Identification of Human Postural Dynamics

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Abstract—**The study consists of an analysis of measurements related to human postural dynamics, which was investigated in six healthy subjects by means of a force platform recording body sway induced by vibrators attached to the calf muscles. The model of body mechanics adopted was that of an inverted pendulum; the dynamics of postural control being assumed to be reflected in the stabilizing forces exerted on the platform by the feet as a result of complex muscular activity subject to state feedback of body sway and position. The approach to signal processing has been that of parametric identification of a transfer function representing the stabilized inverted pendulum.**

Posture control was quantified in three variables—swiftness, stiffness, and damping. It is shown that the identification fulfills ordinary statistical validation criteria, and it is conjectured that the state feedback parameters identified are suitable for use in assessing ability to maintain posture.

INTRODUCTION

HUMAN posture control is maintained by proprioceptive, vestibular, and visual feedback integrated within the central vestibular and locomotor system. Lesions to the sensory feedback system, or to the central nervous system, may impair postural control and equilibrium. It is therefore of interest to assess the ability of postural control by measuring the displacement of the body center of gravity. Recordings of the amplitude and frequency of spontaneous oscillations around the equilibrium position may describe the sway and thus, by extension, the control of posture.

Normally, spontaneous oscillation appears in healthy individuals during stance and the oscillating behavior of the body sway is often irregular or complex. Thus, a traditional static analysis is insufficient. It is also difficult to estimate a characteristic period of oscillation. Another problem is to analyze response to an external disturbance in the presence of spontaneous motion. Therefore, it makes sense to analyze the spectrum of oscillations rather than particular frequencies only. To understand the biological correlations of the posture control variables, it is also desirable to make a model-based analysis of the control system.

Several methods of dynamic analysis of spontaneous body sway have been reported [18], [20], [27]. One experimental setup [18] involved measurements obtained from a force platform and light-emitting diodes to detect position; the feedback characteristics were identified in closed-loop control, but without taking measured external disturbances into consideration. A variety of external disturbances, such as movable platforms and visual stimuli [32], [33], have been introduced to study dynamic aspects and improve the identifiability of posture control. A perturbation evoked by a movable platform, however, disturbs both proprioceptive and vestibular feedbacks simultaneously.

For the present study, we developed a model for posture control based on exposure of the subject to erroneous proprioceptive input. The stimulus is produced by vibration of the calf muscles, which results in activation of muscle spindles [22], [28]. Vibration is believed to activate the muscle spindles, as occurs during passive muscle stretch which causes a reflex contraction [11], [12]. Eklund [6]–[8] has studied the physiological effects of vibration-induced body sway on various muscle groups with respect to vibration frequency and intensity (amplitude). Eklund found that the effects could be elicited with any of a range of vibration amplitudes and frequencies.

In the present experiments, the stimulus used is vibration of the calf muscles. Body sway is measured with a force platform. The model adopted is that of the standing human body as an inverted pendulum, equipped with a servomechanism for balance. The model is designed so that the spectral analysis is compatible with a dynamic systems approach of signal processing and control theory. Laplace transform methods are used for transient input-output analysis. Parametric estimation is done with maximum likelihood estimation of coefficients in ARMAX-models. Model fitness and parameter uncertainty are analyzed statistically. The aim of the present study was to identify feedback parameters useful in evaluating ability to maintain posture control.

METHODS

Material

Tests were done on naive human subjects, three males and three females (mean age 28, range 23–39 years), none of whom had any history of vertigo, central nervous disorder, ear disease, or 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 hours.

Equipment and Experimental Setup

The equipment consisted of a square force platform connected up with a computer for data recording and com-

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putation. The force platform was developed at the Institute of Occupational Health, Helsinki, Finland, and the ENT Clinic of Lund, Sweden [29]. The platform is equipped with strain gauges to measure vertical force at each corner at four symmetrically located points. Measurements obtained from the strain gauges are recorded by the computer, and represent the differential distribution of forces exerted by the feet on the platform. The equipment allows simultaneous recording of body sway both in the saggittal and frontal planes. The stimulus is produced by vibration of the calf muscles at frequencies of 60 and 100 Hz, and of 0.4 mm in amplitude. The subject stood with heels together on the platform while staring at a spot on the opposite wall. A small vibrator was attached to the calf muscle in each leg with elastic straps. The subject stood erect but not at attention, either with closed or open eyes as instructed, and the recording was started. First, spontaneous sway was recorded. Then, the vibrators were turned on/off and modulated pseudorandomly (PRBS) according to a program executed in the computer while recording continued.

The frequency of the vibrators depended linearly on the input voltage v which had been checked for all vibrators before use.

As part of routine laboratory practice, it was verified that there was no interference (aliasing) between the sampling frequency and the vibration frequency. The test sequence took 150 s.

Modeling of the Posture Control System

When exposed to a saggittal perturbation a subject may regain equilibrium by two different strategies [10]: "ankle strategy," in which muscular forces rotate the body around the ankle joint, or "hip strategy," involving flexion at the hips and knees [25]. The ankle strategy is sufficient to counteract minor perturbations which occur during natural stance and fits the model of posture control as an inverted pendulum. Hip strategy has to be employed when the vertical projection of the body center of mass falls in front of the subject's feet.

In hip strategy, there is a potential problem with shear forces against the supporting surface. However, the force platform has been constructed so that shear forces do not interfere with the recorded signal. The moment of inertia may also change in pronounced movements because the center of body mass will be lowered. Thus, it is arguable that where gross compensatory movements are concerned, e.g., preventing the subject's imminent fall, the inverted pendulum model may be insufficient. However, in the natural stance and in the minor perturbations induced by the vibratory stimulus used here, the inverted pendulum model is fully adequate to account for the corrective movements used to control body posture.

The model is formulated for dynamics in the saggittal plane with the body conceived of as an inverted pendulum. The inverted pendulum has an unstable equilibrium point at $\theta = 0$ (Fig. 1) which means that active stabilizing forces must compensate for deviations in position in order to maintain posture. The balancing forces exerted are the result of a complex event invoking all body muscles acting in concert. A model of balance as a servomechanism need not, however, be more complicated than suffices to describe the resulting behavior as reflected in the measurements.

