Changes in Multi-joint Performance with Age

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Running title: multi-joint control changes with age
Abstract

The purpose of this study was to determine whether elderly adults exhibit deficits in the performance of multi-joint movements. Two groups of subjects (mean ages 68.9, 30.1 years respectively) participated in this experiment. Subjects performed planar arm pointing movements to various targets. One target could be achieved via elbow extension only, while the remaining three required both elbow extension and horizontal shoulder flexion, thus requiring coordination at the two joints. In contrast to the young adults, the elderly adults produced movements that became less smooth and less accurate with increasing shoulder joint contribution. The results imply a selective coordination deficit for the elderly adults. In addition, the elderly adults coactivated opposing muscles more than the young adults for the single-joint movement. The elderly adults reduced coactivation at both joints for the two-joint actions, however, while the young adults did not. These data suggest a relationship between high coactivation levels and good performance for elderly adults. It may be more difficult for the elderly to implement high coactivation levels for multi-joint movements because of the increased energy costs and complexity of planning required in comparison to the single joint actions. Thus, elderly persons appear to use coactivation in a manner that is fundamentally different than young adults to achieve motor performance.
Introduction

Aging is accompanied by increases in movement duration (Seidler-Dobrin & Stelmach, 1998; Seidler & Stelmach, 1995; Welford, 1984) and variability (Contreras-Vidal, Teulings, & Stelmach, 1998; Cooke, Brown, & Cunningham, 1989) as seen in a variety of tasks. Investigations into how aging affects movement control have yielded data to support that increased reliance on feedback control is a major factor in movement slowing (Seidler-Dobrin & Stelmach, 1998; Haaland, Harrington, & Grice, 1993; Pohl, Weinstein, & Fisher, 1996). Furthermore, it appears that much of the increase in movement variability that occurs with age may be explained by reorganization of motor units, resulting in a reduced ability to finely grade muscle force amplitude (Seidler-Dobrin, He, & Stelmach, 1998; Galganski, Fuglevand, & Enoka, 1993; Erim, Beg, Burke, & de Luca, 1999). Despite these advances in our understanding of how aging affects the control system and control processes, little attention has been paid to whether coordination deficits occur with increasing age (Greene & Williams, 1996; Stelmach, Amreihn, & Goggin, 1988; Swinnen et al., 1998). This is surprising, as both inter- and intralimb coordination are required to achieve adequate manual dexterity and for virtually all activities of daily living. On average, 13% of adults over the age of 65 years experience difficulties with performing an activity of daily living, and 35% of those over 85 years experience difficulty (Schultz, 1992).

The few studies that have been conducted on aging and coordination have examined bimanual reaching movements (Stelmach et al., 1988; Greene & Williams, 1996). Stelmach and colleagues (1988) found greater reaction time (RT) differences between the two hands for elderly adults than for young adults when performing
bimanual aiming movements. In contrast, Rothstein and colleagues (1989) found no age effects on the interlimb differences in movement initiation and termination times. Greene and Williams (1996) demonstrated interlimb differences in movement amplitude for the elderly when performing bimanual rhythmic movements. The literature on aging and coordination remains inconclusive, and has also been more focused on bimanual tasks rather than intralimb coordination. Thus, the existence and extent of coordination impairments in the elderly remains unclear.

The potential factors that could contribute to reduced coordination in the elderly have not been fully identified either. Swinnen and colleagues (1998) demonstrated that older adults are less able to alter preferred coordination patterns in order to acquire new ones. These authors suggested that this may be due to the differential degeneration that occurs in the frontal lobes with increasing age (Kramer, Humphreys, Larish, Logan, & Strayer, 1994; West, 1996). Alternatively, Leonard and colleagues (Leonard, Matsumotor, Diedrich & McMillan, 1997) have demonstrated that the elderly exhibit delays in reflex modulation and inconsistent coupling between antagonist pairs of muscles. This could contribute to reduced intra- and interlimb coordination. It may also be that the problems contributing to increased slowing and movement variability observed in simple, single joint movements in the elderly are compounded for multi-joint movements. For example, we have previously shown that the heightened coactivation demonstrated by the elderly in single-joint movements is likely compensation for increased force variability (Seidler-Dobrin et al., 1998). The greater muscle force variability in the elderly appears to be associated with the motoneuron death that occurs with increasing age (Campbell, McComas, & Petito, 1973; Kanda & Hashizume, 1989).
Some of the muscle fibers that were previously innervated by these motoneurons are reinnervated into the remaining motoneurons. These newly grouped motor units produce greater force, due to the increased number of muscle fibers/motor unit. Thus the elderly are left with less fine control over force gradation than the young adults, especially at lower force output levels (Galganski et al., 1993; Erim et al., 1999). We demonstrated via experimental and simulation work that greater muscle coactivation is effective at damping out this muscle force variability in the elderly (Seidler-Dobrin et al., 1998) and smoothing movement trajectories. Similarly, in young adults, the magnitude of antagonist coactivation increases with the amount of instability (Hogan, 1984; Milner & Cloutier, 1993).

