Age differences in knee extension power, contractile velocity, and fatigability

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Age-related sarcopenia leads to accelerated reductions in skeletal muscle mass and strength beyond the fifth decade at a rate of 1–2% per year (1, 44). Because muscle strength is an important determinant of functional performance for tasks such as standing from a chair and climbing stairs (4), sarcopenic atrophy is considered a primary risk factor for dependence and disability in older adults (13, 18). The majority of age-related weakness appears to be directly attributable to reduced muscle mass; however, older muscle may also be weak for its size (44). To compare differences between groups, strength or power is often normalized to muscle size (15), yielding “specific” strength or power. Any age-related losses in this ratio in vivo may be influenced by numerous anatomic, mechanical, or neural factors (12, 15). Studies of age differences in the ratio of strength to whole muscle size have yielded equivocal results with evidence of both reduced (9, 29, 42) and unchanged specific force associated with age (15, 21). Differences in estimates of muscle size may account for these inconsistent findings.

Age-related declines in muscle power occur at a greater rate (3–4% per year) than strength (1–2% per year) (5, 6, 36). Muscle power is an important factor for mobility and ambulation, particularly in tasks like walking, stair climbing, and standing from a chair (5, 7, 14, 44). Losses of muscle power may also increase risk of falls (37). Age-related reductions in specific power have been recently reported (12). Declines in specific power may be due to an age-related loss of maximal contractile velocity as velocity has been found to be the critical determinant of whole muscle power in older women (11). Single muscle fibers of men and women have also shown an age-related slowing of maximum contractile velocity (23). We are unaware of published studies examining age differences in whole muscle contractile velocity and power in older men.

Muscle fatigability is another important component of performance. Fatigue is typically measured as a loss of force during repeated or continuous activation (38). Various skeletal muscle show no age-related differences in fatigability, including the quadriceps muscles or tibialis anterior (8, 28, 38, 40), whereas the triceps surae exhibit increased susceptibility to fatigue (9) and the adductor pollicis exhibit age-related resistance to fatigue (31). Despite mixed findings on age-related fatigability, muscular fatigue of the quadriceps femoris muscle has been related to increased falls risk in older women (34). Because power may be more indicative of falls risk than strength (37), it would be pertinent to evaluate fatigability based on the rate of decline in power with repetitive contractions rather than a drop in force. We are not aware of any published findings on age differences in muscle fatigability using power as the criterion measure.

The purpose of this study was to examine age and gender differences in knee extensor strength, power, and fatigue by using open- and closed-chain testing procedures. We employed both open- and closed-chain testing because open-chain testing is heavily used in research and rehabilitation evaluation, whereas closed-chain testing is considered more functional and closely simulates daily activities (2, 39). Performance data were adjusted for thigh lean mass (TLM) to estimate specific strength and specific power. We tested the hypothesis that

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specific strength would not differ by age, whereas age differences in specific power and fatigue would remain consequent to blunted maximal contractile velocity in older adults. We also tested for potential gender differences in each of these muscle performance variables.

METHODS

Subjects. Fifty-two adults were recruited from the Birmingham, Alabama metropolitan area into two age groups. Age ranges were 60–75 yr for the older group (13 men, 11 women) and 20–35 yr for the younger group (12 men, 16 women). All subjects completed a detailed health history appraisal, and all older subjects passed a comprehensive physical examination conducted by a geriatrician and a diagnostic stress test monitored by a cardiologist. Subjects were free of any musculoskeletal or other disorders that might have affected their ability to complete testing for the study. Subjects were not obese (body mass index < 30) and had no leg resistance training experience within the past 5 yr. None of the subjects were being treated with exogenous testosterone or other pharmacological interventions known to influence muscle mass. The study was approved by the Institutional Review Boards of both the University of Alabama at Birmingham and the Birmingham Veterans Affairs Medical Center. Written, informed consent was obtained before participation.

Body composition. Thigh lean mass, total body lean mass, and body fat percent were determined by dual-energy X-ray absorptiometry (DEXA) using a Lunar Prodigy (model 8743, GE Lunar, Madison, WI). Analyses were conducted according to manufacturer’s instructions using enCORE 2002 software (version 6.10.029). For age and gender comparisons, strength and power results were adjusted for TLM to yield estimates of specific strength and specific power. The TLM-to-total body lean mass ratio was utilized as an index of preferential lower limb atrophy in older adults. This ratio is defined as relative TLM.

