organization Test in Healthy Young Adults,” *Arch. Phys. Med. Rehabil.*, vol. 88, no. 8, pp. 1049–1054, 2007.
- [71] M. Cenciarini, R. J. Peterka, M. Cenciarini, and R. J. Peterka, “Stimulus-Dependent

- Changes in the Vestibular Contribution to Human Postural Control Stimulus-Dependent Changes in the Vestibular Contribution to Human Postural Control,” *J. Neurophysiol.*, vol. 95, no. 5, pp. 2733–2750, 2006.
- [72] M. Godi, F. Franchignoni, M. Caligari, A. Giordano, A. M. Turcato, and A. Nardone, “Comparison of Reliability, Validity, and Responsiveness of the Mini- BESTest and Berg Balance Scale in Patients With Balance Disorders,” *Phys. Ther.*, vol. 93, no. 2, pp. 158–167, 2013.
- [73] A. M. Myers, P. C. Fletcher, A. H. Myers, and W. Sherk, “Discriminative and Evaluative Properties of the Activities-specific Balance Confidence (ABC) Scale,” *J. Gerontol. Med. Sci.*, vol. 53, no. 4, pp. 287–294, 1998.
- [74] R. H. Fortinsky, V. Panzer, D. Wakefield, and F. Into, “Alignment between balance confidence and fall risk in later life: Has over-confidence been overlooked?,” *Health. Risk Soc.*, vol. 11, no. 4, pp. 341–352, 2009.
- [75] T. Steffen and M. Seney, “Form Health Survey , and the Unified Parkinson Disease Rating Scale in People With Parkinsonism,” *Phys. Ther.*, vol. 88, no. 6, pp. 733–746, 2008.
- [76] C. J. Winstein and R. a. Schmidt, “Reduced frequency of knowledge of results enhances motor skill learning.,” *J. Exp. Psychol. Learn. Mem. Cogn.*, vol. 16, no. 4, pp. 677–691, 1990.
- [77] R. A. Schmidt, “Frequent augmented feedback can degrade learning: Evidence and interpretations,” in *Tutorials in motor neuroscience*, Dordrecht, The Netherlands Springer, 1991, pp. 59–75.
- [78] R. B. Burns, D. Crislip, P. Daviou, A. Temkin, S. Vesmarovich, J. Anshutz, C. Furbish,

- and M. L. Jones, "Using telerehabilitation to support assistive technology," *Assist. Technol.*, vol. 10, no. 2, pp. 126–133, 1998.
- [79] D. Kairy, P. Lehoux, C. Vincent, and M. Visintin, "A systematic review of clinical outcomes, clinical process, healthcare utilization and costs associated with telerehabilitation," *Disabil. Rehabil.*, vol. 31, no. 6, pp. 427–447, 2009.
- [80] M. Tousignant, P. Boissy, H. Corriveau, and H. Moffet, "In home telerehabilitation for older adults after discharge from an acute hospital or rehabilitation unit: A proof-of-concept study and costs estimation.," *Disabil. Rehabil. Assist. Technol.*, vol. 1, no. 4, pp. 209–216, 2006.
- [81] J. C. K. Lai, J. Woo, E. Hui, and W. M. Chan, "Community-Based Stroke Rehabilitation," pp. 199–205, 2004.
- [82] R. Llorens, E. Noe, C. Colomer, and M. Alcaniz, "Effectiveness, usability, and cost-benefit of a virtual reality-based telerehabilitation program for balance recovery after stroke: A randomized controlled trial," *Arch. Phys. Med. Rehabil.*, vol. 96, no. 3, p. 418–425.e2, 2015.
- [83] I. Cikajlo, M. Rudolf, N. Goljar, H. Burger, and Z. Matjačić, "Telerehabilitation using virtual reality task can improve balance in patients with stroke.," *Disabil Rehabil*, vol. 34, no. 1, pp. 13–18, 2012.
- [84] P. B. Shull, K. L. Lurie, M. R. Cutkosky, and T. F. Besier, "Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback.," *J. Biomech.*, vol. 44, no. 8, pp. 1605–9, May 2011.

Chapter 4 Vibrotactile Display Design: Quantifying the Importance of Age and Various Factors on Reaction Times

4.1. Introduction

Vibrotactile displays, one common type of haptic display [1], have been applied to various areas of the body to deliver spatial and temporal information in a variety of real-time applications [2], [3]. During driving, vibrotactile displays provide navigation information [4] and warn drivers of potential collisions [5], [6]. During flying, they can provide altitude information, warning signals and simple communication to replace or reinforce visual and auditory cues [7], [8]. Vibrotactile displays have also been applied to virtual reality applications to enhance physical sensations within a virtual environment [9], [10]. More recently, vibrotactile displays have been used for training arm movements, balance, and gait [11]–[19].

Delays caused by human reaction times (RTs) to vibrotactile cues should be considered when designing vibrotactile displays. Depending on the application, RTs to vibrotactile stimuli have been shown to vary from 200 ms to 2000 ms [4], [5], [19]–[22]. For example, surgical applications often require small, carefully controlled movements, therefore a fast RT (less than 250 ms) is desirable [20]. Likewise, fast RTs are also desirable for golf swing training (e.g., initiation of downswing through impact typically takes less than 250 ms [21]) and step response (e.g., step response to unpredictable, abrupt backward translation of a support surface typically

takes less than 300 ms [19]) applications. On the other hand, longer RTs are acceptable for use in driving and flying applications as warning signals (e.g., ~900 ms [5], [22] and ~1600 ms [4], respectively). Lastly, RTs associated with real-time gait training applications vary considerably based on the specific activity; step/stride frequencies for walking and running usually range from 1 Hz to 3 Hz requiring RTs ranging from 300 to 1000 ms.

Numerous individual factors have been shown to affect RTs. Harrar and Harris investigated RTs to vibrotactile stimuli across various body locations (forehead, lip, neck, hand, and foot) and showed that RTs increased proportionally with the distance from the brain with a slope of 45 ms/m [23]. Shull et al. found that RTs for arm movements were ~60 ms faster than leg movements by placing tactors (i.e., vibrotactile transducers) on the arm and leg, respectively [24]. Asseman et al. investigated the effects of location and stimulus-response compatibility on RTs while participants were exposed to an unpredictable translation of the support surface. Their results indicated that 1) RTs to vibrotactile stimuli displayed on the head were ~60 ms faster than those displayed on the sternum during a protective stepping task and 2) RTs to vibrotactile stimuli displayed on the forehead were ~100 ms faster than those displayed on the back of the head for the same task [19]. Peon and Prattichizzo demonstrated that a higher vibrotactile stimuli intensity produced faster (i.e., ~25 ms) RTs compared with a low stimuli intensity [20]. Hick showed that RTs logarithmically increased as the number of different visual stimuli increased in choice experiments [25]. Haggerty et al. showed that verbal and push-button reaction times increased by up to 200 ms when the participants simultaneously were asked to respond to torso-based vibrotactile cues to make postural corrections [26]. Although limited studies have investigated the effects of aging on RTs to vibrotactile stimuli, Taware et al. demonstrated significant increases in RT (i.e., on the order of 60 ms) to both auditory and visual stimuli for

older adults (50-59 yrs) compared to young adults (20-29 yrs) [27]. Ng and Chan reported that multiple factors including age, gender, and level of education had significant effects on RTs for auditory and visual stimuli; in general, aging increased RTs, while females had faster RTs than males, and participants that had completed either a tertiary or secondary education levels had faster RTs than participants that had only completed the primary education level [28].

Additionally, they studied RTs to vibrotactile stimuli applied at the wrist and leg locations; however, no significant differences were observed based on stimuli location [28].

Additional factors including frequency of stimulation and tactor type have been shown to affect human responses to vibrotactile stimuli. Humans are most sensitive to vibrotactile stimuli applied with a frequency of 250 Hz [20], [29]. Vibrotactile displays typically use one of two types of tactors: linear actuators and rotary motor [2], [30]. Lee et al. showed that a C-2 tactor induced larger non-volitional postural shifts in the directions of the applied stimuli compared to a Tactaid tactor [29].

The majority of RT studies have focused on visual and auditory stimuli [31]. Among the studies that have investigated RTs to vibrotactile stimuli, the most commonly performed studies have compared RTs to vibrotactile stimuli with RTs to visual and/or auditory stimuli [4], [23], [28], [31], [32], or have compared RTs to vibrotactile stimuli while varying one-to-two stimuli factors (e.g., location, secondary task) using the same subject population and/or experimental condition [5], [19], [20], [23], [24], [26], [31]. Therefore, there are limited quantitative data that facilitate the determination of the relative importance of vibrotactile stimuli factors on RTs. As previously mentioned, RTs are important to consider when designing vibrotactile displays for various applications; specific applications may necessitate careful selection of tactor locations, number of tactors, and type of tactors. To inform vibrotactile display design for real-time

applications, this study quantitatively examined RTs to vibrotactile stimuli as a function of various factors in young adults and older adults using simple reaction time tasks. The factors investigated in this study included stimulus frequency, type of factor that generates the vibrotactile stimulus, stimulus location, auditory cues generated by the vibration (ACV), number of factors in the same location, number of possible stimulus locations, and secondary task.

4.2. Methods

This study was performed in four parts; the first three experimental parts were performed within a single session on one day and the fourth part was performed during a separate session on a subsequent day. Throughout the experiments, vibrotactile stimuli were presented using different factor types at different locations on the body and factors were secured by Velcro straps. A shim was used to achieve approximately uniform pressure across all body location sites. Participants were asked to confirm that the pressure of the factor against their skin felt similar at all locations.

During the experiments, participants were barefoot. They were instructed to stand with their arms held at their sides with their feet hip-width apart at a 15° lateral rotation angle [29] while looking straight ahead at a stationary, eye-level visual target. Participants were asked to respond as quickly as possible each time they perceived a vibrotactile stimulus by pressing a thumb trigger with their dominant hand. Trigger responses were recorded by a computer and RT was defined as the time when the command to the factor was applied to the time the trigger signal was received by the computer. The delay caused by the signal transmission was not measured given that the objective of the study was to elucidate relative differences in RTs to different vibrotactile display design factors. Participants were not informed of their performance

during the experiments. Before the start of the experiments, participants were asked to familiarize themselves with the stimuli-trigger system by practicing responses prior to each experimental part for approximately 3-4 minutes. All participants were asked to confirm they could feel the vibration at all locations prior to each experimental part.

Ten healthy young adults (YA) (21.9 ± 1.3 yrs; six males, four females) and ten healthy community-dwelling older adults (OA) (68.3 ± 2.7 yrs; four males, six females) participated in the first three parts of this study. All participants completed the three parts in the same order. Nine healthy OA (70.5 ± 5.6 yrs; four males, five females), including six participants who also participated in the first three parts of the study, participated in the fourth part of this study. Individuals were eligible to participate in this study if they reported no neurological conditions or joint replacements, and if they could stand for at least 1 min without assistance. Prior to participation, all participants completed the 10-g monofilament test [33] on the dorsal aspect of the dominant foot and were excluded if they failed the test. Both YA and OA followed the same experimental protocol. The University of Michigan IRB approved the experimental protocol and informed consent was obtained from each participant prior to the start of the experiment. All participants were compensated \$15/hr for their participation in the study.

4.2.1. Part I: Effects of Frequency and Tactor Type

The first part was designed to test the effects of vibration frequency and tactor types on RTs. Four types of vibrotactile stimuli were presented using three different tactors as shown in Table 4.1. The C-2 tactor (EAI, Inc.) operated at either 250 Hz or 200 Hz with a 6.2 cm^2 contact area and 0.5 cm^2 contactor area. Only the contactor vibrated to provide vibrotactile stimuli. The Tactaid VBW32 tactor (Audiological Engineering Corp) operated at 250 Hz with a 3.7 cm^2

contact area. Based on experimental measurements, the coin-style motors used in this study (Precision Microdrives, 310-101 vibration motor) operated at a frequency of 215 ± 5 Hz with an input of 3.7 V. They were encapsulated in a plastic case with a contact area of 6.2 cm^2 , and have previously been used in a smartphone balance trainer device [34]. Both the C-2 and Tactaid tactors were driven by a sinusoidal signal generated at a RMS current of 0.225 A (or peak-to-peak voltage of 4.47 V) by the tactor controller unit provided by EAI, Inc.. The coin-style motors were driven by a battery at 3.7 V. The battery was fully charged prior to each data collection session. The amplitude of the signals was adapted from our prior application-based studies [29].

Table 4.1: The major differences among three different tactors. ⁺: C-2 tactor vibrated at both 250 and 200 Hz; ⁺⁺: The contactor area is 0.5 cm^2 , while the contact area is 6.2 cm^2 .)

Type	Transducer display	Frequency (Hz)	Area (m^2)
C-2 tactor	Linear actuator	250/200 ⁺	0.5 (6.2 ⁺⁺)
Tactaid tactor	Linear actuator	250	3.7
Coin-style motor	Rotary motor	200	6.2

Vibrotactile stimuli were presented to four torso locations aligned with navel, spine, and the left and right sides of the torso at the level of the L4/L5 spinal segment. These specific four locations were selected for inclusion in this study because they are the most commonly used torso locations for balance-related applications [13], [14], [16]. Together, these three tactors and four locations formed four configurations as shown in Figure 4.1. The presentation order of configurations was randomized for each participant. For each configuration, each type of vibrotactile stimuli was presented eight times and each vibration lasted 500 ms [35]. The presentation order of these 32 stimuli was randomized and then grouped into four trials with eight stimuli per trial. A randomized interval of 4-7 s was inserted between stimuli. Participants

were asked to wear foam earplugs and earmuffs to eliminate the auditory cues generated by the vibration (ACV).

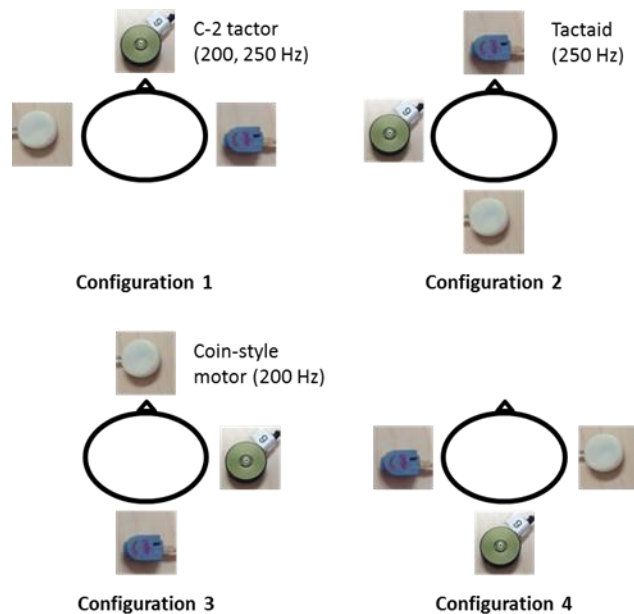


Figure 4.1: Experimental illustration for Part I: Bird's eye view (triangle represents the participant's nose) of the Part I tactor configurations; three tactors were placed on the torso (possible locations included the navel, spine, left and right side).

4.2.2. Part II: Effects of Stimuli Locations and Auditory Cues from Vibration

The second part tested the effects of tactor locations and the ACV on RTs. Vibrotactile stimuli were presented using C-2 tactors operating at 250 Hz at the following five locations on the body: center of the forehead (head), index fingertip on the dominant hand (finger), torso vertically aligned with the navel at L4/L5 spinal segment level (torso), anterior mid-shank on the dominant leg (shank), and dorsal base of metatarsus III on the dominant foot (foot) as shown in Figure 4.2. These five locations were chosen to replicate the locations that are most commonly for vibrotactile stimulation [7], [13], [15], [16], [18]–[20]. This part was divided into two identical subparts except that participants wore foam earplugs and earmuffs in the first subpart, but did not wear earplugs and earmuffs in the second subpart. For both subparts, each tactor was randomly activated for 500 ms eight times, with a randomized interval of 4-7 s between stimuli.

The presentation order of these 40 stimuli was randomized and then grouped into four trials with ten vibrations per trial.

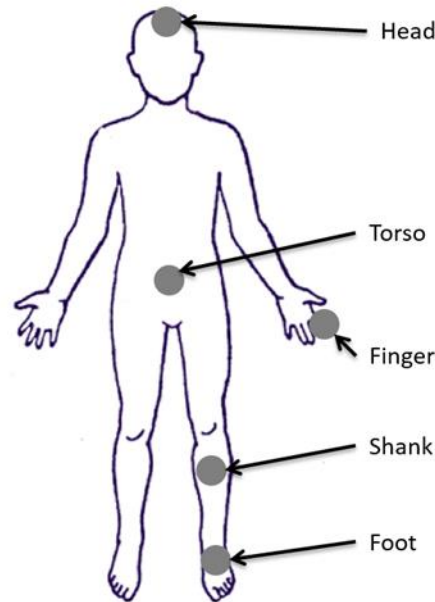


Figure 4.2: Experimental illustration for Part II: Five C-2 tactors were placed at five body locations.

4.2.3. Part III: Effects of Number of Tactors in the Same Location

The third part investigated the effects of the number of tactors in the same location on RTs. In this study, these four C-2 tactors (operating at 250 Hz) were grouped in a 2x2 clustered array at the torso aligned with the navel at the L4/L5 spinal segment to form four vibration patterns as shown in Figure 4.3. The distance between the nearest edges of two adjacent tactors was 1 cm. In pattern 1, only one tactor in the upper left corner of the cluster was activated. In patterns 2 and 3, two tactors in the horizontal direction and vertical direction of the cluster were simultaneously activated, respectively. In pattern 4, all tactors within the cluster were activated simultaneously. Each pattern was presented randomly for 500 ms five times and there was a randomized interval of 4-7 seconds between stimuli. The order of these 20 stimuli was

randomized and they were grouped into two trials with 10 vibrations per trial. Participants were asked to wear foam earplugs and earmuffs.

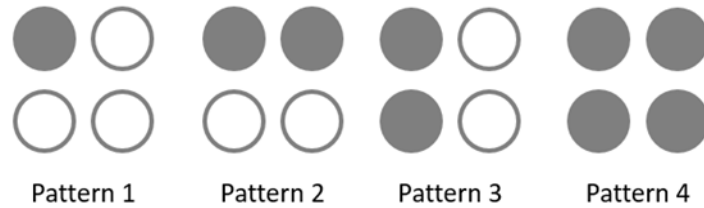


Figure 4.3: Experimental illustration for Part III: Four C-2 factors grouped in a 2x2 array formed four patterns; the solid dots indicate the factors that were activated in the corresponding patterns.

4.2.4. Part IV: Effects of Secondary Task

The fourth part explored the effects of simultaneously performing a secondary task on the RTs to vibrotactile stimuli. Tactor placements were similar to the placement during Part II of this study (i.e., five factors were placed at the five locations shown in Figure 4.2), but participants did not wear earplugs or earmuffs. During this part of the study, participants were also asked to perform a simultaneous secondary cognitive task, which was used to divide the attention and increase the cognitive load [36]. Specifically, participants were asked to continuously count backward by three from an initial number provided by a study team member (randomly assigned, ranging from 90 to 99) [37]. Verbal responses were audio-recorded. If participants counted down to 0 within any given trial period, they were asked to begin counting backwards from the initial number provided to them at the start of the trial. Cognitive task performance was first assessed without the RT task to establish a baseline. The length of the cognitive task was chosen to be 60 s to approximately match the length of each trial in Part II. Participants completed four trials per location.

