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Viscoelastic properties of muscle-tendon units

The biomechanical effects of stretching *

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ABSTRACT

Most muscle stretching studies have focused on defining the biomechanical properties of isolated elements of the muscle-tendon unit or on comparing different stretching techniques. We developed an experimental model that was designed to evaluate clinically relevant biomechanical stretching properties in an entire muscle-tendon unit. Our objectives were to characterize the viscoelastic behavior of the muscle-tendon unit and to consider the clinical applications of these viscoelastic properties.

Rabbit extensor digitorum longus and tibialis anterior muscle-tendon units were evaluated using methods designed to simulate widely used stretching techniques. Additionally, the effects of varying stretch rates and of reflex influences were evaluated. We found that muscle-tendon units respond viscoelastically to tensile loads. Reflex activity did not influence the biomechanical characteristics of the muscle-tendon unit in this model.

Experimental techniques simulating cyclic stretching and static stretching resulted in sustained muscle-tendon unit elongations, suggesting that greater flexibility can result if these techniques are used in the clinical setting. With repetitive stretching, we found that after four stretches there was little alteration of the muscle-tendon unit, implying that a minimum number of stretches will lead to most of the elongation in repetitive stretching. Also, greater peak tensions and greater energy absorptions occurred at faster stretch rates, suggesting that the risk of injury in a stretching regimen may be related to the stretch rate, and not to the actual technique. All of these clinically important considerations can be related to the viscoelastic characteristics of the muscle-tendon unit.

Stretching advocates have published numerous articles,[8, 21, 36, 37, 50–52, 65] books,[3, 9, 35, 61] and stretching technique descriptions[19, 35, 44, 59, 69] that address the positive effects of stretching. These authors feel that stretching can be beneficial in several areas. In sports medicine, stretching has been recommended to prevent injury[8, 9, 17, 19, 23, 24, 31, 35, 37, 45, 61] and to improve performance.[3, 9, 22, 43, 45, 61, 63] In rehabilitation, stretching is used as a method to regain range of motion and to improve overall function.[7, 23, 45, 51, 55, 61] DeVries[20, 21] has recommended stretching as a method to reduce muscle soreness following activity. Stretching has even been put forth as a way to enhance one's general well-being.[3, 9]

General categories of stretching techniques include ballistic, static, passive, and proprioceptive neuromuscular facilitation. Ballistic stretching is characterized by repetitious bouncing movements where the muscles are rapidly lengthened and immediately returned to resting length. Ballistic exercises are considered to be less beneficial than other regimens.[3, 8, 9, 17, 19, 61, 63] The inhibitory effects of stretch reflex on ballistic stretching are frequently mentioned as a reason for the reduced efficacy of this technique.[9, 19] The possibility of injury to the muscle-tendon unit, especially in rehabilitation, has also been associated with ballistic stretching.[3, 8, 37]

Static stretching refers to the technique of stretching a

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o alongamento pode ser benífico em várias áreas. na medicina esportiva, alongamento é recomendado para prevenir lesões, e melhora a performance. na reabilitação, o alongamento é usado como método para aumentar a amplitude de movimento e melhoria de todas funções.

muscle group to a length just short of causing pain. This length is then maintained for an extended period of time (anywhere from 6 to 60 seconds has been recommended).[3,9,55,59] Advocates of this technique feel that the stretch reflex influences will be minimized by gentle motion and the absence of pain. Static exercises are recommended in some cases because they are easy to perform and have little associated injury risk.[3,8,9,21]

Passive stretching can be applied to both ballistic and static techniques. In passive stretching, a partner applies additional external stretching force.

Several techniques use tenets of proprioceptive neuromuscular facilitation (PNF).[35,52,61] PNF was established by Kabat[40] to aid in the therapy of paralyzed patients. PNF uses the concepts of reflex activation and inhibition, and reversal of antagonists in attempts to achieve improvement in active and passive motion.[40,67] Holt[35] applied similar concepts to athletic exercises. He called his technique the 3-S system (Scientific Stretching for Sport). This technique includes voluntary contraction of the muscle group being stretched, contraction of the antagonist muscle group, and finally, passive static stretch of the target muscle group. Voluntary isometric contraction of the stretched muscle group leads to self-inhibition through Golgi tendon organ reflexes. Voluntary isometric contraction of the antagonist muscles causes a subsequent reflex inhibition of the muscle groups being stretched. The muscles are then passively stretched to a new lengthened position.

Throughout this discussion on techniques one finds a great emphasis placed on reflex activity. The goal of many stretching routines is to positively influence reflex activity in order to bring about increased muscle-tendon unit length. However, the basic biomechanical properties that influence how the muscle-tendon unit responds to stretch have received little attention in the sports medicine literature.

Tension in muscle can be considered to be comprised of active and passive components as defined in classical muscle physiology.[5,12,13,33,34,47] Any stretching effects mediated by reflex activity must involve the active component. Tension in the passive component may also respond to stretch, independent of the active component. The passive component resides in part in the connective tissue of the muscle-tendon unit.

