Perturbation-based Detection and Prosthetic Correction of Vestibulopathic Gait

by

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ABSTRACT

While being able to balance is something most of us take for granted, each year approximately 400,000 Americans are diagnosed with a balance disorder. In order to prevent fall-related injuries due to postural instability, it is important to create both diagnosis techniques so that therapy can be applied before a fall occurs and devices which can aid the balance-impaired population. The aims of this research are twofold: 1) to develop metrics that quantify the locomotor stability of individuals with reduced vestibular function and 2) to assess the capability of a noninvasive vibrotactile balance prosthesis for improving postural and gait stability.

The clinical standards of practice for assessing vestibular deficiency include testing postural stability while standing but not during locomotion. This research examines one prospective locomotor-based technique involving the analysis of postural recovery from controlled surface perturbations. The research also investigates the use of a novel wearable vibrotactile sensory substitution device for enhanced postural and locomotor stability. The balance prosthesis is composed of an inertial motion-sensing system mounted on the lower back, a vibrotactile display worn around the torso, and a computer controller. It can serve as a permanent or temporary replacement of motion cues, a tool for vestibular rehabilitation, or an additional sensory channel for military troops, pilots, and astronauts.

This research demonstrates that well-compensated vestibulopathic patients can be differentiated from young and age-matched controls during over ground locomotion based on step width variability. Prior to this research, unilateral and bilateral vestibulopathic patients donning the vibrotactile balance prosthesis have demonstrated increased postural stability during single-axis support surface perturbations using single-axis sway information. This work shows that multi-directional vibrotactile tilt feedback reduces postural sway during multi-directional support surface perturbations, and has both short- and long-term effects on increasing postural stability. Finally, this research demonstrates for the first time that medial-lateral (M/L) tilt feedback can be used by balance-deficient subjects to reduce factors associated with fall risk (M/L tilt and M/L step width variability) during various locomotor tasks.

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THESIS DEFENSE

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To my parents

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(to be submitted to the Journal of Vestibular Research, 2006)

III. Assessment of a vibrotactile sensory substitution device for improving medial-lateral gait stability
(to be submitted to the Journal of Vestibular Research, 2006)

ABBREVIATIONS

ABC	Activities-Specific Balance Confidence
ADL	Activities of Daily Living
A/P	Anterior-Posterior
BVH	Bilateral Vestibular Hypofunction
DHI	Dizziness Handicap Index
CDP	Computerized Dynamic Posturography
COG	Center of Gravity
COP	Center of Pressure
CT	Continuous
DT	Discrete
ENG	Electronystagmography
MCT	Motor Control Test
M/L	Medial-Lateral
NT	No Tactors
RMS	Root-Mean-Square
RVR	Reduced Vestibular Response
SOT	Sensory Organization Test
SWV	Step Width Variability
VOR	Vestibuloocular Reflex

INTRODUCTION

Postural imbalance can result from various vestibular (central and peripheral), neurological, orthopedic and vascular disorders, as well as sensory conflicts, head injuries, infections, medications, aging, and space flight [1]. In America alone, it is estimated that more than 40% of the population will seek medical attention at least once for dizziness [2]. Balance disorders are a major cause of fall-related injuries to the elderly, resulting in patient care costs exceeding $8 billion per year [1]. Of particular concern are statistics regarding post-fall mortality rates: one study of nursing home residents has shown the mortality rate to more than double in the year following a fall [3].

The aims of this research are to develop metrics that quantify the locomotor stability of individuals with reduced vestibular function and to assess the capability of a noninvasive sensory substitution device (also referred to as a vibrotactile balance prosthesis) for improving postural and locomotor stability. Vestibular-deficient patients and age-matched controls will serve as the primary subject populations.

The clinical standards of practice for assessing vestibular deficiency include tests of postural stability while standing but not during locomotion. One prospective locomotor technique involves the analysis of postural recovery from controlled perturbations during normal gait [4]. To develop this technique, a custom-built moveable balance disturber platform (Figure 1-1) was created to facilitate investigation of the fundamental differences in postural and gait recovery to surface perturbations of normal and vestibulopathic subjects [5]. This platform can be automatically or manually triggered to deliver small surface perturbations varying in direction and magnitude to a subject during any phase of the gait cycle. To date, we have shown that individuals with subtle vestibulopathies require more steps to recover from a surface perturbation and make larger medial-lateral foot placements immediately following the perturbation compared to healthy controls [4-7].

Existing therapies for balance disorders include pharmacological treatments, balance rehabilitation, surgical procedures, and balance aids such as canes, walkers, and wheelchairs. The potential benefits of both implantable and non-implantable balance prostheses are currently being explored [8-12]. Both forms of prostheses will likely continue to be developed in parallel given that the National Institutes of Health NIDCD Workshop on electrical stimulation outcome supported research in both arenas - stating that an implant alone is not a complete solution and that a sensory substitution device complements the implant. Non-implantable prostheses such as vibrotactile display of body tilt [11, 13], surface electrode stimulation of the vestibular nerve, electric currents applied to the tongue [8, 9], and audio feedback offer varying degrees of non-invasive self-motion cues. Such devices can serve as a permanent or temporary replacement of motion cues, a tool for vestibular/balance rehabilitation, or an additional sensory channel for military troops, pilots, and astronauts. To date, we have demonstrated increased postural stability for unilateral and bilateral vestibulopathic patients donning a vibrotactile balance prosthesis during computerized posturography experiments [11, 13-16]. The balance prosthesis device, which senses body tilt, consists of a motion-sensing system mounted on the lower back of the subject, a vibrotactile display, and a laptop computer (or wearable watch) with analog and digital interfaces.

Both controlled perturbation during locomotion and sensory substitution in the form of the vibrotactile balance prosthesis described above could potentially be incorporated into current rehabilitation programs.

RESEARCH AIMS

The primary focus of this research is twofold: 1) to develop metrics that quantify the locomotor stability of individuals with reduced vestibular function, and 2) to assess the capability of a noninvasive sensory substitution device for improving postural and locomotor stability.

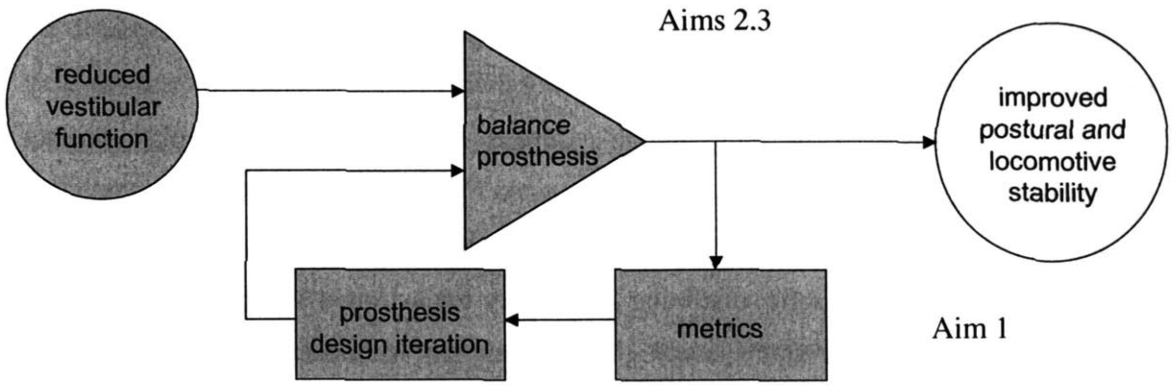

Figure 1. Flow chart of research specific aims. The metrics developed in aim 1 are used to evaluate and refine the design of the prosthesis developed in aims 2 and 3.

Specific Aim 1: Quantification of vestibulopathic gait

Gait instability is characteristic of vestibulopathic patients, astronauts/cosmonauts returning from long-duration space flight, and the elderly. Conventional clinical techniques used to diagnose reduced vestibular function include balance tests during non-perturbed and perturbed standing. This work aims to develop metrics that quantify reduced vestibular function by instead analyzing postural recovery from controlled perturbations during normal gait, which may lead to increased accuracy in identifying vestibulopathies. This research completes the characterization of the medial-lateral (M/L) stability of vestibulopathic patients and healthy young and age-matched controls supported by the National Space and Biomedical Research Institute.

<u>Develop post-processing techniques to identify heel strike events during locomotion</u>

Heel strike and toe off events are gait cycle markers used to identify single and double support phase. The heel strike event is commonly detected with switches or pressure sensors on shoes, force plates, or contact mats placed on the ground. Since such systems may not always be available or applicable to a given study, it is desirable to indirectly detect these events. Heel strike identification will be accomplished using only kinematics and limited force plate data [17]. Single support and double support phase will be estimated based on the heel strike values. We hypothesize that heel strike events can be accurately determined (within 1 sample) without the use of foot switches for both vestibulopathic and healthy individuals during paced gait trials [17].

Characterize the recovery of vestibulopathic subjects and age-matched controls to surface perturbations during locomotion

Postural recovery from controlled perturbations during normal gait will be evaluated as a potential diagnostic of subtle vestibular deficiency for astronauts returning from space flight. Following long-duration space flight, astronaut gait is characterized by exaggerated medial-lateral foot placement, trunk shifts to the side of the supporting leg, and failure to maintain the intended path [18]. Despite returning to preflight postural testing baselines, Shuttle and Mir crewmembers continue to report symptoms indicative of postural instability up to several months following landing. A more sensitive test than computerized dynamic posturography is desired to capture the slow recovery of sensorimotor function.

Well-compensated vestibulopathic patients who tested as normal in computerized dynamic posturography were used as analogues for post flight astronauts. Young healthy individuals were used to represent a preflight astronaut population. Wall et al. [5] showed that well-compensated vestibulopathic subjects had significantly greater changes in their medial-lateral foot placements compared to the young healthy controls; their recovery step following lateral and medial perturbation was widened and narrowed, respectively. These well-compensated vestibulopathic subjects also required a greater number of steps to return to normal pre-perturbation gait baselines in response to surface perturbations [5]. Differences in the sternum and head accelerations between the two groups were not as consistent. However, there was a trend toward greater response deviations in the vestibulopathic group for all of the surface perturbation types tested.

One significant independent variable that remains to be examined is the effect of age. Numerous studies have shown that posture and gait stability are negatively affected by age [19-31]. It is possible that some of the effects we observed in our perturbation study were due in part to age and not purely the vestibulopathy. Age-matched controls will be tested using an identical perturbation protocol. All previously published metrics [4-6, 17] will be calculated for the age-matched controls and compared to the well-compensated vestibulopathic subjects' values. We hypothesize that the differences previously observed between the well-compensated vestibulopathic patients and young healthy controls are not age-dependent. We therefore expect to see similar differences in medial-lateral foot placements as well as the number of steps required to recover from surface perturbations between the well-compensated vestibulopathic patients and age-matched controls.

Additionally, the relative roll between the head, trunk, and pelvis, and average medial-lateral acceleration of the head, sternum, and pelvis during estimated single and double support phases will be compared between the two groups in order to identify further diagnostic measures.

<u>Determine whether vestibulopathic patients can be distinguished from healthy individuals during non-perturbed gait</u>

Although quantifiable differences exist between the responses of vestibulopathic patients and young controls to surface perturbations during gait, a simpler and less provocative test may be more desirable for some severely balance-compromised individuals such as vestibulopathic patients and long-duration post flight astronauts. Sensorimotor functionality varies among patients status post vestibular surgery and astronauts immediately following landing; surface perturbations may not be tolerable or even possible post-op or in the initial post flight testing days. Therefore, if a simple walking test were: 1) capable of indicating vestibular function, and 2) able to be consistently performed throughout the recovery stages of a surgical procedure or long-duration space flight, it would in many cases be preferable to a surface perturbation protocol.

This study seeks to compare the well-compensated vestibulopathic and age-matched controls by examining basic parameters of medial-lateral stability during the non-perturbed non-paced and paced-gait trials to determine whether or not one can distinguish between the two populations without needing to apply a surface perturbation. Medial-lateral stability will be quantified using kinematics to calculate various frontal plane measures. We hypothesize that the perturbation trials will elicit greater differences between the two groups compared to the non-perturbation trials because patients with subtle vestibulopathies have developed sufficient compensatory strategies to cope with the challenges of straight and level locomotion. Unexpected controlled surface perturbations however, are likely to reveal sensorimotor problems; this technique is somewhat analogous to using impulse responses to characterize the dynamics of linear systems.

Compare sampling events for frontal plane measures

Sampling, the selection of one or more events that occur only once per gait cycle, is a common technique used to study locomotor kinematics. There are numerous events throughout the gait cycle that could serve as sampling points for clinical locomotor diagnostics. The sampling event that shows the greatest difference between the control and vestibulopathic patient populations is desirable because it focuses attention on the most likely indicator of reduction in sensorimotor functionality. The most favorable sampling events will be determined for each of the measures developed in Specific Aim 2. Distinct sampling events such as heel strike, anterior-posterior shank crossing, single support phase, and double support phase will be examined. Perturbation trials will be reanalyzed using the heel strike and single and double support phase sampling events (previous analyses [4-6] only used the anterior-posterior shank crossing event). We hypothesize that the single support phase will be the best sampling event for comparing the measures derived in Specific Aim 2 because it captures the least stable dynamics during the gait cycle.

Determine if systematic differences exist between vestibulopathic patients with left and right-sided lesions

Vestibulopathic patients with left and right-sided lesions have directional tilt differences following roll perturbations during quiet standing [32]. Specifically, they lean towards the contra-lateral lesion side. Patients with acute vestibular neuritis show a direction-specific deviation of gait towards their affected ear when instructed to close their eyes and walk slowly [33]. However, when asked to increase their walking speed, they straighten out their path. Preliminary analyses suggest that there is not a statistically significant difference in head, sternum, or pelvis roll during single support phase (unperturbed locomotion) between left and right-sided lesion patients. Perturbations, however, are more likely to elicit a difference in the roll response if such a difference exists. Roll of the head, sternum, and pelvis during heel strike, anterior-posterior shank crossing, single and double support phase will be calculated for both small and large forward right and backward left perturbations. Additionally, peak roll and time to peak roll will be examined during single and double support phase. We hypothesize that if a difference between vestibulopathic patients with left and right-sided lesions exists, it will most likely be detectable by observing the head and sternum roll dynamics in the first two steps following a surface perturbation.

Specific Aim 2: Assessment of multi-directional vibrotactile feedback on postural performance during quiet stance and multi-directional surface perturbations

Research efforts are underway to develop sensory substitution devices to supplement and/or permanently replace motion cues. Modes of delivering postural tilt and orientation include electrotactile, vibrotactile, and auditory biofeedback. The ability of a sensory substitution device to improve postural stability along a single-axis, in the anterior-posterior direction, has been validated for electrotactile, vibrotactile and auditory biofeedback varieties. Single-axis vibrotactile feedback has been shown to significantly reduce the root-mean-square (RMS) sway in vestibulopathic patients during single-axis perturbation [14]. The presence of single-axis feedback in the A/P direction has also been shown to reduce tilt in the M/L direction [34]. The obvious next step in terms of supporting the development of balance prosthesis for eventual commercial application as both a balance aid and rehabilitation tool is to determine whether balance-compromised individuals can derive benefit from a device which supplies complete information about their body orientation. To this end, this research examines the effect of multi-directional vibrotactile biofeedback on postural sway performance during multi-directional surface perturbations and investigates whether an optimal device configuration exists for providing biofeedback during perturbed stance.

The vibrotactile balance prosthesis studied here consists of a motion-sensing system mounted on the lower back of the subject, a vibrotactile display, and a laptop with analog and digital interfaces [11, 13, 14, 16] (Figure 1-2). The inertial motion-sensing system is composed of microelectromechanical (MEMS) gyroscopes that sense angular rate and MEMS accelerometers that sense linear accelerations [11]. The gyroscope and accelerometer signals are processed to obtain a tilt angle estimate accurate to within 2 milliradians over a 0 to 10 Hz bandwidth. Tilt estimates are haptically displayed in the form of vibrations on the subject's torso by three rows of tactors; performance in a modified version of the manual control critical tracking task was not appreciably improved when the prosthesis was equipped with more than three rows of tactors [15]. Tactor firing patterns are set on an individual basis based on the subject's limits of stability. Anterior and posterior tactor activation coding is asymmetrical because the limits of postural stability are smaller in the posterior direction than the anterior direction.

It has been previously shown that anterior-posterior (A/P) display of tilt will reduce A/P sway during Computerized Dynamic Posturography Motor Control Tests involving A/P step perturbations [13, 14]. We hypothesize that multi-axis (4-16 columns of tactors) display of body tilt during multi-directional surface perturbations will reduce sway in all directions. Eight vestibular deficient subjects will be tested. Tilt estimates will be acquired from the motion sensor array mounted on each subject's lower back. Center of Pressure (COP) data will be measured with a force plate. Head, sternum, and pelvis movements (6 degrees of freedom) will be acquired from an optical tracking system. Previous kinematics measures developed to assess head on trunk stability during locomotion will be used to complement tilt and COP data.

<u>Assess the effect of tactor column spatial frequency on the stability of unilateral and bilateral vestibulopathic patients</u>.

The balance prosthesis will be configured with three rows and 16 columns of tactors and programmed to display four different tactor column configurations (no tactors, 4 columns, 8 columns, and 16 columns). Subjects will be trained on how to interpret and use each of the four different display configurations. Continuous and discrete perturbations will be used to evaluate the displays. Balance performance will be assessed for each of the four tactor column configurations for both continuous and discrete perturbations. Depending on the outcome of this analysis, tactor configurations can be programmed to dynamically change based on the user's environment. For example, if the eight-column tactor configuration proves to be most effective during the continuous perturbation experiment, the prosthesis could display eight columns of information when the user is riding on public transportation, etc. Dependent measures for the continuous perturbations include: root-mean-square (RMS) tilt, area of best fit ellipses to A/P and M/L tilt data, tilt pathlength, percentage of time spent outside of a one-degree tactor dead zone, RMS center of pressure (COP), and head/sternum/pelvis pitch and roll dynamics. Dependent measures for the discrete perturbations include: maximum tilt, time to maximum tilt, time to return to subject's baseline sway, and percentage of time spent outside a one-degree tactor dead zone.

<u>Determine the change in upper body postural control strategy used by patients while donning the balance prosthesis versus no prosthesis during quiet stance</u>

Challenging locomotor tasks, certain diseases, and long-duration space flight tend to induce a generally rigid ('en bloc') functioning of the head, trunk, and pelvis [35, 36]. The anchoring index is a previously published parameter for characterizing head and trunk stabilization strategies in the frontal plane during stance or unperturbed locomotion [36-38]. Angular dispersions and anchoring indexes will be computed from roll, pitch, and yaw of the head, sternum, and pelvis. These metrics will be used to determine the effect of tactor column configuration on upper body postural control strategy. The degree to which the 'en bloc' stabilization strategy is reduced might serve as an indicator of the effectiveness of one tactor configuration versus another.

Specific Aim 3: Assessment of a wearable vibrotactile balance prosthesis for improving medial-lateral gait stability

Following the completion of the second research aim, we should have a better understanding of the design requirements necessary for a balance aid that will provide sufficient body posture information during standing. The next logical step is to examine the efficacy of such a device during locomotor activities. On average, humans (healthy elderly) spend roughly three-quarters of their upright time in stance and one-quarter performing locomotor tasks daily [39]. Bipeds are least stable in the A/P direction during stance. However, the postural control challenge shifts from one of controlling the body about the A/P axis to the M/L axis during gait. In fact, A/P instability is a necessary condition for forward progression. Of the electrotactile, vibrotactile, and auditory balance devices being developed, published locomotion data exist only for one of two auditory devices. Hegeman et al. tested six compensated bilateral vestibular loss patients both with and without auditory biofeedback using a battery of tests. The tests included: semi-stance (walking 8 tandem steps with and without foam support surface) and gait tasks (self-paced walking over 3 m while horizontally rotating the head, vertically pitching the head, or with eyes closed; get up from stool and walk 3 m; and walking up and down stairs without handrails). The authors reported that the vestibular-compromised patients were able to perform the gait tasks as well the healthy controls and that auditory biofeedback was not effective in reducing sway during gait tasks, although for some gait tasks a decrease in trunk sway with velocity feedback was observed [40].This purpose of this research aim is to study the effect of providing M/L tilt information to vestibular-compromised patients during various locomotor tasks to determine whether or not real-time vibrotactile tilt feedback can be used to make adjustments to their trunk tilt position.

<u>Validate balance prosthesis during locomotion – Proof of concept</u>

Eight balance-impaired patients will participate in a training/experimental session that explores the usefulness of the balance prosthesis during locomotion. Several locomotor tasks varying in difficulty will be used. These tasks include: slow- and self-paced walking, perturbed walking, narrow stance walking, and walking across a foam surface. The slow-, self-, and perturbed walking trials will be performed on the BALDER platform (please see

General Methodology section). Narrow stance walking will be performed along one edge of the BALDER platform to create a one-sided balance beam. High-density foam sheets will be placed end-to-end to create a long foam surface, which will serve to distort proprioceptive inputs. Two display configurations will be used to test the effect of displaying tilt feedback at different points during the gait cycle. The first configuration will continuously display the subject's M/L tilt throughout the gait cycle. The second configuration displays gated information during only the 200 ms following the detection of the heel strike event. The second display was based on recent findings by Bent et al. that demonstrated that vestibular information is used at heel strike to determine the M/L placement of the subsequent foot placement [41]. Metrics developed in Specific Aim 1 will be used to assess the effectiveness of each feedback configuration.

BACKGROUND

Balance

Standing

The single inverted pendulum is a simple model used to describe postural dynamics assuming small angular deviations about an upright position. An analysis of the system dynamics reveals that the inverted pendulum is unstable (right half plane pole) without feedback. Subsequently, it can be shown that the system cannot be stabilized unless the feedback is comprised of at least the pendulum angular deviation about upright (tilt) and the pendulum tilt rate. The visual, somatosensory, and vestibular systems provide the primary sensory inputs that are fed back to the central nervous system and processed to generate motor commands that stabilize the body [42]. Jenk et al. reports that sensory input provides more accurate information about the body's velocity than its position or acceleration during static stance [43]. Compensatory ankle torque, given the appropriate feedback, is effective for controlling balance if small angular deviations are assumed. Specifically, small deviations about an upright position, enable upright posture to be effectively controlled by angle torque as long as the center of mass (COM) does not exceed the center of pressure (COP). When the COM exceeds the COP, ankle torque is no longer an effective control actuator. In such a case, a hip strategy, combination of ankle/knee/hip torques, or stepping is necessary to recover balance [44].

Previous postural control studies have shown that vestibular loss results in a normal ankle strategy, but a lack of a hip strategy [45]. Central and peripheral vestibular disorders lead to varied fall directions. For example, patients with bilateral vestibulopathy tend to fall in the fore-aft direction due to impaired vestibulospinal postural reflexes while patients with vestibular neuritis tend to fall laterally due to vestibular tone imbalance resulting from horizontal and anterior canal paresis [46].

Walking

The biped gait cycle is defined as the activity that occurs between heel strike of one limb (reference limb) and the subsequent heel strike of the same limb. The gait cycle is divided into three phases: stance phase (or single support phase), swing phase, and double support phase. Stance phase begins and ends with the heel strike and toe off of the reference foot, respectively. Swing phase onset occurs when the reference foot is no longer in contact with the ground. Double support phase is as the component of the gait cycle in which both limbs are in contact with the ground.

Walking, by its nature, requires that the COM exceed the COP in order to propel oneself forward. Therefore, fore-aft dynamics are likely stabilized passively or by low-level control [47, 48]. Fallers and individuals with fear of falling generally increase the portion of the gait cycle spent in double support phase and minimize the perceived unstable single support phase [21]. Townsend developed a model for biped gait that was stabilized via by discrete foot placements based on sensory feedback available prior to the time of foot placement [49]. Kuo's biped gait model showed that lateral balance is actively controlled against dynamic instability via lateral foot placement with visual-vestibular feedback [48]. Reduction of visual and/or vestibular information reduces the amount of sensory information available and is equivalent to increasing the sensor noise. This increase in sensor noise is hypothesized to have a greater impact on lateral than fore-aft foot placement [48]. Bent et al. showed by applying galvanic vestibular stimulation to subjects at heel strike, mid-stance, and toe-off, that vestibular input was most important during heel strike events; foot placement was most affected when stimulation was applied during heel strike and least affected during mid-swing [41].

Kuo's model also indicated that there is a slight stability advantage with wider foot placement. Although this may seem intuitive to one who has walked on ice before or observed elderly gait, Krebs et al. has experimentally shown that wide-based gait alone cannot differentiate between subjects with and without balance impairments [50]. In another study from the same lab, 102 balance-impaired patients and healthy subjects during free and paced gait showed no significant differences in the base of support [51]. Their data suggest that if trends exist toward wider stance width for balance-compromised individuals, the trends are not sufficient to blindly identify an individual with a balance disorder from a healthy normal.

From the standpoint of designing the vibrotactile display for a balance prosthesis, the system must be malleable. It must be able to sense the user's activity mode and dynamically alter its feedback accordingly. Additionally, the system should be able to incorporate additional sensing technologies such that feedback can be delivered in a gait-cycle appropriate manner to aid in foot placement.

Gait Disorders

Clinical Vestibular Disorders

Postural imbalance as a result of various central and peripheral vestibular disorders is a frequent cause of compromised gait in the general population [52]. Subsequently, many central and peripheral nervous system diseases can be initially recognized by their impact on posture and gait [53]. Typically, patients suspected of having balance control disorders resulting from central or peripheral vestibular disorders undergo clinical evaluations which examine posture, postural reflexes, and walking [53]. Stance width, body sway, and one's ability to balance on two legs during quiet standing and locomotion comprise the posture evaluation. Postural reflexes are observed while gently pushing the patient forward and backward on the back and chest, respectively. Additionally, locomotor parameters such as gait initiation stride length, rhythm of stepping, speed of walking, and gait termination and associated synergistic arm movements are assessed. Peripheral vestibular disorders include vestibular neuritis, benign paroxysmal positioning vertigo (BPPV), Ménière's drop attacks, otolith Tullio phenomenon, vestibular paroxysmia, and bilateral vestibulopathy. Central vestibular disorders include vestibular epilepsy, ocular tilt reaction, paroxysmal ocular tilt reaction, lateropulsion (Wallenberg's syndrome), and downbeat nystagmus/vertigo [54].

Acute unilateral vestibular dysfunction can inflict severe vertigo, imbalance, nausea, vomiting, and prostration [55]. The aforementioned side effects associated with acute unilateral vestibular dysfunction generally subside over several days. It is not uncommon for the patient to initially have difficulty detecting the direction of vertigo if symptoms are severe.

Loss of bilateral vestibular function results in gait unsteadiness and oscillopsia induced during walking or head movements. Typically, such patients can be identified by decreases in ocular motor response to caloric stimulation and/or angular accelerations and substantial anterior-posterior postural sway. Patients suffering from complete bilateral loss never recover the ability to maintain postural equilibrium when both vision and proprioceptive sensory information are compromised [52].

Astronauts

Similarly, astronauts returning from long-duration space flight exhibit comparable difficulties regarding balance and gait. Long-duration exposure to weightlessness results in broad-spectrum physiological deconditioning. Despite significant flight experience within the US and Russian space programs, much remains to be learned regarding the fundamental mechanisms triggering the degradation of wide-ranging physiological subsystems. Long-duration exposure to the stimulus of weightlessness, and to some extent, short-term exposure, results in alterations to bone physiology, skeletal muscle, sensorimotor integration, cardiovascular and pulmonary systems, endocrine and immune systems, and psychosocial behavior [56]. The physiological response to the absence of gravity is both appropriate to the immediate environmental requirement and demonstrative of human adaptability. The issue is not whether these changes are appropriate for the weightlessness environment, but rather if the changes will detrimentally affect the safe return to and productivity in a gravito-inertial environment [57, 58].

Neurosensory changes resulting from exposure to weightlessness were of minor concern during the Mercury and Gemini programs because of the limitations in flight duration [59]. During post flight evaluations of cosmonauts participating in the Soyuz missions, however, Russian investigators observed alterations in locomotor behavior, which included distinct post flight performance decrements in gait and jumping behavior. The Soyuz missions ranged in duration from 2 to 63 days and post flight abatement in locomotor performance was generally proportional to the length of the mission [18].

Post flight postural control is characterized by increased dependence on vision throughout the sensorimotor rearrangement process (otolith and proprioceptive) which occurs concomitantly with re-adaptation to a gravito-inertial environment [59]. For example, post flight postural testing performed by the Johnson Space Center (JSC) Neuroscience Laboratory demonstrated balance control deficits in 45 astronauts 1.6-4.5 hours following flight [60]. The recovery profile was fit with two distinct exponential curves; one portraying rapid improvement over the first 8 to 10 hours post flight and the second by a more gradual return from to preflight stability levels over the next 4 to 8 day. Additionally, first time astronauts (rookies) tended to exhibit greater post flight sway compared to veterans when vestibular input was the sole reliable spatial orientation reference cue [60].

Post flight locomotion is characterized by exaggerated medial-lateral leg placement, trunk shifts to the side of the supporting leg, and failure to maintain the intended path [18]. Additional post flight locomotor alterations observed within Shuttle and Mir crewmembers include: perceived sensation of turning while attempting to walk a straight path, impairment of postural stability while turning corners, perception of exaggerated pitch and rolling head movements during walking, compromised gaze control, reduction in head stability, loss of orientation, and overall degradation in locomotor function [18].

One potential life-threatening repercussion of compromised sensorimotor integration upon return to a gravito-inertial environment is the inability to egress from the spacecraft in the event of an emergency. Presently, countermeasures are being pursued to alleviate emergency egress problems and ameliorate long-duration broad-spectrum physiological deconditioning for astro/cosmonauts assigned to International Space Station flights. A balance prosthetic could provide additional sensory information to help prevent unsteady astronauts from falling or provide directional cues for evacuating in an emergency situation.

Gait and Falls in the Elderly

Gait disorders in the elderly include gait that is slow, unsteady, or biomechanically compromised [61]. Susceptibility to falls and fall-related injuries increase with age [20].

Falls are a leading cause of hospitalization and accidental death among the elderly and contribute to a decline in functional mobility by causing injury, limiting activity, and instilling fear of future falls [20, 62]. For example, between 1992 and1995, 147 million injury-related visits to emergency departments in the U.S. were reported [63]. Falls were the leading external cause of injury, accounting for 24 percent of the injury-related emergency department visits and were more common among children under 5 years of age and the elderly than for other ages [63].

Standards of Practice

Clinical Testing

Clinical laboratory tests that assess the integrity of the vestibular system are typically comprised of the following exams: the electronystagmography (ENG) test battery (pursuit/random, saccades, spontaneous nystagmus, positional maneuvers, caloric stimulation), rotation about a vertical axis, Romberg exam, and Computerized Dynamic Posturography. ENG is typically available in a practitioner's office, while the full complement of tests is only available at a specialized balance center or tertiary care center.

Rehabilitation

Rehabilitation is an important component of the recovery process for individuals inflicted with vestibular disorders. An effective rehabilitation program facilitates the ability of the patient's central nervous system to compensate for lesions in the vestibular system by focusing on the development of adaptive capacities for retraining postural stability [52]. Rehabilitation programs should cater to the particular balance deficits of each individual patient instead of providing "blanket" treatments for all vestibular disorders (central and peripheral) [52].

Adaptive generalization, or "learning to learn," and transfer of adaptation are of particular interest for use as clinical and post flight vestibular rehabilitative techniques and countermeasures. Adaptive generalization is defined as the ability to adapt more readily to a novel sensory rearrangement as a result of prior adaptation training [64]. With regard to balance rehabilitation, patients participate in a series of exercises designed to recover, retrain, or develop new sensorimotor strategies to facilitate functional mobility, decrease dizziness, and re-establish effective coordination [52].

Both perturbations during locomotion and sensory substitution in the form of the vibrotactile balance prosthesis described above could potentially be incorporated into current rehabilitation programs. Verbal feedback obtained from subjects participating in the perturbation study suggests that controlled surface perturbations during gait allowed them to realize their balance capabilities in a safe setting. Subjects participating in the vibrotactile balance prosthesis experiment commented that they were able to experience their "true limits of stability" and that following completion of the study, increased their daily activities of living (i.e., one subject stated that she walked outside to pick up the newspaper for the first time in several years).

Vestibular Sensory Substitution Devices

Uses of Vestibular Sensory Substitution Devices

A sensory substitution device for balance can serve as a permanent or temporary replacement of motion cues, a tool for vestibular rehabilitation, or an additional sensory channel for conveying information to military troops, pilots and astronauts. In addition to being a valuable balance aid, this technology has numerous video-gaming and sporting applications.

Types of Vestibular Sensory Substitution Devices

Non-implantable prostheses such as vibrotactile display of body tilt, surface electrode stimulation of the vestibular nerve, electric currents applied to the tongue, and audio feedback offer a non-invasive means of providing self-motion cues compared with implantable devices [13]. Implantable prostheses are presently being tested in animal models. The semicircular canal ampullae, otolith organs, vestibular nerves, and Scarpa's ganglion are potential sites for electrode implantation [11].

GENERAL METHODOLOGY

Data collection took place in the Injury Analysis and Prevention Laboratory in the NeuroMuscular Research Center at Boston University.

Equipment

BALDER

The Injury Analysis and Prevention Laboratory has a unique custom-built moveable BALance DisturbER (BALDER) platform. The 2.1 m square BALDER platform generates a programmable stimulus while the motion of the subject's body is optically tracked. The primary components of the BALDER platform are: a force-plate (ORG-6 AMTI, Newton, MA, USA) imbedded in a wooden platform, two AC-servo motors controlled by two linear servo drivers, two high precision linear position transducers (Novotechnik, Germany), and a 16 channel A/D - two channel D/A data acquisition board (Microstar 3200e/415). The BALDER platform performance characteristics include: movement range of 0.45 m in the horizontal plane, acceleration capability up to 1.2 g, peak velocities of over 1 m/s, positioning accuracy to 1 mm, and three dimensional reaction force and torque measurement capability. Additionally, a long walkway was attached to BALDER to allow

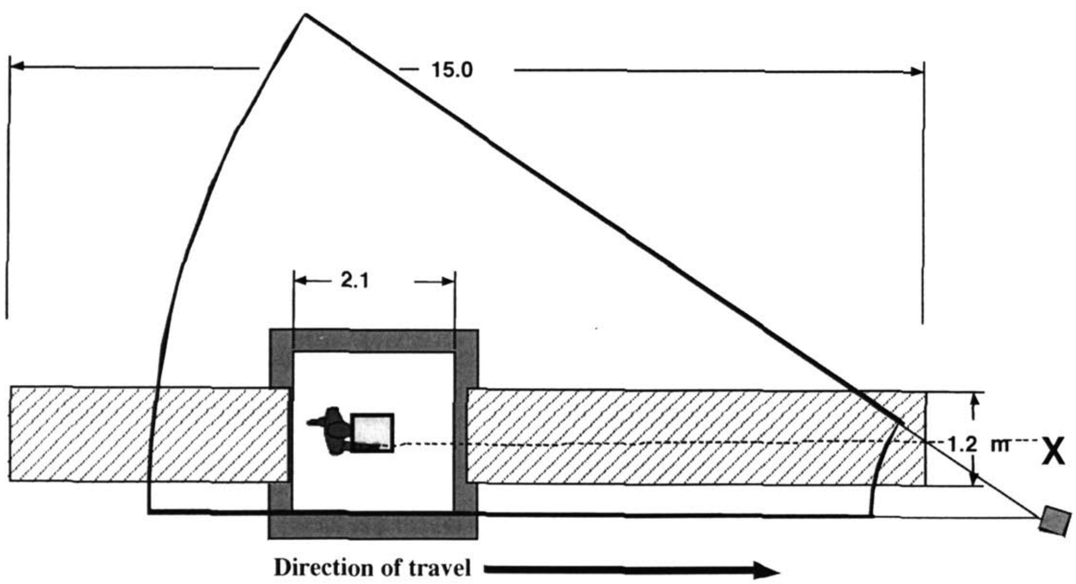

Figure 2. BALDER perturbation platform and wooden walkway

for the study of perturbations during free locomotion (Figure 2).

