Assessment of functional series elastic stiffness of human dorsiflexors with fast controlled releases

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De Zee, Mark, and Michael Voigt. Assessment of functional series elastic stiffness of human dorsiflexors with fast controlled releases. J Appl Physiol 93: 324–329, 2002. First published March 8, 2002; 10.1152/japplphysiol.00696.2001.—The series elastic stiffness (SES) of the human dorsiflexors was investigated in vivo with the fast controlled release method in 8 subjects. The maximum moment of a voluntary contraction (66 ± 17 Nm) was significantly higher than the maximum moment with electrical stimulation of tibialis anterior (34 ± 16 Nm). At an ankle moment of 34 Nm produced with either voluntary or electrical stimulation, we found a significantly different SES of 219 ± 54 and 149 ± 54 Nm-rad⁻¹, respectively. It is proposed that this is due to the fact that, during voluntary contraction, more elastic tissue parallel with each other is involved, because of coactivation of the extensor hallucis longus, extensor digitorum longus, and tibialis anterior. This shows that, for a functional assessment of the SES of the dorsiflexors, one has to include the toe extensors, which is possible with the fast controlled release method. Additionally, our results demonstrated that the SES of the human dorsiflexors at moment levels up to about isometric maximum did not reach an asymptote at which the stiffness is independent of moment, i.e., the series elastic component of the dorsiflexors is during daily activities loaded for the greatest part in the nonlinear part of the stress-strain function.

The series elastic stiffness (SES) of active muscles has a significant influence on both the control and the economy of movement. For making a quantitative evaluation of the functional significance of SES during human movement, accurate and reliable quantitative measures of SES in vivo in humans have to be developed.

The mechanical properties of the tendinous tissue in the human tibialis anterior measured in vivo by use of a method based on ultrasonography have recently received a lot of attention (5, 6, 10–13). This has given important new information about this muscle in vivo. However, for a functional evaluation of the SES of the human dorsiflexors, the ultrasonography method is not sufficient. The ultrasonography method has limitations that do not allow for studies of whole muscles and muscle groups but only of parts of muscles and superficial muscle structures. Therefore, only distal tendon and aponeurosis parts of superficial muscles have been investigated so far with this method.

During natural movement, however, joint moment is generated by muscle synergies, i.e., more than one muscle, and both proximal and distal series elastic component (SEC) structures are involved. Therefore, to get a more functional measure of SES in a muscle synergy controlling a joint, it is necessary to include all the involved series elastic tissue in the synergy in the measurement. It is possible to achieve this in vivo with the fast controlled release method (4, 7). This method is based on muscle shortening at a very high but constant speed. The shortening should be completed before the first reflexes arrive at the antagonist muscles, and the speed should be well above the maximum shortening speed of the muscle to minimize muscle fiber shortening. By recording the decline in moment as a function of joint rotation corresponding to shortening of the active synergy, the SES can be measured and expressed in newton-meters per radian, which we consider to be a useful functional measure for evaluation of the influence of series elasticity in human movement. This measure does not require knowledge about the lengths of the tendon moment arms of the individual muscles, which is a methodological advantage because it simplifies the measurement procedures. However, to translate the rotational measures of SES into linear measures of SES of the individual muscles, detailed information about tendon/aponeurosis dimensions, tendon moment arms, and force sharing is needed.

The aim of this study is to quantify the SES in the human ankle dorsiflexors by using the fast controlled release method with a custom-designed high-pressure hydraulic actuator. In addition to voluntary activation of the dorsiflexors, we also applied electrical stimulation to the tibialis anterior. This allowed us to measure the SES of tibialis anterior alone and to compare our results with data from the literature, especially the study by Maganaris and Paul (11).
METHODS

Equipment. A Servo-controlled custom-designed high-pressure hydraulic actuator (MTS-systems 215.35, 230 Bar) was used for the fast controlled-release experiments (16). The foot of the subject was firmly strapped to a foot adapter with large cable ties. The position of the foot adapter was adjusted such that the anatomical axis of the ankle coincided as well as possible with the center of rotation of the actuator. The releases were imposed over a 30° angular excursion between 20° plantar flexion and 10° dorsiflexion with an angular velocity of about 15 rad/s. For the calculation of the ankle joint angular accelerations and joint moments, the foot adapter was instrumented with linear load cells (Kistler, Slimline) to measure the force components parallel and vertical to the footplate of the foot adapter and accelerometers (Kistler, K-SHEAR Piezotron) to measure the corresponding acceleration components (see Fig. 1). The position of the rotary actuator was monitored with an angular displacement transducer (Transtek DC ADT series 600). Surface electromyograms were recorded from the tibialis anterior and soleus muscles. The signals were sampled at 4,000 Hz.

