

COMPARISON OF MATERIAL PROPERTIES IN FASCICLE-BONE UNITS FROM HUMAN PATELLAR TENDON AND KNEE LIGAMENTS

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Abstract—The fascicle material properties in bone-fascicle-bone units were determined for the anterior and posterior cruciate ligaments (ACL, PCL), the lateral collateral ligament (LCL) and the patellar tendon (PT) from three young human donor knees. Groups of fascicles from each tissue were isolated with intact bone ends and failed at a high strain rate in a saline bath at 37°C.

In each knee tested the load related material properties (linear modulus, maximum stress and energy density to maximum stress) for the patellar tendon were significantly larger than corresponding values for the cruciate and collateral ligaments. Bundles from different ligaments in the same knee were similar to each other in their mechanical behavior. In addition, no significant differences were present in the maximum strains recorded for any of the four tissue types examined.

The results presented have implications in studies of ligament injury. They are also important in the design and use of synthetic and biological ligament replacements and in tissue and whole knee modeling.

INTRODUCTION

An approach to describing the human anterior (ACL) and posterior (PCL) cruciate ligaments has been proposed using structurally motivated mathematical models (Stouffer *et al.*, 1985). These models are based on the assumption that a ligament's mechanical response can be predicted if the geometric organization and mechanical properties of the component fascicles (fiber bundles) are known.

The authors have taken this approach to modeling from the results of two important observations. First a cruciate ligament is composed of fiber bundles having different lengths and orientations (Girgis *et al.*, 1975; Kennedy *et al.*, 1974, 1976; Noyes and Grood, 1976; Norwood and Cross, 1979). This implies that the mechanical response of the ligament depends greatly on the fiber bundle structure which is expected to differ from ligament to ligament within a donor or from donor to donor. Noyes and Grood (1976) highlighted this point, indicating that for a given ligament, non-homogeneous fascicle organization would produce lower ultimate loads than for similar structures where the fibers could be loaded more uniformly. Because of this they recommended that collagen material properties be determined in smaller specimens. Harkness (1968) also suggested that higher material properties could be anticipated when smaller specimens are employed.

Second, nonuniform loading in the total ligament is further accentuated by the orientation of the ligament's bone ends. Alm *et al.* (1974) examined the effects of tibial rotation on the failure loads of the anterior cruciate ligament-bone unit. They found that maximum anterior drawer force was obtained in

neutral tibial rotation and dropped with external or internal rotation. These results suggest that it could be difficult to orient the attachments of a whole ligament on a testing frame to produce consistently uniform loading of all the fiber bundles. It is easier to orient the insertion sites of a small portion of the structure to achieve more uniform loading of the fibers. Furthermore, cruciate fascicle-bone units with cross-sectional areas as small as 1–3 mm² can be prepared and mechanically failed using standard dissection and testing techniques.

Tissue types were selected for this study according to two criteria. First, all tissues had to be taken from the same knee to avoid the issue of biological variability. Second, both tissue ends had to insert into the bone through a zonal insertion site (Cooper and Misol, 1970; Noyes *et al.*, 1974) to reduce the chances for premature attachment site failures. This eliminated the medial collateral ligament which has a distal attachment directly onto periosteum.

The objectives of this work were twofold. First we sought to prepare and fail fascicle-bone units from the ACL, PCL, LCL and PT in the same human knee and correlate their material properties. Second, we wanted to determine the differences in material properties between the same tissue types from donors in a similar age range.

METHODS

Specimen preparation

Three human cadaveric knees were obtained from three young donors within 18 h postmortem. The age, sex, height, weight and cause of death for each donor are shown in Table 1. All knees appeared normal at autopsy. Manual forces were applied to the proximal tibia in an anterior-posterior direction while restrain-

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Table 1. Donor information

Donor No.	Age	Sex	Height (m)	Mass (kg)	Cause of death
1	30	F	1.63	65.8	Drug overdose
2	30	M	1.73	71.7	Electrocution
3	21	F	1.72	83.9	Suicide

ing the thigh. The resulting displacements were small and similar for left and right knees, suggesting integrity of the anterior and posterior cruciate ligaments (Butler *et al.*, 1980). The joints were removed with 15 cm of femur and tibia, doubly-wrapped in plastic bags and placed in a freezer (-30°C) for up to 3 weeks.

