Galvanic Vestibular Stimulation for Analysis of Postural Adaptation and Stability

Rolf Johansson, Måns Magnusson, and Per A. Fransson

***Abstract*— Human postural dynamics was investigated in 12 normal subjects by means of a force platform recording body sway, induced by bipolar transmastoid galvanic stimulation of the vestibular nerve and labyrinth. The model adopted was that of an inverted segmented pendulum, the dynamics of postural control being assumed to be reflected in the stabilizing forces actuated by the feet as a result of complex muscular activity subject to state feedback of body sway and position. Time-series analysis demonstrates that a transfer function from stimulus to sway-force response with specific parameters can be identified. In addition, adaptation to the vestibular stimulus is demonstrated to exist, and we describe this phenomenon using quantification in terms of a postural adaptation time constant in the range of 40–50 s. The results suggest means to evaluate adaptive behavior and postural control in the erect human being which may be useful in the rehabilitation of individuals striving to regain upright stance.**

I. Introduction

THE ABILITY to maintain stability and withstand the effects of gravity on stance and motion is important to a large variety of human activities. The function of human postural control is a most complex function which includes detection of movement as well as evoking and controlling coordinated motor responses. Since the human body is not statically stable, maintaining upright posture requires continuous antigravity action by means of coordinated adjustments of the tone of the antigravity muscles [16], and human postural control can, at least partly, be viewed as a dynamic feedback control system. The feedback information originates in the afferent sensory input from the visual, vestibular, and somatosensory receptors reporting changes in position and velocity of body posture [32], [20], [33]. The afferent sensory information evokes and modifies the motor output at all levels, from the spinal medulla to the cerebral cortex, cf. [51], [21].

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 [25], [26]. 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, such as the Selspot$^{\mathrm{TM}}$ or Elite$^{\mathrm{TM}}$ systems. Postural control has been challenged by movements of the support surface and/or visual surrounds [47], [48], and vibration toward muscles [31], [44].

Although several attempts at applying quantitative methods have been reported [10], [23], [27], [28], [31], [32], the literature has largely described the feedback properties of posture control in qualitative terms—for instance, with force, velocity, and position feedback or—rather vaguely—as "response patterns" [13]. Moreover, findings on several previous studies have demonstrated a preference for evaluation of postural control to movements in the anterior-posterior plane [40], [41]. However, it is obvious that in the lateral plane also postural control is necessary to maintain upright stance.

The importance of vestibular input has generally been evaluated by indirect approaches, such as investigating subjects with congenital or acquired bilateral loss [35], [45] 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 induce vertigo in man [1], [6], [13], [42], [43]. Transmastoid galvanic stimulus of the vestibular nerve and labyrinth induces body movements in the lateral plane [3], [23], [32]. 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 provides reproducible results [35]. For this reason, we chose a constant-current regime of stimulus in our study. The power of the stimulus in the course of the experiment can thus be well defined.

A reduction in response amplitude in the course of an experiment has been clearly demonstrated in previous studies [35] for a variety of stimulus conditions. The reduction in sway-response amplitude or sway variance, often interpreted as adaptation, is related to the function of postural stabilization but can not simply be interpreted as a transient component of a linear-system response.

The aim of the present investigation was to ascertain whether, using a galvanic vestibular stimulus, postural control in the lateral plane can be described and 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 [30], [34], it is also necessary to develop suitable methodology. Another aim of the present study was therefore to develop methodology suitable for data analysis and to integrate feedback and adaptation models and the measurements obtained with a view to further elucidating

Manuscript received July 30, 1993; revised September 13, 1994.
R. Johansson is with the Department of Oto-Rhino-Laryngology, Lund University Hospital, S-221 85 Lund, Sweden, and the Department of Automatic Control, Lund Institute of Technology, S-221 00 Lund, Sweden.
M. Magnusson and P. A. Fransson are with the Department of Oto-Rhino-Laryngology, Lund University Hospital, S-221 85 Lund, Sweden.
IEEE Log Number 9408355.

0018–9294/95$04.00 © 1995 IEEE

the process of postural control and providing the prerequisite basis for developing clinical tests of lateral posture stability.

II. Materials and Methods

Tests were done on 12 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. At the time of the investigation, no subject was on any form of medication or had consumed alcoholic beverages for at least 48 h.

III. Equipment and Experimental Setup

The equipment consisted of a square force platform coupled to a computer for data recording and computation. The force platform was developed at the Institute of Occupational Health, Helsinki, Finland, and the ENT Clinic at University Hospital, Lund, Sweden [44]. The platform is equipped with strain gauges to measure forces, and measurements obtained from the strain gauges are recorded by the computer. The equipment allows simultaneous recording of body sway, i.e., sway forces both in the sagittal and frontal planes (see [31, Appendix] for details). The subject stood with heels together on the platform 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) to produce a spectrum up to 5–10 Hz, according to a computer-driven program. The stimulus and recording of sway forces were carefully synchronized by means of high-quality real-time software.

As part of routine laboratory practice, it was verified that there was no interference (aliasing) between the sampling frequency and the frequencies of stimulation. The test sequence took 183.6 [s] with sampling at a rate of 20 [Hz], which was chosen to permit sufficiently rapid variations in the galvanic stimulus with its spectrum designed to cover up to 5–10 [Hz]. A control experiment was performed for calibration purposes to measure electronic offsets, noise levels, and disturbances. The absence of cross-talk interference between inputs (stimulus) and outputs (force responses) was verified.

IV. Experimental Procedure

During the test, the subject stood erect, but not at attention, with arms across the chest, either with closed or open eyes as instructed, and the recording was started. First, spontaneous sway was recorded for 30 [s]. The stimulus was then started and maintained at a current of 1.00 [mA], during which period the stimulus changed polarity (recordings from a typical experiment are shown in Figs. 1 and 2).

V. Data Analysis

Basic data analysis was made by means of system identification methodology [30], [34]. Autospectra, cross-spectra,

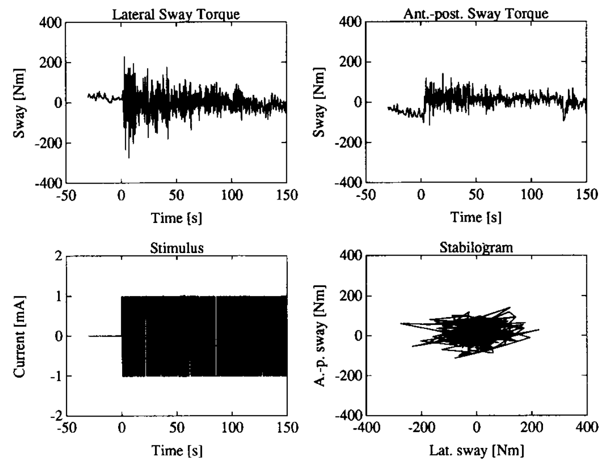


Fig. 1. Sway response to 1 [mA] galvanic stimulation with closed eyes: Galvanic stimulation (*lower left*) starting at time $t = 0$; lateral sway response (*upper left*) and anterior-posterior sway response (*upper right*). Stabilogram (*lower right*) of sway torques [Nm]. Time scale in [s].

