Age-Related Changes in Rate and Magnitude of Ankle Torque Development: Implications for Balance Control

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Background. One of the key components of postural control is the motor system's ability to produce appropriate torques to counteract perturbations that may lead to a loss of balance. Evidence exists to show that there is an age-related decline in absolute strength and in the ability to rapidly produce torque. The relationship between age-related decreases in these voluntary torque production capabilities and the ability to rapidly produce torques in a reactive balance task has not been studied. Thus, the purpose of this study was to examine the magnitude and rate of torque production in younger and older adults under reactive balance conditions.

Methods. Older (OA) and younger (YA) adults received forward and backward surface translations of varying amplitudes and velocities. Maximum ankle muscle torque (maxMa) and rate of change of ankle muscle torque (Ma) following a perturbation were calculated.

Results. Two balance responses emerged: a no-step and a step response. With increasing perturbation difficulty, YA and OA used different responses. The no-step and step responses were examined for age-group differences in the force characteristics. No significant age-group differences were found for maxMa or rate of change of Ma within either no-step or step responses.

Conclusion. The results of this study suggest that neither the magnitude nor rate of ankle muscle torque production, as produced during the initial balance response in this set of reactive balance control tasks, determines the different balance responses seen in younger versus older adults.

FALLING among older adults is a major public health concern. The statistics related to falls in older adults are staggering: approximately one third of adults over 70 years of age fall in a given year, with one fourth of those falling resulting in fall-related injuries (1). In fact, falls are the leading cause of accidental death in older adults (1,2). In order to provide effective preventative strategies, we must understand the mechanisms that cause older adults to fall.

Numerous studies have identified factors contributing to falls. The research points to a multifactorial causation, including postural instability, decreased function of the sensory systems, disease processes, cognitive impairment, and overmedication (1,3-5). The risk for falls increases linearly with the number of risk factors, and the elimination of one or two risk factors can lead to a significant reduction in an individual’s risk for falling (1). Some of the risk factors associated with falling, such as cognitive impairment, disease, and decreased sensory functioning, may be less amenable to treatment. Postural instability is a major risk factor that may be amenable to training, and thus deserves careful attention (6).

One of the key components of postural control is the motor system’s ability to produce appropriate torques to counteract perturbations that may lead to a loss of balance. Strong evidence exists to show that absolute strength exhibits age-related declines (7-10). Both isometric and isokinetic strength testing reveal significantly reduced peak torque production about the knee and ankle in older versus younger adults, and lower extremity weakness is associated with decreased balance ability (8,10-12). It is not clear, however, whether a causal relationship exists between strength and balance.

It is logical, therefore, to ask whether an increase in strength will result in fewer falls. Results of a nationally based study Frailty and Injuries: Cooperative Studies of Intervention Techniques (FICSIT) demonstrated a significant reduction in the number of falls in those groups for which the intervention included some form of exercise as at least one component (6). However, those groups that received resistance exercise as the only intervention did not exhibit a significant reduction in falls. While strength is a necessary component of balance, strength training alone appears insufficient to reduce the incidence of falling.

The FICSIT study demonstrated that maximum voluntary torque production (i.e., strength) is not the critical factor in balance ability in the healthy older population. In many situations, balance requires automatic, not voluntary, responses. Loss of balance typically results from a sudden displacement of the center of mass or base of support (e.g., a trip or slip) to which the individual must react. Further, reactive balance tasks do not seem to require large magnitude torques (13,14). Gu and colleagues (13) found that larger ankle joint torques were produced under voluntary testing than during reactive balance testing. What then is the necessary component to maintain balance? The ability to produce adequate force in a task-appropriate period of time has been suggested as critical in maintaining upright posture in the face of a threat to balance (10).

Recent studies have shown that the ability to rapidly produce
torque declines in older adults. The rate of knee extensor torque production in an isometric task was significantly slowed in older versus younger adults (7,8). Thelen and associates (10) found that the rate of torque production at the ankle was likewise affected by age. In addition, Patla and coworkers (15) have shown a decreased rate of force production during voluntary stepping (a multijoint, complex task) in older adults. Because stepping is often recruited to avoid a fall, this decreased rate of force production may be instrumental in contributing to the increased rate of falling in older adults.

