A Portable Audio-biofeedback System to Improve Postural Control

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Abstract—**This paper introduces a portable audio-biofeedback (ABF) system that encodes the signals provided by a linear accelerometric sensor into a stereo sound. This sound is relayed to the subjects via headphones and can enhance the ability of subjects to perceive trunk accelerations. We tested this system on nine healthy subjects while they stood in three conditions listening to the ABF representation of their trunk accelerations. The ABF significantly improved the subjects' balance in all three conditions. The subjects reported that they were comfortable wearing and using the ABF device. Results suggest that devices such as this ABF system may be used for balance training and balance rehabilitation therapy.**

Keywords—**Accelerometer, acoustic feedback, balance, dynamometric platform, wearable sensor, postural control, prosthesis.**

I. Introduction

A complex interplay of feedback and feedforward control ensures the ability of subjects to maintain balance [1-2]. Visual, vestibular, and somatosensory receptors provide the central nervous system with sensory information on body orientation and motion in space for the maintenance of balance. Balance deficits are sometimes associated with incomplete environmental information supplied by these senses. One method to improve balance, which has been widely used in physical therapy and rehabilitation, involves providing additional environmental information via biofeedback systems [3]. Biofeedback systems, providing visual center of pressure (COP) displacement information, and have been used to stabilize balance in healthy subjects [4] have been applied successfully for rehabilitation therapy in hemiplegic [5] and stroke subjects [6].

These biofeedback systems were not portable and involved the use of expensive laboratory equipment. Currently, technological advances have offered new solutions for portability and cost. Advances in miniaturizing movement sensors and a growing interest in non-invasive patient-monitoring techniques have facilitated the development of many systems that are portable and not cumbersome. In particular, the use of new-generation accelerometers spurred the creation of small sensors that assess body kinematics [7-8]. Accelerometers and gyroscopes have been combined to obtain kinematics data comparable in accuracy to motion analysis data [9].

The validity of trunk-acceleration information sensed by accelerometers for posture analysis was demonstrated by comparing trunk-acceleration information with force-platform information [10]. In addition, accelerometers have been successfully tested to assess simulated body movements [11]. These studies suggest that trunk acceleration is meaningful and reliable to assess postural control. Consequently, trunk-movement information may be a good substitute for COP displacement data in biofeedback systems.

Wall *et al.* [12] developed a wearable vibrotactile feedback device based on trunk-tilt. A sensor combining accelerometric and gyroscopic information detects trunk tilt. Based on this detection, a vibrator matrix is activated, representing tilt estimation. This study showed how vibrotactile feedback is able to improve balance performance for healthy [13] and vestibular [14] subjects. It also showed the possible utility of such a system as a prosthesis for people with pathologies that impair balance.

Using tactile, instead of visual, feedback, subjects can perform several tasks while wearing a biofeedback device. However, visual feedback potentially encodes more information than vibrotactile feedback.

In this paper, we present an audio-biofeedback (ABF) system that provides trunk-acceleration information to subjects through the auditory channel. We modulated a stereo sound to encode the 2-D acceleration of the trunk sensed by a portable accelerometric sensor [7] and fed back to the subjects, the sound. The experiments described here provide evidence that the ABF improves balance. These results suggest that portable and inexpensive devices, such as our ABF, may be used for balance training and balance rehabilitation therapy.

II. The design of the ABF system

A diagram of the ABF system architecture is shown in Fig. 1. The system, designated to be portable, has three major components: a sensory unit, a processing unit, and an audio rendering unit. Figure 1 also shows a force plate that is not part of the ABF system. This force plate was used as a cross-validation device to estimate body sway.

A. System Components

The sensory unit is as described in Giansanti *et al.* [7]. It consists of a cell with two linear uni-axial accelerometers (3031-Euro Sensor, UK). The accelerometers are orthogonally mounted and included in a sensor box (5 cm x 5 cm x 2.5 cm). The sensor box is mounted on the subject's back (L5) to measure the accelerations of the trunk. The accelerometers provide the trunk-acceleration information for the processing unit.

