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RESEARCH ARTICLE

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# The effects of stochastic galvanic vestibular stimulation on human postural sway

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**Abstract** Galvanic vestibular stimulation serves to modulate the continuous firing level of the peripheral vestibular afferents. It has been shown that the application of sinusoidally varying, bipolar galvanic currents to the vestibular system can lead to sinusoidally varying postural sway. Our objective was to test the hypothesis that stochastic galvanic vestibular stimulation can lead to coherent stochastic postural sway. Bipolar binaural stochastic galvanic vestibular stimulation was applied to nine healthy young subjects. Three different stochastic vestibular stimulation signals, each with a different frequency content (0–1 Hz, 1–2 Hz, and 0–2 Hz), were used. The stimulation level (range 0.4–1.5 mA, peak to peak) was determined on an individual basis. Twenty 60-s trials were conducted on each subject – 15 stimulation trials (5 trials with each stimulation signal) and 5 control (no stimulation) trials. During the trials, subjects stood in a relaxed, upright position with their head facing forward. Postural sway was evaluated by using a force platform to measure the displacements of the center of pressure (COP) under each subject's feet. Cross-spectral measures were used to quantify the relationship between the applied stimulus and the resulting COP time series. We found significant coherency between the stochastic vestibular stimulation signal and the resulting mediolateral COP time series in the majority of trials in 8 of the 9 subjects tested. The coherency results for each stimulation signal were reproducible from trial to trial, and the highest degree of coherency was found for the 1- to 2-Hz stochastic vestibular stimulation signal. In general, for the nine subjects tested, we did not find consistent significant coherency between the stochastic vestibular stimulation signals and the anteroposterior COP time series. This work demonstrates that, in subjects who are facing forward, bipolar binaural stochastic galvanic stimulation of the vestibular system leads to coherent stochastic mediolateral postural sway, but it does not lead to coherent stochastic anteroposterior postural sway. Our finding that the coherency was highest for the 1- to 2-Hz stochastic vestibular stimulation signal may be due to the intrinsic dynamics of the quasi-static postural control system. In particular, it may result from the effects of the vestibular stimulus simply being superimposed upon the quiet-standing COP displacements. By utilizing stochastic stimulation signals, we ensured that the subjects could not predict a change in the vestibular stimulus. Thus, our findings indicate that subjects can act as "responders" to galvanic vestibular stimulation.

**Key words** Vestibular system · Posture control · Balance · Cross-spectral analysis · Coherency · Human

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## Introduction

Galvanic vestibular stimulation has proven to be a valuable technique for studying the role played by vestibular information in the control of stance and balance (Njiokiktjien and Folkerts 1971; Coats 1972a, 1973; Nashner and Wolfson 1974; Honjo et al. 1976; Magnusson et al. 1990a,b; Johansson and Magnusson 1991; Iles and Pisini 1992; Britton et al. 1993; Fitzpatrick et al. 1994; Krizkova and Hlavacka 1994; Hlavacka et al. 1995; Inglis et al. 1995; Johansson et al. 1995; Cass et al. 1996; Day et al. 1997). With this technique, small-amplitude galvanic current (less than 4 mA) is delivered transcutaneously to the vestibular afferents that lie directly below the mastoid bones. This serves to modulate the continu-

ous firing level of the peripheral vestibular afferents. Specifically, cathodal (negative) currents increase the firing rate of vestibular afferents, whereas anodal (positive) currents decrease the firing rate of vestibular afferents (Goldberg et al. 1984). Thus, constant bipolar galvanic current produces a tonic vestibular asymmetry. This effect causes a standing subject to lean in different directions depending on the polarity of the current (Coats and Stoltz 1969) and the direction of the subject's head (Lund and Broberg 1983; Hlavacka and Njiokiktjien 1985). In general, a subject will tend to lean toward the anodal stimulus (i.e., in the direction of the vestibular apparatus with reduced afferent activity levels) and/or away from the cathodal stimulus (i.e., away from the vestibular apparatus with increased afferent activity levels).

A considerable number of studies have examined the body-sway response to constant galvanic stimulation of the vestibular system. Coats (1973), for instance, used monopolar monaural constant galvanic stimulation and demonstrated that the amplitude of the body-sway response increases linearly with increasing stimulus current (from 0.2 to 1.0 mA). Hlavacka and Njiokiktjien (1985), on the other hand, used bipolar binaural constant galvanic stimulation and showed that the direction of the evoked sway is approximately in the direction of the intermastoid line. Thus, with bipolar binaural constant galvanic stimulation, lateral sway is produced if a subject's head is facing forward, whereas anteroposterior sway is produced if a subject's head is turned to the left or right (i.e., over the left or right shoulder).

