EFFECT OF AGING ON HUMAN POSTURAL CONTROL AND THE INTERACTION WITH ATTENTION

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The ability to stand upright and walk is generally taken for granted, yet control of balance utilizes many processes involving the neuromuscular and sensory systems. As we age, balance function begins to decline and can become problematic for many older adults. In particular, adults 65 years of age and older exhibit a higher incidence of falls than younger adults, and falls are a leading cause of injury in older adults, contributing to significant medical costs. Without better understanding of the impact of aging on balance and means to ameliorate those effects, this problem is expected to grow as life expectancy continues to increase.

In addition to sensori-motor declines with age that impact balance, another factor known to affect balance, particularly in older adults, is attention, meaning the amount of cognitive resources utilized for a particular task. When two or more tasks vie for cognitive resources, performance in one or more tasks can be compromised (a common example today is driving while talking on a cell phone). Attention has been observed to be a critical factor in many falls reported by older adults. However, it is still not fully understood how aging and attentional demand affect balance and how they interact with each other.

In this dissertation, we conducted dual-task experiments and model-based analyses to study upright standing and the interaction of the effects of age and attention on postural control. The effect of age was investigated by testing two age groups (young and older adults) with no

evident balance and cognitive impairment and by comparing results of the two groups. The effect of attention and its interaction with age was studied by comparing body sway in the two age groups in response to a moving platform, while either concurrently performing a cognitive task (dual-task) or not (single-task). Our findings highlight postural control differences between young and older adults, as quantified by experimental measures of body motion as well as by model parameter values, such as stiffness, damping and processing delay.

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PREFACE

I am very grateful to all the members of my dissertation committee – Profs. Cham, Loughlin, Mao, Redfern, and Sparto – for their support, advice, and continuing trust in my abilities over my years at the University of Pittsburgh. I would like to express particular thanks to Dr. Patrick Loughlin, Mark Redfern, and Patrick Sparto who have always been there when I needed advice both in my academic and personal life; their encouragement and constructive criticism was invaluable. I am particularly thankful to Dr. Patrick Loughlin, my advisor, for his feedback, words of advice, and the general encouragement he continually provided in the past seven years.

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1.0 INTRODUCTION

The present work represents the collection of various experiments that were performed to investigate the interaction between the role of attention and aging in control of human balance. Control of balance is known to be influenced by attention, or the amount of brain resources allocated for a particular task such as counting, listening to sounds, remembering and recalling word and number groups, etc. [1-5]. Aging is another factor that is known to influence control of balance. In this work, balance-healthy subjects were studied, which included people without significant mobility, standing, and cognitive impairments; this is what is referred to as "healthy aging," in this dissertation. The interaction of the effects of aging and attention on balance is of particular interest because the occurrence of balance related accidents increase as we age, and with an increasing elderly population, we can expect overall balance related accidents to also increase.

Our goal is to study the interaction between aging and attention and their effect on balance control, through experiments and model-based analyses. The model used in this dissertation has been previously developed and shown to be able to characterize human postural control in a variety of different conditions [1, 6-10].

The model includes parameters that quantify sensory, motor, and biomechanical aspects of postural control, including stiffness, damping, sensory weights, and a lumped time delay that models delays due to neural transmission, muscle activation, and sensory integration, and cognitive processing. By conducting dual-task balance experiments and then fitting the model to the data, we hypothesized that changes in the model parameters between the two age groups (young and older adults) would allow us to detect and interpret changes in the control system as a result of the aging process, as well as changes due to attentional demands. We investigated both anterior-posterior and medial-lateral sway during perturbed standing (in separate experiments). For simplicity, and to reduce the complexity of the model, all experiments were performed with the subject's eyes closed.

We present results from two main projects. The first project included experiments using support surface tilts that induced body sway in the frontal plane (medial-lateral direction). This sway orientation was chosen because a component of this project involved stimulation of the vestibular system via galvanic vestibular stimulation (GVS). However, the results presented here are for support surface perturbations only (no GVS trials). Changes in the body sway response to the support surface perturbation were found with age. Specifically, the group-mean experimental transfer function gain of the older adults was larger than that of the young adults in the mid-high frequency range, and the older adult group gain curve exhibited a slight peak around 0.7 Hz. We were able to quantify these changes and offer a model-based explanation for the behavior observed. In particular, normalized active stiffness and damping and passive stiffness parameters in the model were found to be significantly higher in the older adult group than in the young adults. Simulations were carried out to further investigate our findings and give a more control-system based reason as to why we may be more prone to falling as we age.

The second project addressed the issue of the influence of aging and attention on control of balance using support surface tilts that induced body sway in the sagittal plane (anterior-posterior direction). The project included a large sample size for both younger and older adults based on a pilot study that was performed prior to the start of this project [1]. Two age groups (young and older adults) of healthy subjects were included in this project to study how age affects postural control and the interaction with attention. Dual task experiments involving support surface perturbations and attentional tasks were conducted. The older adults were characterized on average by a higher transfer function gain than that of the young adults in the mid-high frequency range (>0.3 Hz) and presented a slight peak just below 0.6 Hz. The group-average phase curve of the older adults indicated that this group had phase lag as compared to the young in the mid-frequency range (0.2-0.8 Hz). These changes in the experimental frequency response characteristics of the postural control system were reflected by an increase in the mean value of controller parameters, the system time delay, and the passive damping parameter in the older adult group as compared to the young adults.

The findings presented here have the potential to suggest new directions in the assessment of balance performance done in clinical settings as discussed in Chapter 4.5. However, further investigation is required to design and test possible physical exercises or activities that older adults could practice to improve their overall postural and balance control.

2.0 BACKGROUND

2.1 MOTIVATION AND SIGNIFICANCE

A large percentage (at least 30%) of adults age 65 and older is recorded to fall every year [11]. Falls are a leading cause of injury in older adults and it is estimated that 10% to 20% of falls are responsible for sever injuries and this contribute to significant medical costs [12]. Falls also constitute one of the major contributors to the causes of death in older adults [13]. Research studying the epidemiology of aging has pointed out that the population of older adults (age of 65 and older) is increasing and it is projected to be over 20% of the US population by the year 2030 [14]. Accordingly, without a better understanding of balance function and interventions to reduce the incidence of falls in older adults, injuries from falls, and the associated medical costs as well as lost mobility and independence, will increase.

In addition to sensori-motor declines with aging that impact balance, attention has been shown to have a complex effect on balance [2, 15-18]. In a review on this topic, Wollacott and Shumway-Cook (2002) [18] reported that older adults exhibit an increased need of attention for control of posture, and that a mental task performed concurrently with a postural task has a more negative effect on balance control in older adults than in young adults. However, the effect varies with the postural task and with the cognitive demand of the attentional task performed, and the interaction of attention and postural control is not well understood [18].

These findings demonstrate a complex interaction between aging, attentional demand, and control of balance, and point to the need for further study. Improving our understanding of this problem has the potential to allow us to determine possible procedures to improve balance stability and reduce the likelihood of falls in older adults. These are the motivations for the research undertaken in this dissertation.

2.2 HUMAN STANCE, SENSORIMOTOR INTEGRATION, AND MODELING

Human upright stance, the focus of this work, is characterized by a bipedal configuration that is unstable from a biomechanical perspective and requires continuous control by the neuromuscular system to maintain upright balance [10]. In particular, when the center-of-mass (COM) of the body deviates from earth-vertical (such as when we lean slightly forward or backward), gravity acts on the body COM to produce a torque about the ankles that tends to drive the body toward the ground. A fall would occur if counteracting torques were not produced by the muscles about the ankle joint.

While simple in concept, it is not entirely clear how the postural control system operates and accomplishes the task of torque generation to maintain upright balance. Several mechanisms have been proposed over the past three decades. Some advocate open loop mechanisms [19] while others advocate sensory feedback mechanisms [6, 9, 10, 20] for the control of balance. The sensory feedback view of postural control is the prevailing, if not unanimous, point of view, and it has been shown that passive stiffness or open loop stiffness (with no delay) is not enough to control posture [6, 21] as also discussed in [22].

In the last three decades, the availability of new devices to study balance and analytical techniques to design experiments and process the data have led scientists to formulate and test different hypothetical models of the postural control system [6, 23-28]. The common goal is to develop an adequate control model that accounts for the observed human postural behavior(s) and is able to characterize the mechanisms involved in human control of posture.

Generally, a control system includes two main parts: the controlled and the controlling systems. The controlled system can be modeled by equations characterizing its dynamic properties and the level of detail on the characterization is based on the assumptions made. The controlling system, for a feedback design, comprises sensors to detect the state of the system (e.g. body kinematics), a controller (e.g. nervous system) to integrate all the sensory information and determine the required actions to perform, and actuators (e.g. muscles) to perform the action commanded by the controller. In this work, the human body is characterized as a single-link inverted pendulum, which is a widely accepted simplification of the biomechanics yet is adequate to characterize human body motion behavior during quiet stance [19, 23, 24, 27, 29] and also under small perturbations [6, 9, 10].

Two main types of control strategy have been used in postural control modeling: open-loop control, and closed loop (feedback) control. Researchers favoring the open-loop view suggest that upright stance is maintained by passive stiffness control, wherein the neural controller acts mainly to set the ankle joint stiffness without continuously regulating the generation of a control torque in response to sensed body sway [19, 30]. However, a more widely accepted view is that postural control is a reactive feedback system, utilizing sensory information about body position and motion to generate corrective torques [6, 8, 10, 21, 23, 24, 31]. This latter view is adopted in the work presented in this dissertation.

A major difficulty in proposing a postural control model is the characterization of the underlying neuromuscular control processes involved to maintain upright posture. The neuromuscular system is clearly non-linear [6, 27], time varying on a short scale [8] as well as on a long scale like the changes involved with aging [3, 15], and task dependent [3, 15, 26, 32]. Despite this complexity associated with the postural control system, the model used in this project (Figure 1) and originally developed by Peterka has been able to explain experimental observation and findings under a variety of conditions [6, 8, 9, 16, 17].

