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Stiffness and Damping in Postural Control Increase with Age

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Abstract

Upright balance is believed to be maintained through active and passive mechanisms, both of which have been shown to be impacted by aging. A compensatory balance response often observed in older adults is increased co-contraction, which is generally assumed to enhance stability by increasing joint stiffness. We investigated the effect of aging on standing balance by fitting body sway data to a previously-developed postural control model that includes active and passive stiffness and damping parameters. Ten young (24 ± 3 y) and seven older (75 ± 5 y) adults were exposed during eyes-closed stance to perturbations consisting of lateral pseudorandom floor tilts. A least-squares fit of the measured body sway data to the postural control model found significantly larger active stiffness and damping model parameters in the older adults. These differences remained significant even after normalizing to account for different body sizes between the young and older adult groups. An age effect was also found for the normalized passive stiffness, but not for the normalized passive damping parameter. This concurrent increase in active stiffness and damping was shown to be more stabilizing than an increase in stiffness alone, as assessed by oscillations in the postural control model impulse response.

INTRODUCTION AND BACKGROUND

From a biomechanical perspective, human stance represents an unstable system, in that gravity acting on the body generates a torque that drives the body away from vertical upright. A fall would occur in the absence of stabilizing torques generated by the postural control system to counter the effects of gravity. It is generally held that postural control for upright stance involves active and passive mechanisms [1], [2], [3]. The active mechanisms consist of neurally-mediated sensory-based feedback control that utilizes perceived body position and movement in space to generate corrective torques [1], [2]. There are three main sensory systems used to provide feedback for control of upright balance: the vestibular, visual, and proprioceptive systems [4], [5]. The vestibular system detects head orientation and motion in space; proprioception detects orientation and motion of each body segment with respect to each other; and vision detects head orientation and motion in space.

Information provided by each sensory system is neurally processed and combined to extract

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overall information of body orientation and movement in space. An important aspect of this sensory integration is the ability of the postural control system to adapt to external perturbations by adjusting the relative importance (weighting) placed on the information from the various sensory systems [5], [6], [7], [8].

The passive mechanisms involved in postural control arise from mechanical properties of muscles and tendons around the joints, such as the intrinsic visco-elasticity of muscles and stiffness of tendons, which provides some gravity-counteracting forces in much the same way a spring generates a counter-force when it is displaced from its resting equilibrium [9], [10], [11]. Previous studies have shown that the contributions of passive mechanisms to maintenance of upright stance while being subject to postural perturbations such as floor tilts, moving scenes or galvanic stimulation, are relatively minor in comparison to active control mechanisms [5], [7].

Effects of Aging on Postural Control

The ability to maintain a stable upright posture is known to decrease with age. One potential mechanism of aging on postural control is reduced sensory function of one or more sensory systems that affect stance stability through ankle impedance characteristics [12]. Ho and Bendrups have shown that older adults have generally different stiffness than young adults, and moreover, when the older adults are separated into fallers vs. non-fallers, the fallers show increased ankle stiffness compared with the non-fallers [12]. Co-contraction has been shown to be one mechanism for older adults to increase stiffness when exposed to changes in the sensory information [13]. Others have suggested that passive properties of muscle fibers and tendons also impact stiffness and damping properties [14], [15]. Ishida et al. [15] showed that the spectral characteristics of ankle impedance (ratio of external torque to ankle angle) changed as the amplitude of the perturbation was varied (small vs large) and as the available sensory information (eyes open vs closed) was altered. The frequency-dependent changes observed were interpreted by the authors as evidence that increased low frequency impedance (mainly stiffness) is effective for enhancing stability in response to slowly changing perturbations, but as the changes become more rapid, it is important to study ankle impedance over a broad frequency range and not only talk about stiffness alone when studying the overall stability of the posture system [15].

In this paper, we provide new experimental results and model-based interpretations that complement and add to the findings of Ishida and others, regarding increased stiffness and damping. Specifically, we present new results showing that older adults employ increased active stiffness and damping as a means of improving their stability in response to platform perturbations, and that this is more effective than increasing active stiffness alone. The data presented are part of a larger experiment investigating the impact of aging on postural control. In this paper, we only report the findings from the postural control stiffness and damping modeling efforts of body sway response to platform tilts alone. These results were presented in part at the IEEE Engineering in Medicine and Biology Conference, 2009.

