J. Human Ergol., **29**: 15-25, 2000

THE EFFECTS OF VISUAL INPUT ON POSTURAL CONTROL MECHANISMS: AN ANALYSIS OF CENTER-OF-PRESSURE TRAJECTORIES USING THE AUTO-REGRESSIVE MODEL

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New measures to characterize center-of-pressure (COP) trajectories during quiet standing were proposed and then utilized to investigate changes in postural control with respect to visual input. Eleven healthy male subjects (aged 20-27 years) were included in this study. An instrumented force platform was used to measure the time-varying displacements of the COP under each subject's feet during quiet standing. The subjects were tested under eyes-open and eyes-closed conditions. The COP time series were separately analyzed for the medio-lateral and antero-posterior directions. The proposed measures were obtained from the parameter estimation of auto-regressive (AR) models. The percentage contributions and geometrical moment of AR coefficients showed statistically significant differences between vision conditions. The present COP displacements under the eyes-open condition showed higher correlation with the past COP displacements at longer lag times, when compared to the eyes-closed condition. In contrast, no significant differences between vision conditions were found for conventional summary statistics, e.g., the total length of the COP path. These results suggest that the AR parameters are useful for the evaluation of postural stability and balance function, even for healthy young individuals. The role of visual input in the postural control system and implications of the findings were discussed.

INTRODUCTION

Human upright stance is regulated by a complex control system involving several different sensory systems, i.e., the visual, vestibular, and somatosensory systems. This control system accommodates postural stability and steadiness during upright standing. Maintenance of postural stability is critical to everyday life. Deficiencies in the postural control system can lead to instability, falls, and injury. Many falls occur because of impaired balance function or balance disorders, while others are due solely to environmental factors. Falls are the leading cause of accidental death and injury, particularly among the elderly (Panzer et al., 1995). Due to the importance of evaluation and rehabilitation of balance function for elderly persons, many researchers have investigated age-related changes in postural stability using a variety of methods and measures (for the details of reviews, see Prieto et al., 1996). In contrast, there are few studies focusing on the effects of the characteristics of living environments on the quality of the sensory inputs participating in postural control (Cohn and Lasley, 1985; Owen, 1985; Simoneau et al., 1999). Hashizume et al. (1986) suggested that elderly persons more depend on the visual information in order to maintain postural stability comparing with young

Received for publication June 27, 2000

individuals. Deficiencies of visual function with aging may probably cause an increased incidence of falls among the elderly. Thus, studies on the role of vision in the postural control system are very important to provide new ways to reduce the risk of falling for elderly population. More scrutiny of the effects of visual input on postural control is also needed even for healthy young individuals.

In order to properly appreciate the relative contributions of the visual system to the maintenance of balance, the Romberg test has been generally used (e.g., Black et al., 1982; Paulus et al., 1984; Paulus et al., 1989). This test involves the comparison of an individual's quiet standing postural sway under eyes-open and eyes-closed conditions. Postural sway is typically quantified by using a force platform to determine the maximum displacements, root mean squared displacements, and/or total excursion traversed by the center-of-pressure (COP) under a subject's feet (Goldie et al., 1989; Kirby et al., 1987; Murray et al., 1975). It is presumed that increased postural sway, as measured by the aforementioned COP summary statistics, represents decreased postural stability or poor balance. Prieto et al. (1996) reported that many of the traditional COP measures showed significantly larger values under the eyes-closed condition for both young and elderly populations. They concluded that balance instability increases under the eyes-closed condition. Collins and De Luca (1995) reported that for nearly half of the young subjects tested in their experiments no significant differences were found between the visual conditions for any of the traditional COP measures such as the maximum displacements and total sway path. These inconsistent results may be due to lower reliability of the traditional COP summary statistics, particularly for assessment of the effects of visual input on postural control. The traditional COP summary statistics, in general, cannot be interpreted in a physiologically meaningful way. Therefore, it is necessary to develop a reliable, consistently useful approach or technique for extracting physiologically significant features from the COP time series.

In this study, we apply the auto-regressive (AR) model to the modeling of the COP time series during quiet standing. Our approach is based on an assumption that the COP time series can be modeled as a stochastic process. The AR model predicts the present output by the linear combination of the past inputs (Kantz and Schreiber, 1997). The present COP really represents the neuromuscular response to the past imbalances of the body (Winter, 1990). Thus, this time domain approach to characterize the COP profiles will provide a further understanding of the internal dynamics in the postural control system including the closed-loop feedback mechanisms.

