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It is shown that the identification fulfils ordinary statistical validation criteria. Furthermore it is conjectured that the identified state feedback parameters may be utilized to estimate the ability to maintain posture.

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IDENTIFICATION

OF

HUMAN POSTURE DYNAMICS

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ABSTRACT

The present study contains an analysis of measurements related to human posture dynamics. The control of posture was investigated by recording vibration induced body sway with a force platform in six normal subjects. Vibrators attached to the calf muscles caused perturbations and the stabilizing forces of the feet actuated on a square platform were measured with strain gauges technique. The body mechanics is modelled as a stabilized inverted pendulum with all complex muscular activity resulting in stabilizing forces actuated by the feet. Control with state feedback of body sway and position was considered. Signal processing was made with the approach of parametric identification of a transfer function representing the stabilized inverted pendulum. The control of posture could be quantified by three parameters.

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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 produce deviated postural control and equilibrium. It is therefore of interest to access the ability of postural control by measuring the displacement of the body center of gravity. Registration of amplitude and period of spontaneous oscillations around the equilibrium position may describe the sway and thus the control of posture.

Normally, spontaneous oscillation appears in healthy individuals during stance and the oscillating behaviour 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 the response to an external disturbance in the presence of spontaneous motion. Therefore, it is motivated to analyse the spectrum of oscillations rather than only particular frequencies. To understand the biological correlates of the posture control parameters it is also desirable to make a model based analysis of the control system.

Several dynamic analyses of spontaneous body sway have been reported [11], [13], [19]. The experimental set-up in [11] involved measurements from a force platform and light emitting diodes to detect position. The feedback characteristics were identified in closed loop control without measured external disturbances. Different external disturbances such as movable platforms and visual stimulus [23], [24] have been introduced to study dynamic aspects and improve the identifiability of posture control. A movable platform, however, will inflict a simultaneous disturbance on both proprioceptive and vestibular feedbacks.

In the present study we develop a model for body posture control when loaded with misinforming proprioceptive feedback. The stimulus is produced by vibration of the calf muscles which results in activation of muscle spindles and hence a misleading sensory information [14]. Measurements of body sway are made with a force platform. The body is modelled as an inverted pendulum with a servo mechanism model of balance. The spectral analysis is made compatible with a dynamic systems approach of signal processing and control theory. Transient input-output analysis is made via Laplace transform methods. Estimation of parameters is made by maximum likelihood estimation of coefficients in ARMAX-models. A statistical analysis of the model goodness and the parameter uncertainty is made. The aim of the present study is to identify feedback parameters useful for evaluation of the ability of postural control.

METHODS

Material

Tests were done on three male and three female (mean age 28, 23 - 39 years) naive subjects. No one had any history of vertigo, central nervous or ear diseases and no one had suffered from injury of the lower extremities. At the recording no subject was on any form of medication or had consumed alcoholic beverages for at least 48 hours.

Equipment and experimental setup

The equipment consists of a square force platform and a computer for registration and computation. The force platform has been developed at Institute of Occupational Health, Helsinki, Finland, and the ENT Clinic of Lund, Sweden, see Pyykko et al [20]. The platform is equipped with strain gauges to measure vertical force in each one the four corners at four symmetrically located points. Measurements μ obtained from the strain gauges are recorded by the computer. The measurements represent the differential force distributions exerted by the feet on the platform. The equipment allows simultaneous registration of body sway in the saggittal plane and in the frontal plane.

During measurements the subject is told to stand with heels together on the platform while staring at a spot on the opposite wall. A small vibrator is attached to each one of the calf muscles with elastic straps. The subject is told to stand with either closed or open eyes and the recording starts. The reaction to vibratory stimulation is recorded after initial registration of the spontaneous body sway. The vibrators are turned on/off and modulated pseudorandomly (PRBS) according to a program executed in the computer. The frequency of the vibrators depends roughly linearly on the input voltage v which had been confirmed for all vibrators used. It was verified experimentally as a part of standard laboratory practise that there is no interference (aliasing) between the sampling frequency and the vibration frequency. The test sequence takes 150 $[s]$.

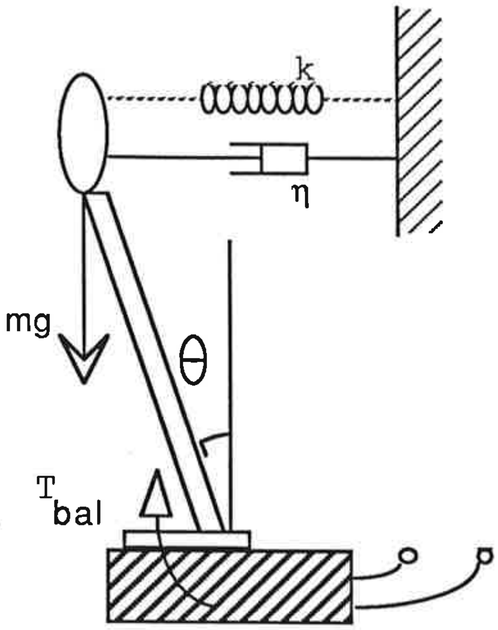

Figure 1. Inverted pendulum model of human posture dynamics (a) without balance and (b) with balancing torque T_{bal} with properties similar to that achieved of a spring (k) and a dashpot (η).

MODELLING OF THE POSTURE CONTROL SYSTEM

When exposed to a saggittal perturbation a subject may regain equilibrium by two different strategies of body movement, see Golliday [5]. In 'ankle strategy' muscular forces rotate the body around the ankle joint [5]. In 'hip strategy' there is flexion in the hip and in the knees. The 'ankle strategy' is sufficient for response to the small perturbations which occur during natural stance, see Nashner [17], and fits the modelling of posture control as an inverted pendulum. Hip strategy has to be employed when the projection of the body center of mass falls in front of the feet of the subject. In hip strategy there is a potential problem with shear forces against the supporting surface. The platform is, however, 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 can be argued that in large compensatory movements, preventing an immediate fall of the subject, the inverted pendulum model may be insufficient. However, in natural stance and in the small perturbations inflicted by the vibratory stimulus used, the inverted pendulum model is well in order to explain the corrective movements used to control body posture.

The model is formulated for dynamics in the saggittal plane with the body modeled as an inverted pendulum. The inverted pendulum has an unstable equilibrium point at $\theta = 0$, see figure 1. This means that active stabilizing forces must compensate for deviations in the position in order to maintain posture. The balancing forces are executed by all body muscles in a complex cooperation. A model of the balance as a servo mechanism need however not be more complicated than what is motivated by the resulting behavior as seen in the measurements.

The model consists of an inverted pendulum to explain the pure body mechanics and a control system which acts like a shock absorber of a vehicle. The suspension is characterized by a spring constant k and a damping η which keeps the body in upright position with ability to counteract a disturbance. The response to an impetus is determined by the values of k and η as well as body weight m and the body center of mass on distance l from the ground.

