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A Technique for the Measurement of Tension in Small Ligaments

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7.1 Introduction

The goal of this chapter is to present a method for the measurement of the *in situ* tensile force in small ligaments, the ligament tension transducer (LTT), and demonstrate its utility by displaying an application to measuring the properties of the ligaments of the volar side of the wrist. This method allows the static *in situ* force within the bulk of a ligament to be determined without disturbing its functional performance. Before presenting the technique, the significance and history of the study of the biomechanical properties of ligaments will be reviewed; the general mechanical properties of ligaments will be presented since these properties affect the methods by which measurements are made; and the advantages and shortcomings of other techniques will be discussed. The LTT technique, its performance and limitations, and an example application will then be covered.

The significance of studying the biomechanical properties of ligaments stems from the benefits provided. Such studies have increased our understanding of ligament behavior, helped to identify key ligaments requiring restoration after injury, and have assisted in identifying materials and tissues with appropriate characteristics that can be used as replacements. For instance, classic studies of the properties of the anterior cruciate ligament of the knee and various materials used for its replacement after injury have allowed selection of materials with appropriate strength and stiffness characteristics. This has led to a high success rate for this common procedure.[24]

7.2 Background

A Short Summary of Experimental Techniques in Ligament Biomechanics

Knee ligament studies have dominated the literature, probably because trauma to the knee is very painful and disabling, yet common, and the ligamentous structures are large and easily identified.[29] Since the majority of knee loads are supported by four ligaments, any ligament tear is functionally disabling due to increasing joint instability. In 1974, Warren et al.[33] published an *in vitro* study of knee medial collateral ligament biomechanics using a radiographic technique to determine ligament strain during functional positioning. In 1975, Noyes et al.[25] described the biomechanics of the ACL of the rhesus monkey. Their study correlated tensile force with strain rate, using isolated bone-ligament-bone preparations mounted to a materials testing machine.

During the 1980s, new transducer techniques that emerged allowed a shift from *in vitro* to *in situ* testing. In 1982, Lewis[22] described an *in situ* study on the human cadaver knee anterior cruciate ligament (ACL) using a buckle transducer to measure tensile force. In 1983, Stone et al.[30] performed an *in vitro* study on the human ACL and an *in vivo* study of the canine ACL using the liquid metal strain gage (LMSG) to measure strain in biological tissue. Also in 1983, Arms et al.[4] published an *in situ* study of the human cadaver medial collateral ligament (MCL) of the knee using a Hall effect transducer. The buckle transducer measures force while the LMSG and Hall effect transducers measure strain and will be described in detail later in the discussion. These devices not only allowed the measurement of *in situ* force and strain, but also could be applied to smaller ligaments. The emergence of the buckle transducer pushed biomechanical ligament analysis ahead in understanding ligament function by directly measuring the tensile force carried by the ligament; unfortunately, application of the transducer prestresses the ligament and changes its operating range.[6,22]

In the late 1980s and early 1990s, studies of ligament function were expanded to the ankle and wrist, with the adaptation of instrumentation used in the large ligaments of the knee to the smaller ligaments of these other joints. In 1988, Renstrom and Arms[26] used the Hall effect transducer to measure *in situ* strain in cadaver ankle ligaments. In 1990, de Lange et al.[15] performed an *in situ* study of the strain in a number of ligaments of the human cadaver wrist. Their group used a biplanar radiography method by which the three-dimensional positions of tantalum balls placed within the ligamentous substance were determined during functional loading of cadaveric wrists. This method produced a large amount of detailed information since strain within different regions of the ligament could be detected. In 1993, Acosta et al.,[1] adapted a smaller Hall effect transducer for use in measuring the *in situ* strains of wrist ligaments. In 1994, Kristal et al.[21] and Weaver et al.[34] developed the ligament tension transducer (LTT) for application to measuring the functional strains in eight ligaments of the volar side of the wrist. This device allowed the study of very small ligaments, less than 1 cm in length, and provided an *in situ* static measurement of force that did not change the function of the ligament. All other techniques described measured ligament strain and only provided indirect indications of mechanical function. On the other hand, the other transducers allow continuous measurement so that dynamic testing can be performed.

Apart from the need to further study smaller ligaments experimentally, mathematical models can be used to describe ligament properties. At the macromolecular level, both tendons and ligaments are primarily made of type I collagen. Considerable attention has been paid to models of tendon mechanical function, but there has been little focus on ligaments. If the cross-sectional shape of a ligament varies during loading, changes in the overall material and mechanical properties occur.[10,36] Ligaments have many different cross-sectional shapes and thicknesses which makes modeling challenging and indicates that experimental measurement will continue to provide a significant source of information.

Comparison of *In Situ* and *In Vitro* Models

In vitro and *in situ* models have been used to evaluate the properties of ligaments. An *in situ* measurement is taken on a ligament that has not been removed from its anatomic setting, while an *in vitro* measurement

is taken on a ligament that has been harvested. For determining stress-strain behavior, the *in situ* model comes closer to simulating the *in vivo* behavior. When using an *in vitro* approach, measurement of the initial *in situ* ligament length should be made before removal of the ligament. This defines the operating condition of the ligament, for example, its prestress condition. *In vitro* testing must consider the anatomic directions in which the load is applied, which may not necessarily be along the axes of the ligament fibers. Another difference between the two approaches is that ligamentous specimens tested *in vitro* experience end effects from clamping to the mechanical testing machine. Such enforced boundary conditions change local stress fields about the anchor points, and may cause differences in mechanical behavior. Therefore, one can see that an *in situ* experimental model approximates the *in vivo* condition better than the *in vitro* model does.

Biomechanical Properties of Ligaments

Ligaments do not follow the laws of continuum mechanics, so they cannot be modeled as ideal elastic solids.[17] In this section, solid continuum mechanics aspects are discussed since they provide a framework for understanding ligament behavior. Then, ligament viscoelastic or time dependent properties are demonstrated since they, too, have significant effects on measured properties.

An ideal elastic solid can be modeled using Hooke's law, which states that stress is directly proportional to strain and Young's modulus. From the theory of elasticity, any ideal isothermic and isotropic elastic-solid can be three-dimensionally modeled by the following equations.

$$\frac{\partial T_{ij}}{\partial X_j} + \rho b_i = \rho \frac{\partial^2 u_i}{\partial t^2}; \begin{matrix} i = 1-3 \\ j = 1-3 \end{matrix} \tag{7.1}$$

$$T_{ij} = \lambda E_{kk} \delta_{ij} + 2\mu E_{ij} \tag{7.2}$$

$$E_{ij} = \frac{1}{2}\left(\frac{\partial u_i}{\partial X_j} + \frac{\partial u_j}{\partial X_i} \right) \tag{7.3}$$

Eq. 7.1 represents three equations of motion which satisfy force equilibrium. The first term represents the sum of traction vectors expressed in three orthonormal directions. The second term is the sum of all body forces acting on an object. The last term is the sum of all the resultant accelerations; ρ is the mass density, T_{ij} is the stress tensor, and u_i is the displacement vector. Eq. 7.2 is Hooke's law, rewritten in indicial notation. The first term is the Cauchy-Green stress tensor. The second term identifies volumetric strain, and the third term identifies shear strain. E_{ij} is the strain tensor; μ and λ are Lamé constants. Eq. 7.3 is the set of geometric compatibility equations. The strain tensor is a function of orthonormal displacements and lengths. Fifteen unknowns are presented in Eqs. 7.1 through 7.3: 6 stresses, 6 strains, and 3 displacements.

