Neural and Muscular Contributions to the Age-Related Reductions in Rapid Strength

GENA R. GERSTNER[1,2], BRENNAN J. THOMPSON[3], JOSEPH G. ROSENBERG[4], ERIC J. SOBOLEWSKI[5], MICHAEL J. SCHARVILLE[4], and ERIC D. RYAN[1,6]

[1]Neuromuscular Research Laboratory, Department of Exercise and Sport Science, University of North Carolina at Chapel Hill, Chapel Hill, NC; [2]Human Movement Science Curriculum, University of North Carolina at Chapel Hill, Chapel Hill, NC; [3]Department of Kinesiology and Health Science, Utah State University, Logan, UT; [4]Quintiles, Durham, NC; [5]Department of Health Sciences, Furman University, Greenville, SC; [6]Department of Allied Health Sciences, University of North Carolina at Chapel Hill, Chapel Hill, NC

ABSTRACT

GERSTNER, G. R., B. J. THOMPSON, J. G. ROSENBERG, E. J. SOBOLEWSKI, M. J. SCHARVILLE, and E. D. RYAN. Neural and Muscular Contributions to the Age-Related Reductions in Rapid Strength. *Med. Sci. Sports Exerc.*, Vol. 49, No. 7, pp. 1331–1339, 2017. **Introduction**: The purposes of this study were to investigate the age-related differences in absolute and normalized plantarflexion rate of torque development (RTD) at early (0–50 ms) and late (100–200 ms) time intervals and to examine specific neural and muscular mechanisms contributing to these differences. **Methods**: Thirty-two young (20.0 ± 2.1 yr) and 20 older (69.5 ± 3.3 yr) recreationally active men performed rapid plantarflexion isometric muscle actions to examine absolute and normalized RTD and muscle activation using EMG at early and late time intervals. Ultrasonography was used to examine medial gastrocnemius muscle size, echo intensity (EI), and muscle architecture (fascicle length [FL] and pennation angle [PA]). **Results**: The older men were weaker (23.9%, $P < 0.001$) and had lower later absolute and normalized RTD ($P = 0.001$–0.034) variables when compared with the young men. The older men also had higher EI ($P < 0.001$), smaller PA ($P = 0.004$), and lower later EMG amplitude values ($P = 0.009$–0.046). However, there were no differences in early RTD and EMG amplitude values, muscle size, or FL between groups ($P = 0.097$–0.914). Lower late RTD values were related to higher EI, smaller PA, and lower EMG amplitude values ($r = -0.28$–0.59, $P = 0.001$–0.044); however, late RTD values were no longer related to PA after normalizing to peak torque. **Conclusions**: Age-related alterations in muscle quality (EI), architecture, and muscle activation may influence rapid torque production at late time intervals (≥100 ms) from contraction onset. These findings highlight specific neuromuscular factors that influence the age-related reductions in RTD, which has been shown to significantly influence function and performance in older adults. **Key Words:** RATE OF TORQUE DEVELOPMENT, EXPLOSIVE STRENGTH, PLANTARFLEXORS, NEUROMUSCULAR FUNCTION

It is projected that by 2030, the number of older adults ages 65 yr and older will represent nearly 20% of the total population (30). A large percentage of older adults experience functional limitations with everyday tasks and a high incidence of injuries resulting in a significant economic burden (17). For example, one out of three older adults experience a fall each year (40), an event that often times results in accelerated deteriorations in health (4). Age-related changes in neuromuscular function have been proposed as the potential cause of these life-threatening conditions (10,12,14–16,20,24,39).

Many previous studies (5,12,16,37–39) have demonstrated that aging is associated with losses in maximal muscle strength (i.e., peak torque [PT]). However, rapid strength, or the rate of force/torque development (RFD/RTD), has been reported to decrease at a greater magnitude than maximal strength (12,16,38,39). These findings are clinically important given the findings of recent papers (28,31), which have suggested that RTD measured during the initial 200 ms from contraction onset is more functionally relevant than PT. For example, RTD has been shown to be an important discriminator of fall history (6) in older adults and influences maximal walking speed (9). Thus, determining mechanisms contributing

Address for correspondence: Eric D. Ryan, Ph.D., Department of Exercise and Sport Science and Department of Allied Health Sciences, University of North Carolina at Chapel Hill, 209 Fetzer Hall, CB #8700 Chapel Hill, NC 27599-8700; E-mail: edryan@email.unc.edu.
Submitted for publication September 2016.
Accepted for publication January 2017.

0195-9131/17/4907-1331/0
MEDICINE & SCIENCE IN SPORTS & EXERCISE®

DOI: 10.1249/MSS.0000000000001231

to the age-related reductions in RTD is an important approach to help identify key areas that could help diminish age-related reductions in overall function and reduce the risk of fall-related injuries.

Few studies have examined specific neural and muscular variables contributing to the age-related reductions in rapid strength. For example, it has been suggested that the primary mechanism underlying RTD is voluntary activation (i.e., motor unit firing rates) (20), yet contrasting findings exist regarding a lack of change in muscle activation between healthy young and older adults (38). In addition, other muscle-specific factors (i.e., muscle size, muscle quality, and architecture) have been shown to be altered with aging (14,19,24,29,35,36) and may also contribute to changes in RTD as supported by previous studies that have demonstrated associations between muscle thickness and echo intensity (EI) (33,41) with RTD in older adults. However, a recent paper (25) has suggested that muscle architecture may also be an important variable contributing to RTD that requires further examination. Accordingly, there is a need for a comprehensive simultaneous examination of the potential mechanisms contributing to the age-related reductions in RTD.

