

ALMA MATER STUDIORUM
UNIVERSITÀ DI BOLOGNA

DEIS – Department of Electronics, Computer Science, and Systems

Biofeedback Systems for Human Postural Control

A method for understanding sensory integration and improving motor training

Marco Dozza

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Supervisor

Prof. Angelo Cappello

Co-supervisors

Prof. Lorenzo Chiari

Prof. Fay B. Horak

Reviewer

Prof. Carlo Marchesi

(Relatore: Marco Dozza)

(Coordinatore: Prof. Angelo Cappello)

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Thesis Abstract

Reliable sensory information and correct integration of sensory information are necessary for the control of posture. When sensory information is inadequate, such as in vestibular loss subjects, balance and postural control are impaired. Further, gradual loss of sensory information is also a consequence of the natural process of aging and is one of the major causes leading to falls in the elderly population.

Sensory information can be augmented by using biofeedback (BF) devices. BF devices are artificial systems able to provide additional movement information to their users. Although BF devices for postural control have been experimented since the 70s, the extent to which BF devices can substitute for missing sensory information for the control of posture is still unknown. Further, although BF devices have been suggested to be helpful for rehabilitation, no study to date has provided conclusive evidence that practice with BF is better at improving retention of motor performance than practice without BF.

The purpose of this dissertation are 1) to design, set-up and validate new-generation, portable, low-cost BF devices, 2) to determine how the design features of such BF devices influence postural control during static and dynamic motor tasks in vestibular loss and healthy subjects, 3) to elucidate how movement information from BF devices is integrated with sensory information for the control of posture, and 4) to understand the relationship between the effect of BF and spontaneous motor learning on postural control.

We implemented several types of BF devices that coded postural sway from bi-axial accelerometers, a combination of accelerometers and gyros, and a force plate into a stereo sound, vibrotactile stimulation to the trunk and/or a visual representation using different coding algorithms. By comparing such devices, we demonstrated how crucial the design of a BF device is, since it influences both the motor performance and the postural strategy of its user. In addition, we showed how vestibular loss and healthy subjects can use our audio-BF device to reduce sway by augmenting postural control without increasing muscular stiffness. Further, we found that audio-BF increases closed-loop control of posture and does not influence the open loop control of posture.

By testing bilateral vestibular loss and healthy subjects in several conditions of limited or inadequate sensory information, we showed how audio-BF efficacy is related to the individual dependency of each subject on vestibular, somatosensory, and visual information. In addition, we showed that audio-BF improves posture also in dynamic tasks such as standing on a randomly rotating surface and that the extent of these postural improvements is proportional

to the amount of movement information coded into the sound. Also, we showed that spontaneous motor learning and audio-BF affect different ranges of frequency of postural control during standing on a randomly rotating surface.

Furthermore, unilateral vestibular loss subjects were tested during tandem gait using a cross-over design to understand whether tactile-BF of trunk tilt could improve postural performances during a complex, dynamic motor task such as gait. Results from this experiment showed that tactile-BF of trunk tilt acts similarly to natural sensory feedback in immediately improving dynamic motor performance and not as a method to recalibrate motor performance to improve dynamic balance function after short-term use.

Our results have many implications for the design of BF devices, for the understanding of motor control and sensory integration, and for the design of the protocols to be used with BF devices. More specifically, our results suggest that BF 1) needs a customized design for each subject and each task to optimally improve postural motor performance without facilitating undesirable postural strategies, 2) improves motor control in static and dynamic tasks by augmenting motor information and substituting for missing sensory information, 3) must be equipped with training protocols able to favor motor learning in order to become a helpful tool for balance and motor rehabilitation and training.

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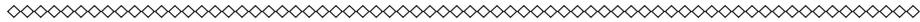
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Chapter 1

Sensory Integration and Augmentation for the Control of Posture



Abstract

This first chapter is an introduction to sensorimotor integration and augmentation. Many evidences of sensorimotor integration are presented from studies often focused on sensorimotor illusions. These evidences support the hypothesis that the central nervous system is continuously and unconsciously able to 1) integrate and re-weight sensorimotor information, 2) create an internal representation of the body in space based on sensorimotor information, and 3) re-calibrate sensorimotor information.

Some examples of movement disorders related to impaired sensorimotor integration are also reviewed. Specifically, vestibular loss is presented as one of the pathology which could more likely benefit from sensory information augmentation both for motor improvements and for rehabilitation . In addition, studies aimed at demonstrating sensorimotor integration impairments in subjects with peripheral neuropathy, Parkinson disease, and other movement disorders are briefly reported.

Finally, sensory augmentation using biofeedback is reviewed. Many applications of biofeedback are reported with an emphasis on biofeedback systems for postural control. The main issues related with the design of biofeedback systems for postural control and with the design of experimental protocols aimed at valuating biofeedback system effectiveness are also discussed. In addition, a brief review on the application of virtual reality for postural control assessment and improvement is presented and proposed as a highly desirable feature for next-generation biofeedback systems for postural control.

Evidence of Sensory Integration for the Control of Posture



Control of posture is a crucial task with two main goals: equilibrium and orientation [1]. Although we often take the postural system for granted because it operates primarily at a non-cognitive level, it actually depends on a complex and active interaction among the sensory, muscular, and nervous systems. To appreciate the importance and the level of accuracy that this interaction can reach in humans to maintain equilibrium and spatial orientation, imagine a circus performer walking on a 10-meters-high steel wire while juggling clubs. Now, imagine the same juggler being suddenly in the dark, or losing the sensation of the wire under his/her feet and, if that is not enough, losing also the perception of gravity. This example highlights the importance of sensory information and its integration for the achievement of effective postural control

Photos: Michalska and Venturska street, Bratislava, June 2006.

Concurrence and Interference of Sensory Information

The control of equilibrium and orientation depends on *concurrent feedback of motion information* from the vestibular, somatosensory, and visual senses. The importance of sensory feedback is evident, for example, from the sway increase occurring when sensory information becomes unavailable during simple quiet stance. In fact, during quiet stance, the largest increase in postural sway occurs when somatosensory information is unavailable [2]. The next largest increase, when vestibular information is unavailable, and the smallest, when vision is unavailable [1;3;4]. Limitation of vestibular, visual, and somatosensory information, which is part of the ageing process, is a major factor leading to falls in elderly people [5;6] and a major health problem [7].

Information from vestibular, somatosensory, and visual senses is *redundant*. Redundancy of sensory information is crucial, for example, for walking in the dark (when vision is not available), or on a compliant surface (when somatosensory information is inaccurate). In fact, in these two situations, the central-nervous-system must rely only on two senses (vestibular and somatosensory, and vestibular and visual, respectively). Thus, the ability to walk in the dark or on a compliant surface does not only depend on sensory information redundancy but also on the central-nervous-system ability to 1) evaluate and compare sensory information, 2) distinguish between reliable and unreliable information and 3) combine the sensory information into a integrated representation of the environment. This central-nervous-system process is known as **sensory integration**. The prevailing theory states that the various sources of sensory information are integrated to form an internal model of the body that the central-nervous-system uses to plan and execute motor behaviors [1]. This internal model must be adaptive, to accommodate changes associated with growth and development, aging, and injury [1]. Thus, this internal model needs to be continuously recalibrated so that it can weigh differently the motion information coming from the different senses [8;9].

One evidence of sensory interaction and remapping of an internal model of spatial orientaton, based on vestibular sensory information, can be foreseen in the *oculogyral and audiogyral illusions*. Oculogyral and audiogyral illusions are experienced by a subject rotating with a constant angular velocity and are due to ambiguity of sensory information. When a subject is seated on a chair rotating with a constant angular velocity both visual and auditory spatial localization change [10;11]. In fact, if the subject is in the dark and a head-fixed visual target is lit, the subject perceives the target as moving with his/her body, changing the apparent position in space but leading the body as well in the direction of the acceleration (oculogyral illusion). In a similar way, a head-fixed auditory target will be heard by the same subject as moving in opposite direction with reference to the angular acceleration (audiogyral illusion). Whiteside et al. (1965), [12], explained the oculogyral illusion as due to an error in body visual localization due to the fixation of the target overriding the vestibular nystagmus being misinterpreted as an eye deviation. Other illusions, known as oculographic, somatogravic, and audiogravic, occur when a subject is exposed to unusual patterns of gravito-inertial acceleration [13-15]. These illusions are due to a misinterpretation of gravitational vertical when a subject is exposed to centrifugal acceleration in a rotation chamber. Howard and Templeton, [16], explained these illusions as due to the inability of the otholith to distinguish between gravitational and inertial acceleration. In 2001, another explanation, suggesting a more complex mechanism of central remapping of sensory localization, was also proposed by DiZio et al. [17].

As the otolith organ in the inner ear can influence the perception of the orientation of the head and the gravito-inertial acceleration, muscle spindles have been found to influence the perception of position of body segments [18]. In fact, by vibrating postural muscles, it is possible to activate the muscle spindle so that *illusions of motion* are elicited. For example,

by vibrating the Achilles tendons of a subject, it is possible to elicit a pitch rotation at the ankles. In addition, if a fixed, visual target is showed to the subject during the vibration, this subject will see the target as if it were moving in the direction of apparent self-motion [19]. Another example of motion illusion comes from an experiment of Karnath et al. (1994) who found that when neck muscles are vibrated, perception of head rotation is elicited [20]. Similar illusions can be elicited also for other body segments [21]. In addition, if a visual or auditory target is presented to subjects during muscle vibration, this target will be perceived to move according to the motion illusion experienced by the subject.

The influence of haptic information on posture is extremely important and can also induce perceptions of self-motion. *Light-touch*, haptic information from the index finger of a hand touching a firm surface without any mechanical support (force applied is less than 100g) stabilizes posture by reducing sway up to 50% in blindfolded subjects [22]. Furthermore, light-touch improves postural stability in all subjects tested up to now such as elderly, cerebellar, neuropathic, and labyrinthine defective subjects [23]. When the touched surface is oscillated, subjects sway entrained to this oscillation, then trusting the haptic information more than the other motion sensory information [24] by responding to an illusion of motion. Light-touch information cancels out the destabilizing effect of tonic vibration reflexes in leg muscles [25] as well as the illusion of self-displacement and airplane displacement during parabolic flights [26].

When sensory information is ambiguous, as in the example of illusions described above, *cognitive knowledge* and assumptions can influence the subject's behavior and the extent to which subjects perceive the illusions. For example, if subjects are aware that the surface used for light-touch is oscillated, they may show a smaller correlated oscillation in their sway than if they did not know about this surface motion. Cognitive knowledge is also important, for example, to neglect the otolith information elicited by the centrifugal force when a sharp sudden turn occurs [27]. Internal models created by cognitive knowledge have also been suggested to be used by the central-nervous-system to resolve ambiguity in sensory information from the otoliths and, specifically, for distinguishing between inertial and gravitational acceleration based also on information from the semicircular canals [28].

Tuning and Calibration of Sensory Information for Internal Model Representation

To achieve sensory integration, the central-nervous-system needs to compare continuously the sensory information from different senses, so that matches or mismatches among sensory information can be accurately detected and internal **model of localization** conveniently tuned up and calibrated. Evidence that interaction with hands may help achieve *spatial calibration* of the body comes from another illusion described by Lackner et al. (1988) [21]. In Lackner's experiment, a subject was holding his/her nose when the biceps brachii muscle of the arm was vibrated. The illusion of arm extension due to the vibration, led the

subject to feel his/her nose was elongating. This 'Pinochio illusion' suggests that body spatial calibration may start from tactile interaction with the environment.

When subjects are exposed to an artificial gravity environment, head, arm, and leg movement control, as well as locomotion, can be promptly adapted if the same motion task is attempted repeatedly [29]. During the adaptation process, the subjects create a new model of the environment where they integrate the new *Coriolis* acceleration due to the artificial gravity. Once the adaptation is completed, Coriolis acceleration is not cognitively perceived anymore. However, as soon as the subjects are back to natural gravity environment, the Coriolis force, associated with the artificial gravity environment and integrated in the subjects' internal model of the environment, is consciously perceived again as influencing the subjects' movement until a new process of adaptation to natural gravity has been completed. Until the adaptation process is completed, subjects show mirror-image error in movement control compared to the ones experience in the first process of adaptation to artificial gravity. Thus, these results suggest that the body is dynamically calibrated by its force environment, and that movements within it feel virtually effortless.

The adaptive tuning of the body internal model of gravity must also take into account the self-generated Coriolis forces experienced during common daily movements [30]. In fact, whenever a natural turn-and-reach movement is performed, the simultaneous occurrence of trunk movement and the arm forward velocity generate very high Coriolis acceleration on the reaching arm. The preservation of reaching accuracy suggests that the central nervous system is able to predict the Coriolis acceleration and compensate it with anticipatory forces generated during the task.

Coriolis acceleration can be generated also by making pitch head movement during passive rotation [31]. In this case, an illusory tumbling sensation is elicited by the Coriolis acceleration and the subjects feel nauseous. Surprisingly, during orbital flights subjects performing the same head rotation do not experience nausea. Further experiments on this interesting result suggest that the lack of motion sickness during orbital flights may be due to the lack of internally represented body displacement. However, motion sickness remains not totally understood. Many theories have been proposed to explain motion sickness such as sensory information conflict [32], however the only firm result is that subjects without functioning labyrinths have not been made motion sick although several protocols have been tried [33].

Movement Disorders and Their Relation to Sensorimotor Integration



Sensorimotor integration is the ability of using sensory information properly for assisting motor program execution. Whenever sensorimotor integration is impaired, movement disorders are experienced by subjects with a variety of different symptoms such as vertigo, dizziness, and bradykinesia. The relationship between sensorimotor integration and pathologies involving motor impairment is intuitive in case of vestibular or somatosensory loss. However, a connection between sensorimotor integration and movement disorders has also been suggested in other motor pathologies such as Parkinson's disease.

Photo: A subject with Parkinson's disease during an experiment at the Biomechanics Laboratory of the University of Bologna in Cesena, Italy. January 2005.

Sensory Loss and Aging

One-third to one-half of the population over age 65 reports some difficulty with balance or ambulation [34-37]. The most common cause of impaired postural stability is the loss of accurate and/or adequate sensory information from vestibular, somatosensory, and visual systems [38-40]. In Europe, approximately one-third of community-dwelling adults over 65 years and fifty percent of those over 80 years fall at least once a year [41]. Twenty to thirty percent of those who fall suffer injuries that reduce mobility and independence and increase the risk of premature death [42-43].

Acute peripheral, **vestibular loss** can be caused by damage of the vestibular organ or of the vestibular nerve and results in sensations that reflect abnormal information about head motion [44]. The vestibular system is a bilateral organ that consists of 3 semicircular canals and 2 otoliths on each side [45]. This complicated organ is able to provide the central nervous system with information about the linear accelerations and angular velocities of the head in space. The central nervous system processes this sensory information and integrates it with the other sensory information to determine the gravitational vertical direction [1]. The most common cause of damage of the vestibular organs (and, consequently, of bilateral vestibular

loss) is a toxic reaction to antibiotics such as gentamicin, which selectively damages the vestibular hair cells. Exposure to gentamicin causes bilateral vestibular loss in 3-4% of cases [46]. Unilateral vestibular loss is common when the vestibular nerve is damaged. Specifically, neuritis is the most common cause of unilateral vestibular loss [47]. Unilateral damage of the vestibular nerve causes an asymmetry in the vestibular nerves firing rates. The central nervous system interprets this asymmetry as a head rotation toward the contralesional ear. This results in spontaneous nystagmus, with slow components in the direction of the lesioned ear and fast in the direction of the contralesional ear.

Nystagmus is related to the vestibular organ via the vestibulo-ocular reflex [48]. A clinical measure of vestibular function is based on the vestibular-ocular reflex and observed nystagmus is an indicator of vestibular function [49]. For example, the "head thrust test" is based on the knowledge that when the vestibular-ocular reflex is functioning normally, the eyes move in the direction opposite to the head movement to stabilize gaze in space [50]. The vestibulo-ocular reflex gain and phase are used to quantify vestibular loss and are normally measured in the laboratory by recording eye movement in the dark when the subjects are rotated in the horizontal plane. However, the vestibular-ocular reflex gain and phase during horizontal body rotation are only indicators of the horizontal canal function and not of the whole vestibular system. Recently, a new diagnostic method, based on the vestibular-evoked myogenic potentials has been used to also measure saccular otolith function [51].

Loss of vestibular function can occur slowly as in the aging process or suddenly, as in the case of ototoxicity and neuritis described above. When vestibular loss occurs suddenly, balance disorders are immediately evident and subjects need to go through a rehabilitation period where they learn how to compensate for the vestibular loss before they can walk or comfortably perform daily life motor tasks again. During this period, subjects learn how to rely more on visual and somatosensory information to compensate the lack of vestibular loss [52]. Classical symptoms occurring after sudden vestibular loss include: vertigo and dizziness due to the abnormal perception of self-motion. These symptoms disappear spontaneously over time [53]. However, some abilities as riding a bike or playing tennis may not be ever achieved again. Even after being fully compensated, unilateral vestibular loss subjects may show abnormal postural alignment [54], asymmetric weight distribution [55], and inability to stand on one foot or walk with a narrow base of support [56]. In addition, whenever unilateral or bilateral loss vestibular loss subjects are exposed to condition of altered visual or somatosensory information, their ability to maintain balance is especially impaired [57].

Vestibular rehabilitation for subjects with unilateral deficits consists of exercises to enhance the gain of the vestibular-ocular reflex, static and dynamic exercise with augmented sensory information from a therapist, and activities to tolerate movement of the head and the body [58]. Such rehabilitation therapies have been proven to be effective in helping those with acute vestibular neuritis return to normal activities of daily living [59]. Vestibular

rehabilitation was found to be successful also in subjects with bilateral vestibular loss [60]. Vestibular rehabilitation may be useful also for elderly subjects who may not be aware of their vestibular loss because it occurred gradually. In fact, when vestibular loss occurs gradually, although there may be no dizziness, subjects may be unstable and fall when in an environment requiring vestibular information for balance, i.e. in the dark on an unstable surface.

With aging, *peripheral somatosensory nerve deficits* also become more common. Peripheral sensory nerve deficits lead to delay, distortion and loss of somatosensory information from the muscles, joints, and skin which can be assessed by measuring the vibratory sensation and ankle stretch reflex. Both vibratory sensation and the ankle stretch reflex are commonly impaired in the elderly population [61;62]. Peripheral neuropathy can be the consequence of several causes such as diabetes mellitus, alcoholism, nutritional deficiencies, infections, malignancies, and autoimmune diseases [63]. Environmental and pharmaceutical agents, as well as some hereditary factors can also lead to peripheral neuropathy. However, only in 72% of the adults manifesting the syndromes of peripheral neuropathy, a specific cause can be identified [64].

Some changes in the structure and function of peripheral nerves may be the result of the aging process itself [65-68]. Peripheral neuropathy has been related to impaired balance and falls by many studies [69-73]. Loss of sensory information from neuropathy can challenge sensory integration in the elderly, leading to falls, or simply limiting elderly subjects' activities that facilitate a premature functional decline.

Augmentation of Sensory Information for Postural Control



Photo: A subjects using audio- and visual-biofeedback at the Balance Disorders Laboratory of the Oregon Health & Science University, Portland (OR) USA, July 2005.

Sensory loss or abnormal, inadequate sensory information from the vestibular, somatosensory and visual senses can jeopardize the central-nervous-system ability to control postural stability. In this case, providing the central-nervous-system with substitutive, artificial, sensory information may help restore the ability to control posture. Artificial, sensory information can also be used to augment sensory information during rehabilitation sessions when brain plasticity and adaptation are crucial and depends on the extent and accuracy of the sensory information available. Augmentation of sensory information normally implies the use of an external device able to provide information about body motion through biofeedback, eventually presented in a virtual reality environment. However, proprioception has been proven to be augmented also by a simple sole or knee vibration ([74] and [75], respectively).

Overview on Biofeedback Experimentation

Biofeedback has been applied since the 50s [76] and can be defined as a process in which a person learns to reliably influence physiological responses of two kinds: either responses which are not ordinarily under voluntary control or responses which ordinarily are regulated but for which regulation has broken down [77]. Since the early 60s, many studies have reported the use of biofeedback in many areas such as instrumental conditioning of automatic nervous system responses, psychophysiology, behavior therapy and medicine, stress research and stress management strategies, biomedical engineering, electromyography, consciousness, electroencephalography, cybernetics, and sports [76].

The application of biofeedback to improve postural control began in the 70s with visual biofeedback of electromyogram, positional, or force parameters [78;79]. Studies with

electromyograms showed how subjects with sensorimotor deficits can volitionally control single muscle activation and become more aware of muscular contraction when muscle activation could be seen or heard [80;81]. Studies on positional and force parameters showed how subjects could improve control of posture by actively responding to visual cues indicating surface reactive forces provided by the biofeedback systems [82].

The neurological mechanisms underlying the effectiveness of biofeedback are still mainly unknown. However, some hypotheses have been suggested. Two hypotheses come from Basmajian (1982) [83] who believed that, either new pathways, or a new feedback loops recruiting cerebral and spinal pathways already existing are used as a consequence of exposure to biofeedback. In addition, Wolf (1983) [79], suggested that auditory and visual stimuli from biofeedback can activate synapses that were not used before. Although not totally understood, the effects of biofeedback seem to favor brain plasticity and, as a consequence, shows a noticeable potential for motor rehabilitation applications.

Design of Biofeedback Systems

Three main parts essential for the design of a biofeedback system for postural control are: 1) a sensor or an instrument able to measure some aspect of human motion, 2) a restitution device, able to convey the biofeedback information to the subject (e.g. via the auditory, visual, or tactile sense), and 3) some circuit or a computer able to implement a conversion algorithm which transposes the information sensed by the sensor into a convenient activation of the restitution device (Figure 1).

Several combinations of sensors and restitution devices, concerted by simple or complex algorithms, have been implemented and tested to determine whether they could improve motor control. Biofeedback systems have been proven to be effective in many areas, despite the intrinsic difference of the wide variety of biofeedback systems designs and of their application fields. For example, biofeedback systems were found to be effective in improving sportive performance [84] by decreasing stress and anxiety during training in many sports such as gymnastic [85], swimming [86], basketball [87], judo [88], archery [89], shooting [90], and golf [91]. The use of biofeedback for improving control of posture in subjects with motor disorders has been more oriented to provide augmented movement information of body

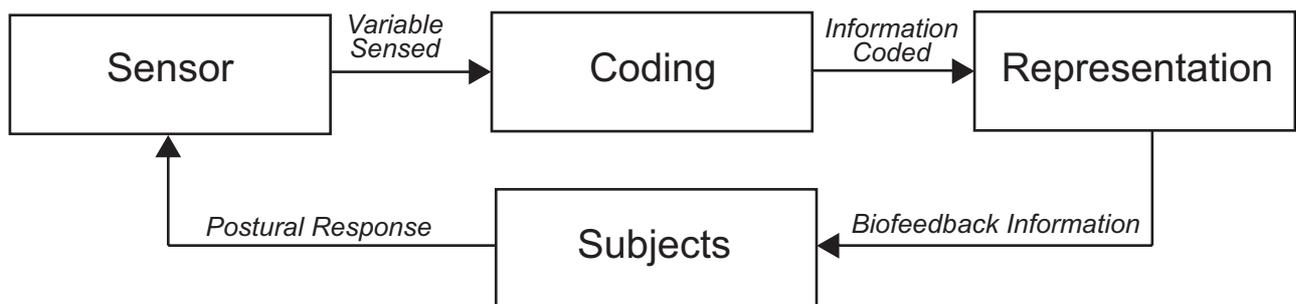


Figure 1 – *Diagram for the design of a biofeedback system.*

Table 1 – *Published studies on the experimentation of biofeedback systems for postural control*

Article	Feedback Variable	Restitution	Subjects	Application
Halpern et al., 1970	Head Position	Tactile/Mechanical	Children with Cerebral Palsy	Head Control
Hlavackha., 1973	Center of Pressure	Visual	Healthy	Standing
Harris et al., 1974	Head/Limb Position	Visual/Auditory	Athenoid Children	Head/Limb Control
Woolridge and Russel 1976	Head Position	Visual/Auditory	Children with Cerebral Palsy	Head Control
Wannstedt and Herman, 1978	Center of Pressure	Auditory	Hemiplegic	Symmetry of Standing
Walmsley et al. 1981	Head Position	Auditory	Mentally Retarded Children	Head Control
Leiper et al., 1981	Head Tilt	Auditory	Children with Cerebral Palsy	Head Control
Wolf and Binder-McLeod, 1983	Muscle Activity	Auditory	Hemiplegic	Symmetry in Standing and Locomotion
Catenese and Sanford, 1984	Head Position	Visual/Auditory	Children with Spstic Quadriplegia Childre with Diplegia or	Head Control
Bertoti and Gross, 1988	Head Position	On/Off Movie Play	Quadriplegia	Head Control
Schumway-Cook et al., 1988	Center of Pressure	Visual	Hemiparetic	Symmetry of Standing
Winstein et al., 1989	Weight Distribution	Visual	Hemiplegic	Symmetry in Standing and Locomotion
Domaraki et al., 1990	Head Position	Auditory	Children with disabilities	Head Control
Clarke et al., 1990	Center of Pressure	Visual	Healthy	Weight Shifting
Hamann and Krausen, 1990	Center of Pressure	Visual	Vestibular	Standing
Jobst, 1990	Center of Pressure	Visual	Brainstem and Cerebellar Lesions	Quiet Standing and Voluntary Sway
Hamman et al., 1992	Center of Gravity	Visual	Healthy	Dynamic Tasks
Barona et al., 1994	Center of Pressure	Visual	Healthy	Voluntary Sway
Edgardt, 1994	Weight Distribution	Auditory	After Stroke	Symmetry of Standing
Pertersen et al., 1996	Center of Pressure	Auditory	Stroke	Stance Perturbed by Muscle Vibration
Easton et al., 1997	Center of Pressure	Auditory	Blind	Standing
Wong et al., 1997	Weight Distribution	Visual/Auditory	Hemiplegia/Brain Injuries	Standing
Wu, 1997	Center of Gravity	Visual	Neuropathic	Perturbed Stance
Aruin et al., 2000	Knee-to-Knee Distance	Auditory	Stroke/Pelvic Instability	Gait
Rougier et al., 2002	Center of Pressure	Visual	Healthy	Standing
Tyler et al. 2003	Head Tilt	Elcetro-Tactile	Vestibular	Standing
Dault et al., 2003	Center of Pressure	Visual	Elderly and Young Healthy	Standing
Sihvonen et al., 2004	Center of Pressure	Visual	Elderly Women	Standing and Weight Shifting
Kentala et al. 2003	Trunk Tilt	Tactile	Vestibular	Standing
Dozza et al., 2005	Trunk Acceleration	Auditory	Vestibular	Standing
Heageman et al. 2005	Trunk Angular Velocity	Auditory	Vestibular	Standing and Dynamic Tasks
Chiari et al., 2005	Trunk Acceleration	Auditory	Healthy	Standing
Dozza et al., 2006	Trunk Acceleration	Visual	Healthy	Standing

segments than to reduce psychological stress. Table 1 summarizes some of the studies on biofeedback aimed at improving postural control. From Table 1, it is possible to appreciate the wide variety of biofeedback designs implemented up to now and the multitude of pathologies they have been tested on.

As shown in Table 1, visual-biofeedback of center of pressure displacement has been the most popular biofeedback system design for improving postural stability. Visual-biofeedback has been extensively used for balance rehabilitation of subjects after stroke [92] in order to reduce postural asymmetry. In addition, visual biofeedback from force plate measurements is the only biofeedback system commercially available and diffused. In fact, systems made by Neurocom (<http://www.onbalance.com/>), such as Balance Master, which are currently used for balance training and rehabilitation, are equipped with visual-biofeedback. Recently, partly based on work in this thesis, the interest in biofeedback design is moving from the visual-biofeedback of force plate measurements to audio- and tactile-biofeedback of inertial sensors measurements [93;94]. This new trend in the design is driven by the intent of producing new cost-effective and portable systems for balance training and rehabilitation. In fact, tactile and auditory feedback do not rely on some expensive and cumbersome monitor, and do not require power supply cabling; further, inertial sensors are one thousand times less expensive than force plates and much smaller, portable, and sturdy.

Biofeedback is thought to have a relevant potential for rehabilitation applications [92;95;96]. In fact, biofeedback can help subjects re-educate their motor control system during dynamic tasks with functionally goal-oriented exercise which help the subject to explore the environment and solve specific motor problem [95]. However, the design of such biofeedback systems and of the most effective clinical protocol is challenging.

One of the first challenges to be faced is the determination of the variable to be fed back. This variable should depend upon the motor control mechanism, training task, and therapeutic goal [92]. For example, since there are studies suggesting that hand kinematics in reaching movement is either controlled by equilibrium point shifting [104] or by creating a virtual trajectory of end-point [98], instead of scaling muscle activity [99], a biofeedback system for this task should use, as feedback variable, some kinematics information instead of electromyographic information. Successful reaching also requires control of alignment of finger-thumb opposition [100;101], as a consequence, a biofeedback system designed to help reaching should also provide the subject with this information. The presence of more than one relevant parameter to be controlled in the tasks presents another challenge for the biofeedback system design. In fact, a multi-sensing, task-oriented biofeedback system (Figure 2) should be able to feed back all information relevant for the task without distracting or overwhelming the subject. Determining how to combine different information into one variable, that can be fed back without too highly cognitively demand for the subject, is a necessary feature for the design of an optimal biofeedback system. A possible help in this

matter is the use of biomedical models to calculate and feed back several variables in real-time [102]. Other challenges for the design of a biofeedback system regard 1) the design of an algorithm able to correctly and efficiently represent the feedback variable in a way easy to learn and understand for the subjects [103], 2) the choice of a convenient representation for the feedback variable that does not interfere with the task performance [103].

Experimentation of Biofeedback Systems: Protocol Design

The biofeedback systems and the protocols described in the studies reported in Table 1 are very different from each other in their designs, which were customized to encounter the needs of different pathologies. Nevertheless, all of these studies report some beneficial effect of the biofeedback intervention. However, in some studies [104;105] not every subjects improved. Nevertheless, it was always possible to find a subgroup of subjects who significantly improved their postural performance by using biofeedback. This suggests that, depending on the different pathologies and personal characteristics, some subjects may be more suitable than others for benefiting from biofeedback.

Although it was always possible to show some performance improvement in at least a restricted set of the subjects exposed to biofeedback, previous studies have not quantified or reported positive results about learning, retention, and transfer effects as a consequence of biofeedback training. These effects are relevant because experimentation of biofeedback for postural control has, as a major future repercussion, the use of such devices for rehabilitation. In a rehabilitation process, it is more important for a subject to learn a task than for the same subject to be able, in some controlled situation with some temporary artificial biofeedback help, to reach an outstanding performance. At the same time, the goal of a rehabilitation process is to restore the subject's postural ability, which requires the retention of the postural improvement achieved during the rehabilitation session. Finally, a rehabilitation exercise is the more useful the more the improvements, acquired by a subject practicing that specific task, transfer to other motor tasks

Despite some positive results in terms of retention and transfer effect due to biofeedback intervention have been reported (e.g. [104;106]), many studies do not demonstrate that biofeedback therapy leads to significant motor function recovery [92;107-109]. This lack of conclusive results can be due to some intrinsic challenge in the experimentation of biofeedback. One of these challenges can be foreseen in the presence of subjects that, for personal characteristics not well understood yet, do not show any improvement (or even get worst) when practicing with biofeedback. The presence of such subjects [104;105] affects the experimental results, hiding the potential beneficial effect of biofeedback. Understanding the reasons why some subjects do not benefit from biofeedback may help determining, a priori, which subjects are suitable for benefiting from biofeedback.

Another challenge for biofeedback studies is the difficulty of determining the extent to

which improved performance with biofeedback intervention is due to biofeedback efficacy or to the natural, spontaneous learning process induced by repetitive practice of the task [108]. A possible solution to this issue is the implementation of an experimental design in which trials are randomized and two groups of subjects are included, so that one of the groups is exposed to biofeedback and the other to the simple repetition of the task. The difference between the two groups will then be a more accurate indicator of biofeedback success.

Finally, another challenge, that concerns the retention effect, is that, practicing static tasks, such as quiet standing, seems to have less potential to transfer performance improvements to other motor tasks than practicing dynamic tasks [108;110]. However, it has also been reported how practicing dynamic tasks does not improve static tasks such as quiet stance [111]. This later finding suggests that two different biofeedback therapies, one aimed to quiet stance improvements and one aimed to dynamic tasks improvements, may be necessary in the rehabilitation process.

In conclusion, two double-blinded, experimental designs with randomized trials [79], one during dynamic tasks and one during static tasks, seem to be the best protocol to determine the effectiveness and potential impact in the rehabilitation field of biofeedback systems. However, such protocols require a larger number of subjects, longer time, and more resources than any simple protocol aimed to describe the immediate, overall effect of biofeedback systems on postural control.

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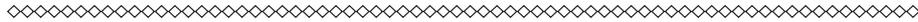
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Chapter 2

Audio-Biofeedback for Balance Improvements: an Accelerometry-Based System



Most of the content of this chapter has been published in: L. Chiari, M. Dozza, A. Cappello, F. B. Horak, V. Macellari, and D. Giansanti, "Audio-biofeedback for balance improvement: an accelerometry-based system," IEEE Trans Biomed Eng, vol. 52, no. 12, pp. 2108-2111, Dec.2005.

Abstract

This paper introduces a prototype audio-biofeedback system for balance improvement through the sonification of trunk kinematic information. In tests of this system, normal healthy subjects performed several trials in which they stood quietly in three sensory conditions while wearing an accelerometric sensory unit and headphones. The audio-biofeedback system converted in real-time the two-dimensional horizontal trunk accelerations into a stereo sound by modulating its frequency, level, and left/right balance.