Since most ligaments are tested with uniaxial loading, the theory of elasticity can be reduced to Eq. 7.4, where E is Young's modulus, T_{xx} is uniaxial stress and ε_{xx} is uniaxial strain. This is shown experimentally in Fig. 7.1.

$$T_{xx} = E\varepsilon_{xx} \tag{7.4}$$

We cannot accurately model solid biological tissues as Hookean solids. Biosolids differ from Hookean solids because of their nonlinear characteristics, viscoelasticity, and plasticity. Three phenomena define viscoelasticity: hysteresis, creep, and stress relaxation. Hysteresis, shown in Fig. 7.1, occurs when the stress-strain curve shifts during cyclic loading, since less energy is returned during the unloading phase of the test. Creep, as shown in Fig. 7.2A, occurs when a body continues to deform under a constant

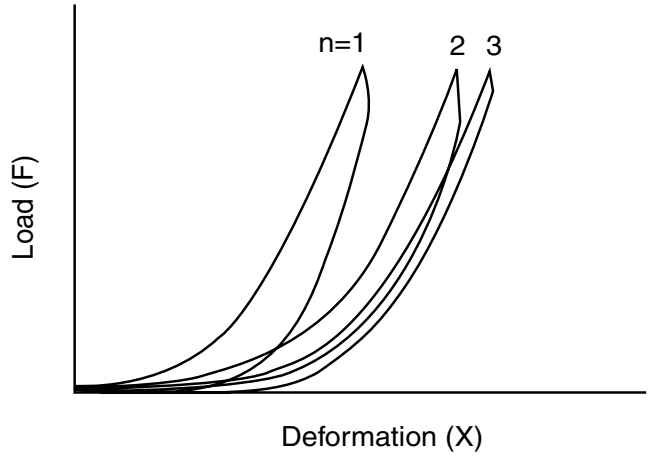


FIGURE 7.1 Example of the results of uniaxial tensile testing of a ligament demonstrating nonlinear response and hysteresis or energy loss with unloading. (*Source*: Fung, Y.C., *Biomechanics: Mechanical Properties of Living Tissues*, Springer-Verlag, NY, 1981. With permission.)

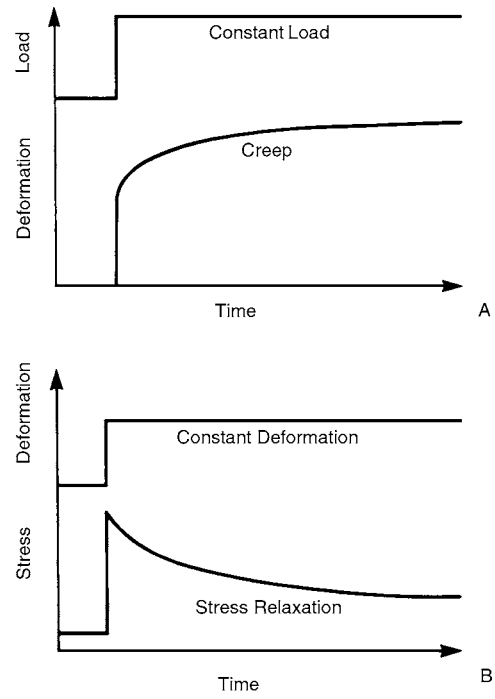


FIGURE 7.2 A. Typical response of a ligament to a step load demonstrating creep or continued deformation. B. Response of a ligament to a step deformation demonstrating stress relaxation. (*Source*: Mow, V.C. and Hayes, W.C., Eds., *Basic Orthopedic Biomechanics*, Raven Press, NY, 1991. With permission from Lippincott Williams and Wilkins.)

stress, preventing the establishment of equilibrium. Stress relaxation, as shown in Fig. 7.2B, occurs when the deformation is maintained constant and the stress decreases.

Fung introduced a mathematical framework to characterize viscoelastic behavior in soft tissues.[18] Fung's law is known as the quasi-linear viscoelasticity law, and shown in Eq. 7.5.

$$\sigma(t) = f\{\varepsilon(t)\} + f'\{\varepsilon(t - \tau);\ t,\ \tau\} \tag{7.5}$$

From Fung's law, $\sigma(t)$ and $\varepsilon(t)$ signify stress and strain at any given time t. The term $f\{\varepsilon(t)\}$ represents a function of time-dependent strain, and the $f'\{\varepsilon(t - \tau);\ t,\ \tau\}$ term represents a function of the whole time history. Haut and Little have modified this equation in analysis of the biomechanics of rat tail tendon.[19] Later sections of this chapter will present a more in-depth study of ligament viscoelasticity. For more information outside of the scope of this chapter, consult Fung's and Viidiks' chapters in *Handbook of Bioengineering* ,[31] and *Biomechanics of Diarthrodial Joints* .[32]

7.3 Measuring Biomechanical Properties of Ligaments *In Situ*

We strongly believe that *in situ* tensile load determination is a more direct measurement of ligament function than *in situ* strain since force must be inferred indirectly from the measured strain.[34] This force estimation may be achieved through the use of Fung's law that was modified by Haut and Little and Butler et al.[11,19] Another indirect method of determining load-carrying capabilities was performed by Huiskes (1991, unpublished). After ligament strain was measured *in situ*, each ligament biomechanical unit was removed from the wrist and force-displacement curves were measured *in vitro*. A direct *in situ* tensile force measurement technique eliminates the potential uncertainties associated with measuring *in situ* strain and then converting the data to force. In the next section, the methodology, strengths, and weaknesses of the measurement techniques for *in situ* strain and force measurement in ligaments are discussed. The description of the ligament tension transducer concludes this chapter.

Liquid Metal Strain Gage

The liquid metal strain gage (LMSG) transducer system is the combination of an LMSG as the primary sensing element and its supporting electronic hardware. The LMSG is an electromechanical transducer; it reads a length change and outputs a voltage. The LMSG is a mercury-filled silastic tube incorporated into electrical wire. This simple configuration is a powerful feature because mercury is a naturally occurring liquid-element that is very conductive and the system is highly compliant while accommodating large strains. The LMSG has a linear response when the operating range is kept below 40% strain, due to direct extension of the length of the silastic tube and a corresponding decrease in tube cross-sectional area, both of which change resistance across the gage (Fig. 7.3). If an LMSG is stretched above 140% of its total length, the electrical response deviates into nonlinearity. The LMSG can be used *in vivo*, to measure strain history in dynamic loading. Brown reported the LMSG dynamic response to be flat to 50 Hz and without phase shift.[9]

The electrical resistance of the LMSG changes with the change in length and cross-sectional area of the mercury column within; therefore, the voltage drop across the supporting electronic hardware will correlate to a specific length change. Stone et al.[30] have derived this relationship, which is shown in Eq. 7.6, where R is the resistance of the gage, ΔR is the change in resistance due to strain, and ε_l is the axial strain along the length of the gage.

$$\frac{\Delta R}{R} = 2\varepsilon_l \tag{7.6}$$

The LMSG can be used in either of two testing configurations: a Wheatstone bridge or a series circuit.[23] In the Wheatstone bridge, the LMSG is placed in series with one arm of the bridge. The series-circuit configuration has the LMSG in series with a drop-down resistor. The outputs of both circuits must be amplified to increase resolution. Meglan[23] pointed out that the series-circuit configuration is ten times more sensitive than the Wheatstone bridge; however, the output of the series circuit is not truly linear. The Wheatstone bridge has great linear response, but lacks sensitivity. The sensitivity of the series-circuit configuration can be enhanced by increasing the current passing through the system, but that would