The present study investigated vision-related changes in postural control during quiet standing. The secondary goal was to determine the usefulness of AR parameters for assessment of postural stability in young healthy individuals by examining COP time series during quiet standing.

MATERIALS AND METHODS

Subjects

Eleven healthy male subjects with similar age (mean age of 22.2 years, SD 1.83) and body dimension (mean body weight 65.7 kg, SD 9.42; mean body height 168.7 cm, SD 5.36) were included in the study. All subjects had no evidence or known history of gait, postural or skeletal disorders. Informed consent was obtained from each subject prior to participation.

Experimental procedures

Postural stability was evaluated by using a force platform (ANIMA, Model G1822S) to measure the time-varying displacements of the COP under a subject's feet. Subjects participated in a set of postural tasks. The first task (eyes-open condition) consisted of standing barefoot in a standardized stance on the platform with the arms comfortably at the sides, downward. In the standardized stance, the subject's feet were abducted 30 degrees and their heels were separated mediolaterally by a distance of about 3 cm. This stance was given to minimize the influence of foot position on the amount of postural sway and the mean position of COP (Kirby et al., 1987). Subjects were instructed to look at a red colored small object (1.5 x 1.5 cm) placed on a white screen, at eye height, about a distance

2 m apart from the platform. The second task (eyes-closed condition) was the same as the first one except that the subjects performed the task with their eyes closed and fully covered with an eye mask. The data for each postural task were collected for a trial duration of 60 s and sampled at a frequency of 50 Hz. Subjects were asked to stand still on the platform for about 5 s before data collection began for each trial. Ten trials at each of the two postural conditions were conducted for the first subject. This large number of tests was required to assess the reliability of the proposed statistical methodology. Two trials for each postural condition were conducted for each of the remaining 10 subjects. The order of the presentation of the vision conditions was counterbalanced across the subjects. Rest periods of 3 min were provided between each trial.

Figure 1-a shows a stabilogram for a representative trial under the eyes-open condition. The origin corresponds to the mean coordinates of COP trajectories along the medio-lateral (ML) and antero-posterior (AP) directions. The ML and AP time series are shown in Figures 1-b and 1-c, respectively. The COP time series data consist of displacements from the mean COP in each direction.

Mathematical models

In an AR model, the present outcome is a linear combination of the signal in the past (with a finite memory), plus additive noise, as shown in Figure 2. The AR model of order M for a time series of the COP displacements x_t is given as

$$x_t = \sum_{k=1}^{M} a_k x_{t-kT} + e_t, \qquad [1]$$

where a_k are AR coefficients, k is the number of order, M equals the maximum number of order, and T equals the data interval. e_t indicates a prediction error of the AR model. The sequence of the AR coefficients a_k ($1 \le k \le M$) is defined to minimize the squared prediction error. When the prediction error e_t is obtained from the optimum AR coefficients, it equals white Gaussian noise. The model on the basis of auto-correlation functions is given as

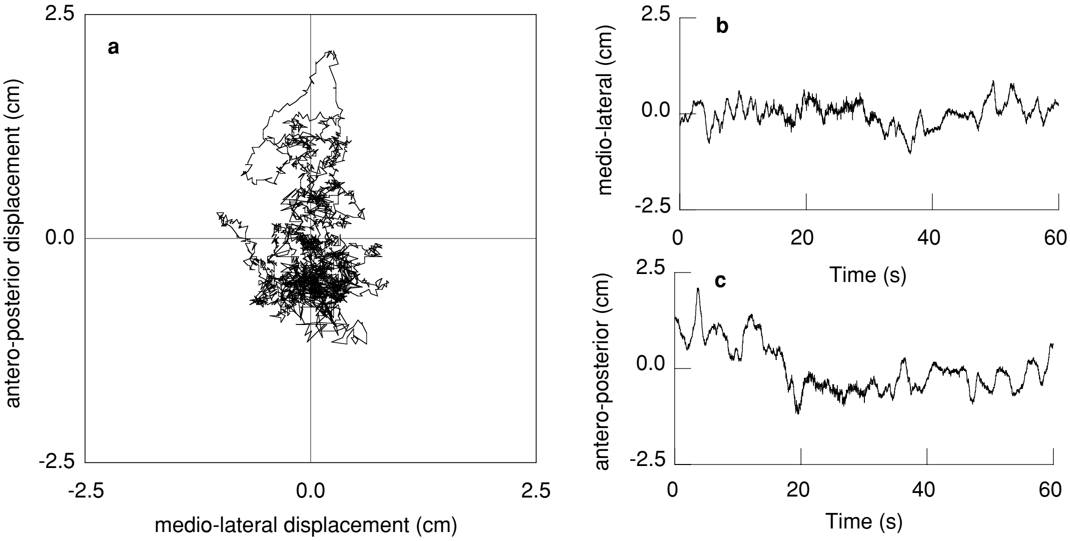

Fig. 1. (a) A typical 60 s stabilogram for a representative trial under the eyes-open condition. (b) and (c) The medio-lateral and antero-posterior directions time series corresponding to the stabilogram, respectively.