Optotrak

Kinematics are collected using the Optotrak 3020 system (Northern Digital, Waterloo, Ont.). Rectangular arrays consisting of six infrared emitting diodes (IRLED) can be placed on the subject's legs, pelvis, sternum, and head. The IRLED sampling rate is 1500 Hz and the array positions are estimated at 40 or 100 Hz depending on the number of arrays used in a particular study. The 3020 is placed at the far end of the walkway or 4 m away from BALDER to accommodate gait or standing studies, respectively. The maximum viewing distance is over 12 m when the IRLED intensity is set to its maximum output. However, the IRLEDs can only be used for short periods of time at this intensity and the technical specifications have only been validated within the manufacturer-defined 3020 system optimal viewing range (6 m volume). The root-mean-square IRLED position error over the viewing volume is on the order of 1 mm as indicated by the calibration procedure provided with the system software. The 3020 specs describe accuracy of up to 0.1 mm and resolution of 0.01 mm. The 3D IRLED translations are recorded and converted to 6D data using the Data Analysis Package provided with the system. Optotrak data is saved and formatted into files that are compatible with MATLAB version 5.2 (The MathWorks, Natick, MA) for subsequent filtering and processing.

Balance Prostheses

Two versions of the vibrotactile balance prosthesis are used in this research. The tethered, two-axis device is used for the investigations involving standing. The wireless, three-axis device is used for gait studies (Figure 3). The vibrotactile balance prostheses in general, consist of a motion-sensing system mounted on the lower back, a vibrotactile display, and a laptop computer with analog and digital interfaces. The inertial motion-sensing system is composed of microelectromechanical (MEMS) gyroscopes that

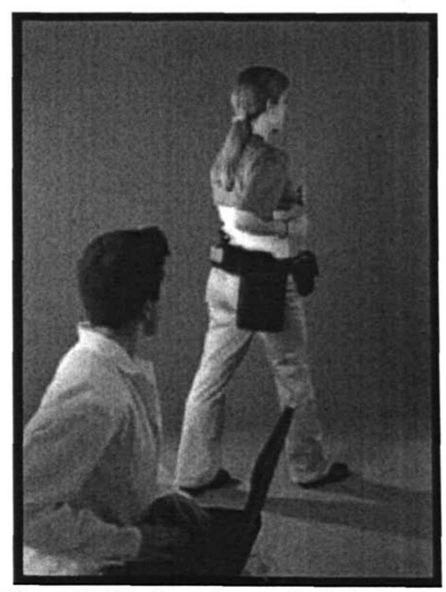

Figure 3. Wireless vibrotactile balance prosthesis

sense angular rate and MEMS accelerometers that sense linear acceleration. Tilt estimates are displayed on the subject's torso in the form of vibrations. The prosthesis will be equipped with a 3 row by 16 column tactor array and will be customized on an individual basis using an elliptical fit to four static leaning values. The lowest, middle, and highest rows activate when the subject leans ~1°, ~3-5°, and ~ 5-7° respectively outside of a user-defined cone of stability (Figure 4). Subjects are instructed to always move to null out the vibration. In a previous study involving the average response of six subjects to SOT 5 & 6 tests, significantly lower root-mean-square tilts were observed when using proportional plus derivative feedback compared to proportional or derivative feedback alone. Predictor feedback did not result in significantly better values. Therefore, the subjects in this research, unless otherwise indicated will be provided with proportional plus derivative estimates of their tilt.

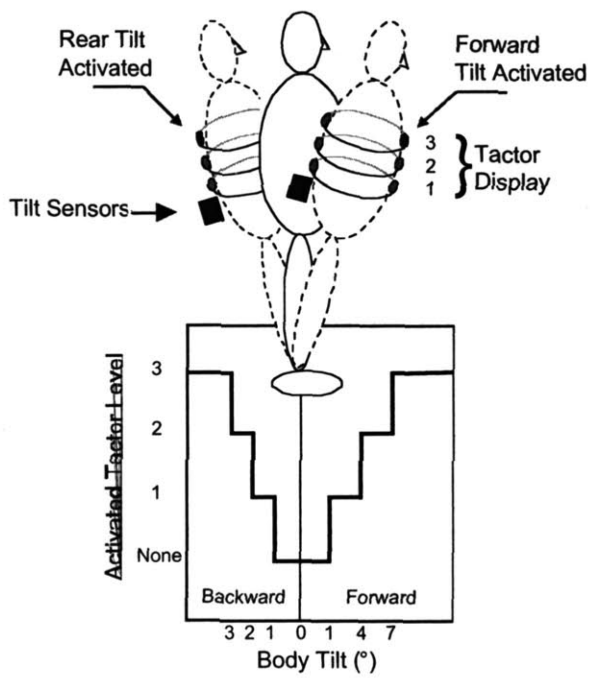

Figure 4. Tactor activation scheme

Miscellaneous

A video camera will be used to record the experimental sessions. The foam used in this study is 10 cm thick medium density foam (Sunmate Foam, Dynamic Systems, Inc., Leicester, NC) and when placed end-to-end creates a 4 ft wide by 24 ft long walkway.

Subjects

Poorly compensated vestibulopathic patients will be referred by Drs. Lewis and Rauch, a neuro-otologist and oto-neurologist, respectively in the Harvard Medical School Department of Otology and Laryngology at the Massachusetts Eye & Ear Infirmary. Poorly compensated is defined as those patients who fail the NeuroCom EquiTest computerized dynamic posturography Sensory Organization Tests (SOT) 5 & 6. During SOT 5, the subject's eyes are closed and the posture platform is sway referenced (Figure 5). SOT 6 is performed with the subject's eyes open and both the platform and visual surround sway referenced. Patients with histories of mental illness, migraines, or disorders that significantly affect motor or sensory systems (i.e., advanced diabetes) will be excluded. Additionally, obese individuals will be excluded due to the size constraints of the vibrotactile balance prosthesis.

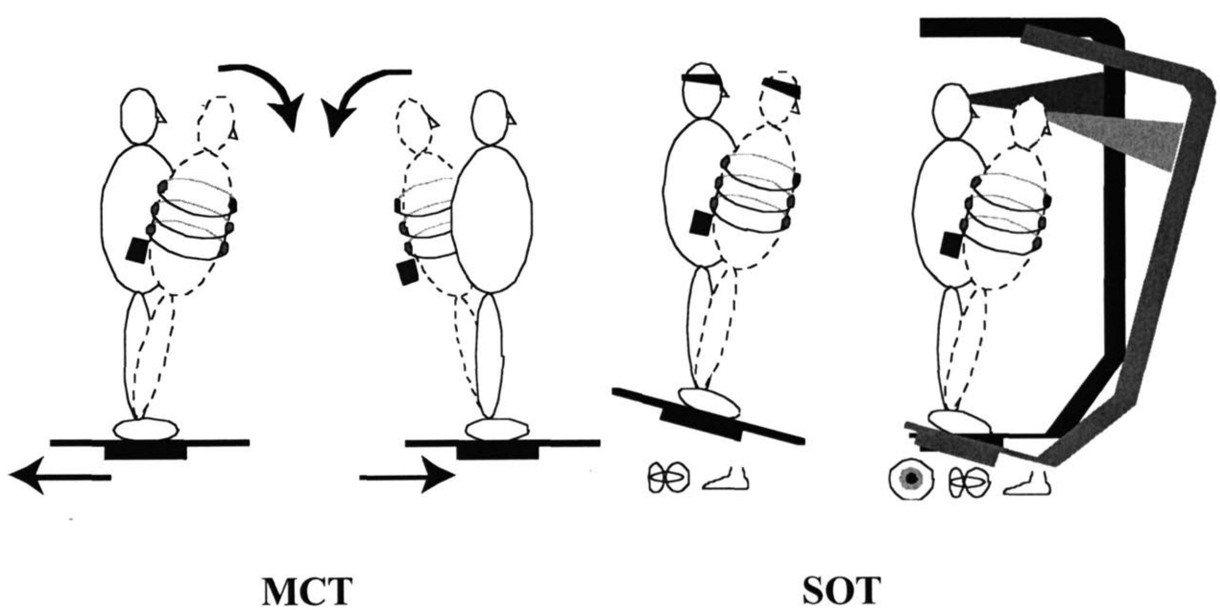

Figure 5. NeuroCom EquiTest Computerized Dynamic Posturography

Standard Protocols

Perturbed gait

The following perturbation protocol has been used in two previously published studies [4, 5]. This protocol was followed exactly to collect the age-matched data for the first research aim. An abbreviated version of this protocol was used to test several of the subjects in the third research aim. Details of the modified protocol can be found in Study 3.

Subjects are instructed to walk at a pace of 100 steps per minute (pace maintained by an electronic metronome) along the walkway while fixating on a visual target (positioned at eye-level beyond the end of the walkway). The complete walkway consisted of a staging area measuring either 3.7-m leading up to the balance disturber platform (BALDER) followed by an additional walkway extension measuring 6.1-m. Both the staging area and the walkway extension are 1.2-m wide.

Four different perturbations and one control case (no perturbation) are included in the experimental protocol. The perturbations are applied at two different amplitudes (5-cm and 10-cm) and in opposite directions (+45°, +225° as measured clockwise from the subject's direction of march). The onset of the BALDER translation occurs 100-120 ms after the detection of right heel-strike for the to ensure that the subject's left leg is in its swing phase. The platform is programmed to accelerate and decelerate at a constant rate of 5 m/s^2. Maximum velocity is reached halfway through the perturbation and its magnitude is the consequence of the acceleration and displacement settings.

Perturbation trials are rehearsed as needed to familiarize subjects with the novel stimulus issued by the BALDER platform; practice trials using both 5 and 10 cm magnitude perturbations are repeated until "stutter stepping" (defined as a quick corrective step) is eliminated. Three trials of non-paced and paced walking without perturbation were collected prior to the start of the experimental protocol. The experimental protocol consists of three trials of each perturbation type (+45°/+225° at either 5-cm or 10-cm translation) and twelve control cases applied in a random order.

Optotrak Data Analysis

3D positional data are converted into 6D data using Northern Digital RIGMAKER and Data Analysis Package RIGID software. Due to instrumentation inconsistencies between the three major data collection sessions (1999, 2001, 2004), all data have been reprocessed using identical modified processing schemes. Supplemental Matlab scripts were developed by Balkwill and Sienko to identify and remove mis-sampled IRLED 3D marker data on a sample-by-sample basis prior to conversion to 6D data using RIGMAKER. Figure 6 shows two frames (two samples) of Optotrak data. The plot on the left compares a properly sampled frame with a mis-sampled frame (IRLED x and z positions) on the right. Once 6D data are obtained, another set of scripts allows the user to inspect the integrity of the data and select the regions over which interpolation, if any, will be performed. For example, if a rigid body marker were obscured for a brief instant while executing a ballistic trajectory, one could justify interpolating over that region since the two end points are known and a non-linear trajectory is statistically unlikely (if considering a sine wave, imagine a small portion of the signal > 0 and $< \pi/4$). However, if the rigid body marker were obscured during a transition from, for example, a left-ward tilt to upright stance, it would be impossible to know the peak tilt achieved and the exact trajectory characteristics and therefore the data should not be interpolated (if considering a sine wave, imagine the concave region of the signal near $\pi/4$).

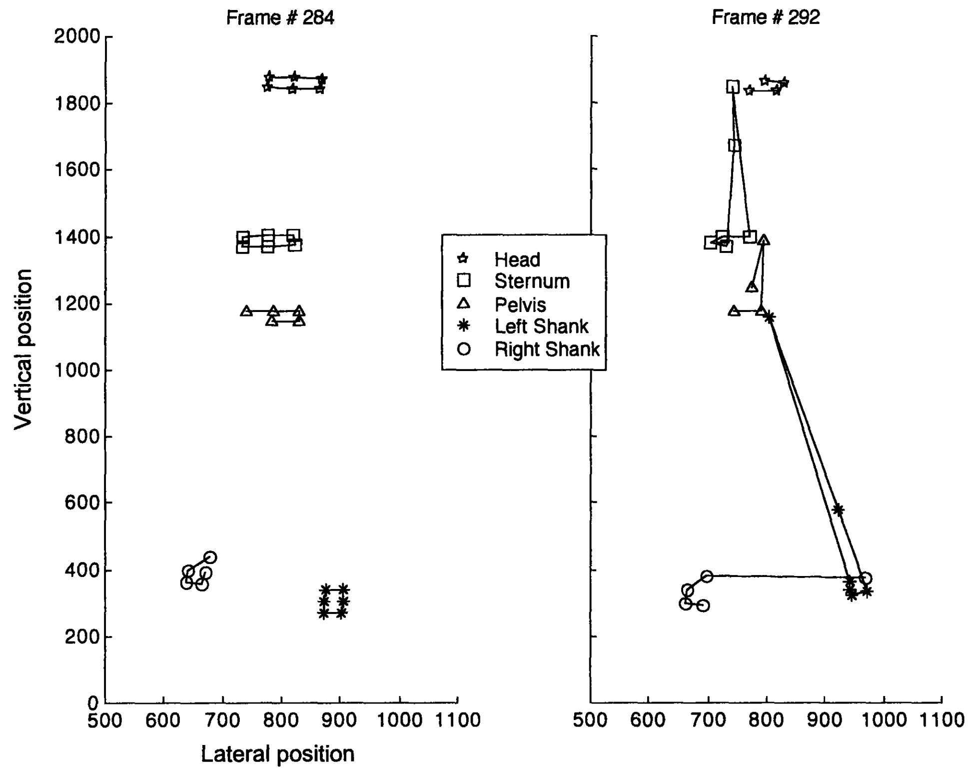

Figure. 6 Example of mis-sampled optotrak position data. The plot on the left shows one frame of properly sampled data from an illustrative subject. The plot on the right shows a mis-sampled frame for the same subject. The stars indicate head markers, the squares sternum markers, the triangles pelvis markers, and asterisks left shank markers, and the circles right shank markers.

RESULTS SUMMARY

Study I: Quantification of vestibulopathic gait

We investigated the recovery trajectories of young controls, vestibulopathic-deficient patients, and age-matched controls in response to support surface perturbations during locomotion. The vestibulopathic subjects employed in this investigation had subtle clinical findings of postural instability despite their known vestibulopathy. M/L moment arms defined by the horizontal difference in sternum and shank position of the stance leg were used to characterize foot placement during perturbation trials. Step width variability was used to characterize foot placement during natural and paced-gait trials. Upper body dynamics were captured by head, sternum, and pelvis angular dispersions and head and sternum anchoring indices, which describe stabilization strategies.

The M/L moment arm responses among groups were dependent on the choice of sampling event. The large BL perturbation revealed a significant difference among groups when moment arms were sampled using the A/P shank crossing event. The moment arm at the 6th shank crossing event was significantly different for both the large (p>0.0011) and small perturbations (p>0.0042). When sampling the M/L sternum and shank positions at the A/P shank crossing events, the results corroborated with the previously published data and our expectations; the first moment arm following the perturbation was significantly smaller (more exaggerated) for the vestibulopathic group compared to the young controls. However, although the age-matched controls had larger moment arms compared to the vestibulopathic group, they were still significantly smaller than the young controls' responses. This finding did not hold when calculating the M/L moment arm at the estimated heel strike event immediately following the perturbation. In the heel strike sampling case, the age-matched controls had significantly smaller (more exaggerated) moment arms compared to the vestibulopathic patients. The forward right perturbations did not reveal significant differences among groups.

Step width variability sampled at heel strike events during non-perturbation paced gait trials was the best discriminator among subject groups, with vestibulopathic patients exhibiting a significantly greater variability compared to both age-matched and young controls. Additional measures showed significant results between pairs of groups, but not among all three groups. Vestibulopathic patients had a smaller, but not significantly smaller, average sternum RMS roll sway during natural and paced gait trials compared to young and age-matched controls. On the other hand, vestibulopathic patients' sway was greater over a data segment spanning the first five recovery steps following all perturbation directions and magnitudes compared to the young and age-matched controls. Neither the head nor sternum roll anchoring indices were significant among groups by trial type. In all three non-perturbation trial sets however, the median head anchoring index was least negative and not significantly different from zero for the young controls, which suggests neither a head stabilization in space nor with respect to their trunk strategy. Age-matched and vestibulopathic patient sternum and pelvis pitch angular dispersions were consistently significantly larger than those corresponding to the young controls' values for all trial types. The trend in the pelvis angular dispersion was reversed with the young controls exhibiting larger yaw dispersions compared to the vestibulopathic patients and age-matched controls. Pelvis yaw dispersions however, were larger for the vestibulopathic and age-matched subjects compared to the young controls across all trials.

In summary, step width variability during paced gait trials and the M/L moment arm calculated on the step immediately following a large backward left perturbation were the most sensitive measures for discriminating among the vestibulopathic, age-matched control and young control groups.

Study II: Use of multi-directional vibrotactile feedback during support surface perturbations

Single-axis vibrotactile feedback has been shown to significantly reduce the root-mean-square (RMS) sway in vestibulopathic patients during single-axis perturbation. This research demonstrated that multi-directional vibrotactile biofeedback can be used to improve postural sway performance during multi-directional surface perturbations. Eight well-compensated vestibular-deficient patients donned a multi-axis vibrotactile prosthesis that mapped tilt estimates onto the patients' torsos in a 3 row by 16-column vibrotactile tactor array. The number of tactor columns displayed was varied depending on the test condition to assess the effect of spatial resolution on several postural stability performance parameters. Root-mean-square tilt ($p< 0.0003$), elliptical fits to trajectory areas ($p<0.0000$), and percentage of time spent activating the middle ($p<0.0006$) and top rows of tactors ($p<0.0025$) were significantly decreased when the device provided tilt feedback during trials of continuous support surface perturbations compared to the no tilt feedback condition. Tilt pathlength computed by summing the square root of the squared sum of roll and pitch tilt revealed a significant difference ($p < 0.0058$) only between the first trial performed with the device off compared to all device on display trials. RMS COP did not significantly change as a function of device state (on/off). There were no significant differences for head, sternum, or pelvis angular dispersion values or head and sternum anchoring indices suggesting that subjects did not merely stiffen up when the device was turned on, but rather that the subjects employed similar stabilization strategies regardless of the device state. There was no significant difference in the time to peak tilt displacement, peak tilt, or time to reenter the one degree dead zone among display configurations for discrete perturbation trials. RMS tilt and percentage time spent outside the dead zone, calculated over a 5 second interval starting three seconds after the onset of the discrete perturbations were significantly greater in the device off versus the device on configuration. The results show that among the displays evaluated in this study, there is not an optimal tactor column configuration for standing tasks involving continuous and discrete surface perturbations. However, subjects expressed individual display preferences. Furthermore, subjects performed worse when nonsensical information was presented. Both short and long-term reductions in RMS sway and other parameters were observed. This finding suggests that rehabilitation balance training is one possible application of vibrotactile tilt feedback.

Study III: Use of a vibrotactile balance prosthesis during locomotion

Eight vestibular deficient subjects participated in a four hour pilot study that explored the usefulness of a wearable vibrotactile balance prosthesis during locomotion. Subjects trained for approximately 45 minutes with a vibrotactile balance prosthesis that provided real-time medial-lateral (M/L) tilt feedback. Two proof-of-concept feedback displays were evaluated during various locomotor tasks, which included slow and self-paced walking, perturbed locomotion trials, walking on a foam surface, and walking along a narrow walkway. The average tilt offset and root-mean-square (RMS) tilt was calculated for all locomotor tasks. Additionally, during slow and self-paced walking trials, stance width and upper body dynamics were assessed. A modified five point Likert scale was used to assess the subject's impression regarding the usefulness of the device in improving stability. Three well-established balance-related subjective questionnaires were also completed. Use of roll tilt feedback resulted in a significant decrease in roll sway for the narrow stance walking task. Non-significant roll sway decreases were observed with the use of roll tilt feedback for slow, self-paced, and foam walking tasks. RMS roll tilt for the posttest device off trials (following sets of trials performed with the device on) tended to be lower than the average RMS roll tilt of the pretest device off trials. Several subjects completed more than the one standard set of vibrotactile feedback trials during the foam walking task. For these subjects, their average RMS roll tilt was lower during the second set of tests compared to the first set. Step width and step width variability were significantly reduced during vibrotactile feedback trials compared to trials without feedback. A significant correlation between the Dizziness Handicap Index score and the percent change in roll sway provided some insight into the type of patients that could potentially derive the greatest benefit from vibrotactile tilt feedback. The most severely sensory-deficient subject presented provided one example of how a visible asymmetrical roll tilt of the trunk could be eliminated when vibrotactile feedback signaling roll tilt was provided. No significant difference was identified between the two device displays evaluated. However, subjects expressed preferences for one over the other on an individual basis. This pilot study showed that vestibular-deficient patients can decrease their M/L RMS tilt and significantly decrease their step width variability during challenging locomotor tasks by using vibrotactile tilt feedback.

DISCUSSION AND CONCLUSION

As with any body of research, more questions are raised than are answered by the time that the final experiment is completed. I have chosen to combine the general discussion of the results and the future work in one section since it is only through the discussion of results, that the future work is formulated. Below is a brief discussion of the overall implications of this research and some of the questions that deserve future attention.

The main goal of this thesis was to develop metrics that quantify the locomotor stability of individuals with reduced vestibular function and to assess the capability of a noninvasive sensory substitution device for improving postural and locomotor stability. The work in Studies II and III is important because it furthers the development of a vibrotactile balance prosthesis, which could greatly impact the quality of life for balance-impaired patients, serve as a tool to enhance balance rehabilitation/training, and potentially reduce the risk of falls. The metrics developed in Study I will permit gross and subtle changes in gait to be evaluated during the developmental stages of a "walking" balance prosthesis. Study III shows for the first time that tilt biofeedback can be processed and acted upon during locomotor tasks. Additional subjects with moderately to serve sensory deficits should be evaluated with the vibrotactile feedback device during locomotor activity to determine if the trends seen in M/L sway reduction are indeed significant.

Study II demonstrates for the first time that multi-directional vibrotactile tilt feedback can be used to improve postural stability. The electrotactile and auditory tilt feedback devices currently being used in research studies only provide tilt information along a single-axis (A/P). Additionally, tilt feedback has never been evaluated during multi-directional surface perturbations. To date, studies have focused on quiet standing and single-axis linear translations and rotational perturbations. The results from Study II showed that a multi-directional tilt display was effective in reducing the resultant tilt in response to multi-directional support surface translations.

Of the four perturbation types (two directions, two magnitudes) delivered in Study I, the large backward left perturbations were most effective in discriminating among subject groups. However, the M/L moment arm results were sensitive to the gait cycle sampling event. Although not considered in the original comparison between young controls and vestibulopathic patients, step width variability (SWV), appeared to be a good metric for differentiating among all three subject groups. Since we did not systematically step through all phases of the gait cycle in our analysis, it remains to be seen whether step width variability or M/L moment arms resulting from surface perturbations are the most effective means of discriminating among subject populations. It is worthwhile mentioning though, that a locomotor test measuring step width variability is an easier test to implement in a postflight or clinical setting.

One goal of sensory substitution during locomotor activities is to reduce the risk of falls. Narrowed stance width has been correlated with an increased risk of falling [65]. Given that SWV increases with subtle vestibulopathy and age as shown by the well-compensated vestibulopathic patient and age-matched control results from Study I, decreasing SWV variability may have a role in reducing postural instability and subsequently, one's fall risk. Study III demonstrated that SWV can be decreased with the use of vibrotactile roll tilt feedback during locomotor tasks.

<u>Vestibulopathic patients as long-duration astronauts analogues</u>

Study I arose from reviewer's critique of a previous manuscript that compared the recovery trajectories of vestibulopathic patients to young healthy controls. The original study, funded by the National Space Biomedical Research Institute as part of the Neurovestibular Countermeasure Initiative in 1998, aimed to determine whether or not surface perturbations could be used to differentiate post-flight astronauts that appeared to have completed re-adaptation to the Earth's 1-G environment based on standard computerized dynamic posturography performance to those that had not. Although astronauts returning from long-duration spaceflight exhibited test scores indicative or preflight balance status, they continued to complain of balance difficulties and visual disorientating illusions. Since astronauts are hand-selected with great care and consideration, such oral reports suggesting residual postural instabilities could not be taken lightly. It was decided that additional avenues of sensorimotor evaluation would be explored to determine if a more sensitive indicator of recovery could be established. Due to the enormous costs associated with spaceflight, and given that traditional techniques of simulating some of the physiological changes that take place in a micro-gravity environment such as parabolic flight, 6° head down tilt bed rest, and water immersion, do not suitably mimic the sensorimotor changes that occur as a function of spaceflight, Earth-based patient populations were considered as potential analogues.

Postural and locomotor responses of astronauts returning from long duration space flight have been qualitatively and anecdotally compared to those of vestibulopathic patients. Astronauts and vestibulopathic patients alike have been described as exhibiting deficits in head-trunk-pelvis roll compensation with their upper body dynamics mimicking a single inverted pendulum (locking of body segments to one another) instead of hierarchal stability of the head with respect to space (to provide a stable visual platform) as seen in young, healthy individuals.

The precursor to Study I compared well-compensated vestibulopathic patients to young healthy subjects. The young subjects were selected as controls for the astronaut population. The average age of the current astronaut corps is in the early forties. Therefore, the particular age group that was selected to serve as the control group was justified based on the question that was being addressed – are well-compensated vestibulopathic patients analogues for long duration post-flight astronauts. However, posture and gait stability are negatively affected by age [19-31]. It is possible that some of the significant differences that were originally observed when comparing recovery responses to surface perturbations between vestibular-compromised patients and young healthy controls were due in part to age and not purely the vestibulopathy. Based on the results of Study I, well-compensated vestibulopathic patients who cannot be distinguished from controls during clinical standing tasks can be distinguished from controls during gait. Step width variability was the best discriminator among subject groups. Neither the head nor sternum roll anchoring indices were significant among groups in Study I. Therefore, these data cannot categorically support the notion that well-compensated vestibulopathic patients exhibit 'en-bloc' head-trunk stabilization in the frontal plane. However, age-matched and vestibulopathic patient sternum and pelvis pitch angular dispersions were consistently significantly larger than those corresponding to the young controls' values. The trend in the pelvis angular dispersion was reversed with the young controls exhibiting larger yaw dispersions compared to the vestibulopathic patients and age-matched controls.

Balance prostheses

Changing from the diagnostic to the treatment perspective, vestibulopathic patients are a generally underserved patient population in comparison with patients suffering from other forms sensory loss such as vision and hearing. They suffer from what can be termed an "invisible disease." Often times, their symptoms are interpreted by their family, friends, and work colleagues as purely psychological, which can lead to additional social stresses. Symptoms run the gamut from vertigo to postural instability and gait unsteadiness. Canes, walkers, and wheelchairs can serve as either light touch support to provide verticality cues or biomechanical support for patients who require weight-bearing assistance as well. Assistive devices can be cumbersome to transport and draw undue attention to individuals with intact musculoskeletal systems. Perhaps of more concern, they have the potential to induce or contribute to a fall by constraining medial-lateral foot placement, an important control mechanism for maintaining balance during gait. A need exists for both an implantable device for bilateral loss patients and a non-implantable sensory substitution device for temporary replacement of motion cues [13].

The next design challenge

The next major challenge to the development of the balance prosthesis is determining the optimal means of aiding an individual during locomotor tasks. Study III demonstrated that roll tilt can be reduced by vibrotactile tilt feedback during challenging locomotor tasks such as walking with a narrow stance or walking on a surface that distorts proprioceptive information. Based on the findings of Bauby et al. [48] and Bent et al. [41], it seems logical to try and provide the wearer with information on where to laterally place the subsequent foot. The most appropriate or optimal information to feedback during gait remains to be determined in future investigations.

The ideal device

The ideal sensory substitution device would provide meaningful body motion and orientation information in a discreet manner. The device's inertial sensing unit should be capable of providing accurate information throughout the course of the day. The power source should be rechargeable and last for a 12-hour period. The device should be able to detect the state of the individual and automatically switch modes to provide the appropriate type of motion feedback for the given task. For example, we have yet to investigate the type of feedback necessary to assist an individual when arising from a seated position. The display and feedback signal may differ in this situation compared with the information provided during steady state gait. The device should be intuitively easy to use and customizable to provide meaningful information to the specific user. This brings to mind the inside-out versus outside-in conundrum in aviation displays [66]. Both displays have value and represent the status of the aircraft, but one does so with the aircraft as the reference and the other with the horizon as the reference. A similar display design scenario can be applied to the balance device. Some users may prefer to have the direction they are tilting indicated to them while other would prefer to be told which way they should move in order restore an upright posture. Based on the results from Studies II & III, personalized feedback settings (spatial resolution, tactor activation pattern, etc.) will be an important design principle to incorporate into a commercial prototype. Other considerations for a commercial device include data logging capabilities that can monitor and report balance events to physicians and physical therapists, and training modules that can by "played " on the device taking the patient through a series of balance rehabilitation exercises at home.

Perceived versus objective benefit

A single fall can have a significant psychological impact on an individual. In addition to physical health complications, a fall can lead to decreased participation in activities due to fear of falling. A decrease in activity level subsequently reduces one's ability to maintain balance and can lead to future falls [67]. This sequence of events is known as "the vicious cycle." A sensory substitution device has the potential to impact the patients' quality of living by providing patients with the confidence to perform tasks that they might not otherwise do. One anecdote related to this issue springs to mind as a result of interacting with patients from Studies II and III. Many of the patients that have participated in one of our studies involving the vibrotactile balance prosthesis have subsequently requested to participate in additional studies. Patients typically cite a perceived improvement in balance following completion of the experimental session using the device as their primary reason for wanting to return. Following training with the device, some patients claim to increase their level of activity at home because they feel more confident about their balance capability. Part of the reason why patients may feel this way is because we challenge them to explore their limits of stability in a safe environment. For example, patients participating in the perturbation protocol (Study I) have reported similar feelings of improved postural stability and activity following participation in the study. Patient reports such as these serve as a reminder that a perceived benefit of using the vibrotactile balance prosthesis should not be disregarded.

End users

Three patient populations who suffer from balance impairment due to sensory loss or degradation include vestibular patients, elderly, and peripheral neuropathy patients. Each year, there are approximately 150,000 new vestibular patients diagnosed, approximately 250,000 elderly identified as high-risk fallers, and approximately 20,000 diabetic patients diagnosed with peripheral neuropathy [68]. In theory, all of these patient populations could derive benefit from additional sensory information to supplement or replace their impaired sensory channels. The elderly for example, experience balance impairment due to partial loss of vestibular function (degeneration of vestibular hair cells), vision (formation of cataracts that cloud vision), and/or proprioception (loss of sensory capabilities in our lower limbs). Patients with peripheral neuropathies suffer symptoms ranging from numbness, loss of feeling, and reduced ability to stand or walk.

Who will be the end user of such devices? Vibrotactile sensory orientation feedback was pioneered by Rupert et al. [69] for aviation purposes to provide a true gravity vector cue.

This form of sensory substitution is not limited to providing only motion cues for balance-impaired. Its greatest utility to date is in providing a cue of verticality, which can be used by healthy individuals to supplement their natural senses in extreme situation. For example, imagine a soldier who has just traversed the ocean and is suffering from sea legs. He is then flown to a desert location and parachutes to land in a dark sandy environment in which both his vision and proprioception are compromised in addition to the lingering effects of sensorimotor disruption from time at sea.

The device also has potential for use as ergonomic aids, interactive entertainment, and sports related applications such as snowboarding and golfing. Rehabilitation and teaching can be enhanced through the use of such a device if the motions of the rehabilitation specialist or teacher are mapped directly to the patient or student in the form of vibrotactile cues.

Cognitive workload

Cognitive workload has not been a problem for subjects using the balance device during standing tasks. However, our pilot study suggested that it may be an issue during gait tasks. Subjects in Study III tended to decrease their gait velocity when the device was turned on because they wanted to try and make use of the feedback information to make trunk roll tilt corrections. Some subjects mentioned that the vibrations were occasionally a distraction. These subjects would prefer not to receive information if their M/L tilt fell within the nominal range. Others preferred the constant lower row of tactors rhythmically vibrating twice per gait cycle because it indicated to them that the device was functioning (and therefore they could rely on obtaining information in a balance crisis). Study III seemed to suggest that two levels of magnitude feedback are enough for walking tasks. This may also be the case for standing tasks and the data collection from Study II could be reanalyzed to address the percentage of times that the third level of tactors were actually used to initiate a corrective response. If a total of three states are sufficient (off, small tilt, large tilt), different tactors could be used that encode magnitude using frequency (e.g. sum and difference tones to create beat frequencies). This would be beneficial from an economic perspective because it would limit the number of tactor elements needed per device and decrease the number of parts that could potentially incur damage with use. Cell phone vibration modes employing multiple tone qualities have proven to be discernable by the user (style/tone of vibration indicates who is calling). Research in the MEEI Jenks Vestibular Diagnostic Laboratory is currently testing the reaction times of frequency encoding tactors.

Implantable device

In theory, both an implantable device and wearable sensory substitution device could be used simultaneously in certain situations given that neither an implant nor a sensory substitution device alone are a complete solution to a vestibular-comprised system. Therefore, it would be interesting to investigate postural behavior when both are used simultaneously. Wall et al. [70] have successfully elicited vestibuloocular reflexes in response to multiphasic pulse trains of electric stimulation to the posterior ampullary nerve of two subjects prior to translabyrenthine labyrinthectomy at the University Hospital of Geneva. Video oculography was recorded subjects' eye movements. Vestibulo-ocular reflexes were observed when subjects were stimulated following surgical exposure of the nerve canal (estimated bone thickness of approximately 100 microns). The stimulating pulse was characterized by an initial 200 microsecond negative phase, followed by neutral and positive phases of the same duration. The duration of the fourth and final phase of each multiphasic pulse was systematically varied (pulse repetition rates ranged from 25 to 400 pulses per second). This stimulus produced a robust vertical nystagmus. No change in the slow component velocity of the horizontal slow component was observed. This experiment is a first step in demonstrating the feasibility of a vestibular prosthesis in humans. Perhaps it will be possible in the near future to electrically stimulate the nerve bundle of the vestibular portion of the 8th cranial nerve in a patient under local anesthesia prior to a vestibular nerve section while performing posturography. Postural responses due to externally applied electrical stimulation could be analyzed and compared to trials both with and without vibrotactile feedback.