4.2.5. Data Analysis

RTs faster than 100 ms were considered as errors and discarded [20], [23], [38]–[40]. The maximum RT was a function of the time interval between stimuli (e.g., 4-7 s) and was considered as lack of response. The lack of a response prior to the subsequent stimulation was considered a missed response and was not included in the analysis. All data were logarithmically transformed to achieve a normal distribution. The repeated measures of RTs for same condition within same participant were averaged to get a more reliable RTs.

T-tests were used to compare the differences in average RTs per condition between YA and OA. To analyze the effects of different vibration factors on RTs, a linear mixed effect model (LMM) was used. The benefits of using a LMM are that it accounts for the random effects of subjects into the model [41].

The LMM was built for each experimental part separately. Fixed factors included the type of vibrotactile stimuli (Part I), tactor location and presence of ACV (Part II); stimulation pattern (Part III), and location of tactors and presence of secondary task (Part IV). When only one factor was included in the model, it was considered a fixed effect factor. If two factors were included, the combinations of these factors were considered as fixed effect factors. The differences among participants were always considered as random effects for all analyses. All LMM analyses were performed as stratified analyses. The analysis was done in R (r-project.org). The significance was defined by p-values less than 0.05.

The performance of the secondary cognitive task was evaluated by calculating the number of verbal responses per minute and the number of mistakes made per minute. The correctness of counting was not evaluated in this study. The significance was analyzed using the LMM by replacing ‘Reaction Times’ with ‘Number of Verbal Responses’.

4.3. Results

Only one data point was faster than 100 ms; the participant mentioned that he might have accidentally pushed the thumb trigger during that trial. The slowest RT was found during the dual-task condition and was less than 2 s. All other RTs were less than 1 s for all parts of the study. For YA, 0.3%, 0% and 0% of RTs were considered as missing data points for Part I, Part II, and Part III, respectively. For OA, the missing data points increased to 2.3%, 0.5%, and 0.5% for Part I, Part II, and Part III, respectively. Also, most of the missing RTs occurred when the participants were presented with stimuli at the spinal location in Part I. There were several possible reasons for missing points including: the participants were not able to sense the vibration, the participants did not push the trigger hard enough to trigger a response, and/or the stimulation system did not work properly. Only OA participated in Part IV. The percentage of missing data points increased to 8.9% for Part IV, where the stimulation at the foot location contributed to more than half (56%) of the total missing data points. An increase in missing data points was expected due to the dual-task paradigm. Among all OA participants, two out of the nine participants were more likely to miss stimuli than the other participants.

The results of tested factors were reported for each experimental part. Across all experimental parts, RTs for YA were significantly faster than OA to the same type of vibrotactile stimuli (31-122 ms, $p < 0.01$). It was also noted that that the variance of RTS within the OA population was significantly higher than within the YA population for the same vibrotactile stimuli ($p < 0.0001$).

4.3.1. Effects of Frequency and Types of Tactors (Part I)

The mean RTs to the four types of vibrotactile stimuli are shown in Figure 4.4. The difference between YA and OA were 66 ms, 68 ms, 82 ms and 62 ms for C-2 tactor at 250Hz, C-2 tactor at 200Hz, Tactaid, and coin-style motor respectively.

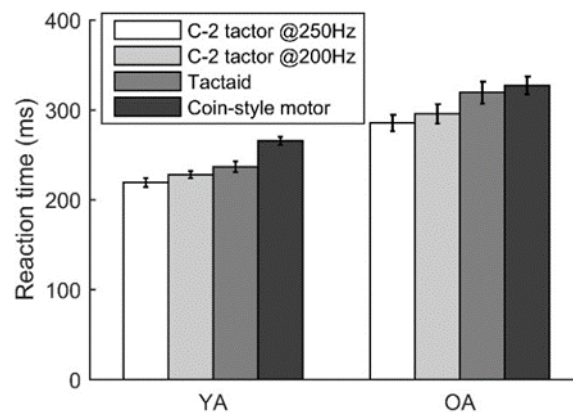


Figure 4.4: RTs to four different types of vibrotactile stimuli at torso locations; error bars present standard errors of means across participants.

For both YA and OA, RTs to the C-2 tactor at 250 Hz were fastest (9-46 ms, $p < 0.05$ and 10-42 ms, $p < 0.05$, respectively) and RTs to the C-2 tactor at 200 Hz were significantly faster than RTs to the Tactaid (9 ms, $p < 0.05$ and 24 ms, $p < 0.05$, respectively) and coin-style factors (37 ms, $p < 0.05$ and 32 ms, $p < 0.05$, respectively). For YA, RTs to the Tactaid tactor were significantly faster than those to the coin-style tactor (29 ms, $p < 0.01$); but for OA, there were no significant differences between the Tactaid and coin-style factors.

4.3.2. Effects of Tactor Locations and ACV (Part II)

The mean RTs to the vibrotactile stimuli at the five body locations are shown in Figure 4.5. Among YA, regardless of ACV, RTs were significantly faster at the head location than the other four locations (19-43 ms, $p < 0.05$). Without ACV, there were no significant differences

among these four locations. With ACV, RTs were significantly faster at the finger location than the torso location (12 ms, $p < 0.05$) and RTs were slowest at the foot location (11-42 ms $p < 0.05$). Compared with RTs without ACV, RTs with ACV were 1) significantly faster at the finger location (8 ms, $p < 0.05$); 2) not significantly different for the other four locations.

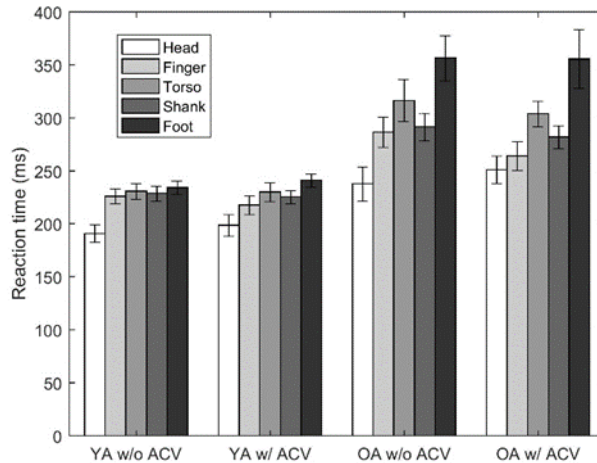


Figure 4.5: RTs to vibrotactile stimuli at five different body locations for both YA and OA and with or without ACV; error bars present standard errors of the means across participants.

Among OA, without ACV, RTs at the head location were fastest (49-119 ms, $p < 0.05$) and RTs at the foot were slowest (40-119 ms, $p < 0.05$). RTs at the torso were significantly slower than RTs at the finger (30 ms, $p < 0.05$) and shank (25 ms, $p < 0.05$). With ACV, RTs were significantly different among all five locations (20-99 ms, $p < 0.05$) except the comparison between the head and finger locations (14ms, $p = 0.052$). Compared with RTs without ACV, RTs with ACV were 1) significantly slower at the head location (13 ms, $p < 0.05$); 2) significantly faster than at the finger location (22 ms, $p < 0.05$); 3) not significantly different at other three locations.

4.3.3. Effects of Number of Tactors in the Same Location (Part III)

RTs for the pattern-based vibrotactile stimuli presented at the navel location are shown in Figure 4.6. For YA, RTs to pattern 4 were significantly faster than the other patterns (20 ms, $p < 0.05$), but there were no significant differences among the other three patterns. For OA, RTs to pattern 4 and pattern 2 were significantly faster than RTs to patterns 1 (31 ms, $p < 0.01$ and 20 ms, $p < 0.05$, respectively).

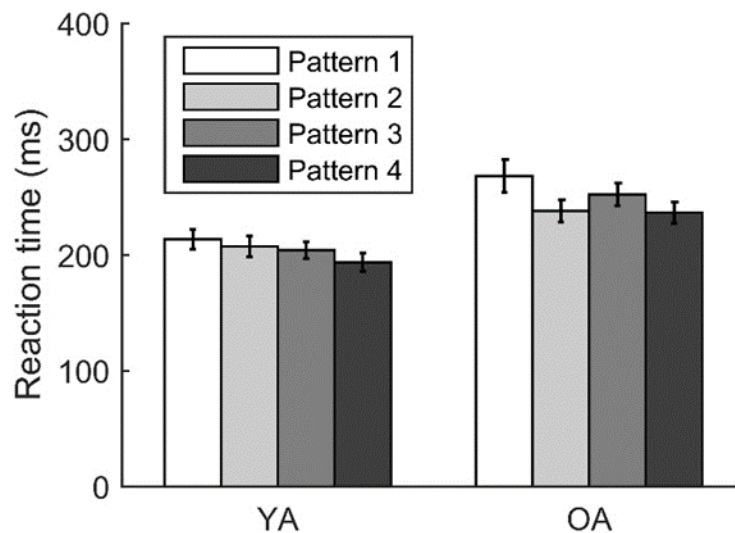


Figure 4.6: RTs to the four stimulation patterns for both YA and OA; error bars present standard errors of means across participants.

4.3.4. Effects of the Secondary Task (Part IV)

The effects of the secondary task on RTs to vibrotactile stimuli and the performance of the secondary task were examined.

RTs for the nine participants who participated in Part IV are shown in Figure 4.7. With the secondary task, the RTs at the finger were significantly faster than the RTs at the other four locations (87-182 ms, $p < 0.01$). The RTs at the head were significantly faster than at the torso (78 ms, $p < 0.01$) and foot (95 ms, $p < 0.01$).

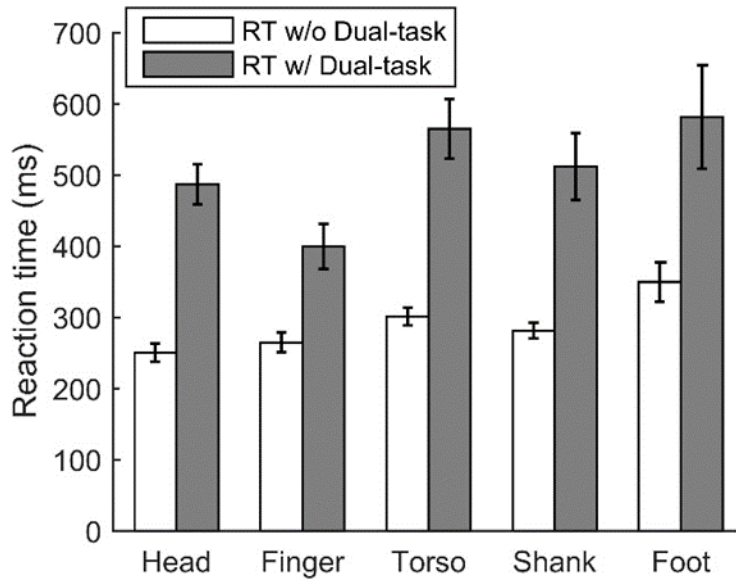


Figure 4.7: RTs to the vibrotactile stimuli at five different body locations with or without the secondary task; error bars present standard errors of means across participants.

RTs with and without the secondary task were compared for each location for the six participants who completed the experiments on both days. At all locations, RTs to vibrotactile stimuli significantly increased when the secondary task was performed (135-264 ms, $p < 0.01$).

All nine participants had slower counting speeds when performing two tasks simultaneously; on average, the number of verbal responses per minute significantly decreased from 39.7 ± 2.5 to 34.1 ± 1.8 ($p < 0.01$).

4.3.5. Effects of the Number of Tactor Locations

By design, there was one common C-2 tactor placed at the navel location for Parts I-III (Part I: three tactors at four locations around the torso; Part II: five tactors at five locations across the body; Part III: four tactors in a cluster only at the navel location) to facilitate comparisons of RTs to the same location with different number of potential tactor locations. The results are shown in Figure 4.8.

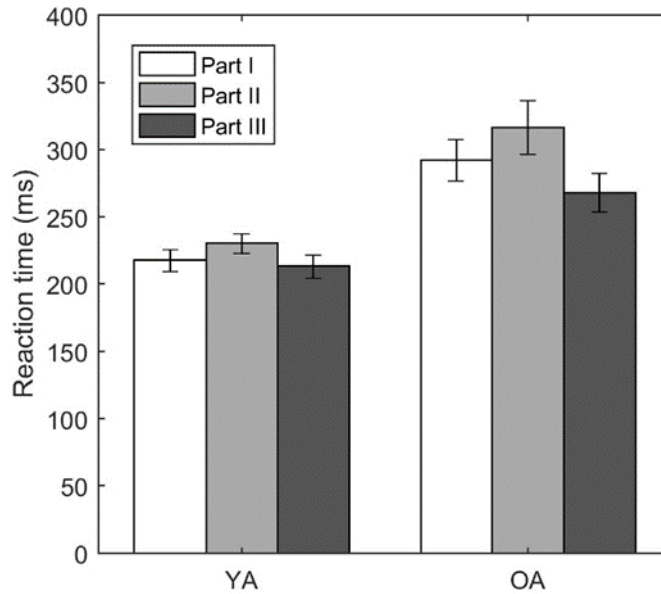


Figure 4.8: RTs for the common factor located at the navel for three experimental parts; Part I: three factors at four locations around the torso; Part II: five factors at five locations across the body; Part III: four factors in a cluster at the navel. Error bars present standard errors of means across participants.

Among both YA and OA, for the same factor type (single C-2 factor with 250 Hz) and location (navel location), RTs in Part II were significantly slower than RTs in Parts I (13 ms, $p < 0.05$ and 24 ms, $p < 0.05$) and Part III (17 ms, $p < 0.05$ and 48 ms, $p < 0.05$). For OA, the RTs in Part I were significantly slower than those in Part III (24 ms, $p < 0.01$).

4.3.6. Summary of Results

Overall, vibrotactile stimuli factors affected RTs by less than 10 ms to more than 250 ms. Table 4.2 summarizes the various factors' effects (only includes statistically significant differences for comparisons among various factors) based on RTs for both YA and OA.

Table 4.2: Summary of various factors' effects on RTs for YA and OA (* secondary cognitive task was only tested on older adults)

Factors	Comparisons	Absolute Difference in RTs	
		Young adults	Older adults
Stimulus frequency	250 vs 200 Hz	9 ms	10 ms
Auditory cues generated by vibration (with ACV vs without ACV)	@ head location	8 ms	22 ms
	@ finger location	8 ms	13 ms
Number of tactors in the same location	One vs four	20 ms	31 ms
Number of tactor locations	One vs five	17 ms	48 ms
Type of tactors	C-2 vs Tactaid	18 ms	34 ms
	C-2 vs Coin-style motor	46 ms	42 ms
Stimulus location	Head vs finger	19 ms	14 ms
	Head vs torso	31 ms	51 ms
	Head vs shank	27 ms	31 ms
	Head vs foot	42 ms	99 ms
Stimulus location with secondary task	Finger vs head	N/A*	87 ms
	Finger vs torso		165 ms
	Finger vs shank		112 ms
	Finger vs foot		182 ms
Secondary cognitive task (with task vs without task)	@ finger location	N/A*	135 ms
	@ head, shank, foot locations		233 ms
	@ torso location		264 ms

4.4. Discussion

In this study, the effects of various factors, including vibration frequency, type of tactors, tactor location, ACV, number of tactors in the same location and secondary task on RTs to vibrotactile stimuli were investigated for both healthy YA and OA. The difference of RTs between YA and OA was also studied.

4.4.1. Young and Older Adults

On average, RTs for OA were ~60 ms slower than the RTs for YA, which is in agreement with prior RT findings to audio-visual stimuli as a function of age [27]. This observed difference

may result from age-related reduced neuron conduction rates [27], [42], decreased cognitive and motor functions [43], [44], and/or deteriorated sensory function [45]. However, in our study, RT differences between YA and OA were as large as 100 ms depending on the experimental condition. In general, differences in RTs between YA and OA increased when YA had large RTs. For example, YA reacted in ~200 ms to stimuli provided at the head location, while OA reacted in ~250 ms to stimuli at the same location (~50 ms difference); YA reacted in ~230 ms to stimuli at the foot location, while OA reacted in ~330 ms to stimuli at the same location (~100 ms difference).

4.4.2. Stimulus Frequency and Number of Tactors in the Same Location

RTs to changes in stimulus frequency and the number of tactors in the same location were predictable; the fastest RTs were observed when stimuli were operated at 250 Hz compared with 200 Hz and RTs decreased as the number of tactors in the same location increased. Pacinian Corpuscles have an optimal sensitivity at approximately 250 Hz [2], [46]–[48], and sensitivity has been shown to increase as contact area increases for relatively small areas ($< 5 \text{ cm}^2$) [47]. Overall, these two factors have a relatively lower impact on RTs (i.e., 10-30 ms) compared with other factors investigated in this study.

4.4.3. Type of Tactors

RTs to the C-2 tactor were ~50 ms faster than the other two tactors. However, the C-2 tactor is substantially more expensive (i.e., \$250) than a coin-style tactor (i.e., \$5). Thus, there is a trade-off between RT and cost.

4.4.4. Stimulus Location

Harrar and Harris [23] found that RTs to vibrotactile stimuli were linearly related to the distance from the forehead by 45 ms/m for YA. In this study, it was found that as the distance from the head to the tactor location increased, RTs generally increased except for RTs to stimuli applied at the torso. For both YA and OA, although the distance from the forehead to the torso was shorter than the distance from the forehead to the finger and the shank, RTs at the torso were slower than RTs at both the finger and shank. This counter-relationship of distance is likely due to greater skin sensitivity to vibration at the finger and shank than the torso [49]. In the study by Harrar and Harris [23], the tactors were placed on the forehead, lip, neck, finger and foot locations, which are generally more sensitive than the torso [49]. Thus, the RTs could be linearly related to the distance from the forehead for stimulus locations with comparable sensitivities.

By comparing RTs at the head, finger, shank and foot, a linear relationship with slopes of 26 ms/m for YA and 67 ms/m for OA was found, which is faster than the conduction velocities from Harrar and Harris [23]. However, converting our findings for conduction velocity from ms/m to m/s yielded 38 m/s for YA and 15 m/s for OA, which is closer to the conduction velocities measured using physiological techniques [50]. Finally, the slope (ms/m) for the OA was more than twice as steep as the slope for the YA. This implies that the effects of body location on RTs increases with age and that stimulus locations far from the forehead should be taken into careful consideration for certain applications, especially for OA.

4.4.5. Auditory Cue from Vibration

The effects of ACV on RTs were different for different locations. RTs at the finger were significantly faster for both populations when ACV were provided, which is consistent with

previous findings by Diederich and Colonius [51] who found that RTs to combinations of auditory and tactile stimuli were faster than RTs to each type of stimulus alone. However, the presence of ACV did not influence RTs at the torso, shank, or foot for either population. This may be caused by the contact medium (e.g., abdominal skin) and the distance between the sound source and ear. Interestingly, at the head location, wearing earplugs and ear muffs did not affect the RTs for YA and even slowed down the RTs for OA. One potential reason for these findings is the occlusion effect, which assumes that bone conduction increases sensitivity to low frequency sounds when the ear canal is blocked [52], [53]. Thus, when participants were wearing earplugs and ear muffs, the ACV was possibly magnified rather than blocked. Environmental noise might have interfered with the ACV when participants removed the earplugs and ear muffs.

