33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

A balance control model predicts how vestibular loss subjects benefit from a vibrotactile balance prosthesis

Adam D. Goodworth, *Member, IEEE*, Conrad Wall III, and Robert J. Peterka, *Member, IEEE*

Abstract—**A balance control model was applied to interpret how subjects with a severe vestibular loss (VL) used vibrotactile information from a balance prosthesis to enhance balance control. Experimental data were from 5 VL subjects standing with eyes closed and responding to continuous pseudorandom surface tilts of the stance platform. Results showed that vibrotactile feedback information reduced sway at frequencies below ~0.6 Hz, but vibrotactile feedback was less effective in reducing sway as stimulus amplitude increased. This experimental pattern was accurately predicted by the model, which was based on time-delayed sensory feedback control. The model predicted that changes to the vibrotactor activation scheme could improve performance of the prosthesis and demonstrated that further improvements might be possible if motor learning, acquired by practice and training, could increase VL subjects' reliance on the prosthesis.**

I. Introduction

THE morbidity and mortality associated with human falls is an important health concern [1]. Because vestibular degradation is a common occurrence in aging populations and is associated with an increased likelihood of falls [2], we were motivated to develop a balance prosthesis that improves balance control in patients with VL as well as other balance disorders. To this end, a vibrotactile balance prosthesis has been developed that provides users with augmented sensory information in the form of a pattern of vibration cues that covey orientation information about body sway relative to earth vertical [3,4,5]. Previous studies demonstrated that subjects were able to use the additional orientation information from the prosthesis to reduce sway during gait [6,7], quiet stance [8], and perturbed stance [9,10].

One of these previous studies [9] used non-parametric system identification methods to quantify the effectiveness of the prosthesis in subjects with normal sensory function and in subjects with severe VL. Experimental frequency response functions (FRFs) were calculated that characterize the sway evoked by pseudorandom surface-tilt stimuli [11]. Both normal and VL subjects showed a reduction in FRF gains at frequencies below about 0.6 Hz when using the prosthesis. A gain reduction indicates that subjects tended to orient more toward earth-vertical, and less toward the tilting surface, when using the prosthesis, thus reducing the amplitude of stimulus-evoked sway. However, the prosthesis appeared to be relatively less effective in reducing stimulus-evoked sway when higher amplitude perturbing stimuli were presented compared to lower amplitude stimuli, and the prosthesis was unable to reduce stimulus-evoked sway in VL subjects to a level that occurs in subjects with normal vestibular function [9, 12].

In a later study, we developed a balance control model that accounted for the specific patterns of FRF gain and phase changes seen in normal subjects using the prosthesis across conditions that varied the pattern of vibrotactile feedback provided by the prosthesis [10]. The purpose of the current study is to determine whether: 1) our balance control model could also explain our previous experimental results in VL subjects, 2) the model could be used to suggest changes in the prosthesis that could improve its performance, and 3) motor learning might contribute to further improvements of balance control in VL subjects.

II. Methods

A. Experimental description

Experimental data used in this study was described previously [9]. Five subjects with severe VL were tested using a protocol approved by the Institutional Review Board at Oregon Health & Science University. VL was verified with measurements of vestibular-ocular reflex gains. Anterior-posterior (AP) sway was evoked in eyes-closed subjects who stood on a tilting surface. Surface tilts occurred continuously according to a pseudorandom waveform that had a peak-peak amplitude of 1, 2, or 4° (see Fig. 1A for sample stimulus waveform). For each stimulus amplitude, two tests were performed: with (Tactors ON condition) and without (Tactors OFF condition) vibrotactile feedback. In all tests, subjects' center-of-mass sway responses with respect to earth-vertical were measured.