As with most biological tissues, muscle is thought to act viscoelastically.[1,17,25,27,28] Therefore, muscle is considered to have both elastic and viscous properties. Elasticity implies that length changes, or deformations, are directly proportional to the applied forces, or loads. Viscous properties are characterized as time-dependent and rate change-dependent, where the *rate* of deformation is directly proportional to the applied forces. In biomechanics, true elasticity is represented by Hooke's model of a perfect spring (Fig. 1A), and viscous elements are represented by Newton's model of a hydraulic piston known as a dashpot (Fig. 1B).[26,68] The perfect spring represents the reversible nature of elastic materials in which deformation depends solely on the applied force. The dashpot represents viscous materials in

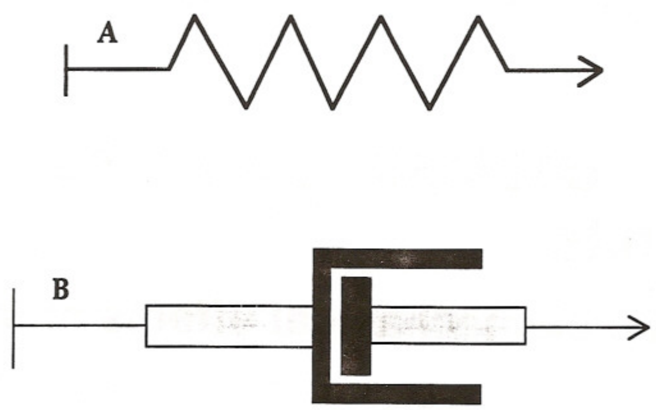

Figure 1. The two fundamental elements of a viscoelastic model. A, the Hookean body: This perfect spring provides a model for elastic behavior. Deformation is proportional only to force. B, the Newtonian body: Here, a hydraulic piston, or dashpot, containing viscous fluid provides a model for viscous behavior. The velocity of dashpot displacement is directly proportional to force.

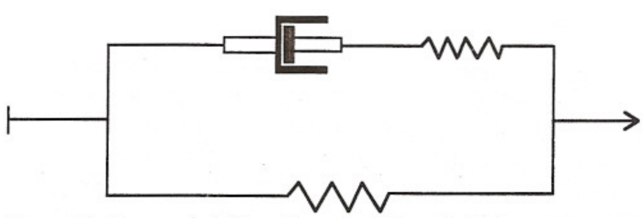

Figure 2. Representative viscoelastic model. The spring and dashpot can be combined in a series or parallel to demonstrate viscoelastic behavior.

which the rate and duration of the application of forces influence the length changes.

Most biologic tissues act with both elastic and viscous properties; in other words, viscoelastically. Biomechanical models attempt to represent viscoelastic characteristics by combining springs and dashpots in various configurations, such as in Figure 2.[26-28,68]

Certain properties are characteristic of viscoelastic materials. If a viscoelastic material is stretched and then held at a constant length, the stress, or force, at that length gradually declines. This decline is called stress relaxation.[26] The behavior of the material is both viscous, because the tension decreases with time, and elastic, because the specimen maintains some degree of tension.

Creep is another viscoelastic property and is characterized by continued deformation at a fixed load.[27] The deformation asymptotically approaches a new length based on the viscoelastic elements of the material.

Hysteresis is the variation in the load-deformation relationship that takes place between loading and unloading a specimen.[27] For viscoelastic materials greater energy is absorbed during loading than is dissipated during unloading.

Viscoelastic materials also demonstrate the property of strain rate dependence. Strain rate-dependent materials ex-

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hibit higher tensile stresses at faster strain rates. Strain rate dependence occurs because slower strains allow for greater relaxation to take place within the tested material.

There have been many studies addressing the viscoelastic nature of muscle[1, 5, 16, 25, 28, 32–34, 48, 49, 56, 57, 60] and tendon.[2, 10, 11, 18, 46, 58, 64, 70–72] Most muscle studies have been aimed at describing the viscoelasticity in active muscle. Few studies are applicable to the behavior of the muscle-tendon unit under passive stretch. The practice of stretching is so widespread in sports, the arts, and in general fitness programs that it is remarkable that no more basic science background exists to guide the rational use of stretching. Realizing this basic lack of understanding, we developed a model that evaluates basic biomechanical properties of the entire muscle-tendon unit in such a way that the findings could have clinical applications. Our objectives in studying this model were the following: 1) to characterize the viscoelastic properties of a muscle-tendon unit, and 2) to consider the clinical applications of these viscoelastic properties with regard to stretching.

MATERIALS AND METHODS

This study was divided into three parts. Part I examined the characteristics of repeated stretching of muscle-tendon units to a predetermined length. In Part II, muscle-tendon units were stretched repeatedly to the same tension and held at a fixed length. In Part III we explored how varying the stretch rates and denervating the muscles affected muscle-tendon unit stretching. A total of 20 tibialis anterior (TA) and 40 extensor digitorum longus (EDL) muscle-tendon units from New Zealand White rabbits (2.75 to 4.5 kg) were used in this study. The TA and EDL were chosen for their ease of access, and because their distinct tendons of origin and insertion simplify length measurements. The TA and EDL are comprised of predominantly fast twitch fibers,[6] and the EDL crosses more than one joint, both of which are characteristics of commonly strained muscles.[14, 15, 30] The TA lies anterior to the EDL in the rabbit hindlimb, originating from the anterolateral surface of the tibia, and inserting medially at the base of the second metatarsal. The EDL originates from the lateral femoral condyle and inserts, after passing under the cruciate ligament on the dorsum of the rabbit's foot, onto the phalanges of the four digits. A branch from the common peroneal nerve innervates both muscles.