4.4.6. Secondary Task

When a secondary task was performed, RTs to vibrotactile stimuli significantly increased by at least 130 ms compared to no secondary task. Slower RTs were coupled with poorer performance of the cognitive task. These findings agree with previous studies [17], [54]. Mohebbi et al. [54] showed large increases in RTs (148 ms) to vibrotactile collision warnings when participants talked on a cell phone. Lin et al. also found a significant increase in RTs to a cognitive task when the participants were asked to perform balance exercises with vibrotactile cues [17].

During the presence of the secondary task, a linear relationship was not observed between RTs and the distance from the forehead; while performing the secondary task, RTs at the finger location were the fastest, followed by RTs at the head location (there were no significant

differences among RTs at the torso, shank, and foot locations). The changes at the head and finger locations may be explained by the stimulus-response compatibility effects, which refer to how compatible the locations of stimuli are with the location of response [55]–[57]. According to Kornblum et al., the high compatibility sets should yield faster responses than the low compatibility sets [55]. In this study, the sets involved stimuli at the head and finger location, and responses at the thumb location with a trigger-push. The finger-thumb response set was more compatible than head-thumb stimuli-response sets which resulted in faster RTs at the finger. These results agree with those by Ho et al. where they found a set-level compatibility between stimulus locations of wrist and foot, and response locations of hand and foot [58]. However, this compatibility effect was not observed during the single task experiments (experimental Part II). One possible explanation is that the compatibility effect does not outweigh the effect of distance from the forehead, which was also found in the study by Ho et al. With the increased cognitive load by the addition of secondary task, the compatibility effect may have been magnified.

The presence of the secondary cognitive task also negatively affected the number of trials completed with appropriate RTs. Two prevalent attention division theories are the capacity-sharing theory (attentional capacity is limited and performing two tasks will cause deterioration of one or both tasks) and the bottleneck theory (one task will be delayed until the other task is completed) [36], [59], [60]. Aligned with the bottleneck theory, participants may have been processing the secondary cognitive task when the 50-ms vibration occurred and were not able to perceive and react to the vibration.

4.4.7. Number of Possible Stimulus Locations

RTs to a common location (navel) increased when there were more potential stimulus locations (five locations from head to foot in Part II, four torso locations in Part I and one navel location in Part III). Hick's law applied to the present study predicts that RTs should increase as the number of stimulus locations increase. Donders's subtraction method segments the reaction to vibrotactile stimuli into four stages: stimulus detection, stimulus discrimination, response selection and motion execution [39]. For recognition and choice RT experiments, it was shown that stimulus discrimination and response selection have larger effects on RTs compared to stimulus detection and motor execution [25], [39], [61]. In this study, participants were asked to push the trigger regardless of the stimuli location. We posit that the difference in RTs at the navel location is due to stimulus detection rather than stimulus discrimination, response selection, or motion execution. Participants' attentional capacities may have been divided across multiple potential stimulus locations during Parts I and II, as compared to Part III and the stimulus detection process may have subsequently been delayed. The effect of age was also notable in this scenario; RTs at the navel for OA differed by ~50 ms between Part II (five stimulus locations) and Part III (single stimulus location), which were approximately three times larger than the differences observed for the YA (~15 ms).

4.4.8. Applications

The results of this study should be considered and applied to the design of vibrotactile displays. Factors such as stimulus frequency, factor type, number of factors in the same location, stimulus location or age could play an important role in applications that require relatively high-frequency motions (i.e., less than 500 ms) including surgical applications [20], real-time gait

training, golf training [21], or fall prevention [19]. For example, when changes were made in the experimental configurations used for investigating the effects of these factors on RTs in this study, RTs varied by as much as 50 ms, which is approximately 10% of the required time to complete the high-frequency motions. Furthermore, a longer RT should be expected if a vibrotactile display involves multiple stimulus locations. Thus, the number of stimulus locations should also be carefully considered.

For other applications where motions are generally more than 500 ms in duration, factors that affect RTs by less than 50 ms may not be as important, but factors such as stimulus location (i.e., head vs foot) or age should be considered because this study has shown that they can affect RTs by at least 100 ms. In applications where secondary cognitive tasks are likely to be performed such as in driving or flying [4], [5], [22], a significant increase in RTs should be expected as the results of this study demonstrated increased RTs of at least 130 ms when a cognitive task was performed.

This study also demonstrated that the selection of stimulus locations should be intentionally chosen to take into consideration delays in RTs as a function of the distance from the forehead, skin sensitivity, and stimulus-response compatibility. For example, in this study, faster RTs to stimuli were observed at the head location compared with the finger location when the participants were asked to perform a simple motor task (i.e., push a trigger using the thumb). However, when the secondary cognitive task was performed, faster RTs were observed at the finger location compared with the head location, potentially due to the higher compatibility between the stimuli-response set. In vibrotactile-based balance exercise training, the two most commonly used display locations to date have been the head [12], [18] and the torso [13], [62], [63]. In this context, trainees typically use vibrotactile cues to prompt postural corrections of

their center of mass (i.e., response location), which is approximately located at the level of the torso-based tactors used in this study. For a single motor task, like the task used in this study, RTs for the head location were faster than RTs the torso location. For this scenario, one might predict that the response-stimuli compatibility will yield RTs as fast if not faster than stimuli applied to the forehead despite the use of an inferior anatomical body segment.

Lastly, even though RTs to visual and auditory cues might elicit faster responses and potentially provide richer informative than vibrotactile cues [64], vibrotactile cues have distinct advantages within the fast-growing field of wearable technologies. Firstly, vibrotactile display have the potential to cause less interference with daily activities. For example, visual and auditory displays may not be suitable for use during scenarios involving simultaneous viewing or listening tasks. [7]. Secondly, vibrotactile displays may provide more flexibility for specific types of applications, e.g., for balance rehabilitation, vibrotactile displays can be used during eyes-closed and head movement exercise configurations [65]. Multi-modal displays may be appropriate in certain scenarios to address disadvantages of individual display modalities, however multi-modal displays may require more complicated and expensive instrumentation [66].

4.4.9. Limitations

There were several limitations of this study. Although contact pressure was roughly controlled between the skin and tactors using a shim, it likely varied somewhat within and across participants, which may have influenced the skin sensitivity [67]. In part III, the C-2 tactors used in this study may have been manufactured with different polarities thereby potentially affecting the results since RTs to parallel and anti-parallel vibrations might be different. Factors such as arousal [68], [69], fatigue [69], prior experience, and physical parameters including height and

weight were not controlled. The environmental noise level in the laboratory setting was not measured, which may have affected the results in Part I. This study did not investigate the effects of increasing increments of age on RTs, nor the effects of a secondary task on RTs for YA. Additionally, only three types of factors were included in this study. Finally, only simple responses to vibrotactile stimuli were studied as opposed to complex vibration patterns and alternative modes of response.

4.5. References

- [1] S. O’Modhrain, N. A. Giudice, J. A. Gardner, and G. E. Legge, “Designing media for visually-impaired users of refreshable touch displays: Possibilities and pitfalls,” *IEEE Trans. Haptics*, vol. 8, no. 3, pp. 248–257, 2015.
- [2] S. Choi and K. J. Kuchenbecker, “Vibrotactile display: Perception, technology, and applications,” *Proc. IEEE*, vol. 101, no. 9, pp. 2093–2104, 2013.
- [3] K. Myles and M. S. Binseel, “The Tactile Modality : A Review of Tactile Sensitivity and Human Tactile Interfaces,” *Army Res. Lab.*, no. May, pp. 1–27, 2007.
- [4] J. B. F. Van Erp and H. a H. C. Van Veen, “Vibrotactile in-vehicle navigation system,” *Transp. Res. Part F Traffic Psychol. Behav.*, vol. 7, pp. 247–256, 2004.
- [5] F. Meng, C. Ho, R. Gray, and C. Spence, “Dynamic vibrotactile warning signals for frontal collision avoidance: towards the torso versus towards the head,” *Ergonomics*, vol. 58, no. March 2015, pp. 411–425, 2014.
- [6] C. Ho, H. Z. Tan, and C. Spence, “Using spatial vibrotactile cues to direct visual attention in driving scenes,” *Transp. Res. Part F Traffic Psychol. Behav.*, vol. 8, pp. 397–412, 2005.
- [7] Y. Salzer, T. Oron-Gilad, A. Ronen, and Y. Parmet, “Vibrotactile ‘on-thigh’ alerting

- system in the cockpit.,” *Hum. Factors*, vol. 53, no. 2, pp. 118–131, 2011.
- [8] H. a H. C. van Veen and J. van Erp, “Tactile information presentation in the cockpit,” *Haptic Human-Computer Interact.*, vol. 2058, pp. 174–181, 2001.
- [9] A. M. Okamura, M. R. Cutkosky, and J. T. Dennerlein, “Reality-based models for vibration feedback in virtual environments,” *IEEE/ASME Trans. Mechatronics*, vol. 6, no. 3, pp. 245–252, 2001.
- [10] A. Bloomfield and N. I. Badler, “Virtual Training via Vibrotactile Arrays,” *Presence Teleoperators Virtual Environ.*, vol. 17, no. 2, pp. 103–120, 2008.
- [11] K. Bark, P. Khanna, R. Irwin, P. Kapur, S. A. Jax, L. J. Buxbaum, and K. J. Kuchenbecker, “Lessons in using vibrotactile feedback to guide fast arm motions,” *2011 IEEE World Haptics Conf. WHC 2011*, pp. 355–360, 2011.
- [12] W. Nanhoe-Mahabier, J. H. Allum, E. P. Pasman, S. Overeem, and B. R. Bloem, “The effects of vibrotactile biofeedback training on trunk sway in Parkinson’s disease patients.,” *Parkinsonism Relat. Disord.*, vol. 18, no. 9, pp. 1017–21, Nov. 2012.
- [13] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.
- [14] E. Kentala, J. Vivas, and C. Wall III, “Reduction of postural sway by use of a vibrotactile balance prosthesis prototype in subjects with vestibular deficits.,” *Ann. Otol. Rhinol. Laryngol.*, vol. 112, no. 5, pp. 404–9, May 2003.
- [15] P. B. Shull, K. L. Lurie, M. R. Cutkosky, and T. F. Besier, “Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback.,”

- J. Biomech.*, vol. 44, no. 8, pp. 1605–9, May 2011.
- [16] P. B. Shull, W. Jirattigalachote, M. a Hunt, M. R. Cutkosky, and S. L. Delp, “Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention.,” *Gait Posture*, vol. 40, no. 1, pp. 11–19, May 2014.
- [17] C.-C. Lin, S. L. Whitney, P. J. Loughlin, J. M. Furman, M. S. Redfern, K. H. Sienko, and P. J. Sparto, “The effect of age on postural and cognitive task performance while using vibrotactile feedback,” *J. Neurophysiol.*, vol. 113, no. 7, pp. 2127–2136, 2015.
- [18] J. A. Goebel, B. C. Sinks, B. E. Parker, N. T. Richardson, A. B. Olowin, and R. W. Cholewiak, “Effectiveness of head-mounted vibrotactile stimulation in subjects with bilateral vestibular loss: a phase 1 clinical trial.,” *Otol. Neurotol.*, vol. 30, no. 2, pp. 210–216, 2009.
- [19] F. Asseman, A. M. Bronstein, and M. A. Gresty, “Using vibrotactile feedback of instability to trigger a forward compensatory stepping response,” *J. Neurol.*, vol. 254, no. 11, pp. 1555–1561, 2007.
- [20] A. R. Peon and D. Prattichizzo, “Reaction times to constraint violation in haptics: Comparing vibration, visual and audio stimuli,” *2013 World Haptics Conf. WHC 2013*, pp. 657–661, 2013.
- [21] R. Neal and B. D. Wilson, “3D Kinematics and Kinetics of the Golf Swing,” *Int. J. Sport Biomech.*, vol. 1, pp. 221–233, 1982.
- [22] C. Ho, R. Gray, and C. Spence, “Reorienting driver attention with dynamic tactile cues,” *IEEE Trans. Haptics*, vol. 7, no. 1, pp. 86–94, 2014.
- [23] V. Harrar and L. R. Harris, “Simultaneity constancy: detecting events with touch and

- vision,” *Exp. Brain Res.*, vol. 166, no. 3–4, pp. 465–473, 2005.
- [24] P. B. Shull, X. Zhu, and M. R. Cutkosky, “Continuous Movement Tracking Performance for Predictable and Unpredictable Tasks With Vibrotactile Feedback,” *IEEE Trans. Haptics*, vol. 1412, no. c, pp. 1–1, 2017.
- [25] W. E. Hick, “On the rate of gain of information,” *Q. J. Exp. Psychol.*, vol. 4, no. 1, pp. 11–26, 1952.
- [26] S. Haggerty, L.-T. Jiang, A. Galecki, and K. H. Sienko, “Effects of biofeedback on secondary-task response time and postural stability in older adults.,” *Gait Posture*, vol. 35, no. 4, pp. 523–8, Apr. 2012.
- [27] G. B. Taware, M. V Bhutkar, P. M. Bhutkar, V. P. Doijad, and a D. Surdi, “Effect of Age on Audio-Visual and Whole Body Reaction Time,” *Am. J. Med. Sci.*, vol. 5, no. 1, pp. 90–94, 2012.
- [28] A. W. Y. Ng and A. H. S. Chan, “Finger Response Times to Visual, Auditory and Tactile Modality Stimuli,” *Proc. Int. MultiConference Eng. Comput. Sci.*, vol. II, pp. 1449–1454, 2012.
- [29] B.-C. Lee, B. J. Martin, and K. H. Sienko, “The effects of actuator selection on non-volitional postural responses to torso-based vibrotactile stimulation.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 21, 2013.
- [30] B. J. P. Mortimer, G. A. Zets, and R. W. Cholewiak, “Vibrotactile transduction and transducers,” *J. Acoust. Soc. Am.*, vol. 121, no. 5, p. 2970, May 2007.
- [31] R. J. Kosinski, “A Literature Review on Reaction Time,” in *Clemson University*, 2008, vol. 10.
- [32] Y. Pan, S. K. Renganayagalu, and S. Komandur, “Tactile Cues for Ship Bridge

- Operations,” *27th Eur. Conf. Model. Simul.*, pp. 177--183, 2013.
- [33] A. J. M. Boulton, D. G. Armstrong, S. F. Albert, R. G. Frykberg, R. Hellman, M. S. Kirkman, L. A. Lavery, J. W. LeMaster, J. L. Mills, M. J. Mueller, P. Sheehan, and D. K. Wukich, “Comprehensive Foot Examination and Risk Assessment: A report of the Task Force of the Foot Care Interest Group of the American Diabetes Association, with endorsement by the American Association of Clinical Endocrinologists,” *Diabetes Care*, vol. 31, no. 8, pp. 1679–1685, 2008.
- [34] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, “Cell phone based balance trainer,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, Jan. 2012.
- [35] R. W. Cholewiak, J. C. Brill, and A. Schwab, “Vibrotactile localization on the abdomen: effects of place and space.,” *Percept. Psychophys.*, vol. 66, no. 6, pp. 970–987, 2004.
- [36] G. Yogev, J. M. Hausdorff, and N. Giladi, “The Role of Executive Function and Attention in Gait,” vol. 23, no. 3, 2009.
- [37] L. A. Brown, A. Shumway-Cook, and M. H. Woollacott, “Attentional demands and postural recovery: the effects of aging.,” *J. Gerontol. A. Biol. Sci. Med. Sci.*, vol. 54, no. 4, pp. M165–M171, 1999.
- [38] B. Forster, C. Cavina-Pratesi, S. M. Aglioti, and G. Berlucchi, “Redundant target effect and intersensory facilitation from visual-tactile interactions in simple reaction time,” *Exp. Brain Res.*, vol. 143, no. 4, pp. 480–487, 2002.
- [39] J. O. Miller and K. Low, “Motor Processes in Simple, Go/No-Go, and Choice Reaction Time Tasks: A Psychophysiological Analysis.,” *J. Exp. Psychol. Hum. Percept. Perform.*, vol. 27, no. 2, pp. 266–289, 2001.
- [40] C. Suied, P. Susini, and S. McAdams, “Evaluating warning sound urgency with reaction

- times.," *J. Exp. Psychol. Appl.*, vol. 14, no. 3, pp. 201–212, 2008.
- [41] C. Krueger and L. Tian, "A Comparison of the General Linear Mixed Model and Repeated Measures ANOVA Using a Dataset with Multiple Missing Data Points," *Biol. Res. Nurs.*, vol. 6, no. 2, pp. 151–157, 2004.
- [42] L. J. Dorfman and T. M. Bosley, "Age-related changes in peripheral and central nerve conduction in man.," *Neurology*, vol. 29, no. 1, pp. 38–44, 1979.
- [43] D. C. Park, "The basic mechanisms accounting for age-related decline in cognitive function," *Cogn. aging A Prim.*, vol. 11, no. 1, pp. 3–19, 2000.
- [44] T. A. Salthouse, T. M. Atkinson, and D. E. Berish, "Executive Functioning as a Potential Mediator of Age-Related Cognitive Decline in Normal Adults," *J. Exp. Psychol. Gen.*, vol. 132, no. 4, pp. 566–594, 2003.
- [45] J. M. Thornbury and C. M. Mistretta, "Tactile sensitivity as a function of age.," *J. Gerontol.*, vol. 36, no. 1, pp. 34–9, 1981.
- [46] S. J. Bolanowski, G. A. Gescheider, R. T. Verrillo, and C. M. Checkosky, "Four channels mediate the mechanical aspects of touch.," *J. Acoust. Soc. Am.*, vol. 84, no. 5, pp. 1680–1694, 1988.
- [47] R. T. Verrillo, "Effect of Contactor Area on the Vibrotactile Threshold," *J. Acoust. Soc. Am.*, vol. 35, no. 12, p. 1962, Jul. 1963.
- [48] J. C. Craig and C. E. Sherrick, "The role of skin coupling in the determination of vibrotactile spatial summation," *Percept. Psychophys.*, vol. 6, no. 2, pp. 97–101, 1969.
- [49] A. Wilska, "On the vibrational sensitivity in different regions of the body surface.," *Acta Physiol. Scand.*, vol. 31, no. 1952, pp. 284–289, 1954.
- [50] G. Macefield, S. C. Gandevia, and D. Burke, "Conduction velocities of muscle and

