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ORIGINAL ARTICLE

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Viscoelastic properties of short calf muscle-tendon units of older women: effects of slow and fast passive dorsiflexion stretches in vivo

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Abstract Changes in connective tissues of the skeletal muscle-tendon unit (MTU) of aging animal muscles have been associated with increased passive viscoelastic properties. This study examined whether similar changes in the viscoelastic properties were present in short calf MTUs of older women in vivo. Fifteen women 68–87 years of age with short calf MTUs, as represented by limited active dorsiflexion (DF) range of motion (ROM) of ≤ 5°, and 15 women 20–26 years of age without decreased DF ROM participated. A Kin-Com dynamometer stretched the MTU from plantarflexion to maximal DF at the slow velocity of 5° s^{-1} (0.087 rad s^{-1}) and the fast velocity of 120° s^{-1} (2.094 rad s^{-1}) with minimal surface electromyogram activity in the soleus, gastrocnemius, and tibialis anterior muscles. Two-way analysis of variance (ANOVA) tests for repeated measures (Velocity × Group) indicated that all women showed greater passive torque, average passive elastic stiffness, and total absorbed passive elastic energy for the fast stretch than for the slow stretch ($P < 0.001$). The older women had greater percent increases for the average passive torque (30%) and total absorbed passive elastic energy (26%) for the fast stretch than the younger women ($P < 0.05$), who had 17.5 and 13% increases, respectively. The older women had less maximal and average passive torque (Nm) and total absorbed passive elastic energy (°Nm), but greater average passive elastic stiffness (Nm$°^{-1}$) at both stretch velocities ($P < 0.001$). The results indicated that short calf MTUs of older women have increased passive viscoelastic properties that could have implications for balance and ambulatory function.

Keywords Aged · Calf muscle-tendon unit · Passive elastic stiffness · Passive elastic energy · Viscoelastic properties · Women

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Introduction

Studies have long shown that the passive resistive forces of skeletal muscles increase in a curvilinear manner as the muscle-tendon unit (MTU) is stretched from a shortened length to its maximal length (Brodie 1895; Haycraft 1904; Banus and Zetlin 1938; Tabary et al. 1972, 1976; Williams and Goldspink 1978). Investigations have shown that a combination of anatomical structures are responsible for this curvilinear increase. Some studies have indicated that the resistance to a passive stretch is borne primarily by the relatively large amounts of perimysium connective tissues that surround muscle fasciculi (Borg and Caufield 1980; Rowe 1974; Purslow 1989) and undergo mechanical deformation as the MTU is lengthened (Rowe 1974; Purslow 1989; Williams and Goldspink 1984). Other studies have indicated that the endomysium surrounding muscle fibers also contributes to the passive resistive forces of a stretched muscle (Purslow and Trotter 1994; Trotter and Purslow 1992). The proposed mechanism for the transmission of forces across these structures is a transfer of shearing forces between muscle fibers (Purslow and Trotter 1994; Trotter and Purslow 1992). Within the muscle fiber, during passive lengthening, strain of the non-contractile protein "titin" of the endosarcomeric cytoskeleton (Funatsu et al. 1996; Linke et al. 1996; Trombitas et al. 1998; Wang et al. 1993), and of the non-contractile protein "desmin" of the exosarcomeric cytoskeleton (Wang and Ramirez-Mitchell 1983; Wang et al. 1993) would also contribute to increased passive resistance.

Extramuscular tendons that anchor skeletal muscles to bone are relatively non-elastic, but recent evidence suggests that they may provide substantial contributions to storing and releasing passive elastic energy (Fukunaga et al. 2001; Kawakami et al. 2002; Kurokawa et al. 2003). The intramuscular tendon, referred to as the aponeurosis of the muscle belly has unique elastic qualities that allow elongation that is greater than the extramuscular tendon (Huijing and Ettema 1988; Zuurbier et al. 1994), and the elastic qualities of the aponeurosis vary within the distal and proximal areas of the aponeurosis (Zuurbier et al. 1994).

