Journal of the Korean Institute of Illuminating and Electrical Installation Engineers (2014) 28(3) : 63~71
http://dx.doi.org/10.5207/JIEIE.2014.28.3.063 ISSN 1229-4691(Print) ISSN 2287-5034(Online)

Biomechanical Analysis of Human Balance Control

Youngkyun Shin* · Gu-Bum Park**

Abstract

A single-inverted-pendulum model is presented to simulate and predict the passive response of human balance control. This simplified biomechanical model was comprised of a torsional spring and damper, and a lump mass. An estimation of frequency response function was conducted to parameterize the complexity. The frequency domain identification method is used to identify the parameters of the model. The equivalent viscoelastic parameters of standing body were obtained and there was good conformity between the simulation and experimental result.

Key Words : Frequency Response Function, Frequency Domain Identification, Open Loop Response, Human Posture Complexity

1. Introduction

The mechanism of human bipedal standing is inherently unstable not only because of the large body mass located high with its center of mass (COM), but also because of maintaining it over a relatively small base of support. A small sway perturbation from the steady-state position might result in the unstable mechanism accelerating further away. Therefore a counteractive torque must be exerted by the combined sensory control systems to maintain its equilibrium control.

* Main author : Instructor, Department of Electrical Engineering, Yuhan University
** Corresponding author : Professor, Department of Electrical Engineering, Yuhan University
Tel : +82-2-2610-0734, Fax : +82-2-2619-9771
E-mail : pgb@yuhan.ac.kr
Date of submit : 2014. 1. 21
First assessment : 2014. 1. 25
Completion of assessment : 2014. 2. 28

In prior studies, human standing posture is often considered to be an inverted pendulum pivoted at the ankles [1-3]. These studies revealed that the inverted pendulum balancing task was broadly equivalent to real standing. Winter proposed an inverted pendulum model with the muscles serving as tunable springs to drive the center of pressure (COP) in phase with the COM, and verified the model for sagittal sway [2, 4]. Morasso and Shieppati [5] investigated active control by the central nervous system (CNS) in this approach, and found that the potential role of ankle proprioception and foot somatosensation allowed for anticipatory control.

The single-link inverted pendulum model of standing postural control assumes that control of postural stability is dependent on the control of a single joint, the ankle [5-6]. When subjects used this ankle strategy, their bodies were rigid and swayed centered about the ankle joints [3-7].

The effects of long-term external vibration are still not clear, however, and the nature of the control mechanism under such condition is still an object of controversy. In addition, there is little quantitative knowledge on viscoelastic parameters of the complexity of the standing postural dynamics. Therefore, to quantify the role of the human sensory systems on stabilizing the standing posture, it is necessary to identify the passive elements of standing posture. Accordingly, it is very important to have a simple model designed for the standing body, since it is very difficult to quantify the active elements of the standing dynamics by the use of a complex model.

Whenever a physical model is designed for the system, an identification approach is required to characterize the parameters of the system. One important point for the identification of a system is to know whether the system is linear or nonlinear. If the system is not fully linear, it is valuable to know how much nonlinear distortion exists. For a linear system with slight nonlinear distortions, the frequency domain identification method is preferred over the time domain method. In addition, this method is useful to remove the contaminating noise of the measured signal, because it allows extraction of the best linear approximation of a nonlinear system in the presence of this noise [8-9].

The complexity of the standing postural dynamics in the present study is not a fully linear system and contains nonlinear distortion. For such a system, it is more appropriate to utilize the frequency domain identification method than the time domain method. To achieve further insight into the postural control behavior revealed in the experimental results, this study used a control model to parameterize the transfer function results. The model, then, consists of a single-inverted-pendulum to explain the pure body mechanisms and a standing balance control system.

2. METHODS

2.1 Measurement

The AC Servo-motor controlled vibrator was designed as a mobile rigid platform (606×406 mm). It consisted of an AC Servo-motor (Sanyo-Denki Co.) and actuator unit (THK Co.), and had a Max. Stroke of 1,200 mm, Max. Frequency of 5 Hz, and Max. Load of 100 kgf. Zero-mean Gaussian random vibration was devised as input to test severe conditions near the limits of standing posture. It was a nominally flat spectrum, and generated a mean stroke and mean acceleration of 0.03 m and 0.44 ms^{-2}, respectively.