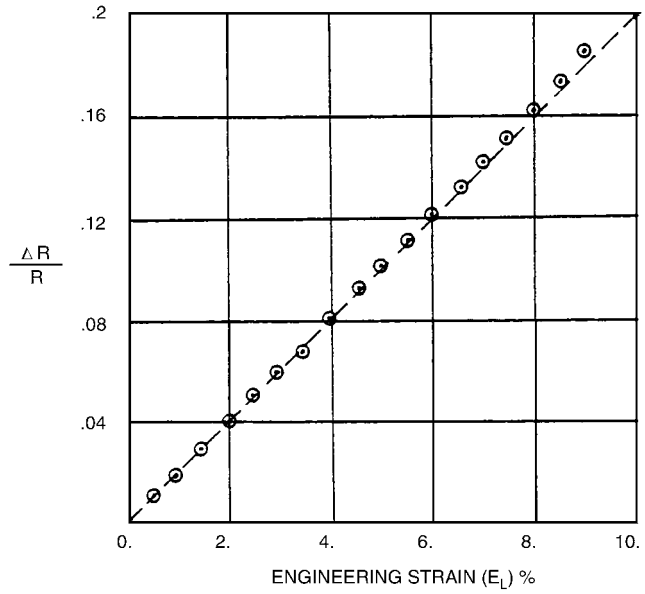


FIGURE 7.3 Liquid metal strain gage (LMSG) performance in terms of change in resistance divided by original resistance with increasing engineering strain. (*Source*: Stone, J.E., Madsen, N.H., Milton, J.L., Swinson, W.F., and Turner, J.L., *Experimental Mech*., 132, June 1983. With permission from Sage Publ.)

increase heating of the LMSG. Too much heat generation causes LMSG response to become more nonlinear. Another caveat of the series-circuit configuration is that the value of the drop-down resistor has to be chosen carefully. The purpose of this resistor is to minimize the effects of electrical heating.

The LMSG has been used on cadaver knees *in situ* for quasi-static tests[23,30] and to measure strains in cruciate ligaments. Gages can be attached to these ligaments by either of two methods. One method is to use a contact cement that bonds well to biological tissue. The lead wires should be secured to the tibia and femur at the ligament insertion points. This action assures parallel alignment of the gage with respect to the ligament. The second method of gage attachment is to suture the lead wires of the gage to the ligament itself. The LMSG should be pre-stressed when attached to the ligament, so it is operating in its linear range. Strain of 5 to 10% is ideal, assuming no compression will take place. This is an important precaution because the LMSG can only measure tensile strains.

The LMSG has several limitations. A great deal of care must be exercised when handling an LMSG. If the silastic tubing is over-stretched, it may rupture and leak mercury into the environment. A typical LMSG has a shelf life of 6 months, because the mercury slowly oxidizes out of the silastic tubing. The anchoring method of the LMSG is not completely reliable. The suture method requires less space to mount the gage, but the ligament must be pierced for anchoring. The act of piercing holes into the ligament changes its properties. The suture method allows potential slack in the LMSG-ligament system, introducing hysteresis. The glue method of attachment is fragile. Also, the LMSG requires a minimum amount of space to operate, but cannot be used on small ligaments in confined spaces. The LMSG records surface strain between its attachment points, not necessarily the average strain throughout an axial cross-section.

The positive characteristics of the LMSG outweigh its limitations. The output of the LMSG is very linear when used with a Wheatstone bridge. The linear operating range of the transducer is very large, so it is suitable for biologic tissue response. The LMSG is inexpensive, easy to use, easy to calibrate, fast to set up, and capable of both static and dynamic strain measurement. The LMSG is also easy to manufacture. Brown et al.[9] made their own because the commercially available products were too large for *in vivo* studies. However, it must be emphasized that this device measures ligament strain, not force, which still must be determined indirectly.

Hall Effect Transducer

The Hall effect strain transducer (HEST) is the combination of a Hall effect transducer and supporting electronics. The HEST is an electromagnetic device; it reads a change in a magnetic field and outputs a voltage drop that is proportional to the magnetic field. As shown in Fig. 7.4A, a simple Hall effect transducer is a small instrument made of only three parts: a magnetic wire, a Teflon casing, and a Hall effect semiconductor. The semiconductor is anchored to the Teflon casing and the magnetic wire is free to slide in and out of the casing. The midrange response of an HEST, from 10 to 40% strain, is linear, as shown in Fig. 7.4B, but measuring at its extremes produces very nonlinear results.[4] The HEST is closely related to the linear variable differential transducer (LVDT) in principle. The Hall effect semiconductor detects the proximity of a permanent magnet; consequently, it produces a voltage drop that is proportional to the strength of the magnetic field.[26]

The HEST device requires only a current source and a precision amplifier. Because the operating range is from 10 to 40%, it is extremely important to anchor the HEST with 20% strain onto the ligament in its rest position. Otherwise, one runs the risk of measuring in a nonlinear range with a linear calibration curve. There are two methods of anchoring an HEST to a ligament: suturing, or piercing the ligament with barbs. Both methods anchor the device by piercing the ligament substance.

The HEST is extremely versatile. Acosta et al.[1] have used the HEST in osteoarthritic human cadaver wrists to measure *in situ* strain in the dorsal and palmar distal radio-ulnar ligaments (DRUL) before and after reconstruction of the distal radio-ulnar joint (DRUJ). Cawley et al.[12] have used the HEST to define *in situ* biomechanical parameters of ankle collateral ligaments during physiologic foot motion. The HEST has been used by Arms et al.[4,5] in cadaver knees to define MCL and ACL properties. Erickson et al.[16] studied dynamic *in vitro* properties of human MCL and ACL in prophylactic knee braces using HEST devices applied *in vivo*.

Buckle Transducer

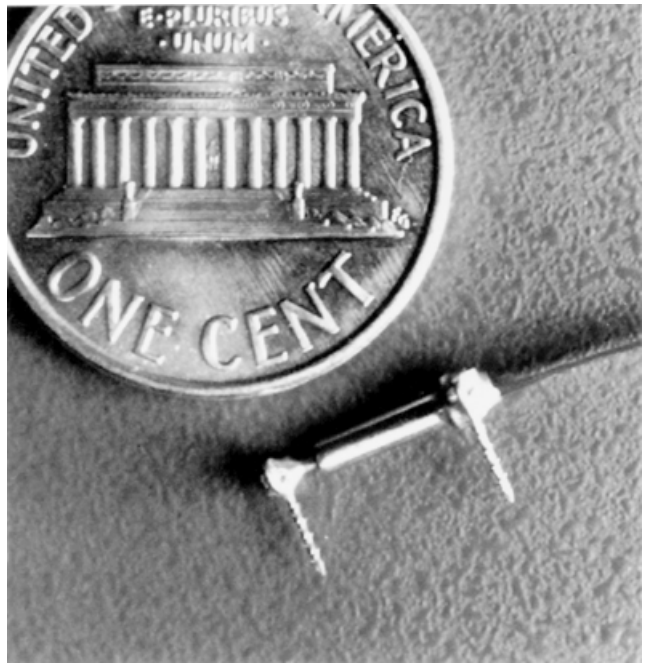
The buckle transducer works by slightly deflecting the normal configuration of a load-carrying flexible element in three-point bending due to interaction with the ligament. Tension in the ligament fibers causes the ligament to straighten, thereby bending the crossbar and frame of the regular buckle transducer and bending the buckle beam of the modified buckle transducer.[6]

