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Strategies and synergies underlying replacement of vestibular function with prosthetic feedback.

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Abstract— **This study investigated changes in movement strategies and muscle synergies when bilateral peripheral vestibular loss (BVL) subjects are provided prosthetic feedback of their pelvis sway during stance. Six BVL subjects performed 3, for them, difficult stance tasks: standing eyes closed, on a firm surface, on a foam surface, and standing eyes open on foam. Movement strategies were recorded as roll and pitch ratios of upper and lower body velocities with body-worn gyroscopes. Surface EMG recordings were taken from two pairs of antagonistic, lower leg and trunk muscles in order to note synergy changes with feedback. Subjects were first assessed without feedback. Then they were provided stance training with vibro-tactile and auditory feedback of pelvis angle sway, and finally reassessed with the same feedback active. For analysis of movement strategies, angle values integrated from angular velocity samples, were split into 3 frequency bands (<0.7, 0.7-3, and >3 Hz). Feedback caused a reduction in pelvis sway angle displacements to values of age-matched healthy controls (HC) for all tasks. Pelvis sway velocity was only reduced for the task with largest angle displacements, standing eyes closed on foam. Movement strategies in each frequency band examined were unaltered by feedback, except for amplitude, and were not different from those of HCs before or after use of feedback. Low frequency motion was in-phase as if the upper and lower body moved as an inverted pendulum, high frequency motion anti-phasic. Amplitudes of EMG were reduced with feedback. Synergies recorded in the form of activity ratios of antagonistic muscle pairs were reduced with feedback.**

This is the first study that demonstrates how vestibular loss subjects achieve a reduction of sway during stance with prosthetic feedback. Unchanged movement strategies with reduced amplitudes are achieved with reduced antagonistic muscle synergies. This study has implications for the choice of feedback parameters (angle or velocity) and patient groups when using prosthetic devices to reduce sway of those with a tendency to fall.

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I. INTRODUCTION

Loss of vestibular function is well-known as a factor underlying an increased tendency to fall in older persons [1]. Furthermore, a number of persons with less than 60 years of age, having either a unilateral or bilateral peripheral vestibular loss (UVL or BVL) have difficulty maintaining their balance particularly for complex gait tasks such as climbing stairs, or for stance tasks for which visual and proprioceptive inputs are reduced, such as standing on a carpet in the dark [2]. For these reasons, a number of investigators have developed balance prostheses to provide these persons a replacement for vestibular sensory information on their center of mass sway. Such prosthetic systems generally rely on vibro-tactile or auditory feedback or both modes, appropriately coded with body sway information [3-9]. Despite variations in where the sway measures are taken, and how the sway signals are processed, and at which body location the feedback is provided, the general conclusion is that such biofeedback helps UVL and BVL patients improve their balance during stance and gait [3,7,9].

The reduction in sway with prosthetic feedback achieved by vestibular loss subjects raises several questions concerning how this reduction is achieved. An important question concerns whether position or velocity feedback should be used. Early results suggest position feedback was more effective than velocity feedback [6] and when employed leads to reduction of angle rather than velocity of sway [5]. Nonetheless others have used a combination of angle and velocity feedback with success [7-9]. Another crucial question is whether, for effective sway reduction, patients need to use a movement strategy for balance corrections that is appropriate for the feedback mode being employed with the prosthesis. One way to answer this question is to examine a patient group that have a similar but exaggerated movement strategy to healthy controls [10] and on whom balance feedback is known to be effective [4,7].

Vestibular loss patients have movement strategies during stance that are similar to those of controls, whereas those with lower-leg proprioceptive loss have different movement strategies [10]. For this reason, it has been suggested that subjects with vestibular loss might be more responsive to biofeedback modes that function well for healthy controls [11]. Thus the first question this study sought to answer was whether improvements in balance control achieved by vestibular loss subjects using artificial sway position feedback were brought about using the same movement strategy as when no feedback was available. This question is by no means as simple as considering body sway during stance as similar to that of an inverted pendulum, because the upper and lower parts of the body move with two modes simultaneously during stance [10,12]. One mode is like an inverted pendulum with in-phase motion of the pelvis and trunk and the other mode is an anti-phase motion of the 2 segments. Thus the muscle synergies possibly underlying reductions in amplitudes of these movement strategies in two modes must at least be driven by muscles acting at the ankle joints and at the trunk, unless as shown by Goodworth et al. [9], improvements are only present in the inverted-pendulum mode of motion. The typical changes in muscle synergies with vestibular loss observed in response to rotating surface perturbations could cause changes in both modes of motion described above for stance. These changes consist of reduced ankle muscle activity but increased trunk muscle activity with respect to healthy controls [13,14]. Thus the second question we have attempted to answer with this study is how muscle synergies are changed when biofeedback of pelvis sway is provided to BVL subjects.