Assumptions

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

Assumption 1: The body is stiff with mass m $[kg]$.

Assumption 2: The body center of mass is located at distance l $[m]$ from the ground.

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. 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 mathematically as well as intuitively that there is no stable equilibrium at $\theta = 0$. A person who does not counteract the torques of gravity will implacably fall in the absence of a stabilizing action. The following two assumptions are introduced to model balance action and influence of disturbances from the environment.

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

Assumption 5: Assume that there is a disturbance torque T_d from the environment

The torque balance now has the form

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

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: $-mgl\sin\theta(t) - kJ\theta(t)$

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

I: $-\rho J\int_{t_0}^{t}\theta(t)dt$

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

The components (P,D) with $k, \eta > 0$ are indispensible for stability according to the Routh criterion of stability, [7]. The integral component (I) accounts for (slow) compensation of bias in θ. It is not a priori necessary for stability and it is therefore a part of the experiment to show its presence.

The parameter k may be interpreted as a spring constant and η is similar to a viscous damping. 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 influence of the vibrating stimulus.

Assumption 7: The vibration v enters the stabilizing system as a 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: $-mgl\sin\theta(t) - kJ\theta(t) + b_1v(t)$

D: $-\eta J\dot{\theta}(t) + b_2v(t)$

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

Transfer function of θ

In this section Laplace transforms and transfer function notions are used as known from control theory, see Hahn [7] (p. 27 ff.). Functions in the variable t indicate time domain functions and functions in variable s and/or capital letters denotes frequency domain functions.

The torque equilibrium of (2) and assumption $A6$ give the two equations

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

$$T_{bal}(t) = -mgl\sin\theta(t) - kJ\theta(t) - \eta J\dot{\theta}(t) - \rho J\int_{t_0}^{t}\theta(t)dt \tag{3}$$

Elimination of T_{bal} gives

$$J\frac{d^2\theta}{dt^2} = -\eta J\frac{d\theta}{dt} - kJ\theta(t) - \rho J\int_{t_0}^{t}\theta(\tau)d\tau + T_d(t) \tag{4}$$

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

$$\theta(s) = \frac{\frac{1}{J}s}{s^3 + \eta s^2 + ks + \rho}T_d(s) \tag{5}$$

With a vibration stimulus v according to assumption 7 there is one more transfer function namely that from stimulus v to θ

$$\theta(s) = \frac{\frac{1}{J}(b_1 + b_2)s}{s^3 + \eta s^2 + ks + \rho}V(s) + \frac{\frac{1}{J}s}{s^3 + \eta s^2 + ks + \rho}T_d(s) \tag{6}$$

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

$$\theta(s) = \frac{\frac{1}{J}(b_1 + b_2)}{s^2 + \eta s + k}V(s) + \frac{\frac{1}{J}}{s^2 + \eta s + k}T_d(s) \tag{7}$$

A transfer function from vibration stimulus V and disturbance T_d to the torque T_{bal} is found via (2), (6) for linearized motion around the equilibrium $\theta = 0$ where $\sin\theta \approx \theta$

$$T_{bal}(s) \approx (Js^2 - mgl)\theta(s) - T_d(s) = \tag{8}$$

$$= \frac{(b_1 + b_2)(s^3 - \frac{g}{l}s^2)}{s^3 + \eta s^2 + ks + \rho}V(s) - \frac{\eta s^2 + (k + \frac{g}{l})s + \rho}{s^3 + \eta s^2 + ks + \rho}T_d(s) \tag{9}$$

It is here of interest to estimate the indispensible positive coefficients k and η and to decide from data if there is any integral action or not.

Forces on the platform

Before making signal processing it is necessary to relate 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.

Such a model is not quite satisfactory for purposes of dynamic analysis. The force center and the projection of center of boy mass are not at the same point in general. The foot may e.g. exert a corrective force on the platform to initiate an angular acceleration of the body.

As described in the appendix it holds that the measurement μ is related to the torque T_{bal} so that

$$\mu(t) = \frac{2\gamma}{a+b} T_{bal}(t) + \gamma \frac{b-a}{a+b} mg \tag{10}$$

for positions a, b and 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.

A dynamic response classification

We have given one interpretation of the coefficients in terms of a mechanical model with a spring k and a viscosity η. Naturally, a more rapid reflex system requires a balanced increase of both spring action and viscosity action. It is therefore desirable to quantify mutually independent characteristics of motion. Normalization of the transfer function (9) with respect to frequency gives for the stimulus dependence

$$T_{bal}(s) = \frac{(b_1+b_2)((\frac{s}{\omega_0})^3 - \frac{g}{l\omega_0}(\frac{s}{\omega_0})^2)}{(\frac{s}{\omega_0})^3 + \frac{\eta}{\omega_0}(\frac{s}{\omega_0})^2 + \frac{k}{\omega_0^2}(\frac{s}{\omega_0}) + 1} V(s) \qquad \omega_0 = \sqrt[3]{\rho} \tag{11}$$

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

- **Swiftness:** $\omega_0 = \sqrt[3]{\varrho} \qquad [rad/s]$
- **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 informs 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 the posture control swiftness because the dependence on ω_0 is eliminated.

A high value of swiftness means rapid response to disturbances and a high value of stiffness means rapid correction to 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. Non-parametric identification [12] as well as parametric identification, [21] (ch.10) or [18] (ch.8-9), can be used. The most established method of parametric estimation is based on time series analysis with ARMAX-models fitted by 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, [25], [7] and Pro-MATLAB [15].

The signal processing were performed in the following order:

Non-parametric identification

1: Autospectrum of
- Stimulus v (vibration)
- 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 ($\chi^2 -$test)
- Normal distribution of residuals

7: Validation by simulation

8: Translation from ARMAX-model to continuous-time transfer function

EXPERIMENTS

Experimental Procedure

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

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 to non-symmetric stimulation.

The following recordings were made with sampling interval 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/off according to pseudo random binary sequence and the subject standing with open eyes.

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

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

Experiment E: A test with a vibration stimulus of 100 Hz of only the right calf and the subject standing with closed eyes. The purpose was to test the sensitivity to non-symmetric stimulation.

Results of the experiments

A detailed presentation of calculations and numerical results is given in appendix 2.

The coherence between stimulus and response was tested for the different experiments. It was found that the coherence was low for all experiments with open eyes (B :) and assymmetric stimulation (E :). The response of frontal sway was also shown to be very low for all subjects. All further computations of transfer functions based on such data must therefore be discouraged.

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.

Computation of the continuous-time pole polynomial (transfer function denominator) of (9) was made. The third order model pole polynomial of (6)

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

was fitted with data from experiments C : and D :. The following results were obtained from experiment D :.