The skeletal MTU demonstrates passive viscoelastic behaviors in response to a stretch. That is, they exhibit elastic behaviors that depend on the load of the applied stretch, and they exhibit viscous behaviors that depend on the velocity of the applied stretch (Le Veau 1992; Simons and Mense 1998; Gajdosik 2001). Increased passive resistance to a fast dynamic (movement) stretch compared to a slow dynamic stretch has been shown in animal muscles (Taylor et al. 1990), as well as in the calf MTUs of healthy, non-disabled younger and middle-aged people (Gajdosik 1997; Lamontagne et al. 1997; McNair et al. 2002; Singer et al. 2003). However, the effect of different velocities of lengthening has not been well documented in older adults. Changes in the connective tissues of the MTU during the aging process would potentially increase the viscoelastic behaviors of the human calf MTU in vivo, but this has not been examined in detail. Studies with animal muscle models have indicated that the amount of collagen of the perimysium and the endomysium increases in aged rat muscles (Alnaqeeb et al. 1984), and increased total collagen (Kovanen 1989) and collagen fiber "cross-linking" has been associated with increased passive stiffness of aged rat muscles (Kovanen and Suominen 1988). Animal muscles dominated by slow twitch muscle fibers (type I) have increased collagen and greater stiffness compared to muscles dominated by fast twitch muscle fibers (type II) (Kovanen 1989; Kovanen et al. 1984), and a shift to slow twitch muscle dominance in aged animal muscles may contribute to increased passive elastic stiffness (Alnaqeeb and Goldspink 1987; Kanda and Hashizume 1989).

Based on the well-established influence of the connective tissues on the passive resistive forces of stretched skeletal muscles, and the changes in the connective tissues as a result of aging that have been shown in animal muscle models, one would expect that aged human calf MTUs would show increased passive viscoelastic properties in response to rapid passive stretching. Accordingly, the present study was designed to examine the effects of a slow stretch and a fast stretch on the passive properties of short calf MTUs of older women in vivo to examine this proposal. The calf MTU was selected because decreased calf MTU length, which is represented by decreased dorsiflexion (DF) range of motion (ROM), is associated with normal aging in both men and women (James and Parker 1989; Vandervoort et al. 1992; Gajdosik 1997; Gajdosik et al. 1996, 1999). Past studies that examined the passive properties of the calf MTU of older people recruited from the general population reported decreased passive resistance and decreased passive elastic stiffness (Gajdosik 1997; Gajdosik et al. 1999), but the subjects in these studies did not have marked calf MTU shortening for their age groups. A follow-up study with a subset of older women with short calf MTUs with active maximal DF ROM $\leq$ 10° showed that the average passive elastic stiffness of their calf MTUs was maintained compared to the calf muscles of younger women, although the length and passive resistive forces were decreased (Gajdosik et al. 2004). The calf MTUs also had significantly greater passive elastic stiffness at 0 and 5° of DF. Studying the passive properties of short calf MTUs in response to fast stretches compared to slow stretches would contribute to understanding the influence of aging on the viscoelastic characteristics and function of the calf MTU for this important subset of older women. The results would also have clinical implications, as the calf MTU is routinely stretched at varying velocities during standing balance and ambulatory activities.

Methods

Subjects

Fifteen older women with goniometric active DF ROM $\leq$ 5° and 15 younger women without limited DF ROM participated. The mean age of the older women was 78.1 years [Standard Deviation (SD)]: 6.5 years, range: 68–87 years), and the mean age of the younger women was 23.8 years (SD: 1.4 years, range 20–26 years). The older women had a maximal active DF ROM of $-0.3 \pm 3.0°$ (range: −4 to 5°) and the younger women had a maximal active DF ROM of $8.7 \pm 4.1°$ (range: 2–18°). A 90° angle between the foot and the leg was defined as 0°, DF degrees were positive, and plantarflexion (PF) degrees were negative. All women were without a history of orthopedic or neurological disorders that could confound the results, and they were minimally to moderately active. All subjects signed an informed consent form for the study, which was approved by the Institutional Review Board for the Use of Human Subjects in Research of The University of Montana.