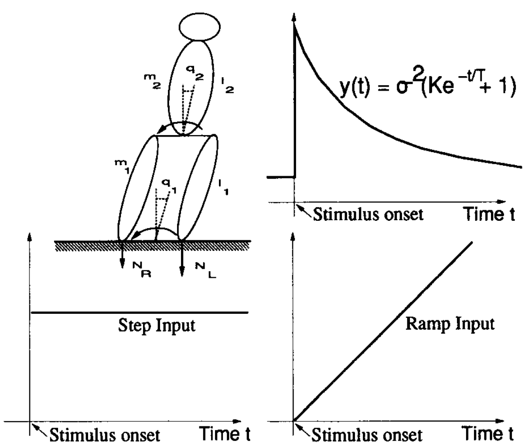


Fig. 2. Biomechanical model with normal forces N_L, N_R of the feet shown (Appendixes I and II). Step and ramp input models modeling onset of stimulus. Exponential $y(t) = \sigma^2(Ke^{-t/T} + 1)$ for $t > 0$ fitted to the residual variance of the estimated model. Responses to the step and ramp inputs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

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 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] (see Figs. 1 and 3).

A. Biomechanical Modeling

We used a multilink inverted pendulum model with coordinates $q = (q_1\ q_2\ \ldots)$ of the links and an associated control model (see Fig. 2). A detailed account of the modeling assumptions is given in Appendix I. It is shown in Appendix III that a linearized transfer function from the stimulus u to the torque responses τ_{bal} for a two-link stabilized inverted

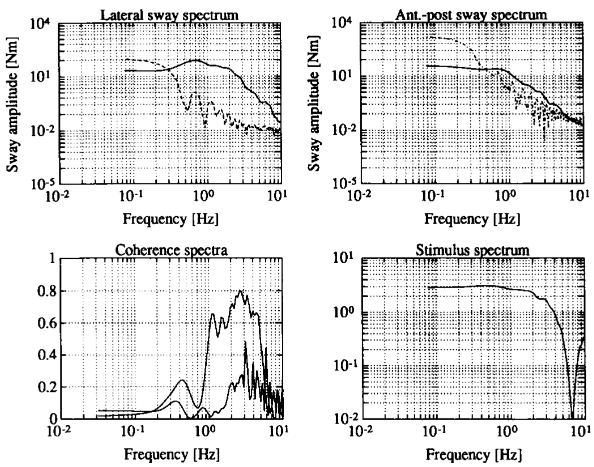


Fig. 3. Spectral analysis of stimulus and response from galvanic stimulation experiment. Solid lines depict the result during stimulation and dashed lines the spontaneous sway. Lateral sway spectrum (*upper left*). Anterior posterior sway (*upper right*). Coherence spectra between stimulus and responses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

pendulum is found as

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

where $G(q)$ designates gravitation forces and $M(q)$ 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); see Appendix I.

B. Assessment of Dynamics and Stability

The methodology to fit a transfer function like (1) to data is known as system identification, or as an inverse problem, and is described elsewhere. Briefly, Matlab$^{\mathrm{TM}}$ [37] was used for time-series analysis [30]. As data exhibited clear nonstationary properties with trends and time varying behavior (see 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 structure

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

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- 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 electronic offsets. Hence, a stochastic time-series model in the form of an autoregressive moving average model with external inputs (ARMAX) was fitted to the experimental data by means of pseudolinear regression, i.e., an iterated least-squares method, for a sequence of increasing model orders $n = 1, 2, 3, 4, 5, 6, \ldots$. Determination of the estimated model of a suitable model order was supported by model validation criteria [30], [34], such as statistical evaluation of the loss function the Akaike information criterion (AIC), the final prediction criterion (FPE), and residual analysis (tests of autocorrelation and cross-correlation between stimulus and residuals) [30], [34]. The goal of identification was to determine a time-invariant model according to (2) with the minimum residual power and a sequence of zero-mean noncorrelated residuals ("white noise" in engineering terminology). The correspondence between the resulting ARMAX model and the biomechanical continuous-time model was done by standard methodology; see [30].

C. Assessment of Adaptation

Adaptation-related properties were quantified by extracting the following information from sequences of squared residuals, i.e., sequences of squared noncorrelated zero-mean residuals, cf. Figs. 3 and 6:

$$\mathcal{E}\{\varepsilon_k^2\} = \begin{cases} \sigma^2, & t < 0 \\ \sigma^2(Ke^{-t/T} + 1), & t > 0 \end{cases} \quad \text{where} \begin{cases} t = kh - 30 \\ h = 0.05 \end{cases} \tag{3}$$

and where

K: K denotes the increase in the residual variance in response to the onset of stimulus, and

T: 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 are shown in Table I.

VI. Results of the Experiments

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 described by means of the step and ramp inputs. 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 for the experiments [29], [30]. (A detailed presentation of the calculation procedure is given in [31, Appendix 3].) 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 in the stimulated sway spectra was demonstrated, as compared to the spontaneous sway (see Fig. 3).

TABLE I

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}

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

The results with closed eyes were quite convincing, with good coherence between galvanic stimulus u and the lateral body sway, whereas the results for open eyes were inconclusive. This finding indicates that there is a substantial response to stimulation in the absence of visual input, whereas visual input tends to attenuate the effect of the stimulus. The continuous-time pole polynomial (transfer function denominator) of (4) was computed with Matlab and a fourth-order characteristic polynomial representing the stabilized body mechanics is obtained as

$$\det(s^2 I_{2\times 2} + sK_d + K_p) = s^4 + p_1 s^3 + p_2 s^2 + p_3 s + p_4. \quad (4)$$

The results obtained all characterize stable systems with pole polynomials which may be examined in their own right to characterize stability properties; see [31].

A stochastic process model in the form of an autoregressive moving average model with external inputs (ARMAX model according to (2)) was fitted to the experimental data with pseudolinear regression, which belongs to the family of least-squares methods. 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) [30], [34]. Also, 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 statistical validation criteria [30, ch. 10].