A limited number of studies have shown that the application of sinusoidally varying bipolar galvanic currents to the vestibular system can lead to sinusoidally varying postural sway (Coats 1972b; Hlavacka and Njiokiktjien 1985, 1986; Petersen et al. 1994, 1995). With sinusoidal galvanic stimulation, as with constant galvanic stimulation, the body tends to sway toward the positive stimulus and away from the negative stimulus (Coats 1972b). For low-frequency stimulation, the frequency of the evoked body sway matches the frequency of the stimulus, whereas the amplitude of the evoked body sway varies from subject to subject (Hlavacka and Njiokiktjien 1986). For high-frequency stimulation, the evoked response is dominated by the body's physical characteristics (Coats 1972b). These results offer some insight into the interaction of the vestibular system with the biomechanics of the body. However, it is important to note that the interpretation of these results may be confounded by the highly predictable nature of sinusoidal stimulation signals, i.e., it is possible that the reported results were influenced significantly by learning or anticipatory physiological mechanisms. The influence of such mechanisms could be reduced or avoided by using stochastic (i.e., aperiodic) galvanic stimulation protocols. [We note that others, such as Guitton et al. (1986), have used stochastic stimuli in motor control experiments in order to avoid the effects of possible anticipatory mechanisms.] To date, only one investigation (Fitzpatrick et al. 1996) has explored the use of a stochastic signal in galvanic vestibular stimulation. In that study, Fitzpatrick and colleagues were interested in measuring the loop gain of postural reflexes in humans. They used stochastic galvanic vestibular stimulation to evoke leg muscle activity and determine the role of lower-limb muscles in postural reflexes.

In this study, our objective was to test the hypothesis that stochastic galvanic stimulation of the vestibular system can lead to coherent stochastic postural sway. To test this hypothesis, we applied stochastic galvanic vestibular stimulation to healthy young subjects. We measured the displacements of the center of pressure (COP) under each subject's feet to determine the subject's response to the vestibular stimulus, and we used cross-spectral measures to quantify the relationship between the applied stimulus and the resulting COP time series.

## Materials and methods

Nine healthy young subjects (6 women and 3 men, aged 18–30 years; height 1.63–1.91 m, mean 1.71 m; body weight 43.1–86.2 kg, mean 62.8 kg) were included in the study. The subjects had no evidence or history of a neurological, gait, postural, or skeletal disorder. Informed consent was obtained from each subject prior to participation. This study was approved by the Boston University Charles River Campus Institutional Review Board.

Postural sway was evaluated by using a Kistler 9287 multicomponent force platform to measure the displacements of the COP under a subject's feet (Fig. 1). (It is important to note that the displacements of the COP do not correspond to the displacements of the body's center of mass, particularly at high frequencies.) Each subject was instructed to stand upright on the platform in a standardized stance; the subject's feet were separated mediolaterally by a distance of 1–2 cm (Day et al. 1997). During the test, the subjects stood barefoot with their arms crossed in front and their head facing forward. Subjects were required to close their eyes and wear headphones to block out visual and auditory cues, respectively.

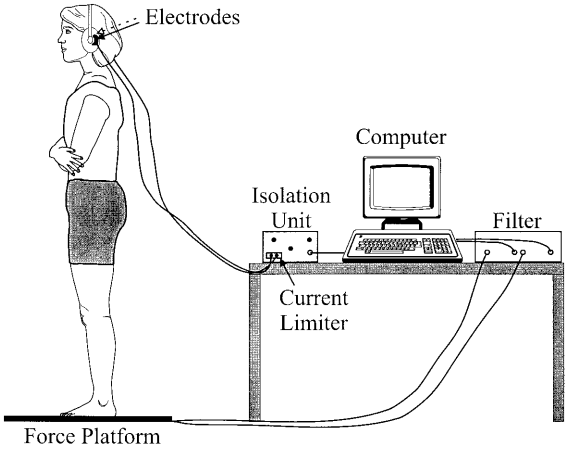


**Fig. 1** A schematic diagram of the experimental setup. Each subject stood on a force platform with their arms crossed in front. Two carbon-rubber surface electrodes were placed on the mastoid bones of each subject, one behind each ear, in order to apply the galvanic vestibular stimulation. The stochastic vestibular stimulus was formed digitally on a computer. The stimulus was transmitted via a D/A board to an isolation unit, which was connected to the electrodes via a current-limiting device. The displacements of the center of pressure (COP) under the subject's feet were measured with the force platform. The COP time series were low-pass filtered during data acquisition and then stored on the computer

Two flexible, carbon-rubber, surface electrodes were placed on the mastoid bones of each subject, one behind each ear, in order to apply the galvanic vestibular stimulation (Fig. 1). A conductive adhesive gel was used to ensure proper conduction between the skin and the electrodes and to keep the electrodes in place. The electrodes were approximately 9 $cm^2$ in area and kidney-shaped to fit comfortably behind the ears. Stochastic current stimuli were applied binaurally and bipolarly to each subject. The anodal electrode was positioned behind the right ear of each subject, and the cathodal electrode was positioned behind the left ear. The stochastic stimulus was formed digitally on a computer. The stimulus was transmitted via a digital-analog (D/A) board to an isolation unit (BAK Electronics; model BSI-1), which was connected to the electrodes via a current-limiting device (Fig. 1).

The stimulus amplitude for individual subjects was determined using the following protocol. Each subject was galvanically stimulated using a sine wave (1–2 Hz), and the amplitude of the stimulus was gradually increased until periodic sway at the input frequency was observable. The subject's stimulation level (range 0.4–1.5 mA, peak to peak) was then used as the maximum amplitude limit during the stimulation trials for that subject.

