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Influence of Feedback Parameters on Performance of a Vibrotactile Balance Prosthesis

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Abstract

We investigated the influence of feedback conditions on the effectiveness of a balance prosthesis. 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 several different combinations of angular position and velocity of body sway in the sagittal plane. Control tests were performed with no tactor activation. Body sway was evoked in subjects with normal sensory function by rotating the support surface upon which subjects stood with eyes closed. Body sway was analyzed by computing root-mean-square sway measures and by a frequency-response function analysis that characterized the amplitude (gain) and timing (phase) of body sway over a frequency range of 0.017 to 2.2 Hz. Root-mean-square sway measures showed a reduction of surface stimulus evoked body sway for most vibrotactile feedback settings compared to control conditions. However, frequency-response function analysis showed that the sway reduction was due primarily to a reduction in sway below about 0.5 Hz, whereas there was actually an enhancement of sway above 0.6 Hz. Finally, we created a postural model that accounted for the experimental results and gave insight into how vibrotactile information was incorporated into the postural control system.

Index Terms

Balance Prosthesis; Feedback Control Model; Posture control; Sensorimotor Integration

I. INTRODUCTION

In recent decades, much research has been performed using vibrotactile cues to convey information [1–3]. This popularity is likely due to the wide variety of information that the brain can understand through vibrotactile interfaces. Some of the more celebrated examples have been the successful coding of alphabetic letters (Opticon) [4] and coding of aircraft orientation for pilots [5]. Recent research has extended the vibrotactile display of aircraft orientation to postural orientation with the development of a noninvasive balance prosthesis [6], [7].

A balance prosthesis is a device intended to ameliorate some of the deficits of stance and movement control associated with abnormal vestibular function and other sensory deficits [8], [9]. This is accomplished by improving balance control through enhancement or restoration of sensory function, or by providing orientation information via an alternative sensory

modality. For a vibrotactile prosthesis, the alternative sensory modality is vibration of mechanoreceptors in skin, coded to convey information about body orientation relative to earth-vertical to the central nervous system (CNS).

The prosthesis used in this study is worn like a vest with small vibrotactile stimulators, called tactors, which convey information about spatial orientation from motion-sensing devices [6]. Previous research used critical tracking tests (CTT) to help determine appropriate tactor locations on the body, tactor spacing, tactor vibration patterns, and motion feedback information [10]. The CTT measured subjects' ability to use a joystick to control and stabilize an inherently unstable object using only vibrotactile cues. Results from the CTT tests suggested that both position information and velocity information must be conveyed through the tactors of a balance prosthesis because both position and velocity feedback were required to stabilize an object that had similar unstable dynamics to postural stance [10]. Furthermore, this conclusion seems reasonable based on a physiological argument that in normal postural control, the CNS processes both position and velocity information to maintain upright stance [11].

However, a systematic investigation of the effect of feedback information on postural control had not been performed. A postural test, contrary to the CTT, requires the subject to react both voluntarily and involuntarily; whereas the CTT only requires voluntary reactions to vibrotactile information. A postural test is, therefore, more complex than a CTT because vibrotactile information must be assimilated into an existing involuntary postural control system.

A recent study tested the vibrotactile prosthesis using a single combination of position and velocity feedback ($2/3$ position + $1/3$ velocity) in a series of postural tests on subjects with normal sensory function and with a bilateral absence of vestibular function [12]. The postural tests elicited body sway responses using a continuous perturbation provided by tilts of the surface upon which subjects stood. Body sway responses were analyzed via an engineering system identification approach that used a frequency domain analysis to characterize the dynamic properties of the postural control system across a wide frequency bandwidth [13]. A system identification approach provides a more detailed representation of postural dynamics compared to simple sway measurements [14] because it reveals a subject's body sway response at specific perturbation frequencies [15], [16]. Results showed that feedback information from the prosthesis helped both normal and severe vestibular loss subjects to better compensate for the postural perturbation as indicated by reduced body sway responses at low frequencies. However, at higher perturbation frequencies the body sway response was slightly increased. A more effective prosthesis should reduce responses at all test frequencies.

We postulated that the dynamic response to the perturbations would vary systemically as a function of the combination of position and velocity feedback information encoded by the vibrotactile prosthesis. The first goal was to identify the particular combination of position and velocity feedback that would give a subject the greatest improvement in balance as indicated by the ability to resist the influence of a postural perturbation.

The second goal was to understand how orientation information from the prosthesis was combined with natural sensory information and subsequently used for postural control. To do this, we modified an existing model of postural control [15]–[17] to include a representation of the balance prosthesis within the postural system.

II. METHODS

A. Subjects

Eight healthy subjects with no history of balance disorders (4 male, 4 female, mean age 26 years $\pm$ 7 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.

B. Protocol

Each subject performed 8 posture tests during which the subject's balance was perturbed by a continuous rotation of the surface upon which the subject was standing. The postural tests included 3 identical control conditions with no vibrotactile feedback (Tactors OFF) and 5 test conditions with vibrotactile feedback (Tactors ON). The Tactors ON conditions included different combinations of vibrotactile feedback gains and activation thresholds (Table I). The first and last postural tests were always Tactors OFF conditions, which were used to examine learning effects. The remaining 6 tests were randomized to minimize the potential biases of fatigue and learning. Preceding each new Tactors ON test, subjects were informed of the new tactile feedback type and were given a brief period (~30 seconds) to become acquainted with the feedback by swaying voluntarily. Subjects were also instructed to make conscious use of the vibrotactile feedback to help them reduce sway and stand upright throughout the test. No subjects felt that a longer time period was needed to understand how to use the information from the prosthesis and a previous study [18] demonstrated immediate improvement as soon as the prosthesis was turned on. However, because the training time and the entire test protocol was limited, we cannot rule out the possibility that extended use of the prosthesis could induce long-term adaptation effects that produce results that differ from the experimental results in this report.

