Strain Inhomogeneity in the Anterior Cruciate Ligament Under Application of External and Muscular Loads

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To determine whether mathematical relations between strains in different bundles and loads would be needed to predict injury of the anterior cruciate ligament (ACL), this work tested the hypothesis that strains developed in two bundles of the ACL were significantly different under the application of a number of loads important to injury etiology of the ACL. To provide the data for testing this hypothesis, liquid mercury strain gages were installed on both the anteromedial (AMB) and posterolateral (PLB) bundles of the ACL of ten specimens, which were then subjected to passive flexion/extension, hyperextension moment, anterior force, internal and external axial moments, quadriceps, and hamstrings forces. Various combinations of these loads were also applied. Flexion angles ranged from 8 deg of hyperextension through 120 deg of flexion. The data were analyzed using a repeated measures analysis of variance. The analyses indicated that significant strain differences existed between the two bundles only for passive flexion/extension. However, the analyses did not support the hypothesis that AMB and PLB strains are significantly different from each other under the application of external and muscular loads. Because noticeable differences (>3 percent) between bundle strains did exist in some load cases for limited ranges of flexion and the PLB strain was consistently higher than the AMB strain, it may be sufficient to consider strain in only the PLB when predicting ligament damage based on strain-load relations.

Introduction

Based on studies that have shown that the anteromedial (AMB) and posterolateral (PLB) bundles reciprocate in function over the range of flexion/extension (e.g., Amis and Dawkins, 1991; Kurosawa et al., 1991; Hollis et al., 1991), it is recognized that the ACL does not behave as a homogeneous structure. This inhomogeneity is potentially important to the study of injury mechanics since ligament failure is governed by the degree of strain which the tissue experiences (e.g., Woo, 1982) and failure may begin in the more highly strained bundle.

A substantial body of previous research has measured strain in the AMB under all of the external and muscular loads implicated in etiology of ACL injury. The effects of hyperextension (e.g., Kennedy et al., 1977), anterior force (e.g., Berns et al., 1992), axial moment (e.g., Pope et al., 1990) and muscle forces (e.g., Renstrom et al., 1986; Draganich and Vahey, 1990) have all been studied. Because strain was measured in the AMB only, however, these studies did not provide information regarding the strain inhomogeneity between bundles.

Other research has examined the effects of various loads on the elongation (i.e., length changes) of both the AMB and PLB. Length changes under axial moments (Amis and Dawkins, 1991), A/P force (Hollis et al., 1991), and quadriceps force (Kurosawa et al., 1991) have all been measured. Inasmuch as absolute strain is a quantity more relevant to ligament injury than length change (e.g., Woo, 1982) and all of these studies measured length changes that cannot be converted to absolute strain, these studies were not directly relevant to injury mechanics. Another limitation of these studies was that nonphysiologic load levels were applied.

Because of the limitations of previous research, the general goal of this research was to advance the study of ACL injury mechanics by quantifying the strain inhomogeneity between the two bundles. To reach this goal, the specific objective was to test experimentally the hypothesis that AMB and PLB strains are significantly different from each other under the application of various external and muscular loads important to ACL injury etiology.

Methods

To test this hypothesis, surface strains of both the AMB and PLB of 10 unembalmed cadaver knee specimens (mean age 49.8 ± 14.7 years) were measured under the application of various loads. These specimens were installed and aligned in a six-degree-of-freedom load application system (LAS) (Bach and Hull, 1995). This apparatus was pneumatically actuated and under full closed-loop control. Each degree of freedom was individually actuated and instrumented for both load and displacement measurement. Also, actuators were included to simulate the effect of quadriceps and hamstrings contractions through force application to the respective tendons. The functional flexion/extension and internal/external axial rotation axes of the specimen were aligned with those of the apparatus using a functional alignment procedure (Bach and Hull, 1995; Berns et al., 1992).

After installation and alignment each specimen was subjected to a consistent preconditioning protocol. Immediately following preconditioning the 0 degree flexion angle was defined by applying a 2.5 Nm extension moment. The specimen was then removed from the apparatus and the knee dissected to allow installation of liquid metal strain gages (LMSG) on both the AMB and PLB.