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II. Methods

Fourteen adult subjects (6 bilateral peripheral vestibular loss (BVL) subjects and 8 age-matched healthy control subjects) were included in this study. The BVL subjects were out-patients at the University Hospital of Basel, Switzerland. The BVL subjects ranged between 45 and 54 years of age (mean 50.0; SEM 2.6) and the controls had a mean age 49.0 (SEM 4.0). Exclusion criteria for the healthy subjects included self-reported sensory, neurological or musculoskeletal impairments that could interfere with balance and inability to stand on one leg, eyes closed, for 20 seconds without falling.

Two gyroscope based systems, Swaystar (Balance International Innovations GmbH, Switzerland), measured pelvis and upper trunk angular velocities in the roll and pitch planes. Trunk measurements were taken at the level of the shoulders. These angular velocities were sampled at 100 Hz with 16 bit accuracy over a range of 327 deg/s and then transferred wirelessly to a PC which computed angle changes via trapezoid integration [15]. Muscle activity was measured with pairs of surface, silver-silver chloride EMG electrodes. These electrodes were placed 3cm apart, along several muscles: left M tibialis anterior, left M soleus, bilaterally at M external obliquus, and bilaterally at M paraspinalis at L4-L5.

A BalanceFreedom feedback system (Balance International Innovations) provided biofeedback of pelvis sway to the participants using signals from the pelvis gyroscopes. Actuators for the feedback were mounted on a head band and were active once angle thresholds for activating the vibro-tactile, auditory and visual actuators were exceeded. Feedback thresholds were based on individual values of the 90% ranges of pelvis sway in the pitch and roll directions for the 70 sec duration of each task in the first assessment. If a loss of balance occurred, the task was repeated a maximum of two times and the trial with the largest duration was used. The thresholds in each direction were set at 40% of the 90% range for vibrotactile signals (that is, a range equal to 80% of the 90% measurement), 80% for the acoustic signals and 150% for the visual threshold**.** Once activated, each feedback signal remained active as long as its threshold was exceeded. The vibrotactile signals were sent to 1 of 8 vibrators in the headband set at 45˚ intervals around the head. A vibrator switched on when sway threshold was reached in the direction of the sway and this direction had the largest sway amplitude. The acoustic feedback consisted of two bone-conducting acoustic actuators placed above the ears at the level of the mastoids. The left actuator was activated with one frequency when the acoustic threshold was reached for sway to the left, the right actuator with another frequency when swaying to the right, and both conductors with a higher and lower frequency when swaying backwards and forwards, respectively. The visual feedback served as a flashing warning signal [5].

Pelvis and trunk sway and EMG signals were recorded while two-legged stance tasks were performed, without shoes, and with the arms hanging alongside the body. Two stance tasks were performed on a foam surface, eyes open and eyes closed and one task eyes closed on a normal surface. For the first assessment, the sway was recorded for 70 s during these tasks, and the individual feedback thresholds were determined from this data. Then, subjects rested for 20 mins before 30 mins of training was provided with biofeedback. The training tasks were the same as the assessment tasks but also included tandem stance on a firm surface, eyes open and closed. After another short pause of 5 minutes, subjects were reassessed on the 3 stance tasks with the feedback active. During all tests, two spotters stood close to, but behind, the subjects in order to prevent a potential fall.

Once the original velocity data was integrated into angular data, the low frequency trend, determined by applying the dynamically parameterized denoise function of the "Rice Wavelet Toolbox" [16], was subtracted from the original data. The resulting data were subsequently filtered and separated into three frequency bands: low pass (<0.7 Hz), high pass (>3.0 Hz) and band pass (0.7-3.0 Hz) – see figure 1. This filtering was implemented using simple 3rd order Butterworth filters running forwards and backwards over the data.