The 5 Tactors ON conditions each had a different combination of position and velocity feedback information given to the subjects through the vibrotactile interface. The 5 different feedback settings include position only feedback (P100V0), 75% position and 25% velocity feedback (P75V25), 50% position and 50% velocity feedback (P50V50), 25% position and 75% velocity feedback (P25V75), and velocity only feedback (P0V100).

Subjects stood with eyes closed on a support surface with their feet placed at a comfortable distance apart. Their ankle joint axes were aligned with the rotational axis of the support surface. A backboard assembly was used to insure that the subjects swayed as a single-link inverted pendulum. Subjects were attached to the backboard by shoulder straps, waist belt, and a belt above the knees. The backboard allowed free movement in the sagittal plane with rotation occurring about an axis aligned with the rotation axis of the support surface and the subject's ankle joint axis. Wearing the backboard assembly was previously shown to result in body sways nearly indistinguishable from natural freestanding body sway when stimulus conditions identical to those used in the current study were used to evoke body sway [15]. Measurements of the backboard rotational position using a potentiometer (Midori America Corp., Fullerton, CA) with respect to earth-vertical and velocity using a rate gyro (Watson Industries, Au Claire, WI) were used to characterize body sway behavior. Backboard rotation position and velocity were also measured via a custom inertial sensor [6]. As in previous experiments, orientation measures from the inertial sensor were used to drive the vibrotactile balance prosthesis [12].

Each test presented a postural perturbation by continuously rotating the support surface according to a pseudorandom stimulus based on a pseudorandom ternary maximal length sequence [15], [19]. The stimulus was periodic with a period of 60.5 s. Six consecutive cycles of the stimulus were presented for each postural test. This test duration has been shown to provide reliable estimates of frequency-response functions without inducing measurable fatigue [15]. The support surface orientation was controlled by a servomotor that could rotate the surface producing sagittal plane tilts of the surface. Peak-to-peak surface rotations of 2° were used for all tests. This stimulus amplitude had previously been shown to evoke large enough body sway to obtain reliable estimates of frequency-response functions while avoiding

large amplitude body sways that would normally require the use of a “hip strategy” in freestanding subjects [11].

Tests were performed with eyes closed for several reasons. First, stance in the dark or with eye closure is physiologically relevant since subjects with balance disorders often have more difficulty in conditions that limit the availability of visual orientation cues. Second, eliminating the visual contribution simplifies the stance control system and would likely enhance the relative contribution of prosthesis-provided orientation information in comparison to orientation information from the remaining natural sensory sources (mainly proprioception and vestibular), thus providing greater sensitivity for measuring postural changes attributable to the prosthesis. Third, testing with eye closure allows for comparisons with results from a previous study [12] that used this prosthesis on normal subjects and subjects with a bilateral vestibular deficit in an eyes closed condition.

C. Vibrotactile Balance Prosthesis

The balance prosthesis used in this experiment has been described previously in detail [12], [20]. Briefly, it contained three major elements: an inertial instrumentation package [6], a computer serving as a signal processor (Macintosh PowerBook, Apple Computer, Inc. running a custom National Instruments Labview program), and a vibrotactile array. The inertial instrumentation consisted of a micro-mechanical linear accelerometer and gyroscope (gyro) that was attached to the backboard about 35 cm above the ankle height. A measure of angular body sway in the sagittal plane was obtained by adding the integrated, high-pass filtered gyro signal to the low-pass filtered accelerometer signal [6]. The angular body sway velocity in the sagittal plane was obtained directly from the gyro.

The vibrotactile display consisted of 12 small tactors (2.7×1.9×1 cm, Audiological Engineering, Somerville, MA), held against the subject by a wide elastic fabric wrapped around the torso. The tactors were configured in 2 vertical columns of 3 tactors each on both the anterior and posterior surfaces of the lower torso. When a tactor was activated, it vibrated at a constant frequency of 250 Hz, with an intensity that all subjects were able to perceive.

At any moment of time either no tactors were active, indicating minimal body sway, or one pair of horizontally adjacent tactors were active. In general, body sway magnitude was encoded and fed back to the subject by activating a tactor pair located higher on the torso when body sway was larger, and lower on the torso when body sway was smaller. Sway direction was encoded by activating tactor pairs on the anterior surface during forward sway and on the posterior surface during backward sway. More specifically, the activation of these tactors was based on a step-wise coding scheme with activation determined in different Tactors ON conditions by different combinations of body sway angular position and velocity. Every 10 milliseconds, body sway position and velocity signals were sampled by the Labview program, and expressed in units of degrees and degrees/s, respectively. The position signal was multiplied by a position gain factor (P_{gain}), the velocity signal was multiplied by a velocity gain factor (V_{gain}), and these multiplied values were added together to obtain the ‘composite response’ (CR). The CR was then compared to 3 previously calculated threshold values (T_1 , T_2 , T_3 , see Table I) to determine which of the tactors, if any, should be activated. This process is described schematically in Fig. 1.

The activation thresholds were set to specific values for each Tactors ON condition (Table I) in order to maximize the effectiveness of feedback by matching the expected dynamic range of the CR for each test condition. Based on preliminary Tactors OFF experiments in 4 subjects (preliminary experiments not included in this paper), a ‘maximum expected CR ’ (CR_{max}) was calculated for each of the different combinations of sway position and sway velocity shown in Table I. To find CR_{max} for a particular combination of position and velocity feedback, first, an

average CR waveform was calculated for each of the 4 subjects by averaging the CR over the last 5 cycles of the pseudorandom stimulus. The peak-to-peak values of these 4 average CR waveforms were measured. The mean of these peak-to-peak values was calculated and CR_{max} was defined as one half of this mean value.