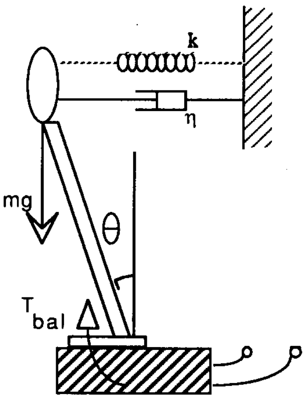

Fig. 1. Inverted pendulum model of human postural dynamics with the balancing torque T_{bal} similar to that achievable with a spring (k) and a dashpot (η).

The model, then, consists of an inverted pendulum to explain the pure body mechanics and a balance control system which acts like the shock absorber of a motor vehicle. The "suspension" is characterized by a spring constant k and a damping η which keep the body in an upright position and capable of counteracting disturbance. The response to an impulse is determined by the values of k and η, as well as m (body weight) and l (distance of the body center of mass from the platform surface).

Assumptions

The following assumptions are made in order to formalize and simplify analysis:

Assumption 1: The body is stiff, and has a mass m (kg).

Assumption 2: The body center of mass is located at distance $l(m)$ from the platform surface.

Assumption 3: There is a dynamic equilibrium between the torque of the foot and the forces acting on the "pendulum."

A person who does not counteract the forces of gravity may be modeled by the force equilibrium of an inverted pendulum. Introduce J as the body moment of inertia around the ankle, and the tangential torque equilibrium for a standing person subject to gravitation g is then

$$J\frac{d^2\theta}{dt^2} = mgl\sin\theta(t), \qquad J = ml^2. \tag{1}$$

It is easy to understand both mathematically and intuitively that there is no stable equilibrium at $\theta = 0$. A person who does not counteract the gravitational torque with a stabilizing response will inevitably fall. The following two assumptions are introduced to model balance action and the effect of disturbances from the environment:

Assumption 4: Assume that there is a stabilizing ankle torque $T_{\text{bal}}(t)$.

Assumption 5: Assume that there is a disturbance torque $T_d(t)$ from the environment.

The torque balance now has the form

$$J \frac{d^2\theta}{dt^2} = mgl \sin \theta(t) + T_{\text{bal}}(t) + T_d(t), \qquad J = ml^2. \tag{2}$$

We assume that PID-control (proportional, integrating, derivative) via the ankle torque T_{bal} is sufficient to represent the nature of the stabilizing control.

Assumption 6: Assume that T_{bal} stabilizes the posture with PID-control with the components P, I, D determined by coefficients k, η, and ρ.

$$P\colon\ -mgl \sin \theta(t) - kJ\theta(t)$$

$$D\colon\ -\eta J\dot{\theta}(t)$$

$$I\colon\ -\rho J \int_{t0}^{t} \theta(t)\, dt.$$

PID-control is chosen here because the proportional, derivative, and integral actions are fundamental modes of control. The PID concept is widely known [1], and contains what is necessary for postural control.

The components (P, D), and that $k, \eta > 0$ are indispensible for stability according to the Routh criterion of stability [15]. The integral component (I) accounts for (slow) compensation of bias in θ; as (I) is not *a priori* necessary for stability, one of the aims of the experiments is to show its presence.

The parameter k may be interpreted as a spring constant, and η might be compared with a viscous damping as obtained with a dashpot. The parameter ρ may be interpreted as a constant for the slow reset action in the control system. Finally, it is necessary to model the effect of the vibration stimulus.

Assumption 7: The vibration v introduces erroneous input into the stabilizing system, causing misperception of the position θ (stretch) and the angular velocity $\dot{\theta}$ (rate) so that the P, D actions of feedback system are modified to

$$P\colon\ -mgl \sin \theta(t) - kJ\theta(t) + b_1 v(t)$$

$$D\colon\ -\eta J\dot{\theta}(t) + b_2 v(t)$$

where it is assumed that v disturbs both stretch and rate perception but at different proportions b_1 and b_2, respectively.

Transfer Function of θ

The concept of transfer function adopted here is that of Laplace transforms as used in control theory [15]. Functions in the variable t indicate time domain functions, and functions in variable s and/or capital letters denote frequency domain functions. The torque equilibrium of (2) and assumption $A6$ give the two equations

$$J \frac{d^2\theta}{dt^2} = mgl \sin \theta(t) + T_{\text{bal}}(t) + T_d(t)$$

$$T_{\text{bal}}(t) = -mgl \sin \theta(t) - kJ\theta(t) - \eta J\dot{\theta}(t) - \rho J \int_{t0}^{t} \theta(t)\, dt. \tag{3}$$

There are three states that affect motion, namely, angular velocity $d\theta/dt$, angular position θ, and the bias compensation. A transfer function from vibration stimulus V and disturbance T_d to the torque T_{bal} is found via (2), (3) (see Appendix I).

$$T_{\text{bal}}(s) = \frac{(b_1 + b_2)(s^3 - g/ls^2)}{s^3 + \eta s^2 + ks + \rho} V(s) - \frac{\eta s^2 + (k + g/l)s + \rho}{s^3 + \eta s^2 + ks + \rho} T_d(s). \tag{4}$$

It is of interest here to estimate the indispensible positive coefficients k and η, and to decide from data whether there is any integral action.

Forces on the Platform

Before signal processing may proceed, it is necessary to establish the relationship of the measurement signal μ to the angular position θ.

A static force equilibrium argument would go as follows: A signal which represents the center of force on the force platform is measured. 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, such a model is not entirely satisfactory for the purposes of dynamic analysis, as the force center and the vertical projection of center of body mass do not generally coincide at the same point. The foot may, for example, exert a corrective force on the platform to initiate an angular acceleration of the body.

As described in the Appendix II, it holds that the measurement μ is related to the torque T_{bal} for a certain body mass m so that

$$\mu(t) = \frac{2\gamma}{a + b} T_{\text{bal}}(t) + \gamma \frac{b - a}{a + b} mg \tag{5}$$

for positions a and b with a gain factor γ. This means that the measurement μ represents the ankle torque T_{bal} except for a gain factor and a bias term. It is part of signal processing to compensate for the gain factor and the bias term in the recorded measurements.