- cutaneous afferents in the upper and lower limbs of human subjects,” *Brain*, vol. 112, no. 6, pp. 1519–1532, 1989.
- [51] A. Diederich and H. Colonius, “Bimodal and trimodal multisensory enhancement: effects of stimulus onset and intensity on reaction time.,” *Percept. Psychophys.*, vol. 66, no. 8, pp. 1388–1404, 2004.
- [52] G. v. Békésy, “Vibration of the Head in a Sound Field and Its Role in Hearing by Bone Conduction,” *J. Acoust. Soc. Am.*, vol. 20, no. 6, p. 749, Jun. 1948.
- [53] S. Stenfelt, “Acoustic and Physiologic Aspects of Bone Conduction Hearing,” *Adv Otorhinolaryngol*, vol. 71, pp. 10–21, 2011.
- [54] R. Mohebbi and R. Gray, “Driver Reaction Time to Tactile and Auditory Rear-End Collision Warnings While Talking on a Cell Phone,” vol. 51, no. 1, pp. 102–110, 2005.
- [55] S. Kornblum, T. Hasbroucq, and A. Osman, “Dimensional overlap: cognitive basis for stimulus-response compatibility--a model and taxonomy.,” *Psychol. Rev.*, vol. 97, no. 2, pp. 253–270, 1990.
- [56] J. R. Simon, “The effects of an irrelevant directional cue on human information processing,” *Adv. Psychol.*, vol. 65, pp. 31–86, 1990.
- [57] R. Proctor and H. Tan, “Implications of compatibility and cuing effects for multimodal interfaces,” *Proc. ...*, 2005.
- [58] C. Ho and C. Spence, “Effectively responding to tactile stimulation: Do homologous cue and effector locations really matter?,” *Acta Psychol. (Amst.)*, vol. 151, pp. 32–39, 2014.
- [59] H. Pashler, “Dual-task interference in simple tasks: Data and theory.,” *Psychol. Bull.*, vol. 116, no. 2, pp. 220–244, 1994.
- [60] H. Pashler and J. C. Johnston, “Attentional Limitations in Dual-task Performance,” *Dual-*

- Task Interf. Simple Tasks*, pp. 155–189, 1998.
- [61] S. Sternberg, “Memory-scanning: mental Processes revealed by reaction-time experiments,” vol. 57, no. 4, pp. 421–457, 1969.
- [62] C. Wall III and E. Kentala, “Effect of displacement, velocity, and combined vibrotactile tilt feedback on postural control of vestibulopathic subjects,” *J. Vestib. Res.*, vol. 20, no. 1, pp. 61–9, Jan. 2010.
- [63] J. Lee, T. Bao, and K. H. Sienko, “Cell phone telerehabilitation device for home-based balance training,” in *International Society for Posture and Gait Research World Congress*, 2015.
- [64] N. A. Giudice, J. M. Loomis, R. L. Klatzky, and C. R. Bennett, “Touch-Screen Technology for the Dynamic Display of 2D Spatial Information Without Vision: Promise and Progress,” *Multisens. Res.*, vol. 27, no. 5–6, pp. 359–378, 2014.
- [65] T. Bao, W. J. Carender, C. Kinnaird, V. J. Barone, G. Peethambaran, S. L. Whitney, M. Kabeti, R. D. Seidler, and K. H. Sienko, “Effects of long-term balance training with vibrotactile sensory augmentation among community-dwelling healthy older adults,” *J. Neuroeng. Rehabil.*, vol. 15, no. 5, pp. 1–13, 2018.
- [66] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, “Trunk sway reductions in young and older adults using multi-modal biofeedback,” *Gait Posture*, vol. 31, no. 4, pp. 465–72, Apr. 2010.

Chapter 5 A Machine Learning Approach to Automatically Evaluate the Performance of Standing Balance Exercises

5.1. Background

Falls caused by balance impairments lead to loss of mobility, anxiety, and reduced quality of life [1]–[6]. To improve balance performance in people with balance deficits (e.g., vestibular disorders, Parkinson’s disease, stroke), balance rehabilitation programs are commonly prescribed [7]–[11]. Within a rehabilitation program, there are more than 200 static standing, dynamic standing and gait balance exercises, with levels of difficulty ranging from easy (1) to hard (5) [12], [13]. Balance exercises are generated by varying the stance, vision, standing surface and head motion conditions [12]. In clinics, physical therapists (PTs) select and customize balance exercises for each individuals [12]. At home, individuals perform balance exercises based on either instructions given by PTs during in-clinic training or paper instructions [14]. However, due to the lack of supervision and consequent loss of motivation, in-home training is not as effective as in-clinic training and therefore does not lead to the same improvements in balance-related function outcomes [15]–[17].

To improve the effectiveness of in-home balance training, researchers have introduced remote supervision (e.g., telerehabilitation) or semi-supervision (e.g., wearable devices with periodic expert input) [18], [19]. Telerehabilitation delivers remote rehabilitation services,

including assessments and intervention, via telecommunication networks [18]. Cikajlo et al. showed that in people with stroke, telerehabilitation with virtual reality tasks led to balance improvements similar to conventional in-clinic balance [20]. Telerehabilitation, however, generally requires remote interactions between the expert and user for the duration of the training session via video conference. This may require longer PT consultation times [21]. Video game-based in-home balance training has been shown to improve clinical measures after a minimum of five weeks of training [22]–[25]. Such training protocols use balance platforms (e.g., Wii Fit balance board) and a display screen to provide visual cues of balance. These restrictions limit the utility of such protocols during balance exercises that require closed eyes, head movements, and altered stances. Bao et al. demonstrated improvements in clinical balance outcomes using a semi-supervised balance training protocol for older adults in their homes; both the experimental group that received vibrotactile sensory augmentation during balance training via a smartphone balance trainer and a control group that performed the exercises unaided showed significant improvements in clinical outcomes [13]. PTs remotely determined exercise progression by reviewing participants' data on a weekly basis. Based on the number of step-outs from the correct position and participants' self-assessments on a 1-5 scale, they prescribed exercises for the following week. However, the participants' self-assessments may not align with balance assessments of the PT [26], [27], thus affecting the exercise progression and limiting the effectiveness of balance training.

More recently, researchers have turned to machine learning (ML) techniques to aid in making clinical assessments based on body motion [28]–[34]. Using motion data from wearable sensors, researchers have successfully leveraged ML techniques to identify activity [28], mine gait patterns [29], and classify gait disorders [30]–[33]. For example, Begg and Owen applied

ML techniques to automatically recognize young versus old gait types with 83.3% accuracy [30]. Using data on minimal foot clearance collected from a motion capture system they successfully trained a support vector machine (SVM) classifier. LeMoyné et al. extracted features from motion data measured by an inertial measurement unit (IMU) on the ankle joint [31]. These features were used to train a neural network to distinguish between older adults and people with Friedreich's ataxia [31]. Mannini et al. used time and frequency information extracted from gait data to train a SVM to classify subjects as older adults, people with stroke, and people with Huntington's disease achieving a classification accuracy of 90.5% [32].

Appropriate progression of balance exercises within a balance-training regimen is critical to achieving improvements in balance outcomes for both clinic- and home-based settings [12]. The extent to which the human (here, the expert/PT) is "in the loop" is diminished as one shifts from a telerehabilitation to a semi-supervised balance rehabilitation program paradigm [35]. The use of ML techniques enables potential further reduction or elimination of the human-in-the-loop with respect to decision-making regarding exercise progression to support home-based balance rehabilitation training. The objective of this study was to apply ML techniques to learn an accurate mapping from trunk sway collected by an IMU to a PT's assessment ratings of balance performance (e.g., Functional Independence Measure [36]). In this pilot study we assessed the feasibility of an ML-based approach for automatically and accurately evaluating balance exercise performance.

5.2. Methods

5.2.1. Data

Sixteen participants (68.2 ± 8.0 yrs, five males, 11 females) with balance disorders or balance concerns performed balance exercises during 18 sessions over the course of six weeks. All participants gave written informed consent, and the study was conducted in accordance with the Declaration of Helsinki. The study protocol was reviewed and approved by the University of Pittsburgh Institutional Review Board (PRO13020399). In each session, participants performed multiple trials of two standing exercises. Exercises were selected by a PT from a set of 60 standard standing exercises for balance rehabilitation [12], [37]. The PT used clinical judgment to select the exercises with moderate difficulty level. The exercises were generated by varying the visual (open/closed), stance (feet apart/feet together/semi-tandem Romberg/tandem Romberg/single leg stance), head motion (none/pitch/yaw), and support surface (firm/foam) conditions [12]. Each participant performed six trials of a given exercise, and each trial lasted for 30 seconds. If participants had to step out or needed help in maintaining balance, the trial was terminated and marked as a “step-out” trial. After each set of six trials, both the PT and the participant who performed the exercise rated balance performance across the six trials on a scale of 1 to 5. The rating scale used by the PT was adapted from the Functional Independence Measure [36], and the rating scale for the participants was adapted from Espy et al. [38]. The guidelines for the ratings for both scales are shown in Table 5.1.

Table 5.1: Rating guidelines for the physical therapist (adapted from Functional Independence Measure [36]) and participants (adapted from Espy et al. [38])

Ratings	Description for physical therapist	Description for participants
1	Independent with no sway	I feel completely steady
2	Supervision with minimal sway	I feel a little unsteady or off-balance
3	Close supervision with moderate sway	I feel somewhat unsteady, or like I may lose my balance
4	Requires physical assistance or positive stepping strategy after 15 seconds	I feel very unsteady, or like I definitely will lose my balance
5	Unable to maintain position with assist or step out in the first 15 seconds of the exercise	I lost my balance

During the training, participants wore a wearable IMU (MTx, Xsens Technologies B.V.) aligned with the L4/L5 spinal segment level on their back. The IMU comprised accelerometers and gyroscopes and customized filters to estimate trunk sway relative to the gravitational vector in both the pitch and roll directions. Figure 5.1 shows an example of trunk sway in the pitch and roll directions recorded during an exercise trial.

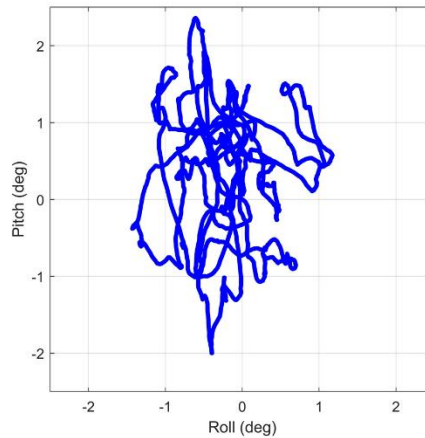


Figure 5.1: An example of trunk sway recorded during an exercise trial (feet apart stance on foam surface with eyes closed). In this particular example, the participant demonstrates more sway (i.e., variation) in the pitch direction relative to the roll direction.

In total, we rating labels were applied to 576 sets of six trials (16 participants x 18 sessions x 2 exercises). However, technical issues resulted in the loss of trunk sway data for three sessions. After discarding those three sessions, we were left with 570 labeled sets performed by

16 participants. Figure 5.2 shows the number of exercises included in the data set as a function of the rating levels provided by the PT and participants.

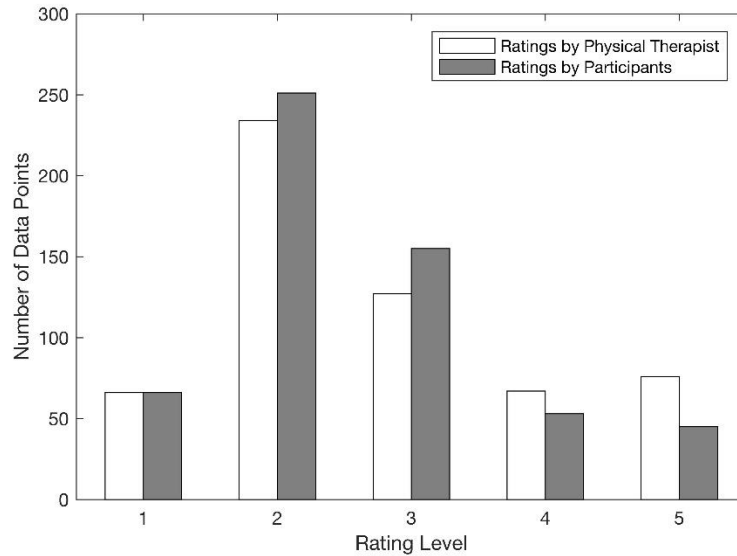


Figure 5.2: Distribution of rating levels provided by the physical therapist and participants.

5.2.2. Data Features

From the trunk sway data, we extracted features to capture balance performance over the course of a set of six trials. These features are described in Figure 5.3. For each trial, we used ten commonly used kinematic metrics to describe the trunk motion [39]–[46]. Since we only obtained one rating for each set of six trials, we summarized the metrics across the set of six trials by computing six statistical descriptors: 1) mean, 2) standard derivation, 3) minimum, 4) maximum, 5) median, and 6) slope (1, 0, or -1) for all ten metrics as shown in Figure 5.3. We added *the number of non-“step-out” trials* across the six trials as the last feature. This resulted in 61 continuous valued features (Figure 5.3) for each set of trials. Since these features lie on different scales, we z-scaled all features to zero mean and unit standard deviation. This preprocessing was done in MATLAB (MathWorks, R2016b).

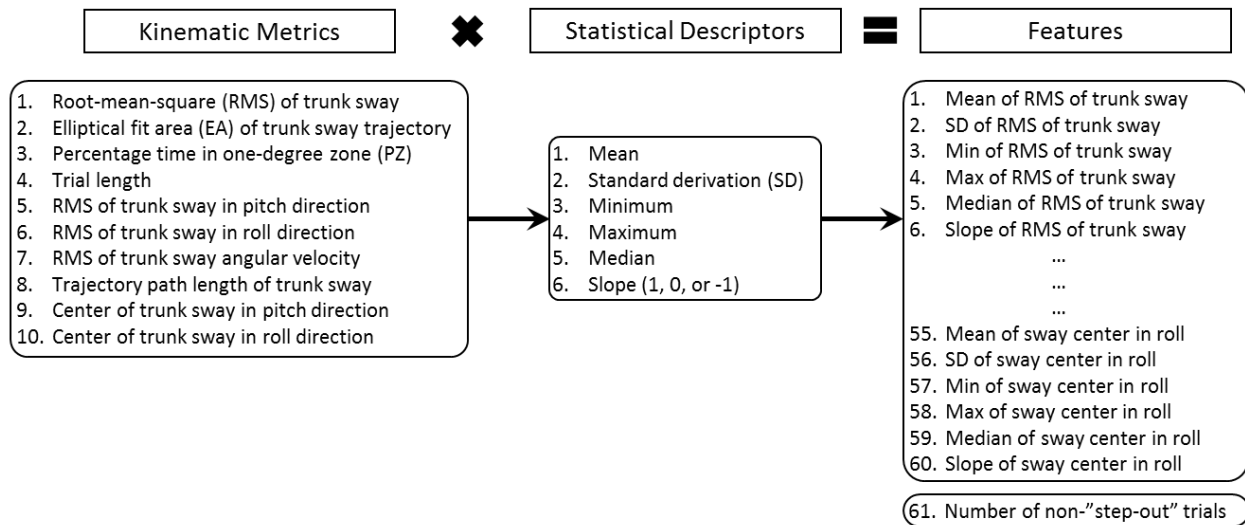


Figure 5.3: We used 61 features to represent each set of six trials. These features are based on ten commonly used metrics and statistical descriptors to summarize performance across the different trials.

5.2.3. Classification

Using the data above, we aimed to predict the PT's ratings based on trunk sway by learning a mapping from the 61-dimensional feature space to the PT's ratings. To learn this mapping, we trained a multi-class support vector machine (SVM) with a linear kernel [47]. We also considered other non-linear techniques (e.g., support vector regression, extreme gradient boosting (XGBoost) [48]). However, these techniques did not yield statistically significantly different results from the linear SVM approach and lacked interpretability. To build the multi-class SVM, we learned a separate classifier for each pair of classes (i.e., we used the one-against-one framework). To account for the variation in frequency across classes, we used asymmetric costs [49]. The asymmetric costs were set to the inverse frequency of each class (i.e., rating level). Given the ten classifiers (e.g., 1 vs. 2, 1 vs. 3, etc.), test examples were then classified by applying each classifier in turn and taking the majority vote.

5.2.4. Validation and Evaluation

We used the leave-one-participant-out method for both validation and evaluation. During the validation phase we tuned the hyperparameters and during the evaluation phase we evaluated model performance. The use of leave-one-participant-out ensures no bias of the classifier, by hiding the test participant's data from the classifier prior to testing [50]. The details of this process are shown in Figure 5.4. The data of each participant, in turn, was marked as the test dataset, and the remaining data were used for training and validation. Within each training dataset, we used leave-one-participant-out cross-validation to tune the hyperparameters. In our framework, we tuned the SVM cost parameter (C), sweeping from $[1e-7, 1e3]$. We averaged the performance across cross-validation folds, and selected the C that led to the best averaged F1 score. We used the averaged F1 score because of the imbalanced distribution of rating levels as demonstrated in Figure 5.2. Given this optimal C , we trained a final classifier on the training data and then applied it to the test dataset. We repeated the process for each participant (i.e., repeated 16 times) to obtain the final evaluation results of overall classification accuracy and averaged F1 score.

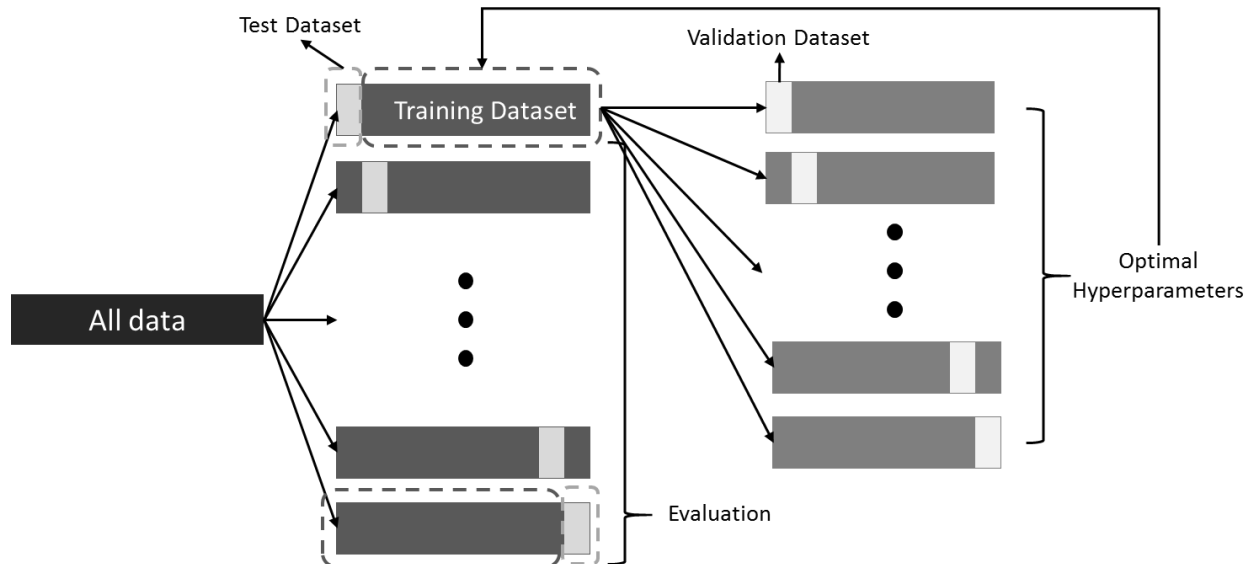


Figure 5.4: Validation and evaluation of the classifier using the leave-one-participant-out cross validation. For visualization, only the first cross-validation is shown.

