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Effect of stretching training on the viscoelastic properties of human tendon structures in vivo

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Kubo, Keitaro, Hiroaki Kanehisa, and Tetsuo Fukunaga. Effect of stretching training on the viscoelastic properties of human tendon structures in vivo. J Appl Physiol 92: 595–601, 2002; 10.1152/japplphysiol.00658.2001.—The purpose of this study was to examine whether stretching training altered the viscoelastic properties of human tendon structures in vivo. Eight men performed the stretching training for 3 wk. Before and after the stretching training, the elongation of the tendon and aponeurosis of medial gastrocnemius muscle was directly measured by ultrasonography while the subjects performed ramp isometric plantar flexion up to the voluntary maximum, followed by a ramp relaxation. The relationship between the estimated muscle force (Fm) and tendon elongation (L) during the ascending phase was fitted to a linear regression, the slope of which was defined as stiffness of tendon structures. The percentage of the area within the Fm-L loop to the area beneath the curve during ascending phase was calculated as a flexibility index. Flexibility index decreased significantly after stretching training (−13.4 ± 4.6%). On the other hand, the stretching training produced no significant change in stiffness but significantly decreased hysteresis from 19.9 ± 11.7 to 12.5 ± 9.5%. The present results suggested that stretching training affected the viscosity of tendon structures but not the elasticity.

THE USE OF STRETCHING EXERCISES to improve flexibility is a widespread practice among competitive and recreational athletes. Numerous stretching studies employing this test have documented increases in the maximal joint range of motion after stretching exercises (5, 14, 26). However, the findings obtained by the technique could be affected by factors such as a raised muscle pain threshold. Alternatively, Toft et al. (20) pointed out that the passive tension measurements would be more objective than the range of motion measurements, because psychological factors did not interfere with results. They found a 36% decrease in passive tension of plantar flexors after a 3-wk stretching program. However, the mechanisms for the acute and chronic changes in joint range of motion and/or passive tension remain ambiguous.

It is widely conjectured that increasing flexibility will promote better performances during various movements (1, 26). For example, Wilson et al. (26), who applied a damped oscillation technique to determine the stiffness of the upper limbs, showed that the rebound bench press performance enhancement observed consequent to flexibility training was caused by a reduction in stiffness of muscle-tendon units, increasing the utilization of elastic strain energy during the rebound bench press lift. Inversely, Rosenbaum and Henning (17) observed reductions of the rate of force development and electromyograph (EMG) amplitudes, and an increase in EMG latencies, after static stretching of the triceps surae. Taken together, these previous results would indicate that the stretching training made the viscoelastic properties of tendon structures change. However, no attempt has been made to investigate the influences of stretching on the properties of human tendon structures in vivo.

Recent reports have shown that ultrasonography can be used to determine the stiffness and hysteresis of human tendon structures in vivo (10, 11, 13). Furthermore, our laboratory showed that the static stretching of plan- tar flexors for 10 min significantly decreased the stiffness and hysteresis of tendon structures in medial gastrocnemius (MG) muscle (10). Hence, we applied this technique to determine the magnitude of elongation in the tendon structure of human MG muscle before and after static stretching training for 3 wk. The purpose of the present study was to investigate the effects of static stretching training for 3 wk on the viscoelastic properties (stiffness and hysteresis) of human tendon structures in vivo. A brief account of this work has been presented previously in abstract form (8).

METHODS

Subjects. Eight healthy men [age 24.6 ± 1.8 (SD) yr, height 173.0 ± 7.1 cm, weight 75.4 ± 14.6 kg] participated as subjects. When the data were collected, the subjects had been...
participating in recreational sports but did not have any
experience in strength training or flexibility training pro-
grams. The subjects were fully informed of the procedures to
be utilized as well as the purpose of this study. Written
informed consent was obtained from all subjects. This study
was approved by the office of Department of Sports Sciences,
University of Tokyo, and complied with their requirements
for human experimentation.

