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ANALYSIS OF CUMULATIVE STRAIN IN TENDONS AND TENDON SHEATHS

STEVEN A. GOLDSTEIN*, THOMAS J. ARMSTRONG‡, DON B. CHAFFIN† and LARRY S. MATTHEWS*

*Biomechanics, Trauma and Sports Medicine Laboratory, Department of Surgery, Section of Orthopaedic Surgery, University of Michigan, Ann Arbor, MI, †The Center for Ergonomics, and ‡Department of Environmental and Industrial Health, University of Michigan, Ann Arbor, MI, U.S.A.

Abstract—Twenty-five fresh frozen flexor digitorum profundus tendons stratified by sex were subjected to uniaxial step stress and cyclic loads in twelve intact human cadaver hands. By attaching specially designed clip strain gage transducers on tendons just proximal and distal to an undisrupted carpal tunnel, the interactions of the tendons, tendon sheath and retinacula were measured.

The elastic and viscous response of the tendon composites to step stresses were found to fit fractional power functions of stress and time respectively. A significant and quantifiable decrease in strain from the proximal to the distal tendon segment was found to be a function of wrist deviation.

The results indicate that an accumulation of strain does occur in tendinous tissues during physiologic loading.

INTRODUCTION

Past studies have shown that cumulative trauma disorders of the upper extremities such as tenosynovitis, tendonitis, bursitis, and carpal tunnel syndrome can be caused, precipitated, or aggravated by repeated exertions with certain hand and wrist positions—particularly in combination with forceful exertions. These findings were based on both epidemiological and biomechanical studies of manual work activities.

The epidemiological studies have shown that morbidity patterns are associated with repeated exertions and certain job-related attributes (Wilson and Wilson, 1957; Thompson *et al.*, 1951; Kendall, 1960; Eichoff, 1927; Armstrong and Chaffin, 1979a; Cannon *et al.*, 1981; Muckart, 1964; Hymovich and Lindholm, 1966). Biomechanical studies have illustrated that work activities involving forceful exertions in combination with certain postures can produce stress concentrations on the tendons and adjacent tissues that correspond with the sites of injury (Armstrong and Chaffin, 1979b; Castelli *et al.*, 1980; Meachim and Roberts, 1969; Muckart, 1964; Eichoff, 1927). These studies treated the tendons and adjacent structures as perfectly elastic, frictionless materials; however, tendinous tissues have been shown to demonstrate viscoelastic properties. Consequently, repeated exertions may produce progressive elongation or creep of the tendons and tendon sheaths.

In a study conducted by Rais (1961), morphological tissue changes resembling peritendinitis crepitans were experimentally induced in rabbits by over exercising the hind limb. The most significant factor related to the severity of tissue abnormalities was the duration of hyperfunction. This work was the first to correlate time and load characteristics to a subsequent cumulative trauma injury.

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Based on these studies, it is hypothesized that the viscoelastic creep responses of the tendons or tendon sheaths is an important etiological factor in cumulative trauma disorders. These disorders are hypothesized to be a physiological response to *cumulative strain* developed in the tendons, tendon sheaths, or retaining ligaments that form the anatomical pulleys. The purpose of this study was to investigate the local biomechanical aspects of tendon–tendon sheath function. Specifically, to determine whether the tendons and tendon sheaths exhibit viscoelastic properties under simulated physiologic loading conditions.

MATERIALS AND METHODS

A procedure was developed to measure strains in tendons under simulated physiological loads *in vivo* in cadaver hands.

Twelve hands from seven subjects, four females and three males, between the ages of 55 and 72, were tested. Twenty-five fresh frozen flexor digitorum profundus tendons were subjected to uniaxial step stress and cyclic loads at physiologic levels.

The hands were cut from the cadaver forearm approximately 8 cm from the distal wrist crease. Careful dissection of the hands was performed to isolate the profundus tendons to digits two, three and four. The carpal tunnel and common flexor sheath were carefully preserved. Proximal and distal to the common sheath, 4 cm segments of superficialis tendons two, three and four were carefully resected to expose the profundus tendons. The surface of the profundus tendons proximal and distal to the carpal tunnel were prepared and clip strain gages made from beryllium copper with two 120 Ω resistive gages were attached to each tendon by passing sutures through the bulk of the tendon and using a cynoacrylic adhesive on the surface. A schematic of a prepared specimen is illustrated in Fig. 1. The force necessary for full scale deflection of the isolated clip gages corresponding to approximately 20% strain was between 0.5 and 0.8 N.