A Dynamic Response Classification

We have given one interpretation of the coefficients in terms of a mechanical model with a spring effect k and a dashpot effect η. Naturally, a more rapid reflex system requires a balanced increase both of spring action and damping action. It is therefore desirable to quantify mutually independent characteristics of motion. Normalization of the transfer function (4) with respect to frequency gives for the stimulus dependence

$$T_{\mathrm{bal}}(s) = \frac{(b_1 + b_2)\left(\left(\frac{s}{\omega_0}\right)^3 - \frac{g}{l\omega_0}\left(\frac{s}{\omega_0}\right)^2\right)}{\left(\frac{s}{\omega_0}\right)^3 + \frac{\eta}{\omega_0}\left(\frac{s}{\omega_0}\right)^2 + \frac{k}{\omega_0^2}\left(\frac{s}{\omega_0}\right) + 1} V(s)$$

$$\omega_0 = \sqrt[3]{\rho}.$$

A more functional characterization of the motion based on the transfer function properties may therefore be formulated using the concepts

- swiftness: $\omega_0 = \sqrt[3]{\varrho}$ (rad/s)
- stiffness: k/ω_0^2
- damping: η/ω_0.

This classification describes the posture dynamics by one swiftness parameter and two stability parameters. The swiftness parameter is a bandwidth (rad/s) and provides information about the highest angular frequency of the disturbance for which the posture control system gives adequate correction. The stiffness and damping are dimensionless stability parameters, independent of posture control swiftness because the dependence on ω_0 is eliminated.

A high value of swiftness means rapid response to disturbance, i.e., rapid compensation for small deviations from equilibrium. A high value of damping means good damping of sway velocity.

Signal Processing

The identification of transfer function from measured data may be done by several methods. Either nonparametric identification [19] or parametric identification, [30, ch. 10], [26, ch. 8, 9], or [21], can be used. The most established method of parametric estimation is based on time series analysis with ARMAX-models fitted by means of linear regression [21], [2]. In the present study, both parametric and nonparametric identification were utilized and two independent program packages were used for interactive identification, namely IDPAC, [34], [14] and Pro-MATLAB [23]. The signal processing were performed in the following order:

Nonparametric Identification

1) Autospectrum of stimulus v (vibration) and response μ (force distribution in direction x).

2) Cross spectrum between v and μ.

3) Coherence between v and μ.

4) Transfer function from v to μ computed from spectra.

Parametric Identification

5) Maximum likelihood identification of an ARMAX-model.

6) Validation by test of residuals:

- changes of signs (χ^2 test),
- autocorrelation (χ^2 test),
- cross correlation between v and residuals (χ^2 test),
- normal distribution of residuals (χ^2 test).

7) Validation by simulation.

8) Translation from ARMAX-model to continuous-time transfer function.

Experiments

Experimental Procedure

A series of experiments was performed with six subjects in order to evaluate the model and the method. The first experiment tested the difference between performance with open and that with closed eyes.

Another set of experiments were performed to test the difference between two choices of stimulation frequency. Finally, a test was made to check the sensitivity of the method by using asymmetric stimulation. Time and frequency domain properties of the stimulus are presented in the results section. The following recordings were made with a sampling interval of 0.04 s, i.e., the sampling frequency 25 Hz.

Experiment A: The empty experiment to measure electronic offsets.

Experiment B: A test sequence with a vibration stimulus of 100 Hz that is switched on and off according to pseudo-random binary sequence (PRBS); the subject standing with open eyes.

Experiment C: A test sequence with a vibration stimulus of 100 Hz that is switched on and off according to a PRBS; the subject standing with closed eyes.

Experiment D: A test sequence with a vibration stimulus of 60 Hz that is switched on and off according to a PRBS; the subject standing with closed eyes.

Experiment E: A test sequence with a vibration stimulus of 100 Hz applied to the right calf only with the subject standing with closed eyes; the purpose being to test the method's sensitivity to asymmetric stimulation.

Results of the experiments

Coherence between the stimulus and the response was tested for the different experiments. (A detailed presentation of calculations and numerical results is given in Appendix II.) It was found that coherence was low for all experiments with open eyes (B) and with asymmetric stimulation (E). Response of frontal sway was also shown to be very low for all subjects. Thus, computations of transfer functions based on such data are not to be recommended.

The results with closed eyes and symmetric stimulation were quite convincing with good coherence between vibration stimulus v and body sway in the saggittal plane. This indicates that there is a reasonable response to vibration in the absence of visual input. The continuous-time pole polynomial (transfer function denominator) of (4) was computed. The third order model pole polynomial

$$A(s) = s^3 + \eta s^2 + ks + \rho$$

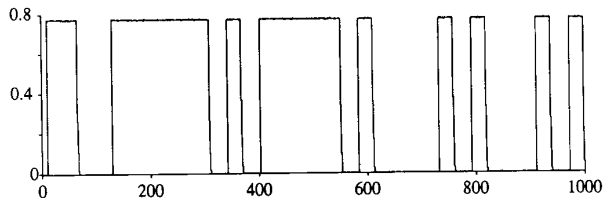

Fig. 2. Experiment D (closed eyes, 60 Hz). Input voltage (10 V) to vibrators versus time. Time scale in units of 0.04 s.

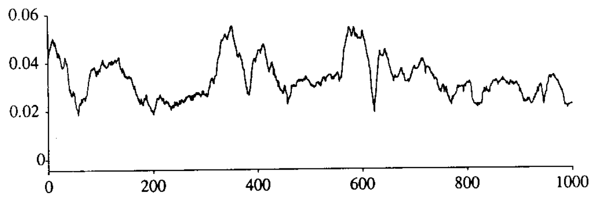

Fig. 3. Experiment D (closed eyes, 60 Hz). Sway (371.7 normal/scale unit) in saggittal plane (x, z) versus time. Time scale in units of 0.04 s.

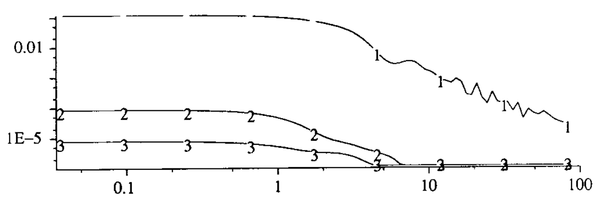

Fig. 4. Experiment D (closed eyes, 60 Hz). Vibrator input voltage power spectrum 1) versus frequency (rad/s). Saggittal sway power spectrum 2) frontal sway power spectrum 3). versus angular frequency (rad/s).

was fitted with data from experiments C and D. The following results were obtained from experiment D (closed eyes, 60 Hz). (See Figs. 2–4.)