The threshold activation levels, shown in Table I, were based on CR_{max} . The thresholds T_3 , T_2 , and T_1 were set to 0.75, 0.5, and 0.25 of CR_{max} , respectively. Because the position gain and velocity gain factors were different for the 5 Tactors ON conditions, the CR_{max} values were different for each condition, and, therefore, the tactor thresholds differed for each Tactors ON condition.

D. Analysis

Frequency-response function and time series statistical analyses were used. The estimation of frequency-response functions relating body sway responses to the pseudorandom surface stimulation has been described previously [15]. Briefly, frequency-response functions were computed from power and cross-power spectra. Power and cross-power spectra were computed using a discrete Fourier transform of the body sway angle with respect to earth vertical and the measured support surface tilt angle. The various power spectra were computed for each of the last 5 stimulus cycles (discarding the first 60.5s stimulus cycle to reduce the possibility of transient responses influencing the analysis), then smoothed by averaging these power spectra together, and then averaging adjacent frequency points [13] so that the final 16 frequency-response function points were approximately equally spaced on a logarithmic frequency scale ranging from 0.017 Hz to 2.2 Hz. Frequency-response functions were expressed as gain and phase functions. The phase values (in degrees) were “unwrapped” using the function “phase” from the Matlab Signal Processing Toolbox (The MathWorks, Natick, MA) so that phase lags exceeding -180° could be displayed.

Root-mean-square (RMS) and peak-to-peak values of body sway angle and angular velocity were measured from the mean body sway responses for each postural test, averaged over the last 5 stimulus cycles. The mean value of each time series was subtracted prior to calculation of RMS values. RMS values were used to investigate overall changes between test conditions. Peak-to-peak and RMS values were also used as constraints in our model predictions.

Statistical comparisons of RMS results and frequency-response function gains across test conditions were performed using a repeated measures one-way analysis of variance, with a 0.05 significance level. Multiple comparison tests were performed using multiple paired t-tests with the Bonferroni correction [21].

E. Model-based Interpretation

1) Model Description—An experimentally validated feedback control model of the postural system (Fig. 2) was used to interpret the experimental results. The model used in this study was based on a version of the “independent channel model” described previously [16] but was modified to include a nonlinear vibrotactile feedback mechanism, exclude the visual channel (all tests were eyes closed), and exclude the passive muscle/tendon properties (muscle/tendon properties are expected to influence mainly high frequency responses that were not included in this model).

Additional model details are provided in previous studies [15]–[17]. Briefly, following the notation in Fig. 2, the model assumes that the body sways as an inverted pendulum in the sagittal plane about the ankle joint. Body-in-space (BS) sway is caused by torque due to gravity acting on an inherently unstable inverted pendulum body and by an external perturbation from tilt of the support surface (SS *input*). Stability is actively maintained by a corrective torque

(T_c) generated based on the time delayed sensory error signal. Specifically, T_c is generated in proportion to the error signal (gain factor K_P) and the derivative of the error signal (gain factor K_D). The sensory error signal is the summed output of various sensory systems related to sensing BS (vestibular system), body-relative-to-surface (proprioceptive system), and T_c (torque feedback system). The vestibular and proprioceptive systems are assumed to provide wide bandwidth orientation information and are therefore represented by gain constants W_{vest} and W_{prop} , respectively. W_{prop} and W_{vest} also represent the relative proportion of orientation information utilized by the postural control system such that $W_{prop} + W_{vest} = 1$. Thus, in the tactors OFF test condition, the model of postural control can be expressed by the analytical equation

$$\frac{BS}{SS} = \frac{W_{prop} \cdot TD \cdot NC \cdot B}{1 + TD \cdot NC \cdot B - TF \cdot TD \cdot NC} \quad (1)$$

where TD is the time delay, NC is the neural controller, B is the inverted pendulum body, and TF is the torque feedback [16].

In the tactors ON test conditions, an additional feedback loop was included in the model to represent the contribution of the balance prosthesis. The model representation of the prosthesis first computes the composite response (CR) from BS and the derivative of BS . The CR is then compared to the tactor activation thresholds (Table I). The output of the model prosthesis, P_{out} , is an integer value ranging from -3 to 3 representing the tactor activation level and neural encoding of the activation levels. The central neural processing of P_{out} is represented by a mathematical integration of P_{out} and multiplication by a gain factor G_{Tactor} . The output from this vibrotactile prosthesis feedback loop is summed with the postural systems' natural sensory feedback signals (proprioceptive, vestibular, and torque related sensors). Note that the G_{Tactor} value is not comparable to the sensory channel weights W_{prop} and W_{vest} because the particular value of G_{Tactor} depends on the arbitrary choice for values of P_{out} .

Other representations of neural processing of tactile information were considered, such as low-pass filtering of P_{out} and an additional time delay in the vibrotactile feedback loop. In particular, inclusion of a first order low-pass filter rather than an integrator in the feedback loop provided an equally good explanation of the experimental data. When model parameters were estimated (see next section) the mean cutoff frequency of the low-pass filter was about 0.02 Hz. Because of this very low cutoff frequency, the dynamic behavior of a low-pass filter resembles a pure integrator over the frequency range of our experimental results. Therefore, we chose to use the simpler representation of a pure integrator because it was the simplest model that described the main features of the experimental results. Furthermore, the inclusion of an additional time delay in the vibrotactile feedback loop did not improve the model's fit to the experimental data.