The design of the transducer is specific to the individual ligament. The design, illustrated in Fig. 7.5, is based on the following parameters: (1) ligament parameters: the ligament thickness t, the length L_l, and the expected maximum tension T; (2) transducer performance parameters: the tolerable amount of ligament shortening due to transducer implantation, S_t; (3) transducer material parameters: Young's modulus and yield strain of the chosen metal must be known, and (4) transducer geometric parameters: the minimum width of the transducer b, and the transducer length L_i.[6]

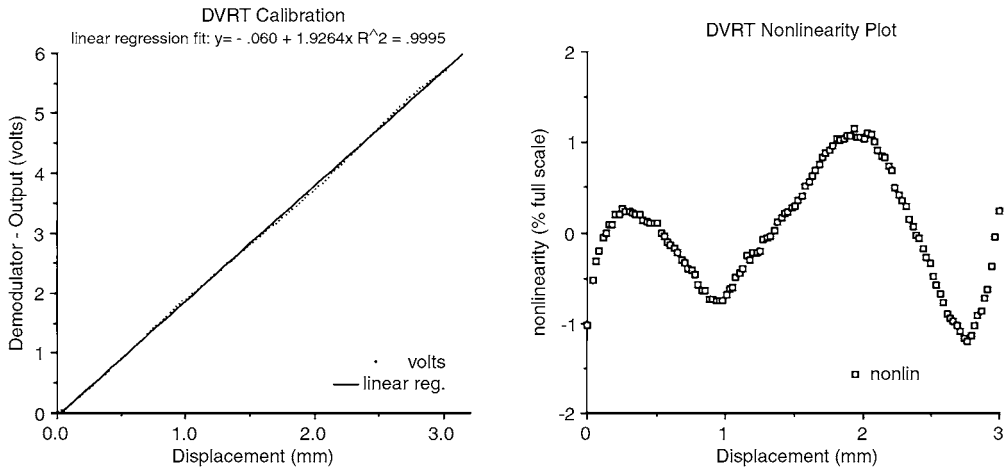
$$\sec\theta = \left[S_t\left(L_l / L_i + 1\right)\right] \tag{7.7}$$

From Eq. 7.7, we see how the transducer offset angle θ is determined from the transducer geometry. The offset angle can be used to calculate the section modulus of the beam, where the maximum strain is set at the beam's midsection. For this calculation, the transducer is modeled as a simply supported beam in bending, affected by an applied load P, as shown in the top portion of Fig. 7.5. The tensile force can be determined from the product of the section modulus and the strain gage output. Eq. 7.8 shows how the dimensions of A and H_c are determined transducer parameters. The d_c is the center of deflection of the transducer, and L_c is the width of the clip.

$$\mathrm{Tan}\Theta = \left[2\left(A - H_c + \mathrm{t} - d_c\right)\right] / (Li - Lc) \tag{7.8}$$



A



B

FIGURE 7.4 A. Photograph of a Hall effect strain transducer (HEST). B. Ligament strain and resulting force for two different ligaments with and without the buckle transducer indicating the pre-stress effect of the transducer itself. (*Source*: An information brochure, MicroStrain, Burlington, VT. With permission.)

The buckle transducer strain gages form two arms of a Wheatstone bridge. Each strain gage is 120 ohms.[2,3,6] The buckle transducer is attached to a ligament simply by snapping both halves of the transducer together, with the ligament between the halves. During installation, it is important to keep in mind that if too much tissue is inserted, excessive ligament shortening occurs. If not enough ligament tissue is inserted, the signal-to-noise ratio will be too small.[6,22]

Once the transducer is placed on the ligament, the transducer can be calibrated *in situ.* This is done by clamping forceps on the ligament, only a few millimeters from the buckle frame, and then looping a string through the forceps. The other end of the string is attached to a calibrated spring scale. Pulling

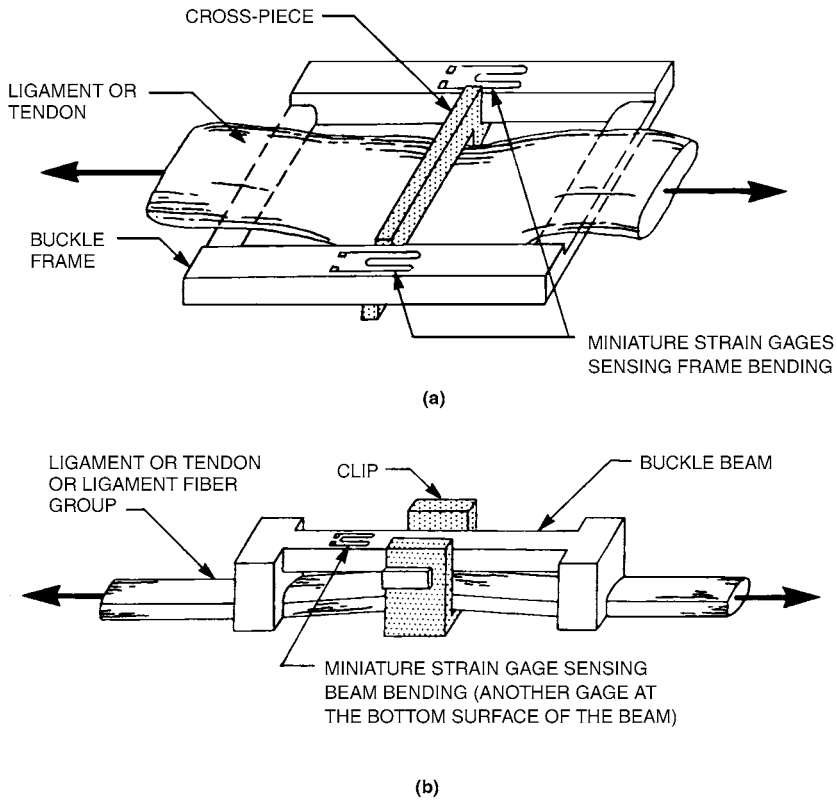


FIGURE 7.5 Schematic diagram of a buckle transducer. (*Source*: Barry, D. and Ahmed, A.M., *J. Biomech. Eng.*, ASME, 108, 149, 1986. With permission.)

on the scale applies a known force through the buckle and ligament, and results in a measurable buckle response.[22]

The effect of a poorly pre-conditioned ligament is more apparent in a buckle transducer than any other device.[6] If the tendon or ligament was not pre-cycled long enough before testing, a noticeable drift in response will be witnessed, as shown in Fig. 7.6. This drift in response is due to the morphological changes of the tissue; moreover, the cross-sectional area changes when the tissue is loaded infrequently, resulting in poor repeatability. The mere act of attaching the buckle transducer onto a ligament causes changes in its length. Once the buckle is locked in place, the resting length of the tissue is shortened because of the path it must take. The presence of the buckle transducer changes the local stresses and boundary conditions at the site to which it is attached.[6] Shortening the ligament changes its stiffness, pre-stressing.[6] Because of these effects imposed on the ligament, it is essential to test the transducer for repeatability during calibration. The main advantage of the buckle transducer is that it measures bulk ligament force directly.