Instrumentation

A Kin-Com isokinetic dynamometer (Kinetic Communicator II 500H, Software Version 4.03, Chattecx Corp., Chattanooga, TN, USA) was used for all tests. The Kin-Com ankle-foot apparatus was used to stretch the calf muscles by passively moving the ankle from PF into DF at $5° s^{-1}$ (0.087 rad s^{-1}: slow stretch) and at $120° s^{-1}$ (2.094 rad s^{-1}: fast stretch), and to immediately move the ankle back into PF at $5° s^{-1}$ after both stretch velocities. Stretching the calf MTU at the slow velocity of $5° s^{-1}$ was used because this is an established stretch velocity for ensuring that muscle activity is not elicited reflexively, thus permitting passive stretches (Gajdosik et al. 1996, 1999; Lamontagne et al. 1997; McNair et al. 2002; Singer et al. 2003). A fast stretch velocity of $120° s^{-1}$ is sufficient to elicit a velocity-dependent viscoelastic response without stimulating reflexive muscle activity in relaxed calf muscles (Gajdosik 1997), while also minimizing the loss of stretch ROM because of acceleration and deceleration artifacts with the Kin-Com at higher velocities. The lever arm was held constant at 20 cm to express the passive resistance in torque (Nm).

Surface electromyography (SEMG) (GCS 67, Therapeutics Unlimited, 2835 Friendship St., Iowa City, IA, USA) was used to monitor the activity of the soleus, the medial head of the gastrocnemius, and the tibialis anterior (TA) muscles during the tests. The bandwidth of the frequency response was 20 Hz–4 kHz. The common mode rejection ratio was 87 dB at 60 Hz, and the input impedance was greater than 25 MΩ. The surface electrodes were round bipolar electrodes made of Ag/AgCl with a diameter of 8 mm and an inter-electrode distance of 20 mm between the electrode centers. Electrode placement was determined with the subjects standing. They were asked to rise up on their toes (active PF) and the medial head of the gastrocnemius muscle and the soleus muscles were identified visually and by palpation. The skin was shaved if needed, cleansed with alcohol, and the pre-gelled electrodes were affixed to the skin with double-sided hypoallergenic tape. For the medial head of the gastrocnemius muscle the electrode was placed over the muscle belly parallel to the longitudinal axis of the leg to approximate the direction of the muscles fibers. For the soleus muscle the electrode was placed in the midline about 2.5 cm inferior to the inferior border of the medial head of the gastrocnemius muscle, and also parallel to the longitudinal axis of the leg to approximate the direction of the muscle fibers. Active DF, visual inspection and palpation were used to locate the TA muscle. The TA electrode was affixed to the skin over the most prominent aspect of the muscle belly longitudinally, and approximately 1/4 the distance from the tibial plateau to the ankle, immediately lateral to the crest of the tibia. A pre-gelled reference electrode was placed over the left fibular head, and the electrode connections were tested using active PF and DF. The above electrode specifications were in accordance with the SENIAM guidelines (Hermens et al. 2000).

The raw SEMG signals are amplified (5000 times), high pass filtered at 20 Hz, and the analogue signals were converted to digital format at a sampling rate of 2000 Hz and recorded using Biopac MP150 data acquisition hardware and AcqKnowledge software Version 3.7.3 (BIOPAC Systems Inc., 42 Aero Camino, Santa Barbara, CA, USA). Angle (°), velocity ($° s^{-1}$) and force (N) signals from the Kin-Com (sampled at 500 Hz) were simultaneously synchronized with the EMG tracings from the three leg muscles.

Procedures and measurements

The subjects first assumed a supine position on an examination table and the axis of the right ankle was estimated using a procedure described by Blanpied and Smidt (1992). The subjects then completed a calf-muscle-stretching regimen of ten static wall stretches held for 15 s each as reported previously (Gajdosik et al. 1999). These exercises decreased "tissue force-relaxation" and helped to prevent a decline in the maximal passive resistance during the data acquisition trials. After stretching, the SEMG electrodes were attached over the appropriate muscle bellies. The subjects then assumed a supine, relaxed position on the Kin-Com table with the right knee extended to test the right calf MTU. With the leg level with the horizontal plane the ankle and foot were secured in the apparatus and the ankle was aligned with the axis of the Kin-Com armature. Stabilization straps secured the right knee and the chest. Using visual oscilloscope tracings and auditory feedback, the subjects learned to recognize SEMG activation and SEMG silence of the muscles, and this helped to ensure that calf muscle activations were minimal during the passive stretch trials. Background SEMG noise activity was collected over a 5 s period with the subject resting quietly. The mean of the root mean square (RMS) SEMG activity within the middle 3 s window of the 5 s was calculated, and this background noise activity was deducted from the RMS SEMG activity recorded during the passive stretch trials. The RMS SEMG activity recorded during the trials was then expressed as a percentage of the RMS SEMG activity recorded during maximal PF or DF isometric contractions (MIVCs) conducted after completion of the passive tests.