Measurement of EMG. EMG activity was recorded during
the ramp isometric contraction (measurement of tendon
properties) and passive stretch (measurement of flexibility).
Bipolar surface electrodes (5 mm in diameter) were placed
over the bellies of MG, lateral gastrocnemius, soleus, and
tibial anterior (TA) with a constant interelectrode distance of
25 mm. The positions of the electrodes were marked on the
skin by small ink dots. These stained dots ensured the same
electrode positioning in each test during the experimental
period. EMG signals were transmitted to a computer (Mac-
tosh Performa 630, Apple) at a sampling rate of 1 kHz. EMG
was full-wave rectified and integrated for the duration of the
contraction and passive stretch to give integrated EMG
(iEMG).

Measurement of flexibility. To assess flexibility, the joint
angle and the passive torque were measured during a passive
stretch of the triceps surae muscles. The subject lay prone on
a test bench, and the waist and shoulders were secured by
adjustable lap belts and held in position. The subjects did not
warm up before the stretch maneuver. The right ankle joint
was set at 0° (anatomic position) with the knee joint at full
extension, and the foot was securely strapped to a footplate
connected to the lever arm of the dynamometer (Myorex,
Asics). The platform of the dynamometer, which was at-
tached to the sole of the subject’s foot, was moved to 25°
of dorsiflexion with a constant velocity of 5°/s. The passive
torque (Nm) during the stretching was detected by the dyna-
rometer. The measured torque (TQ) during isometric plantar
flexion up to maximum (Fig. 1). A marker (X) was placed
between the skin and the ultrasonic probe as the
landmark to confirm that the probe did not move during
measurements. The cross-point (P) between superficial apo-
neurosis and fascicles did not move. Therefore, the displace-
ment of P (L) is considered to indicate the lengthening of
the deep aponeurosis and the distal tendon (10, 11).

The measured torque (TQ) during isometric plantar flexion
was converted to muscle force (Fm) by the following equation

\[ Fm = k \cdot TQ \cdot MA^{-1} \]

where k is the relative contribution of the physiological
cross-sectional area of MG within plantar flexor muscles (3)
and MA is the moment arm length of triceps surae muscles at

Measurement of viscoelastic properties of tendon structures.
Tendon structures behave as a nonlinear viscoelastic struc-
ture. The constant slope is supposed to represent stiffness of
the collagenous material from which the tendon is con-
structured. In this study, “elasticity” is defined as the stiffness
(see below). On the other hand, the loading and unloading
curves during cyclic tensile test produce a loop (hysteresis).
In this study, we will use the term “viscosity” to refer to the
hysteresis.

The posture of the subject and setup were similar to that
for the measurement of flexibility as described above. The
right ankle joint was set at 0° anatomic position. Before the
test, the subject performed a standardized warm-up and
submaximal contractions to become accustomed to the test
procedure. The subject was instructed to develop a gradually
increasing force from relaxation to maximal voluntary con-
traction (MVC) within 5 s, followed by a gradual relaxation
within 5 s. The task was repeated two times per subject with
at least 3 min between trials. Torque signals were analog-to-
digital converted at a sampling rate of 1 kHz (MacLab/8, type
ML780, AD Instrument) and analyzed by a personal com-
puter (Performa 630, Macintosh). The measured values that
are shown below are the means of two trials.

A real-time ultrasonic apparatus (SSD-2000, Aloka) was
used to obtain a longitudinal ultrasonic image of MG at the
level of 30% of the lower leg length, i.e., from the popliteal
crease to the center of the lateral malleolus. The ultrasonic
images were recorded on videotape at 30 Hz, synchronized
with recordings of a clock timer for subsequent analyses.
The tester visually confirmed the echoes from the aponeurosis
and MG fascicles. The point at which one fascicle was at-
tached to the aponeurosis (P) was visualized on the ultra-
sonic image. P moved proximally during isometric torque
development up to maximum (Fig. 1). A marker (X) was
placed between the skin and the ultrasonic probe as the
landmark to confirm that the probe did not move during
measurements. The cross-point (P) between superficial apo-
neurosis and fascicles did not move. Therefore, the displace-
ment of P (L) is considered to indicate the lengthening of
the deep aponeurosis and the distal tendon (10, 11).