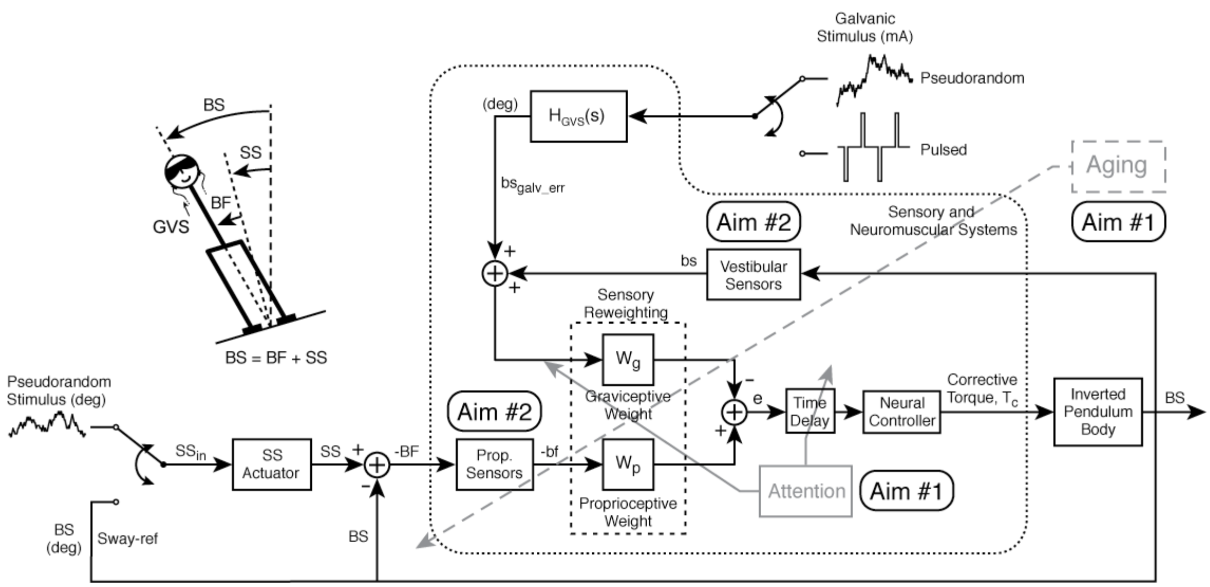

Figure 1. Feedback Model of Postural Control – ML Sway. Independent channel model of sensory integration in postural control for M/L body sway. Sensory channels are limited to vestibular (graviceptive) and proprioceptive sensory channels. The external perturbation provided by the GVS is also included in the schematic. GVS profile set includes: 1) a pseudorandom waveform and 2) an alternating sequence of pulses. The mechanical perturbation provided by the support surface is indicated in the schematic. Two possible perturbations are indicated: 1) a pseudorandom stimulus and 2) sway referenced support surface in which body sway is fed back as input to the SS actuator and ideally canceling out BS input to proprioceptive sensory channel. Eyes are closed, therefore the effect of vision is removed to simplify the overall model. Note the “Aging” box effect that potentially can effect the overall Sensory and neuromuscular system. While attention is thought to only have an effect of the Sensory Reweighting process and the postural control time delay.

This model is linear and the neural controller is assumed to mainly behave as a proportional, derivative, and integrative (PID) controller.

However, more sophisticated control techniques like optimal control and state estimation [25, 26, 33] have shown to be able to characterize experimental data as well. The question about which approach and therefore which model is the most correct one is still open. However, in this proposed project we implement and apply the model proposed in [6] and used in subsequent studies [1, 9, 10], in order to parameterize postural control changes due to aging and due to concurrent execution of attentional tasks.

2.3 INTERACTION BETWEEN AGING, ATTENTION AND POSTURE

We know that postural control is influenced by the surrounding environment and specifically through the interaction of the sensory systems with their individual frame of reference. However, the internal state and the availability of processing resources affect postural performance. This aspect was reported by an early work by [34] in which interaction of posture and execution of cognitive task was studied. Subsequent studies have looked at the interaction between maintenance of posture and execution of different cognitive tasks and a variety of findings have been obtained, depending on the attentional task and postural control challenge performed [15, 34-36]. The emerging evidence from previous work is that postural control in healthy young adults seems to be little affected by performing a cognitive task while standing on a fixed surface. However, as the postural task becomes more challenging, there is a greater effect and interaction between attention and postural control. Additionally, as age increases, the interaction between the execution of the two tasks becomes evident for postural tasks like quiet stance on a fixed surface. Therefore, it can be concluded that maintenance of balance requires more information processing with normal aging [15].

However, when interaction of attention and postural control is studied in older adults, there are other aspects that have to be considered for the changes in the behavioral observations. Specifically, age-related changes in the neuromuscular system can affect postural control performance even in the case of no dual-task conditions. In fact, there is evidence that sensory function degrades over time. For example, cells in the epithelia of vestibular organs decrease with age [37] and the function and sensitivity of the proprioceptive and visual sensory systems degrade with age [38-40]. Furthermore, natural aging has an effect on neural processes involved in integration of sensory information resulting in slower performance of motor activities, reflexive actions, and cognitive processing [41, 42]. Lastly, the musculoskeletal system changes over time by loosing mass, strength, and muscle fiber distribution, which affects dynamic properties of the muscle activation-force generation [43-45]. In summary, changes in any of these components of the postural control system can possibly have an effect on the overall system dynamics as a result of aging.

Therefore, study of the interaction between maintenance of balance and execution of an attentional task considering the effect of age requires two aspects to be considered: first, healthy aging affects postural control due to changes of individual components and second, healthy aging might affect mechanisms that are used for information processing and specifically the ones shared by both postural control and attentional processes. It is not clearly understood how this interaction occurs and what changes over time from young to older healthy adults are responsible for the observed changes.

3.0 STIFFNESS AND DAMPING IN POSTURAL CONTROL INCREASE WITH AGE[1]

3.1 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

[1] The content of this chapter has been published in *IEEE Transactions on Biomedical Engineering*, vol. 57, pp. 267-75, Feb 2010 and it is authored by M. Cenciarini, P. J. Loughlin, P. J. Sparto, and M. S. Redfern.

increase in stiffness alone, as assessed by oscillations in the postural control model impulse response.

3.2 BACKGROUND AND INTRODUCTION

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[7, 46, 47]. The active mechanisms consist of neurally-mediated sensory-based feedback control that utilizes perceived body position and movement in space to generate corrective torques [7, 46]. There are three main sensory systems used to provide feedback for control of upright balance: the vestibular, visual, and proprioceptive systems [6, 48]. 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 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 [6, 8, 9, 49].

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-countering forces in much the same way a spring generates a counter-force when it is displaced from its resting equilibrium [50-52]. 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 [6, 9].

3.2.1 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 [53]. 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 [53]. Co-contraction has been shown to be one mechanism for older adults to increase stiffness when exposed to changes in the sensory information [54]. Others have suggested that passive properties of muscle fibers and tendons also impact stiffness and damping properties [55, 56]. Ishida et al. [56] 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 [56].

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.

3.3 METHODS

3.3.1 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.

3.3.2 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 (Figure 2). 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 device[2] (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 [6, 9] for details).

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[2] Because 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.

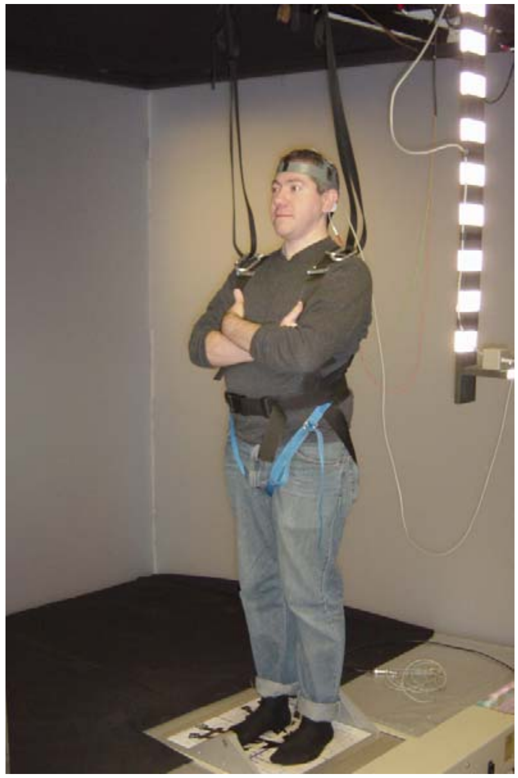

Figure 2. Experimental Setup. A NeuroTest posture platform was used to provide the support surface perturbation (Neurocom, Inc.). A Polhemus Fastrak magnetic-based motion tracking system was used to measure body sway. Subjects stood with eyes closed during experimental trials.

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.

3.3.3 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 [57]. 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.

3.3.4 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,

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

The error made by implementing this approximation is about 1% for angles within ±15°. 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,…,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:

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

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

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

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

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

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 [6, 9]. 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:

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

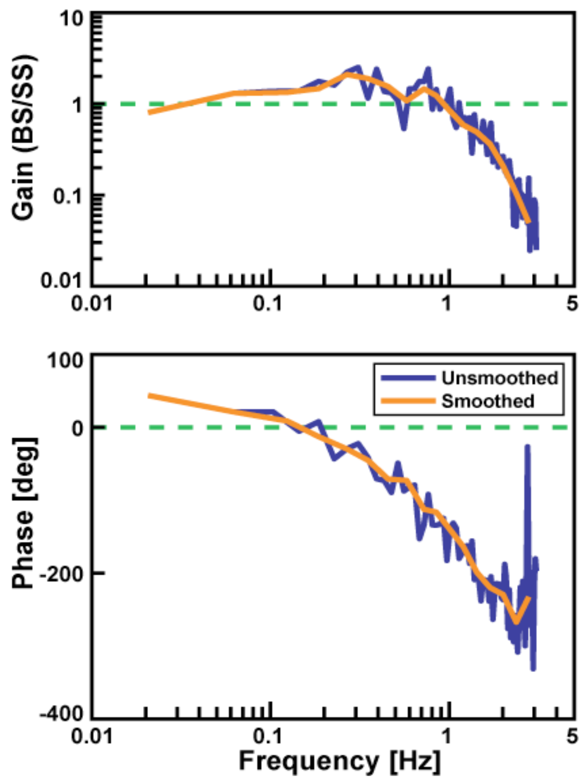

Figure 3. Sample Experimental Transfer Function. An example of the experimental Transfer Function (TF) gain and phase curves computed from smoothed (orange) and unsmoothed (blue) spectral estimates from a representative young subject. Smoothing was applied as described in Methods to reduce variability in the transfer function estimates.

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 Figure 3 for a representative example of the effect of smoothing on the spectral estimates.)

3.3.5 Postural Control Model Fits and Statistical Analyses

3.3.5.1 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

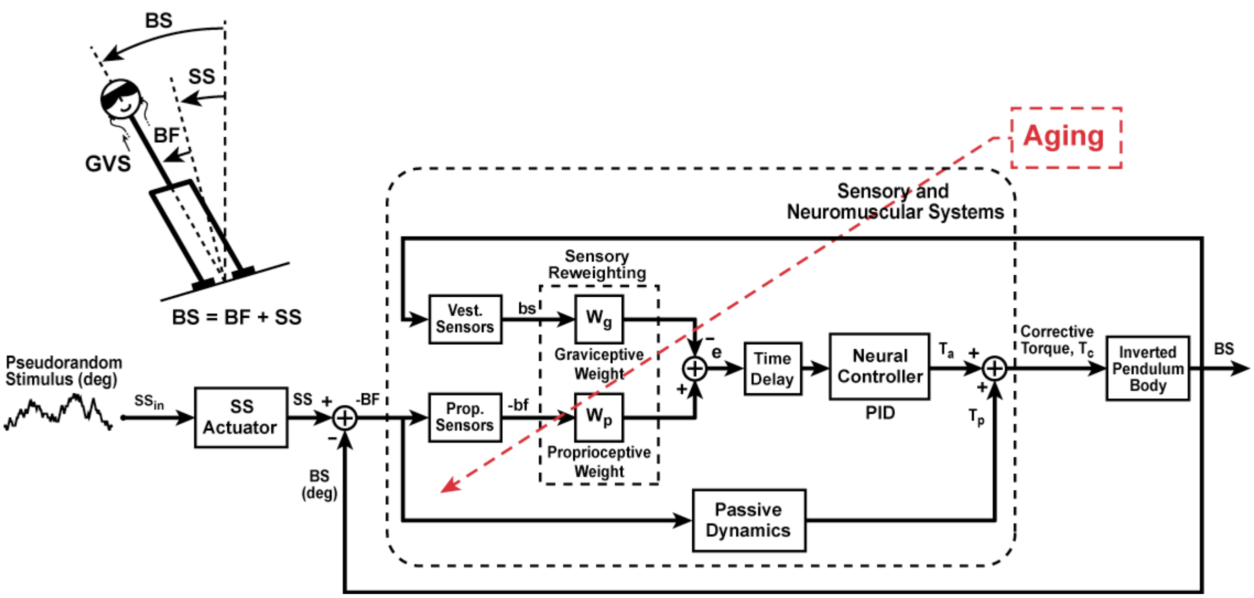

Figure 4. Feedback Model of Postural Control. Sensory channels are limited to vestibular (graviceptive) and proprioceptive sensory channels. The mechanical perturbation provided by the support surface (SS) is indicated in the schematic as a pseudorandom stimulus. Body sway with respect to the feet (i.e. 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 [1, 9].)

simulations in good agreement with postural data under a variety of experimental paradigms, including ones similar to our experiments [1, 6, 8, 9, 16, 17].