The averaged F1 score was obtained by averaging the F1 scores of all rating levels. We calculated the F1 score for each rating level using the precision and the recall at this rating level (Equations 5-1, 5-2, and 5-3).

$$F1 \text{ score} = 2 \cdot \frac{\text{precision} \cdot \text{recall}}{\text{precision} + \text{recall}} \quad \text{Equation 5-1}$$

$$\text{Precision} = \frac{\text{True Positive}}{\text{True Positive} + \text{False Positive}} \quad \text{Equation 5-2}$$

$$\text{Recall} = \frac{\text{True Positive}}{\text{True Positive} + \text{False Negative}} \quad \text{Equation 5-3}$$

5.2.5. Feature Importance

After evaluating model performance, we used backward feature elimination to measure the relative importance of the features. To avoid overfitting, we used a leave-one-participant-out setup similar to the one shown in Figure 5.4. For each training dataset, features were eliminated one by one based on model performance (i.e., averaged F1 score) until only a single feature remained. This gave us a measure of the relative importance of each feature, where the first

feature to be eliminated had a relative importance of 61 (i.e., least important), and the last feature to be eliminated resulted in a relative importance of 1 (i.e., most important). This feature elimination procedure was repeated for each participant (i.e., 16 times). By averaging the relative importance of each feature (i.e., 16 relative importance), we obtained an overall ranking of features based on the relative importance. To understand the relative importance of each metric and each statistical descriptor, we averaged the relative importance of features within each category (i.e., the 10 metrics and 6 statistical descriptors shown in Figure 5.3).

5.2.6. Additional Classification

Based on input from PTs, we considered an additional three-level classification task in which we grouped rating levels 1 and 2 together, and rating levels 3 and 4 together. In general, PTs identified exercises rated as 1 or 2 as easy/safe, exercises rated as 3 or 4 as moderate/suitable, and exercises rated 5 as difficult/unsafe.

5.2.7. Statistical Analysis

To compare the agreement between the SVM and PT assessments (i.e., classification accuracy) with the agreement between PT and self-assessments, we used a paired two-tail t-test with a significance level of 0.05. All classification and statistical analyses were performed using R (r-project.org, version 3.4.1).

5.3. Results

The performance of the five-class and three-class classification models is shown in Table 5.2. For the five-class classification task, the SVM predictions agreed with the PT's ratings with

an accuracy of 64.3% and averaged F1 score of 0.64. The accuracy is 13.8 percentage points ($p < 0.001$) better and the averaged F1 score is 0.18 better than the agreements achieved by the self-assessment ratings. After grouping ratings into three-classes, the accuracy and averaged F1 score improved to 82.0% and 0.81, respectively. Moreover, the SVM still outperformed the participants' self-ratings (13.2 percentage points ($p < 0.001$) for accuracy and 0.19 for the averaged F1 score).

Table 5.2: Results for the five-level and three-level classification models. Standard deviations are reported and the symbol * indicates statistical significance ($p < 0.05$).

Models	Five-class Classification		Three-class Classification	
	Accuracy*	Averaged F1 Score	Accuracy*	Averaged F1 Score
Participant	50.6±08.1%	0.46±0.16	68.8±9.8%	0.62±0.15
SVM	64.3±11.3%	0.64±0.11	82.0±8.4%	0.81±0.07

The confusion matrices for the self-assessments and five-class SVM classifier predictions are shown in Table 5.3. The calculated F1 score, precision, and recall for each rating level are also shown with the confusion matrices. The SVM predictions outperformed participants' self-assessments for all rating levels for the F1 scores.

Table 5.3: Confusion matrix for (A) participants' self-assessments versus the physical therapist's assessments and (B) SVM predictions versus physical therapist's assessments.

A

		Self-assessments					Recall
		1	2	3	4	5	
PT assessments	1	36	29	1	0	0	0.55
	2	27	154	51	2	0	0.66
	3	3	49	53	21	1	0.42
	4	0	15	26	14	12	0.21
	5	0	4	24	16	32	0.42
Precision		0.55	0.61	0.34	0.27	0.71	Accuracy: 51%
F1 Score		0.55	0.64	0.38	0.23	0.53	

B

		SVM Predictions					Recall
		1	2	3	4	5	
PT assessments	1	56	10	0	0	0	0.85
	2	53	142	33	6	0	0.61
	3	3	29	68	26	1	0.54
	4	1	3	12	42	9	0.63
	5	0	0	3	14	59	0.78
Precision		0.5	0.77	0.59	0.48	0.86	Accuracy: 64%
F1 Score		0.63	0.68	0.56	0.54	0.81	

Table 5.4 shows ten features ranked by relative importance, all metrics ranked by relative importance, and all statistical descriptors ranked by relative importance obtained from the backward feature elimination method. The *number of non-"step-out" trials* and *min of RMS in roll* were the two most important features across all 61 features.

Table 5.4: (A) The top ten features and their relative importance; (B) The relative importance of each metric; (C) The relative importance of each statistical descriptor.

A

Rank	Features	Importance Ranking
1	# of Non- "Step-out" Trials	1.1 ± 0.5
2	Min of RMS in Roll	2.3 ± 0.7
3	Slope of RMS	7.9 ± 7.6
4	Mean of Trial Length	10.9 ± 11.3
5	Slope of PZ	12.2 ± 6.7
6	Slope of Path Length	12.8 ± 10.6
7	Slope of Center of Sway in Pitch	15.6 ± 5.3
8	Slope of RMS in Roll	17.3 ± 11.4
9	Slope of EA	18.7 ± 11.0
10	Min of EA	19.7 ± 12.9

B

Rank	Metrics	Importance Ranking
1	RMS in Roll	21.7±12.0
2	PZ	27.2±8.4
3	Trail Length	28.6±12.5
4	Path Length	31.0±11.5
5	Center of Sway in Roll	32.3±7.8
6	EA	33.5±11.9
7	RMS in Pitch	34.6±6.9
8	Center of Sway in Pitch	34.7±13.3
9	RMS of trunk sway	34.8±14.9
10	RMS of Velocity	36.6±6.8

C

Rank	Statistical Descriptors	Importance Ranking
1	Slope	19.4±7.7
2	Min	29.1±11.5
3	Median	31.8±6.8
4	Standard Deviation	33.6±8.4
5	Max	37.0±6.0
6	Mean	38.0±12.0

5.4. Discussion

In this pilot study, we used an SVM-based approach to automatically assess balance performance based on trunk sway collected from an IMU sensor. To the best of our knowledge, this is the first study to develop and validate the use of ML techniques to automatically provide PT-like assessments of balance performance. On held-out data, compared to self-assessment ratings, the ratings generated by the SVM classifier were significantly closer to the PT ratings.

In the five-class rating classification, the SVM outperformed the participants' self-assessments with respect to precision, recall, and F1 score at all rating levels with the exception of precision in rating level 1 and recall in rating level 2. Increased precision indicates a higher confidence when exercises are predicted at a particular rating level. Increased recall indicates that the model correctly captures a larger portion of the exercises at this rating level. Compared to self-assessments, the recall rate increased from 0.42 to 0.78 and the precision rate increased from 0.71 to 0.86 when using the SVM for rating level 5. This indicates that the SVM model is significantly better at identifying difficult/unsafe exercises that could cause the loss of balance than the participants. The recall for exercises rated as level 2 and precision for exercises rated as level 1 decreased slightly compared to the self-assessments. However, upon further inspection we noted that the algorithm re-classified many exercises rated level 2 as level 1. Further investigation is needed to understand the difference between rating levels 1 and 2. When designing a balance rehabilitation program, the inclusion of easy to perform exercises (i.e., exercises with low ratings (1,2)) should be minimized because they produce fewer benefits [51]. Exercises rated 3 or 4 (i.e., exercises most often recommended by PTs for repetitive training) showed both increased precision and recall.

Balance performance is often described by kinematic metrics including the RMS of trunk sway, PZ, EA of trunk sway, trial length, trajectory path length of trunk sway, and RMS of trunk sway angular velocity [39]–[46]. Here, we investigated the relative importance of different kinematic metrics and statistical descriptors to better understand how they are related to a PT's assessments. In this study, RMS of trunk sway, PZ, trial length, and path length were ranked as the top four indicators. Notably, RMS of trunk sway in the roll direction was more important than RMS in pitch direction and overall RMS, which suggests that RMS in roll direction is more closely associated with an exercise's level of difficulty. This observation reaffirms the reality that difficult exercises (e.g., Tandem stance with yaw head movements) challenge balance in roll direction. Melzer et al. also found that fallers had significantly higher trunk sway in the roll direction than the non-fallers for balance exercises performed with eyes open and eyes closed [52].

Among the statistical descriptors, the trend across six trials (i.e., slope) was most important. This suggests that the PT involved in this study considered performance over all six trials when making her evaluation. Average performance (i.e., mean) and worst performance (i.e., max) among the six trials were the least important statistical descriptors, whereas the best performance (i.e., min) among the six trials was the second most important statistical descriptor. The implication of these results is that one poorly performed trial may not affect the overall rating assigned to the performance, while a well-performed trial may drive the overall rating.

This pilot study has three major limitations. First, only one PT provided the ratings that were used as the gold standard. There may be relatively low intra-rater reliability for any given PT, which could reduce classification accuracy. Second, we only investigated standing exercises, but standard balance rehabilitation programs include weight shifting, modified center of gravity,

and gait exercises [10], [12], [13]. Different feature extraction methods are likely needed to automate ratings for these additional types of exercises. Third, PT ratings were unevenly distributed across classes. To mitigate this effect, we considered F1 scores when evaluating classification performance. Still, this could suggest why performance is lower in less prevalent classes.

Automated balance performance assessments have the potential to augment existing in-home balance training programs. In particular, automated assessments could guide participants in the selection of appropriate exercises to perform, potentially allowing for more flexibility compared to a fixed program since multiple exercise options would exist at any given difficulty level. Training performed at an appropriate level has been shown to lead to greater improvements in balance [12], [51]. In addition, the receipt of feedback immediately following the completion of an exercise may increase compliance and motivation, a general problem with traditional in-home training programs [22]. Moreover, accurate identification of difficult/unsafe exercises would likely reduce the performance of unsafe exercises in the absence of a PT, thereby decreasing the risk of falls in the home [53]. Aside from increase losses of balance and fall risk, exercises that exceed an individual's capabilities typically do not have positive effects on training outcomes [51]. A study comparing the effects of long-term in-home balance training with ML-informed exercise progression with telerehabilitation and/or semi-supervised approaches is warranted.

5.5. Conclusion

Using ML techniques, we successfully learned a mapping from trunk sway data from a single IMU sensor to a PT's ratings of balance performance. Compared to self-assessment

ratings, the automatically generated ratings more closely agreed with the PT's ratings. On a three-level scale, the model achieved an accuracy of 82%. The results of this study could be used to provide balance assessments during unsupervised or semi-supervised balance rehabilitation programs. Such automated assessments could lead to a reduction in PT consultation time, an increase in compliance and an overall improvement in the effectiveness of in-home balance training programs.

5.6. References

- [1] S. J. Herdman, P. Blatt, M. C. Schubert, and R. J. Tusa, "Falls in patients with vestibular deficits.," *Am. J. Otol.*, vol. 21, no. 6, pp. 847–851, 2000.
- [2] M. E. Tinetti, "Preventing Falls in Elderly Persons," *N. Engl. J. Med.*, vol. 348, pp. 42–49, 2003.
- [3] A. C. Scheffer, M. J. Schuurmans, N. Van dijk, T. Van der hooft, and S. E. De rooij, "Fear of falling: Measurement strategy, prevalence, risk factors and consequences among older persons," *Age Ageing*, vol. 37, no. 1, pp. 19–24, 2008.
- [4] O. A. Donoghue, H. Cronin, G. M. Savva, C. O'Regan, and R. A. Kenny, "Effects of fear of falling and activity restriction on normal and dual task walking in community dwelling older adults," *Gait Posture*, vol. 38, no. 1, pp. 120–124, 2013.
- [5] W. J. Rejeski and S. L. Mihalko, "Physical Activity and Quality of Life in Older Adults," *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 56, no. Supplement 2, pp. 23–35, Oct. 2001.
- [6] D. Monzani, L. Casolari, G. Guidetti, and M. Rigatelli, "Psychological distress and disability in patients with vertigo.," *J. Psychosom. Res.*, vol. 50, no. 6, pp. 319–323, 2001.

- [7] A. Meli, G. Zimatore, C. Badaracco, E. De Angelis, and D. Tufarelli, "Vestibular rehabilitation and 6-month follow-up using objective and subjective measures.," *Acta Otolaryngol.*, vol. 126, no. 3, pp. 259–66, 2006.
- [8] S. L. Whitney, A. H. Alghadir, and S. Anwer, "Recent Evidence About the Effectiveness of Vestibular Rehabilitation," *Curr. Treat. Options Neurol.*, vol. 18, no. 3, pp. 1–15, 2016.
- [9] A. Shumway-Cook, F. B. Horak, L. Yardley, and A. M. Bronstein, "Rehabilitation of balance disorders in the patient with vestibular pathology," *Clin. Disord. Balanc. Posture, Gait. London, United Kingdom Arnold, Div Hodder Headl. PLC*, pp. 213–220, 1996.
- [10] C. D. Hall, S. J. Herdman, S. L. Whitney, S. P. Cass, R. A. Clendaniel, T. D. Fife, J. M. Furman, T. S. D. Getchius, J. A. Goebel, N. T. Shepard, and S. N. Woodhouse, *Vestibular Rehabilitation for Peripheral Vestibular Hypofunction: An Evidence-Based Clinical Practice Guideline*, vol. 40, no. April. 2016.
- [11] S. L. Hillier and M. McDonnell, "Vestibular rehabilitation for unilateral peripheral vestibular dysfunction.," *Clin. Otolaryngol.*, vol. 36, no. 3, pp. 248–249, 2011.
- [12] B. N. Klatt, W. J. Carender, C. Lin, S. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, "A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular Disorders," *Phys. Med. Rehabil. - Int.*, vol. 2, no. 4, p. 1044, 2015.
- [13] T. Bao, W. J. Carender, C. Kinnaird, V. J. Barone, G. Peethambaran, S. L. Whitney, M. Kabeti, R. D. Seidler, and K. H. Sienko, "Effects of long-term balance training with vibrotactile sensory augmentation among community-dwelling healthy older adults," *J. Neuroeng. Rehabil.*, vol. 15, no. 5, pp. 1–13, 2018.
- [14] H. S. Cohen and K. T. Kimball, "Increased independence and decreased vertigo after

- vestibular rehabilitation,” *Otolaryngol. - Head Neck Surg.*, vol. 128, no. 1, pp. 60–70, 2003.
- [15] V. Lun, N. Pullan, N. Labelle, C. Adams, and O. Suchowersky, “Comparison of the effects of a self-supervised home exercise program with a physiotherapist-supervised exercise program on the motor symptoms of Parkinson’s disease,” *Mov. Disord.*, vol. 20, no. 8, pp. 971–975, 2005.
- [16] C. L. Kao, L. K. Chen, C. M. Chern, L. C. Hsu, C. C. Chen, and S. J. Hwang, “Rehabilitation outcome in home-based versus supervised exercise programs for chronically dizzy patients,” *Arch. Gerontol. Geriatr.*, vol. 51, no. 3, pp. 264–267, 2009.
- [17] A. Lacroix, R. W. Kressig, T. Muehlbauer, Y. J. Gschwind, B. Pfenninger, O. Bruegger, and U. Granacher, “Effects of a Supervised versus an Unsupervised Combined Balance and Strength Training Program on Balance and Muscle Power in Healthy Older Adults: A Randomized Controlled Trial,” *Gerontology*, vol. 62, no. 3, pp. 275–288, 2015.
- [18] R. B. Burns, D. Crislip, P. Daviou, A. Temkin, S. Vesmarovich, J. Anshutz, C. Furbish, and M. L. Jones, “Using telerehabilitation to support assistive technology,” *Assist. Technol.*, vol. 10, no. 2, pp. 126–133, 1998.
- [19] P. Bach-y-Rita, “Tactile sensory substitution studies,” *Ann. N. Y. Acad. Sci.*, vol. 1013, pp. 83–91, May 2004.
- [20] I. Cikajlo, M. Rudolf, N. Goljar, H. Burger, and Z. Matjačić, “Telerehabilitation using virtual reality task can improve balance in patients with stroke,” *Disabil Rehabil*, vol. 34, no. 1, pp. 13–18, 2012.
- [21] D. Kairy, P. Lehoux, C. Vincent, and M. Visintin, “A systematic review of clinical outcomes, clinical process, healthcare utilization and costs associated with

- telerehabilitation,” *Disabil. Rehabil.*, vol. 31, no. 6, pp. 427–447, 2009.
- [22] M. G. Jorgensen, U. Laessoe, C. Hendriksen, O. B. F. Nielsen, and P. Aagaard, “Efficacy of nintendo wii training on mechanical leg muscle function and postural balance in community-dwelling older adults: A randomized controlled trial,” *Journals Gerontol. - Ser. A Biol. Sci. Med. Sci.*, vol. 68, no. 7, pp. 845–852, 2013.
- [23] S. Roopchand-Martin, R. McLean, C. Gordon, and G. Nelson, “Balance Training with Wii Fit Plus for Community-Dwelling Persons 60 Years and Older.,” *Games Health J.*, vol. 4, no. 3, pp. 247–52, 2015.
- [24] A. S. Fu, K. L. Gao, A. K. Tung, W. W. Tsang, and M. M. Kwan, “Effectiveness of Exergaming Training in Reducing Risk and Incidence of Falls in Frail Older Adults with a History of Falls,” *Arch. Phys. Med. Rehabil.*, vol. 96, no. 12, pp. 2096–2102, 2015.
- [25] L. Prosperini, D. Fortuna, C. Gianni, L. Leonardi, M. R. Marchetti, and C. Pozzilli, “Home-Based Balance Training Using the Wii Balance Board,” *Neurorehabil. Neural Repair*, vol. 27, no. 6, pp. 516–525, 2013.
- [26] L. Eskelinen, A. Kohvakka, T. Merisalo, H. Hurri, and G. Wagar, “Relationship between the self-assessment and clinical assessment of health status and work ability,” *Scand J Work Env. Heal.*, vol. 17, no. 1, pp. 40–47, 1991.
- [27] F. D. Duffy and E. S. Holmboe, “Self-assessment in Lifelong Learning and Improving Performance in Practice,” *Jama*, vol. 296, no. 9, p. 1137, 2006.
- [28] A. Mannini and A. M. Sabatini, “Machine learning methods for classifying human physical activity from on-body accelerometers.,” *Sensors (Basel)*, vol. 10, no. 2, pp. 1154–75, Jan. 2010.
- [29] R. Caldas, M. Mundt, W. Potthast, F. Buarque de Lima Neto, and B. Markert, “A

- systematic review of gait analysis methods based on inertial sensors and adaptive algorithms,” *Gait Posture*, vol. 57, no. June, pp. 204–210, 2017.
- [30] R. K. Begg, M. Palaniswami, and B. Owen, “Support vector machines for automated gait classification,” *IEEE Trans. Biomed. Eng.*, vol. 52, no. 5, pp. 828–838, 2005.
- [31] R. LeMoyne, F. Heerinckx, T. Aranca, R. De Jager, T. Zesiewicz, and H. J. Saal, “Wearable body and wireless inertial sensors for machine learning classification of gait for people with Friedreich’s ataxia,” *BSN 2016 - 13th Annu. Body Sens. Networks Conf.*, pp. 147–151, 2016.
- [32] A. Mannini, D. Trojaniello, A. Cereatti, and A. M. Sabatini, “A machine learning framework for gait classification using inertial sensors: Application to elderly, post-stroke and huntington’s disease patients,” *Sensors (Switzerland)*, vol. 16, no. 1, 2016.
- [33] T. Nakano, B. T. Nukala, S. Zupancic, A. Rodriguez, D. Y. C. Lie, J. Lopez, and T. Q. Nguyen, “Gaits classification of normal vs. patients by wireless gait sensor and Support Vector Machine (SVM) classifier,” *2016 IEEE/ACIS 15th Int. Conf. Comput. Inf. Sci. ICIS 2016 - Proc.*, 2016.
- [34] D. T. H. Lai, R. K. Begg, and M. Palaniswami, “Computational intelligence in gait research: a perspective on current applications and future challenges.,” *IEEE Trans. Inf. Technol. Biomed.*, vol. 13, no. 5, pp. 687–702, 2009.
- [35] T. B. Sheridan, *Monitoring behavior and supervisory control*, vol. 1. Springer Science & Business Media, 2013.
- [36] R. A. Keith, C. V. Granger, B. B. Hamilton, and F. S. Sherwin, “The functional independence measure: a new tool for rehabilitation,” *Adv Clin Rehabil*, vol. 1, pp. 6–18, 1987.

