Ultrasonic evaluations of Achilles tendon mechanical properties poststroke

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Submitted 9 September 2008; accepted in final form 27 December 2008

Zhao H, Ren Y, Wu Y, Liu SQ, Zhang L. Ultrasonic evaluations of Achilles tendon mechanical properties poststroke. J Appl Physiol 106: 843–849, 2009. First published December 31, 2008; doi:10.1152/japplphysiol.91212.2008.—Spasticity, contracture, and muscle weakness are commonly observed poststroke in muscles crossing the ankle. However, it is not clear how biomechanical properties of the Achilles tendon change poststroke, which may affect functions of the impaired muscles directly. Biomechanical properties of the Achilles tendon, including the length and cross-sectional area, in the impaired and unimpaired sides of 10 hemiparetic stroke survivors were evaluated using ultrasonography. Elongation of the Achilles tendon during controlled isometric ramp-and-hold and ramping up then down contractions was determined using a block-matching method. Biomechanical changes in stiffness, Young’s modulus, and hysteresis of the Achilles tendon poststroke were investigated by comparing the impaired and unimpaired sides of the 10 patients. The impaired side showed increased tendon length (6%; P = 0.04), decreased stiffness (43%; P < 0.001), decreased Young’s modulus (38%; P = 0.005), and increased mechanical hysteresis (1.9 times higher; P < 0.001) compared with the unimpaired side, suggesting Achilles tendon adaptations to muscle spasticity, contracture, and/or disuse poststroke. In vivo quantitative characterizations of the tendon biomechanical properties may help us better understand changes of the calf muscle-tendon unit as a whole and facilitate development of more effective treatments.

THE ACHILLES TENDON PLAYS AN IMPORTANT role in force transmission and energy storage and return during functional activities such as walking and running, (26) and it interacts with the calf muscle directly (32). Recent studies show that the metabolic activity in human tendon is remarkably high, which affords the tendon the ability to adapt to changing demands (3, 4, 15, 16). Stroke survivors often develop spasticity, contracture, and/or muscle weakness at the ankle, which contribute considerably to their disabilities poststroke (5, 12, 14, 34). About 34% of stroke survivors develop ankle contracture (27, 28, 34). The impaired ankle poststroke was evaluated indirectly during controlled isometric ramp-and-hold and ramping up then down contractions was determined using a block-matching method. Biomechanical changes in stiffness, Young’s modulus, and hysteresis of the Achilles tendon poststroke were investigated by comparing the impaired and unimpaired sides of the 10 patients. The impaired side showed increased tendon length (6%; P = 0.04), decreased stiffness (43%; P < 0.001), decreased Young’s modulus (38%; P = 0.005), and increased mechanical hysteresis (1.9 times higher; P < 0.001) compared with the unimpaired side, suggesting Achilles tendon adaptations to muscle spasticity, contracture, and/or disuse poststroke. In vivo quantitative characterizations of the tendon biomechanical properties may help us better understand changes of the calf muscle-tendon unit as a whole and facilitate development of more effective treatments.

METHODS

Subjects. Ten stroke survivors participated in the study (5 right side impaired and 5 left side impaired; both the left side- and right side-impaired groups contained 1 female subject and 4 male subjects). The inclusion criteria for the patients were having hemiparesis following a stroke at least 1 yr earlier, able to walk with a cane or without any mechanical aid, and able to generate plantar flexion torque using the calf muscles. Exclusion criteria were: having other severe disease, leg musculoskeletal injuries, and/or orthopedic surgeries on the leg. The patients had a mean age of 60.3 ± 8.4 (mean ± SD; range 52.8–78.8) yr, stroke duration of 12.0 ± 5.8 (range 4.5–20.8) yr, a mean height of 170.8 ± 10.2 (range 147.3–181.6) cm, body mass 80.6 ± 15.3 (range 59.9–99.9) kg. Their modified Ashworth scale was 1.6 ± 1.3 (range 0–3). Their passive range of motion was 11.0 ± 4.8 (range 5–19) ° in dorsiflexion, and 42.6 ± 4.5 (range 37–51) ° in plantar flexion, respectively. Three subjects wore an ankle foot orthosis, and one subject was using antispasticity medication. Three subjects used the straight cane and one subject used the quad cane. All participants gave informed consent to participate in this study, which was approved by the Institution Review Board.

