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ORIGINAL ARTICLE

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# Balance control and adaptation during vibratory perturbations in middle-aged and elderly humans

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**Abstract** The objective was to investigate if healthy elderly people respond and adapt differently to postural disturbances compared to middle-aged people. Thirty middle-aged (mean age 37.8 years, range 24–56 years) and forty healthy elderly subjects (mean age 74.6 years, range 66–88 years) were tested with posturography. Body sway was evoked by applying pseudorandom vibratory stimulation to the belly of the gastrocnemius muscles of both legs simultaneously. The tests were performed both with eyes open and eyes closed. The anteroposterior body sway was measured with a force platform and analyzed with a method that considers the adaptive changes of posture and stimulation responses. The results showed that middle-aged people generally used a different postural control strategy as compared to the elderly. The elderly responded more rapidly to vibratory perturbation, used more high-frequency (>0.1 Hz) motions and the motion dynamics had a higher degree of complexity. Moreover, the elderly had diminished ability to use visual information to improve balance control. Altogether, despite having an effective postural control adaptation similar to that of middle-aged people, the elderly had more difficultly in withstanding balance perturbations. These findings suggest that the balance control deterioration associated with aging cannot be fully compensated for by postural control adaptation.

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## Introduction

Humans are continuously faced with new postural control tasks to which they have to adapt and adjust. To detect the movements and co-ordinate voluntary and reflexive muscle responses, the balance system uses sensory information from the visual, vestibular and somatosensory receptors (Johansson and Magnusson 1991; Kleiber et al. 1990). However, decline in the function of the sensory systems is a part of the aging process. The visual system deteriorates (Baloh et al. 1993; Magnusson and Pyykkö 1986) and proprioception (Skinner et al. 1984) and vibration sensation in the lower limbs decrease (Steinberg and Graber 1963). Signs of vestibular dysfunction are also fairly common in elderly individuals (Enrietto et al. 1999). Aging is associated with diminished muscle strength (Thelen et al. 1996), slower ankle joint torque development (Thelen et al. 1996), changes in posture (Woodhull-McNeal 1992) and lower postural stability (Lord and Ward 1994; Wolfson et al. 1992). It is therefore probable that the elderly have a decreased ability to adapt to a disturbance of posture because of the deterioration in the receptor systems and the poorer ability to perform motions.

The study of orthograde posture regulation constitutes an essential topic of motor control because of the universal importance of the mechanisms involved. They are used not only to maintain the static posture, but also to ensure body stability during various locomotory movements (Gurfinkel et al. 1995). Human postural control is maintained by somatosensory, vestibular and visual feedback, integrated within the locomotor and central nervous (CNS) systems. Research about postural control dynamics and stability therefore involves studies of the mechanical aspects of the human body, its sensory systems, and the principles governing coordination in motor control

(Johansson and Magnusson 1991). The manner in which the CNS might use adaptive adjustments to reduce the likelihood of balance loss, during the performance of a postural task in which a perturbation may or may not occur, is of particular relevance to fall prevention (Pai and Iqbal 1999; Pavol and Pai 2002). It is well known that the CNS employs both feedforward (predictive) and feedback (reactive) control to compensate for the perturbations that might occur during movement. Appropriate feedforward compensations, based on an adaptive internal model of the system (Wolpert et al. 1995) and the expected external conditions, can greatly reduce the magnitude of the reactive responses required (Pavol and Pai 2002).

To quantify the adaptive adjustments in biomechanics and postural control we used an approach describing postural control as a dynamic feedback control using system identification procedures. The method provides a mathematical model of the relationship between induced disturbances and counteractive postural adjustments (Fransson et al. 2000). The perturbations were induced by vibration, which, when applied to a muscle or a muscle tendon, increases the firing of the muscle spindles, thus signaling that the muscle is being stretched (Matthews 1986). The stimulated muscle responds to this with a reflexive contraction (tonic vibratory reflex) (Goodwin et al. 1972). Vibration applied to the calf muscles during upright stance induces anteroposterior body sway in healthy subjects (Eklund 1973). The muscular reflexes to vibratory stimulation do not undergo any noticeable impairment in the lower limbs dependent on the subject's age. However, the intensity of the postural responses may decrease with age, which is believed to be caused by deterioration in the higher postural control levels (Quoniam et al. 1995). Repeated exposure to vibratory proprioceptive stimulation generally leads to an adaptive process that gradually decreases the vibration-induced body sway (Fransson et al. 2000; Tjernström et al. 2002).

The aim of this study was to investigate if healthy elderly people respond and adapt differently to a postural disturbance compared to middle-aged people, and examine to what extent the postural control strategy in the elderly and adaptation are affected by the availability of visual information.

