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Responses of the posture-control system to pseudorandom acceleration disturbances

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Abstract—*By means of a servocontrolled force plate, pseudorandom acceleration disturbances were imposed upon a standing subject. By measuring and processing the responses of a subject, we computed the frequency-response functions of the posture-control system. The experimental results are discussed from a viewpoint of control engineering.*

Keywords—*Servocontrolled force plate, Posture-control system*

1 Introduction

THE BODY sway in the human upright posture is regulated by the sensory feedback which comprises visual, vestibular and proprioceptive sensors. The human posture-control system consists of a multifeedback system of which the controlled plant is unstable in itself. To elucidate the basic mechanism of such a complicated posture-control system, the characteristics of the system must be dealt with quantitatively. Quantification of the equilibrium function has been attempted by means of a force plate that measures the sway of the centre of gravity of a standing subject (THOMAS and WHITNEY, 1959; BARON, 1964; KAPTEYN and DE WIT, 1972; DICHGANS *et al.*, 1976). However, as the sway of the centre of gravity during quiet stance is the resultant sum of internal noises generated in the multifeedback, postural-control system, the interpretation of data is difficult.

To identify the characteristic of a feedback system in which internal noise exists, an external signal is necessary. Therefore we developed a servocontrolled force plate by which external disturbances are fed to the posture-control system (ISHIDA, 1978). Movable platforms have been used (GANTCHEV and POPOV, 1973; GURFINKEL *et al.*, 1976; MEYER and BLUM, 1978) but in most cases movement, such as sinusoidal and stepwise movement, was simple and predictable. To elucidate the different properties of the posture-control system, we imposed pseudorandom acceleration disturbances upon an upright subject and computed frequency response functions from ankle-joint moment and sway-angle data. Using a simple model of the posture-control system, we discuss our results from the viewpoint of control engineering.

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2 Method

The servocontrolled force plate is shown in Fig. 1. A print motor is adopted as actuator for this servomechanism. The drive mechanism is a 10-mm per revolution lead screw. The platform is supported by two vertical-force transducers and one spherical bearing which are fastened to the frame, together with the nut of the lead screw. Dimensions of the force plate are as follows:

platform
$500(W) \times 550(D) \times 6(T)$ mm

whole apparatus
$950(W) \times 750(D) \times 145(H)$ mm

maximum stroke of the platform
180 mm

weight
62 kg

The resonance frequency of this force plate with an additive weight of 50 kg is about 11 Hz. Static error of centre of pressure is less than 3 mm.

In the experiment we dealt with a condition where the force plate moved in the anterior–posterior direction. For an ordinary unmovable force plate, the centre of pressure in the anterior–posterior direction is given by

$$mg\bar{x}_{0b} = (R_1 + R_2)l \quad \ldots \ldots \quad (1)$$

where $\bar{x}_{0b}$ and l are the distances of the centre of pressure and the force transducer from the spherical bearing, $R_{1\sim 2}$ are the outputs of the vertical-force transducers, m is the mass of a subject, and g is the acceleration due to gravity (Fig. 2). When the platform is so thin that moments due to horizontal forces acting on the platform are negligible, eqn. 1 applies also to the movable force plare. In what follows the following assumptions are made:

(*a*) the human body can be expressed by a link model,

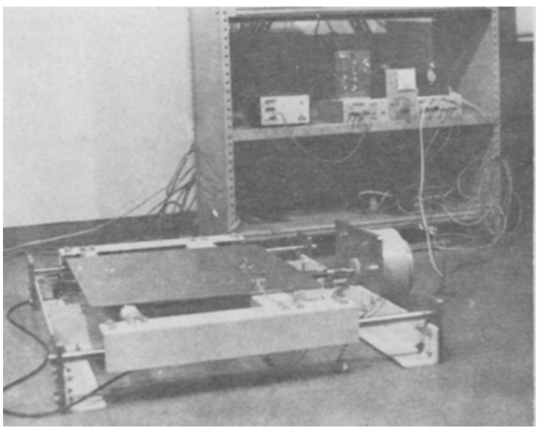


Fig. 1 Servocontrolled force plate

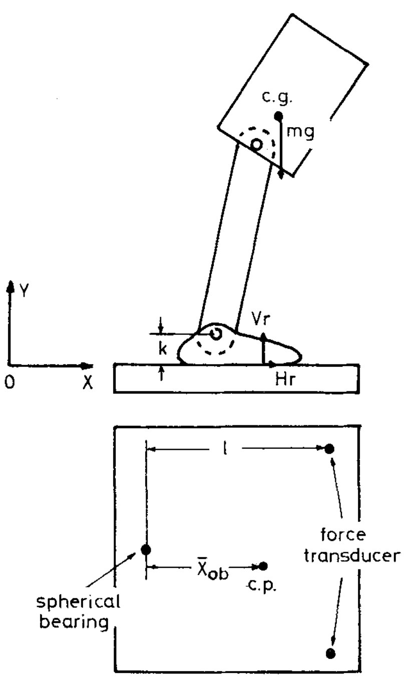


Fig. 2 Measurement of centre of pressure by a force plate c.g.; centre of gravity of a subject c.p.; centre of pressure

(*b*) the movement of a subject is restricted to a saggital plane;

(*c*) the mass of the foot is negligible.

For describing movement, an absolute co-ordinate system $0-XY$ is used; X-anterior–posterior horizontal, Y-vertical (Fig. 2). Subscripts r and l represent the right and the left side, respectively. The equations of motion are given by

$$\begin{aligned} m\ddot{y}_1 &= V_l + V_r - mg \\ m\ddot{x}_1 &= H_l + H_r, \end{aligned} \quad \ldots \ldots (2)$$

where (x_1, y_1) are the absolute co-ordinates of the centre of gravity of a subject, and V_i and H_i are the vertical and horizontal forces acting on each foot. Let us define the absolute co-ordinates of the centre of pressure of each foot and an ankle joint as $(x_{0i}, 0)$ and (x_2, k), then ankle joint moment u_i is given by

$$\begin{aligned} u_l &= H_l k + V_l(x_{0l} - x_2) \\ u_r &= H_r k + V_r(x_{0r} - x_2). \end{aligned} \quad \ldots \ldots (3)$$

Defining the absolute co-ordinates of the resultant centre of pressure calculated by eqn. 1 as $(x_0, 0)$, the following equation is derived:

$$(x_0 - x_2)(V_l + V_r) = (x_{0l} - x_2)V_l + (x_{0r} - x_2)V_r. \quad (4)$$

Using eqns. 2–4, the sum of each ankle joint moment is given by

$$u = u_l + u_r = m(g + \ddot{y}_1)(x_0 - x_2) + m\ddot{x}_1 k. \quad (5)$$

As $\ddot{x}_1$ and $\ddot{y}_1$ are small compared with g under our experimental condition, the ankle joint moment can be approximated by

$$u \simeq mg(x_0 - x_2). \quad \ldots \ldots \ldots (6)$$

It is the acceleration of the movable force plate that disturbs the postural-control system. However, we found it difficult to control the acceleration of the force

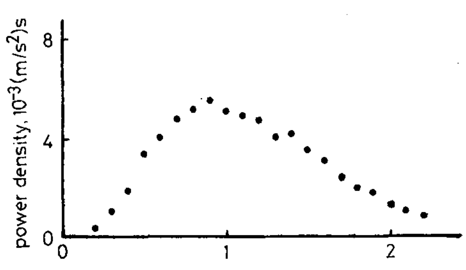


Fig. 3 Power spectral-density function of acceleration of the force plate (mean of 10 trials)

plate directly because of the drift and noise of the accelerometer. We constructed a force-plate speed-control system, the reference of which was given by the integral of the desired value of acceleration. Velocity is detected by a tachometer and slight position feedback is also provided by a potentiometer to prevent a drift in position. A maximum-length pseudorandom sequence is used as the acceleration reference. Because the low-frequency component of acceleration produces a large displacement of the force plate, components in a frequency range below 0·5 Hz are cut off by a filter. The power spectral-density function of acceleration of the force plate is shown in Fig. 3.