The squared value of residuals for one typical subject is shown in Fig. 6. Although the residual sequence is nonstationary with time-varying variance, it exhibits white-noise properties with mutually uncorrelated residuals and residuals

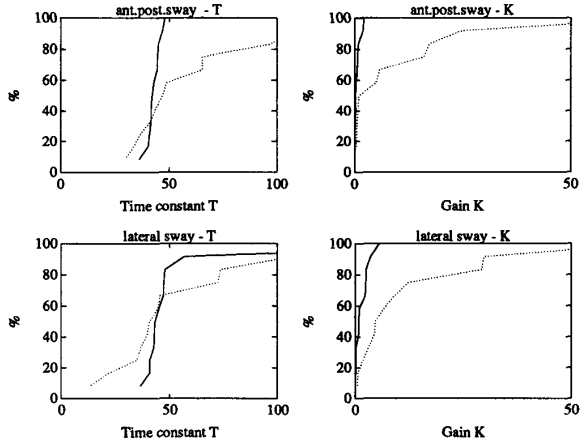


Fig. 4. Empirical distribution of the residual variance adaptation time constant T (*left*) and the initial gain (*right*) for anterior-posterior sway (*upper*) and lateral sway (*lower*). Notice that the adaptation time, constant for open eyes (*solid line*), appears to have a small variation range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

uncorrelated with any of the inputs. As a least-squares type of method is being used, there can be no artifacts arising from assumptions on certain stochastic models.

A. Assessment of Adaptation

Adaptation-related properties were quantified by extracting the following information from the graphs of residual power, cf. Fig. 6:

K: Gain

T: The time constant of the attenuation of residual power from its peak value.

The time of the peak value of residuals and the time constant of the residual attenuation were calculated and are found in Table I. Empirical distributions of K and T for both anterior-posterior and lateral sway responses are shown in Fig. 4.

VII. DISCUSSION

A. Effects of Galvanic Stimulus

A bipolar biaural galvanic stimulus induces vestibular and postural responses [35]. 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 [9], [36]. Galvanic induced vestibular responses in humans reflect the integrity of the vestibular nerve, but not of the labyrinth [42], [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 [42]. A less powerful stimulus may, however, cause asymmetry of vestibulary mediated eye motor reflexes [7]. Lateral postural sway is induced by currents of less than 0.4 mA, which does not evoke sensations from the skin over the mastoid [50]. A bipolar biaural 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 [49]. 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 *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.

B. Interpretation of Experimental Data

A primary effect of the bipolar galvanic stimulus is that the lateral sway dominates over the anterior-posterior sway [35]. It is also apparent from data that the impact on the sway starts with a certain latency, as previously reported by Sekitani [24], [46] and others.

The spectral properties of lateral sway show that there was more prominent high-frequency motion during stimulation in all subjects (Fig. 3), whereas anterior-posterior sway decreased in magnitude in most subjects. This reduction in anterior-posterior sway may reflect some increased attention and effort to maintain balance on the part of the test subject.

The coherence spectra show that the galvanic stimulus affects motion in the frontal plane, whereas much less impact is seen in the sagittal plane. The coherence between galvanic stimulus and lateral sway is high in the frequency range from 0.1–0.2 Hz to 5–10 Hz, whereas coherence is low below this frequency range.

C. Model Complexity

When exposed to sagittal perturbation, a subject may regain equilibrium by two different strategies, "ankle strategy," in which muscular forces rotate the body around the ankle joint, see [18], or "hip strategy," involving flexion at the hips and knees, see [38]. The physiological significance of force, velocity, and position feedback has been discussed by Houk [25]. We have not explicitly modeled the force feedback from the Golgi tendon organ. However, as any (negative) force feedback results in a modified $M(q)$, the mechanical moment of inertia $M(q)$ of (1) apparently increases to facilitate the modulation of force. Verification of the existence of force feedback can be made with our method by comparison of the computed body mechanical moment of inertia $M(q)$ with that of the identification.

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 modeling, 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. Extraction of the dominant motion features corresponding to a simple inverted pendulum allows a crude stability and performance characterization similar to that reported in [31]. 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. The fourth-order dynamics correspond to the dynamics of the compound center of mass, as well as and higher-frequency dynamics originating from the biomechanical interactions between the body segments. Possible force feedback in the sway force response would not change the system order, but only the interpretation of the inertia matrix $M(q)$.

The neurologically modulated muscles provide a constrained motion. Not only do the mechanical properties affect motion, but also the volition. Changes in the desired posture can not be measured, and therefore constitute a source of error in the parameter estimates. However, as we incorporate the artificial step- and ramp-formed inputs and the corresponding responses in the time-series analysis to model such adaptive features as the time-varying shift of the center of pressure after onset of stimulus, we can exclude any major effect of such error. This approach proves successful as the residual sequences of system identification exhibit white noise properties and fulfill other standard validation criteria; see [30, ch. 9].

D. Problems of Adaptation and Control

The possible adaptation of the subjects to the stimulus raises several methodological questions as to the reproducibility of the results and the reliability of time series analysis which is based on assumptions of time-invariant statistical properties.

First, the stimulus has time-invariant statistical characteristics after the 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. As opposed to Courjoun *et al.* [9], 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.

Second, the application in time-series analysis of artificial step- and ramp-formed inputs commencing at the onset of stimulus serve to model the response to the stimulus in the choice of a new equilibrium position, i.e., a new posture, see Fig. 5. The change of posture after onset of stimulation varies in direction for different subjects and tends to be larger for test conditions with closed eyes. The directions of the step- and ramp-response appear to have indefinite correlation.

In addition, we have been relating our results not to those of other experiments but to the performance of the subject before the onset of galvanic stimulus, see Fig. 4. Identification here serves to separate the mechanically determined responses from sway force responses of other origin. Examination of original sway-force responses, as well as residual sequences after separation of mechanically determined responses from data by means of system identification, clearly shows a time-variant behavior.

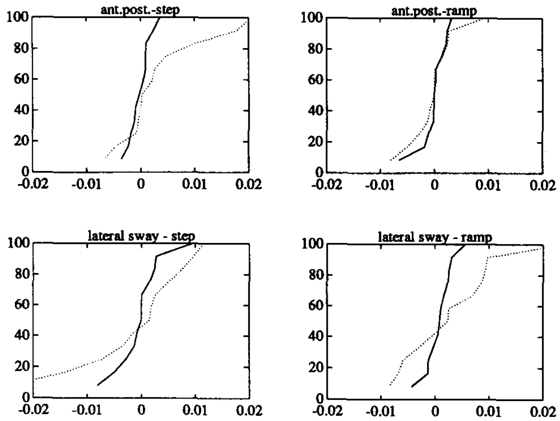


Fig. 5. Empirical distributions of step- and ramp-formed shifts in the center-of-pressure for open eyes (*solid line*) and closed eyes (*dotted line*), respectively.