## Methods

### Subjects

Posturographic tests were performed on 30 middle-aged subjects (18 men and 12 women; mean age 37.8 years, range 24–56 years) and on 40 elderly subjects (13 men and 27 women; mean age 74.6 years, range 66–88 years). The elderly were recruited as volunteers from two senior citizen organizations on the grounds that they considered themselves to be in good health. The subjects had no history of vertigo, central nervous disease, or injury to the lower extremities. At the time of the investigation, none of the subjects were on any form of medication that could affect their balance or had consumed alcoholic beverages for at least 24 h. Before the experiments, all subjects gave written informed consent. The experiments were performed in accordance with the Helsinki declaration of 1975 and approved by the local ethical committee.

### Apparatus

The body sway was induced by simultaneous vibratory stimulation to the belly of the gastrocnemius muscles of both legs. The vibrators were constructed as cylinders 0.06 m in length and 0.01 m in diameter, and kept in place around the calf muscles by straps. The vibratory amplitude was 1.0 mm and the frequency 85 Hz. Forces and torques imposed on the supporting surface were recorded with six degrees of freedom by a force platform. Data were sampled at 10 Hz by a computer equipped with an AD converter and a customized program controlled sampling and stimulation.

### Procedure

The subjects were instructed to stand erect but not to attention, their feet at an angle of about 30° open to the front. Two tests were conducted on each subject; one test where the subjects were instructed to focus on a mark on the wall at a distance of about 1.5 m, and a second test where the subjects were instructed to stand with their eyes closed during the test. Before the vibratory stimulation started, spontaneous sway was recorded for 30 s. The vibratory stimulation were executed according to a computer-controlled pseudorandom binary sequence (PRBS) schedule (Johansson 1993) for 205 s by turning on/off the vibratory stimulation. The PRBS schedule was composed of stimulation shift periods with random duration of between 0.8 and 6.4 s, which yielded an effective bandwidth of the test stimulus in the region of 0.1–2.5 Hz. Thus, the designated PRBS stimuli cover a broad power spectrum and the randomized stimulation reduces the opportunity to make anticipative and preemptive adjustments. The subjects stepped down from the force platform and relaxed for 3 min between the tests.

### Analysis

The measured anteroposterior torque during vibratory stimulation was analyzed with a method that considered the adaptation of postural control. The adaptation analysis method, where multiple time-variant dynamical and biological changes are quantified by iteratively estimated non-linear functions, is described in detail elsewhere (Fransson et al. 2000, 2002). The adaptation analysis method is not sensitive to changes in anthropometrics parameters such as the height and weight of the test subjects. The modeling technique used aims to describe the adaptation of posture as well as the adjustments of stimulation responses during the first 100 s of exposure to stimulation. This information was used to estimate a time-invariant feedback control model that mathematically describes the relationship between the stimulation and measured body sway responses (Fransson et al. 2000). The three components of feedback control, postural and stimulus adaptation are separated in an identification procedure with five steps.

*Step 1: preliminary feedback model*

A third-order ARMAX (Auto Regressive Moving Average with eXternal input) model (D'Asso and Houpis 1988; Johansson 1993) (A, B, C polynomials are of third order) was used to estimate a preliminary feedback model between (input) stimulation and (output) measured torque responses.

*Step 2: preliminary posture adaptation model*

The feedback model contribution in step 1 was removed from original measured torque data and the slow changes of the remaining output data were described by the "Posture adaptation" function (see Eq. 1).

*Step 3: stimulus adaptation model*

The slow changes described by the "Posture adaptation" function in step 2 were removed from the original measured torque. The remaining data were rectified and used for an estimation of the changes in the stimulation–response relationship as described by the "Stimulus adaptation" function (see Eq. 1 below).

*Step 4: feedback model*

Based on the results from steps 2 and 3 the (input) stimulation and (output) torque responses were thereafter compensated for adaptation and used in an estimation of a feedback model describing the steady-state relationship. The input signal amplitude was altered by superimposing the changes found in the stimulus–response according to the "Stimulus adaptation" function. The output signal was modified by removing the changes described by the "Posture adaptation" function. The feedback model was evaluated with increasing model orders until its performance fulfilled the $\chi^2$ criteria of white noise properties. The optimum time delay between input and output was found by using the Akaike Final Prediction Error (FPE) (D'Asso and Houpis 1988; Johansson 1993).

*Step 5: posture adaptation model*

The feedback model contribution in step 4 was removed from original measured data and the slow changes of posture were finally determined by a renewal estimation of the "Posture adaptation" function (see Eq. 1).