The purpose of this study was to determine whether elderly adults exhibit selective performance deficits for two-joint movements in comparison to single-joint actions. We compared the ability of both young and elderly adults to perform one- and two-joint goal-directed arm movements. We hypothesized that older adults, compared to young adults, would exhibit behavioral (kinematic) and neuromuscular (EMG) changes under multi-joint compared to single joint movements. Thus, we predict poorer performance on the two-joint movements for older adults.

**Methods**

(The methods, with small variations, have been described previously in Seidler, Alberts, & Stelmach, 2001.)

**Subjects**

Eight elderly adult subjects with a mean age of 68.9 (3.6) years were recruited from the community and paid $20.00 each for their participation, which took an average
of one hour. Seven young adult control subjects with a mean age of 30.1 (3.4) were recruited from the Arizona State University campus and were provided with course experiment credit for their participation. Subjects were given a health history questionnaire to exclude those who may have had a condition affecting their performance such as a recent history of stroke or arthritis. The elderly adults were given the Mini-Mental State test (Folstein, Folstein, & McHugh, 1975); a minimum score of 28 or higher was required for participation. All subjects were right handed, and all signed informed consent forms in accordance with human subjects policies.

Procedure and Design

Subjects were seated at a table with the chair height adjusted such that when the arm rested on the table, both the upper and the lower arm were parallel with the table surface. The subjects pointed to four targets on the tabletop, initiating each movement from the same start position directly in front of the subject’s midline, 24.5 cm from the table edge. The trial was ended when the subject stopped their movement by making contact with the tabletop. The four targets were arranged such that Target 1 required elbow extension only, with Targets 2-4 requiring increasing amounts of shoulder flexion (cf. Sainburg, Ghilardi, Poizner, & Ghez, 1995). Target 1 was located 45° to the right of the midline, 34 cm from the start position. Subjects used approximately 45° of elbow extension to achieve target 1 (see Table 1 for average joint excursions for each target). Target 2 was 28 cm anterior to the start position; subjects used approximately 45° of elbow extension and 8° of shoulder flexion to achieve this target. Target 3 was 45° to the left of the midline, 25 cm from the start. Subjects used approximately 33° of elbow extension and 15° of shoulder flexion to achieve target 3. Target 4 was 90° to the left of
the midline, 42 cm from the start position. Subjects used an approximately equal amount
of elbow extension and shoulder flexion to achieve target 4 (25° for both). Infrared light
emitting diodes were placed on the trunk (sternum), shoulder, elbow, and index fingernail
of the right arm to record movements. Marker data were acquired at 100 Hz using an
OPTOTRAK 3020 optoelectronic two-camera system. Trunk, wrist, and finger motion
were prohibited with braces. Each subject performed 20 trials to each target, presented in
2 blocks of 10 trials each. Subjects were instructed to move as fast and as accurately as
possible. The sequence of target presentation was counterbalanced across subjects.