- [37] T. Bao, W. J. Carender, C. Kinnaird, V. J. Barone, G. Peethambaran, S. L. Whitney, M. Kabeto, R. D. Seidler, and K. H. Sienko, “Effects of long-term balance training with vibrotactile sensory augmentation among community-dwelling healthy older adults: a randomized preliminary study (accepted),” *J. Neuroeng. Rehabil.*, 2017.
- [38] D. Espy, C. Nathan, W. Tony, K. Natalie, and R. M., “Development of a rating scale for perceived stability during balance training,” in *International Society for Posture and Gait Research*, 2014.
- [39] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.
- [40] K. H. Sienko, M. D. Balkwill, and C. Wall, “Biofeedback improves postural control recovery from multi-axis discrete perturbations.,” *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 53, 2012.
- [41] Z. Halická, J. Lobotková, K. Bučková, and F. Hlavačka, “Effectiveness of different visual biofeedback signals for human balance improvement,” *Gait Posture*, vol. 39, no. 1, pp. 410–414, 2014.
- [42] M. Dozza, F. B. Horak, and L. Chiari, “Auditory biofeedback substitutes for loss of sensory information in maintaining stance.,” *Exp. brain Res.*, vol. 178, no. 1, pp. 37–48, Mar. 2007.
- [43] J. C. Davis, M. C. Robertson, M. C. Ashe, T. Liu-Ambrose, K. M. Khan, and C. A. Marra, “International comparison of cost of falls in older adults living in the community: A systematic review,” *Osteoporos. Int.*, vol. 21, no. 8, pp. 1295–1306, 2010.

- [44] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, "Cell phone based balance trainer," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, Jan. 2012.
- [45] S. B. Lim, B. C. Horslen, J. R. Davis, J. H. J. Allum, and M. G. Carpenter, "Benefits of multi-session balance and gait training with multi-modal biofeedback in healthy older adults," *Gait Posture*, vol. 47, pp. 10–17, 2016.
- [46] M. Mancini, A. Salarian, P. Carlson-Kuhta, C. Zampieri, L. King, L. Chiari, and F. B. Horak, "ISway: a sensitive, valid and reliable measure of postural control.," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 59, Jan. 2012.
- [47] J. A. K. Suykens and J. Vandewalle, "Least Squares Support Vector Machine Classifiers," *Neural Process. Lett.*, vol. 9, no. 3, pp. 293–300, 1999.
- [48] T. Chen and C. Guestrin, "XGBoost : Reliable Large-scale Tree Boosting System," in *Proceedings of the 22nd ACM SIGKDD International Conference on Knowledge Discovery and Data Mining*, 2016, pp. 785–794.
- [49] K. Morik, K. Morik, P. Brockhausen, P. Brockhausen, T. Joachims, and T. Joachims, "Combining Statistical Learning with a Knowledge-Based Approach - a Case Study in Intensive Care Monitoring," *16th Int. Conf. Mach. Learn.*, pp. 268–277, 1999.
- [50] P. E. Taylor, G. J. M. Almeida, T. Kanade, and J. K. Hodgins, "Classifying human motion quality for knee osteoarthritis using accelerometers," *2010 Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBC'10*, pp. 339–343, 2010.
- [51] A. Reinthal, "Getting the Dosage Right in Balance Exercise Prescription: the Intensity Problem," *J. Nov. Physiother.*, vol. 7, no. S4, 2017.
- [52] I. Melzer, N. Benjuya, and J. Kaplanski, "Postural stability in the elderly: A comparison between fallers and non-fallers," *Age Ageing*, vol. 33, no. 6, pp. 602–607, 2004.

[53] F. B. Horak, "Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls?," *Age Ageing*, vol. 35, no. SUPPL.2, pp. 7–11, Sep. 2006.

Chapter 6 Discussion

6.1. Effects of Training with Vibrotactile Sensory Augmentation

Chapter 2 and Chapter 3 studied long-term balance training with and without vibrotactile sensory augmentation (SA). The experimental design for both studies emphasized 1) long-term training (i.e., at least 6 weeks or 18 sessions), 2) clinical tests before and after training, 3) control groups (i.e., experimental group (EG) training with vibrotactile sensory augmentation (SA) and control group (CG) training without SA), 4) reduced feedback frequency for the EG (i.e., vibrotactile SA was provided for four of the six exercise trials), and 5) participant-customized exercise progression. The major differences between the two studies were 1) participant populations (people with unilateral vestibular disorders (UVD) vs. healthy older adults (OA)), 2) balance training locale (in-clinic vs. in-home), and 3) supervision (direct supervision by physical therapist vs. remote semi-supervision by physical therapist).

All participants improved their SOT and Mini-BESTest scores after long-term balance training, which agrees with the findings of prior studies [1]–[9]. People with UVD improved their ABC scale and gait-related tests (i.e., gait speed, DGI, FGA) but healthy OA did not improve. The differing results were attributed to two possible factors: high outcome measures (especially for ABC scale) of healthy OA prior to their enrollment [10], and lack of direct supervision by physical therapists during in-home balance training [11]. Comparing both groups'

pre-training assessments, the two studies suggest that the EG demonstrated greater improvements for the Mini-BESTest after the training than the CG. For healthy OA, the EG also demonstrated greater improvements for the SOT than the CG after the training. However, there was no significant difference between the EG and CG in the gait-related tests, possibly due to the lack of vibrotactile SA provided during the gait exercises.

The Mini-BESTest and SOT scores for both the EG and CG at the pre-, mid- and post-training assessments are shown in Figure 6.1. Both the EG and CG showed larger improvements for the Mini-BESTest and SOT from the pre-training to the mid-training assessments compared to their improvements from the mid-training to the post-training assessments. This finding indicates that improvements slowed down for both groups after three–four weeks of training, which agrees with a study that found a smaller improvement after the first three–four weeks than in the first three–four weeks [9]. After the mid-training assessments, the CG showed little to no improvements, which suggests that the CG reached a functional plateau after completing half of the training (3-week training for people with UVD, 4-week training for healthy OA). However, the EG showed some improvements after completing half of the training. This finding indicates that the training with vibrotactile SA could result in a higher functional plateau.

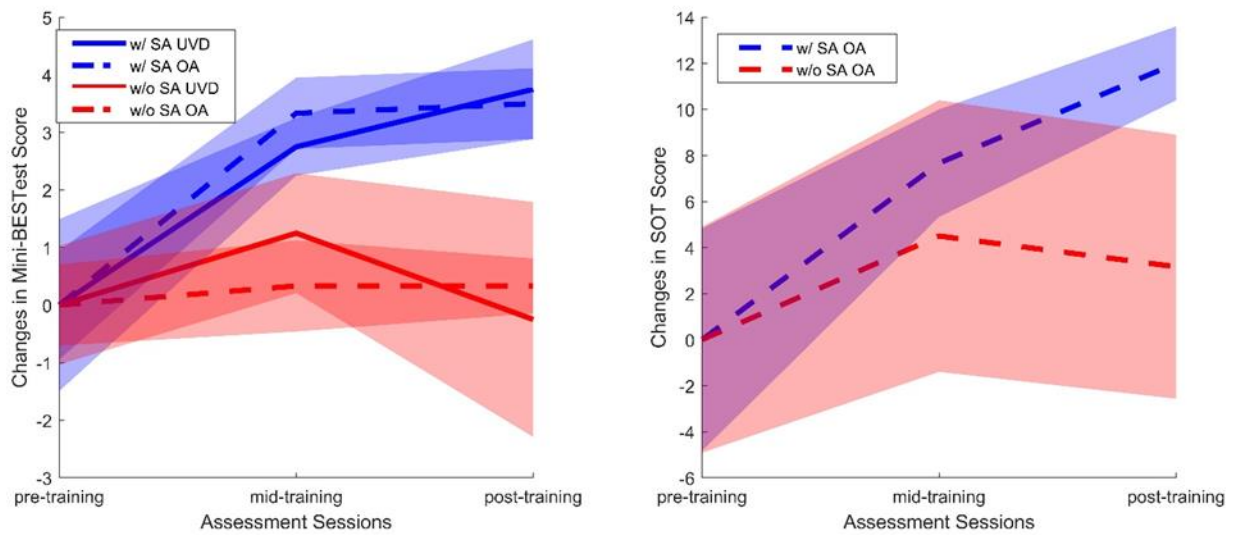


Figure 6.1: The Mini-BESTest and SOT scores for both the EG and CG at the pre-, mid- and post-training assessments. The scores were offset based on the averaged pre-training assessments.

The differences in improvements between the EG and CG for the Mini-BESTest and SOT scores were significant at the post-training assessments, but insignificant at the mid-training assessments, which may indicate that there was no difference in the improvement rates prior to the mid-training assessments. This finding seems to contradict the findings of short-term training controlled studies which showed that OA trained with SA significantly reduced body sway on selected standing exercises compared to OA trained without SA immediately after 30 min of training on the same standing exercises [12]. One explanation is that this dissertation used clinical outcomes to assess participants' functional ability rather than the outcomes of selected standing exercises. This dissertation's small sample size or the lack of balance assessments earlier than three weeks following the initiation of training could also explain the differences.

While this dissertation indicated that training with vibrotactile could lead to a higher functional plateau for the Mini-BESTest and SOT, future work is needed to confirm the differences in the improvement rates.

6.2. Mechanism of Sensory Augmentation

The mechanism(s) which enable(s) individuals using vibrotactile SA to improve their balance performance is not well understood [13]. The hypotheses (see Chapter 1, Section 1.3.4) include 1) sensory reweighting: improved balance performance results from the sensory reweighting mechanism, which aligns with the hypothesis of the traditional vestibular rehabilitation therapy [14], [15], [4], 2) “sixth” sense: CNS interprets SA as a new and distinct sensory input [13], 3) cognition: the additional information provided by SA is cognitively processed and acted upon, but is not used to adjust sensory weights by the CNS [13], 4) context-specific adaptation: a new sensorimotor function is developed only when SA is presented [13], and 5) combined volitional and non-volitional response: improvements result from both volitional and non-volitional responses [16]–[18].

This dissertation provided insight regarding the potential mechanism(s) of SA for balance training applications. Several studies, including this dissertation, have shown that balance improvements can be carried over to clinical outcome measures and may be retained for up to six months after training with SA provide. These studies provide support for the sensory reweighting hypothesis. One study noted that the balance performance improvements may be due to the effect of balance training itself and not the use of SA [19]. However, this dissertation finds that people who trained with vibrotactile SA showed greater improvement in a subset of their clinical outcome tests, e.g., Mini-BESTest and SOT scores, compared to those who trained without vibrotactile SA suggesting that if training improves balance performance, the use of vibrotactile SA may provide additional benefits.

This dissertation also found that both populations demonstrated greater reliance on their vestibular system inputs to maintain postural balance after balance training based on the

computerized dynamic posturography measurements. Although not all outcome measurements were significant, data trends from the two training studies in this dissertation suggest that the use of SA may promote the CNS to reweight its sensory inputs, leading to improved functional balance performance compared with balance training alone. Again, this dissertation's small sample size may account for the unobserved significance and a larger sample size is needed to validate the findings in this dissertation.

6.3. Machine Learning for Balance Rehabilitation

Steady improvements in wireless communications and wearable technology now allow researchers to obtain very accurate measurements, including those captured remotely, of body motion. Unfortunately, data analysis is still limited to simple time-domain and frequency-domain analysis. This dissertation explored the use of machine learning (ML) as an analytical tool to improve unsupervised balance rehabilitation (Chapter 5). Inspired by the positive findings of Chapter 5, we also proposed the potential use of ML as an analytical tool to provide meaningful SA during gait exercises in the section below.

6.3.1. Machine Learning for Unsupervised Balance Rehabilitation

The effectiveness of unsupervised in-home balance rehabilitation was enhanced when telerehabilitation was developed, primarily due to the availability of personal computers and the internet. Telerehabilitation delivers rehabilitation services, including assessments and intervention, remotely via telecommunication networks [20]. The expert, typically a physical therapist, and the exercising individual “talk” remotely throughout the training session [21]. For people with stroke, research has shown that telerehabilitation with virtual reality tasks led to

balance improvements similar to the improvements provided by conventional balance training in clinical settings [22]. Published research has demonstrated the effectiveness of telerehabilitation and the potential economic benefits compared to traditional rehabilitation programs [21], [23]. However, telerehabilitation can require longer consultation times with the expert and user via video conferencing, and increase the cost to the expert (time) and user (money) [21].

Recent developments of SA have improved unsupervised in-home balance rehabilitation. Video game-based in-home balance training uses balance platforms (e.g., Wii Fit balance board) and a display screen to provide visual SA. Video game-based in-home balance training has been shown to improve clinical measures after a minimum of five weeks of training [24]–[27]. Jorgensen et al. and Whyatt et al. demonstrated that OA improved their ABC scores and reduced body sway by using the Nintendo Wii board [24], [28]. However, these devices cannot be used in balance exercises that require closed eyes, head movements, altered stances, and different support surfaces.

Chapter 3 introduced a smartphone balance trainer, which measures trunk sway and provides vibrotactile SA as an alternative device. Instead of interacting via teleconference with the exerciser for an entire training session, the physical therapist makes suggestions based on the performance sent automatically by the smartphone. The drawback is that the current configuration of the smartphone balance trainer only allows the device to send step-out information and the user's self-assessments to the physical therapist, and the information may not align with the therapist's own observations. The decision to only send step-out and self-assessment data in this study was intentional because the body sway parameters (e.g., RMS sway, elliptical fit area, percentage time in one-degree zone as mentioned in Chapter 1 Section

1.1.4) would require additional interpretation (and therefore time) by the physical therapist and the minimum data set needed to support exercise progression decisions is unclear.

Chapter 5 explored a new approach to provide more reliable balance performance assessments when there was limited access to physical therapists. The ML approach is appealing because today's wearable sensors (i.e., small size, low weight, low cost) can collect large amounts of data which ML techniques can convert to easy-to-understand assessment ratings. Automatically generated ratings may enable physical therapists to provide quality remote suggestions regarding exercise progression. Although the agreements between the ML classifier (i.e., support vector machine) and physical therapist' ratings were shown to be better than the agreements between the physical therapist' ratings and participants' self-ratings, the classifier's accuracy remains limited. There are three possible reasons for the relatively low accuracy obtained during the study. First, there was low inter-rater or intra-rater reliability at some rating levels. This dissertation adapted a rating scale from a prior study [29], and only one physical therapist provided ratings. Second, the exercise data were taken from only sixteen participants. The ML algorithm may not be fully trained due to the lack of data. Third, trunk motion was used as the single input for the ML algorithm. The motion of multiple body segments measured by multiple sensors may be needed. Future work that addresses these limitations may lead to improved accuracy.

More automation of unsupervised in-home balance rehabilitation will be possible when the ML algorithms can provide comparable exercise progression recommendations based on the user's prior performance compared to in-person/live recommendations provided by a physical therapist. Although recommender systems have not been applied to balance and gait applications, recommender systems have been successfully used for movie (Netflix), music (Pandora), search

queries (Yelp), and online shopping (Amazon) applications [30]. By building a database with individuals' prior balance performance assessment ratings during in-clinic balance rehabilitation, the recommender system could suggest the next exercise after an exercise is completed. A pilot study, which used a limited amount of data and collaborative models (k-nearest neighbor [31] and singular value decomposition [32]), obtained an accuracy of ~47.5% for the ratings at a discrete scale of five, which was slightly higher than a guess-based case (25%). Therefore, a larger dataset and more complicated models should also be feasible.

6.3.2. Sensory Augmentation for Gait Tasks with Machine Learning Techniques

Vibrotactile SA was not provided during gait-based exercises in the studies described in Chapter 2 and Chapter 3 because the use of vibrotactile SA for locomotion is underdeveloped. Most studies of SA applications have addressed specific body segment(s) motion(s) such as the trunk and knee [33]–[40]. Since gait is more complex than standing balance, providing SA on only one segment could produce unexpected behaviors that negatively affect gait performance. For example, providing vibrotactile SA could reduce trunk sway, but also produce a slower gait during slow-paced walking and narrow stance walking, and stiffening in the coronal plane (i.e., rigid and awkward gait) [37]. Another study found that participants appeared to walk naturally after training to walk with a toe-in gait angle, but they were not necessarily comfortable [40].

Researchers have investigated the use of multiple vibrotactile SA displays for multiple segments [38]. However, longer potential reaction times are one of the shortcomings of this approach. Prior studies have indicated that reaction times to vibrotactile cues linearly increase with the number of vibrations during choice and recognition reaction time tasks [41]. Chapter 4 found that even for simple reaction time tasks, reaction times increased when the number of

possible stimulation locations increased. Given that the usual gait cycle is approximately 1 s, increased reaction times due to multiple displays would likely negatively affect the effectiveness of vibrotactile SA. With respect to SA display modality, visual feedback appears to be one of the best options (if not the best option) for gait-based activities from a reaction time standpoint since more information can be analyzed immediately via visual systems. Franz et al. used visual SA to enhance forward propulsion forces and push-off muscle activities during walking in healthy OA [42]. In their study, visual SA was provided via a large screen based on force plate data and electromyographic signals [42]. However, this type of set-up may not be practical for home-based balance training applications unless a heads-up display or similar technology is used.