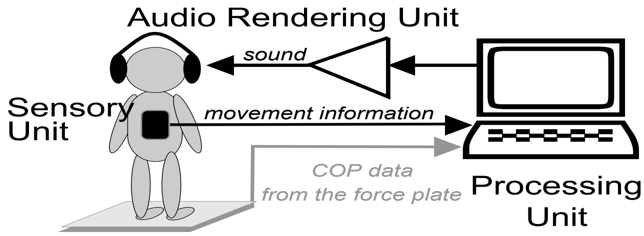

Fig. 1. Diagram of the prototype audio-biofeedback system.

The processing unit consists of a Toshiba commercial laptop computer equipped with a DAQ card (NI 6024E). The accelerations are acquired through the DAQ card using Matlab and its Data Acquisition Toolbox. Using Matlab, the laptop processes in real-time a digital signal that represents a subject's trunk accelerations. The DAQ card then converts the digital control signals into binaural synthetic sounds and sends them to the audio rendering unit.

The audio rendering unit, consisting of an amplifier (Fostex PH-5) and a pair of headphones (Sony SBC HP-140), adjusts the sound for the human ear. The laptop also digitizes and stores all of the 3-D linear and angular data sensed by the sensory unit as well as the force plate data (reaction forces and torques along the three axes).

B. Acceleration and Sound Correlation

ABF maps anterior-posterior (AP) and medial-lateral (ML) trunk accelerations into a stereo sound that is modulated in frequency and amplitude, and adjustable in balance. The right and left channels constituting the stereo sound are independently adjustable. The sound is updated with a 20-*Hz* refresh rate.

The stereo-sound coding functions are shown in Fig. 2. As seen in Figs. 2(A) and 2(C), the larger the absolute value of the AP or ML acceleration, the greater the volume of the sound. Figure 2(B) shows the relationship between pitch and AP acceleration: positive accelerations (corresponding to forward movements) increase the sound frequency in both channels, while negative AP accelerations (corresponding to backward movements) decrease the sound frequency in both channels. ML accelerations, as shown in Fig. 2(D), are encoded as changes in the stereo-sound balance. Leftward trunk accelerations induce a volume increase in the left headphone and a volume decrease in the right one. Rightward trunk accelerations induce a volume increase in the right headphone and a volume decrease in the left one.

C. Implementation of Correlations

During quiet stance, humans maintain balance by swaying in a range of ±1 deg from the vertical [10]. COP displacements that are compatible with this range of sway determine the area of normal sway. We refer to this area as the reference area (RA). While subjects sway inside the RA, ABF provides a constant pure tone with the frequency f_0. The RA, expressed in terms of AP and ML accelerations, represents the threshold for changing the ABF sound, as Fig. 2 shows. The size of RA is slightly different for each subject because the computer determines the size based on a subject's height. In particular, the computer uses an inverted pendulum to model each subject and to estimate the RA. The RA is the estimated area in which the inverted pendulum accelerations are compatible with the inverted-pendulum oscillations in the range of ±1 deg from the vertical. Preliminary calculations showed that human forward sway is about 1.5 times larger than sway in the other directions. Thus, we uses the value obtained from the inverted-pendulum model to set the anterior RA threshold, and we multiply this value by a coefficient of 2/3 to obtain the posterior, left and right RA thresholds (Fig. 2).

A body can maintain its balance in a static condition if its center-of-mass projection falls inside its support base. The human body maintains its balance in a stable condition by keeping the center-of-mass projection inside its support base. The computer calculates the support base of a subject based on feet anthropometry. The boundary of the support base represents the subject's personal limits of stability. This support base, defined by its limits of stability, represents the safety area (SA). We use the interval between SA and RA to determine the dynamic range of the stereo sound, as shown in Fig. 2.

The sigmoid functions (Figs. 2(A) and 2(C)) explain the changes in volume that correspond to the AP and ML accelerations. The sigmoid functions are determined by:

$$V = \frac{Vl\ A_i^k}{A_i^k + m^k} + c \tag{1}$$

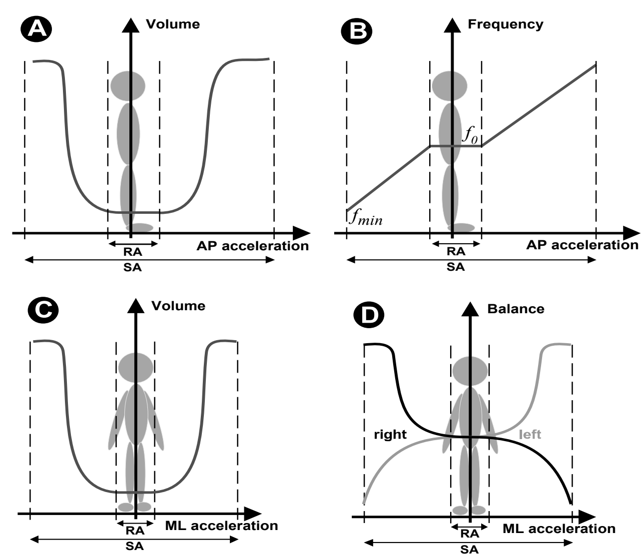

Fig. 2. A. Sigmoid volume-modulation function based on AP accelerations expressed by (1). B. Linear frequency-modulation function based on ML accelerations expressed by (2). C-D. Volume and stereo balance-modulation functions based on ML movements and expressed by (3) and (4).

where V is the volume; A_i is the acceleration sensed along the axes (i=AP, ML); and k, m, Vl, and c are constants that we determined empirically.

The frequency modulation function associated with AP acceleration, shown in Fig. 2(B), follows a piecewise linear law:

$$f = \begin{cases} s_{pos} \dfrac{A_{AP} - SA_{pos}}{RA_{pos} - SA_{pos}} + f_{min} & \text{if } A_{AP} < RR_{pos} \\ f_0 & \text{if } A_{AP} \in RR \\ s_{ant} \dfrac{A_{AP} - RA_{ant}}{SA_{ant} - RA_{ant}} + f_0 & \text{if } A_{AP} > RR_{ant} \end{cases} \quad (2)$$

where s_{pos}, s_{ant}, f_{min}, and f_0 are constants, RA_{ant} is the forward RA threshold, RA_{pos} is the backward RA threshold, SA_{ant} is the forward SA threshold, and SA_{pos} is the backward SA threshold.

The balance between the left and right audio channels is regulated by changing the stereo balance (Fig. 2(D)):

$$w = 1 - e^{-kd} \quad (3)$$

where w represents the balance between the channels, k is a constant, and d represents the distance from the RA edge to the SA edge, that is, how far the subject sways outside the natural sway region, in percentage. The volume from the left and the right channels is computed as:

$$\begin{aligned} V_{left} &= (1 + w)\, V \\ V_{right} &= (1 - w) V \end{aligned} \quad (4)$$

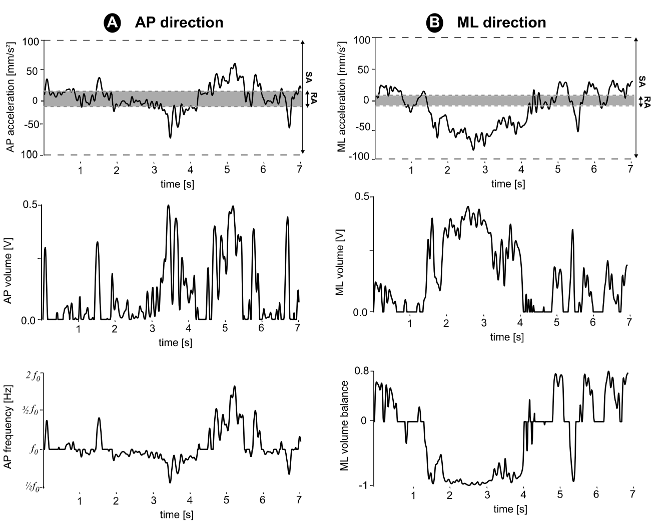

Fig. 3. Computation of ABF volume, frequency, and balance from (A) AP and (B) ML accelerations using equations (1), (2), (3), and (4).

When the ML sway is narrower than the RA width, w is set to 0, that is, the stereo sound presents the same volume in both the left and the right channels.

Figure 3 shows the effect of (1), (2), (3), and (4) on the frequency, volume, and balance of the ABF stereo sound when the AP and the ML accelerations individually change.

III. EXPERIMENTAL PROTOCOL

An experiment was performed to evaluate the influence of ABF on the balance of nine healthy subjects in quiet stance. Our hypothesis is that ABF improves their balance.

