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Identification of Adaptation in Human Postural Control using GARCH Models

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Abstract

Human postural dynamics was investigated in twelve normal subjects by means of a force platform recording body sway induced by bipolar transmastoid galvanic stimulation of the vestibular nerve and labyrinth. We modeled the stabilizing forces actuated by the feet as resulting from complex muscular activity subject to feedback of body velocity and position. Time series analysis demonstrated that a transfer function from stimulus to sway-force response with specific parameters could be identified. In addition, adaptation to the vestibular stimulus was demonstrated to exist. Residual GARCH modeling (generalized autoregressive conditional heteroskedasticity) suggested a postural adaptation time constant in the range of 40–50 s. The results suggest means to evaluate adaptive behavior in postural control and in other physiological contexts.

Introduction

The function of human postural control is a most complex function which includes gravity compensation, controlled coordinated motor responses, detection of movement, stability of motion. Feedback information originates in the afferent sensory input from the visual, vestibular and somatosensory receptors reporting changes in position and velocity of body posture [11, 21]. The afferent sensory information evokes and modifies the motor output at all levels, from the spinal medulla to the cerebral cortex [30]. In understanding the task of posture control feedback, control theory has long been an important source of inspiration to physiologists for physiological modeling and understanding of complex mechanical and neurological interactions between muscle forces, loads, and posture [15]. During the last two decades, a wealth of literature on human postural control has been devoted to the responses to induced perturbations by means of force plate measurements, EMG recordings and movement analyzing systems. Postural control has been challenged by movements of the support surface and/or visual surrounds [27, 14], and vibration toward muscles [18]. Although lateral plane postural control is necessary to maintain upright stance, findings on several previous studies have demonstrated a preference for evaluation of postural control to movements in the anterior-posterior plane [23]. Several attempts at applying quantitative methods have been reported [8, 16, 18, 10],

The importance of vestibular input has generally been evaluated by indirect approaches, such as investigating subjects with congenital or acquired bilateral loss and applying complex stimulus conditions [4]. To assess dynamic feedback control, we would need a well-defined specific stimulus with a primary effect on sensory inputs only. Galvanic stimulus has a long history as a means to affect the vestibular system and to induce vertigo in man [1, 24, 26, 25, 6, 20, 28, 29, 9]. Transmastoid galvanic stimulus of the vestibular nerve and labyrinth induces body movements in the lateral plane [3]. Experience from early experiments using constant-voltage stimulus have shown to have somewhat unpredictable results mainly due to series-impedance and capacitative coupling effects whereas a constant-current regime of stimulus provide reproducible results [19]. In our study, we chose a constant-current regime of stimulus which permits a well defined power of the stimulus. The aim of the present investigation was to find out whether, using a galvanic vestibular stimulus, postural control in the lateral plane can be quantified by means of time-series analysis of stimulus-response data. As standard time-series analysis is poorly suited to analysis of systems with feedback control and adaptation [17], it is also necessary to develop suitable methodology. Whereas parts of this work was previously presented in [19, 10], we here elaborate on adaptation modeling using GARCH models [5].

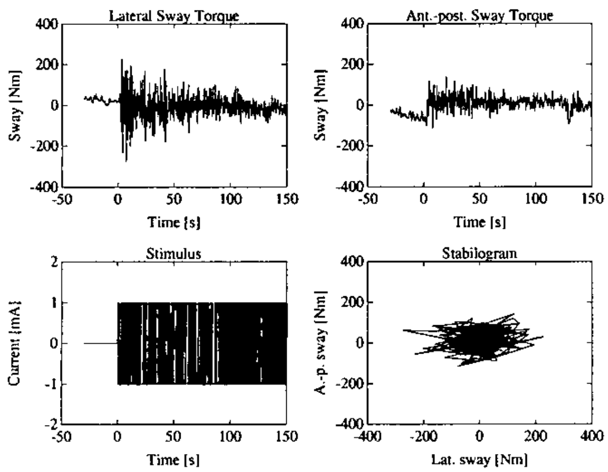

Figure 1: Sway response to 1 [mA] galvanic stimulation with closed eyes: Galvanic stimulation *(lower left)* starting at time $t = 0$; lateral sway response *(upper left)* and anterior-posterior sway response *(upper right)*. Stabilogram *(lower right)* of sway torques [Nm]. Time scale in [s].

