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Determining the effectiveness of a vibrotactile balance prosthesis

Robert J. Peterka[a,*], Conrad Wall III[b] and Erna Kentala[b,c]
[a]*Neurological Sciences Institute, Oregon Health & Science University, Portland, OR, USA*
[b]*Massachusetts Eye & Ear Infirmary, Boston, MA, USA*
[c]*Helsinki University Hospital, Helsinki, Finland*

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Abstract. We present a quantitative method for characterizing the effectiveness of a balance prosthesis based on tactile vibrators. The balance prosthesis used an array of 12 tactile vibrators (tactors) placed on the anterior and posterior surfaces of the torso to provide body orientation feedback related to the angular position and velocity of anterior-posterior body sway. Body sway was evoked in subjects with normal sensory function and in vestibular loss subjects by rotating the support surface upon which a test subject stood with eyes closed. Tests were performed both with (tactor trials) and without (control trials) the prosthesis activated. Several amplitudes of support surface stimulation were presented with each stimulus following a pseudorandom motion profile. For each stimulus amplitude, a transfer function analysis characterized the amplitude (gain) and timing (phase) of body sway evoked by the support surface stimulus over a frequency range of 0.017 to 2.2 Hz. A comparison of transfer function results from the control trials of normal subjects with results from tactor trials of vestibular loss subjects provided a quantitative measure of the effectiveness of the balance prosthesis in substituting for missing vestibular information. Although this method was illustrated using a specific balance prosthesis, the method is general and could be applied to balance prostheses that utilize other technology.

Keywords: Vestibular, balance prosthesis, posture, orientation, vibrotactile, human

1. Introduction

There is increasing interest in developing balance prostheses that can ameliorate some of the deficits of stance and movement control associated with abnormal vestibular function and other sensory deficits [12, 14]. A balance prosthesis is defined as any technological device that is meant to improve balance control through enhancement or restoration of sensory function, or by providing orientation information via an alternative sensory modality (i.e. sensory substitution). Independent of the particular technology that is used for the prosthesis, the need exists for methodology to assess the effectiveness of the prosthesis in compensating for a sensory deficit. The goal of this study was to demonstrate a quantitative method that can characterize the extent to which a balance prosthesis restores a normal pattern of postural control in subjects with absent vestibular function and enhances the ability of subjects with normal vestibular function to remain oriented to earth vertical despite postural perturbations.

One measure of the functional utility of a balance prosthesis is its ability to restore normal control of the upright stance position. The question is how to quantify stance control behavior so that meaningful improvements in balance control due to prosthesis use can be easily determined. Quantification of stance control is often based on spontaneous body sway measures during stance on a stable surface [11], but this method is problematic because there are often only small differences in spontaneous sway between normal

*Corresponding author: Robert J. Peterka, Ph.D., Neurological Sciences Institute, OHSU West Campus, Building 1, 505 NW 185th Avenue, Beaverton, OR 97006, USA. Tel.: +1 503 418 2616; Fax: +1 503 418 2501; E-mail: peterkar@ohsu.edu.

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and vestibular loss (VL) subjects [7]. In more challenging stance conditions presented in dynamic posturography tests [7], such as eyes-closed stance on a sway-referenced surface,[1] large differences between normal and VL subjects are revealed because normal subjects can maintain stance, but VL subjects consistently fall. If VL subjects using a prosthesis were able to maintain stance on a sway-referenced surface with eyes closed, this would provide strong evidence for the effectiveness of the prosthesis. However, if falls persisted, it would not be possible to determine if the prosthesis was completely ineffective, or if it was just not effective enough to restore balance in a very challenging test condition.

What is needed for an evaluation of the prosthesis is a test that demonstrates large differences between normal and abnormal stance control without evoking falls. Such a test is suggested by the results from recent investigations that have shown large and systematic differences between the stance control behavior in subjects with normal sensory function and bilaterally absent vestibular function [8,9]. In these studies, body sway was perturbed by low amplitude, pseudorandom angular rotations of the support surface (SS) upon which the subjects stood. Both normal and VL subjects tended to orient to the rotating SS, but the amount of body sway evoked by SS rotation was greater in VL than in normal subjects. Additionally, as the amplitude of the surface rotation increased, the amplitude of body sway in VL subjects increased in proportion to the SS stimulus amplitude. In contrast, the amplitude of body sway in normal subjects showed a more limited increase with increasing SS stimulus amplitude and reached an upper limit beyond which there was no further increase in sway amplitude with increasing SS stimulus amplitude. The same general pattern of results occurred when eyes were open or closed, but the evoked sway was larger with eye closed.

A transfer function analysis was used to quantify the stimulus-response behavior of the postural control system using sway evoked by SS rotations. Transfer function analysis is a common tool for the characterization of linear control systems [1] and has proved useful for the analysis of human postural control [8,9]. A transfer function is represented by gain and phase functions that vary across frequency. At each frequency, the gain value indicates the ratio of body sway amplitude to SS stimulus amplitude and the phase value indicates the relative timing between body sway and SS rotation. A gain function equal to 1.0 and phase function of zero degrees across all frequencies would indicate that the subject remained perfectly aligned with the SS. Gains of zero across all frequencies would indicate that subjects remained aligned with earth-vertical and, therefore, were not perturbed by the SS stimulus. In actuality, the balance of both normal and VL subjects is perturbed by a SS stimulus with the gain and phase varying with frequency in a manner that characterizes the dynamic behavior of the postural control system [8, 9].