Dynamic strength. Before one-repetition maximum (1 RM) strength assessment, subjects attended two familiarization sessions within 5 days defined as follows. 1) The first session was an introduction and familiarization to the bilateral movements tested (squat, leg press, knee extension). During this session, each machine was configured for a given subject and the configuration settings were recorded. Subjects then completed several full range-of-motion repetitions with minimal loads. 2) In the second familiarization session, subjects performed several more minimal load contractions and completed practice 1-RM strength tests. This session dually served to familiarize subjects with 1-RM testing procedures and to instruct and encourage subjects to exert maximal voluntary effort.

Subjects returned to the laboratory 2–3 days later for 1-RM assessments in the three bilateral movements using established methods. Briefly, 1-RM strength was assessed for closed-chain movements first (squat, leg press), followed by the open-chain knee extension. Before testing, subjects warmed up for 5 min on either a treadmill or cycle ergometer set at a low intensity, and, for each movement, the test protocol was preceded by a set of 8–12 repetitions with a light load. Thereafter, attempts of one repetition with progressively increasing load were performed. Attempts were separated by 90–120 s of rest intervals. 1 RM was defined as the highest load lifted through a full range of motion before two failed attempts at a given load. Tests were performed using free weights (i.e., squat) and resistance exercise stations (Body Masters Circuit Master, Body Masters Sports Industries, Rayne, LA). To accurately compare 1-RM data by age and gender, we determined the actual loads lifted at each station using regression-curve fitting procedures as follows. The actual load equivalent to each increment of the weight stack was determined using a load cell (Omega S Beam Stainless Load Cell, Omega Engineering, Stamford, CT) that we had previously calibrated across a load range up to 2,255 N. Resultant data for each station were analyzed using regression- and curve-fitting software (SigmaPlot 2001 version 7.0, SPSS, Chicago, IL), leading to a load-conversion formula for each station (regression R² values ranged from 0.993 to 0.999). Thus, for each 1-RM test, the weight stack increment was converted to load in newtons for statistical analysis.

Load-power and load-velocity relationships. Peak concentric knee extension power was determined across a load spectrum that included five submaximal loads relative to maximum isometric voluntary contraction (MVC) force. Bilateral MVC was measured first while the subject was seated on the constant-load dynamometer using a calibrated load cell (attached in lieu of the weight stack cable). Knee angle was held constant at ~0.942 rad (54 ± 1°) of flexion as measured by electrogoniometry (model SGI150, Biometrics, Gwent, UK). Subjects were secured to the dynamometer by using 10-cm-wide Velcro straps across the torso and hips, and a 5-cm Velcro strap secured the ankles to the lower leg pads. MVC was accepted as the peak force obtained in two trials. Contractions were 5 s in duration and were separated by 60-s rest intervals. Weight stack loads providing resistance forces equal to 20, 30, 40, 50, and 60% of MVC force were determined by using a load cell-derived force regression equation for the dynamometer’s weight stack (R² = 0.993). Subjects completed three full range of motion knee extension contractions with each load. During each repetition, the concentric phase was performed as rapidly as possible, while the eccentric phase was controlled (~2 s). Test order for the five loads was randomly assigned within subjects before testing. Knee electrogoniometry was used to measure the time interval from 0.873 rad (50°) to 0.349 rad (20°) of knee flexion during the concentric phase of each repetition (knee angle sampling rate = 500 Hz). Angular velocity (rad/s) was then computed. Angular displacement of 0.524 rad was converted to linear displacement on the basis of lever arm length in meters. Work in joules equalled linear displacement (m) × external force (N). Power in watts was computed as work (J) ÷ time in seconds. At each load, the repetition yielding peak power was used for analysis. Power results were adjusted for TLM to yield specific power.

Dynamic fatigue. Fatigability was tested during both open-chain (knee extension) and closed-chain (sit-to-stand) movements. Open-chain fatigue was tested during bilateral knee extension using an external load equal to 40% of MVC force. Ten repetitions were performed in succession on the open-chain dynamometer’s weight stack. As previously, each concentric phase was performed as rapidly as possible followed by a controlled eccentric phase. Velocity, work, and power were computed for each repetition as described above. In these multiple-repetition open-chain tests, we have found that peak power often occurs beyond the first repetition (e.g., in repetition 2 or 3). We therefore define fatigue as a decline in power production from the repetition eliciting peak power to the 10th repetition.