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Using the fabrication techniques of Meglan et al. (1988), gages approximately 0.625 mm in diameter and about 5 mm in length were constructed. LMSGs constructed using these techniques were extremely compliant (stiffness = 0.024 N/mm), allowing them to be mounted directly to the ligament without affecting its properties (Brown et al., 1986). Prior to implantation each gage was calibrated to approximately 30 percent strain using a micrometer index table with graduations of 0.02 mm.

Gages were attached to the most superficial fibers of the ligament via sutures. Each gage was installed with a pre-stretch of 0.5–1.0 mm beyond the slack length of the ligament. The pre-stretch of the LMSG ensured that the gage did not slacken completely before the ligament fibers did. Using the stiffness of the LMSG reported by Meglan et al. (1988), the pre-stretch generated a pre-load in the ligament fibers of approximately 0.024 N.

The anteromedial bundle gage was installed via access gained by reflecting the patella using bilateral parapatellar incisions. This gage was installed on, and aligned with, the anterior-most fibers of the anteromedial bundle (Fig. 1). The proximal end of the gage was located 8–10 mm distal to the femoral insertion of the ligament. Following attachment of both ends of the LMSG, the knee was flexed and extended over the full range of motion while monitoring the gage to ensure that the installation was acceptable. If either the gage appeared to slacken before the AMB did, or there was some other problem with the installation, then the LMSG was removed and reinstalled.

The PLB gage was installed on the most posterior fibers of the posterolateral bundle. Fibers that could be followed from their origin proximally to the superior border of the posterior cruciate ligament (PCL) distally were selected (Fig. 2). Access to the PLB was gained through a small (<1 cm) incision in the posterior capsule. The knee was flexed to approximately 90 deg to facilitate visualization and gage installation. The LMSG was introduced into the joint space and the distal end of the gage secured to the desired fibers proximal to the superior border of the PCL. Once the gage body had been securely attached to the ligament, the installation was checked similarly to the AMB gage. After both gages had been installed, the knee was closed and the specimen reinstalled in the LAS. Described in greater detail in Bach et al. (1997), the procedures used to install the gages did not affect the load-displacement relations of the joint.

For injury prediction, it was necessary to measure absolute strain. The reference lengths of the LMSG's were the lengths at 90 deg flexion for the AMB and 15 deg flexion for the PLB when the joint was unloaded. When the joint either was flexed passively from the angles used to determine the reference lengths or was loaded, the strains were computed from the length change normalized to the reference length. Any negative strains were set to zero.

Fig. 1 Diagram depicting the placement of a LMSG sutured to the AMB of a right knee viewed from the front at full extension

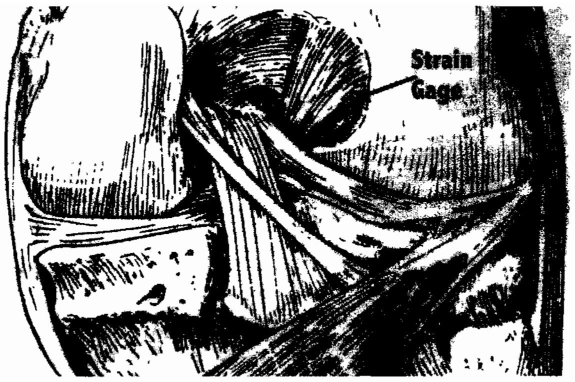

Fig. 2 Diagram depicting the placement of a LMSG sutured to the PLB of a right knee viewed from the back at full extension

Following LMSG installation each knee was forced into hyperextension through application of a 20 Nm extension moment. The specimens were then flexed to 125 deg and returned to full extension. Knee flexion/extension (F/E) moment, joint displacements, and ligament strains in both bundles were recorded continuously during the motion. These measurements were made on ten specimens.

After the passive F/E cycles, the ten specimens were subjected to a detailed loading protocol to examine the effects of externally applied (anterior, internal/external) and muscular (quadriceps, hamstrings) loads, and combinations thereof, on the strains in the ligament. Loads were applied in two to four steps up to their maximum values (Table 1). Flexion angles were randomized as was the order of load application within each flexion angle. Axial moments were not applied at 0 deg due to premature failure of two specimens during internal axial moment loading at this angle. Due to various technical difficulties (e.g., muscle attachment failure), full data sets were recorded for a subset of the remaining eight specimens. To maximize the number of specimens used in the statistical analyses to be described shortly, subsets of these eight specimens were selected for each analysis such that each specimen in the subset provided a full data set for a particular analysis.