Each knee was removed from the freezer and thawed for 24 h in a refrigerator at 4°C . The patellar tendon (PT) was freed proximally by cutting along its medial and lateral borders and by isolating the patella from the quadriceps tendon and retinacular structures. The tendon and patella were then wrapped in saline moistened towels and reflected anteriorly to expose the cruciate ligaments. The lateral collateral ligament was then isolated from surrounding tissues on the lateral side of the knee. The fibula (distal attachment of the LCL) was separated from the tibia and the ligament wrapped as described above. The entire joint capsule and fat pad were then carefully removed so as not to disturb the attachments of the cruciate ligaments. All remaining tissues were then discarded.

Preparation of bone blocks was then performed. The femur was first split into three parts, each containing the proximal attachment of one of the three ligaments to be tested. Special care was taken when separating the cruciate ligaments by starting the bone block cuts distally (inside the joint) and moving proximally up the femur. The tibia was then split into three blocks, one for the patellar tendon and one for each of the cruciate ligaments. Examples of these four tissue types are shown in Fig. 1 prior to final splitting of the tibia. All preparations were then wrapped in saline moistened towelling, and except for the tendon or ligament that was to undergo bundle dissection, stored at 4°C for up to 72 h.

Final dissection was then conducted using an Olympus stereomicroscope ($6\times$). Each specimen, consisting of several collagen fascicles, was prepared by first cutting along primary fiber bundle directions from one bony insertion to the other. While the distal bone block of each tissue was maintained, proximal bone segments were carefully cut for each of the tissue bundles to be tested. After grossly dividing the ligament or tendon into the total number of test specimens, each tissue was further dissected at magnifications ranging from 6 to $30\times$. The dissection was performed in a bath of 0.9 M saline at 4°C during which all damaged fascicles were removed. This permitted isolation of an undamaged core to be used as a test specimen. When dissection was completed, the length and cross-sectional area of each functional unit were determined. Specimen length was taken as the average of four bone-to-bone measurements using a vernier calipers accurate to 0.05 mm . The repeatability in length measurements was less than 1.2% of the average reading. The cross-sectional area was measured at four (ligament) or six (tendon) equally-spaced locations along the length of each specimen using an instrumented, constant pressure area micrometer (accurate to 0.01 mm^2) as previously described (Butler *et al.*, 1983b). Average specimen length and area for each tissue type are shown in Table 2. Specimen bone ends were then prepared. The entire distal bone end (tibia or fibula) was potted in bone cement (polymethylmethacrylate) in either a $1\frac{1}{4}\text{ in.}$ or $\frac{3}{4}\text{ in.}$ O.D. aluminum tube mold. Each of the proximal bone specimen blocks was potted in a similar way, using the upper grip as a mold. Examples of the finished specimen preparations are shown for all four tissue types in Fig. 2.

Table 2. Tissue bundle dimensions

Donor No.	PT	Initial length (mm)			Total	PT	Initial area (mm^2)		
		ACL	PCL	LCL			ACL	PCL	LCL
1 X	44.5	25.4	32.0	50.9	13	1.20	1.29	1.91	1.54
S.E.M.	(1.1)	(3.7)	(4.5)	(2.4)		(0.28)	(0.13)	(0.24)	(0.21)
n*	5	3	3	2					
2 X	57.8	31.7	31.4	52.9	15	1.86	1.17	1.12	2.98
S.E.M.	(1.4)	(2.4)	(0.3)	(1.9)		(0.31)	(0.39)	(0.45)	(0.07)
n	8	3	2	2					
3 X	55.6	29.1	29.5	48.7	14	1.65	1.91	2.40	3.73
S.E.M.	(2.4)	(3.1)	(2.7)	(0.6)		(0.21)	(0.09)	(0.24)	(0.23)
n	6	3	3	2					
					42				

*n = number of bundles per tissue from each donor.