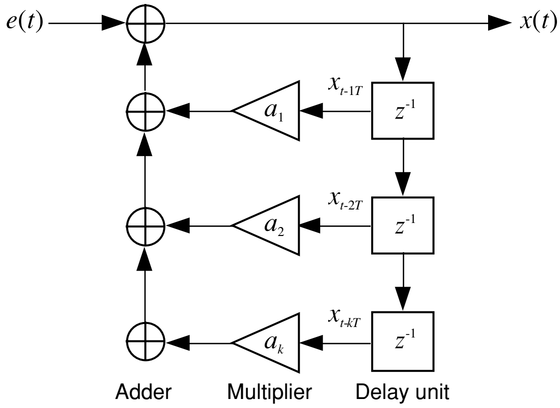

Fig. 2. A schematic diagram of the auto-regressive model.

$$r_t = \sum_{k=1}^{M} a_k r_{t \cdot t-kT} + \sigma^2, \qquad [2]$$

where $r_{t \cdot t\text{-}kT}$ are auto-correlation functions between x_t and $x_{t\text{-}kT}$, and σ^2 equals variance of the white noise, i.e., the residual variance that is "unexplained" by the independent variables in the AR model: Equation [1]. These parameters a_k, $r_{t \cdot t\text{-}kT}$, and σ^2 were estimated using Burg's formulation, where auto-correlation functions $r_{t \cdot t\text{-}kT}$ are not calculated directly from the time series x_t (Hino, 1977).

The AR model was applied to the COP displacements time series for each individual trial and condition, separately for the ML and AP directions. For a preliminary analysis, the order M of the AR models was selected by means of the Final Prediction Error (FPE) criterion (Akaike, 1970). The number of order selected based on the FPE varied from M=3 to M=14 across subjects and vision conditions. Therefore, the order of the AR models was practically fixed at M=20.

The goodness-of-fit of the AR model with order M=20 for each COP time series was also checked by calculating the FPE, i.e., the mean residual between the predicted COP data and the observed COP data. The means and standard deviations of the FPEs based on the total database (n=60; 2 trials x 2 experimental conditions x 10 subjects plus 10 trials x 2 experimental conditions x 1 subject) were 0.019 ± 0.009 cm and 0.010 ± 0.007 cm in the ML and AP directions, respectively. These values are considerably less compared with the amount of COP sway. The FPE showed no significant differences between the vision conditions for both the ML and AP directions. This suggests that the use of the AR model with 20 orders is appropriate for the analysis of the COP time series data observed here.

AR model-based measures

In Equation 2, the r_t equals 1 and then 1-σ^2 equals the total variance of the dependent variable x_t explained by the independent variables $x_{t\text{-}kT}$. The mean variances explained by the independent variables (based on all 20 orders) were 98.0% and 98.9% in the ML and AP directions, respectively. Thus, the AR coefficients and auto-correlation coefficients can be used as a reasonable description of the nature of the COP time series.

In order to examine the relationships between the dependent variable and the independent variables in the AR model, the percentage variance explained (or percentage contribution) by each independent variable with respect to the total variance was defined and calculated as follows,

$$a_k r_{t \cdot t-kT} \times 100\ (\%) \qquad (k = 1, 2, 3, \cdots 20). \qquad [3]$$

That is, these parameters represent the relative contribution of the past COP displacements $x_{t\text{-}kT}$ to the present displacements x_t.

Figure 3 illustrates mean AR coefficient vs. lag time plots for the 10-trial data set in the first subject. These AR coefficients are calculated for each of the ten trials, and the mean data for each vision condition are plotted along the ordinate. The scale of the lag time corresponds to kT (k=1, 2, 3, ... 20), where T is equal to the data interval for 20 ms (i.e., the sampling rate of 1/50 Hz^{-1}). The mean AR coefficient plots continuously decreased with the lag times from 20 to 160 ms for both the eyes-open and eyes-closed conditions. At the lag times over 160 ms, the mean AR coefficients were equal to around zero. In order to quantify differences in the decreasing tendency of the AR coefficients related with increasing the lag times, the geometrical moment of AR coefficients was calculated as follows,

$$G = \sum_{k=1}^{M} a_k kT \, . \qquad [4]$$

When the AR coefficient vs. lag time plot can be approximated to a simple decreasing curve form without any remarkable peaks, the geometrical moment value represents an averaged magnitude of the AR coefficients with the lag times. That is, the larger the geometrical moment value, the larger the magnitude of the AR coefficients at relatively longer lag times, and *vice versa*.