Experimental procedure. The experimental protocol and procedures were approved by the local ethical committee. Three women and five men gave their informed consent to participate in the experiments. They sat in a chair with a knee angle of ~160° and a hip angle of ~100°. After two recordings necessary for correction (see below), two sets of 20 releases were performed. In the first set of 20 releases, the subject produced a steady level of voluntary dorsiflexion moment before each release. The subject could see the produced moment level on an oscilloscope. The initial moment was varied between ~5 Nm and maximum effort, which was presented as a mark on the oscilloscope. In the second set of 20 releases, the moment was produced by electrical stimulation of the tibialis anterior at different levels (stimulator: Isolator 11, Axon Instruments). Electric activation of the tibialis anterior was produced by 1 s of percutaneous stimulation at a frequency of 100 Hz using pulses with a duration of 100 μs. Electrodes (Medicotest neurostimulation electrodes) were placed over the tibialis anterior muscle. The current was varied between the motor threshold and the point at which the subject could tolerate no higher current.

Correction for inertia and passive stiffness. The moment signal had to be corrected for inertia and passive stiffness of the antagonists. The correction method is described in detail in De Zee and Voigt (4). In short, the correction for inertia was based on the assessment of the angular acceleration plus two additional recordings. The first extra recording was a slow release with an angular velocity of 0.1 rad/s with the muscle passive, which gave the passive moment-angle curve. The second extra recording was a fast release (15 rad/s) also with the muscle passive. From this recording, a transfer function H was obtained with the angular acceleration as input and the moment as output. The fast controlled release measurements were then first corrected for inertia by using the acceleration signal. Subsequently the passive moment was subtracted from the total moment. Figure 2 gives an impression of the correction procedure applied.

Correction for CC shortening. If the force decreases below the initial isometric moment, the contractile component (CC) starts to shorten with a certain speed. The speed of shortening (expressed in rad/s) can be calculated from the Hill's force velocity relation (2, 7)

\[ \dot{\phi}_{cc} = \frac{M_0 - M_{cc}}{M_{cc} + nM_0} \]  

where \( \dot{\phi}_{cc} \) is the CC shortening speed, \( M_{cc} \) is the measured moment after correction, and \( M_0 \) is the initial isometric
moment just before the release. The values for the parameters have been adopted from the literature (8, 17), where b = 1.2 and n = 0.12. The shortening speed was integrated to obtain the angular excursion corresponding to CC shortening, and this number was subtracted from the measured ankle angle (see also Fig. 2).

**Shift of release curves and curve fitting.** Because the individual releases start at different initial moment levels, but at the same joint angle, the moment and angle are shifted between the measurements. Therefore, in line with our previous study on plantar flexor SES (4), the corrected, submaximal release curves were angle shifted so the initial moment level corresponded to the moment level on the trial with the highest (M₀,max) initial moment (see Fig. 3). The shifted curves were averaged and fitted with the following nonlinear function (14)

$$\phi = (aM + b)^{0.5} + c$$  \hspace{1cm} (2)

where $\phi$ is angle and M is moment. A nonlinear function gives good fits because tendinous tissue shows nonlinear behavior up to stresses of 30 MPa (1, 15). The values of the constants a, b, and c were obtained by a nonlinear regression using the Marquardt-Levenberg algorithm.

**Statistics.** A paired t-test was used to compare between the voluntary muscle activation and electrical stimulation. A value of $P < 0.05$ was considered significant. Average values are presented as means ± SD.

**RESULTS**

The mean age of the 8 subjects was 35 ± 6 yr, body mass 76 ± 17 kg, and height 174 ± 8 cm. The maximum measured M₀,max just before the release was 66 ± 17 Nm with the voluntary isometric muscle activation and 34 ± 16 Nm with electrical stimulation ($P < 0.001$). See for individual values Table 1. An example of an electromyogram recording during a voluntary contraction is presented in Fig. 4. It shows that the stretch reflex of the soleus is too late to influence the stiffness calculation.