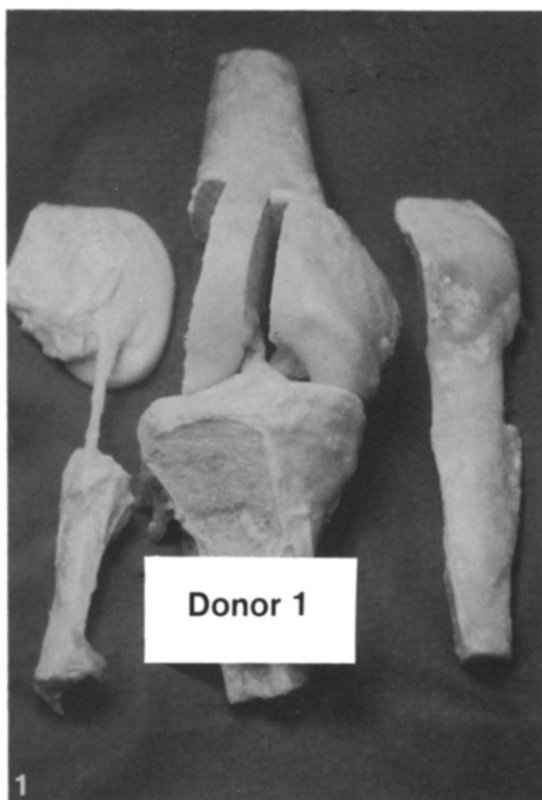
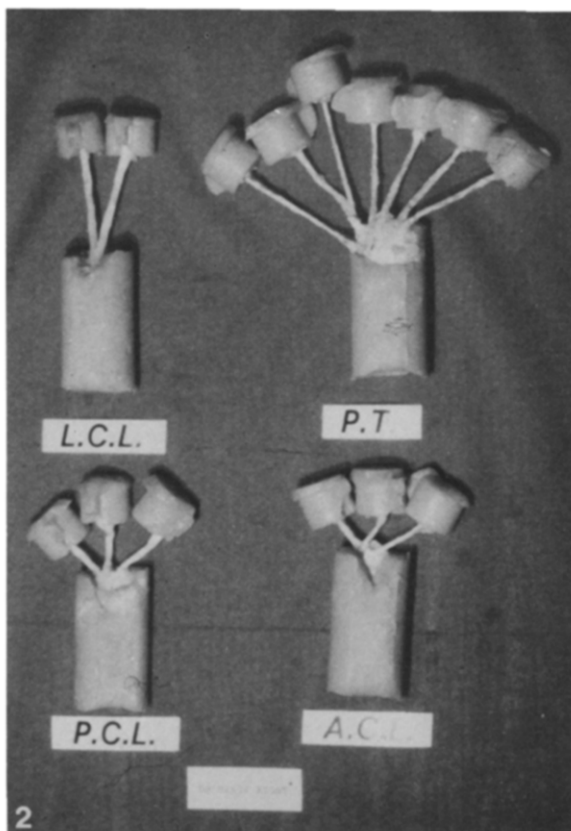


Fig. 1. Ligament and tendon-bone units prior to final bundle dissection. The femur-LCL-fibula preparation is at the left, the femur-cruciates-tibia unit in the center and the patella-pattellar tendon-tibia specimen at the right.

Fig. 2. Multifascicle-bone units from the LCL, PT, PCL and ACL. The bone ends have been potted in polymethylmethacrylate.

The entire tissue preparation was then mounted in a test chamber with plexiglass sidewalls and base, glass front and back sides and styrofoam insulation. The chamber then was filled with 16 l. of buffered 0.9 M saline (pH 7.2–7.4) kept at 37°C using a heating system (two 100 W heaters, temperature regulator, and temperature probe). The distal potted bone end was secured in the lower grip (fabricated from plexiglass and stainless steel) which could be rotated in fifteen increments off vertical to provide more anatomical bundle to attachment site orientations. The PT and LCL were tested with a bundle to grip axis angle of 15° and the ACL and PCL at a more horizontal angle of 30°. Finer improvements were achieved in anatomical alignment by bone block adjustments during potting. The load cell (Interface SM-50) and upper grip were suspended from a servo-controlled Aerotech X-Y table by a universal joint.

The failure tests were conducted using an Instron Model 1321 hydraulic materials testing machine at a strain rate of 100% s⁻¹ by scaling actuator stroke rate to the initial length between bone ends. The tests were driven by, and the data acquired and reduced by, a ModComp II lab minicomputer. Descriptions of the equipment and the data acquisition and reduction procedures can be found elsewhere (Butler *et al.*, 1980, 1983a).

Four material parameters (modulus, maximum stress, energy density to maximum stress and maximum strain) were computed for this study. Modulus was first determined from tissue stiffness. Stiffness was found using the computer by performing a linear regression between the beginning and endpoints of the linear portion of the load-elongation curve. These points were chosen from the experimental curve produced on the screen of a Tektronix 4010 Interactive Graphics Terminal. Modulus was then computed as the stiffness in this linear region times the ratio of the average initial length to average initial cross-sectional area. Maximum stress was determined as the maximum force produced by the specimen divided by the average initial area. Energy density to maximum stress was computed as the area under the stress-strain curve to maximum stress and maximum strain as the actuator displacement up to the point of maximum stress divided by initial length.

