*biodynamics*

# Viscoelastic stress relaxation in human skeletal muscle

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## ABSTRACT

MCHUGH, M. P., S. P. MAGNUSSON, G. W. GLEIM, and J. A. NICHOLAS. Viscoelastic stress relaxation in human skeletal muscle. *Med. Sci. Sports Exerc.*, Vol. 24, No. 12, pp. 1375–1382, 1992. Viscoelastic stress relaxation refers to the decrease in tensile stress over time that occurs when a body under tensile stress is held at a fixed length. The purpose of this study was to demonstrate viscoelastic stress relaxation in human skeletal muscle. Resistance to stretch (tensile force), hip flexion range of motion (ROM), and reflex contractile activity (IEMG) of the hamstring muscle group were measured during a passive straight leg raise. The testing protocol involved a first stretch to the maximum tolerated ROM with the lower extremity held at that point for 45 s (test 1). All 15 subjects tested (9 men, 6 women) had a stretch induced EMG response. The onset of a sustained EMG response occurred at a specific hip flexion angle in 10 subjects. These 10 subjects (6 men, 4 women) underwent a second straight leg raise stretch (test 2) to a ROM 5 degrees below the ROM at which the onset of EMG activity occurred in test 1. The stretch was held at this hip flexion angle for 45 s. There was a significant decrease in force at final ROM during the 45 s in test 1 (11.35 ± 1.75 N, $P < 0.0001$) and in test 2 (4.2 ± 1.55 N, $P < 0.05$). The percent decrease from the force at the respective final ROM was not significantly different between the tests (14.4 ± 2.2% in test 1 and 13 ± 2.3% in test 2). In test 1 there was a significant decrease over time in IEMG of 59.71 ± 16.01 $\mu V \cdot s$ ($P < 0.01$) which was not significantly correlated to the decrease in force. In conclusion, the data from test 2 demonstrates viscoelastic stress relaxation independent of detectable EMG activity. When the stretch was sufficient to elicit an EMG response the relaxation response was similar but had a larger absolute magnitude.

MUSCLE STRETCH, ELECTROMYOGRAPHY, TENSILE STRESS

Muscular flexibility has neurophysiological and mechanical components (19). It has been suggested that the myofibrils bear most of the resting tension in skeletal muscle (12). With elongation the tension in the muscle-tendon unit limiting range of motion is attributed to the viscoelastic components of the connective tissue (8,17,18,20). However, a contractile stretch reflex, originating from the muscle spindle organs in parallel with extrafusal muscle fibers, may also produce resistance to stretch (3,9).

The contribution of the stretch reflex to the resistance encountered during a stretch is unknown. Decreased reflex response to stretch has frequently been used to explain decreased resistance to stretch with various stretching techniques (4,5,9,16,21). In the terminology of Moore and Hutton (14) the four basic stretching techniques described are ballistic stretching, static stretching, contract relax stretching, and contract relax-agonist contract. The latter two techniques are referred to as methods of proprioceptive neuromuscular facilitation (PNF) (23). In an apparent paradox, some studies have demonstrated increased contractile responses in association with stretching techniques that provided greatest ROM (14,15).

The role of the mechanical component of muscular flexibility is more clearly understood (1,2,8,11,17,18,20,22). The mechanical response of muscle to stretch is characterized by viscous and elastic properties (6,22). The term viscoelastic is used to describe a tissue's loading response that shows a combination of elastic and viscous properties. Viscoelasticity is classically described by mechanical models using linear springs (elasticity) in series or in parallel with dashpots (viscosity) (6). In these models elastic deformation is instantaneous and proportional to the load and viscous deformation is proportional to the velocity of load application (6). Thus elongation is a function of load and rate of application.

When a body under tensile stress is held at a fixed length the tensile stress will decrease over time. This is referred to as stress relaxation (1,22) and with regard to human muscle duplicates the static stretch technique. Viscoelastic stress relaxation has been demonstrated *in vitro* in animal models (1,22) and with cadaver speci-

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mens (11) but lacks confirmation in human skeletal muscle.

The purpose of this study was to see whether viscoelastic stress relaxation could be demonstrated in human skeletal muscle, independent of a stretch-induced EMG response. Additionally, we sought to examine how the stretch induced EMG response affects the stress relaxation response to tensile stress.