In clinical settings, physical therapists observing the performance of gait tasks give real-time verbal feedback or specific instructions about a body segment or multiple body segments, but the feedback or instructions are not typically provided on a step-by-step basis. Instead, physical therapists tend to focus on the outcomes of gait performance (e.g., walking speed, walking coordination) based on full body observations. Since ML techniques are able to map high dimensional time series data to categorical or low dimensional outputs, an alternative to providing real-time SA cues to inform postural/gait parameter corrections on a step-by-step basis would be to use ML and the motion data collected by wearable sensors to “learn” the physical therapist’s assessments and provide intermittent feedback or knowledge of results. To date, the exploration of ML for gait analysis has primarily focused on classification of people with gait disorders or on gait pattern recognition [43]–[51]. The high classification accuracy of previously published studies (i.e., above 90%) and the results of Chapter 5 imply the feasibility of using ML to analyze gait/locomotion. Using ML techniques to map the time-series body motion to physical therapists’ feedback should be studied in greater detail.

6.4. Optimizing the Design of Vibrotactile Sensory Augmentation

While many studies, including the studies described in Chapter 2 and Chapter 3, have demonstrated balance improvements with the use of vibrotactile SA, the actual design of vibrotactile SA displays warrant more study. The section below describes some of the design parameters that should be considered.

6.4.1. Scheduling Sensory Augmentation

The studies described in Chapters 2 and 3 provided vibrotactile SA to participants for four of the six trials in order to enhance motor learning. In motor learning, feedback can be provided using full feedback schedules, reduced feedback frequency schedules, and self-controlled feedback schedules [52]–[58]. In a full feedback schedule, feedback is continually provided throughout the training process. However, negative effects may occur if the trainee becomes dependent on, or demands, a constant level of feedback [54]. In a reduced feedback frequency schedule, feedback is provided in some portion of the training. Several studies found that reducing feedback frequency (e.g., 50%, 67%) led to more effective learning than a full feedback schedule [52], [55], [56], but other studies noted that reduced feedback frequency did not result in better learning than a full feedback schedule [57]. In the case of augmented balance rehabilitation, most of the previously published studies, with the exception of the studies described in Chapters 2 and 3 of this dissertation, used full feedback schedules. Lim et al. found that healthy OA following a 2-week balance training protocol with SA and a full feedback schedule indicated few benefits compared to the 2-week balance training protocol without SA [58], unlike the study described in Chapter 3 of this dissertation which found significant improvements for a subset of the outcomes studied for OA who trained with SA compared with

those who trained without SA. It is possible that different feedback schedules account for the variations. However, it is not clear if the 67% feedback frequency schedule used in this dissertation is preferable to other options. The self-controlled feedback schedule, also called feedback as needed, only provides feedback upon demand. Research has shown that although the reduced frequency schedule can enhance learning in the complexity skills, the self-controlled feedback schedule is more useful for simple tasks [59]. Optimal scheduling for SA exercise programs remains to be determined.

6.4.2. Vibrotactile Activation Thresholds for Sensory Augmentation

Determining the feedback/SA activation thresholds for balance rehabilitation exercises is another important issue. Some researchers have used fixed thresholds (e.g., 1-degree zone) across all exercises and all participants [60]–[62]. The advantage of this approach is that it requires little effort to customize the thresholds for each individual and each exercise. However, fixed thresholds may provide limited SA cues for easy exercises or provide overwhelming SA cues for difficult exercises. Other researchers have used a percentage (e.g., 80%) of the maximum motion as the threshold, assuming that customized thresholds may optimize the training effects [12], [58], [63]–[65]. The thresholds are then updated after some period of training (e.g., after each session). The major disadvantage of this approach is the additional time required to determine the thresholds. Both approaches have shown to yield improvements in balance performance during real-time uses of SA. In this dissertation, the thresholds were customized for each exercise, and kept the same for all participants. The results from this customized activation threshold approach based on exercise type suggest that different thresholds for different exercises may increase the training's effectiveness. However, it is not clear whether

customizing thresholds can produce more benefits than fixed thresholds since the studies in this dissertation did not control for this factor.

In addition, for the customized threshold approach, it is not clear whether a narrow threshold (e.g., 30% of max body sway) or a wide threshold (90% of max body sway) is ideal for improving and retaining training. Training effect refers to the real-time improvements demonstrated when vibrotactile SA is provided, and retention effect refers to the improvements demonstrated when vibrotactile SA is removed after a period of training with vibrotactile SA. A narrow threshold, which theoretically provides more frequent cues resulting in individuals making more frequent postural corrections, can also make individuals more dependent on the cues and consequently reduce motor learning [66]. Furthermore, a large amount of information may ultimately be distracting and worsen performance [60]. A wide threshold, on the other hand, may allow individuals to produce too large of a postural movement before they cued to initiate a correction, resulting in a step out of position or a fall. Wide thresholds may not yield better training effects compared to narrow thresholds since sway is likely to be increased, but they may lead to better retention effects because individuals are challenged to rely on their own sensory inputs to initiate corrective movements (i.e., similar to the benefits of a reduced feedback frequency schedule). We are conducting a pilot study using narrow thresholds (0.8-degree zone) and wide thresholds (1.5-degree zone) for selected exercises to test the hypothesis that narrow thresholds lead to better training effects and wide thresholds lead to better retention effects.

6.4.3. Reaction Time to Vibrotactile Sensory Augmentation

The study described in Chapter 4 investigated the effects of various display design factors on reaction time to vibrotactile cues. The findings outlined below can provide a general guideline for designing vibrotactile displays.

First, the effect of tactor type (i.e., C-2 tactor, Tactaid and coin-style motors) on reaction time was less than 50 ms. Considering that reaction time to vibration at the trunk location is around 250 ms, and a control signal with a velocity component (i.e., tilt angle plus half tilt rate) is frequently used for comparison with the preset activation threshold, 50 ms could have little effect on the use of vibrotactile SA during standing balance tasks. Coin-style motors may be preferable in cost-sensitive scenarios (e.g., ~\$1 compared to the ~\$250 cost of the C-2 tactor). For gait tasks or applications with continuous vibrotactile SA [66], the C-2 tactor could be a more attractive choice since the trainee may need to respond to the vibration within 1 s depending on how the vibration is used.

Second, the stimulus-response compatibility effect is an important design factor. It is generally hypothesized that a positive relationship exists between reaction time and the distance between the stimulus location and the forehead (i.e., reaction time at the head location is faster than at the torso location) [67]. The results from this dissertation were aligned with this hypothesis when the reaction time task was simple and there was similar sensitivity at the stimulus locations. However, the hypothesis did not hold when a more complicated task (e.g., secondary task) was involved. Instead, the stimulus-response compatibility effect (i.e., the high stimulus-response set led to a faster reaction time [68]) outweighed the effect of the distance from the forehead. Ho et al. noted similar findings [69]. For standing balance tasks, the trunk is considered the response location for the vibration because the COM is located within the trunk.

Thus, the reaction time to the vibrotactile stimulus at the trunk location may be as fast as the head location or even faster than the head location given the stimulus-response compatibility.

Third, the combination of age and the presence of a secondary cognitive task had the greatest effects on reaction time compared to the other design factors analyzed. The presence of a secondary cognitive task was shown to increase the reaction time by as much as ~300 ms for healthy OA. The increased reaction time for the secondary task condition may require display designers to revisit the vibrotactile activation control signal to further minimize delays.

6.5. Implications

This dissertation has contributed to the study of balance rehabilitation by researching 1) the effects of balance training with SA, 2) the design of vibrotactile SA displays with respect to reaction time, and 3) exploratory ML algorithms for improving unsupervised balance rehabilitation.

First, this dissertation provided preliminary support for the use of SA as a rehabilitation tool to enhance balance rehabilitation, especially postural balance rehabilitation. Improvements in a subset of the clinical outcomes (i.e., SOT, Mini-BESTest, ABC) indicated that individuals showed functional improvements, instead of simply balance performance improvements on specific balance exercises after training with SA. The lack of SA provided during gait exercises could be the limiting factor for the lack of improvements observed in gait-related measures (e.g., gait speed, DGI, FGA). This dissertation also found that the use of SA could lead to higher functional recovery plateaus compared to balance rehabilitation alone. Higher functional plateaus suggest that individuals may attain a health stage higher than that attained with traditional balance rehabilitation. People with UVD showed more functional improvement in the ABC

scale, which is a subjective measure of confidence in performing various ambulatory activities without falling or experiencing unsteadiness. The significant improvement in confidence could lead to a decreased fear of falling and less depression and anxiety [71]. Further functional improvements by OA were reflected in their higher SOT and Mini-BESTest scores. Since both scores significantly relate to fall history, training with SA could improve quality of life and reduce medical costs related to falls for OA [72], [73]. The increase in vestibular reliance for both populations also suggested that SA could facilitate the reweighting of sensory inputs from the vestibular system by the central nervous system. Sensory reweighting could also explain the functional improvements observed. Considering that balance rehabilitation is therapeutic for people with UVD, and preventative for healthy OA, the findings may be generalizable to populations with other types of damaged or deteriorated vestibular function.

Second, the dissertation findings have implications for vibrotactile display design including customized activation thresholds for different exercises and display location to minimize reaction time delays due to a user's age or the presence of secondary tasks (e.g., head movements, cognitive task).

Third, to the best of our knowledge, this dissertation is the first to study the application of machine learning techniques to enhance in-home balance rehabilitation. The preliminary results suggest that remote balance performance assessments might be possible through the use of a wearable IMU and classification algorithms. Providing performance assessments during unsupervised in-home balance rehabilitation has the potential to improve the individual's compliance and motivation, and thus improve the effectiveness of in-home balance rehabilitation.

6.6. Limitations

The findings of Chapters 2, 3, and 5 were based on small sample sizes, which limited the use of statistical analysis and ML techniques. In Chapters 2 and 3, participants who trained with vibrotactile SA showed a higher increase in some clinical tests compared to those who trained without SA, but no significant effects were found. Statistical analyses of larger sample sizes are preferable for detecting significant effects. Chapter 2 mentioned the difficulties of recruiting people with UVD who were able to commit to six consecutive weeks of in-clinic training and three in-clinic assessments for up to six months after completing training. A smartphone balance trainer providing vibrotactile SA for in-home training may be a reasonable option for people who cannot attend in-clinic training. Chapter 3 also described the first attempt by participants to use vibrotactile SA independently within their homes.

An important limitation of this dissertation is that vibrotactile SA was not provided for gait exercises. As discussed in Section 6.3.2, the complexity of gait motion requires more research to understand how to provide meaningful SA to individuals performing gait exercises.

6.7. Future Work

6.7.1. Balance Training with Sensory Augmentation

6.7.1.1. Sample Size

As mentioned, the studies in this dissertation were exploratory in nature and therefore the sample size was limited. Future studies involving larger sample sizes should consider the performance of long-term balance training studies in participants' homes. While the smartphone balance trainer was designed for use by healthy OA, it could be re-designed for use by other populations with balance disorders. However, since participants with balance disorders could

present different symptoms (e.g., blurred vision) than OA participants, it is recommended to use the symptoms to inform the re-design (e.g., use of audio prompts such as Apple Siri instead of visual communication for the human-machine interface). Additionally, future work should consider collaborations with clinics or medical research groups that focus on populations with balance disorders in increase the sample size. Our successful collaboration with physical therapists at the University of Pittsburgh indicated that the vibrotactile SA device could easily be used by other physical therapists and research groups. One long-term study by Basta et al. recruited 105 participants through collaborations with clinicians at multiple locations [19]. Collaborations offer the advantages of increased sample sizes, greater diversity of populations with balance disorders, additional physical therapists and medical research personnel.

6.7.1.2. Understanding the Mechanism of Sensory Augmentation

The reported vestibular reliance in this dissertation was calculated from the SOT protocol. To truly understand the physiological changes after training with SA, techniques such as functional MRI (fMRI), electroencephalography (EEG), and electromyography (EMG) could be used. fMRI measures brain activity by detecting the changes associated with blood flow. Prior studies have found that electrotactile SA upregulated the visual sensitivity to optic flow in people with balance disorders [74], [75]. EEG records the brain's electrical activity. The EEG technique could help measure the activation of different functions (e.g., vestibular function) when balance exercise is performed during training with vibrotactile SA or after training with vibrotactile SA. EMG records the electrical activity produced by skeletal muscles. The EMG technique could help to understand how motor function (i.e., muscle activation) changes by using vibrotactile SA. A prior study found that muscle activity was reduced when vibrotactile SA was provided [76].

6.7.1.3. Sensory Augmentation for Gait Tasks with Machine Learning techniques

As mentioned in Section 6.3.2, to date, there has been limited evidence of effective SA strategies for improving gait performance in individuals with vestibular deficits and older adults [33]–[39]. Inspired by assessing balance performance for standing exercises using machine learning techniques, SA for gait tasks could resemble the verbal feedback given by physical therapists. For example, wearable sensors could record multiple body segment motions (e.g., trunk, leg, foot, arm, head) and ML techniques could be used to map the body motions to physical therapists’ assessments during gait tasks. SA could be given explicitly in the form of assessments (multi-class classifications), or implicitly by “telling” the individual if the previous step or the previous gait task was performed correctly (binary classification). It is possible that implicit feedback could lead to improved motor learning [77].

6.7.2. Design of Vibrotactile Sensory Augmentation

To improve the design of vibrotactile SA, three important design factors should be considered: activation thresholds for providing vibrotactile SA, schedules for providing vibrotactile SA, and control signals for providing vibrotactile SA. First, as mentioned, it is not clear whether the fixed threshold is as effective as the threshold customized to an individual’s balance performance for a specific exercise [12], [19], [58], [60]–[64]. It is also not clear whether a narrow threshold (e.g., 30% of the maximum sway) or a wide threshold (e.g., 80% of the maximum sway) improves motor learning. Both short-term and long-term training studies are needed. Second, the frequency of providing SA needs more study, because the majority of published studies have investigated training effects by considering SA as a real-time aid. Having provided preliminary support for using SA as a rehabilitation tool, we suggest studying the

frequency schedules of SA to enhance motor learning. Third, the current control signal (i.e., tilt angle plus half tilt angular velocity) should be studied by changing the weight of the tilt angular velocity based on our finding that reaction times to vibrotactile cues for different populations and/or the presence of second tasks varied from 100 ms to 300 ms.

6.7.3. Smart Balance Training

6.7.3.1. Smartphone Balance Trainer

Chapter 3 developed a smartphone balance trainer for real-time, in-home use by OA. Although use of this device led to balance improvements, the design needs further refinement. The iPod touch sensing unit could potentially be replaced with smaller wearable devices worn on the wrist (e.g., Apple Watch, Fitbit). A smaller sized, lightweight belt holding the necessary accessories of the current smartphone balance trainer could also be designed. Second, the current version of the smartphone balance trainer did not provide balance performance assessments to the individuals trained at home. It may be desirable to integrate the ML algorithms into the smartphone balance trainer to provide assessment ratings directly to the individual based on his/her body sway to further engage and motivate the individual.

6.7.3.2. Balance Exercise Assessment Rating

In Chapter 5, the labels (i.e., ground truth) used for the trained classifier were supplied by one physical therapist. The rating scale was adapted from a prior study, but not enough is known about the inter-rater reliability associated with the scale. Therefore, we suggest that several physical therapists assess the balance performance of exercises with different difficulty levels based on the rating scale. If the inter-rater reliability is low, the classifiers could be trained based

on the preferences of physical therapists or a new rating scale could be developed to obtain higher inter-rater reliability.

6.7.3.3. Smart In-home Balance Training

To improve in-home balance training in the absence of physical therapists, we propose the development of a recommender system that recommends exercises to an individual based on the history of the individual and the collective assessments of exercises previously performed by multiple individuals. The two major types of recommender system are collaborative filtering and content-based filtering. Collaborative filtering builds models from the user's past behavior (e.g., ratings, preferences, decisions) and other users' behaviors to predict the user's behavior on other unseen items [78]. Content-based filtering builds models from the properties of an item to recommend additional items with similar properties. Collaborative filtering uses two types of approaches: memory-based and model-based. The memory-based approach calculates the similarities between users or items to make recommendations, and the model-based approach uses models (e.g., Bayesian networks, singular value decompositions and Markov decision processes) to make recommendations [79].

If this recommender system is developed, we could use vibrotactile SA to provide real-time knowledge of performance, classification techniques to assess balance performance, and recommender system techniques to suggest the optimal exercise progression for in-home balance rehabilitation. Use of a smart in-home balance rehabilitation device could potentially benefit the increasingly aging population in the United States. Smart in-home balance rehabilitation could also improve the availability of physical therapists for people who require out-patient care, and

could potentially lower the cost of administering preventative and therapeutic treatments, lower fall risk, and decrease dependency on family members and caregivers.

6.8. References

- [1] S. L. Whitney and M. M. Rossi, “Efficacy of vestibular rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 33, no. 3, pp. 659–672, Jun. 2000.
- [2] J. L. Cowand, D. M. Wrisley, M. Walker, B. Strasnick, and J. T. Jacobson, “Efficacy of Vestibular Rehabilitation,” *Otolaryngol. Clin. North Am.*, vol. 118, pp. 49–54, 1998.
- [3] S. L. Whitney, A. H. Alghadir, and S. Anwer, “Recent Evidence About the Effectiveness of Vestibular Rehabilitation,” *Curr. Treat. Options Neurol.*, vol. 18, no. 3, pp. 1–15, 2016.
- [4] M. B. Badke, J. A. Miedaner, C. R. Grove, T. A. Shea, and G. M. Pyle, “Effects of vestibular and balance rehabilitation on sensory organization and dizziness handicap,” *Ann. Otol. Rhinol. Laryngol.*, vol. 114, no. 1 I, pp. 48–54, 2005.
- [5] A. Meli, G. Zimatore, C. Badaracco, E. De Angelis, and D. Tufarelli, “Vestibular rehabilitation and 6-month follow-up using objective and subjective measures,” *Acta Otolaryngol.*, vol. 126, no. 3, pp. 259–66, 2006.
- [6] M. E. Tinetti and C. S. Williams, “The Effect of Falls and Fall Injuries on Functioning in Community-Dwelling Older Persons,” *Journals Gerontol. Ser. A Biol. Sci. Med. Sci.*, vol. 53A, no. 2, pp. M112–M119, Mar. 1998.
- [7] K. Hauer, B. Rost, K. Rüttschle, O. Hedda, N. Specht, P. Bärtsch, P. Oster, and G. Schlierf, “Exercise Training for Rehabilitation and Secondary Prevention of Falls in Geriatric Patients With A History of Injurious Falls,” *J. Am. Geriatr. Soc.*, vol. 49, no. 1, pp. 10–20, 2001.

