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Effect of age on detecting a loss of balance in a seated whole-body balancing task

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Abstract

Background. Most falls are attributed to a loss of balance without a quantitative definition of the term. It has been proposed that a loss of balance is detectable as an unusually large (anomalous) value of the system control error. The hypotheses were tested that age will not affect the detection of control error anomaly, or prediction of the associated compensatory response, in a challenging balancing task.

Methods. Twenty healthy older adults were asked to sit and balance a chair over its rear legs for as long as possible. The dominant foot's ground reaction force and the chair's sagittal-plane acceleration represented the system input and output, respectively. Control error was the difference between actual and expected acceleration output from a self-identified forward internal model of the system. A control error anomaly was detected once the error crossed a threshold set at three standard deviations (3-Sigma) above the mean of baseline data. Results from five trials were compared to published results in 20 healthy young adults.

Findings. A control error anomaly was successfully detected in 91% of 91 older adult trials, statistically similar to the 92% success rate obtained previously in young adults. A response was predicted in 57% of the 77 older adult trials with responses, significantly less than the 92% obtained in the young adult trials (age effect significant: $P < 0.005$).

Interpretation. The condition leading to uncontrolled backward acceleration of the chair was reliably detected in both groups. While the young waited to respond to this condition, older subjects responded prematurely.

Keywords: Loss of balance; Postural control; Aging; Internal model; Control error; Falls

1. Introduction

Unintentional falls are usually attributable to a 'loss of balance'. We have proposed that a loss of balance is required for the central nervous system to trigger a compensatory response and prevent the ensuing fall (Ahmed and Ashton-Miller, 2004). For standing postures the response usually involves a rapid change in body configuration, such as a compensatory step taken before the center of gravity passes outside the base of support (Maki and McIlroy, 1997). The present study addresses how age may affect the ability of the central nervous system to determine that a loss of balance has occurred.

Central nervous system control of posture has been modeled as a mechanical system with an input signal, a controller, a plant, and feedback of the output signal and system states (Kuo, 1995; van der Kooij et al., 1999). In this paper we ask not how the central nervous

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system controls and maintains postural stability, but how it detects a failure to do so and is thus forced to compensate and recover using an alternate control strategy. Loss of balance is therefore defined as a *loss of effective control* of balance, detectable, both internally by the central nervous system and externally, as a control error anomaly (CEA). In the event of CEA, a change in control strategy would be required to regain control over the system. A model-reference adaptive controller and failure-detection algorithm have been used to represent central nervous system decision-making and identify what triggered the change in strategy based on system input and output signals in healthy young adults (Ahmed and Ashton-Miller, 2004) (Fig. 1).

In this model the central nervous system sends the control input to a forward internal model of the plant, which calculates an expected output. The forward internal model is boot-strapped from initial steady-state data using system identification techniques. Control error, defined as the residual generated when the actual system output is compared to the predicted output, is calculated by a sub-component of the controller we call the 'CEA detector'. The CEA detector monitors the residuals and compares them to the maximum allowable limits using a failure-detection algorithm. A CEA is detected when an unusual event occurs as defined by the error exceeding a threshold three standard deviations (3σ[1]) beyond the mean baseline signal. This information is relayed to the central controller, triggering the execution of a control compensatory strategy. A loss of balance, defined as CEA, predicted a compensatory response in over 90% of 197 balancing trials performed by young adults (Ahmed and Ashton-Miller, 2004).

Based on our theory and corresponding model, advancing age could adversely affect the successful detection of a CEA in a number of ways. The error signal is partly formed from visual, vestibular or somatosensory sensory signals. However, with age, visual acuity, contrast sensitivity and peripheral resolution are reduced by 20%, 28%, and 6% per decade respectively, while vestibular hair cell densities, vestibular nerve fibers, and vibration and joint proprioception thresholds also degrade by at least 29%, 37%, and 100% respectively (Anderson and McDowell, 1997; Barrack et al., 1983; Bergstrom, 1973; Gilsing et al., 1995; Merchant et al., 2000; Rubin et al., 1997; Wiles et al., 1991). Increased sensory noise and elevated sensory thresholds can affect the detection of a CEA by decreasing the accuracy of the internal model, its predictions, and the control error calculated. Not only are the separate sensory systems adversely affected by normal aging, but their integration can also decline. Visuo-vestibular interaction in the linear vestibulo-ocular reflex has been shown to have prolonged latencies and reduced sensitivities with age (Tian et al., 2002). There is also a decline in the integration of visual, vestibular, and somatosensory information, as well as a delay in cognitive-motor responses (Mattay et al., 2002; Perrin et al., 1997; Teasdale et al., 1991). The detection of CEA is essentially a decision making process involving a choice reaction time, which is also known to increase with age (for example, Luchies et al., 1999). Thus impaired sensory integration and an increased choice reaction time may result in an inability to detect CEA in a timely manner and respond appropriately in order to successfully prevent a fall.