Electromyographic data were recorded from the biceps brachii, medial head of the
triceps muscle, anterior and posterior deltoid muscles. Although the biceps muscle is
both an elbow and a shoulder flexor, it has been shown in this type of pointing task to act
primarily as a brachioradialis synergist (i.e., as an elbow flexor, Karst & Hasan, 1991).
Pre-amplified bipolar Ag-AgCl surface electrodes were placed one third of the distance
between the center of the muscle belly and the distal tendon-muscle insertion point.
Electrode placement was verified by examining whether isometric contraction caused
activity at the agonist electrode but not the antagonist. The electrodes were 35 mm in
length, 15 mm wide, and had an interelectode distance of 21 mm (center to center
distance). The amplifier used in the data collection was a Therapeutics Unlimited model
EMG-67 with an input impedance of >25 Megohms, a common mode rejection ratio of
87 db at 60 Hz, and noise of <1.5 V RMS (referred to input). A high pass filter with 20
Hz cutoff was used during collection to reduce motion artifacts. The EMG data were
sampled at 1000 Hz starting at least 500 ms before the movement began and continuing
throughout the trial, and were acquired synchronously with the kinematic data. A
submaximal effort was recorded for each muscle to allow amplitude normalization. These efforts consisted of holding a 2.5lb. weight in a static posture for each muscle.

Static postures chosen to elicit agonist activity without any antagonist activation were the following: 1) biceps brachii- elbow flexed to 90° with lower arm parallel to the floor, 2) triceps medial head- upper arm positioned vertically, next to the subject’s ear and the elbow flexed to about 45°, 3) anterior deltoid- arm extended straight out to the side and parallel to the floor, 4) posterior deltoid- initial position as in 3) but then internal shoulder rotation combined with horizontal shoulder extension of about 45 degrees.

Data Analysis

Kinematics. Position data from collected trials were subjected to a residual analysis to determine the appropriate smoothing cut-off frequency (Winter, 1990). All data were filtered at 8 Hz with a dual pass Butterworth digital filter to eliminate phase shift and differentiated to obtain angular velocity and acceleration. As sampling was initiated simultaneously with the go signal rather than upon movement initiation, the optimal algorithm of Teasdale, Bard, Fleury, Young & Proteau (1993) was used to determine movement onset. The algorithm works as follows: Locate the sample at which the time series first exceeds 10% of its maximum value (Vmax); working back from this point stop at the first sample (call it S) less than or equal to (Vmax/10)- (Vmax/100); find the standard deviation of the series between sample 1 and sample S (call this sd); working back from S stop at the first sample less than or equal to S-sd; this is the onset sample. The same algorithm was used in reverse to determine movement offset.
Elbow and shoulder joint angular excursions, velocities, and accelerations were determined. Extension was assigned the positive direction for joint motion and flexion was assigned as the negative direction. Movement smoothness was assessed by calculating the jerk score for the endpoint (fingertip) trajectory and normalizing it for movement distance and duration as follows:

\( \sqrt{\frac{1}{2} \int_j^2(t) \, dt \cdot \frac{d}{l^2}} \),

where \( j \) is the third time derivative of the position data, \( d \) is the movement duration, and \( l \) is the movement amplitude. The value is thus unit-free, normalized for the amplitude and duration of the movement. Movement straightness was analyzed by computing a straightness error for the endpoint (fingertip) trajectory and normalizing it for both movement distance and duration as follows:

\( \frac{1}{l} \cdot \sqrt{\sum (y(t) - y_d(t))^2/N} \),

where \( l \) is the movement length, \( y_d(t) \) is the least squares fit line through all of the data samples, \( y(t) \) is the array containing all of the data samples, and \( N \) is the number of samples.

Electromyography. The EMG signals were processed by removing the DC bias, full wave rectifying, and then low pass filtering at 20 Hz. The data were then amplitude-normalized by dividing by the average amplitude of the submaximal effort for each muscle. The peak amplitude and the time of occurrence of the peak was determined for each muscle. Coactivation scores were computed as follows: each EMG trace was time normalized to 500 samples and divided into 50 bins of ten samples each. Coactivation scores were determined by assigning a value of two to bins in which both muscles were active, otherwise the score for the bin was zero. If activation level for a muscle was
greater than 50% of the submax average amplitude across a bin, then the muscle was assessed as “on” during this bin. This allowed the determination of when within the movement subjects were coactivating. Peak coactivation scores were determined for each trial.