Prior to dissecting and testing, all animals were anesthetized with an intramuscular injection of a mixture of ketamine, 100 mg/kg (Vetalar, Parke-Davis Laboratories, Morris Plains, NJ), xylazine, 12.5 mg/kg (Rompun, Mobay Corporation, Shawnee, KS), and acepromazine, 3 mg/kg (PromAce, Aveco Company, Ft. Dodge, IA). Each hindlimb was shaved, and an anterior incision was made extending from 2 cm above the knee to the level of the metatarsals on the dorsum of the foot. The TA and EDL muscle-tendon units were exposed and the distal tendons were transected near their insertion points.

Kirschner wires were placed transversely through both the femoral condyles and the proximal tibia, and were used to immobilize the hindlimb in a rigid frame. The frame was, in turn, mounted onto the specimen table of an Instron materials testing machine. An Instron Table Model Universal Testing Instrument (Instron Corp., Canton, MA) was used in Parts I and II of the study, and a Model 1321 Instron Servohydraulic Materials Testing System was used in Part III. After clamping the distal tendon to the load cell of the Instron, each specimen was stretched to an initial tension of 1.96 N (Fig. 3). The length of the specimen at 1.96 N was defined as the resting length, L_o. The tension of 1.96 N corresponds to the acceleration due to gravity acting on a 200 g weight. A 200 g weight provided the smallest accurately detectable measurement on the Instron chart recorder. Additionally, the muscle-tendon unit lengths at 1.96 N were similar to the lengths of the intact muscles in the rabbit at rest. Length measurements were made using precision dial calipers accurate to 0.1 mm.

A 37°C saline irrigation was used during all testing procedures to keep the muscle-tendon units moist and within physiologic temperatures. Care was taken to ensure that the neurovascular supply remained intact during testing, and specimens were eliminated from the study if there was any slipping of the tendons in the clamp. Additional anesthesia was administered throughout the study as needed.

Part I: Repeated stretching of muscle-tendon units to 10% beyond resting length

Eight EDL muscle-tendon units were used in Part I. The stretching regimen in the first part of our study was designed to simulate stretching to a certain position followed by an immediate return to the neutral position. This is similar to a cyclic stretching technique to a given length.

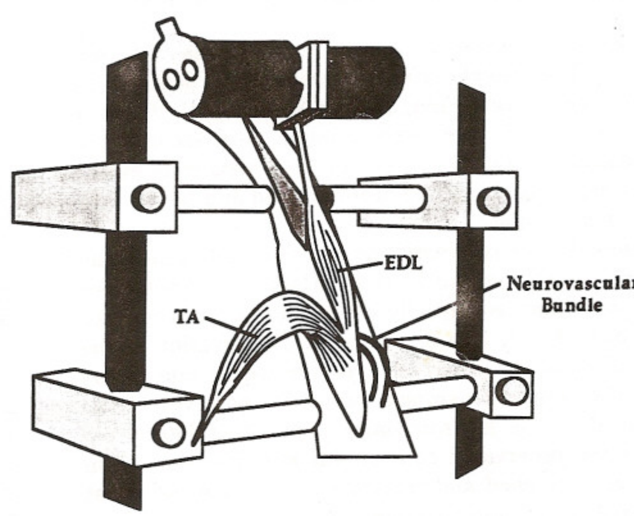

Figure 3. Immobilized rabbit hindlimb during stretching of the EDL muscle-tendon unit. The tendons of insertion are clamped to the Instron load cell. The neurovascular supply of the EDL and TA was maintained throughout testing except during the denervation studies in Part III.

The EDLs were stretched at a rate of 2 cm/min to 10% beyond L_o, and then immediately returned to L_o. Each cycle was repeated 10 times. A 10% length increase was selected because it provided a significant deformation in the "elastic" region of the load-deformation relationship and did not create any irreversible injury.[54] An Instron chart recorder documented tension and length changes for the stretched specimens.

Part II. Repeated stretching of muscle-tendon units to a set tension.

Twelve EDL muscle-tendon units were used in Part II. The EDLs were stretched from a starting tension of 1.96 N at a rate of 2 cm/min to a tension of 78.4 N. The tension of 78.4 N was selected because it also resulted in a significant deformation in the "elastic" region of the load-deformation relationship and did not create any irreversible injury.[54] Based on other studies 78.4 N is approximately 65% of the tension required to rupture a passively stretched EDL.[66]

Once the tension of 78.4 N was reached, stretching was discontinued, and the muscle-tendon unit was maintained at a fixed length for 30 seconds. After 30 seconds the tendon was returned to a tension of 1.96 N. This cycle was repeated 10 consecutive times for each specimen. This stretching regimen was designed to simulate the static technique in which an individual repeatedly stretches to a certain position, holds that position, and then returns to the initial resting position.

Length measurements were derived from chart recordings of the EDL load-deformation relationships.

Part III: Stretching of muscle-tendon units at varying rates and comparisons of innervated and denervated muscle

Twenty EDL and twenty TA muscle-tendon units were used in Part III. The stretching in this section was designed to investigate two areas. First, the biomechanical effects of varying the stretch rates were studied, and second, the contributions of reflexes in our animal models were evaluated by comparing stretching of innervated and denervated muscle-tendon units.