RESULTS

Material parameters

Shown in Fig. 3 are typical stress-strain curves for fascicle-bone units from the patellar tendon and three ligaments. All four tissues show distinct linear regions, smooth loading curves to maximum stress and abrupt declines after max stress. The PT specimen develops a much higher modulus and maximum stress than the three ligaments, which are quite similar in shape and magnitude to each other. All four tissues appear to achieve maximum stress at similar maximum strains, however.

Shown in Figs 4 to 7 are the four material parameters (mean $\pm$ S.E.M.), calculated from the measured force and stroke data. The number of test units from each donor also appears in Table 2.

The moduli shown in Fig. 4 (mean $\pm$ S.E.M.) for the three ligaments (LCL, PCL, ACL) are not significantly different from each other with the mean ranging from 278 to 447 MPa for all three donors. By contrast, the patellar tendons have a significantly larger average modulus with a mean range of 597–842 MPa, or 1.3 to

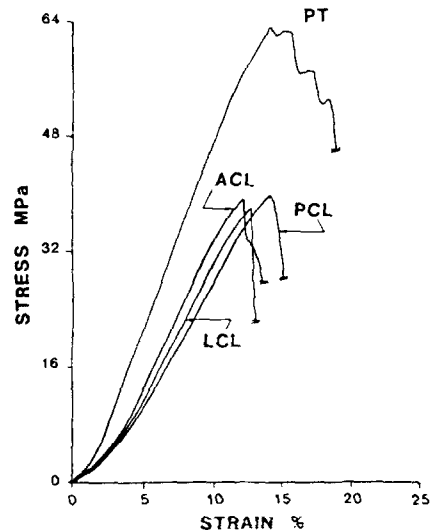


Fig. 3. Typical stress-strain curves for the PT, ACL, PCL and LCL fascicle-bone units. Note the much larger modulus and maximum stress for the PT specimen. The strains at maximum stress are similar for all four tissues, however. The portions of the curves beyond maximum stress are not shown for clarity.

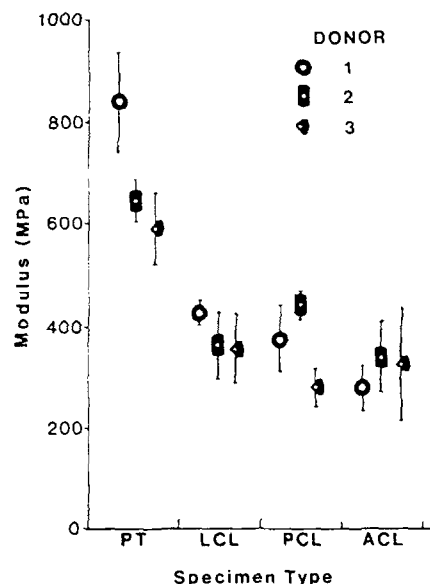


Fig. 4. Elastic modulus for each of the four specimen types from the three donors. Mean $\pm$ 1 S.E.M. are shown.

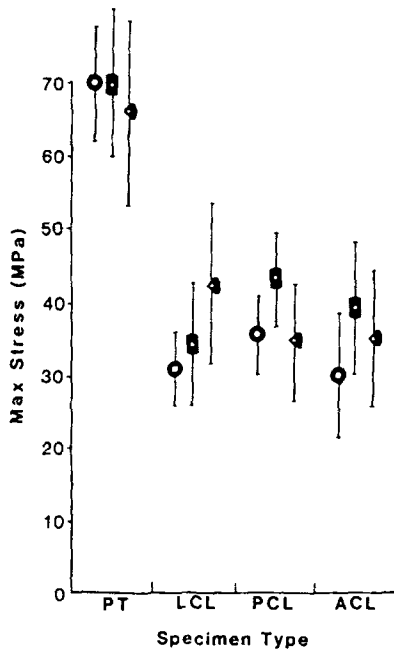


Fig. 5. Maximum stress ($\bar{X} \pm \text{S.E.M.}$) vs specimen type. Donor symbols are given in Fig. 4.