Preliminary results showed that subjects improved balance using this audio-biofeedback system and that this improvement was greater the more that balance was challenged by absent or unreliable sensory cues. In addition, high correlations were found between the center of pressure displacement and trunk acceleration, suggesting accelerometers may be useful for quantifying standing balance.

Introduction

A complex interplay between feedback and feedforward control results in the ability of the human body to stabilize and to maintain balance in an upright stance and during movement [1]. Visual, vestibular, and somatosensory receptors provide the central nervous system (CNS) with sensory information about body orientation and motion in space in order to maintain balance. Balance deficits are frequently associated with diseases, disorders, and conditions in which there is either incomplete environmental information supplied to the CNS by the senses, such as in vestibular disorders, or a deterioration of the circuitry of the CNS, such as in stroke or Parkinson's disease. One approach to improving balance, which has been widely used in physical therapy and rehabilitation, involves feeding back to the CNS supplementary environmental information about body motion. This supplemental information may be coming from artificial sensors, a therapist, or laboratory equipments [2], [3].

In the past few years, increases in the speed of microprocessors, advances in miniature devices, and a growing interest in noninvasive patient monitoring and management have stimulated the development of real-time portable biomedical systems that are compact and have low cost [4], [5]. One promising application of such systems is biofeedback, which can be used to enhance human perception of automatic biological processes, such as movement and balance [6], [7]. Recently, Giansanti et al. [8] developed a portable sensor consisting of three accelerometers and three gyroscopes that estimate three-dimensional (3-D) kinematic information of a body segment. This sensor became part of the audio-biofeedback (ABF) prototype device presented in this paper. In this paper, we will 1) describe the architecture and the functioning principle of this ABF system, and 2) present the results of a preliminary study that tested the hypothesis that ABF benefits normal, healthy subjects most when sensory information is partly compromised.

Materials and Methods

In this study, we used our customized ABF device, and a force plate (AMTI OR6-6, Watertown, MA) to estimate body sway by means of center of pressure (COP) data. The ABF device has three major component: 1) a sensory unit, 2) a processing unit, and 3) an audio-output unit. The force plate was used for cross-validation and is not a component of the ABF system.

Sensory Unit

The prototype uses a portable sensory unit described elsewhere [8], weighing about 100 grams. Briefly, the sensory unit incorporates a cell with two linear uni-axial accelerometers (3031-Euro Sensor, UK), packaged into a 7.5x7.5x3.5 mm³ module. The accelerometers have the following specifications: range = $\pm 2g$, sensitivity = 3 mV/g, linearity = 0.08 %FS, frequency response = 0-350 Hz, and peak-to-peak noise = 0.15 mg over the entire bandwidth. The accelerometers are aligned with an orthogonal reference frame rigid with the cell, and they measure the linear accelerations of the trunk in the anterior-posterior (AP) and medial-lateral (ML) directions.

The performance of the accelerometers in the sensory unit was previously evaluated during several postural tasks related to activities of daily living. Results from these studies were compared with simultaneous recordings from an optoelectronic stereo-photogrammetric system. The sensory unit performed at a maximum error of about 10^{-4} g in horizontal accelerations.

Processing Unit

The acceleration outputs of the sensory unit are analog-to-digital converted by a DAQ board (NI-6024E, National Instruments, Austin, TX), and processed on a Toshiba laptop computer (CPU: Intel Celeron 2.0 GHz) running Matlab Data Acquisition Toolbox (Mathworks, Natick, MA). Digital processing in the laptop computes the proper frequency, level, and left/right (L/R) balance of the audio output signal. The laptop also digitizes and stores additional signals for future analysis. Signals such as the complete 3-D linear and angular trunk kinematics are recorded by the portable sensor. Ground reaction forces and moments are recorded by the force plate on which subject stands.

Audio Output Unit

After digital processing in the laptop, the DAQ board converts the audio output signal into a binaural, synthetic feedback signal flowing through a common audio amplifier (Fostex PH-5, Japan) into headphones (Philips SBC HP-140, The Netherlands) that the user wears.

Algorithm for ABF Sound Generation

The algorithm for ABF sound generation is designed to convey spatial information about the horizontal movements of the user's trunk to the headphones by means of sinusoidal tones. The audio signal maps AP and ML accelerations into stereo sound modulated in frequency, level, and L/R balance. The ABF system uses independently modulated right and left output channels for sound representation, with a 20-Hz refresh rate.

To avoid an overload of sensory information presented to the user, the ABF evaluates a region around a user's natural stance posture where subtle, spontaneous sway with small accelerations always occurs, even in normal, healthy individuals who are able to use all of their senses available. We refer to this region as the reference region (RR). The RR is considered to be the area in which an individual sways while standing but still does not need any extra information to stabilize upright posture. When swaying outside this region, an individual receives sensory feedback to correct sway to within the RR in order to stabilize upright posture. Our goal is to help an individual correct sway to within the RR, therefore, stabilizing upright posture by using ABF.

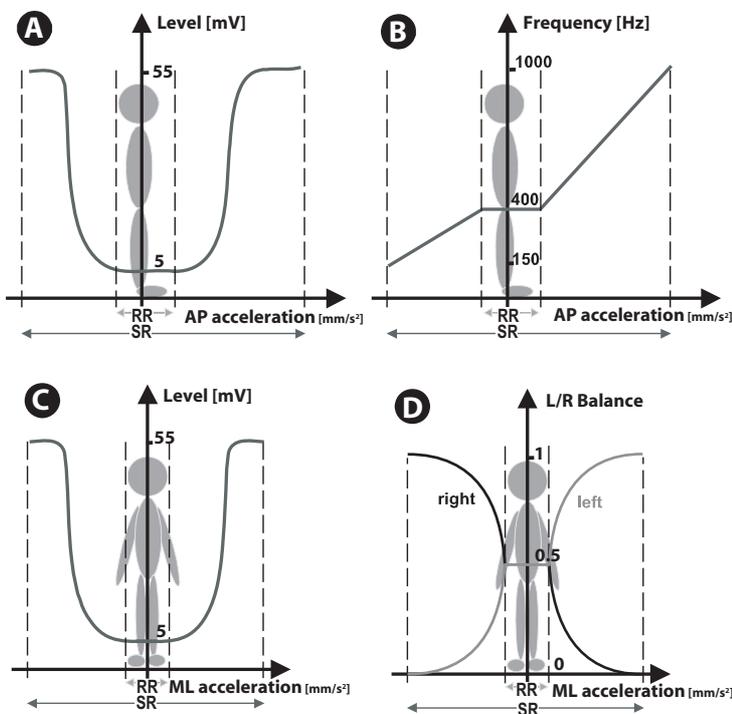


Figure 1 – (a) Level and (b) frequency modulation functions based on anterior/posterior (AP) accelerations. (c) Level and (d) left and right balance modulation functions based on ML accelerations.

The size of an individual's RR is subject-specific and is defined as a function of the person's height. To calculate RR, we use an inverted pendulum model and assume, as RR threshold, an acceleration that keeps the angular sway within ± 1 deg from initial position [9]. Because forward sway is usually larger than in any other direction, we use the value obtained from the inverted pendulum model to set the anterior threshold of the RR, and we empirically assign a coefficient of 2/3 to obtain the posterior, left, and right thresholds.

To determine an upper bound for acceptable accelerations to help

an individual stabilize upright posture, the ABF processes the limits of the stability of each user, which we term the safety region (SR). The SR is defined as a function of the user's feet dimensions. Since a body can maintain its balance while standing in static conditions if its center of mass (COM) projection falls inside its support base, we can roughly estimate an individual's limits of stability, i.e. his/her SR as a function of the size of the support base. The borders of the SR represent the maximum acceleration of the user's trunk just before the COM projects outside the base of support delimited by the dimensions of the person's feet.

The ABF device is designed to take advantage of human hearing, which recognizes differences in sound frequency more easily if a reference sound is given for comparison [10]. In our ABF system, when a user sways within the RR, the ABF sends a stereo, low-volume (a few dBs above the hearing threshold), pure tone ($f_0 = 400$ Hz) almost equivalent to the G above the middle C to the user via the headphones. However, when the user sways outside the RR, the ABF sends different tones which signal to the user that sway needs correcting and how to correct it.

We used the interval between SR and RR to design the dynamic range of the audio output, as shown in . 1. The sigmoid functions of Fig. 1(a) and Fig. 1(c) represent the coding laws for the generation of the sound level (expressed as input voltage to the headphones) based on the accelerations recorded along the AP and ML directions, respectively.

The general equation of the sigmoid functions is

$$(1) \quad L = \frac{L_0 a_i^k}{a_i^k + b^k} + c$$

where $a = \max(a_{AP}; a_{ML})$ is the maximum dimensionless ratio between the actual amount of acceleration exceeding the RR threshold and the RR-to-SR acceleration excursion in AP and ML; $k = 3$ for the anterior direction; $k = 2.5$ for the posterior and ML directions; $b = 0.3$; $L_0 = 50\text{mV rms}$ defines the sound range; and $c = 5\text{mV rms}$ sets the minimum signal level in the headphones. The consequent range of the output level may be as wide as 20 dB-SPL. The frequency modulation associated with AP acceleration follows the piecewise linear law [see (2)] shown in Fig. 1(b).

$$(2) \quad f = m a_{AP} + f_0$$

where $m = 250$ Hz outside RR backward, $m = 0$ Hz inside RR, and $m = 600$ Hz outside RR forward. The amplitude and sign of the ML acceleration regulate the L/R balance between the audio channels [see Fig. 1(d)]. Given the weighting function

$$(3) \quad w = 1 - e^{-10 a_{ML} \operatorname{sgn}(a_{ML})}$$

where $\operatorname{sgn}(\cdot)$ represents the signum function, the left and right levels are computed as

$$(4) \quad \begin{aligned} L_L &= [1 + \operatorname{sgn}(a_{ML}) w] L / 2 \\ L_R &= [1 - \operatorname{sgn}(a_{ML}) w] L / 2 \end{aligned}$$

When the subject's ML sway is inside the RR, $a_{ML} = 0$ and $w = 0$. Hence, the L/R levels are equal. Fig. 2 shows an example of the ABF variables during a representative experiment as processed in real-time by the computer, based on (1)–(4).

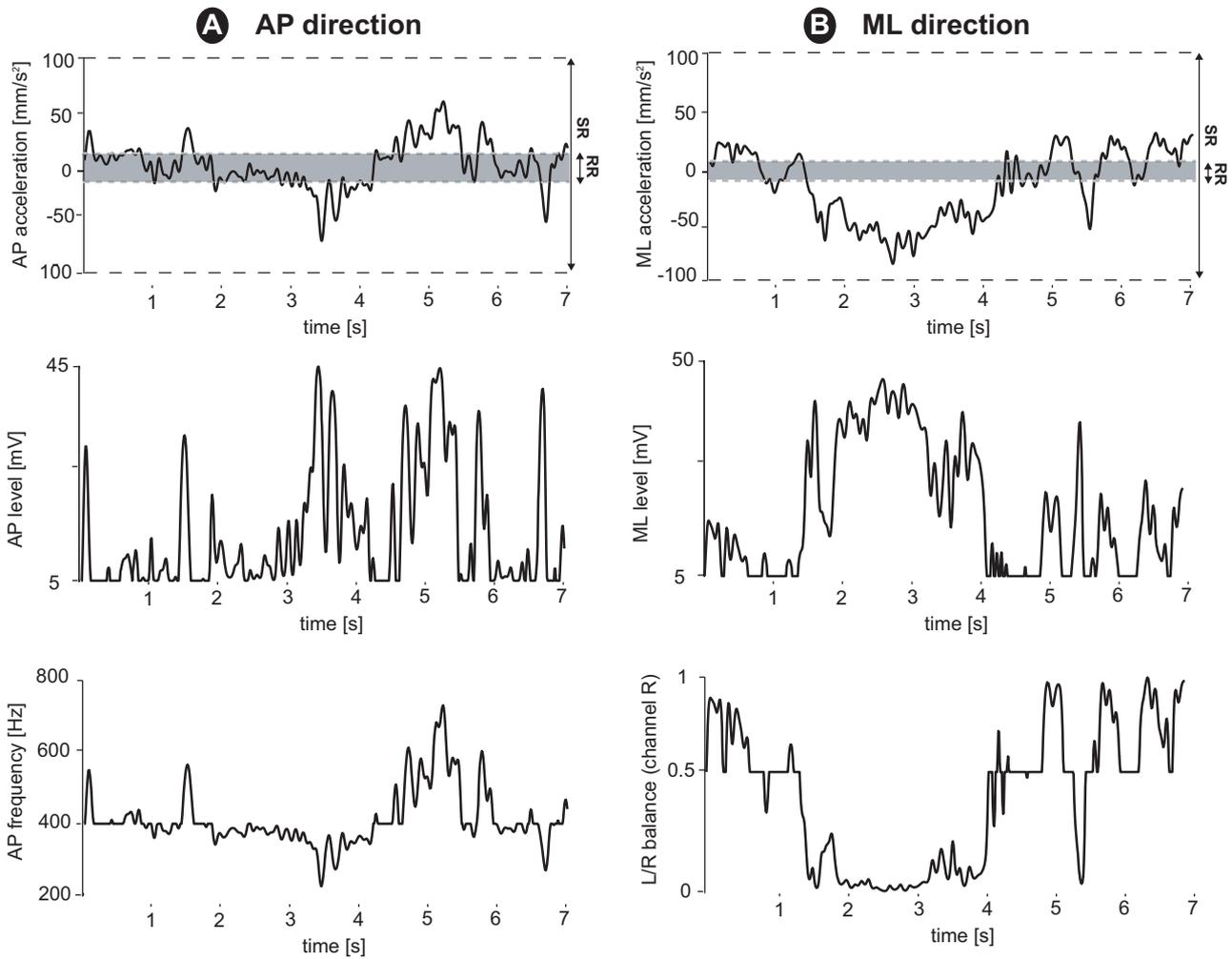


Figure 2 - Example of sound coding for part of a representative trial. (A) AP raw acceleration (top) is converted into sound level (middle) by (1) and into sound frequency (bottom) by (2). (B) ML raw acceleration (top) is converted into sound level (middle) by (1) and into left and right balance (bottom) by (3) and (4).

Experimental Protocol and Results

Several pilot experiments were performed to develop the ABF system. Critical steps in the design phase involved 1) defining the RR, the SR, and the functions to relate sound and body movements and 2) developing the digital sound generation process.

The validity and usefulness of the ABF system were evaluated in a preliminary experiment in which nine normal, healthy subjects used the ABF device to maintain balance while standing quietly. Their mean age, height, and weight were 55 (33-71 years), 167 cm (151-180 cm), and 73 kg (65-86 kg), respectively.

Each subject performed 13 trials (60 s each) with ABF while standing quietly on a force plate, in three different conditions: five trials with eyes closed (EC), five trials with eyes open and with foam under feet (EOF), and three trials with eyes closed and with foam under feet (ECF). Each subject also performed the same trials without ABF, for a total of 26 trials. The eyes-closed conditions eliminated visual information. The foam-under-feet conditions, achieved by covering the force plate with a 10-cm-thick, medium density Temper foam (Kees Goebel Medical, Inc, Hamilton, OH), made somatosensory information from the surface unreliable. The order of the trials was randomized.

For all trials, the sensory unit was mounted on the subject's back, as close as possible to the body COM by taking the subject's navel at the height of L5 as a reference. The first 10 s of each trial, regardless of sensory condition, were used for hardware re-calibration to reduce the effect of any possible drift of the sensors. A two-dimensional bubble placed on the sensory unit helped correct the alignment of the sensor. To maximize the repeatability of the procedure, the same experimenter mounted the sensory unit on all subjects.

For the trials with ABF, the subjects were instructed to keep the reference sound as constant as possible, thus indicating that postural sway was maintained within the RR. Before recording the trials, each subject performed one practice trial 1-min long to experience the relation between sound and movement, and to gain confidence with the ABF system.

During each trial, COP data from the force plate and accelerations from the portable sensor were recorded at a 100-Hz sample rate. Comparisons among the three sensory conditions concentrated on the following five COP variables: root mean square distance (RMS), mean velocity (MV), frequency containing 95% of the power (F95%), frequency dispersion (FD), and direction of maximum sway variability ($|\theta_{90-Mdir}|$) [11]. The same five variables were

computed also from the acceleration signals.

Our initial analysis assessed the relationship between the COP displacement and trunk acceleration. Thus, we performed a correlation analysis in the time domain and a coherence analysis in the frequency domain between the two signals in all three of the sensory conditions, with and without ABF. We also performed a correlation analysis between the five COP and five acceleration variables. Not surprisingly, COP displacement and trunk acceleration were largely mutually dependent (Fig. 3) [12]. As expected for an inverted pendulum model of postural sway, the correlation coefficients found between the COP and trunk acceleration signals along the AP and ML axes were high in all three sensory conditions ($0.7 < r < 0.9$). Regarding the effect of ABF on correlations, the change in correlation coefficient r was largely negligible, except in the ECF condition, where ABF reduced r slightly ($r = 0.87 \pm 0.02$ without ABF, $r = 0.78 \pm 0.03$ with ABF) but systematically in both the AP and ML directions.

The coherence between COP displacement and trunk acceleration along the AP and ML axes was high (>0.8) for frequencies below 1 Hz, peaking at 0.5 Hz. This finding is in agreement with the low-pass nature of the biomechanical filter that relates trunk (and body) motion and the location of the COP [1].

COP displacement RMS and acceleration RMS were the variables with the strongest correlations ($r = 0.74$), while the other parameters had lower correlations: MV: $r = 0.36$, F95%: $r = 0.36$, FD: $r = 0.62$, and $|90\text{-Mdir}|$: $r = 0.50$.

Fig. 4 shows the percentage change due to ABF observed in the COP-based parameters in all three sensory conditions. Using ABF in the EC conditions, all nine subjects swayed less, as reflected by the reduction of COP displacement RMS (statistically significant in EC, $p < 0.05$; in ECF, $p < 0.01$). In addition, using ABF, most of the subjects applied more postural corrections to their sway, as shown by the increase in MV (statistically significant in EOF, $p < 0.01$)

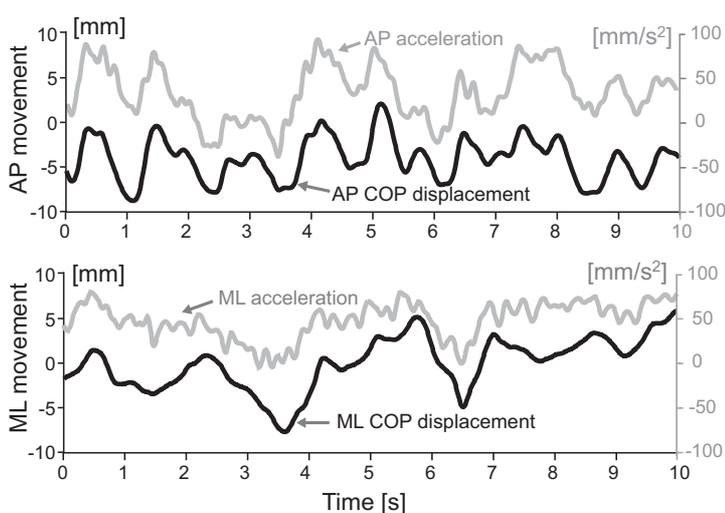


Figure 3 - Correlations between COP and trunk acceleration signals. AP (top) and ML (bottom) components of COP displacement (black line) and acceleration (gray line) for a part of a representative trial are shown.

and F95% (consistently statistically significant across conditions, $p < 0.01$). ABF had no clear influence on FD and $|90\text{-Mdir}|$. The more challenging the sensory condition, the more that ABF affected both stability and postural corrections. In fact, ABF benefited subjects' maintenance of stance within the RR the most in the ECF condition. The corresponding values of the COP parameters, expressed as mean (\pm SD), were: without ABF: RMS = $14.8 (\pm 3.9)$ mm, MV = $27.7 (\pm 11.3)$ mm/s, F95% = $1.59 (\pm 0.18)$ Hz, FD = $0.77 (\pm 0.05)$, $|90\text{-$

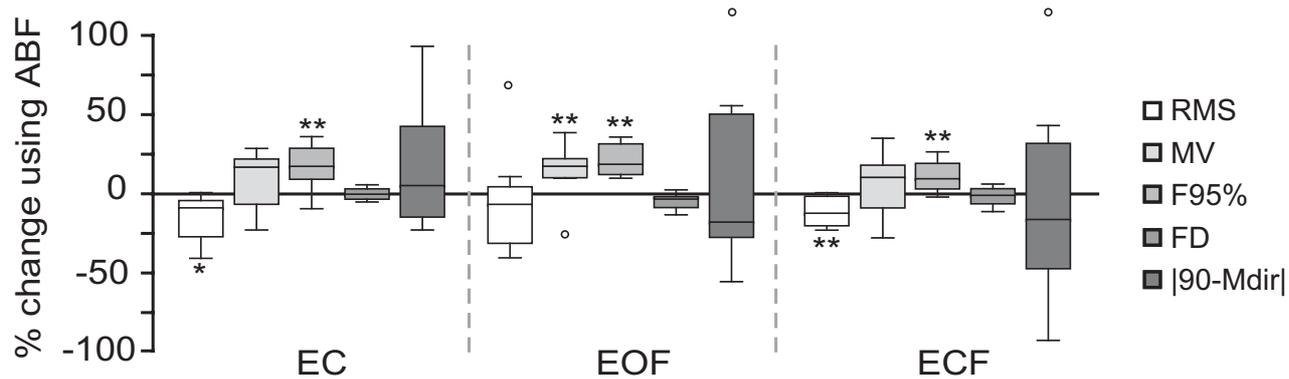


Figure 4 - Effects of ABF on COP parameters in all three sensory conditions tested. Boxplots describe the distribution of the percent changes of the five parameters across the population. Small circles indicate outlying values. *: $p < 0.05$; **: $p < 0.01$.

Mdir| = $17.0 (\pm 15.9)$ deg; and with ABF: RMS = $12.5 (\pm 3.6)$ mm, MV = $28.1 (\pm 7.6)$ mm/s, F95% = $1.73 (\pm 0.15)$ Hz, FD = $0.75 (\pm 0.03)$, and |90-Mdirj| = $19.0 (\pm 14.9)$ deg.

Discussion

We have developed and preliminarily tested an ABF system that sends trunk acceleration information to users to help them correct postural sway during stance. This acoustic information helped subjects reduce postural sway, especially when visual and sensory information were compromised by eye closure and stance on foam. The instrument met requirements for an adequate biofeedback system: adequate bandwidth and sensitivity, convenient feedback signal generation, and lightweight portability. None of the subjects had problems learning how to use the ABF system, and the 1-min practice trials were adequate to teach them how to use ABF to reduce their sway while quietly standing. The efficacy of ABF appears to depend on the availability of alternate sensory information since the more subjects were unstable in a sensory condition, the more that they improved their balance with ABF. This finding suggests that subjects use ABF to partially substitute for the lack of visual information and/or for the unreliability of somatosensory information while they try to maintain postural control.

The results reported here were from experiments with normal, healthy subjects who have extensive sensory and functional redundancy in their postural system. We hypothesized that our ABF device would help subjects with sensory deficits improve postural sway even more, and subsequent studies of ABF experiments with bilateral vestibular loss subjects confirmed this hypothesis [13]. In the present study the improvements in stance were probably due to a change in postural control strategies because sway variables measured with ABF were consistent with smaller (see decrease in RMS) and more frequent (see increase in MV and F95%) postural corrections [14]. In accord with this result is also the decrease in correlation between COP and acceleration signals observed in the ECF condition. This decrease may reflect a moderate decline in the simple ankle strategy to maintain balance [15] in favor of more complex control; experiments aimed at investigating this hypothesis are in progress. However, it is possible that the attention of these subjects (which was not measured in the protocol reported here) may also have contributed, at least in part, to their improved balance while using ABF.

Many earlier biofeedback systems used audio alarms to notify the user of abnormal values of monitored parameters (e.g. [16]). The present ABF system is novel in the use of nonlinear coding functions and in the customization of these functions to each subject and task. Preliminary results suggest this ABF device may be a useful tool for rehabilitation in the clinic, home-care setting, and community during mobility training. The use of ABF may

become attractive for rehabilitation, especially if it is found to favor neural plasticity in motor control [2]. In other words, a person with impaired abilities to control posture could practice with ABF to achieve better postural control when not using ABF.

Plans are underway to improve the current ABF system by making it wireless for increased portability and for enabling remote control and remote monitoring. Different sonification procedures will also be tested in the near future. In particular, 3-D generated sound with a headrelated transfer function or immersive sound will be investigated. In addition, since the current ABF system may interfere with hearing for communication purposes or may be unsuitable for people with hearing deficits. Other sound-delivery processes will be investigated, including bone mastoid vibration.

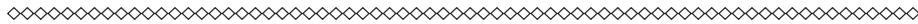
The strong correlation between COP and acceleration signals suggests that the sensory unit could be developed for use as a portable, miniaturized force plate [17], which may be helpful for remote monitoring such as for elderly persons and persons with postural and mobility disorders.

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Chapter 3

Influence of a Portable Audio-Biofeedback Device on Structural Properties of Postural Sway



Most of the content of this chapter has been published in: M. Dozza, L. Chiari, B. Chan, L. Rocchi, F. B. Horak, and A. Cappello, "Influence of a portable audio-biofeedback device on structural properties of postural sway," J Neuroeng Rehabil, vol. 2, 2005.

Abstract

Good balance depends on accurate and adequate information from the senses. When sensory information is limited or unreliable balance may become critical. One way to substitute missing sensory information for balance is with biofeedback systems. We previously reported that audio-biofeedback (ABF) has beneficial effects in subjects with profound vestibular loss, since it significantly reduces body sway in quiet standing tasks

In this paper, we present the effects of a portable prototype of an ABF system on healthy subjects' upright stance postural stability, in conditions of limited and unreliable sensory information. Stabilogram diffusion analysis, combined with traditional center of pressure analysis and surface electromyography, were applied to the analysis of quiet standing tasks over a Temper foam surface.

These analyses provided new evidence that ABF may be used to treat postural instability. In fact, the results of the stabilogram diffusion analysis suggest that ABF increased the amount of feedback control exerted by the brain for maintaining balance. Interestingly, the resulting increase in postural stability was not at the expense of leg muscular activity, which remained almost unchanged.

Examination of the stabilogram diffusion analysis and the EMG activity supported the hypothesis that ABF does not induce an increased stiffness (and hence more co-activation) in leg muscles, but rather helps the brain to actively change to a more feedback-based control activity over standing posture.

Introduction

Maintaining balance is a complex task accomplished by the brain through the fusion and interpretation of sensory information. When sensory information from vestibular, somatosensory, and visual systems [1-3] are not accurate and/or adequate, balance will be compromised. Although, in many cases, the loss of peripheral sensory information is not curable or reversible, the brain can compensate for the loss of sensory information by relying more on the other sensory channels [4;5].

The purpose of biofeedback (BF) systems for postural control is to provide additional sensory information about body equilibrium to the brain [6]. In the last few years, different sensors, encoding algorithms, and information restitution devices have been combined to develop promising BF systems for postural control [7-9]. The major design goals were focused on portability, usability, economy, and effectiveness in balance improvements [8;10-12].

The development of these BF systems has been facilitated by the availability of lightweight, miniaturized, and economical sensors such as accelerometers, inclinometers, and gyroscopes [13]. The use of these sensors makes BF devices inexpensive, unsusceptible to shadowing effect, and not limited in the measurement field, in contrast to dynamometric platforms and motion analysis systems, which are commonly used in laboratory settings [14;15]. In addition, due to their size and weight, these sensors can measure body segment movement without hindering natural motor execution.

More detail is needed for understanding how biofeedback information interacts with the brain or, from a neuroscience perspective, how the brain uses artificial BF information and combines it with natural sensory information. We believe that understanding this interaction is fundamental for further developing effective BF systems.

An interesting analysis in the understanding of how the brain may use BF information for postural control was proposed by Collins and De Luca in 1993 [16]. These authors developed a statistical-biomechanics method for analyzing force platform data recorded during quiet standing, called stabilogram diffusion analysis (SDA). SDA was applied to center of pressure (COP) data and it disclosed that COP tends to drift away from a relative equilibrium point over short-term observation intervals (less than 1-second long), whereas COP tends to return to a relative equilibrium point over long-term observation intervals. These results took Collins and De Luca to suggest that the motion of the COP is not purely random, and that SDA may be

able to give insight on the amount of open-loop and closed-loop postural control applied by the central nervous system for maintaining balance [17]. SDA was used several contexts, e.g. to evaluate the effect of spaceflight [18], visual input [19;20], and age-related changes [21;22] on postural stability. In 2000, Chiari developed and validated a new nonlinear model for extracting parameters from SDA diagrams, reducing from 6 to 2 the number of the parameters used to characterize the structural properties of COP [20]. In 2004, Rocchi found that these new parameters may be useful adjuncts to evaluate postural control strategies in patients with Parkinson's disease and may allow the comparison of different deep brain stimulation electrode sites based on their effect on structural properties of the COP [23].

In this paper, we investigate the effect on postural stability of a portable, accelerometry-based, audio biofeedback (ABF) system recently developed by the authors [9]. Standing with eyes closed on Temper™ foam will be used to evaluate the effects of artificial auditory cues to enhance the reduced (from the eyes) and masked (from the feet) natural sensory information. Measurements include COP recorded by a force platform under the feet, trunk acceleration measured by the ABF sensors, and EMG signals from the leg muscles. SDA according to Chiari et al. [20], traditional COP analysis [24], and muscle activation analysis according to Olney & Winter [25] were performed in order to evaluate the effect of ABF on healthy young subject's upright posture.

These analyses were aimed to answer two questions: (1) do structural properties of postural sway change with ABF? And, if so, (2) in which way will this help in understanding the mechanisms underlying ABF efficacy and in improving the design of a rehabilitation strategy for balance disorders? In this paper, we present evidence that supports the hypothesis that ABF does not simply induce a purely biomechanical increase in stiffness (and hence more co-activation) in the leg muscles, but rather ABF helps the brain actively adapt its control activity over standing posture.

Methods

Participants

Eight healthy subjects participated to this experiment (5 males and 3 females, aged 23.5 ± 3.0 yrs, range 21-28 yrs). All participants were free from any neurological, orthopedic, hearing, or vestibular disorder. Informed consent form was obtained from each subject. The form was prepared in accordance with the Oregon Health & Science University Ethical Committee and respected the declaration of Helsinki, 1964.

Apparatus and procedure

Subjects performed 10, 60-second trials standing with eyes closed on Temper™, 4"-thick foam. COP displacement was recorded with an AMTI OR6-6 force plate. An ABF system [9] was used to provide subjects with additional balance information related to trunk acceleration. The ABF system used a sensor, based on 2-D accelerometers (Analog Device ADXL203) mounted on the subject's back (L5), to create an audio stereo sound representing the acceleration sensed along the anterior-posterior (AP) and the medial-lateral (ML) direction. A laptop, Toshiba Celeron 2.3 GHz, was dedicated to convert the accelerations into stereo sounds. Commercial headphones were used by the subjects to listen to the ABF sound. The ABF system is described in detail in [9] and illustrated in Figure 1. In short, the stereo sound provided by the ABF system consisted of two sine waves, one for the left ear channel and one for the right ear channel. Pitch, volume and left/right balance of the stereo sound were modulated to represent the 2-D acceleration information. Specifically, when the subject swayed forward, and consequently the acceleration increased in the anterior direction, the sound got louder in volume and higher in pitch. When the subject swayed backward, and consequently the acceleration increased in the posterior direction, the sound got louder in volume and lower in pitch. When the subject moved right and, consequently, the acceleration increased in the right direction, the sound got louder in the right ear channel and lower in the left one. When the subject moved left, and consequently the acceleration increased in the left direction, the sound got louder in the left ear channel and lower in the right one. The sound dynamics was optimized for each trial by taking as a reference the first 10-second recordings of each trial. The equations used for the pitch, volume, and left/right balance modulation can be found in [9]. Each subject was instructed to maintain balance during the trials by taking advantage of the ABF information, when available. Five trials with ABF and 5 trials without ABF were

performed in randomized order by each subject. Before the experimental session, the subjects were instructed on how ABF codes trunk acceleration into sound, and performed free-movement trials until they felt confident in performing the full experiment.

Data recording

For each standing trial, ground reaction forces and torques were recorded from the force plate with a 100-Hz sample frequency. COP displacement was processed offline from the force plate data after applying a 10-Hz cut-off, low-pass Butterworth filter. Accelerations along AP and ML direction were collected with a 100 Hz sample frequency after applying a low-pass filter with a 20-Hz cut-off. EMG was recorded from right leg muscles, tibialis (TI), soleus (SO), and gastrocnemius (GA) with two surface electrodes fixed about 2 cm apart along the length of each muscle belly; the ground electrode was fixed on a bony area of the right hallux. The EMG signals were amplified 20000 times, band-pass filtered (71-2652 Hz), integrated and full-wave rectified with a 6th order Butterworth low pass with a cut-off of 100Hz.

Data analysis

From AP COP data, the root mean square distance (COP-RMS) and the frequency comprising the 95% of the power (F95%) were extracted according to Prieto et al. [24].

From the acceleration sensed at trunk level along AP direction we computed the root mean square value (Acc-RMS).