Numerous studies (2,18,26,38,39) have examined RTD at multiple time intervals (i.e., 30, 50, 100, and 200 ms) from the onset of contraction because different time intervals may be governed by unique physiological parameters. For example, early RTD intervals are primarily influenced by muscle activation, whereas the late RTD intervals are more closely related to maximal strength (3). Furthermore, normalizing RTD variables to PT has been suggested to represent more qualitative characteristics independent of peak strength (2). However, contrasting findings exist regarding the influence of normalization on the age-related differences in RTD (18,26,37–39). To our knowledge, we are aware of no previous studies that have simultaneously examined the neural and muscle-specific factors contributing to the age-related reductions in RTD while using many of the recommendations for RTD assessments as suggested in a recent review by Maffiuletti et al. (25). Therefore, the purpose of this study was to determine specific mechanisms contributing to the age-related reduction in absolute and normalized plantarflexion RTD at early (0–50 ms) and late (100–200 ms) time intervals, which have been suggested to be governed by unique physiological parameters (2,3). On the basis of a similar study examining the plantarflexors (38), we hypothesized that there would be a reduction in both absolute and normalized RTD at early and late time intervals. Furthermore, we hypothesized that early and late RTD intervals would be influenced by muscle activation (2,13,20,25) and EI (muscle quality) (33,41), whereas late RTD variables would also be influenced by muscle size (25) and architecture (2,25).

METHODS

Participants. Thirty-two young (mean ± SD, age = 20.0 ± 2.1 yr, body mass = 73.6 ± 9.4 kg, stature = 174.8 ± 7.2 cm) and 20 older (69.5 ± 3.3 yr, 79.7 ± 7.9 kg, 176.9 ± 5.7 cm) men volunteered to participate in this study. All participants were considered recreationally active because of their self-reported weekly exercise habits (young = 6.5 ± 4.1 $h \cdot wk^{-1}$, older = 5.4 ± 2.0 $h \cdot wk^{-1}$). None of the participants reported any metabolic or neuromuscular diseases or musculoskeletal injuries sustained within the past 3 months specific to the low back, hip, knee, or ankle. All participants completed and signed an approved consent form and a health history and exercise status questionnaire. This study was approved by the university institutional review board.

Experimental design. Each participant visited the laboratory on two separate occasions separated by 2–7 d at the same time of day (±2 h). The first visit was a familiarization trial where all participants practiced the isometric strength assessments. The second visit was the experimental session that included all the ultrasound (US) assessments, followed by the maximal isometric strength assessments. All participants refrained from any vigorous physical activity 24 h before testing.

US assessments. Before conducting the US assessments, participants laid prone for a 10-min rest period to allow for fluid redistribution to minimize measurement error (8). The US imaging assessments of the medial gastrocnemius (MG) were performed on the right limb at approximately one-third the distance from the articular cleft between the femur and the tibia condyles to the lateral malleolus (34). The distance of the lower limb was measured using a Gulick tape measure (AliMed, Dedham, MA) and marked while the participants were seated with their leg relaxed. The US images were obtained with a portable brightness mode (B-mode) US imaging device (LOGIQ e 5; General Electric Company, Milwaukee, WI) and a multifrequency linear array probe (12L-RS; 5–13 MHz; 38.4 mm field of view) (General Electric Company). All US measurements were taken with the participants lying prone with their legs fully extended and relaxed. To prevent rotation and movement of the leg, the right foot was attached securely to a vertical post at a 90° joint angle between the foot and the leg with an adjustable strap as described previously (34). Water-soluble transmission gel was applied generously to the skin to enhance acoustic coupling and reduce possible near-field artifacts. Image quality was ensured by having the same experienced sonographer perform all of the assessments as well as having a minimum of two investigators present during imaging to verify image quality.

Panoramic US images of the MG were taken to determine anatomical muscle size (cross-sectional area [CSA]), subcutaneous fat corrected EI to represent muscle quality, pennation angle (PA), and fascicle length (FL). The US was set to the musculoskeletal mode before testing with equipment settings optimized for image quality. Panoramic images of muscle size and quality (i.e., CSA and EI) were obtained with settings that were held constant across participants and were as follows: gain (68 dB), depth (4 cm), and frequency (10 MHz). A custom-made probe support composed of high-density

foam padding was positioned perpendicular to the longitudinal axis of the plantarflexor muscles and secured with an adjustable strap (34). This ensured that the US probe was moved perpendicular to the skin and along the transverse plane (from medial to lateral) during the panoramic imaging assessments. Panoramic images of muscle architecture (i.e., PA and FL) were obtained in the fascicular plane, and equipment settings were adjusted if image quality needed to be optimized. For each assessment, the probe was moved in a slow, continuous movement while consistent minimal pressure was applied to the skin to avoid any muscle compression. Real-time panoramic cross-sectional images of the MG were generated using LogicView (GE Logiq e) software.

Muscle size and quality analysis. ImageJ software (version 1.46r; National Institutes of Health, Bethesda, MD) was used for all US imaging analyses. Using the straight-line function, each image was converted from pixels to centimeters. To determine CSA of the MG, the polygon function was used to select the region of interest, including as much of the muscle as possible without any surrounding fascia. Using the same preselected region of interest, the mean EI values were assessed by computer-aided gray scale analysis with the standard histogram function to determine muscle quality, ranging between 0 and 255 arbitrary units (black = 0, white = 255). Values of EI were then corrected for subcutaneous fat (42), which was measured using the straight-line function from the skin to the superficial aponeurosis at half the distance between the medial and the lateral boarder of the MG. Previously, our laboratory has reported test–retest reliability statistics that have demonstrated intraclass correlation coefficients and SEM (expressed as a percentage of the mean) values of 0.914 and 0.720 and 5.830% and 3.680% for CSA and EI, respectively (34).

Fascicle length and PA analysis. Fascicle length was determined as the length of the fascicle between the superficial and the deep aponeuroses just inside the surrounding fascia of the MG. Two fascicles from the middle of each image were measured and averaged for subsequent analyses. Using the same fascicles, PA was determined as the angle between the fascicle and the deep aponeurosis, and the average of the two PA measurements was used for subsequent analyses. We were unable to obtain these data from one participant in the older group. Previously, our laboratory has reported test–retest reliability for FL and PA, which yielded intraclass correlation coefficients and SEM values of 0.945 and 0.809 and 4.25% and 6.52%, respectively.