2) Model Parameters—Model parameters related to body mass, m , (not including the feet but including the backboard mass), moment of inertia (J , including the body and backboard), and height of the center of mass (h) of the body and backboard above the ankle joints were estimated from body measurements on each subject [22] and from measurements of the backboard assembly. The average values across subjects were $J = 90.2 \text{ kg.m}^2$, $m = 83.3 \text{ kg}$, and $h = 0.95 \text{ m}$. The remaining model parameters were estimated from experimental results for each test condition (Tactors OFF and Tactors ON) using a constrained nonlinear optimization routine 'fmincon' (Matlab Optimization Toolbox, The MathWorks, Natick, MA) to minimize the mean-squared-error (MSE) between the frequency-response functions resulting from model simulations and the experimental frequency-response functions. The MSE was normalized to the magnitude of the simulated frequency-response functions so that each frequency point was equally important in estimating parameter values.

In the Tactors ON test conditions, nonlinearity in the processing of prosthesis information (due to the step-wise tactor activation scheme) eliminated the possibility of expressing the model as an analytical equation. Therefore, in Tactors ON test conditions, model derived frequency-response functions were obtained by analyzing the time series of body sway obtained from model simulations. Model simulations were performed using Simulink (The MathWorks, Natick, MA). Support surface input (SS input) was identical to the pseudorandom stimulus used in experiments. To ensure that the frequency-response function fit identified parameters consistent with a stable system, the optimization applied additional time domain constraints requiring the model simulation to be within 25% of the experimental RMS and peak-to-peak sway and sway velocity.

Previous studies [15] showed that model fits to experimental frequency-response functions could be improved at frequencies greater than about 1.5 Hz by including additional parameters representing contributions to corrective torque due to passive muscle/tendon properties. To simplify the model analysis, these additional parameters were not included in the model and the model fits included only the lowest 13 frequencies ranging from 0.017 Hz to 1.17 Hz.

III. RESULTS

A. Evoked Sway With and Without Tactor Feedback

Support surface stimulus angle (identical for all test conditions) and the experimental body sway responses for several test conditions are shown in Fig. 3. The experimental body sway response waveforms in Fig. 3 are the mean body sway from the last 5 stimulus cycles of each test, averaged over all subjects. In all test conditions, the body sway responses generally followed the support surface rotation angle, indicating that subjects tended to align themselves to the support surface. While some differences between the test conditions can be seen by visual inspection of these plots, experimental results can be better appreciated after further data processing in both the time and frequency domains.

To illustrate the general effect of the tactor feedback, box plots of the RMS body sway and sway velocity for the different test conditions are presented in Fig. 4. The most effective feedback conditions for reducing RMS body sway were conditions that included a majority of position feedback information (P100V0 and P75V25). The P100V0 and P75V25 conditions resulted in statistically significant reductions of 51% and 48%, respectively, in mean RMS sway compared to the Tactors OFF condition. The P50V50 condition also resulted in a significant RMS sway reduction of 41% compared to the Tactors OFF condition. There was only a small and insignificant reduction in RMS sway for the test conditions that provided predominantly velocity feedback (P0V100 and P25V75) compared to the Tactors OFF condition.

Box plots of RMS body sway velocity showed different results (Fig. 4b). Tactors ON conditions yielded minimal changes in mean RMS sway velocity compared to the Tactors OFF condition. None of the Tactors ON conditions resulted in statistically significant reductions in mean RMS sway velocity.

Subjects were tested under control conditions (Tactors OFF) at the beginning and the end of each test session to test for learning or habituation effects to the pseudorandom stimulation. No learning effects were identified based on two-tailed paired t-tests of RMS sway ($p=0.21$) and sway velocity ($p=0.16$).

B. Frequency-response Function Analysis

1) Gain curves—Frequency-response function gain curves are plotted in Fig. 5a (left). To more clearly show the gain changes relative to the Tactors OFF condition, the ratio of gains in

the Tactors ON to the Tactors OFF conditions are plotted in Fig. 5b (left). All feedback settings reduced frequency-response function gains at low frequencies (≤ 0.23 Hz) compared to Tactors OFF condition. However, all feedback settings also increased body sway response gains at high frequencies (≥ 0.9 Hz). We quantified these body sway gain changes with test condition by analyzing the average gain at the low (≤ 0.23 Hz) and high (≥ 0.9 Hz) frequencies for statistical significance. At the low frequencies (≤ 0.23 Hz), mean gains were significantly reduced for the P100V0, P75V25, P50V50, and P25V75 test conditions compared to Tactors OFF. At the high frequencies (≥ 0.9 Hz), mean gains were significantly increased for the P100V0, P75V25, and P50V50 test conditions compared to Tactors OFF.

It can also be seen in Fig. 5 that each test condition resulted in a unique characteristic gain curve. No single feedback setting resulted in the lowest gains across all frequencies; but each feedback setting, at some frequency point, resulted in the lowest gain relative to the other feedback settings. Results from the P100V0 and P0V100 conditions illustrate the range of frequency-response function changes produced by the balance prosthesis. At low frequencies (≤ 0.1 Hz), the P100V0 setting produced a large reduction in frequency-response function gain compared to the Tactors OFF frequency-response function, more than any other setting. However, the P100V0 setting also resulted in a gain increase compared to Tactors OFF at mid- and high-frequencies (≥ 0.27 Hz). In comparison, the P0V100 feedback setting resulted in the lowest gains at mid- to high-frequencies (≥ 0.5 Hz) of any Tactors ON condition. However, the P0V100 condition produced only small gain changes across all frequencies relative to the Tactors OFF condition.