Figure 5A presents 20 release curves obtained during voluntary activation from one subject. The signals are corrected for inertia, passive stiffness, and CC shortening. Figure 5B presents the results of the shift of the release curves. Finally, an average of the curves in 5B is presented in Fig. 5C (solid line).

The average curves of Fig. 5C were fitted with Eq. 2. The dashed line in Fig. 5C is an example of the fit. For all subjects, good fits were obtained (voluntary muscle activation, $R^2 = 0.98 ± 0.025$; electrical stimulation, $R^2 = 0.91 ± 0.051$). To calculate the SES for the situation with voluntary muscle activation and the situation with electrical stimulation, Eq. 2 was differentiated. SES is moment dependent, so for the sake of comparison the SES had to be calculated at the same moment for the two situations. The moment of 34 Nm was chosen, which was the average maximum in the case of electrical stimulation. The SES at this moment of 34 Nm was significantly different between the trials, with voluntary muscle activation and electrical stimulation (voluntary muscle activation, SES₃₄Nm = 219 ± 54 Nm·rad⁻¹; electrical stimulation, SES₃₄Nm = 149 ± 54 Nm·rad⁻¹; $P < 0.001$).

Figure 6 shows families of dorsiflexor SES vs. moment curves ($0 - M₀,max$) from all the subjects during

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**Table 1. Individual subject data.**

<table>
<thead>
<tr>
<th>Sex</th>
<th>Age, yr</th>
<th>Body mass, kg</th>
<th>Height, cm</th>
<th>M₀,max, Nm</th>
<th>SES at 34 Nm, Nm·rad⁻¹</th>
<th>M₀,max, Nm</th>
<th>SES at 34 Nm, Nm·rad⁻¹</th>
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<tbody>
<tr>
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<td>29</td>
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<tr>
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<td>149§</td>
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<td>8</td>
<td>17</td>
<td>54</td>
<td>16</td>
<td>54</td>
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</tbody>
</table>

M₀,max, maximal measured moment just before the release; SES, series elastic stiffness; F, female; M, male. Values marked with * and § are significantly different (for both values, $P < 0.001$).
voluntary (Fig. 6A) and electrical (Fig. 6B) activation, respectively, obtained by differentiation of the fitted series elastic release curves. It is clearly seen that the stiffness of SEC between 0 and \( M_{0,\text{max}} \) never reaches an asymptote, i.e., a constant stiffness.

**DISCUSSION**

**Comparing voluntary muscle activation with electrical stimulation.** The maximum measured moment during voluntary dorsiflexion was much higher than during electrical stimulation. This is most probably because during voluntary dorsiflexion not only the superficial tibialis anterior was active but also the toe extensors, i.e., extensor digitorum longus and extensor hallucis longus. In the case of electrical stimulation, the moment was produced solely by the tibialis anterior. This also implies that during voluntary activation of the dorsiflexors more elastic tissue lying parallel is involved than during electrical stimulation, which should be reflected in the measured stiffness. Our results indeed showed a significantly higher stiffness during voluntary activation compared with electrical stimulation at a certain moment. The total SES at a certain moment is, however, not just proportional with the total cross section of the elastic tissue, but also dependent on the stress. In case of electrical stimulation, the whole moment is taken up by the elastic tissue in series with the tibialis anterior muscle fibers, whereas in the case of voluntary activation the moment is distributed over the elastic tissue of more than one muscle. Hence, during electrical stimulation, the involved elastic tissue is working at a higher stress in the nonlinear stress-strain curve than the involved elastic tissue during voluntary contraction. Although the involved elastic tissue during voluntary contraction is working at lower stresses with lower stiffnesses, the total stiffness increases because of the larger total

Fig. 4. A: angle of ankle joint during a fast controlled release recorded with the angular displacement transducer. 0 rad, actuator position where the ankle joint was in neutral position. B: electromyogram (EMG) of the soleus during a fast controlled release. C: EMG of the tibialis anterior with a voluntary contraction of the dorsiflexors. Note the stretch reflex in soleus around 60 ms, which occurred after the end of the fast controlled release. Hence, the stretch reflex will not influence the stiffness measurement.