Quadriceps forces were applied in combination with anterior force and internal/external (I/E) moment. Quadriceps plus hamstrings loads were also applied along with anterior force. The quadriceps and hamstrings loads were always applied so

Table 1 Loads and load combinations (maxima)

Flexion Angles (deg.)	Forces or Moments	Levels (N, Nm)
0	Anterior	300
	Quadriceps	1000
	Quads +Hams	750+?
	Anterior + Quads	300+1000
	Anterior + Quads + Hams	300+250+?
15,45	Anterior	300
	Internal	10
	External	10
	Quadriceps	1000
	Quads + Hams	750+?
	Internal + Quads	10+750
	External + Quads	10+750
30, 60, 90, 120	Anterior	300
	Internal	10
	External	10
	Quadriceps	1000
	Quads + Hams	750+?
	Anterior + Quads	300+1000
	Internal + Quads	10+750
	External + Quads	10+750
	Anterior + Quads + Hams	300+250+?

as to maintain flexion/extension equilibrium of the joint. The ratio of loads to be applied was determined for each of the flexion angles tested. For these combined loads, the following procedure was followed: muscle forces (quadriceps or quadriceps + hamstrings) were applied to the desired level and maintained. For the quadriceps + hamstrings loading conditions the two muscle loads were increased simultaneously. The external load was then applied to its desired level and returned to zero. The muscle forces were increased to the next level and the external load reapplied and removed. This was repeated until the maximum muscle loads had been tested. Finally the muscle forces were returned to zero.

To ensure that there was no progressive damage to the joint occurring during the various loading steps, a repeatability evaluation was included. Immediately after each flexion angle change, a set of anterior–posterior (± 200 N) and varus–valgus (± 10 Nm) loads was applied (repeatability loads). The desired loading sequence for that flexion angle was then applied. After this loading cycle was completed, but before the flexion angle was changed to the next position, the repeatability loads were reapplied. Following the experiment, the data from these repeatability loading cycles (corresponding to one flexion angle) were compared to ensure that the protocol loading cycles did not change the mechanics of the joint. If it was apparent from this comparison that some damage to the structures of the joint had occurred, then results from that loading cycle, and all subsequent ones, were deleted from further analysis.

To test the hypothesis that the strains in the AMB and PLB were significantly different under the application of external and muscular loading, the data from these experiments were subjected to several repeated measures analysis of variance (RANOVA) procedures. The first analysis included hyperextension and passive F/E rotations only. For this passive flexion/extension analysis, the data included two within-subject effects, flexion angle and bundle, as well as the interaction between the two. Although strain data were taken continuously during the passive trial, the flexion angle effect was tested at 11 levels (Fig. 3). The data from all ten specimens were used in this analysis.

A second RANOVA examined the effect of muscle loads only on the bundle strains. The analysis included three within-subjects effects (flexion angle, bundle, and load) and all four possible interactions of the three within-subject effects. Data input to this analysis were obtained from five specimens.

The third and fourth RANOVA's tested the effects of anterior force applied both with and without muscle forces. The analysis for anterior force alone included three within-subjects effects (flexion angle, bundle, and anterior force) and all possible interactions of the three within-subject effects. The analysis for combined anterior and muscle forces included the additional effect

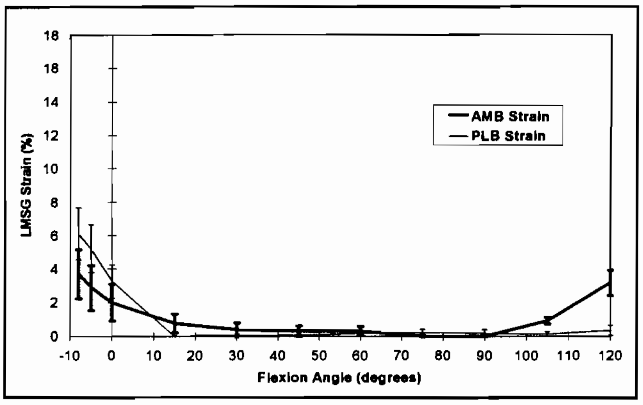

Fig. 3 Mean ACL strains and standard errors versus flexion angle for passive flexion and hyperextension

of muscle force and all possible interactions of the four within-subject effects. Data derived from five and four specimens were used in the analyses for anterior force alone and the combined forces, respectively.