The subjects were encouraged to maintain "flat SEMG tracings" (muscle silence) during the test session, which was conducted in a quiet room with the lights dimmed. The maximal passive DF angle was first determined blindly by manually moving the ankle slowly into DF without observing the angle displayed on the monitor. After several trials, the end point of the DF stretch was defined just prior to the point that caused pain or discomfort, by a marked increase of SEMG activity in the calf muscles (Gajdosik et al. 1996, 1999), or when the heel moved out of the ankle-foot apparatus (Singer et al. 2003). The subjects' perceived tolerance to the maximal passive stretch and increased SEMG activity were the primary criteria used to define the maximal DF angle, which operationally defined the maximal length of the calf MTU. All subjects reported a stretching sensation in the muscle bellies of the calf muscles at the terminal stretch angle.

After defining the maximal DF angle, the ankle-foot apparatus was moved to 45° of PF. The ankle was then stretched passively by the Kin-Com from this PF position to the maximal DF angle and immediately returned back into PF. Three stretch and return trials at each of the randomly ordered stretch velocities of 5 and 120° s^{-1} were performed. The ankle was returned into PF at the slow velocity of 5° s^{-1} to ensure subject comfort and safety. Subjects who could not relax sufficiently showed obvious SEMG activity throughout the stretch and return ROMs, so these subjects were not included in the final analyses.

After the passive tests, the subjects completed three trials of PF MIVCs held for 5 s with the ankle-foot apparatus held at 0° of DF in order to record the maximal SEMG activity of the two calf muscles. This was followed by three trials of DF MIVCs also held for 5 s to record the maximal SEMG activity of the TA muscle.

Data reduction

All force and EMG recordings were recorded in relation to the angular displacement of the stretch trials. The maximal passive DF angle was then defined at 4° less than the actual maximal passive DF angle, which accounted for the deceleration artifact present during the 120° s^{-1} stretch. The maximal passive resistive force was measured at this adjusted maximal DF angle for both stretch velocities. The initial passive DF angle was identified for both stretch velocities after the acceleration phase of the 120° s^{-1} stretch, and after deducting the force associated with the inertia of the apparatus. The difference between the initial DF angle and the maximal DF angle defined the full stretch ROM. Using this method allowed the 5° s^{-1} stretch trials to be matched to the 120° s^{-1} stretch trials. The passive force measurements (N) were then expressed as passive torque measurements (Nm) by calculating the ankle moment (0.20 m×N).

A 3rd order polynomial function ($Y = A + BX + CX^2 + DX^3$, where Y is the passive torque, X the angular displacement, B the slope, and A, C and D are constants) was used to approximate the curvilinear passive curves (mean coefficient of determination of $R^2 = 0.998$). From the fitted curves, the average passive elastic stiffness was calculated as the slope, or the change in torque divided by the change in angle (Δ Nm/Δ °). The maximal DF angle was constant for both stretch velocities and tabulated for both groups. The maximal and mean passive torques (Nm), average passive elastic stiffness ($Nm°^{-1}$), and the absorbed passive elastic energy [integrated area under the full stretch curve (°Nm)] were measured.

After deducting the mean of the RMS background SEMG noise activity, the RMS SEMG activity during the stretch trials was calculated as the percent RMS SEMG relative to the RMS SEMG activity recorded over a middle 3 s period during the 5 s MIVC that produced the greatest SEMG activity. Stretch trials were defined as passive if the mean RMS SEMG activity was less than 5% of that which was recorded during the MIVCs and if there was no influence on the shape of the passive curves. The actual mean RMS SEMG activity during the stretch was calculated as $<0.5\%$ (range 0.05–0.48%) of the mean RMS SEMG activity recorded during the MIVCs of all three muscles for both groups. The SEMG activity present in some subjects was primarily at the end of the stretch ROM, which was not included in the analysis because the last 4° were eliminated. No obvious increased SEMG activity was observed above baseline within the early stage of the stretch ROM.