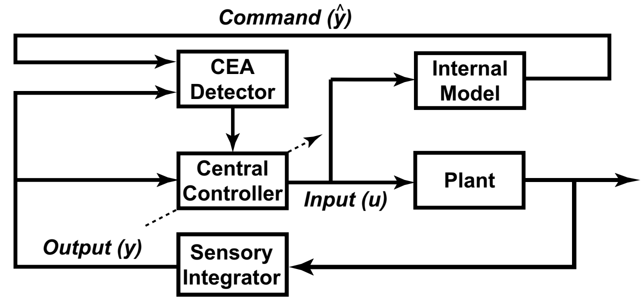

Fig. 1. Block diagram representation of the feedback control model of balance with CEA detail.

The objective of this paper is to measure the reliability of CEA detection and response prediction in older adults and compare them to previously published results from healthy young adults. We will test the primary null hypothesis that there would be no effect of age on the successful detection of CEA using a 3σ threshold criterion on the control error signal. A secondary (null) hypothesis was tested that age would not affect the reliability of CEA in predicting the occurrence of any compensatory response occurring at least 100 ms later.

2. Methods

2.1. Theory

To test these hypotheses we considered the situation in which a person is attempting to balance themselves over the two rear legs of a chair (Fig. 2). We have described our approach in an earlier paper (Ahmed and Ashton-Miller, 2004) and so will only provide a brief outline here. In this balancing task, the control input is restricted to the modulation of ground reaction forces (R_x, R_y) acting on the person's dominant foot. In terms of a feedback control system, the controller represents the central nervous system, which sends the reaction force system inputs to a plant (chair and body) and a forward internal model. The minimal relative motion between the subject and the chair allows them to be

[1] '3σ' is used here in preference to '3Σ' originally used in Ahmed and Ashton-Miller (2004) because σ is more commonly used to denote standard deviation.

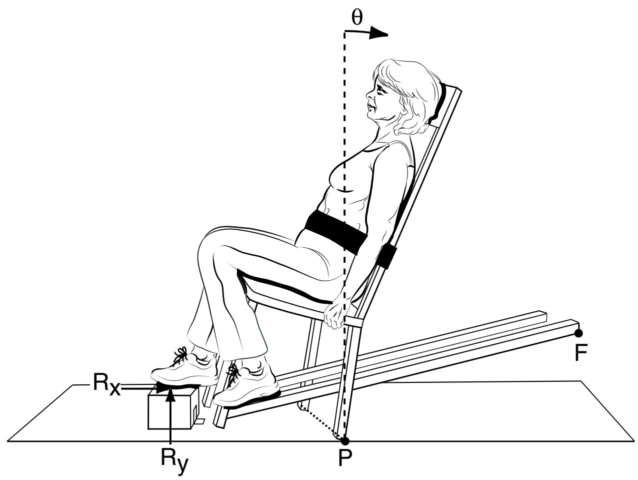

Fig. 2. Schematic of a subject starting her attempt to balance over line 'P' by varying the unipedal reaction forces R_x and R_y. A 20 cm-wide waist belt was used to strap the subject to the chair for safety.

modeled together as a simple inverted pendulum pivoting about the line between the rear feet of the chair, P, with the following equation of motion:

$$I\ddot{\theta}(t) = mgl\sin\theta(t) + T(t). \tag{1}$$

The input, $T(t)$, is the torque about the pivot point due to the resultant ground reaction force, the angle, $\theta(t)$, is the angle of the center of mass from the vertical and the output signal, $y(t)$, is the angular acceleration, $\ddot{\theta}(t)$. The system mass, moment of inertia, acceleration due to gravity, and distance from the pivot point to center of mass are respectively denoted by m, I, g and l. Both the input, T, and the output, $\ddot{\theta}(t)$, are available to the central nervous system through corollary discharge and vestibular afference, respectively. An advantage of this setup is that both these variables can be measured externally.

The command signal, $\hat{y}(t)$, is the resultant movement expected by the central nervous system due to the applied torque and is calculated using an internal model. The internal model is a parameterization of the system equation of motion in the form of a simple, first-order polynomial,

$$\hat{y}(t) = \hat{\ddot{\theta}}(t) = c_0 + c_1\tilde{T}(t). \tag{2}$$

The parameters c_0 and c_1 are identified using linear, least-squares optimization, and are regressed from a steady-state interval of 2 s at the start of the balancing task. The internal model represents an approximation of system dynamics about a local operating point that is moving with respect to time. Its movement is defined by the low-frequency trajectory of the system. Accordingly, the control input to that model, $\tilde{T}(t)$, is a linearization of the original input $T(t)$ about the local operating trajectory. An approximation of θ as a constant, due to the minimal changes in its value, is included in the parameter c_0. Once identified parametrically, the internal model can be used without modification to calculate the command signal (expected output) given the measured input for the duration of the task. The ensuing analysis is based upon the calculation and monitoring of the control error signal, $e(t)$, the difference between the output and command signals:

$$e(t) = y(t) - \hat{y}(t). \tag{3}$$

A CEA is detected when the control error crosses a threshold set at three standard deviations (3σ) above the mean of the baseline performance data (Fig. 3). Data from a fixed 2-s window, a, at the beginning of the trial is used to identify the model parameters. Baseline performance data is obtained from b, a 2-s forward moving trailing window, which lags the current time instant, t, by 100 ms. The moving threshold, $e_{\text{thresh}}(t)$, is calculated