The postural control system is modeled by a linear feedback controller (Figure 4). Body dynamics are represented by a single-link inverted pendulum, as in other studies [6, 8, 23, 24].

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 [9]. Similar results have also been reported for AP conditions [6].

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 [6, 8]. 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 (Figure 4). By "passive" we mean as used in previous studies [6, 7, 9], 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, $W = w_p + w_g = 1$

[6, 9], 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 [1, 16] 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.).

$$M_{SS}(s) = \frac{BS(s)}{SS(s)} = \frac{w_P \cdot (K_D s^2 + K_P s + K_I) \cdot e^{-T_D s} + Bs^2 + Ks}{Js^3 + W \cdot (K_D s^2 + K_P s + K_I) \cdot e^{-T_D s} + Bs^2 + (K - mgh_{COM})s} \qquad (6)$$

3.3.5.2 Statistical Analyses

A mixed-factor repeated-measures ANCOVA was performed on all model parameters, using a significance level of α=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 1). Furthermore, body size differences (mass, height) are known to be correlated with stiffness and damping [6], and thus any observed differences in these parameters may be explained by differences in body size.

Table 1. Summary of anthropometric data from the subjects included in the study. – ML sway

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

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).

3.4 RESULTS

3.4.1 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 Figure 5, 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 (Figure 6 and Table 2). In particular, the active stiffness K_P was significantly larger in the older adult group compared to the young adults.

Table 2. Estimated model parameters (mean ±SD) for young and older adult groups - ML

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

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.

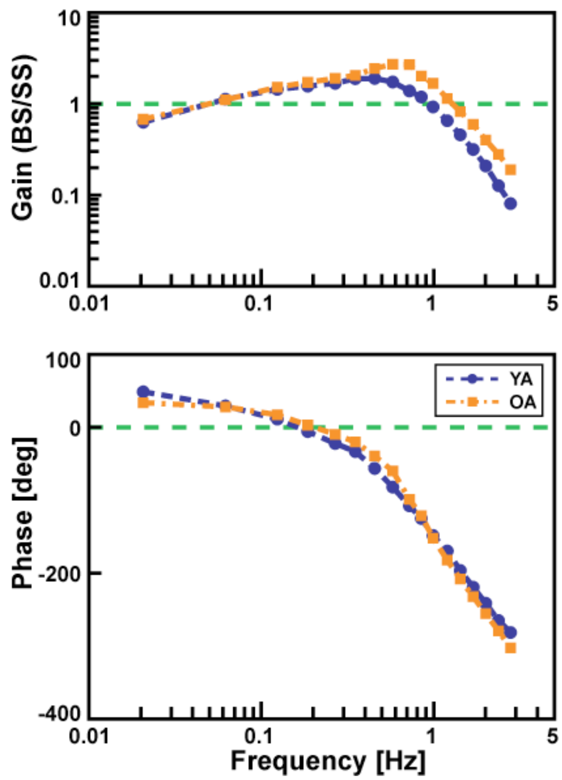

Figure 5. Group Average Experimental Transfer Function. Experimental Transfer Function (TF) estimated from body sway response to SS rotations for young adult (YA) and older adult (OA) age groups. Averages were taken across the three sessions for PRTS-SS trials alone. YA and OA gain curves were similar 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.

Passive control parameters (K and B) tended to be much smaller in value than the corresponding active parameters (Table 2). 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.

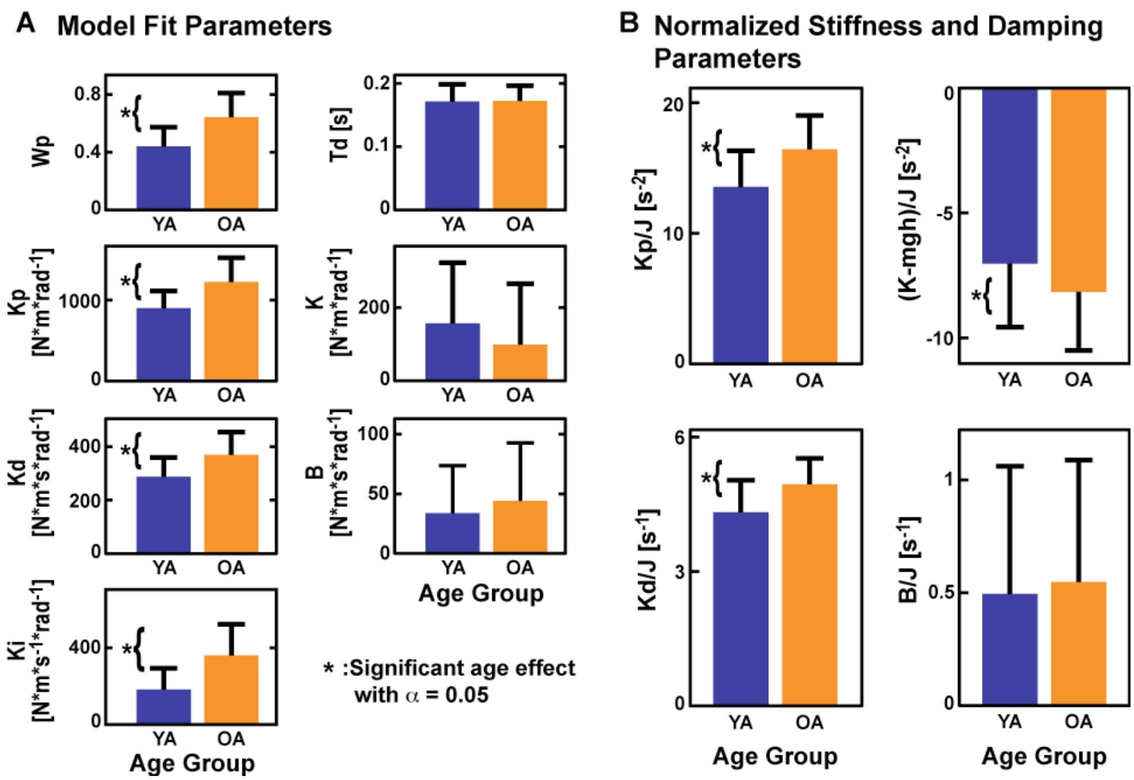

Figure 6. Summary of Model Parameter Estimates. A. Model parameter values estimated by fitting the postural control model to the experimental transfer functions (also see Table 2). Bar plots show average results (mean ± SD) for young adults (YA) and old adults (OA). Significant age differences ($p<0.05$) are indicated by *. B. Normalized stiffness and damping (also see Table 3).

3.4.1.1 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 obtained the normalized parameters K_P/J, K_I/J, K_D/J, B/J and $(K\text{-}mgh_{COM})/J$. 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 [6]. Accordingly, we repeated our statistical analyses on K_P/J, K_I/J, K_D/J, B/J and $(K\text{-}mgh_{COM})/J$, 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 3 and Figure 6).

Table 3. Normalized model parameters (mean ±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 ± 2.5	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

Thus, the differences observed are not attributable solely to differences in body size between the two populations.

3.5 DISCUSSION

Aging is known to affect human balance, with increasing degradation in motor and sensory function [58-60]. 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 [61]. 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

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

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.

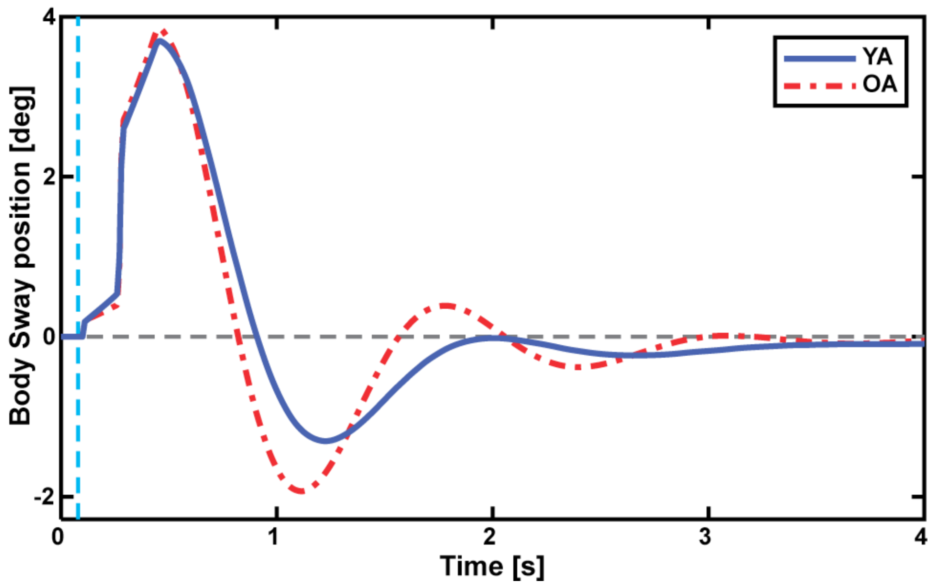

Figure 7. Simulated Model Impulse Response for YA and OA. Plots of the impulse response of the postural control model for nominal values of active and passive stiffness and damping in the young adult (YA) group (solid curve) versus the older adult (OA) group (dash-dot curve). The first peak is similar 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.