Each muscle was preconditioned by cyclically stretching the muscle to 10% beyond L_o at 2 cm/sec for 20 cycles. All muscles were then sequentially stretched to 10% beyond L_o at rates of 0.01, 0.1, 1, and 10 cm/sec. Denervation of the TA and EDL muscles was then performed in one leg by dividing the common peroneal nerve at the level of the lateral head of the gastrocnemius. The contralateral leg served as the innervated control. All muscles were again sequentially stretched and returned to 10% beyond L_o at rates of 0.01, 0.1, 1, and 10 cm/sec.

Data was captured on a Nicolet Model 3091 oscilloscope (Nicolet Instrument Corporation, Madison, WI) and stored on an IBM Personal System/2 Model 80 computer (IBM, Armonk, NY) using a Nicolet Waveform BASIC IBM/PC Software Package (Blue Feather Software, New Glarus, WI). The maximum force and energy absorbed by each muscle was measured for each stretch to 10% beyond L_o. Statistical analysis was performed using a multifactorial mixed model analysis of variance. .

RESULTS

Part I

Figure 4 depicts the results of repeated EDL stretching to 10% beyond L_o. There is a progressive decrease in tension with each stretch. Overall, there was a 16.6% decrease in the peak tension from the 1st to the 10th cycle. Most of the decrease in peak tension occurred early in the series. Duncan's multiple range test was applied to the series of peak tensions in order to establish statistical significance. Each of the peak tensions from the first, second, third, and fourth stretches showed a statistically significant difference ($P < 0.05$) from the other nine peak tensions. Each of the peak tensions for stretches 7 through 10 failed to show a statistically significant difference relative to the peak tensions of the immediately preceeding and immediately subsequent stretching cycles.

Part II

Figure 5 shows the relaxation curves that follow each stretch for the EDLs tested in Part II. There was a statistically significant ($P < 0.05$) difference between the relaxation curve following the first stretch and the other nine relaxation curves when analyzed using Duncan's multiple range test. The second relaxation curve also showed a statistically significant difference from the other nine curves. There were no statistically significant differences in Relaxation Curves 4 through 10 relative to each preceeding and subsequent curve. As a percentage of the initial length, the EDL increased in length 3.46% ± 1.08% following the 10 stretching cycles. Figure 6 depicts the lengthening of the muscle-tendon unit associated with each stretch as derived from the chart recorder.

Relaxamento ao stress

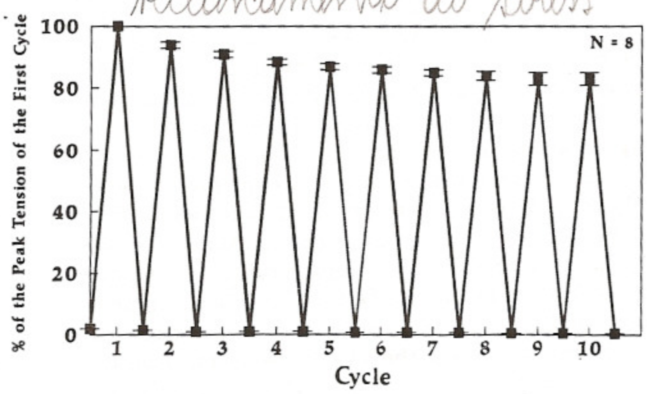

Figure 4. Tension curves of EDL muscle-tendon units repeatedly stretched to 10% beyond resting length. Each of the peak tensions for the first four stretches showed a statistically significant ($P < 0.05$) difference from the other peak tensions. The overall tension decrease was 16.6%.

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Relaxamento ao stress

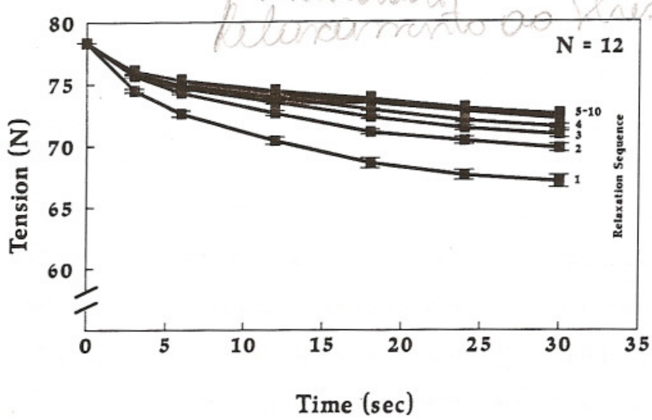

Figure 5. Relaxation curves for EDL muscle-tendon units stretched repeatedly to 78.4 N. The relaxation curves of the first two stretches demonstrated statistically significant differences from the other curves. There were no significant differences in curves 4 through 10.

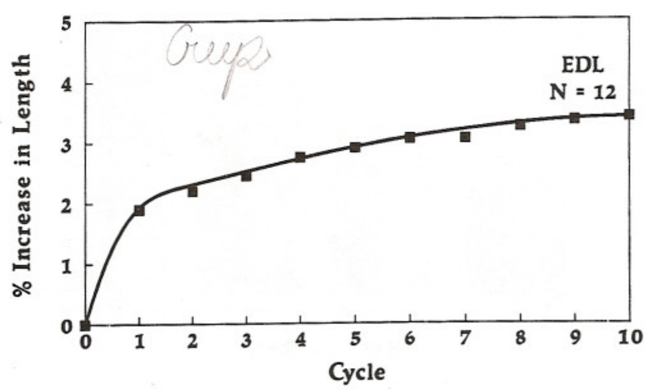

Figure 6. Graphic representation of EDL lengthening with repeated stretching to the same tension. Eighty percent of the length increase occurred during the first four stretches.