The reliability and the precision of the method used to determine the maximal passive DF angle and the maximal passive torque were examined and reported previously (Gajdosik et al. 1999). The intraclass correlation coefficient (ICC) and the standard error of measurements (SEM) for measuring the maximal DF angle were 0.91 and ±1.2°, respectively. The ICC and SEM for measuring the maximal passive torque were 0.90 and ±3.9 Nm, respectively. The ICCs and the SEMs indicated excellent reliability and precision for these measurements.

Statistics

The mean and SD were tabulated for the following measurements: (1) maximal passive DF angle, (2) maximal passive DF torque, (3) average passive torque for the full stretch ROM, (4) the average passive elastic stiffness for the full stretch ROM, and (5) the integrated area under the full stretch curve (total absorbed passive elastic energy). The adjusted maximal DF angle was analyzed with a one-way analysis of variance (ANOVA) to examine group differences. The mean of the three trials of the remaining measurements were analyzed by separate two-way ANOVA tests for repeated measures (Velocity × Group) to primarily examine the effects of the 120° s^{-1} fast stretch compared to the 5° s^{-1} slow stretch, but differences between the groups were also examined. Differences between the two stretch velocities were expressed as a percent change, which also was examined for group differences. Statistical significance was set at $P \leq 0.05$.

Results

The passive curves for the full stretch ROM at both stretch velocities for both groups are depicted in Fig. 1a and b, respectively. The curvilinear shape was similar for both groups, but the curves for the older women were smaller, steeper and truncated because the maximal passive DF angle was shifted to the left and the maximal

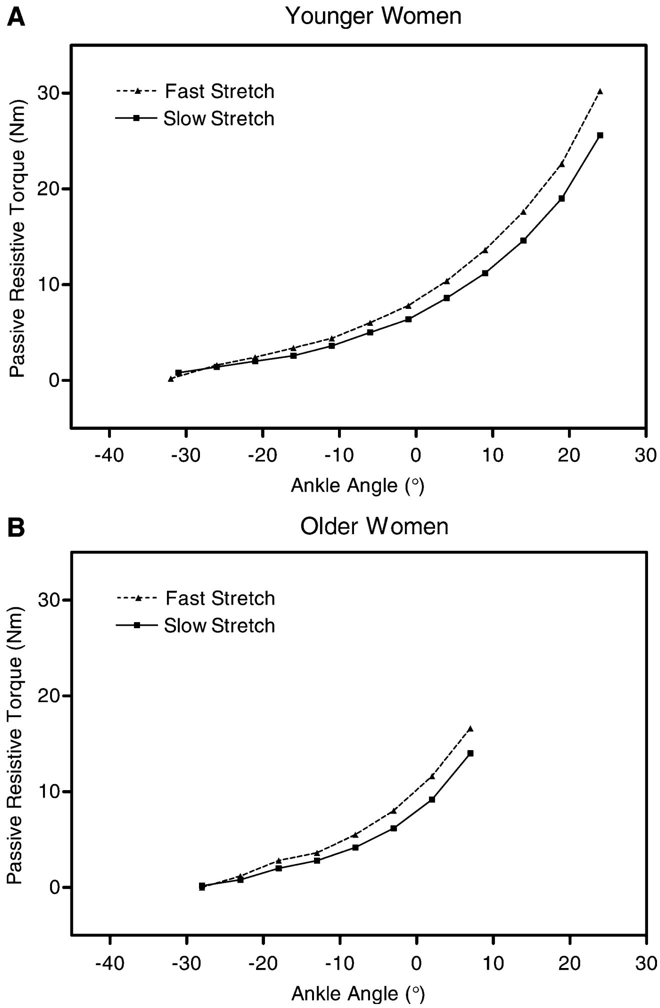

Fig. 1 Passive stretch curves for the slow 5° s^{-1} and the fast 120° s^{-1} stretches for: **a** the younger women (n = 15), and **b** the older women (n = 15). Both groups showed increased average passive torque in response to the fast stretch compared to the slow stretch ($P<0.001$). *Note* Standard deviation *error bars* were deleted for clarity

passive torque was of lesser magnitude at both velocities. The curves for the fast stretch for both groups showed greater passive resistance compared to the curves for the slow stretch.