A subset of results from an earlier study [8] illustrates the large differences between transfer functions from normal and VL subjects. Figure 1(A) shows a set of 4 transfer function gain and phase curves estimated by taking the mean results from eight subjects with normal sensory function during eyes-closed stance on the rotating SS. The 4 transfer functions were obtained using different amplitudes of pseudorandom SS stimulus motion ranging from 1° to 8° peak-to-peak. The gain curves showed a systematic decrease with increasing SS amplitude consistent with these subjects being less responsive to the SS stimulus as the SS amplitude increased. As shown in a previous study [8], this gain decrease is consistent with normal subjects relying to a greater extent on vestibular orientation information (signaling body orientation in space) and less on proprioceptive information (signaling body orientation relative to the SS) as the amplitude of the surface motion increased. Consistent with this interpretation and as shown in Fig. 1(B), a subject with bilateral VL does not show this gain decrease with increasing SS stimulus amplitude.

The large disparity in postural dynamics between normal and VL subjects shown in Fig. 1 is the basis for our quantitative method for assessing a balance prosthesis. Specifically, if a balance prosthesis was able to fully restore or substitute for absent vestibular function, then a VL subject using that prosthesis would show the same pattern of transfer function changes with changing SS stimulus amplitude as a subject with normal sensory function. If the prosthesis was able to improve, but not fully restore normal balance control, then this would be revealed by the extent to which the transfer function changes matched those of normal subjects. In addition, because the transfer function analysis provides a representation of changes across a range of fre-

[1] Surface sway-referencing refers to a method in which the support surface continuously rotates in direct proportion to the body sway angle. Sway-referencing maintains a nearly constant ankle joint angle and thereby decouples the normal relationship between ankle angle and body sway. This decoupling greatly reduces the contribution of proprioceptive orientation cues to balance control.

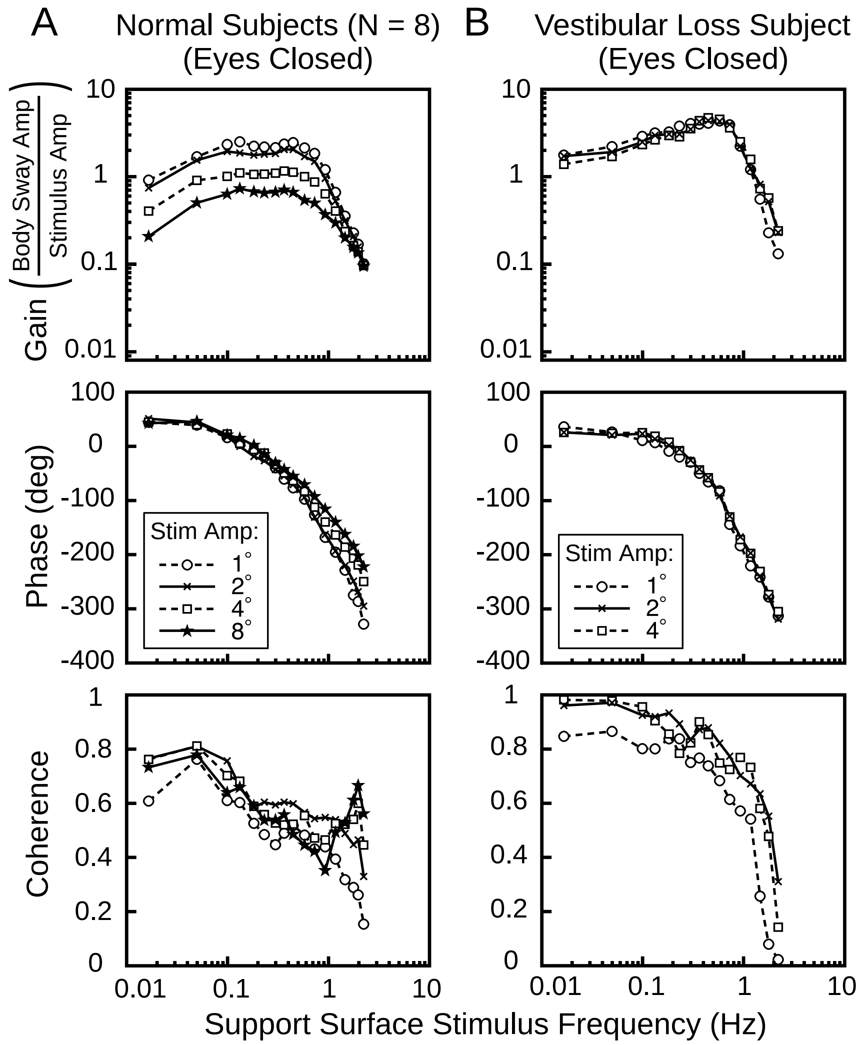

Fig. 1. Family of transfer functions that characterize body sway evoked by support surface rotations during eyes closed stance. A. Transfer functions and coherence functions obtained from subjects with normal sensory function (mean of 8 subjects – data from [8]) in response to surface rotations with peak-to-peak amplitudes of 1°, 2°, 4°, and 8°. B. Transfer functions and coherence functions obtained from a subject with bilaterally absent vestibular function in response to surface rotations with peak-to-peak amplitudes of 1°, 2°, and 4°.

quencies, our proposed method offers the potential for revealing the frequency-dependent actions of a balance prosthesis that could give insight into the nature of information provided by the prosthesis.

Our quantitative method for characterizing the effectiveness of a balance prosthesis is demonstrated by evaluating changes in postural control in normal and VL subjects due to use of a vibrotactile prosthesis [6, 15,16]. This prosthesis is based on the principle of sensory substitution whereby information about body orientation and motion is conveyed to the subject via patterns of vibration applied through an array of tactile vibrators held against the surface of the subject's torso.

2. Methods

2.1. Subjects

Six healthy subjects with no history of peripheral or central vestibular disorders (3 male, 3 female, mean age 35 years ± 8 SD) and 5 subjects with severe VL (2 male, 3 female, mean age 52 years ± 11 SD) participated in this study. Subjects gave their informed consent prior to being tested using a protocol approved by the Institutional Review Board at Oregon Health & Science University. Table 1 provides additional information about the VL subjects.