## METHODS

**Testing protocol.** The response of the hamstring muscle group during a passive straight leg raise held at a fixed angle was studied. Subject consent was given in accordance with the policy statements of the American College of Sports Medicine. The stretch was applied to the right lower extremity of 15 subjects (9 men, 6 women) mean age 28.7 ± 1.2 yr. During the stretch the contralateral lower extremity was fixed to the table with a strap to limit pelvic rotation (Fig. 1). Hip flexion range of motion (ROM), stretch-induced EMG response of the hamstring muscle group (IEMG), and resistance to stretch (N) were measured. Each subject underwent two testing procedures. In test 1 the lower extremity was stretched to the maximum tolerated ROM and held at this point for 45 s. Maximum tolerated ROM represented the hip flexion angle at which the discomfort was such that the subject could not comfortably tolerate any further increase in ROM. All subjects demonstrated a stretch-induced EMG response during the stretch before final ROM was reached. For 10 subjects the onset of a sustained EMG response was identifiable. In test 2 the straight leg raise was brought to 5 degrees below the ROM at which the onset of EMG activity occurred in test 1 and held at this point for 45 s. The subjects with no identifiable point of onset of EMG activity demonstrated a stretch-induced EMG response even when the hip was flexed to 50% the ROM of test 1 and were not included in test 2. Ten subjects (6 men, 4 women) were used in test 2. We defined a decrease in tensile force over the 45 s period in test 2 (no sustained EMG response) as representing viscoelastic stress relaxation. The tester attempted to apply the same slow rate of stretch in both tests and between all subjects. The mean rate of stretch for each test is reported.

A 45-s relaxation period was chosen for three reasons. Previous *in vitro* work with animal models showed a plateau in the relaxation curve by 30 s (22). In clinical practice stretching protocols generally involve single stretch durations of less than 45 s. Since the stretch was applied manually we wanted to ensure that tester fatigue did not affect the ability to maintain a constant ROM. A digital display of ROM enabled the tester to maintain a constant ROM.

With this testing protocol it was possible that quad-

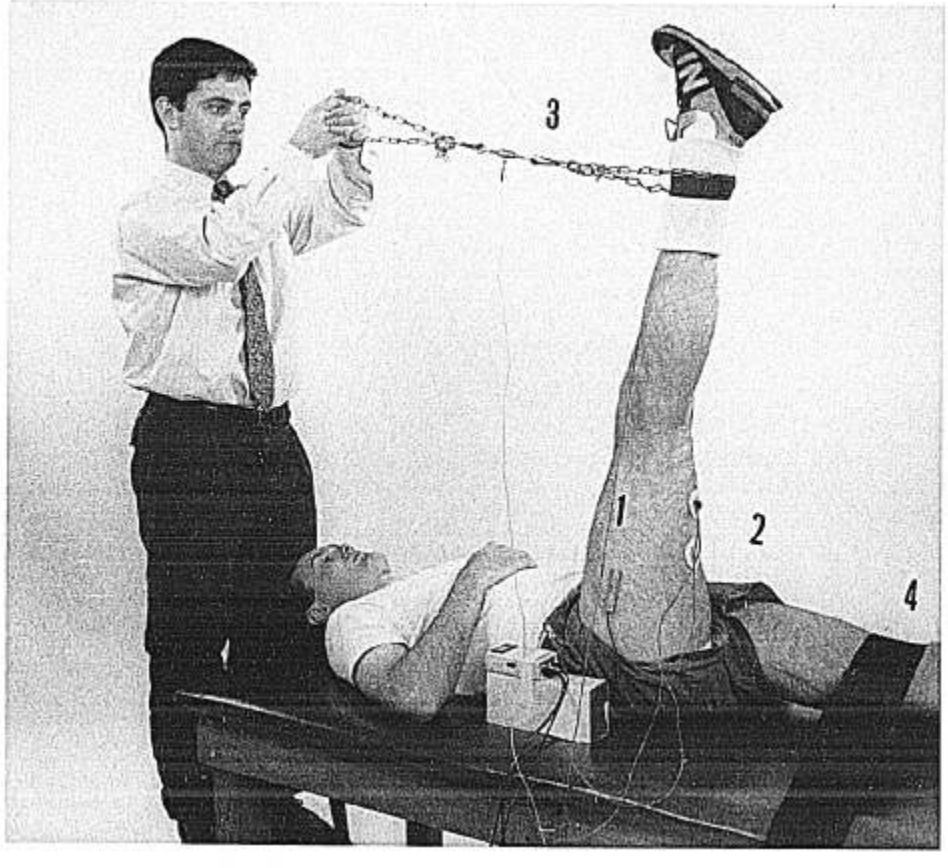


**Figure 1—Test apparatus for applying stretch. (1) Electrogoniometer attached to the hip. (2) Surface electrodes over medial hamstring. (3) Load cell at 90 degrees to the leg. (4) Strap to limit pelvic motion.**