Closed-chain power and fatigue were tested during a 10-repetition sit-to-stand test. Subjects were instructed to stand (concentric) as rapidly as possible (without jumping) from a seated position. Jumping was not allowed for two reasons: 1) to standardize foot position across all repetitions and 2) as a safety precaution. Because subjects were instructed not to jump, force and velocity data were extracted early in the standing movement (before deceleration). Return to the seated position (eccentric) occurred more slowly. Position of the upper limbs was standardized (arms across chest), and subjects were not allowed any assistance during the movements. A simulated park bench of standard height (45 cm) straddled left and right foot force plates (RoughDeck model 1.66 × 4.5 sp-1K, Rice Lake Weighing Systems, Rice Lake, WI). We standardized bench height rather than starting knee angle (for individuals of different statures) to maximize the applicability of the results. Forces were digitized at 500 Hz, and knee angle electrogoniometry was digitized at 500 Hz using Biometrics Limited software (version 2.0). For each standing (concentric) repetition, peak force under the right foot and right knee angle at peak force were determined during the first 0.524 rad (30°) of knee extension from the seated position. Peak force was found to typically occur within the first 0.262 rad (15°) of knee extension.
Table 1. Descriptive characteristics of each group

<table>
<thead>
<tr>
<th></th>
<th>Young Women (n = 16)</th>
<th>Young Men (n = 12)</th>
<th>Older Women (n = 11)</th>
<th>Older Men (n = 13)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age, yr</td>
<td>27.1 ± 0.9</td>
<td>26.8 ± 1.1</td>
<td>62.9 ± 1.0</td>
<td>64.2 ± 1.3</td>
</tr>
<tr>
<td>Height, cm</td>
<td>164.0 ± 1.9</td>
<td>181.1 ± 1.7</td>
<td>160.2 ± 2.2</td>
<td>174.1 ± 3.2</td>
</tr>
<tr>
<td>Weight, kg</td>
<td>75.3 ± 8.4</td>
<td>80.4 ± 3.2</td>
<td>62.9 ± 3.7</td>
<td>84.1 ± 2.8</td>
</tr>
<tr>
<td>Body fat, %</td>
<td>37.0 ± 1.3</td>
<td>20.0 ± 2.4</td>
<td>39.6 ± 1.9</td>
<td>33.3 ± 1.7</td>
</tr>
<tr>
<td>Total LM, kg</td>
<td>38.7 ± 1.4</td>
<td>61.1 ± 1.1</td>
<td>35.9 ± 1.5</td>
<td>53.9 ± 2.3</td>
</tr>
<tr>
<td>TLM, kg</td>
<td>93.0 ± 4.4</td>
<td>15.4 ± 0.4</td>
<td>7.7 ± 0.4</td>
<td>12.8 ± 0.8</td>
</tr>
<tr>
<td>Relative TLM, %</td>
<td>23.9 ± 0.4</td>
<td>25.1 ± 0.3</td>
<td>21.5 ± 0.5</td>
<td>23.6 ± 0.8</td>
</tr>
</tbody>
</table>

Values are means ± SE; n, no. of subjects. LM, lean mass; TLM, thigh lean mass. Relative TLM (%) = TLM/total LM × 100. †Main gender effect, P < 0.05. ‡Main age effect, P < 0.05. **Main gender interaction, P < 0.05.

Peak power per repetition was determined across a small window beginning with peak force and ending 0.087 rad (5°) later. Average total force (N) across this interval was determined by summing the average right foot and left foot forces. Because the interval selected was based on angular displacement about the knee, and because foot position was fixed, the lever arm in this case was the femur. We therefore measured right femur length (m) (right knee used for goniometry) by DEXA. Angular displacement was then converted to linear displacement in meters for computations of vertical work (J) performed. Peak power (W) per repetition was then determined from vertical work and the time of the 0.087 rad (5°) range of motion interval.

As in the open-chain tests of power and fatigue, we identified peak power for each subject as the highest concentric power output of this 10-repetition standing test. Differences in peak standing power were tested between groups. To test fatigability during repetitive standing movements, the repetition eliciting peak concentric power was compared with power production during the 10th repetition.

Relative muscle activation. In a standing task separate from the tests of power and fatigue described above, we used surface electromyography (EMG) to evaluate the magnitude of quadriceps neural activation (relative to maximum) required to stand from a chair. The methodologies applied in this test differ markedly from the methods used to test standing power. First, it was critically important to control the rate of rise from the bench as well as the rate of descent. We used an audible and visual metronome to standardize the cadence of each phase to 2 s, and several practice trials were performed before administration of the test. Second, the data of interest were analyzed at the knee angle equivalent to the knee angle during isometric MVC (peak power and force were not relevant because velocity was controlled). This was required to accurately interpret standing EMG data normalized to maximum EMG. Third, data were analyzed during both eccentric descent and concentric ascent.