Using specially designed clamps, the tendons were coupled to a yoke on the loading machine. Tensiometer load cells were placed in series with each tendon. The yoke allowed three tendons to be adjusted for initial tension independently and loaded simultaneously. The specimens were rigidly fixed in a constant temperature test chamber which was maintained at 38°C and 100% humidity by securing the radius and ulna to a wooden block using cortical bone screws.

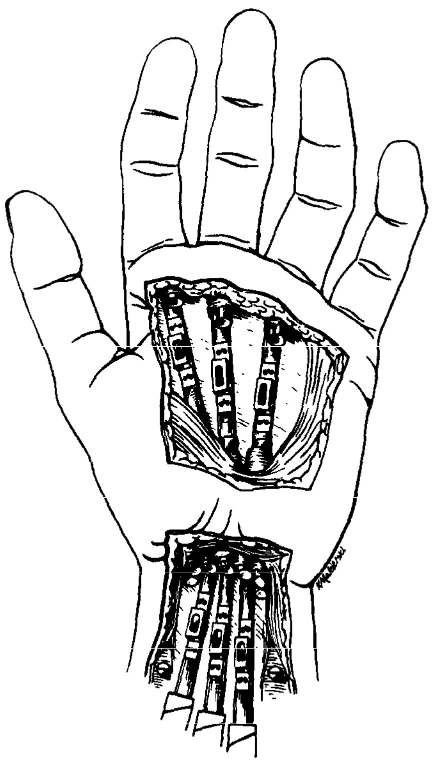

Fig. 1. The cadaver hands were prepared as illustrated above. After careful excision of the proximal and distal segments of the superficialis tendons, the clip gages were attached to the profundus tendons using cynoacrylate cement and sutures.

The loading machine consisted of a specially adapted pneumatic cylinder and regulated air supply. The air supply was fed to the cylinder through an electronic valve which provided accurate and reproducible loading parameters. On–off control was provided by a micro-processor which made variations to step and cyclic loading possible (Fig. 2).

The system provided a means for measuring input load, as well as strain *proximal and distal to the osseofibrous canal.* The pneumatic system was chosen because of its time to load characteristics. From studies by Milner-Brown and Stein (1975) and Close (1972) it can be estimated that the time of contraction of skeletal muscle to maximum tension is approximately in the range of 20–120 ms. The pneumatic loading system (time to maximum load is 40–60 ms) simulated these physiological loading conditions.

Calibration of the strain gage transducers was performed before each test using a simple micrometer based device. All transducers were linear within the experimental range and no hysteresis was observed. Since it was difficult to determine the effect of cementing the clip gages to the tendons, and to account for nonhomogeneous effects of measuring strain in two different places, a second calibration of the clip gages was performed at the end of each experiment. This calibration was performed by dissecting out the tendons tested, with the gages still intact, and loading them individually while monitoring deformation with a linear voltage differential transformer (LVDT). The tendons were cut at their insertion in the phalanges and secured in a second tendon clamp fixed to the wall of the test chamber. The proximal end was maintained in the loading machine. The LVDT was mounted on the loading cylinder and measured displacement of the piston. No slippage was observed at either clamp

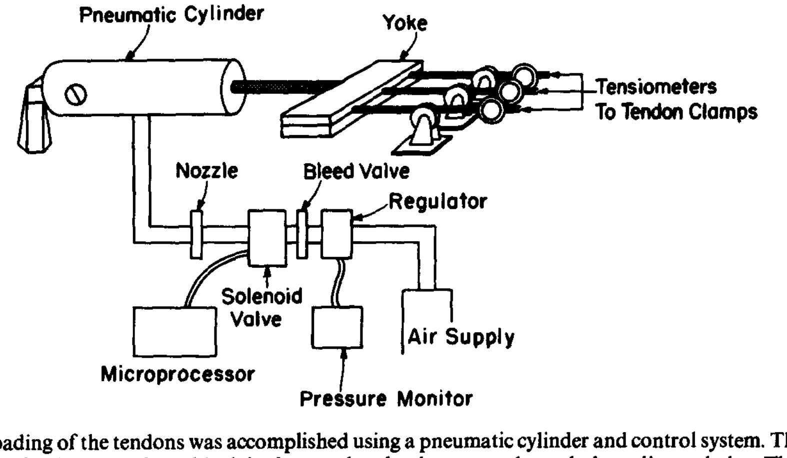

Fig. 2. Loading of the tendons was accomplished using a pneumatic cylinder and control system. The hands were rigidly fixed to a specimen block in the test chamber by screws through the radius and ulna. The fingers were securely fixed to the frame of the chamber in a power grip position with the wrist in either 65° of extension, neutral or 65° flexion.

interface. A correction factor, which had a range from 0.95 to 1.19, was then applied to the data. The LVDT was accurate to 0.0063 mm.