Vibrotactile feedback was delivered through 12 small tactile vibrators arranged in three rows and two vertical columns held against the anterior and posterior surfaces of the torso. Tactor activation was based on 1) using inertial motion sensors to measure AP body sway, 2) making an on-line calculation of a composite response (CR) as a weighted combination of 0.67 times the instantaneous sway angle (in

Manuscript received March 26, 2011. This work was supported in part by the National Institutes of Health grant R01-DC6201 and training grant T32-DC005945.
A. D. Goodworth is with the Center for Health, Care, and Well-being, Department of Physical Therapy, University of Hartford, West Hartford, CT 06117 USA (phone: 860-768-5571; e-mail:goodworth@hartford.edu).
C. Wall III is with the Department of Otology and Laryngology, Harvard Medical School, Massachusetts Eye & Ear Infirmary, Boston, MA 02114 USA. (e-mail: cwall@mit.edu.).
R. J. Peterka is with the Department of Biomedical Engineering, Oregon Health & Science University, Beaverton, OR 97006 USA (e-mail: peterkar@ohsu.edu)

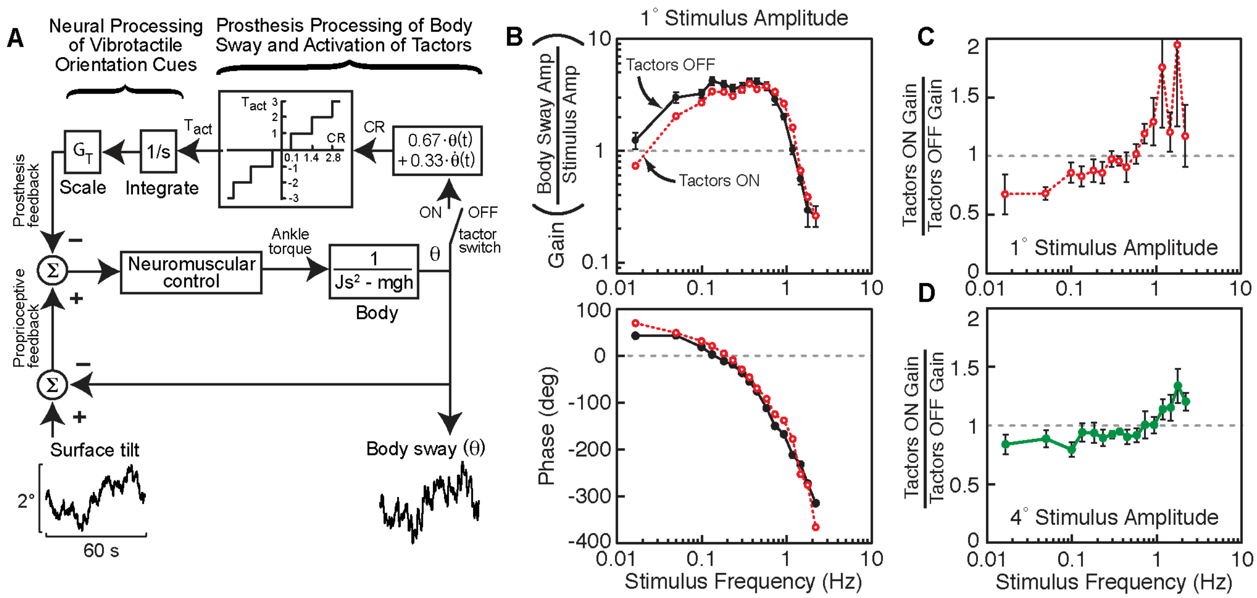

Fig. 1. Balance control model to explain experimental results of subjects with severe vestibular loss. A) Model describes balance system in eyes-closed conditions with and without prosthesis feedback (Tactors ON/OFF). *J* represents the moment of inertia, *mgh* represents the body mass x gravitational acceleration x center-of-mass height, and *s* is the Laplace variable. B) Experimental frequency response functions showed reduced gains in the Tactors ON condition at frequencies below ~0.6 Hz. C and D) Gains in Tactors ON conditions normalized by gains in the Tactors OFF condition. Modified from figures originally published in [9], Copyright (2006), with permission from IOS Press.

degrees) plus 0.33 times the instantaneous sway velocity (in deg/s), and 3) using a step-wise coding scheme to activate tactors based on the *CR* value. The particular weighted combination was selected because both position and velocity feedback is needed to stabilize an inverted pendulum body. A pair of tactors in each row was turned on if the *CR* reached certain threshold values (*CR* threshold values ±0.1, ±1.4, ±2.8; see Fig. 1A), and only one of the 6 pairs of tactors was turned on at any point in time.