The results from the adaptation analysis method can be divided into three categories:

- Adaptation of body leaning and induced body sway by the repeated stimulation.
- Motion dynamics and motion complexity.
- Stimulus-induced body sway and spontaneous body sway

## Adaptation of body leaning and induced body sway by the repeated stimulation

Two exponential functions were used to describe the adaptive changes in response amplitudes and slow changes in posture. The "Stimulus adaptation" function describes the adaptive changes in body sway amplitude induced by the repeated stimulation over time and the "Posture adaptation" describes the slow adaptive change of posture, such as adopting new body leaning. The function modeling gain adaptation consists of a sum of two exponential terms and is formulated as:

$$Y = A_1 e^{-t/\tau_1} + A_2 e^{-t/\tau_2} + C \tag{1}$$

where $\tau_1$ and $\tau_2$ denote the time constants (in s), the exponential term with the shortest time constant subscripted "1" and the other "2"; $A_1$ and $A_2$ denote amplitude (in Nm); $C$ is a constant term (in Nm); and $Y$, the measured adjustment pattern (in Nm). The parameters obtained were evaluated and terms with negligible or time-invariant influence were removed before the statistical evaluation. The number of negligible or time-invariant terms was used to classify whether the adjustment pattern had properties that were best described with a constant value, one time constant or two time constants (Fig. 1). An exponential term was considered time-invariant if the time constant was longer than 100 s or shorter than 0.1 s (one sample interval). A term was also considered of negligible influence if the magnitude was more than 100 times lower in gain than the other exponential term presupposed that the other exponential term had a time constant within the acceptable time range ( > 0.1 s to < 100 s). If both exponential terms, according to the two criteria above, were considered time-invariant or had negligible influence the adjustment pattern was classified to be best described with a constant value. If one exponential term was excluded according to the criteria the pattern was classified to be best described with one time constant and subsequently if none of the terms were excluded according to the criteria the pattern was classified to be best described with two time constants.

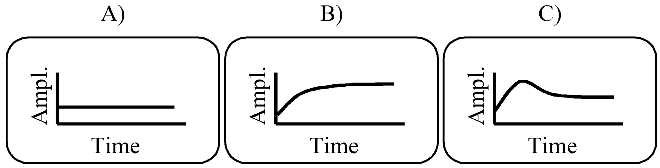


**Fig. 1A–C** Schematic examples of adjustment patterns classified by **A** a constant value, **B** one time constant, **C** two time constants

## Motion dynamics and motion complexity

The relationship between the stimulation and the recorded body sway responses was described with an ARMAX feedback model, see identification procedure step 4 (Fransson et al. 2000). This model evaluated the dynamical properties of the movements induced by the stimulation and estimated the latency between the individual stimulation pulses and recorded motion responses. The model also evaluated the dynamical complexity of the body sway induced by the stimulation, in terms of the degree of parameters needed to describe the relationship between stimulation and motion responses.

The dynamics of the estimated ARMAX feedback model were analyzed in terms of three normalized dynamical parameters, swiftness, stiffness and damping, which was obtained by normalization of the parameters from a third-order ARMAX model (Johansson et al. 1988). If the estimated feedback model was of higher model order, model-order reduction (Johansson 1993) was used to obtain a third-order ARMAX model before the normalization procedure. The dynamical parameters correspond to the parameters of a PID (i.e., proportional, integrative and derivative) control used in automatic control theory (D'Asso and Houpis 1988; Johansson 1993). Swiftness corresponds to the integrative control and a high swiftness value means that the adjustments to a disturbance are rapid and that the subject quickly returns to the chosen equilibrium body position after a perturbation. Stiffness describes the reaction to a deviation from the assumed equilibrium position and a high stiffness value means that the subject reacts strongly to a small deviation of body position. Damping describes the control action dependent on the velocity of the body sway and a high damping value means fewer oscillations of lower velocity around the chosen equilibrium position after a perturbation.

The complexity of the body movements induced by the stimulation are reflected in the degree of A, B and C polynomials in the feedback model. A feedback model with few parameters is sufficient to describe the movements if the body moves strictly like a single-link inverted pendulum during the period analyzed. However, it is necessary to increase the degree of model parameters if the body movements contain multisegmental motions in hip and knees or if the movement characteristics are changed by adaptation during the test period.