The stimulus $x(t)$ used for galvanic vestibular stimulation was a realization of a stochastic process, given by the first-order autoregressive difference equation (Brockwell and Davis 1991):

$$x(t)=ax(t-1)+\varepsilon(t),\ \varepsilon(t)\sim N(0,\ \sigma^2) \tag{1}$$

From a physical standpoint, this process describes a relaxator that is driven by white noise $\varepsilon(t)$, with variance $\sigma^2$. The spectrum $S(\omega)$ of this process is continuous (i.e., it contains all frequencies) and its power is distributed such that it is inversely related to frequency $\omega$ (Brockwell and Davis 1991) according to the expression:

$$S(\omega)=\frac{\sigma^2}{2\pi}[1+a^2-2a\cos(\omega)]^{-1} \tag{2}$$

The second-order spectral properties of this process are thus similar to those of quiet-standing COP data, the power spectra of which decrease with increasing frequencies (Collins and De Luca 1993). This process is, therefore, a more "natural" choice for a stochastic posture stimulus than a white noise signal, whose power is distributed equally over all frequencies. The shape of the process spectrum (i.e., the distribution of the process variance over the frequency) is determined by the relaxation time $\tau$, which is related to the parameter $a$ (Eq. 1) by the expression $a=\exp(-\Delta t/\tau)$, where $\Delta t$ denotes the sampling interval. In this study, we used a relaxation time of 1 s and a sampling interval $\Delta t$ of 0.01 s. Note that the variance $\sigma^2$ of the driving noise $\varepsilon(t)$ is not of importance here, since the stimulation levels were determined by the stimulation hardware as described in the previous paragraph.

The autoregressive process was filtered, by multiplying a box with smoothed edges with the Fourier transform of $x(t)$ and then transforming it back to the time domain (for a detailed description of digital filters, see Hamming 1989). Three stimulation signals were created, each with a different frequency content: 0–1 Hz, 1–2 Hz, and 0–2 Hz. Each of the signals contained a part of the continuous spectrum, e.g., the 0- to 1-Hz stimulus contained the entire frequency band from 0–1 Hz, with the shape of the aforementioned autoregressive process. Note that the filter we used is not of special importance given that it is linear and given that we only used it to limit the bandwidth of the stimulus signal to the respective frequency band. Each of the three stimulation signals (duration 60 s) was used in five different trials. Each trial was 60 s in duration and subjects were galvanically stimulated throughout each trial. In addition to the stimulation trials, five 60-s quiet-standing trials, without galvanic stimulation, were conducted on each subject. Thus, in total, 20 trials were conducted on each subject – 15 stimulation trials and 5 control (no stimulation) trials. The presentation order of the stimulation and control trials was randomized. The displacements of the COP during each trial were measured with the force platform. To prevent aliasing effects, the COP data were low-pass filtered at 30 Hz during data acquisition. All data were sampled at 100 Hz and stored on a computer for off-line analysis.

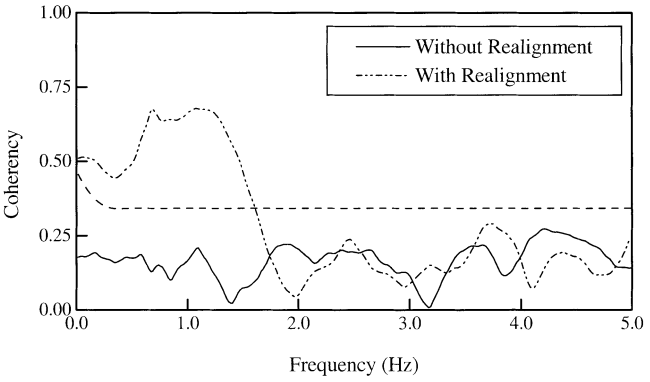


**Fig. 2** Plots of the coherency between the 0- to 2-Hz stochastic vestibular stimulation signal and the resulting mediolateral COP time series for a single 60-s trial from one subject. Results are shown for the two time series without and with a realignment of 0.62 s. The *dashed line* indicates the level of significance, *s*, for $\alpha$=0.95 (see Eq. 6)

## Data Analysis

The cross-spectrum $CS(\omega)$, where $\omega$ is frequency, of two stationary, zero-mean time series $x(t)$ and $y(t)$ is defined as the Fourier transform (FT) of the cross-correlation function $CCF(t')=\langle x(t)y(t-t')\rangle$, where $\langle\cdot\rangle$ denotes expectation. The coherency spectrum $Coh(\omega)$ is defined as the modulus of the normalized cross-spectrum $CS(\omega)$

$$Coh(\omega)=\frac{|CS(\omega)|}{\sqrt{S_x(\omega)S_y(\omega)}} \tag{3}$$

where $S_x(\omega)$ and $S_y(\omega)$ denote the power spectra of $x(t)$ and $y(t)$, respectively, the FT of the respective autocorrelations (Brockwell and Davis 1991; Timmer et al. 1996, 1998). The coherency can be interpreted as a measure of linear predictability (Priestley 1989; Brockwell and Davis 1991) – it equals 1 whenever $x(t)$ is a linear function of $y(t)$.