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Experimental setup. The experimental setup consisted of four major parts: a custom knee-ankle joint-driving device, an ultrasound imaging system, an electromyography (EMG) system, and a personal computer.

A custom knee-ankle joint-driving device was used to position the leg and the ankle and to measure the joint torque. The motor at the knee was fixed rigidly to a frame anchored to the ground, and a leg linkage was mounted to the knee motor through a torque sensor. The ankle motor was mounted at the distal end of the leg linkage, and a footplate was mounted to the ankle motor through another torque sensor. The ankle motor and footplate could be adjusted along the leg linkage so that the ankle and knee motors were aligned with the ankle and knee flexion axes, respectively (Fig. 1).

The ankle plantar flexion torque was measured and displayed on the screen in real time as a visual feedback. Bagnoli EMG system (Delsys, Boston, MA) was used to monitor the EMG signal from the tibialis anterior muscle. A trigger signal was sent to the ultrasound imaging system to initiate the ultrasound image recording. The sampling rate for both torque and EMG data acquisition was 1,000 samples/s.

A commercial ultrasound imaging system, GE LOGIQ-9 with a 12-MHz high-resolution matrix probe M12L (GE Healthcare, Waukesha, WI), was used to obtain images of the Achilles tendon and monitor the movement of the soleus-Achilles MTJ.

Experimental design. The experiment for every subject consisted of two visits. Each of the legs was tested in one of the two visits, following the same protocol. The sequence of the side being tested was selected randomly. Time between the two visits for each subject was within 1 wk. For the patients who were taking antispastic medications, the experiment was done at least 4 h after the medicine was taken.

The subject was seated upright with the thigh and trunk secured using Velcro straps. The leg and foot were securely attached to the leg linkage and footplate, respectively (Fig. 1), with the knee and ankle at full extension and 0° dorsiflexion, respectively. After several warm-up isometric contractions in plantar flexion, the subject gradually increased the plantar flexion torque until reaching a specified level and then held this for a period of time (ramp-and-hold), following the target and actual torques displayed in real time on the monitor. The ramping speed was set so that the subjects could carry out the task properly and the MTJ displacement could be tracked in the ultrasound images. The task was repeated three times with a 1-min rest in between. The subject then performed a ramp up and down (gradually increased the torque to a specified level and then gradually decreased it) isometric plantar flexion following the target torque for three times separated by a 1-min rest. The torque levels for both tasks were set (~15 N·m) so that the subject could finish the task properly without losing control or inducing much fatigue during the isometric contractions. The ultrasound video data, ankle joint torque, and EMG signals were recorded throughout both tasks.

Measurement of Achilles tendon length. The Achilles tendon insertion into the calcaneus notch, and the soleus-Achilles MTJ were located with LOGIQ-9 and marked at the corresponding positions on the skin surface. The soleus-Achilles MTJ was defined as the location
where Achilles tendon (or so-called “free tendon”) (25) divides into soleus aponeurosis and gastrocnemius tendon. In an ultrasound image (Fig. 2, bottom), the MTJ was the intersection between the most distal muscle fascicles of the soleus muscle and the Achilles tendon. A line (defined as “line A”) was drawn along the Achilles tendon on the skin surface to connect these two points (Fig. 2, top). The probe was placed perpendicular to the skin and moved smoothly along line A, scanning in the sagittal plane with the subject relaxed. A special function of LOGIQ-9, called LOGIQView, was used to extend the field of view and register the images covering the full tendon length (Fig. 2, bottom). The scan was repeated 10 times to minimize the measurement error. ImageJ (National Institutes of Health, Bethesda, MD) was used to measure the distance from the soleus-Achilles MTJ to the calcaneus notch insertion along the tendon line of action (LOA) in the LOGIQView images.