METHODS

Subjects

The experimental protocol was approved by the Institutional Review Board at the University of Pittsburgh, and all subjects gave their informed consent to participate in this study. Data were obtained from seven older adults (three males and four females, ages 68 to 81 years (mean 75 ± 5 years SD)) and ten young adults (four male and six females, ages 21 to 30 years (mean 24 ± 3 years SD)). Prior to experimental tests, all subjects completed a set of screening examinations to ensure absence of any balance abnormality. The screening

procedures consisted of standard vestibular function, oculomotor, and balance testing, including caloric and rotational tests, vibration and cutaneous pressure sensation, and computerized dynamic posturography.

Experimental Apparatus and Design

A dynamic posturography platform (NeuroTest, Neurocom International, Inc., Clackamas, OR) was used to induce rotational platform perturbations and acquire center of pressure data in response to those perturbations (Fig. 1). A harness system was used to prevent injury from falling during testing. The harness did not impede sway, or give any positional feedback to the subject. Measures of body sway were obtained using a magnetic tracking device¹ (Fastrak, Polhemus, Colchester, VT) with two sensors; one positioned on the lower back, at the height of the iliac crest, and one positioned at the top of the head.

For the data used in modeling stiffness and damping that is reported here, subjects were only exposed to movements of the underfoot platform. The platform (or support surface (SS)) condition consisted of a randomly moving platform whose velocity followed a pseudo-random ternary sequence (PRTS-SS). Subjects stood on the platform such that the platform rotated about an anterior-posterior axis in between the medial malleoli, thus inducing ML sway. Subjects were asked to stand comfortably with their feet close together. Foot placement was marked at the start of the first trial so that the same position was maintained from trial to trial and across visits for each subject. The average distance (± 1 SD) between the middle of the heels was 15.0 ± 1.6 cm and 19.9 ± 3.9 cm for the young and older adults, respectively.

During the PRTS-SS condition, the platform rotated pseudo-randomly according to the integral of a PRTS, with peak-to-peak amplitude of 4° and a cycle duration of 48.4 s (see [5], [7] for details). Recorded postural sway in response to the perturbations were used to estimate dynamic properties of the postural control system over a range of frequencies (0.02 – 2.80 Hz) in response to surface tilts. The PRTS-SS condition allows for an input-output analysis and estimation of model parameters. To ensure adequate steady-state conditions for this analysis, three consecutive cycles of the PRTS-SS were presented during a trial, for a total perturbation interval of 145.2 s.

Experimental Protocol

Subjects were tested over three sessions. On the first testing visit, prior to the start of trials, anthropometric measurements were taken so that the body inertia (J), mass (m), and center-of-mass height (h_{COM}) used in the model could be obtained [16]. During each visit, subjects were tested under a variety of conditions, with the PRTS-SS condition alone being one of the conditions. This resulted in three PRTS-SS trials for each subject. During experimental testing, subjects stood upright, with eyes closed, on the posturography platform while performing the trials for each visit, with a three minute seated rest in between each trial, and at least two days between visits. All subjects were given the following instructions: “Maintain a relaxed upright stance position with your eyes closed and arms folded across your chest.” The duration of each trial was 205.2 s, consisting of a 145.2 s perturbation interval (PRTS-SS motion) preceded and followed by a 30 s quiet stance period on a fixed SS.

¹Because the Fastrak system is sensitive to surrounding metal and the NeuroTest platform is made of steel and aluminum, we calibrated the accuracy of the Fastrak measurements. We used a laser pointer to accurately position the sensor at various locations, at approximately the mean lower back height it would be above the platform during experiments. Our calibration measurements showed inaccuracies of less than 2.5% over a range of 10 cm.

Data Measurement and Analysis

Medial-lateral (ML) displacement of the lower back measured with the electromagnetic sensor was used to estimate body sway (BS) angle in the frontal plane by using the small angle approximation,

$$Hip_{ML}(t) = h_{COM} \sin BS(t) \approx h_{COM} BS(t) \quad (1)$$

The error made by implementing this approximation is about 1% for angles within $\pm 15^\circ$. This angular range corresponds to a lower back displacement of ± 25 cm, assuming $h_{COM} = 100$ cm, which is about 5 times larger than the excursions we observed.

