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Human Postural Dynamics in Response to the Horizontal Vibration

Young-Kyun Shin, Mohammad A. Fard, Hikaru Inooka, and Il Hwan Kim*

Abstract: The dynamic responses of human standing postural control were investigated when subjects were exposed to long-term horizontal vibration. It was hypothesized that the motion of standing posture complexity mainly occurs in the mid-sagittal plane. The motor-driven support platform was designed as a source of vibration. The AC Servo-controlled motors produced anterior/posterior (AP) motion. The platform acceleration and the trunk angular velocity were used as the input and the output of the system, respectively. A method was proposed to identify the complexity of the standing posture dynamics. That is, during AP platform motion, the subject's knee, hip and neck were tightly constrained by fixing assembly, so the lower extremity, trunk and head of the subject's body were individually immovable. Through this method, it was assumed that the ankle joint rotation mainly contributed to maintaining their body balance. Four subjects took part in this study. During the experiment, the random vibration was generated at a magnitude of $0.44m/s^2$, and the duration of each trial was 40 seconds. Measured data were estimated by the coherence function and the frequency response function for analyzing the dynamic behavior of standing control over a frequency range from 0.2 to 3 Hz. Significant coherence values were found above 0.5 Hz. The estimation of frequency response function revealed the dominant resonance frequencies between 0.60 Hz and 0.68 Hz. On the basis of our results illustrated here, the linear model of standing postural control was further concluded.

Keywords: Coherence function, frequency response function, non-predictive horizontal vibration, resonance frequency, standing postural control.

1. INTRODUCTION

The analysis of human postural dynamics allows not only better diagnoses and treatment of disorders in regard to balance control but also facilitates the design artificial control systems for patients with sensor-imotor deficits. Many of the earlier studies have attempted to perturb the standing postural system in a large number of ways in order to quantify the human balance, such as perturbations to the proprioceptive system [1,2], perturbations to the visual system [3,4], and perturbations to the vestibular system [5,6]. These researches have focused on a significant role in elucidating the contribution made by different subsystems to the overall balance control.

In regards to the external perturbation, one approach that has proved useful for investigating human equilibrium control is employing transient perturbation to evoke characteristics of postural responses [7-9]. The transient perturbation may be useful to demonstrate trigger specific and equally transient motor programs with respect to the standing posture mechanism. For instance, a sudden translation or a short acceleration [10] demonstrated that the control mechanism is expressed by mainly reflexive responses. However, the effects of prolonged acceleration have not been elucidated, thus it may not be directly related to the continuous regulation of postural balance control. Hence, the use of continuously applied perturbation may be appropriate to the study of postural control behavior, which itself is a continuously maintaining process.

The inverted pendulum model is often used for standing posture study. When standing body is exposed to the anterior-posterior (AP) perturbation, the ankle joint firstly generates reactionary movement

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Young-Kyun Shin and Hikaru Inooka are with the Graduate School of Information Sciences (GSIS), Tohoku University, Sendai, Japan (e-mails: ykshin@yuhan.ac.kr, inooka@niche.tohoku.ac.jp).

Mohammad A. Fard is with the Nissan Motor Co., 560-2, Okatsuko ku, Atsugi 243-0912, Japan (e-mail: fard@mail.nissan.co.jp).

Il Hwan Kim is with the Department of Electrical and Electronic Engineering, Kangwon National University, 192-1 Hoyza2-dong, Chunchon, Kangwon 200-701, Korea (e-mail: ihkim@ kangwon.ac.kr).

* Corresponding author.

for balance control. Horak and Nashner [9] established a theoretical basis for the ankle and hip strategies, McCllum and Leen [11] also showed that stiffening the body acts as a one-segment as well as a two-segment inverted pendulum, i.e., as in the ankle strategy or in the hip strategy, respectively.