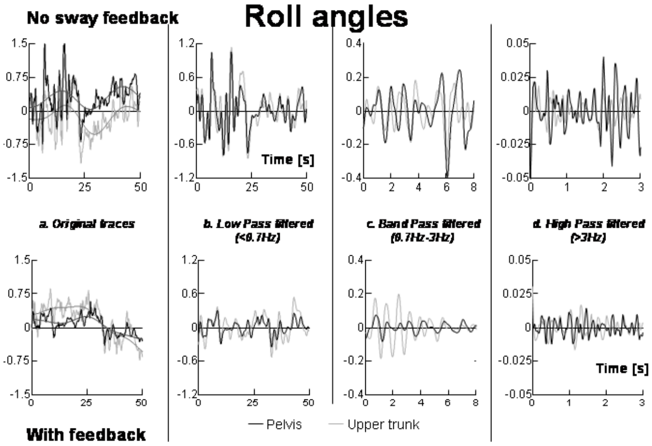

Figure 1. Improvement in roll sway for a BVL subject when provided biofeedback of pelvis sway The upper four panels show the sway angles in degrees of the pelvis and upper trunk while standing eyes closed on foam with no biofeedback, the lower panels with biofeedback. a. 50 seconds of the original traces of upper trunk and pelvis sway with general trend lines. b. The same traces as in a. after removing the general trend and low pass filtering. c. 20 seconds of the same traces as in a. after removing the general trend and band pass filtering. d. 5 seconds of the same traces as in a. after removing the general trend and high pass filtering.

For angle samples in each frequency band, total least-squares regression lines were calculated with respect to the x-y plots of roll respectively pitch data [17 – see figure 3]. The angle regression slopes were further processed and visualized by applying circular statistics [18]. For calculations the «Circular Statistic Toolbox» update 2010b published by P. Behrens [19] was used. In the visualization plots (unit circles) an outcome of, for example -70º, was depicted as 110º, using tan(x) = tan (180+x). Additionally, with the help of the MATLABs Signal Processing Toolbox Version 6.13 (R2010A) we calculated power spectral densities (PSDs) and PSD ratios for the EMG data (after 100 Hz low-pass filtering) data. Fast Fourier transformation was performed on a window size of 2048 samples (20s) with an overlap of 1024 samples.

III. RESULTS

Across all bandwidths considered combined vibro-tactile and auditory feedback reduced sway angles at the pelvis and the upper trunk. Figure 1 provides an example of the improvement for a typical BVL subject who had considerable difficulty to stand eyes closed on the foam surface. Across stance conditions, almost all of the improvement was in the angular position rather than in the angular velocity of sway. The exception was for the condition eyes closed on foam. Figure 2 shows the mean BVL population 90% ranges of sway at the pelvis for the eyes closed and open on foam compared to mean values of the healthy controls (HC).

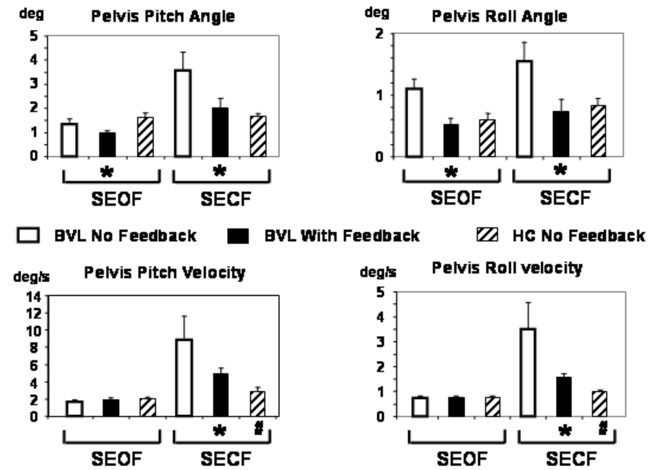

Figure 2. Populations angle and angular velocity means of 90% pelvis sway ranges with and without feedback. The column height represents the mean population pelvis angle or angular velocity for each task. Values are shown for stance tasks on foam with eyes open (SEOF) or closed (SECF). The vertical line above the column indicates the standard error of the mean. BVL stands for bilateral vestibular loss population, HC for the healthy control population. BVL means with feedback marked with * have a significant decrease compared to means with no biofeedback. If the biofeedback means of BVL subjects remained significantly greater than healthy controls with feedback the HC values are marked with #.

The sway amplitudes of BVL subjects in roll and pitch were significantly reduced with feedback to levels that were not different from those of HCs. There was no change in velocities when BVL subjects stood with feedback eyes open on foam, but the velocities were not different from HCs even without feedback. In contrast, velocities standing with eyes closed on foam were reduced with feedback, but levels were still greater than those of HCs (figure 2).