Subject	η	k	ρ
Subject 1	6.09	49.25	18.67
Subject 2	4.46	43.99	10.46
Subject 3	3.64	32.15	14.85
Subject 4	2.90	10.44	4.39
Subject 5	6.89	47.79	28.68
Subject 6	4.97	49.45	31.99

These parameters characterize a very well damped regulation system. The dynamic response classification describes posture dynamics by one swiftness parameter and two stability parameters (see above).

Subject	Swiftness	Stiffness	Damping
Subject 1	2.65	7.00	2.29
Subject 2	2.19	9.20	2.04
Subject 3	2.46	5.32	1.48
Subject 4	1.64	3.90	1.77
Subject 5	3.06	5.10	2.25
Subject 6	3.17	4.91	1.57

The results of experiments are listed with comments on good (+) or poor (−) properties of the present approach in estimating ability to maintain posture control. The arguments for these conclusions are given in the previous section and in Appendix II.

+ There is acceptably strong coherence in saggittal plane motion with closed eyes. The power of the oscillation increases by a factor of two, which means that there is a reasonable response to the vibration stimulus.

+ There is weak coherence with open eyes.

+ There is weak coherence to sway in the frontal plane.

+ The data fit very well to a linear model.

+ It is possible to identify the feedback parameters with very good accuracy.

+ The residual signal has a small oscillative component of 0.2–0.3 Hz which may correspond to breathing.

− The method is sensitive to assymmetry in stimulation?

Discussion

Inverted pendulum models in connection with parametric estimation have earlier been formulated by Östlund [27] and by Ishida and Miyazaki [18] where spontaneous sway was analyzed. There are, however, systematic statistical difficulties with analyses of spontaneous motion in closed loop control systems [21], [30]. These problems have been avoided in the present work by exposing the subject to vibration stimulus. To the best of our knowledge, the third-order model (4) with parametric identification for experimental model verification represents a new approach.

The identified coefficients k, η, and ρ of assumption $A6$ represent different aspects of the posture control system. The amplitude of body sway may become large for a small k whereas a large k gives good postural control of the angular position. The parameter η represents the damping of body sway. Too small of an η value means low damping of body sway whereas a large value means rigidity. The parameter ρ represents the automatic reset, i.e., compensative action to eliminate bias in the angular position.

With a combination of the parameters k, η, and ρ, a large variety of body sways patterns can be described. The proportional and derivative actions represented by the parameters k and η are indispensible to maintain stability. The third-order model is statistically validated; it is accurate and explains data well. The strong cross covariance of the estimates of k and ρ, however, constitutes a practical difficulty.

We have given one interpretation of the coefficients in terms of a mechanical model with a spring effect k and a dashpot effect η. The integral component of (3) and assumption $A6$ is responsible for a slow reset action [5], an action that is biologically feasible considering the anatomical and physiological background. Vestibular, visual, and somatosensory information reaches the spinal motoneurons from the vestibular nucleus *via* several vestibulospinal and reticulospinal tracts with or without modulation in the cerebellum [35]. Spinal motoneurons are also influenced by interneurons with information from antagonistic muscles [9]. An induced perturbation changes the visual, vestibular, and somatosensory inputs which affects the spinal motoneurons at different latencies [35].

Naturally, a more rapid reflex system requires a balanced increase both of spring and damping action. It is therefore desirable to quantify mutually independent characteristics of motion. A more functional characterization of the motion based on the transfer function properties may be formulated via normalized parameters by means of the concepts, swiftness, and damping. A high value of swiftness means rapid response to disturbances of equilibrium, and a high value of stiffness means small deviations from equilibrium. A high value of damping means good attenuation of sway velocity.

With the model presented here, the effect of vibration on muscle stretch perception cannot be distinguished from that on rate perception. The use of coherence functions makes it possible to quantify the relative importance of visual feedback vis à vis vestibular and proprioceptive feedback in different frequency ranges.

The choice of suitable experimental conditions for future clinical development is not self-evident although those applied here have proved reasonably satisfactory. The following aspects deserve further consideration: test duration, vibration amplitude (intensity), vibration frequency, vibration pattern, and the possibility of stimulating other muscle groups. A shorter test duration may be preferable for clinical purposes, as might different choices of vibration amplitude and frequency. Eklund [6]–[8] found that the effects of induced body sway could be elicited with a range of vibration amplitudes and frequencies (20–160 Hz). The amplitude must be chosen so that the vibration stimulus does not induce any excessive sway or falling reactions. Thus, application of the method is naturally limited to patients who are able to tolerate additional loading of their postural control.

Other choices of frequency and amplitude may give different but statistically acceptable results for each testing condition (Fig. 5), although standardization of test conditions is, of course, necessary to permit comparison of results.

The stimulus in the present experiments is produced by vibration of the calf muscles, sway being recorded in the saggittal and frontal planes. The vibration caused only insignificant motion in the frontal plane. Vibration to other muscles may provide tests which induce two-dimensional motion. The choice of a pseudorandom vibration pattern with a flat power spectrum is not dictated by methodological considerations, although there is a certain advantage in having a stimulus that is unpredictable by the subject.

A future clinical application of the present approach is patients where defect postural control may be suspected. A large test material of normals and subjects with well-defined lesions has to be analyzed to determine the reliability, sensitivity, and discriminatory power of the parameters. However, the coherence function is sufficiently large for good reproducability to be expected.

Conclusions are only drawn from the coefficient values of the denominator polynomial, which means that attention is focused on effects of recovery from a perturbation, rather than the onset of perturbation. This is important because the stimulus intensity may vary and there may be substantial interindividual variation in the primary effect of perturbations. The parameters of swiftness, stiffness, and damping presented here may therefore prove useful for interindividual comparison both in clinical practice and in research.

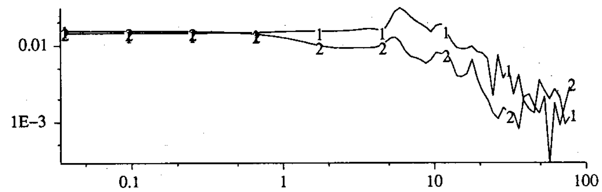

Fig. 5. Comparison between transfer function from experiments with different vibrating stimuli; gain graphs of experiments C (closed eyes, 100 Hz) 1) and D (closed eyes, 60 Hz) 2) versus angular frequency (rad/s).