A surface-mounted accelerometer (Crossbow CXL04LP3) was attached to the platform to measure horizontal acceleration of the input. It was a DC accelerometer having range of ±4 G. An angular rate sensor (Murata ENC-03J) was adhered to the trunk (Clavicle) to measure mainly in the sagittal plane. It had a small, rapid response up to 50 Hz, and a wide range of ±300 deg/sec. 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 by a 4th-order Butterworth filter.

Experiments were performed on four healthy participants. None of the participants reported any history of neurological, otological, or orthopedic abnormalities. Since this study hypothesized that the ankle joint was the major joint for compensation for external perturbation, the other joints, i.e., knee, neck, were firmly fix (Fig. 1). A number of fixing assemblies (medical appliances) were used, including medical splints and casts, and Velcro

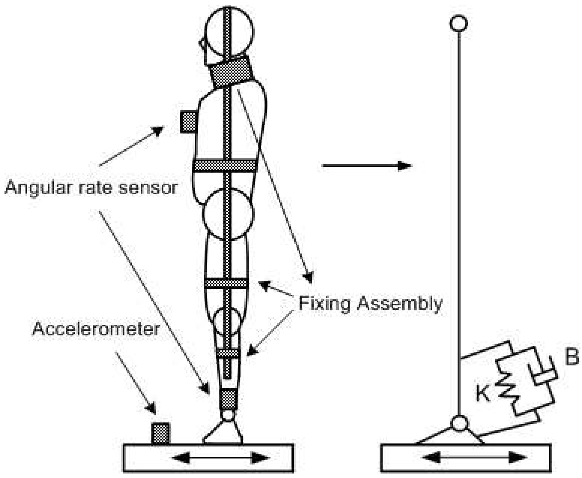


Fig. 1. The schematic of human standing posture exposed to horizontal vibration

straps. Under this condition, the head, trunk and lower extremities of the subjects were not movable separately, therefore the subject posture was controlled as an inverted pendulum. The duration of each trial was 40 s and sufficient intervals were given to avoid the exhaustion of the subject after each trial.

The participants were informed to stand upright and were barefoot. The feet were placed slightly less than shoulder width apart by a distance about 10 cm, and kept together so that the left and right ankle joints rotated about the same axis. Their arms were folded comfortably across the chest, and the head faced forward. This was to eliminate the possibility of arm and head sway entering into the dynamics. Across all trials, the room lights were off, and the participants were blindfolded and instructed not to resist or apply any voluntary response.

2.2 Signal Processing

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

$$H(f)=\frac{G_{xy}(f)}{G_{xx}(f)} \tag{1}$$

where $G_{xx}(f)$ denotes the autospectral density function of input, and $G_{xy}(f)$ is cross-spectral density function [9]. The estimated transfer function takes into account the linearly correlated proportion of the output with the input. The obtained transfer functions for each subject are averaged to represent a unique function with better accuracy than those that resulted from each test trial. The adopted averaging method here was that of the Geometric mean [10-12]. It is defined by

$$\overline{H(f)}=\prod_{k=1}^{n}\sqrt[n]{H_k(f)} \tag{2}$$

where $H(f)$ indicates the complex form of transfer function for the experiment number k, which is calculated from Eq. (1), and the value of n indicates the number of repeated experimental trials (here, n = 4).

This study assumed that the frequency domain noises in the measured signals are normally distributed. 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 [12]. Hence, in this condition, the Geometric mean of the transfer functions as obtained from Eq. (2) likely reduce the effects of noise-corruption and give an unbiased estimation of the transfer function better than that of the arithmetic mean.

2.3 Modeling and identification

A single-degree-of-freedom model (a two-dimensional inverted pendulum) was considered as a model for the standing posture complexity. This model was comprised of a torsional spring, a torsional damper and a lump mass, and it was hypothesized as follows for simplification.

First, it was assumed that the motion of the standing posture complexity mainly occurs only in the mid-sagittal plane. Second, the standing posture complexity only has flexion/extension motion, so the translational motion of the standing posture complexity is assumed to be negligible. Third, during anteroposterior (A/P) platform motion, the subject's knee, hip and neck were tightly fixed, so that ankle joint rotation was assumed to contribute to the maintaining of body balance.