Table 1
Bilateral vestibular loss subjects

Age (years)	Duration of loss (years)	Horizontal VOR Gain* 0.05 Hz	Horizontal VOR Gain* 0.2 Hz	Cause of Loss
36	5	0.14	0.42	Auto Immune
48	11	0.03	0.05	Viral
55	9	0.01	0.03	Viral
59	8	0.18	0.35	Gentamicin Ototoxicity
63	1	0.02	0.08	Viral

*Normal VOR gain range is 0.39–1.02 for 0.05 Hz and 0.40–1.02 for 0.2 Hz [10].

2.2. *Support surface stimulus*

Subjects stood on a SS with their feet placed at a comfortable distance apart and with the ankle joint axes aligned with the rotational axis of the SS. The SS orientation was controlled by a servomotor operating under closed-loop position feedback control that could rotate the surface producing anterior-posterior (AP) tilts of the surface (critically damped 5 Hz bandwidth). The wide bandwidth and servo control motor insured that the SS stimuli used in this study were accurately reproduced.

A backboard assembly was mounted on the SS to insure that subjects swayed as a single-link inverted pendulum (Fig. 2). The weight of the backboard was supported on bearings whose axes were aligned with the SS rotation axis and the ankle joint axes. Subjects wore this backboard like a backpack with straps around the shoulders and waist, with additional straps above the knees, and with a head rest. The AP rotational position of the backboard with respect to the SS was measured by a potentiometer and the rotational velocity by a rate sensor (Watson Industries, Au Claire, WI). The rotational position and velocity of the backboard with respect to earth-vertical were also measured by a custom inertial sensor [16], and the signals from this sensor were used to drive the vibrotactile balance prosthesis. The body sway measures from the backboard and the inertial sensor were completely redundant because of the simple biomechanics imposed by the use of the backboard. Specifically, the backboard insured that all body sway occurred about the ankle joint axis so that the angular tilts of the legs, pelvis, trunk, and head were always equal. The backboard also insured that the inertial sensor, used to drive the balance prosthesis, was providing a signal related to the body center-of-mass (COM) angular motion. Without changing the principles of our proposed method for prosthesis evaluation, future test variations could include freestanding test conditions and/or the use of multiple inertial sensors from which body COM could be derived.

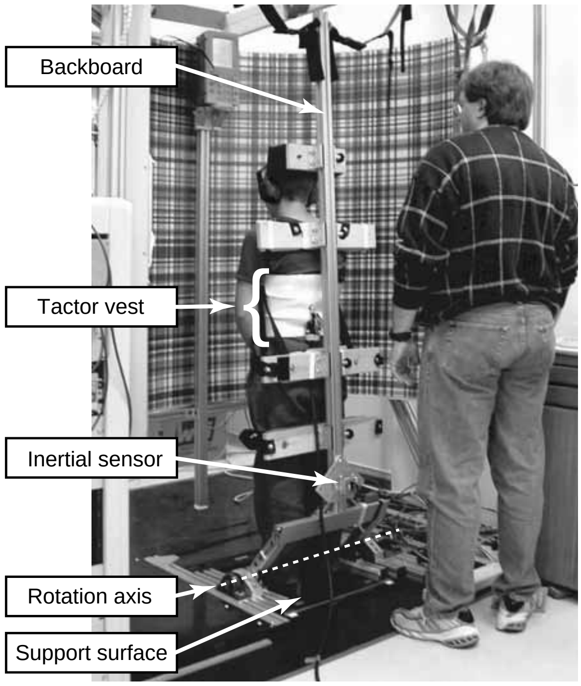

Fig. 2. Experimental setup showing a backboard-supported subject wearing a "tactor vest" that includes an array of tactile vibrators. The subject stands on a support surface with her ankle joint axis aligned with the rotation axis of both the backboard and the support surface (dashed line). The backboard constrains the subject to sway as a single-link inverted pendulum with sway occurring only in the anterior-posterior (AP) direction. A servomotor mounted below the support surface generates a rotational motion of the support surface that perturbs the subject's balance. Body sway is measured by an inertial sensor mounted to the backboard. Signals from a linear accelerometer and rate gyro in the inertial sensor are processed to provide measures of AP body tilt angle and tilt velocity. Tilt and tilt velocity measures are used to drive the tactors in order to convey information about body sway to the subject.

To characterize postural dynamics over a wide bandwidth of frequencies, the support surface was tilted according to a pseudorandom stimulus based on a pseudorandom ternary sequence [3]. The SS stimulus had an identical time course to the pseudorandom stimulus used previously to assess postural dynamics [8]. The stimulus was periodic with a period of 60.5 s and 6 consecutive cycles of the stimulus were presented for each test. A sampled pseudorandom waveform was delivered to the SS servo mechanism at a rate of 100 samples/s, and the actual angular rotation of the surface was measured with a potentiometer and sampled at 100/s. The backboard angular position and velocity was also sampled at 100/s.

All test trials were performed with eyes closed. Different SS stimulus amplitudes were used on different trials. Normal subjects performed trials with 2° and 8° peak-to-peak surface rotations. VL subjects performed trials with 1°, 2°, 4°, and 8° peak-to-peak surface rotations. None of the VL subjects were able to complete 8° trials without falling early in the trials. VL subjects occasionally fell on 4° trials, but were able to complete these 4° trails with repetition.

Normal subjects always performed the 2° trials prior to the 8° trials. VL subjects performed trials sequentially from 1° to 8°, and then repeated 2° trials to test for learning effects. Each test trail of a given amplitude was repeated twice with one trial being performed without the use of the balance prosthesis and the other with the prosthesis activated. The ordering of these 2 trials (prosthesis on before off, or off before on) was kept the same for each subject, but the ordering was reversed for the next subject tested.