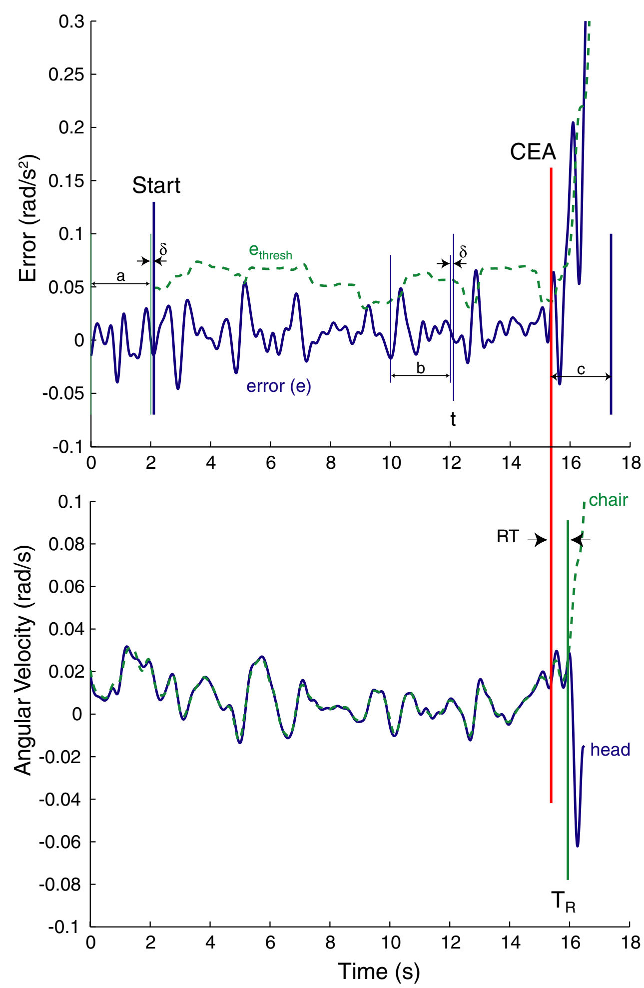

Fig. 3. (Top) Sample time history of the control error signal (solid) along with the moving threshold (dashed). Timing marks illustrate the salient variables described in the text. Constraints on the velocity and acceleration vestibular thresholds prevented CEA detection at the earlier threshold crossings. (Bottom) Corresponding velocity of head (solid) and chair (dashed). A response is detected once the head moves off the chair (T_R).

from the mean, μ_b, and standard deviation, σ_b, obtained from window b (Ahmed and Ashton-Miller, 2004). Calculation of the moving threshold commences at 'Start', initially using the data in a as baseline data with a 100 ms delay (δ) to allow for neural processing (van der Kooij et al., 1999). The moving 2-s window, b, continues to move through the trial, calculating the threshold 100 ms later. A CEA is detected when e crosses e_{thresh}.

In addition to the threshold criterion on the error signal, system angular acceleration and velocity at time t must also exceed published vestibular sensory thresholds: 0.00087 rad/s^2 and 0.017 rad/s, respectively (Gusev and Semenov, 1992; Nashner, 1971).

2.2. Experimental methods

Twenty older adult volunteers were tested (10 males and 10 females: mean [SD]: age: 73.9 [3.78] years and 76.67 [6.36] years; height: 175.15 [5.13] cm and 161.9 [5.54] cm; mass: 75.5 [4.86] kg and 61.37 [5.74] kg). The protocol was approved by the institutional Internal Review Board. All subjects provided written, informed consent.

Subjects were seated in a four-legged, high-backed chair in a sagitally symmetric posture. They were asked to balance themselves and the chair over its rear legs, P (Fig. 2), by pushing down with their dominant foot in order to push themselves slowly backwards until they were perfectly balanced for as long as possible over the rear legs with no foot-ground contact. A two-axis force-transducer was placed underneath their dominant foot. Subjects were directed to always maintain contact between their head and its support. In order to prevent a backward fall and potential injury, the horizontal rungs of the chair were extended backwards by 1 m, F (Fig. 2). Each subject performed five trials with eyes open. No practice trials were allowed.

The CEA algorithm was applied to the older adult data, obtaining both its reliability in successfully detecting task failure and predicting a compensatory response. These results were compared with previously published data from 20 young adults using the same CEA algorithm (10 males and 10 females: mean [SD]: age: 21.4 [3.17] years and 20.9 [5.78] years; height: 176.6 [4.84] cm and 167.55 [5.02] cm; mass: 73.06 [6.33] kg and 63.33 [9.08] kg) (Ahmed and Ashton-Miller, 2004). The published results reported CEA algorithm reliability based on data from 10 trials performed by the subject. However, analysis of the first five trials provided almost exactly the same results with no significant differences. Thus, in the interest of minimizing the duration of the experiment and subject fatigue, the older adults were only asked to perform five trials. To maintain similar sample sizes, older adult results were only compared with results from the first five trials of the published young adult data. Accordingly, the young adult success rates and response times in this paper are not exactly the same as those in Ahmed and Ashton-Miller (2004). Moreover, further analyses of their data are reported for the first time in the present work.