Statistics. A within subjects MANOVA (group × target direction) with repeated measures on target direction was used to determine how performance varied across the four targets for the two groups. The Huynh-Feldt epsilon (Huynh and Feldt 1970) was evaluated to determine whether the repeated measures data met the assumption of sphericity ($\Sigma > .75$). In cases where sphericity was met, the univariate tests were used to maintain power. Otherwise, the repeated measures were treated as multivariate. Note that the significance of the F ratio is assessed using different degrees of freedom depending on whether the univariate or the multivariate tests are used. The observed power was computed for all effects, as was $\eta^2$, an estimate of the total population variance that is explained by the variation due to the treatment (Keppell, 1991). Its value does not depend on sample size or power of the experiment. Its values can range between 0.0 and 1.0, with negative values a possibility when the associated F value is less than 1.0. Cohen suggests that a small effect is comparable to an $\eta^2$ of .01, a medium effect is .06, and a large effect is .15 or greater (Cohen, 1977). These standards were employed in our assessment of treatment effect sizes.

Results

Table 1 summarizes the characteristics of movements made to each target for the two groups. The resultant distance traveled by the fingertip did not differ between the two groups ($F_{1,13}=2.2, p>.10$), although it did vary by target number ($F_{3,11}=215.1, p<.01$,
$\omega^2=.80$, large effect size). There was little change in the angular distance achieved with the elbow joint across targets, with the exception of target #4 where the elderly adults covered a slightly smaller distance than the young adults (group x target interaction, $F_{3,11}=7.6, p<.01, \omega^2=.54$, large effect size). There were no group differences in the distance traveled about the shoulder joint ($F_{1,13}=1.5, p>.10$). The magnitude of shoulder joint motion did increase across targets ($F_{3,39}=228.9, p<.01, \omega^2=.96$, large effect size). Total movement duration also increased with target number ($F_{3,39}=57.5, p<.01, \omega^2=.86$, large effect size) and was greater for the elderly adults than for the young adults ($F_{1,13}=13.2, p<.01, \omega^2=.47$, large effect size). Overall, the young subjects produced smoother movements than the elderly did, resulting in a group main effect for normalized jerk score (Figure 1, $F_{1,13}=20.5, p<.01, \omega^2=.58$, large effect size). Moreover, although the interaction was not significant, the elderly exhibited a quadratic trend across target number ($F_{1,7}=4.8, p=.06$). That is, they increased jerk score when going from the single- to the two-joint movements. In contrast, the elderly adults produced straighter movements than the young adults for movements to target #1, and there were no group differences in normalized straightness error for the remaining targets (Figure 1, group x target interaction, $F_{3,39}=6.6, p<.01, \omega^2=.38$, large effect size). The elderly adults also increased endpoint error with increasing target number while the young adults did not (Figure 1, group x target interaction, $F_{3,39}=4.7, p<.01, \omega^2=.28$, large effect size).

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Insert Table 1, Figure 1 here

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Sample EMG profiles for a representative young and elderly subject for movements to target #1 and #4 are presented in Figure 2. The EMG data has been rectified and normalized to a percentage of the submaximal activation for the figure. The electromyographic variables changed as a function of target number, reflecting the varying coordination demands for movements to each target (Table 2). For both groups, the maximum triceps activity decreased with target number ($F_{3,11}=9.1$, $p<.01$, $\omega^2=.41$, large effect size) while the maximum biceps activity was greater for movements to target #4 ($F_{3,11}=4.6$, $p=.02$, $\omega^2=.26$, large effect size). There was no change in the posterior deltoid maximum burst. The young adults increased maximum anterior deltoid burst with target number, while the elderly adults did not ($F_{3,11}=4.3$, $p=.03$, $\omega^2=.53$, large effect size). Maximum elbow and shoulder coactivation scores for the two age groups varied as well (Figure 3). The young increased elbow joint coactivation with increasing target number, while the elderly adults decreased it ($F_{3,39}=4.6$, $p<.01$, $\omega^2=.28$, large effect size). Although not significant, there was a trend for the elderly group to decrease shoulder joint coactivation with increasing target number while the young did not change ($\omega^2=.07$, medium effect size). Additionally, peak shoulder coactivation occurred later in the movement for the young than the elderly adults, in particular for movements to targets 2 and 3 ($F_{3,39}=3.9$, $p=.01$, $\omega^2=.24$, large effect size). Thus the elderly adults had greater coactivation for movements to target 1 than the young adults, and decreased the magnitude of this coactivation across the remaining target numbers to a level that was comparable to that of the young adults.