In addition, two stochastic parameters were included in the analyses. These parameters characterize a previously developed model that describes with continuity the transition among the different scaling regimes found in the COP time series [26]. The model is described by the following equation:

$$V(\Delta t) = K \Delta t^{2H(\Delta t)}$$

where $V(\Delta t)$ is the variance of COP displacement, computed at time-lag Δt , and H is the scaling exponent, also called Hurst exponent. This is assumed to follow a sigmoid law in the time interval (Δt):

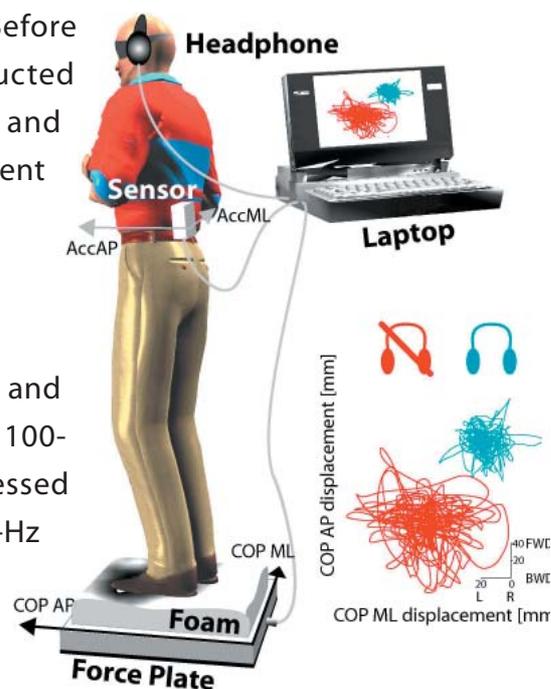


Figure 1 – ABF system device and protocol
The ABF consisted of (1) a sensor mounted on the trunk and sensing acceleration along AP and ML axes, (2) a laptop acquiring acceleration from the sensor and processing the ABF sound, (3) a pair of earphones the subject wears for listening to the sound. In this figure is also shown the protocol where a healthy normal subject is standing on a foam placed on a force plate with eyes closed. Finally, in the figure bottom right statokinesigrams in condition with and without ABF from a representative subject are shown.

$$H(\Delta t) = \frac{\log 2}{\log [2 (1 + \Delta t / \Delta T_c)]}$$

In this way, the features extracted from COP data are the following (see [20] for more details):

- K is an estimate of the diffusion coefficient of the random process obtained by sampling the COP time series at the sampling frequency $1/\Delta T_c$.

- ΔT_c represents the time-lag at which the real process corresponds to a purely random behavior, and where it switches from a persistent (positively correlated, and hence interpreted in terms of feed-forward control) to an anti-persistent (negatively correlated, and hence interpreted in terms of feedback control) behavior [16].

Mean muscular activity was calculated from the full wave rectified EMG of each muscle. For each subject and each muscle, muscle activity was expressed in percentage in reference to the trial with maximal activity recorded. This made possible the comparison of muscle activity among the different subjects. The EMG signals were further processed applying a low pass-filter with a 2 Hz cut-off in order to obtain tension curves according to [25]. These tension curves were cross-correlated to determine the amount of co-activation between the muscles recorded.

Statistical analysis

Paired T-tests were performed to determine the effect of ABF on the different parameters extracted from COP, acceleration and EMG data collected. The threshold for statistical significance was set to $p=0.05$.

Results

Subjects' confidence and comfort

All participants reported ABF sound was comfortable and its way of representing the information was intuitive. In fact, none of the subject needed more than two, free-movement trials before feeling ready to start the experiment.

Subjects' sway

ABF significantly influenced subjects' performance on the foam. The percentage change induced by ABF on all sway parameters, either measured at the trunk level with the accelerometer or at the feet level with the force platform, is shown in Figure 2. Figure 2 also reports significance levels of the parameter changes occurred while using the ABF. The general results shown in Figure 2 are specified in detail in the following.

Center of Pressure analysis

Center of pressure displacement in the AP direction was significantly influenced by ABF. T-tests results revealed significant effects of ABF on COP-RMS ($p=0.015$). This effect is shown by a consistent reduction of COP-RMS for 7 out of 8 subjects as shown in Table 1 (column 7). Average reduction of COP-RMS was 10.7%. Columns 1 and 4 of Table 1 also show the subject-by-subject values of COP-RMS without and with ABF, respectively. The last three subjects (#6, #7, #8) were females and showed smaller COP-RMS, as expected considering their smaller heights [26].

F95% increased with ABF for 7 out of 8 subjects (Table 1, column 8) but this result was not significant ($p=0.42$). The values of F95% are also reported for each subject in both conditions (Table 1, columns 2 and 5). Average increase of F95% due to ABF was 6.2% as shown in Figure 2.

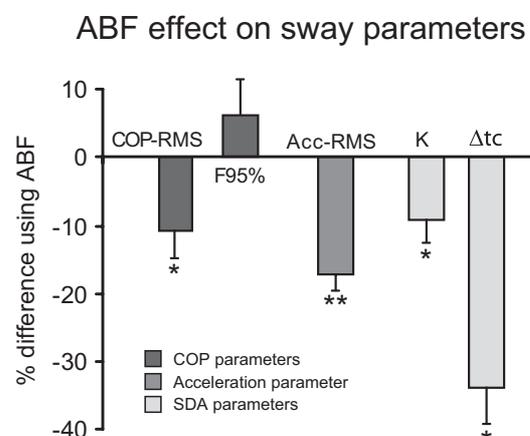


Figure 2 - Effect of ABF on sway parameters. The effect of using ABF on the sway parameters is reported in percentage. COP-RMS and F95% were extracted from the AP COP displacement according to [24]. Acc-RMS was extracted from AP acceleration recorded at trunk level (L5). K and Δtc were derived by applying the method proposed by [20] on the SDA diagrams [16]. Asterisks indicate statistical significance: * $p<0.05$ and ** $p<0.01$. The reductions of K, COP-RMS and Acc-RMS are a consistent evidence of the reduction of sway amplitude shown by the subject using ABF. The increasing of F95% suggests that the postural control applied by the CNS when ABF is available was increased. The reduction of ΔTc suggests a major active closed-loop postural control exercised by the CNS.

Table 1 – ABF effect on sway parameters. Standard deviations are indicated in parenthesis.

	COP-RMS (NO - ABF) [mm]	F95 % (NO - ABF) [Hz]	Acc-RMS (NO - ABF) [mm/s ²]	COP-RMS (ABF) [mm]	F95 % (ABF) [Hz]	Acc-RMS (ABF) [mm /s ²]	% COP-RMS difference	% F95 % difference	% Acc-RMS difference
Subj. #1	10.79 (2.84)	0.99 (0.05)	137 (48)	9.57 (1.86)	1.18 (0.16)	118 (13)	-11.2	19.1	-14.1
Subj. #2	9.91 (2.77)	1.20 (0.29)	142 (27)	9.50 (2.26)	1.30 (0.20)	120 (23)	-4.1	8.7	-15.6
Subj. #3	9.21 (2.94)	1.16 (0.14)	121 (23)	8.61 (1.42)	1.37 (0.07)	113 (21)	-6.5	18.0	-7.0
Subj. #4	10.23 (1.50)	1.43 (0.08)	117 (30)	8.80 (1.74)	1.49 (0.12)	100 (12)	-13.9	4.1	-14.6
Subj. #5	8.50 (0.93)	1.49 (0.22)	143 (46)	6.90 (1.35)	1.53 (0.28)	115 (19)	-18.8	2.6	-19.3
Subj. #6	9.62 (1.55)	1.34 (0.30)	126 (43)	7.35 (0.88)	1.34 (0.09)	89 (20)	-23.6	0.0	-29.2
Subj. #7	6.37 (1.48)	1.60 (0.07)	64 (8.3)	5.19 (0.59)	1.94 (0.12)	51 (4.7)	-18.5	20.8	-20.1
Subj. #8	6.08 (1.19)	1.78 (0.25)	48 (6.3)	6.75 (1.41)	1.37 (0.16)	39 (3.8)	10.9	-23.1	-17.3
Average	8.84 (1.75)	1.37 (0.26)	112 (36)	7.83 (1.54)	1.44 (0.15)	93 (31)	-10.7 (10.9)	6.2 (14.4)	-17.2(6.3)

It is worth noting that subject #8 behaved as an outlier (Figure 3), compared to the other subjects since she was the only one who showed opposite changes in COP-RMS and F95% while using ABF. Performing the T-Tests with this outlier eliminated increased the effect of ABF on COP-RMS ($p=0.002$), and on F95% ($p=0.02$). These results better match the results already published in [9]. The outlying behaviour of subject #8 will be investigated further in the discussion.

Acceleration analysis

Acceleration sensed at trunk level (L5) in AP direction was significantly reduced by ABF. T-test results also revealed significant effects of ABF on Acc-RMS ($p=0.0009$). Acc-RMS was reduced by ABF across all subjects, as shown in Table 1 (last column). Average reduction of Acc-RMS was 17.2% (Figure 2). Columns 3 and 7 of Table 1 also show the subject-by-subject values of Acc-RMS without and with ABF, respectively. The last three subjects were females and showed smaller Acc-RMS, as expected considering their smaller heights [26].

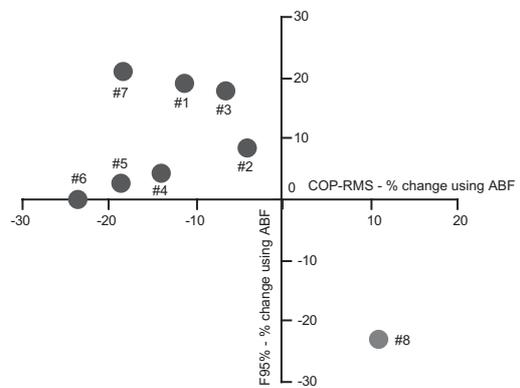


Figure 3 – Antithetic behavior of subject #8. On the horizontal axis COP-RMS percentage change using ABF is reported whereas on the vertical axis F95% percentage change using ABF is reported. The values of each subject from Table 1 are plotted. Subject #8 behaves antithetically to the other subjects.

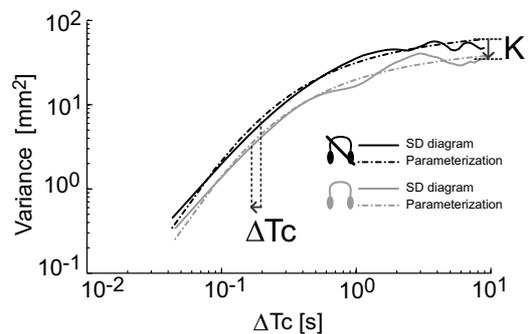


Figure 4 – Effect of ABF on open-loop and closed-loop control. SDA diagrams for one representative subject. Two conditions are reported: without ABF (black) and with ABF (gray). The behavior of the parameters K and ΔTc used to parameterize the SDA diagrams is also shown. This figure suggests that, using ABF, subjects decrease the amount of sway by increasing the closed-loop (feedback) posture control.

Stabilogram diffusion analysis

SDA diagrams plotted from AP COP data, were also significantly influenced by ABF (Figure 4). As a consequence, the parameters K and ΔTc characterizing the SDA diagram, were both significantly decreased by ABF (Figure 2). Average K reduction was 9.3% ($p=0.02$), whereas average ΔTc reduction was 33.9% ($p=0.018$). Table 2 reports the subject-by-subject values of K and ΔTc in both conditions tested. Subject #8 and subject #7 are the only ones who showed a slight increase in K .

Muscle activity analysis

Muscle activity of TI, GA, and SO was not influenced by ABF. Overall, the mean activity, expressed as a percentage of the maximal activity recorded from each single muscle across all the trials of a subject, did not change significantly due to ABF (see Figure 5A). TI activity showed a trend toward increasing in trials with ABF ($p=0.17$) but this change was particularly clear only for subjects #4 and #7.

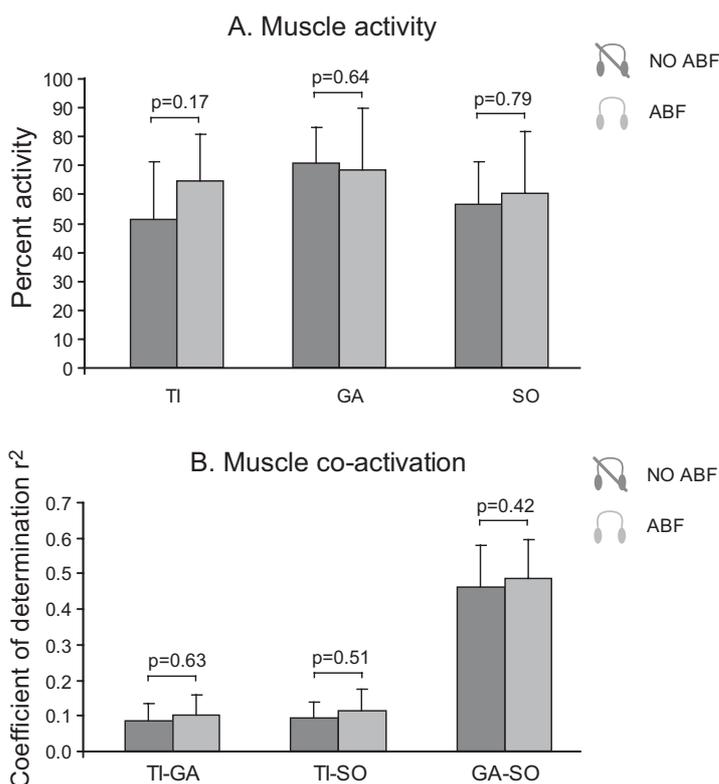


Figure 5 – Effect of ABF on muscle.

Estimates of muscular co-activation (Fig. 5A) for different pair of muscles (TI-GA, TI-SO, GA-SO) and muscle activity (Fig. 5B) are shown. Average values are reported for trials with (light gray) and without (dark gray) ABF. Error bars represent standard deviation. As shown in Figure 5A, using ABF does not change significantly the co-activation between the muscles analyzed (p values from T-Test are reported). This suggests that the major amount of postural corrections induced by ABF does not involve a major co-activation of the muscles TI, GA, and SO in the leg. As shown in Figure 5B, using ABF does not change significantly the activity of the muscles analyzed (p values from T-Test are reported). This suggests that the major amount of postural corrections induced by ABF does not involve a major average activity of the muscles TI, GA, and SO in the leg.

Table 2 – ABF effect on SDA parameters. Standard deviations are indicated in between parenthesis.

	K (NO-ABF) [mm ²]	Δtc (NO-ABF) [s]	K (ABF) [mm ²]	Δtc (ABF) [s]	% K difference	% Δtc difference
Subj. #1	100 (57)	0.42 (0.21)	86 (15)	0.38 (0.17)	-14.6	-9.9
Subj. #2	70 (29)	0.51 (0.31)	66 (20)	0.41 (0.34)	-7.4	-20.5
Subj. #3	75 (41)	0.52 (0.29)	65 (20)	0.29 (0.12)	-13.3	-45.3
Subj. #4	80 (21)	0.81 (0.46)	70 (14)	0.39 (0.14)	-11.1	-52.0
Subj. #5	47 (13)	0.32 (0.08)	39 (10)	0.26 (0.16)	-18.1	-19.7
Subj. #6	64 (12)	0.27 (0.08)	61 (9)	0.20 (0.09)	-5.7	-26.1
Subj. #7	32 (7)	0.17 (0.06)	34 (9)	0.09 (0.01)	6.6	-47.4
Subj. #8	35 (14)	0.29 (0.09)	38 (13)	0.19 (0.06)	5.8	-34.3
Average	63 (23)	0.41 (0.20)	57 (18.5)	0.27 (0.11)	-9.3 (9.2)	-33.9 (15.3)

Muscle co-activation of ankle agonists-antagonists did not change significantly due to the ABF (see Figure 5B). Co-activation between TI and GA was small both with ($r^2=0.11$) and without ($r^2=0.08$) ABF. Similarly small was the co-activation between TI and SO with ($r^2=0.14$) and without ($r^2=0.09$) ABF. As expected, co-activation between GA and SO was instead large ($r^2=0.39$ in trials with ABF and $r^2=0.46$ in trials without ABF). Figure 5B reports the coefficient of determination r^2 , which indicates the amount of muscular co-activation, for all pairs of muscles analyzed in trials with and without ABF.

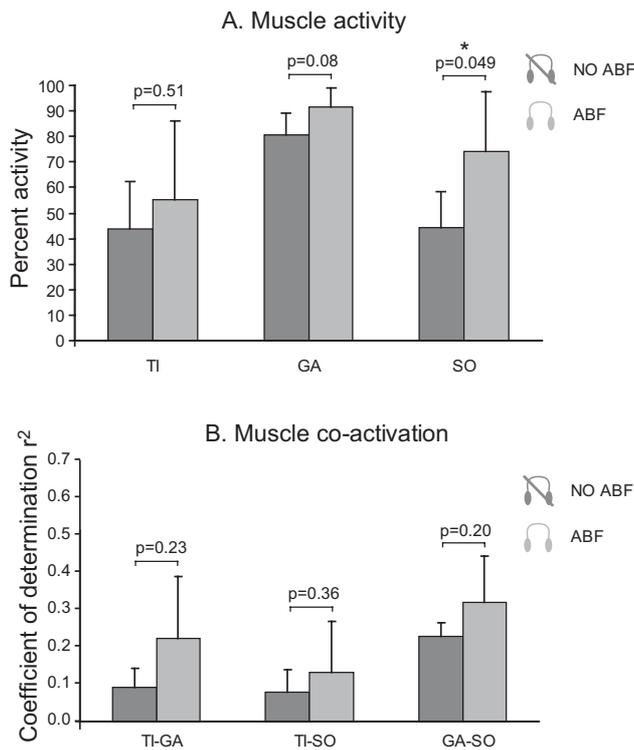


Figure 6 – Muscle activity and co-activation in subject #8.

The antithetic behavior of subject #8 for muscles co-activation, (Fig.6A), and for muscles activity, (Fig. 6B) is shown. Figure 6a reports the estimates of muscular co-activation for different pair of muscles: TI-GA, TI-SO, and GA-SO. Average values are reported for trials with (light gray) and without (dark gray) ABF. Error bars represent standard deviation. Even if co-activation looks higher in trials with ABF for all couples of muscles while using ABF, muscles co-activation does not change significantly (p values from T-Test are reported; since the number of samples is five it is convenient to report also the powers which were respectively: 0.20, 0.14, 0.23). This suggests that a major amount of co-activation of the muscles TI, GA, and SO was exercised by this subject while using ABF. Figure 6B reports the estimates of muscular activity for TI, GA, and SO muscle. Average values expressed in percentage are reported for trials with (light gray) and without (dark gray) ABF. Error bars represent standard deviation. The percent activity was calculated taking as one-hundred-percent reference the trial with the highest muscle activation recorded. Even if muscles activity looks higher in trials with ABF for all muscles, only SO activity changed significantly while using ABF (p values from T-Test are reported; since the number of samples is five, it is convenient to report also the powers which were respectively: 0.09, 0.41, 0.53). This suggests that a major amount of activity of the muscles TI, GA, and SO was exercised by this subject while using ABF.

Discussion

Using the proposed ABF device, all healthy subjects included in this study could sway less when standing in a particularly challenging condition, with vision unavailable and somatosensation partly unreliable. All subjects, in fact, reduced their AP Acc-RMS (see Table 1). In this way, subjects were further from their stability limits and, consequently, more stable. Trunk stabilization also entailed the need of smaller corrective torques at the ankles, and hence smaller COP displacements. This is proved by the fact that all subjects but one (Subj. #8) showed a significant decrease in AP COP-RMS (Fig. 2). During ABF, postural corrections in leg muscles were likely smaller but more frequent in number, as suggested by the increase in F95% of the COP, even if the EMG signals available did not clearly confirm this possibility. Future studies involving more sophisticated techniques for the acquisition and analysis of the EMG signals will be needed to validate this hypothesis. This result suggests that ABF may partially substitute for the lack of visual and somatosensory information for postural control by taking the postural control system towards a new steady state associated with a different control strategy.

Examination of the SDA and the EMG activity supported the hypothesis that ABF does not simply induce an increased stiffness (and hence more co-activation) in leg muscles, but rather helps the brain to actively change to a more feedback-based control activity over standing posture. Representative SDA diagrams reported in Figure 4 suggest that ABF contributes to a general reduction of both the diffusion coefficient K and the transition time ΔT_c . Downward shifts of the SDA diagrams, described by smaller diffusion coefficients, reflect a reduced stochastic activity of the COP, and hence a more tightly regulated control system [16]. Shorter transition times reflect an earlier switching between persistent and antipersistent behaviors, and hence more prompt reactions to perturbations of the postural control system [27]. In summary, these results disclose, as a consequence of ABF: 1) an increase in stability, and 2) a more prominent role for feedback control over feed-forward control. Hence, the solution proposed by the brain after ABF seems to involve more feedback control for a more stable sway.

Interestingly, this result is partly different from the one observed by Rougier in quiet stance experiments with visual BF [28]. In that condition, with BF, SDA diagrams only changed some local properties (local slopes) over short or long observation intervals but did not shift significantly, meaning that one may expect that K is not changing that much. Further, closed-loop control operated over longer observation-times, suggesting that feed-forward control is

expanding over feedback control. Such a different behavior may find an explanation in the peculiar role, not just a simple redundancy, of different senses in multi-sensory integration for the control of posture [29]. Whereas vision induces alertness of the outer environment and hence pushes towards predictions of forthcoming events in the scene (feed-forward control) [30]. In contrast, hearing, compared to vision, may be more important for postural reactions to disturbing stimuli (feedback control). This result can also be related to the different processing times required by the central nervous system for visual and auditory stimuli with auditory reaction times significantly faster than visual reaction times. Finally, another factor which may explain the different outcomes of the two BF-studies is the selection of two, different, input variables (COP for visual BF and Acceleration from the trunk for ABF). It is widely accepted that upper- and lower- body segments are controlled separately [31].

Both predictive (feed-forward) and reactive (feedback) control need to be used in order to have an adequate interaction with the environment. For this reason, it's hard to tell if ABF is preferred to visual BF, or vice versa. Rather, the point is that it could be important, in a rehabilitation setting, to identify which one of the two components of postural control need more reinforcement or substitution in a particular patient, and consequently design an optimized BF treatment.

The outlying results observed for Subj. #8 need to be discussed individually. This woman in fact did not decrease COP-RMS and K, and did not increase F95%, even if, similarly to the other subjects, she decreased Acc-RMS and ΔT_c (these changes were consistent across the whole population). Hence, with ABF she actually swayed less and she showed the same increase of feedback control. Nonetheless, either due to her small body size or to a slightly different control scheme, she obtained these goals with a different strategy. Figure 6 reports her muscle activities and co-activations. It can be seen how she generally improves muscle activity with ABF (Figure 6A), in particular with a large increase in the activity of posterior muscles, GA and SO. It should be noted, however, that also the estimated co-activations (Figure 6B) look pretty dissimilar compared with the ones of the other subjects, shown in Figure 5B. Particularly low is the co-activation of agonists muscles GA-SO without ABF, which ABF partly contributes to enlarge. For all these reasons her postural behavior in the proposed task should be looked as an outlying behavior and more analyses are needed, on a larger population, to assess the real influence of body size or usual control setting on the responsiveness to ABF.

Many earlier biofeedback systems used audio alarms to notify the user of abnormal values of monitored parameters (e.g. [32]). The present ABF system is novel in the use of nonlinear coding functions and in the customization of these functions for each subject and task. Although the current ABF system may interfere with use of hearing for communication, it may be quite useful during the rehabilitation and training process. Plans are underway to improve the current ABF system by making it wireless for increased portability and equipping it with a communication module for remote control, recording, and monitoring. Different

sonification procedures will also be tested and compared in a near future. Specifically, 3-D generated sound with a HRTF (Head Related Transfer Function) or immersive sound may be even more effective signal for improving stance balance.

In conclusion, we have investigated the attributes of a portable instrument that feeds back trunk acceleration to help subjects reducing their postural sway during stance. The instrument meets requirements for an adequate biofeedback system that may find interesting applications not only as a rehabilitation device in the clinic, but also in the home care setting, and when doing community mobility training outside the traditional clinic setting. In fact, it has appropriate bandwidth and sensitivity, smoothness and delay of the acoustic signal generator, and portability. Acoustic information related to trunk movement allowed subjects in the present experiment to increase postural stability when sensory information from vision and the surface were compromised by eye closure and stance on foam. We provided evidence that the balance improvement was not simple stiffening at the ankle, but rather the brain actively adapted its control activity over standing posture with more feedback-based control.

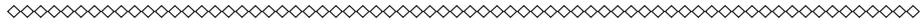
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Chapter 4

Direction Specificity of Audio-Biofeedback for Postural Sway



Abstract

Sway reduction induced by use of biofeedback devices has been widely documented in postural control research. However, the extent to which subjects use a generalized versus a direction-specific mechanism to reduce sway is unknown. In this study, we investigated the effects of audio biofeedback related to medial-lateral trunk acceleration or to anterior-posterior trunk sway on medial-lateral and anterior-posterior center-of-pressure displacement during stance. Results show that direction-specific, audio-biofeedback allowed subjects to reduce their center-of-pressure displacement by increasing the frequency of their postural corrections in the specific direction of the audio-biofeedback. The direction-specific reduction of center-of-pressure displacement and increase of its frequency bandwidth associated with direction-specific biofeedback found in this study suggests that subjects do not reduce center-of-pressure displacement by a general stiffening strategy but by increasing closed-loop control of posture.

Introduction

Different mechanisms have been suggested to reduce sway in subjects attempting to control their stance under a variety of experimental conditions. Sway reduction during stance has been reported to be due to (1) a noise reduction in sensory feedback loop associated with an increase in availability of sensory information [1], (2) an enhanced feedforward control from repetitive balance training [2], (3) a generalized cognitive interference from the performance of a dual task [3], and (4) a change in postural alignment and generalized muscle stiffness associated with a threat of a fall [4]. Understanding the mechanisms used by subjects to reduce their sway under different conditions is fundamental in order to determine how the central nervous system is involved in this process.

The mechanisms applied by subjects to reduce postural sway when using biofeedback have not been investigated. A better understanding of how the central nervous system uses artificial sensory information to reduce postural sway can be exploited to improve biofeedback systems. In this paper, we argue that sway reduction (in terms of center-of-pressure displacement and acceleration at trunk level) associated with audio biofeedback related to direction of postural sway is not the consequence of a simple, generalized mechanisms but rather the consequence of an increase in active, directionally-specific neural control of postural stability.

Materials & Methods

Eight healthy adults participated in this study (22-44 years old, 4 females and 4 males, age 33 ± 7 years, weight 71 ± 16 kg, and height 175 ± 11 cm). The subjects were divided into 2, 4-person groups and were gender- and age-matched between the groups. Subjects were excluded if they reported: a use of medications and/or a history of surgeries that may have affected their balance or their hearing, sensory loss, hearing deficits, and neurological disorders. The rights of the participants were protected according to the Declaration of Helsinki. Each subject signed an informed consent form in accordance with the OHSU Institutional Review Board regulations for human subjects.

The subjects were instructed to maintain balance while standing with eyes closed on a force plate (AMTI OR6-6) with feet 2-cm apart from each other (narrow stance). A prototype ABF system [5] was used to provide subjects with trunk acceleration information via earphones. The ABF system provided direction-specific information: either anterior-posterior (AP) or medial-lateral (ML) information about the subject's trunk movements. The AP and ML ABF were customized for each subject and for each trial by calculating the mean and the standard deviation (SD) of the subject's acceleration during the first 10 seconds of each trial [6].

While the subjects' acceleration at trunk level was inside a 2-SD range from their mean acceleration, which was calculated in the first 10 seconds of each trial, a 400-Hz, low-volume, pure tone was provided to the subjects in both earphones. As soon as they exceeded the 2-SD range, the stereo sound was modulated in pitch and volume in order to represent the subject's acceleration at trunk level, and the subjects were encouraged to adjust their sway in order to return within the 2-SD range. The AP information was encoded by modulating the pitch and the volume of the ABF sound. Specifically, when the subjects swayed forward (AP acceleration increased in the anterior direction), pitch and volume increased in both earphones whereas, when the subjects swayed backward (AP acceleration increased in the posterior direction), pitch decreased and volume increased in both earphones. The ML information was encoded by modulating the left/right balance of the stereo ABF sound. Also, the ABF sound became louder the more the subject leaned far from the vertical (ML acceleration increased). Thus, when the subjects swayed leftward (ML acceleration increased in left direction), the volume increased in the left earphone and decreased in the right one, and when they swayed rightward (ML acceleration increased in right direction), the volume increased in the right earphone and decreased in the left one. The equations used to create

the ABF sound are described in detail in [5].

Before the experiment, subjects were told how to use the ABF and practiced with the ABF system until they felt confident in performing the experiment. All subjects performed a total of 16, 1-minute trials: 4 with the AP ABF, 4 with the ML ABF, and 8 with no ABF. Trials alternated between those with and without ABF. The first group of subjects performed all AP ABF trials first; the second group performed all ML ABF trials first.

Center of pressure (COP) displacement in the AP and ML directions was calculated from the forces and torques sensed by the force plate. From the ABF system, acceleration sensed at the trunk along the AP and ML directions was also recorded. All data were acquired with a 100-Hz sample rate, using a NI-DAQcard 6024E and ABF custom-made software [5].

Trunk acceleration root mean square was post-processed for the AP direction and the ML direction, and for both directions combined (RMS_{AP} , RMS_{ML} , and RMS , respectively). Root mean square of the acceleration was intended as an indicator of the subject's sway area because it is highly correlated with the COP root mean square [5] (see figure 1A), which is traditionally used to quantify the stability of postural sway [7]. From COP data, the frequency comprising the 95% of the COP power spectrum [7] was post-processed for the AP and ML directions and for the two directions combined ($F95\%_{AP}$, $F95\%_{ML}$, and $F95\%$, respectively). These last parameters are computed as the frequency comprising the 95% of the power of the signal spectrum [7]. As a consequence, they are an approximation of the signal bandwidth. An increase of these parameters suggests the power is shifting toward higher frequencies. Under a physiological standpoint, this can be explained as an increase in the amount and intensity of postural corrections. The mean position of COP displacement was also calculated for each trial.

A 2-way, repeated measure, mixed, factorial ANOVA was performed on the data, with the group (first or second) being the between factor and the ABF mode (AP, ML, off) being the within factor. Bonferroni post-hoc tests were performed to discriminate the effects of the different ABF modes on the parameters extracted from the COP and the acceleration data. Paired T-test were used to verify if mean position of COP displacement changed while subjects used the ABF. The threshold for statistical significance was fixed at $p=0.05$.

Results

Direction-specific ABF reduced subjects' sway (in terms of center-of-pressure displacement and acceleration at trunk level) in the specific direction of the ABF by increasing the frequency of postural corrections in the direction of the ABF. For both AP ABF and ML ABF, sway decreased and postural corrections increased in the direction of the feedback twice as much as in the direction without feedback. Figure 1A shows raw AP data from COP and trunk acceleration from one representative subject in two conditions, without ABF (dark gray) and with AP ABF (light gray). The direction of ABF main factor was statistically significant for all the parameters ($p < 0.05$ for RMS_{ML} and $p < 0.01$ for all the other parameters). However, there was no statistical significance found for any parameters between the group that began with the AP ABF trials and the group that began with the ML ABF trials. In addition, there was no significant interaction found between group and ABF mode. Post-hoc analysis verified that both AP and ML ABF significantly reduced RMS and increased F95%. In addition, AP

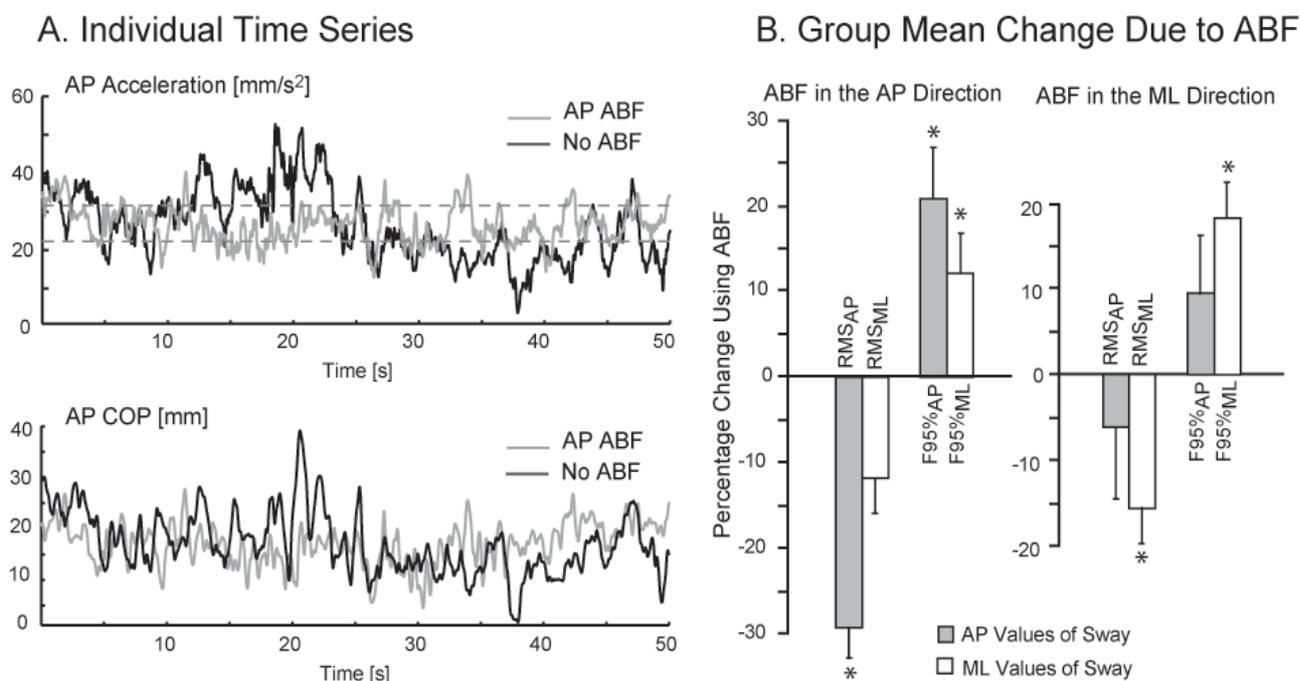


Figure 1 – Figure 1 – Panel A: Acceleration (top) and COP raw data (bottom) in the AP direction from a representative subject are illustrated. The light gray lines in panel A represent the subject's sway when using ABF; the dark gray lines represent the subject's sway when not using ABF. The threshold used for ABF was based on standard deviation and it is represented as a dashed light gray line in panel A (top). Panel B: percent changes from the condition without ABF of RMS_{AP} , RMS_{ML} , $F95\%_{AP}$, and $F95\%_{ML}$ while using ABF in the AP direction (left) and in the ML direction (right). Both AP and ML ABF reduced sway and increased the frequency of postural corrections in the specific direction of the ABF. (* indicates $p < 0.05$).