riceps contraction could have affected our ability to measure the tension applied to the hamstring muscle group. In preliminary work with two subjects, we recorded the level of EMG activity from the quadriceps during the straight leg raise stretch. No quadriceps EMG activity was evident during the stretch with either subject. Although we asked all subjects to relax the lower extremity being tested some subjects initially actively assisted. Any active assistance could be detected by visible quadriceps contraction and was also immediately apparent on the recording of the force measurement, which would show an abrupt drop off in tension in an otherwise smooth sloping line. If a subject actively assisted, the lower extremity was immediately returned to the table and the procedure was repeated with the subject relaxed.

### Instrumentation

**Range of motion measurement.** An electrogoniometer (Penny and Giles, Gwent, U.K.) was placed over the hip joint. The distal endblock was aligned with the femur and attached directly to the skin. The proximal endblock was attached to a brick aligned with the subject's trunk. The electrogoniometer had a sensitivity of 1 degree. Range of motion ROM is reported in degrees of hip flexion with the knee extended. Change in joint angle reflects elongation of the muscle-tendon unit of the hamstring muscle group.

**Tension measurement.** The force required to raise the lower extremity was measured by a load cell (Kistler Instruments, Winterthur, Switzerland) in series with a chain attached to the subject's ankle. Direction of pull was kept at 90 degrees to the leg throughout the ROM so that tensile forces could be accurately determined throughout the ROM. The load cell was calibrated with a 78 N load for 45 s to ensure there was no stress relaxation in the piezo-electric crystal of the load cell. This calibration force was comparable to the forces measured during the experiment. The tension measurement had a sensitivity of 0.25 N.

In this model the measured resistance to stretch is a gross force representing the weight of the lower extremity at the given hip flexion angle (limb weight) plus the resistance to stretch (net force). The net force represents the tensile force applied to the hamstring muscle group and can be computed as follows:

$$\text{Net Force} = \text{Gross Force} - [\cos(\text{hip flexion ROM}) \times \text{Limb Weight}].$$

Limb weight was defined as the force required to hold the lower extremity horizontally with the hip in neutral position. This measurement was made from the chart recording as the force at which the ROM recording left baseline. Figures 2 and 3 show the same limb weight measurement for one subject in test 1 and test 2, respectively. The force measurements reported in the results are net forces.

Using a free-moving metal arm with a fixed axis of rotation, we tested the accuracy of our estimation of the limb weight component. The load cell in series with the chain was attached to the distal end of the arm, and the electrogoniometer was attached at the axis of rotation. The arm was raised from the horizontal (0 degrees) to the vertical (90 degrees). Care was taken to ensure that the direction of pull was perpendicular to the arm throughout the ROM. ROM and the weight of the arm were continuously recorded, simulating the straight leg raise without skeletal muscle resistive forces. The chart recording of this test can be seen in Figure 4. The force is highest at 0 degrees and zero at 90 degrees. Using the baseline measurement of force at 0 degrees, we calculated the force at 10-degree increments up to 90 degrees based on the cosine of the respective angles. We then computed the measured force from the chart recording at 10-degree increments. The accuracy of our testing technique is based on the measured force equaling the calculated force. The linear regression of the calculated force versus the measured force is shown in Figure 5. In a perfect system the slope should equal one and the intercept should equal zero. Figure 5 shows our measurement technique to be accurate and emphasizes the importance of making a correction for the limb weight component.

Accurate determination of the tensile force applied to the hamstring muscle group required the knee to be at full extension throughout the procedure. With a few subjects there was noticeable flexion of the knee (approximately 10 degrees) approaching final ROM in test 1. In preliminary work we attempted bracing the knee in full extension but found that placement of the brace interfered with the EMG response because of contact with the electrodes. Bracing did not totally eliminate the knee flexion and caused additional discomfort, which affected the subjects ability to relax. We chose not to brace the knee for this experiment since most subjects knees remained in extension during the straight leg raise. During the 45-s relaxation period, the ROM at the knee appeared to remain constant, allowing accurate determination of the decline in force over time. A change in ROM at the knee during the relaxation period would have caused a resultant change in hip flexion ROM since the tester was holding the distal leg at a fixed position. Hip flexion ROM remained constant during the relaxation period for all subjects.