These impulse responses are plotted in Figure 7, 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 Figure 8. In particular, Figure 8 shows three plots, corresponding to the ``nominal'' impulse response of the young adults, using their mean parameter values, repeated from Figure 7, 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

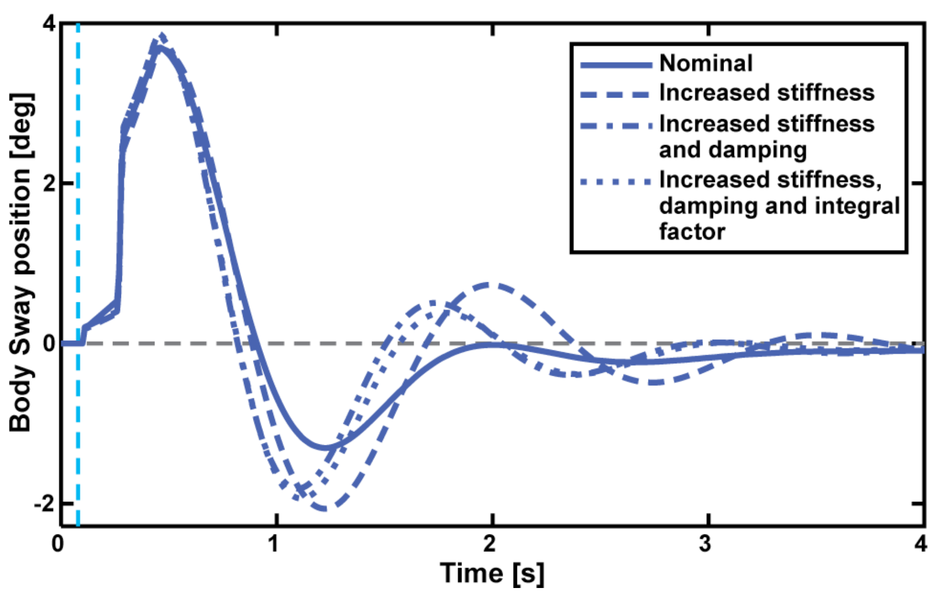

Figure 8. Effects of Stiffness and Damping on Impulse Response Dynamics. Plots of the impulse response of the postural control model for nominal values of stiffness and damping in the young adult population (solid curve), versus increased stiffness only (dashed curve), versus increased stiffness and 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.

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 4. 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, Figure 8, and 4th col., Table 4) 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.

Table 4. 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
1st peak [deg]	3.70	3.68	3.85	3.86	3.86
2nd peak [deg]	-1.30	-2.06	-1.81	-1.94	-1.93
3rd 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

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 [61], it is important to consider active impedance (which is characterized by active stiffness and damping), and not only active stiffness [50, 55, 56]. In a recent study [56], 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 [56]. 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 [6, 9]. 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 3). 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.

3.6 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.

4.0 INTERACTION BETWEEN AGE AND ATTENTION IN THE CONTROL OF POSTURE

4.1 INTRODUCTION

In this chapter, we report findings from model-based analysis of dual-task experiments involving balance and information processing (IP) tasks performed on young and older adults. The aim was to investigate the interactions among aging, attention and sensorimotor integration for balance control. Studies have suggested that dynamic sensory regulation (also called “sensory re-weighting”) of the postural sensory systems (vision, vestibular and proprioception) is a fundamental aspect of balance control, and is a mechanism underlying adaptation to external sensory perturbations [6, 8, 25, 48, 62-65]. Numerous studies have also found that older adults require increased cognitive resources (i.e. attention) for balance [2, 3, 15, 34, 66-72]. However, an understanding of the mechanisms behind this aspect of aging has not been developed.

We investigated the effects of aging and attention on standing balance by fitting body sway data from balance-healthy young and older adults to a previously-developed and validated postural control model [6, 8, 9] that incorporated these two aspects into the model components. In particular, aging and attention are believed to have an effect on the lumped time-delay parameter of the model, and on the dynamic regulation of sensory weights responsible for stance control.

The postural control model incorporating sensory re-weighting sets forth a quantitative framework for exploring this process. This model has been shown to fit experimental data in a variety of conditions [6, 8]. We have explicitly added “attention” to the model and its influence on sensory integration via the sensory weight and time delay parameters of the model (see Figure 9 on p.46). The model suggests that the processing of sensory inputs for balance regulation requires time, and that attention influences this processing time. We use the dual-task paradigm to experimentally examine these attentional effects in the model. An information processing (IP) task that is performed concurrently with a balance task should increase the delay time in the model if the two processes are competing for attentional resources (given that capacity for this processing is limited, i.e. subject to attention). If the IP task does not engage balance-relevant processes, then the time factor in the model will be unaffected. A first test of the applicability of the model is to show that one or more IP tasks do induce increased estimates of the delay in the model. Varying the IP task presented concurrently with the balance task then provides further information on processes shared between balance and information processing. Examining both young and older adults further tests the utility of the model. Older individuals are known to have decrements both in their attention capacity, i.e. capability to perform multiple tasks concurrently [73-75], and in their capability to adjust to balance challenges [34, 66]. The delays in the model should be differentially manipulated by tasks that vary in their processing requirements.

Another aspect of the interaction of attention and postural control that lends itself to study via our proposed model-based approach is sensory selection. In particular, based on experimental findings involving dual-task conditions, Redfern et al. [15] proposed that attention can modulate sensory channels to facilitate specific sensory modalities. The present study further examines this idea and the hypothesis that limited cognitive resources in older adults compared to young adults will result in a larger effect of attention on sensory selection, as manifest in the sensory re-weighting component of the model.

The approach in this study was to: 1) conduct dual-task experiments using specific balance perturbations and concurrent cognitive information processing (IP) tasks; 2) fit the postural response sway data with the proposed model (Figure 9); and 3) use these model fits to test the following hypotheses regarding attention and postural control:

H1) Age will not have an effect on the time delay of the postural control system during postural conditions when there is no concurrent IP task.

H2) Performing an auditory choice reaction-time task (CRT) concurrent with a balance perturbation will have no effect on the time delay in young adults. The same task in older adults will increase the time delay of the postural control system.

H3) Performing more demanding IP tasks (an auditory “vigilance” task (VT) and an auditory “memory” task (MT)) concurrent with a balance perturbation will compete for cognitive resources and manifest as an increase in time delay in the postural control system in both young and older adults. The increase in time delay will be greatest for the MT task and this effect will be greater in older subjects compared to young subjects.

H4) No changes in sensory re-weighting will be observed in young subjects during postural challenges with concurrent IP tasks, compared to baseline (no IP task) conditions, but older adults will exhibit changes in sensory re-weighting during IP tasks. Specifically, older adults will decrease proprioceptive gain, Wp, during concurrent IP tasks.

4.2 BACKGROUND

Upright balance is believed to be maintained through control mechanisms of an ``active'' and ``passive'' nature, which have been shown to be impacted by aging. By ``active control'' we mean neurally-mediated corrective action, generated from sensory information about body position and motion, in response to external disturbances that perturb balance; as such, this action includes significant time delay (on the order of 100-200 ms). In contrast, ``passive control'' arises from reflexive responses, with no higher cognitive function and hence minimal time delay. The passive mechanisms involved in postural control include the intrinsic visco-elasticity of muscles and stiffness of tendons, providing some gravity-countering forces in much the same way a spring generates a counter-force when it is displaced from its resting equilibrium [50-52]. The contributions of passive mechanisms to maintenance of upright stance are relatively minor in comparison to active control mechanisms, during perturbed conditions such as floor tilts, moving scenes or galvanic stimulation [6, 9].

As we age, balance function starts to decline and control of stance can become difficult for many older adults [2]. Decline of balance function is considered one of the factors likely to be responsible for falls in a large percentage of older adults [2, 11]. Results from previous studies have suggested that diminished balance function can induce increased use of attention to maintain upright stance [2]. However, the exact underlying mechanisms that are involved in the control of posture and the interactions between aging, attention and sensory integration for postural control are not fully understood yet.

A model-based approach using a variety of different sensory perturbations [1, 6, 8-10], has been shown to be effective for studying the postural control problem and unveiling some of the mechanisms involved in stance control. Very recently, this approach has been used to investigate the effect of aging and attention on control of balance and their interaction [1]. This original study found that attention towards a concurrent IP task has an impact on the time delay associated with processing speed of the sensorimotor integration process for control of stance. In particular, it was found that older adults have a comparable processing speed to that of the young adults in mildly challenging balance conditions, but when attentional requirements increase due to dual-task conditions, the time delay associated with the sensory integration process changes depending on the cognitive demand of the attentional task and depending on age [1].

Based on the results of this original work [1], the present study was proposed with the intent to further expand the current knowledge about the interactions among attention, aging and sensorimotor integration for posture control. This study uses a dual-task approach consisting of moving platform perturbations presenting with concurrent cognitive (information processing (IP)) tasks that included: 1) a choice reaction time (CRT), in which subjects need to press a button as fast as possible in response to a high or low frequency tone; 2) a vigilance task (VT) in which subjects are asked to count silently the number of occurrences of either the high or low frequency tones, which requires a sensory focus and use of memory to store and count, but not a motor action like the CRT; and 3) a memory task (MT) in which subjects must memorize a sequence of number-word pairs presented prior to the start of the trial, and report these pairs back at the end of the trial. The protocol also included balance (moving platform) conditions alone, without IP tasks, as well as seated IP tasks, as control conditions.

Estimation of the experimental frequency response of the postural control system are presented and key results on the effect of age, attentional condition and visit number (time effect) are described. Model fits to the data were performed following the methodology proposed in [6] and model estimates are reported and discussed in the following sections of this chapter.

4.3 METHODS

4.3.1 Subjects

Thirty five young adults (18 males and 17 females, 23 ± 3 years, range: 18 – 30) and 25 older adults (11 males and 14 females, 74 ± 5 years, range: 66 – 88) participated in this study. Subjects gave informed and written consent before participating in this study. The experimental protocol was approved by the Institutional Review Board at the University of Pittsburgh. All subjects were required to complete and pass a set of screening examinations prior to any experimental tests, to ensure absence of any balance abnormality. The screening procedures consisted of standard clinical testing for vestibular, oculomotor, and balance function, including caloric and rotational vestibular tests, vibration and cutaneous pressure sensation, and computerized dynamic posturography.

4.3.2 Experimental Setup and Protocol

4.3.2.1 Experimental Setup

The experimental procedures took place in the balance testing laboratory of the Medical Virtual Reality Center at the University of Pittsburgh. The laboratory is equipped with a dynamic posturography platform (NeuroTest, Neurocom International, Inc., Clackamas, OR). The platform was used to induce rotational platform perturbations during standing and acquire center of pressure data in response to those perturbations (Figure 2). A harness system was used to ensure subject safety and injury prevention from falling during testing. The harness did not impede sway, or give any positional feedback to the subject. Body sway was measured using a magnetic tracking device (Fastrak, Polhemus, Colchester, VT) that tracked the position of two sensors on the subject's body: one positioned on the lower back, at the height of the iliac crest, and one positioned at the top of the head.

To induce balance perturbations, subjects were exposed to rotational movements of the underfoot platform. The platform, or support surface (SS), rotated about the ankles with a random-like motion; the velocity profile of this motion followed a pseudo-random ternary sequence (PRTS). The PRTS has been shown to have a wide spectral bandwidth over the range of frequency of interest for standing. In particular, the velocity profile of the PRTS-SS has spectral and statistical properties similar to a white noise signal [6, 9]. Another type of SS rotation tilt was also used to induce body sway, in which the platform followed the subject's body sway angle with a 1:1 ratio. This condition, known as sway-referencing SS (SR-SS), is known to induce increased reliance on sensory orientation cues mainly that are of vestibular origin [6, 8, 9], particularly in an eyes close condition. Data from the SR-SS condition are not presented here; processing was only performed on data from the PRTS-SS condition.