The gain reductions afforded by the prosthesis at low frequencies are a desirable effect for a balance prosthesis because a low gain signifies that the surface stimulus was relatively ineffective in perturbing the body from an upright orientation; whereas a gain increase is an undesirable effect signifying that the subject became more sensitive to the perturbing stimulus. The functional significance of these frequency dependent gain increases and decreases and how these FRF gains in general relate to RMS results will be further reviewed in Section A of the Discussion.

2) Phase Curves—The frequency-response function phase curves (Fig. 5a, right) were also altered by use of the prosthesis. Fig. 5b, right, quantifies the change in frequency-response function phase relative to the Tactors OFF condition. At low frequencies (≤ 0.3 Hz), the P100V0 setting produced the largest phase advance of all Tactors ON conditions relative to Tactors OFF. This phase advance extended to the lower frequencies for the P100V0, P75V25, and P50V50 conditions, with an increasing amount of phase advance being associated with an increasing position feedback in the Tactors ON conditions. In contrast to the P100V0 setting, the P0V100 setting produced small phase lags at low frequencies (≤ 0.2 Hz). In the mid-frequency range (0.2–1.0 Hz), all Tactors ON conditions produced a phase advance relative to the Tactors OFF condition. For all Tactors ON conditions, phase values were close to the phase in the Tactors OFF condition at 1.5 Hz.

C. Model Results

Our goal in modeling was to gain insight into how orientation information provided by the prosthesis was incorporated into the postural control system using a model-based interpretation of the experimental results. The model can be thought of as representing a quantitative hypothesis about how the system is organized. While many different model structures could potentially predict the experimental results, our strategy was to begin with a model that previously was shown to describe experimental frequency-response data [15]–[17] and then to modify this model to include a representation of the prosthesis and the neural processing of orientation information provided by the prosthesis. We explored many possibilities, but present

only the simplest model version (Fig. 2) that was found to provide a description consistent with the experimental results. In addition, we also considered the extent to which we could reduce the number of model parameters that needed to be adjusted from the parameter values identified in the Tactors OFF condition (Table II) in order to account for the frequency-response function variations as a function of prosthesis feedback in the Tactors ON conditions.

A comparison of the normalized experimental and model frequency-response functions (Fig. 5b and Fig. 6a) shows that the main features of the experimental data were accounted for by the Fig. 2 model when only the value of G_{Tactor} was allowed to vary across the prosthesis feedback conditions. That is, the model parameters for the neural controller, torque feedback, time delay, and vestibular and proprioceptive weights in the Tactors OFF condition were first identified. For each of the Tactors ON conditions the model prosthesis CR thresholds were set to values actually used in the experiments, all of the parameters identified in the Tactors OFF condition (Table II) were held constant, and only the value of G_{Tactor} was allowed to vary in order to optimize the fit of the frequency-response function calculated from the model simulation to the experimental frequency-response function.

It was found that G_{Tactor} values were similar for the P100V0 and P75V25 conditions, but then monotonically decreased with higher proportions of velocity feedback (Table III). The value of G_{Tactor} was about five times larger for the P100V0 and P75V25 conditions compared to the P0V100 condition indicating that subjects made much greater use of orientation information provided by the prosthesis when the feedback was predominantly position related.

The frequency-response function features predicted by the model included the systematic pattern of decreased gain (relative to the Tactors OFF condition) for low frequencies (<0.25 Hz), a gain increase at mid-frequencies (depending on the particular Tactors ON condition), a large phase advance at frequencies less than 0.2 Hz for the position dominant feedback condition but not for the velocity dominant conditions, and an increase in phase advance for the velocity dominant feedback conditions in the 0.2 to 1 Hz range. The model was least accurate in predicting that frequency-response function gain values remained higher than the Tactors OFF condition for frequencies above 1 Hz. There were differences in specific patterns of the phase data for frequencies below about 0.2 Hz although the general trends were correctly predicted. The model also predicted the time domain responses (Fig. 3, dotted lines).

In order to investigate whether the main features of the model frequency-response functions were due to the quantization nonlinearity or to the inclusion of mathematical integration dynamics in the model's prosthesis feedback loop, the frequency-response functions derived from a fully linearized version of the feedback model were calculated. The linearized model is given by

$$\frac{BS}{SS} = \frac{W_{prop} \cdot TD \cdot NC \cdot B}{1 - TF \cdot TD \cdot NC + \left(1 + \left(\frac{540 \cdot G_{Tactor}}{\pi \cdot CR_{max}}\right) \left(V_{gain} + \frac{P_{gain}}{s}\right)\right) TD \cdot NC \cdot B} \quad (2)$$

where s is the Laplace transform variable, other model components are defined as in equation (1), and specific model parameters are given in Table I, Table II, and Table III. The calculated normalized frequency-response functions are shown in Fig. 6b. These calculated frequency-response functions show that it is not necessary to include the quantization nonlinearity in the model in order to account for the experimental frequency-response functions, but it is necessary to include integration dynamics in the prosthesis feedback loop.

The frequency response functions from equation (2) show that the general pattern of enhanced low frequency phase advance and reduced low frequency gain is expected when a negative

feedback loop with an integrator is added to a system that otherwise has wide bandwidth characteristics. The addition of feedback integration produces a system whose closed-loop dynamic characteristics are that of a high-pass filter with reduced gain and phase advance at low frequencies.

IV. DISCUSSION

A. Predicting Changes in RMS Sway and Sway Velocity

We used a continuous support surface perturbation and a system analysis approach to determine the extent to which feedback settings in a vibrotactile balance prosthesis enhanced postural control. Our first goal was to identify the particular combination of position and velocity feedback would result in the greatest improvement in postural control as determined by the greatest reduction in sway measures (RMS values of sway and sway velocity). However, we found that no single combination of position and velocity feedback reduced stimulus-evoked sway across all of the RMS response measures.