## Stimulation-induced body sway and spontaneous body sway

The body sway content was evaluated by three variance ratio values (Fransson et al. 2000, 2002). The stimulation-induced sway value shows the proportion of total measured torque $y(t)$ that can be explained by the analysis method as responses to the stimulation and as adaptive changes. The quotient defining the amount of stimulus-induced sway $V_{si}$ (Eq. 2) is calculated from the variance of measured torque $y(t)$ and variance of the model error $y_e(t)$, i.e., the remaining part of the body sway that cannot be explained by the model in terms of posture motion $y_p(t)$ and stimulus-response motion $y_{feed}(t)$.

$$\begin{aligned} y_e(t) &= y(t) - y_{feed}(t) - y_p(t) \\ v_{si} &= \frac{var[y(t)] - var[y_e(t)]}{var[y(t)]} \end{aligned} \tag{2}$$

The spontaneous body sway is the remaining part of the body sway, which is not related to the stimulation or to adaptive adjustments. The spontaneous high-frequency motion value shows the proportional size of the spontaneous body sway due to high-frequency motions above 0.1 Hz. Somewhat arbitrarily, we chose to consider motions above 0.1 Hz as high-frequency motions and motions below 0.1 Hz as low-frequency motions. A reason for this frequency choice is that the cut-off frequencies of the vestibular and visual sensory systems are around 0.1 Hz (Diener and Dichgans 1988; Diener et al. 1986). The quotient defining the spontaneous high-frequency motion value $V_{fe}$ (Eq. 3) is calculated from the variance of the model error $y_e(t)$ and variance of the high-frequency part of the error above 0.1 Hz. The high-frequency data $y_{fe}(t)$ are extracted by using a fifth-order low-pass filter with a cut-off frequency of 0.1 Hz. The filter uses a Butterworth FIR design (Proakis and Manolakis 1989) and the filtration is performed twice, once forward and thereafter reversed to achieve a zero-phase distortion.

$$V_{fe} = \frac{var[y_{fe}(t)]}{var[y_e(t)]} \tag{3}$$

The residual rate values $V_r$ (Eq. 4) describe the prediction performance of the estimated model from the variance of measured torque $y(t)$ and the variance of the feedback model residual $\epsilon(t)$. A higher residual rate value indicates that a larger part of the recorded body sway is not induced by the individual stimulation pulses, thus that a larger part of the body sway is either spontaneous sway or adaptive adjustments. The presence of any information remaining in the residuals is a clue that the model might be insufficiently complex or otherwise inappropriate.

$$V_r = \frac{var[\varepsilon(t)]}{var[y(t)]} \tag{4}$$

Statistical analysis

The differences between the middle-aged and elderly groups were analyzed with the Mann-Whitney non-parametric test and the difference between tests performed with eyes closed and with eyes open was analyzed with the Wilcoxon non-parametric test. Non-parametric tests were used since the values were not normally distributed after logarithmic transformation (Altman 1991). Normality of distribution was tested with the Shapiro-Wilk test. In all tests $p < 0.05$ was considered to be statistically significant. The exponential term with subscript 2 was not statistically evaluated since this term was excluded in approximately 40% of the cases.

## Results

At the onset of vibratory proprioceptive stimulation the subject move their center of pressure backwards compared to the quiet stance posture (Fig. 2). This backwards lean is then gradually decreased during the first 30–40 s of stimulation to a posture where the subject leans somewhat more forwards than the initial quiet stance posture. The amplitude of the individual stimulation-induced sway responses is apparently reduced during the first 30–40 s. Both these biomechanical adaptive adjustments were described by the analysis method.

### Adaptation of body leaning and induced body sway by the repeated stimulation

The “Stimulus adaptation” parameters in Table 1 describe the adaptive changes in body sway amplitude induced by the repeated stimulation over time and the “Posture adaptation” parameters describe the slow adaptive change of posture, such as adopting new body leaning. The variations in time constants, amplitudes and in the adjustment complexity indicate a substantial inter-individual variation in the way the adaptive