While the findings of an age-related decline in the rate of torque production seem robust, there has been no examination of the functional consequences of such a decline. The relationship between a decreased voluntary ability to rapidly generate torque and the ability to rapidly produce torque in a reactive balance task has not been studied. There is a need to study the contribution of rate of torque development in the functional task of maintaining upright balance to determine critical elements for fall prevention.

It has been suggested that ankle strength is preferentially affected by aging, and that this preferential decline plays an important role in the loss of balance control in older adults (12). Thus, we examined age-related changes in ankle muscle torque characteristics under reactive balance conditions. Based upon earlier work, we hypothesized that younger adults would produce larger ankle muscle torques than older adults under reactive conditions, and that the rate of development of ankle muscle torque would be greater for younger adults than for older adults.

METHODS

The sample.—Participants in this study included 19 young adults (YA) aged 21–35 years and 21 older adults (OA) between 65 and 85 years (see Table 1 for participant demographic data). The subjects were screened to exclude neurological, cardiovascular, or orthopedic disorders. All participants reported maintaining at least moderate activity levels [defined as more than 10 minutes per week of strenuous exercise; (16,17)]. They received 18 consecutive, pseudo-randomized, forward and backward perturbations of increasing demand. The amplitude of platform movement ranged in 10 cm/s increments from 10 cm/s to 80 cm/s for the forward trials and 10 cm/s to 60 cm/s for the backward trials.

Testing.—Perturbations were supplied by a single forceplate (61.5 cm × 30.5 cm) capable of moving forward or backward at a range of speeds and amplitudes. The platform was hydraulically driven to provide a balance disturbance. Four strain gauges, one in each corner of the forceplate, measured the vertical and horizontal forces under the feet. Analog data were collected by computer and sampled at 500 Hz. Onset of plate movement was determined based on the analog signal of plate position. The platform moved following a 1 s delay after the start of the trial. Participants were unaware of the 1 s time delay.

Position–time data were collected by a two-camera, WATSMART optoelectronic digitizing system (Northern Digital Inc., Waterloo, Canada) at 100 Hz. Data were collected for 5 s for each trial. Infrared light emitting diodes (IREDS) were placed over the following joint landmarks on the right side of the body: the fifth metatarsal head, midpoint of the calcaneus, the lateral malleolus, lateral femoral condyle, greater trochanter, midpoint of the acromion process, the temporal mandibular joint, a point 3–4 cm directly above the previous IRED, and on the forehead.

Electromyographic (EMG) data were collected to identify onset of muscle activity. Bipolar surface electrodes (Delsys Inc., Boston, MA) were placed on the skin over the right tibialis anterior (TA) and medial gastrocnemius (MG) and sampled at 500 Hz. EMG data were rectified, but not filtered for analysis. A customized interactive computer program was used to determine muscle onset latencies occurring within 50–250 ms of onset of platform movement. Criteria for determining muscle onset latencies were that EMG activity was greater than the baseline mean (determined prior to plate movement) plus three standard deviations, and that half of the EMG sample points in the burst remained above this level for at least 40 ms. Preliminary analyses (repeated measures analysis of variance [RM ANOVAs] for the forward and backward directions) revealed no age-group differences in EMG onsets ($p > .05$).

Participants stood on the force platform in a relaxed manner, looking straight ahead, with arms placed across the chest. Participants were instructed to maintain their balance without taking a step, if possible, and were reminded of the task goal at regular intervals. Participants first performed five voluntary sway trials (which were not used in this analysis), and then received 18 consecutive, pseudo-randomized, forward and backward perturbations of increasing demand. The amplitude of platform movements for the backward and forward trials ranged from 5 cm to 15 cm in 5 cm increments. The velocity of platform movement ranged in 10 cm/s increments from 10 cm/s to 60 cm/s for the forward trials and 10 cm/s to 80 cm/s for the backward trials.

From the 18 perturbation trials, 3 forward and 3 backward perturbation trials were selected for analysis. The selected forward perturbation parameters were 10 cm displacements at 10, 20, and 40 cm/s. The selected backward perturbation parameters were 10 cm displacements at 10, 40, and 80 cm/s. This selection of perturbations was based on our goal of creating an increasing scale of balance disturbances in both the forward and backward directions that would elicit a range of behavioral responses.