Phases representing movement strategies present between the lower and upper body were examined using correlation plots of pelvis and trunk angle divided into low (<0.7 Hz), middle (0.7 to 3 Hz), and high (>3 Hz) band widths. In-phase movement-synergies between the trunk and pelvis would imply that the body moved as an inverted pendulum. Correlation plots would be lines with a slope of 45°. The example low pass plots of figure 3 indicate near in-phase low pass (LP) movements with pitch more in phase than roll. This trend is shown for the population values shown in figure 4a. The mid frequency (0.7 to 3 Hz) movements were restricted to mostly trunk movements on a fixed pelvis.

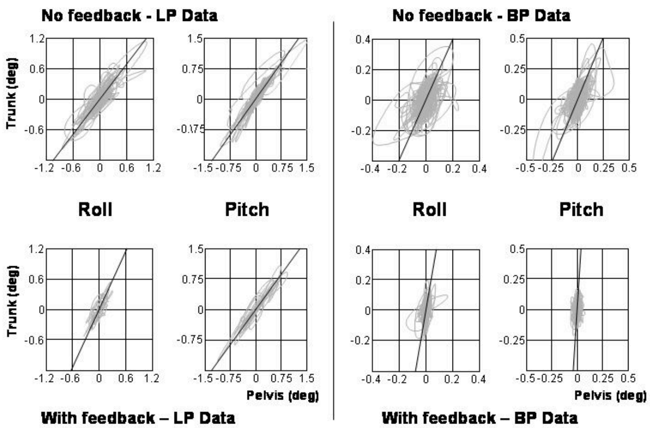

Figure 3. Regression slopes of trunk versus pelvis movements of a BVL subject standing eyes closed on foam with and without feedback. Regression slopes after low-pass filtering (LP) and band-pass filtering (BP) of trunk and pelvis sway angle traces are shown. The upper panel shows the slopes without biofeedback, the lower ones with biofeedback. On the left are the low pass regressions, on the right the band-pass regressions.

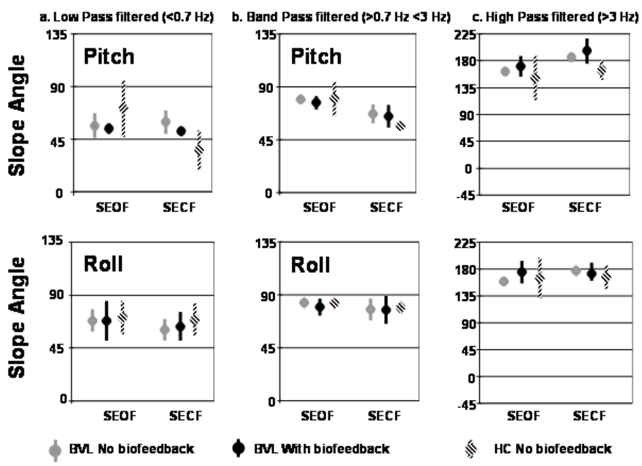

Figure 4. Mean regression slopes of trunk angles with respect to pelvis angles. Mean regression slopes are depicted for the two foam tasks, eyes open and closed, *without* and *with* biofeedback. The upper panels show the slope angles in pitch, the lower panels in roll. a. slope values for low-pass filtered angle values, b. band-pass filtered angles and c. for high-pass filtered values. The bullet symbol marks the mean value of the slope; the vertical line depicts the 95% confidence intervals of the mean.

High frequency (>3 Hz) movements were characterized by anti-phase motion (regression lines values closer to 135°). When BVL population values of the phase relationships were examined in each frequency band without and with feedback, no changes in these characteristics were observed when feedback was provided. Further, the phase relationships did not differ from those of HCs.

The data of BVL subjects described in figures 2-4 is consistent with a decrease in amplitude modulation when sway feedback is provided rather than any change in movement strategy. The question arises how these changes in modulation are brought about. The patterns of underlying muscle activity shown in figure 5 suggest three mechanisms. One mechanism appears to involve changes in the level of background activity in trunk muscles participating in roll and pitch movements (see figure 5 left). The second mechanism is the change in the depth of modulation seen in both trunk and ankle muscles as seen in figure 5, and the third mechanism was a reduction of tibialis anterior/soleus and paraspinal/external obliquus activation ratios to levels of healthy controls.