The trend from the RMS results indicated that the type of feedback information the subject received was the type of response the subject tended to minimize. That is, RMS sway position was most reduced when the prosthesis feedback was mainly position related, and RMS sway velocity was most reduced when the prosthesis feedback was mainly velocity related. However, position feedback was relatively more effective than velocity feedback in reducing overall sway magnitude. Specifically, RMS sway velocity was only minimally reduced in all of the feedback conditions relative to control whereas RMS sway position was reduced to about half of the control value in the position-dominant prosthesis feedback conditions.

The frequency-response function analysis provided a more detailed understanding of why the prosthesis appeared to be more effective in reducing RMS sway position than velocity. The PRTS stimulus velocity waveform used in this study had a power spectrum with approximately constant amplitude components out to a frequency of about 1.3 Hz. The output power spectrum of sway velocity, $G_{yy}(f)$, is predicted by the following equation

$$G_{yy}(f) = |H(f)|^2 G_{xx}(f) \quad (3)$$

where f is frequency, $G_{xx}(f)$ is the input power spectrum of the stimulus velocity waveform, and $H(f)$ is the frequency-response function between input and output that we measured experimentally (Fig. 5a) [13]. Ignoring spontaneous body sway caused by noise sources internal to the postural control system, the measured RMS sway velocity measures would be directly predicted from (3). Specifically, the integral of $G_{yy}(f)$ across all frequencies is equal to the mean squared value of sway velocity, and RMS sway velocity is the square root of this mean squared value.

The frequency-response functions varied as a function of the prosthesis feedback conditions. The low frequency frequency-response function gains were lowest for the position-dominated feedback conditions, and therefore resulted in a reduced sway velocity (relative to the Tactors OFF condition) at these low frequencies. However, the position-dominant feedback conditions also showed gain enhancement at higher frequencies, and this gain enhancement began at lower frequencies for the position-dominant feedback conditions (Fig. 5b, left). This gain enhancement increased the stimulus-evoked sway velocity at these higher frequencies, effectively canceling out the lower frequency sway reduction afforded by the prosthesis. The net result was very little change in the RMS sway velocity relative to the Tactors OFF condition for the position-dominated feedback conditions. For the velocity-dominant feedback conditions, there were only minimal frequency-response function gain reductions or

enhancements relative to the Tactors OFF condition, and therefore the RMS sway velocity was also similar to the Tactors OFF condition.

The RMS sway position results can also be predicted by (3). In this case, $G_{xx}(f)$ is considered to be the power spectrum of the stimulus position waveform and $G_{yy}(f)$ is the power spectrum of the body sway position waveform. The amplitude of the power spectral components of the PRTS stimulus position waveform was largest at the lowest frequency (0.017 Hz) with higher frequency components decreasing in proportion to the square of their frequency. Because the position-dominant feedback settings produced the greatest reduction in the low frequency frequency-response function gains of any of the prosthesis feedback conditions, (3) predicts that these position-dominant feedback settings should produce a large reduction in stimulus-evoked RMS sway position. The higher frequency gain enhancements in the frequency-response functions had little effect over the RMS sway position measures because the amplitudes of the stimulus position power spectral components at these higher frequencies were small compared to the lowest frequency components.

The ability to predict time-domain sway measures using (3) and the experimentally identified $H(f)$ suggests that our results could be generalized to predict outcomes of experiments that included more natural external perturbations or internal disturbances (e.g. noise in sensory and motor systems that evoke spontaneous body sway during quiet stance). For example, if a disturbance was predominantly composed of lower frequency components, as is consistent with internal disturbances that produce spontaneous body sway [23], then our frequency-response function analysis predicts that use of the prosthesis with position-dominant feedback settings and with appropriate tactor thresholds would be effective in reducing both RMS sway position and velocity. In particular, the power spectrum of spontaneous sway, measured by center-of-pressure displacement, decreases with increasing frequency and 50% of the power is below about 0.3 Hz [14]. Because the spectral bandwidth of spontaneous center-of-mass displacement is even lower than center-of-pressure displacement, the experimental frequency-response functions (Fig. 5) indicate that the prosthesis with primarily position feedback should reduce spontaneous sway by about 50% and the high frequency frequency-response function gain enhancement would have almost no detrimental effect.

A more desirable outcome would have been that a feedback setting could have been identified that significantly reduced both RMS sway position and sway velocity measures both for the stimuli used in this study and for any arbitrary internal or external postural perturbations. This desirable outcome would have occurred if the prosthesis could have reduced the frequency-response function gain across a wide bandwidth. However, no feedback combination reduced frequency-response function gain across all frequencies. Thus, the overall conclusion, based on the RMS sway measures and the frequency-response function analysis, is that the prosthesis settings did not afford full control over postural improvements. The modeling results, discussed next, provide additional insight into what limited the effectiveness of the prosthesis.

B. Model-based Interpretation of Experimental Results

Our second goal was to understand how natural and prosthesis derived orientation information is combined and used for postural control. Our approach was to develop a model that accounts for the major features of the experimental data and explains, via the model structure, how orientation information from the prosthesis was incorporated into the postural control system. Parsimony was considered to be more important than achieving an exact accounting for every possible detail of the experimental results. Our base model was a version of an experimentally validated model that previously had been shown to account for human postural control under similar experimental conditions [15]–[17]. We first verified that model parameters could be found that provided a good fit to the experimental frequency-response function results from the Tactors OFF condition and that these parameter values were consistent with previous

results. A number of variations on this basic model were considered to account for the systematic changes in the experimental frequency-response functions as a function of the prosthesis feedback conditions. The simplest alteration to the basic model that accounted for the main experimental features was to add a separate feedback loop that included a representation of the balance prosthesis and of the sensorineural processing of the tactile information. This processing was represented by a simple mathematical integration of the tactile signal multiplied by a tactor weight (G_{Tactor}). The value of G_{Tactor} characterized the extent to which information from the prosthesis was used by the postural control system. To account for experimental results from the different Tactors ON conditions, it was only necessary to vary the value of G_{Tactor} .