As reduction in sway response amplitude or variance over time cannot simply be interpreted as a transient component of the feedback response or a mechanically determined response, there is obviously a change of gain at some level of the control system. This circumstance supports the hypothesis that the body adapts to the stimulus during the course of the experiment.

E. 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 strength with an adaptation time constant in the range of 40–50 [s] (Table I), whereas vestibular habituation is generally characterized by a much longer time course when studied in the vestibulo-ocular reflex [51]. This raises some intriguing questions: Is the short time-related decrease of sway in the lateral plane, and earlier observed also in the sagittal plane [35], an effect of physiological habituation to the stimulus, or is it an effect of a decrease in the perceived intensity of the vestibular stimulus? It is known that, after a short period of constant galvanic stimulation, the induced nystagmus disappears [15]. In the present study, however, there was a discontinuous stimulation in the sense that the polarity changed pseudorandomly during the course of the experiment. Moreover, a decrease of stimulus intensity due to polarization of the tissues can not account for the decrease in response, as a constant-current generator was used to generate the stimulus, and it is known from animal experiments that galvanic stimulation of the vestibular nerve does not lead to a reduction in sensory afferent input in the nerve.

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.,

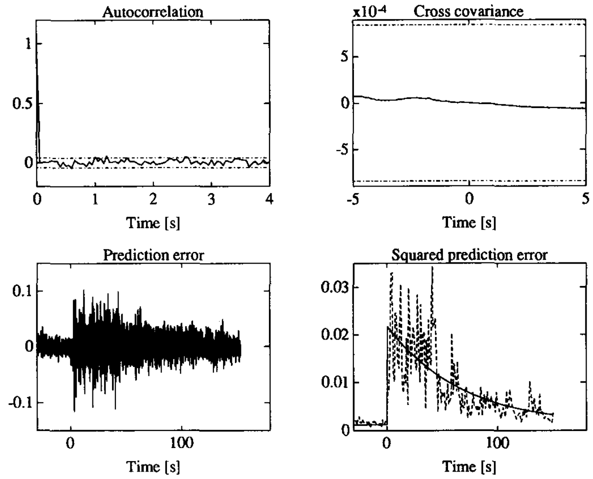


Fig. 6. Residual analysis versus time [s] of the full time series with residual autocorrelation and cross-covariance between residuals and stimulus (the dashed lines in the upper graphs indicate the confidence intervals ($p < 0.05$) for normally distributed residuals). The lower graphs show the prediction error (*left*) and the squared prediction error (*right graph, dashed line*) with a fitted exponential (*right graph, solid line*). Time $t = 0$ indicates the onset of the galvanic stimulus. The spontaneous body sway was recorded for 30 s prior to the onset of the stimulation.

adaptation. The assumption that the adaptive reduction of postural responses is dependent on comparison with other sensory information is supported by the observed decrease rate in galvanically induced body sway after hypothermal anesthesia of the soles of the feet [35].

Results in Table I 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. Moreover, the sensitivity to neuromuscular dysfunction, pharmacological agents, and disease processes may elucidate the physiological basis of the adaptation time constant and will be subject to future study.

F. Test Application

The present test duration of 183.6 [s] may be too long for the purposes of clinical testing. Computations of data from the first part of the time series show that the test duration can be reduced without seriously affecting the quality of parameter estimates. We proceed in our clinical research to evaluate the relevance of the demonstrated adaptation in various contexts of neuromuscular dysfunction.

VIII. Conclusion

We have designed a methodology for quantitative investigation of feedback properties for characterization of human lateral posture stability and adaptation to vestibular stimulus.

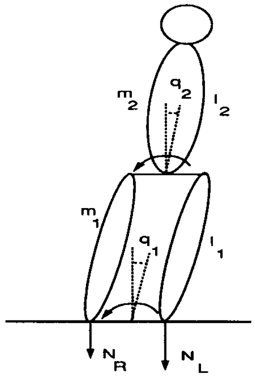


Fig. 7. A two-segment model of lateral posture in the form of an inverted double pendulum. The subject is standing with the feet at a distance 2δ apart. The two body segments have masses m_1, m_2 and are of lengths l_1, l_2, respectively. Normal forces N_L, N_R of the feet are shown; see Appendixes I and II for biomechanical analysis.

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 noncorrelated but time-variant residual sequence permitted extraction of a time constant of 40–50 s to model the decrease of the residual variance in the course of the experiment.

The method is intended for testing of balance disorders, but further investigation of physiological and clinical relevance is required prior to application.

APPENDIX I
MODELING OF FORCE RESPONSES IN POSTURAL CONTROL

The model is formulated for dynamics in the frontal plane with the body conceived of as an inverted double pendulum with possible motion at the ankle and hip. The inverted double pendulum has an unstable equilibrium point at $q_1 = q_2 = 0$ (see Fig. 5), which means that active stabilizing forces must compensate for deviations in position in order to maintain upright stance. The balancing forces exerted are the result of a complex event invoking all body muscles acting in concert. In order to formalize and simplify analysis, the following assumptions are made:

1) Assumption 1: The body is considered to be composed of stiff segments with articulation at the hip. The body segments have masses m_1, m_2 [kg], respectively.

2) Assumption 2: The body center of mass is located at distance l [m] from the platform surface, the body segments being of lengths l_1, l_2 [m], respectively.

3) Assumption 3: There is a dynamic equilibrium between the torque τ of the body muscles and external forces acting on the "double pendulum" with orientation described by the angular coordinates $q = (q_1\ q_2)^T$ of Fig. 7.

Introduce M as the body moment of inertia around the ankle and the hip. According to the laws of classical mechanics, it follows that the torque equilibrium for a standing person, subject to gravitation $G(q)$, is then

$$M(q)\ddot{q} + C(\dot{q}, q) + G(q) = \tau, \quad \text{where } q = \begin{pmatrix} q_1 \\ q_2 \end{pmatrix}. \quad \text{(A1.1)}$$

The first term of (A1.1) denotes inertia forces, the second term C denotes the centripetal and Coriolis forces, and the third term $G(q)$ denotes gravitation forces and torques. The sum of inertia forces, centripetal and Coriolis forces, and gravitation forces equals the applied forces τ, e.g., muscular forces and disturbances from the environment:

A subject who does not counteract the gravitational torque with a stabilizing response will inevitably fall, and both mathematically and intuitively, it is easy to understand that there is no mechanically determined stable equilibrium at $q = 0$ and $\dot{q} = 0$ which represent upright stance. The following two assumptions are introduced to model balance action and the possible effect of disturbances from the environment:

4) Assumption 4: That there are stabilizing joint torques τ_{bal} at the ankle (τ_1) and the hip (τ_2) with

$$\tau_{\text{bal}}(t) = (\tau_1\ \tau_2)^T. \quad \text{(A1.2)}$$

5) Assumption 5: That there is a disturbance torque, $\tau_d(t)$, from the environment.