During the PRTS-SS condition, the platform rotated pseudo-randomly according to the integral of a PRTS, with peak-to-peak amplitude of 2° and a cycle duration of 60.5 s (see [1, 6] for details). This amplitude was chosen to prevent any saturation in the response of the postural control system as reported in [6, 9]. Recorded body sway in response to the SS perturbations were used to estimate dynamic properties of the postural control system over a range of frequencies (0.016 – 2.230 Hz). The PRTS-SS condition allows for an input-output analysis and estimation of model parameters [1, 6, 9]. To ensure adequate steady-state conditions for estimating postural control characteristics, while avoiding long trial durations in order to guard against fatigue effects, particularly in older adults, two consecutive cycles of the PRTS-SS were presented during a trial, for a total perturbation interval of 121 s. The same trial condition was presented to the subjects three times within a session in a randomized order.

Subjects stood on the platform with the axis of rotation of the SS aligned about a medial-lateral axis passing through the malleoli, thereby inducing body sway in the sagittal plane (anterior-posterior (AP) direction) in response to SS rotations. Subjects were asked to stand with a comfortable stance width for their feet. Foot placement was marked at the start of the first trial to maintain the same position from trial to trial and across visits for each subject.

The posturography system was also equipped to allow concurrent information processing tasks (IP) during the postural conditions described above (SS tilt). Three IP tasks were performed, consisting of 1) a choice reaction time task (CRT) that involved auditory tonal stimuli presented via a headphone system using ear inserts (Etymotic, Inc.); 2) a vigilance task (VT), that involved the same tone stimuli as in the CRT task, however, in the VT subjects were asked to silently count either the total number of high or low frequency tones they would hear during a trial and report the number at completion of the trial; and 3) a memory task (MT), involving number-word pair memorization and recall. The IP tasks were selected based on past research and to explore specific cognitive processing that might interfere with posture control and that is known to be associated with each of these three tasks. The CRT task requires auditory stimulus detection, making a choice, and a quick reaction. The emphasis is on simple, but fast, processing that involves a motor component (pressing hand-held buttons). Slowing of processing speed is a common finding with age [42]. In contrast, the VT emphasizes accuracy of detecting the high/low frequency tone and provides a sensory focus, with a memory requirement and no musculo-skeletal action (unlike the CRT task). The VT has shown a different interaction effect with age and postural control performance than the CRT task [1]. In addition to the CRT and VT studied previously [1], the MT was added in this study to test whether memory may be involved in control of posture. A more detailed description of IP tasks 1) and 2) (CRT and VT, respectively) is reported in [1], while that of IP task 3) (MT) is presented below.

4.3.2.2 Experimental Protocol

To participate in the testing experiments, subjects were required to pass a battery of screening procedures to ensure absence of any neurological and cognitive impairment that would affect their control of balance and performance of the attentional (IP) tasks. After screening and prior to experiments, a training visit was first conducted to familiarize subjects with the IP tasks and also to determine baseline performance values. Baseline values were used to ensure subjects were at similar performance levels at the start of actual testing. Three experimental sessions on three separate days were performed after the training sessions, with at least two days between each visit. One IP task, selected randomly, was conducted per testing visit. Prior to the start of experimental tests, subjects rehearsed the IP task for that visit, to re-familiarize them with the task.

As indicated earlier, description of the CRT and VT tasks is reported in [1]. The Memory Task (MT) used in this study was adapted from a test proposed by Brooks [76] and then later applied to postural research by Maylor et al. [5]. They found that memorization had a negative effect on stability and that ability to recall a given set of words decreased with age. In our protocol, for each MT trial, subjects were verbally given a pre-recorded sequence of number-color pairs (e.g. “One – blue, two – green, …”) to be memorized before the start of the trial, and instructed to keep this sequence in mind for recall at the end of the balance trial. The rationale for this task was to induce a focus on cognitive activity that involved memory while the subject also performed a balance task. As shown in [5], subjects’ ability to recall decreased with age.

Accordingly, we chose to normalize the cognitive load across subjects, by determining during training the number of pairs that subjects could memorize and recall with no more than two mistakes. This error rate was achieved during training as follows. Initially, five number-color pairs were presented to the young adults, and three were presented to the older adults. If no error was made during the recall phase, then the number of pairs was increased by two for the next trial. If one or two errors were made during recall, then the number of pairs was kept the same for the next trial. If the number of errors was three or higher, then the number of pairs was decreased by one for the next trial. This training protocol continued until the subject converged to the number of pairs for which the subject made 1-2 mistakes during three consecutive trials. In general, subjects performed about 8-10 trials before completing training. The maximum number of colors available for pairing was 12, allowing for a maximum of 12 different pairs during a trial. Only a few subjects in the young adult group were able to reach this limit. (The 12 colors used were White, Black, Blue, Red, Green, Orange, Brown, Yellow, Purple, Pink, Silver, and Gold).

At the completion of the balance trials involving the MT task, subjects were asked to recall the MT sequence by verbally stating the color corresponding to the numbers of the sequence, which were played back using custom made software in a random order (e.g., "Three - _____, one - ____, four - ____, …"). Subjects were allowed up to 5 seconds to recall the color, and the number of correct responses was tallied.

On the first testing visit, prior to the start of trials, anthropometric measurements were taken so that the body inertia (J), mass (m), and height (h) used in the model could be obtained [57]. During each visit, subjects underwent a setup phase and a short IP task refresher to bring the subject performance close to that shown during training (approx. 30 minutes). The testing phase followed and lasted for about 1.5 hours. Each trial was characterized by a combination of a balance condition (seated or standing on the SS) and an IP condition (NoIP (control; balance only), CRT, VT, MT) for the task chosen for that day. For each visit, there were a total of 15 trials resulting from two platform conditions (PRTS-SS and SR-SS) x two IP conditions (NoIP and visit-specific IP (CRT, VT or MT) x three repetitions per condition, plus three seated trials in which subjects performed only the IP task. Trial conditions were pseudo-randomly presented to each subject, in that IP conditions were randomized across visits, and platform and IP/NoIP conditions were randomized across trials, with the caveat that subjects could not experience consecutive identical platform-IP trials.

During each experimental trial, subjects stood upright, with eyes closed, on the posturography platform for the duration of each trial; a three minute seated rest followed each trial. 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 181 s, consisting of a 121 s perturbation interval (PRTS-SS motion) preceded and followed by a 30 s quiet stance period on a fixed SS.

4.3.3 Data Measurement and Analysis

In this work we present analyses and results only from the PRTS-SS trials for each subject, consisting of three moving platform trials without any IP task (NoIP condition), and three moving platform trials per IP task per visit. Data from these trials were used to estimate body sway characteristics in response to platform perturbation with and without a cognitive dual task: IP vs NoIP condition.

Body sway (BS) angle and velocity in the anterior-posterior (AP) direction (sway in the sagittal plane) from COP data were processed based upon methods in [1]. Angular displacement of the COM in the sagittal plane was estimated from the COP signal, using a small angle approximation [1, 6, 10],

$$Hip_{AP}(t) = h_{COM} \sin BS(t) \approx h_{COM} BS(t) \tag{8}$$

Body sway and platform rotation measurements for the 121 s of pseudo-random platform motion were divided into two segments of 60.5 s duration (PRTS-SS cycle length). For each cycle, the power and cross power spectrum were estimated via the discrete Fourier transform (DFT) of each time series; for further details see Chapter 3.0 or [10]. These power spectra estimates were then averaged across the two cycles. Further spectral smoothing was also applied to reduce variability in the spectral estimates at each frequency, especially at higher frequencies, as described in [6, 9, 10]. The resulting smoothed power and cross-power estimates contained 17 data points over a range of frequency from 0.016 to 2.23 Hz that were evenly spaced in a log-scale.

4.3.4 Model Fits to Experimental Transfer Function

A least-squares fit of the model (Figure 9) to the smoothed frequency response data was made to estimate model parameters for each trial condition and subject. (See Chapter 3.0 and [10] for details.)

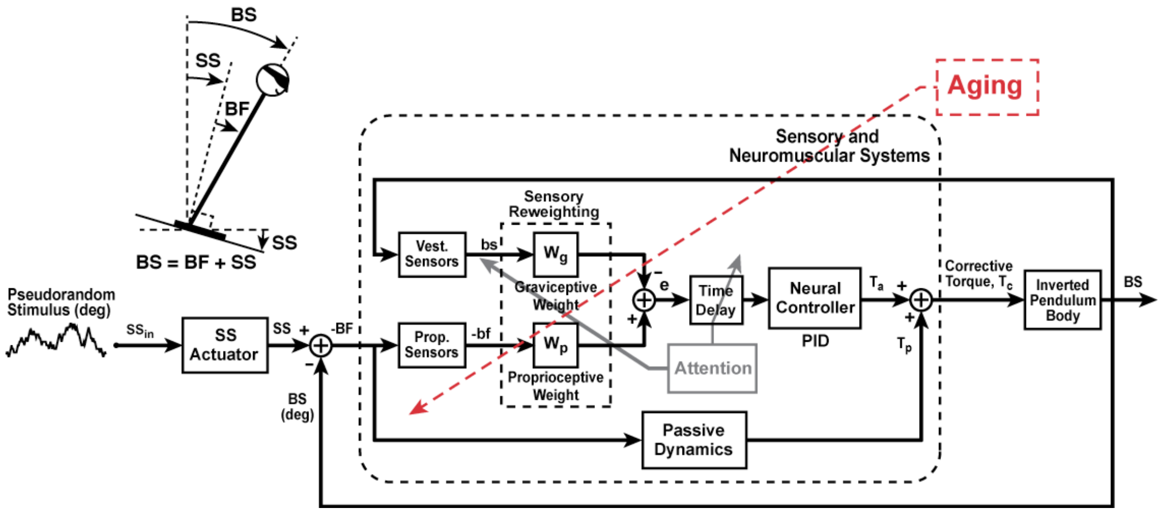

Figure 9. Feedback Model of Postural Control – AP Sway. Sensory channels are limited to vestibular (graviceptive) and proprioceptive sensory channels. The mechanical perturbation provided by the support surface (SS) is indicated in the schematic as a pseudorandom stimulus. Body sway with respect to the 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." While attention is thought to only have an effect on the "Sensory Reweighting" process and the overall postural control time delay. (Model schematic adapted from [1, 10].)

4.3.5 Statistical Analysis

An independent sample t-test was used to test for significant age differences at each of the 17 frequencies of the experimental transfer function estimates. A method suggested by Benjamini and Hochberg [77] was applied to the resulting p-values for both the gain and phase functions of the TF estimates, to control for false discovery rate, as similarly used in [78, 79].

A mixed-factor repeated-measures ANOVA was performed on all model parameters. The independent factors were Age (Young and Older Adults) and Visit (1, 2, and 3). Data were separated by IP condition (NoIP and IP). Data for the NoIP (control) condition were analyzed to test the hypothesis of a significant difference in the postural control characteristics from visit to visit (effect of time). If a difference were to be found then NoIP and IP data would be analyzed separately. Another repeated-measures ANOVA analysis was performed on all model parameters in which the independent factors were Age (Young and Older Adults) and attentional task (CRT, VT, and MT). Data were again separated by IP condition (NoIP and IP) as when we tested for the effect of time. Data for the NoIP condition were analyzed to test whether there was a difference across visits due to some unknown and indirect interaction with the task performed in each visit. If a difference were to be found, then NoIP and IP data would be analyzed separately.