Body sway and platform rotation measurements during the 145.2 s of pseudo-random platform motion were divided into the 3 PRTS-SS cycles of 48.4 s duration. For each cycle, the power and cross power spectrum were estimated via the discrete Fourier transform (DFT) of each time series; in general, for discrete-time series $x(t)$ and $y(t)$, $t=0,1,\dots,N-1$, the DFT $[X(\omega_k), Y(\omega_k)]$, power spectra $[S_{XX}(\omega_k), S_{YY}(\omega_k)]$ and cross-spectrum $[S_{XY}(\omega_k)]$ were computed as:

$$X(\omega_k) = \frac{1}{N} \sum_{t=0}^{N-1} x(t) e^{j(2\pi/N)t\omega_k} \quad (2a)$$

$$Y(\omega_k) = \frac{1}{N} \sum_{t=0}^{N-1} y(t) e^{j(2\pi/N)t\omega_k} \quad (2b)$$

$$S_{XX}(\omega_k) = |X(\omega_k)|^2 \quad (3a)$$

$$S_{YY}(\omega_k) = |Y(\omega_k)|^2 \quad (3b)$$

$$S_{XY}(\omega_k) = X^*(\omega_k) Y(\omega_k) \quad (4)$$

These functions were computed for each cycle of the respective time series measurements, and then averaged across the three cycles. Further spectral smoothing was also applied to reduce variability in the spectral estimates, especially at higher frequencies, as described in [5], [7]. The resulting smoothed power and cross-power estimates had 17 data points ranging from 0.021 to 2.79 Hz, evenly spaced in a log scaled frequency axis. From these smoothed ensemble averages, the experimental frequency response (transfer function) to PRTS-SS was estimated by computing:

$$H_{XY}(\omega_k) = \frac{\bar{S}_{XY}(\omega_k)}{\bar{S}_{XX}(\omega_k)}, \quad (5)$$

where the overbar denotes the smoothed ensemble power and cross-power spectra, and subscripts X and Y denote the SS and BS time series, respectively. (See Fig. 2 for a representative example of the effect of smoothing on the spectral estimates.)

Postural Control Model Fits and Statistical Analyses

Modeling of postural control—A least-squares fit of a previously developed and validated postural control model was made to the smoothed experimental frequency response functions. The model has been shown to produce simulations in good agreement with postural data under a variety of experimental paradigms, including ones similar to our experiments [5], [6], [7], [17],[18], [19].

The postural control system is modeled by a linear feedback controller (Fig. 3). Body dynamics are represented by a single-link inverted pendulum, as in other studies [5], [6], [20], [21]. For the experimental conditions as used here, the single-link dynamics model has been shown to be accurate; in particular, Cenciarini and Peterka showed that the experimental transfer function during free-standing trials was indistinguishable from that of trials in which subjects wore a back-board to insure single-link mechanics [7]. Similar results have also been reported for AP conditions [5].

For eyes-closed stance, body motion is sensed by the vestibular and proprioceptive systems, which are modeled by scalar constants to represent the relative contribution of each sensory system to balance control. This sensory information is combined and used to generate a controlling torque about the ankles, via a “neural controller” modeled by a proportional, integral, and derivative (PID) controller [5], [6]. The sensory feedback loop includes a lumped time-delay that accounts for neural transmission, sensory processing, and muscle activation delays in the postural control system. The neural controller, the time delay, and the sensory weights represent the active mechanisms used by the CNS to control posture.

The model also includes a passive pathway that contributes to torque generation, based on the position and velocity of the ankle joint angle, as obtained via the passive stiffness (K) and damping (B) parameters in the model (Fig. 3). By “passive” we mean as used in previous studies [2], [5], [7], namely the generation of a stabilizing torque without any processing delay, in contrast to the “active” pathways of the model, which are neurally-mediated and include time delay.