The external torques acting at the joints now have the form

$$\tau(t) = \tau_{\text{bal}}(t) + \tau_d(t). \quad \text{(A1.3)}$$

Active balance of any inverted pendulum requires proportional and derivative stabilizing action at the joints. The components P (proportional) and D (derivative) are indispensible for stability, according to the Routh criterion of stability [22], and we assume that PD-control via the joint torques τ_{bal} represents the minimum complexity necessary of the stabilizing control.

6) Assumption 6: That τ_{bal} stabilizes the posture with PD-control with the components P, D determined by the feedback coefficients of K_p, K_d and anti-gravity action and anti-Coriolis compensation.

$$\begin{aligned} P&: \quad -G(q) - M(q)K_p q \\ D&: \quad -C(\dot{q}, q) - M(q)K_d \dot{q}. \end{aligned} \quad \text{(A1.4)}$$

The model, then, consists of an inverted double pendulum to explain the biomechanical interactions and a balance control system. PD-control is chosen as a conjecture because the proportional and derivative actions are fundamental modes of control [17] and contain the minimum complexity necessary for stabilizing postural control of a segmented inverted pendulum. The component P contains anti-gravitational forces and compensates for position errors. The component D contains an anti-Coriolis compensation and also provides damping action. The parameter matrix K_p may be interpreted as a stiffness matrix (spring constant) arising from passive and active muscular forces, whereas K_d might be compared with a viscous damping, as obtained with a dashpot. The postulated relationship between the torques τ_{bal} and the angular positions q, angular velocities $\dot{q}$ is then

$$\tau_{\text{bal}}(t) = -G(q) - C(\dot{q}, q) - M(q)K_p q(t) - M(q)K_d \dot{q}(t). \quad \text{(A1.5)}$$

Suffice it to mention that this is the minimum complexity anticipated for a stabilized inverted pendulum.

Finally, it is necessary to model the effect of the galvanic stimulus.

7) Assumption 7: The galvanic stimulation u introduces erroneous input into the stabilizing system, causing misperception of the position q and the angular velocity $\dot{q}$ (rate), so that the P, D-actions of feedback system are modified to

$$\begin{aligned} P&: \quad -G(q) - M(q)K_p(q(t) - b_1 u(t)) \\ D&: \quad -C(\dot{q}, q) - M(q)K_d(\dot{q}(t) - b_2 u(t)). \end{aligned} \tag{A1.6}$$

It is assumed that the galvanic stimulus u possibly disturbs both position and velocity perception at the different proportions, b_1 and b_2, respectively [51]. The applied torques at the hip and ankle are then

$$\begin{aligned} \tau = &- G(q) - M(q)K_p(q(t) - b_1 u(t)) \\ &- C(\dot{q}, q) - M(q)K_d(\dot{q}(t) - b_2 u(t)). \end{aligned} \tag{A1.7}$$

A calculation (see Appendix II) using the torque equilibrium of (2) and *Assumption 7* gives the transfer function from the galvanic stimulation u to the response in joint positions q and velocities $\dot{q}$.

$$Q(s) = \left(s^2 I_{2\times 2} + sK_d + K_p\right)^{-1}(K_p b_1 + K_d b_2)U(s) \tag{A1.8}$$

where $Q(s), U(s)$ are the Laplace transforms of $q(t), u(t)$, respectively. The concept of transfer function adopted here is that of Laplace transforms as used in control theory and signal processing [22], [17]. Functions in the variable t indicate time domain functions, and functions in variable s and/or capital letters denote frequency domain functions in complex frequency s.

A. Forces on the Platform

Ground reaction forces are naturally characterized with three forces F_x, F_y, F_z and torques τ_x, τ_y, τ_z with respect to coordinates x, y, z in Cartesian space [31]. In particular, the ankle torques τ_x, τ_y represent the forces associated with stabilized anterior-posterior and lateral motion around the ankle joint according to our test conditions. With static equilibrium between the force on the platform and the body weight, it follows that the force center also represents the projection on the platform of the body center of gravity. However, as shown in [32], there is an important discrepancy between the primary sway force responses measured in Nm and the assumed stabilogram (measured in units of distance). Such a model is therefore not entirely satisfactory for the purposes of dynamic analysis, as the force center and vertical projection of the center of body mass do not generally coincide at the same point. For example, the foot may exert a corrective force on the platform to initiate an angular acceleration of the body.

It is shown in Appendices II and III that a linearized transfer function from the stimulus u to the torque responses τ from the force platform is found as

$$\begin{aligned} \tau_{\mathrm{bal}}(s) \approx &\left(M(0)s^2 - \left.\frac{\partial G(q)}{\partial q}\right|_{q=0}\right) \\ &\times \left(s^2 I_{2\times 2} + sK_d + K_p\right)^{-1}(b_1 + b_2)U(s). \end{aligned} \tag{A1.9}$$

Note that (A1.9) provides a fourth-order linear model and that this approximation of linearization does not affect the pole loci but only the zeros.

Appendix II

Euler-Lagrange equations [19] of motion are derived via expressions for kinetic and potential energy and Lagrange functions. We have used the following notations where time arguments have been omitted. The body segments are assumed to have their center of gravity located at distance $l_1/2, l_2/2$ from the respective segment endpoints. Modifications to other locations of center of gravity are trivial, and thus not crucial to further calculations.

q	Joint angular positions $q = (q_1\ q_2)^T$	(dim $q = 2 \times 1$)
$\dot{q}$	Joint angular velocities	(dim $\dot{q} = 2 \times 1$)
$\ddot{q}$	Joint angular acceleration	(dim $\ddot{q} = 2 \times 1$)
τ	Joint torques	(dim $\tau = 2 \times 1$)

and the short notation $c_2 = \cos(q_2), s_{12} = \sin(q_1 + q_2)$ etc.