- [8] L. D. Gillespie, M. C. Robertson, W. J. Gillespie, C. Sherrington, S. Gates, L. M. Clemson, and S. E. Lamb, "Interventions for preventing falls in older people living in the community," in *Cochrane Database of Systematic Reviews*, no. 9, L. D. Gillespie, Ed. Chichester, UK: John Wiley & Sons, Ltd, 2012.
- [9] M. Lesinski, T. Hortobágyi, T. Muehlbauer, A. Gollhofer, and U. Granacher, "Effects of Balance Training on Balance Performance in Healthy Older Adults : A Systematic Review and Meta-analysis," *Sport. Med.*, vol. 45, no. 12, pp. 1721–38, Sep. 2015.
- [10] A. M. Myers, P. C. Fletcher, A. H. Myers, and W. Sherk, "Discriminative and Evaluative Properties of the Activities-specific Balance Confidence (ABC) Scale," *J. Gerontol. Med. Sci.*, vol. 53, no. 4, pp. 287–294, 1998.
- [11] C. L. Kao, L. K. Chen, C. M. Chern, L. C. Hsu, C. C. Chen, and S. J. Hwang, "Rehabilitation outcome in home-based versus supervised exercise programs for chronically dizzy patients," *Arch. Gerontol. Geriatr.*, vol. 51, no. 3, pp. 264–267, 2009.
- [12] J. R. Davis, M. G. Carpenter, R. Tschanz, S. Meyes, D. Debrunner, J. Burger, and J. H. J. Allum, "Trunk sway reductions in young and older adults using multi-modal biofeedback.," *Gait Posture*, vol. 31, no. 4, pp. 465–72, Apr. 2010.
- [13] K. H. Sienko, S. L. Whitney, W. J. Carender, and C. Wall III, "The role of sensory augmentation for people with vestibular deficits: real-time balance aid and/or rehabilitation device?," *J. Vestib. Res.*, vol. 27, no. 1, pp. 63–76, 2017.
- [14] B. N. Klatt, W. J. Carender, C. Lin, S. Alsubaie, C. R. Kinnaird, K. H. Sienko, and S. L. Whitney, "A Conceptual Framework for the Progression of Balance Exercises in Persons with Balance and Vestibular Disorders," *Phys. Med. Rehabil. - Int.*, vol. 2, no. 4, p. 1044, 2015.

- [15] C. D. Hall, S. J. Herdman, S. L. Whitney, S. P. Cass, R. A. Clendaniel, T. D. Fife, J. M. Furman, T. S. D. Getchius, J. A. Goebel, N. T. Shepard, and S. N. Woodhouse, *Vestibular Rehabilitation for Peripheral Vestibular Hypofunction: An Evidence-Based Clinical Practice Guideline*, vol. 40, no. April. 2016.
- [16] B. Lee, B. J. Martin, and K. H. Sienko, “Comparison of Non-Volitional Postural Responses Induced by Two Types of Torso Based Vibrotactile Stimulations,” pp. 195–198, 2012.
- [17] B.-C. Lee, A. Ho, B. J. Martin, and K. H. Sienko, “Effects of co-vibrotactile stimulations around the torso on non-volitional postural responses.,” *Conf. Proc. ... Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. IEEE Eng. Med. Biol. Soc. Annu. Conf.*, vol. 2012, pp. 6149–52, Jan. 2012.
- [18] B.-C. Lee, B. J. Martin, and K. H. Sienko, “The effects of actuator selection on non-volitional postural responses to torso-based vibrotactile stimulation.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 21, 2013.
- [19] D. Basta, M. Rossi-Izquierdo, A. Soto-Varela, M. E. Greters, R. S. Bittar, E. Steinhagen-Thiessen, R. Eckardt, T. Harada, F. Goto, K. Ogawa, and A. Ernst, “Efficacy of a vibrotactile neurofeedback training in stance and gait conditions for the treatment of balance deficits: a double-blind, placebo-controlled multicenter study.,” *Otol. Neurotol.*, vol. 32, no. 9, pp. 1492–9, Dec. 2011.
- [20] M. Janssen, R. Stokroos, J. Aarts, R. van Lummel, and H. Kingma, “Salient and placebo vibrotactile feedback are equally effective in reducing sway in bilateral vestibular loss patients.,” *Gait Posture*, vol. 31, no. 2, pp. 213–7, Feb. 2010.
- [21] R. B. Burns, D. Crislip, P. Daviou, A. Temkin, S. Vesmarovich, J. Anshutz, C. Furbish,

- and M. L. Jones, "Using telerehabilitation to support assistive technology," *Assist. Technol.*, vol. 10, no. 2, pp. 126–133, 1998.
- [22] D. Kairy, P. Lehoux, C. Vincent, and M. Visintin, "A systematic review of clinical outcomes, clinical process, healthcare utilization and costs associated with telerehabilitation," *Disabil. Rehabil.*, vol. 31, no. 6, pp. 427–447, 2009.
- [23] I. Cikajlo, M. Rudolf, N. Goljar, H. Burger, and Z. Matjačić, "Telerehabilitation using virtual reality task can improve balance in patients with stroke.," *Disabil Rehabil*, vol. 34, no. 1, pp. 13–18, 2012.
- [24] M. Tousignant, P. Boissy, H. Corriveau, and H. Moffet, "In home telerehabilitation for older adults after discharge from an acute hospital or rehabilitation unit: A proof-of-concept study and costs estimation.," *Disabil. Rehabil. Assist. Technol.*, vol. 1, no. 4, pp. 209–216, 2006.
- [25] M. G. Jorgensen, U. Laessoe, C. Hendriksen, O. B. F. Nielsen, and P. Aagaard, "Efficacy of nintendo wii training on mechanical leg muscle function and postural balance in community-dwelling older adults: A randomized controlled trial," *Journals Gerontol. - Ser. A Biol. Sci. Med. Sci.*, vol. 68, no. 7, pp. 845–852, 2013.
- [26] S. Roopchand-Martin, R. McLean, C. Gordon, and G. Nelson, "Balance Training with Wii Fit Plus for Community-Dwelling Persons 60 Years and Older.," *Games Health J.*, vol. 4, no. 3, pp. 247–52, 2015.
- [27] A. S. Fu, K. L. Gao, A. K. Tung, W. W. Tsang, and M. M. Kwan, "Effectiveness of Exergaming Training in Reducing Risk and Incidence of Falls in Frail Older Adults with a History of Falls," *Arch. Phys. Med. Rehabil.*, vol. 96, no. 12, pp. 2096–2102, 2015.
- [28] L. Prosperini, D. Fortuna, C. Gianni, L. Leonardi, M. R. Marchetti, and C. Pozzilli,

- “Home-Based Balance Training Using the Wii Balance Board,” *Neurorehabil. Neural Repair*, vol. 27, no. 6, pp. 516–525, 2013.
- [29] C. Whyatt, N. A. Merriman, W. R. Young, F. N. Newell, and C. Craig, “A Wii bit of fun: A novel platform to deliver effective balance training to older adults,” *Games Health J.*, vol. 4, no. 6, pp. 1–11, 2015.
- [30] P. B. Kantor, L. Rokach, F. Ricci, and B. Shapira, *Recommender systems handbook*. 2011.
- [31] S. A. Dudani, “The Distance-Weighted k-Nearest-Neighbor Rule,” *IEEE Trans. Syst. Man Cybern.*, vol. SMC-6, no. 4, pp. 325–327, 1976.
- [32] Y. Koren, R. Bell, and C. Volinsky, “Matrix Factorization Techniques for Recommender Systems,” *Computer (Long Beach, Calif.)*, vol. 42, no. 8, pp. 42–49, 2009.
- [33] P. B. Shull, W. Jirattigalachote, M. a Hunt, M. R. Cutkosky, and S. L. Delp, “Quantified self and human movement: A review on the clinical impact of wearable sensing and feedback for gait analysis and intervention.,” *Gait Posture*, vol. 40, no. 1, pp. 11–19, May 2014.
- [34] C. Wall, “Application of vibrotactile feedback of body motion to improve rehabilitation in individuals with imbalance.,” *J. Neurol. Phys. Ther.*, vol. 34, no. 2, pp. 98–104, Jun. 2010.
- [35] L. L. Verhoeff, C. G. C. Horlings, L. J. F. Janssen, S. a Bridenbaugh, and J. H. J. Allum, “Effects of biofeedback on trunk sway during dual tasking in the healthy young and elderly.,” *Gait Posture*, vol. 30, no. 1, pp. 76–81, Jul. 2009.
- [36] F. B. Horak, M. Dozza, R. Peterka, L. Chiari, and C. Wall, “Vibrotactile biofeedback improves tandem gait in patients with unilateral vestibular loss.,” *Ann. N. Y. Acad. Sci.*, vol. 1164, pp. 279–81, May 2009.
- [37] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “The effect of vibrotactile

- feedback on postural sway during locomotor activities.,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 93, Jan. 2013.
- [38] P. B. Shull, K. L. Lurie, M. R. Cutkosky, and T. F. Besier, “Training multi-parameter gaits to reduce the knee adduction moment with data-driven models and haptic feedback.,” *J. Biomech.*, vol. 44, no. 8, pp. 1605–9, May 2011.
- [39] A. V Dowling, D. S. Fisher, and T. P. Andriacchi, “Gait modification via verbal instruction and an active feedback system to reduce peak knee adduction moment.,” *J. Biomech. Eng.*, vol. 132, no. 7, p. 71007, Jul. 2010.
- [40] P. B. Shull, R. Shultz, A. Silder, J. L. Dragoo, T. F. Besier, M. R. Cutkosky, and S. L. Delp, “Toe-in gait reduces the first peak knee adduction moment in patients with medial compartment knee osteoarthritis.,” *J. Biomech.*, vol. 46, no. 1, pp. 122–8, Jan. 2013.
- [41] W. E. Hick, “On the rate of gain of information,” *Q. J. Exp. Psychol.*, vol. 4, no. 1, pp. 11–26, 1952.
- [42] J. R. Franz, M. Maletis, and R. Kram, “Real-time feedback enhances forward propulsion during walking in old adults.,” *Clin. Biomech. (Bristol, Avon)*, vol. 29, no. 1, pp. 68–74, Jan. 2014.
- [43] R. Begg and J. Kamruzzaman, “A machine learning approach for automated recognition of movement patterns using basic, kinetic and kinematic gait data.,” *J. Biomech.*, vol. 38, no. 3, pp. 401–8, Mar. 2005.
- [44] A. Mannini, D. Trojaniello, A. Cereatti, and A. M. Sabatini, “A machine learning framework for gait classification using inertial sensors: Application to elderly, post-stroke and huntington’s disease patients,” *Sensors (Switzerland)*, vol. 16, no. 1, 2016.
- [45] G. Williams, D. Lai, A. Schache, and M. E. Morris, “Classification of Gait Disorders

- Following Traumatic Brain Injury,” *J. Head Trauma Rehabil.*, vol. 30, no. 2, pp. E13--E23, 2015.
- [46] L. Bao and S. S. Intille, “Activity Recognition from User-Annotated Acceleration Data,” in *Pervasive Computing*, 2004, pp. 1–17.
- [47] A. Mannini and A. M. Sabatini, “Machine learning methods for classifying human physical activity from on-body accelerometers.,” *Sensors (Basel)*, vol. 10, no. 2, pp. 1154–75, Jan. 2010.
- [48] R. Caldas, M. Mundt, W. Potthast, F. Buarque de Lima Neto, and B. Markert, “A systematic review of gait analysis methods based on inertial sensors and adaptive algorithms,” *Gait Posture*, vol. 57, no. June, pp. 204–210, 2017.
- [49] R. K. Begg, M. Palaniswami, and B. Owen, “Support vector machines for automated gait classification,” *IEEE Trans. Biomed. Eng.*, vol. 52, no. 5, pp. 828–838, 2005.
- [50] R. LeMoyne, F. Heerinckx, T. Aranca, R. De Jager, T. Zesiewicz, and H. J. Saal, “Wearable body and wireless inertial sensors for machine learning classification of gait for people with Friedreich’s ataxia,” *BSN 2016 - 13th Annu. Body Sens. Networks Conf.*, pp. 147–151, 2016.
- [51] T. Nakano, B. T. Nukala, S. Zupancic, A. Rodriguez, D. Y. C. Lie, J. Lopez, and T. Q. Nguyen, “Gaits classification of normal vs. patients by wireless gait sensor and Support Vector Machine (SVM) classifier,” *2016 IEEE/ACIS 15th Int. Conf. Comput. Inf. Sci. ICIS 2016 - Proc.*, 2016.
- [52] P. M. van Vliet and G. Wulf, “Extrinsic feedback for motor learning after stroke: What is the evidence?,” *Disabil. Rehabil.*, vol. 28, no. 13–14, pp. 831–840, 2006.
- [53] S. K. Subramanian, C. L. Massie, M. P. Malcolm, and M. F. Levin, “Does provision of

- extrinsic feedback result in improved motor learning in the upper limb poststroke? A systematic review of the evidence.,” *Neurorehabil. Neural Repair*, vol. 24, no. 2, pp. 113–124, 2010.
- [54] R. A. Schmidt, “Frequent augmented feedback can degrade learning: Evidence and interpretations,” in *Tutorials in motor neuroscience*, Dordrecht, The Netherlands Springer, 1991, pp. 59–75.
- [55] C. J. Winstein and R. a. Schmidt, “Reduced frequency of knowledge of results enhances motor skill learning.,” *J. Exp. Psychol. Learn. Mem. Cogn.*, vol. 16, no. 4, pp. 677–691, 1990.
- [56] Q. Lai and C. H. Shea, “Generalized motor program (GMP) learning: Effects of reduced frequency of knowledge of results and practice variability,” *J. Mot. Behav.*, vol. 30, no. 1, pp. 51–59, 1998.
- [57] C. Winstein, A. Merians, and K. Sullivan, “Motor learning after unilateral brain damage,” *Neuropsychologia*, vol. 37, no. 8, p. 975--87., 1999.
- [58] S. B. Lim, B. C. Horslen, J. R. Davis, J. H. J. Allum, and M. G. Carpenter, “Benefits of multi-session balance and gait training with multi-modal biofeedback in healthy older adults,” *Gait Posture*, vol. 47, pp. 10–17, 2016.
- [59] R. Sigrist, G. Rauter, R. Riener, and P. Wolf, “Self-Controlled Feedback for a Complex Motor Task,” in *BIO Web of Conferences*, 2011, vol. 1, p. 84.
- [60] K. H. Sienko, M. D. Balkwill, L. I. E. Oddsson, and C. Wall, “Effects of multi-directional vibrotactile feedback on vestibular-deficient postural performance during continuous multi-directional support surface perturbations,” *J. Vestib. Res.*, vol. 18, pp. 273–285, 2008.

- [61] C. Wall III and E. Kentala, "Control of sway using vibrotactile feedback of body tilt in patients with moderate and severe postural control deficits.," *J. Vestib. Res.*, vol. 15, no. 5–6, pp. 313–325, 2005.
- [62] C. Kinnaird, J. Lee, W. J. Carender, M. Kabeto, B. Martin, and K. H. Sienko, "The effects of attractive vs. repulsive instructional cuing on balance performance," *J. Neuroeng. Rehabil.*, vol. 13, no. 1, p. 1, 2016.
- [63] J. H. J. Allum, M. G. Carpenter, B. C. Horslen, J. R. Davis, F. Honegger, K.-S. Tang, and P. Kessler, "Improving impaired balance function: real-time versus carry-over effects of prosthetic feedback.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2011, pp. 1314–8, Jan. 2011.
- [64] M. Rossi-Izquierdo, A. Ernst, A. Soto-Varela, S. Santos-Pérez, A. Faraldo-García, A. Sesar-Ignacio, and D. Basta, "Vibrotactile neurofeedback balance training in patients with Parkinson's disease: reducing the number of falls.," *Gait Posture*, vol. 37, no. 2, pp. 195–200, Feb. 2013.
- [65] A. Domingo and D. P. Ferris, "The effects of error augmentation on learning to walk on a narrow balance beam.," *Exp. brain Res.*, vol. 206, no. 4, pp. 359–70, Oct. 2010.
- [66] B.-C. Lee, A. Fung, and T. A. Thrasher, "The Effects of Coding Schemes on Vibrotactile Biofeedback for Dynamic Balance Training in Parkinson's Disease and Healthy Elderly Individuals," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 1, pp. 153–160, 2018.
- [67] V. Harrar and L. R. Harris, "Simultaneity constancy: detecting events with touch and vision," *Exp. Brain Res.*, vol. 166, no. 3–4, pp. 465–473, 2005.
- [68] S. Kornblum, T. Hasbroucq, and A. Osman, "Dimensional overlap: cognitive basis for stimulus-response compatibility--a model and taxonomy.," *Psychol. Rev.*, vol. 97, no. 2,

- pp. 253–270, 1990.
- [69] C. Ho and C. Spence, “Effectively responding to tactile stimulation: Do homologous cue and effector locations really matter?,” *Acta Psychol. (Amst)*, vol. 151, pp. 32–39, 2014.
- [70] C. Wall III and E. Kentala, “Effect of displacement, velocity, and combined vibrotactile tilt feedback on postural control of vestibulopathic subjects,” *J. Vestib. Res.*, vol. 20, no. 1, pp. 61–9, Jan. 2010.
- [71] K. M. C. Talley, J. F. Wyman, and C. R. Gross, “Psychometric properties of the activities-specific balance confidence scale and the survey of activities and fear of falling in older women,” *J. Am. Geriatr. Soc.*, vol. 56, no. 2, pp. 328–333, 2008.
- [72] S. L. Whitney, G. F. Marchetti, and A. I. Schade, “The relationship between falls history and computerized dynamic posturography in persons with balance and vestibular disorders,” *Arch. Phys. Med. Rehabil.*, vol. 87, no. 3, pp. 402–407, 2006.
- [73] C. S. L. Tsang, L.-R. Liao, R. C. K. Chung, and M. Y. C. Pang, “Psychometric Properties of the Mini- Balance Evaluation Systems Test (Mini-BESTest) in Community- Dwelling Individuals With Chronic Stroke,” *Phys. Ther.*, vol. 93, no. 8, pp. 1102–15, 2013.
- [74] J. C. Wildenberg, M. E. Tyler, Y. P. Danilov, K. A. Kaczmarek, and M. E. Meyerand, “Sustained cortical and subcortical neuromodulation induced by electrical tongue stimulation,” *Brain Imaging Behav.*, vol. 4, no. 3–4, pp. 199–211, 2010.
- [75] J. C. Wildenberg, M. E. Tyler, Y. P. Danilov, K. A. Kaczmarek, and M. E. Meyerand, “High-resolution fMRI detects neuromodulation of individual brainstem nuclei by electrical tongue stimulation in balance-impaired individuals,” *Neuroimage*, vol. 56, no. 4, pp. 2129–2137, 2011.
- [76] F. Honegger, I. M. A. Hillebrandt, N. G. A. van den Elzen, K.-S. Tang, and J. H. J. Allum,

- “The effect of prosthetic feedback on the strategies and synergies used by vestibular loss subjects to control stance,” *J. Neuroeng. Rehabil.*, vol. 10, no. 1, p. 115, 2013.
- [77] R. Richards, J. C. van den Noort, J. Dekker, and J. Harlaar, “Gait Retraining With Real-Time Biofeedback to Reduce Knee Adduction Moment: Systematic Review of Effects and Methods Used,” *Arch. Phys. Med. Rehabil.*, vol. 98, no. 1, pp. 137–150, 2016.
- [78] J. B. Schafer, D. Frankowski, J. Herlocker, and S. Sen, “Collaborative Filtering Recommender Systems,” *Int. J. Electron. Bus.*, vol. 4321, no. 1, pp. 291–324, 2007.
- [79] Y. Koren, “The bellkor solution to the netflix grand prize,” *Netflix Prize Doc.*, no. August, pp. 1–10, 2009.