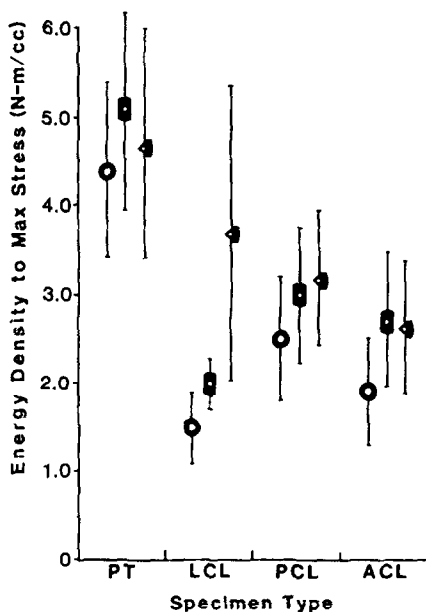


Fig. 6. Energy density to maximum stress ($\bar{X} \pm \text{S.E.M.}$) vs specimen type. See Fig. 4 for donor symbols.

3.0 times those of the ligaments ($p < 0.001$). Similar trends can be seen in Fig. 5 for the maximum stress, with the PT developing mean stresses of 65–70 MPa, 1.5 to 2.3 times those of the ligaments (30–44 MPa) ($p < 0.001$).

The energy densities at maximum stress (Fig. 6) show similar trends. Values for the ligaments ranged from 1.6 to 3.7 Nm cm^{-3} , significantly less ($p < 0.005$)

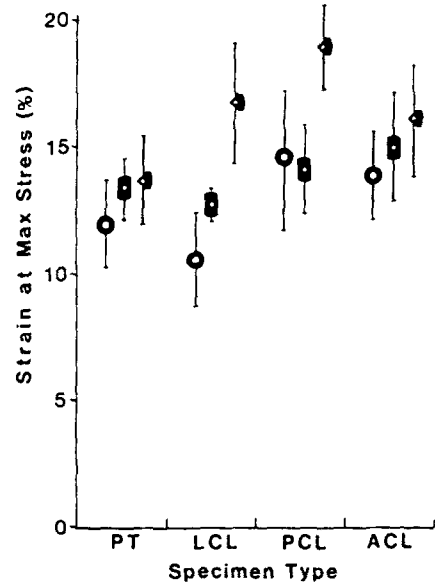


Fig. 7. Strain to maximum stress ($\bar{X} \pm \text{S.E.M.}$) vs specimen type. See Fig. 4 for donor symbols.

than the patellar tendon values (4.3–5.2 Nm cm^{-3}). By contrast the strains to maximum stress (Fig. 7) were similar for all tissues (11–19%), the tendon having values in the low end of the range. Strains from the 21 yr old donor were slightly larger for all four tissue types examined.

Among the ligament data, no significant donor-to-donor or tissue-to-tissue differences were found for any one of the material properties examined. Thus, the response for all ligaments were grouped and compared with tendon behavior. Shown in Table 3 is this comparison of cross donor means for nineteen patellar tendon and 23 ligament bundle preparations. Note that the modulus, maximum stress and energy density of the tendon bundles were significantly greater than those for all ligament bundle units. By contrast, no patellar tendon to ligament difference was found for the strain at maximum stress. It should also be noted that when inspected after the test, each specimen appeared to fail through soft tissue with no evidence of bone avulsion.

DISCUSSION

Before discussing the significance of the findings, it is important to cite a number of limitations in the study.

(1) The data presented are for tissues from donors only in the second and third decades. Properties are expected to differ when tissues from older or younger donors are tested (Noyes and Grood, 1976; Torp *et al.*, 1975; Viidik, 1979; Yamada, 1970). (2) Damage may have occurred during specimen preparation, although each bundle was carefully examined under the microscope before testing. (3) The lower grip/bone combination was used several times while testing bundles to

Table 3. Cross-donor means

	PT ($\bar{X} \pm \text{S.E.M.}$)	Ligaments (ACL, PCL, LCL) ($\bar{X} \pm \text{S.E.M.}$)	Significance*
Modulus (MPa)	643.1 (53.0)	345.0 (22.4)	$p < 0.001$
Maximum stress (MPa)	68.5 (6.0)	36.4 (2.5)	$p < 0.001$
Energy density to maximum stress (Nm cm^{-3})	5.03 (0.66)	2.53 (0.25)	$p < 0.005$
Strain at maximum stress (%)	13.5 (0.7) $n = 19$	15.0 (0.8) $n = 23$	—