Fourier methods were applied to stimulus (surface tilt) and response (body sway angle) data to calculate FRFs [11]. The FRFs were expressed as gain and phase values over frequencies ranging from 0.016 Hz to 2.2 Hz with the gain values at each frequency giving the ratio of body sway to surface tilt amplitude and phase values expressing the relative timing between body sway and surface tilt [12]. Reductions in gains with vibrotactile feedback indicated improvement in balance control.

B. Model description

A block diagram of the model used to describe sway behavior in VL subjects is shown in Fig. 1A. In this model, the prosthesis is represented by the block that calculates *CR* from body sway and sway velocity, and then generates a tactor activation signal, T_{act}, based on the value of *CR* in relation to preset threshold values. The T_{act} signal has only integer values ranging from -3 to 3, such that larger magnitude *CR* values correspond to larger magnitude T_{act} values and the sign of T_{act} encodes the direction of body sway. The nervous system's interpretation of the tactor signal is represented by a mathematical integration ($1/s$) of T_{act} and a gain factor, G_T. The combination of prosthesis information with orientation information from other sensory sources is represented by a simple summation. For VL subjects standing with the eyes closed, the other sensory information used for balance control is limited to only proprioception signaling body sway relative to the surface [12]. The combined sensory orientation information is processed by the "neuromuscular control" block that generates torque about the ankle joint. The neuromuscular control includes: 1) proportional plus derivative control action, 2) a torque feedback component that accounts for low frequency dynamic behavior of the balance control system, and 3) time delay representing all delays in the control system (see [13]). The ankle torque acts on the body, represented as a single-link inverted pendulum, to control body sway.

The model in Fig. 1 is identical to a previously validated model for healthy control subjects [10], with the one exception that in the current model for VL subjects, vestibular feedback was eliminated.

To select model parameters that predict Tactors OFF VL sway, model parameters for the neuromuscular control and body were set equal to those obtained from Goodworth et al. 2009. The justification for using these values is based on the previous finding that neuromuscular control parameters scale with body mass/moment of inertia and are essentially the same in healthy and VL subjects [12]. Fixed parameters for the body were: $J = 90.2$ kg·m^2, $m = 83.3$ kg, and $h = 0.95$ m. Fixed parameters for the neuromuscular control system were: proportional control gain = 1008 N·m·rad^{-1}, derivative control gain = 361 N·m·s rad^{-1}, torque feedback gain = 0.0011 rad·N^{-1}·m^{-1}, torque feedback low-pass filter time constant = 14.2 s, and system time delay = 0.175 s. To account for Tactors ON VL sway, G_T was selected to match the model-predicted FRFs to the experimental FRFs while all other parameters remained at values for the Tactors OFF condition.

III. Results

At all stimulus amplitudes, vibrotactile feedback reduced VL subjects' sway at frequencies below ~0.6 Hz (Fig. 1B top, C, D), increased sway above ~0.6 Hz, and resulted in slight phase leads (Fig. 1B bottom). The gain reductions at low frequencies were more prominent in sway responses to the low stimulus amplitude compared to the high stimulus amplitude (compare Fig. 1C with 1D). Thus, as stimulus amplitude increased, vibrotactile feedback was less effective in reducing postural sway. This pattern of amplitude-dependent gain reduction is opposite to what occurs in subjects with normal vestibular function performing similar tests without a prosthesis [12]. Specifically in normal subjects, gain reductions increase with increasing stimulus amplitude and, additionally, the gain reductions occur over a wider bandwidth.