Torque assessments. The isometric strength assessments were conducted on a calibrated HUMAC Norm dynamometer (Computer Sports Medicine Inc., Stoughton, MA) using a custom-designed steel foot plate (length = 36 cm, width = 17 cm, thickness = 0.9 cm) to examine plantarflexor torque production. Participants were seated at a 135° angle between the torso and the thigh, with restraining straps placed over the chest, pelvis, and thigh, and they were required to keep their arms crossed in front of their chest during testing. The participants' right leg was fully extended (0° below the horizontal plane) with the ankle joint angle in a neutral position (0° of dorsiflexion). The foot was secured in a thick rubber heel cup with straps placed over the toes and metatarsals, aligning the lateral malleolus of the fibula with the axis of rotation of the dynamometer.

Before the maximal strength testing, participants performed a warm-up composed of two submaximal isometric voluntary contractions at 50% and 75% of their perceived maximal effort for 3–4 s. Each participant then performed 2–3 rapid and maximal voluntary contractions (MVC) for 3–4 s with 2 min of recovery between trials. Over the duration of each MVC, participants received strong verbal encouragement in which they were instructed to plantarflex "as hard and fast as possible" without any preceding countermovement as previously described (38,39). To ensure quality of the MVC, each MVC was visually inspected for pretension or countermovement. The slope of the baseline (100 ms) before contraction onset was calculated to determine whether there was pretension or a countermovement before the MVC. During analyses, if the slope values exceeded ± 5 N·m·s^{-1} during baseline (preceding 100 ms), the MVC was discarded. Consequently, three participants (two young and one older) were excluded before analyses as all of their MVC attempts exceeded this threshold. The average baseline torque slope values of the remaining participants were 0.07 ± 1.06 N·m·s^{-1}. In addition, to determine baseline fluctuations, 3 SD of the baseline mean was not different between groups ($P = 0.838$), and the mean value across groups was (0.11 ± 0.09 N·m), which was 0.09% MVC.

EMG. A preamplified, bipolar surface electrode (TSD150B; Biopac Systems Inc., Santa Barbara, CA; gain = 350 and interelectrode distance of 20 mm) was placed parallel to the muscle fiber orientation on the midbelly of the MG (38). A pregelled, disposable reference electrode was placed over the tibial tuberosity on the right leg (Ag-Ag Cl Quinton Quick Prep; Quinton Instruments Co., Bothell, WA). Preceding the electrode placement, the skin was shaved, lightly abraded, and cleaned with isopropyl alcohol to reduce interelectrode impedance and increase the signal-to-noise ratio.

Signal processing. The EMG (μV) and torque (N·m) signals were sampled simultaneously at 2.5 kHz with a Biopac data acquisition system (MP150WSW; Biopac Systems, Inc.) and stored on a personal computer (Think Pad T420; Lenovo, Morrisville, NC). A custom-written software (Labview 8.5; National Instruments, Austin, TX) was used to process all of the signals offline. All EMG signals were filtered with a fourth-order, zero phase shift Butterworth filter with a band pass of 20–400 Hz, whereas the torque signals were corrected for baseline passive tension and filtered using a fourth-order, zero phase shift low pass Butterworth filter with a 50-Hz cutoff frequency. The identification of the torque and the EMG onsets were separately determined by the same experienced investigator using a high-resolution *x*- and *y*-axis scale as recommended by Maffiuletti et al. (25). For EMG and torque onset, a horizontal line was plotted at 3 SD

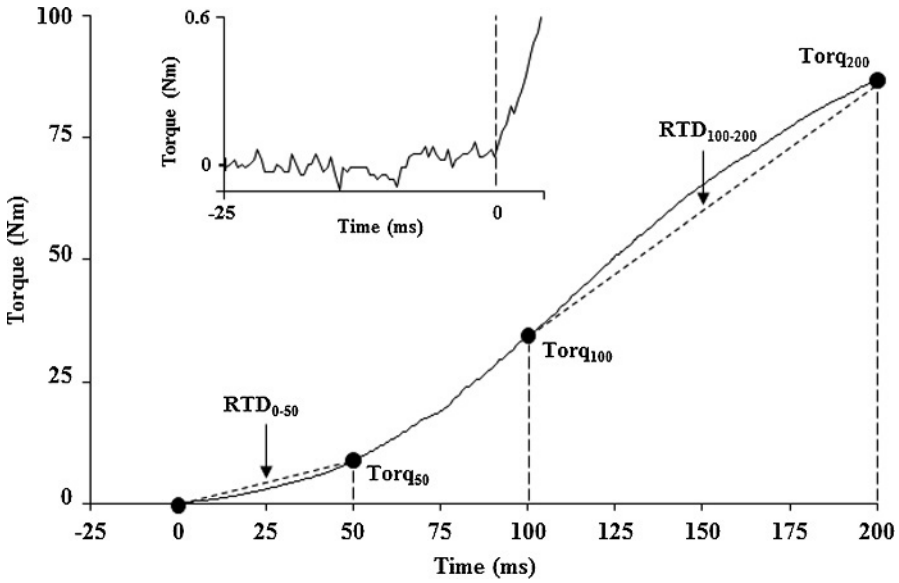

FIGURE 1—An example of a torque–time curve during an isometric MVC of the plantarflexors. Variables of interest of the torque–time curve include torque at 50, 100, and 200 ms ($Torq_{50}$, $Torq_{100}$, and $Torq_{200}$) and early and late RTD intervals ($RTD_{0–50}$ and $RTD_{100–200}$). A magnification of the torque–time signal is provided to demonstrate the manual onset detection defined as the last trough before the signal deflects from baseline as recommended by Maffiuletti et al. (25).

above baseline and was used as a guide for manually determining where the signal deflected from baseline. The investigator zoomed in near the signal onset, including the horizontal line, and placed a vertical cursor at the point at which the respective signal deflected (i.e., last trough before signal deflection; Fig. 1) from baseline (25).

Isometric PT was determined as the highest 500-ms epoch during the 3- to 4-s MVC plateau. The rapid torque variables were calculated from the torque–time curve at 50, 100, and 200 ms from onset ($Torq_{50}$, $Torq_{100}$, and $Torq_{200}$, respectively; Fig. 1). The absolute and normalized RTD variables were quantified from the slope of the absolute and normalized torque–time curves, respectively, at time intervals of 0–50 ms (aRTD$_{0–50}$; nRTD$_{0–50}$) and 100–200 ms (aRTD$_{100–200}$; nRTD$_{100–200}$), similar to Thompson et al. (38). These specific time intervals were chosen to represent nonoverlapping early (RTD$_{0–50}$) and late (RTD$_{100–200}$) torque–time characteristics (Fig. 1), which may represent different physiological parameters according to Andersen and Aagaard (3). It is important to note that Andersen and Aagaard (3) examined RFD in 10-ms overlapping time intervals ranging from 0 to 250 ms from the onset of contraction.