Once the AR model with order M=20 was fitted to each COP time series, the AR coefficients and auto-correlation coefficients were computed. They were used to calculate the percentage contributions and geometrical moment.

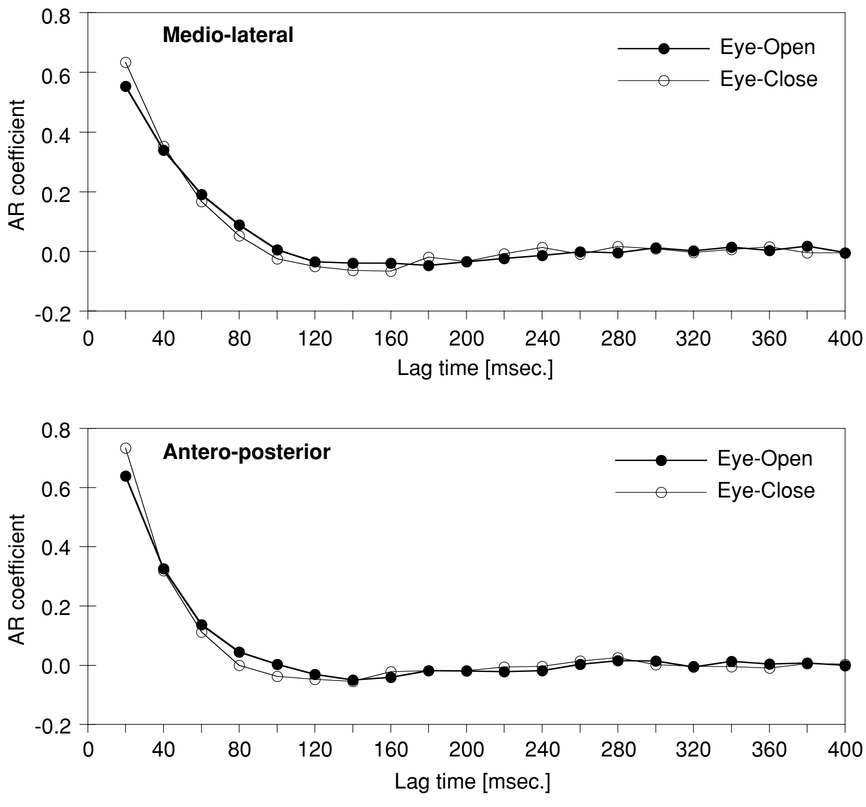

Fig. 3. Mean AR coefficient vs. lag time plots in the ML and AP directions for the 10-trial data set in a particular subject under eyes-open and eyes-closed conditions.

Conventional COP measures and statistical analyses

The following commonly used COP measures were also calculated from the stabilogram time series: the average ML distance from the mean COP, average AP distance from the mean COP, total length of the COP path, and mean velocity of the COP. These conventional COP measures and the aforementioned AR model-based measures were computed for each subject trial. Then, for the 10-subject data set the respective values were averaged for each set of two trials to obtain two resultant measures for each parameter: one for the eyes-open condition and one for the eyes-closed condition. A one-way repeated measures ANOVA was used to test for significant differences in these measures between the eyes-open and eyes-closed conditions.

RESULTS

The average ML distance from the mean COP (MLDIST), average AP distance from the mean COP (APDIST), total length of the COP path (TOTEX), and mean velocity of the COP (MVELO) representing the nature of body sway were computed for the 10-subject data set. Means and standard deviations for these measures are shown in Table 1. The mean value of MLDIST under the eyes-closed condition tended to be larger than that under the eyes-open condition. In contrast, the mean value of APDIST under the eyes-closed condition tended to be less than that under the eyes-open condition. The TOTEX and MVELO under the eyes-closed condition showed larger mean values when compared to the eyes-open condition. Nevertheless, there were no significant differences between the eyes-open and eyes-closed conditions for any of the four conventional COP measures.