Table 1. Means (and Standard Deviations) of Participant Demographics for Age, Height, and Mass

<table>
<thead>
<tr>
<th>Demographics</th>
<th>Younger Adults</th>
<th>Older Adults</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>25.7 (4.3)</td>
<td>72.4 (5.8)</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>165.8 (6.0)</td>
<td>136.7 (6.4)</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>65.7 (10.2)</td>
<td>61.1 (9.6)</td>
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Dependent variables.—The muscle torque at the ankle (Ma) was determined using the mathematical technique of inverse dynamics based on a four-segment biomechanical model (feet, lower legs, thighs, and head-arm-trunk). Ankle muscle torque was calculated using Newtonian equations of motion using the collected ground reaction forces from the platform, individual anthropometrics, and kinematic data (19,20). The Ma reported...
is the combined torque for both ankles. Ma measured during testing was normalized with respect to the torque value just prior to onset of plate movement. Normalization was performed by subtracting the torque value just prior to perturbation onset from subsequent torque values. During quiet stance, there is a net plantarflexor torque counteracting the tendency of gravity to create a forward rotation of the body. By normalizing with respect to this baseline torque, Ma achieved during platform testing represents the magnitude of torque required to maintain upright posture in excess of that used in quiet stance. Negative values represent a change toward a dorsiflexor muscle torque, and positive values represent a change toward an increasing plantarflexor muscle torque.

Maximum ankle muscle torque (maxMa) was defined as the maximum plantarflexor or dorsiflexor torque following onset of plate movement. If a step occurred during the trial, the maximum torque achieved prior to step initiation was chosen. Step initiation was defined as the onset of a medial or lateral deviation of the center of pressure (COP; derived from forceplate data) just prior to the step (21). Onset of step initiation was defined as the beginning of the final COP trajectory shift toward the stance limb.

The rate of change of ankle muscle torque was defined as the slope of Ma calculated for two, 60 ms bins. The difference between the first and last torque values over the 60 ms bin was calculated. Muscle onset latencies were used to differentiate the two bins into passive and active phases of torque development. The passive phase of rate of change of Ma was calculated from onset of plate movement for 60 ms. The active phase of rate of change of Ma was calculated for the first 60 ms following onset of agonist muscle activity (e.g., medial gastrocnemius in a backward trial; Figure 1). The active and passive phases were examined separately because of potentially different age-related mechanisms acting on each. The passive phase may be affected by the age-related increase in muscle tissue stiffness, and the active phase is potentially affected by age-related changes in strength (7-10, 22). If a step was initiated during the timeframe of either bin, the slope was not calculated.

Data analysis.—In this study the questions of interest were the age-related changes in magnitude and rate of Ma production following increasing balance perturbations. Thus, we first employed six (3 conditions for each of 2 directions), Fisher’s Exact Chi-Square tests (p < .05) to examine age-group differences in rate of success with increasing balance demand. Successful completion of the task (per instructions) required that a nonstepping strategy be used, although two balance responses emerged: a no-step and a step response.

Under the slowest forward and backward conditions, all participants employed a nonstepping strategy, so age-group differences were examined using two (forward at 10 cm/s and backward at 10 cm/s), one-way MANOVAs (p < .05). To investigate whether response (i.e., step vs no-step) and/or age-group differences existed on the torque characteristics, four, 2 × 2 (Response × Group) MANOVAs were conducted (p < .05). The dependent variables included the ankle muscle torque characteristics: maxMa, passive rate of change of Ma, and active rate of change of Ma. Significant multivariate findings were followed up with appropriate univariate statistics (p < .05). Three, 2 × 2 (Condition × Group) RM ANOVAs (p < .05) were used to examine the effect of increasing backward perturbation at 10 cm/s and 40 cm/s on each dependent variable. As reported in Results, significant differences were found between the step and no-step torque characteristics in the forward conditions. Thus, for the forward condition we could not examine the condition effect without confounding the response effect.

Results
Older adults were less successful than younger adults in performing the balance task. Under the forward at 10 cm/s and backward at 10 cm/s conditions, all participants were successful (Figure 2). Under the forward at 20 cm/s condition, 42% of the YA were able to perform the task without stepping, whereas only 29% of the OA were able to do so. Under the forward at 40 cm/s condition, 16% of the YA were successful, and only 5% of the OA were successful. Age-group differences were significant (p = .012) only for the backward at 80 cm/s trials, where 89% of the YA were successful, but only 24% of the OA were successful.