The pattern of amplitude-dependent reduction in stimulus-evoked sway in VL subjects was explained by the model shown in Fig. 1A. Model-predicted FRFs are shown in Fig. 2A. The model was able to predict the general features of experimental results from VL subjects in both Tactors OFF and Tactors ON conditions. In Tactors ON conditions, model-predicted FRFs showed gain reductions at frequencies below ~0.4 Hz and gain increases at frequencies 0.4-1.5 Hz. The gain reductions were greater for low amplitude surface-tilt stimuli compared to high amplitude stimuli. Normalized gain curves shown in Fig. 2B show greater detail of the model-predicted gain reductions afforded by vibrotactile feedback at frequencies below ~0.4 Hz. The model-predicted normalized gain reductions at lower frequencies compare favorably with the experimental data (Fig. 2C). The model performed better at predicting sway features at lower frequencies (< ~0.4 Hz) compared to higher frequencies (> ~0.6 Hz) where experimental gains were larger than the model-predicted gains.

These model predictions were obtained with a fixed G_T value (0.0004 unitless) across stimulus amplitudes. This fixed value of G_T implies that VL subjects' reliance on prosthesis information did not vary across stimulus amplitude. Therefore, in Tactors ON conditions, changes in FRFs across stimulus amplitude were due to the particular choice of threshold values used to process the prosthesis information that was fed back to the user.

These results suggest that tactor thresholds could be selected that produce FRF gain reductions, across a range of different stimulus amplitudes, that more closely resemble those that occur in normal subjects without a prosthesis [12]. We systematically varied thresholds in the model in order to identify a set of thresholds that achieved a desired pattern of FRF gain reductions relative to the Tactors OFF condition. By selecting thresholds of $CR = \pm 0.6, \pm 1.1, \pm 1.4$, the pattern of gain reductions shown in Fig. 2 was reversed such that gains were now lower in 4° compared to the 1° surface-tilt stimuli (Fig. 3A). This new pattern of decreasing gains with increasing stimulus amplitude more closely approximates results in subjects without vestibular deficits [12]. Because

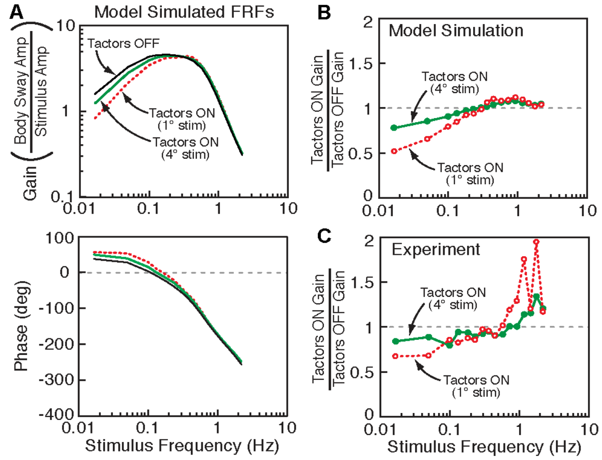

Fig. 2. Model-predicted frequency response functions. A) Model accounts for general pattern of FRFs in Tactors ON and Tactors OFF conditions and predicted reduced effectiveness of vibrotactile feedback as stimulus amplitude increases. B) Normalized model-predicted gains curves. C) Normalized experimental gain curves. Fig. 2C modified from figures originally published in [9], Copyright (2006), with permission from IOS Press.

the predicted maximum gain reduction from 1° to 4° was only ~20% at the lowest stimulus frequency (0.016 Hz), changing thresholds offers only a limited benefit when a fixed value of $G_T = 0.0004$ is maintained.