The tendons were removed from the hands after the experiment was complete and set to dry at room temperature for 30 min. Using a finely sharpened microtome knife, the tendons were sectioned into 1/4 in. segments. The segments were then placed on a ruled positioning block over parallel pins of equal diameter. Thirty-five millimeter slides of the mounted segments were then projected and the areas were calculated using planimetry.

Sixteen tendons from eight hands were subjected to step stress tests. Four of the hands were from males and four were from females. Two tendons, flexor profundus digit two and flexor profundus digit four were tested simultaneously on each hand. Two variables were measured on each tendon: proximal and distal strain. All tendons were subjected to three loads (23 N, 65 N, 80 N) at three positions: neutral wrist, 65° flexed and 65° extended. The experiment provided 288 strain response curves to variations in six factors: load, position, sex, strain gage location, digit and time. Each step stress test was held for 5 min. Before beginning testing on a particular tendon, it was pre-loaded to 15 N to ensure stability of clamp fixation.

Twelve tendons from four hands were subjected to a variety of cyclic load conditions with the wrist in neutral position. Each tendon was subjected to various combinations of load (23 N, 46 N, 65 N), load duration times (0.25 s, 1 s, 4 s, 8 s, 12 s) and period (5 s, 10 s, 15 s). These combinations of time parameters provided variations in frequency, load durations, and recovery periods. Each tendon was preconditioned to 30 N and then loaded for 500 cycles.

The loads selected for study were derived from the measurements of work attributes in occupations requiring repetitive tasks. For example, in a study by Armstrong *et al.* (1982), hand and grasp forces between 70 and 140 N were documented from workers performing a cutting operation in a poultry processing plant. Chao *et al.* (1976) calculated that for a grasp position, the force exerted by the tendons would be multiplied by a factor of 2.77–4.32 per unit force exertion. If the profundus tendon supplies at least 50% of the load of a particular digit and the digit accounts for 25% of the total hand exertion (Hazelton *et al.*, 1975), then the force of the tendon would be expected to be between 24 and 75 N.

RESULTS

For analysis purposes, it was assumed that the total strain, was equal to a viscous component plus an elastic component. It was further assumed that during load applications ($0 < t < t_{load\ max}$), at physiological rates, the viscous components of strain are negligible with respect to elastic components. Therefore:

$$\varepsilon_{total} = \varepsilon_{elastic} \qquad 0 < t < t_{load\ max} \tag{1}$$

$$\varepsilon_{total} = \varepsilon_{elastic} + \varepsilon_{viscous} \qquad t_{load\ max} < t. \tag{2}$$

As noted by Fung (1967) in nonlinear materials such as tendons, the tangent elastic modulus can be described by a linear function of stress. The linear relationship between the elastic modulus and stress of the proximal segment was modeled using least squared regression analysis. The resultant equation of the regression line was found to be as follows

$$A = 112\ (B) + \text{error}$$

where A = modulus in MPa, B = stress in MPa ($r = 0.94$, $p < 0.0001$). The nonlinear elastic response was also found to be significantly dependent on sex; female tendons were significantly stiffer than male tendons ($p < 0.01$), (see Fig. 3).

Least squared error forward stepwise regression procedures were used to evaluate the main effects and first order interactions of digit, wrist position, and gender in the relationship between strain and stress. Only gender was found to be significant at $p < 0.01$; the following model was obtained

$$\varepsilon = A\sigma^{0.495} \tag{3}$$

where A = constant dependent on sex ($A = 0.228$ for males; $A = 0.119$ for females).

As noted in the methodology section, strain was measured from a tendon segment proximal and distal to the osseofibrous canal. Any difference in strain from the proximal to the distal segment indicates a decrease in uniaxial load along the tendon caused by an increase in shear traction between the tendon, tendon sheath and anatomical pulley. A quantifiable decrease in strain from the proximal to distal segment was found to be dependent on load and wrist deviation.