There were significant differences in the dependent variables (maxMa, passive rate of change of Ma, and active rate of change of Ma) between the step and no-step responses for the forward at 20 cm/s [Wilks' lambda = .659, F(3,25) = 9.806, p < .001] and forward at 40 cm/s [Wilks’ lambda = .569, F(3,19) = 4.791, p = .012] conditions only. Follow-up univariate testing revealed that under the forward condition at 20 cm/s, maxMa [F(1,27) = 5.280, p = .03] and active rate of change of Ma [F(1,27) = 28.083, p < .001] contributed significantly to the response difference. In each case, the variables were larger for the no-step response (Figure 3). Under the forward condition at 40 cm/s, maxMa [F(1,21) = 20.460, p < .001] contributed significantly to the response difference. MaxMa was greater with a no-step response (Figure 3).

No significant age-group main effects were found for the Ma characteristics across all conditions (p > .05), except for forward at 40 cm/s [Wilks’ lambda = .667, F(3,19) = 3.160, p = .049; Figure 4]. Correlations among the variables (maxMa, passive rate of change of Ma, and active rate of change of Ma) ranged from -.60 to .51. Closer examination of the medium forward condition revealed that only one OA performed the task without stepping, whereas three YA performed the task without stepping. Re-analysis of this condition with that older participant removed resulted in no significant age-group differences (p > .05). Further analysis of this condition is based on the data without this older adult. There were no significant Age Group × Response interactions (p > .05).

RM ANOVAs revealed a significant condition effect for the dependent variables of maxMa and passive and active rates of change of Ma (F = 69.579; F = 79.218; F = 119.631; p’s < .001, respectively) with increasing backward perturbation size. Follow-up pairwise comparisons revealed a significant increase in magnitude and rate (both passive and active phases) of torque production across the three backward perturbations (Figure 5). The significance level for all comparisons was p < .001, except for maxMa backward at 40 cm/s and backward at 80 cm/s, which were significantly different at p = .001. There were no significant Condition × Group interactions for any of the dependent measures.
Figure 1. A: An exemplar ankle muscle torque (Ma) time series following a backward perturbation (perturbation onset delineated by the vertical dashed line). The passive and active phases of torque development are marked. The rate of change of Ma was calculated independently for both the passive and active phases. The passive phase was the first 60 ms after perturbation onset. The active phase was 60 ms following agonist muscle activation. B: Medial gastrocnemius EMG associated with Ma in the top graph. Plate onset is marked by the vertical dashed line, and EMG onset is marked by the solid vertical line. Plate position is delineated with the solid line.
Figure 2. Percent success (i.e., no-step response) by age group and perturbation condition. The perturbations included forward conditions at 10, 20, and 40 cm/s, and backward conditions at 10, 40, and 80 cm/s. YA = younger adult; OA = older adult. *Fisher’s Exact Test = .012.

Figure 3. Step versus no-step comparison for (top) maximum ankle muscle torque (maxMa); (middle) rate of change of ankle muscle torque (Ma), passive phase; and (bottom) rate of change of Ma, active phase. Numbers associated with each bar represent number of participants included in the analysis. For each dependent variable, perturbations included forward conditions at 10, 20, and 40 cm/s, and backward conditions at 10, 40, and 80 cm/s. Positive torque values are plantarflexor, and negative torque values are dorsiflexor.

Figure 4. Age-group comparison for (top) maximum ankle muscle torque (maxMa); (middle) rate of change of ankle muscle torque (Ma), passive phase; and (bottom) rate of change of Ma, active phase. Numbers associated with each bar represent number of participants included in the analysis. For each dependent variable, perturbations included forward conditions at 10, 20, and 40 cm/s, and backward conditions at 10, 40, and 80 cm/s. Positive torque values are plantarflexor, and negative torque values are dorsiflexor. YA = younger adult; OA = older adult.