Neural activity of the right knee extensors was measured for each of the three superficial quadriceps muscles [vastus medialis (VM), vastus lateralis (VL), and rectus femoris (RF)] by using preamplified, bipolar electrodes (20-mm interelectrode distance) (model SX230, Biometrics). EMG signals were digitized at 1,000 Hz and electrogrotnometer inputs at 500 Hz, and data from each input channel were analyzed simultaneously using Biometrics DataLINK PC software version 2.0.

Before this test, unilateral (right leg) MVC was measured during 5-s maximal contractions at a knee angle ~0.942 rad (54 ± 1°) using the same protocol described previously for bilateral MVC testing. Three MVCs were performed with 60-s rest intervals, and the trial yielding peak force was used for this analysis. A thorough explanation of the procedure was followed by multiple practice trials before recording of the the three MVCs. Each raw EMG signal was full-wave rectified and converted to its root mean square (RMS) by using a 100-ms sliding window. RMS (μV) was determined for each muscle across a 100-ms window centered on peak force. RMS values for VM, VL, and RF were then averaged within subjects, and this average was considered to be representative of maximal voluntary EMG amplitude.

One trial of the sit-to-stand test consisted of three repetitions. During both concentric and eccentric phases, RMS EMG for each muscle was determined across a 100-ms window centered on the specific knee angle (within subjects) at which unilateral MVC was assessed. Within repetition phase (concentric or eccentric), RMS values for VM, VL, and RF were averaged. RMS results within repetition phase were then averaged across all three repetitions. Results were normalized to maximum RMS EMG (during MVC) to yield indexes of relative muscle activation during the concentric or eccentric phase of a sit-to-stand task. Higher values indicate a greater muscle activation requirement. In other words, the task would be performed at a higher intensity (i.e., with more difficulty).

Statistical analysis. All statistical analyses were performed using STATISTICA 6.1 (StatSoft, Tulsa, OK). We used two-factor (age, gender) ANOVAs to detect group differences on 1) anthropometric measures (height, weight, percent body fat, lean mass, TLM), 2) strength (absolute and specific strength for knee extension, leg press, and squat), 3) power measures (peak and specific power, peak contractile velocity), and 4) relative muscle activation (sitting and standing). For load-power and load-velocity curves, two (age, gender) × five repeated-measures ANOVAs were used to detect differences in peak power, specific power, and peak contractile velocity across five loads (20–60% MVC). Open- and closed-chain fatigue tests were evaluated by 2 (age, gender) × 2 repeated measures (peak and 10th repetition) ANOVA. When appropriate, post hoc comparisons were conducted using Tukey’s HSD tests. All results are reported as means ± SE. Significance was set at P < 0.05.

Table 2. Absolute and specific strength of each group

<table>
<thead>
<tr>
<th></th>
<th>Young Women (n = 16)</th>
<th>Young Men (n = 12)</th>
<th>Older Women (n = 11)</th>
<th>Older Men (n = 13)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-RM strength, N</td>
<td></td>
<td></td>
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<tr>
<td>Knee extension*</td>
<td>376.6 ± 19.6</td>
<td>632.6 ± 26.5</td>
<td>266.8 ± 16.7</td>
<td>371.7 ± 24.5</td>
</tr>
<tr>
<td>Leg press*</td>
<td>872.8 ± 42.2</td>
<td>1,531.9 ± 51.0</td>
<td>725.7 ± 48.1</td>
<td>1,084.7 ± 47.1</td>
</tr>
<tr>
<td>Squat*</td>
<td>579.6 ± 25.5</td>
<td>938.5 ± 49.0</td>
<td>485.5 ± 24.5</td>
<td>665.9 ± 28.4</td>
</tr>
<tr>
<td>Specific strength measurements, 1-RM (N)/TLM (kg)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Knee extension*</td>
<td>41.2 ± 2.0</td>
<td>41.2 ± 1.0</td>
<td>35.3 ± 2.0</td>
<td>29.4 ± 2.0</td>
</tr>
<tr>
<td>Leg press*</td>
<td>94.2 ± 2.9</td>
<td>100.0 ± 2.9</td>
<td>96.1 ± 4.9</td>
<td>86.3 ± 2.9</td>
</tr>
<tr>
<td>Squat*</td>
<td>63.7 ± 2.9</td>
<td>61.8 ± 3.9</td>
<td>64.7 ± 2.0</td>
<td>53.9 ± 2.9</td>
</tr>
</tbody>
</table>