A measure of muscle activation was computed for each time interval of interest as the root-mean-square (RMS) amplitude of the EMG. Each value was normalized to the peak RMS amplitude value defined as the highest RMS value (500-ms epoch) during the MVC. The EMG variables were quantified at 50-ms time intervals that included 0–50 ms (EMG$_{0–50}$), 100–150 ms (EMG$_{100–150}$), and 150–200 ms (EMG$_{150–200}$). These epochs were chosen to correspond with the time intervals for early (0–50 ms) and late (100–200 ms) RTD variables.

Statistical analyses. Independent samples t-tests were used to examine any differences between the young and the older men for all demographic data (body mass, stature, and subcutaneous fat thickness), muscle-specific variables (CSA, EI, PA, and FL), PT, rapid torque variables, absolute and normalized RTD variables, and muscle activation. Normality was assessed using the Shapiro–Wilk test, and nonnormally distributed variables were log-transformed before analysis. A Pearson product moment correlation coefficient was used to examine the relationship between each of the absolute and normalized RTD variables and PT, muscle architecture, and the muscle activation variables. All analyses were performed with SPSS (IBM SPSS Statistics for Windows, Version 21.0., Armonk, NY) with an alpha of $P \leq 0.05$ to determine statistical significance.

RESULTS

There was a significant difference in age ($P < 0.001$) and body mass ($P = 0.020$), but not in stature ($P = 0.287$), self-reported weekly exercise ($P = 0.270$), or subcutaneous fat thickness of the MG (young = 0.287 ± 0.023 cm, older = 0.322 ± 0.368 cm, $P = 0.397$) between the young and the older men. The young men exhibited greater PT, $Torq_{100}$, and $Torq_{200}$ ($P = 0.001–0.040$) values than the older men; however, no difference was observed for $Torq_{50}$ ($P = 0.429$) between groups (Table 1, Fig. 2).

For the absolute RTD variables, the young men exhibited greater aRTD$_{100–200}$ ($P < 0.001$) values than the older men; however, no difference was observed for aRTD$_{0–50}$ ($P = 0.395$) between groups (Fig. 3A). For aRTD$_{0–50}$, there was a significant relationship with EMG$_{0–50}$ ($r = 0.44$, $P = 0.001$);

TABLE 1. Mean ± SD values for torque, muscle architecture, and muscle activation variables in the young and older men.

Variable	Age-Group	Mean ± SD
PT (N·m)	Young	142.4 ± 29.1*
	Older	108.4 ± 19.2
$Torq_{50}$ (N·m)	Young	8.9 ± 4.4
	Older	8.8 ± 6.1
$Torq_{100}$ (N·m)	Young	35.3 ± 14.4*
	Older	26.6 ± 14.5
$Torq_{200}$ (N·m)	Young	80.4 ± 22.1*
	Older	55.9 ± 17.6
CSA (cm^2)	Young	12.6 ± 2.8
	Older	13.1 ± 3.4
EI (arbitrary unit)	Young	86.8 ± 7.5*
	Older	102.0 ± 9.5
PA (°)	Young	21.1 ± 3.2*
	Older	18.2 ± 3.4
FL (cm)	Young	5.5 ± 0.8
	Older	5.4 ± 1.1
$EMG_{0–50}$ (%PEMG)	Young	46.4 ± 30.6
	Older	62.3 ± 36.5
$EMG_{100–150}$ (%PEMG)	Young	112.2 ± 31.3*
	Older	96.0 ± 20.8
$EMG_{150–200}$ (%PEMG)	Young	96.6 ± 23.7*
	Older	78.8 ± 21.6

PEMG, peak electromyographic amplitude.
*$P \leq 0.05$, significant difference between the young and the older men.

however, no relationships were observed between $a\text{RTD}_{0–50}$ and any of the other variables ($r = -0.11$ to 0.21, $P = 0.144–0.788$) (Table 2). For $a\text{RTD}_{100–200}$, there were significant relationships with PT, EI, PA, $\text{EMG}_{100–150}$, and $\text{EMG}_{150–200}$ ($r = -0.39$ to 0.66, $P = 0.001–0.016$); however, no relationships were observed between $a\text{RTD}_{100–200}$ and CSA, FL, and $\text{EMG}_{0–50}$ ($r = -0.23$ to 0.11, $P = 0.101–0.985$) (Table 2).

For the normalized RTD variables, the young men exhibited greater $n\text{RTD}_{100–200}$ ($P = 0.034$) values than the older men; however, no difference was observed for $n\text{RTD}_{0–50}$ ($P = 0.914$) between groups (Fig. 3B). For $n\text{RTD}_{0–50}$,

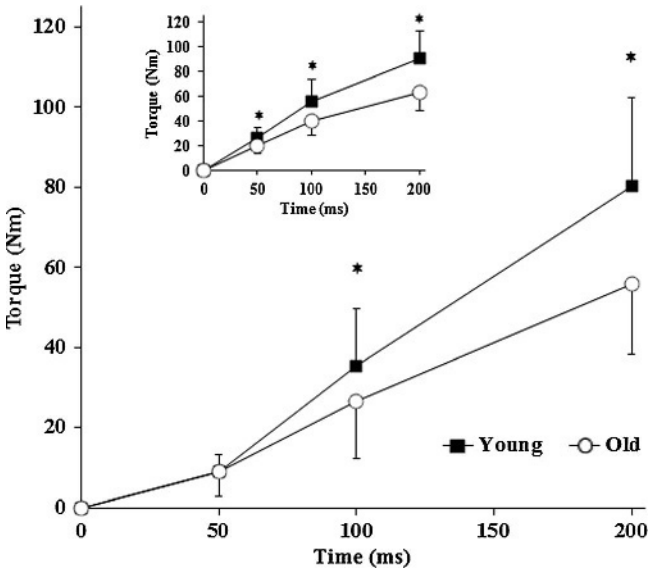

FIGURE 2—Isometric torque at 50, 100, and 200 ms for the young and older men during rapid plantarflexion isometric contractions determined by manual onset. The isometric torque values determined by the automated onset technique (4 N·m) are displayed in the upper left graph. *$P \leq 0.05$, significant difference between the young and the older men. Data are presented as mean ± SD.