ABF significantly reduced RMS_{AP} but did not significantly influence RMS_{ML} . Similarly, ML ABF significantly reduced RMS_{ML} but did not significantly influence RMS_{AP} . Figure 1B shows the averaged effect of AP and ML ABF on RMS_{AP} and RMS_{ML} . AP ABF significantly increased $F95\%_{AP}$ more than $F95\%_{ML}$ for all but one subject. Furthermore, ML ABF significantly increased $F95\%_{ML}$ but did not significantly influence $F95\%_{AP}$. Figure 1B shows the averaged effect of AP and ML ABF on $F95\%_{AP}$ and $F95\%_{ML}$. Also, the mean position of COP displacement did not significantly change ($p>0.5$) when subjects used the ABF.

Discussion

This study shows that ABF providing direction-specific information about trunk acceleration with respect to gravity reduced subjects' sway in the specific direction provided by the ABF by increasing the frequency of postural corrections in that direction. In fact, for both AP and ML biofeedback, sway parameters were affected twice as much in the direction of ABF than in the orthogonal direction. AP ABF influenced all sway parameters more than ML ABF probably because AP sway has a larger range of motion and consequently, a larger tolerance for parameters that can change. In addition, the fact that $F_{95\%_{ML}}$ significantly increased with AP biofeedback may have been induced by a higher activity of the TIB muscles. In fact, a co-activation of TIB muscles to move the subject forward increases ML stiffness as a result of the not orthogonal force exerted by these muscles. This higher ML stiffness may have been reflected in our study by an increase of $F_{95\%_{ML}}$.

There are several different mechanisms by which ABF could have influenced postural sway but we favor a mechanism involving increased sensory feedback control [6]. It is unlikely that sway was reduced as a consequence of generic auditory stimulation because auditory stimulation unrelated to body sway has been found to increase, not decrease, postural sway [8-10]. It is also unlikely that the dual task required by attending to auditory cues while balancing was responsible for sway reduction, because cognitive tasks usually increase postural sway [11] and direction-specific sway reduction due to a secondary cognitive task has not been reported [3].

ABF may be able to reduce postural sway by generalized muscle co-contraction. A generalized co-contraction, according to an inverted pendulum model, would increase sway area and increase sway frequency in both AP and ML direction. In our previous study with AP and ML ABF while standing on foam, subjects reduced sway area and increased sway frequency without increasing muscular co-contraction [6].

Lengthening and activating the Tibialis Anterior muscle can result in decrease AP but not ML sway. Carpenter and Frank (2001) showed that subjects may decrease sway area and increase sway frequency in AP, but not ML direction, when faced with the threat of a fall from standing while facing the edge of a high support surface. Although ABF and the threat of falling similarly affected the standard deviation of COP displacement (-10% from fear and -5% with ABF) and increase the mean frequency of COP (+15% from fear and +18% from ABF), the mechanisms differ. With threat of a forward fall, the changes in AP postural sway appear to

be a consequence of a backward shift of the mean COP which increased the magnitude and the duration of activity in the Tibialis Anterior muscle. In our study, the mean COP position did not significantly change with ABF. In other studies, we also found this ABF device that ABF did not alter leg muscle activity or co-contraction, so increased muscle stiffness cannot explain sway reduction due to ABF [6].

Postural sway can be controlled with both feedback and predictive, feedforward mechanisms [12;13]. Using stabilogram diffusion analysis, we previously showed that the short-term, "closed loop" component was increased whereas, the long-term, "open loop" component was decreased by ABF [6]. Although ABF appears to reduce sway primarily via an increase in sensory feedback control in our studies, it is possible that with more practice, the biofeedback task may become more automatic so that subjects could rely more on feed forward control provided by the trunk acceleration signals [2;14;15].

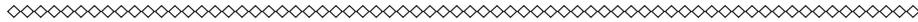
Direction-specific ABF was found to induce direction-specific reduction in postural sway by increasing the frequency of postural corrections. These results are consistent with an active integration of ABF with other sensory information by the nervous system to enhance postural control.

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Chapter 5

Audio-Biofeedback Improves Balance in Patients With Bilateral Vestibular Loss



Most of the content fo this chapter has been published in: M. Dozza, L. Chiari, and F. B. Horak, "Audio-biofeedback improves balance in patients with bilateral vestibular loss," Arch Phys Med Rehabil, vol. 86, no. 7, pp. 1401-1403, July2005.

Abstract

The extent to which subjects with loss of sensory information can substitute audio information to control body sway is unknown. We developed an audio-biofeedback (ABF) system to investigate its effect on postural stability during stance.

Audio biofeedback consisted of soundwaves representing 2D trunk cinematic (position, velocity and acceleration) information. When the subject sway was outside a 1° threshold, frequency and amplitude modulation signaled anterior-posterior trunk sway and left-right ear volume balance signaled left-right sway. Nine subjects with bilateral loss of vestibular function and nine age-matched control subjects attempted to use this biofeedback to minimize postural sway in stance with eyes closed and with foam under their feet.

Balance stability was evaluated according to the following parameters: the root mean square of (1) the center of pressure (COP) displacements and of (2) the trunk accelerations; the COP bandwidth; the time spent by the participant within $\pm 1^\circ$ threshold from their baseline COP position; and the mean accelerations of the trunk while the participant was swaying outside this $\pm 1^\circ$ threshold.

Participants with BVL had significantly larger postural sway than did unaffected participants. Those with BVL, while using ABF, decreased sway area by $23\% \pm 4.9\%$, decreased trunk accelerations by $46\% \pm 9.9\%$, and increased time spent within $\pm 1^\circ$ sway threshold by $195\% \pm 34.6\%$. In conclusion, ABF improved stance stability of participants with BVL by increasing the amount of postural corrections.

Introduction

The brain relies on the visual, somatosensory (proprioceptive, cutaneous), and vestibular systems to obtain reliable sensory information to control balance in stance [1]. The more accurate this sensory information, the better is postural stability [2]. When head linear- and angular-acceleration information are lost because of vestibular pathology, postural stability in stance is compromised, particularly in environments lacking adequate visual and surface somatosensory information [3]. Approximately 20% of the general population is affected by a vestibular disorder. Patients with vestibular disorders suffer from poor balance, spatial disorientation, and ataxia and they lack balance confidence, especially when other sensory references are limited [4]. Recently, new technologies have produced inexpensive, small sensors that transduce body-motion information normally provided by the human senses [5]. Such sensors have been used to provide vibrotactile information for improving balance in normal healthy individuals [6] and in people with vestibular loss [7]. This article reports on the effects of using a new prototype audio-biofeedback (ABF) system based on accelerometric sensors. This system uses the auditory input to provide sensory information, similar to that provided by the vestibular system, to people with bilateral vestibular loss (BVL). It was hypothesized that ABF sound coding of torso acceleration improves postural stability of people with BVL because this additional information, which is closely related to otolith information, may at least in part substitute for the lack of vestibular function that is the cause of balance deficits in people with BVL [8;9].

Methods

Participants

Nine individuals (4 men, 5 women) with severe BVL and 9 age- and sex-matched controls performed the experiment. Participants used as controls had no balance deficit or a history of surgeries that could affect their balance or hearing. People with BVL who had other pathologies or a history of surgeries that could affect their balance or hearing were also excluded from this study. All participants with BVL had bilaterally absent caloric responses and horizontal vestibular ocular reflex gains between .005 and .140 for rotations at .05Hz. Diagnosis for participants with BVL included 5 with gentomycin ototoxicity, 1 with Ramsey Hunt syndrome, 1 with autoimmune disorder, and 2 with idiopathic vestibular loss. Participants with BVL were referred to our lab by neuro-otologists. The mean age of participants with BVL was 55 years (range, 38–73yrs); mean height, 171cm (range, 160–193cm); and mean weight, 71kg (range, 51–115kg). The mean age of the control subjects was 55 years (range, 33–71yrs); mean height, 167cm (range, 151–180cm); and mean weight, 70kg (range, 65–86kg). All participants were protected according to the 1964 Declaration of Helsinki and signed an informed consent form before performing the experiments.

Procedures

All participants were instructed to stand with eyes closed and without footwear on an AMTI OR6-6 force-plate with medium-density, 4-inch Temper foam in 2 conditions: with and without ABF. The ABF system prototype (fig 1), which we developed, was equipped with a small (3x3x1.5cm) sensor that detected antero-posterior (AP) and medio-lateral (ML) linear accelerations at the trunk level when the sensor was applied to the torso near the body center of mass. A laptop computer acquired the signals from the sensor and generated a stereo sound encoding body-sway information. ML acceleration was encoded as the balance between the volume in the left and right channel whereas AP acceleration was encoded via both pitch and volume. As the participant leaned forward, the pitch increased, and as the participant leaned backward, the pitch decreased. The volume always increased as the participant leaned away from the vertical in all directions. For example, if participants swayed diagonally forward and to the left, they heard a sound increasing in pitch in both ears and becoming louder in the left ear and quieter in the right ear. The ABF system changed pitch or volume only when participants exceeded their baseline sway by $\pm 1^\circ$ [10]. A 1-minute training

phase was enough for all participants to understand the ABF representation of sway. Participants were instructed to use the biofeedback sound during trials, to correct their postural sway. Each participant performed three 1-minute trials with ABF and three 1-minute trials without ABF, in random order. A force-plate recorded the center of pressure (COP) displacement under the feet. COP is the imaginary point on the floor at which participants exert the net reaction force to control balance. To quantify postural stability, 5 parameters were calculated: (1) the root mean square (RMS; mm) and (2) the bandwidth (Hz) of the COP displacement [11], (3) the time spent within $\pm 1^\circ$ sway threshold (s), (4) the RMS of torso acceleration (mm/s^2), and (5) the mean torso acceleration outside the $\pm 1^\circ$ threshold (mm/s^2). Our hypothesis was that ABF would decrease the postural sway (RMS of COP and of torso acceleration in both AP and ML directions) and increase the time spent inside the $\pm 1^\circ$ sway threshold, especially for participants with BVL.

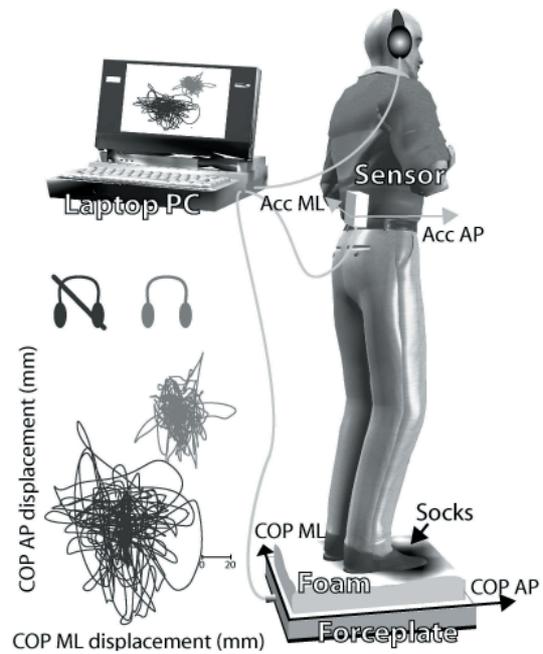


Figure 1 - Experimental setup. The center of pressure (COP) displacements illustrated are from 1 participant with BVL standing on the foam with eyes closed, with (light gray) and without (dark gray) ABF information available. The smaller the dimension of the COP displacement in the graphs, the smaller is the participant's sway. Consequently, the graphs show how, using ABF, people can reduce their postural sway. Abbreviation: Acc, acceleration.

Statistical Analysis

A 2-way, repeated-measure analysis of variance (ANOVA) was performed on each dependent variable to determine the effect of ABF, the difference in postural sway between subjects with BVL and controls, and the interaction between ABF and pathology. The criterion for statistical significance was p less than .05.

Results

Table 1 summarizes the results. Subjects with BVL had significantly larger postural sway while standing on foam than did controls, and both groups decreased postural sway using ABF in both AP and ML directions. Without ABF, 2 subjects with BVL were unable to complete the trials standing on the foam with eyes closed, even after many attempts. However, 1 of them was successful in all trials when using ABF. Two other participants with BVL were unable to remain standing during the first 2 attempts to perform a trial on the foam, eyes closed, without ABF. However, with ABF they were able to remain standing throughout all trials. For the participants with BVL who could stand on the foam with eyes closed, their COP displacements were 65% larger than those of the controls and their torso accelerations were 22.6% larger than those of controls without ABF. Time spent within the $\pm 1^\circ$ threshold did not differ statistically between the 2 groups without ABF. ABF significantly reduced postural sway, as reflected by reductions in both torso acceleration and COP displacement in both AP and ML directions. The significant interactions indicated that participants with BVL reduced COP displacement and acceleration with ABF significantly more than did controls. Using ABF, participants also increased the time spent inside a $\pm 1^\circ$ sway threshold and decreased their sway acceleration while outside this threshold.

Table 1 - Effects of Using ABF on COP and Acceleration Parameters

	Means of the parameters extracted during quiet stance on foam		Percentage changes in postural parameters with versus without ABF		Statistics, 2-way ANOVA		
	Control Mean (SD)	BVL Mean (SD)	Control % change (SEM)	BVL % change (SEM)	BVL Pathology	ABF	2 Factors Interaction
RMS	14.8 (3.9) [mm]	24.3 (8.7) [mm]	-15.9 (3.4)	-23.0 (4.9)	0.013	0.000	0.000
RMS AP	11.9 (2.7) [mm]	18.8 (7.5) [mm]	-15.4 (4.4)	-22.2 (4.4)	0.015	0.000	0.019
RMS ML	8.5 (3.4) [mm]	15.2 (4.8) [mm]	-15.0 (2.9)	-23.6 (6.1)	0.010	0.000	0.000
F95	1.58 (0.18) [Hz]	2.28 (0.81) [Hz]	9.3 (3.4)	8.4 (5.6)	0.000	0.060	0.830
RMS-Acc	65.2 (26) [mm/s ²]	115 (81) [mm/s ²]	-32.1 (10.3)	-46.2 (5.7)	0.002	0.000	0.005
RMS-Acc AP	54.6 (22) [mm/s ²]	100 (52) [mm/s ²]	-38.2 (10.9)	-49.8 (5.0)	0.042	0.000	0.002
RMS-Acc ML	32.9 (16) [mm/s ²]	52.6 (16) [mm/s ²]	-29.8 (10.8)	-35.5 (7.2)	0.060	0.001	0.030
Mean-Acc	40.9 (14) [mm/s ²]	69.9 (29) [mm/s ²]	-16.0 (5.9)	-25.3 (7.5)	0.001	0.053	0.387
Time-in -Thresh.	3.18 (2.82) [s]	1.91 (1.23) [s]	653.2 (336.9)	195.3 (20.1)	0.076	0.001	0.549

Discussion

These results indicate that ABF reduced postural sway and was more effective for subjects with BVL than for the control participants for most parameters during quiet stance on the foam. Thus, sound may substitute, at least partially, for the lack of vestibular sensory information to control postural sway in stance. Because participants significantly increased time spent within the $\pm 1^\circ$ [10] sway threshold using the ABF, our conclusion is that sway reduction was a consequence of additional postural control triggered by the audio information [12].

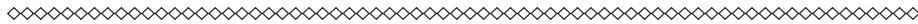
In conclusion, these results suggest that a biofeedback system, such as ABF, may help people with BVL improve balance when attempting to stand in environments with surface somatosensory and visual information inadequate for postural control. Also, this ABF device may be useful for balance training rehabilitation, as it has been found in other studies of postural biofeedback [13-15]. Future studies are needed to determine (1) whether people, after practicing with ABF, can use this additional information more automatically, without focused attention on feedback or postural control and (2) whether ABF is useful for stabilizing dynamic balance in tasks such as gait.

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Chapter 6

Auditory Biofeedback Substitutes for Loss of Sensory Information in Maintaining Stance



Most of the content of this chapter was published in: M. Dozza, F. Horak, and Chiari L., "Auditory Biofeedback Substitutes for Loss of Sensory Information in Maintaining Stance," Exp. Brain Res, 2006.

Abstract

The importance of sensory feedback for postural control in stance is evident from the balance improvements occurring when sensory information from the vestibular, somatosensory, and visual systems is available. However, the extent to which also audio-biofeedback (ABF) information can improve balance has not been determined. It is also unknown why additional artificial sensory feedback is more effective for some subjects than others and in some environmental contexts than others.

The aim of this study was to determine the relative effectiveness of an ABF system to reduce postural sway in stance in healthy control subjects and in subjects with bilateral vestibular loss, under conditions of reduced vestibular, visual, and somatosensory inputs. This ABF system used a threshold region and non-linear scaling parameters customized for each individual, to provide subjects with pitch and volume coding of their body sway.

ABF had the largest effect on reducing the body sway of the subjects with bilateral vestibular loss when the environment provided limited visual and somatosensory information; it had the smallest effect on reducing the sway of subjects with bilateral vestibular loss, when the environment provided full somatosensory information. The extent that all subjects substituted ABF information for their loss of sensory information was related to the extent that each subject was visually-dependent or somatosensory-dependent for their postural control.

Comparison of postural sway under a variety of sensory conditions suggests that patients with profound bilateral loss of vestibular function show larger than normal information redundancy among the remaining senses and ABF of trunk sway. The results support the hypothesis that the nervous system uses augmented sensory information differently depending both on the environment and on individual proclivities to rely on vestibular, somatosensory or visual information to control sway.

Introduction

The control of postural sway depends on continuous feedback of sensory information from the vestibular, somatosensory, and visual senses. The largest increase in postural sway in stance occurs when somatosensory information is compromised [1]. The next largest increase occurs when vestibular information is lost, and the smallest, when vision is eliminated by eye closure [2-4]. These increases in postural sway suggest that the central nervous system (CNS) relies primarily on somatosensory information, less so on vestibular information, and even less so on visual information to control postural sway during quiet stance. In fact, a linear sensory interaction model predicts such postural sway in adults during stance by proposing a 70% dependence on somatosensory information from a firm surface, 20% on vestibular information, and 10% on visual information [5]. However, several studies support the notion that the CNS re-weighs its relative dependence on sensory information when the availability of information from different senses changes [6-8]. For example, when healthy subjects stand on an oscillating surface with eyes closed, they increasingly depend on vestibular information and visual information and decrease dependence on somatosensory information from the surface as the amplitude of the surface rotations increases [5].

It is as yet unknown the extent to which the CNS reweighs its relative dependence on sensory information in presence of augmented sensory information. Augmentation of sensory information, such as auditory information, could be useful for rehabilitation of balance in patients with sensory loss, especially if the CNS proportionately integrates this information with the natural sensory information depending on the sensory demands of the task.

One type of augmentation to reduce postural sway—auditory information in the form of biofeedback—has received minimal investigation. When audio-biofeedback (ABF) was investigated, it was usually in conjunction with visual biofeedback [9;10]. In studies of ABF and visual biofeedback, the sound constituting the ABF was a simple alarm signal [11;12] that was used to augment the visual biofeedback. However, another type of ABF, able to represent a complex information and not limited to an alarm signal, may be especially useful to augment postural feedback since auditory cues: (1) are easy to integrate with the remaining senses in sensory-impaired individuals, such as those with vestibular losses [13], (2) do not interfere with visual information, and (3) are capable of signaling spatial information [14;15]. To illustrate this last point, humans use hearing for spatial localization whenever we turn our heads to locate the source of a sound. In addition, it has been shown that novice pilots can

learn how to fly in a flight simulator using either visual information or auditory tracking for turns, bank angles, and tilt [16], and it was subsequently determined that healthy subjects can use auditory information nearly as accurately as visual information to detect body orientation and motion in space [14].

Auditory and vestibular information are both transmitted to the brain via the VIII cranial nerve, which projects to the temporal lobe. Auditory cues automatically (subconsciously) influence postural alignment, and postural alignment automatically alters the ability to locate auditory cues in the environment [17;18]. Even stationary auditory cues were found to reduce the body sway of control and blind subjects when the cues were from stereo speakers in close proximity to both ears [19].

Recently, it has been found that subjects with a loss of vestibular information were able to use both ABF [20;21] and tactile biofeedback [22;23] that map their body movement in order to reduce postural sway. However, subjects with and without vestibular loss varied widely in their ability to reduce sway with augmented sensory ABF and vibrotactile biofeedback. The reasons for this inter-subject variability are unknown. However, similar inter-subject variability was also found when subjects with and without vestibular loss relied on their three natural sources of sensory information (visual, vestibular, and somatosensory) to control postural sway [7;24]. For example, 50% of subjects with neuromas on the VIII cranial nerve increased their postural sway in stance with eyes closed, but 50% decreased or did not change their sway with eyes closed [24]. After surgery to remove the neuroma, the same subjects, who were visually dependent (i.e., relied more on visual than on somatosensory information to maintain balance) before the surgery, no longer increased their sway with eyes closed, whereas those subjects who were not visually dependent increased their sway with eyes closed after surgery. Further, as people age or are exposed to weightlessness in space for a long time, many, but not all, increased their relative dependence on visual and somatosensory information to maintain balance [25;26]. Sensory compensation for pathological loss of sensory information has also been found to vary among subjects with profound bilateral loss of vestibular information (BVL, bilateral vestibular loss). Fifty percent of these subjects were able to significantly reduce body sway during surface oscillations by opening their eyes, whereas the other fifty percent could not [27]. Studying BVL subjects using a custom-made ABF, Hegeman [21] reported balance improvements when they stood with eyes open on firm surface but not on foam surface or with eyes closed. However, Hegeman [21] did not perform any analysis aimed at understanding how and why individual subjects were able or unable to use the ABF information to improve their stability in the different postural tasks.

In the study described here, we investigated how individual subjects' relative dependence on a particular sensory channel influenced their ability to reduce postural sway in stance when they used ABF information to control body sway. The objectives of this research were (1) to determine the extent to which ABF information helps control postural sway given

limited visual, vestibular, and surface somatosensory information and (2) to account for why the relative effectiveness of ABF varies among individuals across sensory environments. We used an ABF system, which we designed to mimic aspects of otolith vestibular information by monitoring accelerations in the transverse plane [28].

Methods

Participants

Nine subjects, four men and five women, with profound BVL and nine age- and gender-matched, healthy control subjects participated in this study. There were no significant age, height, and weight differences ($p > 0.05$) between the BVL and control subjects, respectively: age 55 years (38–73) versus 55 years (33–72); height 171 cm (160–193) versus 167 cm (151–180); and weight 71 kg (51–115) versus 73 kg (65–86). Table 1 summarizes the BVL subjects' pathologies, ages, duration of their vestibular loss, and their horizontal vestibulo-ocular reflex (VOR) gain at 0.05 Hz. Normal VOR gains range from 0.7 to 1 for the control subjects. All BVL subjects had a bilaterally absent response to warm and cold water on caloric tests and a VOR gain of less than 0.3 across a range of oscillations between 0.01 and 0.1 Hz, indicating severe loss of vestibular function [2]. In addition, each BVL subject fell without an apparent postural response soon after the start of surface sway-referencing trials with eyes closed, consistent with their BVL [1]. All of the BVL and control subjects were free of hearing, orthopedic, and neurological diseases or disorders, except the vestibular pathology for BVL subjects. Written informed consent was obtained from all subjects prior to their participation. The rights of the participants were protected according to the 1964-Declaration of Helsinki.

Apparatus

For all experiments, the BVL and the control subjects wore a custom-made ABF system [28] while standing on an AMTI OR6–6 force plate. The ABF system provided auditory information to the subjects about their body sway while they stood on the force plate.

The ABF system is comprised of three main parts: the sensory unit, the sensory processing unit, and the audio output unit [28]. The sensory unit consists of a small ($1.5 \times 3 \times 3 \text{ cm}^3$) sensor that is mounted on the subject's back at L5 with a Velcro belt. The sensory unit uses 3031 Eurosensor accelerometers (range $\pm 0\text{--}50$, resolution $2 \times 10^{-4} \text{ g}$, noise

Table 1 – Characteristics of BVL Subjects including the vestibulo-ocular reflex gain (VOR)

Subject ID	Age	Diagnosis	Duration of Loss	VOR
1	46	Ototoxicity	7	0.030
2	50	Idiopathic	12	0.006
3	56	Idiopathic	14	0.005
4	60	Ramsey Hunt	3	0.020
5	61	Ototoxicity	10	0.047
6	38	Auto Immune Disease	7	0.140
7	53	Ototoxicity	7	0.260
8	56	Ototoxicity	10	0.007
9	73	Ototoxicity	9	0.022

1 μV p-p, temperature error-zero $-0.05 \text{ mV}/^\circ\text{C}$) to sense the linear accelerations along the anterior–posterior (AP) and medial–lateral (ML) directions near the body center of mass. L5 was chosen because its position is minimally affected by movement artifacts, such as respiration, heartbeat, and voluntary head or limb movement. The processing unit consists of a laptop computer (Intel Celeron 2.4 GHz) equipped with an A/D board (DAQCard NI 6024E). It acquires, records, and processes the AP and ML accelerations sensed by the sensory unit and encode them into two analog sine waves that constitute the ABF stereo sound. The closed-loop delay introduced by the processing was estimated to be 5 ms. We developed the software for the processing unit using Matlab© 6 R12 and Matlab Data Acquisition Toolbox [28]. The audio output unit consists of an amplifier (Fostex PH- 5) that boosts the two sine waves provided by the computer so that the subjects are able to hear tones through the earphones (Philips SBC HP-140), with the tones representing the degree and direction of the body accelerations.

The force plate estimates body sway in the AP and ML directions by recording forces and torques under the subject's feet. In certain testing conditions, the force plate was covered with a 10 cm-thick, medium density Temper™ foam (indentation force deflection at 25%: 116 N, tensile strength: $125 \text{ kN}/\text{m}^2$, elongation: 109%, when temperature is 22.2°C and relative humidity is 50%) to reduce somatosensory information about body sway from the feet. When a subject stands on the foam, the distance between the subject's feet and the force plate continuously changes due to the compliance of the foam itself. As a consequence, the estimation of the center of pressure (COP) displacement was theoretically not as accurate as without foam. However, the error of estimation was calculated in post-process and found to be smaller than 10%. Linear accelerations from the sensory unit, as well as forces and torques from the force plate, were acquired with a 100-Hz sample rate.

Figure 1 shows, from a top-down perspective, four directions of sway and the relative ABF stereo sound changes in each earphone, for each direction. The ABF left–right balance and the volume in the earphones change according to ML body sway, and the pitch and volume of the stereo sounds change according to AP body sway [28]. In this study, all sounds were dynamically adjusted for each subject based on unique definitions of: (1) the region of natural sway [29], and (2) the area of the support base that is the region of a safe sway. Using an inverted-pendulum model [30], the region of natural sway and the region of safe sway were uniquely calculated for each subject. Specifically, the region of natural sway was determined by the range of AP and ML accelerations compatible with an oscillation of $\pm 1^\circ$ around the vertical, which depended upon the subject's height. The region of the safe sway was determined by the range of AP and ML accelerations compatible with the subject's COM projection on the ground, not exceeding the subject's base of foot support. Thus, the region of natural sway and region of safe sway were used to customize and to optimize the ABF tones for each subject.

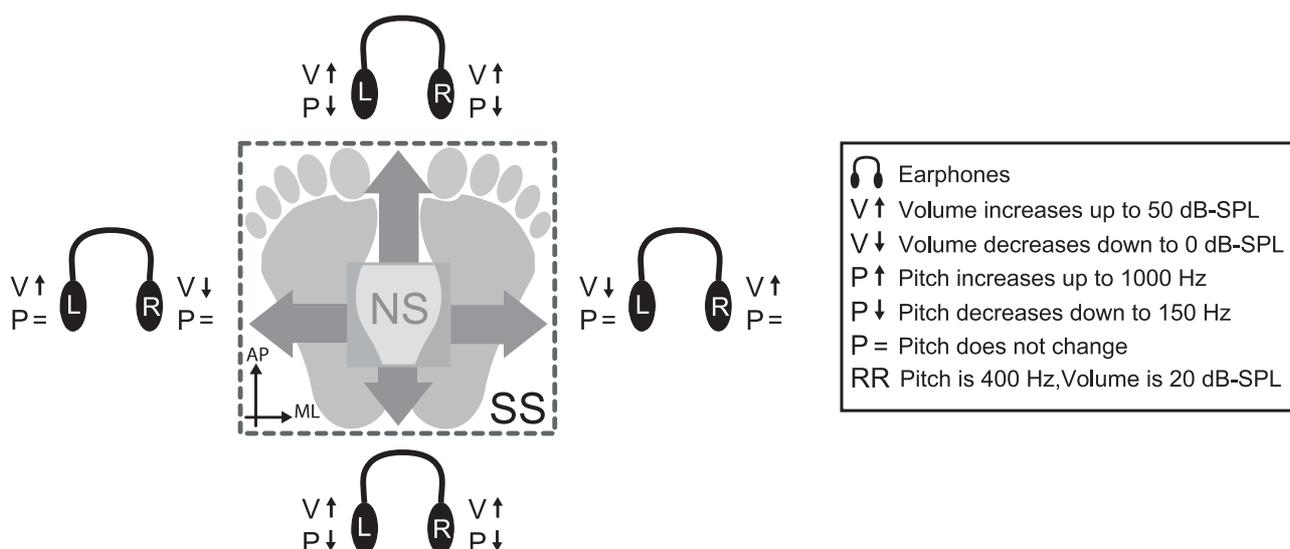


Figure 1 – ABF sound dynamics encoding postural sway. Pitch and volume change in the two earphones, depending on the direction of sway. The arrows in the middle of the force plate (outlined) indicate the direction of sway. The regions of natural sway (NS) and safe sway (SS) were customized for each subject.

The ABF system was designed so that the tones changed, depending on the subject's sway relative to the calculated region of natural sway. When a subject swayed within his or her calculated region of natural sway in the ML and AP directions, the same constant, low-volume (20 dB-SPL), 400-Hz tone was fed back to the subject through each earphone. However, when a subject swayed outside his or her region of natural sway in the ML direction, the tones in the earphones simultaneously became louder in the ear corresponding to the direction of body sway and quieter in the other ear. When the subject swayed outside the region of natural sway in the anterior direction, the tones changed equally in both ears and became louder (up to 50 dB-SPL) in volume and higher in pitch (following a linear function up to 1000 Hz). When the subject swayed outside the region of natural sway in the posterior direction, the tones changed equally in both ears and became louder in volume and lower in pitch (following a linear function down to 150 Hz). When the subject swayed outside the region of natural sway in an oblique direction, for example in the anterior-left direction, the tones became higher in pitch in both ears, louder in volume in the left ear, and quieter in volume in the right ear. All the equations used to generate the ABF sound using sigmoidal function are reported in detail in Chiari et al. [28].

Procedure

Subjects stood on the force plate and kept their feet 15° externally rotated and their heels 1 cm apart (narrow stance position). They were instructed to maintain quiet stance throughout all testing when using and not using the ABF device. Before the experimental protocol began, subjects practiced with the ABF system for a few minutes on a firm surface with eyes open by voluntarily swaying at different angles and directions, and listening to the corresponding changes in tones in the earphones until they understood how the trunk

information was coded into the ABF sound. The subjects were instructed to correct their body sway by using the tones, i.e., to maintain their sway within the region of natural sway by achieving a constant 400-Hz tone in each earphone. Once they understood how to change their body sway to achieve the constant 400-Hz tone, they performed three practice trials with eyes closed and without ABF, followed by three practice trials with eyes open on foam and without ABF. The purpose of the practice trials was for the subjects to gain confidence in standing with eyes closed or standing on the foam-covered force plate without falling, and to minimize the initial effects of standing on the foam. Data from these practice exercises and trials were not considered in the analyses.

BVL subjects repeated a block of six conditions three times (18 trials total), and the control subjects repeated the same block of six conditions five times (30 trials total). For each of these blocks, the six conditions were presented in random order; three conditions were with and three conditions were without ABF. Conditions one and two were: eyes closed on a firm surface without ABF and with ABF. Conditions three and four were: eyes open on foam surface without ABF and with ABF. Conditions five and six were: eyes closed on foam surface without ABF and with ABF. We did not test the eyes-open on firm-surface condition since, in this condition, the sway of both the BVL and the control subjects is expected to be inside the region of natural sway, so there is no need for additional ABF information [1]. The BVL subjects performed fewer trials to limit fatigue. Each trial lasted 1 min.

Data and statistical analysis

From the 2D, planar COP displacement, we quantified postural sway with two independent parameters [31-33]: the root-mean-square distance (COP-RMS) and the frequency below which the 95% of the power of the signal is included (F95%). From the 2D, planar acceleration measured by the sensory unit, we computed the RMS (Acc-RMS). To determine the effect on sway of subject groups, conditions, and ABF, we performed a three-way ANOVA, 2 groups (BVL and control) x 3 sensory conditions (vestibular, somatosensory, and visual), repeated (eyes closed, eyes open on foam, and eyes closed on foam) x 2 ABF conditions, repeated (ABF on and off) for each parameter (COP-RMS, F95%, and Acc-RMS). The threshold for statistical significance was $p = 0.05$.