The estimation of the power and cross spectra is achieved by a direct spectral estimation (Brockwell and Davis 1991, Priestley 1989), based on the discrete FT of the recorded data. The periodogram, which is the squared modulus of the discrete FT, is smoothed by a window function $W_j$ to obtain a consistent estimator of the spectra (Bloomfield 1976; Priestley 1989; Brockwell and Davis 1991). The simplest form of such a procedure is a sliding average. We chose a triangular window (i.e., the so-called Bartlett estimator) to calculate the spectra because its statistical properties are superior to those of a sliding average. We then estimated the coherency by replacing the spectra in Eq. 3 with their respective estimated quantities. For each trial, we investigated the coherency between the stochastic vestibular stimulation signal $x(t)$ and the resulting COP time series (mediolateral and anteroposterior, respectively). It is possible, however, that estimation bias due to misalignment (Hannan and Thomson 1971, 1973; Carter 1987; Priestley 1989) results in an underestimation of coherency. To control for this effect, we realigned all time series, i.e., $x(t)$ and the resulting COP time series, using an iterative procedure described by Bloomfield (1976). In short, we performed all calculations using $x(t-d)$ instead of $x(t)$, since it is expected that the COP time series lags $x(t)$ by a certain delay $d$. The delay $d$ was estimated using the phase spectra $\Phi(\omega)$ defined by the relationship (Brockwell and Davis 1991)

$$CS(\omega)=|CS(\omega)|\exp[i\Phi(\omega)] \tag{4}$$

Figure 2 provides plots of the coherency between the 0- to 2-Hz stochastic vestibular stimulation signal and the resulting mediolateral COP time series for a single 60-s trial from one subject, without and with a realignment of 0.62 s. It can be seen that realignment resulted in a significant increase in the amount of coherency found between the two time series.

At first glance, the estimated realignment delay $d$ might be interpreted as the delay between the input and the measured output signal, which is a physiologically meaningful parameter. However, the term "delay" can be misleading from a physiological standpoint and thus the obtained results for $d$ should be interpreted carefully. One reason for this is that the realignment delay $d$ might be frequency-dependent, i.e., there might be some nontrivial dispersion relation (Hamon and Hannan 1974; Nakano and Tagani 1988). In such a case, a constant estimated realignment delay $d$ would represent a "best fit" of a straight line to a function exhibiting some nontrivial curvature. In other words, if we assume a linear system, the output $y(t)$ can be written as

$$y(t) = \int_d^\infty b(t')x(t-t')dt' \quad (5)$$

where $b(t')$ is the so-called impulse response that characterizes the system. If and only if $b(t')=\delta(d-t')$, where $\delta(\cdot)$ is Dirac's delta distribution, can the realignment delay $d$ be interpreted as a real delay between the input and output (Nakano and Tagani 1988; Brockwell and Davis 1991; Timmer et al. 1998). For the general case of Eq. 5, the phase spectrum $\Phi(\omega)$ is given by a straight line, determined by $d$, plus a second term, which is the argument of the Fourier transform of $b(t')$. Therefore, if $b(t')$ is not known, as in our case, an interpretation of the estimated realignment delay $d$ as the delay between the input and output would lead to a spurious delay, which is not physiologically meaningful. Finally, because the errors of the phase spectrum are directly related to the coherency (i.e., the higher the coherency, the lower the errors in the estimated phase), the estimated $d$ does not provide a meaningful value in the case of a nonsignificant coherency (Priestley 1989; Brockwell and Davis 1991).

To test each output trial for linear independence from the input stimulus, we used the critical value $s$ for the null hypothesis of zero coherency for a given significance level $\alpha$:

$$s = \sqrt{1-\alpha^{\frac{2}{v-2}}} \quad (6)$$

where $v$ is the so-called equivalent number of degrees of freedom, which depends on the direct spectral estimator (i.e., on $W_j$) and the tapering used (Bloomfield 1976; Brockwell and Davis 1991; Timmer et al. 1998). The advantages of coherency-based tests for linear independence over the more commonly used cross-correlation technique are described in detail by Timmer et al. (1998).

It is important to note that with coherency-based tests it is often not sufficient to consider simply the value $s$. The reason is that the derivation of the underlying statistics that lead to a test based on Eq. 6 assumes that the cross spectrum is approximately constant over the width of the window function $W_j$ used in the direct spectral estimation. Asymptotically, this assumption is always true given the required properties of a valid smoothing window function $W_j$ (Brockwell and Davis 1991). If, however, a cross spectrum of a finite series exhibits a high curvature, then the confidence interval is no longer valid. To overcome this problem, investigators (Priestley 1989; Brockwell and Davis 1991) commonly use a technique known as *prewhitening*, in which one (or two) of the series is linearly filtered so that the cross spectrum of the resulting, filtered series is flat. This can be done because a linear filter applied to one or both of the signals does not modify the coherency (Brockwell and Davis 1991). In this study, we prewhitened the stochastic vestibular stimulation signal before we calculated the coherency. Since we know the parameter $a$ in Eq. 1, we are able to prewhiten $x(t)$ simply by inverting the filter of Eq. 1.