2.3. Data acquisition

An Optotrak 3020 motion analysis system was used to monitor head and chair location and orientation in three-dimensional space to the nearest 0.1 mm. A total of five markers were used: three markers were mounted on the head; one marker was attached to the bottom of the chair's rear leg; and one to the headrest.

Kinematic and force data were recorded at 100 Hz and low-pass filtered with a cutoff frequency of 3 Hz using a fourth-order Butterworth filter. Kinematic data were differentiated to obtain velocity and acceleration data. All filtering routines were employed forward and backward to minimize phase shift artifact.

2.4. Data analysis

CEA was detected using the algorithm and 3σ threshold described earlier, and designated T_{CEA}. For successful detection, points 'F' on the chair must strike the ground, representing task failure, within c, a 2-s post-CEA interval (Fig. 3).

The occurrence of a natural righting response, a rapid flexion of the head (relative to the chair), was defined as a compensatory response, and external evidence of CEA perception. Response time (RT) was defined as the latency of this response and could occur no earlier than 100 ms after CEA. The time of response initiation, T_R, was determined by monitoring, post hoc, the difference between the angular velocity of the chair and the subject's head (Fig. 3) (Ahmed and Ashton-Miller, 2004). Trials with excessive head movement were removed from the analysis of both hypotheses. Trials where the subjects exhibited no response were removed from the test of the secondary hypothesis.

We also compared the performance of the CEA algorithm in predicting compensatory responses to the performance of alternative methods based solely on body kinematics; like that of Wu (2000), these methods do not take into consideration the control input or control error. The alternative methods were implemented using an algorithm that detected a loss of balance when the angular velocity or acceleration of the system exceeded a pre-defined fixed threshold.

A chi-squared analysis, with a P-value less than 0.01 being considered statistically significant, was used to test the primary and secondary hypotheses. It was also used to compare the reliability of the CEA algorithm with the reliability of the velocity and acceleration algorithms for each hypothesis and age group.

In addition to the analyses underlying the tests of the primary and secondary hypotheses, the following output variables were calculated for each trial and tested for age effects. The mean and standard deviation of the angular acceleration, angular velocity, and control error signals and the mean torque input from the start of the trial to T_R were calculated. We also calculated the angular acceleration, angular velocity, and control error at a time instant 100 ms before the onset of any response. The head movement relative to the chair was also obtained as the difference between the head and chair angular velocity from the start of each trial to T_R. Trial duration, accuracy of the identified internal model (R^2), and T_F were also compared. Multiple two-tailed *t*-test comparisons were used to investigate age effects on these variables using a Bonferroni corrected *P*-value of less than 0.0036 as being statistically significant.

3. Results

Since no gender differences were found ($P > 0.05$), the results were pooled into two groups: young and older adults. The primary hypothesis tested for age effects in the successful detection of CEA. In order to be detected successfully, CEA must be followed by task failure by at most 2 s. The primary hypothesis was supported in that CEA was successfully detected in 91.2% of the 91 older adult trials, compared to the previously obtained 91.9% of the 99 young adult trials (no significant age difference: χ^2: $P > 0.995$). All failures were false positives in that CEA was prematurely detected more than 2 s before the chair hit the ground.

The secondary hypothesis tested for age effects in the reliability of response prediction by CEA. Only trials with successful CEA detection and a compensatory response were included in the analysis. The secondary hypothesis was rejected in that a compensatory response was successfully predicted using the 3σ algorithm in only 44 of 77 older adult trials with responses (57.14%), significantly less than the 76 successful predictions out of 83 young adult trials with responses (91.57%) (age effect significant: χ^2: $P < 0.005$). The trials with successfully predicted responses will hereafter be referred to as "H2S" trials. All failures, which we have dubbed "H2F" trials, were false negatives: the algorithm detected CEA after the response was initiated. Applying a lower threshold (2σ) to the H2F trials did successfully predict responses in a further 12% of older adult trials and 3% of young adult trials. But there remained a significant age effect with these added successes. The secondary hypothesis was not supported in two older adult subjects in particular (none of the five trials were predicted successfully). Removing these subjects from the analysis, however, did not account for the significant age effect, and a lowered threshold only predicted three of the 10 responses. The remaining H2F trials occurred across 13 different subjects.

There was no significant age effect in RT for the H2S trials ($P = 0.357$). Older adults had a mean RT of 0.406 s ± 0.162 s, similar to the young adult mean RT of 0.458 ± 0.183 s.

A sensitivity analysis demonstrated that the optimal threshold level in the older adults was 2.25σ; lower levels resulted in more false positives, while higher levels resulted in delayed CEA detection times (Fig. 4). This is lower than the optimal threshold level of 3σ obtained for the young adults. Even using the older adult success rates obtained with the optimal threshold level of 2.25σ, there remained, however, a significant age difference when compared with the young adult 3σ results in the tests of both hypotheses ($P < 0.01$).