The potential energy $\mathcal{U}$ of the segmented body according to Fig. 7 is found as the sum of the potential energy of each one of the segments.

$$\begin{aligned} \mathcal{U}(q) &= 2m_1 g\left(\frac{1}{2}l_1 c_1\right) + m_2 g\left(l_1 c_1 + \frac{1}{2}l_2 c_2\right) \\ &= (m_1 + m_2)g l_1 c_1 + \frac{1}{2}m_2 g l_2 c_2. \end{aligned} \tag{A2.1}$$

The kinetic energy $\mathcal{T}$ is

$$\begin{aligned} \mathcal{T}(q, \dot{q}) = 2 \cdot \frac{1}{2}m_1 &\left(\left(\frac{l_1}{2}\dot{q}_1 c_1\right)^2 + \left(\frac{l_1}{2}\dot{q}_1 s_1\right)^2\right) \\ &+ \frac{1}{2}m_2\left(\left(-\frac{l_2}{2}\dot{q}_2 c_2 + l_1\dot{q}_1 c_1\right)^2\right. \\ &\left. + \left(-\frac{l_2}{2}\dot{q}_2 s_2 - l_1 \dot{q}_1 s_1\right)^2\right). \end{aligned} \tag{A2.2}$$

The body moment of inertia $M(q)$ is therefore defined via the kinetic energy (A2.2) and the relation

$$\begin{aligned} \mathcal{T}(q, \dot{q}) &= \frac{1}{2}\dot{q}^T M(q)\dot{q} \\ &= \frac{1}{2}(\dot{q}_1 \quad \dot{q}_2)^T \\ &\quad \times \begin{pmatrix} (\frac{1}{2}m_1 + m_2)l_1^2 & -\frac{1}{2}m_2 l_1 l_2 c_{12} \\ -\frac{1}{2}m_2 l_1 l_2 c_{12} & \frac{1}{4}m_2 l_2^2 \end{pmatrix}\begin{pmatrix} \dot{q}_1 \\ \dot{q}_2 \end{pmatrix}. \end{aligned} \tag{A2.3}$$

The Lagrangian L is according to methods of classical mechanics [6]

$$L = \mathcal{T} - \mathcal{U} \tag{A2.4}$$

and the Euler-Lagrange equations of motion are found as

$$\frac{d}{dt}\frac{\partial L}{\partial \dot{q}_1} - \frac{\partial L}{\partial q_1} = \tau_1 \quad \frac{d}{dt}\frac{\partial L}{\partial \dot{q}_2} - \frac{\partial L}{\partial q_2} = \tau_2. \tag{A2.5}$$

$$\begin{pmatrix}\tau_1\\ \tau_2\end{pmatrix} = \begin{pmatrix}(\frac{1}{2}m_1 + m_2)l_1^2\ddot{q}_1 - \frac{1}{2}m_2 l_1 l_2 c_{12}\ddot{q}_2 + \frac{1}{2}m_2 l_1 l_2 s_{12}\dot{q}_1^2 - (m_1+m_2)g l_1 s_1\\ -\frac{1}{2}m_2 l_1 l_2 c_{12}\ddot{q}_1 + \frac{1}{4}m_2 l_2^2\ddot{q}_2 + \frac{1}{2}m_2 l_1 l_2 s_{12}\dot{q}_2^2 - \frac{1}{2}m_2 g l_2 s_2\end{pmatrix}. \tag{A2.6}$$

Evaluation of the partial derivatives of (A2.5) gives (A2.6) shown at the top of the page. Identification of terms of (A2.6) gives

$$\tau = M(q)\ddot{q} + C(\dot{q}, q) + G(q) \tag{A2.7}$$

with the moment of inertia $M(q)$ of (A2.3) and the centripetal forces

$$C(\dot{q}, q) = \begin{pmatrix}\frac{1}{2}m_2 l_1 l_2 s_{12}\dot{q}_1^2\\ \frac{1}{2}m_2 l_1 l_2 s_{12}\dot{q}_2^2\end{pmatrix}. \tag{A2.8}$$

The gravitational forces are

$$G(q) = \frac{\partial \mathcal{U}(q)}{\partial q} = \begin{pmatrix}-(m_1+m_2)g l_1 s_1\\ -\frac{1}{2}m_2 g l_2 s_2\end{pmatrix}. \tag{A2.9}$$

The natural eigenfrequencies are determined from $M(q)$ and $G(q)$, with the slower eigenfrequencies roughly corresponding to periodic variations in the compound center of gravity.

Appendix III

The body mechanics are described by (A2.7) and the applied torques at the hip and the ankle are assumed to be

$$\begin{aligned}\tau = &- G(q) - M(q)K_p(q(t) - b_1(t))\\ &- C(\dot{q}, q) - M(q)K_d(\dot{q}(t) - b_2 u(t)).\end{aligned} \tag{A3.1}$$

Elimination of τ between (A2.7) and (A3.1) gives the closed-loop control

$$M(q)\ddot{q} + M(q)K_d\dot{q} + M(q)K_p q = M(q)(K_p b_1 + K_d b_2)u. \tag{A3.2}$$

Elimination of the multiplicative inertia matrix factor $M(q)$ gives

$$\ddot{q}(t) + K_d\dot{q}(t) + K_p q(t) = (K_p b_1 + K_d b_2)u(t). \tag{A3.3}$$

Laplace transformation of this linear ordinary differential equation gives

$$s^2 Q(s) + sK_d Q(s) + K_p Q(s) = (K_p b_1 + K_d b_2)U(s) \tag{A3.4}$$

where s denotes the Laplace variable of complex frequency. A transfer function from the galvanic stimulation U to the lateral body orientation Q is obtained as

$$Q(s) = \left(s^2 I_{2\times 2} + sK_d + K_p\right)^{-1}(K_p b_1 + K_d b_2)U(s). \tag{A3.5}$$

A linearization around the equilibrium point $q = 0$, $\dot{q} = 0$ gives a suitable approximate transfer function from u to the torques τ as

$$G(q) = G(q_0) + \left.\frac{\partial G(q)}{\partial q}\right|_{q=q_0}(q - q_0) + \cdots$$

where $G(q_0) = 0$ and

$$\left.\frac{\partial G}{\partial q}\right|_{q=q_0} = \begin{pmatrix}-(m_1+m_2)g l_1 & 0\\ 0 & -\frac{1}{2}m_2 g l_2\end{pmatrix} \tag{A3.6}$$

so that

$$\tau_{\text{bal}}(s) \approx \left(M(0)s^2 - \left.\frac{\partial G}{\partial q}\right|_{q=q_0}\right)Q(s)$$

or

$$\begin{aligned}\tau_{\text{bal}}(s) \approx &\left(M(0)s^2 - \left.\frac{\partial G}{\partial q}\right|_{q=0}\right)\left(s^2 I_{2\times 2} + sK_d + K_p\right)^{-1}\\ &\times (K_p b_1 + K_d b_2)U(s).\end{aligned} \tag{A3.7}$$

Notice that there is no approximation with respect to the poles of the system. The characteristic equation is given as

$$\det\left(s^2 I_{2\times 2} + sK_d + K_p\right) = 0. \tag{A3.8}$$

Factorization of (A3.8) into

$$\det\left(s^2 I_{2\times 2} + sK_d + K_p\right) = s^4 + p_1 s^3 + p_2 s^2 + p_3 s + p_4 = 0 \tag{A3.9}$$

for some feedback coefficients $p_1, \ldots, p_4$. Identification of the transfer function (A3.7) with standard linear system methods [30], [34] gives the characteristic polynomial coefficients.