In addition to the above tests, we also determined a mean coherency between the respective vestibular stimulation signals and the significantly dependent COP time series for each subject. The mean was taken for all values within the broadest contiguous frequency band of significant coherency. If the contiguous frequency band showing significant coherency was smaller than 0.5 Hz (which was the width of the spectral estimator $W_j$), then the bandwidth of the stochastic stimulation signal (i.e., 0–1 Hz, 1–2 Hz, or 0–2 Hz) was taken by default.

## Results

The 0- to 2-Hz stochastic vestibular stimulus and the resulting mediolateral COP time series for a single 60-s trial from one subject are shown in Fig. 3a. This figure demonstrates the difficulty in determining by visual inspection whether there is a relationship between the two time series. The coherency plot for the two series in Fig. 3a is shown in Fig. 3b. The dashed line indicates the level of significance, $s$, for $\alpha$=0.95 (see Eq. 6). It can be seen that there is significant coherency between the vestibular stimulus and the mediolateral COP time series at frequencies less than 2.0 Hz, i.e., at frequencies less than the upper limit of the filtered input stimulus.

Figure 4 provides the coherency results for the three different stochastic vestibular stimulation signals, i.e., signals that were bandlimited between 0–1 Hz (Fig. 4a), 1–2 Hz (Fig. 4b), and 0–2 Hz (Fig. 4c), for the subject in Fig. 3. It can be seen that, for each single trial, there is significant coherency between the vestibular stimulus and the mediolateral COP time series at frequencies less than the upper limit of the filtered input stimulus. In addition, it can be seen that the coherency results for each stimulation signal were reproducible from trial to trial (Fig. 4), i.e., the coherency plots for the five trials for a given stimulus have similar shapes. As expected, the position of the maximum coherency varied with the frequency band of the different stimulation signals and was observed within

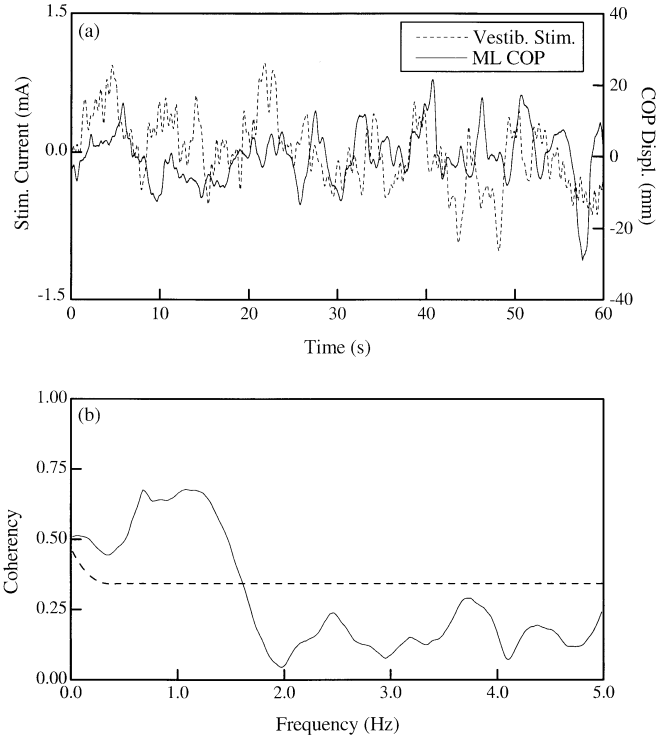


**Fig. 3** **a** The 0- to 2-Hz stochastic vestibular stimulation signal and the resulting mediolateral (ML) COP time series for a single 60-s trial from the subject of Fig. 2. **b** The coherency between the vestibular stimulation signal and the COP time series in **a**. The *dashed line* indicates the level of significance, $s$, for $\alpha$=0.95 (see Eq. 6)