The mathematical expression for the frequency characteristics of the model is given in Eq. (6). This equation expresses the model transfer function for body sway in response to a PRTS-SS perturbation, with eyes-closed and under steady-state conditions, for which the sum of the proprioceptive (w_p) and graviceptive (w_g) sensory channel weights is unity, $W = w_p + w_g = 1$ [5], [7], such that the sensory weights represent the relative contribution of the respective sensory channel to the total sensed body sway. Least-squares curve fits of the model transfer function to the experimental transfer functions were made as in [17], [19] to obtain the parameters of the model, namely proprioceptive sensory channel weight (w_p), active stiffness (K_p), active damping (K_D), integral factor (K_I), time delay (T_D), passive stiffness (K) and passive damping (B). (Note that because $w_p + w_g = 1$, w_g is given by $1 - w_p$).

$$M_{ss}(s) = \frac{BS(s)}{SS(s)} = \frac{w_p \cdot (K_D s^2 + K_p s + K_I) \cdot e^{-T_D s} + B s^2 + K s}{J s^3 + W \cdot (K_D s^2 + K_p s + K_I) \cdot e^{-T_D s} + B s^2 + (K - mgh_{COM}) s} \quad (6)$$

Statistical Analyses—A mixed-factor repeated-measures ANCOVA was performed on all model parameters, using a significance level of $\alpha=0.05$. The independent factors were Age (Young and Older Adults) and Visit (1, 2, and 3). In addition, we included the moment of inertia, J , as a covariate factor in our statistical analysis, because the estimated moment of inertia was larger in the older adults than in the young adults (Table I). Furthermore, body size differences (mass, height) are known to be correlated with stiffness and damping [5], and thus any observed differences in these parameters may be explained by differences in body size. This statistical approach allows us to determine if the postural control model parameters change significantly with age, while accounting for differences in body size (i.e., moment of inertia, J , which depends on mass and height).

RESULTS

Experimental transfer functions and modeling results

There was no effect of visit nor interaction between visit and age on the model parameters; therefore we focus on the effect of Age. In Fig. 4, the mean group (young vs old) experimental transfer function gain and phase curves of BS to PRTS-SS are plotted. Below about 0.3 Hz, there is little difference between the two groups. Above 0.3 Hz, the gain was higher for the older adults compared to the young and exhibited a slight peak around 0.7 Hz. No apparent systematic differences between young and old were observed for the phase.

Model fits to the subjects' experimental transfer functions yielded parameters that quantify differences between the young and older adult groups (Fig. 5 and Table II). In particular, the active stiffness K_P was significantly larger in the older adult group compared to the young adults. Similarly, the active damping K_D was significantly larger in the older adults than in the young adults. The integral gain parameter K_I also increased significantly with age.

Passive control parameters (K and B) tended to be much smaller in value than the corresponding active parameters (Table II). Moreover, no significant age effect was found in the passive parameters.

No significant age effect on the time delay (T_D) parameter was found, which was on average the same for both age groups. The proprioceptive sensory weight parameter (w_p) was significantly larger in the older adults than in the young adults; consequently, since $w_p + w_g = 1$, the graviceptive (vestibular) sensory weight was smaller in the older adults than in the young adults.

Accounting for body size differences—Another way to account for the influence of body size (i.e. body mass and height) on the parameter values is to normalize the parameters by the moment of inertia (which is a function of body mass and height; for example, it is proportional to mh^2 for the inverted pendulum). This normalization is suggested by the form of the transfer function in Eq. (6): namely, divide the numerator and denominator by J to obtain the normalized parameters K_P/J , K_I/J , K_D/J , B/J and $(K-mgh_{COM})/J$. In a study of young adults, Peterka showed that stiffness and damping normalized in this way were uncorrelated with and insensitive to differences in body mass and height [5]. Accordingly, we repeated our statistical analyses on K_P/J , K_I/J , K_D/J , B/J and $(K-mgh_{COM})/J$, without the effect of the covariate. This analysis revealed that, with the exception of passive damping, the normalized parameters showed similar significant changes between the young and older adults (Table III and Fig. 5). Thus, the differences observed are not attributable solely to differences in body size between the two populations.