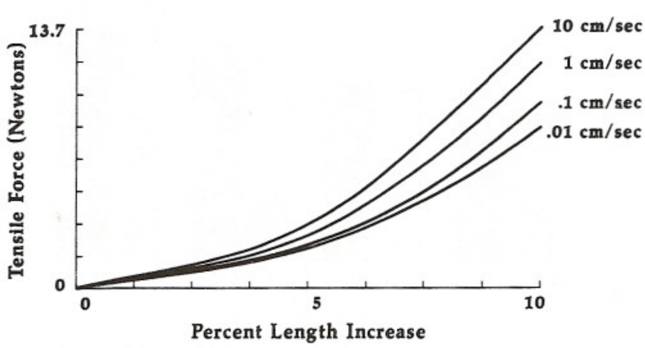

Figure 7. Representative force-length relationships demonstrating the effect of stretch rate on a single TA muscle-tendon unit.

Part III

Figure 7 depicts representative waveforms from a single TA muscle during the first series of stretches. Peak tensile forces were measured directly from these curves. Energy absorbed by the muscle-tendon unit during each stretch was determined by calculating the area beneath the curve. Without exception, all muscles tested in this part of the study (N = 40) exhibited increased peak tensile forces and greater absorbed energies when stretched at faster rates.

A summary of the biomechanical data is found in Table 1. Data are reported as least square mean ± least square standard error of measurement (LSM ± LSSEM). There were statistically significant differences ($P < 0.0001$) in the peak tensile forces and the energies absorbed at each stretch rate in each of the stretching cycle groups (first, innervated, and denervated).

Graphic representations of peak tensile force and energy absorbed in innervated and denervated TA and EDL muscle-tendon units are shown in Figures 8 and 9. The results for innervated and denervated muscle-tendon units showed almost no difference between the two test groups. A comparison of the effects of innervation and denervation revealed no significant difference between the two groups when analyzed statistically using a multivariate repeated measures analysis of variance. This held true for both the TA and EDL muscle-tendon units. The P value in comparing the peak tensile forces of the TA between the innervated and denervated groups was $P = 0.9328$. The corresponding P value for the EDL was $P = 0.9130$. The P value in comparing energy absorbed by the TA between the innervated and denervated groups was $P = 0.8294$. The corresponding P value for the EDL was $P = 0.5899$.

DISCUSSION

Experimental model

The approach taken in this study was to evaluate stretching in a manner in which we could apply our biomechanical findings to the clinical practice of stretching. There were three elements of the study design which were directed toward this approach.

First, we examined two experimental stretching methods that were similar to clinically well-known techniques. The repeated stretch to a 10% length increase is similar to a cyclic stretching technique with a stretch to the same length, e.g., toe-touching exercise. The stretch to a fixed tension and subsequent length maintenance is very similar to static stretching. Using these two methods we could examine biomechanical properties of clinically performed techniques.

Second, we examined the properties of the entire muscle-tendon unit. This approach provided better correlation to the clinical state than studying an isolated element of the muscle-tendon unit.

Third, the muscle-tendon unit was maintained in a near physiologic condition. With the exception of the denervated muscle-tendon units in Part III, the neurovascular supply of each specimen was unharmed during testing. As a result, there should have been little change in the homeostasis of the muscle.

While our techniques allowed for correlation to clinical experience, they also resulted in some differences from con-

TABLE 1
Biomechanical data

Stretching series	Peak tensil force (Newtons)				Energy absorbed (Newtons × cm)			
	0.01 cm/sec	0.1 cm/sec	1 cm/sec	10 cm/sec	0.01 cm/sec	0.1 cm/sec	1 cm/sec	10 cm/sec
First TA series ($N = 20$)	8.35 ± 0.31	9.25 ± 0.36	11.60 ± 0.42	13.81 ± 0.39	2.14 ± 0.09	2.38 ± 0.10	3.03 ± 0.14	3.76 ± 0.14
Second TA series:								
Innervated TA ($N = 10$)	8.35 ± 0.33	9.60 ± 0.35	12.21 ± 0.40	14.89 ± 0.45	2.18 ± 0.10	2.39 ± 0.11	3.13 ± 0.14	3.92 ± 0.17
Denervated TA ($N = 10$)	8.62 ± 0.33	9.69 ± 0.35	12.45 ± 0.40	14.81 ± 0.45	2.14 ± 0.10	2.53 ± 0.11	3.29 ± 0.14	4.15 ± 0.17
First EDL series ($N = 20$)	42.09 ± 0.16	44.85 ± 0.15	51.09 ± 0.13	56.65 ± 0.18	10.46 ± 0.07	11.04 ± 0.10	13.23 ± 0.10	17.85 ± 0.19
Second EDL series								
Innervated EDL ($N = 10$)	40.45 ± 0.09	43.01 ± 0.09	49.14 ± 0.23	54.64 ± 0.17	9.70 ± 0.06	10.24 ± 0.13	12.28 ± 0.13	16.66 ± 0.27
Denervated EDL ($N = 10$)	40.50 ± 0.09	43.22 ± 0.09	49.57 ± 0.23	55.03 ± 0.17	9.89 ± 0.06	10.76 ± 0.13	12.74 ± 0.13	17.55 ± 0.27

