AGE RELATED CHANGES IN POSTURAL MUSCLE RESPONSES WITH INCREASING PERTURBATIONS TO THE UPPER BACK

Luis Rosado, Christopher J. Hasson, Richard E.A. Van Emmerik, and Graham E. Caldwell
University of Massachusetts Amherst, MA, USA, lrosado@kin.umass.edu

INTRODUCTION

Older adults display reduced postural stability associated with age-related neuromuscular deficits (Woollacott et al., 1986). If their base of support is suddenly moved backwards, older adults take longer to activate muscles needed to stop their forward rotation (Peterka & Black, 1990). Magnifying the support platform perturbation velocity alters muscular onset latencies and activity magnitudes in older subjects (Lin & Woollacott, 2002).

An external force perturbation applied to the upper back will also induce sudden forward rotation. In these and other perturbations, rapid balance recovery relies on various sensory inputs including muscle stretch receptors and the vestibular system. However, the force perturbations to the upper back likely cause greater head accelerations and neural responses from sensory mechanisms in the cranium. Because balance recovery responses are sensitive to the dynamics of the perturbation (Brown et al., 2001), external force perturbations might produce different age-related responses than platform motions.

Therefore, our purpose was to examine the relation between perturbation magnitude and plantarflexor muscle responses in young and old adults subjected to a series of increasing force perturbations to the upper body.

METHODS AND PROCEDURES

Healthy young [Y, n=9; 27±3 yrs, 1.73±0.11 m, 71±15 kg] and older [O, n=9; 71±5 yrs, 1.71±0.10 m, 82±16 kg] subjects stood on a force plate while wearing a safety harness anchored to the ceiling. Subjects were strapped to a backboard designed to constrain motion to the sagittal plane about the ankle joints. During quiet stance, a weighted pendulum was released to impact the subject on the upper back. Subjects were instructed to resist the subsequent forward sway and re-establish upright stance quickly without stepping. The perturbation magnitude was controlled by the pendulum release angle, which was increased sequentially until the subject could not recover without stepping. The perturbation level was quantified as the instantaneous pendulum angular velocity upon impact, computed from pendulum motion tracked at 200 Hz with a Qualisys camera system.

A force transducer in the pendulum striker was sampled at 1000 Hz to measure the perturbation impact. Activity levels for the gastrocnemius (GA) and soleus (SO) muscles were measured with electromyography (EMG) sampled at 1000 Hz. The EMG data were full wave rectified and low pass filtered at 50 Hz. EMG onsets were defined manually with the aid of a nominal threshold three standard deviations above baseline activity. Muscle onset latencies were calculated as the time from perturbation impact to EMG onset. Muscle activity magnitude was represented by the peak EMG value in the time period after onset, scaled relative to the largest value in the perturbation trials for that subject.

Correlations between pendulum impact velocity and muscle response variables were computed for each subject. Differences between age groups were assessed for maximum impact velocity before stepping, onset latencies, and correlations with t-tests, using $p < .05$ to determine significance.
RESULTS

Younger subjects were able to withstand greater pendulum velocities before stepping than the older adults (Mean±SD; Y: 496±137 deg/s; O: 380±107 deg/s; p = .032). Onset latencies for both muscles were longer in the older group than in the young (GA: p = .024; SO: p < .001; Fig. 1).

Correlation coefficients for onset latencies and impact velocity were negative for both muscles and age groups (Table 1), indicating shorter latencies with larger perturbations. In contrast, correlations between peak EMG and perturbation size were positive; as pendulum velocity increased so did the muscle activities for both age groups (Fig. 2). There were no age group differences in correlations for either onset latencies or EMG magnitudes.

Table 1. Correlations (r) of muscle response variables with pendulum velocity (Mean±SD).

<table>
<thead>
<tr>
<th></th>
<th>GA</th>
<th>SO</th>
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</thead>
<tbody>
<tr>
<td>Onset Latency</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young</td>
<td>-.43 ± .43</td>
<td>-.31 ± .53</td>
</tr>
<tr>
<td>Older</td>
<td>-.39 ± .38</td>
<td>-.07 ± .33</td>
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<tr>
<td>p</td>
<td>.844</td>
<td>.263</td>
</tr>
<tr>
<td>EMG Magnitude</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young</td>
<td>.71 ± .24</td>
<td>.84 ± .09</td>
</tr>
<tr>
<td>Older</td>
<td>.46 ± .45</td>
<td>.68 ± .28</td>
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<tr>
<td>p</td>
<td>.160</td>
<td>.114</td>
</tr>
</tbody>
</table>

DISCUSSION


Likewise, the correlations with perturbation magnitude for both EMG onset latency and activity agree with Lin & Woollacott (2002).

These similar findings occurred despite the use of a very different perturbation method than in these moving platform studies, and the findings of Brown et al. (2001) who showed how responses are modulated by the details of platform accelerations. The pendulum used here delivered a brief forward acceleration, with subsequent movement dictated solely by the subject response. Platform motion causes backward acceleration of the feet followed by subsequent deceleration that may occur within the response period. Onset latency and peak EMG measures may reflect only the first acceleration phase shared by both protocols.

REFERENCES


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