DISCUSSION

Aging is known to affect human balance, with increasing degradation in motor and sensory function [22], [23], [24]. As muscle and sensory properties change, the strategies employed by older adults to maintain balance may change to compensate for degraded function. For example, a common view is that elderly adults are “stiffer,” exhibiting increased co-contraction of muscles during more challenging balance tasks, or in response to balance perturbations [25]. However, from a control systems standpoint, increased stiffness alone is not necessarily a good compensatory response. In particular, it is well known that increasing the stiffness in a mechanical mass-spring-dashpot system can result in resonant (oscillatory) behavior. A similar result occurs in the linearized, second order, stable inverted pendulum model with stiffness K , damping B , mass m , center-of-mass height h_{COM} and moment of inertia J , which has poles given by

$$s_{1,2} = \frac{-B}{2J} \pm \sqrt{\left(\frac{B}{2J}\right)^2 - \frac{K - mgh_{COM}}{J}} \quad (7)$$

Note that, if the damping B is held fixed, increasing K will eventually cause the term under the square root to become negative, resulting in complex poles and hence an oscillatory (i.e., underdamped) response to a perturbation. In the frequency response gain, this effect is manifest as a peak in the neighborhood of the damped natural frequency, indicating that the system is more susceptible (i.e. will respond more) to perturbations near this frequency. From a stability standpoint, this response is not necessarily desirable. A concurrent increase in the damping would reduce the frequency response gain and hence the magnitude of the oscillations, and moreover would result in a faster settling time to steady state after a perturbation.

Our experimental results showed that, indeed, older adults are stiffer than young adults, but that they also have increased damping. This concurrent increase in damping has the beneficial effect of reducing oscillations in response to external perturbations. To see this effect, let us first consider the impulse response from our postural control model, Eq. (6), using the mean parameter values for the young and older adults, respectively. These impulse responses are plotted in Fig. 6, from which it is clear that the older adults exhibit larger, more sustained, and slightly higher frequency body sway in response to a perturbation.

Now consider the effect of increasing the stiffness alone, versus increasing the stiffness and damping, which is shown in Fig. 7. In particular, Fig. 7 shows three plots, corresponding to the “nominal” impulse response of the young adults, using their mean parameter values, repeated from Fig. 6, along with a plot of the response obtained by increasing stiffness (normalized to body size) to that seen in the older adults, and a plot of the impulse response obtained by increasing stiffness and damping (normalized to body size) to that seen in the older adults. These plots show that increased stiffness alone results in larger overshoots and longer settling times in the impulse response than does a concurrent increase in stiffness and damping. We also observed that increasing stiffness alone did not seem to affect the timing of the peaks as compared to a concurrent increase of stiffness and damping. The differences in overshoot and settling time can be quantified by measuring the first three peaks in the impulse response (max, min, max), and the area under the curves, as reported in Table IV. Lower peaks and smaller area are indicators of greater stability, in the sense that the system is better able to resist external perturbations. Note, however, that while the concurrent increase in damping with stiffness does yield a more stable impulse response than that arising from increased stiffness only, it does not yield an impulse response as stable as that of the young adults, who have lower damping and lower stiffness.

Substantial increases in the integral gain, K_I , were also observed in the older adults, and it is therefore worthwhile to investigate this effect as well. From a control systems perspective, the integral term is not needed to stabilize the inverted pendulum; a proportional-derivative (PD) controller (i.e., a controller with gains K_P and K_D) is sufficient to yield a stable response. The effect of introducing an integral correction term in a PD controller has the potential benefit of improving tracking ability and reducing steady-state error. In our simulations, increasing K_I in the simulated response of young adults to the value seen in older adults (dotted curve, Fig. 7, and 4th col., Table IV) had some effect, most notably in a slight reduction in the amplitude of the third peak, consistent with the notion that integral control helps to reduce steady state error. However, from a stability perspective, the increases in K_P and K_D seem to have more of an effect on the impulse response than does the (comparatively larger) increase in K_I .

Our experimental results and model-based analyses are consistent with and complement recent work by others on active stiffness and damping (i.e., active impedance) in balance and motion. While co-contraction, resulting in increased active stiffness, has been cited as a compensatory response to balance perturbations in the elderly [25], it is important to consider active impedance (which is characterized by active stiffness and damping), and not only active stiffness [9], [14], [15]. In a recent study [15], the authors concluded that if elderly subjects use co-contraction to compensate for loss of sensory and motor function due to aging, then this could explain why older adults are more prone to falls than are young adults, especially in response to rapid perturbations. They speculated that older adults are able to respond adequately to very slow perturbations or changes in the environment, but are unable to adequately counteract rapid perturbations. Considering changes in the relative stiffness and damping may explain why in particular situations the postural control system responds adequately but in others may fail and a fall occurs.