Roentgenstereophotogrammetric Analysis

Stereophotogrammetry is the use of multiple two-dimensional pictures of three-dimensional objects to reassemble a three-dimensional image.[27] The term stereo indicates the reconstruction process of 3-D image building and the prefix *roentgen* indicates that X-rays are used to obtain the image. Roentgenstereophotogrammetry analysis (RSA) is a three-dimensional radiographic technique used to study joint motion pathways. While rigid body joint motion is the primary focus of this technique, it can also be

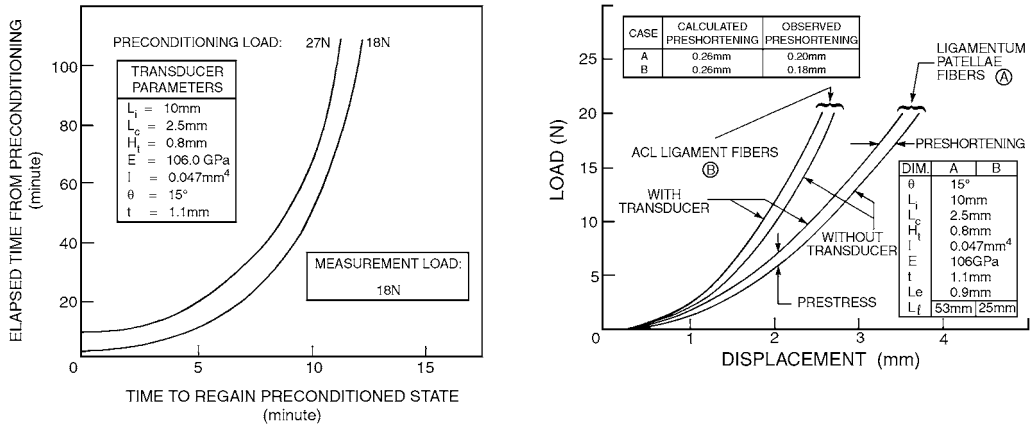


FIGURE 7.6 A. The relationship of the time required for a ligament with a buckle transducer attached to regain its pre-conditioned state based on the time elapsed from pre-conditioning. B. Ligament strain and resulting forces for two different ligaments with and without the buckle transducer indicating the pre-stress effect of the transducer. (*Source*: Barry, D. and and Ahmed, A.M., *J. Biomech. Eng*., ASME, 108, 149, 1986. With permission.)

used to determine *in situ* strains in soft tissues. Tantalum pellets are used as X-ray markers because of their excellent radiopaque characteristics and biocompatability.[28]

The measurement is performed in two steps (Fig. 7.7). In the first step, after using calibration objects of known shape to locate the two X-ray sources, the intersection between the vectors from the X-ray source to the same point on the X-ray in each of the two planes defines the three-dimensional coordinates of the object to be reconstructed. In the second step, the changes in position of the object after loading can be defined using standard kinematic techniques. For ligament strain measurements the tantalum balls placed into the ligament substance are considered as points and the magnitude of the translation vector divided by its initial (unloaded) magnitude defines the strain of that tissue segment.[14,35]

An experimental setup performed by de Lange et al. is shown in Fig. 7.8. Two roentgen tubes (D) are used to radiograph the specimen. A hand-wrist joint specimen (A) is placed in front of a reference plate (C). Hand movements are controlled by a motion constraint device, and springs (B) are used to load the tendons during testing.[15] RSA has been used successfully in the knee,[7,8,20] wrist,[13-15] and the foot for *in vivo*, *in vitro*, and *in situ* studies.

The successful use of the RSA technique requires accurate knowledge of the locations of the X-ray sources. Therefore, the precision of the calibration process is of fundamental importance. The process is performed on a structure that has known dimensions and is outfitted with tantalum markers; moreover, it is recommended that nine markers which are not coplanar with each other be used.[28] The markers in the test cage function as calibration points, and are X-rayed on the same film as the object. Calibration markers and object markers are exposed from the two separate roentgen foci. The cage markers are of two kinds: fiducial marks and control points. The fiducial marks are used for projective transformations of the image points to the laboratory coordinate system. The control points are used for determining the roentgen foci positions in the same (fiducial) coordinate system. Finally, the three-dimensional coordinates of an object in the test cage can be determined by locating the intersection of the vectors between the roentgen foci and the transformed image points.[27]

This technique has several advantages and disadvantages. The calibration procedure is complex and long. Roentgen film cassettes are not uniformly flat, and that will affect the geometry of the system. It is difficult to maintain specimen alignment throughout an entire range-of-motion recording. The extreme markers must be in the same locations, from one specimen to another.[20] The pellets must be inserted into the ligament by opening a space and gluing the pellets in place.[20] Finally, only static measurements can be made. The system is expensive, and a risk of radiation exposure exists.

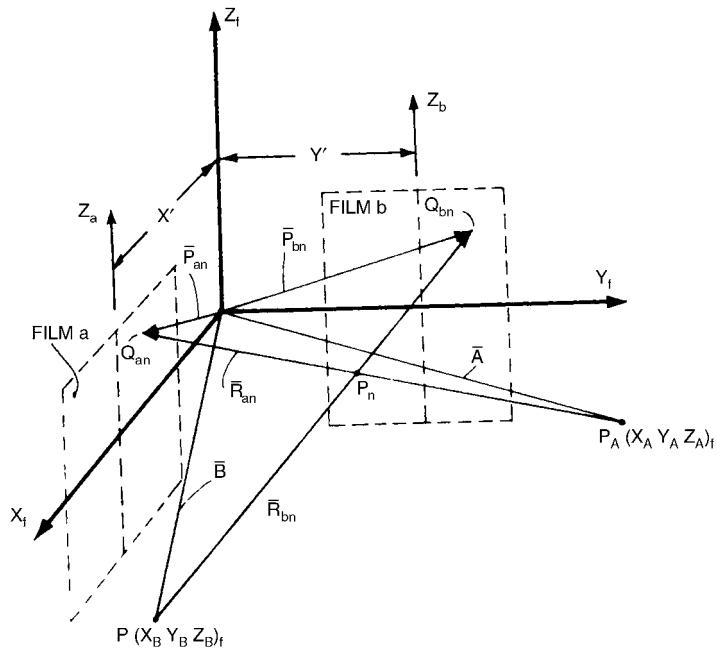


FIGURE 7.7 Schematic diagram of determination of location of a marker point in 3-D space using roentgenstereophotogrammetric analysis (RSA). P and P_A are ideal locations of the X-ray point sources. The vectors Q_{an} and Q_{bn} connect the X-ray sources and the image of the point on each radiograph. P_n is the point in space. (*Source*: Huiskes, R., Kremers, J., Lange, A., de Woltring, H.J., Selvik G., and van Rens, T.J.G., *J. Biomech.*, 18, 559, 1985. With permission from Elsevier Science.)

RSA has two major advantages that all the other transducer systems lack. First, it has been used successfully to make *in vivo* measurements since the placement of tantalum balls into the bones of volunteers has been well tolerated. Second, other techniques only measure bulk tissue strain at the location of the transducer. Arms et al.[4] have shown that the MCL has consistently different strain patterns between the proximal, middle, and distal segments of the anterior and posterior borders. Butler et al. have shown similar findings in the ACL.[11] With RSA, one can measure the local strain wherever two tantalum markers exist. RSA allows the biomechanist to determine complete ligament strain, including bending of the ligament around a bony prominence. Further, RAS has no effect on ligament strain due to application of the technique, unlike the buckle transducer which pre-strains the ligament with insertion.