The equation of motion for this model, linearized at the equilibrium point ($\theta=0$), is given by the following form [2, 7, 11, 13-14],

$$J\ddot{\theta}+B\dot{\theta}+(K-mgl)\theta=-ml\ddot{y} \quad (3)$$

where θ is the angular displacement of the system, m is the mass of the body, and J is the mass moment of inertia of the body (excluding feet) around the ankle, l is the distance between COM and center of rotation, and g is the gravitational acceleration. The $\ddot{y}$ is the input (acceleration), K indicates the viscoelastic coefficient of muscles around the ankle, while B is the viscous coefficient of muscles around the ankle. According to Eq. (3), the controlled object is stable where $B>0$ and $K>mgl$. It is easy to understand both mathematically and intuitively that there is no stable equilibrium at $\theta=0$. This means that the inverted pendulum model is unstable and easily falls if the subjects do not try to stabilize it. Again, a person who does not counteract the gravitational torque with a stabilizing response will inevitably fall.

Notably, ankle torque dominates the body movement in this equation. Backward ankle torque is continuously applied to the body to prevent it from falling forward, because COM is located in front of the ankle joint. The ankle flexor activities are rare and ankle extensors are considerably activated [3, 15], therefore it can be said that ankle extensors contribute the most towards control of the ankle joint torque; as a result, the body is able to maintain itself in standing posture.

The m was then derived from

$$m=M-M_f \quad (4)$$

where M and M_f indicate mass of the whole body and the mass of the feet, respectively. It has assumed that the feet had a constant length-density across all subjects [2]. The mass moment of inertia of the body around the ankle, J, was derived from Eq. (4) [7, 13] by assuming that the ratio of the masses and the ratio of radiuses of gyration of the body were equal to the ratio of their lengths [7, 16].

$$J=ml^2 \quad (5)$$

The distance of center of mass from the ankle, l, was measured by the following equation [2, 14].

$$l=0.575\times H \quad (6)$$

where H is the height of the subject. The inertia parameters of the subjects' standing bodies are shown in Table 1. The s-domain transfer function

Table 1. Physical characteristics and inertia parameters of subjects. Note that the # sign indicates the subject number

Subject	#1	#2	#3	#4	Mean(SD)
$Age(yr)$	23	23	32	32	27.5(5.2)
$M(kg)$	65	58	64	75	65(7.05)
$M_f(kg)$	1.89	1.68	1.86	2.18	1.9(0.21)
$m(kg)$	63.11	56.32	62.14	72.82	63.6(6.84)
$H(m)$	1.70	1.68	1.74	1.74	1.72(0.03)
$l(m)$	0.978	0.966	1.0	1.0	0.986(0.017)

Table 2. The results of identification with physical characteristics of four subjects

Subject	#1	#2	#3	#4	Mean(SD)
α	0.9775	1.0732	0.7038	0.8221	0.894(0.164)
β	1.9089	1.9291	2.2616	2.0952	2.049(0.165)
γ	14.688	18.835	12.151	10.957	14.158(3.485)
$l(m)$	0.978	0.966	1.0	1.0	0.986(0.017)
$J(kgm^2)$	60.36	52.56	62.14	72.82	61.97(8.34)
$K(Nm/rad)$	1511.27	1574.02	1354.78	1497.85	1484.48(92.62)
$B(Nms/rad)$	117.76	106.31	139.52	151.12	128.68(20.34)

$H(s)$ of the system between the trunk angular velocity and input acceleration was obtained from Eq. (3) [2, 7, 11, 13-14].

$$H(s) = e^{-sTd}\frac{s}{\alpha s^2 + \beta s + \gamma} \tag{7}$$

where $s = j2\pi f$,

$$\alpha = \frac{J}{ml} \tag{8}$$

$$\beta = \frac{B}{ml} \tag{9}$$

$$\gamma = \frac{K - mgl}{ml} \tag{10}$$

T_d indicates the pure time delay of the system, and was estimated prior to identification. The best value, which improved the quality of the identification, was found to be $50\,ms$. Therefore, three unknown parameters remained, α, β, and γ, for the frequency domain identification method [8, 11] in Eq. (7). The averaged frequency response function (Eq. (2)), corresponding to each subject was used as a single set of experimental data for the identification [8]. In other words, an attempt was made to fit the model to the averaged value of the experimental data.