The velocity- and acceleration-based algorithms in the older adults used a fixed threshold level of 0.04 rad/s and 0.16 rad/s^2, respectively for CEA detection. Young adult velocity- and acceleration-based algorithms used fixed threshold levels of 0.04 rad/s and 0.11 rad/s^2, respectively. These values were determined empirically, and were the optimum values for each algorithm. For the older adults, CEA performed better in the test of primary hypothesis than did the velocity or acceleration thresholds, while there was no significant difference between all three algorithms in the test of the secondary hypothesis (Fig. 5, $P > 0.04$). In the young adults there was no significant difference in the outcome of testing the primary hypothesis using the CEA and velocity algorithm ($P = 0.271$). For the secondary hypothesis, however, the CEA algorithm was significantly more reliable than the velocity and acceleration algorithms in young adults.

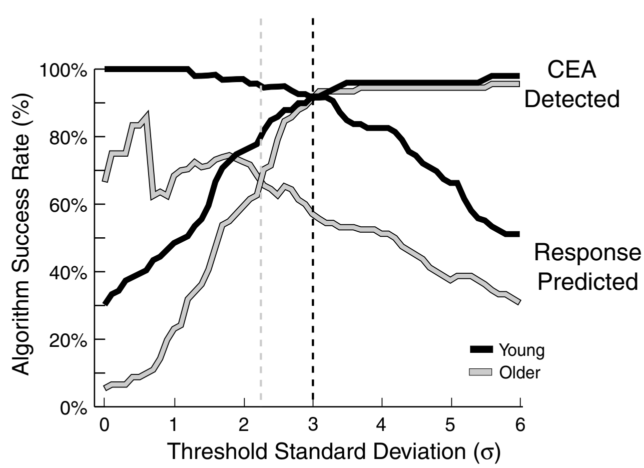

Fig. 4. Results of algorithm sensitivity to changes in σ threshold level. As the threshold increases success of CEA detection increases and success of response prediction decreases in both young (black) and older (gray) adults. Age-specific optimal threshold levels are determined from the intersection of the color-matched lines (dashed vertical lines).

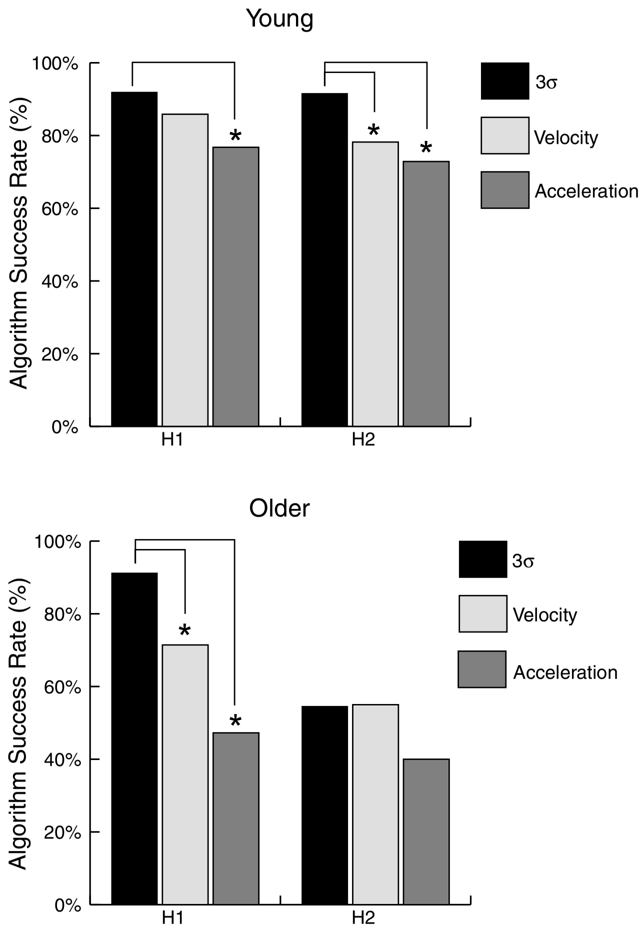

Fig. 5. A comparison of algorithm performance using CEA with a 3σ criterion and two examples of a fixed kinematic threshold: Velocity and Acceleration (*$P < 0.01$). Results for both primary (H1) and secondary (H2) hypotheses are presented.

Upon comparison of all subject trials (ALL), excluding those disqualified because of head movement, older adults had significantly greater torque variability than did the young adults ($P < 0.0036$, Table 1).

There were few differences between H2S and H2F trials within each age group. The most notable difference was the significantly lower head angular velocity in older adult H2F trials, 100 ms prior to their response, than in older adult H2S trials.

4. Discussion

The results suggest that the CEA algorithm can be used by an external observer to reliably detect a CEA in both young and older adults performing this seated balancing task. There were, however, age-related differences in a healthy subject's ability to internally detect a CEA because the 3σ threshold did not predict a response in older adults as accurately as in young adults. Failures were due to the older adults responding earlier than predicted.