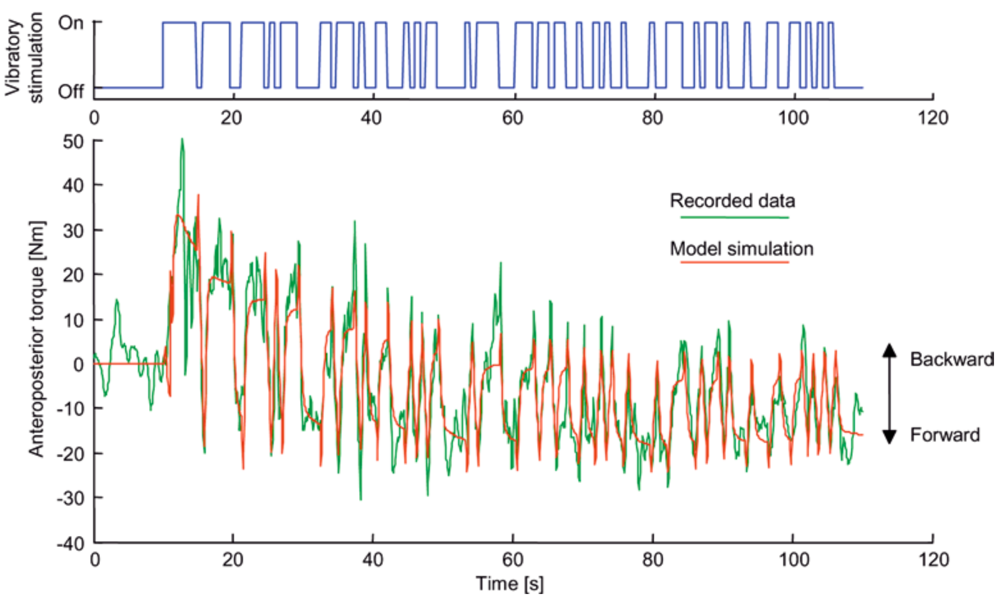


**Fig. 2** Model simulation values and measured torque in the anteroposterior direction from an elderly subject exposed to vibratory stimulation with her eyes closed. Note the change in center of gravity during the initial phase of the stimulation sequence and the reduced body sway responses to the stimulation over time. Observe also the high accordance between the model simulation values from the adaptation analysis method (*red*) and recorded body sway (*green*)

**Table 1** Mean and standard error of mean (SEM) for the absolute amplitude and time parameter values across subjects obtained from the "Posture adaptation" and "Stimulus adaptation" functions for the middle-aged and elderly groups. The adjustment pattern value shows the percentage of (second-, first-, zero-order) patterns during test conditions and for the age group, i.e., the percentage of adaptation patterns best described by a function with two time constants, one time constant or a constant value

| Adaptation type | Age group | Amplitude $A_1$ | Time constant $\tau_1$ | Amplitude $A_2$ | Time constant $\tau_2$ | Adjustment pattern (2,1,0) |
|---|---|---|---|---|---|---|
| Posture adaptation | | | | | | |
| Closed | Middle-aged | 35.9 (9.8) | 6.0 (1.6) | 36.8 (19.1) | 42.3 (5.8) | (47, 40, 13) |
| | Elderly | 32.2 (8.0) | 9.1 (3.2) | 34.3 (17.6) | 59.9 (6.2) | (33, 57, 10) |
| Open | Middle-aged | 47.6 (12.3) | 12.5 (3.7) | 55.3 (15.4) | 37.1 (5.2) | (68, 16, 16) |
| | Elderly | 32.0 (7.9) | 14.6 (3.4) | 57.0 (25.6) | 42.9 (6.9) | (30, 65, 5) |
| Stimulus adaptation | | | | | | |
| Closed | Middle-aged | 32.6 (7.5) | 9.9 (2.4) | 29.9 (8.6) | 21.8 (4.1) | (80, 20, 0) |
| | Elderly | 22.4 (4.9) | 8.0 (1.7) | 20.0 (6.6) | 35.2 (4.7) | (63, 32, 5) |
| Open | Middle-aged | 29.6 (8.7) | 11.9 (1.8) | 32.6 (10.8) | 24.3 (4.8) | (70, 23, 7) |
| | Elderly | 25.8 (5.8) | 11.6 (1.9) | 27.0 (7.5) | 29.5 (4.4) | (70, 27, 3) |

adjustments were performed. However, the properties of adjustments of posture ("Posture adaptation", Table 1), and those of the adaptive adjustments of the stimulation response ("Stimulus adaptation", Table 1) were within the same range for middle-aged and elderly people. Thus, the statistical comparison of the parameter values showed no significant differences in amplitude and time constants between the middle-aged and elderly groups in any of the tests.

There was a clear trend of shorter adaptation time constants with eyes closed compared to eyes open (Table 1). The "Posture adaptation" time constant was significantly shorter with eyes closed compared with eyes open for the elderly ($p < 0.05$) and time constant values for the "Stimulus adaptation" were significantly shorter with eyes closed compared to eyes open for the middle-aged ($p < 0.05$).

## Motion dynamics and motion complexity

The dynamical parameter values show that the elderly and middle-aged subjects utilized visual information differently (Fig. 3). The elderly responded more rapidly (swiftness parameter) to the perturbation with eyes closed compared to eyes open ($p < 0.05$). The average swiftness value was 54% larger with eyes closed compared to eyes open in the elderly group. Moreover, the stiffness, i.e., the responses to the induced body deviations, was on average 31% higher with eyes closed for the elderly compared to the middle-aged though this difference was not significant.