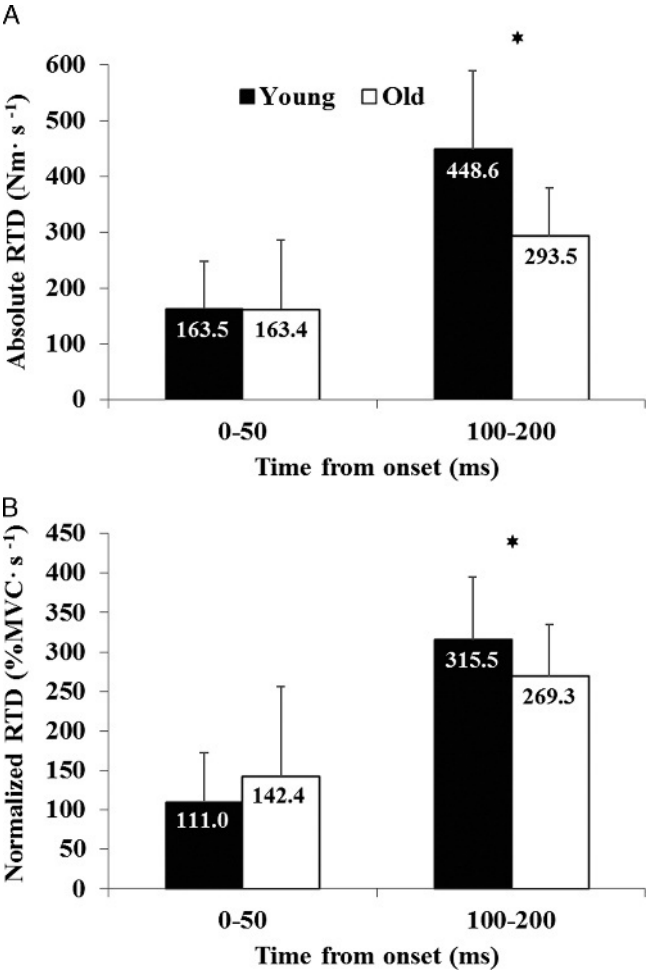

FIGURE 3—Absolute (A) and normalized (B) RTD for 0–50 and 100–200 ms for the young and older men during rapid plantarflexion isometric contractions. *$P \leq 0.05$, significant difference between the young and the older men. Data are presented as mean ± SD.

there was a significant relationship with $\text{EMG}_{0–50}$ ($r = 0.52$, $P < 0.001$); however, no relationships were observed between $n\text{RTD}_{0–50}$ and any of the other variables ($r = -0.11$ to 0.21, $P = 0.110–0.985$) (Table 2). For $n\text{RTD}_{100–200}$, there were significant relationships with EI, $\text{EMG}_{100–150}$, and $\text{EMG}_{150–200}$ ($r = -0.28$ to 0.59, $P = 0.001–0.044$); however, no relationships were observed between $n\text{RTD}_{100–200}$ and CSA, PA, FL, and $\text{EMG}_{0–50}$ ($r = -0.21$ to 0.16, $P = 0.147–0.767$) (Table 2).

TABLE 2. Bivariate correlation coefficients between maximal strength, muscle architecture, and muscle activation variables and absolute and normalized RTD variables.

	Absolute		Normalized	
Time (ms):	0–50	100–200	0–50	100–200
Maximal strength				
PT	0.21	0.66*	—	—
Muscle architecture variables				
CSA	−0.04	0.11	−0.17	−0.11
EI	−0.09	−0.39*	−0.00	−0.28*
PA	−0.11	0.34*	−0.23	0.16
FL	0.20	0.00	0.08	−0.21
Muscle activation variables				
$EMG_{0–50}$	0.44*	−0.23	0.52*	−0.04
$EMG_{100–150}$	0.08	0.54*	0.13	0.59*
$EMG_{150–200}$	0.14	0.48*	0.09	0.49*

*$P \leq 0.05$, significant difference between the young and the older men.

For the muscle factors, the older men exhibited higher EI (poorer muscle quality) values compared with the young men ($P < 0.001$). The young men exhibited greater PA compared with the older men ($P = 0.004$), and no differences were observed for CSA and FL ($P = 0.644$–0.803) between groups (Table 1). For the muscle activation variables, the young men exhibited greater $EMG_{100–150}$ and $EMG_{150–200}$ ($P = 0.009$–0.046) values than the older men. No difference was observed for $EMG_{0–50}$ ($P = 0.097$) between groups (Table 1). In addition, there was no difference in the passive baseline torque values between groups (young = 13.25 ± 2.90 N·m, older = 11.63 ± 3.82 N·m, $P = 0.091$). To determine whether the baseline passive torque values influence RTD, an analysis of covariance was performed on all RTD variables (with passive torque as the covariate), and the analyses resulted in similar findings for all RTD variables.

DISCUSSION

The primary findings of the present study indicated that later (100–200 ms), but not early (0–50 ms), rates of torque development were significantly lower in the older men when compared with young men, both before and after normalizing to PT (Figs. 2 and 3). The older men also had higher EI values, lower PA, and lower later EMG amplitude values when compared with the young men, whereas muscle CSA, FL, and early EMG amplitude values were not significantly different between groups. Interestingly, lower $a\text{RTD}_{100–200}$ values were related to higher EI, smaller PA, and lower later EMG amplitude values; however, later RTD values were no longer related to PA after normalizing to PT. These findings add to previous studies (1,5,12,16,18,20,26,37–39) that have investigated specific mechanisms contributing to the age-related reductions in rapid strength.