Values are means ± SE; n, no. of subjects. 1-RM strength, 1-repetition maximum strength. *Main age effect, P < 0.05. †Main gender effect, P < 0.05. **Age × gender interaction, P < 0.05.
height and weight with higher values for men \((P = 0.05)\). Of the older postmenopausal women, 6 of 11 were on estrogen replacement therapy. Percent body fat was lower in young men compared with all other groups \((P < 0.05)\). Age and gender effects for total lean mass and TLM \((P < 0.05)\) resulted from higher levels of total lean mass and TLM in men vs. women and young vs. older participants. Within gender, height did not differ by age; thus these age differences in lean mass and TLM are indicative of lean mass atrophy with aging. This apparent atrophy in older men led to no significant differences in lean mass \((P = 0.63)\) and TLM \((P = 0.20)\) between older men and young women despite a 10-cm difference in height. Relative TLM was significantly affected by both gender and age \((P < 0.05)\). Post hoc analysis revealed that these differences were primarily driven by the older women with a significantly lower ratio than young men and women \((P < 0.05)\).

### Dynamic strength

As shown in Table 2, significant age \(\times\) gender interactions exist for all absolute measures of strength using 1 RM \((P < 0.05)\). For knee extension, leg press, and squat, young men were substantially stronger \((41–137\%)\) than all other groups \((P < 0.05)\). After adjustment for TLM, age differences in specific strength for leg press and squat were no longer detected \((P = 0.11\) and 0.30, respectively); however, a main age effect for knee extension specific strength remained \((P < 0.05,\ Table 2)\). An age \(\times\) gender interaction was detected for leg press specific strength \((P < 0.05)\). Post hoc analyses revealed no significant differences between any two age or gender groups, but we noted a strong trend \((P = 0.07)\) toward greater leg press specific strength in young men compared with older men. A main gender effect for squat specific strength was driven by the relatively low value in older men \((P < 0.05)\), as post hoc analyses revealed no differences between young men and either female group \((young, P = 0.96; older, P = 0.90)\).

### Skeletal muscle power

Figure 1A displays the unadjusted knee extension load-power curve for each group across five loads. For all loads \((20, 30, 40, 50, \text{and} 60\%\) of MVC\), main effects of age and gender were noted with greater peak power in young participants and in men \((P < 0.05)\). Age and gender differences exist for the relative load eliciting peak absolute power \((young women, 50\%; young men, 60\%; older women, 60\%; \text{and} older men, 40\%)\), resulting in significant age \(\times\) gender interactions at 50 and 60% MVC. Adjustment for TLM did not negate age differences in knee extension power as shown in Fig. 1B. There were significant main age and gender effects at all loads \((20 – 60\%\) of MVC\) with the young group.
exhibiting greater specific power than the older group \( (P < 0.05) \) and men showing greater specific power than women \( (P < 0.05) \). Similar to absolute power, there were group differences in the load eliciting peak specific power (young women, 50%; young men, 60%; older women, 60%, older men, 40%), resulting in an age \( \times \) gender interaction at 60% MVC. With increasing load, older men lost power and were reduced to a level similar to older women by 50% \( (P = 0.57) \), and 60% \( (P = 0.99) \) MVC. This marked decline in specific power with increasing load was unique to older men, leading to no detectable difference in power generation within this group at 60 vs. 20% MVC \( (P = 0.10) \).

The load-velocity relationship is shown in Fig. 2. This curve displays the velocity component of each knee extension power value in Fig. 1, A and B. Age exerted a significant main effect on peak velocity at each load because shortening velocity was higher in young participants at all loads \( (P < 0.05) \). The expected decline in peak shortening velocity with increased load was revealed in all groups as a significant load effect \( (P < 0.05) \).

Peak power was also assessed during a closed-chain, explosive sit to stand movement (Fig. 3A peak repetition). There were main effects of age and gender on peak power during the sit-to-stand test because the young group generated more power than the older group \( (P < 0.05) \) and men generated more power than women \( (P < 0.05) \). Older men generated a striking 101% more power than older women, whereas the gender difference in young adults was only 22%. Examining the components of sit-to-stand showed a main gender effect \( (P < 0.05) \) on peak force (Fig. 3B). There was an age effect on velocity (Fig. 3C) similar to the results with knee extension peak velocity, because peak velocity was higher in the young group during the sit-to-stand movement than in the older group \( (P < 0.05) \).