2.3. *Vibrotactile balance prosthesis*

The balance prosthesis contained three major elements: an inertial instrumentation package [16], a computer serving as a signal processor (Macintosh PowerBook, Apple Computer, Inc.), and a vibrotactile array. All input, output, and signal processing was programmed using LabView (National Instruments Corporation, Austin, TX). The inertial sensor and vibrotactile array used in this experiment were designed to sense and provide feedback only for body sway in the AP direction.

The inertial instrumentation consisted of a micromechanical linear accelerometer and rate gyroscope (gyro) that was attached to the backboard about 35 cm above ankle height. Details regarding the methods and motivation for processing the signals provided by the inertial instrumentation are provided in [15]. Briefly, the accelerometer and gyro provided linear acceleration and angular rate information, respectively, about the subject's AP body tilt. A tilt estimate was calculated by adding the integrated, high-pass filtered gyro signal to the low-pass filtered accelerometer signal. If the gyro signal alone were integrated to estimate body tilt, the output would have a "drift" error due to the integration of an unwanted noise or "bias" term inherent in rate-gyro sensors. If the accelerometer signal alone were used as a tilt estimate, the output would have an error due to inclusion of a tangential linear acceleration component in addition to the desired tilt component due to sensing the inclination of the accelerometer relative to the gravitational force vector. High-pass filtering of the gyro signal removed the drift error, but preserved high-frequency tilt information. Low-pass filtering of the linear accelerometer signal removed the error due to tangential linear acceleration, but preserved low-frequency tilt information. These two filtered signals, when combined, provided the final measure of angular body tilt. This tilt measure has been shown to give a very accurate and nearly drift free estimate of orientation of the instrument package with respect to gravity vertical over the frequency range of normal body sway [15]. The combination of this tilt measure with the tilt velocity measure from the rate gyro was used to drive the vibrotactile display.

The vibrotactile display consisted of 12 small tactile vibrators ($2.7 \times 1.9 \times 1$ cm) called tactors (Audiological Engineering, Cambridge, MA), held against the anterior and posterior surfaces of the torso by a wide elastic fabric wrapped around the torso. The tactors were arranged in pairs (3 pairs anterior and 3 pairs posterior). On both the anterior and the posterior surfaces of the torso, the lowest pair was located at approximately the height of the umbilicus with the tactors located 3.8 cm to the right and to the left of the midline. The second and third pairs were located 3.8 cm and 7.6 cm, respectively, above the lowest pair on both the anterior and the posterior surfaces of the torso. Thus there were 2 vertical columns of 3 tactors each on both the anterior and the posterior surfaces of the lower torso.

We felt that it would simplify the subject's task of generating a corrective response if the information provided by the tactor array could be mapped directly into the torque required for postural corrections. Therefore, a combination of tilt and tilt velocity was used as an input signal to drive the tactor activation because it is well known that a control system designed to stabilize an inverted pendulum must generate corrective torque in proportion to a combination of both tilt and tilt velocity [5].

A step-wise coding scheme was used to activate pairs of tactors. Each pair of tactors was activated in an all-or-none manner with activation determined by the sum of the measured tilt and one half of the measured tilt velocity. Only one pair of tactors was activated at a time, and increasing tilt and tilt velocity activated pairs located higher up on the torso. Tactors on the anterior surface of the torso were activated when the sum of the tilt angle and tilt velocity was greater than a threshold value (as generally occurred during forward sway), and tactors on the posterior surface of the torso were activated when the sum was less than a threshold value (generally during backward sway). The range over which a tactor pair was activated was set individually for each subject, based on their maximum forward and backward tilt range. To account for normal body motion, no tactors fired in a dead-zone defined by a threshold value corresponding to a static (zero velocity) tilt of one degree in both the anterior and posterior directions. With increasing body tilt and tilt velocity the activated tactors moved from the bottom pair to the top pair in a stepwise manner. The top pair of tactors was set to fire when the subject reached 80% of their limits of stability. Only the spatial location of the tactor pair was used to represent the information proved by the sum of tilt and tilt velocity. Vibration intensity, as determined by amplitude and frequency of the input signal to the tactors, was kept constant and was set at the highest current rating (200 mA) of the tactile vibrators, as recommended by the manufacturer. With these parameters, all subjects were able to perceive the tactor vibrations, and no subjects reported aversive effects.

2.4. Transfer and coherence function analysis

The estimation of transfer functions relating body sway responses to the pseudorandom surface stimulation has been described previously [8]. Briefly, after discarding the first stimulus cycle to avoid transient behavior, power and cross-power spectra were computed for each of the remaining 5 cycles using results from a discrete Fourier transform of the sampled COM body sway angle with respect to earth vertical and measured SS tilt angle. The various power spectra were smoothed by averaging results from the 5 cycles and by averaging adjacent frequency points [1] so that the final 16 transfer function points were approximately equally space on a logarithmic frequency scale ranging from 0.016 Hz to 2.2 Hz. Each transfer function point was represented by a gain and phase value. The phase values (in degrees) were "unwrapped" using the function "phase" from the Matlab System Identification Toolbox (The MathWorks, Natick, MA) so that phase lags greater than -180° could be displayed.