To evaluate the correlation between severity of vestibular loss and the effect of ABF on sway amplitude in the eyes closed on foam condition, a robust regression correlation analysis was performed between the VOR gain and the percentage reduction in COP-RMS, with and without ABF for BVL subjects. To assess whether ABF was effective in helping subjects reduce body sway in proportion to each subject's level of dependency on visual and somatosensory information, a robust regression correlation analysis was performed between the levels of sensory dependency and the effect of ABF on COP-RMS when only visual (eyes open on foam

with ABF condition) or only somatosensory information (eyes closed with ABF condition) was available. The levels of visual dependency and somatosensory dependency were estimated for each subject as the percentage of the body sway reduction occurring when visual or somatosensory information was added (visual information, in the eyes open on foam condition and somatosensory information, in the eyes closed condition) and were compared to the reference eyes closed on foam condition (when neither visual and somatosensory information was available).

Results

Center of pressure displacement

For BVL and control subjects, body sway increased as natural sensory information or ABF information became absent or unreliable. Further, COP-RMS was significantly larger in the eyes open on foam condition than in the eyes closed condition ($p < 0.05$). COP-RMS was also significantly larger when eyes were closed than when eyes were open while subjects stood on foam without ABF ($p < 0.01$). In the eyes closed, eyes open on foam, and eyes closed on foam conditions, BVL subjects' COP-RMS was significantly larger than the control subjects' COP-RMS ($p < 0.001$). Figure 2 shows the anterior–posterior versus lateral COP displacements of one representative BVL subject (Fig. 2a) and one representative control subject (Fig. 2b), in all six conditions. Table 2 reports the COP-RMS values in the eyes closed, eyes open on foam, and eyes closed on foam conditions for both subject groups.

In the three ABF conditions, both groups benefited from ABF. That is, ABF significantly decreased COP-RMS for both the BVL and control groups ($p < 0.05$). The percentage of changes in COP-RMS due to ABF is shown in Table 3. No significant interaction was found between the groups and the conditions tested since COP-RMS was larger in BVL subjects than in control subjects in every condition. In addition, there was no significant interaction between the groups and ABF as both groups improved in the conditions tested. A significant interaction was found between the condition factor

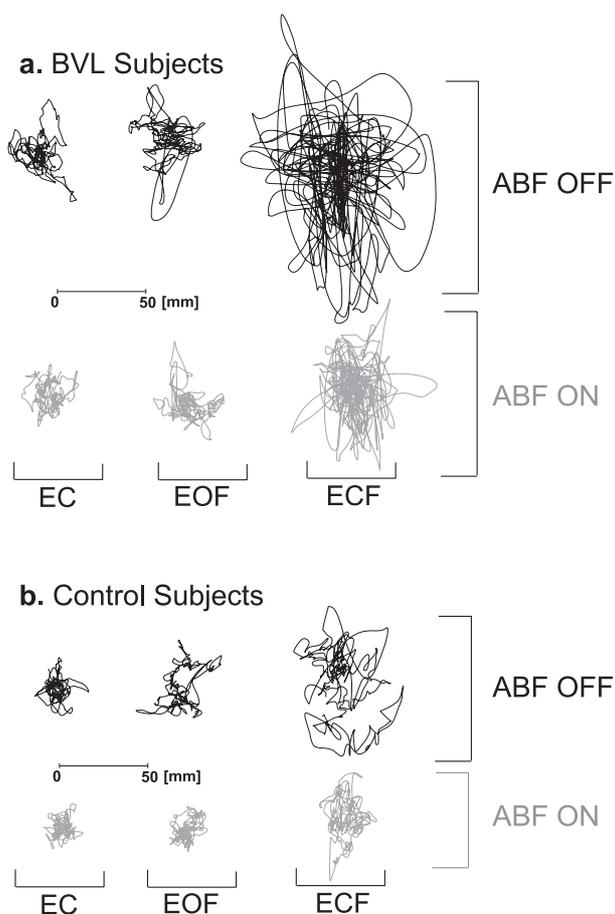


Figure 2 – ABF sound dynamics encoding postural sway. Pitch and volume change in the two earphones, depending on the direction of sway. The arrows in the middle of the force plate (outlined) indicate the direction of sway. The regions of natural sway (NS) and safe sway (SS) were customized for each subject.

Table 2 – Mean values and standard deviation (in parenthesis) of postural parameters for bilateral vestibular loss (BVL) and control subjects in the three conditions tested without audio-biofeedback (ABF). Root mean square distance (RMS) is reported for the center of pressure displacement (COP) and for the acceleration sensed at trunk level (Acc). Also, the values of frequency, below which the 95% of the power of the COP signal is included, are reported.

Parameter	Eyes closed		Eyes open on foam		Eyes closed on foam	
	BVL	Control	BVL	Control	BVL	Control
COP-RMS (mm)	13.82 (8.9)	8.31 (2.8)	14.01 (9.7)	9.34 (1.2)	24.66 (7.58)	14.92 (3.7)
F95% (Hz)	1.85 (0.55)	1.31 (0.15)	1.87 (0.52)	1.39 (0.19)	2.51 (0.31)	1.59 (0.18)
Acc-RMS (mm/s ²)	14.12 (8.07)	12.61 (2.4)	16.79 (9.50)	13.48 (1.9)	56.09 (19.13)	21.84 (5.60)

and the ABF factor ($p < 0.001$) due to ABF decreasing COP-RMS more in the eyes closed on foam condition than in the eyes closed or eyes open on foam condition (Table 3). For the BVL subjects in the eyes closed on foam condition, a significant interaction was found among all three ANOVA factors ($p < 0.001$) due to ABF decreasing COP-RMS the most in the eyes closed on foam condition for all BVL subjects.

Figure 3 shows the average COP-RMS reduction when BVL and control subjects used ABF on foam with eyes closed. As shown in Fig. 3, all but one of the BVL subjects able to perform the eyes closed on foam condition benefited from ABF in this condition. In addition, BVL subject #2 fell a few times in the eyes closed on foam condition, but she never fell in this condition while using ABF. BVL subject #1 fell consistently in the eyes closed on foam condition but also never fell in this condition while using ABF. BVL subject #8 benefited from ABF, although minimally when compared to the other BVL subjects. BVL subject #5 (Fig. 3) was not able to stand in the eyes closed on foam condition, with or without ABF, although he benefited from ABF in the other conditions (eyes closed and eyes open on foam). Also as shown in Fig. 3, all control subjects benefited from ABF in the eyes closed on foam condition.

Frequency spectrum

For BVL and control subjects,

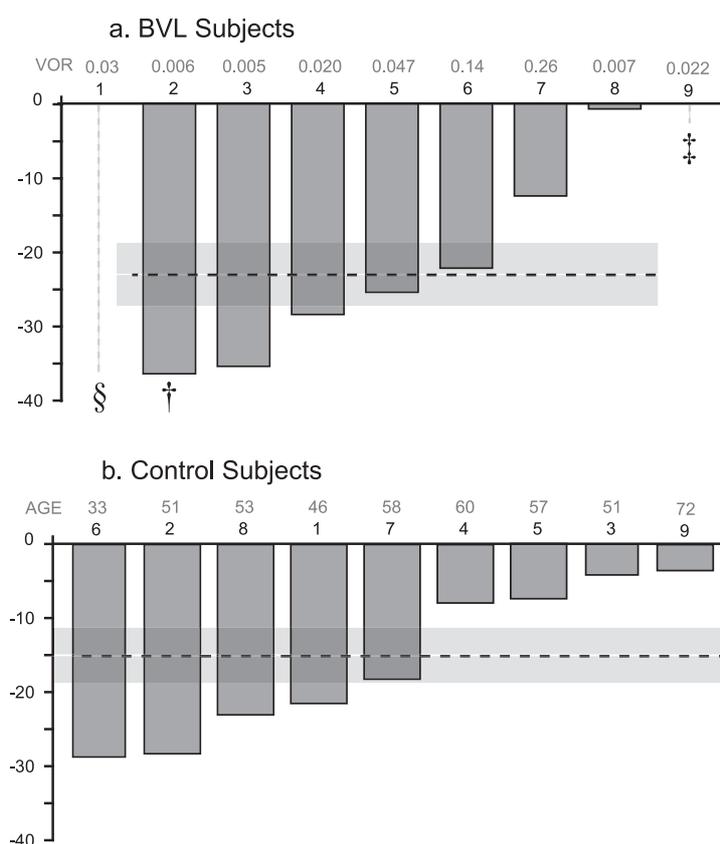


Figure 3 – The percentage of COP-RMS reduction using ABF is reported for each bilateral vestibular loss (a) and control (b) subject in the condition eyes closed on foam. Data were ordered by percentage improvement using ABF. Subject numbers indicate matching subjects between the groups. † BVL Subject 2 fell twice without ABF but never fell during trials using ABF. ‡ BVL Subject 9 fell repeatedly with and without ABF. § BVL Subject 1 could stand only with the help of ABF. Black, dashed lines represent the mean reduction using ABF. Gray, shadowed areas represent the standard error of the reduction using ABF.

Table 3 – Mean percentage difference of each postural parameter with and without audio-biofeedback (ABF) for bilateral vestibular loss (BVL) and control subjects. Root mean square distance (RMS) is reported for the center of pressure displacement (COP) and for the acceleration sensed at trunk level (Acc). Also, the values of frequency, below which the 95% of the power of the COP signal is included, are reported.

Parameter	Eyes closed		Eyes open on foam		Eyes closed on foam	
	BVL	Control	BVL	Control	BVL	Control
COP-RMS (mm)	-3.24	-10.87	-9.98	-5.42	-23.07	-15.90
F95% (Hz)	21.90	23.01	10.54	18.89	8.38	9.28
Acc-RMS (mm/s ²)	-20.82	-35.24	-27.38	-40.56	-46.18	-32.15

the amount of postural corrections (indicated by the parameter F95%) decreased as natural sensory information became available or reliable and increased when ABF information was available. Specifically, the frequency spectrum components of the COP were significantly affected by the different test conditions, with the power at the higher frequencies increasing when visual and/or somatosensory sensory information was reduced ($p < 0.001$). F95% was higher in the eyes closed on foam condition than in the eyes open on foam condition ($p < 0.05$), and higher in the eyes open on foam condition than in the eyes closed condition ($p < 0.05$). F95% was also higher for the BVL subject group than for the control group in all conditions ($p < 0.001$). Table 2 reports F95% values in the three conditions tested without ABF for the BVL and control subjects. The use of ABF significantly increased F95% for both the BVL and control subjects in all conditions ($p < 0.001$). Table 3 shows the percent of increase in F95% when controls and BVL subjects used ABF in each condition. There was a significant interaction ($p < 0.05$) between the condition tested and the presence of a vestibular deficit, with F95% increasing in the BVL subject group more than in the control group, particularly in the eyes closed on foam condition.

Sensory substitution

Subjects benefited from ABF information in relation to the lack of natural sensory information. For most BVL subjects, the extent that they reduced their body sway with ABF in the eyes closed on foam condition correlated with the extent of their vestibular loss ($r = 0.76$; $p < 0.05$). Table 1 shows the VOR gains and percentage of improvement in sway for all of the subjects using ABF. One subject with very low VOR gain (#9) could only stand with the ABF in this condition so the percentage of improvement could not be calculated.

For both the BVL and control groups, the effectiveness of ABF in reducing body sway was related to how dependent each subject was on visual or somatosensory information, but not on the amount of sway in the baseline eyes closed on foam condition. Somatosensory-dependent subjects benefited the most from ABF when somatosensory information was missing, and vision-dependent subjects benefited the most from ABF when visual information was missing. Figure 4 shows the relative dependence of each subject on visual or somatosensory information versus the amount of benefit that each received from ABF under conditions in which visual or somatosensory information was limited (i.e., the eyes closed

and eyes open on foam conditions). A linear relationship for the BVL subjects and the control subjects was found between the degree of benefit from ABF and their dependence on visual and somatosensory information, shown by the greater number of circles in the top-right and bottom-left quadrants of Fig. 4. The circles in the top-right quadrant represent the subjects who were somatosensory-dependent and benefited the most from ABF when somatosensory information was missing. The circles in the bottom-left quadrant represent subjects who were vision-dependent and benefited the most from ABF when visual information was missing.

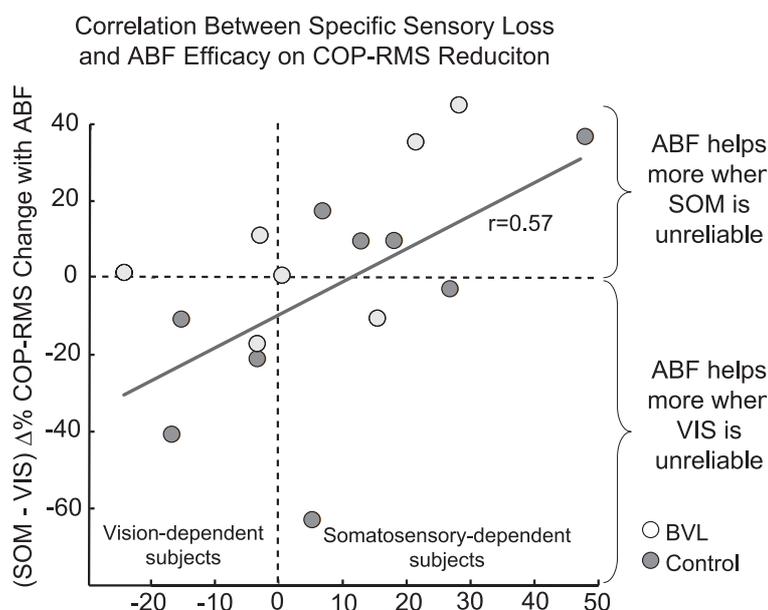


Figure 4 – Subjects in terms of their vision and somatosensory dependency. There is a correlation between the use of ABF in the eyes closed and eyes open on foam conditions, and visual and somatosensory dependency. Each subject's tendency to rely, more on vision or somatosensory information is reported on the horizontal axis. Negative values imply a dependency on vision more than on somatosensory information, whereas positive values imply a dependency on somatosensory more than on vision information (a zero value on the horizontal axis indicates a subject who relies on vision as much as on somatosensory information to maintain balance in stance). The vertical axis shows the effect of ABF for each subject. Positive values imply ABF reduces sway more when somatosensory information is made unreliable by standing on foam, negative values imply ABF reduces sway more when visual information is missing (a zero value on the vertical axis indicates a subject who, when using ABF, reduces sway when vision information is limited as much as when somatosensory information is inadequate). The Pearson coefficient for the regression line is $r=0.57$ comprising data from both group and is statistically significant ($p<0.05$). The Pearson coefficients reported in the figure for the two groups of subjects separated ($r=0.62$ and $r=0.65$ for bilateral vestibular loss and control subjects, respectively) are not statistically significant ($p>0.05$), however they are close to statistical significance $p=0.06$

Discussion

ABF efficacy in reducing sway is related to the availability of sensory information

Results from this study show that the amount that ABF compensates for missing sensory information depends on the extent of sensory loss. When somatosensory information was reduced (the eyes open on foam condition) and the more that BVL and control subjects were somatosensory-dependent, the more they benefited from ABF and were able to reduce their sway. When visual information was not available (the eyes closed condition) and the more that BVL and control subjects were visually dependent, the more they also benefited from ABF and were able to reduce their sway. When both somatosensory information and visual information were limited (the eyes closed on foam condition), both BVL and control groups showed the most benefit from ABF. Thus, we hypothesize that the degree to which subjects benefit from ABF to reduce postural sway depends on their degree of visual, somatosensory and vestibular loss [24;34]. Our results also showed a trend in which the more severe the vestibular loss, the more subjects benefited from ABF. This trend needs further testing with more subjects in order to show statistical significance. Our findings are consistent with other studies that also reported that control and BVL subjects were able to reduce postural sway with visual, tactile, and audio-biofeedback [21;35]. However, our study, for the first time, has identified a potential relationship between benefits from ABF information and dependency on sensory information.

Both BVL and control subjects' postural sway increased when sensory information was limited, confirming the commonly held hypothesis that the control of postural sway depends on the amount of available sensory feedback that is available [5;36;37].

Our BVL subjects showed significantly larger sway than did our control subjects in all conditions tested, in agreement with other studies [7;38;39]. However, the BVL subjects' degree of sway reduction via ABF when either visual information or somatosensory information was available was not related to the extent of their vestibular loss. This finding may be due to the subjects' hesitance to rely on novel sensory information (available via ABF) when ordinary sensory information normally and extensively used to compensate the loss of vestibular information [24] was also available. However, this finding may also be explained by the ABF information not yet being integrated with the subjects' existing somatosensory and visual information since they used ABF for only 15 min or less during testing. This lack of integration

is also supported by another study in which we found that the use of ABF requires a larger number of rapid postural corrections [40]. Lack of integration may be the consequence of the subjects' paying excessive attention to the ABF, thus interfering with the attention paid to other sensory information. It has been shown how dual-task interference decreases with practice over time when tasks become quasi-automatic [41]. Consequently, it may be possible for ABF information to become more integrated with other sensory information as when ABF is used after a longer period of time than just the few minutes in our study [42].

Attention to natural sensory information may have limited ABF efficacy in BVL subjects

Although BVL subjects reduced their sway more than the control subjects did in the eyes closed on foam condition, they did not in the eyes open or in the firm surface conditions. In contrast, Hegeman et al. [21] found that BVL subjects reduced sway in stance using ABF only with eyes open on a firm surface, but not with eyes closed and/or when on foam. This different effect of ABF may be related to differences in: (1) the design of the ABF systems, (2) the use of trunk angular velocity instead of linear acceleration that was fed back to the subjects, (3) the linear algorithm chosen to map trunk movement into sound, (4) subject selection, and (5) how postural sway was measured and quantified. In our study, the high degree of attention that BVL subjects normally pay to visual and somatosensory information in the eyes closed and eyes open on foam conditions may have limited their ability to use ABF since the initial use of ABF requires some a degree of attention to the tones in the earphones [40]. Indeed, during the rehabilitation period of BVL subjects, they are taught to pay more voluntary attention to visual and somatosensory information than would be the case if they did not have the BVL, to compensate for the vestibular loss [43;44]. Consequently, focusing more on visual information and somatosensory information available in the eyes open on foam and eyes closed conditions, may have interfered with their ability to concentrate on the ABF [45;46]. However, in the eyes closed on foam condition, when visual information and somatosensory information were limited, subjects could focus their attention on the ABF. Another explanation for subjects' decreasing their sway with ABF is that their use of ABF and the headphone equipment influenced them to pay more attention to their sway. However, in studies in which subjects were instructed to deliberately focus their attention on their body sway and to increase their control of posture, they did not reduce their sway [47]. Thus, we believe that the large sway reduction induced by ABF in BVL subjects was not likely only due to the subjects' paying more attention to their sway.

Use of ABF reduced BVL subjects' inter-subject Variability

We found a high inter-subject variability among BVL subjects for all the parameters analyzed, which agrees with findings from many other studies [38;39;48]. Indeed, two of the nine subjects did not benefit from ABF in the eyes closed condition. Some of this variability

may be explained in terms of how individual BVL subjects compensate for the vestibular loss, which is by increasing reliance on either visual or somatosensory information [24;49]. If inter-subject variability depends on the degree of visual or somatosensory dependency, we may expect inter-subject variability to decrease when visual information and somatosensory information are limited (the eyes closed on foam condition). Indeed, we found a consistent decrease in inter-subject variability in this condition, when BVL subjects exhibited relatively smaller standard deviations (Table 2), although their sway was larger than in the eyes open on foam and eyes closed conditions [50]. Our BVL subjects showed significantly higher frequency of postural corrections (F95%) than did our control subjects in all conditions tested. This result suggests that BVL subjects were using a different mode of controlling their balance than were the control subjects [51]. However, without kinematic measures, we cannot distinguish between ankle and hip sway strategies, as it was done by Creath et al. [51]. The higher frequency of postural corrections that the BVL subjects exhibited may also be related to the higher sensory noise due to the vestibular loss that BVL have compared to control subjects.

ABF redundancy with sensory information was higher for BVL than for control subjects

In order to better highlight the difference in the use of ABF information between BVL and control subjects, we performed a meta-analysis which combined the results from BVL and control subjects in all the condition presented in this study in terms of sensory information redundancy using Venn diagrams. Redundancy of sensory information occurs when the same information is provided by more than one sensory channel. Sensory integration for balance is driven by—that is, is dependent on—redundancy of natural sensory information from somatosensory, visual, and vestibular channels [52]. Extensive redundancy of sensory information provides persons with a better estimate of body segment position and kinematics, which results in smaller postural sway [53;54].

To quantify sensory redundancy among the natural sensory information and ABF, we averaged the sway reduction occurred in the conditions tested (when natural and ABF sensory information was available) and represented these averages using Venn diagrams. Figure 5 shows two Venn diagrams (one for the BVL subjects and one for the control subjects) that represent the contributions when all or some of the sensory information channels were contributing sensory information to control sway. The size of each diagram and their percentages represent the percent of COP sway reduction occurred from a condition in which ABF, somatosensory, and visual information are all limited (by turning off the ABF device, by using foam, by closing the eyes, respectively; i.e., the eyes closed on foam condition without ABF) and a condition when only one of these information is available.

The redundancy between the ABF contribution in reducing sway and the contribution from each of the other sensory information was larger for BVL subjects (Fig. 5a) than for control subjects (Fig. 5b). For BVL subjects, ABF reduced sway 46% (4, 11, and 31%)

compared to 32% (12, 7, and 13%) for control subjects (each of the three percentages in parenthesis is the amount of redundancy between ABF information and visual, somatosensory, and both visual and somatosensory, respectively). From these analyses, for BVL subjects, the redundancy among somatosensory, visual, and ABF information was higher (31%) than for control subjects (13%). The greater redundancy in BVL than control subjects suggests that compensating for vestibular loss depends on more extensive sensory redundancy between visual and somatosensory information. Figure 5 shows that ABF information can also be redundant with visual and somatosensory sensory information, suggesting that the CNS may treat ABF information similarly to natural sensory information. Also, since redundancy between ABF information and other sensory information is greater for BVL subjects than for control subjects, BVL subjects may benefit more from the ABF information than may control subjects, especially in sensory-deprived situations. In fact, with more practice, ABF information may also facilitate a more accurate integration and calibration of sensory information, induced by the CNS continually comparing natural sensory information to ABF information.

The use of foam to limit somatosensory information may have limited in the accuracy of sensory redundancy estimation. In fact, when determining the role that the somatosensory information plays in reducing sway (Fig. 5), we did not include all somatosensory information that the CNS received from the entire body but only the somatosensory information from the subject's feet which was restricted by using the foam. Even with these qualifications, Fig. 5 provides new insight into the mechanisms of sensory redundancy and sensory re-weighting during human stance. In conclusion, we found that the BVL and the control subjects used ABF information about their trunk acceleration to control sway, in proportion to the extent that their other sensory information was reduced. In addition, all subjects used ABF differently, depending on their individual proclivities to rely on vestibular, somatosensory, or visual information in order to control sway. Redundancy between sensory information from different sensory channels and ABF information was larger in BVL subjects than in control subjects, suggesting that ABF information may help subjects compensate for vestibular loss by facilitating the CNS's integration of sensory information.

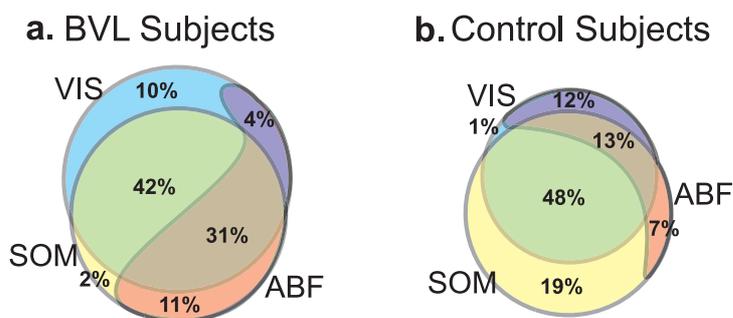


Figure 5 – Subjects in terms of their vision and somatosensory dependency. There is a correlation between the use of ABF in Fig. 5 a, b. In the form of Venn diagrams the contributions of somatosensory (SOM yellow/lighter-colored circle), visual (VIS blue/darker-colored circle), and (ABF orange/dark-gray diagram) information in reducing COP-RMS during quiet stance for bilateral vestibular loss and control subjects, respectively. Percentages indicate the size of the different areas and represent the COP-RMS reduction experienced by the subjects when that information was available. Overlapping areas represent redundancy of information across the sensory systems.

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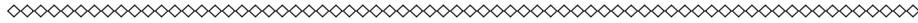
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Chapter 7

Effects of Linear versus Sigmoid Coding of Visual or Audio Biofeedback for the Control of Upright Stance



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Abstract

Although both visual and audio biofeedback (BF) systems for postural control can reduce sway during stance, a direct comparison between the two systems has never been done. Further, comparing different coding designs of audio and visual BF may help in elucidating how BF information is integrated in the control of posture, and may improve knowledge for the design of innovative BF systems for postural control.

The purpose of this paper is to compare the effects of linear versus sigmoid coding of trunk acceleration for audio and visual BF on postural sway in a group of eight, healthy subjects while standing on a foam surface.

Results showed that sigmoid-coded audio BF reduced sway acceleration more than did a linear-coded audio BF, whereas a linear-coded visual BF reduced sway acceleration more than a sigmoid-coded visual BF. In addition, audio BF had larger effects on reducing center of pressure (COP) displacement whereas visual BF had larger effects on reducing trunk sway. These results suggest that audio and visual BF for postural control benefit from different types of sensory coding and each type of BF may encourage a different type of postural sway strategy.

Introduction

Biofeedback (BF) systems for postural control are aimed at providing sensory information to supplement the natural sensory information to improve human balance [1]. Experimentation with visual BF for postural control has been in progress since the 1970s [2] and, traditionally, involved the visualization of subjects' center of pressure (COP) displacement on a monitor placed in front of the subjects. Using visual BF, subjects see the movement of their COP displacement on the computer monitor and use this information to decrease their postural sway [3]. A few studies also reported how repetitive use of visual BF may be a valid rehabilitation or training tool for subjects with neuropathy [4], stroke subjects [5] and healthy elderly subjects [6;7]. However, it is still uncertain whether training with biofeedback has a carry-over effect without biofeedback [8].

Audio BF has received much less attention than visual BF. This lack of attention to audio BF is probably due to its relative design complexity. Whereas, visual BF could be actualized with a standard oscilloscope connected to a force plate, audio BF requires customized computer algorithms for their coding. In the last few years, interest in audio BF for postural control has been renewed [9;10], partially due to advances in technology for real-time processing and movement sensing and to new trends in wireless portable devices that can be worn during daily activities. These new BF devices are not meant to be used only in a laboratory setting, and offer more advantages in terms of costs and portability than visual BF devices [9-12].

It is difficult to evaluate the relative merits of the different types of BF because each one of these, new BF system has a unique, complex design. Specifically, each one uses a different movement sensor which assesses a different aspect of the subjects' sway. Further, this movement information is then fed back to the subjects by using a different coding algorithm and through a different sensory modality; Figure 1. These substantial differences in the design make it nearly impossible to determine which different variables in the design is responsible for different results obtained with the different BF systems; even if these results were obtained from similar protocols.

The strategies that subjects use to alter postural sway with different types of BF are also unknown. Postural sway can be reduced via a number of different strategies, including 1) a general stiffening via muscle co-contraction, 2) moving the body about the ankle joints with little motion at the knees or hips (ankle strategy; [13]), or 3) moving the body about many joints (multisegmental strategy such as the hip strategy; [13]). Nashner and colleagues

hypothesized that vestibular or visual inputs favored a top-down hip strategy, whereas somatosensory inputs favored an inverted pendulum-like ankle strategy [14] so visual and auditory BF may favor different ways of reducing postural sway. A general stiffening from co-contraction of muscles around joints due to fear of falling has been shown to reduce COP displacement, but it is thought to be an undesirable way of reducing postural sway because it doesn't improve the ability to respond quickly to external perturbations [15],

Thus, the limited knowledge to date on the effect of BF on postural control does not allow us to determine: (1) which is the optimal algorithm to code body motion into a sensory signal for reducing postural sway, (2) whether different postural control strategies are favored by different designs of visual or audio BF, and (3) whether and when visual or audio BF is the more effective in controlling sway. This study starts to address these questions for the first time, by comparing the effects on COP displacement, trunk acceleration, and muscular activity of two designs of visual and audio BF. The results shown in this paper provide evidence that (1) different types of coding may be optimal for visual and audio BF, and (2) visual and audio BF may favor different postural strategies for the control of upright stance.

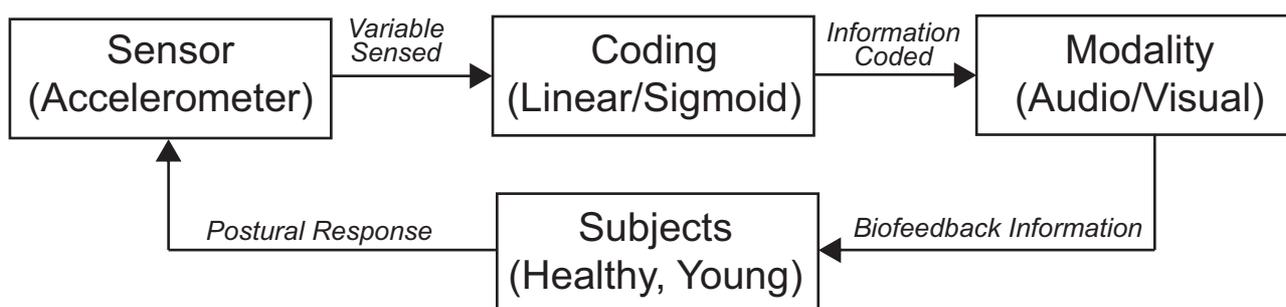


Figure 1 – Box diagram representing the loop design of a biofeedback system for postural control and its application. The features of the biofeedback system used in this study are reported in parenthesis for each box

Methods

Participants

Eight, healthy, young adults (6 men and 2 women) participated in this study after providing informed consent. Average and standard deviation of age, height, and weight of the participants were, respectively, 23 ± 3 yrs, 173 ± 7 cm, and 62.5 ± 12.5 kg. All participants indicated that they had no known neurological, orthopedic, hearing, or balance disorders. None needed prescription glasses. The experimental protocol was approved by the OHSU Ethics Committee and followed the recommendations of the Declaration of Helsinki for Human Experimentation.

Apparatus

Subjects were asked to stand on an AMTI OR6-6 force plate that was covered with a 10cm-thick Temper™ foam (Indentation Force Deflection at 25%: 116N, Tensile Strength: 125 kN/m², Elongation: 109%, when temperature is 72F and relative humidity is 50%) while wearing the BF movement sensor. The foam was used to alter the somatosensory information from the bottom of the feet and its usefulness for maintaining balance. An electromyographic (EMG) custom-made device recorded leg muscles activity. The EMG signals from the electrodes were amplified 20000 times, band-pass filtered (71-2650 Hz), full-wave rectified, and integrated with a 6th order Butterworth low-pass filter with a cut-off of 50Hz. EMG signals were recorded from the Tibialis anterior (TIB), and medial Gastrocnemius (GAS) of the dominant leg.

A custom-made BF system was used to provide subjects with two different designs of either visual or audio BF of the acceleration sensed at trunk level (L5). This acceleration was sensed using a 2D accelerometer (Analog Device ADXL-203), low-pass filtered (50Hz) to cut off high-frequency noise, and amplified 4.5 times. The accelerometer was mounted on the subject's back using a Velcro belt. Visual BF was generated in real-time based on this processed acceleration signal. A red, 1.5-cm-wide, 5-point star, representing the instantaneous acceleration values along AP and ML axes, was plotted on a 15-inch, LCD monitor (resolution 1024x768 pixels; Figure 2A) that was located 50-cm away from the subjects' eyes and adjusted to the subject's height. The red star subtended about 1.5 degrees of visual arc.