Subject #	η	k	ϱ
Subject #1	6.0853	49.2467	18.6723
Subject #2	4.4556	43.9899	10.4562
Subject #3	3.6408	32.1469	14.8532
Subject #4	2.9045	10.4387	4.3851
Subject #5	6.8936	47.7854	28.6838
Subject #6	4.9686	49.4480	31.9890

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

Subject #	Swiftness	Stiffness	Damping
Subject #1	2.6530	6.9970	2.2938
Subject #2	2.1867	9.1996	2.0376
Subject #3	2.4581	5.3202	1.4811
Subject #4	1.6368	3.8963	1.7745
Subject #5	3.0611	5.0996	2.2520
Subject #6	3.1744	4.9070	1.5652

The results of experiments are listed with comments on good (+) or bad (−) properties of the present approach in estimating the ability to maintain posture control. The arguments for these conclusions are given in the previous section and in appendix 2.

+	There is acceptably strong coherence in direction x with closed eyes. The power of the oscillation increases by a factor two. This 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

The identified coefficients k, η, and ρ of assumption 6 represent different aspects of the posture control system. The amplitude of body sway may become large for small k while a large k gives good postural control. The parameter η represents the damping of body sway. A too small value η means low damping of body sway while a large value means rigidity. The parameter ρ represents the automatic reset i.e. a compensative action to eliminate bias in the position.

A combination of the parameters k, η, ρ can describe a large variety of body sways patterns. The proportional and derivative actions represented by the parameters k and η are indispensible to maintain stability. The third order model is statistically validated, is accurate and explains data well. The strong cross covariance of the estimates of k and ρ is however a practical difficulty. We have given one interpretation of the coefficients in terms of a mechanical model with a spring k and a viscosity η. Naturally, a more rapid reflex system requires a balanced increase of both spring action and viscosity 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 the concepts swiftness, stiffness 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.

The presented model cannot separate between vibration influence on the stretch or rate perceptions, respectively. The use of coherence functions makes it possible to quantify the relative importance of visual feedback compared to vestibular and proprioceptive feedback in different frequency ranges.

Conclusions are only drawn from the coefficient values of the denominator polynomial. This means that attention is focused on effects of recovery from a perturbation rather than the onset of perturbations. This is important because there may be large interindividual variations in the primary effect of perturbations. The presented parameters of swiftness, stiffness, and damping may therefore proove valuable for interindividual comparison in clinical practice and research.

CONCLUSIONS

Quantitative analysis of a postural test on a force platform has been made by a new method. The proposed model oriented transfer function approach also allows computation of angular position θ or the displacement of the body center of gravity as well as velocity of sway from the measurements of 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_{bal} represents the body's feedback control to maintain stability. It is emphasized that the force platform measurement may be best understood as the feedback control actuated by the body.

- A quantitative analysis of the feedback properties of posture control is made. The control action is analysed 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 presented results of computation show that the proposed quantifiers of posture k, η 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 suggested model is compatible with earlier attempts to represent measurements of the posture dynamics by spectral analysis, see [19]. Analysis of spectra 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 respons can be made with higher confidence than parametric analysis of spontaneous motion. The coherence function gives a measure of the dependence of the respons to variations in the stimulus.

Appendix 1

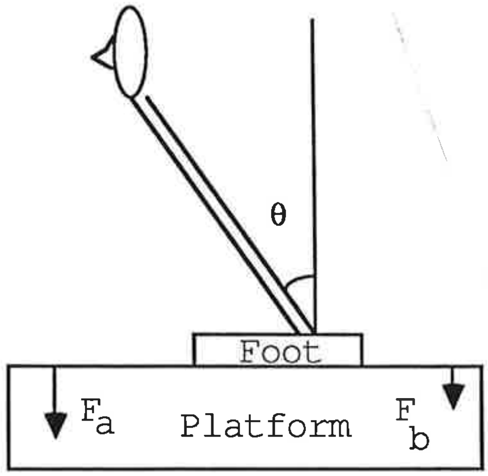

Figure 1. Anterior force F_a and posterior force F_b on the force plate.

The distances a and b denote distances from the ankle to each one of the edges of the force plate. Let F_{foot} denote the pressure of the soles exerted on the force plate. Let furthermore Ω denote the area of contact between the feet and the force plate.

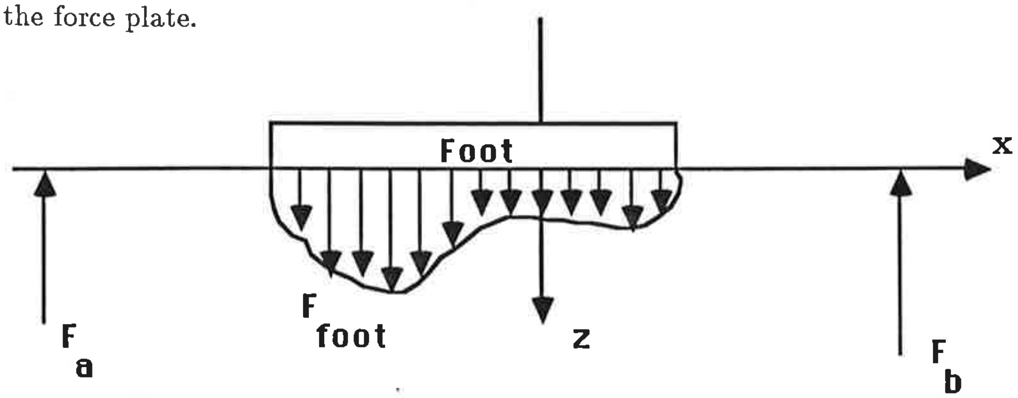

Figure 2. Sole pressure F_{foot} on the area Ω of the force plate in the $xz-$plane.

The forces F_a and F_b represents the support forces at the edges of the force plate. The measurements are force differences given by

$$\mu = \gamma(F_a - F_b) \tag{A1}$$

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

Force equilibrium

On the body

$$\int_\Omega \int F_{foot}(x, y)dxdy = mg \tag{A2}$$

On the force plate

$$\int_\Omega \int F_{foot}(x,y)dxdy = F_a + F_b \tag{A3}$$

Torque equilibria

The force plate equilibrium is

$$\int_\Omega \int F_{foot}(x,y)xdxdy = T_y = T_{bal} \tag{A4}$$

$$\int_\Omega \int F_{foot}(x,y)ydxdy = T_x \tag{A5}$$

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

$$-F_a a + F_b b + T_{bal} = 0 \tag{A6}$$

The ankle torque equilibrium results in body sway given by

$$ml^2 \frac{d^2}{dt^2}\theta(t) = mgl \sin\theta(t) + T_{bal}(t) \tag{A7}$$

Relation between measurement μ and T_y

From (A1) and (A6) we find

$$F_a + F_b = mg \qquad \gamma(F_a - F_b) = \mu \tag{A8}$$

Solving these equations with respect to F_a and F_b gives

$$F_a = \frac{1}{2}mg + \frac{1}{2\gamma}\mu \qquad F_b = \frac{1}{2}mg - \frac{1}{2\gamma}\mu \tag{A9}$$