To compute the actual tensile stress applied to the hamstring muscle group, the tensile force must be divided by the cross sectional area of the tissue bearing tension. This computation was not possible here given the nonlinearity of muscle cross sectional area and our *in vivo* approach.

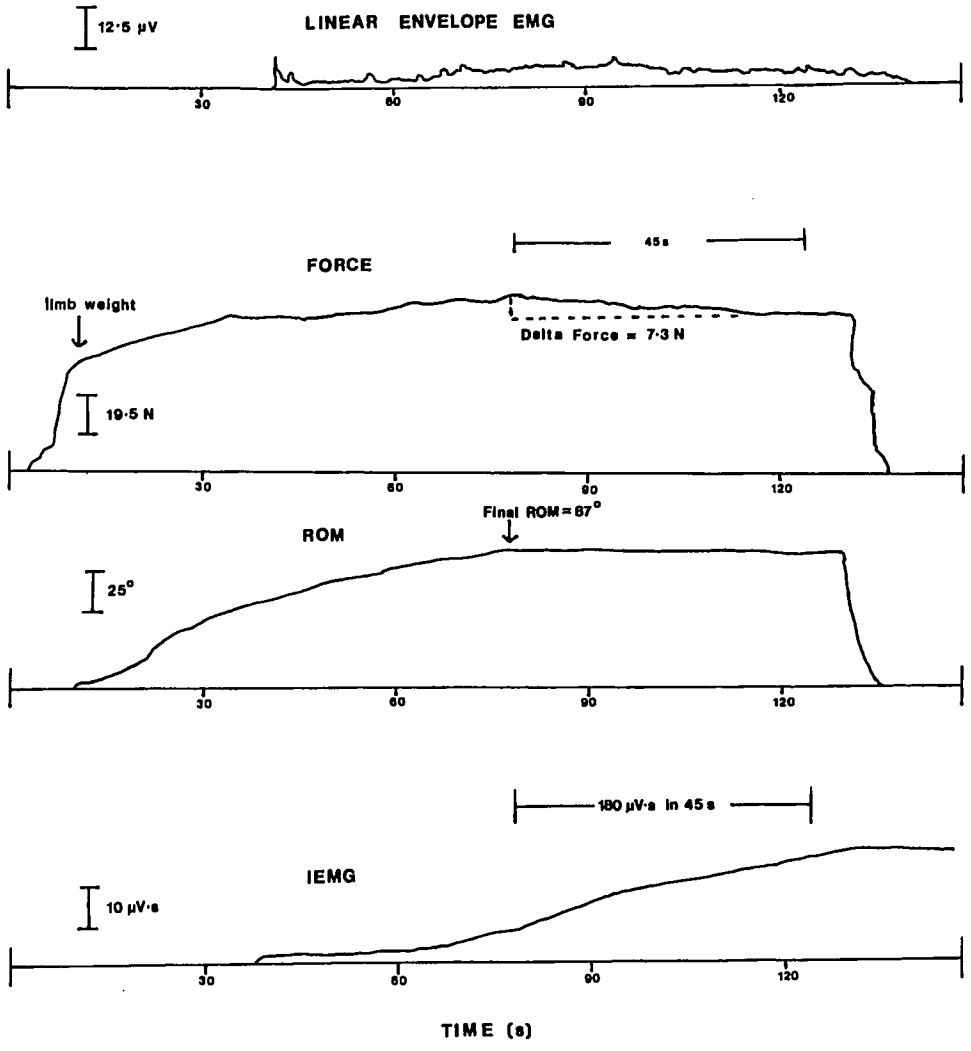


Figure 2—Chart recording of test 1 for one subject. Top line showing linear envelope EMG with onset of EMG activity at approximately 60 degrees. The recording second from top shows the gross tensile force response to stretch (without correction for limb weight). Limb weight is identified as the point at which the ROM recording leaves baseline corresponding with the lower extremity being lifted from the table. The relaxation response over the 45-s period is indicated by delta force. The third recording shows hip flexion ROM during the straight leg raise with final ROM identified as the plateau in this line. The bottom line shows the integrated EMG activity during the stretch identifying the total activity for the 45-s relaxation period.