To quantify the role of the human sensory systems on stabilizing the standing posture, it is firstly required to analyze the passive responses (open-loop responses) of the standing posture. Accordingly, it is very important to have a simply designed model for the standing body, since it may be very difficult to quantify the active responses (closed-loop responses) of the standing dynamics by the use of a complex model. This is a major reason that many works concerning standing posture have preferred to utilize a single-inverted pendulum model rather than a complex model.

The single-link inverted pendulum model of postural control assumes that control of postural stability in standing is dependent on the control of a single joint, the ankle [4,12,13]. This assumption is based on several experimental findings. First, the amplitude of ankle movement in standing surpasses movements in other joints; second, the projection of the body mass center is located about 5 cm anterior to the ankle joints and consequently the static moment at these joints is large; and thirdly, the changes of activity of muscles around ankle joints are dynamic in standing.

Nevertheless, due to the complexity of the biomechanics of human standing posture dynamics, studies regarding freestanding body as an inverted pendulum might oversimplify the mechanism of postural control, and it may lead to a misunderstanding of postural control behavior. Hence, the restriction of the subject standing body condition as a low degree of freedom is essential to understand the basis of the biomechanics of standing posture complexity.

The resonant behavior of the different parts of the human body in some frequencies can further suggest taking the frequency-domain responses of the human body into consideration.

In this study, human upright standing posture was investigated when subjects were exposed to the horizontal vibration in the sagittal plane. The AC Servo motor controlled vibrator was designed, and the continuous acceleration was applied as a source of external vibration. As the input and the output of the postural system, the platform acceleration and the trunk angular velocity were used.

Two significant assumptions were adopted in our study. Firstly, the motion of standing posture complexity mainly occurs in sagittal plane. Secondly, during AP platform motion, the subject's knee, hip and neck were tightly fixed, so the ankle joint rotation was assumed to contribute to maintaining their body balance. For the analysis of dynamic behavior of human upright standing control, the coherence function and the frequency response function were estimated over a frequency range from 0.2 to 3 Hz.

2. MATERIALS AND METHODS

2.1. Subjects

Four healthy subjects participated in this study (mean age ± SD, 27.5 ± 5.2 years). Their physical characteristics are shown in Table 1. No subjects had any evidence or history of neurological, gate, postural or musculoskeletal impairments. All participants gave their informed consent prior to inclusion in the study.

All subjects completed all trials without falling, stepping or reaching for a stable support. During the experiment, each subject was instructed to stand upright on the platform in a standardized stance, and the feet of the subject were separated medio-laterally by a distance of about 10 cm, and kept together so that the left and right ankle joints rotate about the same axis.

Except for the ankle joint, the other joints that may compensate against external perturbation, i.e., knee, hip, neck, were tightly fixed (Fig. 1). For this condition, a number of medical appliances (fixing assembly) were used, i.e., Velcro straps, medical splints and casts. As contrasted with freestanding, since the subject's bodies were extremely restricted, they were not able to tolerate such experimental conditions for any length of time.

Under these conditions, the lower extremity, trunk and head of the subject's body were individually immovable; the subject's posture would be controlled as an inverted pendulum. After each trial, intervals were included to avoid subject's exhaustion. All subjects stood barefoot with their arms folded comfortably across the chest and their head facing forward and upright. This arm position was adopted in order to eliminate the possibility of their arms swaying entering into the dynamics. The condition of

Table 1. Physical characteristics of the subjects (# sign indicates the subject number).