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

$$T_{bal} = aF_b - bF_a = \frac{a-b}{2}mg + \frac{a+b}{2\gamma}\mu \tag{A10}$$

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

$$\mu = \frac{2\gamma}{a+b}T_{bal} + \gamma mg\frac{b-a}{a+b} \tag{A11}$$

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

Appendix 2 - Calculations and Analysis

The results of computation is presented in this section together with certain conclusion which are also presented in a more compact form in a later section. The presentation essentially follows the order of computation $(1-8)$ given in the section on signal processing. References to capital letters $(A - E)$ are made with respect to experiments presented in the previous section. Graphs of registrations are given in electrical units ($[\cdot 10 \quad V]$) versus sampling instant. Time axis is given in units of sampling time ($\cdot 0.04 \quad [s]$).

Graphical presentation of experimental results

The followings two graphs show the result of experiment B where the subject keeps the eyes open.

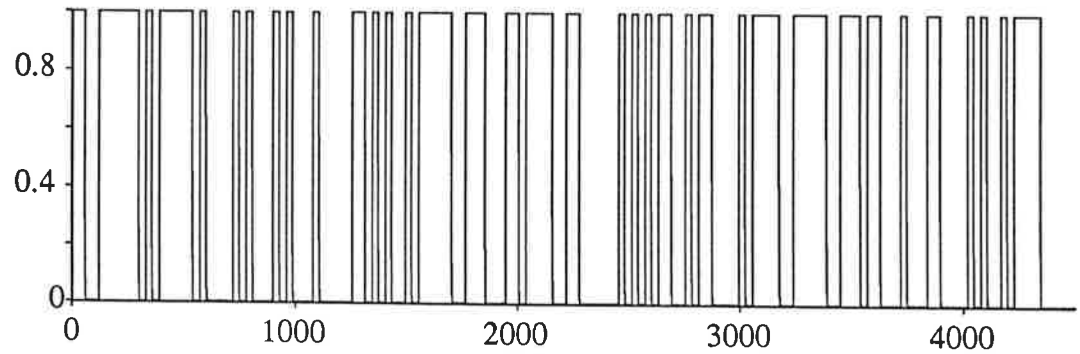

Figure 1. Experiment B : Input voltage $[x10 \quad V]$ to vibrators vs. time. Time scale in units of 0.04 $[s]$.

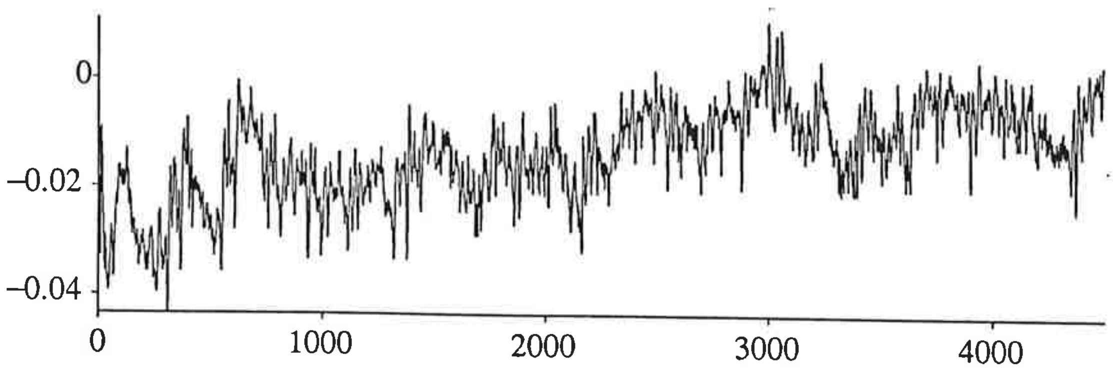

Figure 2. Experiment B : Sway (371.7 $[Nm/scale \quad unit]$) in saggittal plane (xz) vs. time. Open eyes. Time scale in units of 0.04 $[s]$.

The corresponding experiment results with closed eyes is shown below.

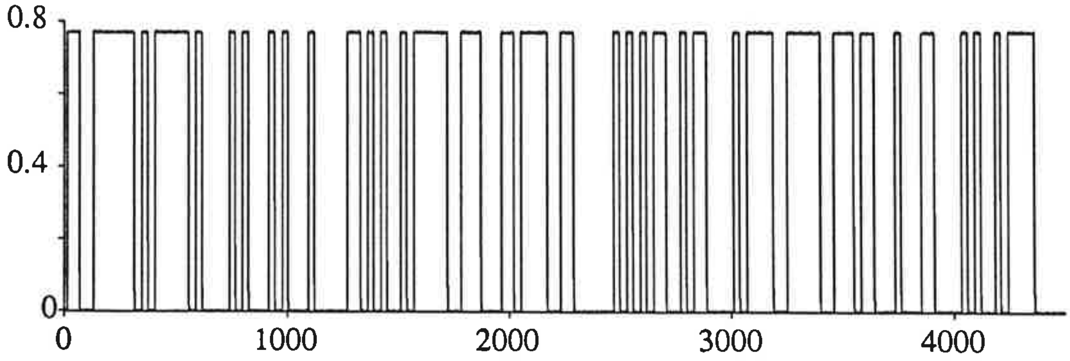

Figure 3. Experiment D : Input voltage to vibrators vs. time. Time scale in units of 0.04 $[s]$.

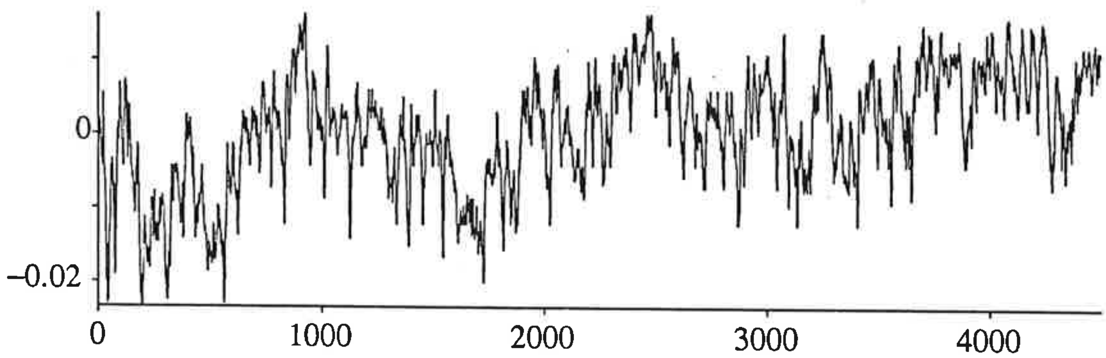

Figure 4. Experiment D : Sway (371.7 $[Nm/scale \quad unit]$) in saggittal plane (xz) vs. time. Time scale in units of 0.04 $[s]$.