The maximal passive DF angle for the older women (mean 6.8°, SD 4.3°) was less than the maximal passive DF angle for the younger women (mean 24.1°, SD 7.0°) ($F=66.69$, $P<0.001$). The mean (± SD) for the maximal passive DF torque, average passive torque, full stretch ROM passive elastic stiffness and total absorbed passive elastic energy for both stretch velocities for both groups are reported in Table 1. The repeated measures ANOVA analyses indicated that all measurements increased for the fast stretch for both groups ($P<0.001$, Table 1). The fast stretch maximal passive DF torque increased 18% for the younger women and 18.6% for the older women, which were not significantly different. The older women, however, had a significantly greater percent increase (30%) than the younger women (17.5%) for the average passive torque for the fast stretch full ROM ($P=0.029$) (Fig. 2). The maximal passive DF torque and the average passive DF torque were less for the older women compared to the younger women at both stretch velocities ($P<0.001$).

The passive elastic stiffness for the older women was greater at both stretch velocities ($P<0.001$), but the percent increases in the passive elastic stiffness for the fast stretch for the older women (14.3%) and the younger women (17.5%) did not differ statistically. Figure 3 provides a visual comparison of the passive curves for the older and younger women for the slow stretch to illustrate the differences in the passive elastic stiffness (slope) between the two groups. The absorbed passive elastic energy was less for the older women compared to the younger women at both stretch velocities ($P<0.001$), and both increased for the fast stretch ($P<$ 0.001). The percent increase for the fast stretch, however, was significantly greater for the older women (26%) compared to the younger women (13%) ($P=0.005$) (Fig. 4).

Discussion

The shift to the left for the maximal passive DF angle and the decreased magnitude of maximal passive DF torque for the older women indicated a marked decrease in the maximal MTU length and the ability to withstand a passive load as tolerated. The older women also had significantly less average passive torque, and less total absorbed passive elastic energy, which were similar to the results of previous studies (Gajdosik et al. 1999, 2004). These results also are in-line with the anatomic and physiologic changes that result from aging, which brings about a loss of functional motor units (Brown et al. 1988; Campbell et al. 1973), and a decrease in the number and size of both slow twitch (type I) and fast twitch (type II) muscle fibers (Lexell et al. 1983, 1988). The reduction in the number of functional motor units and muscle fiber size (sarcopenia) partially account for the decreased muscle mass and the strength deficits reported in the muscles of older people (Doherty et al. 1993; Lexell 1995). The loss of muscle mass, combined with the decreased MTU length related to aging would decrease the calf MTUs ability to withstand a maximal passive stretch.

The calf MTUs of the older women showed greater average passive elastic stiffness compared to the younger women, which suggested that their short calf MTUs probably had internal structural changes that compensated for the decreased length. The increased calf MTU passive elastic stiffness measured in vivo also is in-line with the results of animal studies that have shown increased amounts of collagen of the perimysium and endomysium associated with greater skeletal muscle stiffness in older muscles compared to younger muscles (Alnaqeeb et al. 1984), and increased total collagen

Table 1 Mean and standard deviation (±SD) for the maximal dorsiflexion (*DF*) torque, average DF torque, average passive elastic stiffness and area under the full stretch passive curve (absorbed passive elastic energy) for the younger women ($n=15$) and the older women ($n=15$) for the slow and fast stretch

Measurements	Mean ± SD	Effects of fast stretch	
		F Ratio	*P* Value
Maximal DF torque (Nm)			
Slow stretch (5° s^{-1})			
Younger	25.6 ± 5.0		
Older	14.0 ± 4.3		
Fast stretch (120° s^{-1})		182.37	< 0.001
Younger	30.2 ± 5.6		
Older	16.6 ± 5.2		
Average DF torque (Nm)			
Slow stretch (5° s^{-1})			
Younger	7.9 ± 1.4		
Older	4.5 ± 1.3		
Fast stretch (120° s^{-1})		142.01	< 0.001
Younger	9.2 ± 1.4		
Older	5.7 ± 1.5		
Passive elastic stiffness (Nm°$^{-1}$)			
Slow stretch (5° s^{-1})			
Younger	0.40 ± 13		
Older	0.71 ± 31		
Fast stretch (120° s^{-1})		44.80	< 0.001
Younger	0.47 ± 15		
Older	0.82 ± 27		
Area under passive curve (°Nm)			
Slow stretch (5° s^{-1})			
Younger	466 ± 122		
Older	165 ± 70		
Fast stretch (120° s^{-1})		74.10	< 0.001
Younger	526 ± 131		
Older	207 ± 89		