To estimate the sway angles of a body and legs, we measured the horizontal position of lamps attached to the subject's knee, hip and shoulder by means of a t.v. measuring system (Fig. 4). As the sway angle of legs θ_1 and body θ_2 are small, they are given by

$$\begin{aligned} \theta_1 &= c_1(e_2 - e_1) \\ \theta_2 &= c_2(e_3 - e_2) \end{aligned} \quad . \quad . \quad . \quad . \quad . \quad (7)$$

where $e_{1\sim 3}$ are the outputs of the television measuring system which correspond to the horizontal position of subject's knee, hip and shoulder, and $c_{1\sim 2}$ are the proportionality constants. The static error of this measuring system in horizontal position is less than 1 mm. However, the dynamic characteristics include dead time of about 0·09 s due to persistence of a camera tube.

After acquisition of u and $\theta_{1\sim 2}$, we computed frequency response functions and coherence functions between each variable and acceleration of the force plate, d. In the cases of u, computation is given by

$$\begin{aligned} H_u(f) &= \frac{S_{du}(f)}{S_{dd}(f)} \\ \gamma_u^2(f) &= \frac{|S_{du}(f)|^2}{S_{dd}(f) S_{uu}(f)} \end{aligned} \quad . \quad . \quad . \quad . \quad . \quad (8)$$

where $S_{dd}(f)$ and $S_{uu}(f)$ are the power spectral-density functions of d and u, respectively, and $S_{du}(f)$ is the cross spectral density function between d and u. Errors of sway angles due to dead time of the t.v. measuring system were compensated in the frequency domain. To obtain the power spectral-density functions, we

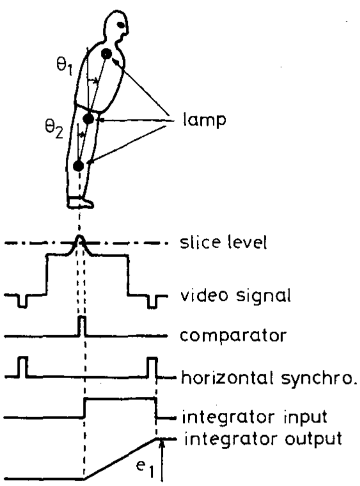


Fig. 4 Measurement of sway angle by a t.v. measuring system

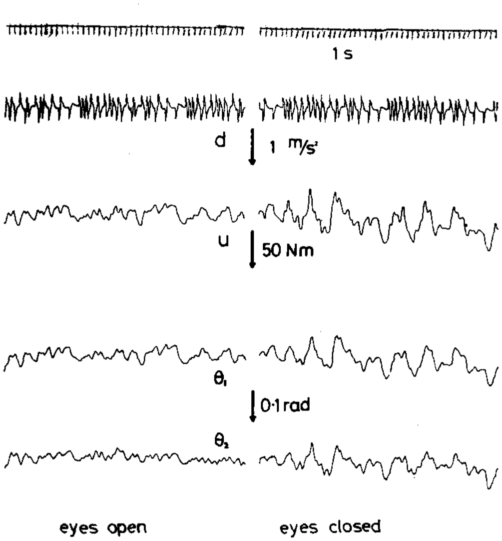


Fig. 5 Experimental raw data (subject S.S.). Refer to Fig. 9 for the direction of arrows

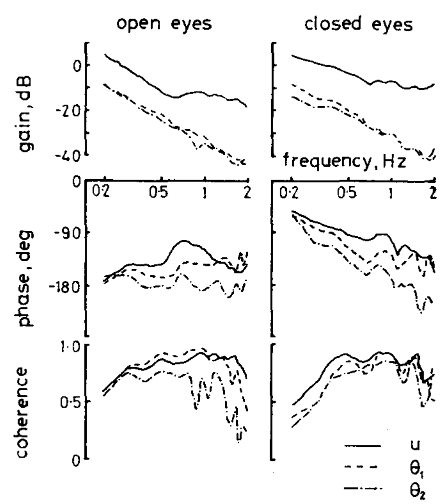


Fig. 6 Frequency-response functions between each variable and acceleration of the force plate (subject S.S.). Gain of u is shifted downward by 40 dB. Units of each variable are as follows. u(N · m), θi(rad), d(m/s²)

adopted f.f.t. algorithms and chose the record length of one trial as 60 s, and the sampling interval as 39·1 ms.