During all trials with visual BF, subjects were instructed to keep the red star inside a green ellipse. Anterior-posterior (AP) acceleration was represented by vertical movements of

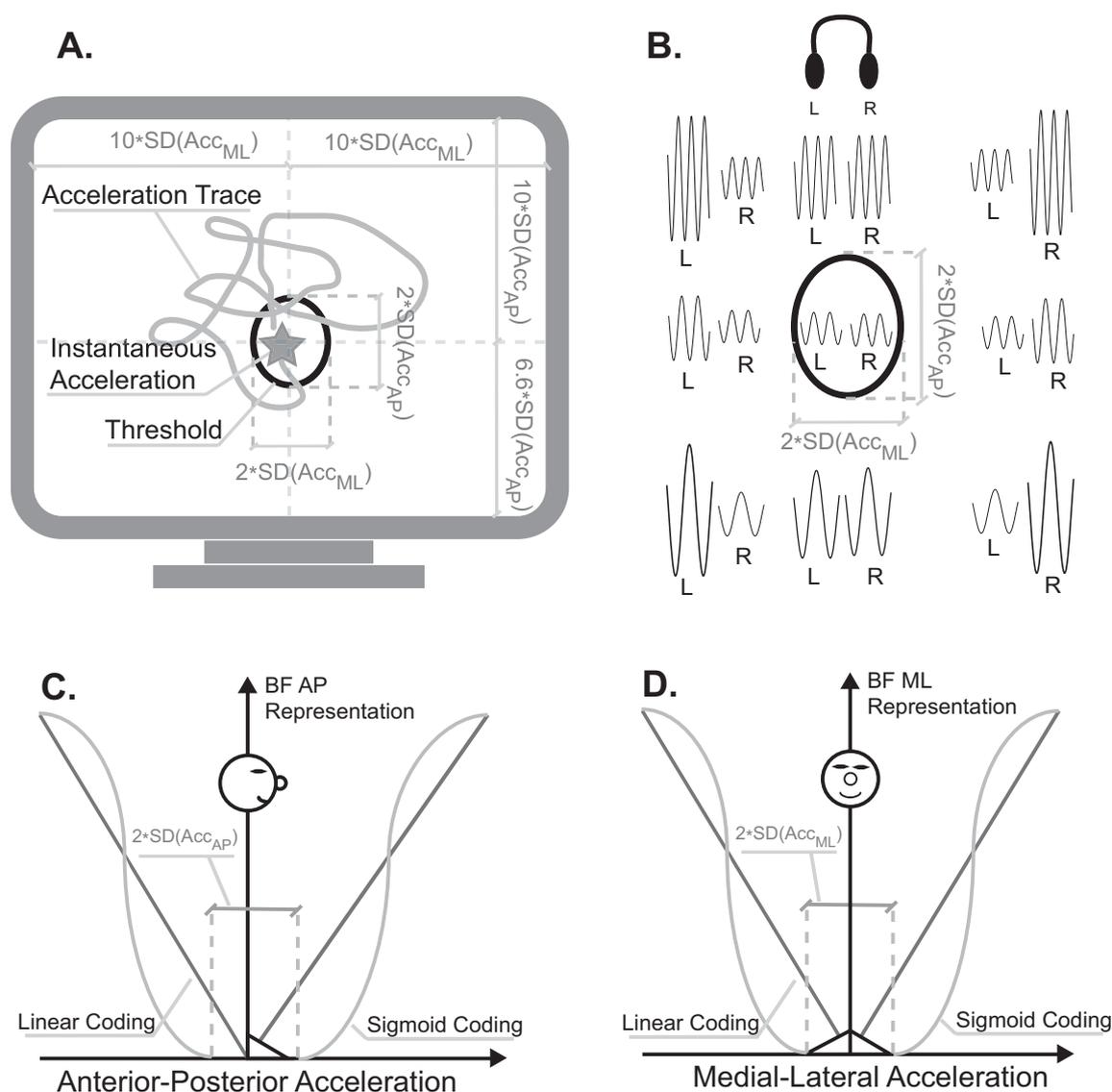


Figure 2 – Panel A: representation of visual BF, a red star (dark gray in this figure) moves on the screen instantaneously representing the trunk acceleration. A blue (gray in this figure) trace represents the trajectory of the acceleration. A green ellipse (black in this figure) represents the target for the subject to pursue during the experiment. The visual BF was scaled on the SD of the acceleration in the first 10 seconds of each trial. Panel B: schematic representation of the dynamics of the ABF sound depending on the subject's direction of sway. The movements of the subject in AP and ML directions induce changes in frequency and volume for the left (L) and right (R) channels of the stereo sound. Although during the experiment, the changes in the stereo sound characteristics were continuous, this panel shows a qualitative representation for each direction. When the subject was inside the threshold, the L and R channel had constant frequency and the lowest volume. An anterior movement induced a frequency and volume increase in both channels (top side), whereas a posterior movement induced a frequency decrease and volume increase in both channels (bottom side). Also, a movement to the left induced a higher volume in the L earphone channel (left side) whereas a movement to the right induced a higher volume in the R earphone channel (right side). Panels C and D: linear and sigmoid codings of BF are represented along AP (C) and ML (D) directions. For the sigmoid coding, a threshold was also implemented so that the subject could get a feedback about his/her movement only when exceeding this threshold.

the star while medial-lateral (ML) acceleration was represented by horizontal movements of the star. During the experiment, a blue trace showed the star trajectory over time. Standard deviation (SD) of the AP and ML acceleration of each subject was obtained in each trial, from the first 10 seconds of recording and used to scale the visual BF. A green ellipse was then displayed on the screen, with its axes aligned with the monitor axes. The vertical axis of the

ellipse was equal to twice the SD of AP acceleration and the horizontal axis, to twice the SD of ML acceleration. The ellipse subtended about 3 degrees of visual arc and was plotted on the screen so that 1) the abscissa of its center was on the vertical axis of the monitor; and 2) the ordinate of its center was plotted such that its distance from the upper edge of the monitor was 1.5 times the distance from the lower edge. These plotting rules were implemented to take into account that subjects have larger sway dynamics in the anterior direction than in the posterior direction [16]. The distance between the center of the ellipse and the left and right edges of the screen were 10 times the SD of the ML acceleration in the first 10 seconds (Figure 2A), The distance between the center of the ellipse and the upper and lower edges of the screen were 10 times and 6.6 times the SD of the AP acceleration in the first 10 seconds, respectively (Figure 2A).

The audio BF was based on the same AP and ML trunk acceleration coding algorithms as the visual BF. A full description of the audio BF software and hardware can be found in [9]. Briefly, a PC laptop was used to generate a stereo sound coding the subjects' trunk accelerations sensed by a bi-axial accelerometer. In this study, the accelerometer was upgraded from the one described in [9]. This new accelerometer was preferred because of its small size, light weight, and portability. During the trials with audio BF, the subjects stood on the force plate while wearing a pair of earphones. The stereo sound provided by the audio BF system consisted of two sine waves, one for the left earphone and one for the right earphone. Pitch, volume, and left/right balance of the stereo sound were modulated to represent the AP and ML acceleration information (Figure 2B).

Specifically, the stereo sound got (1) louder in volume and higher in pitch when the subjects swayed forward (e.g. acceleration increased in anterior direction; volume increased from 20 to 50-dB-SPL, frequency increased from 400Hz to 1000Hz), (2) louder in volume and lower in pitch when they swayed backward (e.g. acceleration increased in posterior direction; volume increased from 20 to 50-dB-SPL, frequency decreased from 400Hz to 150Hz), (3) louder in the right ear channel (volume increased from 20 to 50-dB-SPL) and lower in the left one (volume decreased from 20 to 0-dB-SPL) when they moved to the right (acceleration increased in right direction), and (4) louder in the left ear channel (volume increased from 20 to 50-dB-SPL) and lower in the right one (volume decreased from 20 to 0-dB-SPL) when they moved to the left (acceleration increased in left direction). The first 10 seconds of each trial were used to scale thresholds and limits for the dynamics of the audio BF logically and numerically equal to the one described above, in terms of green ellipse and of screen dimensions, for the visual BF.

Two designs of both the visual and the audio BF were presented, these two designs were obtained by using two different coding functions, logically and numerically similar for both the BF modalities. The simplest one was a linear function (Figure 2C-D) which mapped the acceleration into a movement of the red star on the screen (visual BF) or a pitch and/or

volume sound modulation in the earphones (audio BF) using a fixed, constant gain. With this coding function there was a continuous and proportional effect of movement on the visual/audio BF. The second coding function used a variable gain, following a sigmoid law, with a further nonlinearity due to the presence of the threshold described above, so that subjects did not receive any feedback information while their acceleration was below the threshold (Figure 2C-D). The sigmoid coding of visual BF used equations equivalent to the one described in [9] for the audio BF and shown in Figure 2C-D (see [9]). The sigmoid coding function introduced 2 major characteristics: 1) the feedback was given only when movement exceeded a threshold (i.e. when it was most needed) 2) as soon as the threshold was exceeded the sigmoid function guaranteed a very sensitive BF modulation followed by saturation.

All software for BF and signal acquisition was implemented using Matlab and its Data Acquisition Toolbox. An analog/digital converter (NI-DAQCard 6024E) was used to record the accelerations from the BF system sensor, the muscle activity signals from the EMG device, and the forces and moments from the force plate. All data were sampled with a 100-Hz frequency.

Procedure

All participants performed 30, 55-s long trials standing barefoot on foam. Subjects were instructed to keep their feet as close as possible but without their feet or any part of their legs touching. A few marks on the foam helped the subjects keep their foot position across the trials. After each trial, subjects stepped off the foam surface and waited for the foam to return to its original shape before standing on it again for the next trial. Trials were started 5-10 seconds after the subjects stood on the foam. The 30 trials consisted of five repetitions of six conditions. These six conditions consisted of two BF modalities (audio and visual) each one performed in 3 different modes (linear, sigmoid, off). The off modes conditions were used as reference conditions and consisted of trials without sound and eyes closed for the audio BF modality and of trials with the red star moving randomly for the visual BF modality. These two reference conditions were chosen in order to minimize the potentially misleading effects of attention [17] and the effect of dynamic acoustic cues on sway [18]. During the reference condition for visual BF trials, the subjects were asked to pay attention to the movement of the star without correcting their sway based on the random visual BF. This condition was preferred to a blank screen, because it kept subjects paying attention to the visual task. Since it has been confirmed that paying attention to a second task may induce sway reduction in healthy young subjects [17], this random feedback reference condition assured that visual BF trials were not biased by the attention devoted to a visual task. The reference condition for audio BF also was designed to minimize external phenomena which could have reduced sway. Since Raper & Soames (1991) [18] suggested that a random BF of sound can enlarge postural sway, a silent reference condition was chosen. Off BF conditions were announced to

the subjects so they were aware no BF information would be provided during the trial.

The order of the conditions was randomized in five repetition-blocks. Thus, the subjects performed a full set of six conditions randomized before repeating any of them.

Subjects had their eyes closed during all audio BF modes and open during all visual BF modes. During the trials with BF, subjects were asked to correct their sway according to the feedback, by keeping the red star inside the green ellipse for visual BF and keeping the volume as low and as balanced as possible for audio BF.

Data Analysis

To compare efficacy of BF to reduce postural sway, for each trial, the root mean square (RMS) was calculated for the 2D trunk acceleration and the 2D COP displacement. The RMS reflects extent of sway displacement [19]. These parameters were calculated according to Prieto et al., (1996) [20] and were chosen because, according to Rocchi et al., (2004) [21] and Maurer et al., (2005) [22] they complement each other in characterizing sway displacement. Trunk acceleration reflects body COM acceleration because so much of the COM is in the trunk [13], and it is highly correlated to the COP displacement [9] when subjects use an ankle strategy to maintain balance. The COP displacement reflects body tilt as well as forces the subject exerts into the ground to move the body COM [23]. To further determine the effect of different types of BF on the strategies subjects use to control postural sway, the mean activity of TIB as measured by EMG signals, and the level of co-contraction between TIB and GAS were calculated as an indication of a stiffening strategy. According with Olney and Winter 1982 [24], co-contraction was quantified as the correlation coefficient between the low-pass filtered EMG signals. The AP shear force vector was measured as a reflection of the extent of hip strategy used to correct postural sway [13;25]. Also, the correlation between trunk acceleration and COP displacement was calculated along AP axis to determine whether subjects were moving with an inverted pendulum (ankle) strategy (high correlations), or as more complex, multi-segmental (hip or other) kinematics strategy, (low correlations).

Paired T-tests were used to compare the effects of the sigmoid and linear designs for coding the visual or audio BF on the parameters above. All comparisons were made on percent change due to BF from baseline conditions because the baseline condition without visual and without audio BF differed (eyes open with peripheral view of the room for vision and eyes closed for audio with significantly more sway $p < 0.01$).

Results

Effectiveness of Biofeedback

The BF induced a significant reduction ($p < 0.05$) in the RMS of trunk acceleration in all but the linear coding of audio BF condition. Linear coding of visual BF reduced trunk acceleration more than sigmoid BF ($p < 0.05$). In contrast, sigmoid coding of audio BF reduced trunk acceleration more than linear BF ($p < 0.05$). Figure 3 shows the percent changes induced by the different modalities and coding of BF on the RMS of the trunk acceleration. Figure 4A shows the raw, AP trunk acceleration data from a representative subject while using linear and sigmoid audio and visual BF.

The effect of BF on COP RMS also depended on the BF modality and its coding. Only the sigmoid coding of audio BF significantly reduced COP RMS ($p < 0.05$). Figure 3B shows the percent change of COP RMS induced by the different modalities and coding of BF. Figure 4B shows the raw, AP COP data from a representative subject while using sigmoid and linear audio and visual BF. For all subjects, the RMS of (1) trunk acceleration and (2) COP displacement, were lower in all conditions with eyes open (linear, sigmoid, and off mode of visual BF) than with eyes closed (linear, sigmoid, and off mode of audio BF). Table 1 shows the average absolute values of all parameters analyzed in the 6 different conditions tested.

Postural Strategies

As Table 1 shows, the TIB mean activity was significantly greater during trials with BF

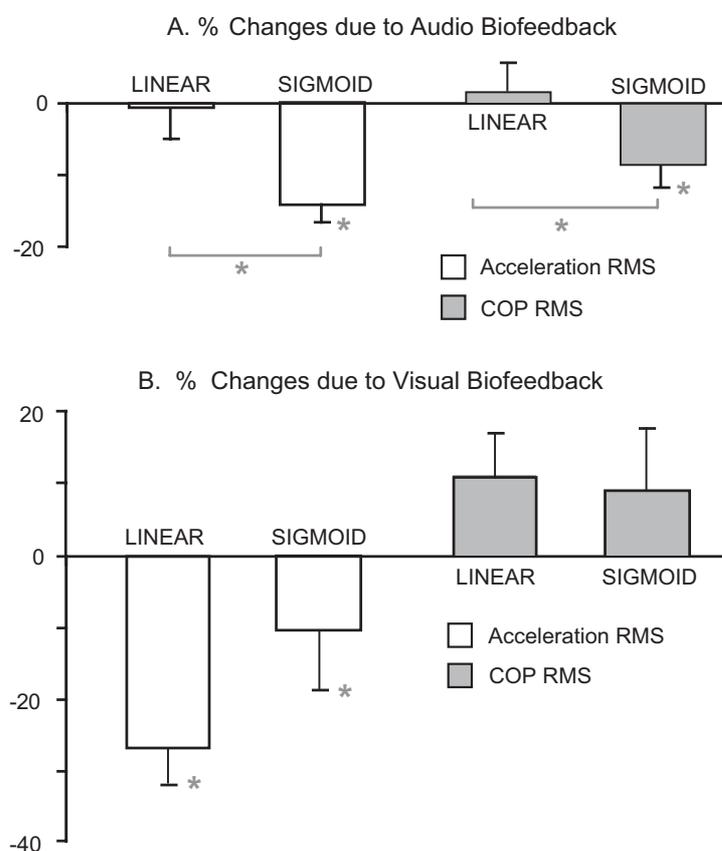


Figure 2 – Average percentage changes of acceleration RMS and COP RMS while using audio BF (panel A) and visual BF (panel B) referenced to the relative, off conditions (eyes closed for audio BF and eyes open for visual BF). The effect of BF on acceleration RMS is represented in white, whereas the effect of BF on COP RAM is represented in gray. The asterisks, which are close to brackets, indicate statistical significant difference ($p < 0.05$) between histograms. The asterisks, which are close to the histograms, indicate statistical significance ($p < 0.05$) from the BF condition represented by the histogram and the respective off condition.

Table 1 – Average Values of The Parameters Analyzed in the Six Conditions Tested

Parameter	Visual			Audio		
	Off	Linear	Sigmoid	Off	Linear	Sigmoid
RMS COP [mm]	5.67(±1.34)	6.30(±1.90)	6.22(±2.38)	12.31(±2.48)	12.27(±2.81)	10.96(±2.02)
RMS Acc [mm/s ²]	90.8(±26.9)	64.4(±22.5)	67.6(±23.1)	139.5(±35.3)	138.8(±40.9)	119.2(±29.6)
RMS Shear [N]	0.09(±0.03)	0.09(±0.02)	0.10(±0.06)	0.19(±0.04)	0.21(±0.03)	0.19(±0.03)
AP Acc-COP correlation	0.62(±0.13)	0.50(±0.20)	0.47(±0.22)	0.68 (±0.12)	0.75(±0.16)	0.71(±0.13)
TIB mean activity	0.52(±0.62)	0.99(±1.25)	1.00(±1.08)	1.02(±1.19)	2.17(±1.50)	1.82(±2.30)
TIB-GAS co-contraction	0.09(±0.04)	0.13(±0.07)	0.12(±0.05)	0.09(±0.05)	0.06(±0.04)	0.10(±0.06)

($p < 0.01$) than in trials without BF. The EMG co-contraction between TIB-GAS did not significantly change with BF.

The correlation between trunk acceleration and COP displacement in the AP direction was significantly larger in trials with audio BF than in trials with visual BF ($p < 0.01$, see Table 1). However, the correlation between trunk acceleration and COP displacement was not significantly different between the two reference conditions (eyes open and eyes closed). The shear forces were no different for the different BF modalities or coding but were significantly larger for all the eyes closed conditions (none, linear audio, sigmoid audio BF) than for all the eyes open conditions (none, linear visual, sigmoid visual BF).

Although linear visual BF and audio sigmoid BF both decreased trunk acceleration RMS (in percentages 28.24 ± 4.79 and 14.38 ± 2.22 , respectively), linear visual BF increased COP RMS whereas audio sigmoid decreased COP RMS (11.2 ± 6.6 and -8.59 ± 3.29 , respectively).

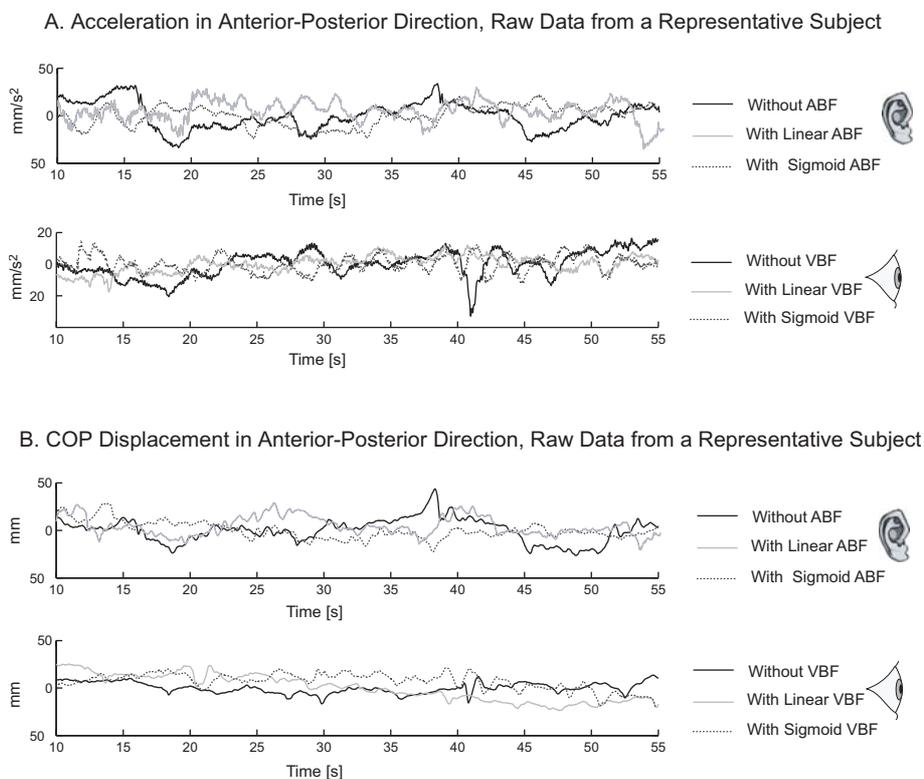


Figure 4 – Panel A and panel B show trunk acceleration and COP displacement, respectively, from a representative subject during all conditions tested.

Discussion

The results reported in this paper showed how both visual and audio BF of acceleration sensed at the trunk level reduced postural sway during upright stance. This sway reduction with audio and visual BF is consistent with the sway reduction reported previously for visual BF [2], for tactile BF [11;12], and for other types of audio BF [26;27]. In this study, two different ways of coding trunk acceleration into BF presentation (linear and sigmoid) were also tested.

The results reported in this paper show that sigmoid coding for audio BF and linear coding for visual BF were the most effective to reduce postural sway in stance. Our results that a different BF coding induces a different extent of sway reduction suggest that customized BF coding for different modalities of BF will make BF information optimally usable. A more sophisticated and accurate exploration of the possible coding between postural sway and BF may result in an even larger sway reduction. For example, Rougier and colleagues found that by adding a delay (>600 ms) and increasing the gain in the BF loop optimized the effects of visual BF on postural stability [28].

Differences between how the nervous system naturally processes audio and visual inputs for detecting postural sway may explain why different coding of BF are needed. Sigmoid coding may be the best for audio BF because subjects can easily detect velocity of sway away from initial posture by the rate of change (velocity) of pitch and volume. Coding feedback with a sigmoid function results in very small changes in BF near the baseline, upright posture with an increasing rate of change of BF as the subject leans toward their limits of stability. Jeka et al. [29] suggest that velocity feedback from somatosensory and vestibular inputs is critical for control of postural stability. Also, allowing a small area with no BF information of sway near upright, as in the sigmoid coding, has the advantage of driving the subjects' attention to the BF only when it was needed. Some models of postural control suggest that natural postural control includes a passive sway area without postural corrections until a threshold is reached, when automatic postural adjustments are triggered [30].

Linear coding may be the best of visual BF because it depends upon subjects detecting the difference in position of a visual signal (in this case, a star) relative to the position of a target representing the initial, upright postural goal (in this case, an ellipse). This detection of error in body versus target position for visual BF, and the relatively slow reaction times elicited from visual inputs compared to auditory inputs [31], may be why a linear coding for

visual BF was optimal.

As expected, all the eyes closed conditions resulted in larger COP excursions, larger trunk acceleration, and larger shear forces, consistent with other studies [19]. Because the availability of a stable visual surround in the periphery has such a large effect on postural sway, we normalized the effects of visual and audio BF to separate reference conditions with eyes open, and eyes closed, respectively. Our previous study showed that audio BF has a very limited effect when healthy subjects or vestibular loss subjects are standing on a firm surface with eyes open, probably because of a ceiling effect, and because subjects are reluctant to switch dependence from preferred sensory reference frames to novel sensory input for posture [32]. However, the amount that subjects use auditory BF to reduce postural sway depends upon how much it is needed based on the sensory context and the extent of their sensory pathology with the maximum effect when vestibular loss subjects stand on a compliant surface with eyes closed [32]. Thus, the optimal sensory mode for effective BF is likely to vary under different sensory conditions, pathology and age. For example, trunk acceleration BF information may be most effective for subjects who have lost otolith information whereas COP BF may be more effective for subjects who have lost sensitivity to pressure under their feet due to pathology.

While visual linear BF and sigmoid audio BF had the largest impact on postural sway, each mode of BF appeared to facilitate a different type of postural sway movement strategy. The visual BF mainly reduced trunk acceleration, whereas the audio BF mainly reduced COP. These results, along with the greater correlation between COP and trunk acceleration that was found in the audio BF condition, suggest that the two BF presentations induce different postural, kinematics strategies. In fact, an inverted pendulum model of postural sway is consistent with the effects of sigmoid audio BF. In contrast, linear visual BF, resulted in an increase in COP displacement with a decrease in trunk acceleration and a lower correlation between COP and trunk acceleration which is consistent with a multi-segmental model of body sway [33]. The necessity of using two different kinematics models to explain the change in postural movement strategy associated with audio and visual BF suggests that visual BF pushes the control of posture more toward a “hip strategy” (multi-segmental model), and the audio BF pushes the control of posture more toward an “ankle strategy” (inverted pendulum model; [34;35]).

Our results also suggest that the eyes open reference condition was associated with a larger contribution of hip strategy than the eyes closed reference condition [29]. In the visual BF reference (eyes open) condition, the correlation between trunk acceleration and COP displacement was lower than in the audio BF reference trial (eyes closed). In the eyes open reference condition for visual BF, the strategy used to control posture may have had a higher contribution of hip strategy [36] to fix the distance in space between head and monitor whereas, in the eyes closed, reference trials for audio BF, the strategy used may have had a

higher contribution of ankle strategy such that movement of the head and ears correlated with movement of the body COM.

An alternative explanation could be that visual and auditory BF may not induce a different strategy for the control of posture but, perhaps, simply enhance the natural postural strategy already used by the central nervous system in that particular condition (eyes open and eyes closed). If this hypothesis is confirmed, it could be further speculated that BF increases the reliability of other sensory information by adding redundancy and providing a reference which increases the signal-to-noise ratio in the control of posture sensory feedback loop [37;38]. In other words, the central nervous system may use the BF information not only by itself, but also in combination with the other sensory information to increase the precision of the estimation of the body posture.

Neither visual nor auditory BF appeared to reduce postural sway via a stiffening strategy since there was no increase in co-activation of muscles around the ankle joints. This suggests that the added sensory information about body sway enhanced the natural, direction specific, automatic postural control strategies rather than superimposing a generalized stiffening. The increase in background TIB EMG activity during use of BF would reduce the threshold when this ankle dorsiflexor would be recruited to resist backward and backward-lateral body sway but was not associated with a change in background COP position in our subjects [39].

In conclusion, this study showed how reduction of postural sway in stance using BF depends on the modality and coding of the BF of trunk acceleration. Linear visual BF and sigmoid audio BF induced the largest reduction in postural sway although via different postural kinematics strategies.

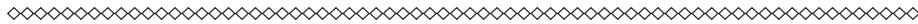
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Chapter 8

Postural Responses Elicited by Auditory-Biofeedback of Center of Pressure during Perturbed Stance



Most of the content of this chapter will be submitted as: M. Dozza, L. Chiari, R.J. Peterka, C. Wall III, and F.B. Horak, "Postural Responses Elicited by Auditory-Biofeedback of Center of Pressure during Perturbed Stance," to Human Movement Science.

Abstract

Biofeedback is known to improve postural control by augmenting movement information. However, the relation between amount of biofeedback information and postural control improvements is still unknown. A few biofeedback-based products are now on the shelf and promise to be effective for motor rehabilitation. However, the interaction between spontaneous motor learning and biofeedback effect, which is the basis for the usefulness of biofeedback in rehabilitation, is still unknown.

In this study, an audio-biofeedback system, providing different amounts of movement information, was used to improve subjects' performance during repetition of perturbed stance.

Higher amount of audio-biofeedback information resulted in higher postural stability in the beginning of the experiment. However, overtime, motor learning normalized the effects of the different amount of audio-biofeedback information. Nevertheless, motor learning did not neutralize the effect of audio-biofeedback at low frequencies (<0.2 Hz) of sway. Analysis of postural responses transfer functions verified that audio-biofeedback affected prevalently the low frequencies of sway (<0.4 Hz) whereas motor learning affected prevalently the high frequencies of sway (>0.4 Hz).

Introduction

The concept of biofeedback is well known since the 50's [1]. However, only in the last few years, the interest on biofeedback systems for postural control has renewed partially due to the advance in technology. This renewed interest is evidenced by several recent publications showing the efficacy of biofeedback in improving motor performances [2-5].

Despite these publications increase our knowledge about biofeedback, many questions are still open about 1) biofeedback design, 2) biofeedback experimental protocols to be used for rehabilitation, and 3) the mechanisms by which biofeedback system may induce postural improvements and retention of motor performance.

The first challenge in the development of a biofeedback device is its design [6]. The design of a biofeedback system should optimize the efficacy of its three main parts: 1) the sensor unit, which acquires the information to be fed back; 2) the elaboration unit, which processes and converts this biological information into new information; and 3) the restitution unit, which conveys this new information to the user. However, to improve the design of the whole biofeedback device, it is relevant to determine the amount of information that is actually needed by the user, and the amount of information that the user is able to handle. To date, there are no studies reporting on this issue.

Another challenge in the development of biofeedback devices is the protocol design to be used for the device validation [7]. In fact, the experimental protocols at this stage of the development should be aimed at evaluating the interaction between motor improvements due to biofeedback and the motor improvements due to other mechanisms such as spontaneous learning. This distinction is fundamental to evaluate retention and transfer of motor performance after exposure to biofeedback and, finally, biofeedback efficacy for rehabilitation. To date, very few studies reported on this issue, which is well known to be a crucial one for the evaluation of biofeedback devices [7;8].

Up to now, biofeedback efficacy was determined, in most of the published studies, by looking at some general balance indicators such as center of pressure, trunk angular velocity, and head tilt which were also the feedback variables (e.g. [4;5;9], respectively). However different biofeedback designs can induce different postural response strategies [10]. As a consequence, to evaluate a biofeedback system, it is necessary to record a high number of variables so that, not just the performance, but also the mechanisms and postural strategy

used to achieve a better performance can be evaluated. For this evaluation, biofeedback devices can take advantage of systems already available and purposely developed for analyzing postural responses such as the one designed by Peterka [11]. Such device is able to quantify postural response at different frequencies of induced sway so that a further insight on the mechanism of sensory reweighting taking place during the exposure to biofeedback can be achieved [12].

In this study, the amount of biofeedback information necessary to stabilize subjects in perturbed stance and the interaction between the effect of biofeedback and spontaneous learning during the practice of this task have been evaluated. Analyses from Peterka's system verified that biofeedback and motor learning affect different frequency intervals of postural responses.

Materials & Methods

Participants

Thirteen healthy subjects, age 33 ± 7 yrs, height 175 ± 10 cm, and weight 78 ± 18 Kg, participated to this study. All subjects responded to the following inclusion criteria: 1) no hearing deficits, 2) no history of traumas or surgeries to the muscular-skeletal system, and 3) no history of orthopedic or neurological diseases or disorders. All subjects signed an informed consent before the experiment took place. This informed consent was approved by the OHSU Ethical Committee and guaranteed the subjects' rights according to the Declaration of Helsinki (1964).

Protocol

All participants stood on a rotating force-plate able to destabilize their posture in the medial-lateral (ML) plane (Fig. 1). The force-plate rotated accordingly to a pseudorandom function [13] with a 4-degree peak-to-peak amplitude over a frequency range of 0.017 to 2.2 Hz [11]. In each trial, subjects were exposed to three cycles of the pseudorandom perturbation. Each cycle was 60.5 s long, so that the total length of each trial was 181.5 s. All participants were asked to maintain balance while the force-plate rotated and to respond to the information from an audio-biofeedback (ABF) when available. This ABF was able to inform the participants about their ML center of pressure (ML-COP) displacement according to four different ABF modalities with different extent of information about ML-COP displacement. The ML-COP displacement was recorded by the rotating force-plate. Two bi-axial accelerometers (Analog Device ADXL202) were mounted on the subjects at C7 and L5 and were oriented so that they could sense acceleration along the subjects' anterior-posterior and ML direction. In addition shoulder and hip position in the ML plane were recorded via two potentiometers. Each subject was tested during three blocks of five randomized conditions. Four out of the five conditions corresponded to the four different modalities of ABF, whereas the fifth condition corresponded to a control condition where the subjects were not provided with any ABF.

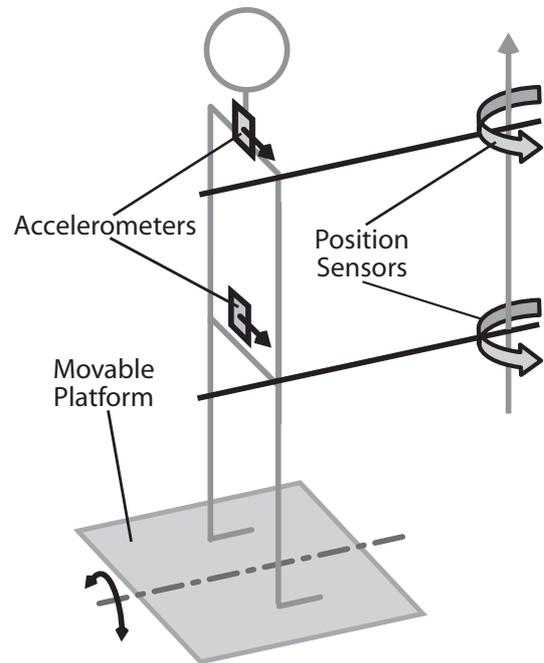


Figure 1 – Experimental set-up.

ABF modalities

The four ABF modalities differed in the amount of ML-COP information which was fed back. Specifically, in modality 1, both the direction and the magnitude (full information) of the ML-COP displacement were fed back to the subjects. In modalities 2 and 3, only direction and only magnitude, respectively, was fed back to the subjects. Finally, in modality 4, the ABF was limited to an alarm signal which informed the participants whenever their ML-COP was exceeding a reference threshold (RT) in either the left or right direction. This RT was determined for each subject based on their ML-COP displacement. This RT corresponded to 1 standard deviation of the ML-COP displacement recorded during the 10 seconds before each trial. In all 4 modalities the ABF was provided only when the subject was exceeding this threshold.

The ABF sound consisted of a 400-Hz sine wave modulated in volume so that changes in volume could provide the information about the above-mentioned ML-COP displacement. When subjects were inside the RT, the ABF volume was constant at 20 dB. The relations between ABF volume and ML-COP displacement in the 4 different modalities are shown in Figure 2. In particular, the algorithm controlling the relation ML-COP/volume in the **first** modality (Fig. 2A) is the same described in Chiari et al., 2005 [14]. Briefly, when the subjects move *left/right*: 1) the sound in the *left/right* earphone increases (20 to 50 dB) according to a sigmoid function, and 2) the LR balance changes according to an exponential function so that the sound in the earphone *right/left* earphone decreases (20 to 0 dB). In this way, both the information about the direction and magnitude of the ML-COP displacement were provided to the subjects. In the **second** modality, Fig. 2B, the volume of the sound was always the same in both earphones and increased according to the same sigmoid function as used in the first modality depending only on the magnitude of the ML-COP displacement. In the **third** modality, Fig. 2C, the ABF volume changed according to a step function so that: 1) when the subjects exceeded the RT in the *left* direction, the volume suddenly increased (0 to 50 dB) in the left earphone and decreased (20 to 0 dB) in the right earphone; and 2) when the subjects exceeded the RT in the *right* direction, the volume suddenly increased (0 to 50 dB) in the right earphone and decreased (20 to 0 dB) in the left earphone. Thus, in this modality, the only direction of the ML-COP displacement was provided to the subjects. Finally, in the **fourth** modality, Fig. 2D, as soon as the subjects exceeded the RT the volume in both earphones increased (20 to 50 dB) accordingly to a step function. Thus, the only information provided to the subjects was if their ML-COP was inside or outside the RT. During the experiment, the participants were asked to pay attention to the sound and to try to minimize its volume which, lately, implied reducing their ML-COP displacement.