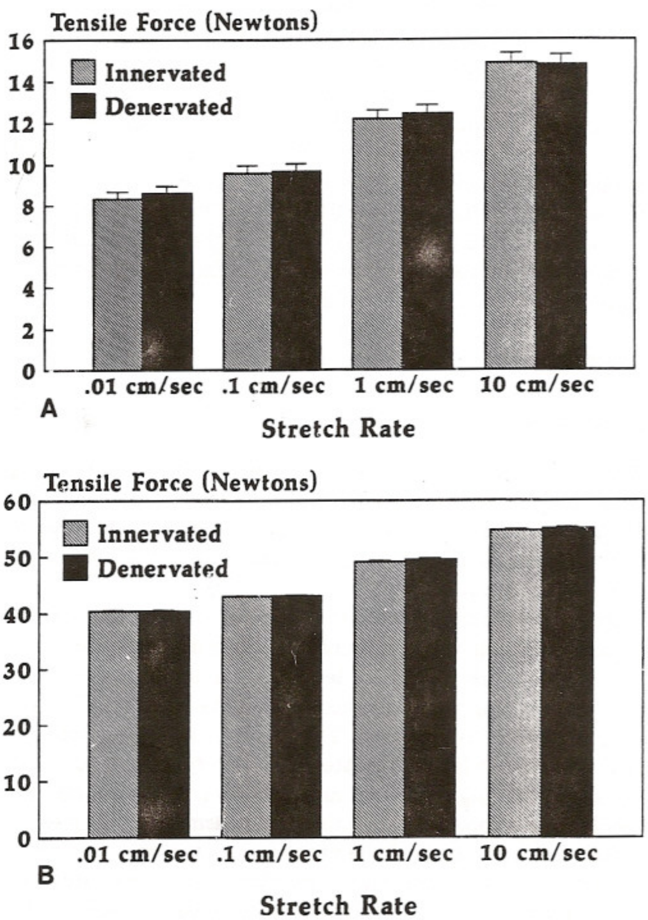

Figure 8. Peak tensile force in innervated ($N = 10$) and denervated ($N = 10$) TA muscle-tendon units (A), and innervated ($N = 10$) and denervated ($N = 10$) EDL muscle-tendon units (B). Values are LSM ± LSSEM. The effect of stretch rate is significant at the $P < 0.0001$ level. Muscle-tendon units which underwent acute denervation prior to stretching responded similarly to innervated units when pulled to a set length at increasing rates.

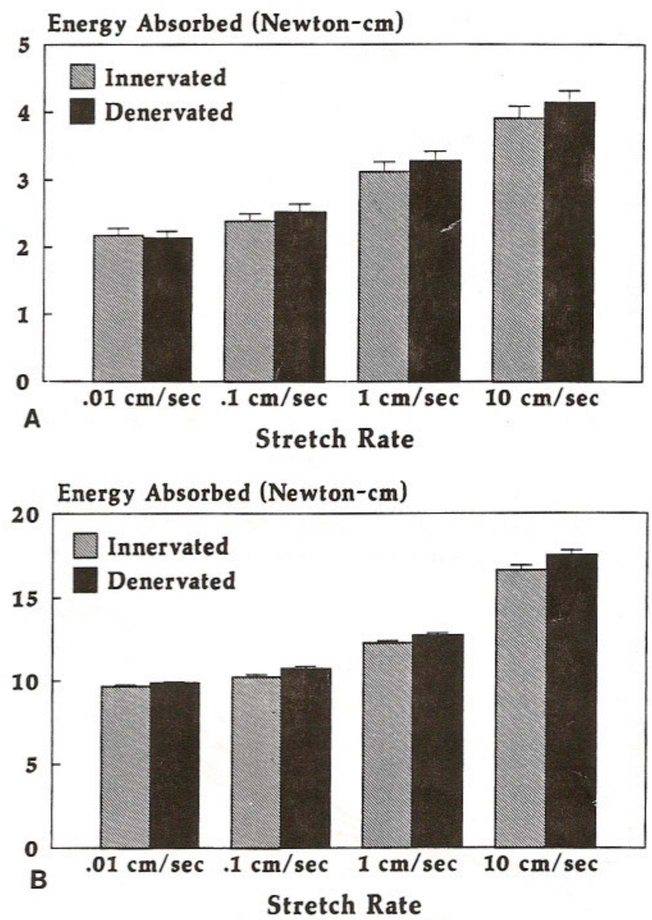

Figure 9. Energy absorbed in innervated ($N = 10$) and denervated ($N = 10$) TA muscle-tendon units (A), and innervated ($N = 10$) and denervated ($N = 10$) EDL muscle-tendon units (B). Values are LSM ± LSSEM. The effect of stretch rate is significant at the $P < 0.0001$ level. Acute denervation did not significantly alter the amount of energy absorbed when compared to innervated TA and EDL muscles.

ventional biomechanical studies. For example, certain accepted biomechanical conditions could not be evaluated. The standard biomechanical property of stress was not measured in the muscle-tendon unit because the cross-sectional area varies throughout the specimen. Evaluating load versus deformation, however, still provided important basic biomechanical information. Additionally, in Part III we elected to use stretch rates instead of strain rates. This allowed us to report the rate of elongation in clinically meaningful units (centimeters per second) instead of the conventional biomechanical units (percent per second).