Subject	Age (yr)	Height (m)	Weight (kg)
# 1	23	1.70	65
# 2	23	1.68	58
# 3	32	1.74	64
# 4	32	1.74	75
Mean (SD)	27.5 (5.2)	1.72 (0.03)	65.5(7.05)

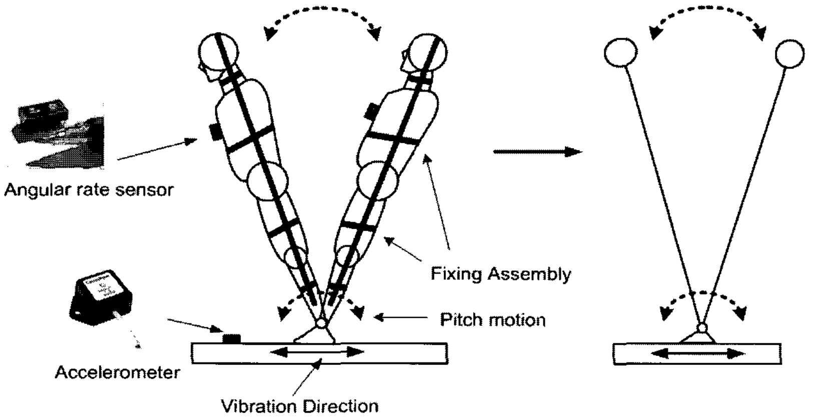

Fig. 1. The schematic of human standing posture exposed to the horizontal vibration.

the feet was assumed not to slide or lift at the heel or toe, and since the triangular foot model is considered, it also did not enter into the dynamics of the system. During the experiment, the room lights were off, and the subjects were blindfolded and instructed not to resist or apply any voluntary response.

2.2. Experimental setup

The AC Servo-motor controlled vibrator was designed as a mobile rigid platform. The vibrator system consists of AC Servo-motor (Sanyo-Denki Co.) and actuator unit (THK Co.). It has Max. stroke of 1200 mm, Max. frequency of 5 Hz, and Max. Load of 100 kgf. The platform had a size of 606 × 406 mm. Zero mean Gaussian random vibration was devised as the input. It generated mean stroke of 0.03 m, and generated mean acceleration of 0.44 m/s^2. The power spectrum of the input signal is presented in Fig. 2. This input was devised for testing a severe condition that could generate near the limits of human upright standing posture. The duration of each trial was 40 seconds. The initial period was not acquired to eliminate from the acquisition non-stationary events possibly induced by the onset of platform translation. So data from the latter 35*s* were subjected to the subsequent analysis.

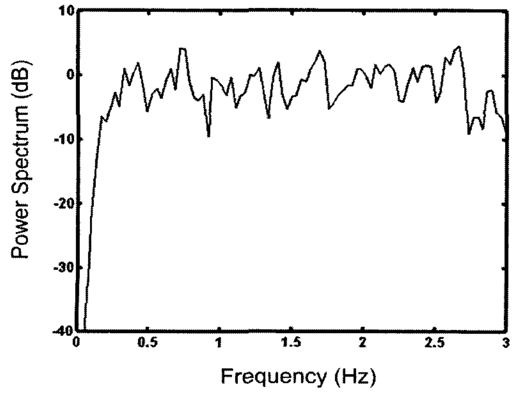

Fig. 2. Power spectrum of the input acceleration.

A surface-mounted accelerometer (Crossbow CXL 04LP3) was attached to the platform to measure horizontal acceleration of the input exerted to the upright standing posture. The accelerometer is a DC accelerometer having a range of ± 4G, which can measure not only accelerations caused by force imposed on a segment but also static accelerations such as gravitational acceleration. These characteristics are suitable for postural calculation as a tilt reference. The stretch adhesive bandages were used for the accelerometer to improve the congruence of body motion. An angular rate sensor (Murata ENC-03J) was applied to measure angular velocity of a subject's trunk. The gyroscope has small and quick response up to 50 Hz, and wide range of ± 300 deg/sec. This sensor was lightweight, pasted on the trunk (Clavicle) to measure mainly in sagittal plane and to provide high-resolution measure of upright stance posture angular velocity.

Each subject underwent horizontal vibration four times individually, but for one of them, one more set of data was collected to use for validating the results. The initial data were preprocessed prior to sampling by a 13 Hz analog low-pass filter. The measured platform acceleration and the trunk angular velocity were sampled at 100 Hz through an A/D converter, and band-pass filtered at 0.2 Hz to 3 Hz.