Several previous studies have demonstrated that rapid force/torque capabilities are significantly reduced in older adults when compared with younger adults (5,12,16,18,20,26,37,39). Our findings, specifically for the later rapid torque characteristics, are consistent with these previous studies. However, few studies have examined the difference in specific time intervals of rapid torque development (i.e., 30, 50, 100, and 200 ms) between young and older individuals (18,26,38,39). The majority of these studies were specific to the knee extensors and flexors (18,26,39), with only one study examining the plantarflexors (38). Although the aforementioned studies (18,26,38,39) used varied time specific intervals between 100 and 200 ms, each was significantly reduced (30.6%–47.1%) in the older group. Our findings show a similar reduction (34.6%) in $a\text{RTD}_{100–200}$ between young and older men. In addition, all of these studies demonstrated that $a\text{RTD}_{0–50}$ was significantly reduced in the older group for the knee extensors (22.0%–67.6%) (18,26,39). Thompson et al. (38) who also examined the plantarflexors found a 43.0% decline in $a\text{RFD}_{0–50}$ in older men. Our findings, which indicate that there is no difference in $a\text{RTD}_{0–50}$ values between groups, may differ from these previous studies because of the similar exercise habits between groups and/or the onset detection method used. In contrast to these studies, which all used an automatic absolute detection method for the determination of the onset of contraction (3–7.5 N·m for the knee extensors; 4 N for the plantarflexors) (18,26,38,39), the present study used a systematic manual onset detection method as suggested in a recent review by Maffiuletti et al. (25). When comparing our plantarflexor values to Thompson et al. (38), our $a\text{RTD}_{100–200}$ values are more similar to the earlier time point values by these authors, suggesting that the manual onset may occur earlier than automated methods. When our data were analyzed using the automated method (4 N·m onset), Torq_{50} was significantly different between groups (young = 25.81 ± 8.35 N·m, older = 20.13 ± 7.07 N·m, $P = 0.015$; Fig. 2), reflecting the results of the above-mentioned studies (18,26,38,39). As shown in Fig. 2, the Torq_{50} manual onset values are lower than our Torq_{50} automated onset values, suggesting that they occurred earlier.

Our normalized RTD variables demonstrated similar results, showing that RTD (independent of PT) was lower in the older men compared with the younger men in the late interval (100–200 ms), but not the early interval (0–50 ms) (Fig. 3). Previous studies have also shown a decrease in normalized rapid force/torque production (20,21,38), albeit these studies also found a decrease in the early time interval (0–50 ms) (21,38). Alternatively, there is evidence that normalized RTD/RFD is preserved in older adults, specifically in the plantarflexors (37), dorsiflexors (37), and knee extensors (18,39) and flexors (39). Previous authors (1,2) have suggested that normalized RTD values are useful to examine physiological characteristics such as motor unit recruitment and frequency, fiber type, muscle architecture, or tendon stiffness. The discrepancies between our findings and these previous studies may be due to 1) the onset identification technique as discussed earlier, 2) the time intervals examined (early, late, or peak RTD), and/or 3) a muscle specific difference, where the distal muscles (i.e., plantarflexors) may undergo relatively greater motor unit remodeling when compared with more proximal muscles (i.e., quadriceps) (23).

Muscle activation has been suggested to be an important factor influencing RTD (11,20). This supports our results, which indicated that there was lower muscle activation at the later time intervals ($EMG_{100–150}$ and $EMG_{150–200}$) in the older men as compared with the younger men, whereas there was no difference during the early time interval ($EMG_{0–50}$). Furthermore, our absolute and normalized RTD variables were significantly correlated ($r = 0.44$–0.59) to their respective EMG time intervals (Table 2). These findings are similar to those reported by Klass et al. (20) who reported surface EMG activity was similar between young and older adults at early time intervals (25 and 50 ms) but differed at a later time interval (~75 ms). The authors (20) also reported that the average instantaneous discharge frequency of single motor units were reduced and progressively decreased to a greater extent in the older adults. These findings and the results of the current study may suggest that age-related reductions in RTD

may be influenced by reductions in the maximal discharge frequency of motor units and their ability to sustain these high rates after the onset of activation.

In the present study, MG CSA was not different between groups (Table 1) or related to the absolute and normalized RTD variables (Table 2). Previous authors have reported differences in muscle size between young and older adults (24,36), whereas other authors (19,38) have shown that there are no differences between groups. Potential differences between our data and these studies may be due to the differences in physical activity status (19) and/or muscle-specific differences (22), where the MG may not experience similar age-related changes in size when compared with the vastus lateralis because of the plantarflexors greater usage during locomotor activities (22). Previously, CSA has been shown to be positively correlated with maximal strength (27) and has been suggested to be related to later RTD time intervals that are closer to peak strength (2). However, our data demonstrated that CSA was not related to early or late RTD but was significantly correlated to PT ($r = 0.332$, $P = 0.016$). Despite no differences in MG CSA, MG EI was significantly different between groups, which may suggest that older adults had poorer muscle quality. This has been reported by previous authors (19) who found that young and older adults had similar total CSA of the tibialis anterior, but the younger adults had a greater amount of contractile tissue and a lower amount of noncontractile tissue within the given CSA when compared with the older adults. Furthermore, MG EI was related to the later absolute and normalized RTD variables (Table 2). These findings may suggest that the age-related changes in rapid torque characteristics are influenced by alterations in muscle quality, whereby older adults experience increases in intramuscular fat and/or connective tissue (10,15,23,24). Similar to the findings in this study, muscle quality in older adults has also been correlated with RTD in previous studies (33,41). Wilhelm et al. (41) found that the EI values from all four quadriceps muscles were related to later RTD time intervals (200 ms), but only the EI of the vastus intermedius was associated with early RTD (50 ms) in older adults. Rech et al. (33) also found significant correlations between the EI of all four quadriceps and the later RTD time intervals (100–300 ms) in older women. The reduction in muscle quality (increase in EI) in the older men may be indicative of the age-related changes in motor unit remodeling resulting in lower relative type II muscle fiber area and increases in fat and fibrous tissue (23,24). Rahemi et al. (32) have suggested that the age-related increase in intramuscular fat may resist muscle fiber shortening and transverse bulging of the muscle resulting in reductions in force production. However, it is important to note that the present study did not measure muscle size and architecture during the contractions. Future studies should consider examining these variables using high frame rate US.