Conclusions

A postural test involving a force platform has been analyzed quantitatively by means of a new method. The proposed model-oriented transfer function approach also allows angular position θ (or displacement of the body center of gravity), as well as sway velocity to be computed from the measurements recorded with the force platform. Parameters to quantify the body's ability to maintain posture have been proposed and the following conclusions are made:

- The ankle torque T_{bai} represents the body's feedback control to maintain stability. It is emphasized that the force platform measurement may best be understood as the feedback actuated by the body.
- A quantitative analysis of the feedback properties of posture control is made. The control action is analyzed with classical control concepts. It is shown that there is corrective action with respect to angular position θ, angular velocity $\dot{\theta}$, and a slow reset control of bias in θ.
- The results of computation show that the proposed quantifiers of posture k, η, and ρ may be estimated with good accuracy according to generally accepted statistical validation criteria.
- The model complexity is chosen as a linear system of order three, which is sufficient to explain the outcome of measurements.
- The method is sensitive to symmetry of stimulation.
- The proposed model is compatible with earlier attempts to represent measurements of the posture dynamics by spectral analysis [27]. Spectral analysis supported by parametric identification is advantageous because it allows quantitative statistical analysis as well as physiological interpretation.
- The approach with parametric identification of a transfer function between stimulus and response can be made with higher confidence than can parametric analysis of spontaneous motion. The coherence function gives a

measure of the dependence of the response on variations in the stimulus.

Appendix I
Transfer Function

The concept of transfer function adopted here is that of Laplace transforms as used in control theory [15, p. 27]. Functions in the variable t indicate time domain functions, and functions in variable s and/or capital letters denote frequency domain functions. The torque equilibrium of (2) and assumption $A6$ give the two equations:

$$J\frac{d^2\theta}{dt^2} = mgl\sin\theta(t) + T_{\text{bal}}(t) + T_d(t) \tag{2}$$

$$T_{\text{bal}}(t) = -mgl\sin\theta(t) - kJ\theta(t) - \eta J\dot{\theta}(t) - \rho J\int_{t0}^{t}\theta(t)\,dt. \tag{A1.1}$$

Elimination of T_{bal} gives

$$J\frac{d^2\theta}{dt^2} = -\eta J\frac{d\theta}{dt} - kJ\theta(t) - \rho J\int_{t0}^{t}\theta(\tau)\,d\tau + T_d(t). \tag{A1.2}$$

There are three states that affect motion, namely angular velocity $d\theta/dt$, angular position θ, and the bias compensation. A Laplace transformation and algebraic simplification gives the transfer function

$$\theta(s) = \frac{\frac{1}{J}s}{s^3 + \eta s^2 + ks + \rho}T_d(s). \tag{A1.3}$$

With a vibration stimulus v, according to assumption $A7$ there is one more transfer function, namely that from stimulus v to θ

$$\theta(s) = \frac{\frac{1}{J}(b_1 + b_2)s}{s^3 + \eta s^2 + ks + \rho}V(s) + \frac{\frac{1}{J}s}{s^3 + \eta s^2 + ks + \rho}T_d(s). \tag{A1.4}$$

A reduced model without any integrating compensation ($\rho = 0$) gives the simplification

$$\theta(s) = \frac{\frac{1}{J}(b_1 + b_2)}{s^2 + \eta s + k}V(s) + \frac{\frac{1}{J}}{s^2 + \eta s + k}T_d(s). \tag{A1.5}$$

A transfer function from vibration stimulus V and disturbance T_d to the torque T_{bal} is found via (2), (A1.4) for linearized motion [15] around the equilibrium $\theta = 0$ where $\sin\theta \approx \theta$ and

$$\begin{aligned}T_{\text{bal}}(s) &\approx (Js^2 - mgl)\,\theta(s) - T_d(s)\\ &= \frac{(b_1 + b_2)\left(s^3 - \frac{g}{l}s^2\right)}{s^3 + \eta s^2 + ks + \rho}V(s) - \frac{\eta s^2 + \left(k + \frac{g}{l}\right)s + \rho}{s^3 + \eta s^2 + ks + \rho}T_d(s).\end{aligned} \tag{A1.7}$$

It is of interest here to estimate the indispensible positive coefficients k and η, and to decide from data if there is any integral action.

Appendix II
The Ankle Torque

The distances a and b in Fig. 6 denote horizontal distances from the ankle point of rotation to each one of the support points at the edges of the force plate. Let F_{foot} denote the pressure of the soles exerted on the force plate and Ω denote the area of contact between the feet and the force platform.

The forces F_a and F_b in Fig. 7 represent the support forces at the edges of the force plate. The measurements μ are force differences given by

$$\mu = \gamma(F_a - F_b) \tag{A2.1}$$

with γ as a gain factor due to strain gauges and the electronics.

Force Equilibrium

On the body,

$$\iint_\Omega F_{\text{foot}}(x, y)\,dx\,dy = mg. \tag{A2.2}$$

On the force plate,

$$\iint_\Omega F_{\text{foot}}(x, y)\,dx\,dy = F_a + F_b. \tag{A2.3}$$

Torque Equilibria

The force plate equilibrium is

$$\iint_\Omega F_{\text{foot}}(x, y)x\,dx\,dy = T_y = T_{\text{bal}} \tag{A2.4}$$

$$\iint_\Omega F_{\text{foot}}(x, y)\,y\,dx\,dy = Tx. \tag{A2.5}$$

The forces F_a and F_b act on the distances a and b from the origin, with the resulting torque

$$-F_a a + F_b b + T_{\text{bal}} = 0. \tag{A2.6}$$

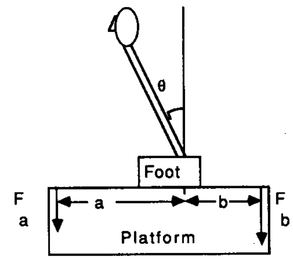

Fig. 6. Anterior force F_a and posterior force F_b on the force plate.

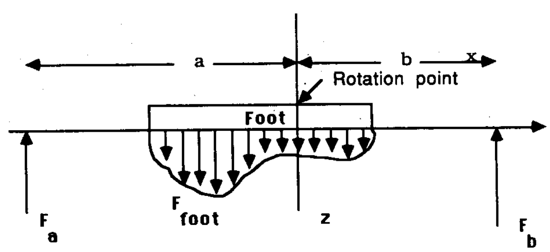

Fig. 7. Sole pressure F_{foot} on the area Ω of the force plate in the x, z plane.