The AR coefficients based on the 10-subject data set were averaged for the eyes-open and eyes-closed conditions in order to generate mean plots that would allow for a visual assessment of the typical pattern of vision-related changes in the COP time series profiles. Figure 4 displays mean AR coefficient vs. lag time plots for the 10-subject data set. The mean data for each vision condition are plotted along the ordinate. The horizontal scale in these plots has been magnified to display the variability of the AR coefficients at lag times shorter than 160 ms. The AR coefficients continuously decreased with the lag times for both the ML and AP directions. Differences between the eyes-open and eyes-closed conditions in the AR coefficients were visually identified, as indicated with the solid arrows in Figure 4. The mean AR coefficients at the lag times from 20 to 40 ms under the eyes-open condition were less than those under the eyes-closed condition. In contrast, the mean AR coefficients at the lag times from 60 to 160 ms tended to be larger under the eyes-open condition when compared to the eyes-closed condition for both the ML and AP directions. Although the lag times over 160 ms are not displayed here, the whole changing patterns of the AR coefficient vs. lag time plots were identical to Figure 3 for both the ML and AP directions. That is, at the lag times over 160 ms the values of the AR coefficients under the eyes-open condition were very similar to those under the

Table 1. Means, standard deviations, and F-ratios for conventional measurements of the COP trajectories under eyes-open and eyes-closed conditions (n=10)

	Eye-Open		Eye-Close		
	Mean	S.D.	Mean	S.D.	F -ratio
MLDIST	3.54	0.97	3.88	1.03	1.42
APDIST	5.22	1.63	4.56	1.06	2.13
TOTEX	1945.03	99.82	2069.12	276.13	2.19
MVELO	32.42	1.66	34.49	4.60	2.19

MLDIST (mm), the average medio-lateral distance from the mean COP; APDIST (mm), the average antero-posterior distance from the mean COP; TOTEX (mm), the total length of the COP path; MVELO (mm/sec), the mean velocity of the COP

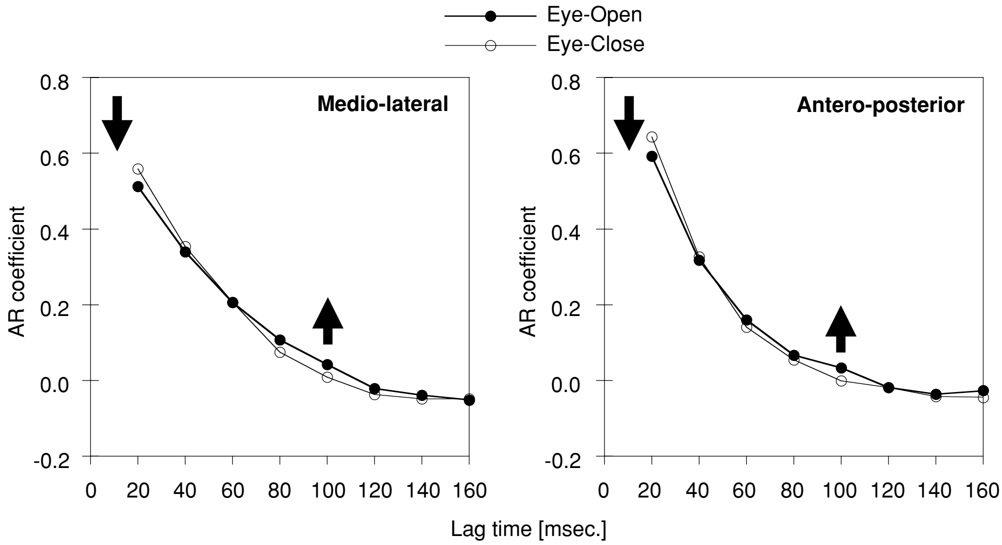

Fig. 4. Mean AR coefficient vs. lag time plots in the ML and AP directions for the 10-subjects data set under eyes-open and eyes-closed conditions. Coefficients at lag times over 160 ms were eliminated to display the variability at shorter lag times. Arrows depict the direction of the differences as viewed from the eyes-closed to the eyes-open conditions.

eyes-closed condition, and their mean values were equal to around zero.

Differences between vision conditions that were visually identified from the mean AR coefficient vs. lag time plots were statistically tested. Tables 2 and 3 show means and standard deviations for the percentage contributions at each of the lag times in the ML and AP directions, respectively, based on the 10-subject data set. The twenty percentage contributions were analyzed separately using a one-way repeated measures ANOVA. The mean percentage contributions at the lag times of 20 and 100 ms showed statistically significant differences between vision conditions in both the ML and AP directions. There were also statistically significant differences between vision conditions at the lag times of 40 and 80 ms in the ML directions. The mean percentage contributions at the lag times from 20 to 40 ms were larger under the eyes-closed condition than those under the eyes-open condition. In contrast, the mean percentage contributions at the lag times from 80 to 100 ms were remarkably larger under the eyes-open condition than those under the eyes-closed condition. The number of lag times with significant values was larger for the ML direction data than for the AP direction data, indicating that the postural control in the ML direction can be more sensitive to visual information.