7.4 Ligament Tension Transducer System

The ligament tension transducer system (LTTS), shown in Fig. 7.9, is based on the qualitative test for ligament integrity performed in surgery which consists of simply pulling the ligament in question in a direction transverse to its long (functional) axis, and estimating its tension. In addition, this method of displacing a cable segment of known length transversely and measuring the transverse force and deformation is used for the quantitative measurement of cable tension in cable rigged structures (such as sailboat masts). In the ligament testing version, a linearly variable differential transformer (LVDT) is used to measure the small transverse deformation applied, and a small load cell provides the force required to do so. During testing, the transducer and specimen must be fixed in space. The probe is placed beneath the ligament being studied, and the displacement screw is turned to first engage and then displace the ligament. The LTTS has been used in two wrist ligament studies. Kristal et al.[21] used it on five ligaments in seven cadaveric hand specimens to determine which ligaments act as key passive motion limiters. An expanded study by Weaver et al.[34] tested eight wrist ligaments to increase the comprehensiveness of the

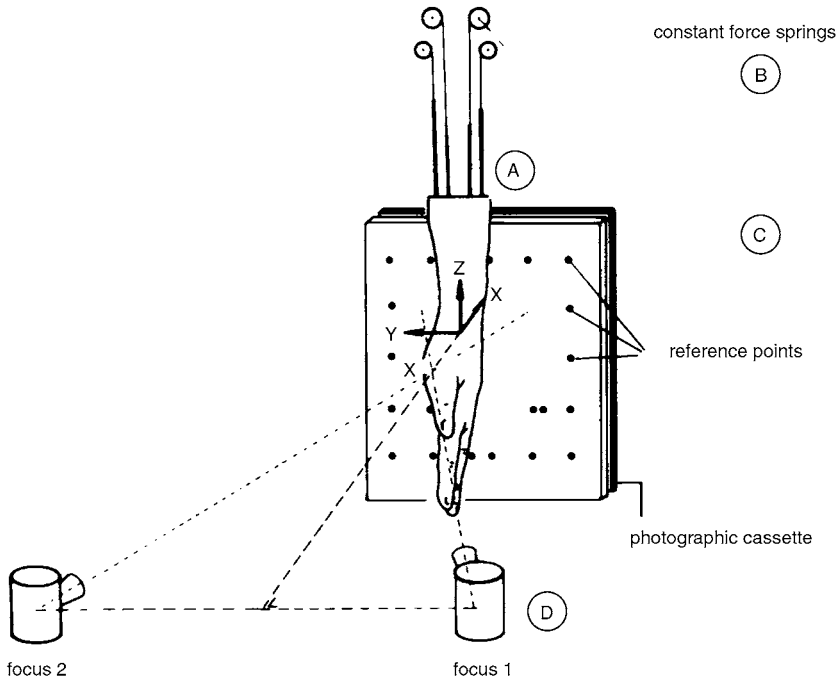


FIGURE 7.8 Example of an experimental setup of the X-ray cartridge, sources, and specimen for RSA. (*Source*: Huiskes, R., Kremers, J., Lange, A., de Woltring, H.J., Selvik G., and van Rens, T.J.G., *J. Biomech.*, 18, 559, 1985. With permission from Elsevier Science.)

experiment. It is important to note that most of the ligaments tested were very small, less than a centimeter in length.

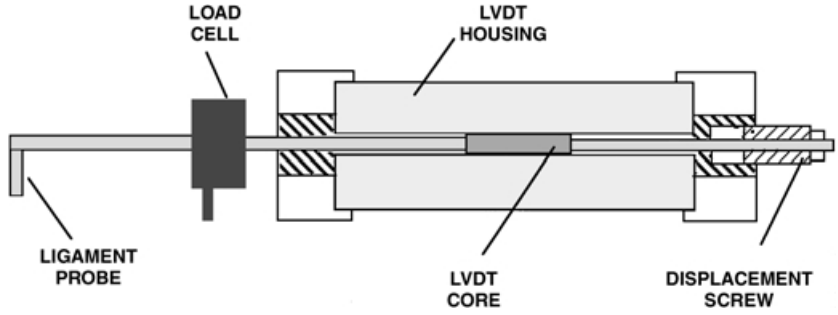
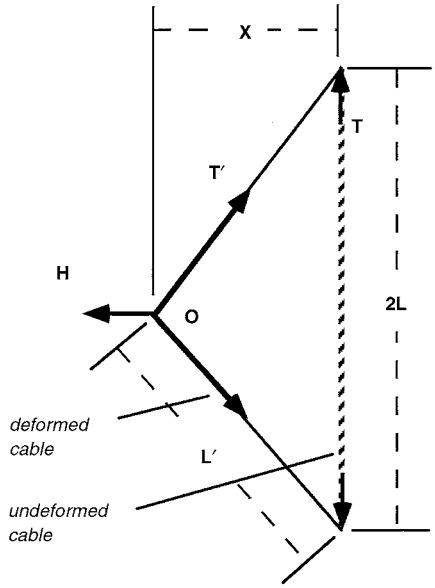


FIGURE 7.9 Schematic diagram of the ligament tension transducer. The probe tip fits behind the ligament. The load cell measures the force required to displace the ligament transversely. The LVDT measures the displacement of the probe which is controlled by the displacement screw. (*Source*: Kristal, P., Tencer, A.F., Trumble, T.E., North, E., and Parvin, D., *J. Biomech. Eng.*, ASME, 115, 218, 1993. With permission.)

The LTTS determines the tensile load in a ligament of known length by measuring the magnitude of the deflection and the force required to do so. For a cable that has a circular cross-sectional area, (Fig. 7.10), the tension in the deformed cable is given in Eq. 7.9.

$$T' = T + K(L' - L) \tag{7.9}$$



where:
T′ = tension in the deformed cable
T = tension in the undeformed cable
K = stiffness of the cable
L′ = length of the deformed cable
L = length of the undeformed cable
*cable is deformed by pulling it laterally

At point O where the perpendicular deforming load is applied:

T′ = 1/2 * (L′ / X)* H,

where:
X = imposed perpendicular deformation
H = applied perpendicular force

The tension in the cable is:

T = 1/2 * (L′/X) * H - K (L′ - L)

FIGURE 7.10 Resolution of ligament tensile force using the ligament tension transducer system (LTTS). (*Source*: Weaver, L., Tencer, A.F., and Trumble, T.E., *J. Hand Surg.*, 19A, 464, 1994. With permission from W.B. Saunders.)

where, T' is the tension in the deformed cable, T is the true tension in the undeformed cable, K is the stiffness of the cable, L is the length of the undeformed cable, and L' is the length of the deformed cable. Then,

$$T' = H/2\left(L'/X\right) \tag{7.10}$$

Considering equilibrium at the point where the lateral load is applied, Eq. 7.10 states that the sum of the forces is zero. H is the applied lateral load, and X is the imposed lateral deformation. By combining Eqs. 7.9 and 7.10, the tension in the cable can be determined from Eq. 7.11.

$$T = H/2\left(L'/X\right) - K\left(L' - L\right) \tag{7.11}$$

The first term describes the force balance at the point when H is applied. The second term, the ligament elongation term, describes how the deformed length and stiffness of the cable add to the initial tension in the cable.