3. Results

The formal measure of correlation between the input and output of the system in the frequency domain, that is, the coherences [9], are shown in Fig. 2. Each graph is related to one subject. This shows the mean values of coherency functions across four subjects in the frequency range from 0.2 Hz to 3 Hz, where 88 % of the mean values were found to be greater than 0.7. This indicates that the input and output of the standing postural system were highly correlated, and consequently it was acceptable to use a linear model for the complexity of the system. The total mean value for four subjects was obtained as 0.817. However, 98 % of the mean values of the coherences were obtained to be greater than 0.7 from 0.5 Hz to 3 Hz, and the total mean value was found to be 0.845. Hence, for the frequency range from the 0.2 Hz to nearly 0.5 Hz, the coherences showed relatively lower values compare to the other frequency range.

The unknown parameters (α, β, and γ) for the frequency domain identification method, and inertia and viscoelastic parameters of the models are summarized in Table 2. The Geometric mean values of the parameters were derived by averaging the corresponding values for different subjects. The frequency response of the trunk angular velocity to the platform horizontal acceleration that corresponded to each of the four subjects had a resonance frequency between 0.60 Hz and 0.68 Hz (Fig. 3). The magnitude of the experimentally-derived transfer functions at the dominant resonance frequencies for

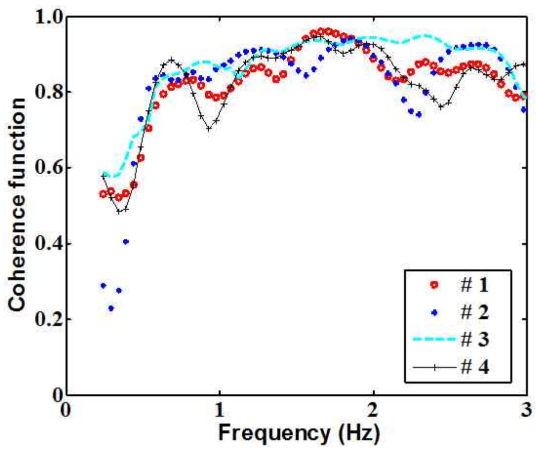


Fig. 2. The mean values of coherency functions for four subjects. Each graph is related to one subject

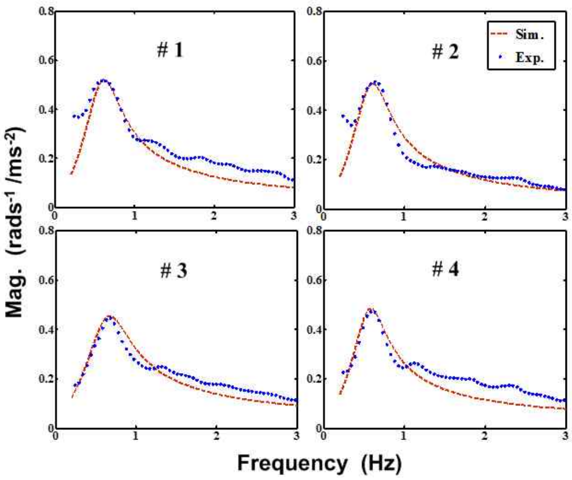


Fig. 3. Comparison of the experimental (Exp.) and simulated (Sim.) results of magnitudes of the transfer functions for four subject. Note that the # sign indicates the subject number

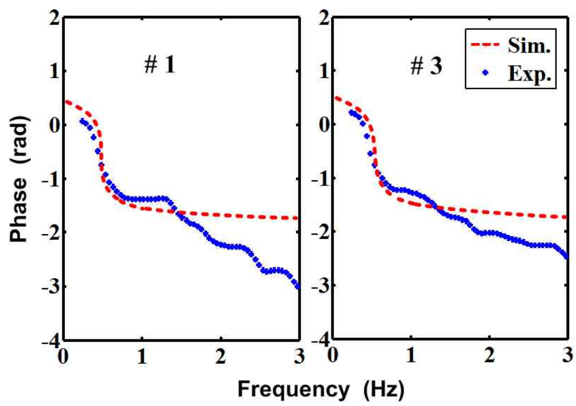


Fig. 4. Comparison of the experimental (Exp.) and simulated (Sim.) results of the phases of the transfer functions for two subjects. Note that the # sign indicates the subject number