2.3. Data analysis

The coherence function between the input $x(t)$ and the output $y(t)$ of the experimental results is defined by

$$\gamma^2(f) = \frac{|G_{xy}(f)|^2}{G_{xx}(f)G_{yy}(f)}, \tag{1}$$

where $G_{xx}(f)$ and $G_{yy}(f)$ denote the autospectral density function of input and output, respectively, and $G_{xy}(f)$ is cross-spectral density function. The coherence function provides a formal measure of the correlation between two signals in the frequency domain. In the ideal case of a constant-parameter linear system with a single clearly defined input and output, the coherence function will be unity. If $x(t)$ and $y(t)$ are completely unrelated, the coherence function will be zero. The coherence function of 1 indicates that the phase shift between the waveform at a given frequency is constant, and the amplitude of the signals at that frequency has a constant ratio [14]. The coherence function can be interpreted as a measure of linear predictability [15,16] – it equals 1 whenever $x(t)$ is a linear function of $y(t)$, and it is also used in the calculation of 95% confidence limits on the transfer function data [17].

The transfer function (frequency response function) of the system $H(f)$ is then estimated by

$$H(f)=\frac{G_{xy}(f)}{G_{xx}(f)}. \tag{2}$$

The estimated transfer function takes into account the linearly correlated proportion of the output with the input. The transfer function characterizes the dynamic behavior of a system by showing how the output response sensitivity changes as a function of the input. Here for estimating both coherence function and transfer function, a frequency resolution 0.1 Hz was considered.

The obtained transfer functions for each subject are averaged to represent a uniqueness with better accuracy than those resulted from each test trial. The adopted averaging method here is that of the Geometric mean [18,19]. It is defined by

$$\overline{H(f)}=\prod_{k=1}^{n}\sqrt[n]{H_k(f)}, \tag{3}$$

where $H(f)$ indicates the complex form of transfer function for the experiment number k, which is calculated from (1). The value of n indicates the number of repeated experimental trials. Here the value of n is 4. We have assumed that the frequency domain noises in the measured signals are normally distributed. It is worthwhile to mention that even the noises of the signals in the time domain are not normally distributed, by transferring into the frequency domain; they can be modeled as normal distributed noises [20].

Hence, in this condition, the Geometric mean of the transfer functions, obtained from (3), likely reduce the effects of noise-corruption and give an unbiased estimation of the transfer function better than that of the arithmetic mean.

3. RESULTS

The coherence functions results from four subjects, i.e., the correlations in the frequency domain between acceleration of platform and angular velocity of trunk, which are used as the input and the output of the system, are shown in Fig. 3. Each individual graph indicates one trial, and totally there are four trials per subject. Though the position of the maximum coherence varied with the frequency band, in general, similar shapes of the coherency plots for each subject were found across all trials. It can be seen that there is significant coherence between the input and the output.

The common fall of the coherences are observed at the frequencies less than 0.5 Hz. In the frequency ranges above 0.5 Hz, the coherence functions results indicate that the behavior of the system is quasi-linear.

Moreover, since the experimental data have been averaged across four trials of data, the amount of noise corruptions as well as bias should approximately be lower than each individual data. Hence, it may be concluded that the coherence is quite satisfactory for estimating the transfer function of the system. The transfer functions from four different subjects between platform acceleration and trunk angular velocity are revealed in Fig. 4. Each individual graph indicates the transmissibility of the system, i.e., magnitudes of the transfer functions. There is a similar correspondence between subjects in the four trials. For each single trial, though it is varied, there is the prominent resonant peak. It is distributed with small deviations within its frequency band and magnitude. This suggests that there is a high degree of repeatability for the obtained transmissibility. At frequencies greater than resonance frequencies, the system dynamics allow for significant reductions in transmissibility.