Dizzy Patients

To date, none of the published groups working on electrotactile, vibrotactile, or auditory biofeedback have presented data from patients suffering from vertigo, affectionately termed here as "dizzy" patients. To gain insight into this question, we invited one idiopathic dizzy patient who complained of vertiginous symptoms to try the vibrotactile balance prosthesis during slow-paced, self-paced, and perturbed walking. Partway through the experiment, the patient complained of feeling dizzy but continued with the gait trials. As the symptoms increased in intensity, the patient verbalized that although she was aware that the vibrotactile elements were vibrating, she was no longer paying any attention to the information because she was already overwhelmed dealing with her vertigo. The patient reported a neutral rating regarding the usefulness of the device. The usefulness of a balance aid in mitigating vertiginous symptoms is unknown. Based on approximately twenty vestibular patient interviews, half of which suffered from vertiginous attacks, several individuals reported that they use physical tactile cues of verticality when asked what strategy, if any, helped them cope with the debilitating effects of dizziness. Two reported strategies included: 1) women grabbing a hold of their purse straps because "purse straps hang down" (straps align with the gravitational vector) and 2) both men and women putting their hands in their pockets because their pockets represent "down" (a known vertical direction indicator). Therefore, perhaps patients who use verticality as a cue to reorient themselves during a vertiginous attack may derive benefit from a wearable device that confirms that they are not actually tilting/spinning.

In conclusion, sensory substitution devices offer a promising means of temporarily or permanently replacing missing or impaired information about body orientation and motion. The results to date indicate that such information can be perceived, processed, and acted upon during tasks involving stance and gait. Numerous research questions must be addressed before a commercial device is made available to clinical populations. This emerging field of research has the potential to improve our society's ability to manage patients with balance-related deficits in addition to offering exciting possibilities in the rehabilitation, military, sports, and entertainment arenas.

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Appendix I Human Subjects

Justification for use of human subjects

The use of human subjects is justified in this investigation because the fundamental scientific questions posed require bipedal biomechanical gait responses to perturbations during locomotion, which preclude the use of animal models. Additionally, the technology under development is intended solely for the assessment and diagnosis of patients with abnormal gait and astronauts.

Subjects

Normal healthy subjects employed for use in addressing Specific Aims 1-3, which will be conducted at Boston University's Neuromuscular Research Laboratory, will be adults selected from the graduate student and staff population at Boston University, Massachusetts Institute of Technology and Massachusetts Eye & Ear Infirmary and/or the greater Boston area.

Vestibulopathic patients for Specific Aims 1 through 3 will be selected from the greater Boston area based on recommendations from collaborating physicians Drs. Rauch and Lewis at Massachusetts Eye & Ear Infirmary. Patients with acute unilateral and bilateral vestibular pathologies will be selected according to clinical vestibular examination results.

Subjects will be aged matched to the astronaut population (between 28 and 55 years old). Subjects will not be excluded on the basis of their gender, race, or ethnicity. Subjects will receive a thorough briefing regarding the experimental protocol and associated potential risks before being presented legal documentation indicating informed consent. Informed consent will be obtained on the day of the scheduled experiment and subjects will be instructed that they may withdraw from the study at any time without penalty. All subjects should be competent to give informed consent. Normal subjects will be screened for vestibular, neurological, cardiovascular, orthopedic or traumatic disorders. Additionally, all subjects will be asked to abstain from caffeine and alcohol 24 and 48 hours preceding the experiment, respectively.

Subject Information and Consent Forms will be approved by the Massachusetts Eye & Ear Infirmary, Boston University, and Massachusetts Institute of Technology for all experiments associated with Specific Aims 1-3. The original copy of the Consent Form will be confidentially stored with the subject's records at the Jenks Vestibular Laboratory. All subjects will receive a photocopy of their informed consent documentation and paperwork describing the experimental protocol and potential risks.

Subject's written medical records and video tapes will be stored in locked files or locked cabinets and will be released only with written permission from the subject. Information extracted from medical records (if applicable) will be treated in the same fashion as data gathered during experimentation; subjects will be identified by name. Code sheets associating test results with a particular subject will be stored in locked files. Loss of confidentiality is therefore highly unlikely.

Risks

The identifiable risks fall into the category of reasonable physical risks. "Reasonable risk" means that the probability and magnitude of harm or discomfort anticipated in the research are greater in and of themselves than those ordinarily encountered in daily life or during the performance of routine physical or psychological examinations or tests, but that the risks of harm or discomfort are considered to be acceptable when weighed against the anticipated benefits and the importance of the knowledge to be gained from the research. No alternative procedures that provide the same information exist to our knowledge. There are no known psychological, social, or legal risks associated with the experiments described in Specific Aims 1-3.

Potential Hazard: Fall during posturography testing
Causes: Disorientation
Effects: Subject/personnel injury
Assessment: Severity = Critical, Probability = Moderate for normals, high for patients
Protection to minimize risks:
A spotter will support the subject while the test is performed. The subject will also be wearing a harness that can be used to support the subject if a fall occurs. These clinical tests are routinely conducted on patients at the Jenks Vestibular Diagnostic Laboratory (>1000 patients/year).

Potential Hazard: Motion sickness
Causes: Conflicting sensory stimuli during posturography test and/or vestibulo-ocular reflex test and/or BALDER test
Effects: Discomfort, nausea, headache, pallor, sweating, fatigue
Assessment: Severity = Low; Probability = Moderate for normals, high for patients
Protection to minimize risks:
The posturography test will be conducted using the NeuroCom EquiTest apparatus. The vestibulo-ocular reflex (VOR) tests used in the Jenks Vestibular Diagnostic Laboratory consist of rotation in the dark and optokinetic stimulation. The locomotion and postural stability tests consist of standing on or walking across a moveable force plate at Boston University's NeuroMuscular Research Laboratory (BALDER). Subjects will be allowed to rest during testing if motion sickness levels reach an unacceptable level and testing will resume when symptoms subside. Water will be available upon their request.

Potential Hazard: Electric shock
Causes: Electro-oculographic (EOG) electrodes used to record eye movements during benign paroxysmal positional nystagmus and vertigo tests (BPPN) and vestibulo-ocular reflex clinical examinations
Effects: Subject/personnel injury
Assessment: Severity = Critical; Probability = Extremely Low
Protection to minimize risks: The risk of electrical shock from EOG recording systems has been minimized by appropriately designing and testing isolation amplifiers and other

recording circuits in compliance with established standards. Recording systems are routinely inspected and regularly maintained to insure that in the event of equipment failure, the potential electrical current passing from the equipment to the subject would be significantly less than the level necessary to inflict physical damage or pain. All amplifiers are isolated. The EOG recording system is routinely used on patients at the Jenks Vestibular Diagnostic Laboratory (>1000 patients/year). It is therefore highly unlikely that subjects will receive electrical shocks during testing.

Potential Hazard: Stress to the neck and back
Causes: Rapid movement of the subject from a sitting position to a supine head orientation during benign paroxysmal positional nystagmus and vertigo (BPPN) testing
Effects: Subject/personnel injury
Assessment: Severity = Critical; Probability = Extremely Low
Protection to minimize risks: Subjects with a history of neck or back strain, dysfunction or injury will not undergo Hallpike examination. The Hallpike Maneuver is routinely conducted on patients at the Jenks Vestibular Diagnostic Laboratory (>1000 patients/year).

Potential Hazard: Fall(s) during BALDER testing
Causes: Disorientation from perturbation
Effects: Subject/personnel injury
Assessment: Severity = Critical; Probability = Extremely Low
Protection to minimize risks: The locomotion and postural stability tests consist of standing on or walking across a moveable force plate at Boston University's NeuroMuscular Research Laboratory (BALDER). A spotter will follow the subject while the test is performed. The subject will be wearing a harness that can be used to support the subject if a fall occurs. The moving BALDER platform at Boston University has been used for at least three years without injury to any subject. The BALDER testing apparatus is equipped with a safety harness similar to the one used for routine clinical testing.

Study I

Recovery trajectories of age matched subjects after perturbations during locomotion

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Abstract
Postural recovery from controlled perturbations during locomotion was recently evaluated as a potential diagnostic of subtle vestibular deficiency. Wall et al. showed that well-compensated vestibulopathic subjects that had only subtle clinical test findings had significantly greater changes in their medial-lateral (M/L) foot placements compared to young healthy controls following surface perturbations; their recovery step following lateral and medial perturbation was widened and narrowed, respectively. This research completes the characterization of the M/L stability of vestibulopathic patients in response to surface perturbations during locomotion by comparing their recovery trajectories to those of age-matched controls. Surface perturbations were delivered in one of two directions to the right stance foot during paced walking trials. Additionally, subjects completed sets of non-perturbed natural and paced gait trials. M/L stability was quantified using kinematics data to calculate (1) the lateral projection of the mechanical moment arm formed between the stance shank and the sternum, (2) the root-mean-square trunk sway, (3) the within-trial step width variability, and (3) the head and trunk stabilization strategies. M/L moment arm responses differed among groups based on where in the gait cycle the moment arms were calculated. At the anterior-posterior (A/P) shank crossing event following the large backward left perturbations, vestibulopathic patients had significantly smaller (more exaggerated) responses compared to young controls (but not age-matched controls). On the other hand, age-matched controls had significantly smaller (more exaggerated) responses compared to young controls (but not vestibulopathic patients) at the heel strike event following perturbation. Step width variability during non-perturbation paced gait trials showed the most pronounced difference among subject groups, with vestibulopathic patients exhibiting a significantly greater variability than either age-matched or young controls. Thus it appears that step width variability may be a more effective discriminator than recovery from surface perturbations for detecting subtle vesibulopathies.

Background

Various central and peripheral vestibular disorders beget postural imbalance and lead to compromised gait characterized by unsteadiness, direction-specific deviation, and falls [1, 2]. Gait instability resulting from compromised vestibular function is not only characteristic of patients with central and peripheral vestibulopathies, but also the elderly (gradual decrease in vestibular hair cell densities as a function of age [3]), and astronauts returning from long-duration space flight (altered sensory weighting schema). Subsequently, many central and peripheral nervous system diseases can be initially recognized by their impact on posture and gait [4]. Inadequate sensory information necessary to trigger and modify postural responses is a major contributor to disequilibrium and can identified by clinical evaluation of sit-to-stand, standing, response to perturbation and turning tasks [5]. Typically, patients suspected of having balance control disorders stemming from central or peripheral vestibular disorders undergo clinical evaluations that examine posture, postural reflexes, and gait parameters such as stance width, body sway, and gait velocity [4]. Clinical tests that assess the integrity of the vestibular system are typically comprised of the following exams: electronystagmometry (ENG) test battery (pursuit/random, saccades, spontaneous nystagmus, positional maneuvers, caloric stimulation), rotation about a vertical axis, Romberg exam, and computerized dynamic posturography.

Postural and locomotor responses of astronauts returning from long duration space flight have been qualitatively and anecdotally compared to those of well-compensated vestibulopathic patients. Long-duration exposure to the stimulus of microgravity, and to some extent, short-term exposure, results in alterations to sensorimotor integration [6]. Post-flight postural control is characterized by increased dependence on vision throughout the sensorimotor rearrangement process (otolith and proprioceptive), which occurs concomitantly with re-adaptation to a gravito-inertial environment [7]. Following long-duration space flight, astronaut gait is characterized by exaggerated medial-lateral foot placement, trunk shifts to the side of the supporting leg, and failure to maintain the intended path [8]. Recovery usually occurs in a step-wise manner; rapid improvement over the first 8 to 10 hours post-flight and a more gradual return from to pre-flight stability levels over the next 4 to 8 days [9]. Despite returning to preflight postural testing baselines, Shuttle and MIR crewmembers continue to report symptoms indicative of postural instability up to several months following landing. A more sensitive laboratory test is desired to capture the slow recovery of sensorimotor function and quantify the qualitative post-flight reports of postural instability.

Oddsson et al. recently proposed a standardized, repeatable and safe gait perturbation protocol as a potential test for eliciting subtle vestibular and/or sensorimotor deficits [10]. Wall et. al demonstrated that it was possible to quantify the locomotor stability of well-compensated vestibulopathic patients by examining their locomotor recovery from controlled perturbations during gait [11]. Medial-lateral (M/L) stability was quantified by estimating the length of the M/L stance (change in the support moment arm) between the support foot and the sternum, an approximation of the postural control system's impulse response during locomotion. The vestibulopathic group had significantly greater changes in their moment arm responses compared to young healthy controls and required a greater number of steps to return to normal pre-perturbation gait.

One significant independent variable that remains to be examined is the effect of age. Numerous studies have shown that posture and gait stability are negatively affected by age [12-24]. Specific to this study, elderly subjects have been shown to have decreased M/L control of compensatory stepping movements [17] and increased trunk roll stiffness [25]. Therefore, it is possible that some of the effects observed by Wall et al. were in part due to age and not purely vestibulopathy. This study seeks to compare the recovery responses of well-compensated vestibulopathic patients to age-matched controls in order to comprehensively evaluate controlled surface perturbations as an indicator of subtle sensory deficits.

Methods

Equipment

Data were collected in the Injury Analysis and Prevention Laboratory in the NeuroMuscular Research Center at Boston University. A custom-built moveable BALance

DisturbER (BALDER) 2.1 square meter platform generated a programmable stimulus and formed part of the 18 m raised wooden walkway (Figure 1). The primary components of the BALDER platform include a force-plate (ORG-6 AMTI, Newton, MA, USA) imbedded in a wooden platform, two AC-servo motors controlled by two linear servo drivers, two high precision linear position transducers (Novotechnik, Germany), and a 16 channel A/D - two channel D/A data acquisition board (Microstar 3200e/415). The raised wooden walkway consisted of a short staging area leading up to the BALDER platform followed by a long additional walkway extension. Both the staging area and the walkway extension were 1.2-m wide. Kinematics were collected using the Optotrak 3020 system (Northern Digital, Waterloo, Ont.). Rectangular arrays consisting of six infrared emitting diodes (IRLED) were placed on the subjects' mid tibias, pelvis, sternum, and head (Figure 2). A single marker was placed on the BALDER platform to confirm perturbation timing, direction and magnitude during data analysis. The IRLED sampling rate was 1500 Hz and the array positions were estimated at 40 Hz. The 3020 camera was placed at the end of the BALDER platform walkway and captured a viewing range of approximately 12 m.

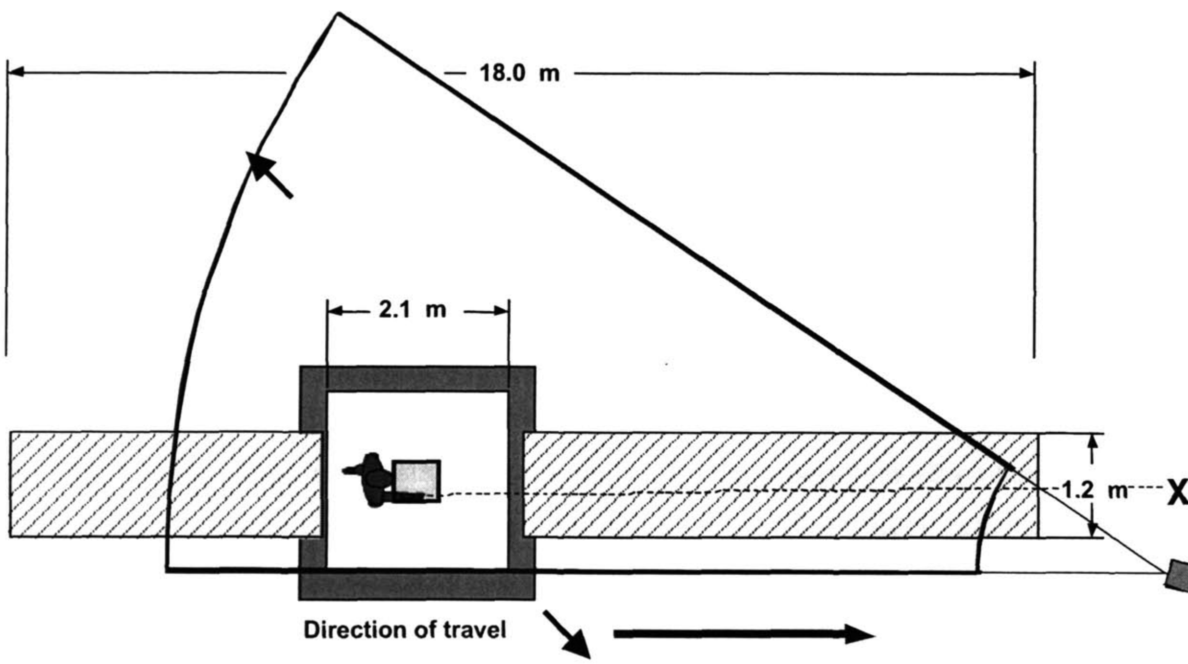


Figure 1. BALDER platform and wooden walkway

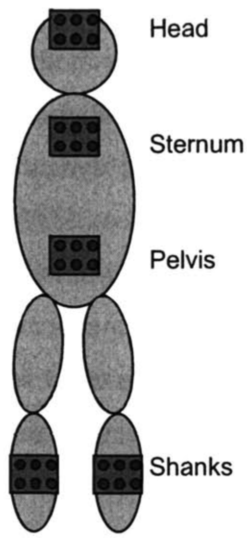


Figure 2. Schematic of IRLED marker placements

Subjects

Eight subjects (6 male, 2 female) with no known history of orthopedic, neurological, or vestibular disorders were selected for the age-matched control group. The mean age of this group was 57.87 ± 10.3 years (age range 45-74 years). Subjects underwent standard vestibular diagnostic testing at the Massachusetts Eye and Ear Infirmary (MEEI) Jenks Diagnostic Vestibular Laboratory for vestibular function assessment. Subjects were excluded from the age-matched control group if they had abnormal findings ENG test battery (pursuit/random, saccades, spontaneous nystagmus, positional maneuvers, caloric stimulation), rotation, or Computerized Dynamic Posturography findings. Informed consent was obtained prior to participation in this study. The Massachusetts Eye & Ear Infirmary, Boston University, and Massachusetts Institute of Technology Institute Review Boards approved the experimental protocol.

Data from two prior experiments involving vestibulopathic patients [11] and young healthy controls [10] were reprocessed and reanalyzed for comparison with the age-matched controls. Eight vestibulopathic subjects (5 males, 3 females) with only subtle findings in clinical tests of postural stability, composed the patient group. Their mean age was 53.4 ± 12.6 years (age range 31 - 68 years). All patients had unilateral vestibular loss (100% Reduced Vestibular Response asymmetry from the caloric test) resulting from surgery for vestibular schwannoma. Detailed information about the vestibulopathic subjects are presented in Table 1. Despite their known vestibulopathy however, all subjects scored within the normal range on the computerized dynamic posturography (CDP) Sensory Organization Tests (mean score 72.67). The mean SOT 5 and SOT 6 scores (parentheses indicate 5th percentile) for persons aged 20-59 yrs is 69 (52) and 67 (48), respectively. These tests are designed to make A/P sensory inputs unreliable while standing. The young control group included 12 healthy subjects (11 male, 1 female) with a mean age of 35 +- 9 years (age range 26-57 years). Age and gender information about the young and age-matched controls are listed in Table 2.

Table 1. Vestibulopathic patient demographics

Age	Gender	Side of Tumor	Time since Surgery (mo)	SOT Score	MCT Score	%RVR	VOR Midrange Gain	VOR Time Constant (s)	UVH or (pBVH)
58	F	right	98	69	155	100	0.79	3.3	(-0.036)
56	M	left	24	70	155	100	0.86	6.5	UVH
56	M	right	107	73	148	100	0.79	7.8	UVH
40	M	left	13	70	129	100	0.71	8	UVH
68	M	left	16	68	146	100	N/A	N/A	N/A
31	M	left	5	71	154	100	0.76	6.3	UVH
51	F	right	90	74	144	100	0.61	9.9	UVH
67	F	right	126	78	128	100	0.49	4.7	(-0.0129)

Legend:
RVR – Reduced vestibular response to bilateral, bithermal caloric stimulation. All but one subject had a 0°/s nystagmus response to ice water in the side-of-tumor ear. One subject had a 5°/s response.
SOT – Sensory Organization Test: Normal mean composite scores are 80.2 for 20-59 yrs and 76.9 for 60-69 yrs, 5th percentile (abnormal) limits are 68.5 for 20-59 yrs and 70.0 for 60-69 yrs
MCT – Motor Control Test: Normal mean composite scores are 143.0 for 20-59 yrs and 151.8 for 60-69 yrs
5th percentile (abnormal) limits are 161.0 for 20-59 yrs and 170.8 for 60-69 yrs
VOR – Vestibuloocular reflex
N/A – Not available
* See Dimitri *et. al.*, 1996, UVH or (pBVH) – Unilateral (UVH) or bilateral vestibular hypofunction, based upon Dimitri *et. al.*, 2002. If patients scored as bilateral hypofunction (BVH), then the probability of this occurring by chance is given in parentheses.

Protocol

The perturbation protocol consisted of 30 paced walking trials, during which surface perturbations were delivered at two different amplitudes (5-cm and 10-cm) and in opposite directions. The directions were +45° (forward-right, FR) and +225° (backward left, BL) as measured clockwise from the subject's direction of march (refer to black arrows in Figure 1). The perturbations were applied to the right foot during single leg stance following a right heel-strike force-plate threshold triggered delay (100 ms or 120 ms depending on the translation direction) to ensure that the subject's left leg was in its swing phase. One control case (no perturbation) was included in the experimental protocol to make perturbation trials unpredictable. Perturbations occurred in approximately half of the trials. Subjects were neither informed of when a perturbation would occur nor the exact number of trials to be performed during the experimental session to prevent subjects from predicting perturbation direction. Table 3 shows the perturbation protocol. BALDER was programmed to accelerate and decelerate at a constant rate of 5 m/s^2. Maximum velocity was reached halfway through the perturbation and its magnitude was the consequence of the acceleration and displacement settings.

Age	Gender	Group
34	M	Young
29	M	Young
34	M	Young
30	M	Young
40	M	Young
26	M	Young
29	M	Young
45	M	Age-matched
33	M	Young
31	M	Young
57	M	Age-matched
39	M	Young
24	M	Young
24	M	Young
24	M	Young
63	M	Age-matched
49	F	Age-matched
50	M	Age-matched
70	M	Age-matched
74	M	Age-matched
55	F	Age-matched

Table 2. Young and age-matched control demographics

Table 3. Perturbation protocol

Trial Number	Trial Type	Trial Number	Trial Type	Trial Number	Trial Type
1	Large FR	11	Paced Gait	21	Large BL
2	Paced Gait	12	Paced Gait	22	Paced Gait
3	Large FR	13	Small BL	23	Paced Gait
4	Paced Gait	14	Paced Gait	24	Small BL
5	Small BL	15	Large BL	25	Paced Gait
6	Paced Gait	16	Large FR	26	Large FR
7	Small FR	17	Paced Gait	27	Small BL
8	Paced Gait	18	Small FR	28	Paced Gait
9	Small FR	19	Paced Gait	29	Small FR
10	Large BL	20	Paced Gait	30	Large BL

The laboratory was dimly lit to minimize visual cues. Prior to the start of the perturbation protocol subjects completed three natural gait (non-paced, non-perturbed) and three paced non-perturbation (paced gait) trials. For paced gait trials, subjects were instructed to walk at a pace of 100 steps per minute (pace maintained by an electronic metronome) along the walkway while fixating on a faint visual target (positioned at eye-level beyond the end of the walkway).

Perturbation trials were practiced prior to data collection to familiarize subjects with the novel stimulus produced by the BALDER platform. Practice trials using both 5 and 10 cm magnitude perturbations were repeated until stutter stepping (defined as a quick corrective step) were eliminated. A safety spotter stood alongside the BALDER platform during all perturbation protocol trials.

Data Analysis

Three-dimensional (3D) linear position data were converted into six-dimensional (6D) linear and angular position data using Northern Digital RIGMAKER and Data Analysis Package RIGID software. Due to instrumentation inconsistencies between the present and the past two studies (2004, 2001, 1999), all data were reprocessed using an identical processing schemes. Supplemental Matlab scripts were developed to identify and remove inappropriately sampled (mis-sampled) IRLED 3D marker data on a sample-by-sample basis prior to converting to 6D data. Once 6D data were obtained, the data were interpolated with cubic spline curves and low pass filtered with a 3rd order phaseless butterworth filter (Matlab function filtfilt.m) with a corner frequency of 10 Hz to remove high frequency noise.

Since neither foot switches nor pressure sensors were used to detect heel strike events, two gait cycle events were used to sample the subjects' 40 Hz kinematics data. One event was estimated using shank position information while the other event used its first derivative, shank velocity. The first event, termed anterior-posterior (A/P) shank crossing occurred once per swing phase (twice per gait cycle) at the instance when the A/P positions of the two shanks were equal. Figure 3 shows the A/P shank position traces, sampled M/L shank and sternum positions, and BALDER vertical force data for one subject's trial. The second gait cycle event was estimated heel strike. Given proper foot placement on the BALDER force plate, vertical force accurately indicated one heel strike and toe off event per trial.

The force plate data were sampled at 400 Hz (1000 Hz for the vestibulopathic patient data collections). Correlations of body marker linear and angular positions, velocities, and accelerations were correlated with the single known vertical force indicator of heel strike and toe off for all subjects' trials. The A/P linear velocity trajectories of the two shanks' intersection - when differences between the A/P shank velocities equaled zero – corresponded most closely with the vertical force indicating heel strike [26]. Heel strike was reconfirmed on a step-by-step basis for each subject. Toe off could not be reliably approximated by correlating any single kinematics data with the force plate event. Figure 4 displays one subject's A/P shank velocity traces, sampled M/L shank and sternum positions, and BALDER vertical force data.

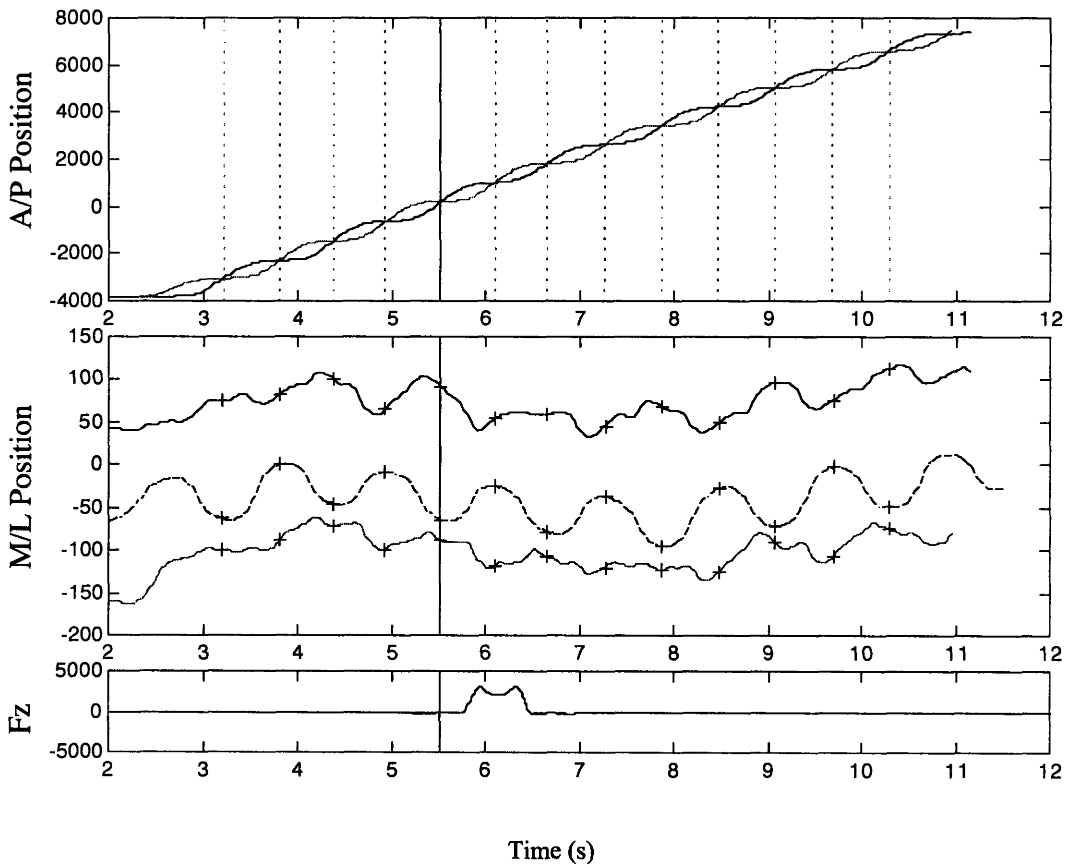


Figure 3. A/P shank crossing event detection. Top panel: A/P shank positions for the right (dark trace) and left (light trace) shanks. The solid black vertical line indicates the shank crossing event during left single stance phase immediately prior to the right heel strike on the BALDER platform. Middle panel: The M/L position traces of the right shank (dark trace), sternum (dashed trace), and left shank (light trace). Plus signs correspond to the sampled position. Bottom panel: Balder vertical force data.

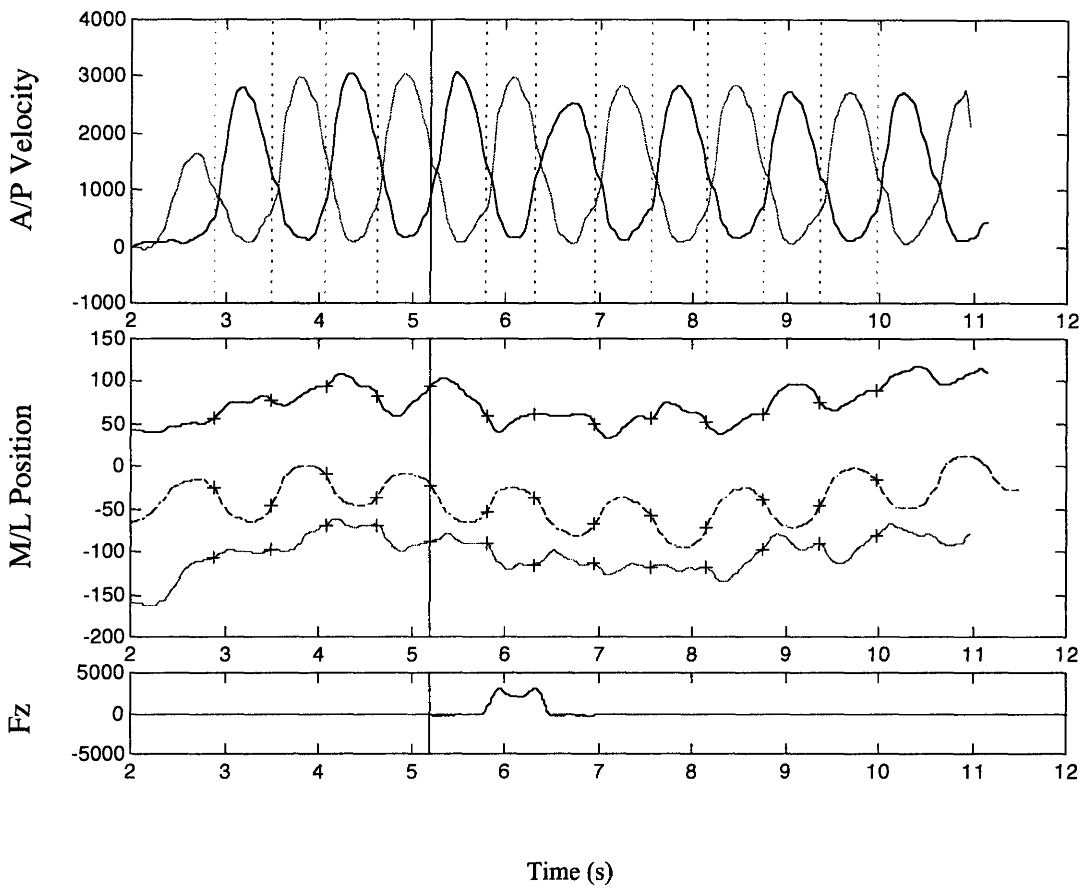
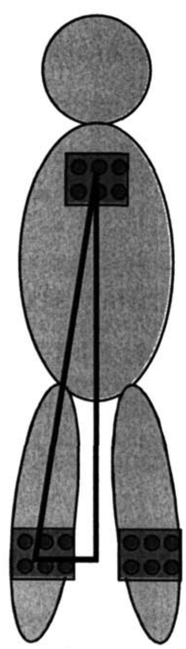


Figure 4. Estimated heel strike event detection. Top panel: A/P shank velocities for the right (dark trace) and left (light trace) shanks. The solid black vertical line indicates the left heel strike event immediately prior to the right heel strike on the BALDER platform. Middle panel: The M/L position traces of the right shank (dark trace), sternum (dashed trace), and left shank (light trace). Plus signs correspond to the sampled position. Bottom panel: Balder vertical force data.

Short steps and stutter steps were identified by comparing the time between the perturbation step and the compensatory step (step immediately following the perturbation) to the average time between steps across the trial. A short step and stutter step were defined as occurring within half and one-third of the average step time interval, respectively. Stutter steps were characterized by a quick, corrective step with the swing foot following the perturbation. Trials in which either 1) a stutter step occurred following the perturbation, 2) a left heel strike occurred on the BALDER force plate or 3) a right foot toe-off triggered the perturbation (misplaced right foot on platform) were excluded from subsequent analysis. This was done to ensure that fundamentally different perturbation responses were not intermixed.

At each above mentioned sampling event, the M/L distance between the sternum and the shank of the stance leg was calculated to provide an estimate of the M/L moment arm (Figure 5). The moment arm values for each perturbation type were averaged for each subject on a step-by-step basis. These values were then normalized using the mean values from the twelve non-perturbation paced gait trials comprising part of the perturbation protocol, in order to reduce the individual differences in moment arm magnitude between subjects. The normalized, subject mean values were then averaged on a step-by-step basis over each subject group.

Figure 5. M/L moment arm estimation



The root-mean-square (RMS) M/L sternum position was calculated from the seventh heel strike event (first recovery step following perturbation) until the twelfth heel strike to assess trunk sway. Step width was calculated by taking the difference between the shank M/L positions. Step width variability was characterized by within subject trial standard deviation and coefficient of variation, the ratio of standard deviation divided by the mean value and expressed as a percentage.

The anchoring index [27-29] was used to characterize head and trunk stabilization strategies during non-perturbed locomotion trials. Angular position data from the head, sternum, and pelvis markers spanning the fifth to the twelfth heel strike events were used for this measure. The index describes the relative angular distribution of the body segment being considered with respect to axes linked to an inferior anatomical segment. The anchoring index is defined as:

$$\mathrm{AI} = [(\sigma_r) - (\sigma_a)]/[(\sigma_r) + (\sigma_a)]$$

where, σ_a is the angular dispersion of any body segment and σ_r is the standard deviation of the relative angular position of the body segment being considered with respect to axes linked to an inferior anatomical segment. For example, a positive head anchoring index indicates a tendency for head stabilization in space rather than stabilization with respect to the trunk, whereas a negative value would indicate a tendency for head stabilization on the trunk rather than with respect to space.

Results

All post-processing was performed using Matlab (The MathWorks, Natick, MA). Statistical analyses were conducted using STATA (StataCorp LP, College Station, TX). One-way, repeated-measure analysis of variance (ANOVA) were performed on each dependent variable. The level of significance was set at $p< 0.05$.

Figures 6-9 show the mean normalized M/L moment arms for the large FR and BL perturbations sampled at both A/P shank crossing and heel strike events. Standard deviation bars were omitted to avoid clutter. The sixth A/P shank crossing corresponded to the shank crossing event when the right foot was in single support stance phase during the perturbation. The sixth heel strike corresponded to the right heel strike on the BALDER force plate that triggered the perturbation.

Statistical comparisons among groups were performed on the 7th through 10th A/P shank crossing and heel strike event moment arm magnitudes for the for small and large FR and BL perturbations.