Table 1
Comparison of mean [standard deviation] task variables by age group (young/older) and trial type (All/H2S/H2F)

Age	Young	Older	P	Young	Young	P	Older	Older	P
Trial type	All	All		H2S	H2F		H2S	H2F	
Outcome variable									
Angular velocity (rad/s)	0.016 [0.008]	0.009 [0.004]	<0.0001*	0.017 [0.008]	0.012 [0.007]	0.253	0.008 [0.004]	0.009 [0.007]	0.469
Angular acceleration (rad/s^2)	−0.001 [0.003]	0.000 [0.002]	0.284	−0.001 [0.004]	−0.002 [0.003]	0.768	0.000 [0.002]	−0.001 [0.004]	0.403
Control error (rad/s^2)	0.003 [0.009]	−0.004 [0.009]	0.0140	0.003 [0.009]	0.006 [0.020]	0.630	−0.005 [0.009]	−0.003 [0.010]	0.706
Angular velocity variability (rad/s)	0.012 [0.003]	0.017 [0.007]	0.0147	0.012 [0.004]	0.016 [0.008]	0.210	0.018 [0.010]	0.018 [0.006]	0.967
Angular acceleration variability (rad/s^2)	0.043 [0.013]	0.080 [0.032]	<0.0001*	0.042 [0.013]	0.055 [0.027]	0.154	0.085 [0.046]	0.082 [0.032]	0.875
Control error variability (rad/s^2)	0.025 [0.006]	0.068 [0.078]	0.019	0.025 [0.006]	0.038 [0.015]	0.006	0.084 [0.110]	0.053 [0.034]	0.303
Torque variability (N m)	3.674 [1.173]	7.586 [2.782]	<0.0001*	3.611 [1.182]	4.894 [3.487]	0.169	7.475 [3.203]	8.042 [3.173]	0.614
Angular velocity at T_R (rad/s)	0.080 [0.062]	0.048 [0.026]	0.040	0.097 [0.073]	0.027 [0.009]	0.045	0.064 [0.036]	0.030 [0.010]	0.001*
Angular acceleration at T_R (rad/s^2)	0.193 [0.169]	0.086 [0.129]	0.030	0.245 [0.218]	0.034 [0.039]	0.045	0.157 [0.156]	0.030 [0.072]	0.007
Control error at T_R (rad/s^2)	0.187 [0.158]	0.0798 [0.133]	0.026	0.232 [0.197]	0.025 [0.096]	0.037	0.136 [0.162]	0.031 [0.078]	0.029
Time to perform task (s)	13.223 [7.682]	19.256 [5.866]	0.008	12.835 [7.797]	17.332 [9.217]	0.276	19.482 [6.232]	17.82 [6.117]	0.448
Relative head movement (rad/s)	0.001 [0.001]	0.003 [0.003]	0.013	0.001 [0.001]	0.001 [0.000]	0.174	0.003 [0.004]	0.003 [0.003]	0.574
Internal model R^2	0.804 [0.106]	0.803 [0.173]	0.981	0.801 [0.096]	0.722 [0.212]	0.216	0.838 [0.119]	0.792 [0.210]	0.4316
T_F (s)	1.237 [0.415]	1.883 [1.589]	0.087	0.968 [0.144]	0.922 [0.175]	0.547	1.170 [0.164]	1.028 [0.236]	0.049

* Significant age difference ($P < 0.0036$).

Although the older adults had a lower optimal threshold than the young, optimal older adult algorithm performance was still significantly less reliable than the corresponding optimum for the young. Thus a lower threshold does not completely explain what appears to be a more cautious performance by the older adults.

It is noteworthy that the velocity- and acceleration-based detection algorithms did not perform better than the CEA algorithm in either age group or hypothesis. Although a velocity detection algorithm may have performed similarly in the young for the primary hypothesis, and the differences between older adults H2S and H2F trials were also discernible from the velocity characteristics, the CEA algorithm still provides a distinct advantage over simple kinematic (output) thresholds (Wu, 2000). Considering both system input and output, it affords a framework by which we can attempt to elucidate the physiological mechanisms behind the observed differences.

4.1. Why did the older adults respond earlier?

The age difference in the success of the CEA algorithm appears to reflect a change in central nervous system control of subjects' responses, not a change in how well the algorithm performs in detecting CEA. We can assume that the algorithm still worked well because regardless of whether a response followed CEA or not, it was followed by task failure by approximately the same delay. This is evident in the lack of a significant difference in the delay between CEA detection and the instant of task failure for H2S and H2F trials (T_F: Table 1). Thus the consistency in CEA detection success rates in H2S and H2F trials indicates that the algorithm was performing similarly in both trial types. This implies that the H2F trial results were due to an altered response strategy, not a limitation on the part of the CEA algorithm. In fact, the only significant difference between these trial types was in a response-related variable: angular velocity 100 ms prior to the response. This velocity was significantly greater in the older adult H2S trials compared to the H2F trials. Responses in H2F trials rarely occurred when the kinematics were significantly greater than the trial mean, whereas all responses in H2S trials occurred when the kinematics were significantly greater than the trial mean. The CEA algorithm was able to identify these trials despite the different behavior. The above evidence suggests a more cautious response strategy on the part of the elderly, but does not explain the reason for such behavior. In the following paragraphs we discuss possible explanations in the form of torque variability, sensory and motor noise, internal model identification and control error calculation, and cognitive and central processing delays.

4.1.1. Torque variability

Increased torque variability would theoretically lead to an inflated threshold level thereby reducing reliability of CEA detection. In agreement with the literature (vide infra), the older adults exhibited significantly greater torque variability than the young adults in both H2S and H2F trials. Additionally, all the older adult subjects had significantly greater variability than the young adult H2S subjects, but responses were still successfully predicted for over half the trials. Thus the increased variability does not appear to affect the algorithm's ability to predict the older adult responses.