The final two RANOVAs examined the effects of both internal and external axial moments applied both with and without quadriceps force. For these analyses four within-subjects effects (flexion angle, bundle, moment, and quadriceps force) were modeled and all possible interactions were again considered. Data from five and four specimens were used for the internal and external analyses respectively.

Following the RANOVA analyses, the results were subjected to Tukey's Method of Multiple Comparisons procedures (e.g., Neter et al., 1990) to determine significant differences between the levels of the significant factors or interactions. The level of significance was set to 0.05 in the multiple comparison tests.

To determine the probability of a Type II error, the power of the hypothesis tests for the bundle effect was analyzed following the procedures outlined by Neter et al. (1990). A separate but similar analysis was performed for each RANOVA. For all analyses the power was determined by specifying a significance level of 0.05 and a sample size equal to four. The most conservative value of one was used for the ratio of D/s where D is the minimum difference in strain between the two bundles and s is the standard deviation of the RANOVA model. For all analyses the power was greater than 0.95. The high power occurred because the number of strain values measured for each specimen was large. For example in the RANOVA to test for the muscle effects, the number of strain values for each specimen was 48. Hence, it was very unlikely that any one of the null hypotheses indicating no significant differences between bundle strains would be accepted when, in actuality, it should be rejected.

Results

Passive Flexion/Extension. The strains in the AMB and PLB demonstrated reciprocating behavior during passive F/E motion (Fig. 3). As the knee was flexed from hyperextension to approximately 30 deg, the strain in both bundles decreased to zero. As flexion continued beyond 90 deg, the AMB strain increased while the PLB strain remained at zero.

From the results of the first RANOVA, the flexion angle effect was significant ($p = 0.046$) while the bundle effect was not ($p = 0.824$). The interaction between these effects (flexion angle * bundle) was also significant ($p = 0.044$). Because the interaction was significant, the follow-up multiple comparison analysis was necessary to investigate differences between bundles over the range of flexion angles.

The results of the Tukey's Method of Multiple Comparisons procedure indicated that the strain in the PLB was significantly different ($\alpha = 0.05$) from the strain in the AMB only at a flexion angle of 120 deg. The differences at the remaining flexion angles were not statistically significant at the $\alpha = 0.05$ level. For the AMB the strains at -8 and 120 deg were significantly different from the strains at 30, 45, 60, 75, and 90 deg but not from each other. Considering the PLB, the strains at -8, -5, and 0 deg were significantly different from the strains at the remaining flexion angles.

Isolated Muscle Loading. The application of quadriceps force caused strain in both bundles to increase above the passive levels in the flexion range from extension to about 60 deg of flexion with the average strains being comparable for the PLB and the AMB except at 15 deg where the PLB strain was noticeably (>3 percent) larger (Fig. 4). The simultaneous application of the hamstrings force caused a reduction in strain relative to the quadriceps force applied alone.

Although a noticeable difference in strains between the two bundles was apparent qualitatively at the single flexion angle