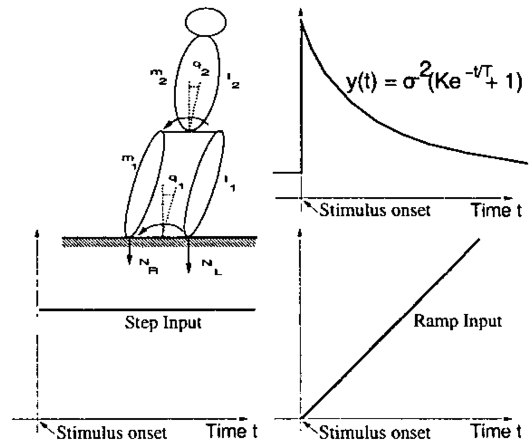

Figure 2: Biomechanical model with normal forces N_L, N_R of the feet shown; Step and ramp input models modeling onset of stimulus; Exponential $y(t) = \sigma^2(Ke^{-t/T} + 1)$ for $t > 0$ fitted to the residual variance of the estimated model; Responses to the step and ramp inputs and fitting the exponential to the residual sequence serve to characterize adaptive behavior.

Materials and Methods

Tests were done on twelve naive human subjects, age 24-44 years (mean 30.9 years, standard deviation 6.6 years) none of whom had any history of vertigo, central nervous disorder, ear disease, or previous injury to the lower extremities. No subject was on any form of medication or had consumed alcoholic beverages for at least 48 hours. The equipment consisted of a square force platform coupled to a computer for data acquisition and computation. The platform was equipped with strain gauges to measure forces. The equipment allowed simultaneous recording of body 'sway'—*i.e.*, forces actuated by the feet on the support surface both in the sagittal and frontal planes [18]. The subject stood erect but not at attention with heels together on the platform and arms across the chest while staring at a spot on the opposite wall. Carbon electrodes (Cefar AB, Lund, Sweden) of dimensions 3.5×2.5 [cm] were attached symmetrically on the mastoid process behind each ear and electronic stimulation was produced by a constant current generator at 1 [mA] with opposite polarity of the two electrodes, *i.e.*, bipolar stimulation. Then, while recording continued, the polarity of the galvanic stimulation was changed pseudorandomly (PRBS) using a real-time software program that carefully synchronized stimulus and measurement without aliasing. The test sequence took 183.6 [s] with stimulus actuation at a rate of 20 [Hz] which was chosen to permit sufficiently rapid and unpredictable variations in the galvanic stimulus with its spectrum designed to cover up to 5-10 [Hz]. The experiment started with spontaneous sway recorded for 30 [s] (Fig. 1).

Basic data analysis was made by means of system identification methodology. Autospectra, cross spectra, and coherence spectra of input (*i.e.*, galvanic stimulus) and outputs (*i.e.*, sway force responses) were made to verify that the signal levels were adequate and that the stimulus spectrum covered the relevant spectral ranges of biological interest in vestibular research—*i.e.*, below 0.1 [Hz] and up to 10 [Hz] (Figs. 1, 3). We used a multi-link inverted pendulum model with coordinates $q = (q_1 \quad q_2 \quad \ldots)$ of the links and an associated control model [19]. A linearized transfer function from the stimulus u to the torque responses τ_{bal} for a two-link stabilized inverted pendulum is found as

$$\begin{aligned} \tau_{bal}(s) &\approx (M(0)s^2 - \frac{\partial G(q)}{\partial q}|_{q=0}) \times \\ &\times \left(s^2 I_{2\times 2} + sK_d + K_p\right)^{-1}(b_1 + b_2)U(s) \end{aligned} \tag{1}$$

where $G(q)$ designates gravitation forces; $M(q)$ inertia matrix. The gain coefficients b_1, b_2 relate to the misperception caused by the stimulus and the stabilizing feedback control is characterized by K_d (damping matrix) and K_p (stiffness matrix).

Matlab^{TM} was used for system identification. As data exhibited clear nonstationary properties with trends and time-varying behavior (Fig. 1), it was not possible to use standard time-series analysis based on stationary stochastic models. To solve the identification problem, we designed a pseudolinear regression that successfully fitted data to the model

$$A(z^{-1})y_k = B(z^{-1})\begin{pmatrix}\text{stimulus}\\ \text{step input}\\ \text{ramp input}\end{pmatrix} + C(z^{-1})w_k \quad (2)$$

which relates the recorded response (or output) $\{y_k\}$ (=body sway force) to the galvanic stimulus, and a postulated sequence of uncorrelated disturbance variables $\{w_k\}_{k=1}^{N}$ that models random disturbances. Outputs corresponding to the additional step-formed and ramp-formed inputs served to model the time-varying shift of the center of pressure of the subject standing on the force platform after onset of the stimulus, whereas the offset accounted for the subject's choice of resting position and for measurement offsets. Determination of a suitable model order was supported by model validation criteria such as statistics of the loss function, the Akaike information criterion (AIC), the final prediction criterion (FPE), and residual analysis (autocorrelation and cross correlation tests between stimulus and residuals) [17].

