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Designing Vibrotactile Balance Feedback For Desired Body Sway Reductions

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***Abstract*— Vibrotactile feedback about body position and velocity has been shown to be effective at reducing low frequency body sway (below about 0.5 Hz) in response to balance perturbations while standing. However, current devices cause an undesirable increase in high frequency body sway. In addition, unlike other sensory prostheses such as hearing aids, which are fine-tuned to the user, current vibrotactile balance prostheses largely employ a "one size fits all" approach, in that they use the same settings (i.e. parameter values) for all subjects. Rather than using a fixed design consisting of position and velocity feedback for all subjects, we propose a "custom design" approach that employs system identification methods to identify the feedback required to achieve a desired body sway frequency response for the subject. Our derivations and simulations show that in order to accomplish this objective, feedback consisting of a *subject-specific* filtered combination of body position, velocity and acceleration is required. Simulation results are provided to illustrate the results.**

I. Introduction

The development of a noninvasive balance prosthesis that can improve balance and gait in persons with abnormal vestibular function has become of increased interest in recent years [10], [14], [15]. Vestibular disorders, especially those that affect the vestibular organs bilaterally, lead to increased risk of falls [5] and adversely affect a persons function and quality of life [12], [17]. Accordingly, a balance prosthesis that can noninvasively provide augmented sensory information that improves balance function has clear clinical potential for ameliorating some of the deleterious effects of vestibular disorders.

Sensory substitution balance prostheses aim to augment sensory information available for balance control in an effort to compensate for lost or diminished natural sensory function. While a variety of sensory substitution devices have been developed, including auditory, electrotactile, and vibrotactile feedback, none has seen widespread clinical use, for various reasons. However, vibrotactile feedback seems especially promising, owing to its relative ease of implementation, portability, noninvasive nature, and demonstrated effectiveness at improving balance (i.e. reducing postural sway) in various subject populations [3], [4], [9–11], [14], [16].

Previous studies have explored the efficacy of vibrotactile feedback for improving balance, and to a more limited extent gait, function. In balance-healthy adults, vibrotactile feedback based on position and velocity of body sway has been shown to improve postural control, as quantified by reductions in RMS body sway in response to external perturbations that disrupt balance [4], [9]. Older adults at risk for falls have also shown improvement in balance and gait function while using vibrotactile feedback, as quantified by their Dynamic Gait Index [16]. A study of five patients with vestibular deficits showed some improvement in balance function using vibrotactile feedback, but to a lesser degree than that seen in balance-healthy adults [9]. Another recent study showed improved postural control in eight vestibular patients during vibrotactile feedback in response to multi-directional platform translations [11]. Improvements in tandem walking during vibrotactile feedback have also been reported in vestibular patients [3].

All of the studies cited above have explored the efficacy of vibrotactile feedback that uses measures of body position and velocity to drive tactors that vibrate on the surface of the torso. Various combinations of position and velocity vibrotactile feedback, ranging from 100% position feeback to 100% velocity feedback, have been explored in balance-healthy adults [4], in efforts to identify the combination of position and velocity feedback that best reduced body sway. These investigations found that no combination of position and velocity feedback reduced sway uniformly over frequency; rather, there was reduction in body sway at low frequencies but increased sway at higher frequencies [4], [9]. Importantly, for a given combination of position and velocity, the parameter values used in the algorithm to activate the vibrotactile device were the same for every subject. While the reduction in sway at low frequencies is desirable, the increase in sway at high frequencies is undesirable and likely counters, to some degree, the stabilizing effects of decreased sway at low frequencies.

In this paper, we approach the design of the vibrotactile device as one of a feedback compensator designed to achieve a desired frequency response. In this way, the vibrotactile prosthesis is custom designed for the user, similar to other sensory prostheses (e.g. hearing aids [13]). To develop this idea, we use a previously developed model of postural control in order to gain insights and quantify the nature of the feedback design requirements necessary to achieve a desired balance response. Our derivations and simulations show that,

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for uniform reduction in sway across frequency, the dynamics required of the vibrotactile device are more complicated than just position and velocity feedback with *a priori* chosen parameters. In particular, for a rudimentary model of the postural control system, we find that the vibrotactile device needs to provide *subject-specific* filtered feedback about body position, velocity, and *acceleration*. The custom parameters required would be obtained clinically or in the laboratory by perturbing the subject's natural standing balance and measuring the corresponding body sway, and then applying system identification methods.