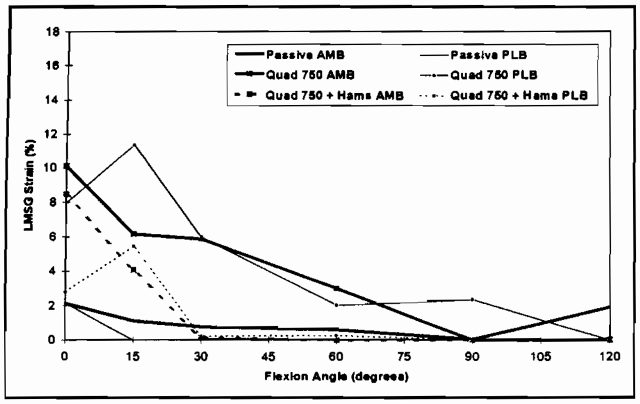

Fig. 4 Mean ACL strains versus flexion angle for muscle forces

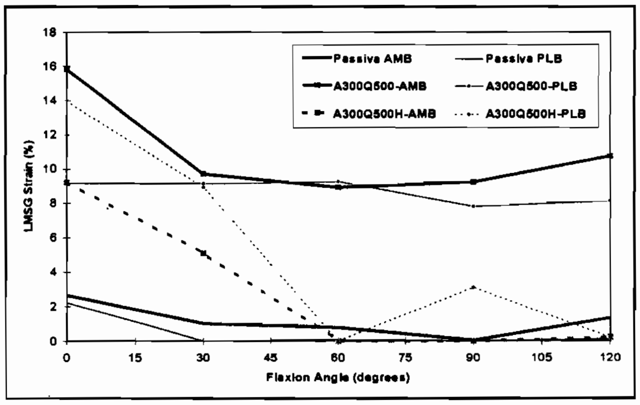

Fig. 6 Mean ACL strains versus flexion angle for anterior force combined with muscle forces

noted above, the RANOVA results, which were based on analysis of the strain differences for all treatments, indicated that the bundle effect was not significant ($p = 0.440$). In contrast to the passive F/E results, neither was the interaction between flexion angle and bundle significant ($p = 0.575$) nor were the other interaction terms, which included the bundle ($p > 0.758$). Only the flexion angle * load two-way interaction was statistically significant ($p = 0.001$).

Anterior Loading. An isolated anterior force resulted in similar strains for both the AMB and PLB, except at 30 deg where the PLB strain was noticeably greater (Fig. 5). The anterior + quadriceps loading (Fig. 6) also showed similar strains between the two bundles except at 0 deg flexion where the AMB strain was noticeably greater. When the hamstrings force was added, the strains in both bundles decreased approximately linearly from 0 deg to 60 deg where the strains reached passive levels.

The repeated measures results for the isolated anterior force indicated that although anterior force caused strain to increase significantly ($p = 0.003$), neither was the bundle effect significant ($p = 0.957$) nor were any of the bundle interaction terms ($p > 0.352$). When a quadriceps force was applied in combination with the anterior force, both anterior load ($p = 0.016$) and muscle ($p < 0.001$) effects were significant as were the flexion angle * muscle load ($p = 0.032$) and anterior load * muscle load ($p = 0.012$) interactions. However, neither the bundle effect ($p = 0.848$) nor any of the bundle interaction terms ($p > 0.299$) were significant. All other interactions were not significant ($p > 0.299$).

Internal/External Loading. Application of an internal axial moment caused strains in both bundles to increase above the passive levels, particularly for small flexion angles with the strain being comparable in both the PLB and the AMB (Fig. 7). When a quadriceps force was applied in combination, the strain in both bundles increased relative to the moment applied individually but remained at comparable levels between the two bundles.

From the RANOVA results for internal axial loading, the bundle effect was not significant ($p = 0.508$) nor were any of the interactions involving the bundle ($p > 0.342$). However, all of the other factors were statistically significant ($p < 0.035$) but none of the interactions between these factors was significant ($p > 0.136$).

In contrast to the effect of an internal axial moment, an external axial moment did not cause any strain increase in either bundle (Fig. 8). When a quadriceps load was applied in combination, however, the strain in both bundles increased above the passive values in the flexion range of 0–60 deg with a noticeably greater increase for the PLB than the AMB at both 15 and 30 deg.

Notwithstanding the noticeable differences apparent between the bundle strains at two flexion angles, the RANOVA results for external axial loading indicated that the bundle effect was not significant ($p = 0.565$) nor were any of the interactions involving the bundle ($p > 0.379$). All other factors were significant ($p < 0.029$) as were all of the interactions between these factors ($p < 0.023$).