4.1.2. Sensory sensitivity and noise

Incorporating age-related decreases in sensory system sensitivity into the algorithm would not improve its prediction of the older adult response. An increased threshold on the perception of sensory feedback would only lead to a delay in CEA detection, or even a failure to detect CEA at all. If this was the case, the older adults would have responded after CEA with greater response times than the young, quite different than the premature responses we observed.

Age-related sensory noise may corrupt the afferent signals used in identifying the internal model and calculating the command and error signals. Injecting gaussian noise of up to 20% of the angular acceleration and velocity feedback would result in the successful prediction of 16 of the 33 older adult H2F trials. Interestingly, the increased sensory noise would induce CEA detection failures that would belie a more cautious response, and result in greater response times in both young and older adults.

4.1.3. Motor noise

The fact that the elderly are known to have less accuracy in their upper and lower extremity motor output, especially at lower forces (Christou et al., 2003; Christou and Carlton, 2001; Laidlaw et al., 2000), might significantly affect the central nervous system perception of the control signal. Thus, in the H2F trials the older adults may not be responding to a different stimulus, but monitoring a different control error signal. Injecting noise, this time to the internal model input signal, should decrease the accuracy of the internal model identification, as well as the subsequent control error calculated. In fact, this resulted in the successful prediction of four of the 33 older adult H2F trials, while the primary hypothesis success rates were not significantly affected in either group. In the former case, the small changes that were observed also caused earlier CEA detection: more false positive and fewer false negative prediction failures.

4.1.4. Internal model identification and control error calculation

Inaccuracy in the internal model itself is also a possible source of error. Independent of the presence of sensory or motor noise, the older adults may identify the internal model in a different manner than the young. A sensitivity analysis revealed, however, that shortening the duration of the model identification window, (Fig. 3), increased the root-mean-square control error without significantly affecting CEA detection or response prediction in either group.

4.1.5. Cognitive and central processing delays

The difference between the young and older subjects does not appear to be a direct consequence of delayed cognitive and central integrative processes. Had this been the case, we should have seen greater response times in the older adults. Instead the older adults responded considerably earlier than expected.

One can speculate that the older adults may have chosen to respond to a lower threshold, a different detection signal, or a combination of signals, due to the age-related changes in sensorimotor, cognitive and central processing already noted. This elevated level of uncertainty could also lead to more cautious decision making. Indeed all early response trials (H2F) appeared to involve responses at lower velocities, although a lower absolute velocity threshold would not have predicted these responses. Some of these trials also involved responses at lower threshold values. This is consistent with previous reports of apparently more cautious elderly responses to experimentally applied postural perturbations. For example, the elderly step at lower perturbation magnitudes and more frequently (Brown et al., 1999; Jensen et al., 2001; Luchies et al., 1994; McIlroy and Maki, 1996).

The CEA hypothesis suggests an alternate explanation. The older adult responses may appear more cautious, but in reality they are still physiologically motivated. Increased sensory or motor noise may induce incorrect assumptions about system states and surrounding environment, thus leading to a false CEA detection and early response. Half of the early response trials would be predicted by increased sensory noise. Increased motor noise could also account for four such trials. The addition of sensory and motor noise together resulted in the correct prediction of 52% of the 33 early responses; this is considerably greater than the 27% of the early trials that a lowered threshold would have predicted.

4.2. Limitations

This study has several limitations. First, variability in subject motivation to balance as long as possible would tend to confound the results. The internal model was identified online at the start of the trial, and did not adapt either with time or as the trials progressed. Future analyses should investigate internal model adaptation and its effect on CEA detection rates. Finally, because we did not identify the control strategy, we could not isolate its possible effects on algorithm performance. The observed age effect was independent of changes in control strategy in the form of increased torque variability. However, an age-related increase in force variability would theoretically increase acceleration, velocity and control error variability, leading to reduced algorithm reliability due to an inflated CEA threshold.

The CEA hypothesis has certain positive attributes. The model is simple, and knowledge of center of mass location is not required to perform the analysis. This makes sense since there is no evidence that the central nervous system actually ever calculates whole body center of mass location. Moreover, the calculation of CEA does not require position feedback, absolute position cues, or the calculation of the boundaries of the base of support. It provides a unique method for analyzing the effects of both afferent and efferent system signals on the initiation of a compensatory response, and understanding the effect of aging on postural control. It has potential applications in the design of fall detectors and clinical tests for the elderly.

In summary, the CEA algorithm was able to detect a loss of balance in these healthy older adults as reliably as in the young in this simple, yet very challenging, balancing task. Predictions of a compensatory response were not as reliable as in the young, perhaps due to increased cautiousness and variability on the part of these older adults.

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References

Ahmed, A.A., Ashton-Miller, J.A., 2004. Is a "loss of balance" a control error signal anomaly? Evidence for three-sigma failure detection in young adults. Gait Posture 193, 252–262.

Anderson, R.S., McDowell, D.R., 1997. Peripheral resolution using stationary and flickering gratings: the effects of age. Curr. Eye Res. 1612, 1209–1214.