**Measurement of moment arm.** A simple method was developed to measure in vivo the Achilles tendon moment arm, defined as the perpendicular distance from the center of rotation (inferior tip of the malleolus) (33) to the tendon LOA. Another line (defined as “line B”) from the center of rotation along the direction perpendicular to the tendon LOA was drawn on the skin surface (Fig. 2, top). The intersection of line A and line B was marked (Fig. 2, bottom), and the total distance from the center of rotation to the intersection projected in the sagittal plane (mₐ) was measured. The corresponding position of the intersection can be found in the LOGIQView image shown in Fig. 2, bottom, by using a marker to produce shadow in the ultrasound image. Then the distance from the intersection to the tendon LOA (mₗ) was measured in the LOGIQView image (Fig. 2, bottom), and the Achilles tendon moment arm (mₐ) was obtained by subtracting mₗ from mₐ (Eq. 1):

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mₐ = mₐ - mₗ
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**Measurement of the cross-sectional area.** Three points dividing line A equally into five sections were marked on the skin surface. The transverse plane at each point was scanned three times using standard ultrasonography. Great care was taken to make the probe and the scanning plane perpendicular to the Achilles tendon. The direction of the transducer was adjusted gradually to get the minimum area, which corresponded to the cross-sectional area (CSA). A representative image of the Achilles tendon cross section is shown in Fig. 3. ImageJ was used to measure the CSA. The average value from the nine measurements was calculated as the Achilles tendon CSA.

**Displacement.** The ultrasound probe was securely attached to the lower leg with a customized probe holder with the soleus-Achilles MTJ in sagittal plane clearly visualized at rest. The probe holder was made of thermoplastic material and custom fit to the shape of the probe at one end and the shape of the posterior lower leg at the other end so that the probe can be mounted to the leg securely using Velcro straps. The properly strapped holder helped keep the probe perpendicular to the skin. When fixed properly, the probe would not move relative to the leg during the plantar flexion effort. The ultrasound video data were continuously recorded at 50 frames/s during the entire plantar flexion effort once LOGIQ-9 received the trigger signal from the personal computer.

The ultrasound video data were analyzed frame by frame to determine the soleus-Achilles MTJ displacement using the MSAD block-matching method, which was proposed to quantify two-dimensional velocities in human (2) and regional myocardial dysfunction in mice (19) and which has been validated using simulation and experimental data. Briefly, this method specifies a tracking pixel block from a previous image frame, and it moves the block around in the following frame to find the best match by comparing the sum of
absolute differences. In the present study, a tracking pixel block size of \(\sim 2 \text{ mm} \times 2 \text{ mm}\) was used to search best match in a \(6 \text{ mm} \times 6 \text{ mm}\) surrounding region in the next frame. Parabolic interpolation was used to estimate subpixel displacements, and a three-tap median filter was used to smooth the estimated displacements between successive image frames. Four regions near soleus-Achilles MTJ were tracked in consecutive frames, and two representative ultrasound images without and with contraction were displayed together with the tracking result of one region in Fig. 4. The arrow points from the original position toward the displaced position based on MSAD block matching. Therefore, the average displacement of the four regions was regarded as the estimated displacement of the soleus-Achilles MTJ. The estimated MTJ displacement during a ramping up then down task was shown in Fig. 5, together with the corresponding ankle joint torque recorded.

Monitoring cocontraction and heel motion. EMG signal from the tibialis anterior muscle was used to monitor the activities of dorsiflexors. The EMG signals recorded at rest were saved as the baseline. The trials with the EMG amplitude higher than the baseline mean amplitude plus three times baseline standard deviation (6) were rejected. A marker was placed on the heel and the motion of the marker was captured by a high-resolution (1,920 \(\times\) 1,080 pixels) digital camcorder (model HDR-SR11, Sony). The camcorder was mounted on a tripod placed close to the marker on the heel. A ruler was placed at the same distance as the marker to show the scale of the picture. After each trial, the video was played back and trials with the heel motion \(>0.3 \text{ mm}\) were discarded.