To account for any possible body size differences that might need to be included as covariates in the ANOVA, an independent sample t-test was performed on subject's body parameters, namely the moment of inertia (J), subject mass (M), subject height of the COM (h_{COM}), and the composite variable mgh. Only h_{COM} was found to be significantly different ($p<0.04$), with young adults being taller on average than older adults. However, because h_{COM} doesn't appear directly in the expression of the model TF, but only via the composite variable mgh, which was not significantly different between the two age populations, there was no need to include any of the anthropometric body parameters (J, M, mgh) as covariates to account for body size differences in the ANOVA.

Table 5. Summary of anthropometric data from the subjects included in the study – AP sway. The * indicates a significant difference across age group with α=0.05.

Age Group	Height of COM* [m]	Mass [kg]	Moment of Inertia (J) [kg·m^2]
Young Adult	0.95 ± 0.05	74.8 ± 17.3	75.6 ± 20.9
Older Adult	0.92 ± 0.04	75.7 ± 13.5	72.0 ± 16.7

One sample t-test on the difference in model parameter estimates between IP and NoIP conditions was used to test whether there was a significant difference between NoIP and IP trials, for each attentional visit (CRT, VT, and MT). This statistical test was run first over all subjects to test the effect of performing a cognitive task, and also individually for each age group.

All statistical analyses were performed using SPSS 11.0 for Mac OS-X with an α-value of 0.05 for determining significance in each test.

4.4 RESULTS

Data from one older subject was excluded because his anthropometric parameters (M and J) were larger than what the motor that drives the floor rotations could handle, as determined by substantial differences in the measured platform motion compared to the desired input PRTS platform profile. The resulting experimental transfer function gain and phase curves deviated from the norm seen in other subjects, and the fitting procedure was not able to provide good fits to the data.

4.4.1 Experimental Transfer Functions

Mean experimental transfer function gain over all PRTS-SS trials for the NoIP condition was higher for the young adult group in the lower frequency range (below 0.3 Hz), while that of the older adults was higher above this frequency and peaked just below 0.6 Hz (Figure 10). Analogous differences were found for the PRTS-SS IP condition trials as well. These results are similar to those found in the ML direction of sway [10] but at a slightly higher frequency (0.7

Hz). Statistical tests on the gain function data resulted in significant differences (α=0.05) between the young and older groups at various frequencies. At 0.05 Hz, the gain of the YA was higher than the gain of the OA. At frequencies of 0.58, 1.90 and 2.23 Hz the gain of the OA was higher than that of the YA (Figure 10). The gain curve of the OA presented a slight peak at a frequency of 0.58 Hz.

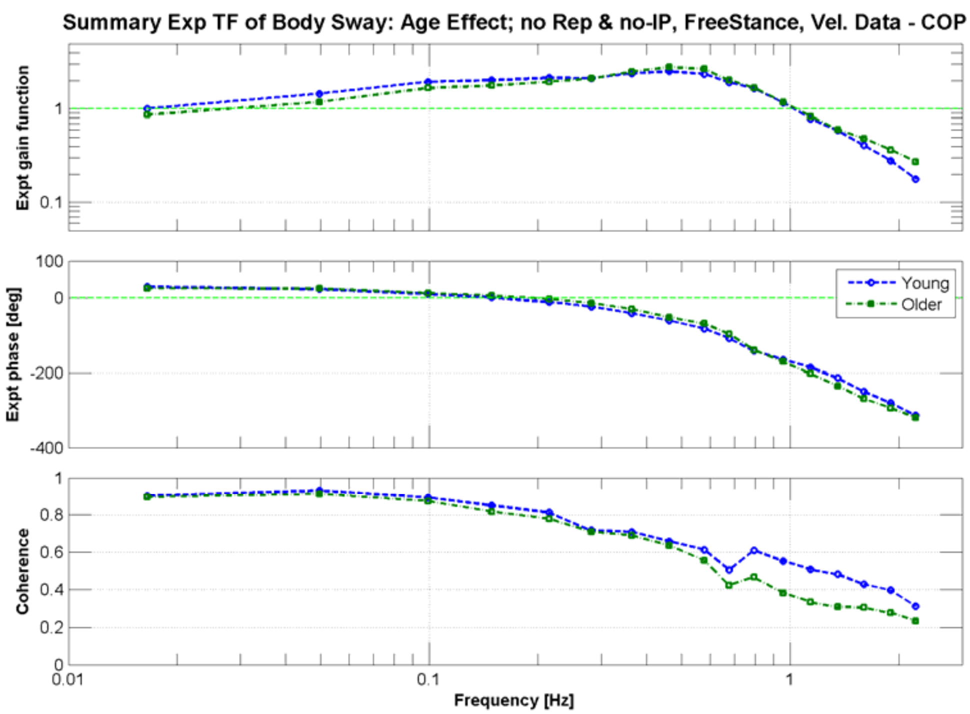

Figure 10. Spectral Estimates of Body Sway Characteristics. Experimental Transfer Function (TF) estimated from body sway response to SS rotations for young adult (Young) and older adult (Older) age groups. Averages were taken across the three sessions for PRTS-SS trials alone. Young gain curve was higher in the low-frequency range (below 0.3 Hz) as compared to the Older curve, 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 0.6 Hz. Phase curves were similar in the very low frequency range (below 0.15 Hz), while above this range till 0.8 Hz the Older phase curve was characterized by less lag than that of the Young. At frequencies above 0.8 Hz, the Older phase lag was greater than that of the Young.

The phase curves of the YA and OA were similar in the very low frequency range (below 0.15 Hz). The phase lag of the OA was less than the YA in the mid-high frequency range (until 0.8 Hz) and then became higher than that of the YA above frequencies of 0.8 Hz. The phase curves were significantly different from one another in the range of 0.15-0.46 Hz, in which the phase of the OA presented less lag than the YA (Figure 10).

The coherence function estimates were relatively high until the frequency of 0.9 Hz indicating a good linear interaction between body sway response and platform perturbation [6, 80].

4.4.2 Model Fits and Parameter Estimates

Experimental transfer function fits were inspected for accuracy of the fits. Mean-square-error (MSE) between model and experimental transfer functions was calculated over the 15 frequencies used for fitting the data (freq. 2 to 16). MSE was on average 0.101 and 0.282 for the YA and OA age groups, respectively, corresponding to 3% and 7% relative errors with respect to the magnitude of the experimental TF, respectively.

4.4.2.1 Visit number and Aging Effects

Repeated measures ANOVA was performed on all the estimated model parameters (Wp, K_P, K_D, K_I, T_d, K, and B) from the PRTS-SS platform condition for the No-IP (i.e. control) task condition. Independent factors "age group" (YA and OA; between-subjects effect) and "visit number" (1, 2, or 3; within-subjects effect) were included in the model. No significant effect of the visit number ($p>0.09$) was found for any of the parameters ($p>0.09$). A significant interaction between age group and visit number was found on the PID parameters (K_P, K_D, K_I)

and on the passive damping parameter (B) ($p<0.04$), for the NoIP condition data. The OA PID parameters progressively increased visit after visit while B was lowest during the first visit and increased and leveled off at the second visit. T_d had a similar trend but no significant interaction effect was found for this parameter ($p>0.27$).

Table 6. Estimated model parameters (mean ±SD) for young and older adult groups for the NoIP condition - AP

		K_P [$N{\cdot}m{\cdot}rad^{-1}$]	K_D [$N{\cdot}m{\cdot}s{\cdot}rad^{-1}$]	K_I [$N{\cdot}m{\cdot}rad^{-1}{\cdot}s^{-1}$]	K [$N{\cdot}m{\cdot}rad^{-1}$]	B [$N{\cdot}m{\cdot}s{\cdot}rad^{-1}$]	T_D [ms]	w_p
V1	YA	1098 ± 288	337 ± 98	152 ± 104	12 ± 40	59 ± 27	196 ± 19	0.62 ± 0.08
	OA	1105 ± 249	331 ± 73	193 ± 133	2 ± 8	63 ± 45	203 ± 19	0.63 ± 0.07
V2	YA	1087 ± 289	338 ± 99	145 ± 88	8 ± 35	58 ± 52	197 ± 21	0.61 ± 0.08
	OA	1172 ± 289	342 ± 79	263 ± 192	0 ± 0	104 ± 115	212 ± 37	0.63 ± 0.13
V3	YA	1063 ± 280	335 ± 97	155 ± 116	10 ± 44	50 ± 30	194 ± 16	0.60 ± 0.07
	OA	1172 ± 284	346 ± 82	296 ± 248	0 ± 0	98 ± 120	211 ± 41	0.63 ± 0.14

A significant age effect was found on the active controller parameters (PID parameters and T_d) and the passive damping (B) ($p<0.05$) for the NoIP condition. The values of those parameters were on average higher for the OA than for the YA group (Table 6).

4.4.2.2 Attention and Aging Effects

Two repeated measures ANOVA were performed on all the model parameters; one for the NoIP (i.e. control) condition and one for the IP (i.e. dual-task) conditions. Independent factors "age group" (YA and OA; between-subjects effect) and "attention task" (CRT, VT, or MT; within-subjects effect) were included in the model. A significant attention task effect was found for the

T_d and B parameters ($p<0.03$) for the IP trials. In addition, the T_d, B and W_p estimates obtained during the NoIP trials were significantly affected by the visit on which the CRT, VT, and MT IP task conditions were performed ($p<0.03$). No significant attention or visit effect was found on the other model parameters ($p>0.1$).