The most prominent difference between middle-aged and elderly people was that the response latency both with eyes closed and with eyes open was on average 43% shorter for the elderly subjects ($p < 0.001$) (Fig. 4). Moreover, the elderly were not able to reduce the motion complexity with eyes open as much as the middle-aged (A, B polynomials $p < 0.01$; C polynomial $p < 0.05$). However, the motion complexity with eyes closed was similar for elderly and middle-aged people.

## Stimulus-induced body sway and spontaneous body sway

The stimulation induced proportionally in percentage terms the same amount of body sway and adaptive

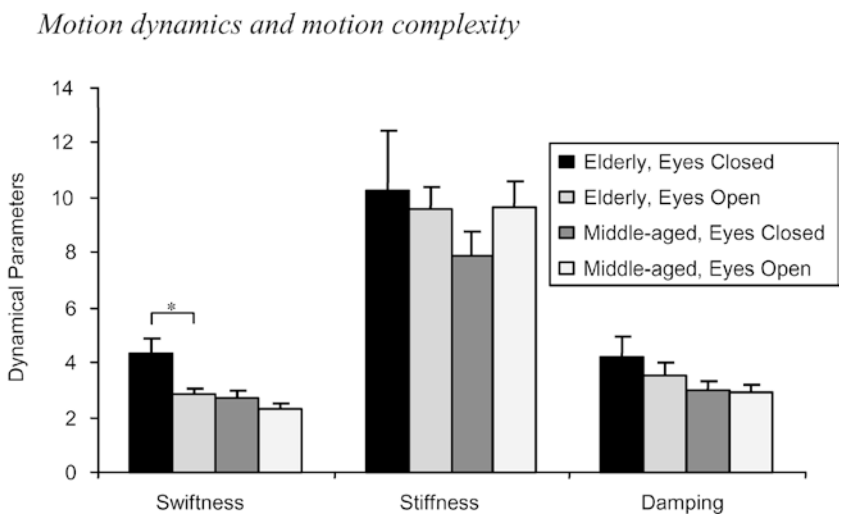


**Fig. 3** The properties of the body sway dynamic when described by the parameters swiftness, stiffness and damping (mean and standard error of mean values, $^{*}p < 0.05$, $^{**}p < 0.01$ and $^{***}p < 0.001$). The dynamics of the body sway were in several cases different between middle-aged and elderly subjects when comparing the responses during eyes open and eyes closed tests

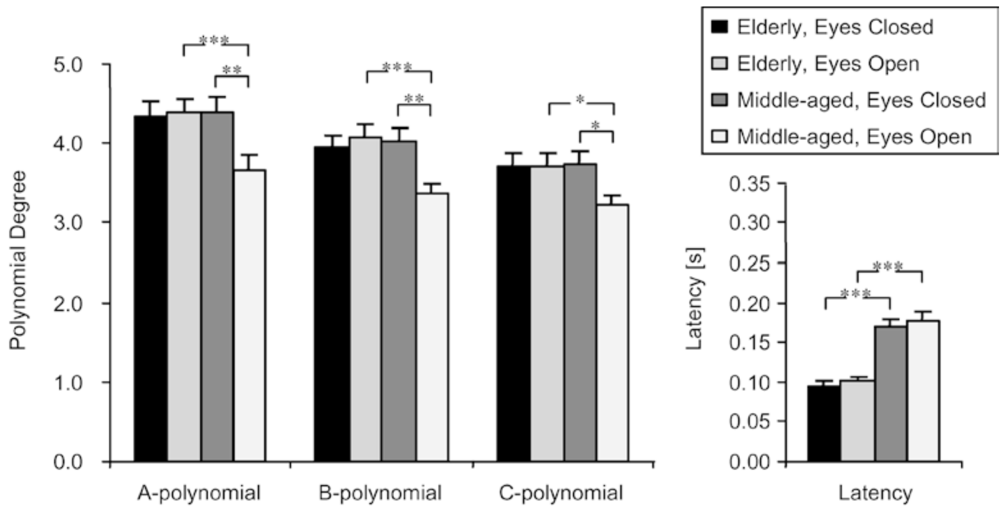


**Fig. 4** The response latency and the motion complexity as reflected by the degree of the A, B and C polynomials needed to describe the complexity of the induced body sway (mean and standard error of mean values, $^{*}p < 0.05$, $^{**}p < 0.01$ and $^{***}p < 0.001$). Both the response latency and motion complexity were significantly different between elderly and middle-aged subjects

adjustments in the elderly and middle-aged subjects (Fig. 5). However, the larger residual rate values and the larger spontaneous high-frequency motion values for the elderly suggest a difference in body sway composition. The residual rate value suggests that proportionally more of the measured body sway in the middle-aged subjects is directly caused by the stimulation and the individual vibratory pulses than in the elderly ($p < 0.001$). The residual rate value was both with eyes closed and with eyes open on average about 108% larger in the elderly compared with the middle-aged subjects. Moreover, the spontaneous body sway variance contained on average between 13% (eyes closed) and 25% (eyes open) less high-frequency motions in the middle-aged group than in the elderly group ($p < 0.001$).