Discussion

An important question to answer in the study of ACL injury mechanics is whether significant strain inhomogeneity exists between the two bundles of the ACL when the knee joint is

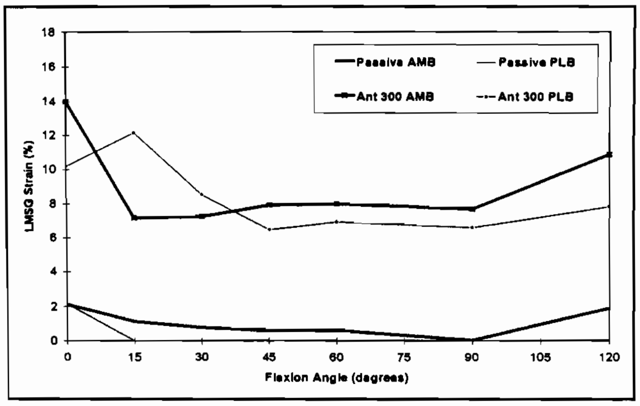

Fig. 5 Mean ACL strains versus flexion angle for anterior force

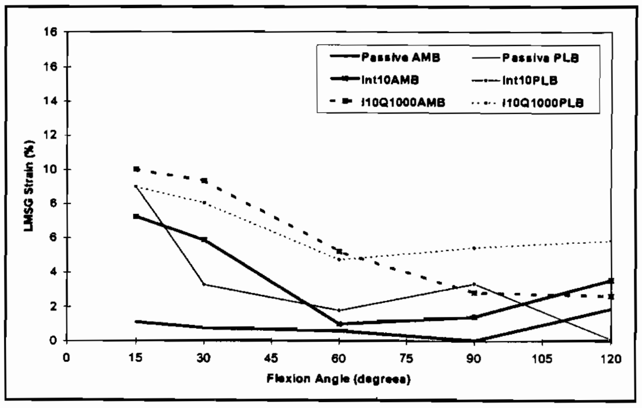

Fig. 7 Mean ACL strains versus flexion angle for internal axial moment both with and without quadriceps force

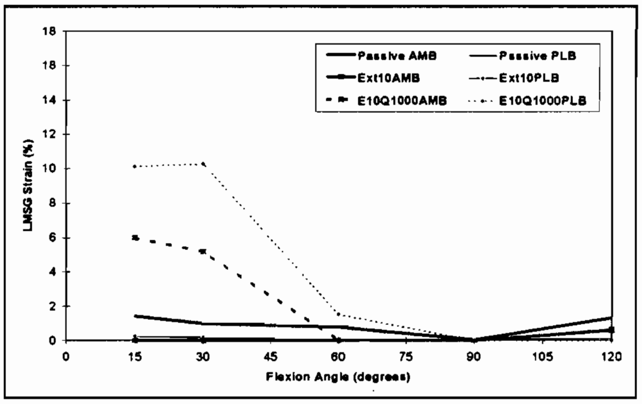

Fig. 8 Mean ACL strains versus flexion angle for external axial moment both with and without quadriceps force

loaded. One reason that this question is important is because of its relevance to reducing the incidence of knee injuries in Alpine skiing. Current mechanical ski bindings do not adequately protect against ACL injuries even when they are adjusted according to widely adopted standards (e.g., Johnson and Pope, 1991; Fischer et al., 1994). To realize a binding that reduces knee injuries, one approach is to base the release–retention decision on ligament strains estimated in real time from a mathematical model that relates the various injury governing variables to ligament strains. Injury governing variables include loads transmitted to the knee joint from ground reactions, loads developed by muscles crossing the joint, the flexion angle, and the loading rate. A programmable ski binding (e.g., Lieu and Mote, 1980; Wunderly et al., 1988) could both solve the model equations based upon measurements of the injury governing variables and release the boot from the ski if the estimated strains approached injury levels. The ability to represent the ACL as a single structure rather than as multiple bundles would simplify the experiments needed to gather the information with which to develop the mathematical model.

To answer the question posed above, the current research project was undertaken to determine the variation in strain between two bundles of the ACL under the application of a number of loads implicated in injury etiology of the ACL during Alpine skiing. These included individual loads such as anterior force (e.g., Bally et al., 1989), quadriceps force (e.g., McConkey, 1986; Paletta and Warren, 1994), hyperextensive moment (e.g., Shino et al., 1987; Paletta and Warren, 1994), and internal and external (I/E) axial moments (e.g., Ettlinger, 1989; Fischer et al., 1994). Also included were combinations of loads consisting of quadriceps force applied simultaneously with anterior force (Read and Herzog, 1992), axial moments, and hamstring forces (Maxwell and Hull, 1989).