References

[1] J. Babinski, "Sur le mécanisme du vertige voltaïque," *C. R. Soc. Biol. (Paris)*, pp. 350–353, 1903.

[2] F. Baldissera, P. Cavallari, and G. Tassone, "Effects on transmastoid electrical stimulation on the triceps brachii EMG in man," *Neuroreport*, vol. 1, pp. 191–193, 1990.

[3] J. B. Baron, N. Ushio, and R. Noto, "Oculo-nuco-vestibulospinal system regulating the tonic postural activity statokinesimetric studies," *Agressologie*, vol. 15, no. 6, pp. 395–400, 1974.

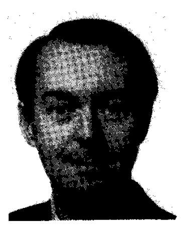
[4] F. O. Black, C. L. Shupert, F. B. Horak, and L. M. Nashner, "Abnormal postural control associated with peripheral vestibular disorders," in *Progress in Brain Research*, vol. 76, O. Pompeiano and J. H. J. Allum, Eds. Amsterdam: Elsevier Science Publishers, 1988, pp. 263–275.

[5] G. E. P. Box and G. W. Jenkins, *Time Series Analysis—Forecasting and Control*. San Francisco: Holden-Day, 1970.

[6] T. Brackman and U. Ranft, "Zum Stellenwert der Galvanischen gleichgewichtsprüfung," *HNO*, vol. 37, pp. 174–178, 1989.

[7] K. Brantberg and M. Magnusson, "Galvanically induced asymmetric optokinetic afternystagmus," *Acta Otolaryngol.*, vol. 110, pp. 189–195, 1990.