With training and practice, it may be possible for VL subjects to increase their reliance on orientation information provided by the prosthesis. In the model, an increase in the value of G_T represents an increased reliance on prosthesis information. The results shown in Fig. 3B indicate that pronounced improvements (~70% maximum gain reduction) could be realized if subjects, through motor learning, were able to increase their reliance on prosthesis feedback. However, if G_T increases without adjustments to any neuromuscular control parameters, large gain increases are predicted at frequencies 0.4-1.5 Hz. These higher frequency gain increases can be reduced if increases in G_T are accompanied by small decreases in the neuromuscular control proportional gain (changes in the derivative gain were not effective in reducing gain increases).

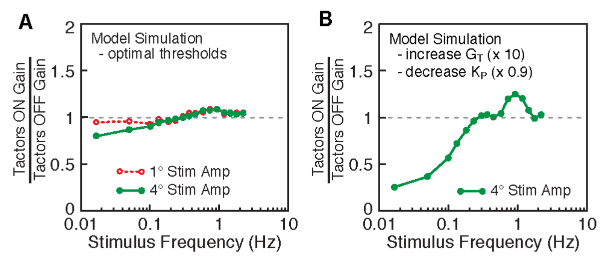

Fig. 3. Model-predicted consequences of altering tactor thresholds and increasing reliance on prosthesis feedback. A) Adjustments to tactor thresholds produced gain reductions with increasing stimulus amplitude that more closely resemble results in subjects with normal vestibular function. B) Potential improvement in low-frequency gain reduction if motor learning increased reliance on prosthesis feedback (increase G_T to 0.004) and slightly decreased neuromuscular control proportional gain (decrease K_P to 907 N·m·rad^{-1}). Tactor thresholds are those used in the original experiments in VL subjects [9].

IV. Discussion

A. Neural processing of vibrotactile information

Results from the current study further validated two concepts, previously described in detail [10], related to neural processing of orientation information provided by the vibrotactile display. First, vibrotactile information is effectively heavily low-pass filtered (modeled as a pure time integration) before being assimilated into the balance control system. In contrast, the processing of vestibular information is considered to provide a wide bandwidth encoding of body tilt with respect to earth-vertical [14]. Although both the vestibular system and the prosthesis encode information about body sway relative to earth vertical, the limited bandwidth of the prosthesis limits its ability to substitute for the missing vestibular information in VL subjects.

Second, modeling results indicate that vibrotactile information is assimilated into the existing balance control system via a sensory addition mechanism rather than a sensory substitution mechanism. Sensory substitution (or sensory reweighting) occurs when subjects shift reliance from one source of sensory information to another [12]. The model-based interpretation of experimental results suggested that VL subjects maintained an essentially identical reliance on proprioceptive information in both the Tactors ON and Tactors OFF conditions and that prosthesis feedback information simply was added to the natural proprioceptive feedback in the Tactors ON condition.

B. Improving the effectiveness of the prosthesis

The model predicted that sway responses of VL subjects to surface-tilt perturbations could be made to more closely resemble response of subjects with normal vestibular function if minor changes were made to tactor activation thresholds and if subjects could be trained to rely more heavily on prosthesis information for balance control.

Model simulations predicted that the values chosen as tactor thresholds can have an important impact on the balance responses. With tactor thresholds of the model set to values used in the experiments with VL subjects, the model accounted for the experimental data from VL subjects showing that the prosthesis was relatively <u>less effective</u> at reducing stimulus-evoked sway at larger compared to smaller stimulus amplitudes. The model predicted that it is possible to select thresholds such that prosthesis becomes relatively <u>more effective</u> at reducing stimulus-evoked sway at larger compared to smaller stimulus amplitudes. The latter pattern of reduced sensitivity to surface-tilt perturbations with increasing stimulus amplitude is seen in subjects with normal vestibular function and has been interpreted to indicate that subjects reweight their reliance on sensory orientation cues by shifting toward increased reliance on vestibular orientation cues and decrease reliance on proprioceptive cues for balance control [12].