Our results provide some corroborating evidence for Ishida's hypothesized explanation for increased risk of falls with aging, especially due to rapid perturbations [15]. In our model based analysis, we were able to estimate stiffness and damping and furthermore to distinguish between active and passive factors. We found that active stiffness and damping dominated the response to external perturbations in both young and older adults, compared to the corresponding passive properties, consistent with previous findings in young adults [5], [7]. Moreover, we found that active stiffness and damping, relative to body size, increased in older adults compared to young adults, and that the normalized passive stiffness of older adults was also more negative (destabilizing, thus requiring more active stiffness in order to achieve enough total stiffness to counter the destabilizing effects of gravity. Perhaps because of this more negative normalized passive stiffness, the increase in active stiffness was larger than the increase in active damping (21% increase in active stiffness compared to young, vs. 14% increase in active damping compared to young -- Table III). As shown in our impulse response plots, while the increase in stiffness and damping was more stabilizing than an increase in stiffness alone, the impulse response of the older adults was still not as stable as that of the young adults. These experimentally derived model-based results provide further insight about the characteristics of the postural control system in older adults that makes them more sensitive to external balance perturbations, compared to young adults.

Of course, extrapolating controlled laboratory findings to real-life situations should be done judiciously. It would be of interest to perform similar analysis under additional conditions, such as eyes-open sway, which is a more real-life scenario than the eyes-closed laboratory condition explored here. Also, it would be of interest to examine multi-link dynamics and model fits particularly if larger perturbations are used. Stiffness and damping may be different at the hips and the ankles across subject groups, which may be uncovered by a multi-link approach.

CONCLUSION

In this paper, we addressed the question of what postural control changes occur with age by examining the frequency response of body sway in response to platform perturbations, and also fitting a control model to the data to obtain stiffness and damping parameters. We observed larger frequency response gain characteristics above 0.3 Hz and a more pronounced peak around 0.7 Hz in older adults compared to young adults. These findings suggest that older adults would experience a more oscillatory response to fast occurring perturbations than they would for slow ones, compared to young adults. Accordingly, older adults may have to be more cautious in situations where they could experience rapid floor movements such as on a bus, train, or escalator.

In addition, the model-fit results showed that older adults had significantly higher active stiffness and damping parameter values as compared to the young adults. Thus, older adults do not just increase stiffness in response to external balance perturbations, but damping as well. This concurrent increase in active stiffness and damping was shown to be more stabilizing than an increase in stiffness alone, as assessed by oscillations in the postural control model impulse response, and the area under the curve. However, the older adults were still less stable than the younger results, as quantified by larger oscillations in the model-fit impulse response.

Further research is needed to understand the possible sources of these changes in the control system and to determine if improved compensation can be achieved with appropriate physical therapy or exercise. Investigations into multi-link behavior, as well as in response to other perturbations and sensory conditions, is also warranted.

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Fig. 1.

Experimental Setup. A NeuroTest posture platform was used to provide the support surface perturbation (Neurocom, Inc.). A Polhemus Fastrak magnetic-based motion tracking system was used to measure body sway. Subjects stood with eyes closed during experimental trials.

Fig. 2.

An example of the experimental Transfer Function (TF) gain and phase curves computed from smoothed (orange) and unsmoothed (blue) spectral estimates from a representative young subject. Smoothing was applied as described in Methods to reduce variability in the transfer function estimates.

Fig. 3.

Feedback Model of Postural Control. Sensory channels are limited to vestibular (graviceptive) and proprioceptive sensory channels. The mechanical perturbation provided by the support surface (SS) is indicated in the schematic as a pseudorandom stimulus. Body sway with respect to the feet (i.e. with respect to a sagittal plane perpendicular to the SS) is indicated by BF. Eyes are closed, hence visual sensory feedback is not included in the model. The “Aging” box is hypothesized to have an effect on the “Sensory and neuromuscular system.” (Model schematic adapted from [7], [19].)