Preconditioning refers to the practice of repeating stretch cycles prior to biomechanical testing of materials.[27] Preconditioning is frequently used in biomechanical testing to improve the reproducibility of data from a test specimen. In Part II of this study, the repetitive stretching cycles (Fig. 5) that would normally be considered preconditioning were studied for their biomechanical characteristics. This was done to simulate the clinical setting of stretching before an activity, which may actually be a form of preconditioning the muscle before activity.

In Part II relaxation was not extended for long periods of time as it often is in many biomechanical studies. The purpose here was to see the biomechanical effects in a situation similar to actual practice and not to determine the minimum value that the relaxation curve would approach. Similarly, the 10% stretch repeated 10 times simulated a physiologic stretching method an individual could perform a reasonable number of times. Thus, the testing techniques used in this study could evaluate certain biomechanical properties of stretching, but in a different fashion than in standard biomechanical tests.

Viscoelasticity of muscle-tendon units

The results of this study demonstrate the viscoelastic nature of the muscle-tendon unit. The time dependency of the load-deformation diagram is demonstrated in both Parts I and II. In Part I the decline in peak tension with each stretch (Fig. 4) shows that the stretching history is relevant to the stretched muscle-tendon unit. The decline in peak tension occurs because the viscoelastic property of stress relaxation leads to an internal change in structure of a specimen during each stretch.

In Part II, stress relaxation is demonstrated more graphically. Figure 5 shows the viscoelastic nature of the muscle-tendon unit in this series of relaxation curves. Figure 5 also demonstrates properties similar to preconditioning in other viscoelastic tissues.[27] With each stretch there is a gradual leveling off of the relaxation curve at a higher tension than the preceeding relaxation curve.

Although the actual property of creep was not measured in our study, elements of Part II demonstrated characteristics of creep. A constant tension was not maintained to cause a length increase, but varying amounts of tension up to a maximum (78.4 N) did lead to elongation. The length increase characteristics shown in Figure 6 were similar to creep properties, forming a curve toward a maximum deformation.

Part III of our study added additional evidence of the viscoelasticity of the muscle-tendon unit. We demonstrated that both peak tensile force and absorbed energy were dependent upon the rate of stretch applied (Figs. 8 and 9). From a biomechanical point of view, stretch rate dependency is explained by the amount of stress relaxation that can occur in a given amount of time. Slower stretches allow for a greater degree of stress relaxation to occur, resulting in lowered peak forces (Fig. 7).

By cyclically stretching muscles to 10% beyond L_o, we were able to demonstrate the viscoelastic phenomenon of hysteresis. Figure 10 shows a single TA muscle-tendon unit cyclically stretched to 10% beyond L_o at different rates. Energy, as measured by calculating the area beneath the load-deformation curve, is absorbed by the muscle-tendon unit during the loading process and dissipated during the unloading process. During any one stretch, the rate at which a muscle-tendon unit absorbs energy is different from the rate at which it dissipates energy. This creates a discrepancy between energy put into the system and energy released from the system. This difference could be accounted for by heat transfer and/or by internal changes within the ultrastructure of the muscle.

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Reflex effects

In addition to providing evidence of viscoelasticity, the third part of our study also validates our original assumption that the behavior of muscle in response to stretch can be explained by viscoelastic properties alone, exclusive of reflex effects. As our experimental model showed, the denervated muscles responded similarly to the innervated muscles for all parameters evaluated. We are not suggesting that the stretch reflex is nonexistent during muscle stretch, but in our model it appears that there are no significant force contributions from a stretch reflex. The reasons for this

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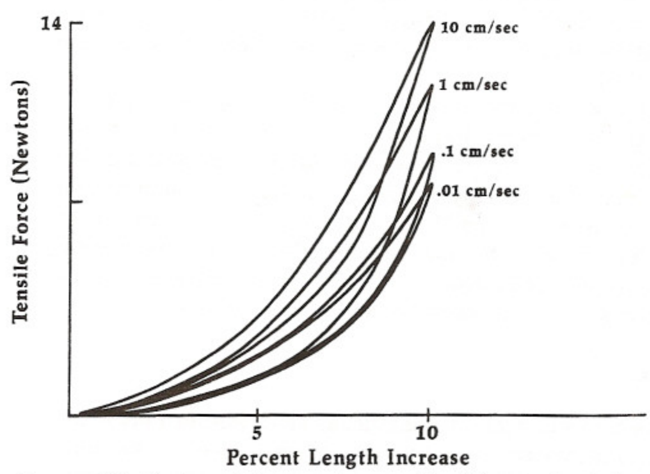

Figure 10. Hysteresis loops observed in TA muscle during loading and unloading at constant rates of 0.01, 0.1, 1, and 10 cm/sec.

may include a nervous system depression from our anesthetic mixture, or possibly that rabbit muscles have very little reflex activity. Regardless, it is apparent that denervation had no significant effect on muscle behavior. Therefore, the behavior of muscle in this model can be attributed to inherent viscoelasticity.

CLINICAL RELEVANCE

The results of muscle stretching in this study are similar to the results of other studies demonstrating the viscoelastic properties of tendons, ligaments, and other connective tissue systems.[2, 4, 10, 11, 18, 29, 41, 58, 64, 70–72] As in other materials composed of connective tissue, the viscoelastic properties of the muscle-tendon unit are responsible for the length increases that occur with stretching. These viscoelastic properties can be applied to several clinical situations.