Barrack, R.L., Skinner, H.B., Cook, S.D., Haddad Jr., R.J., 1983. Effect of articular disease and total knee arthroplasty on knee joint-position sense. J. Neurophysiol. 503, 684–687.

Bergstrom, B., 1973. Morphology of the vestibular nerve. II. The number of myelinated vestibular nerve fibers in man at various ages. Acta Otolaryngol. 762, 173–179.

Brown, L.A., Shumway-Cook, A., Woollacott, M.H., 1999. Attentional demands and postural recovery: the effects of aging. J. Gerontol. A Biol. Sci. Med. Sci. 544, M165–M171.

Christou, E.A., Carlton, L.G., 2001. Old adults exhibit greater motor output variability than young adults only during rapid discrete isometric contractions. J. Gerontol. A Biol. Sci. Med. Sci. 5612, B524–B532.

Christou, E.A., Shinohara, M., Enoka, R.M., 2003. Fluctuations in acceleration during voluntary contractions lead to greater impairment of movement accuracy in old adults. J. Appl. Physiol. 951, 373–384.

Gilsing, M.G., Van den Bosch, C.G., Lee, S.G., Ashton-Miller, J.A., Alexander, N.B., Schultz, A.B., Ericson, W.A., 1995. Association of age with the threshold for detecting ankle inversion and eversion in upright stance. Age Ageing 241, 58–66.

Gusev, V., Semenov, L., 1992. A model for optimal processing of multisensory information in the system for maintaining body orientation in the human. Biol. Cybern. 665, 407–411.

Jensen, J.L., Brown, L.A., Woollacott, M.H., 2001. Compensatory stepping: the biomechanics of a preferred response among older adults. Exp. Aging Res. 274, 361–376.

Kuo, A.D., 1995. An optimal control model for analyzing human postural balance. IEEE Trans. Biomed. Eng. 421, 87–101.

Laidlaw, D.H., Bilodeau, M., Enoka, R.M., 2000. Steadiness is reduced and motor unit discharge is more variable in old adults. Muscle Nerve 234, 600–612.

Luchies, C.W., Alexander, N.B., Schultz, A.B., Ashton-Miller, J., 1994. Stepping responses of young and old adults to postural disturbances: kinematics. J. Am. Geriatr. Soc. 425, 506–512.

Luchies, C.W., Wallace, D., Pazdur, R., Young, S., DeYoung, A.J., 1999. Effects of age on balance assessment using voluntary and involuntary step tasks. J. Gerontol. A Biol. Sci. Med. Sci. 543, M140–M144.

Maki, B.E., McIlroy, W.E., 1997. The role of limb movements in maintaining upright stance: the change-in-support strategy. Phys. Ther. 775, 488–507.

Mattay, V.S., Fera, F., Tessitore, A., Hariri, A.R., Das, S., Callicott, J.H., Weinberger, D.R., 2002. Neurophysiological correlates of age-related changes in human motor function. Neurology 584, 630–635.

McIlroy, W.E., Maki, B.E., 1996. Age-related changes in compensatory stepping in response to unpredictable perturbations. J. Gerontol. A Biol. Sci. Med. Sci. 516, M289–M296.

Merchant, S.N., Velazquez-Villasenor, L., Tsuji, K., Glynn, R.J., Wall, C., Rauch, S.D., 2000. Temporal bone studies of the human peripheral vestibular system. Normative vestibular hair cell data. Ann. Otol. Rhinol. Laryngol. (Suppl. 1813), S3–S19.

Nashner, L.M., 1971. A model describing vestibular detection of body sway motion. Acta Otolaryngol. 726, 429–436.

Perrin, P.P., Jeandel, C., Perrin, C.A., Bene, M.C., 1997. Influence of visual control, conduction, and central integration on static and dynamic balance in healthy older adults. Gerontology 434, 223–231.

Rubin, G.S., West, S.K., Munoz, B., Bandeen-Roche, K., Zeger, S., Schein, O., Fried, L.P., 1997. A comprehensive assessment of visual impairment in a population of older Americans. The SEE Study. Salisbury Eye Evaluation Project. Invest. Ophthalmol. Vis. Sci. 383, 557–568.

Teasdale, N., Stelmach, G.E., Breunig, A., 1991. Postural sway characteristics of the elderly under normal and altered visual and support surface conditions. J. Gerontol. 466, B238–B244.

Tian, J.R., Crane, B.T., Wiest, G., Demer, J.L., 2002. Effect of aging on the human initial interaural linear vestibulo-ocular reflex. Exp. Brain Res. 1452, 142–149.

van der Kooij, H., Jacobs, R., Koopman, B., Grootenboer, H., 1999. A multisensory integration model of human stance control. Biol. Cybern. 805, 299–308.

Wiles, P.G., Pearce, S.M., Rice, P.J., Mitchell, J.M., 1991. Vibration perception threshold: influence of age, height, sex, and smoking, and calculation of accurate centile values. Diabet. Med. 82, 157–161.

Wu, G., 2000. Distinguishing fall activities from normal activities by velocity characteristics. J. Biomech. 33, 1497–1500.