Consequently, it may be concluded that there is a considerable dominant resonance frequency across the experimental results. The geometric mean of the transfer functions, which are obtained from four individual subjects, is illustrated in Fig. 5. And the

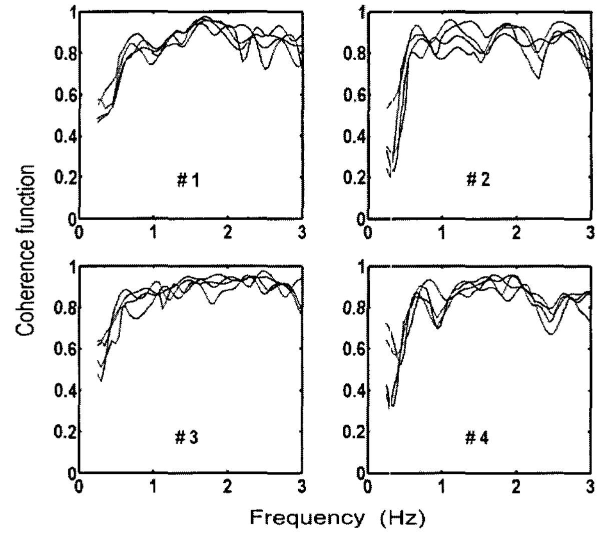

Fig. 3. Coherence functions from four different subjects (# sign indicates the subject number).

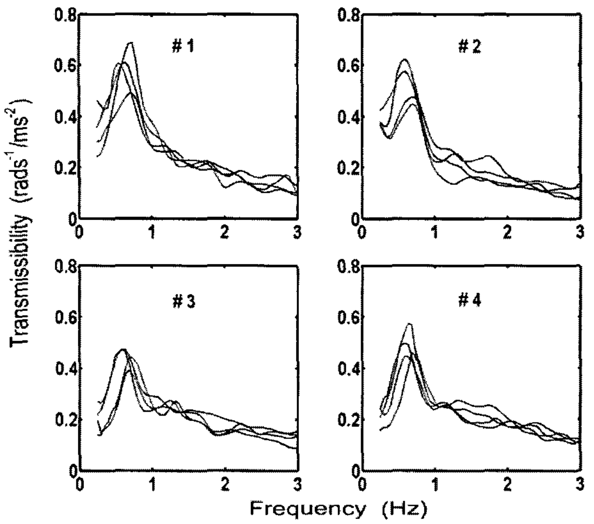

Fig. 4. The transfer functions between the platform acceleration and the trunk angular velocity.

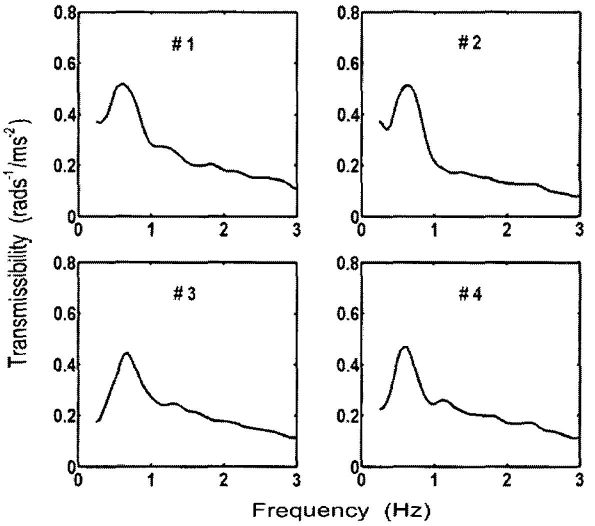

Fig. 5. Geometric mean of the transfer functions for four subjects (subjects # 1 to # 4).

comparison of averaged transmissibility for four subjects is shown in Fig. 6.