Flexibility

The viscoelastic nature of muscle suggests that stretching will result in greater flexibility. Improved flexibility is considered to be related to an increased range of motion of a particular joint.[9, 17, 38, 42, 53, 62] Muscle-tendon units are considered to be the limiting structures preventing greater ranges of joint motion.[36, 39, 61] As shown in this study, especially in the creep characteristics demonstrated in Part II, stretching results in an increased length of the muscle-tendon unit. These length increases should not be rapidly reversible because of the viscous properties of the muscle-tendon unit. Applied clinically, stretching and the subsequent elongation of muscle-tendon units should allow for a greater range of motion of the joint that those muscle-tendon units cross. In other words, stretching should result in greater flexibility of a joint.

Stretching during rehabilitation is one important area in which viscoelastic principles of the muscle-tendon unit may be applied. Limited range of motion secondary to muscle tightness from immobilization or from muscular spasticity might be improved if the known viscoelastic elements are stretched.

Interestingly, the empiric impressions about stretching routines have some basis in viscoelastic principles. Static stretching, recommended by many for its reduction in reflex activity,[3, 8, 9] is actually a clinical example of stretch relaxation, as shown in Part II. Contract-relax exercises like those in PNF or 3-S techniques,[35, 40, 59, 67] also induce stress relaxation by applying tension to the viscoelastic elements during passive stretch and active isometric contraction.

Stretch rate

Some forms of cyclic stretching, as simulated in Part I, can lead to increased flexibility and reduced tensile stress on a stretched muscle-tendon unit as demonstrated by the reduction of the peak load with each stretch. The relaxation of the viscoelastic elements within each stretch may be responsible for this finding. The warnings that ballistic stretching can be dangerous can be explained through viscoelastic properties. Stretch reflexes may also contribute, but they are not considered in this model.

The viscoelastic property of stretch rate dependence, as exhibited in Part III, demonstrates why, in practice, ballistic stretching can be more dangerous than other routines. Ballistic stretching is similar to our cyclic stretching routine except the strain rate is fast and the tension is not held; the muscle reaches relatively high tension during the fast stretch and "bounces" back. Faster stretch rates result in greater tensions and more absorbed energy within the muscle-tendon unit for a given length of stretch (Figs. 8 and 9). Therefore, a ballistic stretch performed at a fast rate has a greater likelihood of causing a strain injury. In addition, since the muscle is not held at the higher tension, there is little chance to allow time-dependent stress relaxation or creep to occur to reduce the tension (at a given length change) or to increase the length (for a given applied force).

Stretching duration and number

The question of how much stretching is necessary to achieve the maximum length increase in the muscle-tendon unit needs to be addressed. How long should a static stretch be held? In theory, static stretching will result in indefinite lengthening of the muscle-tendon unit. However, the amount of stress relaxation that takes place after the first 12 to 18 seconds appears to be much less significant than the changes during the initial 12 to 18 seconds of stretch (Fig. 5).

The number of stretches one should perform is also unknown. Anywhere from 1 to 20 repetitions have been recommended for any particular stretching exercise.[3, 8, 9, 35, 61] In this model, the greatest changes in the muscle-tendon units occurred in the first four stretches for both experimental stretching regimens. When stretched to a set length in Part I, the only peak tensions that showed statistically significant differences from the other peak tensions were the first four.

With static stretching in Part II, 80% of the length increases in the EDLs occurred during the first four stretches (Fig. 6). The greatest change in relaxation curves occurred in the first four cycles (Fig. 5). The final six relaxation curves showed no statistically significant differences from each other. In theory, further stretches would bring about some length increases, but these increases would be small. Additionally, the magnitude of the stretching force or the duration of hold time may have an influence on the ideal number of stretches. However, a minimal number of stretches, four in our model, appear to be effective in bringing about most of the muscle-tendon unit lengthening.

CONCLUSIONS

The extensive clinical interest in stretching and the widespread use of stretching techniques in athletes, the arts, and general fitness programs are contrasted by the paucity of basic science information on stretching. This study presents a model in which clinically relevant biomechanical proper-

A propriedade viscoelástica da unidade músculo-tendão é responsável pelo aumento do comprimento durante o alongamento.

ties of the functional muscle-tendon unit can be studied. Based on our data, muscle-tendon units respond viscoelastically to tensile loads.

The viscoelastic properties of the muscle-tendon unit can be applied to several clinical stretching situations. Different stretching techniques resulted in muscle-tendon unit elongations, implying that greater flexibility can result if these techniques are used in the clinical setting. Also, greater peak tensions and greater energy absorptions occurred at faster stretch rates, suggesting that the risk of injury in a stretching regimen may be related to the stretch rate and not to the actual technique. Finally, it appears that a minimum number of stretches will lead to most of the elongation in repetitive stretching. All of these clinically important considerations are related to the viscoelastic nature of the muscle-tendon unit.

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Prop. visco elastica { organizados em
tempo depend. ⇓ força depend. ⇓ { série e paralelo

1 Relaxamento ao stress { material viscoelástico mantido em uma determinado comp → força/stress diminui

↳ visco → tempo
↳ elast → mantém uma tensão

2 Creep { material viscoelástico mantido sob uma força constante → continua a deformação.

3 Histeresis { liberação de energia devido a uma fricção interna do tecido

força e energia acumulada { são proporcionais ao alongamento { + devagar → resist along. → - força → - energia

ñ existe o papel tão determinante do sistema reflexo sobre a resistência ao alongamento.