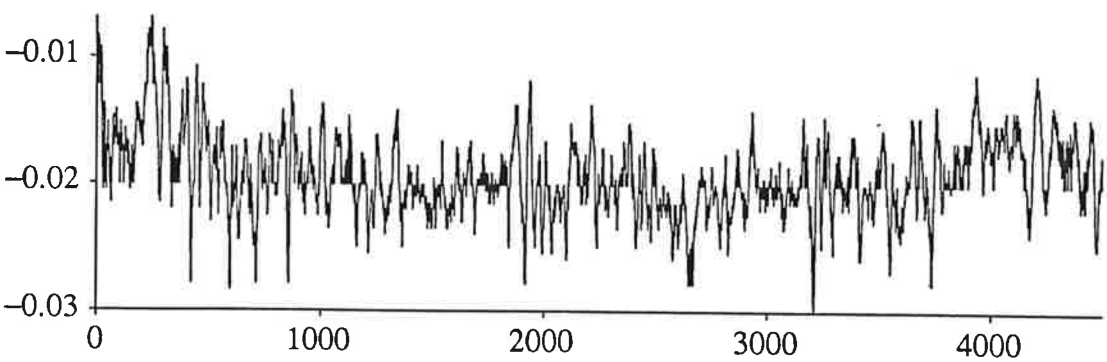

Figure 5. Experiment D : Sway (371.7 $[Nm/scale \quad unit]$) in frontal plane (yz) vs. time. Time scale in units of 0.04 $[s]$.

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

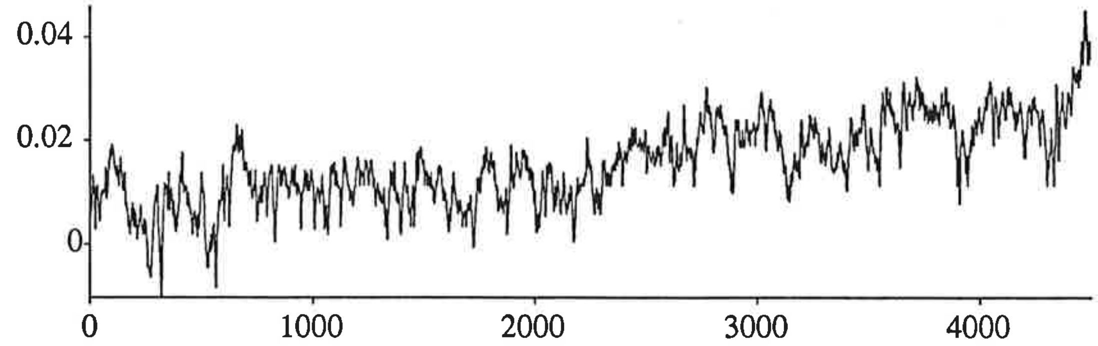

Figure 6. Experiment E : Sway (371.7 $[Nm/scale \quad unit]$) in saggittal plane (xz) vs. time. Time scale in units of 0.04 $[s]$.

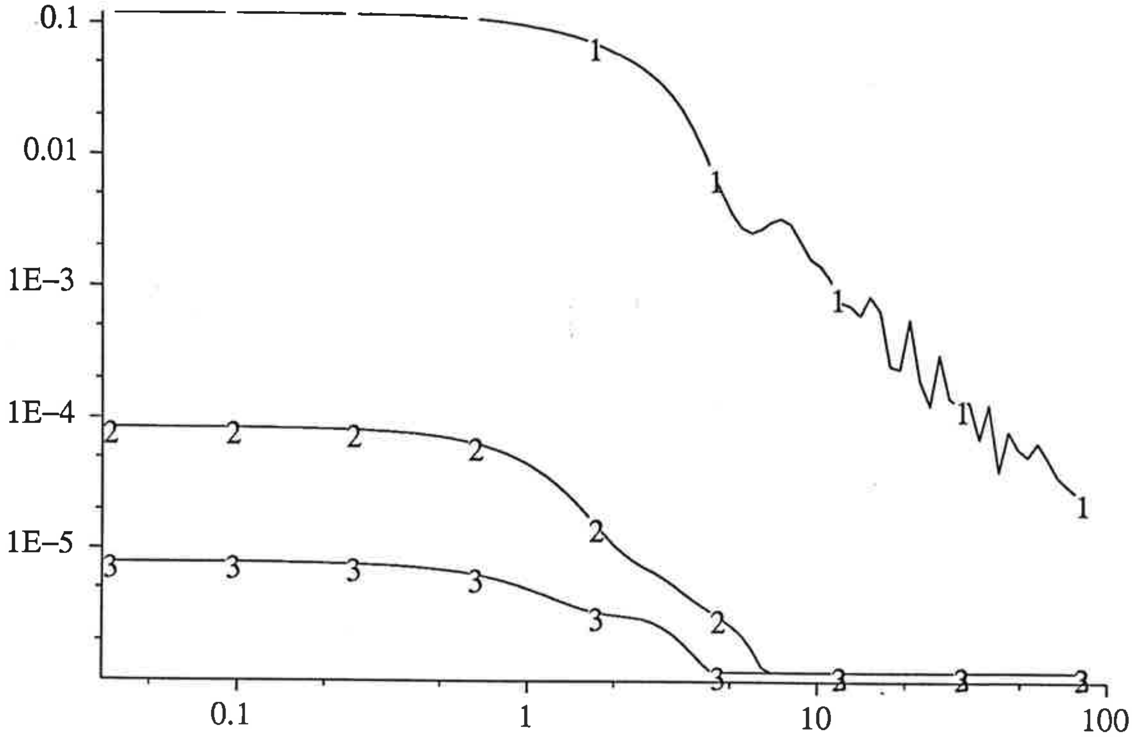

Figure 7. Vibrator input voltage spectrum of experiment D (1) vs. frequency $[rad/s]$. Saggittal sway spectrum of experiment D (2) vs. angular frequency $[rad/s]$. Frontal sway spectrum (3) of experiment D.

Autospectra

The autospectra (power spectra) show the frequency contents of the investigated signals. Notice that the spectrum must not be interpreted as 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 coherence spectrum. A large absolute value close to one indicates that the input v and the output x are correlated. A coherence of 0.5 gives the information 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. This is reasonable because the vibrators are mounted on the calves to stimulate saggittal motion.