The ankle torque equilibrium results in body sway given by

$$J \frac{d^2}{dt^2} \theta(t) = mgl \sin \theta(t) + T_{\text{bal}}(t) \qquad J = ml^2. \tag{A2.7}$$

Relationship between Measurement μ and T_y

From (A2.1) and (A2.6), we find for body mass m (kg) that

$$F_a + F_b = mg \qquad \gamma(F_a - F_b) = \mu. \tag{A2.8}$$

Solving these equations with respect to F_a and F_b gives

$$F_a = \frac{1}{2} mg + \frac{1}{2\gamma} \mu \qquad F_b = \frac{1}{2} mg - \frac{1}{2\gamma} \mu. \tag{A2.9}$$

With F_a and F_b it is possible to express the torque T_{bal} as

$$T_{\text{bal}} = aF_b - bF_a = \frac{a-b}{2} mg + \frac{a+b}{2\gamma} \mu. \tag{A2.10}$$

Solving for μ shows that μ represents T_{bal} via the linear relation

$$\mu = \frac{2\gamma}{a+b} T_{\text{bal}} + \gamma mg \frac{b-a}{a+b}. \tag{A2.11}$$

Calibration experiments give the values $a + b = 0.327$ m and $\gamma = 0.044$ V/N.

Appendix III
Calculations and Analysis

The results of computation are presented in this section together with certain conclusions which are also presented in a more compact form in the section of "Results." The presentation essentially follows the order of computation (items 1–8) given in the section on signal processing. Capital letters (A–E) refer to experiments presented above. Graphs of recordings are given in electrical units (10 V) versus sampling instant. Time axis is given in units of sampling time (0.04 s).

Graphical Presentation of Experimental Results

The following two graphs (Figs. 8 and 9) show the result of experiment B where the subject keeps the eyes open.

Experimental results with closed eyes are shown in Figs. 10–12.

The result of sway in the saggittal plane when one vibrator asymmetrically stimulates the right calf is presented in Fig. 13. The response to vibration stimulus is of lower magnitude than in experiment C.

Autospectra

The autospectra (power spectra) show the frequency contents of the signals investigated (see Fig. 4). Notice that the spectrum should not be interpreted as the vibration frequency. Frequency units are given as rad/s. Divide by 2π to obtain units in c/s or Hz.

Coherence Analysis

Correlation analysis made for each frequency is called a coherence spectrum [2]. A large absolute value close to one indicates that the input v and the output z are correlated. A coherence of 0.5 denotes that half of the output variation may be explained by variations in the stimulus input. It may be concluded from the first graph below that the coherence is quite satisfactory. The coherence is better for sway in the saggittal plane than for sway in the frontal plane, which is understandable as the vibrators are mounted on both calves to stimulate saggittal motion.

It is seen from Fig. 16 that the experimental result contains little information for asymmetrical stimulation. Notice also in Fig. 14 that the coherence is somewhat lower for frequencies below 0.3 rad/s $\approx$ 0.05 Hz.

Transfer Functions from Spectra

Division of the cross spectrum between input v and output x by autospectrum of v gives the transfer function, i.e., gain and phase lag for a range of frequencies (see Fig. 5).

There is some evidence that the muscle spindles react in different ways to different vibration frequencies [31], although this is not a predominant feature in Fig. 5. Different numerical results, however, may be expected.

Estimation of the delay time T_d in the feedback loop is possible by checking the phase lag for high frequencies (Fig. 17). For high-angular frequencies ω it holds that the phase lag is ωT_d. From Fig. 18, we find that with $\omega = 50$ rad/s

$$T_d \approx \frac{\pi}{180} \frac{600}{50} = 0.20 \text{ s}.$$

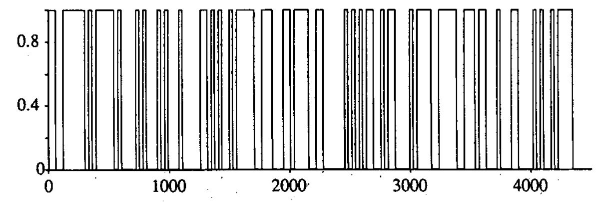

Fig. 8. Experiment B (open eyes, 100 Hz). Input voltage (10 V) to vibrators versus time. Time scale in units of 0.04 s.

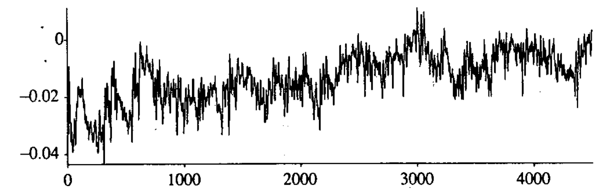

Fig. 9. Experiment B (open eyes, 100 Hz). Sway 371.7 normal/scale unit) in saggittal plane (x, z) versus time. Time scale in units of 0.04 s.

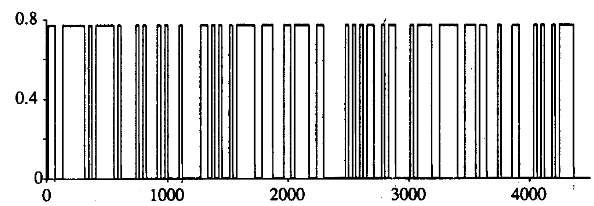

Fig. 10. Experiment D (closed eyes, 60 Hz). Input voltage (10 V) to vibrators versus time. Time scale in units of 0.04 s.

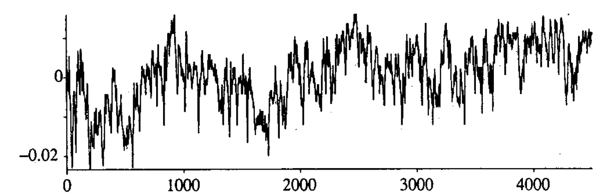

Fig. 11. Experiment D (closed eyes, 60 Hz). Sway 371.7 normal/scale unit) in saggittal plane x, z versus time. Time scale in units of 0.04 s.

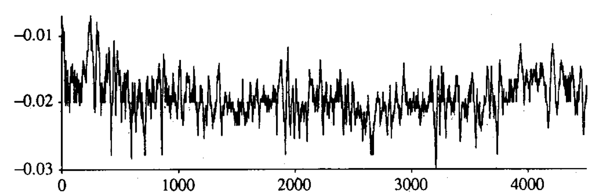

Fig. 12. Experiment D (closed eyes, 60 Hz). Sway (371.7 normal/scale unit) in frontal plane (y, z) versus time. Time scale in units of 0.04 s.