Data collection and analysis

For each trial, the COP displacement, the acceleration at L5 and C7, and the position

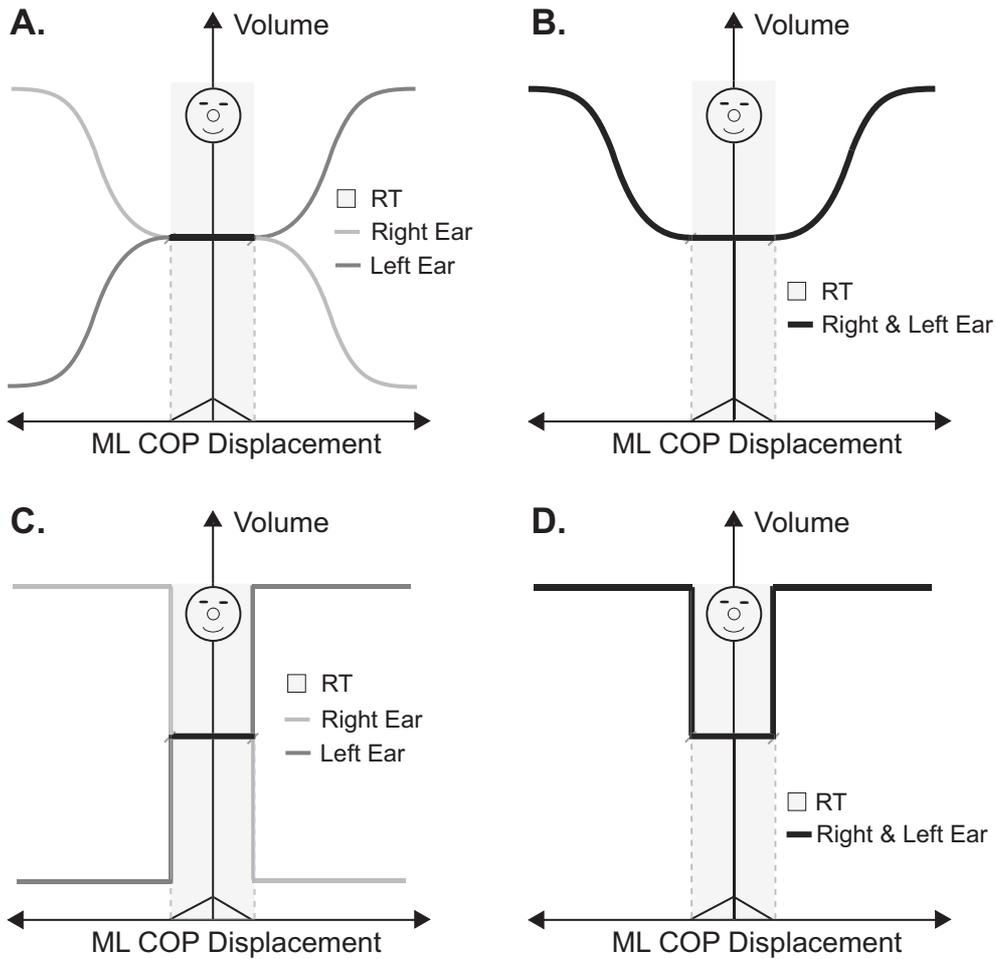


Figure 2 – ABF modalities – A: ABF coding both the magnitude and direction (full information) of COP displacement. B: ABF coding only the magnitude of COP displacement. C: ABF coding only direction of COP displacement. D: ABF coding only for exceeding the RT.

of hip and shoulder were recorded in the ML plane with a 100-Hz sample rate (Fig 1). From ML-COP displacement and acceleration data the SDs were calculated. In addition, from the position of hip and shoulder, the transfer function characterizing the subjects' postural sway responses (PTF) was calculated according to Peterka [11]. Briefly, from the hip and shoulder position and anthropometry, the center of mass (COM) body sway angle with respect to earth vertical was estimated. Then, the COM body sway angle and the measured rotation of the force-plate were used to calculate the power spectra for each cycle of each trial. Finally, the power spectra were averaged across the cycles to obtain a transfer function describing the postural responses to force-plate rotation, in terms of gain and phase (at 16 frequencies evenly spaced in the logarithmic frequency interval 0.016-2.2 Hz). Correlation analyses were performed to verify the relation among COP, L5 acceleration, and C7 acceleration in the ML plane. One-way ANOVA with Newman-Keuls multiple-comparison test was used to verify significant difference ($p < 0.05$) between trials with and without ABF (effect of the ABF modality). Two-tail, paired T-test was used to verify significant difference ($p < 0.05$) between trials in the first and last block (effect of motor learning). Bonferroni correction was applied in case of multiple comparisons.

Results

ML COP and Accelerations

Effect of ABF on ML COP and Accelerations

When using ABF, all subjects exhibited a smaller sway compared to the control condition in all three blocks of trials (Figure 3). All four ABF modalities significantly reduced ML COP displacement and ML acceleration at L5 (Figure 4A-B). ML acceleration at C7 increased for most of the subjects using ABF. However, this last result was not supported by statistical significance. When averaged overtime, the effect of all ABF modalities on ML COP displacement and accelerations was similar (Figure 4).

Effect of learning on ML COP and Accelerations

The amount of sway reduction caused by the four ABF modalities changed overtime. Specifically, in the first block of trials, amount of sway (in terms of ML COP and acceleration at L5; Figure 5A-B) was inversely proportional to the extent of information coded by the ABF. In fact, full-information ABF resulted in the smallest amount of sway; alarm ABF resulted in the largest sway with ABF; and direction and magnitude ABF resulted in a similar amount sway intermediate between the other two modalities. In the second and third block, the amount of sway with the different ABF modalities was similar, even if the full-information ABF resulted in a slightly smaller sway compared to the other modalities. Vice versa, in the first block, ML acceleration at C7 increased proportionally to the extent of information coded by the ABF. However, in the second and third block ML acceleration at C7 was not significantly different in all conditions tested.

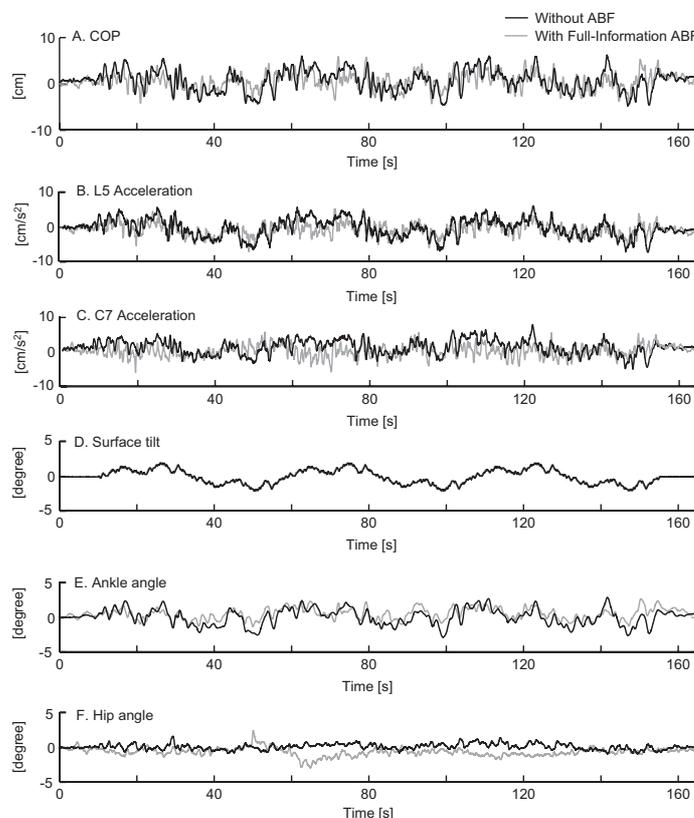


Figure 3 – Raw data from one representative subjects in conditions 1 and 5 (i.e. with full-information ABF and without ABF) from 2 trials performed in the second block of

Effect of ABF Modality on SD of the COP and L5 Acceleration

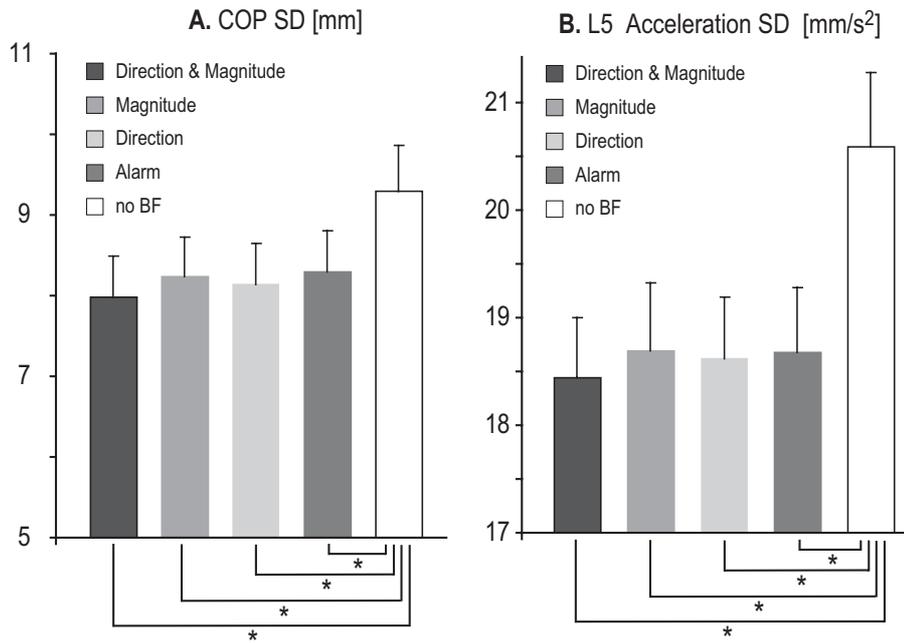


Figure 4 – Averaged SD of A: center of pressure, and B: acceleration at L5 level in all five conditions tested. Asterisks indicate significant difference ($p < 0.05$) from control condition.

Effect of ABF Modalities on SD of COP and L5 Acceleration Overtime

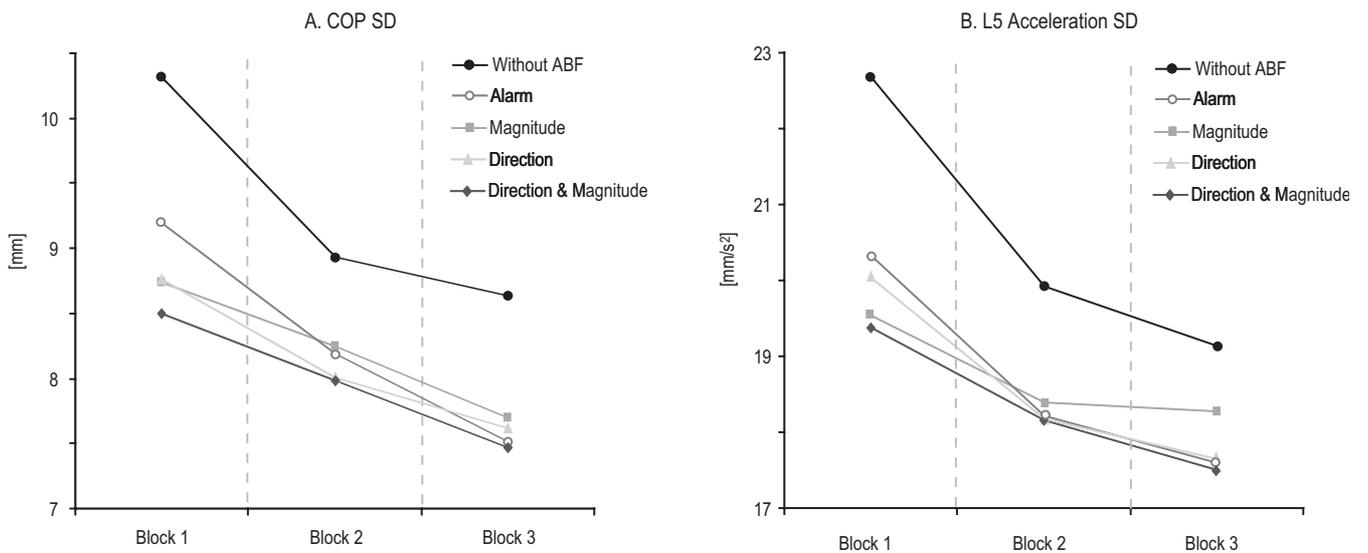


Figure 5 – Standard deviations of A: center of pressure and B: acceleration at L5 level overtime in all conditions tested.

Table 1 – Correlation coefficients (*r*) among center-of-pressure and accelerations.

	Without ABF			With ABF		
	COP-Acc.L5	COP-Acc.C7	Acc.L5-Acc.C7	COP-Acc.L5	COP-Acc.C7	Acc.L5-Acc.C7
1 st Block	0.89 (± 0.04)	0.84 (± 0.09)	0.77 (± 0.12)	0.84 (± 0.06)	0.58 (± 0.21)	0.48 (± 0.18)
2 nd Block	0.90 (± 0.03)	0.77 (± 0.13)	0.69 (± 0.15)	0.84 (± 0.05)	0.71 (± 0.12)	0.56 (± 0.16)
3 rd Block	0.88 (± 0.06)	0.77 (± 0.16)	0.70 (± 0.17)	0.84 (± 0.06)	0.69 (± 0.15)	0.56 (± 0.20)

Effect of ABF and learning on the correlations between ML COP and Accelerations

Correlation among ML COP and acceleration increased when comparing overtime the full-information ABF and the control condition. In the first block of trials, the correlations in the ML plane between 1) COP and C7 acceleration, and between 2) L5 acceleration and C7 acceleration were lower in the full-information ABF than in the control condition (Table 1). However, from the second block of trials, the same correlations in the ML plane between 1) COP and C7 acceleration, and between 2) L5 acceleration and C7 acceleration, increased in the full-information ABF condition and decreased in the control condition (Table 1). Correlation between COP and L5 in the ML plane was high in both the full-information ABF condition and in the control condition and did not significantly change overtime (Table 1). Finally, correlations in the ML plane among COP, L5 acceleration and C7 acceleration did not significantly change between the 2nd and the 3rd block of trials.

Postural Response Transfer Function

Effect of ABF on PTF gain

ABF reduced the gain of the PTF especially at low frequencies. Figure 6A shows data from a representative subject, in the three blocks of trials, comparing the full-information ABF and control condition. Further, the effect of ABF (averaged across subjects) in the first and last block of trials are reported in Figure 7A-B. In the first block full-information ABF significantly ($p < 0.05$) decreased the PTF gain at the very low frequency (0.02 Hz) and in a narrow interval around 1 Hz (Figure 7A). In the third block, full-information ABF significantly ($p < 0.05$) decreased the PTF gain in the wide interval 0.02-0.2 Hz (Figure 7B). Figure 7C compares the effects, in terms of PTF gain reduction, occurred in the first and third block. Specifically, in the first block the largest gain reduction occurred around 0.8 Hz. In the third block the largest gain reduction occurred at low frequencies (< 0.2 Hz).

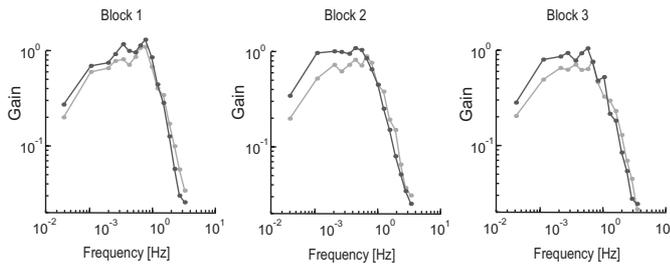
Effect of learning on PTF gain

Subjects reduced the gain of the PTF overtime in all conditions tested. Figure 6B shows data from a representative subject, in all conditions tested, comparing the first and third block of trials. Further, the effect of time (averaged across subjects) in the full-information ABF and control condition are reported in Figure 8A-B. Both the full-information ABF and control condition significantly ($p < 0.05$) decreased the PTF gain overtime at the very low frequency

Effect of ABF and Learning on the Gain of the Postural Reponse Transfer Function

(0.02) and in the interval from 0.2 to 1.1 Hz (Figure 8A-B).

A. Trials without (black) and with ABF (gray) in the three blocks



B. First (black) and third (gray) block in the five conditions

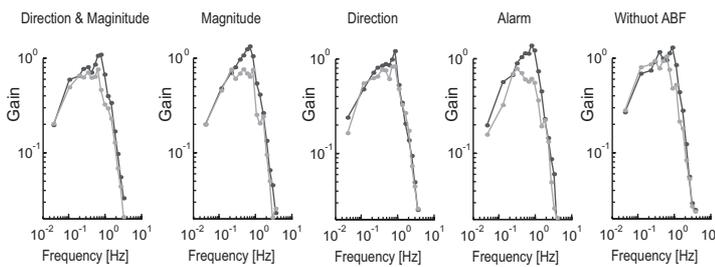


Figure 6 – Gains of the PTF (i.e. ratio between body sway amplitude and stimulus amplitude) from one representative subject. Panel A represent the effect of using full-information ABF (Condition 1) compared to no ABF (Condition 5) in the three blocks of trials. Panel B represents the learning effect i.e. the comparison between trials in the first and last block.

The extent of gain reduction overtime was the largest in a specific, narrow interval of frequencies. In particular, all subjects showed the largest gain reduction in a narrow range of frequencies both in the full-information ABF and control condition. By plotting the difference between the PTF in the first and third block, it was possible to highlight a peak of gain reduction (due to the reduction of gain in the narrow range of frequency) for each subject. Each subject showed the peak of gain reduction at slightly different frequency but always comprehended between 0.2 Hz and 0.9 Hz. This peak was presented both

Effect of ABF: Difference between with and without ABF

Learning: Difference between the First and Third Block

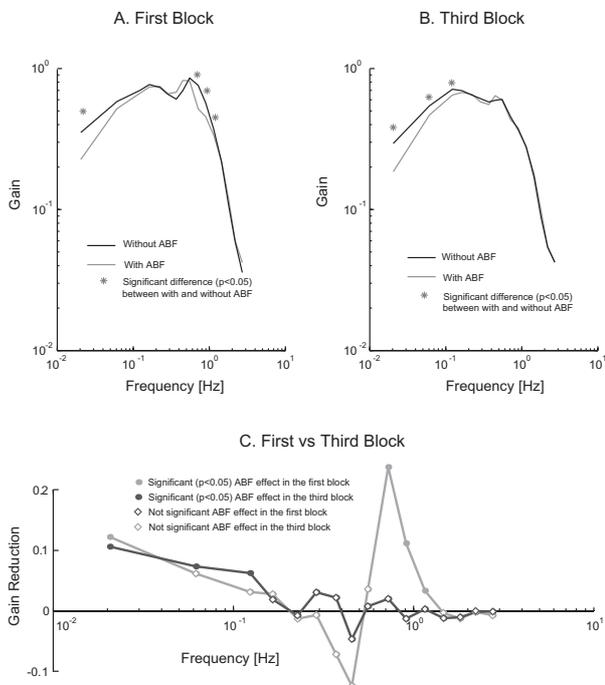


Figure 7 – Effect of full-information ABF on PTF gain (i.e. ratio between body sway amplitude and stimulus amplitude) in the first (panel A) and last (panel B) block. The effects of full-information ABF in the first and third block are compared in panel C.

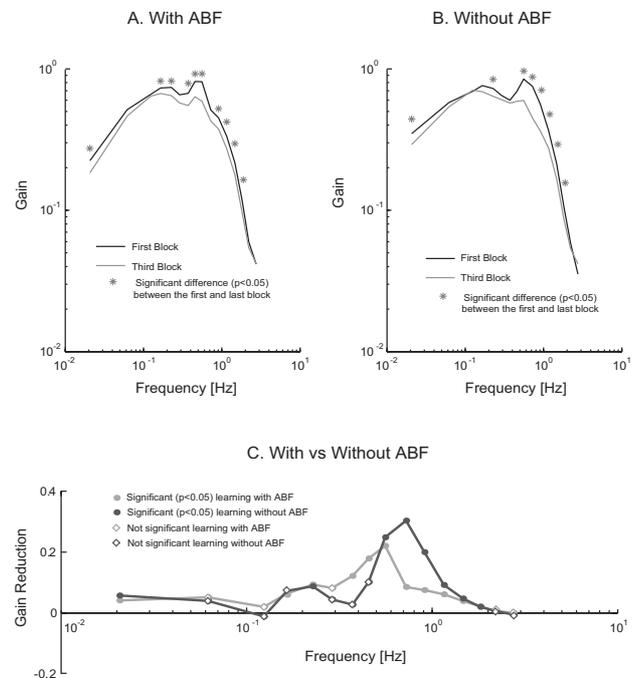


Figure 8 – Effect of learning on PTF gain (i.e. ratio between body sway amplitude and stimulus amplitude) with full-information ABF (Condition 1; panel A) and without ABF (Condition 5; panel B). The effects of learning with and without full-information ABF are compared in panel C.

in the full-information ABF and in the control condition. However, the position and amplitude of the peak was not always the same in all subjects in between these two conditions. The gain reduction, averaged across subjects, is shown in Figure 8C for the full ABF information and control condition. The two low peaks in Figure 8C are the consequence of averaging the individual gain reduction peaks. In the full-information ABF condition, the low peak was lower in amplitude and frequency compared to the low peak in the control condition (Figure 8C). However, this difference between the low peaks was not verified in each subject data.

Effect of ABF and learning on PTF phase

ABF and learning increased PTF phase in different intervals of frequency. Specifically, ABF significantly affected low frequencies (<0.4 Hz) whereas learning significantly affected high frequencies (>0.8 Hz). In addition, the amplitude of PTF phase increase due to ABF was larger than the increase due to learning. Figure 9 shows the effect of ABF and learning on PTF phase.

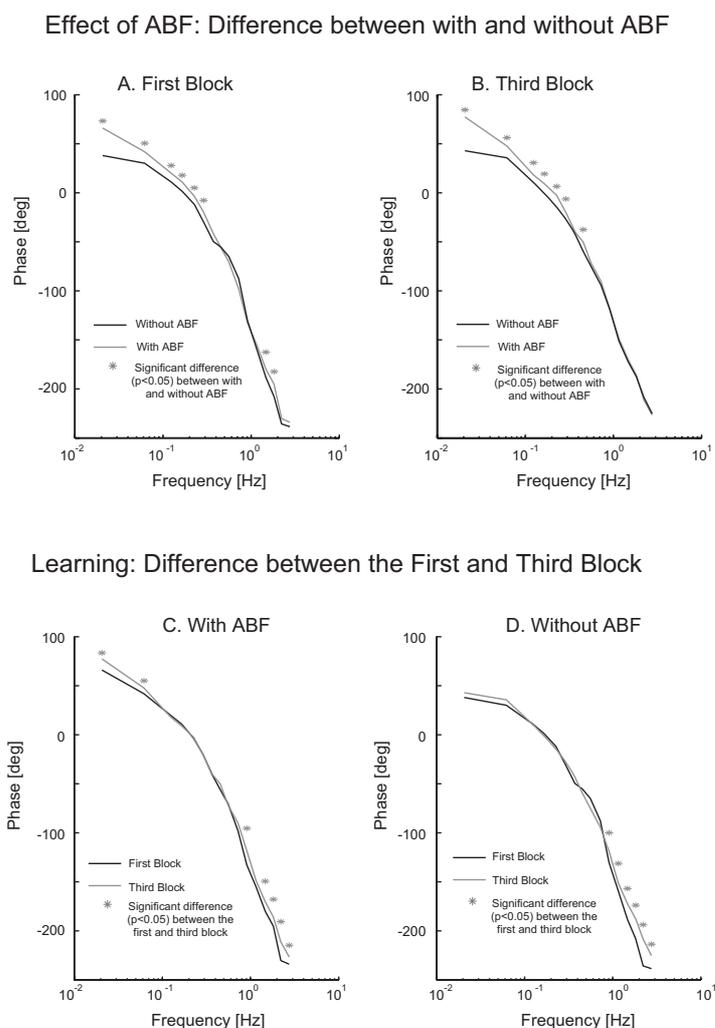


Figure 9 – PTF phase changes due to full-information ABF in the first (A) and third (B) block of trials. PTF phases changes overtime with full-information ABF (C) and without ABF (D).

Discussion

During this experiment, two main factors concurred in decreasing subjects' sway: 1) use of ABF (i.e. augmented sensory information), and 2) repetition of the task (i.e. motor learning). These two factors interacted during the experiment inducing a similar extent of sway reduction. For this reason, they need to be considered combined for the interpretation of the results.

ML COP and Accelerations

With all ABF modalities, subjects' improved their balance even after motor learning occurred (block 3). However, only in the first block of trials, significant differences between the ABF modalities were evidenced by the subjects' performance. In fact, in the first block, the advantage of having ABF with a larger amount of ML-COP information resulted in better performances. In the second block, the difference among the four modalities of ABF became less evident from the subjects' performance, and this difference, then, almost disappeared in the third block. For this reason, the effects of all ABF modalities averaged across time (Figure 4) do look similar. Nevertheless, subjects' exhibited smaller sway only when they received ABF, even in the third block of trials where sway reduction induced by motor learning was the maximum. This result is consistent with our previous results where we showed how this ABF system decrease postural sway in normal and vestibular loss subjects [15].

Sway reduction induced by motor learning is evidenced by the subjects reducing sway overtime in the control condition. However the extent to which motor learning was induced by ABF or was spontaneous is open to debate. In fact, in other experiments, not involving ABF [11], subjects did not show sway reduction overtime by simple practice of standing on the same rotating force-plate. This finding supports the hypothesis that ABF favored motor learning during the experiment resulting in motor retention during control trials.

Subjects performance improved overtime also in all ABF modalities. This result confirms that some learning mechanism occurred during the trial. However, the extent to which improvements in ABF condition were driven by motor learning or by an optimization of the ability to use ABF is still questionable. In fact, a better performance in the task may have been achieved by a more correct interpretation of ABF, which may have been developed overtime by the subjects. Also, the result that, in the third block, subjects could achieve similar performances independently from the modality of ABF, suggests that the subjects were able

to implement faster and more accurate corrections in response to the ABF. Thus, subjects may have completed the partial information from the different ABF modalities with natural sensory information. In this case, improvement of postural performance overtime using ABF may be the result of an improved integration among sensory information and between sensory and artificial information.

In addition, the task tested in this study is not critical for healthy adults – as the subjects participating in this study. Thus, the similar performance, achieved by the subjects overtime, independently from the ABF modality used, may depend on a ceiling effect. In other words, overtime subjects may have been able to optimize their ability of maintaining stance on the rotating surface to the point that further improvements were not possible by simply adding ML COP information.

Movements at the hip level were restricted at the beginning of the experiment and became less restricted overtime in the control condition. In fact, in the control condition, correlation in the ML plane between COP and C7 acceleration and L5 acceleration and C7 acceleration decreased overtime. When using ABF, subjects presented from the very beginning a low correlation in the ML plane between COP and C7 acceleration and L5 acceleration and C7 acceleration which then increased with the optimization of postural responses and the consequent reduction of sway. Thus, using ABF, movements at hip level were not restricted at the beginning and then became more restricted once the task became easier. However, the best performance in all condition was achieved when movements at the hip level were evident (third block).

Also, standard deviation of correlation factors increased overtime in all condition, suggesting that subjects were not converging to a common strategy for the control of posture but, instead, were taking advantage of personalized multi-segmental control to achieve best performances. ABF favored from the beginning multi-segmental control of posture, somehow anticipating what spontaneous control of posture may have developed overtime. Thus, ABF may have favored spontaneous motor learning by inducing the subjects to gain confidence with multi-segmental control of posture from the very beginning of the experiment.

Postural Responses Transfer Function

ABF and motor learning resulted in gain reduction in the PTF – a lower gain is indicator of higher stability [11]. Once again the two factors (ABF and motor learning) causing gain reduction acted contemporarily and with a similar extent on subjects' gain. However, by analyzing the PTF gain and phase changes at the different frequencies, it is possible, to partially discriminate the effect of learning and ABF.

The effect of learning is evident from the gain reduction occurred at 0.02 Hz and in the range 0.2-1 Hz overtime. This gain reduction was found to be similar for trials with and without ABF, suggesting that subjects converged overtime to the same postural control mechanism in

both ABF and control condition. However, only in the first block, the instantaneous effect of ABF regarded the same frequencies affected overtime by learning. In other words, the effect of learning was similar to the effect of using ABF in the first block. Then, by the third block the effect of ABF regarded only the frequencies below 0.2Hz. This result suggests that, overtime, subject retained motor improvements in the range 0.2-1Hz but not at low frequencies (<0.2 Hz) where ABF continued to show an additional reduction effect on top of motor learning. Since gain reduction was found to be an indicator of sensory reweighting from somatosensory to vestibular [11;16], the gain reduction induced by ABF may be also the consequence of a sensory reweighting which favored vestibular information over somatosensory information.

The effect of ABF on sway was also somehow similar to the effect of a vestibular prosthesis based on tactile biofeedback [17] which was recently tested with the same perturbation used in this study [12]. Using this prosthesis both control subjects and bilateral vestibular loss subjects 1) reduced sway, 2) reduced PTF gain at low frequencies (<0.8 Hz), 3) increased PTF gain at high frequencies (>0.8Hz), and 4) did not change PTF phase. Some of the reasons why the results reported in this study differ from the ones reported by Perterka et al. [12] may be due to: 1) the different design of the biofeedback devices used (ABF versus tactile biofeedback), 2) the different direction of biofeedback information and platform perturbation (ML in this study versus anterior posterior in Perterka's study), and 3) the use of a back board only in Peterka's study which constrained the subject to move as an inverted pendulum. Peterka et al. [12] suggest that the PTF gain reduction at low frequencies (which was found also in this study) could be due to the limited bandwidth of the orientation information from the biofeedback. However, another explanation could be that high frequency orientation information had been filtered out by the intrinsic delay of the voluntary postural response to ABF. In other word, high frequency gain reduction would inevitably require short time responses which may not be compatible with the several-hundreds-millisecond dynamic needed for the brain to receive the biofeedback information, elaborate it, and activate the muscles to generate the postural response. In this case, practicing could improve the balance prosthesis performance by making more automatic the postural responses to the orientation information [18;19]; the gain reduction found up to 1.1 Hz in this study after practicing supports this last speculation.

An unexpected result of this study was that subjects did not reduce the PTF gain at all frequencies overtime but, instead, had a pretty narrow and specific range of frequencies that they tended to reduce the most. This narrow range matches the range of frequencies where a small peak, similar to a resonance peak in a second-order system, was also evident in the PTF. The presence of such peaks in a transfer function normally determines more instability for the system in the range of frequencies where the peak is. As a consequence, the reduction of gain in a narrow range of frequencies matching the frequencies of the PTF gain peak seems aimed at improving the system stability where it was more needed. In other words, the peak of reduction showed by the subjects in some narrow range of frequencies may have been

avored by the system being a priori more unstable in that very range of frequencies.

Once motor learning occurred, PTF phase showed a clear difference between the effect of ABF and learning. In fact, the effect of ABF and learning regarded two distinct different intervals of frequency (Figure 9). In particular, ABF anticipated the phase delay at low frequencies (<0.4 Hz) whereas learning anticipated the phase delay at high frequencies (>0.8 Hz). These results suggest that 1) postural responses to low-frequency perturbation were faster when using ABF; and 2) postural responses to high-frequency perturbation became faster with repetition of the task;

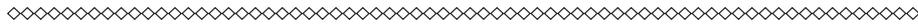
In conclusion, this study showed how motor learning and sensory augmentation concur to sway reduction when humans are practicing a dynamic task, such as perturbed stance, using an ABF system. Higher amount of ABF information resulted in higher postural stability in the beginning of the experiment. However, overtime, motor learning normalized the effects of the different ABFs. Nevertheless, motor learning did not neutralize the effect of ABF at low frequencies (<0.2 Hz) of sway. With learning, subjects increased the variability of postural control by using a multi-segmental strategy. With ABF, subjects used from the very beginning a multi-segmental strategy that was then optimized overtime. PTF analysis highlighted some differences among the mechanisms by which motor learning and ABF caused sway reduction once motor learning occurred. In particular, motor learning favored PTF gain reduction in the 0.2-1 Hz interval and PTF phase increase above 0.8 Hz whereas ABF favored PTF gain reduction for the frequencies below 0.2 Hz and increase of PTF phase below 0.4 Hz

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Chapter 9

Effect of Trunk-Tilt Tactile Biofeedback on Tandem Gait in Vestibular Loss Subjects



Most of the content of this chapter will be submitted as: M. Dozza, R.J. Peterka, C. Wall III, L. Chiari, and F.B. Horak, "Effects of Practicing Tandem Gait with and without Vibrotactile Biofeedback in Subjects with Unilateral Vestibular Loss," to Experimental Brain Research.