Assessment of adaptation: Volatility and state-dependent variance behavior can be modeled using models of *generalized autoregressive conditional heteroskedasticity* (GARCH) [5, 12]. A suitable GARCH model of time-varying variance of uncorrelated residuals would be

$$\begin{aligned}\varepsilon_k &= \sqrt{s_k}w_k && (3)\\ s_k &= \sigma^2 + \alpha_1 s_{k-1} + \cdots + \alpha_m s_{k-m} && (4)\\ &+ \beta_1 v_{k-1} + \cdots + \beta_m v_{k-m}\end{aligned}$$

for some model order m and some 'input' $\{v_k\}$. Such an input may be the onset of stimulus in the context of our adaptation experiments. Adaptation-related properties were quantified by extracting the following GARCH model of model order 1 from sequences of squared residuals(Figs. 1, 5)

$$\mathcal{E}\{\varepsilon_k^2\} = \begin{cases}\sigma^2, t < 0\\ \sigma^2(Ke^{-t/T}+1), t > 0\end{cases}, \begin{cases}t = kh - 30\\ h = 0.05\end{cases} \quad (5)$$

and where K denotes the increase in the residual variance in response to the onset of stimulus, and T denotes the time constant of the attenuation of residual power from the peak value at the start of the stimulus. The time constants of the residual attenuation were calculated and shown in Table 1.

Results

When exposed to the galvanic stimuli, all subjects demonstrated corresponding sway in the lateral plane. At the same time there was a change in anterior-posterior movement (Fig. 1)—usually with a shift in the center of pressure that was effectively modeled by means of the step and ramp responses. No subject reported any discomfort, although a tickling sensation in the skin in contact with the electrodes was sometimes experienced at shifts of polarity during the early part of the stimulus sequence. The subjects could not detect the polarity of the stimuli and were not informed about any expected results of the stimulus. Coherence between stimulus and response was tested. [17]. Response of anterior-posterior sway variance was also shown to be of lower magnitude than the lateral sway variance for all subjects investigated. Both for lateral and anterior-posterior sway, a shift was demonstrated in the stimulated sway spectra as compared to the spontaneous sway (Fig. 3). A time-invariant fourth-order linear model proved to be suitable and the relevance of the fourth order model was supported by evaluation of the loss function, the Akaike information criterion (AIC) and the final prediction criterion (FPE). Residual analysis supported the sufficiency of the fourth order model with 95 % confidence ($p < 0.05$). A third order model was refuted by the same criteria, and a fifth order model exhibited no further improvement as compared to the fourth order model. Cross validation with the first part of the time series—*i.e.*, the spontaneous sway, fulfilled all relevant validation criteria [17].

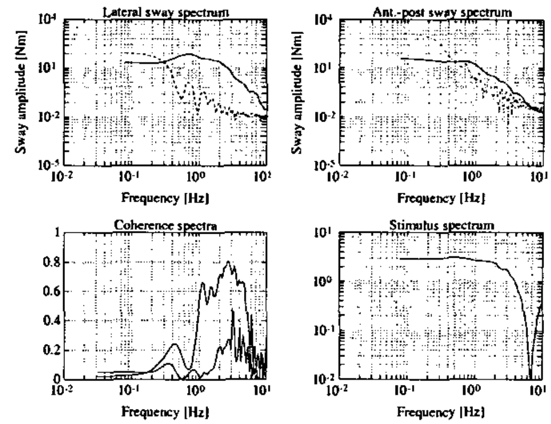

Figure 3: Spectral analysis of stimulus and response from galvanic stimulation experiment. Solid lines depict the result during stimulation, and dashed lines the spontaneous sway. Lateral sway spectrum *(upper left)*. Anterior posterior sway *(upper right)*. Coherence spectra between stimulus and responses *(lower left)* with higher coherence for lateral sway. Stimulus spectrum *(lower right)*. Frequency in [Hz].