The measurement verification process is performed in three steps: verification of the theory using a circular nonbiological cable; *in vitro* comparison of measured to known tension in a typical ligament; and *in situ* ligament tension verification. The test using a circular cross-section cable is necessary to verify the fundamental theory. A nylon cable can be used with a materials testing machine for this step.[21] The load cell of the materials testing machine should be attached to the nylon cable, so recordings of the actual tension in the undeformed cable can be made for comparison to those measured by the LTTS. During this step, it is important to test the effect of nonperpendicular probe orientation.

Bone-ligament-bone preparations should be used for the *in vitro* verification step. Similar to the round cable calibration, the ligament preparations can be placed in a material testing machine, with one end

of the ligament attached to the load cell so that the true bulk ligament load is known. Ligaments are more challenging to test than cables for several reasons. Ligaments are not perfectly round, and typically have varying cross-sections along their lengths. Their viscoelastic behavior causes creep when the transverse load is applied. Another problem involves the stiffness term, shown in Eq. 7.11. If it must be included, then the stiffness of the ligament must be determined separately, adding considerable complexity to the measurement procedure, similar to the problem encountered by techniques that measure only strain. A solution is to choose a transverse deformation[21] that makes the stiffness term insignificant. See Table 7.1.

TABLE 7.1 A Summary Comparison of Different Techniques for Measurement of Ligament Strain and Force

Author	Device	Type	Comments
Stone et al.[30]	Liquid metal strain gage	Strain	Static and dynamic measurements of large strains possible
Arms et al.[4]	Hall effect strain transducer	Strain	Static and dynamic measurements, local strains
Barry and Ahmed[6]	Buckle transducer	Force	Static and dynamic measurements, preload ligament
Huiskes[20]	Roentgenstereophotogrammetry	Strain	Static, noncontact so no loading effect, multilocation possible
Kristal[21]	Ligament tension transducer	Force	Static, small ligaments measurable, does not preload ligament

Studies performed by Kristal et al.[21] have shown that during the *in vitro* ligament calibrations, the LTTS was accurate to within 8%. Kristal also pointed out that the LTTS tends to overestimate higher loads and underestimate lower loads. Nonperpendicular probe orientation increases the force required to laterally deform the ligament. An offset of 10° increases the error by 1%; an offset of 20° increases the inaccuracy by 6% (Table 7.2). For the lengths of ligaments encountered in the wrist studies, a transverse displacement of 0.50 mm was imposed so as to neglect the stiffness term in the force calculation. To test for reproducibility, fresh-frozen specimens were thawed, tested, refrozen, thawed and tested again. Thus testing encompassed specimen setup as well as LTTS errors. The overall mean ratio of measured axial tension between first and second trials of any ligament was found to be 1.05 with a standard deviation of 0.29.[21]

TABLE 7.2 Effects of Error in Ligament Length Measurement and Nonperpendicular Probe Alignment on Measured Ligament Tension

Variable	Estimated Error	Error in Measurement
Ligament length	0.30 mm	5.2%
Probe orientation	10° from perpendicular	1.0%
	20° from perpendicular	6.0%

Several assumptions are made when using this technique. One involves estimating the free length of a ligament which may have a broad attachment area. Typically a pair of modified calipers is slid under the ligament until the jaws contact bone. This free length measurement may underestimate the true free length of the ligament. A second assumption is that the bones to which the ligament attaches do not move during the measurement procedure. This can be tested by placing a displacement gage on the bones to which the ligament is connected and determining whether any displacements occur to the connecting bones during the measurement procedure. A third assumption is that the ligaments do not bend around bony prominences. Since some do, which changes the pure tensile force in the ligament to combined tension and bending, the technique cannot be used for these ligaments.

This technique and the others have certain advantages and disadvantages that are summarized in Table 7.1. The LTTS measures tensile force directly. It has minimal effect on the tissue it measures, for example, and does not cause ligament shortening as the buckle transducer does. The LTTS is not anchored to a ligament in the manner that the LMSG and HEST require for operation, so the ligament is not damaged during testing. It is possible for ligament damage to occur when the probe is placed behind a ligament, but this problem can be avoided if the ligament probe tip is bluntly machined. The LTTS measures an average tensile load, unlike the local tensile force measured by a modified buckle transducer or local strains measured by an LMSG or HEST. The LTTS components are moderately inexpensive to manufacture or purchase, and simple to assemble.

As an example of the potential of this technique, a previous experiment on wrist mechanics is briefly described. Five upper extremity specimens were obtained and eight ligaments on the palmar side of the wrist were identified (Fig. 7.11). The specimen was mounted in a positioning frame (Fig. 7.12), which permitted controlled measurable orientation of the hand and provided a stable platform for the ligament tension measurements. At each position of the hand, the ligament tension transducer was oriented perpendicular to the long axis of the ligament to be tested. The probe was hooked behind the ligament and the displacement screw adjusted until the probe tip was not in contact with the back surface of the ligament. It was then moved outward (i.e., transverse to the axis of the ligament) until contact was achieved. From this point, the load applied to laterally deform the ligament was recorded by a load cell (Model 31, Sensotec Precision Miniature Load Cells, Columbus, OH) and displacement by a linearly variable differential transformer (Model 100 DC-D, Shaevitz Engineering, Pennsauken, NJ). The lateral deflection was stopped at 0.5 mm. At that point the magnitude of the load was monitored until it stabilized (i.e., stress relaxation stopped). Since the actual deforming load was very small, that usually occurred within 30 seconds.

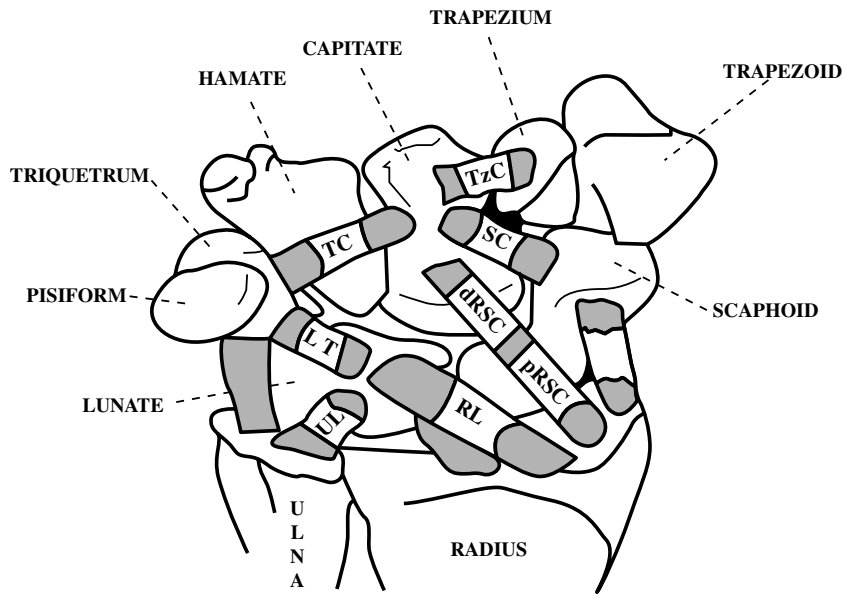


FIGURE 7.11 Ligaments of the palmar side of the wrist tested using the ligament tension transducer system. (*Source*: Weaver, L., Tencer, A.F., and Trumble, T.E., *J. Hand Surg*., 19A, 464, 1994. With permission from W.B. Saunders.)