Fig. 6

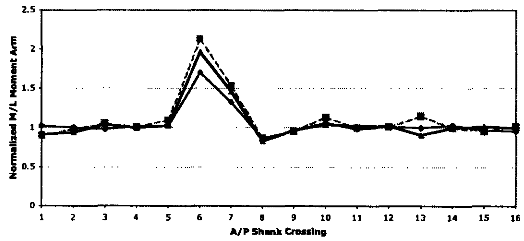


Fig. 7

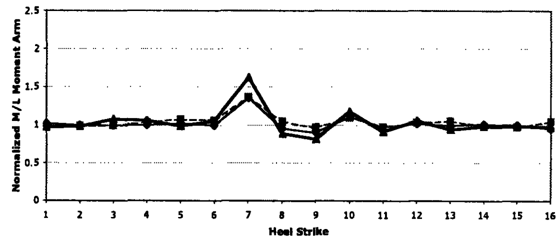


Fig. 8

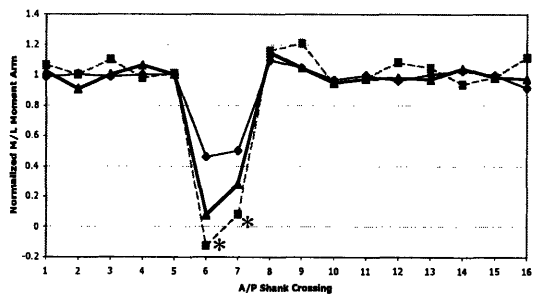
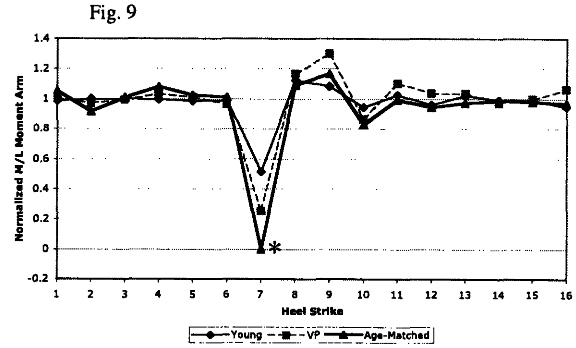


Fig. 9



Figures 6-9. Normalized M/L moment arm during support surface perturbation. The thin solid line with diamond markers represents the average responses for young controls. The dashed line with square markers represents the average responses for vestibulopathic patients. The thick line with triangle markers represents the average response for age-matched controls. The sixth A/P shank crossing corresponded to the shank crossing event when the right foot was in single support stance phase during the perturbation. The sixth heel strike corresponded to the right heel strike on the BALDER force plate that triggered the perturbation. Figure 6. The upper-left plot shows normalized M/L moment arms sampled at A/P shank crossing events for the large FR perturbation. Figure 7. The upper-right plot shows normalized M/L moment arms sampled at heel strike events for the large FR perturbation. Figure 8. The lower-left plot shows normalized M/L moment arms sampled at A/P shank crossing events for the large BL perturbation. Figure 9. The lower-right plot shows normalized M/L moment arms sampled at heel strike events for the large BL perturbation.

Moment arms sampled at A/P shank crossing events

Although not statistically significant, the vestibulopathic group showed a trend toward larger moment arm responses following the small and large FR perturbation (Figure 6) sampled at A/P shank crossing events. Only the large BL perturbation (Figure 8) revealed a significant difference among groups when moment arms were sampled using the A/P shank crossing event. The moment arm at the 6th shank crossing event was significantly different for both the large (p>0.0011) and small perturbations (p>0.0042). For the large BL perturbation, the vestibulopathic patients had a significantly smaller moment arm compared to the young controls. The age-matched subjects' mean moment arm value fell between the vestibulopathic and young groups' values (non-significant difference between age-matched and young groups. Vestibulopathic patients also had a significantly smaller moment arm values compared to the young controls when the BL perturbation stimulus magnitude was halved (p<0.003). Similarly, the 7th shank crossing event was also significantly different for both the large (p>0.0068) and small BL perturbations (p>0.0087). The vestibulopathic patients had significantly smaller moment arm values compared to the young controls.

Moment arms sampled at heel strike events

Although not statistically significant, the age-match group showed a trend toward larger moment arm responses following the small and large FR perturbation (Figure 7) sampled at heel strike. The large BL perturbation (Figure 9) was statistically significant for both the 7^{th} (p>0.0051) and ninth step (p>0.0464) corresponding to the immediate and subsequent left foot placements following the perturbation. The age-matched group had a significantly smaller moment arm compared to the young controls (p<0.004) for the 7^{th} step and a marginally significantly larger moment arm for the vestibulopathic patients compared to the young group (p<0.043) for the 9^{th} heel strike.

RMS sternum sway

Vestibulopathic patients had a smaller average sternum RMS roll sway during natural and paced gait trials compared to young and age-matched controls. On the other hand, vestibulopathic patients' sway was greater over a data segment spanning the first five recovery steps following all perturbation directions and magnitudes compared to the young and age-matched controls. Neither or these findings though, were significant. No consistent trend was observed across trial types among groups for sternum RMS pitch sway over the same five-step segment.

Body segment stabilization

Figure 10 shows sample head, sternum, and pelvis roll data from one young control subject (top panel) and one vestibulopathic patient (bottom panel). These particular paced gait trials were chosen because they clearly illustrate an anti-phase behavior for the young subject's head-sternum versus pelvis and an in-phase behavior for the vestibulopathic patient's head-sternum-pelvis displacements. Angular dispersion and anchoring indices were only calculated for the non-perturbation trials (natural gait, pre-perturbation protocol paced gait abbreviated as "paced gait pre", and interspersed perturbation protocol paced gait abbreviated as "paced gait inter").

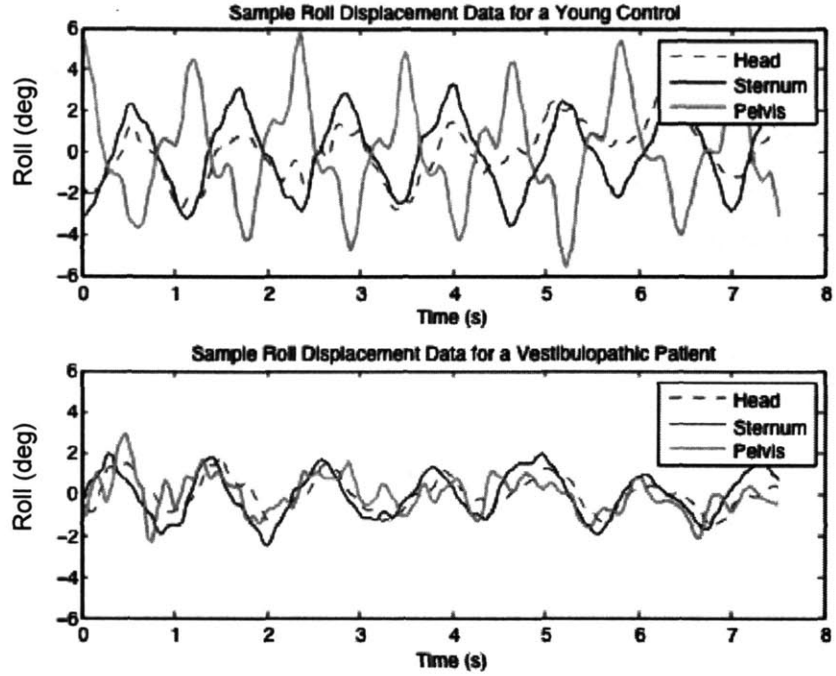


Figure 10. Sample roll displacement time series traces for the head (dotted), sternum (thin solid), and pelvis (thick solid) markers. Top panel: young subject data. Bottom panel: vestibulopathic patient data.

The head, sternum, and pelvis roll angular dispersions (roll variances), which were used to derive the head and sternum anchoring indices for all non-perturbation trial types are tabulated in Table 4.

	Head	**Sternum**	**Pelvis**
Natural Gait			
Young	1.268 (0.556)	1.267 (0.290)	1.474 (1.003)
VP	1.572 (0.470)	1.271 (0.377)	1.378 (0.583)
Age-matched	2.119 (0.625)	1.491 (0.469)	2.065 (0.727)
Paced Gait Pre			
Young	1.241 (0.482)	1.139 (0.325)	1.359 (0.677)
VP	1.467 (0.485)	1.168 (0.430)	1.374 (0.614)
Age-matched	1.967 (0.565)	1.435 (0.488)	1.869 (0.490)
Paced Gait Inter			
Young	1.165 (0.390)	1.230 (0.363)	1.469 (0.822) *
VP	1.464 (0.517)	1.311 (0.416)	1.443 (0.736) *
Age-matched	2.072 (0.496)	1.509 (0.465)	1.992 (0.434)

Table 4. Mean and standard deviation roll angular dispersions by trial type and group

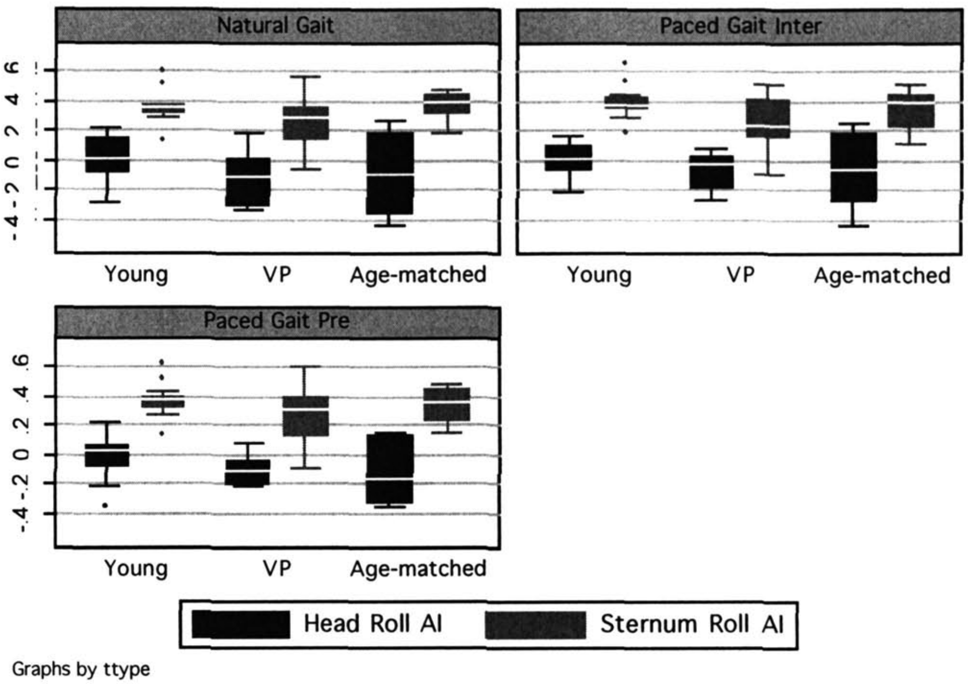


Figure 11. Box plot of the roll anchoring indices by non-perturbation trial type and group. The solid white line marks the median of the sample. The height of each box shows the range within which the central 50% of the values fall, with the box edges at the first and third quartiles. The whiskers (extended vertical lines) indicate an interval that would include 95% of the distribution if the data were normally distributed. The asterisk represents an outlier.

Box plots for the head and sternum anchoring indices results are shown in Figure 11. Neither the head nor sternum roll anchoring indices was significant among groups by trial type. In all three non-perturbation trial sets however, the median head anchoring index was least negative and not significantly different from zero for the young controls, which suggests neither a head stabilization in space nor with respect to their trunk strategy. Table 5 summarizes the significant findings for roll, pitch, and yaw angular dispersions and head and sternum anchoring indices by trial type over group. Age-matched and vestibulopathic patient sternum and pelvis pitch angular dispersions were consistently significantly larger than those corresponding to the young controls' values for all trial types. The trend in the pelvis angular dispersion was reversed with the young controls exhibiting larger yaw dispersions compared to the vestibulopathic patients and age-matched controls. Pelvis yaw dispersions however, were larger for the vestibulopathic and age-matched subjects compared to the young controls across all trials. Head and sternum pitch and yaw anchoring indices results are shown in Figures 12-15.

			Head Dispersion	Sternum Dispersion	Pelvis Dispersion	Head AI	Sternum AI
ROLL	*Natural Gait*	Young VP Age-matched	ns	ns	ns	ns	p<0.0587
	Paced Gait Pre	Young VP Age-matched	ns	ns	ns	ns	ns
	Paced Gait Inter	Young VP Age-matched	ns	ns	p<0.0282	ns	ns
PITCH	*Natural Gait*	Young VP Age-matched	ns	p<0.0000	p<0.0000	p<0.0038	p<0.0538
	Paced Gait Pre	Young VP Age-matched	ns	p<0.0000	ns	p<0.0014	p<0.0208
	Paced Gait Inter	Young VP Age-matched	ns	p<0.0000	p<0.0000	p<0.0000	p<0.0000
YAW	*Natural Gait*	Young VP Age-matched	ns	p<0.0068	p<0.0387	p<0.0507	p<0.0034
	Paced Gait Pre	Young VP Age-matched	ns	p<0.0009	ns	p<0.0017	p<0.0015
	Paced Gait Inter	Young VP Age-matched	ns	p<0.0022	p<0.0111	p<0.0109	p<0.0003

Table 5. Angular dispersion and anchoring indices significant findings

The p value for each comparison indicates a significant finding between groups. The abbreviation ns represents 'not significant'. There were no significant differences between the vestibulopathic and age-matched groups for any measure.

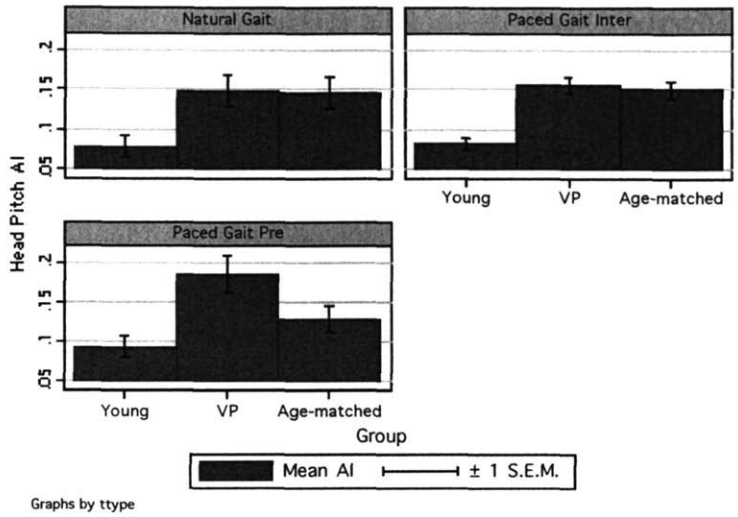


Figure 12. Head pitch anchoring index by trial type over group. Capped lines for all bar plots used in this paper indicate standard error of the mean.

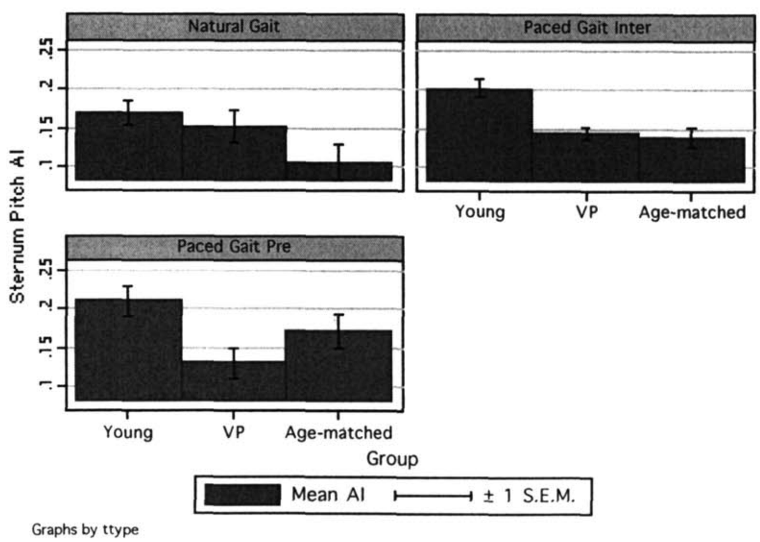


Figure 13. Sternum pitch anchoring index by trial type over group

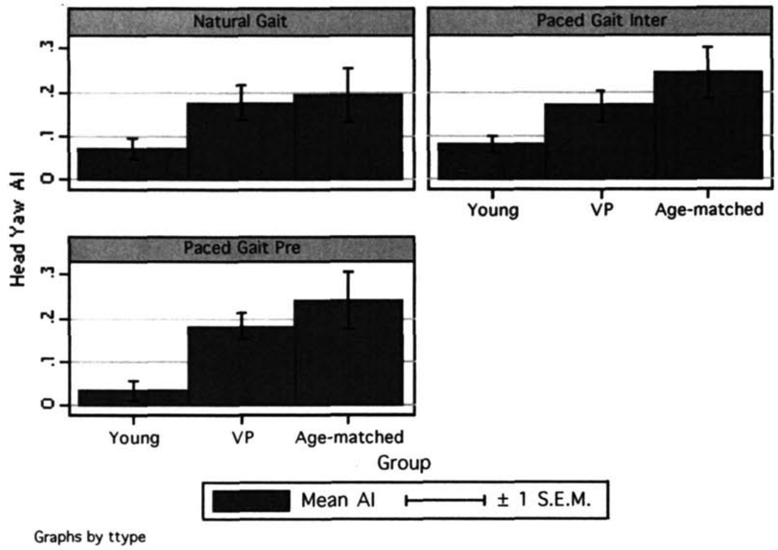


Figure 14. Head yaw anchoring index by trial type over group

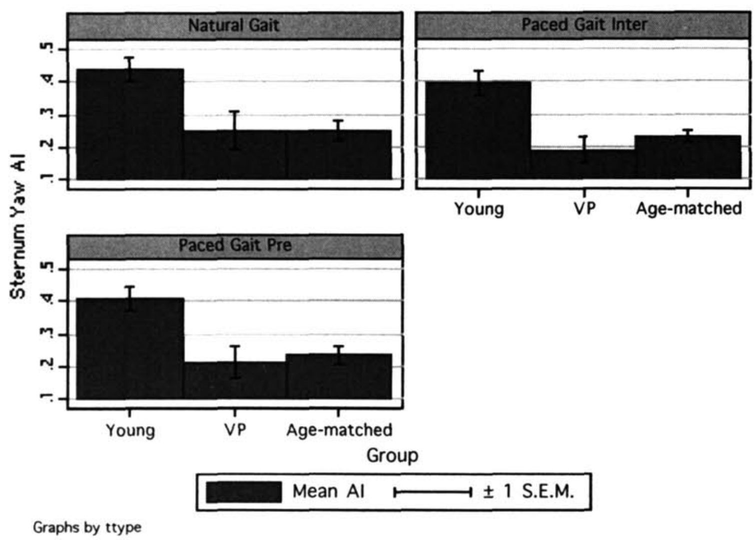


Figure 15. Sternum yaw anchoring index by trial type over group

Step width variability

Both of the step width variability parameters, standard deviation and coefficient of variation were significant among groups for all non-perturbation trials. Figure 16 shows the group standard deviation for the natural gait ($p<0.0001$), paced gait pre ($p<0.0008$), and paced gait inter ($p>0.0002$) trials. The paced gait pre trial type was the only trial type in which there was a significant difference among all groups. The young controls had significantly less step width standard deviation compared to both the vestibulopathic patients ($p<0.000$) and the age-matched controls ($p<0.007$) for natural gait. Similarly, this finding applied to the paced gait pre trials as well ($p<0.001$ and $p<0.033$, respectively). For the paced gait inter trials, the young controls had significantly less step width standard deviation compared to the vestibulopathic patients ($p<0.000$). Figure 17 shows the coefficient of variation for the natural gait ($p<0.0003$), paced gait pre ($p> 0.0018$), and paced gait inter ($p<0.0006$) trials. The young controls had significantly smaller coefficient of variation compared to both the vestibulopathic patients and the age-matched controls for natural gait ($p<0.001$ and $p<0.005$, respectively) and paced gait pre ($p<0.004$ and $p<0.012$, respectively), and paced gait inter ($p<0.001$ and $p<0.035$) trials.

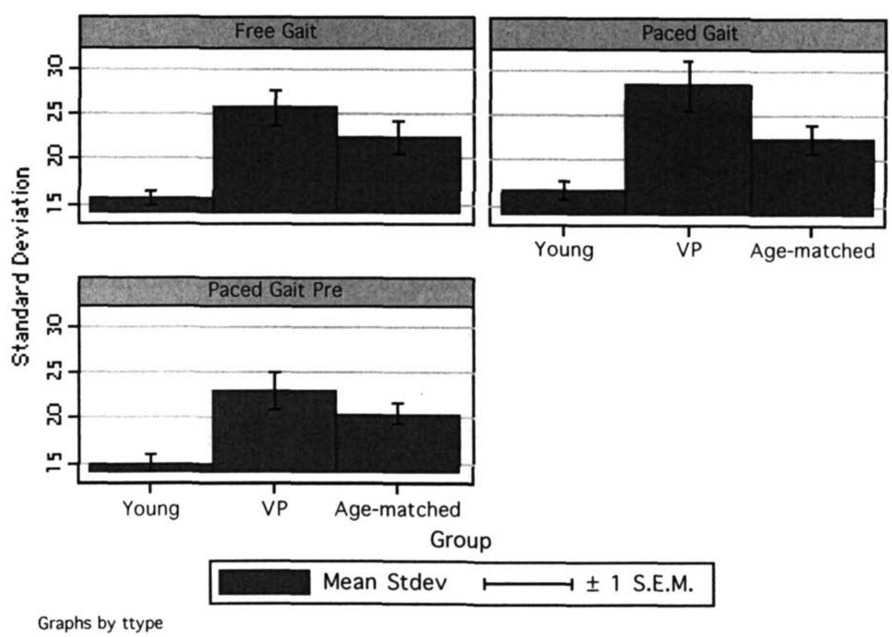


Figure 16. Step width variability

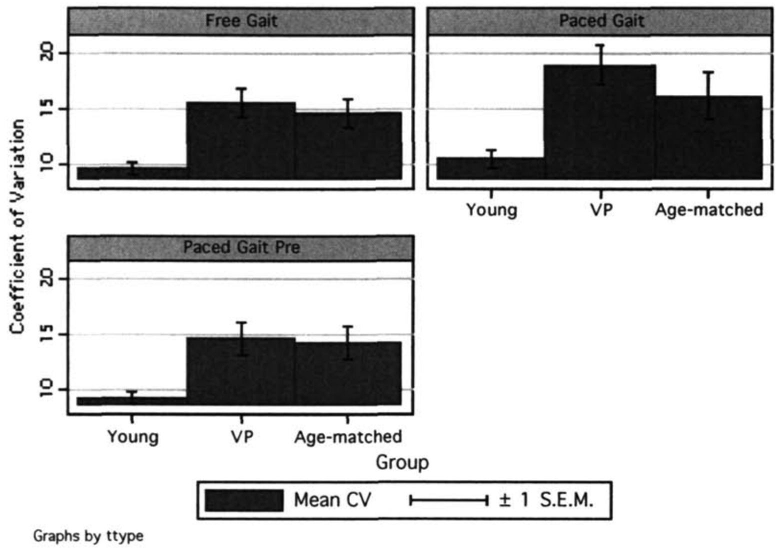


Figure 17. Step width coefficient of variation

Discussion

The purpose of this study was to determine whether or not a previously reported difference in M/L stability parameters following controlled surface perturbations were attributable to vestibulopathy or age. To review, Wall et al. reported that the vestibulopathic patient (mean age 53 yrs.) group had greater changes in their M/L moment arm responses compared to young controls (mean age 35 yrs.).

The BL perturbation elicited the greatest differences in moment arms among groups. The BL perturbation moved the right stance foot backward and to the left causing the ensuing step with the left foot to be brought inward (to the right) to resist the rightward acceleration of the trunk. Therefore, we expected the M/L moment arm immediately following the perturbation to be smaller relative to the pre-perturbation M/L moment arms. Given the known linear relationship between M/L COM acceleration and lateral foot placement [30] and the hypothesis that vestibular patients have greater difficulty in detecting acceleration, we also expected to see a difference in foot placement among groups following a perturbation.

The M/L moment arm responses among groups were dependent on the choice of sampling event. The large BL perturbation revealed a significant difference among groups when moment arms were sampled using the A/P shank crossing event. The moment arm at the 6th shank crossing event was significantly different for both the large (p>0.0011) and small perturbations (p>0.0042). When sampling the M/L sternum and shank positions at the A/P shank crossing events, the results corroborated with the previously published data and our expectations; the first two moment arms following the perturbation were significantly smaller for the vestibulopathic group compared to the young controls. However, although the age-matched controls had larger moment arms compared to the vestibulopathic group, they were still significantly smaller than the young controls' responses indicating that age was a factor in M/L moment arm responses. This finding did not hold when sampling the M/L sternum and shank positions at the estimated heel strike event. In the heel strike sampling case, the age-matched controls had significantly smaller moment arms compared to the vestibulopathic patients.

As opposed to the previous study comparing vestibulopathic patient responses to those of young controls, only the large BL perturbation elicited significant moment arm differences among groups. One reason for this finding was that in order to legitimately compare the results from all three experimental subject groups, the raw data had to be reprocessed and reanalyzed using identical techniques. To this end, a more conservative low-pass filter and interpolation scheme were employed in the reanalysis, which subsequently altered the body marker position traces from which the metrics were derived. Additionally, careful inspection of the toe-off event on the force plate (in hopes of correlating this event with a kinematics parameter) led to the realization that a small number of the BALDER perturbations were initiated by the fore-front of the right foot striking the force plate instead of the heel. If this misstep on the force plate was observed, the trial was eliminated since it resulted in a perturbation stimulus that was delivered during a different phase of the gait cycle. Finally, two additional vestibulopathic and young control subjects were included in the analysis.

It is not completely clear why the sampling event affects would affect the moment arm results. Bent et al. [31] studied whether vestibular contributions during locomotion are gait phase regulated and found significant changes in M/L foot placement timing and magnitude depending on when galvanic vestibular stimulation (GVS) was introduced during the gait cycle. In summary, they showed that changes in foot position were significantly larger at heel strike than when the stimulation was delivered mid-stance. Upper body responses for the head, sternum, and pelvis did not differ based on when the stimulus was delivered. They concluded that the largest vestibular contributions occur during the double support phase and the smallest occur during single support. Extending this conclusion to the present study, M/L moment arm sampling at the shank crossing and heel strike events correspond to single and double support phase, respectively. Interestingly, for the large BL perturbations, vestibulopathic patients had the most exaggerated moment arm response (smallest moment arm) when the data was sampled at the shank crossings events (single support phase). The age-matched subjects had the most exaggerated response (smallest moment arm) on the other hand when the data was sampled at heel strike. Sternum and COM M/L acceleration has been correlated with M/L foot placement [30]. As previously argued in Wall et al., unilateral vestibulopathic patients many have increased measurement noise in their estimate of acceleration due to their vestibulopathy. Since the largest vestibular contributions to M/L stability occur during double support phase (heel strike) and given that the patients in this study may have had measurement error due to sensor noise, additional spatial orientation information (visual and proprioceptive) may not have been completely integrated by mid-stance (A/P shank crossing) immediately following the perturbation stimulus. Therefore, these patients had more exaggerated responses at the shank crossing sampling event, but were able to adequately integrate additional sensory information and make corrective foot placements that more closely resembled those of young controls.

A more obvious explanation might be that vestibulopathic patients have adopted a cautious gait strategy that makes use of the swing limb as ballast – a mechanical moment that can be applied to shift weight (similar to the extension of arms). To gain a better appreciation for the actual swing leg trajectory, a follow-up analysis investigated the step width variability of the swing foot during single support phase of the contra-lateral foot. This analysis revealed that stance width variability during single support phase was significantly smaller for young controls compared to vestibulopathic patients and age-matched controls for all trial types (natural gait, paced gait, and perturbed gait). Although not significant, age-matched controls had less step width variability across all trial types compared to vestibulopathic patients. The large backward left perturbation was the only trial that provided the greatest statistical ($F(2,28)=12.76$, $p<0.0001$) stratification among groups; young controls had a significantly smaller step width variability compared to both vestibulopathic ($p<0.000$) and age-matched controls ($p<0.022$).

We originally proposed that recovery trajectories from controlled surface perturbations would elicit greater differences between the groups compared to the non-perturbation trials because patients with subtle vestibulopathies have developed sufficient compensatory strategies to cope with the challenges of straight and level locomotion. We assumed that unexpected controlled surface perturbations would be likely to reveal sensorimotor problems; this technique is somewhat analogous to using impulse responses to characterize the dynamics of linear systems. Although quantifiable differences exist between the responses of vestibulopathic patients and young controls to surface perturbations during gait, a simpler, less provocative test may indeed be more desirable for some severely balance-compromised individuals such as vestibulopathic patients and long-duration post flight astronauts. Sensorimotor functionality varies among patients status post vestibular surgery and astronauts immediately following landing; surface perturbations may not be tolerable or even possible post-op or in the initial post flight testing days. Therefore, if a simple walking test were 1) capable of indicating vestibular function and 2) able to be consistently performed throughout the recovery stages of a surgical procedure or long-duration space flight, it would in many cases be preferable to a surface perturbation protocol. To this end, lateral shank position variability and body stabilization parameters were analyzed for all non-perturbation trials.

We compared the control and vestibulopathic groups by examining basic parameters of M/L stability during the non-perturbed natural and paced gait trials to determine whether or not one can distinguish between the two populations without applying a surface perturbation. Step-width variability has been found to increase with age [32, 33]. Induced stumbles during treadmill locomotion required a minimum of three recovery [34] steps. Owings and Grabiner showed that step width variability is a more meaningful descriptor of locomotion control than step length variability or step time variability [35]. The standard deviation of step width within trial and a related measure, the step width coefficient of variation were significantly larger for the vestibulopathic patients compared to the young and age-matched controls in this study. Although we observed a difference in our step width variability measures, it is important to note that large number of steps are required to calculate a stable measure of step variability during both overground [36] and treadmill locomotion [37].

Additionally, groups differed in body segment stabilization strategies, although not along the expected axis. Since M/L stability is actively controlled during locomotion, we expected to see differences in roll among subject groups. The well-compensated vestibulopathic patients used in this study did not appear to exhibit dynamics mimicking a single inverted pendulum (locking of body segments to one another) instead of hierarchal stability of the head with respect to space (to provide a stable visual platform) in the frontal plane. However, groups showed the greatest difference along pitch and yaw axes. Mean values for head roll were positive for all subject groups indicating a universal adoption of a head stabilization in space strategy. Interestingly enough, changes in pitch and yaw anchoring indices were most pronounced during the pre-perturbation protocol paced gait trials. During these trials subjects were forced to walk to the beat of a metronome at a fixed and consistent pace. Additionally, they were aware that they had no risk of experiencing a perturbation during these trials.

Vestibulopathic patients with left and right-sided lesions have directional tilt differences following roll perturbations during quiet standing [38]. Specifically, they lean towards the contra-lateral lesion side. Patients with acute vestibular neuritis show a direction-specific deviation of gait towards their affected ear when instructed to close their eyes and walk slowly [39]. However, when asked to increase their walking speed, they straighten out their path. We did not find a statistically significant difference in any of the reported measures between left and right-sided lesion patients in this study. This is likely due to the fact that the pace used in this study was sufficient to straighten out any potential direction-specific deviations existing at slow paces.

In developing a post-flight test, it is remains to be seen whether vestibulopathic patients are appropriate analogues for post-flight astronauts. Based on the analyses performed to date, it is not clear that surface perturbations during locomotion produce more easily observable differences between healthy and well-compensated vestibular-deficient individuals. Step width variability during paced gait showed the most pronounced difference among subject groups.

Acknowledgements

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Study II

Assessment of multi-directional vibrotactile feedback on postural performance during multi-directional support surface perturbations

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Abstract
Single-axis vibrotactile feedback has been shown to significantly reduce the root-mean-square (RMS) sway in vestibulopathic patients during single-axis perturbation. This research examines the effect of multi-directional vibrotactile biofeedback on postural sway performance during multi-directional surface perturbations. Eight well-compensated vestibular-deficient patients donned a multi-axis vibrotactile prosthesis that mapped tilt estimates onto the patients' torsos in a 3 row by 16-column tactor array. The number of columns displayed was varied depending on testing condition. The patients used the additional sensory information provided by the vibrotactile prosthesis to augment their compromised vestibular and native proprioceptive and auditory cues during continuous support surface perturbations. A subset of these patients returned at a later date to complete a similarly structured protocol that involved discrete perturbations. Root-mean-square tilt, elliptical fits to trajectory areas, percentage of time spent outside a one-degree dead-zone, center of pressure, and anchoring index parameters were used to assess the efficacy of the multi-directional vibrotactile prosthesis. Four vibrotactile tactor conditions in addition to the tactors off configuration were evaluated. The findings indicate that the patients had significantly reduced RMS sway ($p < 0.0003$) and significantly smaller elliptical fits of the trajectory area ($p < 0.0000$) in the tactors on configuration versus the tactors off condition. Additionally, patients spent significantly less time outside of a one-degree dead zone in the tactors on condition versus tactors off condition. No significant difference was found among column configurations. The results show that among the displays evaluated in this study, there is not an optimal tactor column configuration for standing tasks involving continuous and discrete surface perturbations. Furthermore, subjects perform worse when erroneous information is displayed and do not appear to alter their head on trunk dynamics when the balance device is activated.

Background

Postural imbalance can result from various vestibular (central and peripheral), neurological, orthopedic and vascular disorders, as well as sensory conflicts, head injuries, infections, medications, aging, and space flight [1, 2]. According to a report by the National Institute on Deafness and Other Communication Disorders, balance problems are among the most common reasons that older adults seek medical attention reported in nine percent of the population aged 65 years and older [3]. Even more astounding, 40% of individuals in the United States are estimated to experience dizziness requiring medical attention. Balance disorders are a major cause of fall-related injuries to the elderly, resulting in patient care costs exceeding $8 billion per year [1]. Of particular concern are statistics regarding post-fall mortality rates: one study of nursing home residents has shown the mortality rate to more than double in the year following a fall [4]. According to the U.S. Census Bureau 2000, America's "baby boomers," those born between 1946-1964, comprise the largest aging population in the United States [5]. Given the large percentage of both vestibulopathic patients and older individuals that experience balance problems and the desire of such individuals as a whole to live independent lives, there is a need to identify ways to mitigate their symptoms of postural instability and reduce their risk of falling.

Existing therapies for balance disorders include pharmacological treatments, balance rehabilitation, surgical procedures, and balance aids such as canes, walkers, and wheelchairs. In the last ten years, significant efforts concerning the potential benefits of both implantable and non-implantable balance prostheses have been explored [6-13]. Non-implantable prostheses such as vibrotactile display of body tilt [9, 14], surface electrode stimulation of the vestibular nerve, electric currents applied to the tongue [6, 7], and audio feedback [11-13] offer varying degrees of non-invasive self-motion cues. Such devices can serve as a permanent or temporary replacement of motion cues for the vestibulopathic patient and aging, a tool for vestibular/balance rehabilitation, or an additional sensory channel for military troops, pilots, and astronauts.