Since each tendon was removed whole from the hands with the gages still attached and a secondary calibration adjustment factor was applied to the data (as described in Materials and Methods), nonhomogeneous effects as well as small fluctuations in cross-sectional areas were accounted for. These differences in strain from proximal to distal segments are only due to

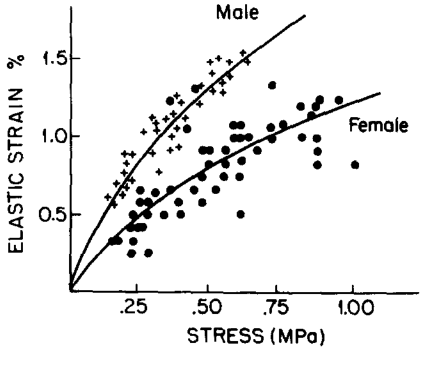

Fig. 3. The nonlinear elastic strain response was found to be significantly different for males and females.

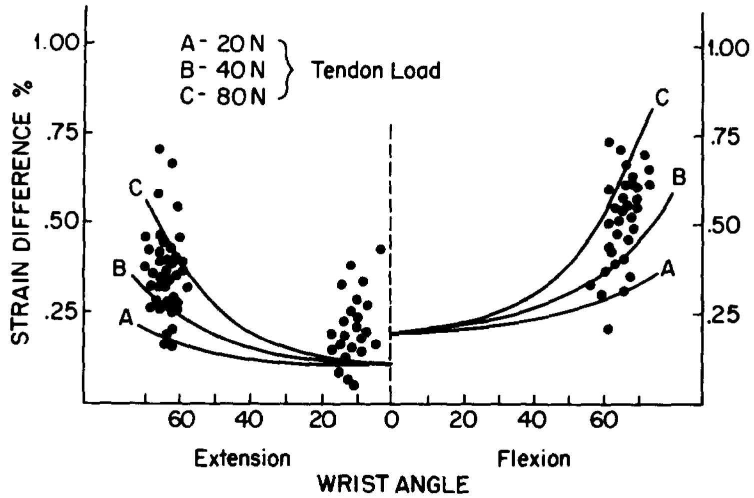

Fig. 4. Strain differences from the proximal to distal segment were significantly related to wrist deviation. Prediction curves based on nonlinear regression analyses are shown for three arbitrary loading conditions superimposed over all the experimental data. Although the prediction lines are based on only three observation regions (65° flexed, neutral, 65° extended), they provide a reasonable estimate of the relationship between strain difference and wrist angle ($R^2 = 0.48$).

shear traction effects. From a repeated measures analysis of variance, the decrease in strain and therefore increase in shear traction forces was significantly higher in wrist flexion and extension than at neutral position, and significantly greater in flexion than extension (Fig. 4). The decrease in distal strain which ranged from 0 to 40% of the proximal strain, demonstrated that significant stresses exist at the tendon, tendon sheath and retinacula interface and that these stresses increase significantly with increasing wrist deviation.

The creep response to the tendons was measured as the increase in strain with time, after the elastic strain. Analysis of the viscous strain response curves as illustrated in Fig. 5 revealed two fairly distinct regions of creep after the elastic response: a very early dynamic or transition creep followed by a fairly constant fibrillar creep rate which continued for several minutes. As indicated by several investigators, the response of tendons to various loads have been shown to

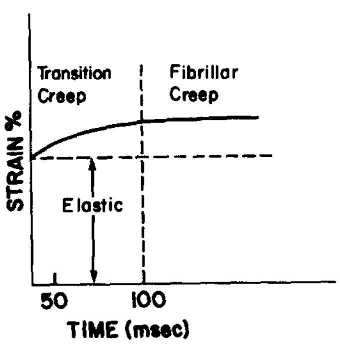

Fig. 5. The experimental response curves were found to have three distinct response components, an elastic response, transition region and fibrillar response region.

correlate well with structural interactions (Betsch and Baer, 1980; Chu and Blatz, 1972; Hooley and Cohen, 1979; Viidik, 1978; Lanir, 1978, Kastelic *et al.*, 1980). Hooley and Cohen (1979) showed a creep behavior model based on structures that fit the experimental data very well. The authors found two distinct regions of creep: (1) an initial shearing of the gel at low strains as the tendon fibers straightened out and (2) a transition from gel behavior to a gradual linear region of fibrillar creep. Based on these findings, we proposed an analogous analysis for the early creep response and hypothesized that the creep response to a cyclic loading condition of short duration may be more accurately modeled by using the dynamic or transition creep rate early in the response curve. Therefore, the experimental creep parameters in this study were calculated by fitting the creep strain *defined* by the first 100 ms of viscous response to a power function of time.