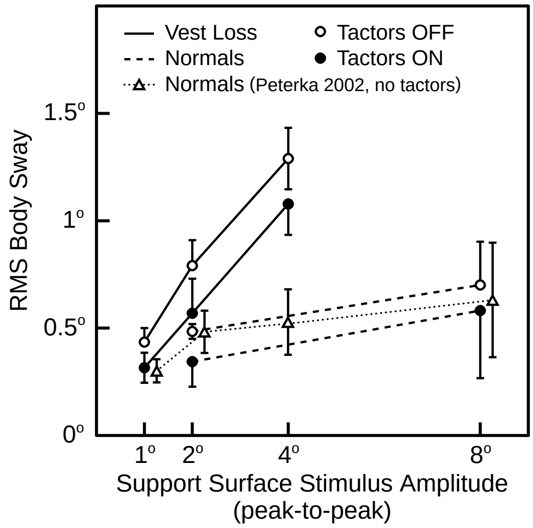

Fig. 3. RMS sway amplitudes in normal and VL subjects evoked by pseudorandom support surface rotations during eyes closed stance (mean $\pm$ SD). Data represented by open symbols were obtain obtained without the vibrotactile prosthesis activated, and closed symbols are with the prosthesis activated. Triangles are data from 8 normal subjects from a previous study [8].

Power spectra were also used to calculate coherence functions [1] which are the frequency-domain equivalent of linear correlation coefficients. Coherence function values vary from 0 to 1 with values of 1 indicating a perfect linear relationship between stimulus and response with no noise in the system or measurements.

3. Results

3.1. Effects on overall body sway levels

The mean ($\pm$ SD) root mean squared (RMS) body sway as a function of SS stimulus amplitude is shown in Fig. 3 for 5 VL and 6 normal subjects in conditions with the prostheses activated (Tactors ON) and inactivated (Tactors OFF). Also shown in Fig. 3 is the mean RMS sway measures from 8 normal subjects from a previous study [8] obtained at 1°, 2°, 4°, and 8° SS amplitudes under conditions equivalent to the Tactors OFF condition. Consistent with previous results in VL and normal subjects (Fig. 4 in [8]), body sway increased with increasing SS stimulus amplitude in both normal and VL subjects, but the increase was much greater for VL than normal subjects.

Body sway levels were consistently and significantly reduced with the use of the balance prosthesis at all SS stimulus amplitudes for both normal and VL subjects.

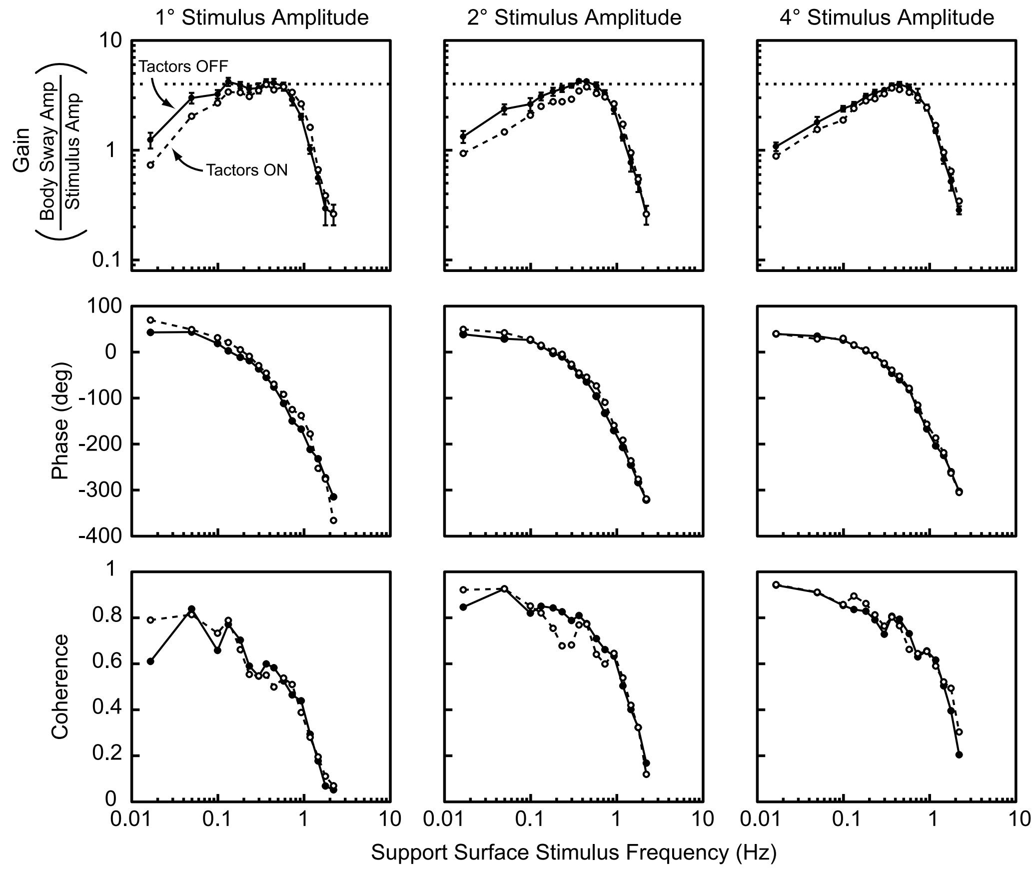

Fig. 4. Mean transfer functions and mean coherence functions for the 5 vestibular loss subjects obtained with the vibrotactile prosthesis activated (Tactors ON, open circles and dashed lines) and inactivated (Tactors OFF, filled circles and solid lines). The left, center, and right columns are from responses to 1°, 2°, and 4° (peak-to-peak) pseudorandom support surface rotations, respectively. Error bars on the gain data show ±1 SE. Peak transfer function gains were approximately the same for the 3 support surface stimulus amplitudes (dotted lines).