There were three conclusions from the modeling study. First, the sensorineural processing of the gravity-related postural information encoded in the vibrotactile balance prosthesis had very different dynamics compared to the processing of gravity-related information encoded by the vestibular system. Specifically, the prosthesis information was transformed via a rather severe low-pass filtering which we modeled as a pure time integration. The origin of this filtering is unknown and could be due to peripheral mechanical properties, peripheral sensory encoding, or CNS processing. In contrast, the processing of vestibular information was considered to provide a wide bandwidth encoding of body tilt with respect to earth-vertical.

A second conclusion from the modeling study was that subjects incorporated vibrotactile information into their existing postural system predominantly through a process of “sensory addition” as opposed to a “sensory substitution” or “sensory reweighting” process. That is, the modeling results indicate that orientation information from the vibrotactile prosthesis was fed back and *added* to the natural vestibular and proprioceptive feedback without changing the reliance on existing natural sensory orientation information or substituting for the natural sensory feedback.

A potential alternative to sensory addition might have been sensory reweighting or substitution. That is, the subjects might have combined the gravity related orientation information from the prosthesis and the vestibular system. The improved quality of the gravity-related information might have resulted in a shift toward increased reliance on gravity-related orientation cues and a corresponding decreased reliance on surface-related proprioceptive cues. If this process had occurred, the overall result could be described as a substitution of prosthesis information for proprioceptive information and therefore a reweighting toward increased utilization of gravity-related orientation information.

However, there was no evidence for a reweighting process. Due to the severe low-pass filtering in the prosthesis feedback pathway, the bandwidth of orientation information from the prosthesis was very limited compared with the wide-bandwidth vestibular information. This bandwidth mismatch likely precluded a direct combination of the prosthesis and vestibular information by the nervous system. The inability of subjects to use prosthesis information in a sensory reweighting mode is a limiting factor since the postural control system normally uses sensory reweighting among natural sensory systems to control posture as environmental conditions change [15].

A third conclusion from the modeling study was that subjects exhibited a higher reliance on the balance prosthesis information when position-related information was encoded by the prosthesis compared to velocity-related information. This conclusion was based on the fact that the identified values of G_{Tactor} changed systematically as a function of prosthesis feedback conditions with the largest G_{Tactor} values in position-dominant feedback conditions (P100V0 and P75V25) and the smallest values in the velocity-dominant feedback conditions (P25V75 and P0V100).

In a separate study, we investigated whether it was possible to predict this pattern of G_{Tactor} changes as a function of the prosthesis feedback conditions [24]. We hypothesized that systematic changes in G_{Tactor} were related to the optimization of a performance goal with G_{Tactor} changes predicted by minimizing a cost function. A cost function, consisting of a linear combination of body sway angle and the 3rd derivative of body sway (jerk), was able to predict the observed changes in G_{Tactor} with changing prosthesis feedback conditions. The two terms in this cost function are reasonable in that the overall cost function represents a compromise that promotes stability (by limiting the angular excursion of the body) and insures a smoothness of movement (by minimizing jerk [25]).

C. Effect of Increasing the Spatial Resolution of the Vibrotactile Display

We previously hypothesized that the low spatial resolution (3 levels) of the vibrotactile display was a limitation, and that performance might be improved with a higher resolution display providing a more continuous representation of body orientation [12]. To explore this idea, the Fig. 2 model was modified to include 7 tactor levels rather than 3. For the P100V0 setting, which produced the largest frequency-response function changes of all feedback settings, the 7 tactor-level model produced only a slight decrease in gain at low frequencies (<0.1 Hz), no change at high frequency gains, and no changes in the phase curves compared to the 3-level model. Therefore, simply making the vibrotactile display more continuous would not be expected to improve performance unless the more continuous display allowed subjects to change the way their nervous system processed the available vibrotactile information. This finding is in agreement with a previous critical tracking task experiment in that 7 tactor levels did not significantly decrease the effective operator delay compared to the use of 3 tactor levels [10].

D. The Balance Prosthesis as a Sensory Substitution Device for Subjects with Bilateral Vestibular Loss

An ‘ideal’ balance prosthesis for vestibular loss subjects would provide gravity-related orientation information in such a way that this information could substitute for absent or degraded information from the vestibular system, and that sensory reweighting between proprioceptive cues and the gravity-related cues provided by the prosthesis could be used to adjust to changing environmental conditions.

Although vestibular loss subjects were not tested in the current study, one could predict from the current results that sensory substitution and reweighting would not be available to vestibular loss subjects using the vibrotactile prosthesis due to the limited bandwidth of the gravity-related information from the prosthesis. That is, there is a bandwidth mismatch between the wide-bandwidth orientation information available from natural proprioception and the limited bandwidth information from the prosthesis. Although our results predict that vestibular loss subjects would not be able to use sensory substitution or reweighting, they would benefit from the vibrotactile information through sensory addition.

