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Skiing Trauma and Safety: Twelfth Volume

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Foreword

This publication, *Skiing Trauma and Safety: Twelfth Volume,* contains papers presented at The Twelfth International Symposium on Skiing Trauma and Safety held in Whistler Blackcomb, British Columbia, Canada, on 4–10 May 1997. The symposium was sponsored by ASTM committee F27 on Snow Skiing and the International Society of Skiing Safety (ISSS). Robert J. Johnson, University of Vermont, was editor of this publication.

J. M. Bach[1] *and M. L. Hull*[2]

Strain of the Anterior Cruciate Ligament Increases Linearly With Quadriceps Contraction

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ABSTRACT: This work was designed to answer two important questions concerning injury mechanics of the anterior cruciate ligament (ACL). One question was whether strain of the ACL increased linearly with increasing quadriceps contraction, and the other was whether strains in the anteromedial (AMB) and posterolateral (PLB) bundles were different at any point in the flexion arch.

To answer these questions experiments were performed to determine the relationship between increasing quadriceps force and strains in both the AMB and PLB. Five unembalmed knee specimens were instrumented on both bundles of the ACL using liquid mercury strain gages. Quadriceps loads of 0, 250, 500, 750, and 1000 N were applied at flexion angles of 0, 15, 30, 60, 90, and 120°. Data were analyzed using repeated measures analysis of variance, as well as linear and quadratic regressions. The results indicated that there were no significant differences in strains between the two bundles and that the relationship between quadriceps load and ACL strain was highly linear for flexion angles of 60° and less, where the quadriceps caused markedly increased strains above those developed during passive motion.

KEYWORDS: quadriceps, anterior cruciate ligament

Injury to the anterior cruciate ligament is a serious problem in Alpine skiing [*1*]. Although studies of injury etiology have identified a number of mechanisms, most of which occur during falls, it has been proposed that ACL injury can occur without the skier falling. The skier, trying to avoid a fall and regain balance, strongly contracts the knee extensors without a contraction of the hamstrings, resulting in a muscularly induced anterior tibial translation [*15,16,23*].

Because current ski bindings respond only to the loads generated between the boot and ski, they are unable to account for the effects of muscularly induced loads. If the quadriceps can generate sufficient loads to rupture the ACL, then future bindings need to account for quadriceps contraction in making release decisions. To develop bindings that are better able to protect the knee from this injury mechanism, it is necessary to identify the degree to which quadriceps load impacts ACL strains and to perhaps develop a model relating quadriceps load to ligament strain.

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A potential complication to an injury prediction model might be the need to include terms for the strains in multiple sites in the ACL. Several studies have shown that the anteromedial (AMB) and posterolateral (PLB) bundles reciprocate in function over the range of passive flexion/extension [*2,21*]. Although these studies measured length changes and not strains, the AMB is most highly strained in flexion and the PLB in extension [*4*]. It could be expected that the bundle experiencing greater strain varies with flexion angle when the tibia is subjected to anterior shear as a result of quadriceps action. Thus, one objective of this work was to test the hypothesis that strains in the two bundles are different depending on flexion angle when physiologic levels of quadriceps loading are applied.

Previous research that has studied the effect of levels of quadriceps load greater than 100 N on ACL strain has limited measurements to the AMB [*12,13,29*]. These studies demonstrated that strain decreases under isometric quadriceps load with increasing flexion. This behavior is a consequence of the change in orientation of the patellar tendon with the tibia [*34*]; as flexion angle increases, the included angle between the tendon and tibia decreases, thus decreasing the anterior component of the quadriceps force.

Just as the strain due to quadriceps contraction varies with flexion angle, it may vary with quadriceps force. As the quadriceps force increases, the tibia will translate anteriorly, thus reducing the anterior shear component of the quadriceps force. The quadriceps force will also have a large component that acts to compress the joint surfaces, thereby increasing the stiffness of the joint [*31*]. These two effects may reduce the increases in ACL strain due to an increasing quadriceps force. Although the effect of increasing quadriceps force on ACL elongation has been studies at nonphysiologic force levels [*21*], these results can not be extrapolated because the effects previously noted would not be of consequence at such force levels. Thus, a second objective of this work was to test the hypothesis that the rate of strain increase in both the AMB and PLB decreases with increasing quadriceps force at physiologic force levels.

Methods

To test the hypotheses, liquid mercury strain gages (LMSGs) were installed on the surface of both the AMB and PLB of five unembalmed cadaver knee specimens (mean age 46.8 ± 16.5 years). These specimens were installed and aligned in a six-degree-of-freedom load application system (LAS) [*3*]. 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. Only the quadriceps actuator was used in the present study. 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 [*3,7*].