The RMS sway was reduced by a mean value of 0.15° (95% confidence limits are ± 0.053° about the mean) across all subjects and SS amplitudes. The mean reduction was slightly greater for VL subjects (mean 0.18°) than for normal subjects (mean 0.10°). At the 1° and 2° SS amplitudes, the sway reduction afforded by the prosthesis gave RMS sway values in VL subjects that were very close to the RMS sway in normal subjects who performed tests without the prosthesis. However, at the 4° SS amplitude, a sway reduction of about 0.77° would have been necessary for VL subjects to match the mean RMS sway in normal subjects without a prosthesis, but the actual mean sway reduction was 0.21°. In general, the prosthesis was unable to restore a normal level of body sway at larger SS stimulus amplitudes as indicated by the results from the 4° SS stimulus trials and by the fact that none of the VL subjects in the Tactors ON condition were able to maintain stance during an 8° SS stimulus.

3.2. *Transfer function analysis in VL subjects*

Results from the transfer function analysis provided additional information about how the balance prosthesis contributed to a reduction in sway. Figure 4 overlays mean gain, phase, and coherence functions from the 5 VL subjects in the Tactors ON and Tactors OFF conditions. For frequencies below about 0.5 Hz, there was a consistent gain reduction in the Tactors ON versus OFF condition. For the 1° SS stimulus, there was a small phase advance in the Tactors ON condition compared to the Tactors OFF condition for frequencies up to about 1 Hz. There was no consistent change in phase for the 2° and 4° SS trials. There was no consistent change in the mean coherence function in Tactors ON versus OFF conditions.

Graphs of the ratio of gains in the Tactors ON condition to the gains in the Tactors OFF condition provide a clearer illustration of gain changes associated with use

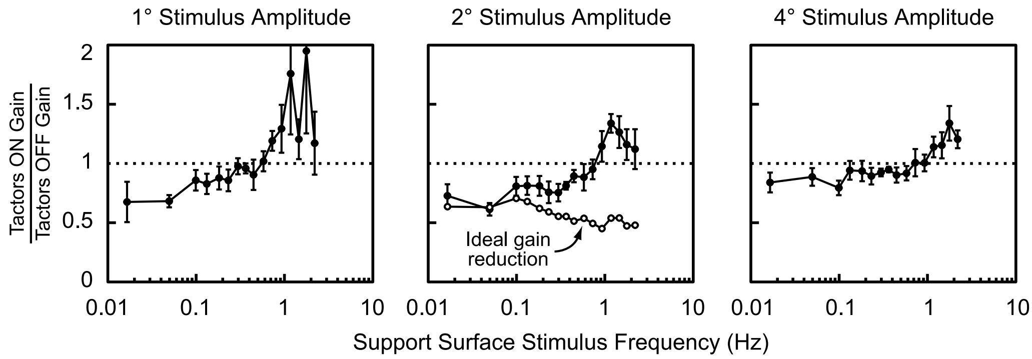

Fig. 5. Ratio of transfer function gains with the balance prosthesis activated (Tactors ON) to transfer function gains with the prosthesis inactivated (Tactors OFF). Solid circles are the mean (±SE) gain ratios for the 5 vestibular loss subjects. The left, center, and right columns are from responses to 1°, 2°, and 4° (peak-to-peak) pseudorandom support surface rotations, respectively. Open circles shown for the 2° stimulus data represent the ideal gain ratios expected if the prosthesis had reduced sway of vestibular loss subjects to that of normal subjects in the 2° stimulus condition. Dotted lines represent the expected gain ratio if the prosthesis had no effect on balance.

of the prosthesis (Fig. 5). Tactor use was associated with a gain reduction (gain ratio less than 1.0) up to frequencies ranging from 0.6 Hz to 0.8 Hz. The gain reduction at these lower frequencies had similar magnitudes for the 1° and 2° SS stimuli, but the gain reduction was slightly less for the 4° SS stimulus. However, at frequencies greater than about 0.8 Hz, the gain ratios were greater than 1.0 indicating that the SS stimuli evoked a greater sway amplitude in the Tactors ON condition than the Tactors OFF condition – an undesirable feature for a balance prosthesis.

Also plotted in Fig. 5 for the 2° SS amplitude is the "ideal gain ratio". This ideal gain ratio is the ratio of gains in normal subjects to the gains in VL subjects, with gains in both normal and VL subjects measured in the Tactors OFF condition. That is, if the prosthesis had restored normal balance control to the VL subjects, then the Tactors ON to Tactors OFF gain ratio in the VL subjects should overlay the ideal gain ratio curve. The ideal gain ratio for the 2° SS stimulus was approximately constant across the full 0.017–2.2 Hz bandwidth of test frequencies. The gain ratio for VL subjects was close to the ideal at the lowest two stimulus frequencies (0.017 and 0.05 Hz), but there was a progressively increasing gain ratio (i.e., less gain reduction) with increasing frequency, and there was a gain enhancement at higher frequencies.

3.3. Transfer function analysis in normal subjects

Transfer functions comparing Tactors OFF and Tactors ON results are show in Fig. 6. The overall transfer function gains were smaller for normal subjects than for VL subjects, consistent with the ability of normal subjects to make use of vestibular cues to reduce the influence of the SS perturbation on postural control [8]. In the Tactors ON condition, normal subjects were able to further reduce their gains over the lower frequency portion of the SS stimulus bandwidth. Specifically, gains were reduced for frequencies below about 0.8 Hz and 0.4 Hz for the 2° and 8° SS stimuli, respectively. Similar to results from VL subjects, phase and coherence functions were minimally influenced by the balance prosthesis.

Graphs of the ratio of gains in the Tactors ON condition to the gains in the Tactors OFF condition (Fig. 7) show results similar to those seen for VL subjects (Fig. 5). The gain ratios were smallest (i.e. greatest gain reduction) at the lowest stimulus frequencies, increased with increasing frequency, and were consistently greater than 1.0 for frequencies greater than about 0.8 Hz.