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Insert Figure 2, 3, Table 2 here
Discussion

The elderly adults demonstrated a differential decline in performance when going from the single- to the two-joint movements, compared to young adults. As movement coordination requirements increased (e.g. greater shoulder joint displacement) older adults produced significantly less smooth movement trajectories and made larger errors. For the single-joint actions, in which performance was similar for the elderly and young adults, the elderly exhibited greater coactivation levels than the young adults. This potentially served to damp out their movement variability (Seidler-Dobrin et al., 1998), allowing them to produce single-joint movements that were kinematically very similar to those of the young. For movements to target 1 (elbow joint extension only), the elderly subjects actually produced spatial trajectories that were straighter than those of the young subjects. Indeed, it has been demonstrated with young subjects that joint stiffness and stability can be increased with muscle coactivation (Hogan, 1984; Milner & Cloutier, 1993). The group difference in coactivation level disappeared for the two-joint actions as the elderly adults exhibited reducing coactivation levels at both the elbow and the shoulder joint, bringing them to the same magnitude as the young adults.

It may be difficult to implement high coactivation levels at each joint participating in a multi-joint movement because the energy costs and complexity of planning required would increase with additional degrees of freedom. This is compounded by the fact that the maximum muscle force output for elderly adults can be up to one third below that of young adults (Schultz et al., 1992). Evidence of difficulties with multi-joint coactivation can be seen in the way that the young and the elderly adults differed in how EMG
activation patterns were altered to achieve the transition from single- to two-joint movements. The young adults increased the magnitude of the anterior deltoid burst in order to increase shoulder horizontal flexion across the target numbers. In contrast, burst magnitudes were already large for the elderly adults even under the single joint movement because of the high level of co-contraction. Thus, it would be potentially difficult and fatiguing for elderly adults to further increase anterior deltoid activation. Instead, the elderly adults tended to decrease shoulder joint coactivation levels. This would reduce shoulder joint stiffness to allow more motion with increasing target number (Hogan, 1984; Milner & Cloutier, 1993).

It is somewhat surprising that the elderly subjects demonstrated heightened jerk scores but not straightness errors in comparison to the young adults. A previous investigation of handwriting in young and elderly adults found the elderly to show higher scores for both measures (Contreras-Vidal et al., 1998). These authors suggested that increased jerk scores were a result of inconsistent modulation of force during the movement, while increased straightness errors arose due to impaired timing control between the multiple joints contributing to the movement. It may be that aging affects the ability to coordinate joint timing for fine motor control tasks such as handwriting, but has less of an impact for the larger arm movements produced in the current study.

We have previously demonstrated that coactivation in elderly subjects is a functional means of damping out movement variability and improving movement performance (Seidler-Dobrin et al., 1998). It is notable that the elderly adults in the current study also had similar performance to the young adults for the single joint movement, in comparison to the two-joint actions. This was found despite the differing
tasks used for the two experiments. In the current study, subjects performed planar pointing movements unencumbered while in the previous study they made aiming movements with the arm resting on a low-friction manipulandum. The greater coactivation exhibited by the elderly subjects in this study appeared to have a positive impact on performance, as well.

Movement amplitude was varied across the targets in this study in order to achieve varying combinations of elbow and shoulder joint motion. One of our recent experiments examined the ability of elderly adults to produce goal-directed actions when task difficulty was manipulated via changing movement amplitude and target size (Ketcham, Seidler, van Gemmert, & Stelmach, 2001). Their data demonstrated that elderly adults did not modulate or scale velocity to the degree shown by young adults when movement amplitude was increased. The task difficulty effects observed in Ketcham et al. (2001) however appear to be overshadowed in the current experiment by the effects of switching from a single to a two-joint movement. Thus it seems that changing joint involvement may have a larger impact on performance for the elderly adults than modulating movement amplitude.