Since each of the AR coefficient vs. lag time plots was approximated to a continuous decreasing curve form without any remarkable peaks, the geometrical moment was calculated. Figure 5 illustrates means and standard deviations for the geometrical moment in the ML and AP directions under the eyes-open and eyes-closed conditions. The geometrical moment showed statistically significant differences between vision conditions in both the ML and AP directions for the 10-trial data set; $F(1, 18)=27.7$ and 26.5 for the ML and AP directions, respectively, $P<0.01$ (Figure 5-a). For the 10-subject data set, statistically significant differences between vision conditions were also detected; $F(1, 9)=11.1$ and 20.4 for the ML and AP directions, respectively, $P<0.01$ (Figure 5-b). The geometrical moment values under the eyes-open condition were significantly larger than those under the eyes-closed condition. This means that the magnitude of AR coefficients under the eyes-open condition was on the average larger at relatively longer lag times, when compared to the eyes-closed condition.

Finally, the results obtained from the 10-trial data set for a particular subject were coincident with those based on the 10-subject data set, as mentioned above. This fact indicates that sample

Table 2. Means, standard deviations, and *F*-ratios for the percentage contribution (%) in the ML direction under eyes-open and eyes-closed conditions (n=10)

Lag time (msec.)	Eye-Open		Eye-Close		F -ratio [1]		ES [2]
	Mean	S.D.	Mean	S.D.			
20	50.67	6.50	55.32	5.16	12.27	**	0.75
40	33.44	2.88	34.94	2.34	5.34	*	0.56
60	20.22	2.11	20.29	1.61	0.01		0.04
80	10.44	3.90	7.29	2.90	7.84	*	0.85
100	4.12	3.09	0.80	2.61	18.23	**	1.02
120	-2.06	2.98	-3.53	2.63	2.81		0.52
140	-3.66	3.19	-4.61	2.28	1.09		0.35
160	-4.85	2.12	-4.55	2.17	0.14		0.14
180	-3.11	2.56	-5.08	2.08	3.10		0.79
200	-3.77	1.74	-2.88	2.25	0.70		0.44
220	-1.69	2.17	-1.51	1.97	0.04		0.09
240	-1.76	2.79	-0.73	1.59	0.84		0.45
260	-1.89	1.86	0.28	1.90	7.97	*	1.01
280	0.13	2.50	0.53	3.00	0.19		0.15
300	-0.35	1.85	0.59	2.43	1.50		0.44
320	-0.50	1.73	-0.25	2.62	0.08		0.11
340	0.50	1.26	1.29	2.20	1.25		0.44
360	0.27	1.49	0.64	1.38	0.26		0.27
380	0.34	1.55	0.26	1.83	0.01		0.05
400	1.78	1.69	-0.62	1.30	11.99	**	1.25

[1] Statistical significance was determined with a one-way repeated measures ANOVA (** P< 0.01, * P< 0.05).
[2] Effect size

Table 3. Means, standard deviations, and *F*-ratios for the percentage contribution (%) in the AP direction under eyes-open and eyes-closed conditions (n=10)

Lag time (msec.)	Eye-Open		Eye-Close		F -ratio [1]		ES [2]
	Mean	S.D.	Mean	S.D.			
20	58.96	7.27	63.93	7.44	13.26	**	0.65
40	31.56	3.46	32.33	3.18	0.41		0.24
60	15.85	3.57	13.91	4.99	3.04		0.45
80	6.66	3.18	5.31	2.36	2.25		0.48
100	3.27	2.71	-0.11	3.24	10.19	*	1.00
120	-1.79	3.00	-1.78	2.85	0.00		0.00
140	-3.55	2.43	-4.03	3.22	0.31		0.17
160	-2.55	2.46	-4.27	1.70	5.66	*	0.77
180	-3.72	1.55	-3.03	2.56	1.09		0.33
200	-3.55	1.95	-1.79	1.96	2.37		0.84
220	-1.55	2.04	-1.63	1.30	0.03		0.05
240	-0.92	2.08	-0.01	1.92	1.30		0.46
260	-1.44	1.49	-1.34	2.95	0.02		0.05
280	0.58	2.48	0.98	1.47	0.47		0.20
300	-0.85	2.60	-0.75	1.63	0.02		0.05
320	0.50	1.93	0.82	2.15	0.30		0.16
340	0.24	2.29	-0.20	2.10	0.28		0.20
360	0.01	1.74	1.31	1.64	4.81		0.73
380	0.63	2.45	-0.59	1.59	1.13		0.58
400	0.84	1.15	-0.09	1.25	2.21		0.74