We have repeatedly demonstrated increased postural stability for unilateral and bilateral vestibulopathic patients donning the vibrotactile balance prosthesis during computerized posturography experiments [14-17]. The vibrotactile feedback device, which senses body tilt, consists of a motion-sensing system mounted on the lower back of the subject, a vibrotactile display, and a laptop computer (or wearable watch) with analog and digital interfaces. All experiments with the vibrotactile balance prosthesis to date have used an input to the vibrotactile display that is the summation of a body tilt estimate and one-half its first derivative (tilt rate). This feedback scheme is supported by a previous investigation that showed the greatest reduction of root-mean-square (RMS) tilt was achieved during computerized dynamic posturography with proportional plus derivate feedback versus proportional or derivative feedback alone (manuscript in preparation). We have shown reduced anterior-posterior (A/P) tilt in vestibulopathic patients when A/P tilt is displayed during quiet standing and 1-axis perturbed standing trials [15, 17]. Additionally, medial-lateral (M/L) tilt has been reduced when M/L tilt was displayed. During test conditions that induced a mild 2-axis random platform motion, all subjects significantly reduced their A/P sway when A/P tilt was displayed. However, the change in M/L sway was not significant suggesting direction-specific control [14].

Other modalities are also being explored for biofeedback including display of body position via a lingual stimulator [6, 7, 10] and auditory biofeedback [11-13]. Dozza et al. has shown that audio biofeedback does not simply increase stiffness, but aids in the CNS actively adapting its control activity over standing posture [12]. Although lingual, auditory, and visual feedback displays are all valid and effective means of reducing sway in vestibulopathic patients, it is the authors' opinion that sensory substitution in the form of a vibrotactile display is preferential to the other modalities of delivery because vibrotactile stimulation does not compete with tasks that involve speaking, eating, hearing, etc.

During erect natural bipedal stance, feet positioned side-by-side approximately hip-width apart, the postural control challenge is predominantly, approximately twofold, in the A/P versus the M/L direction [18]. However, as the feet are brought closer together nearing a tandem or Romberg configuration, the challenge shifts to controlling instability predominantly in the M/L direction. Our original prosthesis provided vibrotactile feedback in the A/P direction alone. In a design study examining vibrotactile display coding, performance in a modified version of the manual control critical tracking task was not appreciably improved when the prosthesis was equipped with more than three rows of position-based tactors [19]. Therefore, in all subsequent versions of our balance prosthesis, we used three rows of tactors to encode magnitude of tilt. Circumferential spatial resolution becomes an issue when providing multi-directional tilt information. This research seeks to determine the optimal number of columns of tactors for achieving postural stability. An argument can be made for having the greatest spatial resolution allowable by two-point discrimination in order to supply the operator, in this case, our vestibulopathic patient, with the maximum amount of information regarding his/her tilt. On the other hand, there is the issue of cognitive workload; the more information that is provided to the patient, the more computations that need to be performed in order to interpret and use that information. We hypothesized that multi-axis (4-16 columns of tactors) display of body tilt during multi-directional surface perturbations would reduce sway in all directions and that the 16-column configuration would result in the lowest RMS tilt, smallest trajectory area, and least amount of time spent outside of the one-degree dead zone.

Methods

Subjects

Eight poorly compensated vestibulopathic patients referred by the Massachusetts Eye & Ear Infirmary (MEEI) Department of Otolaryngology clinicians were used in this study. Poorly compensated was defined as those patients who failed the NeuroCom® EquiTest® computerized dynamic posturography Sensory Organization Tests (SOT) 5 and 6. During SOT 5, the subject's eyes are closed and the posture platform is sway referenced (moves in synchrony with the subject's A/P body sway). SOT 6 is performed with the subject's eyes open and both the platform and visual surround are sway referenced. Patients with histories of mental illness and motor deficits were excluded. Additionally, obese individuals were excluded due to the size constraints of the vibrotactile balance prosthesis. Table 1 shows the subjects' vestibular test results and relevant demographic information. Informed consent was obtained from each subject. The Massachusetts Eye & Ear Infirmary, Boston University, and Massachusetts Institute of Technology Institute Review Boards approved the experimental protocol. The subjects wore a safety harness that was suspended from the ceiling for the entirety of the experiment. Freitas et al. has shown that the contact of the safety harness with the body does not affect sway during quiet stance [20]. A sufficient amount of slack in the safety harness system was provided to account for platform displacements during the perturbation protocol. Subjects verbally confirmed that they could not perceive support from the safety harness prior to the start of the experiment (i.e., harness was not pulling on them). Additionally, a safety spotter stood on the platform directly behind the subject.

Subject Demographics				Computerized Dynamic Posturography				Diagnosis	Rotation	Caloric
Patient ID	Age	Gender	Participation	SOT Score	Sot 5	SOT 6	MCT Score	UVH or BVH	VOR midrange gain	RVR
1	52	M	CT, DT	49	Fall, Fall, Fall	Fall, Fall, Fall	N/A	N/A	N/A	N/A
2	55	F	CT	32	Fall, Fall, Fall	Fall, Fall, Fall	128	N/A	N/A	0
3	44	M	CT, DT	38	Fall, Fall, Fall	Fall, Fall, Fall	158	BVH	0.196	0
4	55	M	CT	45	Fall, Fall, Fall	Fall, Fall, Fall	128	BVH	0.841	N/A
5	58	M	CT, DT	N/A	N/A	N/A	N/A	BVH	0.02	0
6	32	M	CT, DT	46	Fall, Fall, Fall	Fall, Fall, Fall	151	BVH	0.514	0
7	51	F	CT, DT	56	Fall, 26, 45	Fall, Fall, 45	158	N/A	0.956	-4
8	45	M	CT, DT	49	Fall, Fall, Fall	Fall, Fall, Fall	130	BVH	0.899	-11

Table 1. Subject demographics

Legend:
SOT – Sensory Organization Test: Normal mean composite scores are 80.2 for 20-59 yrs and 76.9 for 60-69 yrs, 5^{th} percentile (abnormal) limits are 68.5 for 20-59 yrs and 70.0 for 60-69 yrs
MCT – Motor Control Test: Normal mean composite scores are 143.0 for 20-59 yrs and 151.8 for 60-69 yrs 5^{th} percentile (abnormal) limits are 161.0 for 20-59 yrs and 170.8 for 60-69 yrs
VOR – Vestibuloocular reflex
N/A – Not available
RVR – Reduced vestibular response to bilateral, bithermal caloric stimulation. All but one subject had a 0°/s nystagmus response to ice water in the side-of-tumor ear. One subject had a 5°/s response.
UVH – Unilateral vestibular hypofunction
BVH – Bilateral vestibular hypofunction

Equipment

Data were collected in the Injury Analysis and Prevention Laboratory in the NeuroMuscular Research Center at Boston University. A custom-built moveable BALance DisturbER (BALDER) platform [44] 2.1 m square, generated a programmable stimulus while the motion of the subject's body was optically tracked. The primary components of the BALDER platform include a force-plate (ORG-6 AMTI, Newton, MA, USA) imbedded in a wooden platform, two AC-servo motors controlled by two linear servo drivers, two high precision linear position transducers (Novotechnik, Germany), and a 16 channel A/D - two channel D/A data acquisition board (Microstar 3200e/415). Kinematics were collected using the Optotrak 3020 system (Northern Digital, Waterloo, Ont.). Rectangular arrays consisting of six infrared emitting diodes (IRLED) were placed on the subject's pelvis, sternum, and head. The IRLED sampling rate was 1500 Hz and the array positions were estimated at 100 Hz. The 3020 was placed 3.5 m from the BALDER platform within the system's optimal viewing area. The 3D IRLED translations were recorded and converted to 6D data using the Data Analysis Package provided with the system.

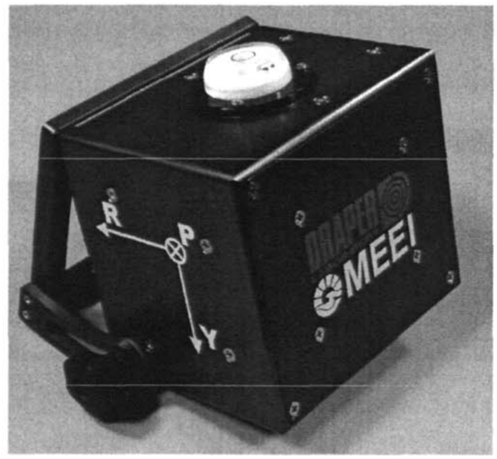

Figure 1. Draper inertial sensor assembly

The vibrotactile balance prosthesis consisted of a two-axis motion-sensing system mounted on the lower back of the subject, a vibrotactile display, and a laptop with analog and digital interfaces. The inertial motion-sensing system was composed of microelectromechanical (MEMS) gyroscopes that sense angular rate and MEMS accelerometers that sense linear accelerations [9] (Figure 1). The gyroscope and accelerometer signals were processed to obtain a tilt angle estimate accurate to within 2 milliradians over a 0 to 10 Hz bandwidth. Tilt estimates were haptically displayed on the subjects' torsos via a 3 row by 16-column tactile vibrator array; rows of the array display estimated tilt magnitude (Figure 2) and columns display tilt directions (Figure 3). The tactile vibrators (Tactaid, Cambridge, MA), referred to as tactors, operated at a constant amplitude (200 mA) and frequency (250 Hz). All subjects reported that they were able to feel the tactor vibrations. The tactor firing range was set on a subject-by-subject basis. An elliptical fit to the four maximum static tilt values during quiet standing was used to map the lowest, middle, and highest tactor row activation thresholds to approximately 1°, 3-5° (50% of maximum static tilt angle), and 5-7° (85% of maximum static tilt angle), respectively. Anterior and posterior tactor activation coding was asymmetrical because the limits of postural stability are smaller in the posterior direction than the anterior direction. No tactors were activated within a dead zone to account for normal body sway. With increasing body sway, tactor firing progressed from the bottom to top tactor along the appropriate tactor column (see below) in a stepwise fashion. Subjects were instructed to always move to null out the vibration to stay within the dead zone; zone 1, 2, and 3 were defined as the region in which the first, second, and third row of tactors were active,

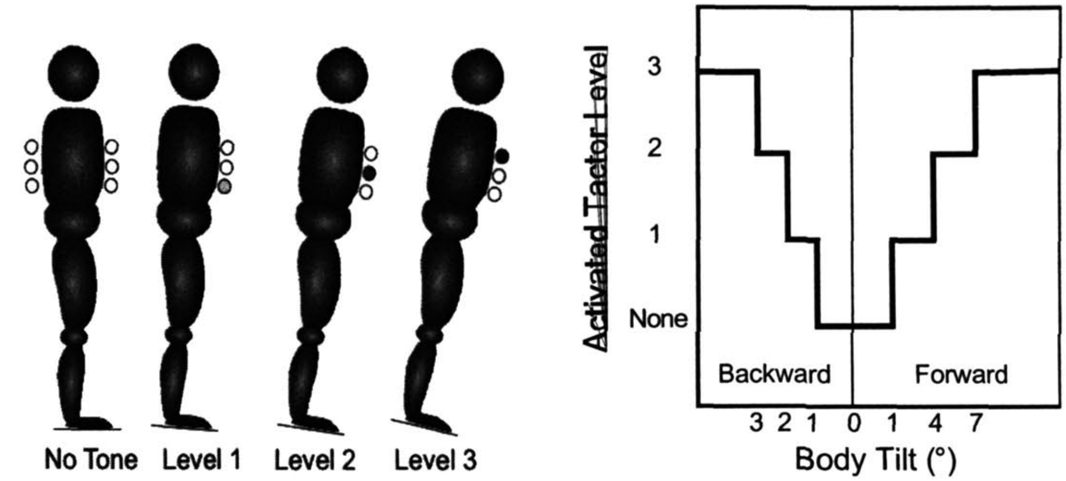

Figure 2. Tactor activation scheme

respectively. The subjects in this study were provided with proportional plus derivative estimates of their tilt.

The tactor coordinate system was defined as shown in Figure 3 (0° corresponds to the axis perpendicular to the intra-aural axis and 90° to the subject's right as viewed from above). Four prosthesis tactor configurations were evaluated in this study: 16 columns (tactors placed every 22.5°), 8 columns (tactors placed at 0°, 45°, 90°, 135°, 180°, 225°, 270°, 315°), 4 columns (0°, 90°, 180°, 270°), and no vibrotactile feedback. Some subjects had an additional display with 6 columns (tactors placed at 0°, 67.5°, 112.5°, 180°, 247.5°, 292.5°) presented to them after the completion of the standard protocol. The standard directional display scheme operated on a "nearest neighbor" principle. That is, the direction of the subjects' tilt was compared to the particular tactor column configuration in use, and the best matching column was activated. One alternative directional display scheme was used for the 4-column independent tactor configuration based on the principle of interpolated position; two columns were activated as long as the tilt direction was not aligned with the coordinate axes.

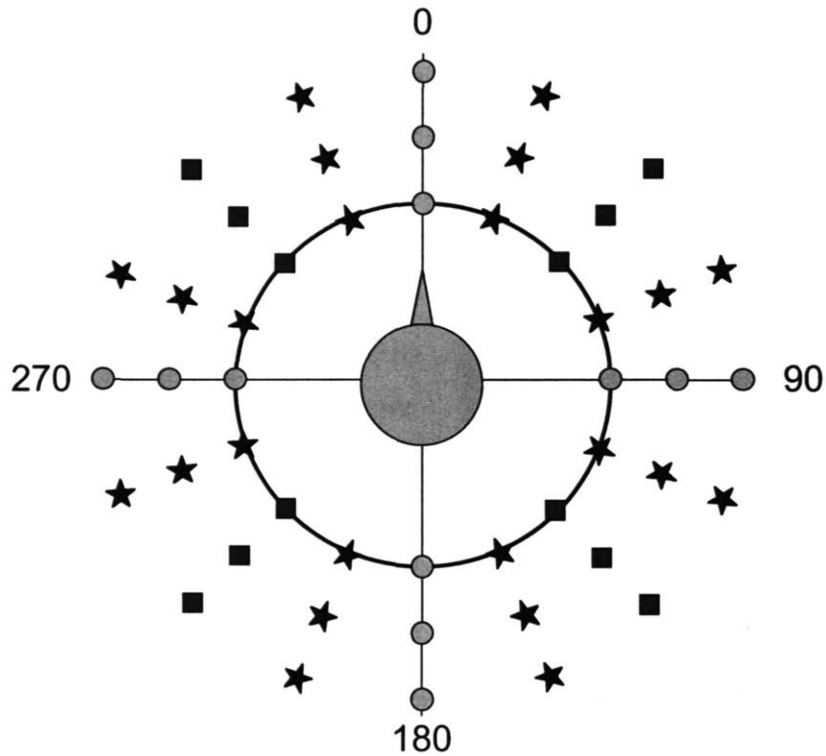

Figure 3. Tactor column displays. Circles indicate tactor column positions for 4 column and 4 column independent displays. Circles and diamonds indicated tactor column positions for the 8 column display. Circles, diamonds, and stars indicate tactor column positions for the 16 column display. The coordinate system is indicated.

Platform Stimuli

Both continuous and discrete surface perturbations were used in this experiment to assess multi-directional vibrotactile feedback on postural performance. Continuous perturbations were selected to represent long duration postural challenges such as riding upright on a bus or subway car while discrete perturbations were selected to characterize responses to abrupt postural disturbances.

Continuous Perturbations

Pseudorandom translation of the support surface was selected for the continuous motion stimulus based on the previous use of a ternary sequence by Peterka to investigate sensorimotor integration in human postural control [21]. The pseudorandom stimulus is advantageous for identifying linear systems because it is a deterministic periodic signal with autocorrelation properties approximating quasiwhite properties (Marmarelis, 1978). A pseudorandom pentary sequence (PRPS) of numbers was used instead of ternary sequence in order to generate a sequence of sufficient duration and directional variety (Davies, 1970). A future analysis will rely on this choice of stimulus to determine transfer functions for the subjects both with and without the balance prosthesis. The method used to generate the pseudorandom sequence waveforms is shown in Figure 4. A linear velocity command sequence was created from a 624-length PRPS sequence by assigning fixed values of +2v, +v, 0, -v, or -2v°/s to the four stage/modulo 5 addition shift register output for a state duration of Δt = 0.09 s. The total duration of each sequence was 60 s. Three sequences were concatenated to compose the continuous perturbation stimulus. The stimulus waveform was then integrated to create a position waveform. The position waveforms were used to drive the x and y BALDER servo motors. Four unique sequences were generated using this technique; an independent set (not correlated) was generated for the x and y velocity signals for the training trials and a second distinct pair was generated for the testing trials. Subjects were trained with 3 repetitions of the one-minute training sequences. The first repetition was performed with their eyes open, the second repetition with their eyes open for the first 30 s followed by their eyes closed for the last 30 s, and the final repetition with their eyes closed for the entirety of the trial. Subjects were tested with three one-minute concatenated sequences for each tactor configuration. Figure 5 shows an overhead view of one cycle of the continuous PRPS test stimulus. The stimulus was scaled in magnitude during the training session based on the subject's balance capabilities; the magnitude was selected such that 1) a step was not elicited and 2) a verbal score of perceived balance difficulty of 7 out of 10, where 10 was defined as the subject's most difficult balance challenge. In general, the stimulus was mild enough such that to a visual observer, it appeared that subjects primarily used an ankle strategy (small corrective torques generated at the ankle) to maintain postural stability. The stimulus appeared to be unpredictable to the test subjects as per their reports.

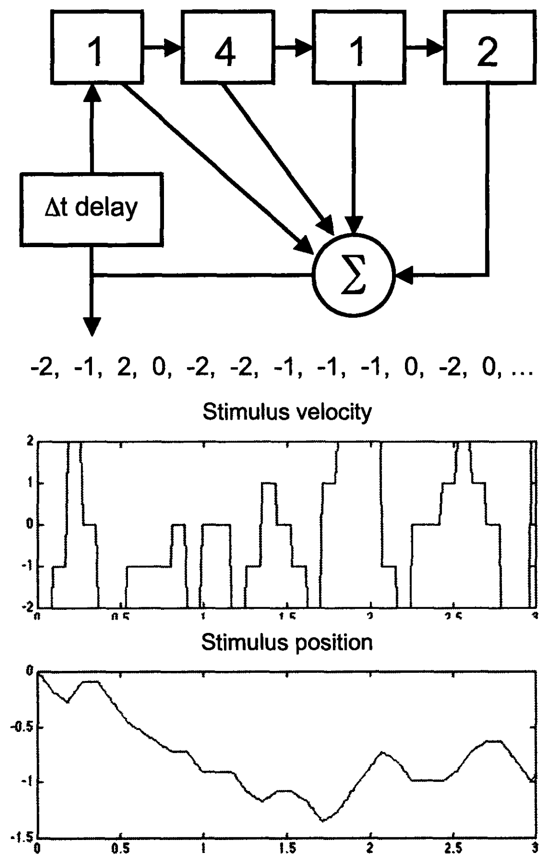

Figure 4. Continuous perturbation stimulus profile. Four stage shift register with feedback generates a pseudorandom pentary sequence (PRPS) using modulo 5 addition. At each time increment ($\Delta t = 0.09s$) the value of each register is shifted. The PRPS sequence is converted to a stimulus velocity. The stimulus velocity and its time integral are shown.

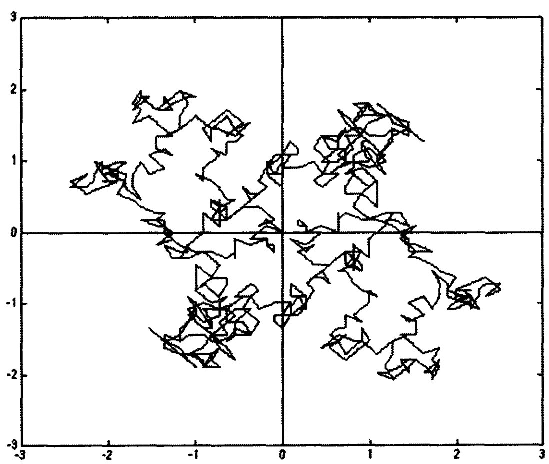

Figure 5. Overhead view of the idealized platform motion

Discrete Perturbations

The discrete perturbations perturbed the subjects in six primary directions: 90°, 135°, 169°, 180°, 315°, and 349°. Two of the perturbation directions (90°, 180°) were in alignment with tactor columns present in all display types. Two of the perturbation directions (135°, 315°) were in alignment with only the 8 and 16 column displays. Finally, two perturbation directions (169°, 349°) did not coincide with any tactor column among display types. The platform was programmed to reset in one of four directions: 0°, 34°, 214°, and 270°. The non-cardinal direction perturbations were chosen such that when viewed from a standard Cartesian coordinate system, subjects' recovery trajectories traversed quadrant II and provided off-tactor column information. Data were collected for 10 sec following the perturbation. The subject and the safety spotter were not informed to the perturbation direction. The appropriate perturbation magnitude was determined by trial and error during the training session; the perturbation was not designed to cause a fall or elicit a stepping response. Subjects were trained with 3 repetitions of a one-minute training sequence (first repetition with eyes open, second repetition with eyes open for 30 s followed by eyes closed for 30 s, third repetition with eyes closed). Subjects were tested with a four-minute long sequence of the aforementioned perturbation directions. The sequence included 23 perturbations and was ordered to minimize predictability of perturbation direction while operating within the BALDER platform range of motion. Subjects completed one four-minute long sequence for each tactor display configuration.

Experimental Design

Latin Squares

A Balanced Latin Squares design was used with tactor configuration as the primary factor (Table 2). For example, Group 1 subjects completed the 4 column standard firing scheme first for all test modalities, followed by the 4 column interpolation scheme, the 8, and finally the 16 column standard scheme. Each group consisted of two subjects.

Group 1	Group 2	Group 3	Group 4
A	B	C	D
B	D	A	C
D	C	B	A
C	A	D	B

Table 2. Balance Latin Squares design

Legend:
A = 4 columns (0°, 90°, 180°, 270°), standard firing scheme
B = 4 columns (0°, 90°, 180°, 270°), independent firing scheme
C = 8 columns (0°, 45°, 90°, 135°, 180°, 225°, 270°, 315°), standard firing scheme
D = 16 columns (every 22.5° starting at 0°), standard firing

Session 1 (8 subjects)

All groups performed a tactors off trial before and after completing the four tactor configuration tests, rested for 20 minutes and repeated the tactor off test to evaluate short-term retention effects. All eight subjects were tested using the continuous protocol during Session 1.

Session 2 (6 out of 8 subjects)

Six of the eight subjects returned on a subsequent date ranging between one day and two months later to be tested using the discrete perturbation protocol. All discrete perturbation trials were completed according to the group's designated order. During Session 2, subjects initially performed two continuous perturbation trials as per the Session 1 protocol. The first of the two trials was conducted in the tactors off configuration as part of a long-term retention study. The second of the two trials was performed using the 16-column display. Subjects did not practice using the prosthesis prior to this trial. This trial was added to assess the intuitiveness of the device and the subjects' ability to perform the task of using the device without recent practice. Table 2 indicates the varying times between the first and second testing dates. After completing the continuous trials, subjects were thoroughly retrained on how to use the prosthesis before beginning their discrete perturbation acclimatization training.

In addition to short and long-term retention trials, subjects completed trials in which erroneous tilt information was fed back to them in order to examine a potential placebo effect (i.e., increased subject performance based solely based on their desire for the prosthesis to be helpful). These trials consisted of playback of the subject's own sway that was recorded immediately following the training session. In the case of the continuous stimulus, a separate pair of x and y PRPS sequences were used to perturb the subject and their sway trajectory was recorded. In the case of the discrete stimulus, six new perturbation directions were explored while their sway was recorded. In both cases, the subjects believed that they were completing an additional training trial. If performed, the erroneous playback trial (succinctly termed "junk" trial) was the final trial during each session in order to prevent the subjects from losing confidence or questioning the validity of the prosthesis in subsequent trials. The tilt information played back to the subject was in the form of vibrations that were uncoupled to the platform motion.

Protocol

Informed consent was obtained from all subjects prior to testing. Subjects were provided with a standardized lightweight polyester tee-shirt (Patagonia silk weight Capilene™) and instrumented with the vibrotactile prosthesis and optical markers. The scaling of the tactor magnitude display algorithm (rows) was customized according to each subject's limits of stability. Subjects participated in a 30-minute training session. All tactor configurations were practiced during quiet stance in the eyes open and closed configurations. Subjects were instructed to move to null out the vibrations regardless of the tactor column configuration.

During test sessions, the subjects were not told which tactor configuration they were using unless it was a tactors off trial. They were only told to move such that they null out the vibrations and to stand as upright as possible.

Subjects were instructed to close their eyes for all trials and their feet were positioned in a standard configuration on the BALDER force plate (slightly less than hip-width apart and slightly skewed outward). Five-minute rest breaks were consistently taken following the completion of two trials.

All trials were performed with the subject's eyes closed and arms placed at their side. The harness was adjusted such that 1) no haptic orientation cues were supplied to the wearer, and 2) the body was prevented from impacting the platform in the event of a fall.

A modified five point Likert scale [22] was used to assess the subject's impression regarding the usefulness of the device in improving stability. Subjects could select the following responses when asked to complete the statement: "I found the device to be": (1) very unhelpful; (2) moderately unhelpful; (3) neutral-neither helps nor hurts; (4) moderately helpful; and (5) very helpful. In addition to the Likert question, subjects were verbally asked to rate their perception of task difficulty and fatigue level on an analog scale of 1 (very easy balance task, no fatigue, respectively) to 10 (most difficult balance task, completely fatigued, respectively) following every trial. The perception of task difficulty was used to adjust the perturbation magnitude to the subject's ability so that each subject found the task to be of similar challenge (7 out of 10). The fatigue scale was used to determine when additional rest periods were to be taken (greater than 6 out of 10).

Results

The data were sampled at 100 Hz. All post-processing was performed using Matlab (The MathWorks, Natick, MA). Statistical analysis was conducted using STATA (StataCorp LP, College Station, TX). Following data collection, the data were low pass filtered with a 3rd order phaseless butterworth filter (Matlab filtfilt.m) with a corner frequency of 10 Hz to remove high frequency noise.

<u>Continuous Perturbations</u>

Tilt data

Figure 6 shows an example of low pass filtered tilt data of a representative subject. The first subplot depicts the sway trajectory of the subject in the tactors off configuration. The second subplot shows the sway trajectory when the device is turned on and the 4-column display (next neighbor principle tactor activation) is used. The third subplot shows the sway trajectory during the junk trials. It is clear from the raw data that the subject has approximately a two-degree pitch forward bias without the tactors on. When the tactors are turned on, the subject is able to maintain balance about the vertical. Although the subject did not fall during the junk trial, it is evident that the sway excursion increased in both the A/P and M/L directions suggesting that merely turning the device on is insufficient to maintain controlled posture as per the experimental instructions.

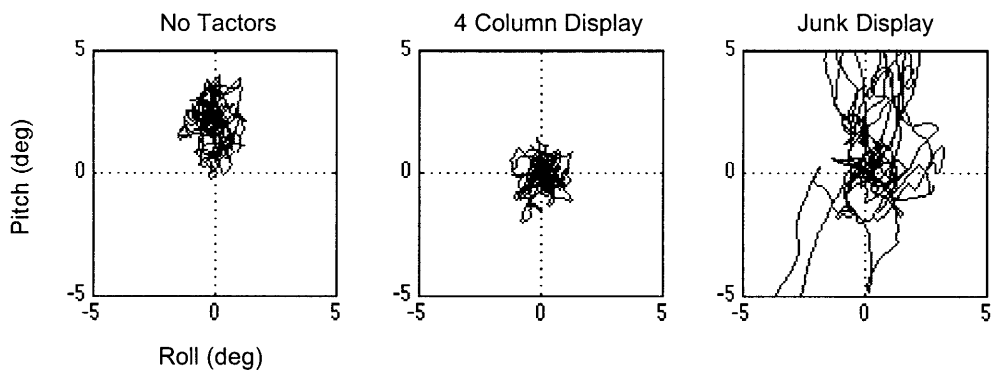

Figure 6. Sample data from a single subject, one cycle of continuous stimulus

In order to capture the difference in trajectory area, the resultant tilt vector (square root of the squared sum of roll and pitch) was fit with 95% Confidence Interval ellipses. Recall that each continuous perturbation stimulus comprises three identical concatenated sequences (subsequently referred to as cycles). The ellipse area was calculated as the product of pi, the length of the minor axis, and the length of the major axis. Cycle ellipse areas and parameters were averaged. The average of the first trial cycles, which were performed without vibrotactile feedback, were used to normalize the ellipse area of the resultant tilt vector for all subsequent trials. Figure 7 shows NT denotes "no tactors" trials. NT1 is the pretest trial and NT2 is the posttest trial. NT3 occurred during Session 1 after a 20-minute rest following the completion of the NT2 trial. NT4 occurred during Session 2. The error bars represent the standard error of the mean. A one-way analysis of variance was performed with normalized ellipse area as the response variable and tactor configuration as the factor variable. The tactors off normalized ellipse areas were significantly larger ($p<0.0013$) than all four tactor on conditions however, there was not a significant difference among tactor configurations. Subjects had significantly larger Junk area ellipses compared to both the tactors off and tactors on configurations. Note, the Junk and day 2 testing comprised only 6 subjects.

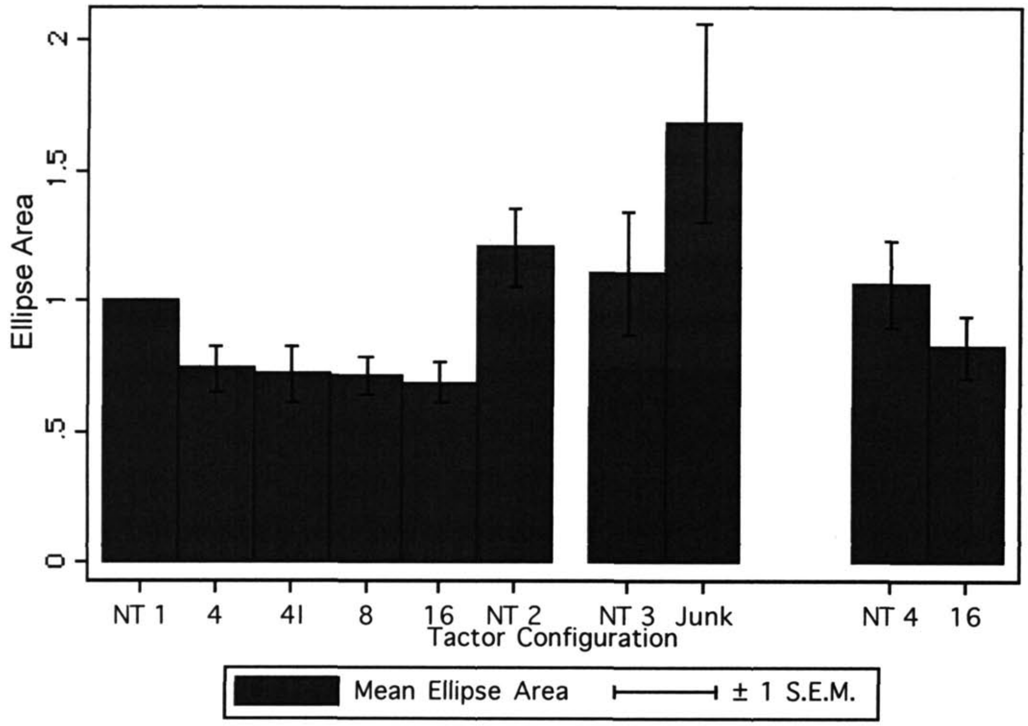

Figure 7. Normalized elliptical fits of tilt trajectory areas by display configuration. NT denotes "no tactors" trials. NT1 is the pretest trial and NT2 is the posttest trial. NT3 occurred during Session 1 testing after a 20-minute rest following the completion of the NT2 trial. NT4 occurred during Session 2. The error bars represent the standard error of the mean.

The root-mean-square (RMS) of the resultant tilt vector (square root of the squared sum of roll and pitch) was calculated for all trials. The three cycles of the first trial, which were performed without vibrotactile feedback, were averaged and used to normalize the RMS of the resultant tilt vector for all subsequent trials. Figure 8 shows the normalized RMS findings for continuous perturbation protocol. The tactors off normalized RMS values were significantly larger in the tactors off pre and posttest ($p < 0.0003$) compared to all four tactor on conditions however, there was not a significant difference among tactor configurations. Broken down into tilt components, the RMS M/L and A/P tilt was significantly larger for the tactors off pre and posttest compared to the tactor on configurations ($p<0.0069$ and $p<0.0005$, respectively). The posttest and short-term retention trials, NT2 and NT3 respectively, have lower RMS tilt values than the pretest. This finding suggests that some learning has occurred throughout the experiment. Although slightly higher than the pretest value, the Junk trial was not significantly different than the tactors off conditions. Compared to the previous two no tactor trials though, the Junk trial RMS is substantially higher; especially if it is accepted that the

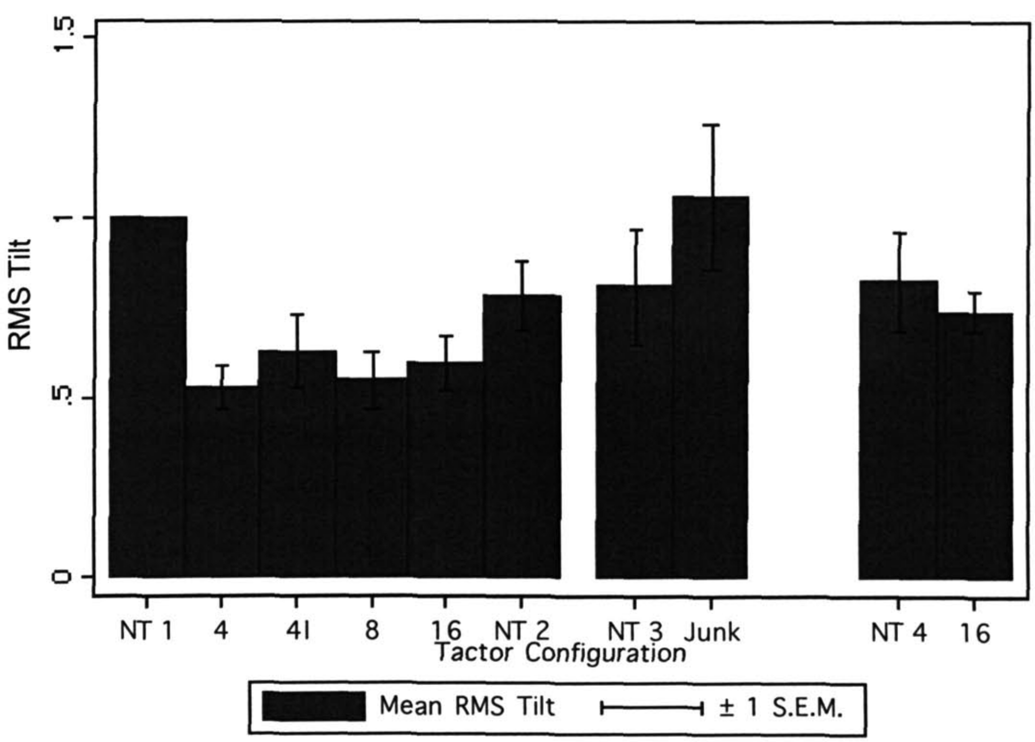

Figure 8. Normalized RMS resultant tilt by display configuration

posttest values offer a more realistic baseline for comparison than the pretest tactors off trial since some learning/adaptation was evident. The long-term retention trial shows a similar value compared to the two posttest values on the first day of testing. The 16 column configuration trial on the second day of testing is not significantly different than the tactors off trial on that same day, but is lower indicating either some retention or same day learning after the initial exposure to the continuous perturbation stimulus.