Analysis of the creep strain data illustrated a large variance in the data as would be expected from the evaluation of the transition creep region. Despite this observation, the creep strain was found to be significantly related to sex and wrist position and could be modeled by the following equation

$$\varepsilon_{creep} = K(t)^{0.409} \tag{4}$$

where K is a constant dependent on sex and wrist position (see Table 1).

Table 1. Values of K

	Position		
	Flexed	Neutral	Extended
Male	0.014	0.01	0.014
Female	0.013	0.008	0.012

The creep strain was not dependent on the stress levels used in this experiment.

The results from the cyclic loading experiments illustrated that significant creep can occur in the physiological loaded tissues. For severe loading regimes (long loading duration and short recovery times, i.e. 10 s period, 8 s load duration) the strain can increase 40 % from cycle 100 to cycle 500, as illustrated in Fig. 6.

DISCUSSION

The first conclusion that can be made from the results is that the tissues illustrate non-linear elastic and viscoelastic properties for a physiologic range of loads similar to those reported by many earlier investigators. From the experimental step stress data, the range of elastic strain measured was from 0.2 to 1.8 %. This is in good agreement with physiologic strains estimated in the literature of 0 to 3 percent by Abrahams (1967) and Rigby *et al.* (1959), Hirsch (1974), Kear and Smith (1975), Barnes and Pinder (1974) and Elliott (1965).

The elastic strain response of the tendons was found to fit a power function of the input load. The value of the exponent on stress, 0.495, was in good agreement to the constitutive model proposed by Haut and Little (1972). In their study, the elastic stress response to applied strain was found to be accurately represented by

$$\sigma_{elastic} = C\varepsilon^2. \tag{5}$$

The elastic strain response was also found to have a statistical dependence on sex. For a given stress or load, the strain response of a female tendon was less than that of a typical male. This consequence implies that the female tendons were stiffer than the male tendons.

While many hypotheses could be proposed to account for these differences, the fairly large variance of the data as well as the problem of extrapolating cadaver experiments to physiologic conditions, no truly conclusive statements can be made concerning the relationship between these findings and the cumulative trauma disorders (Goldstein, 1981).

Significant interactions between the tendons and tendon sheaths are illustrated by the factors affecting the creep strain model and the proximal–distal differences in strain. Since the protocol adjusted the proximal and distal segment responses to account for nonuniaxial strain differences by secondary calibration, these results signify that the stress transmitted to the sheath during tendon exertions is significant and quantifiable. The magnitude of the difference in strain would seem to indicate that the shear interactions were not caused by degeneration of the tissues or loss of lubrication in the cadaver model. These effects, however are unknown. These findings suggest that contrary to the simple pulley belt models, the functional interactions between the tendons and their anatomic pulleys are significant.

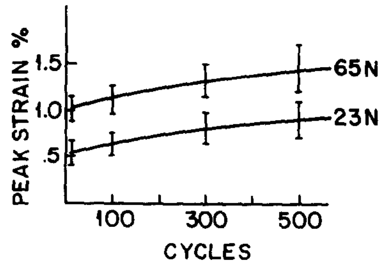

Fig. 6. An example of the response to cyclic loading is illustrated above. The loading period was 10 s with the load duration at 8 s. Two magnitudes are illustrated.

In conclusion, this study demonstrates that the tendon–tendon sheath composite when loaded at physiologic levels exhibits nonlinear viscoelastic behavior.

This study also illustrates that when subjected to the type of hand and wrist exertions typically found in some industrial assembly lines, viscoelastic creep was measured in the tendons.

As predicted by biomechanical models, the stresses at the tendon–tendon sheath and retinacular interface are significant and dependent on wrist deviation. The difference in proximal and distal segment strain and therefore shear traction forces are greater with the wrist flexion and extension than in neutral position.

This study provides some evidence that creep strain in collagenous tissues may be an important etiological factor of chronic cumulative trauma disorders. These studies who also suggest that future animal studies are needed to validate these findings in a living field in order to develop specific recommendations for usage pattern alterations.

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