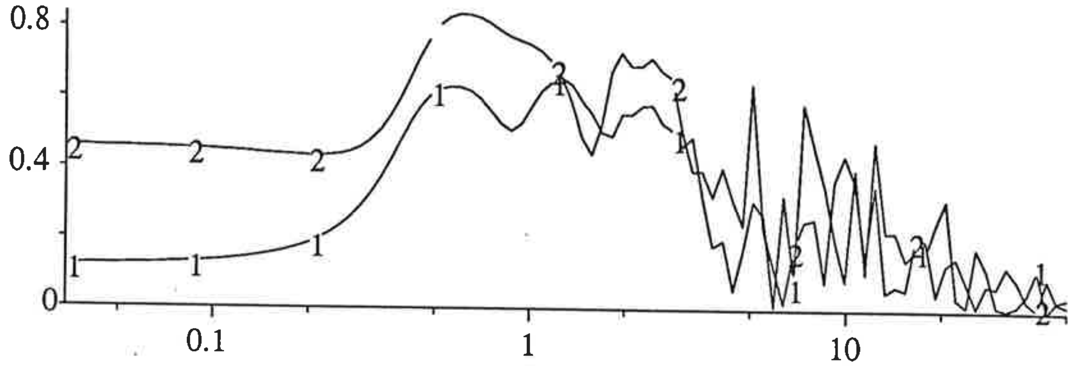

Figure 8. Experiment B : Coherence of sway in saggittal plane with open eyes (1) and experiment D : Coherence of sway in the saggittal plane with closed eyes (2) vs. angular frequency $[rad/s]$.

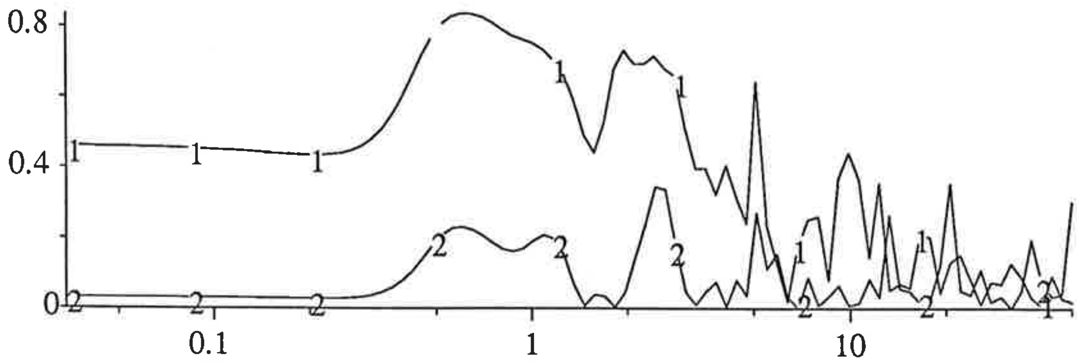

Figure 9. Experiment D : Coherence of sway in the saggittal plane (1) and frontal plane (2) vs. angular frequency $[rad/s]$.

It is seen below that the experiment result contains little information for assymmetrical stimulation.

From the spectra above it is noticed that coherence is low for frequencies above 3 $[rad/s] \approx 0.5[Hz]$. This indicates that the biological sensors (muscle spindles?) affected by the vibration stimulus operate with a bandwidth up to 0.5 $[Hz]$.

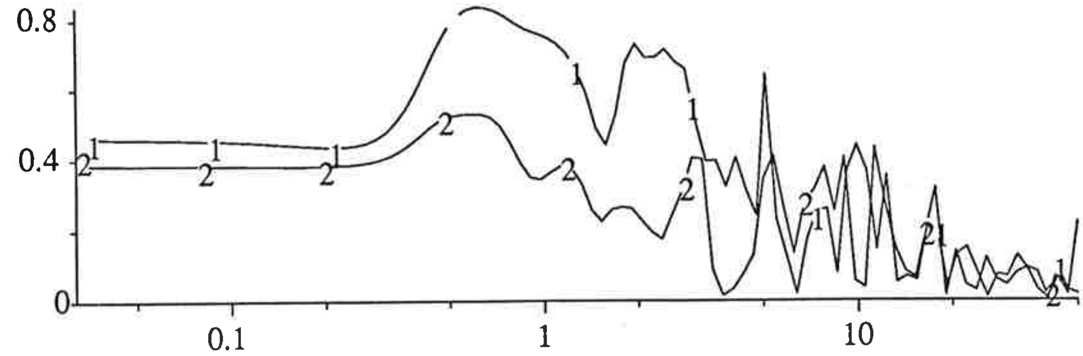

Figure 10. Experiment D : Coherence of sway in the saggittal plane with symmetric stimulation (1) and experiment E : Coherence of sway in the saggittal plane with asymmetric stimulation (2) vs. angular frequency $[rad/s]$.

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.

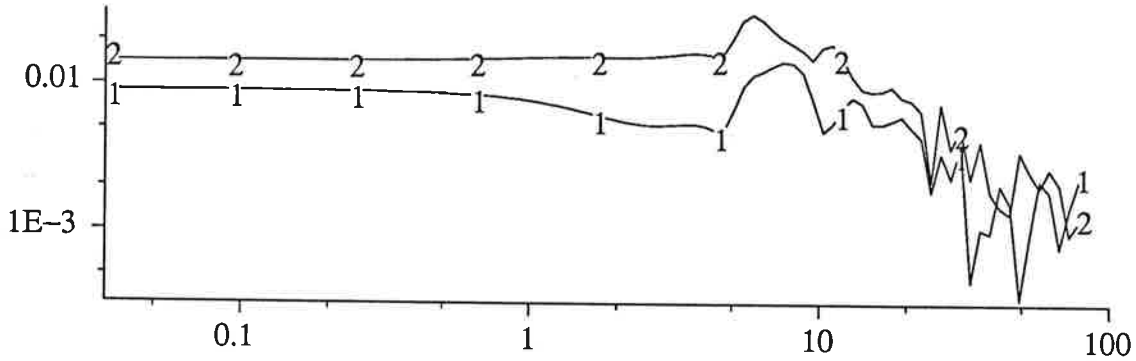

Figure 11. Transfer function from spectra of experiment B : with open eyes (1) and experiment C : with closed eyes (2). Gain graph.

There is some evidence that the muscle spindles react in different ways to different vibration frequencies, see [18]. This is however not a dominant feature in the graph above.