**Skeletal muscle fatigue.** Figure 4 displays the fatigability results associated with 10 explosive repetitions of knee extension performed against an external load equivalent to 40% of MVC force. In this model, the pertinent dependent variable used to test fatigue was concentric velocity through a standardized range of motion. External load was constant within participants, and the lever arm for the external load was held constant both within and between subjects; thus power across repetitions was influenced only by contractile velocity. We found a main effect of repetition, indicating that contractile
velocity changed across repetitions ($P < 0.05$). After the first repetition, peak velocity tended to increase for three repetitions in young and two repetitions in the older subjects before beginning to decline. We detected an age × repetition interaction ($P < 0.05$) because maximum concentric velocity (relative to the peak repetition) declined 24% ($P < 0.05$) by the 10th contraction in the older group, whereas there was a trend but no significant decline in peak velocity detected in the young group ($P = 0.08$). This interaction, which indicates an impaired resistance to fatigue in the older adults, is illustrated in Fig. 5 by representative data from two individual subjects. Shown in Fig. 5 are the knee goniometer recordings from one young subject (A) and one older subject (B) during the knee extension fatigue test. The repetition eliciting peak velocity (3rd repetition for both young and older subject) is shown on the left, and the 10th repetition is shown on the right. On the basis of the time interval from 0.873 rad (50°) to 0.349 rad (20°) of knee extension, the young subject’s maximum concentric velocity dropped only slightly from a peak of 3.40 rad/s to 3.19 rad/s by the 10th repetition, whereas the older subject’s peak velocity of 3.03 rad/s fell substantially to 1.89 rad/s by the 10th repetition.

Muscle fatigability was also measured during an explosive 10-repetition sit-to-stand test (Fig. 3, A–C). Fatigue was assessed by testing for declines in concentric power in repetition 10 compared with the peak repetition while controlling for knee angle and range of motion. Each group reached peak power on the second repetition. Both components of power in this model, force and velocity, were free to vary, and thus each were analyzed. Main effects of repetition revealed significant declines in concentric power, force, and velocity ($P < 0.05$, all), indicating that the 10-repetition sit-to-stand movement was a valid test of fatigue. No age differences in the rates of decline were noted. The mean drop in power was 27% resulting from decreases in both force plate force (12%) and concentric velocity (20%).

**Sit-to-stand performance.** Neural activation relative to maximum voluntary muscle activation required for sitting and standing are displayed in Fig. 6. These ratios of task-specific EMG amplitude to EMG amplitude during maximum voluntary contraction were used as measures of relative muscle activation, with higher values indicating a higher intensity activity. During the sit-to-stand test (concentric contraction), there was a significant age effect with the older group requiring a greater proportion of maximum voluntary muscle activation (i.e., EMG amplitude) to complete the movement ($P < 0.05$). During the stand-to-sit test (eccentric contraction), we found both main age and main gender effects. Men required less relative muscle activation to sit than women ($P < 0.05$), and the young group required less relative muscle activation than the older group ($P < 0.05$).

**DISCUSSION**

Our primary finding was that deficits in concentric power persisted after adjustment for TLM as maximum contractile velocity was markedly lower in older subjects. Older adults were also less capable of sustaining maximum concentric velocity during repetitive contractions. The age-related differences in contractile strength were mainly attributable to apparent age-associated muscle atrophy.

Our sample showed age-related declines in absolute and specific muscle power across a wide range of preloads during knee extension contractions. These preloads were determined within subjects as relative percentages of each individual’s maximum isometric force; thus the power decrements in older adults were velocity dependent even at the same relative preloads between age groups. Both men and women exhibited reduced specific power with age. The older males reached peak power during knee extension at a lighter load (40% MVC) compared with all other groups and were the only group to show significantly decreased power with incremental load increases above 40% MVC. Consequently, across a preload range of 20–60% MVC, we detected the descending portion of the load-power curve only in older males. The load-dependent decline in specific power among older men was particularly revealing at 60% MVC, as power in older men dropped to the level of older women. The marked drop in power from 40 to 50% MVC was seen in 10 of 13 older men, and 12 of 13 older men lost power from 50% to 60% MVC. This is a novel finding in older men that warrants further investigation.

As displayed on the load-power relationship (Fig. 1), there was a trend for older men to decline in maximum concentric power at a faster rate than the other groups when working against loads greater than 40% MVC. With external loads fixed, a loss of contractile velocity is the obvious determinant of the decline in power seen at 50–60% MVC in older men. Others have described contractile velocity as a critical determinant of power in older women (11). The downward shift in the load-velocity curve (Fig. 2) is similar to the shift in the force-velocity curve previously seen in older adults (40). This finding is consistent with the age-related slowing of muscle, which, among other factors, may be caused by selective atrophy of type II myofibers with aging (27), impaired ATPase
activity within a given myofiber type (25), and/or excitation-contraction “uncoupling” (32). Irrespective of the underlying mechanism(s), our data clearly show that age-related losses of muscle power are primarily driven by impairments in explosive contractile velocity.