The ages of the subjects ranged from 33 to 71 years, with an average of 55. Their height ranged from 151 to 180 cm, with an average of 167. Their weight ranged from 65 to 86 kg, with an average of 73.

Each subject performed 13, 60-second trials standing on the force plate in three different conditions. Each subject performed 5 trials with eyes closed (EC), 5 trials with eyes open while standing on foam (EOF), and 3 trials with eyes closed while standing on foam (ECF). The EC condition was used to limit balance information provided by sight. The subjects stood on foam to limit somatosensory information from the feet. The subjects also performed these same 13 trials wearing headphones and listening to the ABF stereo sound. For all three conditions, each subject was instructed to stand inside the RA (normal sway area), that is, to keep the sound corresponding to the RA constant in the headphones during the trials. The order of all 26 trials was random. Before the experiment, the subjects practiced for 1 minute while listening to the ABF system to experience the relationship between sound and movement.

During each trial, trunk accelerations from the sensory unit and COP data from the force plate were recorded with a 100-Hz sample rate. Data analysis was focused on COP-based parameters. In particular, following Rocchi et al. [15], we analyzed these parameters: root mean square distance (COP-RMS), mean velocity (COP-MV), frequency comprising 95% of the power (COP-F95%), frequency dispersion (COP-FD), and the direction of maximum variability (COP-|90-MDir|).

IV. RESULTS & DISCUSSION

The subjects were most unstable in condition ECF, more stable in EOF, and the most stable in EC. ABF reduced COP-RMS and increased COP-F95% in all conditions ($p<0.05$). ABF did not consistently affect COP-FD and COP-|90-MDir|. In addition, ABF increased the mean COP-MV but not statistically significantly. The effects of ABF on all parameters were greater in condition ECF, less so in EOF, and the least in EC. Figure 4 shows the effects of ABF on the COP parameters analyzed in condition ECF. In condition ECF, ABF reduced COP-RMS by 15%,

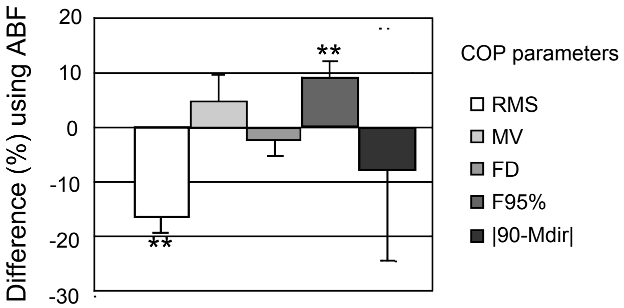

Fig. 4. Effects of ABF on COP parameters in the ECF condition. ABF stabilizes subjects, improving their balance. Error bars indicate standard errors. **: $p<0.01$.

increased COP-MV by 5%, and increased COP-F95% by 9.5%. ABF had no substantial influence on COP-FD and COP-|90-Mdir|.

Using ABF, subjects swayed less, as shown by the reduction of COP-RMS. In other words, they were further from their stability limits and, consequently, more stable. In addition, using ABF, subjects applied more postural corrections to their sway, as shown by the increase of COP-F95%. As shown by COP-RMS and COP-F95%, subjects using ABF increased their stability and postural corrections and, consequently, improved their balance. The more that subjects were unstable in the three conditions, the more they improved their balance with ABF. Thus, the efficacy of ABF in stabilizing subjects was proportional to their lack of sensory information. This result suggests that ABF substituted for the subjects' lack of visual (eyes closed) and somatosensory (standing on foam) balance information. In addition, subjects reported that the ABF system was comfortable and the ABF sounds were not bothersome.

It is noteworthy that subjects improved their balance after practicing for only 1-minute, suggesting that subjects can readily apply ABF. With longer practice, it may be possible for subjects to improve their balance even more. It is unknown how multiple, 1-minute practice trials may affect subjects' retention of improved balance. From preliminary studies (unpublished), subjects who used ABF for 15, continuous, 1-minute practice trials retained their improved balance immediately after using ABF. Additional studies are under way to determinate the length and nature of this retention.

V. Conclusion

The balance improvements induced by the ABF system, as well as the simplicity and low cost of this system, suggest that ABF devices may be suitable for balance training, rehabilitation therapy, activity monitoring, and assessment of body balance.

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