Model simulation also predicted that, if subjects could learn to increase their reliance on prosthesis feedback, low frequency sway would be greatly reduced. Further, if subjects could increase reliance on prosthesis feedback and reduce neuromuscular stiffness, then undesirable high frequency gain increases would be minimized. It may be possible that with experience and training, users would be able to better incorporate vibrotactile information into their existing balance control system. This incorporation over time would likely be influenced by the particular weighted combination of position and velocity feedback encoded by the prosthesis [10]; and may be subject specific. Experimental results from the current study were derived from subjects that performed the tests in one 2-hour block of time. Although no learning effects were detected from the beginning to the end of the 2-hour block of time, if longer-term learning could take place, then despite the dynamic mismatch in bandwidth between vibrotactile information and natural vestibular information, large reductions in lower frequencies of sway are predicted by the model if subjects were able to increase their reliance of prosthesis feedback.

References

[1] J.A. Stevens, P.S. Corso, E.A. Finkelstein, T.R. Miller, "The costs of fatal and nonfatal falls among older adults," *Inj. Prev.*, vol. 12, pp. 290–295, 2006.

[2] Y. Agrawal, J.P. Carey, C.C. Della Santina, M.C. Schubert, and L.B. Minor, "Disorders of balance and vestibular function in US adults," *Arch Intern Med*, vol. 169, pp. 938-944, 2009.

[3] C. Wall 3rd, M. S. Weinberg, P.B. Schmidt, and D.E. Krebs, "Balance prosthesis based on micromechanical sensors using vibrotactile feedback of tilt," *IEEE Trans. Biomed. Eng.*, vol. 48, no. 10, pp. 1153-1161, Oct. 2001.

[4] C. Wall 3rd and M. S.Weinberg, "Balance prostheses for postural control,"*IEEE Eng. Med. Biol. Mag.*, vol. 22, no. 2, pp. 84–90, Mar.–Apr. 2003.

[5] P. P. Kadkade, B. J. Benda, P. B. Schmidt, and C. Wall 3rd, "Vibrotactile display coding for a balance prosthesis," *IEEE Trans. Neural. Syst. Rehabil. Eng.*, vol. 11, no. 4, pp. 392–399, Dec. 2003.

[6] M. Dozza, C. Wall 3rd, R. J. Peterka, L. Chiari, and F. B. Horak, "Effects of practicing tandem gait with and without vibrotactile feedback in subjects with unilateral vestibular loss," *J. Vestib. Res.*, vol. 17, pp. 195-204, 2007.

[7] C. Wall 3rd, D. M. Wrisley, and K.D. Statler, "Vibrotactile tilt feedback improves dynamic gait index: a fall risk indicator in older adults," *Gait Posture*, vol. 30, no. 1, pp. 16-21, 2009.

[8] C. Wall 3rd and E. Kentala, "Control of sway using vibrotactile feedback of body tilt in patients with moderate and severe postural control deficits," *J. Vestib. Res.*, vol. 15, pp. 313-325, 2005.

[9] R. J. Peterka, C. Wall III, and E. Kentala, "Determining the effectiveness of a vibrotactile balance prosthesis," *J. Vestib. Res.*, vol. 16, pp. 45–56, 2006.

[10] A. D. Goodworth, R. J. Peterka, and C. Wall 3rd, "Influence of feedback parameters on performance of a vibrotactile balance prosthesis," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 17, no. 4, pp. 397-408, 2009.

[11] R. Pintelon and J. Schoukens. *System Identification: A Frequency Domain Approach.* Piscataway, NJ: IEEE Press, 2001.

[12] R. J. Peterka, "Sensorimotor integration in human postural control," *J. Neurophysiol.*, vol. 88, pp. 1097–1118, 2002.

[13] R. J. Peterka, "Simplifying the complexities of maintaining balance," *IEEE Eng. Med. Biol. Mag.*, vol. 22, no. 2, pp. 63–68, Mar./Apr. 2003.

[14] D.E. Angelaki, M. Q. McHenry, J.D. Dickman, S.D. Newlands, B.J. Hess, "Computation of inertial motion: neural strategies to resolve ambiguous otolith information," *J. Neurosci.* 19: 316–327, 1999.