There was an age-related effect on contractile velocity during the closed-chain movement of standing from a chair as rapidly as possible, similar to the findings on contractile velocity during knee extension. This age difference was clearly driven by older women, because they had the slowest contractile velocity during the sit-to-stand movement, 32% lower than their young female counterparts. Although not significant, this velocity difference was associated with a 44% difference in peak power between older and young women. Because load (body weight) during the sit-to-stand movement represents a higher proportion of peak muscle capacity in older women, these women stood at a slower velocity of contraction on the basis of the force-velocity relationship (17). This may be an important consideration in testing of older adults, because a significant number of lower limb daily activities are closed-chain movements. The blunted ability of older women to rapidly contract antigravity muscles while weight bearing may contribute to increased risk of falls and fracture.

In this study, we report a novel method of assessing fatigability on the basis of the decline in contractile velocity across repetitive contractions. In both the open (knee extension)- and closed (sit-to-stand)-chain tests of fatigue, concentric velocity declined with repeat efforts. Age had a significant effect on muscle fatigue during knee extension but not during the sit-to-stand movement. During repetitive knee extension contractions, older adults experienced 24% fatigue, which was not found in young adults by the 10th repetition, indicating that older adults were less capable of sustaining maximal contractile velocity across repetitions. Our subjects were instructed and encouraged to contract as rapidly as possible with each repetition to generate maximum muscle power. This technique not only produced fatigue as measured by velocity but also enabled us to statistically discriminate older from young adults during knee extensions. By contrast, however, young and older adults experienced similar rates of fatigue during the closed-chain standing task. We speculate the lack of an age effect on this test of fatigue may have been due to task specificity; the standing task is commonly performed by all ages during weight-bearing daily activities, whereas the knee extension movement may have been less familiar to the older subjects. Although we found no statistically significant differences in
fatigue rates by age during repetitive sit to stands, it is worth noting that decreases in power from peak to 10th repetition were 34% in older women and only 21% in young women. Furthermore, these rates of fatigue may have occurred via different mechanisms in young and older women. In young women, peak power declined primarily as a result of reduced velocity (−29.0%) as force production dropped to a lesser extent (−9.1%). On the other hand, a drop in force (−25.2%) seemed to be the primary driving force leading to fatigue in older women (velocity only fell 12.3%). Declines in force and velocity tended to be somewhat lower among the men, and the two age groups responded similarly. For tasks in which body weight represents the resistance force, these data suggest that our group of 64-yr-old men functioned quite well, whereas the 63-yr-old women may have suffered from a low leg strength-to-body weight ratio.

The fall in velocity among older adults reduces specific power, which appears to be important for the completion of weight bearing tasks. Numerous activities in the daily lives of independent older adults require sufficient movement velocity such as stair climbing, crossing intersections, standing from a chair, rising from the floor, recovering from a potential fall, etc. A fall in contractile velocity with multiple contractions would presumably lead to increased difficulty in completing tasks that require repeat efforts. Fatigue is typically measured as a decline in force during sustained isometric contraction or a drop in peak force across repetitive contractions. Existing data on force-dependent fatigue with aging are equivocal. Velocity-dependent fatigue may prove to be a valuable and sensitive measure of fatigueability in future studies of muscular aging.

To examine age-related changes in maximum dynamic force production, we examined 1 RM for three different concentric movements; knee extension, leg press, and squat. The majority of strength loss with age can be accounted for by sarcopenic loss of lean mass (12). Our data confirm this finding because both men and women lost 17% of TLM with age. The women’s age-related loss of 17% TLM occurred concurrent to losses of absolute strength in the leg press (17%) and squat (16%). The men lost 17% TLM concurrent to a 29% loss of strength in both the leg press and squat. We recognize that limitations exist for the use of DEXA to detect muscle mass or predict strength compared with more accurate but expensive measures of magnetic resonance imaging or computerized axial tomography (3, 33). However, it has been shown that DEXA does provide a valid measure of fat free soft tissue that can reliably predict skeletal muscle mass (30, 35). Despite the possible reduced accuracy of DEXA compared with other criterion measures, we still detected age-related differences in TLM. The age differences in our sample were greater than the standard errors of estimate typically reported for DEXA measurements (26, 30, 43). Despite any limitations, the differences in TLM for our sample were robust enough to be detected from estimates of muscle size based on DEXA measurements.