Each single graph indicates the transmissibility curves obtained from four trials for each subjects, and it is averaged across all trials. The geometric mean of the transfer functions reveals aclearer resonant peak. The prominent resonant peaks are located at 0.60 Hz, 0.63 Hz, 0.68Hz, 0.61 Hz, with respect to subject numbers 1 to 4. The magnitude of the transfer functions at the dominant resonance frequencies for subjects number 1 to 4 are 0.52, 0.51, 0.44, and 0.47 $rads^{-1}/ms^{-2}$, respectively. From these results, it can be surmised that Fig. 4 illustrates the distribution of transfer functions obtained from four subjects restrained to sway as a single link inverted pendulum.

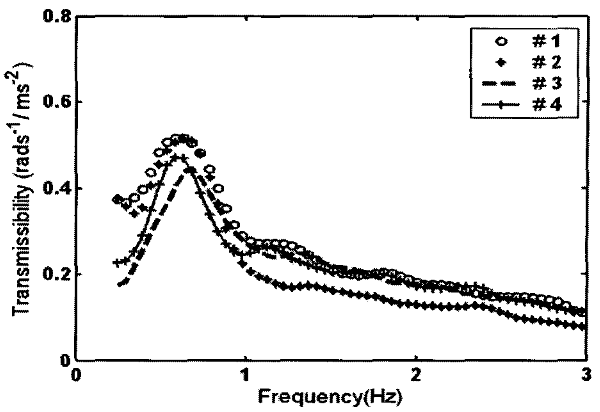

Fig. 6. Comparison of averaged transmissibility's for four subjects (subjects # 1 to # 4).

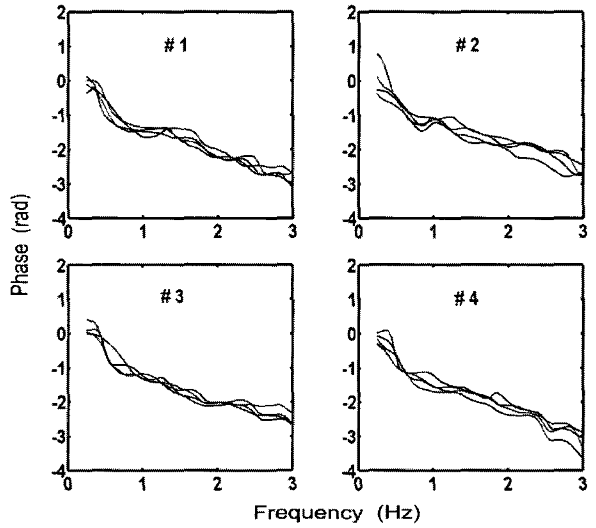

Fig. 7. Phase of the transfer functions for four subjects. Each graph indicates one experiment.

The phase results of transfer functions obtained from four different subjects are revealed in Fig. 7. Each single graph corresponds to a single trial. The inclination angle of the phase graphs at around the dominant resonance frequency is seen to be greater than other frequencies.

It should be noted that a nearly similar pattern was found for all subjects. This indicates that the phase relationship between the input and output of the system are relatively constant. Again, these features suggest that there are remarkable phase drops at around resonance frequencies, which is in accordance with the general characteristics of the phase graphs at resonance frequencies. To illustrate the general behavior of the standing posture control system, the evaluation is processed. The platform acceleration, the trunk angular velocity and the trunk angular displacement are shown in Fig. 8, for a representative subject (the subject # 3). The magnitude of the acceleration input, which is excited to the platform, is measured from -1.58 to 1.28 m/s^2, and the angular velocity and the angular displacement are measured, -0.49 to 0.72 rad/s, -0.1 to 0.06rad, respectively. The

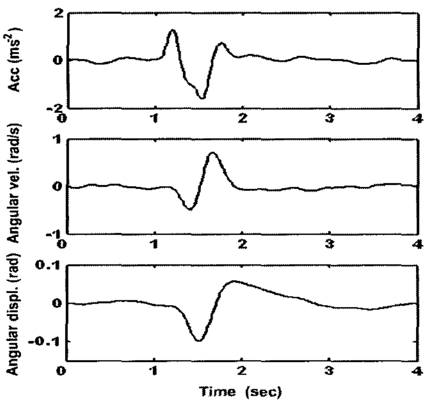

Fig. 8. Transient response of standing postural system (the subject # 3).

angular position of the trunk in upright standing before the platform moved is regarded as the zero position. Forward platform movement caused the subject body to incline backwards, and vice versa.