4. Discussion

Various approaches have been suggested for the development of balance prostheses. These include "sensory substitution" devices whereby sensory information, normally available from a particular sensory system, is encoded in an alternative sensory system [15, 16], "sensory enhancement" devices that improve the sensitivity of an existing sensory systems [12,13], and "sensory restoration" devices that attempt to restore function to a damaged sensory system [4].

For all types of balance prostheses, it is desirable to determine the extent to which the prosthesis is able to improve and, ideally, restore normal balance function.

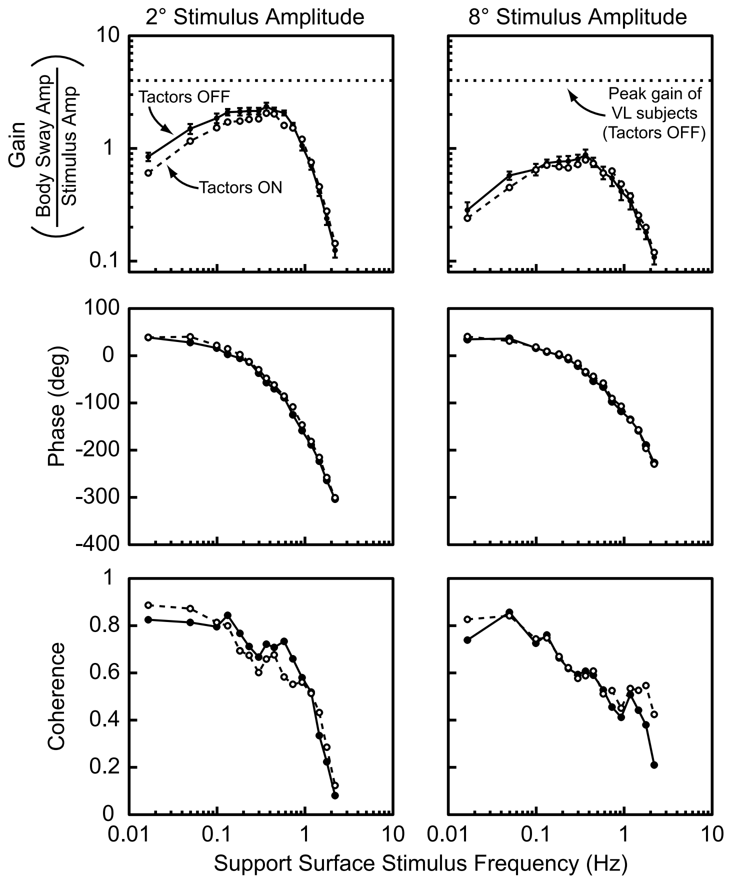

Fig. 6. Mean transfer functions and mean coherence functions for the 6 normal subjects obtained with the vibrotactile prosthesis activated (Tactors ON, open circles and dashed lines) and inactivated (Tactors OFF, filled circles and solid lines). The left and right columns are from responses to 2° and 8° (peak-to-peak) pseudorandom support surface rotations, respectively. Error bars on the gain data show ±1 SE. Dotted lines show the peak gain for vestibular loss subjects.

Previous evaluation methods have relied primary on the use of various spontaneous sway measures to show that the prosthesis reduced sway in different test conditions [6,12]. While reduction in spontaneous sway provides some measure of effectiveness, we advocate the use of much stronger criteria for evaluation of a prosthesis based on methods that characterize stimulus-evoked sway.

We demonstrated a method based on transfer function analysis that characterizes the dynamic behavior of postural control in response to pseudorandom SS tilts of varying amplitudes. We chose to use an eyes closed condition so that the postural responses depended primarily on how subjects utilized proprioceptive and vestibular sensory information, as well as information provided by a prosthesis. Our method provides a strong test for the effectiveness of a balance prosthesis because stimulus-evoked sway in normal subjects shows distinct and consistent changes with changing SS stimulus amplitudes (Fig. 1A), and there are large differences between the pattern of results seen in normal and VL subjects (compare Figs 1A and B). Specifically, transfer functions from normal subjects show a progressive reduction in gain with increasing SS stimulus amplitude while transfer functions from VL subjects did not. If a balance prosthesis was able to fully restore normal balance function in VL subjects, these VL subjects should show a progressive decrease in their transfer function gain with increasing SS amplitude (Fig. 1A), and this gain reduction should be approximately uniform across the 0.017–2.2 Hz bandwidth of the SS stimulus. Thus, our transfer function analysis of responses to a different SS stimulus amplitudes provides information about both amplitude-related and frequency-related changes in postural behavior.

The amplitude-related changes are particularly important for demonstrating the effectiveness of a prosthesis that is meant to restore or substitute for missing

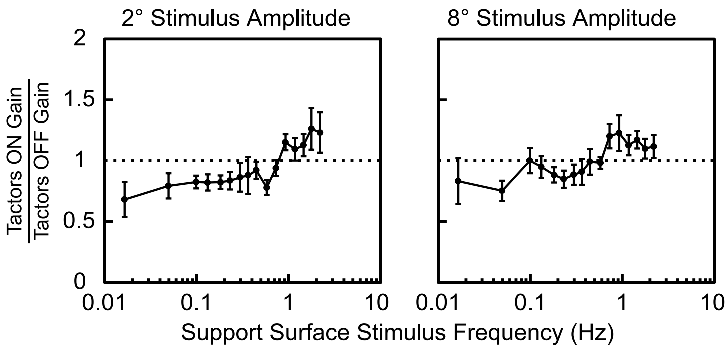

Fig. 7. Ratio of transfer function gains with the balance prosthesis activated (Tactors ON) to transfer function gains with the prosthesis inactivated (Tactors OFF). Solid circles are the mean (±SE) gain ratios for the 6 normal subjects. The left and right columns are from responses to 2° and 8° (peak-to-peak) pseudorandom support surface rotations, respectively. Dotted lines represent the expected gain ratio if the prosthesis had no effect on balance.