Adjustment for TLM minimized age-related declines in strength for the leg press and squat; however, age differences in maximal knee extension specific strength persisted. In our sample, the DEXA-determined thigh lean tissue measurement included most of the mass of the hip extensors and knee flexors (i.e., hamstrings) as well as the bulk of the hip adductors. Knee extension specific strength was therefore underestimated because the surrogate for muscle size included relatively quiet (i.e., inactive) muscles in addition to the agonist for knee extension, the quadriceps femoris muscle group. We speculate the antigravity quadriceps muscle group may atrophy with age at a more rapid rate than the remaining thigh musculature. If this were true, quadriceps mass would comprise a lesser proportion of TLM, which is the denominator in our specific strength calculation. Although further investigation is needed, this is one potential explanation for the considerably lower knee extensor specific strength ratio in older adults. Although there are no published data to support this hypothesis, it is interesting to note that preferential quadriceps atrophy occurs in knee osteoarthritis and joint injury (19, 41, 45). Additionally, our laboratory has previously shown that age-related sarcopenia differentially affects the strength of load-bearing extensor vs. non-load-bearing flexor muscle compartments (24). Whereas it is certainly possible that other factors could have contributed to the reduced specific strength, such as a decrease in maximal central nervous activation, reduced motivation, or increased connective tissue within the muscle belly (16, 38), these alternative causes should have also impacted the leg press and squat strength-to-lean mass ratios.

The leg press and squat movements require activation of the hip extensors, and to some degree, the adductors; thus the muscle mass activated during leg press and squat may be more appropriately quantified by the assessment of TLM using DEXA. In support of this, our data clearly show that adjustment for declines in TLM account for age-related losses in both the leg press and squat. The age × gender interaction detected for leg press suggests that older men lose strength at a greater rate than lean mass; however, there was no significant difference between older and young men. The gender effect detected during squat is most likely due to the greater distribution of...
body weight above the waist of men, resulting in greater loads lifted but not accounted for by the measurement of external load placed on the squat bar. On the basis of DEXA measurements, we calculated the upper body mass to total body mass ratio and found that men had a significantly greater proportion of body weight above the waist compared with women (data not shown). Older men had the highest proportion of body weight above the waist and also had the lowest specific strength in the squat. Thus total load lifted by the hip and knee extensors during the squat was most underestimated in older males, which may have led to the significant gender effect on squat specific strength.

Strength and power are important determinants of functional ability in older adults particularly for activities involving ambulation and mobility (5, 6, 14). Age had a significant effect on quadriceps neural activity during standing and sitting movements, because older adults used a greater proportion of maximal voluntary neural activation to complete the tasks of sitting and standing compared with young adults. There was also a main gender effect during the eccentric stand-to-sit movement with women requiring higher relative neural activity compared with men. Because the sit-to-stand EMG data were normalized to “maximum” EMG, we recognize that any group differences in voluntary activation during the isometric maximum efforts would lead to a group bias in our normalized sit-to-stand EMG data.

Some controversy exists as to whether older adults have a lesser ability than young adults to maximally activate muscle during voluntary contractions. This may partially be due to variations in assessment techniques, because De Serres and Enoka (10) demonstrated a slightly lower activation level in biceps brachii in elderly (95%) vs. young (97.8%) subjects using traditional superimposition methods but complete activation in both elderly and young using extrapolation procedures. Others have also found no effect of age on maximum voluntary agonist activation in young vs. old men during elbow flexion (98 vs. 96%) and elbow extension (98 vs. 99%) (20). If agonist activation is blunted in older adults, it may result from increased antagonist cocontraction (22). This possibility cannot be ruled out because we did not assess cocontraction. On the basis of these previously published data, however, if age differences in voluntary activation exist they appear to be too small to account for the rather large differences in our normalized sit-to-stand EMG amplitude data. Furthermore, our laboratory has previously reported age differences similar to the present findings in older vs. young women (24). The age and gender differences may be attributable to the ratio of TLM to total body mass because older adults, particularly women, had relatively less recruitable TLM to accomplish the vertical work during the sit-to-stand movement. Men had a higher relative TLM, suggesting a greater capacity for movement of body weight. Furthermore, the significant age-related decline in relative TLM suggests a preferential loss of lean mass from the lower limb musculature.

These data support the concept that age-related declines in contractile strength result primarily from muscle atrophy, whereas deficits in contractile power persist after adjustment for muscle mass, because maximum contractile velocity falls markedly with aging. When contractile velocity is used as the criterion measure of fatigue, we have demonstrated that older adults are more susceptible to fatigue during repetitive contractions. These age-associated impairments likely contribute to mobility loss and increased risk of falls among older adults. Interventions aimed toward improving contractile velocity and, therefore, the capacity to generate power, should be pursued in older adults.

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