**EMG measurement.** We chose to use surface electrodes to measure gross activity of the hamstring muscle group in response to a straight leg raise stretch. It was not within the scope of this study to identify the activity of specific muscles. Ag/AgCl surface electrodes (3M, St Paul, MN) were placed approximately 6 cm apart, midway between the gluteal fold and the knee joint, over the semitendinosus muscle belly. In preliminary experiments this electrode placement provided greatest EMG response with active knee flexion. The EMG signal was measured by a wide band A.C. preamplifier and integrator (model 7P3 B) polygraph channel (Grass Instruments, Quincy, MA) and integrated (IEMG) by a summating integrator channel (model 7P10). The main amplifier had a gain of 50 $mV \cdot cm^{-1}$. The linear envelope of the EMG signal and its cumulative integration were recorded. The EMG activity was recorded with a bandwidth of 3–500 Hz with a common mode rejection ratio of not less than 1600:1. The recorded linear envelope EMG had a sensitivity of 12.5 $\mu V \cdot cm^{-1}$. The IEMG had a sensitivity of 100 $\mu V \cdot s \cdot cm^{-1}$. Force, ROM, IEMG, and linear envelope EMG were continuously recorded. Figures 2 and 3 show these recordings for one subject for test 1 and test 2, respectively.

**Statistical analysis.** All data are displayed as means and standard error of the mean (SEM). Peak force refers to the resistance to stretch (net force) at respective final ROM. Final force refers to the resistance to stretch at final ROM after the 45-s relaxation period. Delta force refers to the difference between peak and final force. Total IEMG refers to the EMG activity over the 45-s period and does not include EMG activity prior to peak force. With respect to test 1 only, initial 6-s IEMG refers to the EMG activity during the first 6 s after end of ROM is reached. Final 6-s IEMG refers to the EMG activity during the last 6-s of the 45-s period. Delta EMG refers to the difference between initial and final 6-s IEMG. Paired Student's *t*-tests were used to measure differences between peak force and final force in test 1 and 2, and initial 6-s IEMG and final 6-s IEMG in test 1. Nonpaired Student's *t*-tests were used to measure gender differences. Linear regressions were performed with ROM as the dependent variable.