vestibular information. Specifically, previous studies have shown that a pattern of gain reduction with increasing SS amplitude indicates that a process of sensory reweighting is occurring such that the subject is becoming more reliant on vestibular cues and less reliant on proprioceptive cues with increasing SS amplitude [2,8]. A failure to show this gain reduction is indicative of abnormal utilization of vestibular information (e.g. results shown in Fig. 1(B) for VL subjects). The ability of a balance prosthesis to restore a normal pattern of gain reduction with increasing SS amplitude is, therefore, indicative of the effectiveness of the prosthesis in restoring or substituting for vestibular information. Furthermore, even subjects with quite severe balance deficits can be tested successfully using low amplitude SS stimuli (1 or 2° peak-to-peak rotations), and differences in transfer function gains between normal and VL subjects are evident when using these low amplitude stimuli. Therefore, meaningful quantitative evaluations of the effectiveness of a prosthesis can be obtained using our proposed method even when subjects are unable to perform more challenging tests using higher amplitude SS stimuli. However, failure to maintain stance at higher SS stimulus amplitudes would be indicative of an incomplete ability of the prosthesis to restore normal balance function.

Frequency-related changes in postural behavior also provide evidence for the effectiveness of the prosthesis and can potentially provide insight into the characteristics of information provided by a prosthesis. For subjects with normal sensory function, the transfer function gain decrease with increasing SS amplitude is approximately uniform across the frequency bandwidth of our SS stimulus (see Fig. 1(A) and "ideal gain reduction" curve in Fig. 5). In the eyes-closed condition, a previous study showed that a uniform gain change was consistent with a reweighting of orientation information from the vestibular and proprioceptive systems, and that each of these sensory systems was providing the postural control system with a wide-bandwidth encoding of body sway [8]. Therefore, the failure of a prosthesis to produce a uniform gain change across frequency could imply that the bandwidth of orientation information provided by the prosthesis is limited. For example, a gain reduction that was limited to low frequencies might imply that the prosthesis was conveying only low-frequency orientation information.

We demonstrated our analysis methods using a specific vibrotactile prosthesis, but the analysis would also be expected to provide detailed information about the effectiveness of any type of balance prosthesis. For the vibrotactile prosthesis, the analysis showed only a partial restoration of a normal pattern of postural control, and revealed two specific deficiencies of the prosthesis. First, the gain reduction was not uniform across the frequency bandwidth, and in fact there was even a gain enhancement at higher SS stimulus frequencies (Fig. 5). Second, the prosthesis provided a similar gain reduction factor at each of the 3 SS amplitudes tested in VL subjects (Figs 3 and 5), whereas an increasing gain reduction factor with increasing SS amplitude would have been necessary to approach a normal response pattern.

While the simpler measure of RMS body sway (Fig. 3) provided some information regarding the effectiveness of the prosthesis, the frequency domain information provided by the transfer function analysis gave additional information that is likely to be useful in the overall evaluation of the prosthesis and in guiding the selection of prosthesis parameters to achieve optimal results. For example, the vibrotactile prosthesis improved balance only over a limited bandwidth.

We might hypothesize that this limitation was caused by the small number of tactors used to encode body sway. This hypothesis could be tested quantitatively by comparing of the transfer function results using different numbers of tactors. Furthermore, frequency-dependent effects are a likely consequence of using different schemes for encoding body sway motion. The selection of the correct combination of motion feedback information (e.g., combinations of position-related or velocity-related sway) could be optimized using transfer function analysis.

Transfer function phase changes associated with the use of the balance prosthesis were minimal for the particular prosthesis that we investigated. However, this may not be true for all prostheses or for different implementations of a given type of prosthesis. For example, changing the character of the motion information encoded by the prosthesis (i.e. position, velocity, or combinations of position and velocity) might change the phase characteristics of responses to the SS stimulus. In general, a full evaluation of a prosthesis should include the calculation of transfer function phase and comparison with the phase characteristics of subjects with normal postural control.

The fairly long duration of test trials used in our method for prosthesis evaluation raises the concern that habituation to the SS stimulus or to the prosthesis could influence the results. A diminishing body sway response to the SS stimulus over time would be indicative of habituation to the SS stimulus, and an increasing amplitude of body sway could be indicative of habituation to the prosthesis. Previous results [8] showed no evidence for habituation to the pseudorandom SS stimulus used in the current study. Specifically, there was no significant difference among cycle-by-cycle measures of RMS sway across the 6 cycles of the SS stimulus. Because the SS stimulus is pseudorandom rather than random and repeated cycles of the pseudorandom stimulus are presented, it is quite easy to detect systematic changes in body sway across time. In the current study, similar cycle-by-cycle comparisons showed no evidence for habituation to information provided by the vibrotactile prosthesis.

Previous results have shown that pseudorandom tilts of a visual surround evoke sway that can be analyzed using the same methods used for responses to SS tilts, and that responses to visual stimuli provide information about the extent to which subjects use visual cues for balance control [8]. Therefore, similar methods could be applied using visual stimuli, or combinations of visual and surface stimuli, to investigate how a balance prosthesis facilitates postural control in more complex environments where visual, proprioceptive, and vestibular cues interact.

Results in normal subjects for the vibrotactile prosthesis showed that they were also able to use information from the prosthesis to reduce sway evoked by the SS stimulus. Furthermore, the gain reduction amplitude and the frequency dependency of the gain reduction (Fig. 7) were similar to results in VL subjects (Fig. 5). If it was generally true that prostheses based on sensory substitution or sensory enhancement produced similar effects in normal and VL subjects, then the majority of the tests needed to optimize a particular prosthesis could be performed in normal subjects rather that in subjects with balance disorders.

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