Both elderly and middle-aged subjects used proportionally between 28% (elderly) and 41% (middle-aged) less high-frequency motions during trials with eyes open compared to during trials with eyes closed ($p < 0.001$) and the body sway was during trials with eyes open to a larger extent mostly caused by the stimulation ($p < 0.001$). However, the elderly could not use visual information to change the motion response pattern as much as the middle-aged subjects did ($p < 0.001$).

## Discussion

In daily life, elderly and middle-aged people are continuously faced with new control tasks and new environmental constraints to which they have to adapt and adjust. Our findings in this study imply that the elderly respond to postural disturbances in a way that both quantitatively and qualitatively differs from the way middle-aged people respond. The amount of body sway induced by the stimulation and by adaptive adjustments was proportionally, in percentage terms, about the same for the elderly and middle-aged subjects. However, the variance ratio values, the motion complexity values and the shorter latency time between stimulation and the

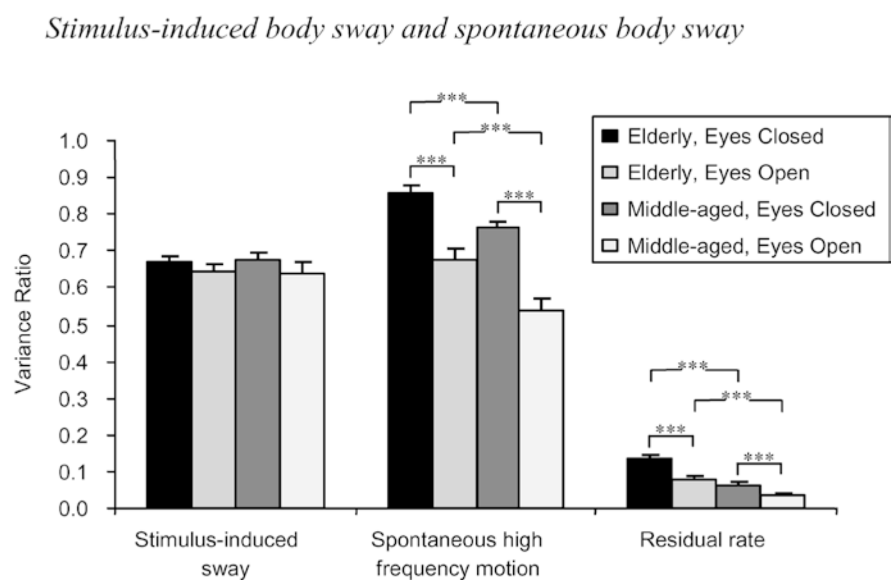


**Fig. 5** Stimulation-induced and spontaneous body sway as reflected by variance ratios (mean and standard error of mean values, $^{*}p < 0.05$, $^{**}p < 0.01$ and $^{***}p < 0.001$), see Eqs. 2, 3 and 4. The variance ratios were significantly different between middle-aged and elderly subjects and between eyes open and eyes closed tests

motion response suggest that elderly use a more complex motion pattern to withstand the perturbations and maintain balance, which contain more and larger high-frequency motions. The above findings of increased high-frequency sway in the elderly are in agreement with other reports (Kristinsdottir et al. 2001; Prieto et al. 1996). Kristinsdottir et al. (2001) found a strong correlation between increased high-frequency sway and diminished vibration perception. Vibration perception (Steinberg and Graber 1963) and proprioception (Skinner et al. 1984) have been reported to decrease with increasing age. Our findings suggest that the adaptive responses were of the same size and time duration within the elderly and middle-aged groups. However, the parameters describing the balance control imply that the body sway characteristics were significantly different between the age groups. Thus, the elderly were able to identify the characteristics of the balance perturbations and act in accordance with the conditions to increase their stability and suppress the perturbation effects. Nonetheless, the elderly seemed to have more difficulty in withstanding balance perturbations compared to the middle-aged and were unable to improve their balance control beyond a certain limit. Our results suggest that this deficit was not associated with inactive or inappropriate adaptation ability, but was associated with the status and accuracy of the elderly subjects' basic balance control. This finding is in line with other reports showing that the sensory and motor systems decline with age (Baloh et al. 1993; Enrietto et al. 1999; Lord and Ward 1994; Skinner et al. 1984; Thelen et al. 1996; Woodhull-McNeal 1992).