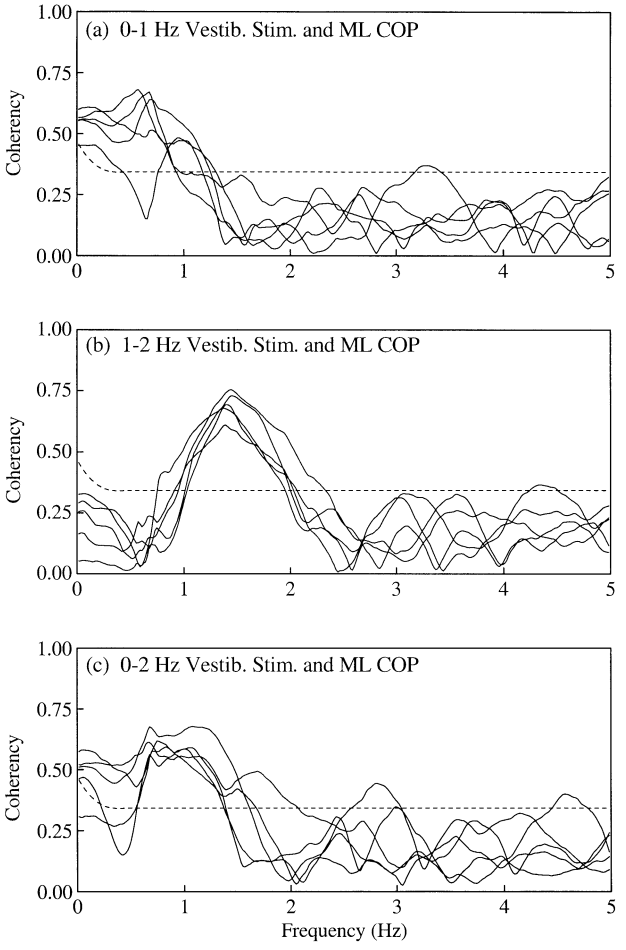


**Fig. 4a, b** Plots of the coherency between the stochastic vestibular stimulation signal and the resulting mediolaterial COP time series for each trial from the subject of Figs. 2 and 3. Shown are the results for the **a** 0- to 1-Hz, **b** 1- to 2-Hz, and **c** 0- to 2-Hz vestibular stimulation signals. Five trials were conducted for each stimulation signal. The *dashed line* indicates the level of significance, *s*, for $\alpha$=0.95 (see Eq. 6)

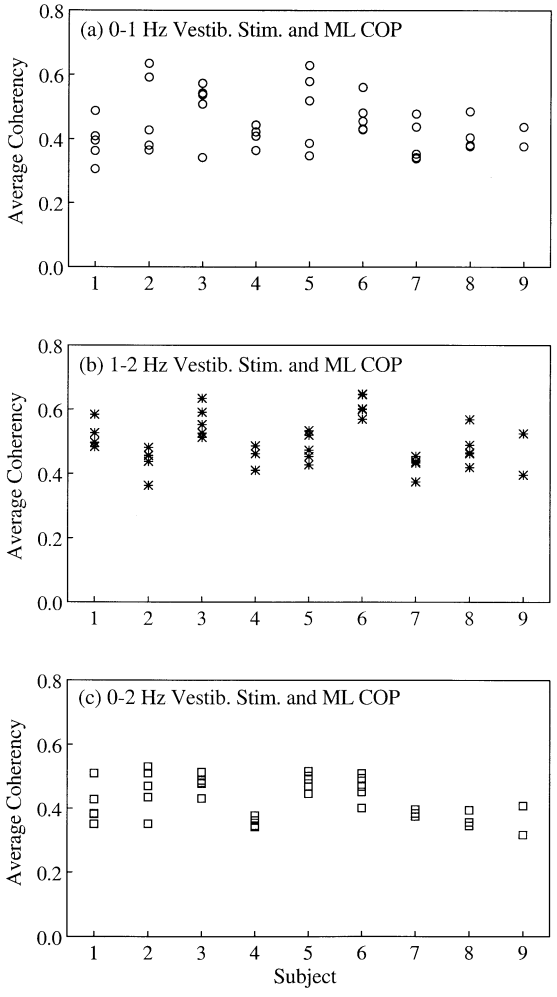


**Fig. 5a–c** The mean coherency values (see text for an explanation) between the respective vestibular stimulation signals and the resulting mediolateral COP time series for the significant coherent trials from each of the nine subjects. Shown are the results for the **a** 0- to 1-Hz, **b** 1- to 2-Hz, and **c** 0- to 2-Hz vestibular stimulation signals

the respective frequency band. These general results were found in eight of the nine subjects tested. In particular, significant coherency between the stochastic vestibular stimulation signal and the resulting mediolateral COP time series was found in 12–15 trials (out of a possible 15) for each of these subjects. The ninth subject only exhibited significant coherency in six trials (two trials for each stimulus). Note that a coherency value of 0.6 is significant for $\alpha$ as high as 0.99999.

Figure 5 shows the mean coherency values for the different trials from each of the nine subjects. The number of points plotted for each subject corresponds to the number of significant coherent trials for that subject. Note that the values plotted in Fig. 5 are slightly lower than the peak values (see Fig. 4), since they correspond to a mean over a frequency band. It should also be noted that for each subject the mean coherency for a given stimulation signal was consistent from trial to trial. Moreover, in general, the highest degree of coherency was found for the 1- to 2-Hz stochastic vestibular stimulation signal. For the control trials, we calculated the coherency between the COP time series and a randomly selected stochastic vestibular stimulation signal. Only 2 of the 45 control trials exhibited significant coherency, as expected for a test with a correct size against the 5% confidence level.

The 0- to 2-Hz stochastic vestibular stimulus and the resulting anteroposterior COP time series for a single 60-s trial from one subject are shown in Fig. 6a. The corresponding coherency plot for that trial is shown in Fig. 6b. It can be seen that there is no significant coherency between the vestibular stimulus and the anteroposterior COP time series. Similar results were obtained for all subjects. Specifically, we found significant (albeit low – the highest value was 0.48 – and inconsistent) coherency between the vestibular stimulus and the anteroposterior COP time series in only 28 of the 135 trials across the subject population.