It is also possible to estimate the delay time T_d of the feedback loop by checking the phase lag for high frequencies. For high angular frequencies ω it holds that the phase lag is $\omega T_d + 90^o$. From figure 12 we find for e.g. $\omega = 50 \quad [rad/s]$ that

$$T_d \quad \approx \quad \frac{\pi}{180}\frac{600-90}{50} = 0.18[s]$$

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

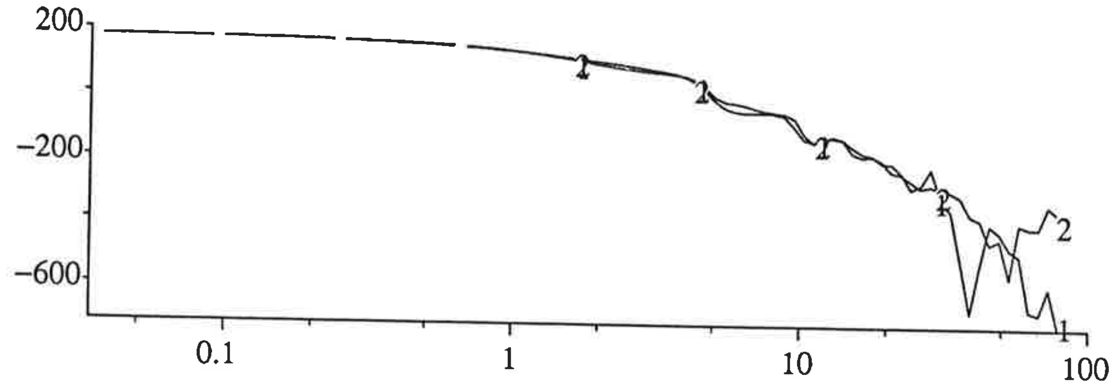

Figure 12. Transfer function from spectra of experiment B : with open eyes (1) and experiment C : with closed eyes (2). Phase graph.

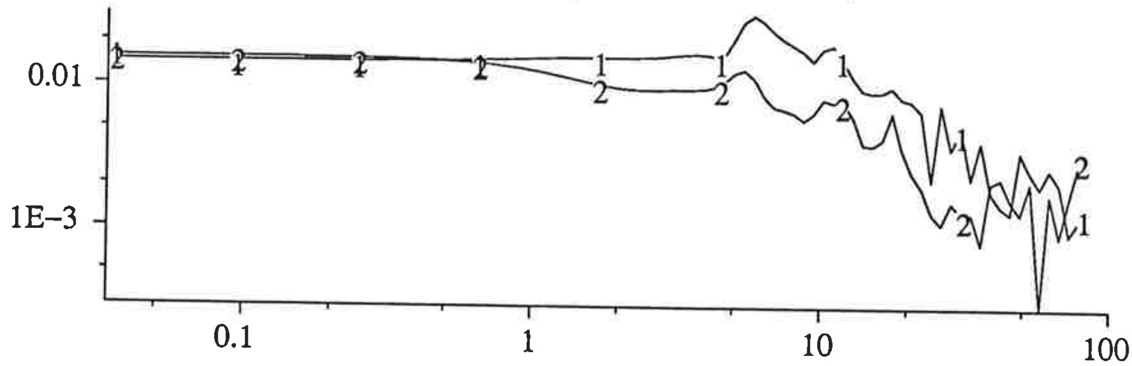

Figure 13. Comparison between transfer function from experiments with different vibrating stimulus; Gain graphs of experiments C : (1) and D : (2).

Maximum likelihood identification

The time delay was estimated to $\approx 0.15[s]$ and it is therefore desirable to estimate model parameters at sampling rate of this order of magnitude. The following ARMAX-models have all a sampling interval $0.20[s]$ which is obtained by extraction of every fifth sample from original time series.

Parameter identification with estimation of initial values is made for model orders two, three, and four. Statistical test 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 is therefore chosen to be 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
The polynomial coefficients and their standard deviations are