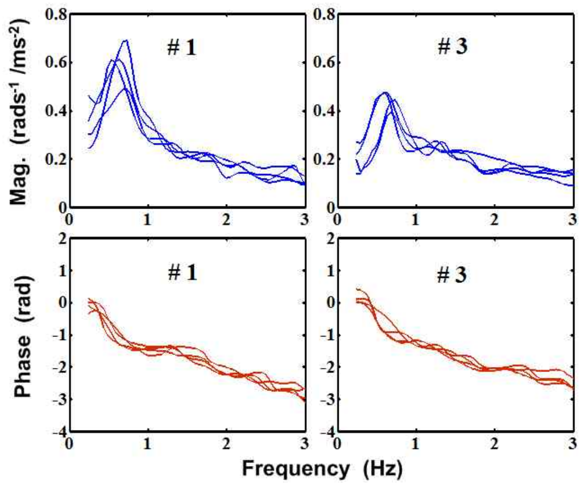


Fig. 5. Comparison of the magnitudes and phase angles of the transfer functions for two subjects. Note that the # sign indicates the subject number

subjects number 1 to 4 were found to be 0.52, 0.51, 0.44, 0.47 $rads^{-1}/ms^{-2}$, respectively. Close agreement was found between the magnitude of the frequency response functions of the models and the magnitudes of the Geometric mean values of the experimentally-derived frequency response functions.

The phase angles of the models corresponding to all four subjects were also found to be consistent with the corresponding phase angles of the Geometric mean values of the frequency response functions. Therefore, the phase angles for subjects number 1 and 3 are shown in Fig. 4. The results of the other subjects were similar to subjects number 1 and 3, showing close agreement between the experimental and simulated phase angles. The scatter of the magnitudes and phase angles of the experimentally-derived frequency response functions were small. This proved that the obtained frequency response of the complexity of the standing posture system corresponding to each of the subjects were replicable. Hence, the magnitudes and phase angles corresponding to subjects number 1 and 3 are shown in Fig. 5. Each panel of this figure consists of four graphs derived from four test runs. The variables of the results within subjects as corresponding to the other two subject were fairly similar.

4. Discussion

The coherence values were found to be relatively low in the frequencies lower than 0.5 Hz. In addition, some discrepancy has been observed between the model and the measured result in the frequency range. This was likely due to the low-frequency ($<$ 0.5 Hz) voluntary movements of the standing body, e.g., slowly changing the position of the standing body in the mid-sagittal plane, during experiments. Since the simulation of the voluntary movements of the standing dynamics is not the aim of our model, this observation cannot reduce the validity of the model at even very low frequencies.

The limitation of the excitation source that generated the input with lower signal-to-noise ratio (S/N) is the second cause of this discrepancy [12]. Therefore, the designed model presents the passive motion of the standing posture at all frequencies lower than 3 Hz.

This study preferred to apply the frequency domain identification method to the time domain method for identification of the complexity of the standing posture. As the standing posture complexity was not a fully linear system and the measured signals were corrupted by noise, the frequency domain identification method was more effective to characterize the system in an appropriate frequency band, as well as to reduce the adverse effects of the noise on the identification than the time domain methods [8, 11].

The cross-spectral function method only takes into account the linearly correlated proportion of the output with input. Therefore, it estimates the best linear approximation to the global system, including the nonlinear part of the system [8, 16].

The purpose of this study was to provide explicit results of passive behavior for the standing postural dynamics with a simplified model. It was found that a second order model (single inverted pendulum) was sufficient to simulate the system in a frequency range less than 3 Hz, where the resonance frequency of the system existed. This was the main reason why this study chose the low frequency range for identification and modeling.

The Gaussian random vibration was used as an excitation signal to the standing postural dynamics. This type of input helps the subjects to better follow the instructions in reducing vestibular, visual, and voluntary responses, and was therefore more appropriate than a periodic excitation for the purpose of this study. In addition, since it was intended to measure the open-loop responses of the complexity of the standing posture to the vibration, therefore, it was necessary to consider a non-predictive random excitation signal. The non-predictive vibration can reduce the influences of the human vestibular and somatosensory system. As clinical research shows, the contribution of the vestibular and somatosensory reflexes is negligible when the input is a non-predictive random vibration.

The viscoelastic parameters (spring and damping coefficients) of the standing posture that were found in this study are very useful, since currently little is known regarding these values in the literature. The mean values of the normalized viscoelastic parameters can be used by adjusting the values in accordance with the weight of the subject.