After all testing was completed, the minimum lengths between ligament attachment sites were measured using a caliper (Enco Manufacturing Co., Santa Ana, CA) whose knife edge jaws were slid under the ligament and expanded until they encountered the bone ligament junction. This gave the minimum length between bone attachment sites. The error in ligament length from repeated measurements was

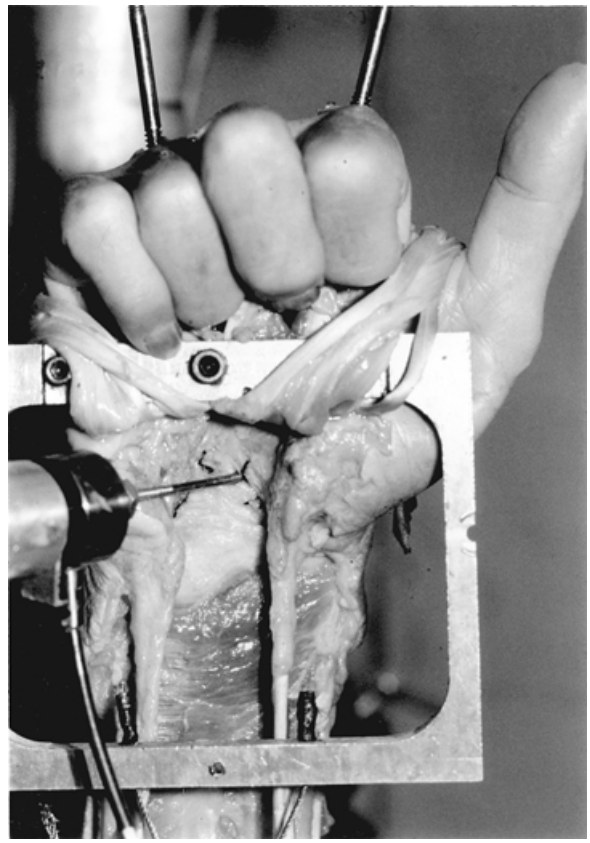


FIGURE 7.12 Experimental setup for measurement of ligament tension using the LTTS. *Source*: Weaver, L., Tencer, A.F., and Trumble, T.E., *J. Hand Surg*., 19A, 464, 1994. With permission from W.B. Saunders.)

0.30 mm, resulting in a 5.2% error in axial tension. An example of the results is shown in Fig. 7.13. As the hand was moved from radial deviation (the thumb points outward with the palm of the hand facing upward) to ulnar deviation (the thumb points inward and movement from radial to ulnar deviation involves motion in the plane of the palm), the radiolunate and ulnolunate ligaments appear to be the key stabilizers to excessive motion. The radiolunate ligament tension increases with maximum radial deviation while the tension in the ulnolunate ligament is relatively small, and the converse is true as the hand moves to maximum ulnar deviation. The following conclusions were made from the study: (1) the palmar ligaments of the wrist have inherent tension, even in the neutral positioned and unloaded wrist; (2) various ligaments play roles as passive stabilizers at the ends of the ranges of motion of the wrist, and (3) some ligaments have significantly greater tensions than others in any position.

7.5 Summary

A variety of techniques have been developed for measurement of soft tissue functional properties. *In situ* testing causes the least disturbance and should therefore provide the most accurate representation of ligament function. Measurement of strain provides only an indirect measure of the load carrying function of the ligament. Of more benefit is the measurement of ligament load directly. Of two transducers capable of measuring load directly, both the buckle transducer and the ligament tension transducer system (LTTS) have advantages. The buckle transducer can measure dynamic loads in a ligament but its installation pre-stresses the ligament tested. The LTTS can only measure static loads; however, it can be used on very small ligaments (less than 1 cm) and does not pre-stress the ligament.

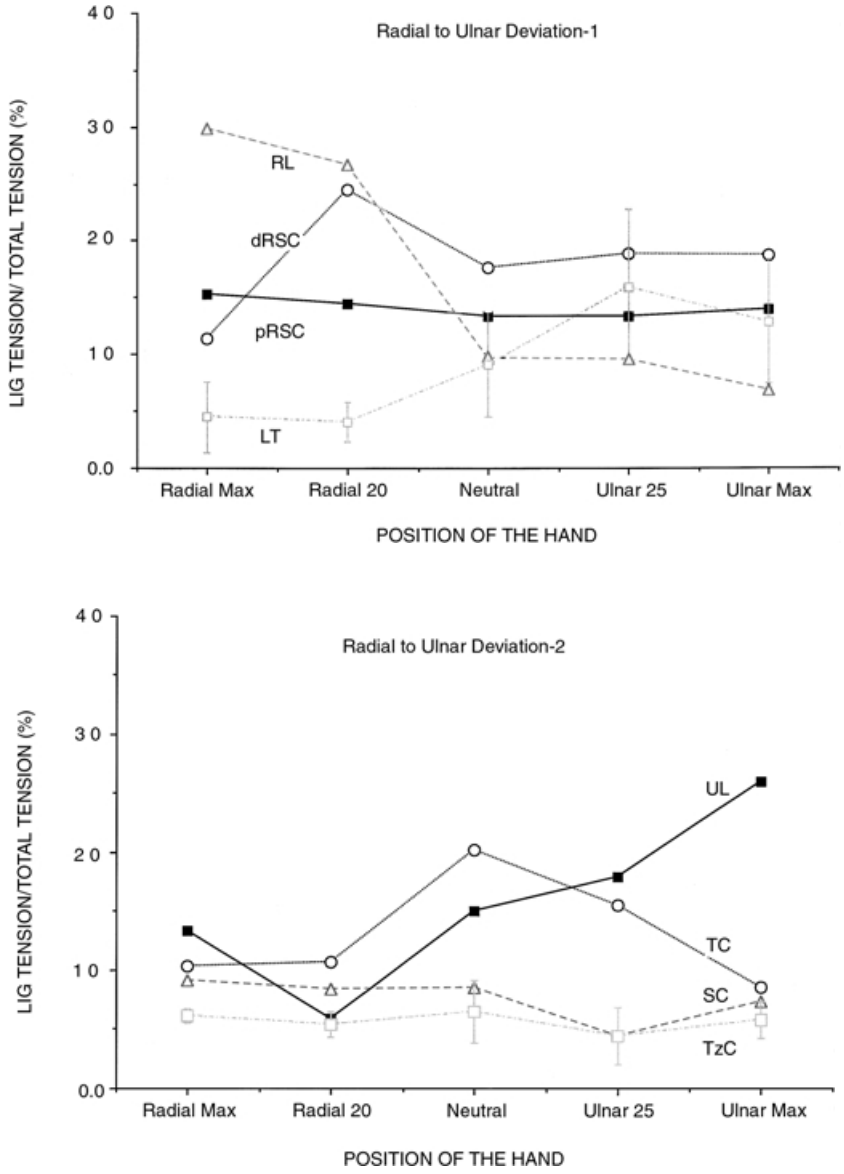


FIGURE 7.13 Example data from testing radial and ulnar deviation. (*Source*: Weaver, L., Tencer, A.F., and Trumble, T.E., *J. Hand Surg*., 19A, 464, 1994. With permission from W. B. Saunders.) *(© 1994 by The American Society for Surgery of the Hand. This material may not be reproduced, stored in a retrieval system, or transmitted in any form or by any means without the prior written permission of the publisher.)*

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