Tilt pathlength was computed by summing the square root of the squared sum of roll and pitch tilt over the duration of each cycle. Cycle pathlengths were averaged and the three cycles of the first trial, which were performed without vibrotactile feedback, were averaged and used to normalize the pathlength of the resultant tilt vector for all subsequent trials. Figure 9 shows the results of the pathlength analysis, which revealed a significant difference ($p < 0.0058$) between only the pretest tactors off trial compared to all four tactor on conditions. There was not a significant difference among tactor configurations. First day posttests and the Junk trial were not significantly different from the tactor on configurations suggesting that the average pathlength, or total tilt excursion did not differ among trials outside of the pretest trial. On the second day of testing, both the tactors off and on configuration showed significantly higher pathlength values compared to all of the first day testing trials. This is likely due to the fact that on the second day of testing, the subjects were tested without any practice on the perturbation platform as opposed to the first day of testing where NT1 was conducted after acclimating to the perturbation stimulus.

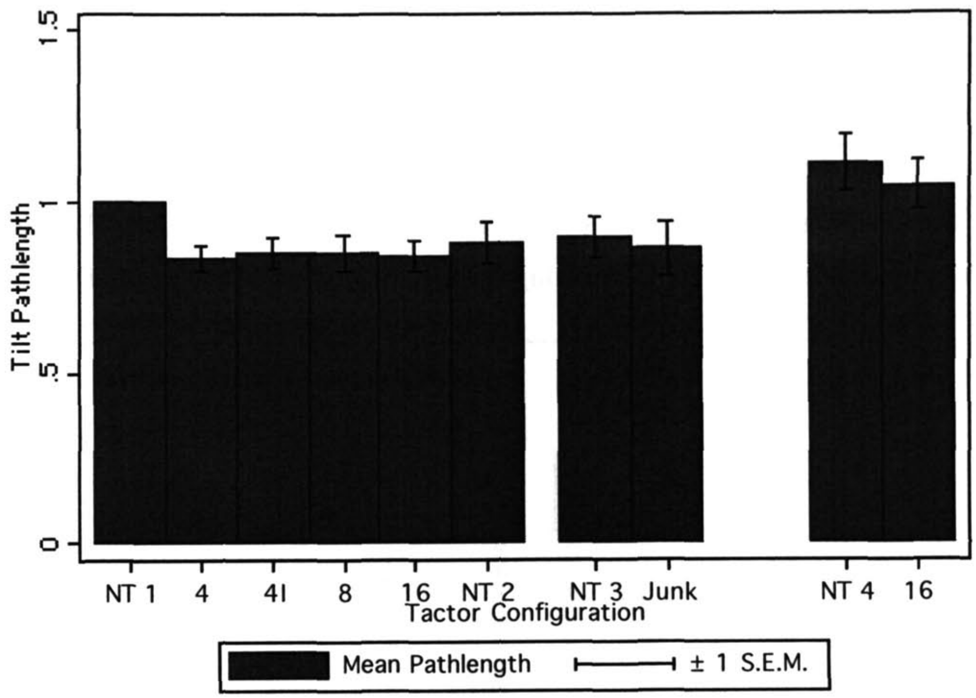

Figure 9. Normalized tilt pathlength by display configuration

The percentage of time spent in the various tactor zones was computed based on binning the tilt values for a give cycle of a given trial on a sample by sample basis using the subject's individual defined tactor activation levels. For example, the dead zone was defined as the region where no tactors were firing. Zone 1 was the region where the first row of tactors fired, zone 2 was the region where the second row of tactors fired, and the zone 3 was the region where the third row of tactors fired. Figure 10 depicts the results from the zonal analysis. Ideally, one would hope to see a reduction in the percentage of time spent in zone 2 and zone 3 as a function of device status and this is exactly what is observed; the percentage of time spent in zones 2 and 3 is significantly greater ($p<0.0006$ and $p<0.0025$, respectively) in the pre and post tactor off configurations compared to the tactor on configurations. Additionally, the percentage of time spent in zone 3 is significantly greater in the Junk trial compared to the tactor off and tactor on configurations. The only zone that did not show a significant change was zone 1. The severity parameter was defined as a linear weighted sum of percentage of time spent in the various zones (Figure 11).

$$Severity = Z_1 + 2 \times Z_2 + 3 \times Z_3$$, where Z designates zone

This parameter was significantly higher ($p<0.0009$) for the pretest tactors off trial compared to all four tactor on conditions. There was not a significant difference between tactor configurations. Although the posttest tactors off value is approximately one third of the pretest value, it is still significantly higher than the tactor on configuration values.

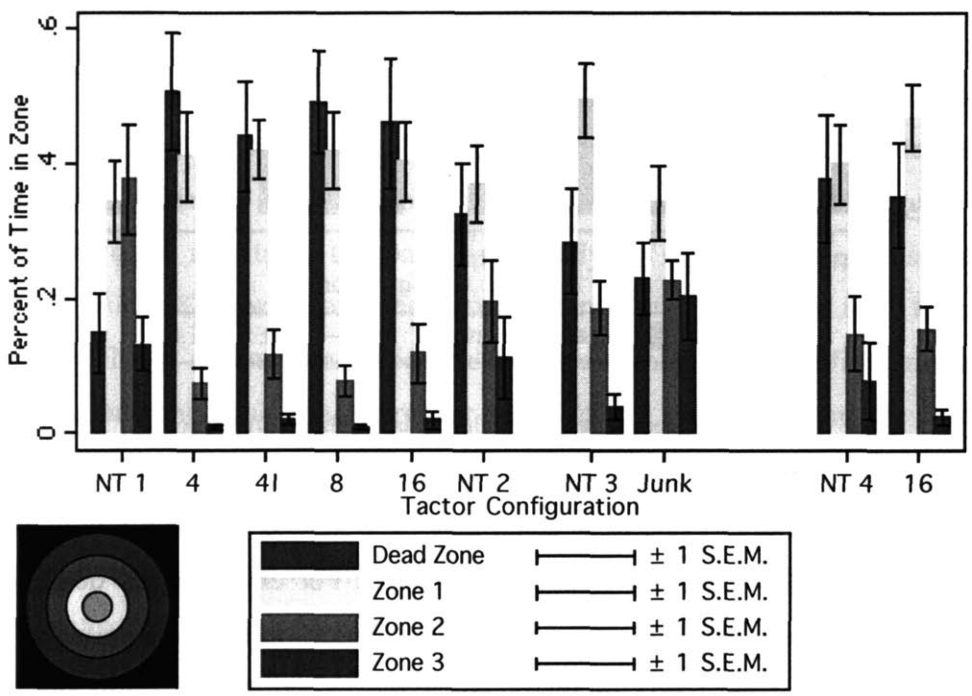

Figure 10. Percentage of time spent in tactor firing zone by display configuration

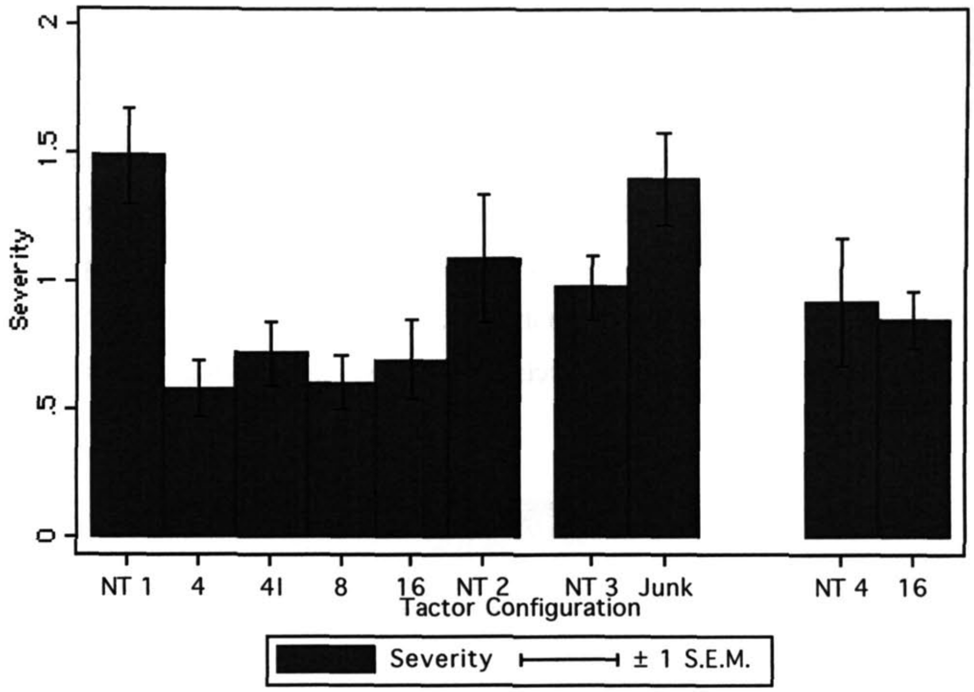

Figure 11. Severity score by display configuration

Optotrak data

The anchoring index is a previously published parameter for characterizing head and trunk stabilization strategies in the frontal plane during unperturbed locomotion [23-25]. The index describes the relative angular distribution of the body segment being considered with respect to axes linked to an inferior anatomical segment. The anchoring index is defined as:

$$AI = [(\sigma_r) - (\sigma_a)]/[(\sigma_r) + (\sigma_a)]$$

where, σ_a is the angular dispersion of any body segment and σ_r is the standard deviation of the relative angular distribution of the body segment being considered with respect to axes linked to an inferior anatomical segment. A positive value of the anchoring index indicates a tendency for trunk stabilization in space rather than on the hip, whereas a negative value would indicate a tendency for trunk stabilization on the hip rather than in space. In theory, this index reveals whether an individual adopts an "en bloc" or inverted-pendulum like stabilization strategy. Figure 12 shows a box plot of both the roll and pitch anchoring indices for both the head and the stern. The head anchoring index explores the relationship of the head to the sternum and sternum anchoring index explores the relationship of the sternum with respect to the pelvis. The anchoring indices are compared for the tactors off configurations (pre and post test) and the tactors on configurations (4, 4 independent, 8, and 16 columns). There were no significant differences for any angular dispersion values or anchoring indices suggesting that subjects did not merely stiffen up when the device was turned on, but rather that the subjects employed similar stabilization strategies in both situations.

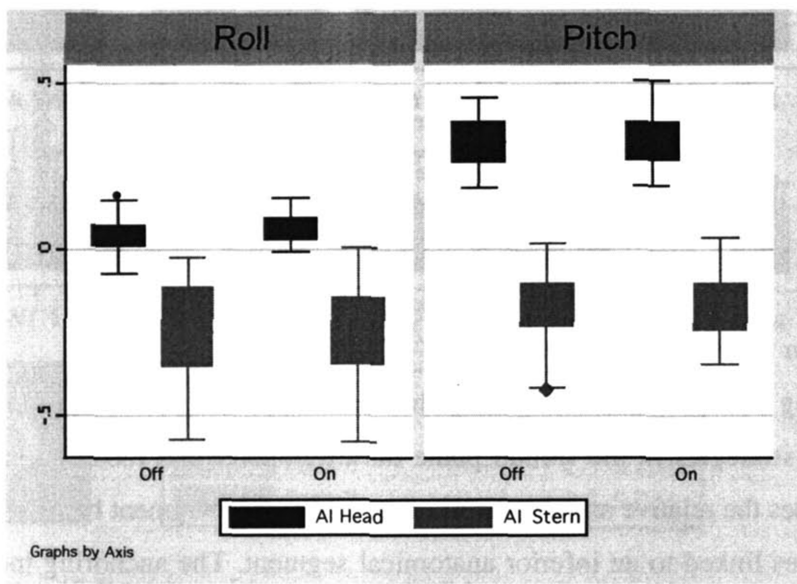

Figure 12. Head and sternum anchoring indices

Center of Pressure data

The RMS of the double limb support COP_{net} (square root of the squared sum of M/L and A/P COP values) was calculated for all trials. The three cycles of the first trial, which were performed without vibrotactile feedback, were averaged and used to normalize the RMS of the COP for all subsequent trials. Figure 13 shows the normalized RMS COP findings for the continuous perturbation protocol. Statistically, there was not a significant difference between the tactors off and tactors on configurations. Figure 14 shows the results of the computing the COP pathlength, or sum of the total COP excursion as seen by the force plate. Cycle pathlengths were averaged and the three cycles of the first trial, which were performed without vibrotactile feedback, were averaged and used to normalize the pathlength of the resultant tilt vector for all subsequent trials. The posttest and Junk trials had significantly shorter pathlength values compared to the pretest and tactor on configurations. The second day of testing reported greater values for both the tactors off and tactors on trials.

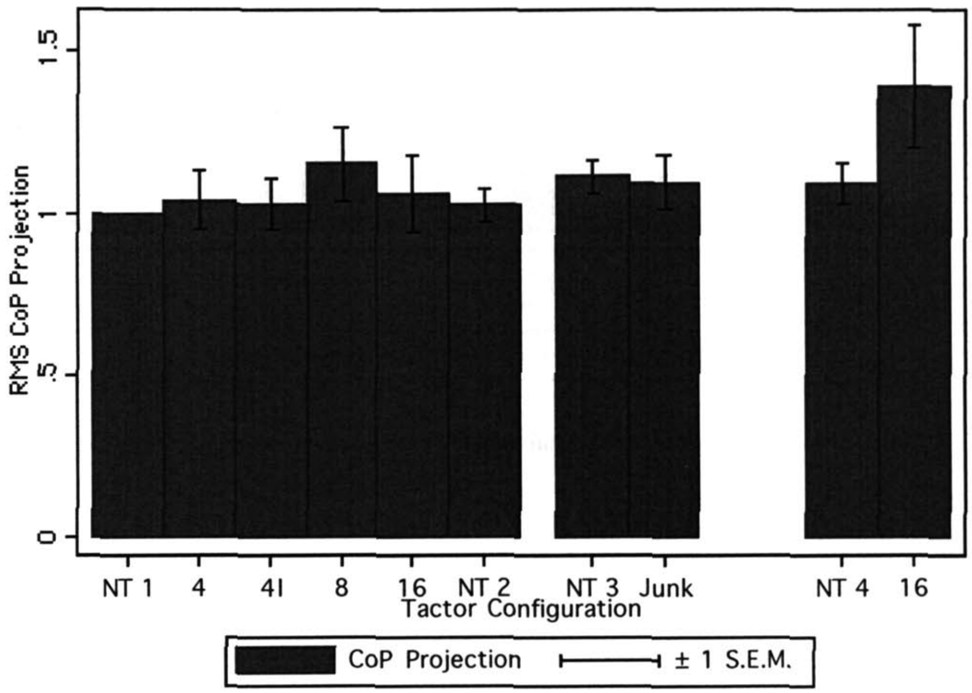

Figure 13. Normalized RMS COP projection

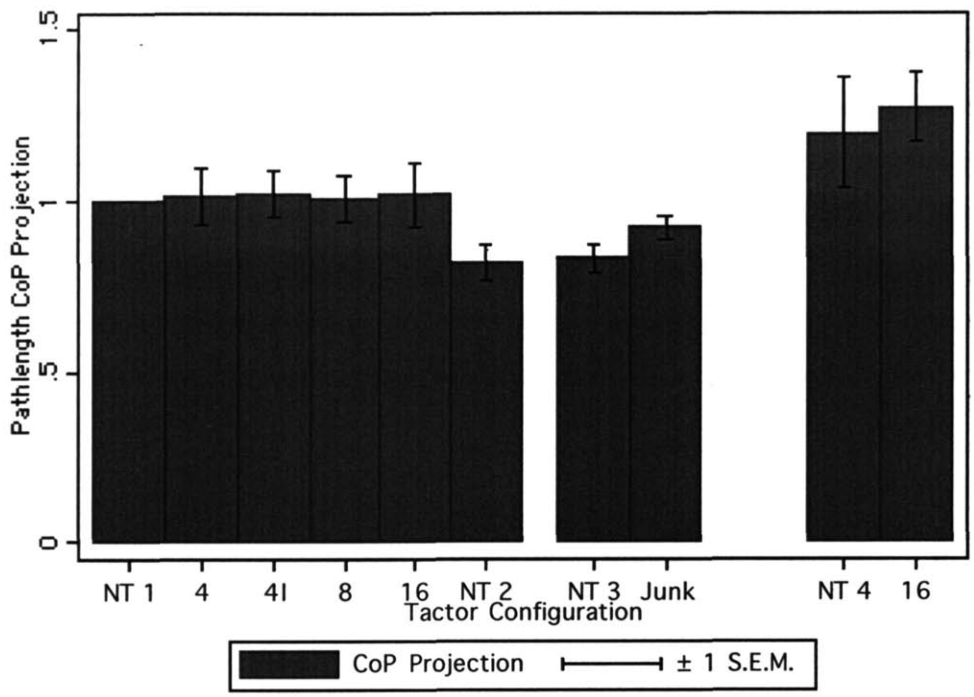

Figure 14. Normalized pathlength COP projection

Discrete perturbations

Figure 15 shows a representative sample of the resultant tilt data for a discrete perturbation. Several parameters were extracted and tabulated for each discrete perturbation including time to peak tilt displacement, peak tilt, time to reenter the one degree dead zone, RMS tilt after returning to dead zone, and percentage of time spent outside of the one degree dead zone. Parameter values for similar perturbation directions were averaged within a given trial for a given subject. The pretest tactors off trial value was used to normalize subsequent trial values so that subject results could be compared with one another. A one-way analysis of variance was performed with the aforementioned variables as response variables and tactor configuration as the factor variable. There was no significant difference in the time to peak tilt displacement, peak tilt, or time to reenter the one degree dead zone among tactor configurations. RMS tilt and percentage time spent outside the dead zone, calculated over a 5 second interval starting three seconds after the onset of the perturbations were significantly greater in the tactors off versus the tactors on configuration. No difference was observed between tactor display configurations.

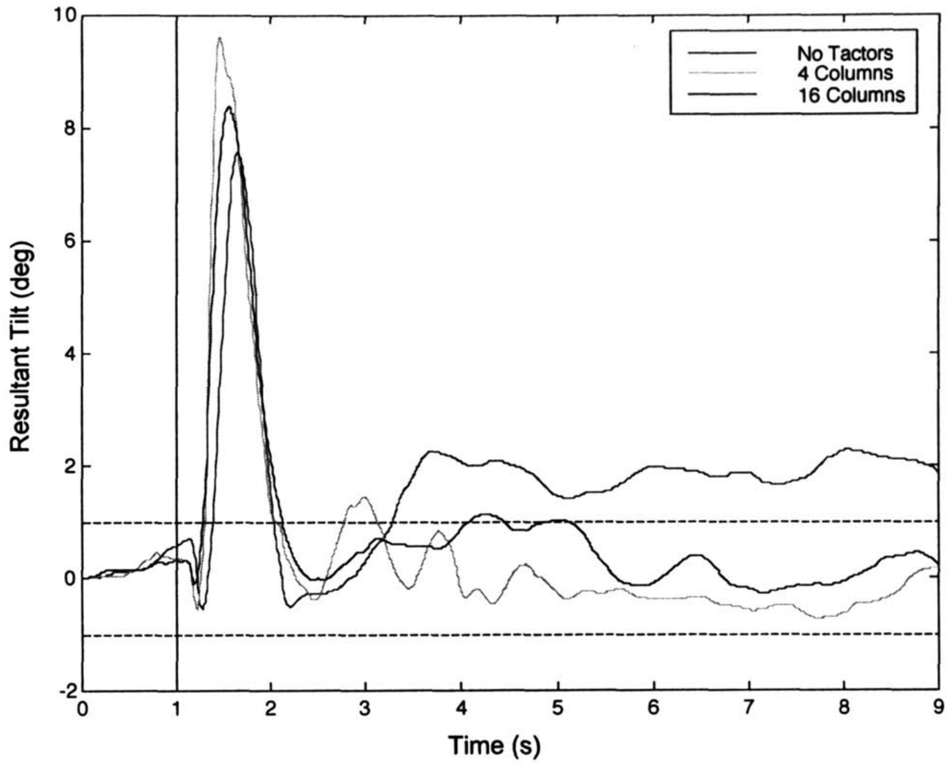

Figure 15. Discrete perturbation illustrative data

Discussion

These results show that subjects can use vibrotactile feedback to control their body sway tilt during multi-directional planar continuous and discrete support surface perturbations. Additionally, this is the first time that multi-directional vibrotactile feedback has been used to supplement body orientation information.

<u>On vs. Off</u>

Based on the results of this experiment, which used well-compensated vestibulopathic patients, no optimal tactor display configuration emerged. Overall, the tactors on configuration elicited improved postural performance compared to the tactors off configuration. On an subject by subject basis, individual performance varied as did personal preference for tactor display. The most common complaint about the 16 column configuration was that too much information was being provided. At the same time, the most common acclaim for the same display was that the subject felt confident that he/she was receiving the best and most complete information about his/her body movements. This study is the first to demonstrate that multi-directional vibrotactile feedback reduces multi-directional sway. Analyses were also performed on the components of sway, A/P and M/L tilt. In all cases where the resultant tilt was significant, A/P tilt was also significant. However, M/L tilt, although reduced in the tactors on configuration, was not significantly lower in all cases.

Why no difference between display types

The obvious question is why there wasn't an optimal tactor configuration in terms of superior performance identified across all subjects. Many reasons exist as to why this might be the case including: 1) limitations of biomechanics, 2) cognitive ability to process information, 3) not sensitive enough measures, and 4) individual variability. It has been shown that the hip is the primary means of controlling M/L sway while the ankles are predominantly used to control A/P sway [18]. Because the A/P component of sway dominates instability in natural bipedal stance, it begs the question of whether or not providing information only in that plane would be sufficient for replacing missing vestibular information during surface perturbations. Based on the discrete perturbation results, this argument seems to be supported. Spatial resolution of 22.5° may be excessive. When one more closely examines the physical trajectories of the subjects to the various off-axis perturbations presented in this experiment, one sees that the peak trajectory is not in line with the actual perturbation. Furthermore, the recovery trajectory has a dominant A/P component. The subjects tend to first make a M/L correction about the sagittal plane followed by oscillations about the coronal plane (intra-aural axis). [Have to try and figure out if there is literature that supports this response in humans and if it is attributable to faster hip responses (M/L) than ankle or if it is a function of limitations in degrees of freedom]. Figure 16 illustrates this phenomenon by showing four subjects' responses to a 45° perturbation. The first row shows the tactors off configuration, the second row the 4 column display, and the third row the 16 column display. The asterisks mark the peak tilt displacement of the various subjects' trials. Given the associated time delays of receiving the vibrotactile sensation, processing the information and responding with an appropriate motor command, it is possible that the subjects have simply made an initial ballistic correction to the discrete perturbation instead of using the information supplied by the device. The device becomes useful, as is evidenced by the discrete perturbation results, during the steady state control problem – remaining within the dead zone following the initial recovery from the perturbation. In this case, we do observe the effect of the device in terms of a significantly reduced time spent outside of the dead zone and a smaller RMS tilt value in the five-second interval following recovery.

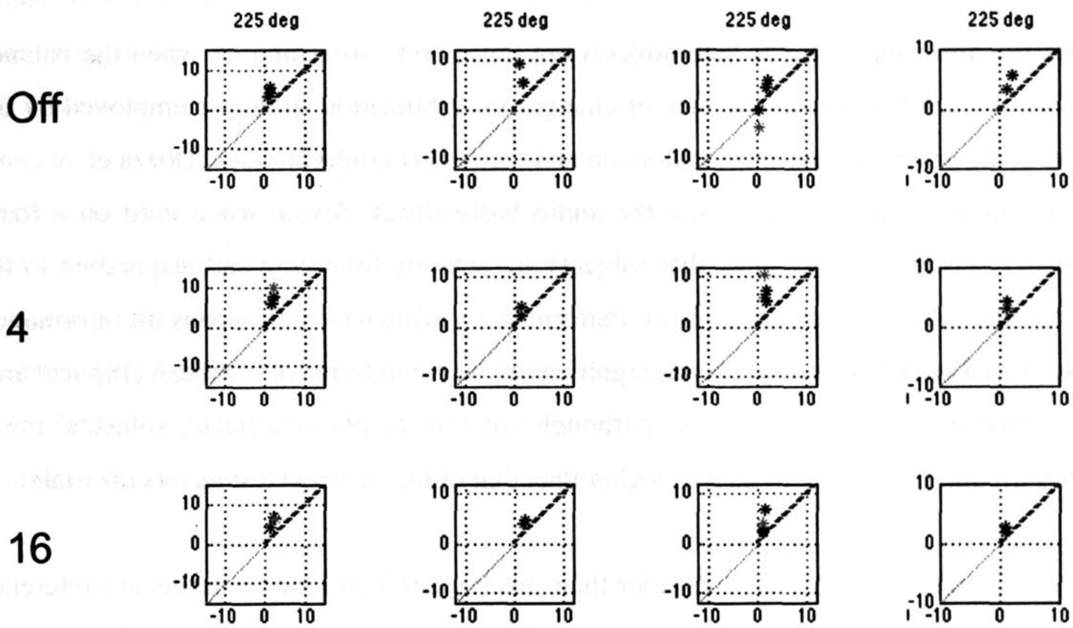

Figure 16. Peak displacements for four subjects during discrete perturbations

Top row: Four subjects' peak tilt displacements without tilt feedback
Middle row: Four subjects' peak tilt displacements with the 4 column display
Bottom row: Four subjects' peak tilt displacements with the 16 column display

The results from the continuous perturbation study suggest that subjects preserve the reduced RMS tilt values following both repeated exposure to the stimulus and training with the balance device. This result could have substantial impact in terms of balance rehabilitation training. If repeated exposure to a continuously moving platform coupled with the use of a vibrotactile biofeedback device were used, patients might be expected to experience improved postural control both immediately following the training as evidenced by the short-term retention findings and over a period of time up to a month as shown by the second day testing results.

The COM results, approximated by the motion-sensing system mounted on the lower back of the subject, did not correspond dynamically with the COP results. This is likely due to the fact that the latter reflects corrective torques, and therefore does not accurately mirror the movements of the COM, which the subject is tasked to control.

This research demonstrates that subjects are not merely stiffening up when the balance device is turned on due to the lack of changes in stabilization strategies employed by the head and trunk during both the tactors on and tactors off configurations. Dozza et. al came to a similar conclusion regarding the audio biofeedback device when used on a foam surface. Our results also show that subjects are actively following and responding to the information that is presented to them. Performance during which erroneous tilt information was played back to the subjects was significantly worse in terms of the mean elliptical area encompassed by the subjects' tilt. Although not statistically significant, subjects' mean RMS tilt during the junk trials was higher than that of the two post test tactors off trials.

Some subjects were able to rank order their display configurations in terms of preference while others were not. Four of the eight subjects preferred the 4 column display, two preferred the 16 column display, one preferred the 8 column, and one preferred the 4 independent column display. Preferences did not correlate with performance; one subject preferred the 16 column display, but it was that subject's worse display in terms of RMS tilt and ellipse area. All subjects who participated in the study felt that they would like to have a device for use in their daily lives. Two specific examples mentioned of when the device might be useful in activities of daily living included: 1) walking the dog on a soft grassy/muddy surface at night and 2) navigating in the dark after getting out of bed at night on a carpeted surface. It is interesting to note that both examples involve reduced visual and altered proprioceptive inputs.

Ultimately, any device that is manufactured will have to be customizable to the individual wearer's needs. For example, although we have used the device to display the actual tilt of the individual and have left it up to him/her to figure out the movement/direction necessary to null out the vibration, some may respond better to a display that indicates the necessary movement. Additionally, one must consider all of the possible locomotor activities in which the device may be used– not just standing applications. To date our research has focused on validating the device in quiet and perturbed normal stance. Additional research must look into the displays necessary to address the postural control issues during narrowed stance (Romberg) and gait; the challenge becomes one of maintaining control of M/L balance.

It remains unclear what effect if any such a device will have on reducing the risk of falling in vestibular compromised and the elderly. The research involving the vibrotactile balance prosthesis published thus far has shown that the vestibular compromised, to varying degrees ranging from severe to well compromised, can use the information displayed by the vibrotactile balance prosthesis to reduce their sway and sway area. Whether or not this translates to a reduced risk of falling remains to be seen.

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Study III

Assessment of a vibrotactile sensory substitution device for improving medial-lateral gait stability

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Abstract
Eight vestibular deficient subjects participated in a four hour training and experimental session that explored the usefulness of a wearable vibrotactile balance prosthesis during locomotion. Subjects trained for approximately 45 minutes with a vibrotactile balance prosthesis that provided real-time feedback regarding their medial-lateral (M/L) tilt. Two feedback configurations were evaluated in a proof-of-concept study during various locomotor tasks, which included slow and self-paced walking, perturbed locomotion trials, walking on a foam surface, and walking along a narrow walkway. The average tilt offset and root-mean-square (RMS) tilt was calculated for all locomotor tasks. Additionally, during slow and self-paced walking trials, stance width and upper body dynamics were assessed. A modified five point Likert scale was used to assess the subject's impression regarding the usefulness of the device in improving stability. Three well-established balance-related subjective questionnaires were also completed. Use of roll tilt feedback resulted in a significant decrease in roll sway for the narrow stance walking task. Step width and step width variability were significantly reduced during vibrotactile feedback trials compared to trials without feedback. A significant correlation between the Dizziness Handicap Index score and the percent change in roll sway provided some insight into the type of patients that could potentially derive the greatest benefit from such a device. The most severely sensory-deficient subject presented, provided one example of how a visible asymmetrical roll tilt of the trunk could be eliminated when vibrotactile feedback signaling roll tilt was provided. No significant difference was identified between the two device displays evaluated however, subjects expressed preferences for one over the other on an individual basis. This preliminary study shows that vestibular-deficient patients can decrease their M/L RMS tilt and significantly decrease their step width variability during challenging locomotor tasks by using vibrotactile tilt feedback.

Background

Sensory substitution, conveyed in the in the form of visual, tactile, and audio cues has proven successful in numerous applications to supplement or in some cases completely replace missing sensory information. Commonly employed tactile cues for example, include Braille characters and vibrating pagers and cell phones.

Canes, crutches, and walkers are mobility aids traditionally used to reduce the risk of falls and mitigate balance problems for elderly individuals [1]. Aside from offering mechanical support, light fingertip contact with mobility and assistive devices provide sensory information about body orientation. Jeka and Lackner have shown that light touch contact was as effective in reducing postural sway as vision of surroundings or force contact, when compared to the no contact eyes closed condition during Romberg stance [2]. Jeka et al. also demonstrated that with light touch, bilateral vestibular loss patients could significantly reduce postural sway when standing in the dark [3]. In fact, the cane has been found to have a similar effect on postural stability as light touch with the fingertip in both healthy and vestibular-deficient individuals [1, 4]. Concurrent with this finding is that canes are regularly prescribed for patients with balance disorders, because patients with balance deficits often use light touch of the surrounding surfaces to stabilize themselves while standing and walking [1, 5].

Assistive devices however, are not without their drawbacks. Wright et al. argues that the use of assistive devices such as canes and walkers increase the risk of falling due to the increased cognitive demands associated with attending to multiple tasks simultaneously [6]. Additionally, a cane or a walker used for sensory information versus biomechanical support can potentially interfere with compensatory stepping mechanisms. Bateni et al. [7] found that compensatory stepping behavior was significantly altered in a group of healthy controls when a walker or cane was used. Collisions between the swing-foot and the assistive device occurred 60% of the time when using a walker and 11% of the time with the cane when medial-lateral (M/L) support surface translations were provided. Significant reduction in average lateral step length [7], the primary control mechanism for maintaining

M/L stability and recovering M/L stability following a surface perturbation [8-11] was also reported. Unrestricted M/L movement of the legs during locomotor activities is preferable for those individuals with intact musculoskeletal systems. Finally, assistive devices often draw undue attention to individuals with balance disorders.

The medical, social and economic impacts of falls are devastating. Complications related to falls can cost health care providers/insurers approximately $19,000 per fall for a total aggregate annual expenditure upwards of $20 billion [12]. Fear of falling leads to a greater increase in balance, gait, and cognitive disorders [13], affects one's ability to perform activities of daily living without assistance [14, 15], and leads to deteriorating quality of life [14]. Hip fractures, in which falling is a necessary but not sufficient condition for the occurrence of a hip fracture [16], resulting from falls are debilitating and can result in nursing home placement, need of a wheelchair, and morbidity. Perhaps of some surprise given the public's perception of the role of osteoporosis in hip fractures, Hayes et al. reports that falls to the side resulting in hip impact raise the risk of hip fracture from 6- to 30-fold, compared to threefold increases in risk associated with 1 standard deviation reduction in hip bone mineral density [17]. Greenspan et al. found that the only significant fall characteristic associated with a hip fracture was fall direction; whether or not an individual fell to the side. Incidence of hip fracture compared between fracture patients and controls did not significantly differ as a function walking, falling from standing height, or the use of the hands to break the force of the fall [16].

A wearable balance prosthesis could supplement or replace missing sensory information by providing an external cue of verticality without restricting M/L leg motion. Such a device has the potential to increase an individual's confidence and improve their ability to perform activities of daily living. Additionally, as opposed to the undue attention that canes and walkers draw to individuals with balance disorders, balance prostheses has the advantage of being concealed under or integrated into the clothing.

Various modes of delivering sensory substitution including electrotactile, vibrotactile, and auditory body sway biofeedback have been effective in improving postural stability during stationary tasks while simultaneously being fairly easy to use. The challenge increases however, when designing a device that provides meaningful information during activities involving gait. Limited studies have been performed during gait tasks and those reported performed have been limited to heel-to-toe walking and walking over a 3 m range. Hegeman et al. [18] tested six compensated bilateral vestibular loss patients both with and without auditory biofeedback in a battery of tests that included semi-stance (walking 8 tandem steps with and without foam support surface) and gait tasks (self-paced walking over 3 m while horizontally rotating the head, vertically pitching the head, or with eyes closed; get up from stool and walk 3 m; and walking up and down stairs without handrails). The authors reported that the vestibular-deficient patients were able to perform the gait tasks as well the healthy controls and that auditory biofeedback was not effective in reducing sway during gait tasks, although for some gait tasks a decrease in A/P trunk sway with velocity feedback was observed.

This pilot study examines the effect of providing M/L tilt information to vestibular-deficient patients during various locomotor tasks to determine whether or not real-time vibrotactile tilt feedback can be used to make adjustments to their trunk tilt position.

Methods

Subjects

Eight subjects with histories of vestibular deficit were recruited to participate in this study. All had previously participated in standing studies involving earlier versions of the balance prosthesis [19, 20]. Obese subjects were excluded from this experiment due to sized limitations of our device. All subjects had computerized dynamic posturography (EquiTest™) Sensory Organization Test composite scores less than 75. These tests are designed to make A/P sensory inputs unreliable while standing. Posturography scores, ages, and gender are noted in Table 1.