Abstract

Subjects with unilateral vestibular loss exhibit motor control impairments as shown by body and limb deviation toward the affect side during gait. Biofeedback devices have been showed to improve postural control, especially when sensory information is limited by environmental conditions or pathologies such as unilateral vestibular loss. However, the extent to which BF could improve motor performance or learning while practicing a dynamic task such as narrow gait is still unknown. In this study 9 unilateral vestibular loss subjects practiced narrow gait in 2 practice sessions with and without wearing a trunk-tilt biofeedback device. The biofeedback device informed the subjects of their medial lateral angular tilt and tilt velocity during gait via vibration of the abdomen. From motion analysis and tilt data, the performance of the subjects practicing tandem gait were evaluated overtime and with and without biofeedback.

By practicing tandem gait, subjects reduced their trunk-tilt, center of mass displacement, variability of stepping, and frequency of stepping error. In both groups, use of biofeedback consistently increased postural stability during tandem gait. Use of tactile biofeedback consistently improved performance of unilateral vestibular loss subjects while they practiced narrow gait. However, one session of practice with biofeedback did not result in conclusive after-effects consistent with retention of motor performance without this additional biofeedback. Tactile biofeedback acts similar to natural sensory feedback in improving dynamic motor performance and not as a method to recalibrate motor performance to improve function after short-term use.

Introduction

Integration of vestibular, visual, and somatosensory information is fundamental to maintain balance and to perform motor tasks [1]. When sensory information is missing, as for example in subjects with unilateral vestibular loss (UVL), postural control is impaired and subjects show an increased reliance on visual and somatosensory information [2]. UVL is often a consequence of unilateral vestibular neurectomy to remove an acoustic neuroma [3]. After the neuroma removal, subjects undergo a period during which the central nervous system relearns how to cope with mismatching sensory information from vestibular, proprioceptive, and visual senses [4]. Although subjects show improvement of balance after surgery, [4], most UVL subjects, even years afterwards, continue to show balance and vestibular disorders such as 1) inability to stand with eyes closed on a sway-referenced surface [5], 2) body and limb deviation toward the affected side with eyes closed [2] and 4) difficulty balancing with eyes closed a) on one foot, b) in tandem stance, and c) in tandem gait [6].

Sensory information can be augmented by using a biofeedback (BF) system [7]. BF systems have been suggested to be beneficial when aimed to improve daily living tasks such as gait [8]. However, most of the published studies to date have investigated the use of BF systems during static or “quasi-static” tasks such as quiet or perturbed stance [9-13].

BF systems for postural control aim to encode some crucial kinematic or kinetic information not normally accessible to subjects into information useful for nervous system control of the task [7]. For example, during gait, information about trunk movement in the medial-lateral plane is crucial for postural stability [14].

Visual, acoustic, and tactile BF systems have been used successfully to improve stance balance in subjects lacking vestibular, visual, and somatosensory information [9], [15], [16], respectively. However, use of visual and acoustic BF systems, could interfere with the ability to deal with visual and acoustic information important for daily living. Thus, tactile BF may be more suitable than visual or acoustic BF for providing additional feedback to improve balance during daily living activities [17].

Improvements in specific motor tasks after practice with BF have been reported in many studies [18]. However, practice of a specific motor task itself stimulates brain plasticity and improves motor performances [2]. Thus, unless a control group is used to determine the extent of spontaneous learning, the effects of BF on retention of motor performance remain

inconclusive [19]. Knowing the extent to which BF practice facilitates retention of postural performance improvements could help determine if BF intervention should be temporary (used only during exercise sessions) or permanent (used as a prosthesis device).

In this study, the effect of augmented medial-lateral trunk tilt information via tactile BF during repetition of a tandem gait task was assessed in subjects with UVL. Further, a cross-over design in the experimental protocol was used to limit order effect when comparing the short-term retention effects of practicing tandem gait with and without trunk tilt BF.

Materials and Methods

Subjects

Nine UVL subjects (5 males and 4 females, age: 49 ± 11 yrs, height: 172 ± 10 cm, and weight: 89 ± 21 kg) participated in this experiment after signing an informed consent. This informed consent was approved by the academic, ethic committee and guaranteed the subjects'

rights according to the Declaration of Helsinki. All subjects were free from orthopedic and neurological diseases or disorders, except for the total vestibular loss on either the left (6 subjects) or the right (3 subjects) side. Subjects' demographics and pathologies are shown in Table 1.

As part of the cross-over experimental design, the subjects were divided into two groups such that the differences between averaged ages, heights, and weights in the two groups were not statistically significant ($p < 0.05$) when compared with a 2-tailed t test.

Apparatus

During the experiments, the subjects were asked to tandem-walk heel to toe, on a firm surface while a commercial metronome was set to "beep" at 30 beats per minute (0.5 Hz). To assure consistent cadence, subjects were asked to take one step for each beep. The subjects' kinematics was acquired using a Motion Analysis system with 8 Falcon cameras. A symmetric set of 20 markers was used (Figure 1). The markers were fixed above the eye, on the jaw joint, and on the acromion, elbow, wrist, great trochanter, knee, malleolus, fifth metatarsal, and hallux of each side of the subject. During all trials, the subjects were wearing a vibrotactile BF system [17] constituted of a vest with 4 columns of tactors (3 tactors per column) and a one-axis tilt sensor unit. The vest was placed around the trunk of the subject with an elastic girdle so that 2 columns of tactors were in contact with the left side of the subject's trunk and the other 2 columns in contact with the right side of the subject's trunk. The sensor unit was aligned so that it could sense the subject's medial-lateral (ML), trunk tilt. The sensor

Table 1 – Details on the UVL Subjects Involved in this Study.

Subject ID	Age	Sex	Years Post-surgery	Pathology	Side Affected	Group
1	60	F	12	Ac. Neuroma	Right	1
2	26	M	10	Skull Injury	Left	1
3	43	F	8	Ac. Neuroma	Left	1
4	46	F	8	Ac. Neuroma	Right	1
5	53	M	4	Ac. Neuroma	Right	1
6	56	M	3	Ac. Neuroma	Left	2
7	63	F	n/a	Ac. Neuroma	Left	2
8	42	M	7	Ac. Neuroma	Right	2
9	49	M	n/a	Labyrinthitis	Right	2

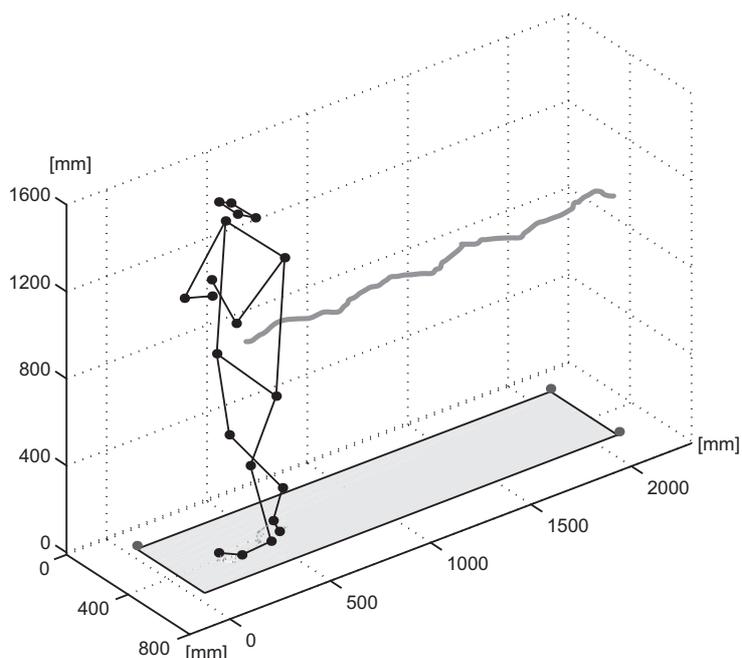


Figure 2 – 3D reconstruction of the experimental set-up. Markers from Motion Analysis are represented as black spheres. Trace of the center-of-mass, calculated from the Motion Analysis data and the subject's anthropometric measures, is also represented.

a Velcro™ belt. This position was preferred because it is close to the center of mass (COM) and minimally affected by artifacts such as breathing and heart beat. A computer (Macintosh Powerbook G3) was used to activate the tactors on the vest depending on the subject's ML, trunk tilt detected by the sensor unit. The tactors on each side were activated in pairs using a step-wise scheme depending on a combination of angular tilt and angular tilt velocity [21]. The lowest pair was activated when the sum of the measured tilt and one half of the measured tilt velocity exceeded a 2 degree "dead-zone", switching to the middle pair at 7 degrees and to the highest when exceeded 12 degrees (Figure 2). During the experiment, all subjects wore exactly the same type of polyester T-shirt so that the intensity of the vibration was as similar as possible for each subject. Data were acquired with a 120-Hz sample frequency from the tilt sensor and 60-Hz from the Motion Analysis system.

Procedure

Before starting the data collection, all UVL subjects learned how to perform tandem gait safely and correctly during a 5- to 10-min-long training period. During this training, all UVL subjects gradually learned how to take one step for each beat from the metronome while keeping their eyes closed and arms crossed. All subjects, at first, were very skeptical about their ability to tandem walk with eyes closed. However, after this very short training, all subjects were able to successfully complete all trials. At the very beginning of the training period, subjects had difficulty maintaining balance and made large lateral trunk movements. Subjects attempted to compensate by using wider lateral foot placements during gait. This

unit consisted of a rate gyroscope and a linear accelerometer. A specially developed algorithm combined these two inputs to produce an estimate of the subject's orientation to the vertical that was accurate to within 0.2 degrees. In particular, the angular velocity sensed by the gyroscope was high-pass filtered, integrated, and then summed to the low-pass filtered acceleration sensed by the accelerometer [20]. Before each experimental trial, the software allowed the experimenter to "zero" the instrumentation while the subject stood quietly in a vertical position. The sensor unit was mounted on the right side of the subjects at L5 level using

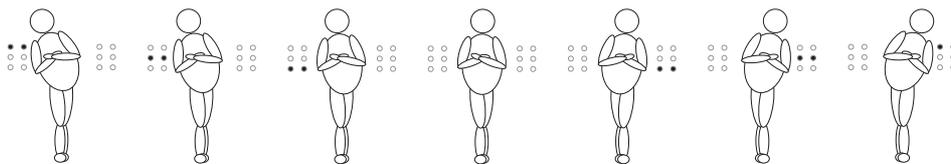


Figure 2 – Qualitative effect of trunk tilt on tactors activation. Circles on the left and right side of the human figure indicate the tactors. Dark-filled circles indicate tactors on, empty circles indicate tactors off.

effect was controlled by monitoring the ML deviation of the foot placement across the steps and providing verbal feedback to correct performance. Once subjects gained more confidence with narrow stance, they tended to walk as fast as possible so that their own body inertia helped to maintain balance. This effect was controlled by monitoring the actual frequency of step.

Following training, each subject performed a test session of 30 trials of tandem walking barefoot with eyes closed and arms crossed, taking one step for each beep of the metronome. A second identical test session of the experiment was performed two weeks after the first one. We refer to these two test sessions as “practice sessions” because we hypothesized that motor learning would occur during each session and that results from the second session would be influenced by practicing tandem gait in the first session.

In both sessions, the first 3 and the last 3 trials were performed with the tactile-BF device turned off. A cross-over design was used for the 24 middle trials. Group 1 performed the 24 middle trials of the first session with the tactile-BF device turned off and the 24 middle trials of the second session (two weeks later) with the device on. For Group 2 this order was reversed.

Each walking trial was 2.5 meters long so that the subjects could take at least 5 complete steps. Before the first session of the experiment, the subjects practiced the task for 5 to 10 minutes in order to get familiar with tandem walking. Subjects started practicing with eyes open and without the metronome, then with eyes closed, and finally with eyes closed and the metronome. Data collection started once the subjects understood the task and they demonstrated that they were able to perform such a challenging task. At the beginning of the practice period, all the subjects stated they would never be able to perform tandem walking with eyes closed, however all of them actually could achieve this for a couple of meters after the 10-minute practice period. During the experimental session, a safety spotter from our laboratory walked on one side of the subjects to catch them in case they lost balance.

Data- and Statistical- Analysis

From the kinematics data and the anthropometric measures of each subject, the 3D-coordinate of the COM during each trial was calculated according to [22-24]. Trunk tilt and

COM data were synchronized via recording of trigger signals from Motion Analysis. Steps were recognized from the position in time of the 6 markers on the feet. The first and the last stance phases of stepping were neglected for the calculation of the following parameters.

The ML SD of the COM was calculated for each trial and used as an indicator of subjects' ML postural stability. The standard deviation (SD) of ML tilt from the BF system sensor was calculated for each trial and used as an indicator of how much the subjects were able to limit their movement based on this feedback. In addition, the mean frequency error (i.e. the difference between the subjects' actual frequency of stepping and 0.5Hz) was calculated for each trial as well as the mean across steps of the feet ML distances during the double-stance phases. This mean ML feet distance and the mean frequency error were used as an indicator of the accuracy of the subjects' in performing tandem-walking. The parameters were averaged across the two cross-over groups to minimize the influence of possible order effect.

Linear regression was used to determine the statistical significance of change in the across practice trials. Simple paired t tests were also used to determine any significant short-term retention effect in terms of percentage change of the parameters between before and after each practice sessions (with and without BF). The parameters were considered independent for statistical purposes since they were obtained from independent measures, as a consequence Bonferroni correction was used only when paired t test were repeatedly applied to the same set of parameters. T-tests verified also that the percentage change of each parameter occurred across the two sessions of the experiment was not statistically significantly different between the two groups. In other words, that the changes in the parameters occurred after the subjects were exposed to both the session of the experiment were not significantly different.

Results

Immediate Effects of BF

All subjects in both groups improved their stability as soon as the BF device was turned on. Specifically, in the first three trials with BF, COM displacement was significantly reduced by 3.8%, trunk tilt by 17.8%, and mean ML feet distance by 20% compared to the previous three trials without BF. Frequency error was the only parameter that increased (by 34.5%) when BF was turned on. Figure 3 shows raw data of COM displacement and trunk tilt from one representative subject with and without BF. Note the decrease in variability and amplitude of COM displacement and trunk tilt occurring with BF.

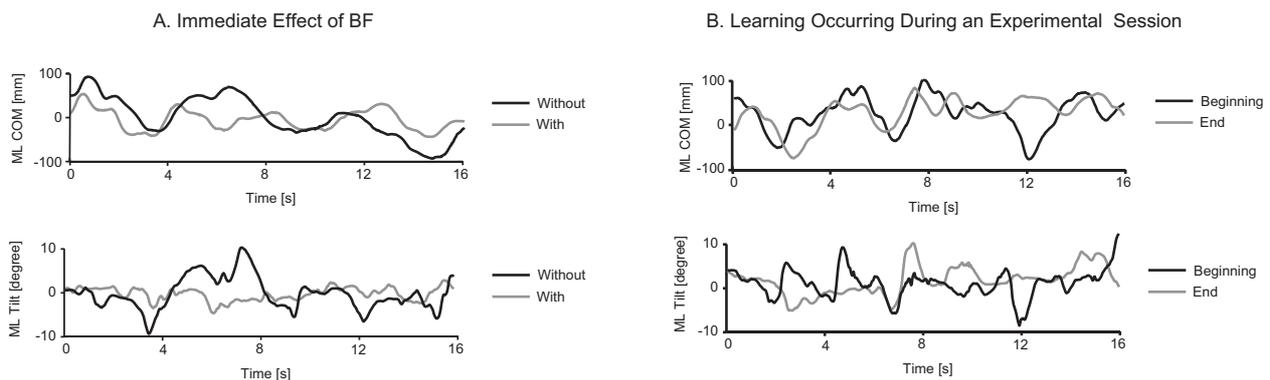


Figure 3 – Panel A shows the immediate effect of BF on lateral trunk tilt and COM displacement from one representative subject. Panel B shows the effect of learning during one experimental session on lateral trunk tilt and COM displacement from one representative subject.

Table 2 – Parameter at the very beginning (before the first practice session) and at the very end (after the second practice session). Each value corresponds to the average of three trials.

Subject #	COM SD [mm]		Tilt SD [degree]		Mean feet distance [mm]		Freq. Error [Hz]	
	Beginning	End	Beginning	End	Beginning	End	Beginning	End
1	44.89	39.65	4.85	3.38	64.71	34.70	0.38	0.01
2	57.01	53.48	5.73	4.25	67.46	53.52	0.13	0.07
3	83.06	35.85	6.92	5.36	71.59	68.90	0.16	0.01
4	51.45	28.31	6.47	2.82	76.14	33.71	0.09	0.01
5	65.67	27.90	4.18	2.85	79.71	48.26	0.33	0.09
6	73.03	62.25	3.84	3.32	59.91	66.98	0.32	0.19
7	27.59	27.23	2.63	2.24	48.45	39.05	0.08	0.02
8	47.66	25.17	2.78	1.75	37.27	36.05	0.08	0.14
9	49.27	41.09	3.25	2.32	70.39	43.47	0.24	0.05
Mean(SD)	55.5(16.5)	37.9(12.9)	4.52(1.58)	3.14(1.11)	64.0(13.6)	47.1(13.4)	0.20(0.12)	0.07(0.07)

Effect of Practicing Tandem Gait on the 4 Parameters Analyzed

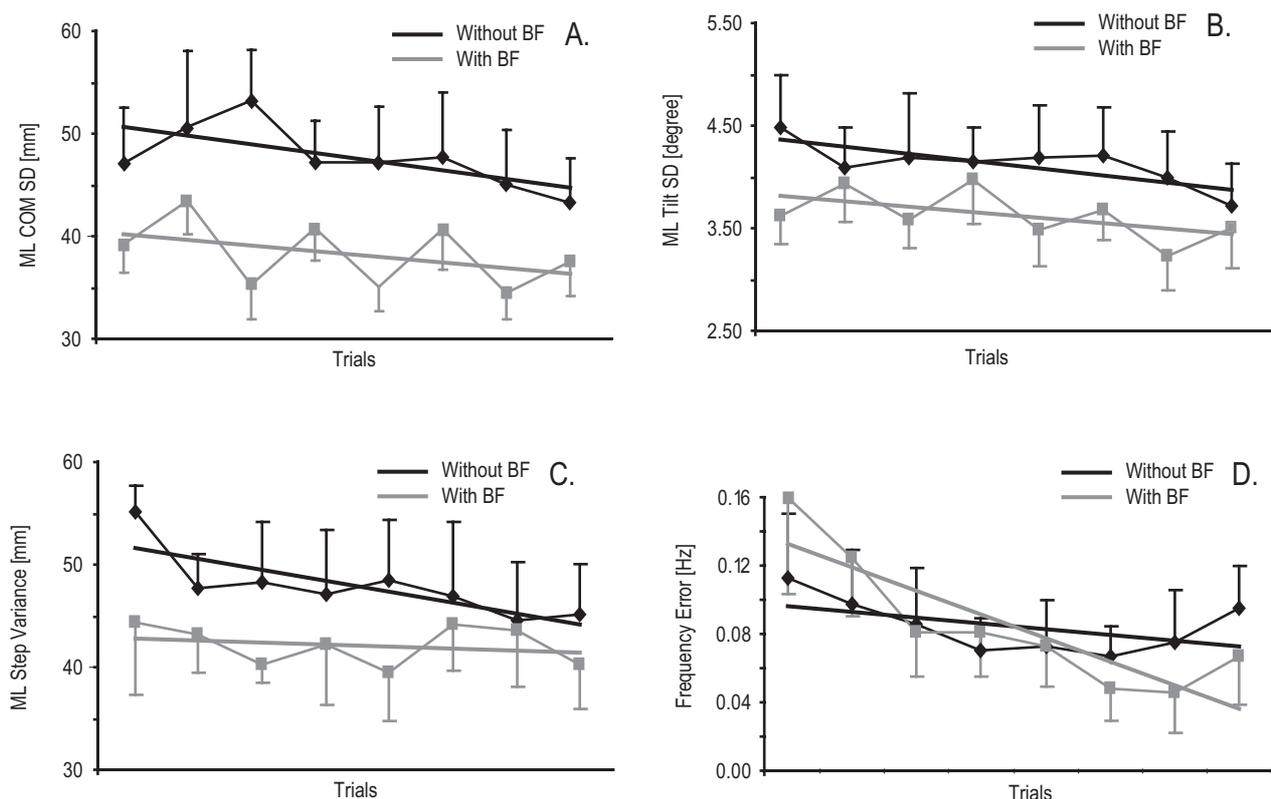


Figure 4 – Effect of practicing tandem gait across trials. Each value represents the average among the subjects of three consecutive trials. Error bars represent standard errors.

Effects of Practicing Tandem Gait

During the experiment, all subjects improved their stability with repetition of tandem gait trials; COM displacement, trunk tilt SD, mean ML feet distance, and frequency error were significantly lower for all subjects in the three trials recorded after the two practice sessions than in the three trials recorded before the two practice sessions (average values are reported in Table 2). Figure 3B shows some raw data of COM displacement and trunk tilt from one representative subject in the first and last trial of the first experimental sessions. Note the decrease in variability and amplitude of trunk tilt and COM displacement occurring with practice.

Effects of Practicing Tandem Gait in the Session without BF

The subjects' COM displacement SD significantly decreased over the course of the session without BF (Fig. 4A). The regression slope was negative and differed significantly from zero ($p < 0.05$), and the linear regression accounted for 70% of the variance. The subjects' SD of trunk tilt also showed significant reduction while practicing without BF (Fig. 4B). The slope of the linear regression coefficient of the tilt SD values across trials was negative and was significantly different from zero ($p < 0.05$), and the linear regression accounted for 60% of the total variation. Subjects' mean ML feet distance (Fig. 4C) also significantly decreased over time while practicing tandem

Table 3 – Short-term retention after practicing tandem gait without and with BF. Each value represents the percentage change occurred between before and after each practice session. * indicate statistical significance of the overall mean percentage change.

Subject #	COM SD %		Tilt SD %		Mean feet distance %		Freq. Error %	
	Without	With	Without	With	Without	With	Without	With
1	-4.40	13.61	-37.34	-7.98	-44.67	-13.74	-86.94	-78.89
2	-18.65	11.14	-22.31	67.03	-25.12	1.76	-71.53	29.26
3	2.40	-9.06	-44.51	-25.83	-64.48	14.40	-91.70	-89.00
4	-32.99	-23.08	-61.07	-26.22	-41.76	14.86	-64.40	-81.27
5	-16.40	-39.05	-44.65	8.88	-61.57	-11.40	-98.06	38.00
6	-7.24	-19.86	-31.83	15.83	-23.38	4.07	-49.50	-13.57
7	-15.21	-1.27	-22.85	5.34	-11.57	-6.32	-84.96	-65.50
8	-4.03	-14.20	-23.70	-1.19	-25.86	3.68	87.60	-29.42
9	-2.31	-9.80	-12.02	-31.44	-3.27	8.30	-41.43	-69.82
Mean(SD)	*-10.9(10.9)	-10.2(16.6)	*-33.4(15.0)	0.5(30.1)	*-33.5(21.1)	1.7(10.4)	*-55.7(57.0)	*-40.0(48.5)

gait without BF, with 60% of the variance accounted for by the linear regression. The subjects' stepping frequency error, however, did not significantly change across trials by practicing tandem gait without BF ($p > 0.05$; Fig4D).

Effects of Practicing Tandem Gait in the Session with BF

During the session with BF, subjects consistently exhibited a smaller trunk COM displacement SD, tilt SD, and mean ML feet distance than in the session without BF ($p < 0.05$; Figure 4A-C). Although the linear regression slope was negative for these three parameters, the statistically analysis showed that the regression slope was not significantly different from zero. The variance accounted for by the linear regression was 17% for COM displacement, 30% for trunk tilt, and 6% for mean ML feet distance. In contrast, the step frequency error did improve with practice (regression slope negative and significantly different from zero, $p < 0.01$). The variance accounted for by the linear regression was 75% for step frequency error.

Short-term Retention Effect of Practicing Tandem Gait

Short-term retention, i.e. the difference between the performances at the beginning and at the end of each session, was higher after practicing without BF than after practicing with BF. Table 3 reports the percentage changes between the averages of the first and last three trials of each session for each parameter. Practicing without BF the significantly reduced trunk COM displacement SD, tilt SD, mean ML feet distance, and step frequency error in performing tandem gait (Table 3). Practicing with BF, only frequency error showed significant improvements (Table 3). COM displacement SD decreased for most of the subjects after practicing with BF, however this change was not significant ($p = 0.08$; Table 3).

Discussion

Motor Learning During Tandem Gait Practice

After practicing tandem gait, all UVL subjects improved their performance in terms of postural stability and stepping accuracy. These improvements included 1) an increased ML stability, shown by the reduction of the trunk tilt and COM SD; and 2) a higher accuracy in maintaining the tandem position of the feet while walking, shown by the reduction of the ML variability of stepping, as well as a high accuracy in stepping to the metronome rhythm. These results suggest that with practice, subjects with UVL can learn to better control their posture during a complex task such as tandem gait. In fact, the lower variability of lateral stepping placement represents reduced stepping deviation toward the affected side which is a typical clinical syndrome of UVL subjects [25] [6]. This improved tandem stepping performance may be due to reduced vestibular-somatosensory conflict and/or increased gain of the proprioceptive postural loop [26] or to improved feedforward control of the complex multi-segmental task [27].

Practice Sessions with and without BF

Thanks to the cross-over design adopted for this experiment, we were able to cancel out the potential effect of session order by averaging across sessions (with and without BF). In other words, the results reported in Figure 4 are not influenced by the order effect of trials with and without BF so that the effect of spontaneous learning, occurring when repeating a task, was equally divided between the 2 sessions. Most previous studies of the effects of BF on postural control did not control for such a practice effect and attributed all of the improvement in performance to effects of BF [11].

During trials with BF, all subjects consistently achieved better performances than in trials without BF. In particular, trunk stability and stepping accuracy were better in trials with BF than in trials without BF. These results suggest that UVL subjects were able to effectively use BF to improve their performance during tandem gait consistent with previous studies with other, less dynamic tasks such as stance [9;28]. Furthermore, this improved performance occurred at the start of the very first trials with the BF device and did not require a period of practice to be effective. This immediate improvement of postural control with BF is consistent with our previous studies of effects of audio-biofeedback on stance posture in subjects with bilateral vestibular loss and controls [10;29]. During the practice trials in the session with BF, UVL subjects

did not increase their relative stability as much as during the practice trials without BF. This result was probably due to the significantly greater stability level induced by the BF leaving a smaller potential for additional improvement (a floor effect). However, BF consistently improved the accuracy of the tandem gait performance across practice trials. Specifically, the frequency error was initially larger in trials with BF than in trials without BF although, in the end, the error was significantly lower (Fig. 4D). The higher error in frequency of stepping shown at the beginning of the session with BF may be due to the subjects' initial inability to pay enough attention to the metronome and the BF at the same time. Over time, however, all subjects could decrease this error to the point that they achieved the best performance, in terms of frequency error, in the trials with BF. This particular result suggests that the use of BF becomes more automatic (i.e. requires less attention) with practice [30].

Short-term Retention of Motor Learning

Immediately after practice, subjects retained their performance improvements achieved by practicing tandem gait without BF in terms of trunk stability and accuracy of foot placement, as shown by the four parameters analyzed in Table 3. This result is further evidence of the extensive potential for motor learning in UVL subjects [31;32]. Only limited short-term retention effects were evident after practice in the session with BF. Only one out of four parameters, the frequency error, was found to retain significant improvements without BF, after practicing with BF (Table 3). This result may suggest that, immediately after turning the BF device off, subjects retained a higher level of cognitive attention; attention that they then focused upon the only remaining external cue, the metronome beat. As a consequence, they more accurately controlled the frequency of stepping.

Three factors may have limited short-term retention of performance in the other three parameters (tilt SD, COM SD, and mean ML feet distance) after practice with BF: 1) the short duration (about 10 minutes) of the practice; 2) the greater number of trials performed without BF (30) than with BF (24); in fact, tandem gait without BF was both the task for practicing and for verifying retention of performance; and 3) the experimental protocol was not purposely designed to facilitate transfer and retention of postural performance. To be more effective, the protocol could have alternated trials with BF and without BF so that, at the beginning, trials with BF were more frequent, and then, over time, trials with BF were gradually diminished [33].

Conclusions

UVL subjects can integrate vibrotactile BF information in their postural control to effectively improve stability and performance accuracy during tandem gait. This improvement occurs as soon as the BF device is turned on and does not require a period of practice. However, this integration of augmented sensory information becomes more automatic with practice over time. Thus, vibrotactile BF acts similarly to natural sensory feedback in improving dynamic motor

performance and not as a method to recalibrate motor performance to improve function after short-term use.

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Chapter 10

Conclusions: Effects of Biofeedback on Postural Control and Potential Impact in Motor Rehabilitation Therapy



Conclusions

Sensory integration is fundamental for the control of posture (see Chapter 1). When sensory information is reduced, such as in vestibular loss subjects, sensory integration is impaired. One way to increase sensory information for the control of posture is using biofeedback devices. The design of biofeedback devices, as well as the design of the experimental protocols aimed at evaluating the effect of biofeedback on learning and retention of postural control performances, face many challenges. Specifically, a two double-blinded, randomized experimental design with both dynamic tasks and static tasks, seems to be the best protocol to determine the effectiveness and potential impact in the rehabilitation field of biofeedback systems.

Visual-biofeedback of center-of-pressure displacement is the biofeedback system that traditionally has received the most interest for experimentation on postural control, and it is currently used for balance rehabilitation in stance. During stance, trunk acceleration is highly correlated with center-of-pressure displacement (see Chapter 2). Thus, an audio-biofeedback system, coding trunk acceleration into a stereo sound modulation, may be an alternative to visual-biofeedback of center-of-pressure displacement. This new, audio-biofeedback device is lighter and more cost-effective than traditional visual-biofeedback systems; further it is portable, so it can be used also during complex dynamic tasks, and does not take over vision.

Using this audio-biofeedback, healthy subjects reduced sway by increasing control of posture when sensory information available is limited (see Chapter 3). More specifically, this sway reduction occurred without increasing muscle activity or muscle co-contraction and was caused by an enhancement of the closed-loop control of posture. Furthermore, the effect of audio-biofeedback on postural sway was found to be direction-specific (see Chapter 4).

Also, bilateral vestibular loss subjects reduced sway by increasing control of posture when sensory information was limited using this audio-biofeedback (see Chapter 5). Furthermore, bilateral vestibular loss subjects could take advantage of audio-biofeedback more than controls when visual and somatosensory information were limited. In addition, the benefit that each subject could take from audio-biofeedback, was related to their relative dependence on visual, somatosensory, and vestibular information (see Chapter 6) suggesting use of audio-biofeedback specifically compensates for lack of vestibular, somatosensory, and visual sensory information.

The efficacy of biofeedback and the strategy of postural responses evoked by the biofeedback were found to depend on the biofeedback design (see Chapter 7). In fact, depending on the representation and coding of the feedback variable, users were able to achieve a different performance level and favor a different postural strategy in response to the biofeedback.

Audio-biofeedback of center-of-pressure improved balance also during dynamic tasks such as stance perturbed by continuously, randomly oscillating surface (see Chapter 8). The amount of information from biofeedback needed by a subject to improve balance was found to depend on the challenge of the task. With practice of stance on a moving surface, motor learning improved subjects' postural responses. However, even after practicing, audio-biofeedback continued to be effective in reducing postural responses at low frequencies (<0.8Hz), suggesting that, with simple motor learning, subjects are not capable to achieve the same level of performance as with audio-biofeedback.

Another dynamic task, tandem gait, was used to determine the effects of a tactile-biofeedback in subjects with unilateral vestibular loss. Tactile-biofeedback on the lateral trunk to indicate lateral postural sway was found to improve subjects' performance while practicing tandem gait (see Chapter 9). However, one session of practice with biofeedback did not result in many after-effects consistent with retention of motor performance without this additional biofeedback. Our results suggest that tactile-biofeedback in tandem gait acts similar to natural sensory feedback in immediately improving dynamic motor performance and not as a method to recalibrate motor performance to improve dynamic balance function after short-term use.

The results and conclusions reported above constitute a brief summary of this thesis. The above-mentioned results show how different biofeedback designs were found to improve balance and motor performance in different postural static and dynamic tasks. During this experimentation, we showed how crucial is the design of a biofeedback system since it determines 1) the improvement that subjects will be able to achieve and 2) which postural strategy will be responsible for this improvement. As a consequence, in order to achieve the best postural performance without eliciting erroneous strategies for the control of posture, the biofeedback design should be customized for each subject and task. Further, results from practicing with biofeedback suggest that biofeedback 1) can still be useful after spontaneous learning occurs and 2) may favor motor learning. However, a customized protocol is necessary to maximize balance improvement and its potential retention for rehabilitation. Finally, the findings presented in this thesis constitute clear evidence that biofeedback 1) can increase basic knowledge about sensory integration and motor control, 2) has the potential, once conveniently customized, to become a helpful tool for balance and motor rehabilitation and training, and 3) needs to be equipped with training protocols able to favor motor learning and control for erroneous control of posture.

The conclusions presented in this thesis, foresee two promising areas of interest for further study on biofeedback, one related to the biofeedback design and one related to the biofeedback application. In particular, biofeedback design can be improved by using virtual reality. Indeed, virtual reality is in essence immersive, multi-modal, attractive, easy-to-understand, intuitive, and entertaining; further, it permits to recreate real life situations. Such features are highly desirable in a biofeedback system for augmenting subjects' motivation and attention which are known to favor brain plasticity and for testing biofeedback in real, controlled daily-life situation. Another promising area of interest for studies on biofeedback regards its experimentation on other classes of subjects with motor impairments, such as Parkinson's or after-stroke subjects. In fact, in this context, biofeedback experimentation could both help understanding the extent to which motor impairments are related to sensory integration deficits (which, to date, is not totally understood for these classes of subject) and help increasing the quality of life of these subjects by improving their postural performances.

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