The average values of Wp and T_d did not present any evident and consistent trend across attention conditions and the changes were different between IP and NoIP data (see Figure 11, top plots). For both IP and NoIP data sets no significant interaction was found between age and attention condition for both these two model parameters ($p>0.05$). T_d was lower during the VT visit for both YA and OA (see Figure 11, top plot, opaque bars); this behavior was more pronounced for the OA group. For the IP condition data (Figure 11, top plot, brighter bars), T_d was highest for the CRT and it went down for the VT and it was the lowest for the MT visit. Again, this trend was more pronounced in the OA group. Because of the significant effect of the attention condition on T_d for both IP and NoIP conditions, in order to test how performing the IP task interacted with the control system, we subtracted each value calculated for the No-IP

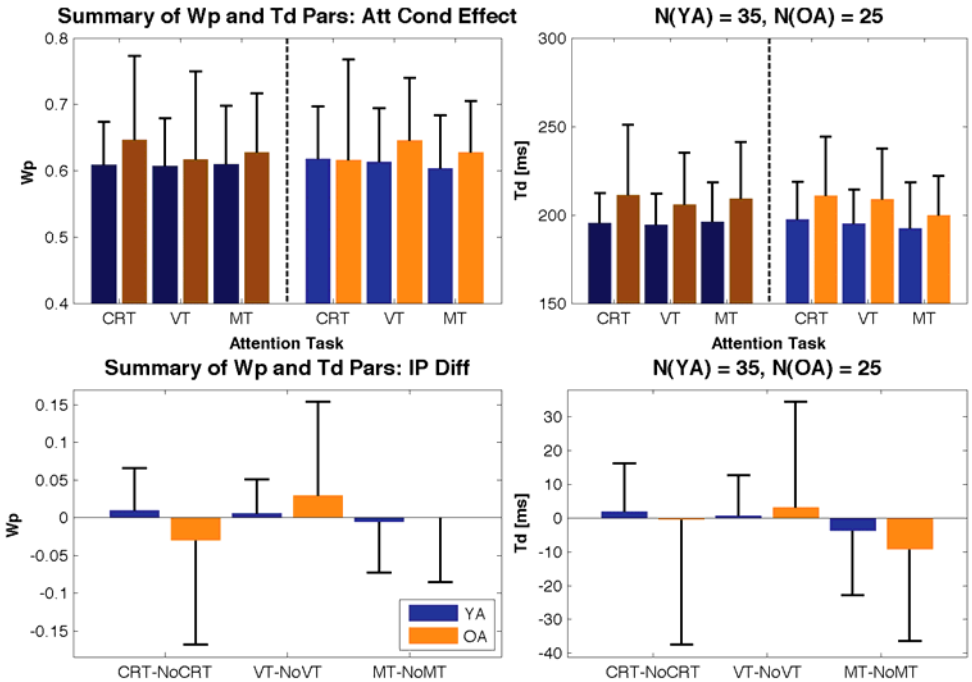

Figure 11. Group Averages of Wp and T_d. Top plots show average (across subjects) of Wp and T_d (± 1SD) for each attention task visit (CRT, VT, and MT) and the plots are separated in NoIP task data (left of the dashed line) and IP task data (right of the dashed line). YA data is reported in blue color while OA in orange. Bottom plots are the difference between estimates of Wp and T_d during IP task minus corresponding estimates during NoIP task conditions.

condition from its corresponding value for the IP condition. By doing this, each subject acted as its own control as well. The repeated measures ANOVA performed on the difference in T_d showed that both age (YA vs OA) and attention task condition (CRT, VT, and MT) did not have a significant effect ($p>0.1$). However, pooling together all the subjects (YA and OA) showed that T_d was on average significantly lower (by 6 [ms]) during the MT task as compared to just standing on the platform ($p<0.05$). The difference was not significant for the CRT and VT visits ($p>0.5$) and the difference was on average 1 and 2 [ms], respectively. When separating the data by age group (YA and OA separately), T_d was on average lower by 4 and 9 [ms], respectively for YA and OA during the MT visit, but these differences were not significant (p>0.09) (see Figure 11, bottom plot). The difference in T_d for CRT and VT was close to 0 [ms] and also not significant (p>0.4).

Repeated measures ANOVA performed on the difference in Wp estimates between the IP and NoIP conditions was not significant (p>0.1). The difference in Wp was also not significant for both of the cases in which data was pooled together from both age groups or when the two age groups were considered separately (p>0.1). The results related to T_d and Wp are in contrast with the earlier work by our group that had a similar protocol [1]. Figure 11 (bottom plots) shows that even though the difference in both Wp and T_d was not zero, the variability in those measures was too high to support a consistent change.

4.5 DISCUSSION

The main aim of this study was to investigate the interactions among aging, attention and sensorimotor integration for balance control. To do so, we adopted a model-based approach in which we fit body sway data from healthy young and older adults to a previously-developed and validated postural control model [6, 8, 9]. This approach enabled us to test our hypotheses that aging and attention would have an effect on the lumped time-delay parameter and on the dynamic regulation of sensory weights in the model. The data presented here showed that the delay time and the reliance on proprioceptive information from the ankle joint included in the model increase with age for body sway in the AP direction. The influence of performing an attentional task was less clear. In particular, while there was an effect of IP task on the time delay T_d and passive damping parameters B, this was confounded by the presence of a visit effect as well on these same model parameters (as well as W_p) for the No-IP control conditions performed on each visit. (Recall that each visit was characterized by one IP condition for that day, and included No-IP control conditions.) Analysis of the *difference* in parameter values for IP vs. No-IP trials per task/visit was not significant.

Previous work by others has shown that attention has an effect on postural control [2, 15-17], and results from an earlier study with a similar protocol to the one used here demonstrated the applicability of the model by showing that some IP tasks did induce increased estimates of the delay in the model [1]. Building on this study by incorporating another IP task and conducting experiments on a larger population, we expected that varying the IP task presented concurrently with the balance task would provide further information on processes shared between balance and information processing, as well as shed insights on these interactions with aging. Older individuals are known to have decrements both in their capability to perform multiple tasks concurrently [73-75], and in their capability to adjust to balance challenges [34, 66]. Lastly, with our proposed model-based approach we could test the idea that attention can influence sensory selection by facilitating specific sensory modalities [15]. The present study examined the hypothesis that limited cognitive resources in older adults compared to young adults would result in a larger effect of attention on the sensory re-weighting component of the model.

Results from this study show that the characteristics of postural control change with age. This result was first evident in the changes seen in the experimental TF gain and phase curves between the two age groups (Figure 10). The average TF gain curve in the OA group was lower than that of the YA in the lower frequencies (below 0.3 Hz) and it presented a slight peak just below 0.6 Hz. In particular, the TF gain curve in the OA was significantly different from that of the YA at 0.58 Hz. The average phase curves were significantly different between the two age groups in the mid-frequency range of 0.15-0.46 Hz in which the phase in the OA group was lagging less than the YA. This result changed above 0.8 Hz, in which the OA group phase lag was larger than that of the YA, but the difference was not significant. Similar changes to the ones seen in the mid-high frequency range were also observed in a study done in the medial-lateral sway direction [10], however, in the present study differences between the two age groups were found also in the lower frequencies for gain and phase curves.

The changes found in the experimental transfer functions curves were analyzed and interpreted via the estimated values of the model parameters. In the present study, the changes observed in model parameters were similar (but not identical) to those found for the ML direction of sway [10]. In particular, K_P, K_D, and K_I for the AP case were significantly different between the two age groups ($p<0.01$) and these parameters were on average higher in the OA than the YA group; however, unlike the ML case, in the AP direction no significant difference was found in the passive stiffness K, and a significant difference was found in the passive damping B between the two age groups ($p<0.04$); on average B was larger in the OA than the YA group.

The time delay, T_d was significantly different in the OA group in both NoIP and IP condition and for all attentional conditions or visit number ($p<0.05$) and this parameter was found to be on average larger in the OA than in the YA group. As we know the main effect of the T_d in a second order system with feedback is seen in an increase of the lag at higher frequencies. We also know that there exists some correlation between T_d and the stiffness and damping parameters in this model [6]. Specifically, in the study reported in [6], stiffness K_P

increased and time delay T_d decreased as the amplitude of the perturbation increased, while damping K_D remained relatively unchanged. From a control perspective, increased stiffness alone is undesirable as it can lead to resonance (sustained oscillations), which can be mitigated by a concurrent increase in damping K_D, as discussed in Chapter 3. Although K_D did not change in the study of [6], the effective damping was increased by a *reduction* in the T_d, which effectively reduced the phase lag at higher frequency as would be seen by increasing K_D [6].

In the present study, we found a concurrent increase in K_D with K_P, along with an increase in T_d. However, as a concurrent increase in K_D with K_P is desirable from a control systems perspective, the increase in T_d is not because it is potentially destabilizing: for fixed K_D and K_P, the model body sway exhibits increased oscillations as T_d increases. Similarly, for fixed T_d and K_P, the model body sway also exhibits increases oscillations as the damping K_D decreases. Thus, body sway becomes increasingly oscillatory as T_d increases, and/or K_D decreases. Therefore, the increase found in K_D for the OA group is desirable, in that it helps to somewhat counter the decreased stability associated with increased time delay T_d. However, it is unknown if there is a causal association between the changes in K_D and T_d, or if from an evolutionary stand point the physiological changes that happen with age in the muscles, sensory systems and the nervous system causes those parameters of the posture control system to change accordingly in a coordinated fashion to prevent instability. We could speculate that T_d could increase with age for a number of reasons. In this model, T_d represents a lumped time delay that includes the time it takes from the information from each sensory systems to reach the CNS, to then process (or integrate) the information to detect body motion and position in space, and to make the necessary adjustment to keep the body upright. Therefore, any reduction in the functionality in any of these components would cause an increase in T_d.

As a final remark about the significant change observed in T_d with age ($p<0.5$) in AP sway is of interest, particularly in regard to a known and not yet solved question about whether control of posture in AP and ML directions can be assumed to be identical. Recall that for ML sway, no difference in T_d was found with age (Chapter 3.0). This could possibly suggest that changes due to age have a more prevalent effect for control of posture in the AP direction than in the ML direction from the time delay involved in the generation of the active component of the control torque. This may be related to the complexity of a multi-segmental structure of the body frame in the frontal plane than in the sagittal plane, as discussed in [61, 81]. Specifically, for the ML direction of sway, a trunk (upper body) roll relative to the lower body is observed [61], and a more complex model is required to characterize trunk stabilization as compared to whole body motion [81]. This additional level of complexity may restrict the allowed change in T_d that we would expect occurring with age resulting from a reduced of functionality in any of the sources responsible for T_d. If increased complexity in the control of posture in ML is assumed to be true, then the control system may not have a luxury to increase the delay without jeopardizing the overall stability of the postural control system in the ML direction. Further investigation is required to properly answer such a question.

A possible explanation for the differences seen between AP and ML direction of sway could be associated to possible differences in the control of balance between ML and AP sway directions. First, body sway mechanics are different in the ML and AP directions. The legs form a parallelogram in the frontal plane (ML sway) and this is not the case for AP sway (malleoli aligned on the same axis of rotation). This difference may affect the dynamics of the response to platform rotations causing differences in control parameters, and how the muscles controlling upright posture are activated. The generation of the control torque at the ankle joints is the result of muscle activation in both legs; however, because of the different structure of the legs in the ML vs. AP direction, changes that may occur in the postural control system with age may not have a large effect in the lower frequencies as found in this study for body-say in the AP direction.

A recent article by Goodworth and Peterka investigated how control of the upper body with respect to the lower body can be represented with a more complex feedback model [81] than the one previously used for whole-body control in the frontal plane [9]. However, in Goodworth and Peterka's study, motion of the lower body was restrained in such a way that support surface tilt was transferred directly to pelvis tilts via a system of carriages as described in detail in [81]. This system allowed one to consider the problem of upper body sway control in a more controlled way without worrying about multi-joint sway involving both lower and upper body as it occurs in free-standing body sway. Even though body sway was constrained to occur only in the upper body, Goodworth and Peterka's findings are important because they were able to show the increased complexity in the control of body sway in the ML direction. In particular, other mechanisms are responsible for spinal stabilization, such as medium- and short-latency control loops. These are in addition to the more central control processes [81] which are the only ones accounted for in the modeling of whole body-sway by using a single lumped time delay that has been shown to be adequate for the AP direction [1, 6].