The measured torque (TQ) during isometric plantar flexion
was converted to muscle force (Fm) by the following equation

\[ Fm = k \cdot TQ \cdot MA^{-1} \]
0° of ankle joint, which was estimated from the lower leg length of each subject as described by Visser et al. (22).

As reported in previous studies using animal and human cadavers (e.g., Ref. 27), the Fm-L relation in the tendon structure was curvilinear, consisting of an initial region (toe region) characterized by a large increase in L with increasing force and a linear region immediately after the toe region. In the present study, therefore, the Fm and L values above 50% of MVC were fitted to a linear regression equation, the slope of which was adopted as stiffness (10, 11).

The Fm-L curves during the ascending and descending phases of force development produced a loop. In the present study, the area of each of the curves under both the ascending and descending phases was calculated. Then, the ratio of the area within the Fm-L loop (elastic energy dissipated) to the area beneath the curve during ascending phase (elastic energy input) was calculated as hysteresis (10).

The test-retest correlation coefficient was 0.90 for stiffness and 0.86 for hysteresis. The coefficient of variation was 5% for stiffness and 11% for hysteresis.

**Static stretching training.** The subjects were randomly assigned to stretch the plantar flexor muscles on one leg while the opposite side served as a control. Two sessions, one in the morning and one in the afternoon, were performed on a daily basis for 20 consecutive days.

During the stretching, EMG activities of all the muscles tested were very small (<1% of MVC), which confirmed the lack of contribution from the contractile component to the measured resistance to stretch. The relationship between passive torque and ankle joint angle during the passive stretch is shown in Fig. 3. In the control side, there were no significant differences in the passive torque values at any ankle angles. For the trained side, the passive torque values at all ankle angles decreased significantly after the stretching training. The flexibility index value decreased significantly from 1.43 ± 0.33 to 1.24 ± 0.30 Nm/° (−13.4 ± 4.6%; P = 0.007).

Figure 4 shows the relationships between Fm and L. In the control and trained sides, there were no signif-
significant differences in \( L \) values at any force levels, and no significant changes in MVC and stiffness were found. For the trained side, on the other hand, the \( \%\text{MVC}-L \) loop became significantly smaller after stretching training (Fig. 5). The measured viscoelastic parameters are shown in Table 1. The stretching training produced no significant change in stiffness (\( P = 0.621 \)) but significantly decreased hysteresis (\( -37.2 \pm 22.2\% \), \( P = 0.009 \)). In addition, there were no significant differences in the activation levels (iEMG) of each plantar flexor muscle before and after training (Table 2). Furthermore, we also confirmed that little cocontraction of the dorsiflexor muscle (TA) occurred during plantar flexion.

DISCUSSION

Before interpreting the results obtained, however, we must draw attention to the limitations and assumptions of the methodology followed. To calculate \( F_m \), we estimated moment arm length and relative contribution of MG to the plantar flexor muscles in terms of physiological cross-sectional areas. The variation in moment arm length and relative contribution of MG among subjects might have caused the large variability in the measured parameters. Moment arm length and physiological cross-sectional areas of respective muscles of each subject would be necessary for an “accurate” absolute muscle force determination. In the present study, however, we aimed to study whether the

| Table 1. Measured variables before and after stretching training |
|---------------------------------|---------|---------|---------|
| MVCL, Nm | Trained | Control | Trained | Control |
| Before | After | Before | After |
| 131 ± 17 | 132 ± 20 | 130 ± 19 | 128 ± 18 |
| Flexibility index, Nm/deg | 1.43 ± 0.33 | 1.24 ± 0.30* | 1.40 ± 0.29 | 1.42 ± 0.32 |
| \( L_{\text{max}}, \text{mm} \) | 25.1 ± 4.4 | 25.2 ± 3.8 | 25.0 ± 3.9 | 24.7 ± 4.3 |
| Stiffness, N/mm | 28.1 ± 4.9 | 27.4 ± 4.8 | 27.7 ± 4.5 | 28.2 ± 4.8 |
| Hysteresis, % | 19.9 ± 11.8 | 12.5 ± 9.5* | 18.6 ± 10.2 | 19.1 ± 11.4 |

Values are means ± SD. MVC, maximal voluntary contraction; \( L_{\text{max}}, \text{maximal tendon elongation} \). *Significantly lower than before, \( P < 0.05 \).
tendon properties changed after the stretching training. In addition, there were no significant differences in the activation levels (iEMG) of each triceps surae muscles before and after stretching training (Table 2). We also confirmed that little cocontraction of dorsiflexor muscle (TA) occurred during plantar flexion (Table 2). Therefore, we considered that this muscle force calculation, based on these assumptions, would be valid to study the changes of the tendon properties after the stretching training.