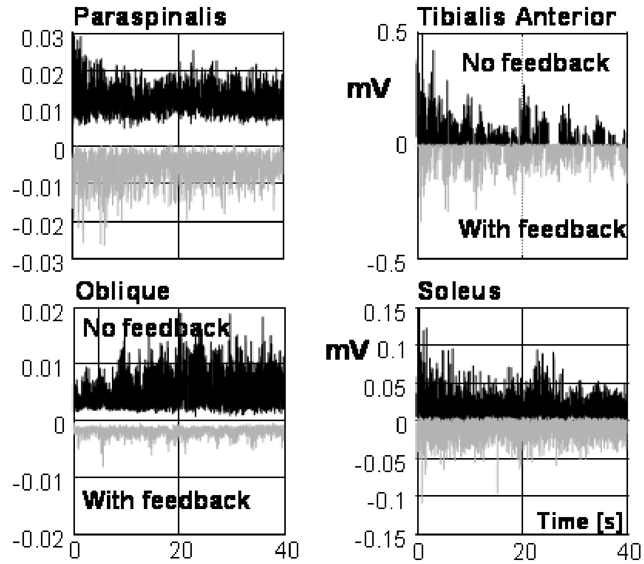

Figure 5. Smoothed muscle activity with and without feedback for the task standing eyes closed on foam. Muscle activity of 2 ankle and 2 trunk muscles without feedback provided is shown in the upper traces of each panel, the activity from the same muscles with feedback is shown inverted in each panel. Data from a BVL subject.

IV. Discussion

The results of this study indicate that for sway amplitudes up to 10 Hz (the high frequency limit of our analysis) combined vibro-tactile and auditory feedback provides improved control of balance for BVL subjects in the form of broad-frequency sway reductions. The same balance correcting movement strategies were employed when feedback of pelvis sway was provided. That is, the same movement strategies as used by normal subjects. Furthermore, these same strategies were employed when feedback was not available. The amplitudes of oscillations for both in- and anti-phase movement strategies were reduced by employing three types of muscle action, reduced background activity, reduced depth of modulation, and reduced muscle activation ratios. Presumably reduced background activity reduces intrinsic muscle stiffness and with it less extensive sway oscillations for the same muscle modulation. However, the action was reinforced by smaller muscle modulation and lower antagonistic activation ratios once sway angle information on CoM motion was provided.

The reductions in sway we observed were mostly in sway angles, but as sway angle became larger with the task of standing eyes closed on foam, velocity reductions occurred as well. This raises the question about the best way to code sway information in a balance prosthesis in order to produce the most effective feedback. Goodworth et al. [9] argued that vibro-tactile feedback provides only low frequency (<0.6 Hz) information on sway. This appears to be counter-intuitive as these authors used 3 pairs of vibrators set with increasing thresholds and combined angle and angular velocity signals to activate the vibrators. However, it is possible that mainly position information could be extracted from our vibro-tactile feedback signals and velocity information was obtained by the acoustic feedback which increased in volume when sway was larger. We noted no difficulties for the patients with combining these 2 types of feedback into motor commands. The reactions to the feedback may even be reflex responses as it is hard to envisage the improvements we noted for movements with content greater than 3Hz (see figure 1d) being due to voluntary reactions. The visual feedback we provided was only active for large sway angles and not available under eyes closed conditions. Another aspect of coding that needs to be considered is whether fixed thresholds (based on population mean values) or individual thresholds are employed. In this study we used individually based thresholds, because otherwise there is a risk that thresholds will be too high for the subjects with less instability [5].

This study may have implications for other patient groups when these receive biofeedback to reduce abnormal body sway. Here we have emphasized that vestibular loss subjects use the same in- and anti-phase movement strategies with feedback as healthy controls and their movement strategies were similar to those of controls even without feedback. That is, vestibular inputs act to provide appropriate modulation of balance correcting strategies which are presumably triggered by proprioceptive inputs [20]. Lower-leg proprioceptive loss subjects use different strategies involving holding the pelvis stable and moving the upper trunk more [11]. For such patients, it is an open question whether these patients need to change their balance correcting strategies to those of healthy controls before they can be aided by the feedback schemes of this and other studies [4,7,8] or whether the feedback characteristics need to be changed appropriately to fit their abnormal balance correcting strategies.

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