One load that was not included in the tests was compression. Although compressive loads would simulate the weight-bearing common during some ACL injuries in Alpine skiing, weight-bearing is not common to all ACL injuries. For example, in situations where the skier catches an edge and the ski externally rotates, the leg is almost completely unweighted. In those situations where the injured leg is weight-bearing, compressive loads would not be expected to contribute to an ACL injury but rather to reduce the strain due to other external loads. Accordingly excluding compression from our testing protocol resulted in worse case strains than if the compression was included.

Because the data used to answer the question of interest to this study were gathered experimentally, it is useful to examine the measurement technique critically before discussing the results. To determine absolute values of strain, the reference length of the gage must be specified as the length that the gage assumes when the ligament is at zero strain. Different methods exist for making this specification. One method is to set the reference length at the inflection point in the load-gage output voltage curve when the joint load is reversed from anterior to posterior force. For length transducers that move freely as the length changes, this method has been shown to provide a reference length that coincides with the onset of zero tension in the ligament fibers (Fleming et al., 1994). However, for length transducers such as LMSG's, which have measurable stiffness (Brown et al., 1986) and which are installed with sufficient pre-stretch so that the gages do not become lax at the onset of ligament fiber unloading, the question arises as to whether the inflection point represents the lax length of the ligament fiber or the lax length of the gage, which may not be the same as the lax length of the fiber. Although it may be argued that the inflection point is in fact that of the ligament fiber owing to the low gage stiffness of about 0.024 N-mm^{-1} (Brown et al., 1986), the method nevertheless introduces uncertainty into the reference length because the inflection point is not clearly defined (Berns et al., 1992).

To avoid the uncertainty mentioned above, an alternative method was used in this study where the reference length was set at flexion angles of 15 and 90 deg for the PLB and AMB, respectively. This method is founded on the measurements of ACL tension during passive motion by Wascher et al. (1993). These authors measured tension in 18 specimens and demonstrated that the ACL tension declined abruptly as the knee was flexed from full extension and consistently reached negligible levels (<5 N) at 15 deg of flexion. Similarly as the knee was extended from full flexion, the tension decreased and consistently reached negligible levels at a flexion angle of 90 deg. In the range between the two angles of 15 and 90 deg, the tension remained at 5 N or less. Inasmuch as the bundles of the ACL reciprocate with the PCL carrying a greater portion of the ACL tension at extension and the AMB carrying a greater portion in flexion (Amis and Dawkins, 1991), a decrease in elongation of the posterior fibers to the reference length is associated with the transition to low tension in flexion, while a decrease in elongation of the anterior fibers to the reference length is associated with the transition to low tension in extension. The small variability in the transition angles between the 18 specimens tested allows the method to be applied confidently to arbitrary specimens.

Regardless of the transducer used, a concern with the measurement of surface strain in the AMB was the potential impingement of the strain gage between the ligament and the intercondylar roof as the knee reached full extension. Other authors have taken steps to prevent impingement of AMB strain gages such as shielding the gage in a rigid hypodermic needle (e.g., Berns et al., 1992) or performing either a notchplasty or roofplasty (e.g., Meglan et al., 1990; Durselen et al., 1995). Any of these techniques would have altered the biomechanics of the joint. In the present study, the decision was made to ensure that the mechanics of the joint were not altered by the measurement technique. Because steps to prevent impingement were not taken and this likely occurred, it is a probable source of error in those load cases where either the knee was hyperextended or the tibia was loaded such that it displaced anteriorly at full extension. Because the AMB strain values at 0 deg were possibly affected by impingement, the statistical analyses were rerun, omitting the data from this flexion angle. Although leaving out these data caused some minor shifts in the p-values, none of these shifts was sufficiently large to cause any change in the conclusions regarding the various hypotheses tested by the analyses.

Since the ACL and the PCL crossed and made contact during knee motion, the potential for PLB gage impingement existed as well. To prevent this from occurring, the PLB gage was installed superior to the crossing point of the two ligaments (Fig. 2). With the knee flexed approximately 90 deg, there was a space of roughly 5 mm between the distal gage attachment and the superior border of the PCL. Flexing the knee an additional 30 deg (to 120 deg) as was done in our experiments was insufficient to result in impingement of the gage by the PCL. Each knee was visualized after gage installation to ensure that the transducer was not impinged during motion.