In conclusion, data from this experiment supports our previous finding that elderly adults coactivate antagonist muscles to a greater extent than young adults for single-joint actions (Seidler-Dobrin et al., 1998). This cocontraction appears to compensate for heightened muscle force variability in elderly adults and smoothes out their movements. Importantly, the current work demonstrates that elderly adults do not apply this compensatory mechanism to movements that are more complex (i.e. two-joint movements). Instead, they decrease coactivation levels when transitioning from single-
to two-joint actions, in contrast to young adults, who increase elbow joint coactivation. Decreasing coactivation is associated with a deterioration in performance for the elderly subjects in the execution of two-joint movements compared to single-joint ones. Increasing coactivation may be difficult for elderly subjects to implement in two-joint actions due to the increase in complexity of movement planning and its associated energy costs. Thus, elderly adults appear to use coactivation in a fundamentally different manner from the young to achieve motor performance.
References


in the healthy old. The Gerontologist, 29, 258A-259A.


West RL (1996). An application of prefrontal cortex function theory to cognitive aging.

   Psychological Bulletin, 120, 272-292.


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Figure Captions

1. Sample spatial trajectories, mean (standard deviation) normalized jerk score, normalized straightness error, and endpoint error describing fingertip motion to each of the targets in plotted for the two age groups. The young adults produced smoother movements than the elderly adults. In contrast, elderly adults produced straighter movements than the young for movements to target 1 (elbow joint only). Additionally, the elderly adults exhibited increasing endpoint error across the targets while the young adults did not.

2. EMG data for one young and one elderly adult making movements to targets 1, 4. YA is a young adult subject, EA is an elderly adult. The biceps and triceps activation is plotted in the left panels and anterior and posterior deltoid in the right panels. The elderly adult exhibits greater coactivation of antagonistic muscle pairs than the young adult.

3. Mean (standard deviation) peak coactivation scores at the elbow and shoulder joint are presented for movements to all targets for each age group. The elderly subjects decreased coactivation with target number while the young either increased it (elbow joint) or kept the level constant (shoulder joint).
Table 1. Mean (standard deviation) performance variables. *target main effect, +group main effect, # group x target interaction

<table>
<thead>
<tr>
<th>DV</th>
<th>Group</th>
<th>Target 1</th>
<th>Target 2</th>
<th>Target 3</th>
<th>Target 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fingertip distance (cm)*</td>
<td>Young</td>
<td>35.0 (1.9)</td>
<td>29.3 (1.5)</td>
<td>26.8 (2.3)</td>
<td>41.7 (2.3)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>34.9 (1.0)</td>
<td>25.6 (9.2)</td>
<td>23.0 (8.4)</td>
<td>40.9 (1.1)</td>
</tr>
<tr>
<td>Elbow angular distance (degrees)#</td>
<td>Young</td>
<td>44.3 (7.9)</td>
<td>52.4 (6.7)</td>
<td>42.9 (5.8)</td>
<td>43.5 (7.9)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>48.6 (5.0)</td>
<td>49.9 (6.7)</td>
<td>38.3 (7.3)</td>
<td>32.9 (7.8)</td>
</tr>
<tr>
<td>Shoulder angular distance (degrees)*</td>
<td>Young</td>
<td>-3.9 (2.8)</td>
<td>-16.6 (2.9)</td>
<td>-26.0 (4.2)</td>
<td>-44.5 (8.1)</td>
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<tr>
<td></td>
<td>Elderly</td>
<td>-2.6 (5.2)</td>
<td>-14.1 (8.0)</td>
<td>-22.9 (8.8)</td>
<td>-36.3 (10.9)</td>
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<tr>
<td>Total movement duration (ms)**+</td>
<td>Young</td>
<td>416 (74)</td>
<td>498 (62)</td>
<td>532 (54)</td>
<td>605 (72)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>615 (114)</td>
<td>695 (119)</td>
<td>709 (121)</td>
<td>738 (130)</td>
</tr>
</tbody>
</table>
Table 2. Mean (standard deviation) electromyographic variables; EMG is presented as percent of submaximal effort, time in ms. *target main effect, +group x target interaction