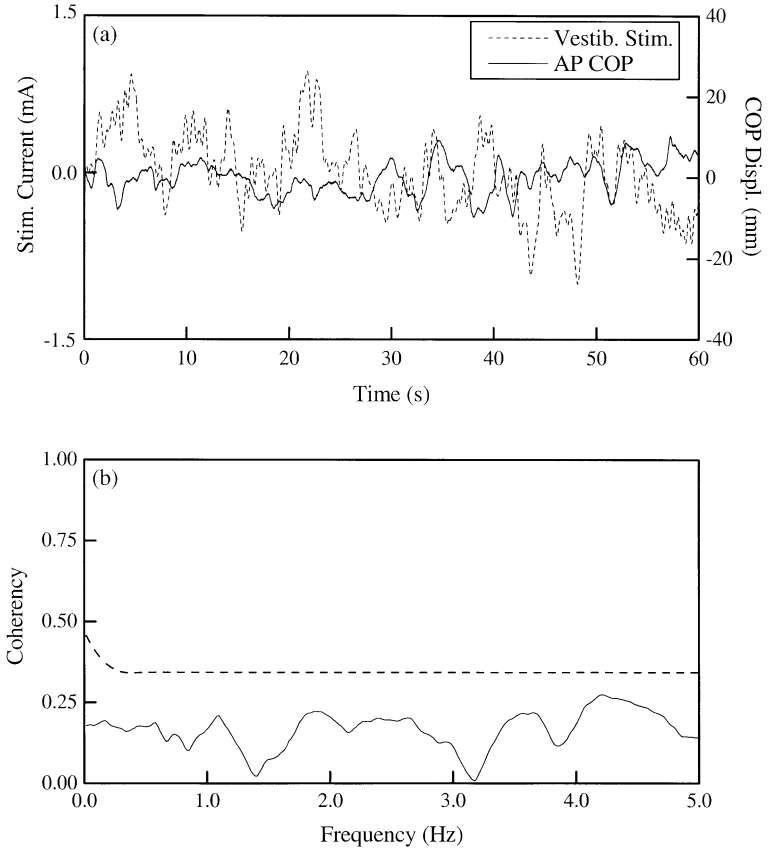


**Fig. 6** **a** The 0- to 2-Hz stochastic vestibular stimulation signal and the resulting anteroposterior (*AP*) COP time series for a single 60-s trial from the subject of Figs. 2 and 3. **b** The coherency between the vestibular stimulation signal and the COP time series in **a**. The *dashed line* indicates the level of significance, *s*, for $\alpha$=0.95 (see Eq. 6)

**Table 1** Realignment delay values (in seconds) obtained for all subjects and stimuli. Each value represents a mean over the significant coherent trials for a given subject and stimulus. The corresponding SEM is given in parentheses. Note that the SEM for the ninth subject is not available because that subject exhibited significant coherency in only two trials for each stimulus. The last two rows contain the group means and their SEM. A one-way, multiple-range ANOVA showed that the realignment delay values obtained for the three different stimuli differ significantly from each other

| Subject | 0–1 Hz stim | 0–2 Hz stim. | 1–2 Hz stim. |
|---|---|---|---|
| 1 | 1.14 (0.18) | 0.48 (0.03) | 0.20 (0.02) |
| 2 | 1.16 (0.24) | 0.77 (0.09) | 0.35 (0.09) |
| 3 | 0.64 (0.13) | 0.64 (0.06) | 0.23 (0.04) |
| 4 | 1.28 (0.06) | 0.27 (0.05) | 0.34 (0.01) |
| 5 | 0.95 (0.07) | 0.71 (0.16) | 0.24 (0.09) |
| 6 | 1.08 (0.09) | 0.61 (0.03) | 0.23 (0.04) |
| 7 | 1.31 (0.28) | 0.47 (0.08) | 0.30 (0.06) |
| 8 | 1.39 (0.05) | 0.52 (0.11) | 0.21 (0.01) |
| 9 | 0.80 (n.a.) | 0.59 (n.a.) | 0.30 (n.a.) |
| Mean | 1.09 | 0.57 | 0.27 |
| SEM | 0.08 | 0.05 | 0.02 |

Table 1 summarizes the values of the realignment delay *d* obtained from the significant coherent trials for each subject. Since the errors of the phase spectra are directly related to the amount of coherency, the most reliable results for *d* are those from subjects who exhibited high coherency (see Fig. 5). The realignment delay values were consistently highest for the 0- to 1-Hz stimulus and lowest for the 1- to 2-Hz stimulus. A one-way, multiple range analysis of variance (ANOVA) showed significant differences ($P$<0.01) between the realignment delay values for all stimuli, according to Sheffé's test (Seber 1977).

## Discussion

In this study, we demonstrated that, in subjects who are facing forward, bipolar binaural stochastic galvanic stimulation of the vestibular system leads to coherent stochastic mediolateral postural sway. Specifically, we found significant coherency between the stochastic vestibular stimulation signal and the resulting mediolateral COP time series in the majority of trials in eight of the nine subjects tested. The coherency values we obtained were up to 0.8 for several trials. Besides the case in which the COP time series and the vestibular stimulus are indeed uncorrelated (which was not the case in our study, as we could show), there could be many reasons as to why we obtained coherency values less than one (Timmer et al. 1998). These include: (1) a nonlinear relation between the vestibular stimulus and the COP time series, (2) other influences on the COP time series that are not related to the stimulus, (3) estimation bias due to misalignment (Hannan and Thomson 1971, 1973; Carter 1987; Priestley 1989), as mentioned above, and (4) the observational noise that is present in COP time series. We excluded item 3 by using

the described realignment procedure (see the Materials and methods section). Item 4 played a minor role because the signal-to-noise ratio for our COP data is generally better than 100:1. With respect to item 1, we ensured that each subject's response to the galvanic vestibular stimulation remained relatively weak by using moderate, subject-specific levels of stimulation. Therefore, it is unlikely that nonlinearities played a major role in our study. This leads us to the conclusion that item 2 (i.e., other influences on the COP time series) is the main reason why we obtained coherency values that were less than 1.