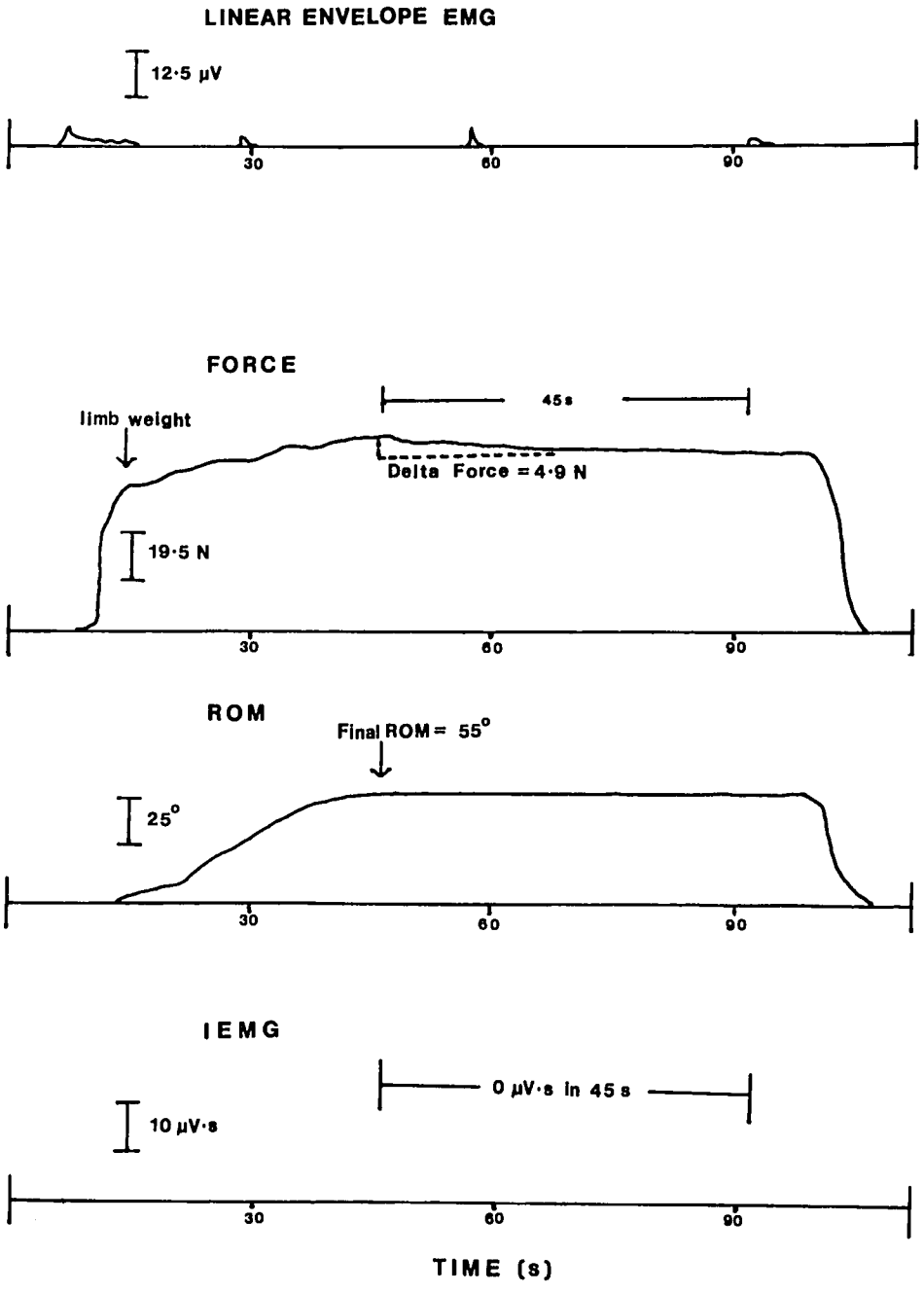


Figure 3—Chart recording of test 2 for the subject used in Figure 2. The top line shows the absence of EMG activity hence zero activity seen for integrated EMG (bottom line). The force recording shows the limb weight to be the same as test 1 and a smaller absolute relaxation response. The ROM recording shows final ROM as 55 degrees (5 degrees below the angle at which the onset of EMG activity occurred in test 1).

## RESULTS

The means ± SEM for peak force, final force, total IEMG, initial 6-s IEMG, final 6-s IEMG (test 1 only), and ROM are given in Table 1. There was a significant decline in force over the 45-s relaxation period in test 1 ($P < 0.0001$) (Fig. 2) and in test 2 ($P < 0.05$) (Fig. 3). For the 10 subjects who completed both tests, the mean percent decrease in force was not significantly different between the tests (14.4 ± 2.2% test 1, 13 ± 2.3% test 2). In test 1, final 6-s IEMG was significantly less than initial 6-s IEMG ($P < 0.01$). The EMG data from one subject in test 1 were excluded because they were periodically greater than the measurement scale. For test 1 percent decrease in IEMG was not correlated with percent decrease in force ($r = 0.001$). The measured IEMG in test 2 ranged from 0 to 40 $\mu V \cdot s$, with a mean of 18 ± 4.73 $\mu V \cdot s$ for the entire 45-s period. This was less than 5% of the total IEMG activity in test 1 and represented negligible EMG activity.

In test 1, final ROM correlated inversely with tensile force at 30 degrees of hip flexion ($r^2 = 0.46$, $P < 0.01$). Force at 30 degrees represented a hip flexion angle below the point of onset of sustained EMG activity. Multiple regression improved the predictability of final ROM when initial 6-s IEMG (negatively related) was

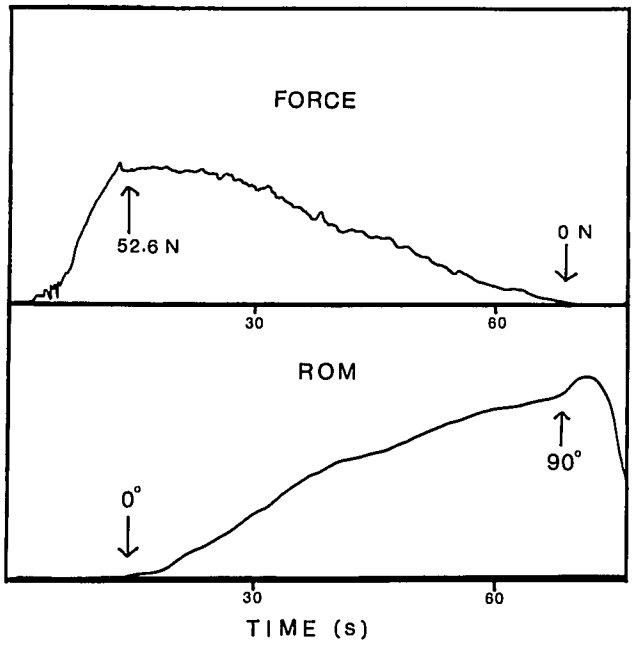


**Figure 4—Sample chart recording of free moving arm validating the limb weight correction for the calculation of tensile force. Force is maximal at the horizontal and zero at the perpendicular.**