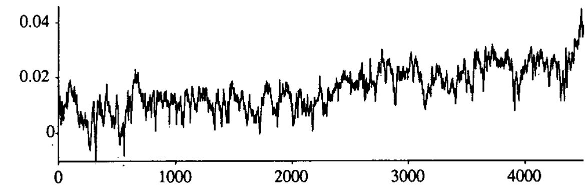

Fig. 13. Experiment E (asymmetric, 100 Hz). Sway (371.7 normal/scale unit) in saggittal plane (x, z) versus time. Time scale in units of 0.04 s.

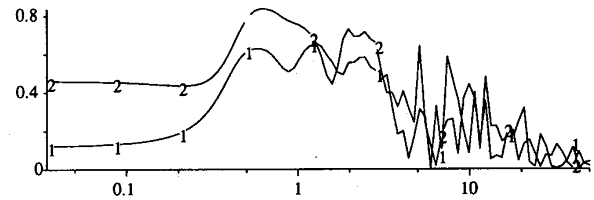

Fig. 14. Experiment B (open eyes, 100 Hz). Coherence between sway in saggittal plane and vibration. 1) Experiment D (closed eyes, 60 Hz). Coherence between sway in the saggittal plane and vibration 2) versus angular frequency (rad/s).

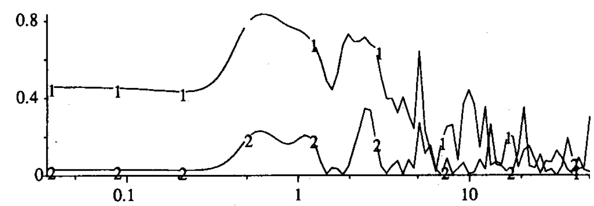

Fig. 15. Experiment D (closed eyes, 60 Hz). Coherence between vibration and sway in the saggittal plane 1) and frontal plane 2) versus angular frequency (rad/s).

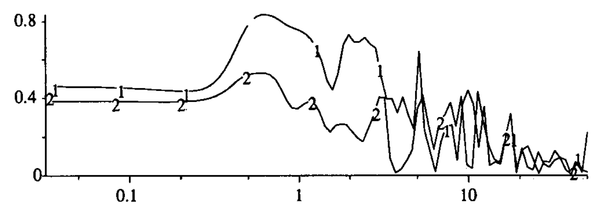

Fig. 16. Experiment D (closed eyes, 60 Hz). Coherence of sway in the saggittal plane with symmetric stimulation 1). Experiment E (asymmetric, 100 Hz). Coherence of sway in the saggittal plane with asymmetric stimulation 2) versus angular frequency rad/s.

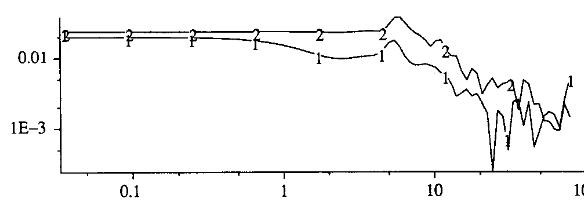

Fig. 17. Transfer function from spectra of experiment B (open eyes, 100 Hz) 1) and experiment C (closed eyes, 100 Hz), 2) gain graph.

This value should be compared to other measures of the time required for a signal to complete a roundtrip in the neurological circuit.

Maximum Likelihood Identification

The time delay was estimated to $T_d \approx 0.20$ s and it is therefore desirable to estimate model parameters as sampling rates of this order of magnitude. The following ARMAX-models all have a sampling interval of 0.20 s which is obtained by extraction of every fifth sample from original time series.

Parameter identification with estimation of initial val-

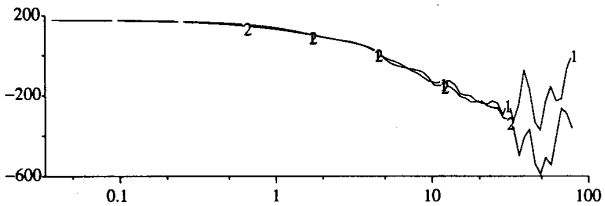

Fig. 18. Transfer function from spectra of experiment B (open eyes, 100 Hz) 1) and experiment C (closed eyes, 100 Hz), 2) phase graph.

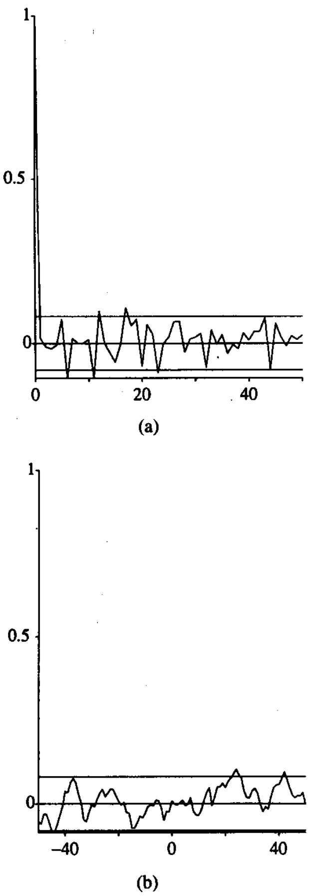

Fig. 19. Test of autocorrelation of residual (left) and cross correlation between stimulus v and the residual (right) of a third order of ARMAX-model fitted to data of D (closed eyes, 60 Hz). Time scale in units of 0.2 s. Confidence interval (95 percent) is displayed.

ues is done for model orders two, three, and four. Statistical tests are satisfied for orders three and four but not for the second order model. The Akaike test criterion (AIC) does not change considerably. The model order of choice is therefore a third order model. Results of a MATLAB output for a third order ARMAX model fitted to saggittal sway data of subject 6 are given below.

present(th6)

This matrix was created by the command ARMAX

Loss fcn: 7.6109e-06 Akaike's FPE: 7.8427e-06

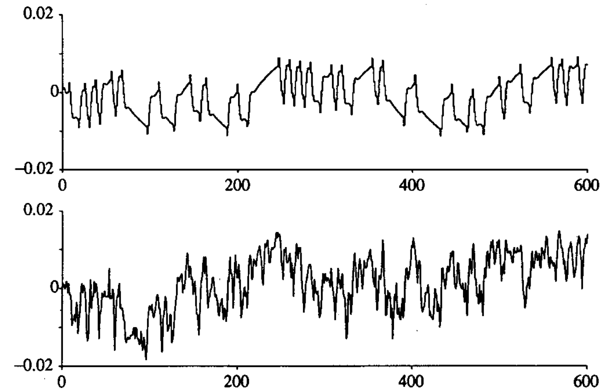

Fig. 20. Saggittal sway of model with input of experiment D (closed eyes, 60 Hz). Simulated output from third order estimated model (upper). Saggittal away of experiment D. Real output (lower). Time scale in units of 0.2 s.