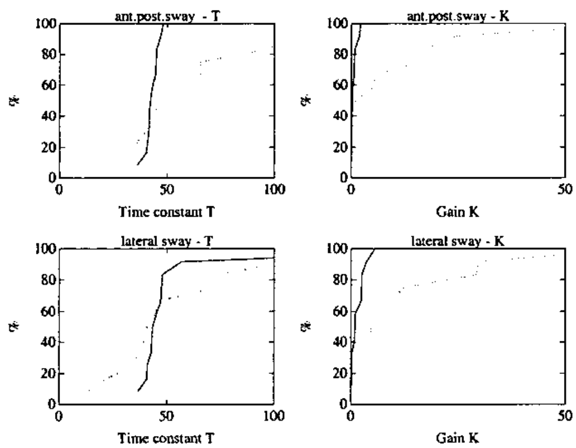

Figure 4: Empirical distribution of the residual variance adaptation time constant T (*left*) and the initial gain (*right*) for anterior-posterior sway (*upper*) and lateral sway (*lower*). Notice that the adaptation time constant for open eyes (*solid line*) appears to have a small variation range (40-50 [s]) for both anterior-posterior sway and lateral sway, as compared to the time constant for closed eyes (*dotted line*).

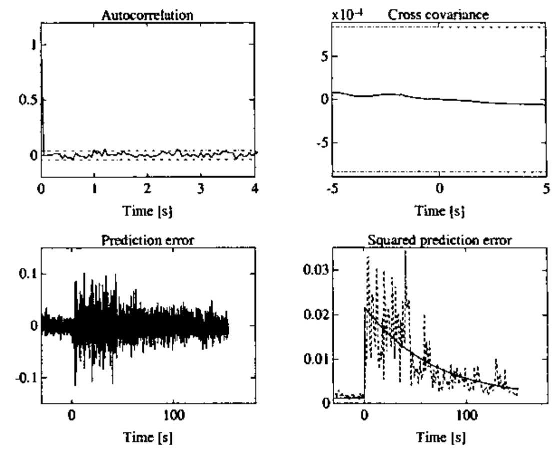

Figure 5: Residual analysis *vs.* time [s] of the full time series with residual autocorrelation and cross covariance between residuals and stimulus (the dashed lines in the upper graphs indicate the confidence intervals ($p < 0.05$) for normally distributed residuals). The lower graphs show the prediction error (*left*) and the squared prediction error (*right graph, dashed line*) with a fitted exponential (*right graph, solid line*). The spontaneous body sway was recorded for 30 [s] prior to the onset at time $t = 0$ of the stimulation.

Assessment of adaptation: The squared value of residuals for one typical subject is shown in Fig. 5. Although the residual sequence is nonstationary with time-varying variance, it exhibits white-noise properties with mutually uncorrelated residuals and residuals uncorrelated with any of the inputs. As a least-squares type method is being used, there can be no artifacts arising from assumptions on certain stochastic models. Adaptation-related properties were quantified by extracting the gain K and the time constant T from the graphs of residual power (Fig. 5). The time of the peak value of residuals and the time constant of the residual attenuation were calculated (Table 1). Empirical distributions of K and T from GARCH model of order one for both anterior-posterior and lateral sway responses are shown in Fig. 4.

Discussion

This study demonstrated that the adaptation of postural control induced similar adjustment patterns during galvanic stimulation irrespective of whether the motion responses were induced in lateral or anteroposterior direction.

The adaptive responses induced by the stronger, repetitive galvanic or vibratory stimulation contained at least two separate processes. One process can be seen in the progressive reduction of body sway during the stimulation. An additional slower simultaneous adaptive process can be seen in the postural displacement. The dynamical properties were found to be different between galvanic and vibratory stimulation, reflected by altered response latencies and altered dynamical properties. Some of these results might be explained by the difference in lateral and anteroposterior biomechanical constraints. However, with one exception only among the test subjects, a low-order GARCH model was sufficient for accurate modeling of volatility behavior in response to the perturbation induced by galvanic or vibratory stimulation.

Effects of galvanic stimulus: A primary effect of the bipolar galvanic stimulus is that the lateral sway dominates over the anterior-posterior sway [19]. It is also apparent from data that the impact on the sway starts with a certain latency, as previously reported by Sekitani [13, 26]. A bipolar bi-aural galvanic stimulus induces vestibular and postural responses [19]. A galvanic stimulus causes an increase of the firing frequency mainly in the irregularly firing neurons of the vestibular nerve on the side of the cathode and a decreased firing frequency on the side of the anode [7, 22]. Galvanic induced vestibu-

Subject	$T_{a.p.}$ [s]	T_{lat} [s]
1:	46.9	40.8
2:	44.4	56.8
3:	42.9	47.2
4:	48.1	47.4
5:	41.7	42.7
6:	41.0	40.8
7:	40.3	43.3
8:	41.7	45.0
9:	45.0	208.3
10:	42.1	43.0
11:	36.2	47.9
12:	44.6	36.6
m	42.9	58.3
s	3.18	47.5
$s.e.m.$	0.92	13.7