II. POSTURAL CONTROL MODEL

A schematic of the postural control model with vibrotactile feedback is shown in Fig. 1. This basic model (without vibrotactile feedback) has been shown to fit experimental balance data reasonably well, under a variety of different conditions and in different subject populations [1], [2], [6–8]. As in [4], we consider the case of eyes closed stance, although the design approach we present is not limited to that case nor to any specific postural control model. We consider small perturbations and angles of body sway, so that the underlying equations of motion are linear. We further assume no adaptation to the perturbation or other time-varying effects, such that the postural control system is also time-invariant. Then, in the absence of vibrotactile feedback (switch open in the figure), the body sway angle (BS) to support surface angle (SS) transfer function is given by

$$G_{cl_0}(s) = \frac{w_p\, G_c(s)\, G_p(s)}{1 + (w_p + w_v)\, G_c(s)\, G_p(s)} \tag{1}$$

where the body is modeled as a linearized single-link inverted pendulum,

$$G_p(s) = \frac{1}{Js^2 - mgh} \tag{2}$$

In order to stabilize the inverted pendulum, a torque in opposition to the gravitational torque must be applied, based on the position and velocity of the pendulum; hence, for the controller, we use proportional-derivative control,

$$G_c(s) = K_p + K_d s \tag{3}$$

With this controller and inverted pendulum model, the transfer function (Eq. (1)) becomes

$$G_{cl_0}(s) = \frac{w_p\,(K_p + K_d s)}{Js^2 + WK_d s + WK_p - mgh} \tag{4}$$

where $W = w_p + w_v$ and under steady-state conditions, $W = 1$ [7]. In the equations above, h is the height to the center of mass, m; J is the moment of inertia of the body (inverted pendulum); $g = 9.8\ \text{m/s}^2$ is acceleration due to gravity; K_p and K_d are scalar constants (stiffness and damping, respectively) that generate a weighted-sum of body angle position and velocity information for balance control; w_p is a scalar constant representing proprioceptive feedback from the ankles; and w_v is a scalar constant representing vestibular feedback.

III. VIBROTACTILE FEEDBACK COMPENSATION

As shown by Goodworth et al. [4], the vibrotactile device adds another feedback path to the postural control system (switch closed ["on"] in Fig. 1). We denote the vibrotactile feedback by the transfer function $H_f(s)$ in Fig. 1, which consists of two parts: one representing the physical vibrotactile device to be designed and that would be worn by the patient ($H_T(s)$), and the other representing the sensory processing of the vibrations on the skin; Goodworth et al. [4] have shown that this sensory processing is well modeled by an integrator with some fixed gain. Accordingly, the vibrotactile feedback path has transfer function

$$H_f(s) = \frac{G_T}{s} H_T(s) \tag{5}$$

In previous designs and experiments [4], [9], [11], [14], [16], the vibrotactile device produced vibrations on the skin based on a weighted sum of measured body position and velocity, such that the tactor device transfer function was

$$H_T(s) = H_{T_{PV}}(s) = P + Vs \tag{6}$$

where P and V are constants, and we use the subscript "PV" to denote the specific case of a position-velocity vibrotactile device. As noted previously, this form of vibrotactile feedback has the benefit of reducing sway at low frequencies, but yields increased sway at higher frequencies.

Rather than *a priori* deciding what the dynamics of the vibrotactile device should be, we solve for the vibrotactile feedback $H_f(s)$ necessary to achieve a desired frequency response (for example, uniform reduction in sway across frequency). To do this, let us first solve for the body angle to platform transfer function when the vibrotactile feedback is present (switch closed in Fig. 1), which is given by

$$G_{cl_1}(s) = \frac{G_{cl_0}(s)}{1 + \frac{H_f(s)}{w_p}\, G_{cl_0}(s)} \tag{7}$$

Hence, in order to achieve a desired response G_{cl_1}, the required vibrotactile feedback transfer function is

$$H_f(s) = w_p\, \frac{G_{cl_0}(s) - G_{cl_1}(s)}{G_{cl_0}(s) G_{cl_1}(s)} \tag{8}$$

Next we consider some specific cases for reducing sway in a desired manner over frequency, and the resulting vibrotactile device transfer function $H_T(s)$ that is required.