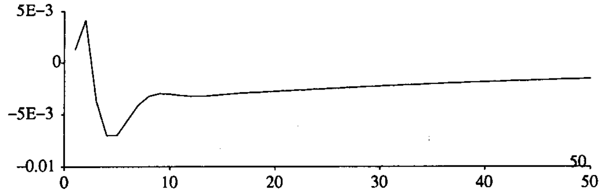

Fig. 21. Saggittal sway. Simulated impulse response of force on the plate from third order estimated model. Time scale in units of 0.2 s.

The polynomial coefficients and their standard deviations are

$$B =$$
$$\begin{array}{rrr} 0.0003 & 0.0013 & -0.0048 \\ 0.0005 & 0.0009 & 0.0007 \end{array}$$
$$A =$$
$$\begin{array}{rrrr} 1.0000 & -1.2265 & 0.7342 & -0.3702 \\ 0 & 0.1060 & 0.1544 & 0.0728 \end{array}$$
$$C =$$
$$\begin{array}{rrrr} 1.0000 & -0.3807 & 0.2742 & -0.0946 \\ 0 & 0.1130 & 0.0926 & 0.0488. \end{array}$$

Validation by Test of Residuals

The purpose of residual tests is to find remaining correlations which indicate whether the model order is adequate. With an adequate model order, the only residual noise is white noise. The residual χ^2 tests for a third order model with 600 data points gives *significant** (95 percent confidence) validation with respect to changes of sign, independence of residuals, normality, and independence between residuals and input.

Validation by Simulation

Real and simulated data have been compared, using the vibration signal as a deterministic input. We studied to what extent the experiment data are explained by the deterministic input–output behavior of the estimated model. The input of D is input to the estimated model.

Conversion to Continuous-Time Parameters

For the third order model, we have estimated an ARMAX pole polynomial

$$A(q) = q^3 + a_1 q^2 + a_2 q + a_3$$

with the following results of parameter values and standard deviations for subject 6:

$$a_1 = -1.227 \pm 0.106 \quad a_2 = 0.734 \pm 0.154$$

$$a_3 = -0.370 \pm 0.073.$$

Statistical multivariate analysis with respect to the coefficients of $A(q)$ presents no problems because the covariance matrix is given in the IDPAC or MATLAB calculations. Conversion to continuous-time parameters requires inverse sampling [1, p. 39] with

$$\frac{1}{h} \log \Phi$$

with a sampling interval h and a matrix Φ. Computation of the continuous-time characteristic polynomial gives

$$A(s) = s^3 + 4.97s^2 + 49.4s + 32.0.$$

This formulation allows identification of the physiological feedback parameters. The third order model pole polynomial of (7)

$$A(s) = s^3 + \eta s^2 + ks + \rho.$$

The coefficients of $A(s)$ determine the postural behavior. Parameters are already normalized in the model with respect to body weight m and body height l, in terms of the moment of inertia J. Results for the test group were as follows:

C: Stimulation frequency 100 Hz; Closed eyes.

Subject	η	k	ϱ
Subject 1	5.24	44.26	45.64
Subject 2	5.18	26.11	14.32
Subject 3	1.37	19.26	1.75
Subject 4	6.00	33.16	26.15
Subject 5	4.56	26.35	17.13
Subject 6	6.53	60.53	99.01

D: Stimulation frequency 60 Hz; Closed eyes.

Subject	η	k	ϱ
Subject 1	6.09	49.25	18.67
Subject 2	4.46	43.99	10.46
Subject 3	3.64	32.15	14.85
Subject 4	2.90	10.44	4.39
Subject 5	6.89	47.79	28.68
Subject 6	4.97	49.45	31.99

E: Asymmetric stimulation 100 Hz; Closed eyes.

Subject	η	k	ϱ
Subject 1	2.91	23.52	10.92
Subject 2	1.85	62.90	10.46
Subject 3	3.67	10.20	1.35
Subject 4	9.18	22.50	6.72
Subject 5	13.64	39.93	15.02
Subject 6	5.87	58.49	12.26

We have given one interpretation of the coefficients in terms of a mechanical model with a spring effect (k) and a dashpot component (η). The more functional characterization of the motion based on the dynamic response is formulated by the concepts

- Swiftness: $\sqrt[3]{\varrho}$
- Stiffness: $k/(\sqrt[3]{\varrho})^2$
- Damping: $\eta/\sqrt[3]{\varrho}$.

This classification describes the postural dynamics by one swiftness parameter and two stability parameters. The swiftness parameter is a bandwidth (rad/s) and provides information about the highest angular frequency of disturbance for which the posture control system gives adequate correction. The stiffness and damping are stability parameters independent of the swiftness in posture control. The results are as follows:

C: Stimulation frequency 100 Hz; Closed eyes.

Subject	Swiftness	Stiffness	Damping
Subject 1	3.57	3.47	1.47
Subject 2	2.43	4.43	2.13
Subject 3	1.21	13.3	1.14
Subject 4	2.97	3.76	2.02
Subject 5	2.58	3.97	1.77
Subject 6	4.63	2.83	1.41

D: Stimulation frequency 60 Hz; Closed eyes:

Subject	Swiftness	Stiffness	Damping
Subject 1	2.65	7.00	2.29
Subject 2	2.19	9.20	2.04
Subject 3	2.46	5.32	1.48
Subject 4	1.64	3.90	1.77
Subject 5	3.06	5.10	2.25
Subject 6	3.17	4.91	1.57

A high value of swiftness means rapid response to disturbance, and a high value of stability means small deviations from equilibrium.

Acknowledgment

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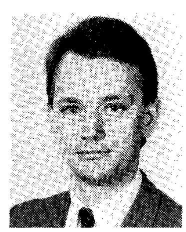

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