Visual information had a prominent effect on the body sway during the tests, both for the elderly and middle-aged subjects. The high-frequency motions and the body sway induced by the perturbations (see Fig. 5) were significantly reduced and for the elderly the stability, as expressed by the dynamical parameters, was significantly increased. Thus, the elderly may have comparatively more balance problems when less visual information is available, for instance when moving in poorly lit surroundings. However, although visual input improved the balance control in elderly subjects, this was not achieved to the same extent as for the middle-aged subjects. The elderly were not able to use visual information to reduce the complexity of the motion responses and change their body sway composition to the same extent as the middle-aged. These findings of decreased ability of the elderly to use vision to reduce body sway are in accordance with other reports (Kristinsdottir et al. 2001; Lord and Ward 1994; Prieto et al. 1996; Teasdale et al. 1991).

Adaptation and habituation are common in many biological systems and effects of adaptation in the human biological system can for example be observed in motor control (Eccles 1986; Ferrel et al. 2000) and in the central nervous system (Robinson 1995). This study is focused on short-term adaptation in the biological and biomechanical systems used for balance control. Figure 2 illustrates the nature of the problem of analyzing adaptation in postural control. The figure shows that the capacity to correctly react to a perturbation, i.e., to adapt, is a gradual process, which develops over time and by gained experience from repeated exposures to the balance disturbances. We have in a series of studies identified two methods we humans seem to use to increase the stability and suppress the effects of perturbations (Fransson et al. 2000; Johansson et al. 1995). One method is to adjust body leaning ("Posture adaptation"). The non-linear characteristics of the biomechanical construction of the human body usually mean that it is favorable to lean slightly forward when submitted to balance perturbations, where the body motions could be controlled with more flexibility and where the stability margins for the motions are larger (Maki et al. 1994; Sinha and Maki 1996). Another method is to learn the effects of perturbation and in a feedforward manner make preemptive adjustments to suppress the balance disturbances, for example by increasing the co-contraction of the muscles in the lower legs ("Stimulus adaptation") (Gatev et al. 1999; Milner 2002). Hence, the adaptive adjustments of postural control have properties that suggest contributions from multiple, partly independent, adaptive processes. These findings are in line with the proposal by Lestienne and Gurfinkel (1988), namely that different systems of postural control are used to regulate balance. First, a reference position for equilibrium is specified by a conservative system. Second, the equilibrium about the pre-selected reference position is maintained by an operative system. These two systems can be manipulated separately and act at different time scales (Fransson et al. 2002; Gurfinkel et al. 1995; Lestienne and Gurfinkel 1988).

Moreover, the effects of adaptation on the body sway seem to be temporary and mostly prominent during the initial phase when submitted to a new kind of perturbation. The last 40 s of the recordings shown in Fig. 2 illustrates that the responses induced by the perturbations after a while form a steady-state behavior pattern, with controlled responses of the same size. Our analysis method is based upon the hypothesis that the described adaptive changes are superimposed upon a basic control mechanism, whose properties and characteristics could be observed and analyzed if the adaptive changes are considered in the analysis. The feedback model in the adaptation analysis method describes this steady-state behavior whereas exponential functions describe the superimposed adaptive responses. The data validation, see Fig. 2 red curve, suggest that it is possible to describe the body motions with high accuracy when a mathematical model based upon the above assumptions is used to analyze the data.

The present study implies that the elderly use the same kind of regulation pattern as the middle-aged, with adaptive response adjustments and "strategic" changes of posture or body leaning, when exposed to perturbations of stance (see Fig. 2, Table 1). The elderly in this study adjusted their responses to stimulation (i.e., the adjustment pattern was above zero-order) in 96% of the posturographic tests and adjusted their posture or body leaning in 92.5% of the tests. These findings are in line with the observations of Maki et al. (Maki et al. 1994; Sinha and Maki 1996), who found a tendency among elderly to lean more forward during continuous perturbation than during quiet stance. Moreover, there was substantial inter-subject variability of the movement patterns used to adapt the posture and to change the stimulation response. We expect that this variability is even larger among patient populations. This variability will affect the results obtained from many of the analysis methods commonly used to evaluate posturography measurements, such as analysis of motion variance, sway velocity, sway path or sway area. For example, a large long-term change of posture will increase all these values, but may not be a true measure of poor balance control. Our findings and other reports (Keshner et al. 1987; Maki et al. 1994) rather imply the opposite: the ability to change posture and body leaning to a more favorable position, thereby reducing the responses to stimulation, might be a sign of good balance control.

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