[1] Statistical significance was determined with a one-way repeated measures ANOVA (** P< 0.01, * P< 0.05).
[2] Effect size

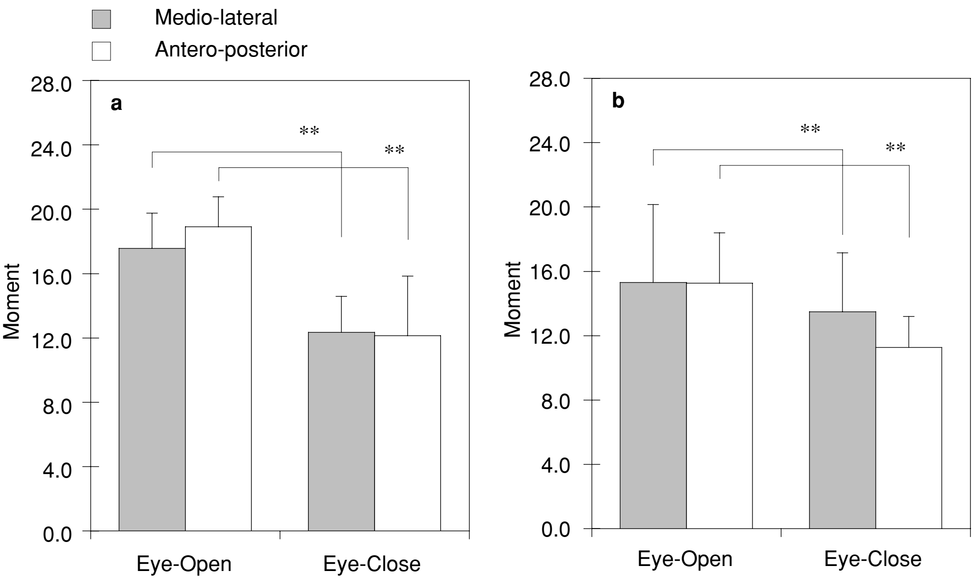

Fig. 5. Means and standard deviations for the geometrical moment in the ML and AP directions under eyes-open and eyes-closed conditions. (a) Differences between vision conditions for the 10-trial data set in a particular subject. (b) Differences for the 10-subject data set. The symbol ** denotes statistically significant differences at P<0.01 levels.

related factors contributed less to the differences between vision conditions with respect to characteristics of the COP profiles.

DISCUSSION

There were no significant differences between the eyes-open and eyes-closed conditions for any of the four different traditional COP measures. This finding does not support the conventional notion that postural sway necessarily increases under eyes-closed conditions. In many past studies, the balance tests using traditional COP measures have been performed on unstable and/or unusual foot positions such as the Romberg stance. The standardized stance in the present examination, where the subject's feet were abducted and their heels were separated mediolaterally, seems to be more stable than that in previous studies. It appears that the traditional COP measures are not useful for assessment of the effects of visual input on postural control during standing in usual stances. Thus, it is necessary to develop an alternative approach or technique to address how visual information affects postural control during quiet standing.

In this study, new measures of the COP based on the AR model, i.e., the percentage contributions and geometrical moment of AR coefficients were proposed. These measures detected the effects of visual input on the performance of the postural control system. The vision-related changes in the AR parameters were observed within 100 ms. The AR coefficients under the eyes-open condition showed relatively smaller values at the lag times ranging from 20 to 40 ms and larger values at the lag times from 60 to 100 ms than those under the eyes-closed condition. Nashner and Berthoz (1978) established that visual input influences involuntary postural adjustment within around 100 ms. This latency is considerably shorter than that of voluntary responses to a stimulus. The latency of musculoskeletal responses based on the proprioceptive system is generally within 30-80 ms. With this information, it is reasonable that the vision-related changes in the AR parameters within 100 ms reflect alterations in the performance of the closed-loop feedback mechanisms involved in maintaining balance. The tendency that the AR coefficients decreased at shorter lag times and increased at longer lag times under the eyes-open condition, was quantified with the geometrical moment of AR coefficients. The geometrical moment values significantly increased under the eyes-open condition. This means that the present COP displacements under the eyes-open condition have higher correlation with the past COP displacements at longer lag times, when compared to the eyes-closed condition. That is, under the visual condition, the COP position moving in a particular direction in the past tended to continue in the same direction for longer times in the future. These results suggest that visual input could enlarge the spatiotemporal threshold of COP fluctuations, after which corrective feedback mechanisms come into play.