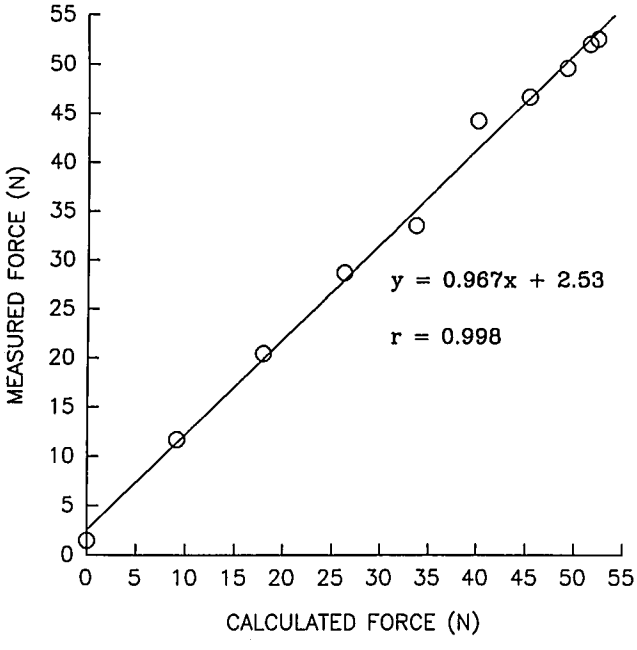


**Figure 5—Linear regression of calculated force with measured force to validate the cosine correction for limb weight. Calculated force = weight of free moving arm × cosine of the given angle. Measured force = force determined from the chart recording in Figure 4 at each given angle.**

combined with force at 30 degrees ($R^2 = 0.74$, $P < 0.001$). That is, greater EMG activity was associated with decreased final ROM. Women had greater final ROM ($P < 0.05$) and less force at 30 degrees ($P < 0.05$) (Table 2) but did not differ from men in peak force, EMG activity or delta force.

TABLE 1. Resistance to stretch and EMG response.

| | Test 1 | Test 2 |
|---|---|---|
| Peak force (N) | 65.8 ± 5.3 | 28.3 ± 4.4 |
| Final force (N) | 54.4 ± 4.6*** | 24.1 ± 3.2* |
| Total IEMG (μV·s) | 542.6 ± 107.1 | 18.0 ± 4.7 |
| Initial 6-s IEMG | 135.5 ± 20.6 | |
| Final 6-s IEMG | 75.8 ± 14.4** | |
| ROM (degrees) | 90 ± 6 | 53 ± 6 |

Values are means ± SEM. *** $P < 0.0001$ significantly different from peak force; ** $P < 0.01$ significantly different from initial 6-s IEMG; * $P < 0.05$ significantly different from peak force.

TABLE 2. Gender differences in response to stretch.

| Test 1 | Men | Women |
|---|---|---|
| Age (yr) | 28.2 ± 1.9 | 29.3 ± 1.1 |
| ROM (degrees) | 80 ± 7 | 104 ± 5* |
| Delta force (N) | 12.6 ± 2.5 | 9.4 ± 2.2 |
| Force at 30 degrees | 15.7 ± 2.2 | 9.0 ± 1.2* |
| Total EMG (μV·s) | 577.5 ± 186 | 496.2 ± 64.7 |
| Delta EMG (%) | 40.4 ± 10.5 | 43.3 ± 7% |

Values in mean ± SEM. * Significant effect of gender $P < 0.05$.

Rate of stretch did not differ between test 1 and test 2 ($N = 10$). Rate of stretch in test 1 ($N = 15$) was 1.63 ± 0.46 degrees·s$^{-1}$ and 2.04 ± 0.19 degrees·s$^{-1}$ in test 2 ($N = 10$).

## DISCUSSION

**Stress relaxation response.** In test 2 (negligible EMG activity) the decrease in force over time represented viscoelastic stress relaxation. Peak force in test 1 was more than twice that in test 2 and was associated with a significant contractile response. In 45 s (relaxation time) there was a significant decrease in EMG activity, in test 1, represented by the difference between initial 6-s IEMG and final 6-s IEMG. The EMG response persisted for the duration of the stretch, as can be seen in Figure 2. The mean activity in the initial 6 s represented only 25% of the total activity over the 45-s period, and the mean activity in the final 6-s represented 14% of the total activity. The lack of any significant relationship between the decrease in EMG activity and the decrease in force in test 1 and the fact that the percent decrease in force in test 1 was similar to the response in test 2 suggest that the contractile response minimally affected the relaxation response we observed.