Fig. 4.

Experimental Transfer Function (TF) estimated from body sway response to SS rotations for young adult (YA) and older adult (OA) age groups. Averages were taken across the three sessions for PRTS-SS trials alone. YA and OA gain curves were similar in the low-frequency range (below 0.3 Hz), while the gain of the older adults was larger in the mid- and high-frequency ranges (above 0.3 Hz) and exhibited a slight peak around 0.7 Hz.

Fig. 5.

A. Model parameter values estimated by fitting the postural control model to the experimental transfer functions (also see Table II). Bar plots show average results (mean $\pm$ SD) for young adults (YA) and old adults (OA). Significant age differences ($p < 0.05$) are indicated by *. B. Normalized stiffness and damping (also see Table III).

Fig. 6.

Plots of the impulse response of the postural control model for nominal values of active and passive stiffness and damping in the young adult (YA) group (solid curve) versus the older adult (OA) group (dash-dot curve). The first peak is similar in terms of timing and amplitude, but the response of the OA group is characterized by larger subsequent peaks, and slightly higher frequency of oscillation, as compared to the YA. Older adults are mainly characterized by larger active stiffness and damping as compared to young adults.

Fig. 7.

Plots of the impulse response of the postural control model for nominal values of stiffness and damping in the young adult population (solid curve), versus increased stiffness only (dashed curve), versus increased stiffness and damping (dash-dot curve), and versus increased stiffness, damping, and integral gain (dotted curve). Note that peak-to-peak oscillations are greatest for increased stiffness alone, and that increasing damping concurrent with an increase in stiffness diminishes these oscillations. Increasing the integral gain has little effect compared to increased stiffness and damping.

Table I

Summary of anthropometric data from the subjects included in the study.

Age Group	Height [m]	Mass [kg]	Moment of Inertia (J) [kg-m ²]
Young Adult	1.72 ± 0.07	67.0 ± 7.4	66.7 ± 12.2
Older Adult	1.72 ± 0.10	75.5 ± 11.4	75.5 ± 19.8

Table II

Estimated model parameters (mean $\pm$ SD) for young and older adult groups.

	K_P [N·m·rad ⁻¹]	K_d [N·m·s·rad ⁻¹]	K_I [N·m·rad ⁻¹ ·s ⁻¹]	K [N·m·rad ⁻¹]	B [N·m·s·rad ⁻¹]	T_D [ms]	w_p
Young	898 $\pm$ 215	288 $\pm$ 71	182 $\pm$ 110	157 $\pm$ 165	34 $\pm$ 40	172 $\pm$ 27	0.44 $\pm$ 0.13
Older	1225 $\pm$ 299	370 $\pm$ 86	359 $\pm$ 163	99 $\pm$ 166	44 $\pm$ 49	173 $\pm$ 24	0.64 $\pm$ 0.17
p-value	0.005	0.042	0.032	0.195	0.943	0.818	0.011

Table III

Normalized model parameters (mean $\pm$ SD) for young and older adult groups.

	K_P/J [s^{-2}]	K_D/J [s^{-1}]	K_I/J [s^{-3}]	$(K - m \cdot g \cdot h_{COM})/J$ [s^{-2}]	B/J [s^{-1}]
Young	13.5 ± 2.8	4.3 ± 0.7	2.6 ± 1.3	-7.0 ± 2.9	0.5 ± 0.6
Old	16.4 ± 2.6	4.9 ± 0.6	4.8 ± 2.1	-8.2 ± 2.4	0.6 ± 0.5
p-value	0.002	0.013	0.019	0.010	0.115

Table IV

Peak amplitudes and area under the curve of the simulated impulse responses.

	Young adults	Increased stiffness	Increased stiffness and damping	Increased stiffness, damping, and integral factor	Older adults
1 st peak [deg]	3.70	3.68	3.85	3.86	3.86
2 nd peak [deg]	-1.30	-2.06	-1.81	-1.94	-1.93
3 rd peak [deg]	0.00	0.73	0.51	0.39	0.39
Area [deg*s]	2.76	3.41	2.83	2.77	2.76