We found that the coherency was highest for the 1- to 2-Hz stochastic vestibular stimulation signal. This result may be due to the intrinsic dynamics of the quasistatic postural control system, in the following manner. It is well known that the COP under the feet of a quietly standing individual continually moves about in an erratic fashion (Collins and De Luca 1993, 1994). If the effects of the vestibular stimulus are simply superimposed upon the quiet-standing COP displacements, then the quiet-standing displacements would be the "other influences" mentioned in item 2 above. Since the power spectra of such COP time series decrease with increasing frequencies, it follows that the "influence" of quiet-standing COP displacements on the effects of a stochastic vestibular stimulus would diminish as the frequency range of the stimulus was increased, assuming that the system's response to the vestibular stimulus remains relatively constant over the investigated frequency range. Within this scenario, the 1- to 2-Hz vestibular stimulation signal would have the largest effect on the displacements of the COP and thus result in the highest coherency.

We obtained coherency values larger than those reported by Fitzpatrick et al. (1996), especially in the low-frequency range. This could be due to an underestimation of coherency with respect to time-series misalignment, an effect that was not controlled for by Fitzpatrick et al. (1996). The discrepancy could also be due to differences in the level of stimulation used in the studies. Fitzpatrick et al. (1996) used the same stimulation level (current value) for each subject. In our study, we assumed that each subject may have a different response threshold, so we determined a stimulation level on an individual basis. By doing so, we avoided using relatively high levels of vestibular stimulation that could cause large postural sway and thereby invoke other undesired postural feedback control mechanisms. Such mechanisms, which would also be "other influences" in the sense of item 2 above, often involve significant nonlinearities and could lead to lower values of coherency between the stimulus and response. However, the most substantial difference between our study and that of Fitzpatrick et al. (1996) is that they did not discuss the significance of their obtained coherency values with respect to the underlying statistics. We, on the other hand, performed the test against the null hypothesis of zero coherency, which leads to a reliable statement concerning the linear independence of the input and output for each trial. Such a test is particularly important for trials exhibiting low but significant coherency.

The realignment delays $d$ could be meaningful from a physiological standpoint if the estimated delays could be interpreted as a time lag between the input and the measured output signal. However, as pointed out in the Materials and methods section, this is rarely the case for complex systems, such as the human postural control system. For this study, a straightforward interpretation of the realignment delay as a constant delay between the input and output is invalid because the values estimated for $d$ differed significantly between the different frequency bands (Table 1). Our finding that the realignment delays were always significantly lower for the 1- to 2-Hz vestibular stimulus indicates that the phase spectra, respectively, the impulse response functions of the system, exhibit a nontrivial curvatured shape.

We also found that in subjects who are facing forward, bipolar binaural stochastic galvanic stimulation of the vestibular system does not lead, in general, to coherent stochastic anteroposterior postural sway. This result is consistent with the findings of Hlavacka and Njiokiktjien (1985), who showed that, with bipolar binaural constant galvanic vestibular stimulation, the direction of the evoked sway is approximately in the direction of the intermastoid line. Thus, it is possible that coherent stochastic anteroposterior sway could be produced with bipolar binaural stochastic galvanic vestibular stimulation if the subject's head is turned to the left or right (i.e., over the left or right shoulder). Interestingly, Magnusson et al. (1990a) have shown that if a subject's head is facing forward, monopolar binaural constant galvanic stimulation of the vestibular system can be used to induce anteroposterior sway in the subject. Thus, it is also possible that coherent stochastic anteroposterior postural sway could be produced with monopolar binaural stochastic galvanic vestibular stimulation. These issues will be addressed in a future study. The few cases in the present study in which we found significant (albeit low and inconsistent) coherency between the stochastic vestibular stimulation signal and the resulting anteroposterior COP time series can be explained by the dependency of the two body directions. Since there is a considerable amount of coherency between the mediolateral and anteroposterior COP time series (due to the biomechanics of the human body), it follows that, even if the anteroposterior COP time series is indeed independent from the vestibular stimulus, there will be a weak relation between the two signals if the coherency between the ML COP time series and the stimulus is high.

Previous studies (Inglis and Macpherson 1995; Inglis et al. 1995; Horak et al. 1994) have suggested that the role of the vestibular system is to modulate the amplitude of the body's postural response. Our results support this notion. In particular, we showed that time-varying galvanic vestibular stimulation can continuously modulate mediolateral postural sway. In addition, by utilizing stochastic stimulation signals, we ensured that the subjects could not predict a change in the vestibular stimulus. Thus, our findings indicate that subjects can act as "responders" to galvanic vestibular stimulation.

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