(Kovanen 1989) and collagen fiber "cross-linking" associated with increased stiffness of aged rat muscles (Kovanen and Suominen 1988). A shift to slow twitch muscle fiber dominance in aged animal muscles also contributes to increased passive elastic stiffness (Alnaqeeb and Goldspink 1987; Kanda and Hashizume 1989), as muscles dominated by slow twitch muscle fibers (type I) have increased collagen and greater stiffness compared to muscles dominated by fast twitch muscle fibers (type II) (Kovanen et al. 1984).

Studies with animal muscles immobilized in shortened positions also have shown increased passive elastic stiffness associated with greater abundance (Tabary et al. 1972) and remodeling (Williams and Goldspink 1984) of the connective tissues of the muscle. Research in

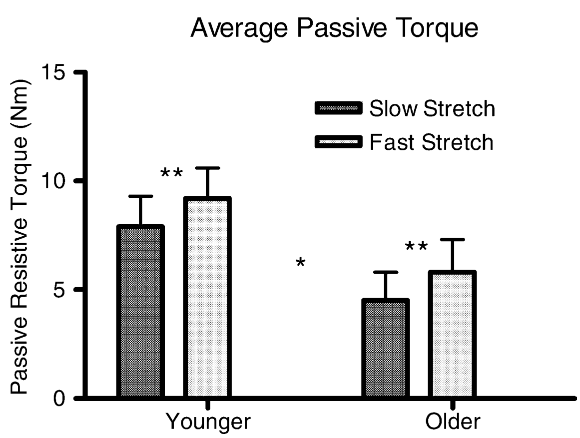

Fig. 2 The average passive torque through the full stretch ROM for the fast stretch compared to the slow stretch. This was greater for both groups for the fast stretch (**$P<0.001$), and although less for the older women (*$P<0.001$), the percent increase was significantly greater for the older women (30%) than for the younger women (17.5%) ($P=0.029$). *Error bars* are ±1 SD

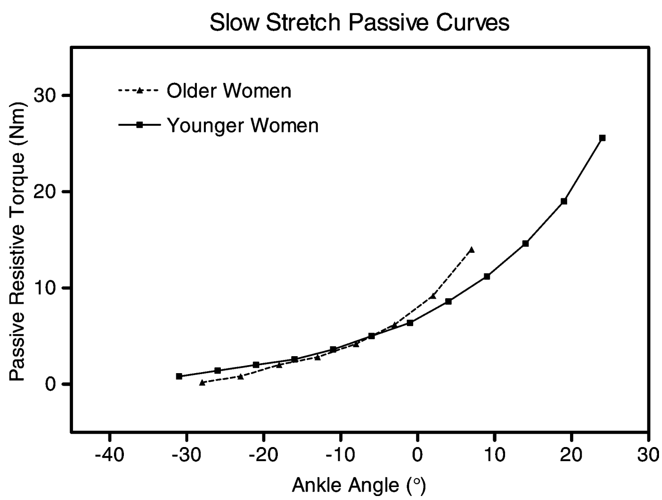

Fig. 3 Passive stretch curve at the slow 5° s^{-1} stretch for the younger women and the older women. The curves for the older women were steeper illustrating greater passive elastic stiffness ($P<0.001$). *Note* Standard deviation *error bars* were deleted for clarity

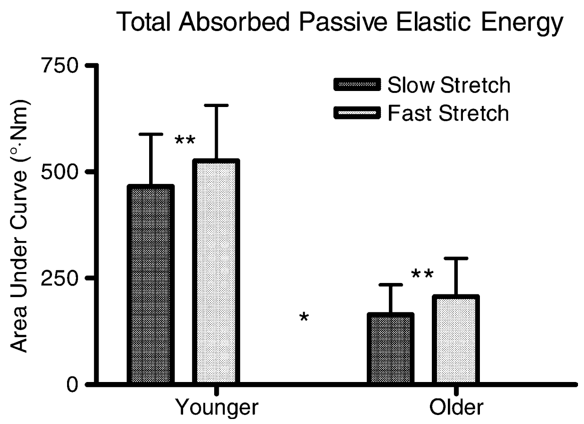

Fig. 4 The total absorbed passive elastic energy through the full stretch ROM for the fast stretch compared to the slow stretch. This increased for fast stretch compared to the slow stretch for both groups (**$P < 0.001$), and although less for the older women (*$P < 0.001$), the percent increase was significantly greater for the older women (26%) than for the younger women (13%) ($P = 0.005$)