Fig. 5. A: 20 release curves of subject 1 with different initial moments produced with voluntary activation. B: release curves in A are shifted to the left in such a way that they coincide with the release curve with the highest initial moment. C: solid line is the average of the curves in B, and dashed line is the nonlinear fit. 0 rad, angular position where the ankle joint was in neutral position.
cross-sectional area of the elastic tissue. This implies that in vivo the total SES of the human dorsiflexors is dependent not only on which muscles are active but also on where on the stress-strain curve the involved elastic tissue is working.

In De Zee and Voigt (4), in a similar experiment on the human plantar flexors, we obtained different results. The maximum moment was higher for the moments produced with electrical stimulation of the triceps surae than for the moments produced with a voluntary contraction. The SES at a certain moment (100 Nm in this case) was not significantly different between the two situations. This shows that the triceps surae is the dominant muscle for both the moment production and the SES in the human plantar flexors. Hence, the influence of the deeper plantar flexors is small, contrary to the dorsiflexors, where the toe extensors play a significant role beside the tibialis anterior. Our results demonstrate that in the case of the human dorsiflexors one cannot neglect the toe extensors in a functional assessment of the SES.

Comparing stiffness values with those from the literature. A recent article written by Maganaris and Paul (11) gives a platform for comparison. They used a combination of the ultrasound technique and magnetic resonance imaging to measure tendon stiffness in the human tibialis anterior in vivo. Electrical stimulation was used to activate the tibialis anterior to make sure that only this muscle was active. This makes it possible to make a comparison with our data obtained with electrical stimulation. Maganaris and Paul (11) found an average tendon stiffness of 161 N/mm at a maximum average moment of 25 Nm. We can convert this stiffness value to a rotational stiffness by using the average moment arm of the tibialis anterior, measured by Maganaris and Paul, of 36 mm.

The result is 209 Nm·rad⁻¹. This number is based on measurement of the free tendon part alone. Our value calculated for an electrical stimulation-produced moment of 25 Nm is 127 Nm·rad⁻¹, which is based on all the elastic tissue in series with the muscle fibers in the tibialis anterior. It is not surprising that our value is much lower than the value of free tendon part alone, because our value includes the free part of the tendon plus the elastic tissue in series with the free part of the tendon. The total stiffness of different elastic structures in series with each other is always lower than the stiffnesses of the individual elastic structures. This shows that for a functional assessment of SES one cannot restrict the measurement to the free part of the tendon but should include all the elastic tissue in series with the muscle fibers.

Nonlinearity of angle-moment curve. If the different curves with different $M_0$ originate from the same amount of elastic tissue, one would expect the curves to coincide when shifting the curves to the left as shown in Fig. 3. As can be seen from Fig. 5B, this indeed was the case. This phenomenon is further supported by research done on animals; the aponeurosis is already totally involved at low forces because of the lateral shear transmission of force (9), and the muscle fibers active at low forces are not concentrated in one part of the muscle but are distributed (3). Because of this, we believe that the average curve in Fig. 5C expresses the total moment-angle curve of the SEC for a particular muscle synergy.

Nonlinear fits gave the best results when the average curves were fitted, which indicates that during maximum isometric dorsiflexions the SEC is still operating in the toe region. This is in line with the study by Maganaris and Paul (13), in which they concluded the same for the tibialis anterior tendon and aponeurosis.

In Fig. 6, it can be seen that for some subjects the stiffness is nonzero or even negative for a zero moment. This is due to a less successful fit in the lower part of these curves, probably because of some noise in the signal of the lower moments. One way to improve this might be to use a weighted nonlinear regression, so that low values of the moment are given less weight.

In conclusion, in the present study, we were able to measure the SES of the tibialis anterior with the fast controlled-release method and electrical stimulation. The measured stiffnesses were comparable with earlier measurements reported in the literature. The big advantage, however, of the fast controlled-release method over the existing methods is the possibility to measure the functional SES during a voluntary contraction, which also takes into account both proximal and distal series elastic structures and the elastic tissue of whole synergies. This information is a prerequisite to understanding the functional role of the dorsiflexor SES in vivo in humans.
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REFERENCES