After installation and alignment, each specimen was subjected to a consistent preconditioning protocol. This protocol consisted of a 20-Nm hyperextension moment followed by a 22-Nm varus/valgus moment and a 330-N anterior/posterior force. The latter two loads were applied at flexion angles of 0, 90, and 120°. All loads were applied for five cycles, which was a sufficient number to gain reproducibility in load-deflection relations. Immediately following preconditioning the 0° flexion angle was defined by applying a 2.5-Nm extension moment.

The specimen was then removed from the apparatus, and liquid mercury strain gages (LMSGs) were installed on the AMB and PLB of the ACL. The AMB gage was installed via access gained by reflecting the patella using bilateral parapatellar incisions. Access to the PLB was gained through a small (<1 cm) incision in the posterior capsule. The LMSG was installed on the posterior fibers of the PLB. After both gages had been installed, the knee was closed and the specimen reinstalled in the LAS. The surgical procedures for the strain gage installation are described in greater detail in Ref *4*.

The knee was then forced into hyperextension through application of a 20-Nm extension moment, flexed passively to 125°, and returned to full extension. Knee flexion/extension (F/E) moment, joint displacements, and LMSG voltage were recorded continuously during the motion. The reference lengths of the LMSGs were set at 90° flexion for the AMB and 15° flexion for the PLB when the joint was unloaded. When the joint was either passively flexed 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.

After the passive F/E cycle, the specimens were subjected to a loading protocol to examine the effects of quadriceps loads on the ACL strains. This loading protocol was executed at flexion angles of 0, 15, 30, 60, 90, and 120°. Quadriceps loads of 250, 500, 750, and 1000 N were applied at each flexion angle.

To determine whether strains were different between the two bundles, the data from these experiments were subjected to a repeated measures ANOVA (RANOVA) procedure using the general linear models (GLM) routines of the SAS System (SAS Institute, Cary, NC). The analysis included both passive and quadriceps forces. Three within-subject variables, flexion angle (five levels with 0° being excluded), load (five levels), and measurement site (two levels), as well as their interactions, were modeled. The results of the RANOVA were subjected to a Tukey's Method of Multiple Comparisons procedure [*27*] to determine significant differences between the levels of the significant factors or interactions.

To describe the relationship between the quadriceps load and strain, the data were also subjected to least squares regression using linear ($\varepsilon_i = b_i + m_i Q$) as well as quadratic (second order, $\varepsilon_i = a_i Q^2 + b_i Q + c_i$) regression models, where Q was the quadriceps load and ε_i was the ACL strain in each bundle, i. The r^2 values from these analysis, along with tests for the significance of the quadratic coefficient, were used to determine which model best fit the data and therefore whether the strain increased at a constant rate with increasing quadriceps force or whether the strain increase decayed with increasing quadriceps force.

Results

For the case of isolated quadriceps loading, the RANOVA results indicated that the strain in the two bundles was not significantly different ($p = 0.809$). Both the flexion angle ($p = 0.0001$), load terms ($p = 0.0001$), and their two-way interaction were statistically significant ($p < 0.0001$). None of the interactions that included the bundle factor was significant.

Tukey's Method of Multiple Comparisons performed using the data from the RANOVA flexion-load interaction term revealed the reasons for the ANOVA results. Flexion angle was a significant factor because the strains averaged in the two bundles consistently decreased as flexion increased when a quadriceps load was applied (Table 1). Quadriceps load was a significant factor because the strains increased as the quadriceps load increased for flexion angles of 15, 30, and 60°. The interaction was significant because the strains remained at zero at 90 and 120° of flexion.

When the strain-load relations were examined for the individual bundles at each flexion angle, these relations appeared almost linear for the AMB at flexion angles of 15, 30, and 60° and for the PLB at flexion angles of 0, 15, and 30° where the strains were more than 2% higher than passive at 1000-N quadriceps force. At the remaining flexion angles, increasing quadriceps force had little or no effect on strain (Figs. 1, 2).

TABLE 1—*Results of Tukey's Analysis of Multiple Comparisons. Strain values in percent are averaged over the two bundles. Values with different letter superscripts are significantly different* ($p < 0.05$).

Flexion Angle (deg)	Quadriceps Load (N) 0	250	500	750	1000
15	$0.7^{\dagger}$	3.0^{def}	6.4^{abcd}	7.6^{abc}	8.5^{a}
30	$0.5^{\dagger}$	3.7^{cdef}	5.1^{abcde}	6.6^{abcd}	8.0^{ab}
60	$0.4^{\dagger}$	$0.9^{\dagger}$	1.7^{ef}	2.9^{def}	3.9^{bcdef}
90	$0.0^{\dagger}$	$0.0^{\dagger}$	$0.0^{\dagger}$	$0.0^{\dagger}$	$0.0^{\dagger}$
120	$0.7^{\dagger}$	$0.6^{\dagger}$	$0.4^{\dagger}$	$0.0^{\dagger}$	$0.0^{\dagger}$

Because of the linear appearance of the strain-load plots, the results of the linear regressions indicated that the linear model fit the data well. At the flexion angles where the quadriceps load markedly increased bundle strain, the minimum R-squared values were 0.94 and 0.96 for the AMB and PLB, respectively. Although the R-squared values for the quadratic model were greater, the coefficient of the quadratic term was positive in half of the regressions, and none of the coefficients was significantly different from zero.