Another potential reason for differences in the changes seen in TF curves with age between AP and ML body sway could be the fact that the stimulus amplitude was different between the two studies. In [10] (see Chapter 3.0) the peak-to-peak amplitude of the SS tilt was 4° as opposed to 2° used in the present study. Previous work by Peterka and colleagues [6, 9] showed that the amplitude of the stimulus induces sensory reweighting and this may be partly associated to the changes seen between the results from the ML and AP direction of sway.

This study was designed partly based on a previous study [1]. Based on the finding of that study, we stated some hypotheses that ultimately were not supported by the findings of the current study. In particular, Mahboobin et al. [1] found a significant change in sensory cues responsible for balance control depending on the dual task performed, and also depending on age. They also found that the overall lumped time-delay T_d changed depending on the task and age, relative to the NoIP condition. While our current study was based on this past work, there were some differences in the set-up and protocol. In the study of [1], the control condition (NoIP) was performed on a separate visit. In the present study, every testing visit included the NoIP condition, to control for possible variation in subjects response from visit to visit. While we found that visit number did not have a significant effect on model parameter estimates, there was an interaction effect between age and visit number that was particularly pronounced in the OA group. This could explain why in [1] a significant change was found in the Wp and T_d across visit and indeed between the control condition and task condition, while this result was not found in the present study. The difference could be due to a change in Td seen in the OA that is not seen in the YA and that the change is also correlated with time for the OA group.

The results from this study showed that the phase lag in the low to mid frequencies was lower in the OA but it then increased in the higher frequency ranges, as compared to the YA group. This can be explained by the combined changes of K_P, K_D and T_d. In fact, stiffness and damping were found to be higher in the OA than in the YA and we know that these two parameters contribute to the system dynamics in the low and mid frequency ranges; this could be why phase lag was less as compared to that of the young in that frequency range. However, as we go to higher frequency the effect of T_d on the phase becomes more pronounced and this could explain the increase in the phase lag seen in OA in the higher frequency range when compared to YA.

One aspect that is always important to keep in mind when using a model-based approach to study real-life systems is to make sure the assumptions made are satisfied. One of our measures of goodness for our fits was the normalized MSE between the experimental and the model TFs that was found to be larger in the OA (7%) as compared to the YA group (3%). This difference could make someone raise the question regarding whether or not the assumptions made or the fitting procedure adopted were adequate for our study. We are confident about the fitting procedure as we have used it and tested in previous studies [1, 9, 10]. However, this study represent one of the first attempt to model the postural control system in older adults; therefore, someone could argue whether the model assumptions made are adequate to study posture in older adults. For instance is the single link assumption valid for older adults as well as it is shown for younger adults [6]? Unfortunately we are not able to answer this question with the data presented here as we have not analyzed the hip and head motion data from the Fastrak system (by Polhemus). However, a paper by Allum and colleagues [61] have studied postural responses to platform perturbation finding similar results to ours. Their work only discussed increase in the stiffness and nothing about damping as in our case. In particular, they showed that older adults tended to exhibit increased trunk stiffness and that in response to a platform tilt they behaved more like an inverted pendulum rather than a two-link system. If the finding from that paper holds true in our case, then we could speculate that the inverted pendulum assumption is satisfied in our OA group. However, only analysis of the hip and head motion data can answer this question. The increase in MSE seen for the OA group could be due to changes that occur with age and that are not included or thought of in our model yet. Perhaps the structure of the controller or the lumped delay may need to be modified to account for some decay that people have found in muscle activation, conduction velocity in the nerves or changes in the CNS that are observed with age. Alternative model structure may be needed to be developed in the future to address other more specific questions and to improve goodness of the fits over the full range of ages of interest in which this model-based approach is adopted to answer some research questions about the postural control system.

In conclusion, this study has shown that aging has a strong impact in the processing delay and overall control parameters important for control of posture. Those changes, as also discussed in [10], suggest that older adults need to be more careful than young adults in situations where the support surface can move rapidly as the changes in the model parameters would suggest a more pronounced body sway response to occur in such a situations. Attention seemed to have had a lesser effect overall, but its effects were generally more pronounced in the older adults.

5.0 CONCLUSIONS AND FUTURE WORK

The work presented here applied model-based analysis to dual-task balance experiments to quantitatively investigate changes occurring in the postural control system due to age and cognitive demand (or "attention"). Findings resulting from this study may be used to develop an intervention approach to reduce the incidence of falls in older adults. Specifically, control parameters estimated from model fits to subject data could be used as metrics to be monitored in assessing the efficacy of a prescribed therapy or exercise regimen, to reduce the changes incurring with age or even induce a change in model parameters in older adults towards values that are more typical of young adults. Also, age-related changes of cognitive processing mechanisms can affect the adaptation required to compensate for changes in the environment and also the interaction with concurrent execution of multiple tasks. Understanding how postural control and attention interact in both young and older adults can potentially help to identify what changes occur with age in their interaction process and what could possibly be responsible to induce a fall-prone situation in older adults.

Both studies presented here, one in the frontal plane (chapter 3.0) and the other in the sagittal plane (chapter 4.0), presented results of frequency response measurements to balance perturbations during dual-task experiments, parameterized by model fits to the data. In both studies, larger gain curves, on average, were observed above 0.3 Hz with a slight peak around 0.5/0.7 Hz (AP/ML) seen in the older adults, which was not present in young adults. From a control systems standpoint, this peak constitutes a resonant oscillatory behavior of the system at that frequency. Accordingly, one could expect that older adults would sway more in response to "fast" perturbations (frequency content >= 0.5 Hz) than they would for "slow" perturbations (frequency < 0.5 Hz), compared to younger adults, and this sway would be characterized by larger initial oscillations and longer decay times.

Model fits to the frequency response data in both studies revealed a significant increase of active stiffness and damping values in older adults compared to young adults. This concurrent increase in active stiffness and damping is more stabilizing than an increase in stiffness alone (Chapter 3.0). However, older adults were still less stable than the younger results, based on model-based analysis of their mean response to an impulsive balance perturbation. Therefore, from the findings of the work presented in this dissertation, it is certainly advisable that older adults be more cautious where rapid floor movements can occur such as on a bus, train, or escalator.

An aspect that emerged from the work presented in this dissertation is that T_d was significantly larger in the older adults in the AP direction of sway (Chapter 4.0) while no significant difference was seen in the ML direction (Chapter 3.0). A possible difference may exist in the control of balance between ML and AP directions, as discussed by Goodworh and Peterka [81] and as shown in [61]. These studies showed that control of balance in the ML direction may be more complex and may limit the effective T_d with age, making the T_d of the YA and OA similar. However, as discussed in Chapter 4.0 , the amplitude of the perturbation differed from the two studies and as shown in [6, 9] T_d was found to be lower as the amplitude of the perturbation increased, which is also what we found. T_d was indeed lower for the ML for which PRTS-SS amplitude was 4-deg peak-to-peak than for the AP direction in which the PRTS-

SS amplitude was 2-deg peak-to-peak. If indeed the increase in the amplitude of the perturbation induced a reduction in the T_d for both age groups, it could be possible the T_d of the OA had to be close to that of the YA for the posture control system to be stable at such a higher amplitude of the perturbation. However, further experiments are needed to fully test this possibility and to better understand the mechanisms involved in the changes of the model parameters and the interaction with age.

This last point of discussion raises the question of whether the interaction between age, attention, and control of posture would change with the amplitude of the perturbation. Peterka and others [6, 9] have shown and quantified sensory re-weighting in response to scaling of the sensory perturbation. However, other model parameters were found to change, too, although to a lesser degree than the sensory weights. Would changes in the amplitude of the sensory perturbation induce different changes in the model parameters as function of age? We don't have an answer to this question at this point and future work is required to address such a question.

Another aspect that was evident from results presented in Chapter 4.0 is that older adult postural control characteristics changed with time, i.e. from one visit to the next. In particular, the PID parameters and passive damping B progressively increased with the visit number, and also T_d increased after the first visit and then stabilized. We don't exactly know the mechanisms involved in these changes but this finding could be used to further our understanding of learning or habituation mechanisms in older adults. It could be that older adults, being perhaps generally less active than younger adults, are not challenged enough in their daily activity, such that exposure to platform perturbations may have provided more pronounced challenges to balance that elicited greater changes in their postural control function with consecutive visits.

As discussed in Chapter 3.0 or [10], older adults are “stiffer” than young adults, but we also found that they had a concurrent increase in damping as well, which is desirable from a control perspective; however, they were still not as stable as young adults. Furthermore, the increase observed with age in the time delay of the postural control system is not desirable either as it has similar effect to reducing damping, which increases the detrimental effect of increased stiffness and therefore would make the system more unstable (as discussed in Chapter 4.0). We don’t have a definite explanation of why those parameters changed in the way seen in the data presented in Chapter 4.0 ; this may be worthy of further investigation to advance our understanding about whether older adults are more adaptable to sensory perturbations than young adults and if this is the case, would the answer to that question help us understand better how age affects control of balance and adaptation to repeated exposure to sensory inputs. This could help in the design of preventive activities or programs to reduce the effect of age that have a negative impact on balance. Poor balance is also associated with fear of falling and slowing the impact of aging on balance function could reduce the occurrence of fear of falling. Characterization of fear of falling might be interesting to include in the model to understand how this aspect affects control of posture in older adults.

The fact that some model parameters changed from visit 1 to visit 3 for the No-IP trials in the older adult group and not for the young adults might raise a question about whether the estimates of the model parameters are stable over time or there is an estimation problem in our methodology. Without running some more in depth analysis we can not answer this question with certainty. However, if there was an estimation problem, why would we find a consistent trend in the model parameters with visit number and only in the older adults? Also, if there was a model fitting or estimation problem, why wasn’t this trend seen in the IP conditions for the

OA? We don’t have definitive answers to these questions but it is worth addressing them with further analysis of the data.

If we assume that the estimation procedure worked well as assumed and tested in previous similar studies [1, 10], then, are the estimates obtained with the frequency-based approach used in this study (i.e. experimental TF and model parameters) stable and robust to exogenous changes in the environment? In other words, if nothing in the postural behavior of a subject changes over time, would we obtain different measures from trial to trial as long as the experimental conditions are the same? We could answer yes to this question as that is what we found in the YA group. However, for the OA we don’t know yet as the question is based on an assumption and if the assumption doesn’t hold true for the OA then we can’t positively answer to this question. This is an important question to be able to answer, as we could foresee this methodology as a great addition to the current clinical tests performed on a posturography platform. The work presented in this dissertation clearly show the potential benefits of this methodology as a way to characterize the aspect of control of posture that relates to control, stability and processing. It allowed us to make inference on the interaction of age, attention and control of posture. This methodology could be used to evaluate changes over time that relate to balance; model parameters can be quantified and potentially checked against norm values. To reach such a clinical application a clinical study should be performed on a larger scale of subjects, collecting data from more people and a broader age range so that norms could be tabulated.

In conclusion, this research has provided some new insights about the aging and attention in the control of balance, while at the same time raising new questions. Clearly, more work has to be done to further our understanding about the mechanisms involved in control of balance and the influence of aging and attention on balance function. A control systems model based approach coupled with well designed human experiments, as undertaken here and in other studies, has helped to improve the progress of knowledge in this field.

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