DISCUSSION

Previously it has been shown that older adults produce less absolute torque than younger adults under voluntary conditions, and that older adults’ ability to rapidly produce torque voluntarily is also impaired (8,10). Based upon this earlier work, we hypothesized that younger adults would produce larger ankle muscle torques than older adults under reactive conditions. The data failed to support this hypothesis. Older adults and younger adults produced equivalent ankle muscle torque magnitudes in all three conditions. In addition, both older and younger adults responded to the progressively demanding backward perturbations by scaling up the ankle muscle torques. This scaling of muscle torques in response to increasing perturbation size is consistent with previous reports of the scaling response seen in younger adults (23). The lack of age-group differences in maxMa in both stepping and nonstepping responses and the scaling effect seen suggest that the mechanical perturbation was matched by a task-appropriate torque response by both older and younger adults.
that the inability of older adults to rapidly produce muscular forces may contribute to the increased incidence of falls (10). However, this hypothesis has been made based on results from voluntary strength testing and not under reactive conditions. The results of this study point out the inherent problem of extrapolating results from voluntary conditions to results under reactive conditions.

An interesting outcome from this study is that older adults tend to use a stepping strategy more frequently than younger adults do. We found some differences in the torque characteristics between a stepping and nonstepping strategy. The nonstepping strategy under the forward conditions at 20 cm/s and 40 cm/s elicited larger magnitudes of Ma and greater slopes for the active phase of rate of change of Ma than the stepping strategy. Despite these response differences, we did not find age-related differences in the execution of each response.

This preference for stepping in older adults is corroborated by several other studies using perturbation paradigms (24,25). Researchers have begun to examine the characteristics of the stepping response. McIlroy and Maki (24) found that the spatial-temporal characteristics of the first step of the balance response were very similar between younger and older adults. Differences begin to appear, behaviorally at least, following the initial response: older adults take multiple steps while younger adults take a single step to recover balance (24,26).

Further examination of the stepping responses is needed to clarify age-related differences. The results of this study suggest that the mechanism leading to different behavioral responses does not appear to be either the magnitude or rate of Ma production during the initial balance response. Based on the literature, we expected to see the most pronounced age-related differences for the ankle muscle torques. First, the balance response under reactive conditions is initiated by activation of the ankle musculature. Second, the ankle musculature has been shown to undergo preferential age-related strength declines (12). Our results lead us to look for alternative hypotheses.

We suggest that the older adults in our study are stepping because of other factors and not because they cannot meet the task demands in terms of distal joint muscle torque production. One alternative hypothesis is that the execution of different and age-related behavioral responses is because of torque production differences in the control of the proximal leg or trunk. Thus, the logical next step is to examine muscle torque production characteristics (magnitude and temporal characteristics) about the knee and hip. A second alternative hypothesis is that older adults are stepping for reasons unrelated to strength [e.g., a fear of falling that biases response selection; (1)]. In such a case, the reactive balance paradigm employed in this study cannot provide the definitive answer. The question that needs to be answered is, What are the physiological limits of muscle under reactive testing conditions? A paradigm in which the potential for falling was removed would get us closer to the answer, but such a paradigm is admittedly artificial.

In conclusion, the results of this study suggest that the force production characteristics of magnitude and rate of Ma development are not the limiting factors for these older adults in this set of reactive balance tasks. This is consistent with previous work that suggests that reactive standing balance control does not require the production of a maximal muscular effort (13). This study is the first documentation that the rate of ankle muscle torque development is not the limiting factor that leads to

![Figure 5](http://biomedgerontology.oxfordjournals.org/)

**Figure 5.** Condition effect across backward conditions at 10 cm/s, 40 cm/s, and 80 cm/s for (top) maximum ankle muscle torque (maxMa); (middle) rate of change of ankle muscle torque (Ma), passive phase; and (bottom) rate of change of Ma, active phase. Positive torque values are plantarflexor. Numbers associated with each dependent measure represent number of participants included in the analysis. For each dependent measure, there was a significant difference across the three conditions (p < .001).

We also predicted that the rate of development of Ma would be greater for the younger adults than for the older adults. The results did not support this hypothesis. Again, there were no significant group differences in the rate of Ma development during the passive or active phases.

In the range of perturbations experienced, the data suggest that both younger and older adults construct the initial balance response similarly: They use equivalent torque magnitudes and rates of torque development, and also scale the magnitude and rate of force development to increasing backward balance threats. These are unexpected results, as it has been suggested that the inability of older adults to rapidly produce muscular
the age-related differences in execution of stepping versus non-stepping balance recovery.

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