It was not within the scope of this study to directly relate the stretch-induced EMG response to a specific level of force generation. However, we measured the EMG response of three subjects for active knee flexion in the standing position in order to have a comparison to the stretch induced EMG response. For three subjects tested, the EMG (linear envelope) response to active flexion of the knee to 15 degrees from the standing position exceeded each subject's peak stretch-induced EMG response for test 1. The EMG response to 90 degrees of active knee flexion was greater than 200 μV

for each of the three subjects. This represents more than 4 times the upper limit of our measurement scale for the stretch-induced EMG response and suggests minimal force generation from the stretch-induced EMG response. The potential force generation contribution of the stretch-induced EMG response has been reported elsewhere (13). However, the contribution of the stretch-induced EMG response to the relaxation response is not entirely clear from this data. Previous work has shown increased EMG responses with PNF stretching techniques to be associated with increases in ROM (14,15).

**Rate of stretch.** Rate of stretch can affect the resistance to stretch through viscous (22) or reflex contractile properties (10). The tester applied a slow passive stretch and attempted to be consistent with rate of application. The coefficient of variation between subjects for rate of stretch in test 1 was 28% and 30% in test 2. A change in rate of stretch of at least an order of magnitude is required to detect changes in the tensile force response (1,22).

**Range of motion.** The range of motion in this experiment ranged from 43 to 125 degrees with no subject showing sustained EMG activity below 30 degrees. The only variable we measured which was independently related to final ROM (test 1) was tensile force at 30 degrees (negatively correlated). With respect to maximum hip flexion ROM, our sample was too small to compare flexible versus inflexible individuals, but the findings suggest that inflexible individuals have greater resistance to stretch throughout the ROM. Final ROM did not show a significant correlation with the decrease in resistance to stretch over time (test 1), suggesting that stress relaxation is unaffected by the "flexibility" of the subject.

Using multiple regression final ROM was best predicted by the combination of force at 30 degrees and initial 6-s EMG (both negatively correlated). Thus, final ROM seems to be determined ($R^2 = 0.74$) by a viscoelastic component (force at 30 degrees = resistance before EMG response) and a contractile component (initial 6-s IEMG = maximum EMG response). This two component model of resistance to stretch has been previously described (13).

**Gender differences.** Women had significantly greater hip flexion ROM and less force at 30 degrees, but the EMG and relaxation responses were remarkably similar with respect to gender. Whether the increased muscle extensibility in these women was a function of smaller muscle mass, a function of joint geometry, or a function of a gender-specific collagenous muscle structure is unknown.

***In vitro* comparisons.** In Taylor et al.'s study (22) all muscle-tendon units were stretched to an initial force of 78.4 N. Although not reported numerically, the decrease in tension graphically appeared to be approximately 11 N over a 30-s period. In our study in test 1, the mean peak force was 65.75 ± 5.31 N with a mean decrease in force of 11.35 ± 1.76 N in 45 s. From Figure 2 it can be seen that most of the stress relaxation occurred within the first 15 s. The actual tensile stress applied to the muscle tendon units was not measured in either study since the cross-sectional area of the muscle was not measured in either case. The hamstring muscle group of the human represents a much larger cross sectional area than the extensor digitorum longus of the rabbit. Thus, the tensile stress applied to the muscles in our study was much less than in Taylor's study.

The architecture of the specific muscle may affect its response to tensile stress. *In vitro* work has shown different responses in the rabbit tibialis anterior and extensor digitorum longus muscles (17,20). The individual muscles of the human hamstring muscle group are characterized by extensive overlapping of the proximal and distal tendons and a significant fibrous raphe in the semitendinosus muscle belly (7). These characteristics may have implications for studying the stretch response of this muscle group compared with other muscles in humans.

## SUMMARY

We have been able to demonstrate viscoelastic stress relaxation in human skeletal muscle, *in vivo*. When a stretch-induced EMG response was elicited, a significant decrease in EMG activity occurred during the relaxation period but was statistically unrelated to the degree of relaxation in the hamstring muscle group. Maximal hip flexion range of motion appeared to be a function of both the viscoelastic and contractile responses with the tighter individuals having greater resistance to stretch at 30 degrees and a greater contractile response at maximal ROM. Women had a significantly greater ROM but had similar subsequent relaxation and contractile responses.

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