In any human isometric tests, the joint angle has been assumed to be constant without the joint angle being directly monitored. Thus, because isometric contraction of muscle about a joint will produce more or less angular joint rotation in the direction of the intended movement, the resulting tendon and aponeurosis displacement is the result of displacement attributed to both joint angular rotation and contractile tensile loading. Magnusson et al. (13) demonstrated the importance of accounting for even small amounts of joint motion: despite a rigid frame that was adjusted separately for each subject, the average plantar flexion motion was 3.6°, which resulted in an overestimation of the displacement by up to ~30%. However, it seems reasonable to suppose that there was no difference in this kind of error between before and after stretching training. In the present study, therefore, we may say that a little overestimation of the displacement in the ankle joint would not affect the present result.

Many previous studies have demonstrated that the maximum range of motion, i.e., flexibility, increased after the stretching training (5, 14, 26). Alternatively, Toft et al. (20) reported that the passive torque of ankle joint decreased after the stretching training for 3 wk. They suggested that the passive torque measurements would be more objective than the range of motion measurements, because psychological factors did not interfere with results. In the present study, therefore, we adopted the passive torque measurements for assessing the flexibility, instead of the measurements of maximum range of motion. The present result showed that flexibility index value decreased significantly after the stretching training (~13.4%). Similarly, the flexibility increase observed in response to the stretching program was within the 11.1–24.6% increase in flexibility of a number of lower body muscle groups, as reported by Wallin et al. (23). Furthermore, in the present study, because during passive dorsiflexion the EMG amplitude was below 1% of MVC (data not shown), it is unlikely that muscle activity contributed significantly to passive torque. Therefore, it is obvious that the static stretching training of ankle joint for 3 wk increases the flexibility of plantar flexor muscles.

There are many sources of the passive torque during passive stretch test: the joint capsule, movement and extension of ligament, synovial fluid movement, and elongation of the connective tissue within the muscle belly. Several researchers have suggested that the major contributors to passive tension are the extensibility of the connective tissue elements in parallel with the muscle fibers, i.e., parallel elastic component (7, 12, 15). The passive tension is influenced by a lengthening deformation of the connective tissues of the endomysium, perimysium, and epimysium of the muscle belly (4). Although all three components of the connective tissues that package the muscle belly contribute to the resistance when a muscle is passively stretched, the relatively large amount of perimysium is considered the tissue that is the major contributor to extracellular passive resistance to stretch (16). From the findings of Purslow (16), the perimysial collagen network, referring to as the parallel elastic component, played a role to prevent overstretching of the muscle fiber bundles. Therefore, the present result implies that the stretching training does not affect the elasticity of tendon structures, i.e., series elastic component, but does affect that of the connective tissue elements in parallel with the muscle fibers, i.e., parallel elastic component.

Recently, our laboratory showed that the static stretching of plantar flexors for 10 min made the tendon structures in the MG muscle more compliant (10). Therefore, it is possible to substantiate the hypothesis that the stretching training alters the stiffness of tendon structures. In the present study, however, no significant change in stiffness of tendon structures was found after the stretching training. Magnusson et al. (13) demonstrated that repetitive static stretches in human skeletal muscle, each lasting 90 s, yielded an immediate decrease in passive tension but that the tension returned to baseline within 1 h. Our laboratory also reported the acute change in stiffness after the static stretching, but this change was relatively small (~8.9%; Ref. 10). Furthermore, we observed that flexibility index was unrelated to the stiffness of tendon structures (9). Considering these findings, it seems reasonable to suppose that the stretching training increases the flexibility but not the elasticity of tendon structures.