Table 1: Experiment results of adaptation to galvanic stimulation with closed eyes. Results formulated in terms of a time constant of residual attenuation T for anterior-posterior sway ($T_{a.p.}$) and lateral sway (T_{lat}). Estimated mean values m, standard deviations s, and standard errors of mean are also shown. If the outlier of subject 9, who exhibits weak or no adaptation, is eliminated, then we find mean $m = 44.7$, $s = 5.24$, and $s.e.m. = 1.58$ for the time constant T_{lat}.

lar responses in humans reflect the integrity of the vestibular nerve but not of the labyrinth [24, 6]. It has been demonstrated that a vestibular nystagmus can be elicited with the fast phase directed toward the cathode, although this requires a current of several mA which generally will cause some pain to the subject [24]. Lateral postural sway is induced by currents of less than 0.4 mA which does not evoke sensations from the skin over the mastoid [29]. A bipolar bi-aural galvanic stimulus of 1 mA causes an asymmetric activation of the soleus muscles increasing EMG activity on the side of the cathode and decreasing it on the side of the anode with a latency of about 100 ms [28]. This asymmetry is modified, however, if the head is turned so that the increased EMG activity is dependent on the direction of the stimulus current relative to the ambient space. Increase in EMG activity appeared in the *m. triceps brachii* on the side of the cathode already at 40 ms after onset of stimulation [2]. Thus, a galvanic stimulus to the vestibular nerve as used in the present experiments can be expected to induce postural movements from the neck down and in the direction of the anode with short latencies.

Model complexity: From a biomechanical point of view it was shown that a fourth-order model is necessary as each link of a two-segment inverted pendulum with velocity and position corresponds to a second order system. Thus, four states require modelling, *i.e.*, $q_1, \dot{q}_1, q_2$, and $\dot{q}_2$, with all body segments strongly coupled in their motion and the resulting behavior is in many aspects like that of a simple inverted pendulum. Moreover, model validation verified that a fourth-order model is sufficient with 95% confidence according to residual analysis. Our results therefore verify that an inverted double pendulum is of relevant biomechanical complexity.

Problems of adaptation and control: The stimulus has time-invariant statistical characteristics after onset of stimulus. However, from a comparison of Fig. 1, it can be seen that the sway response to the galvanic stimulus is time-variant in its statistical properties. Moreover, the time-variant response is not reproducible by means of repeated stimulus on the same subject [35] and, thus, the stimulus causes apparent irreversible changes in the sway response. Opposite to Courjoun *et al* [7], we have not used repetitive stimulus as we want to avoid anticipative (*i.e.*, feedforward) responses from contaminating our result. Instead we have used a pseudorandom stimulus which is unpredictable for the test subjects.

Physiological significance: The vestibular input induces an erroneous signal resulting in perturbations which have to be corrected for in the time series model. The impact of the stimulus loses in strength with an adaptation time constant in the range of 40-50 [s] (Table 1), whereas vestibular habituation is generally characterized by a longer time course when studied in the vestibulo-ocular reflex [30]. Hence, the reduction of body sway appears to be the effect of a physiologic process which involves a decrease of gain at some level of the control system. One possible hypothesis is that if the neural feedback mechanism interprets a reaction on the vestibular stimulus as leading to an increase of error signals from other sources of input, the weight of the vestibular input is reduced. This would result in suppression of the corrective movements induced by the vestibular stimulus—*i.e.*, adaptation. Results in Table 1 exhibit an adaptation time constant with a variation range 36.2–48.1 s for anterior-posterior sway and 36.6-56.8 s for lateral sway and only one subject (# 9) exhibited insignificant or no adaptation in lateral sway in the course of the experiment. Although the time constant of the adaptive response, which is in the range 40-50 s, is similar in magnitude to time constants of eye movement responses following a velocity-step in rotatory stimulation, there is still insufficient support to claim any such relationship.

Conclusions

We have designed a methodology for quantitative investigation of feedback properties for characterization of human lateral posture stability and adaptation to vestibular stimulus. We have verified the relevance of a fourth-order model by means of biomechanical analysis of the test conditions. System identification permitted validation of a fourth-order model by 95% confidence for each subject. The non-correlated but time-variant residual sequence permitted extraction of a time constant of 40-50 s to model the decrease of of the residual variance in the course of the experiment. The heterskedasticity was effectively modeled with a low-order GARCH model.

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