There was also the possibility that the PLB gage might be impinged between the ligament and the intercondylar notch in a flexed knee undergoing axial rotation of the tibia. Again, the gage installation was verified immediately after installation and no such impingement was seen under moderate torsion with the joint flexed to 90 deg.

As with any measurement of strain, the value measured represented an average over the length of the transducer; thus the true local strain was not known. This effect was minimized by using as small a transducer as possible. In this application gages only 4–5 mm long were used as opposed to other studies of the AMB that used gages greater than 20 mm in length (e.g., Berns et al., 1992). The use of these small gages allowed the measured strain to approximate the true local strain in the fibers.

Another factor concerning gage length was the three dimensional nature of the ligament bundles. Since the LMSG was sutured only at its ends, it might have produced a reading for the straight line strain between two points, whereas the ligament likely curved in three dimensions. If the LMSG did not follow this curve, then it would underestimate the true strain. Again the use of such small gages served to minimize these errors.

The results of this study did not support the hypothesis that AMB and PLB strains are significantly different from each other under the application of external and muscular loads. The statistical analyses of the data from the application of quadriceps forces, hamstrings forces, anterior forces, axial load moments, and their combinations did not reveal significant differences in the strains measured in the AMB and the PLB. Neither was the bundle term significant in any of these analyses nor were any of the interaction terms involving the bundle.

This result was unexpected, particularly for anterior and quadriceps loads based on the reciprocating behavior of the bundles during passive motion over the flexion/extension arc. The reciprocating phenomenon was originally identified from studies that measured differences in bundle elongation during passive flexion/extension (e.g., Amis and Dawkins, 1991; Kurosawa et al., 1991). Heretofore, it could only be inferred that the AMB strain would be greater than the PLB strain at full flexion and the inequality reversed at full extension. This was because no previous study known to the authors directly measured strains in both bundles simultaneously. The present study is the first to make such measurements and confirm that the bundle strains are indeed different (Fig. 3). Given that the bundles' lengths are markedly different (e.g., Hollis et al., 1991) with the AMB being roughly twice as long as the PLB and that the origin/insertion sites are different (e.g., Amis and Dawkins, 1991), it was reasonable to expect that the strain differences would be maintained as the joint was subjected to either anterior or quadriceps force, which would cause predominantly an anterior displacement of the tibia on the femur.

The equality of the bundle strains is most probably a consequence of coupled motions. Coupled motions as a result of quadriceps load are not well documented. However, the primary coupled motion under anterior loading is axial rotation (Hollis et al., 1991). Although coupled motions would generally be expected due to changing articular surface geometry as the joint is displaced, the equality of strains suggests that coupled motions also occur so as to self-equalize bundle strains.

The implication of the result that bundle strains are equal is that a model relating these loads to ligament failure could consider the strain in one bundle of the ACL as being representative of the state of strain throughout the ligament. This conclusion must be regarded with caution, however, for two reasons. Although the statistical analyses did not find significant differences between the bundles when all flexion angles were considered, one reason is that qualitative graphic analysis of the data indicated that noticeable differences (>3 percent) did exist in some cases for limited ranges of flexion. The 3 percent value defining a noticeable strain difference was equal to one standard deviation away from the average strain for a particular bundle. When noticeable differences were apparent, the PLB strain was consistently higher than the AMB strain however. The only exception to this was the noticeably greater strain in the AMB at 0 deg flexion, which occurred under simultaneous application of anterior and quadriceps forces (Fig. 6). Inasmuch as this load combination would cause the greatest anterior displacement of the tibia on the femur of all of the loads applied, this result was most likely a measurement artifact due to impingement of the AMB gage on the intercondyler roof. The finding that the PLB strain was consistently higher than the AMB strain suggests that although it might not be necessary to consider two bundles in an injury model, the model should be based upon the PLB. An assumption implicit to this suggestion is that either the failure strains for the two bundles are equal or the PLB fails at a lower strain than the AMB. Measurements of the failure strains for different ACL bundles confirm the validity of this assumption (Butler et al., 1992).

The second reason concerns the proximity of the applied loads to the failure loads. Although the load levels were physiologic, their proximity to the failure threshold is unknown. Because strain-load relations cannot be extrapolated with confidence (Berns et al., 1993), it is possible that relative strains between the AMB and PLB would change at higher loads.

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