Calculation of tendon Young’s modulus, stiffness, and mechanical hysteresis. The MTJ displacement was determined as the Achilles tendon elongation. Achilles tendon strain was obtained by dividing the tendon elongation by the initial tendon length. Tendon force was calculated by dividing the ankle joint torque by the tendon moment arm. Tendon stress was acquired by dividing the tendon force by the average tendon CSA. With the stress-strain curve for the ramp-and-hold task determined, the slope of the curve at 3% strain was defined as the tendon Young’s modulus. Similarly, the slope of the force-elongation curve at the displacement corresponding to 3% strain was defined as tendon stiffness. Force-displacement curves from the ramping up then down task were used to calculate the tendon mechanical hysteresis during loading and unloading, which was defined as the ratio of the area within the loop to the area below the loading curve (22). Representative force-displacement curves from both sides of the same subject are shown in Fig. 6.

Statistics. The various measurements were presented as means \(\pm\) SD. Paired Student’s \(t\)-test was used to test differences in the tendon dimension and biomechanical measures between the impaired and unimpaired sides. Statistical difference was set at a level of \(P < 0.05\).

RESULTS

Achilles tendon length and CSA. The Achilles tendon length measured in the impaired side (68.5 \(\pm\) 13.3 mm) was significantly longer (\(P = 0.04\)) than in the unimpaired side (64.4 \(\pm\) 11.4 mm; Fig. 7A). No significant difference (\(P = 0.19\)) was found between the Achilles tendon CSA measured in the impaired side (53.5 \(\pm\) 11.2 mm\(^2\)) and in the unimpaired side (56.1 \(\pm\) 11.2 mm\(^2\); Fig. 7B).

Tendon young’s modulus and stiffness. The estimated Achilles tendon Young’s modulus in the impaired side (136.4 \(\pm\) 38.1 MPa) was significantly lower (\(P = 0.005\)) than that in the unimpaired side (220.2 \(\pm\) 83.3 MPa; Fig. 7C). Similarly, the estimated Achilles tendon stiffness in the impaired side (105.0 \(\pm\) 14.6 N/mm) was significantly lower (\(P < 0.001\)) than that of the unimpaired side (184.8 \(\pm\) 48.4 N/mm; Fig. 7D).

Tendon mechanical hysteresis. Three subjects were excluded in this test due to obvious cocontraction during the unloading (ramping down) task. From the remaining seven
subjects, the tendon mechanical hysteresis calculated in the impaired side (19.6 ± 3.4%) was significantly higher ($P < 0.001$) than in the unimpaired side (6.8 ± 3.0%), as shown in Fig. 7E, indicating that the impaired limbs had lower energy efficiency than the unimpaired limbs.

**DISCUSSION**

In the present study, changes of Achilles tendon mechanical properties post stroke were evaluated in vivo and noninvasively using ultrasonography combined with biomechanical measurements. Results showed significant changes in the tendon dimension and biomechanical properties, including increased tendon length, lower tendon stiffness and Young’s modulus, and higher mechanical hysteresis in the impaired side compared with the matched unimpaired side of the stroke survivors.

It was important to monitor agonist-antagonist cocontraction and heel motion to achieve accurate evaluations in the present study. With potential dorsiflexors cocontraction, the estimated Achilles tendon force could be lower than the actual force. The subjects in this study were selected from those who still could control their plantar flexors and generate certain levels of plantar flexion torque ($\sim 15$ N·m) in their impaired lower limbs, and the subjects learned to minimize cocontraction after practicing the task multiple times with EMG signals from dorsiflexors displayed as visual feedback. As a result, the subjects managed to avoid cocontraction during the ramp-and-hold task, and there was not obvious cocontraction during the loading or holding phase, especially when the task was set at comfortable levels for the subjects. However, considerable cocontraction still occurred during the unloading phase of the ramp-then-down task in some subjects, and three subjects were excluded for the mechanical hysteresis evaluation due to the reason. In addition, some subjects also tended to lift their heels during plantar flexion effort, which could cause overestimation of tendon displacement. Therefore, the calcaneus motion was monitored using a digital camcorder, and the subject’s foot was tightly secured to the footplate. The subjects were also asked to perform the plantar effort tasks without lifting the heel. Moreover, some subjects tended to flex or extend their knees during the tasks. So the knee and the leg were mounted to the leg linkage tightly to minimize knee movement.