*Statistical significance determined using the Welch Modification of the two tailed Student's *t*-test.

failure. This may have weakened the distal ends of the bundles tested later to some extent. (4) The rather large variations in cross-sectional area (16–21 % for the PT, ACL and PCL) were probably caused by several factors, including intra-donor bundle-to-bundle area differences (14.0–17.9 %), position-dependent differences along each bundle (6.8–8.1 %) and lack of area micrometer repeatability (2.0 %). LCL area variations were only 7.6 %. (5) Testing was performed in 0.9 M saline which may have produced swelling of the preparation over time. (6) Grip-to-grip strains are reported rather than either bone-to-bone or local strains. (7) The number of bundles in each tissue was small preventing a statistical determination of material property variation across the ligament or tendon. (8) Specimen bone ends may not have always been positioned properly in methacrylate to produce uniform loading at the insertion sites. (9) The results for the specific tissues tested probably do not represent the material properties of other tendons or ligaments, even from the same donor.

Within these qualifications, there are a number of important aspects to this study:

(1) Recently, histological and biochemical comparisons have been made among ligaments and tendons from the same animal (Amiel *et al.*, 1984). To our knowledge, however, the data presented in our study provide for the first time material properties of human knee ligaments and patellar tendon in the same knee. This approach has the advantage of eliminating the effects of biological variability between donors caused by differences in age, sex, activity levels, etc.

(2) During mechanical testing, fascicle-bone units have certain advantages compared to whole tissue-bone preparations. Fascicle initial length and cross-sectional area can be more accurately determined due to the smaller, more easily defined insertion sites. Fiber bundles within these specimens tend to be more parallel than in whole tissues (e.g. the anterior cruciate ligament) and the orientation of the fibers into bone is expected to be more uniform. These two factors should result in material properties that are more characteristic of collagen than can be determined from whole tissues. Gripping bone at both ends of the fascicle also eliminates potential damage which can occur when

fascicles or whole tissues are gripped directly.

(3) The data suggests that similar load-related fascicle properties can be employed in mathematical models for the cruciate ligaments and lateral collateral ligament (i.e. for a 90 % confidence interval, a modulus of 300–390 MPa, a maximum stress of 30–40 MPa and an energy density to maximum stress of 2–3 Nm cm^{-3} are reasonable). This data for the PCL and LCL are also, to our knowledge, the first human material property values reported for these ligaments.

(4) The patellar tendon fascicle results are pertinent to the current problem of autograft replacement of the cruciate ligaments. The authors have already reported on the structural and material properties of nine frequently used autograft tissues (Noyes *et al.*, 1984; Butler *et al.*, 1984). These tests were performed using 90 tissues from 20 young human donors. However this data had to be compared to ACL results from an earlier study (Noyes and Grood, 1976). As a result of the current work, it is now possible to state that the patellar tendon fascicle develops almost twice the modulus, maximum stress and energy density of the cruciate ligament bundles (Table 3). Given the percent reduction which occurs in patellar tendon properties after replacement of the anterior cruciate ligament (Butler *et al.*, 1983b), a desired size of patellar tendon graft could be estimated.

(5) It now appears that bundles from all three ligaments tested as well as the patellar tendon fail by a strain-dependent mechanism. The tissues achieve maximum stress in about the same maximum strain range (13–15 %). These results differ from earlier reports (e.g. Noyes and Grood, 1976) which found ligaments to reach maximum stress at much higher levels of 30–45 % strain. The findings are also contrary to the common belief that tendons fail at much smaller strains than ligaments. For example, it has repeatedly been shown that the cruciate ligaments fail by a serial failure mechanism where some bundles are loaded and fail early while others must first become aligned to the loading axis before developing large loads. By testing isolated cruciate ligament bundles we have reduced the chances for nonuniform loading which tend to spread out the loading curve along the strain axis. The patellar tendon does not demonstrate this broad failure curve

due to the more parallel nature of its fiber bundles. Second, many tendons including ones we've evaluated (Noyes *et al.*, 1984; Butler *et al.*, 1984) have been tested as isolated tissues without bone ends. Tested in this way, the tendons may have failed prematurely giving artificially low maximum strains or may have slipped from the grips, resulting in much higher strains. The current study avoided this problem by the careful selection of tissue types.

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