4. DISCUSSION

Significant coherence values were found between the platform acceleration and trunk angular velocity during continuous AP platform motion. It seems that the power of the oscillation increases by a factor of two, i.e., input and output of the system, which means that there is a reasonable response to the horizontal vibration stimulus. The coherence results for each trial were reproducible from trial to trial.

At frequencies between 0.2 to 0.5 Hz, in general, relatively lower values of coherence were obtained, and showed characteristic differences between subjects within its frequency band and magnitude. This is mainly due to the active forces by the ankle muscles of the subject, and it is consistent with a lower signal-to-noise ratio for responses evoked by the very low-amplitude stimulus [21,22]. Besides, at frequencies of low range, the effect of voluntary motion could be improved. Also each subject demonstrates a different biomechanical threshold, so it may contribute to the deviations within its frequency band and magnitude. Prior to estimation of transfer functions, it will be valuable to evaluate the coherence functions for assessment of linear dependence and for detection of disturbance levels.

When the frequency of an input vibration is close to the natural frequency of a human standing postural system, then that system will have a tendency to resonate. The variance of the transfer function estimated at a given frequency band does not decrease as the number of trials grows. Instead, the signal-to-noise ratio determines the accuracy at each frequency [23]. Since the experimental data have been averaged across four trials of data, the amount of noise corruptions as well as bias should approximately be lower than each individual data. The estimation of frequency response function revealed the dominant resonant frequencies across all subjects. The pattern of this frequency response function is consistent in the way that the subjects were restrained to sway as a single-link inverted pendulum. Consequently, it can be seen that each of the responses to the different trials shows a similar degree of linearity though the overall responses of the human postural control system are nonlinear.

There are many earlier studies that mainly addressed the concordance of low-frequency body movements [1,22,24-26]. As contrasted with free-standing, the experimental conditions that were adopted in our study may restrict the change of body center of mass (COM).

The application of the low frequency range (< 3Hz) for the dynamic behavior of standing control was due to the following main reasons: Firstly, under the experimental condition in our study, the movement of standing body COM was relatively restricted, by fixing assembly as well as by being exposed to the non-predictive input. Secondly, the resonance frequency of the standing posture was located at low frequencies. Thirdly, for future applications of this research when it is intended to identify the active motion (closed-loop responses) of the standing posture, a model necessary for the complexity of the standing posture dynamics, which can represent the passive motion (open-loop responses) of the standing posture at low frequencies, as the ankle muscle can be activated and can respond to the vibration at frequencies lower than 3 Hz. Since the model of this study represents the passive behavior of the system at low frequencies (< 3Hz), it is an appropriate model for such applications.

From the present frequency response function results in this study, it is possible to identify the model of postural control system. This issue will be addressed in our future study. For the frequency range adopted here (< 3Hz), we suggest a human postural control model like a single-link inverted pendulum, which controls postural stability in standing by a single joint, the ankle. The schematic of our proposed model is shown in Fig. 9. The model consists of a lumped mass with the rotational spring (K) and damper (B). The θ is the sway angle, m is the mass, and g is the gravity acceleration.