Results from a previous study did test vestibular loss subjects using the vibrotactile prosthesis [12]. In that previous study, the tactor feedback was set to 2/3 position and 1/3 velocity feedback, i.e. P67V33. With this position-dominant feedback setting, the model predicts that, compared to the Tactors OFF condition, frequency-response functions should show a large low frequency gain reduction, gain enhancement at higher frequencies, and phase leads. Results from the previous study are consistent with model predictions (see Fig. 4 and Fig.5 in [12]). Furthermore, there was no evidence that vestibular loss subjects were able to use sensory reweighting to compensate for surface perturbations of increasing amplitudes. The correspondence between the previous experimental results in vestibular loss subjects and the predictions using the model developed from experiments in subjects with normal sensory

function indicate that vestibular loss subjects were using orientation information from the prosthesis in an identical manner. That is, vestibular loss subjects also make use of prosthesis information through a sensory addition mode of operation rather than a sensory substitution mode.

E. Conclusions

Our first goal was to identify the particular combination of position and velocity feedback provided by the vibrotactile prosthesis that would produce the greatest improvement in balance control. Results from measures of body sway and from model-identified prosthesis feedback gains showed that subjects generally were able to utilize position-dominated prosthesis feedback to a greater extent than velocity-dominated feedback. However, our analysis showed that all combinations of position and velocity feedback provided both desirable and undesirable postural effects such that a determination of the combination of feedback that is most effective depends on the frequency distribution of external perturbations or internal disturbances that evoke body sway.

Our second goal was to use a model-based interpretation of the experimental results to understand how orientation information from the prosthesis was combined with natural sensory information and used for postural control. The modeling results showed that orientation information from the vibrotactile prosthesis is effectively heavily low-pass filtered and therefore is very different from the wide-bandwidth dynamic characteristics of the vestibular and proprioceptive systems. The dynamic mismatch between the wide-bandwidth orientation information from natural sensory systems and the low-pass filtered prosthesis information likely precludes a substitution of vibrotactile cues for natural sensory information. Rather, the modeling results indicate that vibrotactile information is incorporated into the existing postural system mainly through a sensory addition mechanism as opposed to a sensory substitution mechanism.

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Schematic of Tactor Activation

Fig 1.

Diagram showing the scheme for controlling tactor activation. At most, only one pair of tactors is activated at a time with the activation depending on the value of the component response (CR) relative to a set of threshold levels (T values). T values were selected such that Upper tactors vibrated when CR reached 75% of the maximum expected CR (CR_{max}). Positive CR values activate front tactors and negative CR values active back tactors.

Fig 2.

Model of human postural control showing contributions of natural sensory system feedback and feedback from the vibrotactile balance prosthesis. The variable s in some blocks is the Laplace transform variable.

PRTS stimulus

Body Sway Response

Fig 3.

Pseudorandom support surface rotation waveform and mean experimental (solid line) and model-predicted (dotted line offset by 1°) body sway response waveforms (N=8 subjects) for several test conditions both with and without prosthesis feedback.

Fig 4.

Box plots showing range and distributions of RMS values of angular body sway (a) and sway velocity (b) for the control condition (Tactors OFF) and for the five different combinations of prosthesis feedback in the Tactors ON conditions.

Fig 5.

Mean experimental frequency-response function gain (left) and phase (right) curves derived from body sway responses to pseudorandom support surface rotations (a). Frequency-response functions show systematic changes as a function of prosthesis feedback settings. Systematic changes are more easily appreciated when frequency-response function gain and phase curves are normalized (b) to the Tactors OFF condition.

Fig 6.

Frequency-response functions derived from model predictions and normalized by the frequency-response function derived from the model prediction for the control (Tactors OFF) condition. In (a), the normalized gain (left) and phase (right) curves show that the main features of the experimental frequency-response functions (compare with Fig. 5b) can be described by the Nonlinear Model which only allowed the value of G_{Tactor} to change across the Tactor ON test conditions. In (b), normalized frequency-response functions from the Linearized Model show qualitatively similar results to the Nonlinear Model, revealing that the quantization of tactor information was not needed to describe the main features of the experimental results.

Table I
Test Conditions

Test Condition	P _{gain}	V _{gain}	Lower Tactor Threshold (T_1) (CR units)	Middle Tactor Threshold (T_2) (CR units)	Upper Tactor Threshold (T_3) (CR units)	Maximum Expected CR (CR_{max}) (CR units)
Control	0.0	0.0	-	-	-	-
P100V0	1.0	0.0	0.42	0.84	1.26	1.68
P75V25	0.75	0.25	0.33	0.66	0.99	1.32
P50V50	0.5	0.5	0.29	0.58	0.87	1.16
P25V75	0.25	0.75	0.30	0.60	0.90	1.20
P0V100	0.0	1.0	0.38	0.76	1.14	1.52

* CR units = Composite Response units. CR is a summation performed in the Labview program of sway position (in degrees) times the position gain factor (P_{gain}) and sway velocity (in degrees/s) times the velocity gain factor (V_{gain}). CR can be considered unitless if P_{gain} has units of degrees⁻¹ and V_{gain} has units of degrees⁻¹.s.

Table II
Model Parameters For Factors Off Condition

Test Condition	K_p (N·m·rad ⁻¹)	Neural Controller	K_D (N·m·s·rad ⁻¹)	τ_d (s)	Time Delay	Sensory Weights	W_{vest}	W_{prop}	T_f (s)	Torque Feedback	K_f (rad·N ⁻¹ ·m ⁻¹)	MSE
Control	1008.4		361.0	0.175	0.40	0.60	0.40	0.60	14.16		0.0011	0.029

Table III G_{Tactor} Model Parameter For Tactors On Conditions

Test Condition	Tactor Gain	MSE
	G_{Tactor}	
P100V0	1.4×10^{-3}	0.074
P75V25	1.5×10^{-3}	0.047
P50V50	1.0×10^{-3}	0.045
P25V75	0.5×10^{-3}	0.036
P0V100	0.2×10^{-3}	0.030