B =

    0.0003    0.0013   -0.0048
    0.0005    0.0009    0.0007


A =

    1.0000   -1.2265    0.7342   -0.3702
         0    0.1060    0.1544    0.0728


C =

    1.0000   -0.3807    0.2742   -0.0946
         0    0.1130    0.0926    0.0488
```

Validation by test of residuals

Residual tests have the aim to find remaining correlations which indicate if the the model order is sufficient. A sufficient model order leaves only white noise as residuals. The residual χ^2−tests for a third order model with 600 data points gives *significant** (95 % confidence) validation with respect to changes of signs, independence of residuals, normality, and independence between residuals and input.

Validation by simulation

Comparison between real and simulated data has been made with the vibration signal as a deterministic input. We study to what extent the experiment data are explained by the deterministic input-output behaviour of the estimated model. The input of D : is input to the estimated model.

Translation to continuous-time parameters

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

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

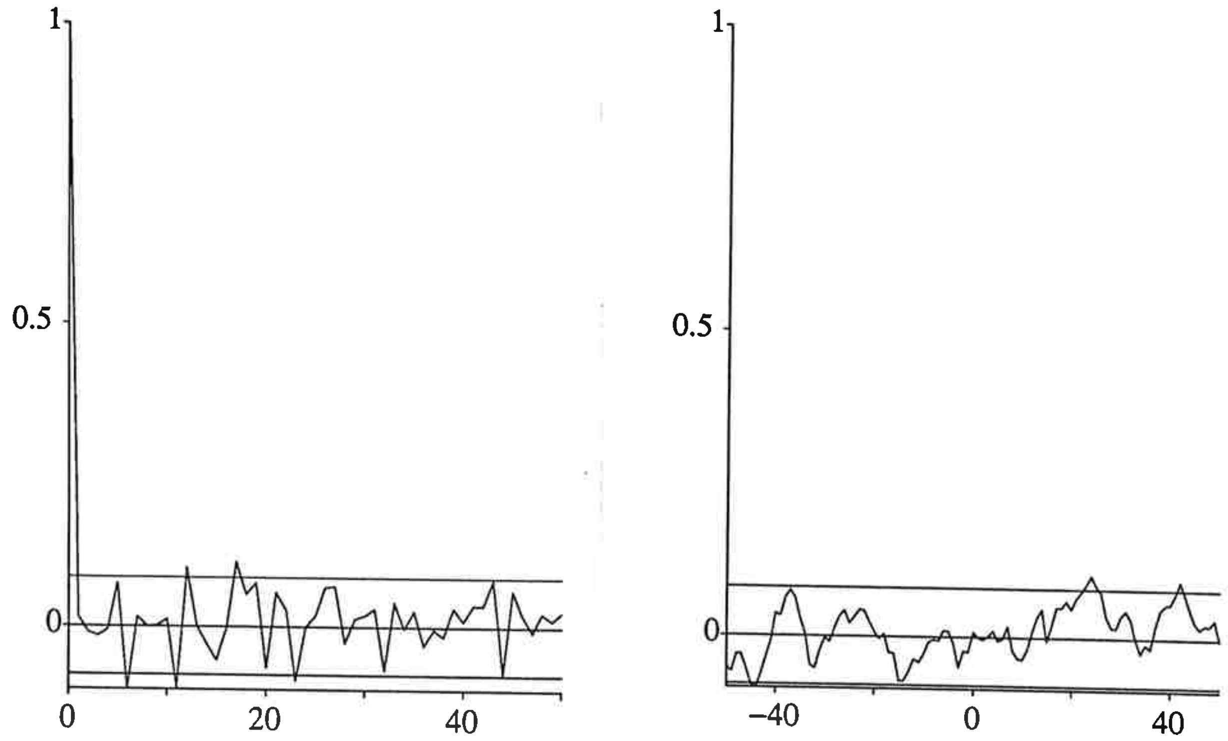

Figure 14. Test of autocorrelation of residual (*left*) and cross correlation between stimulus v and the residual (*right*) of a third order ARMAX-model fitted to data of D :. Time scale in units of 0.2 $[s]$. Confidence interval (95%) is displayed.

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

$$a_1 = -1.227 \pm 0.106 \qquad a_2 = 0.734 \pm 0.154 \qquad a_3 = -0.370 \pm 0.073$$

Statistical multivariate analysis is easy to make with respect to the coefficients of $A(q)$ because the covariance matrix is given in the IDPAC or MATLAB calculations. Translation to continuous-time parameters requires inverse sampling, see e.g. [1] (p.39) with

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

with sampling interval h and a matrix Φ. Computation of 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)

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

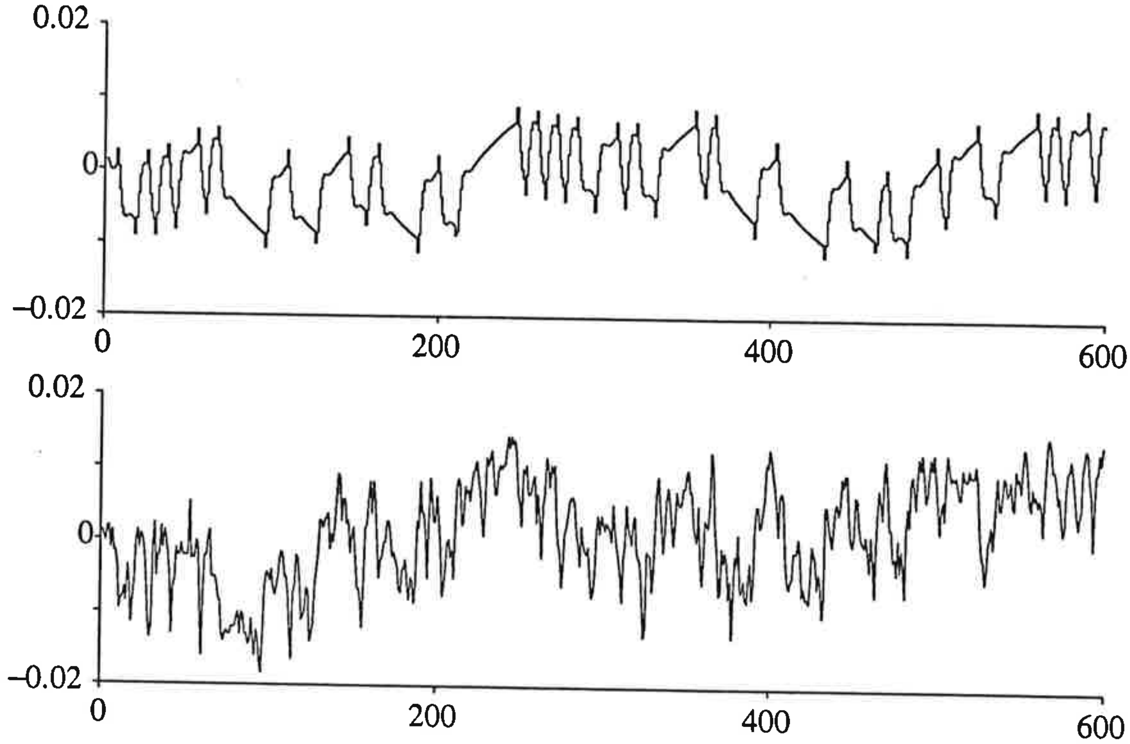

Figure 15. Saggittal sway of model with input of experiment D :. Simulated output from third order estimated model (*upper*). Saggittal sway of experiment D :. Real output (*lower*). Time scale in units of 0.2 $[s]$.

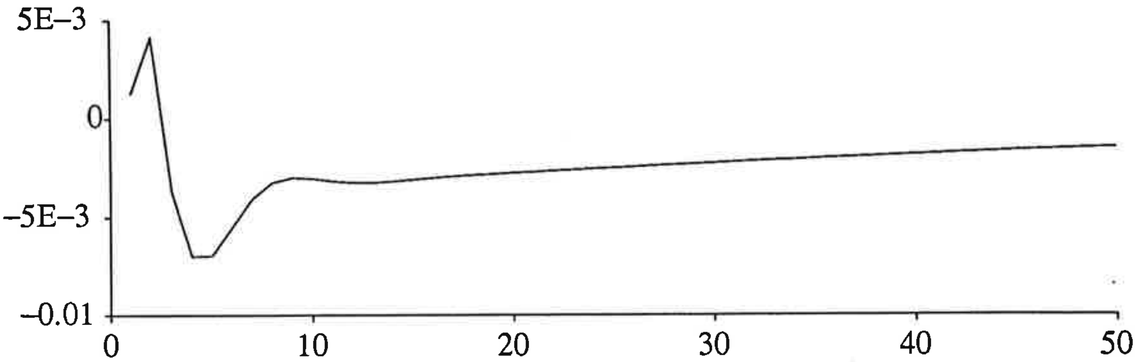

Figure 16. Saggittal sway. Simulated impulse response of force on the plate from third order estimated model. Time scale in units of 0.2 $[s]$.

The coefficients of $A(s)$ determine the posture behaviour. Parameters are already in the model normalized with respect to body weight m and body height l in terms of the moment of inertia J.

Results for the patient group was as follows:

Stimulation frequency 60Hz

Subject	η	k	ϱ
Patient #1	6.0853	49.2467	18.6723
Patient #2	4.4556	43.9899	10.4562
Patient #3	3.6408	32.1469	14.8532
Patient #4	2.9045	10.4387	4.3851
Patient #5	6.8936	47.7854	28.6838
Patient #6	4.9686	49.4480	31.9890

Stimulation frequency 100Hz

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

Assymmetric stimulation 100Hz

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	58.59	12.26

We have given one interpretation of the coefficients in terms of a mechanical model with a spring (k) and a viscous component (η). Naturally, a more rapid reflex system requires a balanced increase of both spring action and viscosity 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 bye the concepts

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

This classification describes the posture dynamics by one swiftness parameter and two stability parameters. The swiftness parameter is a bandwidth $[rad/s]$ and informs 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 posture control swiftness. The results are as follows:

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

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

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