[8] A. C. Coats, "Limit of normal of the galvanic body sway test," *Ann. Otolaryngol.*, vol. 81, pp. 410–416, 1972.
[9] J. H. Courjoun, W. Precht, and D. W. Sirkin, "Vestibular nerve and nuclei unit responses and eye movement responses to repetetive galvanic stimulation of the labyrinth in the rat," *Exp. Brain Res.*, vol. 66, pp. 41–48, 1987.
[10] M. L. Daley and R. L. Swank, "Quantitative posturography: Use in multiple sclerosis," *IEEE Trans. Biomed. Eng.*, vol. BME-28, no. 9, 1981.
[11] J. Dichgans and H. C. Diener, "The contribution of vestibulospinal mechanisms to the maintenance of human upright posture," *Acta Otolaryngol.*, (Stockholm), vol. 107, pp. 338-345, 1989.
[12] H. C. Diener, J. Dichgans, B. Guschbauer, and M. Bacher, "Role of visual and static vestibular influences on dynamic posture control," *Human Neurobiol.*, vol. 5, pp. 105–113, 1986.
[13] V. Dietz, "Human neuronal control of automatic functional movements: Interaction between central programs and afferent input," *Physiolog. Rev.*, no. 1, pp. 33–69, Jan. 1992.
[14] C. Dohlman, "Experimentelle Untersuchungen über die galvanische Vestibularisreaktion," *Acta Otolaryngol.*, (Stockholm), vol. 8, pp. 1–48, 1929.
[15] M. Emami-Nouri and W. Gedlicka, "Ein Beitrag zur galvanischen Stimulation des Vestibularapparates," *Laryngol. Rhinol.*, vol. 53, pp. 200–204, 1974.
[16] C. Eyzaguirre and S. J. Fidone, *Physiology of the Nervous System.* Chicago: Yearbook Med. Publ., 1975, pp. 163–248.
[17] G. F. Franklin, J. D. Powell, and A. Emami-Naeini, *Feedback Control of Dynamic Systems.* Reading, MA: Addison-Wesley, 1986, sec. 3.3.
[18] C. L. Golliday, "Toward development of biped locomotion controls: Planar motion control of kneeless biped standing and walking gaits," Ph.D. dissertation, Columbus, OH, 1975.
[19] H. Goldstein, *Classical Mechanics.* Reading, MA: Addison-Wesley, 1950, ch. 1.
[20] M. Gresty, "Stability of the head: Studies in normal subjects and in patients with labyrinthine disease, head tremor, and dystonia," *Movement Disorders*, vol. 2, no. 3, pp. 165–185, 1987.
[21] S. Grillner, "Neurobiological bases of rhythmic motor acts in vertebrates," *Science*, vol. 228, pp. 143–149, 1985.
[22] W. Hahn, *Stability of Motion.* Berlin-Heidelberg: Springer-Verlag, 1967.
[23] N. G. Henriksson, G. Johansson, L. G. Olsson, and H. Östlund, "Electrical analysis of the Rhomberg test," *Acta Otolaryngol.*, suppl., vol. 224, pp. 272–279, 1967.
[24] S. Honjo, M. Tanaka, and T. Sekitani, "Body sway induced by galvanic stimulation," *Agressologie*, vol. 17, pp. 77–84, 1976.
[25] J. Houk, "Feedback control of muscle: A synthesis of the peripheral control mechanisms," in *Medical Physiology*, vol. 1, V. B. Mountcastle, Ed. Saint Louis: The Mosby Co., 1974.
[26] J. C. Houk, "Motor control processes: New data concerning motoservo mechanisms and a tentative model for stimulus-response processing," in *Posture and Movement*, R. E. Talbott and D. R. Humphrey, Eds. New York: Raven Press, 1979.
[27] A. Ishida and S. Imai, "Responses of the posture-control system to pseudorandom acceleration disturbances," *Med. Biol. Eng. Comput.*, vol. 18, pp. 433–438, 1980.
[28] A. Ishida and S. Miyazaki, "Maximum likelihood identification of a posture control system," *IEEE Trans. Biomed. Eng.*, vol. BME-34, pp. 1–5, 1987.
[29] G. M. Jenkins and D. G. Watts, *Spectral Analysis and its Applications.* San Francisco: Holden-Day, 1969.
[30] R. Johansson, *System Modeling and Identification.* Englewood Cliffs, NJ: Prentice-Hall, 1993.
[31] R. Johansson, M. Magnusson, and M. Åkesson, "Identification of human postural dynamics," *IEEE Trans. Biomed. Eng.*, vol. BME-35, 1988.
[32] R. Johansson and M. Magnusson, "Human postural dynamics," *CRC Crit. Rev. in Biomed. Eng.*, vol. 18, pp. 413–437, 1991.
[33] S. Lundh and C. Broberg, "Effects of different head positions on postural sway in man induced by a reproducible error signal," *Acta Otolaryngol.*, (Stockholm), vol. 117, pp. 307–309, 1983.
[34] L. Ljung, *System Identification—Theory for the User.* Englewood Cliffs, NJ: Prentice-Hall, 1987.
[35] M. Magnusson, H. Enbom, R. Johansson, and J. Wiklund, "Significance of pressor input from the human feet in lateral postural control—The effect of hypothermia on galvanically induced body sway," *Acta Otolaryngol.*, (Stockholm), vol. 110, pp. 321–327, 1990.
[36] L. B. Minor and J. M. Goldberg, "Vestibular nerve inputs to the vestibular-ocular reflex. Functional-abaltion study in the squirrel monkey," *J. Neurosci.*, vol. 11, pp. 43–60, 1991.
[37] C. Moler, J. Little, S. Bangert, and S. Kleiman, *Pro-MATLAB for Sun Workstations—User's Guide.* The Math Works, Inc., 1989.
[38] L. M. Nashner, "Vestibular postural control model," *Kybernetic*, vol. 10, pp. 106–110, 1972.
[39] ——, "Conceptual and biomechanical models of postural control—Strategies for organization of human posture," in *7th Int. Symp. Int. Soc. Posturography Vestibular and Visual Control on Posture and Locomotor Equilibrium*, Igarashi, Black, Eds. Houston, TX: 1983, pp. 1–8.
[40] L. M. Nashner and G. McCollum, "The organization of human postural movements: A formal basis and experimental synthesis," *Behavioral and Brain Sci.*, vol. 8, pp. 135–172, 1985.
[41] L. M. Nashner and P. Wolfson, "Influence of head position and proprioceptive cues on short latency postural reflexes evoked by galvanic stimulation of the human labyrinth," *Brain Res.*, vol. 67, pp. 255–268, 1974.
[42] C. R. Pfaltz, "The diagnostic importance of the galvanic test in otoneurology," *Practica Otorhinolaryngol.*, vol. 31, pp. 193–203, 1969.
[43] K. E. Popov, B. N. Smetanin, V. S. Gurfinkel, M. P. Kudinova, and V. Y. Shlykov, "Prostranstvennoye vospriyatiye *i* vestibulomotornyye reaktsii u cheloveka," *Neyrofisiolog.*, vol. 18, no. 6, pp. 779–787, 1986.
[44] I. Pyykkö, E. Toopila, J. Starck, H. Aaalto, H. Enbom, and H. Seidel, "Computerized posturography: Development of stimulation and analysis methods," in *Vertigo, Nausea, Tinnitus, and Hearing Loss in Cardiovascular Diseases*, C. F. Claussen and M. V. Kirtane, Eds. New York: Excerpta Medica, Elsevier, 1986.
[45] J. W. P. Roos, W. Bles, and J. E. Bos, "Postural control after repeated exposure to a tilting room," in *Posture and Gait: Development, Adaptation and Modulation*, B. Amblard, A. Berthoz, and F. Clarac, Eds. Amsterdam: Elsevier, 1988, pp. 137–144.
[46] T. Sekitani and M. Tanaka, "Test for galvanic vestibular responses—Survey of experimental and clinical investigation for the last 20 years," *Bullet. Yamaguchi Med. Sch.*, vol. 22, pp. 439–452, 1975.
[47] R. E. Talbott and J. M. Brookhart, "A predictive model study of the visual contribution to canine postural control," *Amer. J. Physiol.*, vol. 239, pp. R80–R92, 1980.
[48] R. E. Talbott, "Postural reactions of dogs to sinusoidal motion in the visual field," *Amer. J. Physiol.*, vol. 239, pp. R71–R79, 1980.
[49] T. Tokita, Y. Ito, and K. Takagi, "Modulation by head and trunk positions of the vestibulo-spinal reflexes evoked by galvanic stimulation of the labyrinth, observations by labyrinth evoked EMG," *Acta Otolaryngol.*, vol. 107, pp. 327–332, 1989.
[50] Y. Watanabe, K. Mizukoshi, H. Ohi, K. Ysuda, N. Ohashi, and H. Kobayashi, "Retrolayrinthine disorder detected by galvanic body sway responses in routine equilibrium examinations," *Acta Otolaryngol.*, suppl., vol. 468, pp. 343–348, 1989.
[51] V. J. Wilson and G. Melville-Jones, *Mammalian Vestibular Physiology.* New York: Plenum Press, 1979.



Rolf Johansson received the M.S. degree in technical physics in 1977, the B.M. (medicine) degree in 1980, and the Ph.D. degree in control theory 1983. He was appointed Docent in 1985, and received the M.D. degree in 1986, all from Lund University, Lund, Scandinavia.

In 1985, he spent six months with Laboratoire d'Automatique de Grenoble, Grenoble, France. Since 1986, he has been an Associate Professor of Control Theory at the Lund Institute of Technology. Currently, he acts as a Research Coordinator in Nutek-sponsored robotics research with participants from several departments of the Lund Institute of Technology. In 1993, he published the book *System Modeling and Identification* (Englewood Cliffs, NJ: Prentice-Hall). In his scientific work, he has been involved in research in adaptive system theory, mathematical modeling, system identification, robotics, signal processing theory, and biomedical research. Since 1987, he has participated in research at the Vestibular Laboratory within the Department of Otorhinolaryngology, Lund University Hospital, where he also has been an advisor to students presenting Ph.D. dissertations.

Måns Magnusson was born in 1954 and received the M.D. degree from the University of Lund, Sweden in 1981. As a postgraduate, he received professional and scientific training at the Department of Otolaryngology at the University Hospital of Lund. In 1986, he presented the Ph.D. dissertation on *The Optokinetic Mechanism in Man and Rabbit.*

In 1988, he was appointed Associate Professor and Head of the Vestibular Laboratory. His research interests comprise human postural conrol in health and disease, rehabilitation of disturbed balance, vestibular disorders, and physiology and control of eye movements.

Dr. Magnusson is a member of the Bárány Society and the Society for Posture and Gait Research.

Per A. Fransson was born in Röke, Sweden in 1964. He received the M.E. degree in electrical engineering from Lunds Institute of Technology, Lund, Sweden in 1991. Since 1991, he has been a Ph.D. student working at the Vestibular Laboratory at the Department of Otorhinolaryngology, Lund University Hospital.

His research interests are in the area of posturography, control, and adaptation.