A. Uniform sway reduction

Let us first consider the case where we wish to reduce body sway uniformly across frequency, such that the transfer function with the vibrotactive feedback on is $G_{cl_1}(s) = c\, G_{cl_0}(s)$, where $0 < c < 1$. For that case, we require

$$H_f(s) = \frac{1-c}{c} \frac{w_p}{G_{cl_0}(s)} \tag{9}$$

Thus, by Eq. (4) and (5), the transfer function for the tactor device to implement is

$$H_T(s) = \frac{ks}{K_p + K_d s} \left(Js^2 + K_d s + (K_p - mgh) \right) \tag{10}$$

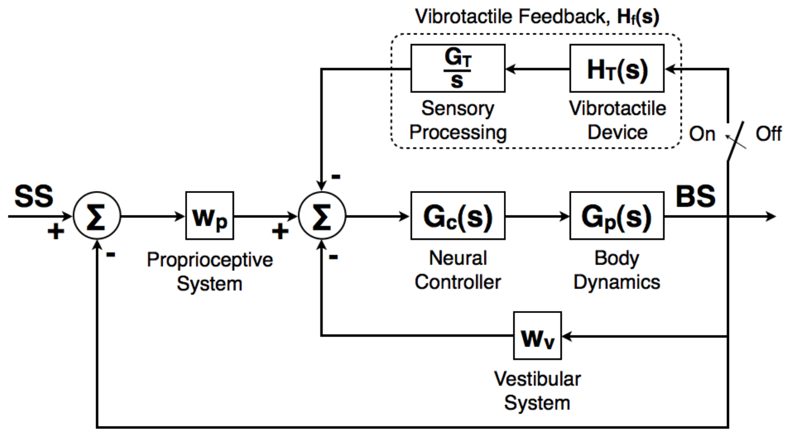

Fig. 1. **Postural control model for eyes-closed stance, with vibrotactile feedback $H_f(s)$, which consists of the actual physical vibrotactile device that would be worn by the subject, represented by $H_T(s)$, and the sensory processing of the tactor vibrations on the skin, represented by $\frac{G_T}{s}$. The body is modeled by $G_p(s)$ and neural processing of sensory information to generate corrective torque is modeled by the controller $G_c(s)$. The scalar constants w_p and w_v represent the relative contributions of proprioceptive and vestibular sensory feedback to postural control during eyes-closed stance. SS denotes possible perturbations to balance produced by movement of the support surface on which subjects stand, and BS is body sway with respect to vertical. [Adapted from [4]]**

where $k = \frac{1-c}{c\,G_T}$. Thus, we observe that, in order to reduce sway uniformly over frequency, the tactor device must provide (high-pass filtered) body motion information consisting of position ($K_p - mgh$ term), velocity ($K_d s$), and *acceleration* (Js^2), with *subject-specific parameters* (J, K_p, K_d, m, h), and not position-velocity (PV) information with subject-independent parameters P and V. For the PV device (Eq. (6)), the gain reduction is non-uniform, with transfer function ratio of tactors on to tactors off given by

$$\frac{G_{cl_1}(s)}{G_{cl_0}(s)} = \frac{s(Js^2 + K_d s + K_p - mgh)}{s(Js^2 + K_d s + K_p - mgh) + G_T(P + Vs)(K_d s + K_p)} \tag{11}$$

A plot of this ratio for varying degrees of position and velocity feedback is shown in Fig. 2. Consistent with previous experimental and analytic findings [4], [9], there is greater gain reduction at lower frequencies, and increased gain at high frequencies for some cases.

B. Frequency-selective sway reduction

Suppose that, instead of uniform reduction of sway across frequency, we wish to reduce the sway more at higher or lower frequencies. For example, to reduce sway more at higher frequencies, we can design the vibrotactile device such that the transfer function ratio of tactors on to tactors off is given by

$$\frac{G_{cl_1}(s)}{G_{cl_0}(s)} = \frac{c}{1 + \tau s} \tag{12}$$

Solving as above for the tactor device transfer function we obtain

$$H_T(s) = \frac{1}{c\,G_T}\frac{s(1 - c + \tau s)}{K_p + K_d s}\left(Js^2 + K_d s + (K_p - mgh)\right) \tag{13}$$

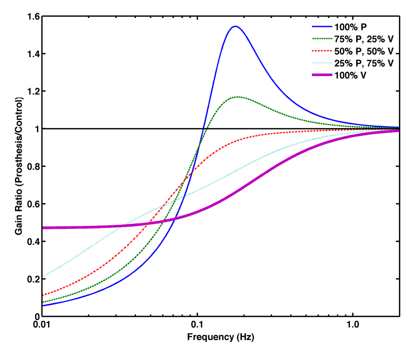

Fig. 2. **Ratio of postural body sway (BS) with vibrotactile feedback to BS without vibrotactile feedback, in response to platform (support surface, SS) perturbation, for the model of Fig. 1 with varying amounts of position-velocity (PV) vibrotactile feedback. Consistent with previous experimental and analytic findings, PV vibrotactile feedback does not decrease sway uniformly over frequency. The body parameter values used in the model simulations were [4]: body mass excluding the feet $m = 83.3\,$kg, body moment of inertia $J = 90.2\,$kg$\cdot$ m^2, and height to center-of-mass $h = 0.95\,$m. The neural controller parameters K_p and K_d, representing stiffness and damping, were 1008.4 Nm$\cdot$ rad^{-1} and 361.0 Nm$\cdot$s$\cdot$ rad^{-1}, respectively. The proprioceptive sensory weight, w_p was set to 0.6. The fixed gain G_T was set to 0.26.**