Table 1. Vestibulopathic patient demographics

Subject ID	Age	Gender	SOT Score	Sot 5	SOT 6	MCT Score
1	61	M	N/A	35, 63, 70	56, 61, 45	148
2	51	F	56	Fall, 26, 45	Fall, Fall, 45	158
3	61	M	N/A	52, 61, 69	17, 71, 46	155
4	33	M	46	Fall, Fall, Fall	Fall, Fall, Fall	151
5	66	F	49	Fall, Fall, Fall	Fall, Fall, Fall	161
6	61	M	70	Fall, 52, 64	66, Fall, 79	N/A
7	55	M	49	Fall, Fall, Fall	Fall, Fall, Fall	N/A
8	40	M	46	Fall, Fall, Fall	Fall, Fall, Fall	138

Legend:
SOT – Sensory Organization Test: Normal mean composite scores are 80.2 for 20-59 yrs and 76.9 for 60-69 yrs, 5th percentile (abnormal) limits are 68.5 for 20-59 yrs and 70.0 for 60-69 yrs
MCT – Motor Control Test: Normal mean composite scores are 143.0 for 20-59 yrs and 151.8 for 60-69 yrs
5th percentile (abnormal) limits are 161.0 for 20-59 yrs and 170.8 for 60-69 yrs
N/A – Not available

Equipment

The vibrotactile feedback device comprises an inertial sensing assembly, a Controller Area Network (CAN) bus with a central processing unit built on a PC104 platform, and a vibrotactile display. The Honeywell HG1920 high performance inertial sensing assembly houses three accelerometers and three gyroscopes based on Draper MEMS technology (Figure 1). A detailed description of this device can be found elsewhere [19, 21-24]. The vibrotactile elements, referred to as tactors (Tactaid, Cambridge, MA), provided a 250 Hz constant amplitude (200 mA) stimulus. Position tilt estimates were displayed on the subject's torso via a 3 row by 2-column tactor array (Figure 2). Rows of the array were used to display estimated tilt magnitude and columns (one mounted on the subject's right side, one on the left) were used to display tilt directions. Prior to testing, the magnitude of the VTTF display was adjusted according to the maximum tilt deviation during self-paced walking. The first tactor row activation threshold was set at 1° and all three rows were activated if the subject tilted 50% of their maximum tilt angle, respectively. The subjects were instructed to move in the opposite direction of the vibration.

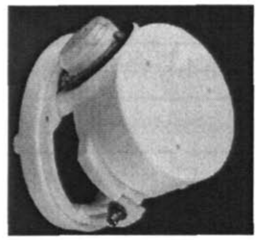

Figure 1. Inertial sensory assembly unit

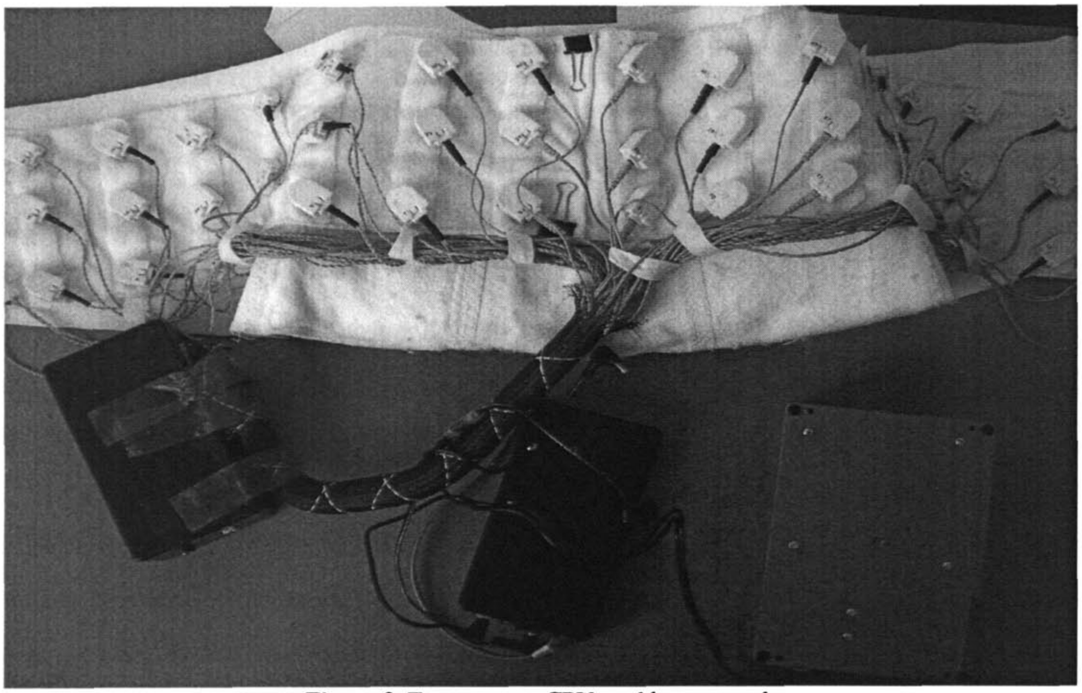

Figure 2. Tactor array, CPU, and battery pack

Three different vibrotactile displays were evaluated; No vibrotactile feedback (tactors off), continuously displayed roll tilt (continuous), and roll tilt displayed only during the heel-strike event (intermittent). The intermittent display provided feedback for 200 ms beginning at the heel-strike event detection. A predefined vertical acceleration threshold detected heel strike. The continuous and intermittent tactor displays were selected to test the effect of displaying tilt feedback at different points during the gait cycle. The intermittent display option was based on recent findings by Bent et al. [25] that demonstrated that vestibular information is used during double support phase to affect the M/L position of the subsequent foot placement.

An 18 m long by 1.2 m wide by 10.2 cm high wooden walkway (Figure 3) was used in this study for four out of the five locomotor tasks. A custom-built moveable 2.1 square meter platform called BALDER (BALance DisturbER) embedded in the wooden walkway generated surface perturbations [26]. Kinematics were collected using the Optotrak 3020 system (Northern Digital, Waterloo, Ont.) for half of the subjects during three of the five

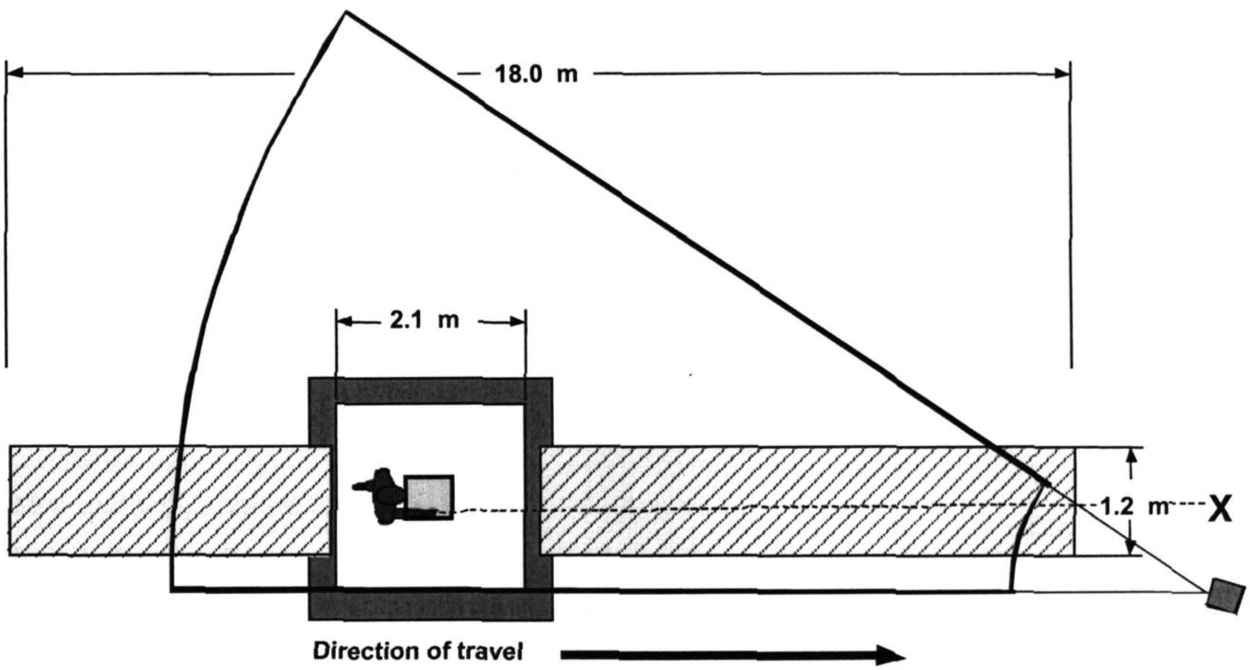

Figure 3. BALDER platform and wooden walkway

locomotor tasks (slow-paced, self-paced, and perturbed walking, see protocol below). Rectangular arrays consisting of six infrared emitting diodes (IRLED) were placed on the subjects' mid-tibias, pelvis, sternum, and head (Figure 4). The IRLED sampling rate was 1500 Hz and the array positions were estimated at 40 Hz. The 3020 camera was placed at the end of the BALDER platform walkway and captured a viewing range of approximately 12 m. The remaining locomotor task was performed on foam. The foam used in this study was 10 cm thick medium density foam (Sunmate Foam, Dynamic Systems, Inc., Leicester, NC) and when placed end-to-end created a 7.3 m long walkway.

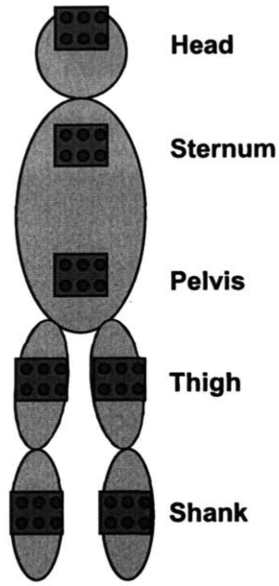

Figure 4. IRLED body marker placement

Protocol

Data were collected in the Injury Analysis and Prevention Laboratory in the NeuroMuscular Research Center (NMRC) at Boston University. The Massachusetts Eye & Ear Infirmary, Boston University, and the Massachusetts Institute of Technology Institute Review Boards approved experimental protocols. Prior to testing, informed consent was obtained.

Locomotor tasks ranging from easy to challenging were purposely selected for this exploratory study since the utility of vibrotactile feedback for improving locomotor postural stability was unknown; prior to this study, vibrotactile feedback had only been evaluated during quiet and perturbed stance. The tasks, ranging from easiest to most challenging based on pilot data from six healthy controls included: self-paced and slow-paced walking, controlled surface perturbations during locomotion, walking with narrowed base-of-support, and walking across foam. Table 2 displays subject participation by locomotor task. Prior to testing, subjects participated in a 45-minute training session which included quiet stance while maintaining the Romberg position [27], heel-to-toe walking

Subject ID	Slow	Self	Perturbation	Narrow	Foam
1	x	x	x	x	
2	x	x	x	x	x
3	x	x	x	x	x
4				x	x
5	x	x		x	x
6	x	x	x	x	x
7	x	x	x	x	x
8	x	x		x	x

Table 2. Subject participation by locomotor task

with eyes open and eyes closed, slow-paced walking, self-paced walking and walking with surface perturbations. Subject wore their own athletic or walking shoes for all locomotor tasks with the exception of foam walking in which case they were sock footed. The laboratory lighting was dimmed to eliminate vertical cues and increase the challenge of the tasks. A safety spotter stood alongside the subjects during all training and testing trials for all locomotor tasks.

Slow and self-paced walking

For platform walking trials, subjects were instructed to walk along the walkway while fixating on a dimly lit visual target that was positioned at eye-level beyond the end of the walkway. Depending on the type of trial, subjects were cued to walk at either their preferred pace or at a fixed pace of 50-60 steps per minute maintained by an electronic metronome. Subjects completed nine trials for each walking pace. The nine trials were broken down into three trials without tilt feedback, three trials with continuous feedback, and three trials with intermittent feedback. Data were collected in sets of three consecutive trials for each tactor display.

Perturbed gait

Two different perturbations and one control case (no perturbation) were included in the perturbation protocol. The perturbation protocol consisted of 27 walking trials, during which unannounced surface perturbations were delivered at one of two amplitudes (5-cm or 10-cm) based on the subject's comfort level and in opposite directions (+45° termed for forward-right [FR] and +225° termed backward left [BL] as measured clockwise from the subject's direction of march). The perturbations were applied to the right foot during single leg stance following a right heel-strike force-plate threshold triggered delay (100 ms or 120 ms depending on the translation direction) to ensure that the subject's left leg was in its swing phase. One control case (no perturbation) was included in the experimental protocol to make perturbation trials unpredictable. Perturbation trials were rehearsed as needed to familiarize subjects with the novel stimulus issued by the BALDER platform; practice trials using both 5 and 10 cm magnitude perturbations were repeated until stutter stepping (a quick corrective step, as opposed to a normal stride length) was eliminated. Subjects were not informed of the exact number of trials to be performed during the experimental session in order to minimize the effects of subjects predicting and counting perturbations. Subjects were instructed to walk at their preferred pace while fixating on a dimly lit visual

Trial Number	Display	Trial Type
1	Continuous	Control
2	Intermittent	Control
3	Intermittent	FR
4	No Tactors	Control
5	Continuous	FR
6	No Tactors	BL
7	No Tactors	Control
8	Intermittent	FR
9	Continuous	Control
10	No Tactors	BL
11	Continuous	FR
12	Continuous	BL
13	Intermittent	Control
14	No Tactors	FR
15	No Tactors	BL
16	Continuous	FR
17	Intermittent	Control
18	Intermittent	FR
19	Continuous	Control
20	Intermittent	BL
21	Intermittent	BL
22	No Tactors	FR
23	No Tactors	FR
24	Continuous	BL
25	No Tactors	Control
26	Continuous	BL
27	Intermittent	BL

Table 3. Perturbation protocol by display and perturbation trial type.

target that was positioned at eye-level beyond the end of the walkway. Table 3 displays the perturbation protocol by prosthesis display configuration.

Narrow stance walking

Subjects completed 9-12 narrow stance walking trials following a short training session both with and without feedback on the narrow stance-width walkway. The boundaries of the walkway were adjusted from 20.5 to 30 cm wide based on the subject's self-assessment and the experimenters' assessment of the task difficulty. The more narrow the width of the walkway, the more difficult the task. The first boundary was formed by the edge of the wooden walkway (10.2 cm rise from the floor). Positioning a long dark mat along the wooden walkway at the appropriate distance from the first boundary formed the second boundary. Subjects were instructed to walk at their preferred pace while fixating on a dimly lit visual target that was positioned at eye-level beyond the end of the walkway. Data were collected in sets of three consecutive trials for each tactor display. The first and last sets were always performed without feedback (tactors off configuration). The intermittent display was occasionally omitted from this protocol because it performed poorly (vertical force threshold was not consistently exceeded) when subjects gingerly placed there feet on the walkway.

Walking on foam

The high-density foam walkway was used to distort proprioceptive inputs. Subjects completed nine foam walking trials following a short training session both with and without feedback. Subjects were instructed to walk at their preferred pace while fixating on a dimly lit visual target that was positioned at eye-level beyond the end of the walkway. Data were collected in sets of three consecutive trials for each tactor display. The first and last sets were always performed without feedback (tactors off configuration). The intermittent display was omitted from this protocol because the foam surface dampened the vertical force such that the vertical force threshold was not consistently exceeded.

Subjective Measures

Subjects were asked to verbally respond to three subjective scales following the third repetition of each vibrotactile display set. A modified five point Likert scale [28] was used to assess the subjects' impressions regarding the usefulness of the device in improving stability. Subjects could select from five responses when asked to complete the statement: "During the last 3 walking trials, I found the device to be": (1) very unhelpful; (2) moderately unhelpful; (3) neutral-neither helps nor hurts; (4) moderately helpful; and (5) very helpful. Subjects were also asked to rate their perception of task difficulty and fatigue level on a verbal analog scale of 1 (very easy balance task, no fatigue, respectively) to 10 (most difficult balance task, completely fatigued, respectively). The perception of task difficulty was used to adjust the perturbation magnitude and width of the narrow-stance support surface so that each subject found the task to be of similar challenge (7 out of 10). The fatigue scale was used to determine when additional rest periods were to be taken (greater than 6 out of 10).

Additionally, each subject completed three standardized surveys including the Dizziness Handicap Index (DHI) [29], the Activities-Specific Balance Confidence Scale (ABC) [30], and the Vestibular Activities of Daily Living Scale (ADL) [31] either prior to arrival, during breaks in the experimental protocol, or after participating in the study. These surveys assessed perception of dizziness, level of independent functioning and the impact of both on the subjects' lives. The DHI is a 25-item self-assessment scale designed to quantify the functional, emotional, and physical effects of dizziness and unsteadiness. The scale runs from 0 (no perceived handicap) to 100 (severe perceived handicap) and has been shown to yield reliable and valid measurements in patients with vestibular disorders [29]. The ABC Scale is a questionnaire that evaluates a patient's confidence in performing 16 activities of daily living. Scores range from 0 indicating no confidence to 100 indicating complete confidence in the subject's ability to complete the task without losing their balance [30]. The ADL is a questionnaire that assesses self-perceived disability in patients with vestibular impairments and includes 27 activities of daily living. Scale ratings range from 1 (independent) to 10 (ceasing to participate in the activity).

Data Analysis

The vibrotactile feedback device data were sampled at 87 Hz. Gait initiation and termination were excluded prior to calculating all parameters. The root-mean-square (RMS) of the resultant roll and pitch tilt, velocity, and acceleration vectors were calculated over a fixed distance for each of the walking trials. The roll RMS for example, was calculated by taking the square root of the squared sum of the estimated roll as derived from the processed inertial sensing assembly data. Percentage of time spent in the each of the tactor firing zones (off, first level activated, all three levels activated) and average tilt (referred to as tilt bias) were also calculated on a trial-by-trial basis. Pacing information was approximated by considering the number of hand counted gait cycles and the sampling rate.

Two gait cycle events were used to sample the subjects' 40 Hz kinematics data. One event was estimated using shank position data while the other used its derivative, shank velocity. The first event corresponded to the instance when the A/P positions of the two shanks were equal. The second event estimated heel strike at the instances when the shank velocities were equal. These events were selected because they occurred once per gait cycle and had been previously used to assess control and vestibulopathic locomotor recoveries to surface perturbations [10, 11]. At each A/P shank crossing and heel strike event, the M/L distance between the sternum and shank position of the stance leg was calculated to provide an estimate of the M/L moment arm. The estimated moment arm illustrates how M/L control is applied by way of lateral foot placement relative to body center of mass during recovery from a surface perturbation. The moment arm values for each perturbation type were averaged for each subject on a step-by-step basis and then normalized using the mean values from the non-perturbation trials in order to reduce the individual differences in moment arm magnitude between subjects.

Step width was calculated by taking the difference between the M/L shank positions. Step length was calculated by taking the difference between the A/P shank positions. Step width variability was characterized by averaging the standard deviation step width values for all single support phases within a trial. Single support phase was estimated from the heel strike event. The RMS M/L sternum position was calculated from the seventh heel strike event (first recovery step following perturbation) until the twelfth heel strike to assess trunk sway.

The anchoring index, a previously published parameter for characterizing head and trunk stabilization strategies in the frontal plane during unperturbed locomotion [32-34], describes the relative angular distribution of the body segment being considered with respect to axes linked to an inferior anatomical segment. The anchoring index is defined as:

$$\mathrm{AI} = [(\sigma_r) - (\sigma_a)]/[(\sigma_r) + (\sigma_a)]$$

where, σ_a is the angular dispersion of any body segment and σ_r is the standard deviation of the relative angular distribution of the body segment being considered with respect to axes linked to an inferior anatomical segment. A positive value of the sternum anchoring index for example, indicates a tendency for trunk stabilization in space rather than on the hip (pelvis), whereas a negative value would indicate a tendency for trunk stabilization on the hip (pelvis) rather than in space.

Results

All post-processing was performed using Matlab (The MathWorks, Natick, MA). Statistical analyses were conducted using STATA (StataCorp LP, College Station, TX). A one-way, repeated-measure analysis of variance (ANOVA) was performed on each dependent variable to determine the effect of the vibrotactile device. The level of significance was set at $p< 0.05$. Overall group results and an in-depth analysis of an illustrative vestibular-deficient subject are summarized below.

<u>Group results</u>

Individual repetitions composing sets of similar tactor display trial types were averaged together on a subject-by-subject basis for each locomotion task. No significant changes were observed for RMS roll and pitch tilt, velocity, and acceleration, percentage time spent in dead zone, and tilt bias among display types (tactors off, continuous, and intermittent) for any of the locomotor tasks. Given the expectation that roll sway was most likely to be affected because roll tilt feedback was provided, the following results focus on the changes observed in RMS roll during the experimental locomotion tasks. The greatest changes in RMS roll were observed in the foam and narrow stance walking tasks. Figure 5 shows the absolute RMS M/L tilt averaged over all subjects for foam walking.

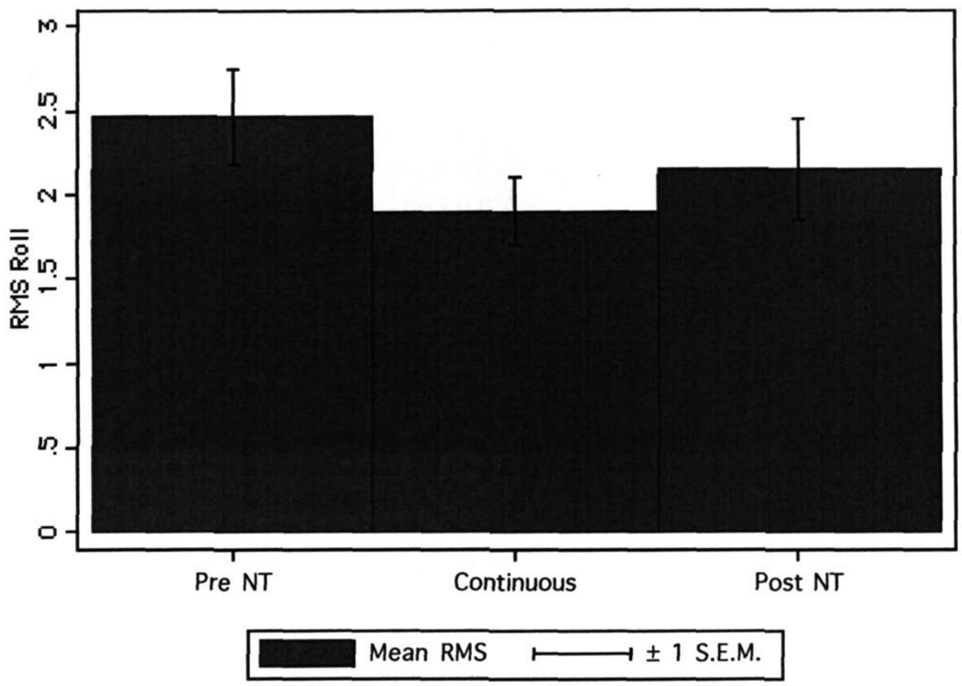

Figure 5. Foam walking average RMS roll

Pre NT denotes the first "no tactors" set completed. Post NT denotes the second set. The error bars represent the standard error of the mean.

Four of the seven subjects that completed the foam-walking task also completed two sets of continuous feedback trials (Figure 6). Although not significantly lower, the second set of three continuous display trials had a lower RMS roll value compared to the first set of three continuous trials.

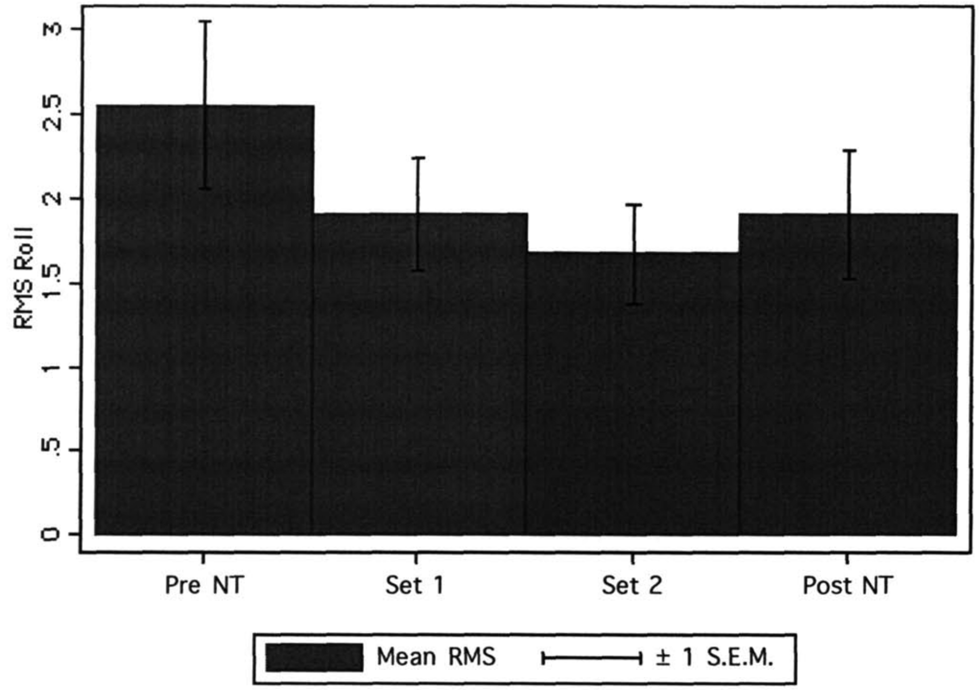

Figure 6. Average RMS roll for two NT and continuous display sets of foam walking trials

Roll tilt bias, the average roll seen over the entire trial, is shown for foam walking in Figure 7. Although not significant, the average roll tilt bias for continuous trials is negligible compared to the first set of tactors off trials. Interestingly, the subjects tended to have a small static lean (tilt bias) in the opposite direction during their last set of tactors off trials compared to their first set.

Figure 8 shows the average RMS roll values for the narrow stance walking trials. The use of tactors (continuous and intermittent) significantly decreased the average RMS roll compared to the tactors off display ($p<0.0297$). Use of the continuous display (n=8) resulted in a slightly lower RMS roll value compared with the intermittent (n=5) display.

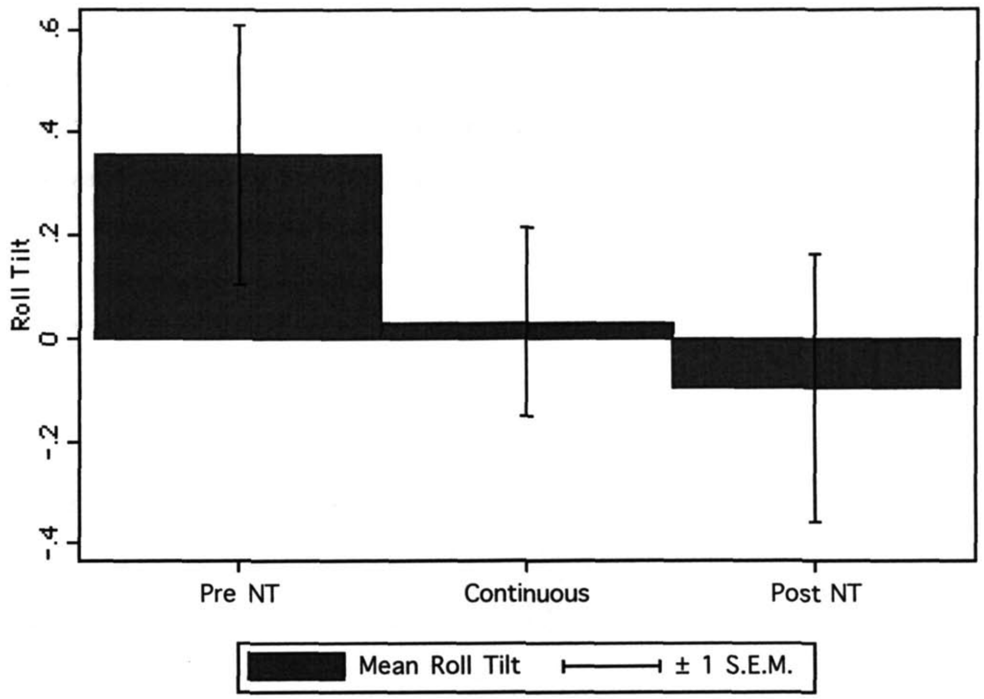

Figure 7. Average roll tilt bias for foam walking trials

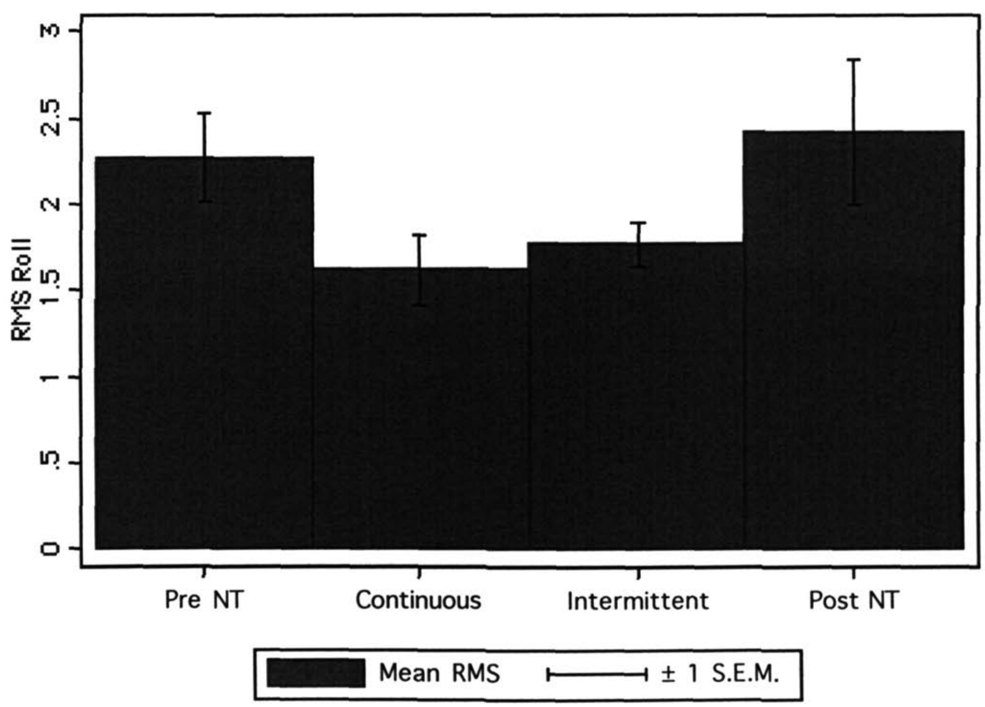

Figure 8. Average RMS roll for narrow stance walking trials

Slow and self-paced RMS roll results are shown in Figures 9 and 10, respectively. Both continuous and intermittent displays slightly reduced RMS roll sway compared to the tactors off display. Continuous feedback had a greater effect on RMS roll sway during self-paced gait compared to intermittent feedback.

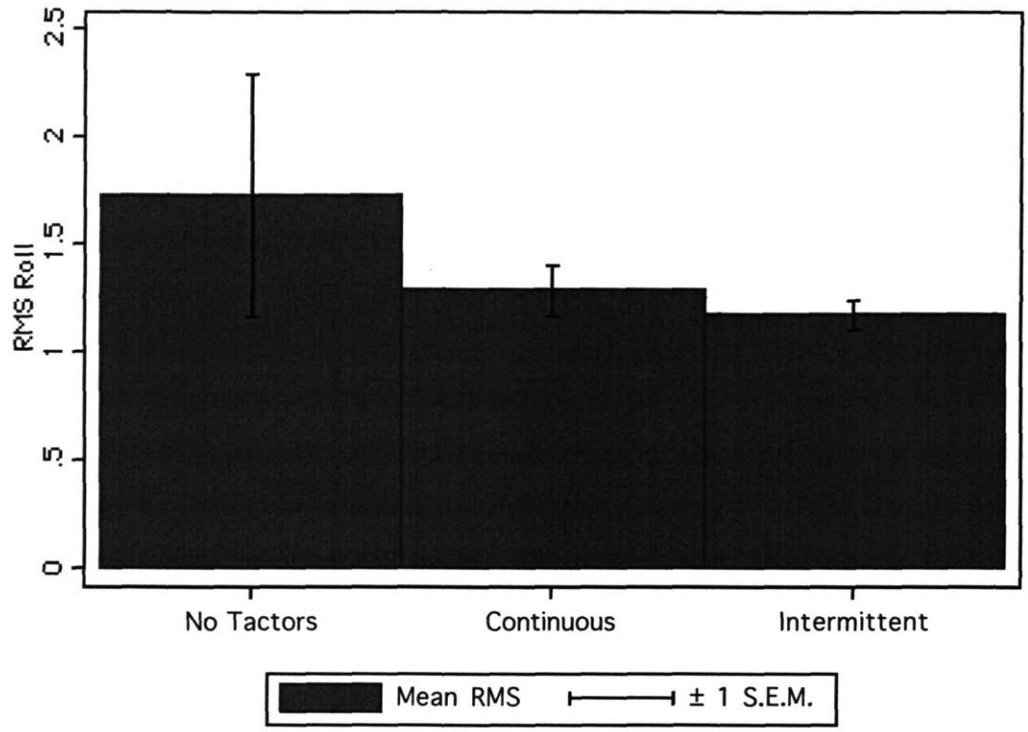

Figure 9. Average RMS roll for slow paced walking trials

Subjects tended to decrease their pace during trials in which the tactors were activated compared to trials in which the device was turned off. This was most notable in the narrow stance, foam, and self paced walking trials. However, although decreased, the change was not significant.

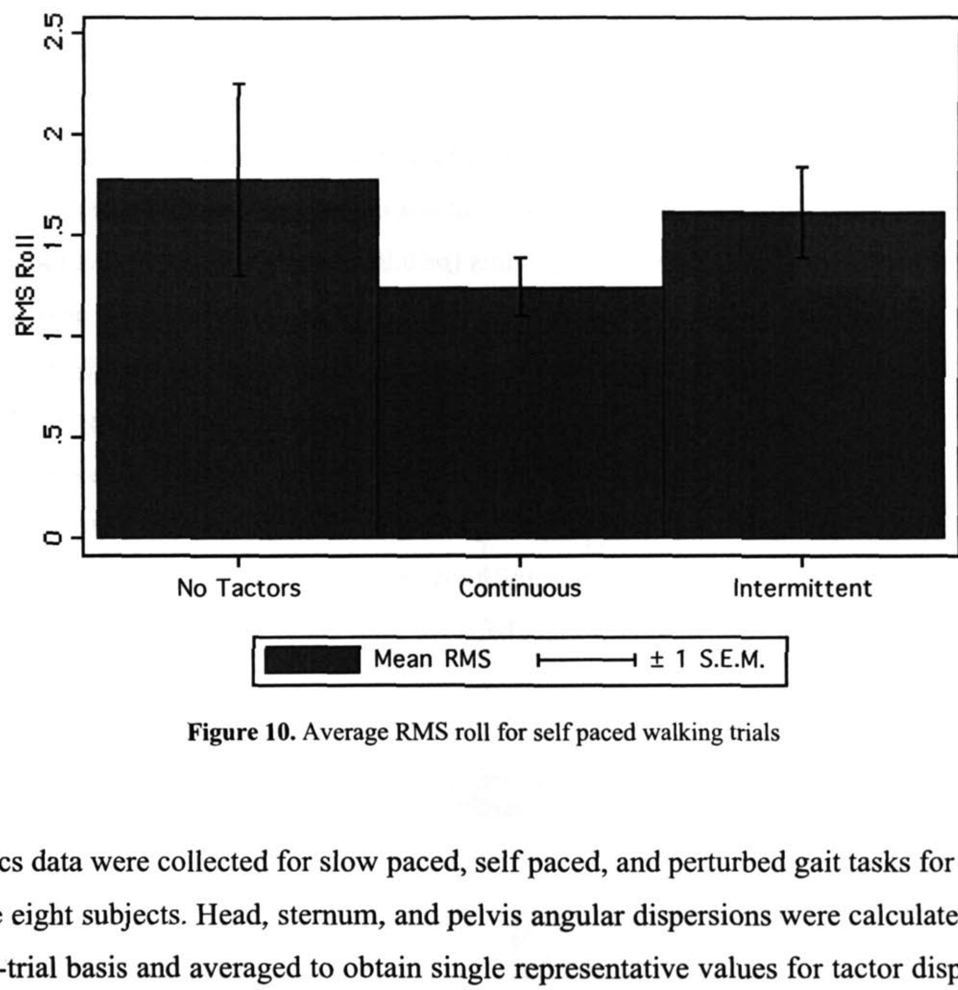

Figure 10. Average RMS roll for self paced walking trials

Kinematics data were collected for slow paced, self paced, and perturbed gait tasks for four out of the eight subjects. Head, sternum, and pelvis angular dispersions were calculated on a trial-by-trial basis and averaged to obtain single representative values for tactor displays for slow and self-paced walking tasks. Head, sternum and pelvis roll dispersions were significantly smaller for continuous and intermittent display trials compared to tactors off trials. Similarly, head, sternum, and pelvis pitch and yaw dispersions also decreased when the prosthesis provided feedback. Anchoring indices for the head and sternum were computed for all axes of rotation. Although the indices significantly differed among tactor displays, no indices changed their sign, which would have indicated a pronounced change in stabilization strategy.