Results showed that tendon length was slightly longer (6%) in the impaired side compared with the unimpaired side. This was probably due to the fact that all the subjects were chronic stroke patients (average 12 yr after stroke) who had developed spasticity/contracture in the plantar flexor muscles for a long time. In related studies, shortening of muscle fascicles was found in the plantar flexors of chronic stroke patients (8, 9, 18). Roughly, the calf muscles and the Achilles tendon act like two springs in series, with the total length of the two springs determined by the leg length. With the calf muscles becoming stiffer and shorter poststroke, the middle point between the two springs (the soleus-MTJ) may shift proximally toward the calf muscles and the Achilles tendon gets elongated poststroke (Fig. 8).

In the process of muscle stiffening and shortening poststroke, it may apply stress onto the Achilles tendon, and creep may also be a mechanism involved in the potential Achilles tendon elongation process. Of note is that such a
possible adaptation poststroke involves relatively low level of loading, which is much lower than the failure loads in previous failure testing of Achilles tendon (36). Although no significant difference was found in the tendon CSA between the impaired and unimpaired sides in the present study, the average CSA in impaired side was slightly smaller (5%) than that of the unimpaired side. The difference might have become significant over a larger sample.

Decreased tendon Young’s modulus (38%) and stiffness (43%) were found in the impaired side of the stroke survivors in this study. Reductions of patella tendon Young’s modulus and stiffness were reported in spinal cord-injured subjects compared with able-bodied controls (23). Although the amount of reductions were greater than what were found in our study, our comparisons were done between the closely matched impaired and unimpaired sides of the same patients poststroke. In addition, the tendon Young’s moduli calculated in the present study were lower than values reported in some other studies with muscles under MVC (21, 22, 24), which may be related to the submaximal contractions within the comfortable torque-generation range of the stroke survivors in this study, considering that the stress-strain relation has an overall nonlinear curve with lower slope in the lower stress (“toe region”) of the tendon stress-strain curve. In the present study, Young’s moduli and stiffness were calculated at 3% strain for fair comparison, which were consistent with those in the corresponding region of the stress-strain curve reported in previous studies (23, 25, 29).

Increased mechanical hysteresis was found in the impaired side (19.6%) compared with the unimpaired side (6.8%). Both values are within the range (3–38%) that has been reported based on tensile testing of isolated specimens, and both values are also comparable to the value of gastrocnemius tendon (18%) reported in Ref. 22. The difference between different studies might be due to the different tendon segments and measurement approaches. Higher hysteresis values showed more energy was dissipated during the loading-unloading cycle, indicating lower energy efficiency in the impaired limbs.

The method in this study could only be used to evaluate patients with the ability of repeatedly performed controlling submaximal isometric plantar flexion tasks, with the subjects trained to perform the tasks following the torque target while avoiding cocontraction. For patients who cannot generate sufficient plantar flexion torque, electric stimulation may be an alternative method for tendon property evaluation (20).

Conclusion. In the present study, Achilles tendon dimension and mechanical properties of impaired and unimpaired sides of 10 hemiparetic stroke survivors were evaluated. Increased tendon length, decreased stiffness and Young’s modulus, and increased mechanical hysteresis were found in the impaired side. These changes may reflect adaptations to the changes of calf muscle fascicles poststroke. A better understanding of the adaptations and changes of the tendon biomechanical properties may help us gain insight into spasticity/contracture and motor impairment poststroke, and this may facilitate development of rehabilitation procedures.

GRANTS
The authors acknowledge the support of the National Institutes of Health.

REFERENCES