<table>
<thead>
<tr>
<th>DV</th>
<th>Group</th>
<th>Target 1</th>
<th>Target 2</th>
<th>Target 3</th>
<th>Target 4</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max biceps activity*</td>
<td>Young</td>
<td>104 (85)</td>
<td>86 (56)</td>
<td>96 (45)</td>
<td>179 (114)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>143 (93)</td>
<td>144 (93)</td>
<td>148 (88)</td>
<td>170 (81)</td>
</tr>
<tr>
<td>Time max biceps activity*</td>
<td>Young</td>
<td>587 (53)</td>
<td>509 (120)</td>
<td>381 (170)</td>
<td>297 (130)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>611 (127)</td>
<td>565 (188)</td>
<td>424 (173)</td>
<td>393 (144)</td>
</tr>
<tr>
<td>Max triceps activity*</td>
<td>Young</td>
<td>175 (147)</td>
<td>186 (78)</td>
<td>123 (48)</td>
<td>109 (57)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>181 (132)</td>
<td>150 (89)</td>
<td>107 (54)</td>
<td>98 (64)</td>
</tr>
<tr>
<td>Time max triceps activity</td>
<td>Young</td>
<td>437 (99)</td>
<td>440 (166)</td>
<td>501 (157)</td>
<td>530 (96)</td>
</tr>
<tr>
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<td>414 (123)</td>
<td>481 (113)</td>
<td>493 (78)</td>
<td>439 (114)</td>
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<tr>
<td>Max anterior deltid activity+</td>
<td>Young</td>
<td>53 (34)</td>
<td>104 (40)</td>
<td>106 (38)</td>
<td>139 (60)</td>
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<tr>
<td></td>
<td>Elderly</td>
<td>106 (77)</td>
<td>108 (69)</td>
<td>93 (47)</td>
<td>86 (34)</td>
</tr>
<tr>
<td>Time max anterior deltid activity*</td>
<td>Young</td>
<td>525 (42)</td>
<td>448 (142)</td>
<td>409 (106)</td>
<td>457 (157)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>494 (73)</td>
<td>452 (107)</td>
<td>416 (118)</td>
<td>442 (176)</td>
</tr>
<tr>
<td>Max posterior deltid activity</td>
<td>Young</td>
<td>72 (54)</td>
<td>73 (37)</td>
<td>67 (35)</td>
<td>59 (33)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>67 (38)</td>
<td>71 (40)</td>
<td>68 (39)</td>
<td>60 (51)</td>
</tr>
<tr>
<td>Time max posterior deltid activity*</td>
<td>Young</td>
<td>475 (117)</td>
<td>605 (72)</td>
<td>594 (74)</td>
<td>486 (184)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
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<td>593 (132)</td>
<td>538 (148)</td>
<td>521 (199)</td>
</tr>
<tr>
<td>Max elbow coactivation+</td>
<td>Young</td>
<td>0.69 (0.79)</td>
<td>0.98 (0.87)</td>
<td>1.12 (1.01)</td>
<td>1.16 (0.98)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>1.81 (0.42)</td>
<td>1.58 (0.60)</td>
<td>1.34 (0.71)</td>
<td>1.06 (0.88)</td>
</tr>
<tr>
<td>Time max elbow coactivation*</td>
<td>Young</td>
<td>543 (181)</td>
<td>314 (394)</td>
<td>365 (170)</td>
<td>433 (292)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>333 (232)</td>
<td>328 (271)</td>
<td>260 (193)</td>
<td>279 (223)</td>
</tr>
<tr>
<td>Max shoulder coactivation</td>
<td>Young</td>
<td>0.23 (0.45)</td>
<td>0.54 (0.67)</td>
<td>0.29 (0.42)</td>
<td>0.28 (0.65)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>1.08 (0.99)</td>
<td>1.10 (0.94)</td>
<td>0.87 (0.84)</td>
<td>0.58 (0.87)</td>
</tr>
<tr>
<td>Time max shoulder coactivation+</td>
<td>Young</td>
<td>428 (97)</td>
<td>583 (63)</td>
<td>524 (184)</td>
<td>438 (24)</td>
</tr>
<tr>
<td></td>
<td>Elderly</td>
<td>276 (243)</td>
<td>268 (255)</td>
<td>305 (288)</td>
<td>295 (310)</td>
</tr>
</tbody>
</table>
Figure 1.
Figure 2.

Movement to Target 1

EMG (% submax)

0 300 600 900

Biceps brachii
Triceps, medial head

Anterior Deltoid
Posterior Deltoid

Time (ms)

Movement to Target 4

EMG (% submax)

0 300 600 900

Biceps brachii
Triceps, medial head

Anterior Deltoid
Posterior Deltoid

Time (ms)
Figure 3.