Many previous studies on the effect of stretching exercise on the properties of muscle and tendon have been limited to animal studies (18, 19, 21, 24). For example, Viidik (21) reported that stretching the tail

Table 2. iEMG values during ramp isometric contraction

<table>
<thead>
<tr>
<th></th>
<th>MG</th>
<th>LG</th>
<th>Sol</th>
<th>TA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trained</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>1.40 ± 0.39</td>
<td>1.52 ± 0.41</td>
<td>1.37 ± 0.39</td>
<td>0.10 ± 0.03</td>
</tr>
<tr>
<td>After</td>
<td>1.38 ± 0.41</td>
<td>1.55 ± 0.25</td>
<td>1.39 ± 0.35</td>
<td>0.11 ± 0.02</td>
</tr>
<tr>
<td>Control</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Before</td>
<td>1.37 ± 0.36</td>
<td>1.48 ± 0.30</td>
<td>1.45 ± 0.35</td>
<td>0.12 ± 0.02</td>
</tr>
<tr>
<td>After</td>
<td>1.39 ± 0.39</td>
<td>1.45 ± 0.27</td>
<td>1.44 ± 0.42</td>
<td>0.10 ± 0.03</td>
</tr>
</tbody>
</table>

Values are means ± SD given in mVs. iEMG, integrated electromyograph; MG, medial gastrocnemius; LG, lateral gastrocnemius; Sol, soleus; TA, tibial anterior. There were no significant differences.
tendon of the rat increased its compliance. Recent studies showed that mean extension of wallaby tail tendons increased slowly during the fatigue test but much faster just before rupture (24). The force-length relation in the tendon structure is curvilinear, consisting of an initial region (toe region) characterized by a large increase in the length with increasing force and a linear region immediately after the toe region (27). The toe region has been assumed to be due to a structural change in collagen fiber organization from a cramped or wavy pattern to a more straightened, parallel arrangement (21). The crimp gives rise to the toe region of the length-tension relation of tendons (21). From the findings of Stromberg and Wiederhielm (18), the collagen fibers follow a wavelike course in the unstrained tendons, but they become aligned or parallel with increasing stress. These studies, which have attempted to establish the influence of stretching on the tendon properties, have mainly investigated the effects of “acute” stretching programs rather than “long-term or chronic” stretching.

In the present study, the hysteresis of tendon structures decreased significantly after the stretching training. Also, we observed an acute decrease in the hysteresis of tendon structures after 10 min of static stretching (10). Frisen et al. (2) and Vidik (21) also showed that repeated cyclic stretches of rat tendons decreased hysteresis, suggesting a reduction of energy dissipation in the tissues after stretching. The mechanisms, which resulted in the decreases of hysteresis after acute and long-term stretching, are unknown. At least for the lowered hysteresis observed in the present study, a change in the structure of the tendons might be involved. However, future work is needed to clarify this phenomenon.

The hysteresis, i.e., the area within the loop, represents the energy loss as heat due to internal damping, whereas the area under the unloaded curve is the energy recovered in the elastic recoil. Hence, the hysteresis of tendon structures should be taken into account when estimating the dynamics of the muscle-tendon complex during human movements. For example, the value of 20% obtained in our measurements (before stretching training) would be representative of the percent energy dissipated as heat in loading-unloading cycles of the tendon during several everyday-life activities. As a result of observations on human in vivo, Wilson et al. (26) reported that flexibility training for the upper limbs induced a significant increase in work during the initial concentric portion of the rebound bench press lift. Also, Dintiman (1) found that the sprint performance was improved when a stretching regimen was included with regular sprint training. These led us to speculate that stretching may be an effective way to increase reused energy during exercise involving a stretch-shortening cycle, by reducing the hysteresis.

In conclusion, the stretching training produced no significant change in stiffness of tendon structures, but it significantly decreased hysteresis. The present results suggested that the static stretching training affected the viscosity of tendon structures but not the elasticity. However, we have no definite information on how the changes induced by stretching training in tendon properties are correlated to performances during stretch-shortening cycle exercises. Further investigations are needed to clear up this point.

REFERENCES