humans has shown that decreased muscle fiber mass in older people may be replaced by increased fat and connective tissue (Lexell 1995; Rice et al. 1989; Sipila and Suominen 1995). Although the calf MTUs of the older women in our study were not immobilized, they probably underwent shortening adaptations that increased their average passive elastic stiffness. These possibilities are worthy of further study.

Both groups showed significant increases in the maximal passive torque and average passive elastic stiffness for the fast stretch compared to the slow stretch, which are in-line with previous studies for younger and middle-aged people (Gajdosik 1997; Lamontagne et al. 1997; McNair et al. 2002; Singer et al. 2003). The marked increases for the older women in the current study, however, differ from the previous study by Gajdosik (1997) that showed only a limited increase in passive forces for older men when their calf MTUs were stretched rapidly. The older women in the current study had clinically short calf MTUs compare to the men tested previously (Gajdosik 1997), which could partially account for their enhanced velocity-dependent response. The lack of a difference between groups for the percent increase in the maximal passive resistive torque and the passive elastic stiffness for the fast stretch contrast with the significant percent increase for the mean passive resistive torque and the absorbed passive elastic energy. Further research is indicated to examine the maximal passive torque and also the importance of measuring the average passive stiffness as a marker for changes in viscoelastic properties.

The significantly greater percent increase in the average passive torque and absorbed passive elastic energy for the older women compared to the younger women suggest that the calf MTUs of the older women had increased viscous properties. Increased amounts and remodeling of the relatively inextensible, non-elastic collagen connective tissues could bring about greater passive torque through the full stretch ROM, particularly if there were changes in the perimysium (Rowe 1974; Purslow 1989; Williams and Goldspink 1984) and endomysium (Purslow and Trotter 1994; Trotter and Purslow 1992). Examining the specific connective tissue structures that could contribute to these differences was beyond the scope of the current study. One might expect, however, that aging changes in the connective tissues throughout the entire MTU would increase the average passive torque and absorbed passive elastic energy measured in vivo. Our results suggested that the enhanced absorbed passive elastic energy after the rapid stretch could enhance active force during a concentric activation, such as during ambulation or in response to perturbations in standing or during ambulation. The effect may be related to the speed of movement, as the stored passive elastic energy is more likely to be reused when the frequency of the movement matches the resonant frequency of the tissues that are stretched (Bach et al. 1983; Wilson et al. 1991). Further research is needed to determine whether very short calf MTUs of older women would show these enhanced active forces after a rapid stretch during functional activities.

The authors acknowledge that restricted DF ROM and passive resistance to a DF stretch can be influenced by structures not directly related to the calf MTU, such as the ankle joint capsule, associated ligaments, neurological tissues, vascular tissues, deep enveloping fascia of the leg, the superficial fascia and the skin. The calf MTU, however, has been shown to lengthen with DF movement (Herbert et al. 2002; Kawakami et al. 2002), and is considered to offer the major resistance to a maximal passive DF stretch (Gajdosik et al. 2004; Singer et al. 2003), especially when the maximal DF stretch angle is defined by increased electrical activity in the calf muscles or by a maximal tolerated stretching sensation in the muscle bellies of the calf muscles (Gajdosik et al. 2004).

Conclusions

The calf MTUs of older and younger women showed a significant increase in the passive torque, passive elastic stiffness, and the absorbed passive elastic energy for a fast stretch compared to a slow stretch. The significantly greater percent increase in the average passive torque and the total absorbed passive elastic energy through the full stretch ROM for the older women suggested that their muscles had increased viscous properties, which would correspond with aging related changes in the collagen of the supporting connective tissues of the MTU. The results also showed that the maximal length, passive torque, and absorbed passive elastic energy for the short calf MTUs of older women were decreased, but the average passive elastic stiffness through the full stretch ROM was increased compared to the younger women. The changes in viscoelastic properties of the calf MTUs of older women could be adaptations that have implications for balance and ambulatory function for the elderly.

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