3 Results

Five healthy male volunteers, ranging in age from 27 to 37 years were our subjects. Each subject stood on the platform with his bare feet at a 30-degree angle to each other. Two trials were performed under each condition of open and closed eyes. Examples of raw data and frequency-response functions are shown in Figs. 5 and 6, respectively. The following frequency-response function features were common to all subjects:

(*a*) the gains of θ_1 and θ_2 appear to have an asymptotic slope of 40 dB/*dec*, but the gain of u shifts from this asymptotic curve beyond about 0·6 Hz;

(*b*) the phase lag of u is smaller than those of θ_1 and θ_2;

(*c*) the gain characteristic of θ_2 is similar to that of θ_1, but the phase lag of θ_2 is larger than that of θ_1.

Feature (*c*) seems to be due to the fact that the mass of a body is larger than that of the legs and the prediction of movement of the platform is difficult because of randomness.

Let us assume that the ankle joint moment u is primarily determined by the information of the sway angle of legs θ_1. Then the difference of the gain and the phase shift between u and θ_1 in Fig. 6 represents the transfer characteristics from θ_1 to u. That is, the gain difference is $20 \log |S_{du}/S_{d\theta_1}|$ (dB) and the phase shift difference is $\angle S_{du} - \angle S_{d\theta_1}$(°). Examples of such transfer characteristics are shown in Fig. 7. One can readily see the derivative characteristics in this graph and this property is derived from the above features (*a*) and (*b*). These derivative characteristics seem to contribute to the stabilisation of the posture-control system, although its degree varies from subject to subject.

To illustrate the effect of eye closure, the frequency response functions under the conditions of open and closed eyes are displayed in Fig. 8. The following features under the closed-eyes condition were common to all subjects:

(*a*) In the low-frequency range, the phase lag with closed eyes is smaller than that with open eyes in all variables.

(*b*) The gain of u with closed eyes increases remarkably.

Feature (*b*) results in an increase in the gain of transfer

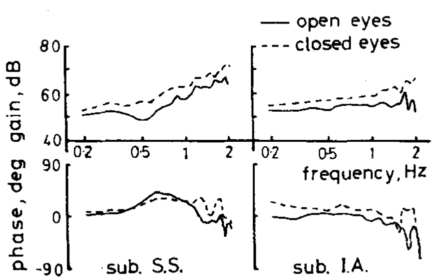


Fig. 7 Transfer characteristics from θ_1 to u

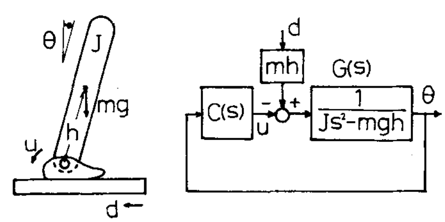
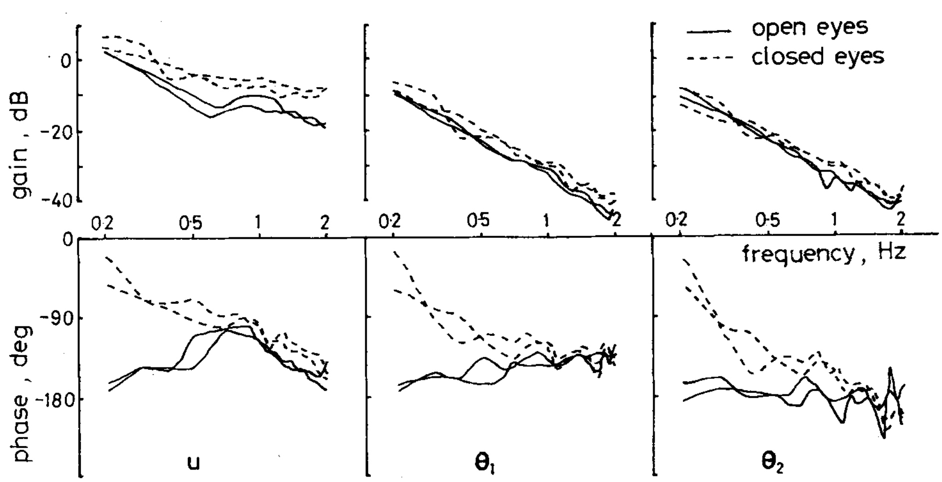


Fig. 9 Model and block diagram of the posture-control system



Frequency response functions with open and with closed eyes (subject S.S.). Gain of u is shifted downward by 40 dB characteristics from θ_1 to u in the condition with closed eyes (Fig. 7).