Kinematics data were also used to evaluate the effect of vibrotactile feedback during recovery from surface perturbations. Neither M/L moment arms sampled at A/P shank crossing nor heel strike events were significantly different among tactor displays following forward right or backward left perturbations. Additionally, RMS M/L sternum displacement spanning the first five steps following perturbation among tactor displays.

Average step length sampled at the heel strike event was significantly reduced during slow pace walking ($F(2,9)=17.66$, $p<0.0008$), but not during self-paced walking. Subjects took significantly shorter steps with continuous ($p<0.011$) and intermittent feedback ($p<0.001$) compared to slow pace walking trials without feedback. Step width was also significantly affected by the use of feedback. The average step width sampled at heel strikes significantly differed among tactor displays for slow pace walking ($F(2,9)=32.03$, $p<0.0001$), but not for self-paced walking. Subjects had a narrowed stance width for both continuous ($p<0.000$) and intermittent ($p<0.000$) displays compared to the tactors off display. The average step width variability during single support phase was significantly less with feedback compared to no feedback for both walking tasks.

Figure 11. Step width variability

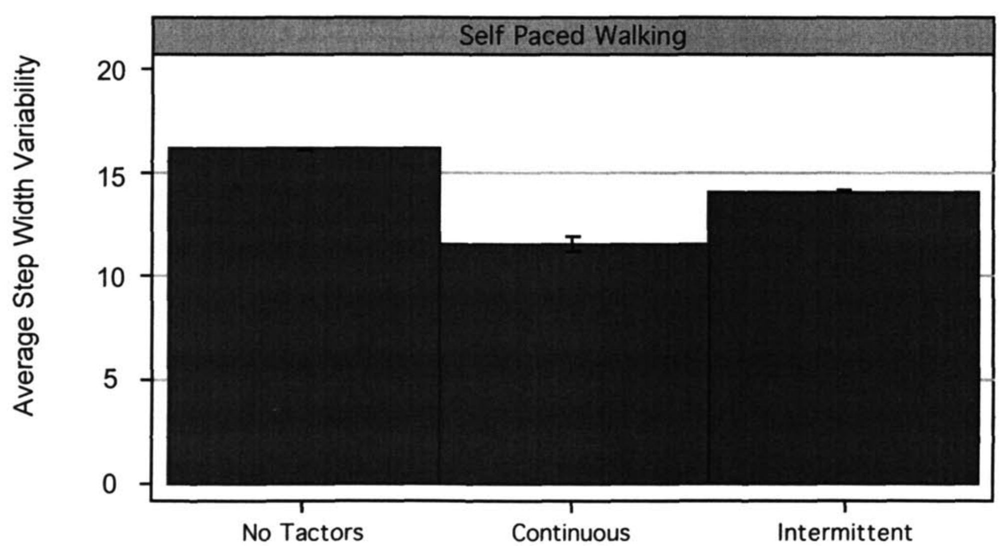

Table 5 shows the tabulated results from the DHI, ABC, and ADL surveys. Eight subjects completed the DHI and ADL questionnaires and five subjects completed the ABC questionnaire. The mean DHI score was 31 (± 22.9 standard deviation) indicating moderate impairment (0 to 30 minimal impairment, 31-60 moderate impairment, > 61 severe impairment). However, one subject's severely impaired score skewed this average. Five out of the eight subjects' DHI scores indicated minimal impairment. The mean ABC score for the subjects was 74.25 (± 12.4) indicating chronic health problems for subset of the group that completed this questionnaire (>80 highly functioning, 50-79 chronic health problems, <50 home bound). Two of subjects' ABC scores indicated chronic health problems and three indicated highly functioning subjects. The mean ADL score was 2.25 (± 1.7) indicating that the subjects that participated in this study are largely independent (1 = independent, 10 = ceasing to participate in the activity).

Subject ID	DHI	ABC	ADL
1	16	73.125	1
2	32	na	2
3	4	na	1
4	46	82.5	2
5	30	53.125	3
6	78	na	6
7	14	81.875	1
8	28	80.625	2

Table 5. Subjective survey results

Figure 12 shows the correlation between the percentage changes in RMS roll when continuous roll tilt feedback was used during narrow stance walking with the subjects' DHI score. Recall that the DHI runs from 0 (no perceived handicap) to 100 (severe perceived handicap). The R^2 value for a simple linear regression was 0.7092, which was significant at the $p<0.05$ level given the sample size considered. The correlation shows that subjects with large DHI scores have the least change in RMS roll tilt with the use of vibrotactile feedback during a challenging task. For the subject with the worst DHI score, the percentage change was positive indicating that an increase in RMS roll tilt was observed with vibrotactile feedback.

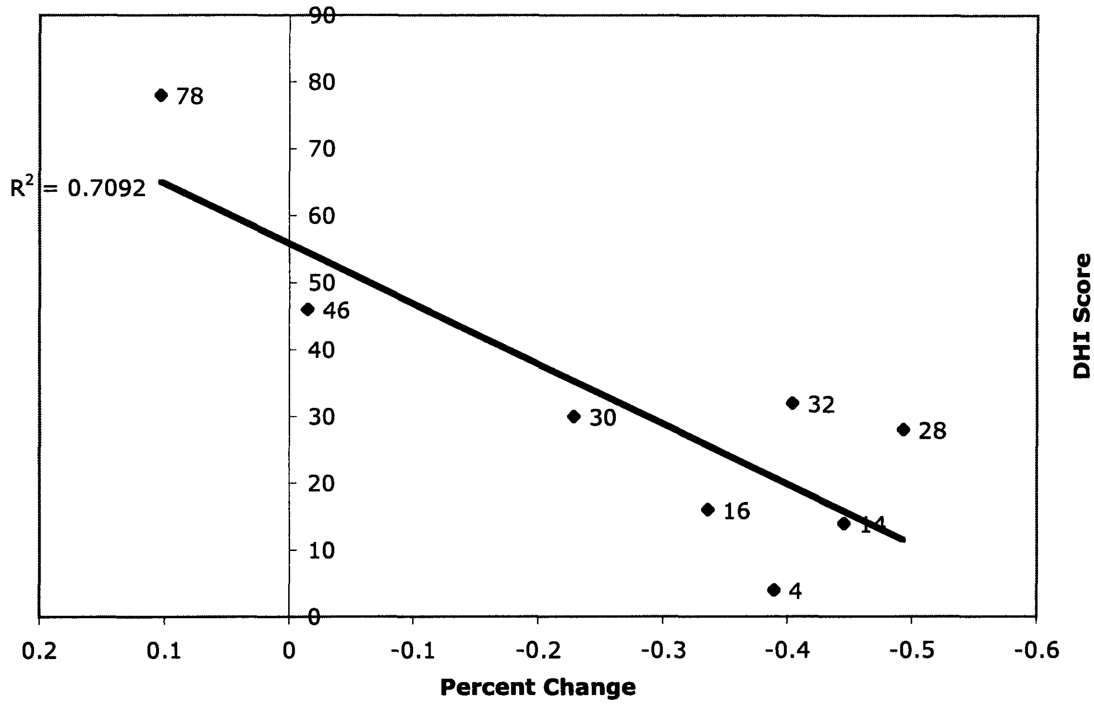

Figure 12. Correlation between percentage change in RMS roll with continuous feedback during narrow stance walking and DHI score.

Illustrative subject

The most severely impaired patient that participated in this pilot study was a 60-year-old male with a history of a right-sided acoustic neuroma, no hearing in the right ear, and reduced vision in the right eye. Functional vestibular testing confirmed no response from his right side. Relevant postoperative symptoms included difficulty with stairs, while walking in poorly lit environments, and during slow-paced walking. His computerized dynamic posturography Sensory Organization Test composite score was 73; a score of 73 means that a subject is performing above the low normal limits, but is below average in performance. This subject completed slow- and self-paced walking, perturbed walking, and narrow stance walking to assess the efficacy of the vibrotactile balance prosthesis. The subject had previously participated in two experiments involving the vibrotactile balance prosthesis during standing tasks [19, 20].

Figure 13 shows the roll tilt in degrees of the subject's trunk using the three different displays during slow-paced walking (60 steps per minute): (1) no tactors; (2) continuous feedback; and (3) intermittent feedback. The horizontal dashed lines represent the second tactor activation threshold; the tilt estimate at which all three rows of tactors are activated. A positive roll angle is defined as the subject's leaning right. From the top plot, it is clear that without the vibrotactile feedback, the subject has considerable lean toward the side of his lesion (right). It is clear from the middle and bottom plots that the subject is using the feedback to control his roll tilt. In both feedback conditions, the subject is able to stay within the ±2° outer limits. However, in the no feedback condition (top panel), the subject ends up oscillating about a 5° rightward tilt. When questioned if he was aware of the tilt bias, he replied "no". He only became aware that he had a tendency to tilt to the right when he received repeated vibrations on his right-hand side during the trials with feedback.

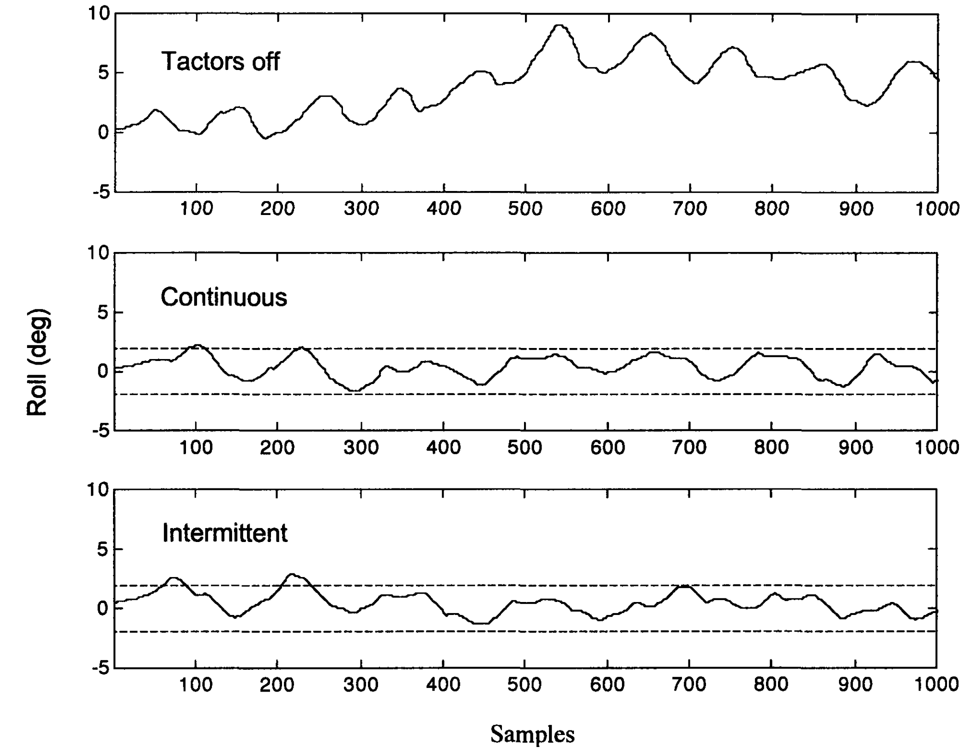

Figure 13. Sample data from illustrative subject

Roll tilt in degrees of the subject's trunk during slow-paced walking. Top panel: No tactors, Middle panel: Continuously displayed feedback, Botton panel: Feedback gated during heel-strike event. The horizontal dashed lines represent the second tactor activation threshold; the tilt estimate at which all three rows of tactors are activated. A positive roll angle is defined as roll to the subject's right.

Figure 14 shows the reduction in RMS roll tilt when the subject used either type of vibrotactile feedback, continuously or intermittent, during narrow stance (beam) walking, slow ($F(2,6)=9.8$, $p<0.0129$), and self-paced walking ($F(2,6)=17.20$, $p<0.0057$). For slow and self-paced walking, both continuous ($p<0.009$, $p<0.038$, respectively) and intermittent ($p<0.021$, $p<0.019$, respectively) feedback resulted in significantly lower RMS roll compared to when no feedback was provided. The effect is more pronounced when the subject is forced to walk at a slower than natural pace. The subject reported feeling most stable when he actively employed a strategy of walking at approximately 110-120 steps per minutes. Figure 15 shows the subject's average roll tilt (tilt bias) by locomotion task. Roll tilt was significantly decreased during slow (F(2,6)=9.59, $p<0.0135$) and self-paced walking (F(2,5)=14.87, $p<0.0079$) with continuous ($p<0.025$, $p<0.010$, respectively) and intermittent feedback ($p<0.030$, $p<0.046$, respectively).

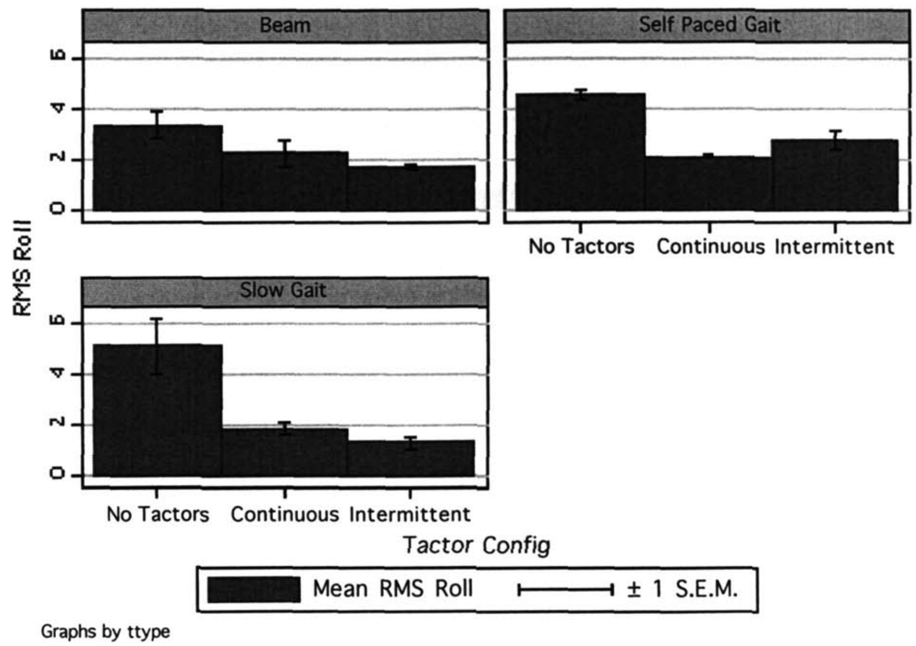

Figure 14. RMS roll during narrow stance (beam), slow, and self paced

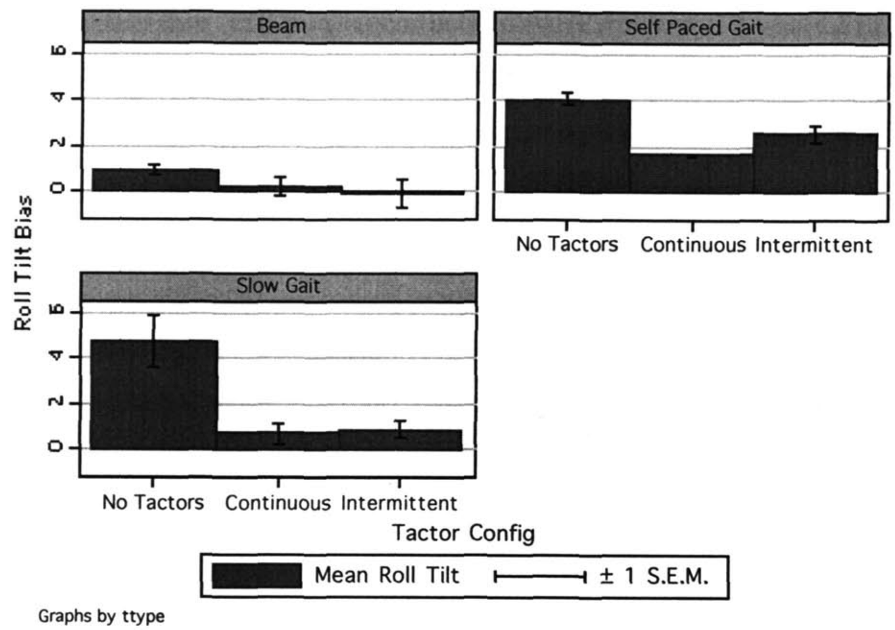

Figure 15. Average roll tilt (tilt bias) during narrow stance (beam), slow, and self paced walking

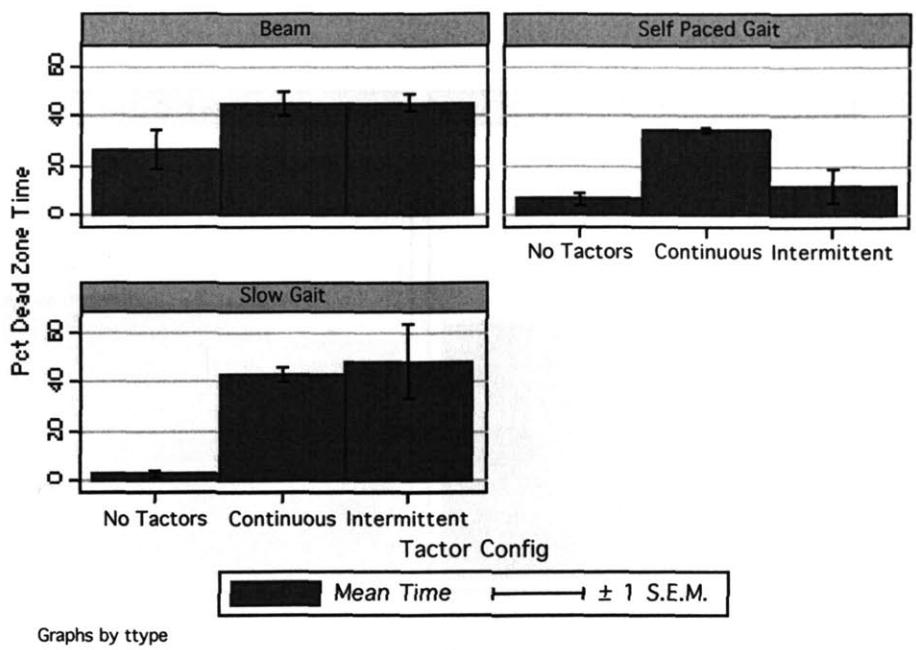

Figure 16. Percentage time spent inside the dead zone for narrow stance (beam), slow, and self paced walking

Figure 16 shows the percentage of time spent within the dead zone (no tactors activated) for narrow stance (beam), slow and self paced walking. Significant increases in time spent inside the dead zone as a function of tactor display were observed for all three locomotion tasks.

Step width variability averaged over single support phases (Figure 17) was significantly reduced with vibrotactile feedback for self-paced walking ($F(2,6) = 11.62$, $p<0.009$); continuous feedback resulted in a significantly ($p<0.009$) smaller step width variability compared with the tactors off display.

Finally, the subject consistently rated the device between a 3 and 5 on the modified Likert scale. For the slow walking task, the subject rated the device as very helpful (5) for the continuous display and moderately helpful (4) for the intermittent display. For the self-paced walking conditions and narrow stance walking, the subject rated the device as neutral (3) for both displays.

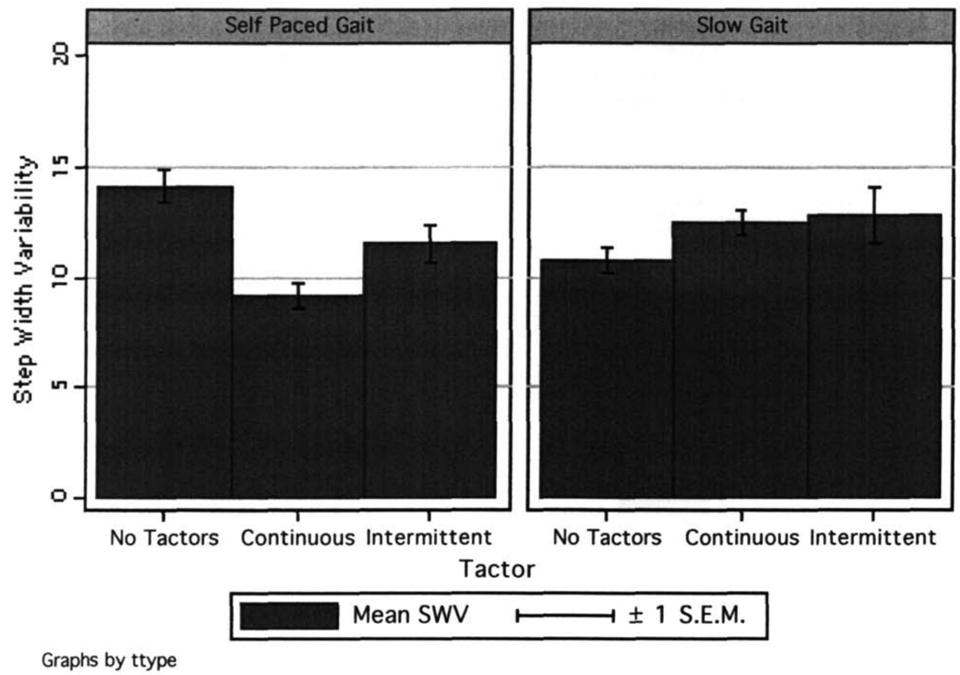

Figure 17. Step width variability average over single support phase for self paced and slow walking

Discussion

Continuous and intermittent vibrotactile feedback based on trunk roll tilt was displayed during various locomotor tasks ranging from self-paced walking to walking across a foam surface that distorted proprioceptive information. Use of roll tilt feedback resulted in decreased roll sway for slow, self-paced, narrow stance, and foam walking tasks. Additional subjects should be evaluated to determine if this trend is significant. Step width and step width variability were significantly reduced during vibrotactile feedback trials compared to trials without feedback. A significant correlation between the DHI score and the percent change in roll sway provided some insight into the type of patients that could potentially derive the greatest benefit from such a device. The illustrative subject provided one example of how a vestibular-deficient patient could potentially benefit from such a balance device (Figure 13). During gait, this individual had a clear and visible asymmetrical roll tilt of the trunk that disappeared almost instantly when vibrotactile feedback signaling roll tilt was provided. No significant difference was identified between the continuous and intermittent device displays on a group basis. However, as evidenced in Figures 14-17, on a subject-by-subject basis, differences were observed.

Based on the subjects' perception of the usefulness of the device, 4.05 and 3.73 out of 5 (where 5 corresponded to very helpful) for the continuous and intermittent feedback displays, respectively, subjects as a whole found the continuous display of information to be more useful. Although there was a slight preference for continuous feedback, it should be noted that subjects reported a higher score (indicating more useful) for continuous feedback across all locomotor tasks. Subjects verbally indicated an increased level of confidence when their tilt information was displayed continuously versus intermittently. However, the subjects' device display preference was not always consistent. When questioned during the experimental protocol, some subjects verbally reported a preference for intermittent feedback. During the post-experiment debrief though, they contradicted their earlier opinion by stating a partiality for the continuous display.

Subjects that preferred intermittent feedback did so based on the notion that their attention could be focused on the device only during balance crises. Subjects that preferred the continuous display felt that the continuous display delivered a better quality of information. The higher quality was attributed to the fact that subjects felt more comfortable knowing that they had the most complete information. One concern that was verbalized about the intermittent display was that it wasn't always clear to the subjects if the device was on. Subjects remarked in post-experiment interviews that they sometimes questioned the device's status if they went walked more than a few steps without receiving a vibrotactile cue. One drawback of the intermittent display was the simple threshold-based algorithm that we used to detect the elevated vertical accelerations of the trunk at the heel strike events. This algorithm worked reliably during the self-paced and perturbation trials. However, when subjects employed a slower, more cautious gait, such as during slow-paced walking and narrow stance walking, the vertical accelerations decreased and therefore the threshold was less sensitive. The worst case was during the foam walking since the high-density foam dampened the vertical accelerations. Vertical accelerations were also dependent on the amount of cushioning present in a subject's shoe.

In general, subjects walked at a slower, although not significantly slower pace when the device was on versus off. Subjects were likely decreasing their gait velocity in an attempt to process and use the feedback information. This finding may suggest that the subjects were not sufficiently practiced at using the device and that long training sessions potentially spanning days may be required for effective use of a biofeedback device during locomotor activities. This is compared to the relatively short training period reported by multiple researchers during standing tasks. Wall et al. reported a 15–30 minute training time for the vibrotactile tilt feedback device [19, 20] and Dozza et al. reported a one-minute training phase [35] for the auditory biofeedback device.

The decreased pace observed across locomotor tasks when the device was turned on also raises the issue of whether or not the presentation of tilt biofeedback creates a more or less stable gait scenario for the subject. Brandt at el. [36] concluded acute vestibular patients are better off running than walking in terms of path deviation. Four patients with vestibular neuritis deviated toward their affected ear when trying to slowly walk in a straight line. On the other hand, when running slowly, the four patients maintained their direction over 10 m and felt more secure. Brandt attributes this finding to the fact that locomotion can be achieved solely by central pattern generators in the spinal cord. If a faster pace is more stable than a slower pace, the implications of slowing down a subject using tilt biofeedback should be further studied to determine if 1) within the walking speed range, does gait velocity affect stability? and 2) can practice attenuate the cognitive workload associated with using the device such that after a sufficient training period with the device, gait velocity can be maintained?

Based on the results of this pilot study, vibrotactile feedback, displayed either continuously or intermittently appears to reduce the RMS roll tilt when subjects are forced to narrow their base of support or walk on a surface that distorts their proprioceptive information. Several questions remain to be addressed before such a device can be commercialized for patient use.

Relationship of decreased M/L tilt to risk of falling

First, does increased roll stability translate to the reduction of falls and/or fall risk? This pilot study does not permit us to draw any conclusions regarding the efficacy of the device to reduce the likelihood of a fall while walking. We can say based on the results that in certain locomotor tasks, RMS M/L tilt is lower when the device is used versus when it is not. Additionally, we showed that in a subject with a pronounced static tilt during gait, use of vibrotactile feedback resulted in a more vertical posture. Greenspan et al. [16] report that the only significant fall biomechanics parameter significantly associated with hip fracture was fall direction – specifically, falling to the side. Mortality aside, if the majority of serious injuries (i.e. hip fractures) resulting from direction dependent falls, one could argue that a balance prosthesis that improves stability in the direction that could be a hip fracture countermeasure.

Goal of sensory substitution during locomotion

The second question is if M/L roll tilt is not the gait parameter to minimize in order to reduce the risk of falls and improve gait steadiness, what should the locomotion metric be? According to Winter [37], during gait, our body's COG is maintained along the medial border of the foot by appropriate placement of the swing foot twice per gait cycle. Bauby et al. showed both theoretically in a simple model and experimentally with healthy subjects that fore-aft gait dynamics are stabilized passively but significant active control must be used in order to stabilize lateral motion [8]. Double support phase allows a brief opportunity for re-stabilization since during this phase, the base of support is not firm; weight is being shifted from one foot to the other and stance support is derived from only the forepart of one foot and the heel of the other. The continuous display configuration could be perceived as superior because it delivers information during both single and double support phase thereby allowing postural corrections to be implemented throughout the complete gait cycle. The intent of the intermittent feedback display on the other hand, was to provide information about M/L tilt such that a M/L foot placement correction could be made on the subsequent step. Bent et al. [25] delivered galvanic vestibular stimulation (GVS) at either heel strike, mid-stance, or toe-off to assess when vestibular information is used during gait. Head, trunk and pelvis roll as well as M/L foot placement were considered. No difference in roll was observed for the upper body segments based on when the stimulation was delivered. However, foot placement was dependent on the time during the gait cycle that (GVS) was provided. Specifically, changes in foot placement were significantly larger when GVS was delivered at heel strike versus mid-stance and toe-off. It should be noted that shorter delays are associated with GVS than receiving, processing, and acting on vibrotactile tilt information provided to the trunk. Additionally, the response to GVS is involuntary while responding to vibrotactile feedback is voluntary. We observed that step width variability was significantly decreased with vibrotactile feedback during slow and self-paced walking. One idea for a device that incorporates both Bauby's and Bent's findings is to provide information on where to laterally place the subsequent foot.

The effect of tilt biofeedback on vertiginous patients

The third question that needs to be addressed during future device development is whether or not such a device will have any effect on patients that are suffering from dizziness and vertigo? To date, none of the published groups working on electrotactile, vibrotactile, or auditory biofeedback have presented data on this patient population. To gain insight into this question, we invited one idiopathic patient who complained of vertiginous symptoms (DHI score equaled 40) to try the vibrotactile balance prosthesis during slow-paced, self-paced, and perturbed walking. Partway through the experiment, the patient complained of feeling dizzy but continued with the gait trials. As the symptoms increased in intensity, the patient verbalized that although she was aware that the vibrotactile elements were vibrating, she was no longer paying any attention to the information because she was already overwhelmed dealing with her vertigo. The patient reported a neutral rating regarding the usefulness of the device. Based on the limited subjects we have evaluated to form our DHI versus percent change in roll sway correlation, this subject fell mid-way on the curve. This suggests that a reduction in roll sway with the device is possible. The usefulness of a balance aid in mitigating vertiginous symptoms is unknown. We hypothesize that patients who use verticality as a cue for reorientation may derive benefit from a wearable device that confirms that they are not actually tilting/spinning. The sheer volume of patients with dizziness problems largely drives the impressive statistics involving the large number of Americans that seek medical attention for balance or dizziness on an annual basis. Therefore, it will be important to understand whether or not this patient population can derive any benefit from biofeedback.

Tilt biofeedback induced sway

The fourth question is whether or not such a device actually induces sway? Hegeman et al. provided a review of vibration-induced postural sway during quiet stance [18]. However, numerous studies involving vestibular-deficient individuals using vibrotactile feedback during quiet and perturbed stance resulted in significant reductions in postural sway [19, 20, 23, 24, 38]. Therefore, despite any local stimulation and response to tactor elements on the trunk, overall body sway was decreased. Perhaps the more relevant issue to consider is whether patients will overreact to biofeedback information (electrotactile, vibrotactile, and auditory) during locomotor tasks and thereby introduce a new potentially unstable situation. We observed such a overreaction during our training sessions – especially among subjects that were intent of perfecting the task; in this case, preventing the device from vibrating. These subjects adopted a rigid and awkward gait to that end. Extra time and coaching had to be provided to these subjects to ensure that they were using the vibrotactile feedback to augment their natural gait. This is one of the primary reasons why we used the lower row of tactors to indicate a slight tilt, but reserved our stronger stimulus (all three rows activated simultaneously) to indicate a more severe tilt. Subjects were told that they could expect and should feel the lower level activating twice per gait cycle.

Cognitive workload – simultaneously attending to balance task and tilt biofeedback information

The final question posed here is whether or not the cognitive workload associated with attending to both gait and prosthesis tasks is doable? In this case, the primary task would be directed gait and the secondary task would be to use the vibrotactile balance prosthesis to make real-time postural corrections. As a first order approximation, human attention can be modeled as limited-capacity channels capable of transmitting small amounts of information per second (on the order of 10 bits/sec) [39]. Humans can use multiple sensory organs to gather and process information in parallel however, typical motor outputs occur in serial; that is, humans are good at taking in several sources of information provided a response to each one is not required [39]. In general, it is hypothesized that people will perform poorer on tasks and make more errors as they are confronted with too much mental workload [40]. Mental workload can be assessed using subjective ratings, secondary tasks, and physiological measures [39]. "The secondary-task paradigm imposes an additional task on the operator in addition to the main task and measures performance on the extra task. Decrements in performance of the secondary task are thought to indicate increased mental workload in the primary task [39]." Several studies have been conducted to evaluate the effect of dual-tasks on postural stability [41]. In summary, age has a significant affect on one's ability to perform a dual-task where the primary task is quiet stance and the secondary task is either a visual, auditory, or verbal response [42-47]. Verbal responses as secondary tasks have been shown to affect postural sway due to articulation [48]. Limited dual-task studies involving locomotion have been conducted. Weerdesteyn et al. (2003) demonstrated that divided attention affects both young and healthy individuals' ability to avoid obstacles while walking [49]. O'Shea et al. (2002) showed that performance of a concurrent motor or cognitive task compromised gait in individuals with Parkinson Disease, although the type of secondary task was not a major factor in the amount of interference with the primary task [50]. Lajoie et al. (1996) found that elderly individuals adopted a slower pace and shorter stride length than young individuals while performing an auditory reaction time task while walking [43]. Examples of cognitive secondary tasks include counting backward from 100 by threes, naming the days of the week in reverse order, arithmetic, recitation tasks, recognition tasks, Stroop Task, etc.

One shortfall of this study, discovered during post-processing, was that only proportional tilt feedback was used due to a programming error that occurred when roll versus phi tilt feedback was displayed. Subsequent gait studies should explore the efficacy of using proportional plus derivative feedback and predictor-based feedback.

In summary, the results indicate that the balance reduced M/L sway during slow, self-paced, foam and narrow stance walking tasks. Step width variability was significantly reduced with vibrotactile feedback. The most severely sensory-compromised subject presented in a more detailed study, showed the greatest amount of improvement among subjects with significant decreases in average roll tilt, RMS roll, percentage time spent outside the tactor dead zone, and step width variability. This pilot study suggests that vestibular-deficient patients can reduce their M/L tilt and decrease their step width variability during challenging locomotor tasks by using vibrotactile tilt feedback. Moderate to severely deficient subjects should be used instead of well-compensated patients in future locomotion experiments in order to gain a better understanding of which gait parameters sensory substitution devices should focus on augmenting. Additionally, such devices should be tested on patients experiencing vertiginous episodes to determine whether or not they can use verticality as a cue to reorient themselves and confirm that they are not actually tilting/spinning.

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