When a stable standing body is exposed to the external perturbation, the reciprocal role of the central nervous system (CNS) and the skeletal muscle system may generate the reactionary movement for maintaining postural balance. By the role of the CNS, the external perturbations are actively damped, or passively damped by viscous properties of the skeletal muscle system. The weight of the anticipatory reaction is probably maximal under the condition of

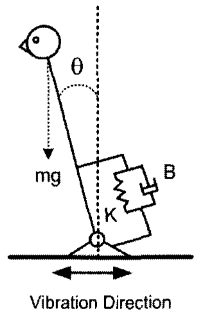

Fig. 9. A single-degree of freedom model of the standing posture.

known type and timing of the platform displacement [27,28]. It is, therefore, important to analyze the posture control system from the point of view of feedback control [10].

We used the Gaussian random vibration as an excitation signal to the standing postural dynamics. Since it was intended to measure the open-loop responses of the complexity of the standing posture to the vibration, therefore, it was required to consider a non-predictive random excitation signal. The non-predictive vibration can reduce the influences of the human vestibular and somatosensory system. The clinical researches show that the contribution of the vestibular and somatosensory reflexes is negligible when the input is a non-predictive random vibration [29]. This kind of input helps the subjects to better follow the instructions in reducing the vestibular, visual, and voluntary responses, and therefore, they are more appropriate than a periodic excitation for the purpose of this study.

In regards to our future research, the investigation at different visual cues, i.e., eyes open and eyes closed conditions, and the comparison between the present results and the experiment results that will be derived by predictive or transient inputs may lead to improve the accuracy of the present data. Besides, it can be facilitated to the issues of the subject's feeling and perceptions, and the mechanism of human sensorimotor integration to postural control.

5. CONCLUSIONS

The dynamic responses of standing postural control were studied where the long-term AP vibration was implied. During AP platform motion, the subject's knee, hip and neck were tightly constrained by fixing assembly. This proposed method allowed that the ankle joint rotation could contribute to maintain their body balance. The obtained coherence function shows that each trial is reproducible. Significant coherence values were found above 0.5 Hz, and it was quite satisfactory for estimating the frequency response function.

The estimation of frequency response function revealed the dominant resonance frequencies between 0.60 Hz and 0.68Hz. The pattern of this frequency response function is consistent in the way that the subjects were restrained to sway as a single-link inverted pendulum. Also, it is coincident with the scheme that we adopted in this study. On the basis of our results in this study, the linear model of standing postural control is further concluded. The present results may provide useful knowledge to identify further application.

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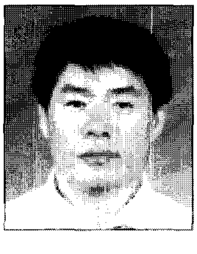
Young-Kyun Shin received the B.S. and M.S. degrees in Electrical and Computer Engineering from Kangwon National University, Korea in 1997 and 1999, respectively. He is currently pursuing a Ph.D. degree in the Graduate School of Information Sciences (GSIS), Tohoku University, Sendai, Japan. His research interests include design, control, and applications of robotics, biomechanics, control of human posture and movement, and human-machine interaction.

Mohammad A. Fard received the Ph.D. in Mechanical Engineering from Tohoku University in 2003. He is currently working for Nissan Motor Co., Atsugi, Japan. His research interests include biomechanical engineering, system identification, influences of vibration to the human body, and automotive NVH analysis.

Hikaru Inooka received the B.E., M.E., and Ph.D. degrees in Mechanical Engineering, from Tohoku University, Sendai, Japan, in 1963, 1965 and 1969, respectively. He joined Tohoku University in 1969 where he was a Professor until 2004. He is currently a Guest Professor at Tohoku University. His current research interests include manual control, human-machine interface, and measurement of human motions.



Il Hwan Kim received the B.S. and M.S. degrees in Control and Instrument Engineering from Seoul National University in 1982 and 1985, respectively and the Ph.D. at the Tohoku University in 1993. In 1995, he joined the Department of Electrical and Electronic Engineering at Kangwon National University where he is currently a Professor. His research interests include control, mechatronics, and human interfaces.