4 Discussion

To discuss the experimental results from a viewpoint of control engineering, a model of the posture-control system is useful. For simplicity, let us consider a single inverted pendulum model as indicated in Fig. 9 and assume that the ankle joint moment is the only controlling input. The equation of motion which is

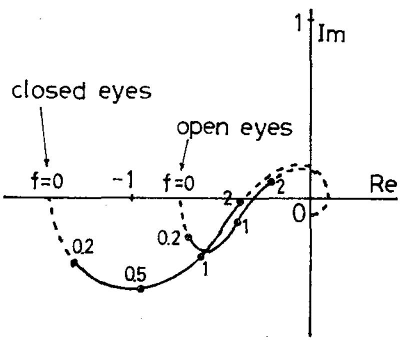


Fig. 10 Vector loci of the posture control system (subject S.S.)

linearised at the equilibrium point ($\theta = 0$) is given by

$$J\ddot{\theta} - mgh\theta = mgd - u\,, \qquad (9)$$

where θ is the sway angle of a body, J is the moment of inertia about an ankle, m is the mass, h is the height of the centre of gravity, d is acceleration of a platform, u is the ankle joint moment and g is the acceleration due to gravity.

Using the above equation, a block diagram of the posture-control system can be derived (Fig. 9). The block $C(s)$ represents the transfer characteristics from θ to u. It contains the characteristics of sensory feedback, information processing in the c.n.s. and muscles. If we substitute the frequency response functions as indicated in Fig. 7 for $C(f)$, $C(s)$ can be approximated by

$$C(s) = K(1 + Ts)e^{-Ls} \qquad (10)$$

Parameters, K, T and L are calculated by curve fitting. Values of parameters for the data of Fig. 7 and the mean values of parameters for all trials are shown in Table 1. Referring to other studies (YAMAZAKI, 1975), parameters J and h can be estimated from the subject's height and weight. For example, $J = 43{\cdot}3\,\text{kgm}^2$ and $mh = 49{\cdot}4\,\text{kgm}$ for subject S.S. Using these values, the vector loci $C(f)G(f)$ of subject S.S. were calculated (Fig. 10). From Nyquist's stability criterion of a feedback system, the number of counterclockwise rotations of the vector locus about the $(-1, j0)$ point for $-\infty < f < \infty$ must be one.

As the low-frequency component of acceleration was cut off in our experiment, it was impossible to obtain frequency-response functions below about 0·2 Hz. However, if we extrapolate the vector loci beyond 0·2 Hz, it is clear that the gain in the low frequency range is insufficient for stability in the condition with open eyes. The posture-control system is stabilised by the sensory feedback comprising visual, vestibular and proprioceptive sensors. As the visual system has a rather long latency, its stabilising effect is restricted in the low-frequency range where the phase lag due to latency is small. Consequently, it seems that the visual system guarantees the necessary gain in the frequency range below 0·2 Hz in the condition with open eyes. On the other hand, to compensate for the lack of visual feedback in the condition with closed eyes, the gain of vestibular and/or proprioceptive sensors increases as indicated in Fig. 7.

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Table 1. Estimated values of K, T and L

subject		K (dB)	T (s)	L (s)
S.S.	open eyes	50	0·42	0·14
	closed eyes	56	0·33	0·10
I.A.	open eyes	50	0·21	0·16
	closed eyes	55	0·25	0·13
total (mean ± s.d.)	open eyes (n = 10)	50 ± 2	0·27 ± 0·13	0·15 ± 0·03
	closed eyes (n = 10)	53 ± 2	0·34 ± 0·07	0·15 ± 0·03

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