A unique aspect of the current study was the additional examination of PA and FL on age-related reductions in RTD. The older men had smaller PA compared with the young men (13.7%), whereas there were no significant differences in FL between the groups. Our findings are similar to previous studies (29,35) who reported smaller (13.2%–14%) MG PA values in older men. However, Stenroth et al. (36) did not find differences in PA between young and older adults. Studies (29,36) regarding FL measurements have normalized values (FL divided by limb length) to account for differences in stature between the young and the older groups; however, the older and younger men in the present study were similar in stature. Kubo et al. (22) found no significant differences in FL, whereas Stenroth et al. (36) and Narici et al. (29) reported significant differences between the young and the older groups. Inconsistencies in FL findings may be due to the measurement method used, as previous authors have estimated FL (22,29), whereas the present study obtained FL measurements using panoramic imaging. In addition, there was a significant relationship between $a\text{RTD}_{100–200}$ and PA; however, there was no relationship between FL and RTD variables (Table 2). Interestingly, $n\text{RTD}_{100–200}$ was not related to PA. This suggests that reductions in PA may result in a smaller physiological CSA (for a given volume of muscle), which may consequently lead to a smaller absolute RTD, especially during later time intervals (25). It has also been hypothesized that reductions in PA may reduce fiber rotation and origin-to-insertion speed resulting in a reduction in RTD (25). Lastly, the present study did not investigate tendon stiffness. The age-related decrease in tensile stiffness of the in-series tendon may also play a role in RTD as Bojsen-Møller et al. (7) noted that tendon stiffness may account for up to 30% of the variance in RTD. However, it is possible that the baseline passive torque values may differ between groups which may have influenced the early RTD variables. We found no difference in the baseline passive torque values between groups and these did not influence the early or late RTD variables.

In summary, absolute and normalized plantarflexion RTD was reduced in healthy older men during the late interval of contraction (100–200 ms) when compared with the younger men. Our findings indicated that the lack of differences in the early interval (0–50 ms) of contraction may suggest that older men initially have similar muscle activation to younger men but are unable to sustain the same rates of muscle activation during the late intervals of contraction (20). Furthermore, additional factors such as poor muscle quality and smaller PA appear to also play a role in the age-associated changes of RTD. It has been speculated that these changes may alter muscle fiber shortening and/or fiber rotation (25,32), which may influence the later interval of RTD. These findings add to our understanding of the importance of various neuromuscular factors that influence the age-related reductions in RTD, which has been shown to significantly influence function and performance in older adults (6,16,28,31).

This work was not supported by funding. The authors do not have any conflicts of interest to disclose.

The authors declare that the results of the study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation and do not constitute endorsement by the American College of Sports Medicine.