5. Conclusions

A simplified method was presented to measure and identify the complexity of the standing postural dynamics in response to the support platform horizontal vibration. It had a resonance frequency at around 0.60 Hz to 0.68 Hz. The the complexity of standing posture dynamics was modeled by the use of a two-dimensional single-inverted-pendulum in a frequency range lower than 3 Hz. A frequency domain identification method was used to estimate the viscoelastic parameters of the standing postural dynamics. The linearity of the system was further examined. Good conformity was obtained between the simulation and experimental data. The presented method may be used to identify the complexity of the standing postural dynamics in response to not only the support platform horizontal vibration, but also in response to the mediolateral (M/L) vibration, i.e., the frontal plane.

References

[1] R. Fitzpatrick and D. I. McCloskey, “Proprioceptive, visual and vestibular thresholds for the perception of sway during standing in humans,” J. Physiol., 478:173-186, 1994.

[2] D. A. Winter, “Biomechanics and Motor Control of Human Movement,” Toronto, John Willey & Sons Inc., 1990.

[3] I. D. Loram, and M. Lakie, “Human balancing of an inverted pendulum: position control by small, ballistic-like, throw and catch movements,” J. Physiol., 540:1111-1124, 2002.

[4] D. A. Winter, F. Pince, J. S. Frank, C. Powell, K. F. Zabjek, “Unified theory regarding A/P and M/L balance in quiet stance,” J. Neurophysiol., 75:2334-2343, 1996.

[5] P. Morasso, M. Schieppati, “Can muscle stiffness alone stabilize upright standing?,” J. Neurophysiol., 83:1622-1626, 1999.

[6] R. J. Peterka and M. S. Benolken, “Role of somatosensory and vestibular cues in attenuating visually induced human postural sway,” Exp..BrainRes., 105:101-110, 1995.

[7] Y. Fukuoka, K. Tanaka, A. Ishida, and H. Minamitani, “Characteristics of visual feedback in postural control during standing,” IEEE Trans. Rehabil. Eng., vol. 7, no. 4, pp. 427-434, 1999.

[8] R. Pintelon, and J. Schoukens, “System identification. A frequency domain approach,” IEEE Press, 2001.

[9] J. S. Bendat, and A. G. Piersol, “Engineering applications of correlation and spectral analysis,” Johan Wiley & Sons, 1980.

[10] R. Pintelon, J. Schoukens, and J. Renneboog, “The geometric mean of power (amplitude) spectra has a much smaller bias than the classical arithmetic (RMS) averaging,” IEEE Trans. Instrum. Meas., vol. 37, no. 2, pp. 213-218, 1988.

[11] I. Kollar, “Frequency domain system identification Toolbox-User's Guide,” The Mathworks, Inc., 2001.

[12] J. Schoukens and J. Renneboog, “Modeling the noise influence on the fourier coefficients after a discrete fourier transform,” IEEE Trans. Instrum. Meas., vol. IM-35, no. 3, pp. 278-286, 1986.

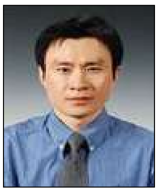
[13] R. Johansson, M. Magnusson, and M. Akesson, “Identification of human postural dynamics,” IEEE Biomed. Eng., Vol. 35, No. 10, pp. 858-869, 1988.

[14] P. De Leva, “Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters,” J. Biomechan., vol. 29, no. 9, pp. 1223-1230, 1996.

[15] P. Gatev, S. Thomas, T. Kepple, and M. Halett, “Feedforward ankle strategy of balance during quiet stance in adults,” J. Physiol., 514:915-928, 1999.

[16] J. S. Bendat, “Nonlinear systems techniques and applications,” Johan Wiley & Sons, 1997.

◇ 저자소개 ◇



Youngkyun Shin

Youngkyun Shin received B.S. and M.S. degrees in Electrical and Computer Engineering from Kangwon National University, Korea in 1997 and 1999, respectively. He is currently an instructor at Yuhan and Inha Universities, and is 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.

Gu-Bum Park

Gu-Bum Park received Bachelor, Master and Ph.D. degrees in Electrical Engineering from Inha University in 1984, 1986 and 1992, respectively. He joined the Department of Electrical Engineering, Yuhan University in 1992, where he is now a Professor. His current research interests include the fabrication of thin films as electrical devices, hard coatings and sensors.