We interpret the present results as suggesting that visual input affects the operational characteristics of the other closed-loop postural control mechanisms, i.e., the proprioceptive and/or vestibular systems. A possible schematic diagram for the postural control model is given in Figure 6. This is partly modified from that proposed by Collins and De Luca (1995). It appears that input from the visual system principally decrease sensitivity to sensory information from the two remaining systems, via a gain modulation factor (g). The gain modulation factors, which may be related with visual information processing in the central nervous system, provide appropriate levels of the sensitivity to sensory information from the proprioceptive and/or vestibular systems. If g<0, e.g., in unstable situations so that the postural control systems should respond to suddenly increased postural perturbation as quickly as possible, the sensitivity will be set to a higher level. On the other hand, if g>0, e.g., in relatively stable situations such as quiet standing, the sensitivity will be decreased. In this scenario, with visual information during undisturbed stance, inputs from the proprioceptive and/or vestibular systems may have been adaptively re-weighted, and thereby reduced their reliance on these feedback systems to correct for ML and AP displacements. The current investigation reinforces this conclusion in that visual input could change magnitudes of correlation between the present postural displacements and the past postural sway in both the ML and AP directions. This hypothesis is consistent with the results of a number of dynamic posturographic studies (Ishida and Imai, 1980; Nashner and Berthoz, 1978).

Collins and De Luca (1995) further hypothesized that an eyes-open balance strategy decreases the stiffness of the musculoskeletal system. The importance of muscle stiffness as a load-compensating mechanism for the regulation of balance has been emphasized. Sato and Fujita (1998) have established that visual input reduces the activity level of lower leg muscles and hence the ankle joint stiffness is decreased during disturbed stance. Reduced stiffness may be accomplished by reducing

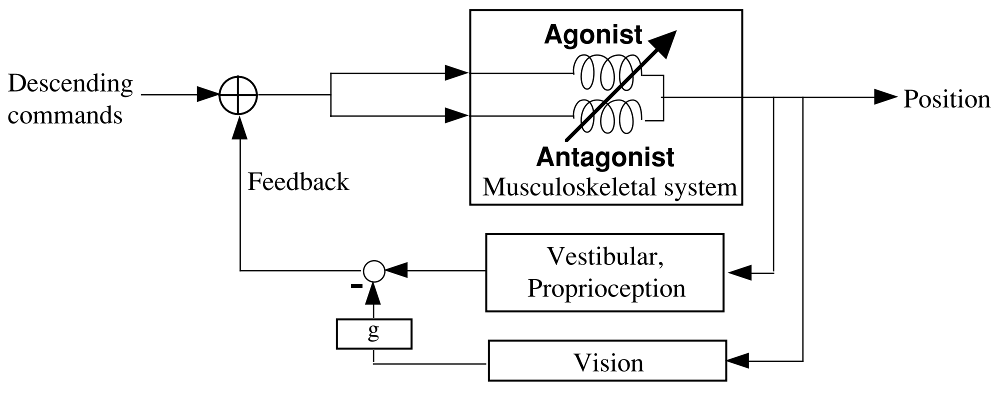

Fig. 6. A possible schematic diagram of how the visual system is integrated into the postural control system. Visual input affects the operational characteristics of the other closed-loop postural control mechanisms, i.e., the proprioceptive and/or vestibular systems. g is a gain modulation factor (partly modified from the model by Collins and De Luca, 1995).

the gain of either or both of the proprioceptive and/or vestibular systems under the vision condition. More physiologically-oriented studies on the relationship between muscular activities and COP displacements during quiet standing are needed to address this hypothesis, especially relating to the latency of muscular responses in controlling posture.

Finally, it should be noted that the present methodology includes a number of limitations: (1) validity of the data sampling rate has not been completely tested on other subjects and (2) application of the AR model parameters is restricted to the lag times ranging from 20 to 400 ms. Within these limitations, the present findings strongly suggest that the AR model is an useful approach for identifying the effects of visual input on the COP profiles during quiet standing, even for young populations. Continuing studies on numerical analysis of the COP time series, utilizing diverse experimental conditions with respect to visual information, will further clarify the role of vision in the postural control system and hence provide new ways to reduce the risk of falling for elderly persons.

This study was supported by a Grant-in-aid for Encouragement of Young Scientists from Japan Society for the Promotion of Science (#12780011). The authors would like to thank Dr. Morihiko Okada, Center for Tsukuba Advanced Research Alliance, Institute of Health and Sport Sciences, University of Tsukuba, who provided available a force platform.

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