REFERENCES

1. Aagaard P, Magnusson PS, Larsson B, Kjaer M, Krustrup P. Mechanical muscle function, morphology, and fiber type in lifelong trained elderly. *Med Sci Sports Exerc*. 2007;39(11):1989–96.
2. Aagaard P, Simonsen EB, Andersen JL, Magnusson P, Dyhre-Poulsen P. Increased rate of force development and neural drive of human skeletal muscle following resistance training. *J Appl Physiol (1985)*. 2002;93(4):1318–26.
3. Andersen LL, Aagaard P. Influence of maximal muscle strength and intrinsic muscle contractile properties on contractile rate of force development. *Eur J Appl Physiol*. 2006;96(1):46–52.
4. Ayoung-Chee P, McIntyre L, Ebel BE, Mack CD, McCormick W, Maier RV. Long-term outcomes of ground-level falls in the elderly. *J Trauma Acute Care Surg*. 2014;76(2):498–503; discussion.
5. Bemben MG, Massey BH, Bemben DA, Misner JE, Boileau RA. Isometric muscle force production as a function of age in healthy 20- to 74-yr-old men. *Med Sci Sports Exerc*. 1991;23(11):1302–10.
6. Bento PC, Pereira G, Ugrinowitsch C, Rodacki AL. Peak torque and rate of torque development in elderly with and without fall history. *Clin Biomech (Bristol, Avon)*. 2010;25(5):450–4.
7. Bojsen-Møller J, Magnusson SP, Rasmussen LR, Kjaer M, Aagaard P. Muscle performance during maximal isometric and dynamic contractions is influenced by the stiffness of the tendinous structures. *J Appl Physiol (1985)*. 2005;99(3):986–94.
8. Cerniglia LM, Delmonico MJ, Lindle R, Hurley BF, Rogers MA. Effects of acute supine rest on mid-thigh cross-sectional area as measured by computed tomography. *Clin Physiol Funct Imaging*. 2007;27(4):249–53.
9. Clark DJ, Manini TM, Fielding RA, Patten C. Neuromuscular determinants of maximum walking speed in well-functioning older adults. *Exp Gerontol*. 2013;48(3):358–63.
10. Csapo R, Malis V, Sinha U, Du J, Sinha S. Age-associated differences in triceps surae muscle composition and strength—an MRI-based cross-sectional comparison of contractile, adipose and connective tissue. *BMC Musculoskelet Disord*. 2014;15:209.
11. Del Balso C, Cafarelli E. Adaptations in the activation of human skeletal muscle induced by short-term isometric resistance training. *J Appl Physiol (1985)*. 2007;103:402–11.
12. Ditroilo M, Forte R, Benelli P, Gambarara D, De Vito G. Effects of age and limb dominance on upper and lower limb muscle function in healthy males and females aged 40–80 years. *J Sports Sci*. 2010;28(6):667–77.
13. Folland JP, Buckthorpe MW, Hannah R. Human capacity for explosive force production: neural and contractile determinants. *Scand J Med Sci Sports*. 2014;24(6):894–906.
14. Frontera WR, Hughes VA, Fielding RA, Fiatarone MA, Evans WJ, Roubenoff R. Aging of skeletal muscle: a 12-yr longitudinal study. *J Appl Physiol (1985)*. 2000;88(4):1321–6.
15. Goodpaster BH, Park SW, Harris TB, et al. The loss of skeletal muscle strength, mass, and quality in older adults: the health, aging and body composition study. *J Gerontol A Biol Sci Med Sci*. 2006;61(10):1059–64.
16. Izquierdo M, Aguado X, Gonzalez R, Lopez JL, Hakkinen K. Maximal and explosive force production capacity and balance performance in men of different ages. *Eur J Appl Physiol Occup Physiol*. 1999;79(3):260–7.
17. Janssen I, Shepard DS, Katzmarzyk PT, Roubenoff R. The healthcare costs of sarcopenia in the United States. *J Am Geriatr Soc*. 2004;52(1):80–5.
18. Jenkins ND, Buckner SL, Cochrane KC, et al. Age-related differences in rates of torque development and rise in EMG are eliminated by normalization. *Exp Gerontol*. 2014;57:18–28.
19. Kent-Braun JA, Ng AV, Young K. Skeletal muscle contractile and noncontractile components in young and older women and men. *J Appl Physiol (1985)*. 2000;88(2):662–8.
20. Klass M, Baudry S, Duchateau J. Age-related decline in rate of torque development is accompanied by lower maximal motor unit discharge frequency during fast contractions. *J Appl Physiol (1985)*. 2008;104(3):739–46.
21. Korhonen MT, Cristea A, Alen M, et al. Aging, muscle fiber type, and contractile function in sprint-trained athletes. *J Appl Physiol (1985)*. 2006;101(3):906–17.
22. Kubo K, Kanehisa H, Azuma K, et al. Muscle architectural characteristics in women aged 20–79 years. *Med Sci Sports Exerc*. 2003;35(1):39–44.
23. Lexell J. Evidence for nervous system degeneration with advancing age. *J Nutr*. 1997;127(5 Suppl):1011S–3.
24. Lexell J, Taylor CC, Sjostrom M. What is the cause of the ageing atrophy? Total number, size and proportion of different fiber types studied in whole vastus lateralis muscle from 15- to 83-year-old men. *J Neurol Sci*. 1988;84(2–3):275–94.
25. Maffiuletti NA, Aagaard P, Blazevich AJ, Folland J, Tillin N, Duchateau J. Rate of force development: physiological and methodological considerations. *Eur J Appl Physiol*. 2016;116(6):1091–116.
26. Mau-Moeller A, Behrens M, Lindner T, Bader R, Bruhn S. Age-related changes in neuromuscular function of the quadriceps muscle in physically active adults. *J Electromyogr Kinesiol*. 2013;23(3):640–8.
27. Maughan RJ. Relationship between muscle strength and muscle cross-sectional area. Implications for training. *Sports Med*. 1984;1(4):263–9.
28. Morcelli MH, LaRoche DP, Crozara LF, et al. Neuromuscular performance in the hip joint of elderly fallers and non-fallers. *Aging Clin Exp Res*. 2016;28(3):443–50.
29. Narici MV, Maganaris CN, Reeves ND, Capodaglio P. Effect of aging on human muscle architecture. *J Appl Physiol (1985)*. 2003;95(6):2229–34.
30. Ortman JM, Velkoff VA, Hogan H. An aging nation: the older population in the United States. *Current Population Reports Series P-25, Population Estimates and Projections*. 2014;1140:1–28.
31. Palmer TB, Thiele RM, Williams KB, et al. The identification of fall history using maximal and rapid isometric torque characteristics of the hip extensors in healthy, recreationally active elderly females: a preliminary investigation. *Aging Clin Exp Res*. 2015;27(4):431–8.
32. Rahemi H, Nigam N, Wakeling JM. The effect of intramuscular fat on skeletal muscle mechanics: implications for the elderly and obese. *J R Soc Interface*. 2015;12(109):20150365.
33. Rech A, Radaelli R, Goltz FR, da Rosa LH, Schneider CD, Pinto RS. Echo intensity is negatively associated with functional capacity in older women. *Age (Dordr)*. 2014;36(5):9708.
34. Rosenberg JG, Ryan ED, Sobolewski EJ, Scharville MJ, Thompson BJ, King GE. Reliability of panoramic ultrasound imaging to simultaneously examine muscle size and quality of the medial gastrocnemius. *Muscle Nerve*. 2014;49(5):736–40.
35. Ryan ED, Rosenberg JG, Scharville MJ, Sobolewski EJ, Tweedell AJ, Kleinberg CR. Pennation angle does not influence the age-related differences in echo intensity of the medial gastrocnemius. *Ultrasound Med Biol*. 2015;41(2):618–21.
36. Stenroth L, Peltonen J, Cronin NJ, Sipilä S, Finni T. Age-related differences in Achilles tendon properties and triceps surae muscle

architecture in vivo. *J Appl Physiol (1985)*. 2012;113(10): 1537–44.
37. Thelen DG, Schultz AB, Alexander NB, Ashton-Miller JA. Effects of age on rapid ankle torque development. *J Gerontol A Biol Sci Med Sci*. 1996;51(5):M226–32.
38. Thompson BJ, Ryan ED, Herda TJ, Costa PB, Herda AA, Cramer JT. Age-related changes in the rate of muscle activation and rapid force characteristics. *Age (Dordr)*. 2014;36(2):839–49.
39. Thompson BJ, Ryan ED, Sobolewski EJ, Conchola EC, Cramer JT. Age related differences in maximal and rapid torque characteristics of the leg extensors and flexors in young, middle-aged and old men. *Exp Gerontol*. 2013;48(2):277–82.
40. Tinetti ME, Speechley M, Ginter SF. Risk factors for falls among elderly persons living in the community. *N Engl J Med*. 1988;319(26):1701–7.
41. Wilhelm EN, Rech A, Minozzo F, Radaelli R, Botton CE, Pinto RS. Relationship between quadriceps femoris echo intensity, muscle power, and functional capacity of older men. *Age (Dordr)*. 2014;36(3):9625.
42. Young HJ, Jenkins NT, Zhao Q, McCully KK. Measurement of intramuscular fat by muscle echo intensity. *Muscle Nerve*. 2015;52(6):963–71.