Thus, to achieve greater sway reduction at higher frequencies, the tactor device again provides subject-specific filtered position, velocity, and acceleration feedback, but the filter characteristics differ from the high-pass filtered feedback for uniform sway reduction. In Fig. 3, the gain ratios for the vibrotactile device with uniform gain reduction, high frequency gain reduction, and position-velocity feedback are plotted, for comparison.

Alternately, we could design the device to attenuate lower frequency sway more than high frequency sway, such that the transfer function ratio is

$$\frac{G_{cl_1}(s)}{G_{cl_0}(s)} = \frac{c\tau s}{1 + \tau s} \tag{14}$$

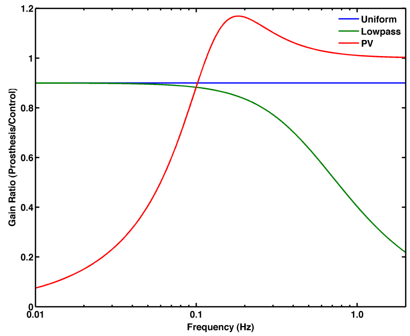

Fig. 3. **Ratio of postural body sway (BS) with vibrotactile feedback to BS without vibrotactile feedback, in response to platform (support surface, SS) perturbation, for the model of Fig. 1 with a device that provides 10% reduction uniformly over frequency ($c = 0.9$), compared to a device designed to attenuate high frequency sway more (labeled "lowpass"), versus a device that provides position-velocity feedback.**

For this case, the tactor transfer function is

$$H_T(s) = \frac{1}{c\,\tau\,G_T}\frac{1+(1-c)\tau s}{K_p + K_d s}\left(Js^2 + K_d s + (K_p - mgh)\right) \tag{15}$$

Again, the device provides subject-specific filtered position, velocity and acceleration feedback.

IV. Conclusions

We proposed a new custom-design approach that utilizes system identification methods to identify the body dynamics and subject-specific parameters required in order for vibrotactile feedback to achieve a desired frequency-specific reduction in body sway. The general form of the vibrotactile device transfer function was found to be

$$H_T(s) = H(s)\left(Js^2 + K_d s + (K_p - mgh)\right) \tag{16}$$

where the parameters (J, K_p, K_d, m, h) are *subject-specific*, and $H(s)$ is a filter that also depends on subject-specific parameters as well as the desired frequency-specific body sway reductions (see Eqs. (10), (13), (15)).

Although a simplified postural control model was used to illustrate the proposed vibrotactor feedback design, the basic custom design idea for vibrotactile feedback is not restricted to the particular postural control model; the solution to achieve a desired frequency response is given by Eq. (8), where G_{cl_0} is the transfer function of the subject without vibrotactile feedback, and G_{cl_1} is the desired response with vibrotactile feedback. In practice, the parameters required for the appropriate feedback strategy could be obtained by perturbing the subject's balance and applying system identification methods to fit a model to the data (e.g. [1]).

The model-based analysis presented here allows one to quantify the inability of position-velocity vibrotactile feedback to achieve uniform sway reduction over frequency, and moreover to discern the elemental form of vibrotactile feedback necessary to achieve sway reduction over a broader range of frequencies. To achieve this objective, our analysis showed that the vibrotactile prosthesis needs to provide subject-specific feedback consisting of filtered body position, velocity, and acceleration.

More sophisticated postural control models can be employed to gain additional insights about the form of vibrotactile feedback needed to achieve desired results. The resulting vibrotactile device may involve more complicated dynamics than (filtered) position-velocity-acceleration feedback. Some of these designs may not be realizable in that the required $H_T(s)$ may be unstable or noncausal. In such cases, one may be able to implement stable or causal approximations. While these issues remain to be explored, our primary conclusion is that, in order to improve the effectiveness of vibrotactile prostheses to reduce body sway, the device must be custom-fit to the user, with subject-specific parameters that can be obtained by applying system identification methods to the subject's response to balance perturbations.

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