The anterior displacement increased the most in the first 250-N increment of quadriceps loading as the knee was moved through the laxity region (Fig. 3). Beyond the first increment, the displacement increased linearly with increasing quadriceps load at a particular flexion angle of 60° or less and decreased with increasing flexion at a particular level of quadriceps load.

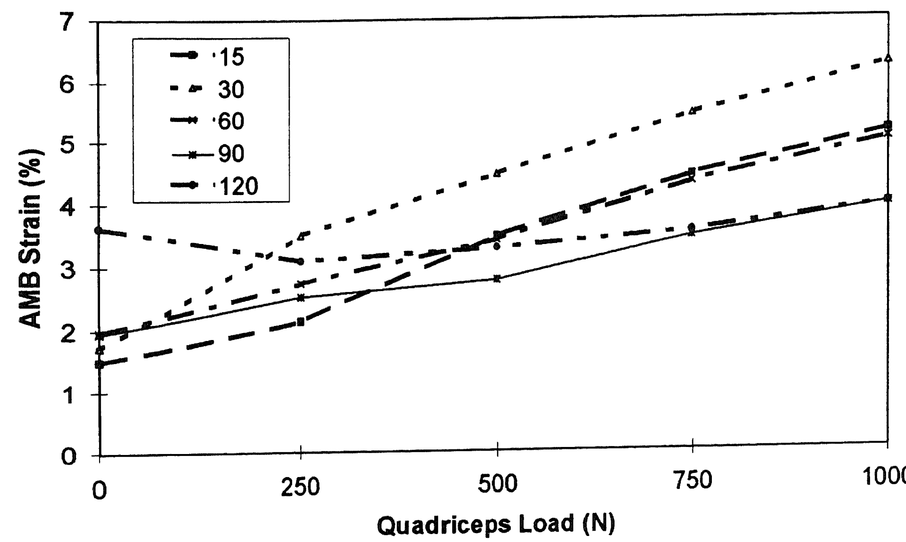

FIG. 1—*Strains in the anteromedial bundle (AMB) of the anterior cruciate ligament caused by quadriceps force applied at different flexion angles. Strains are averaged over the specimens* ($n = 5$).

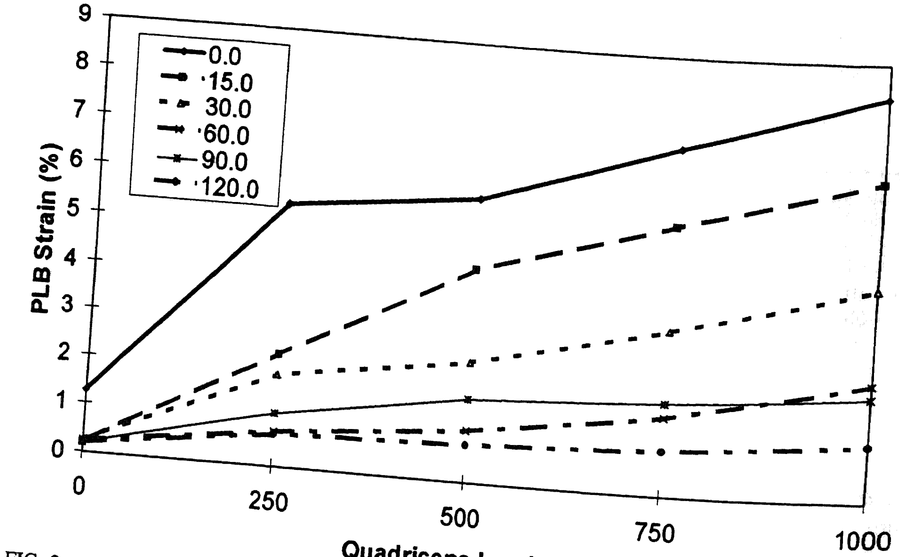

FIG. 2—*Strains in the posterolateral bundle (PLB) of the anterior cruciate ligament caused by quadriceps force applied at different flexion angles. Strains are averaged over the specimens (*n = *5).*

Discussion

Motivated by the desire to answer some basic questions about ACL injury mechanics that may prove useful to design of releasable bindings for Alpine skiing, this study tested two hypotheses. One hypothesis investigated the equality of strains in the AMB and PLB bundles

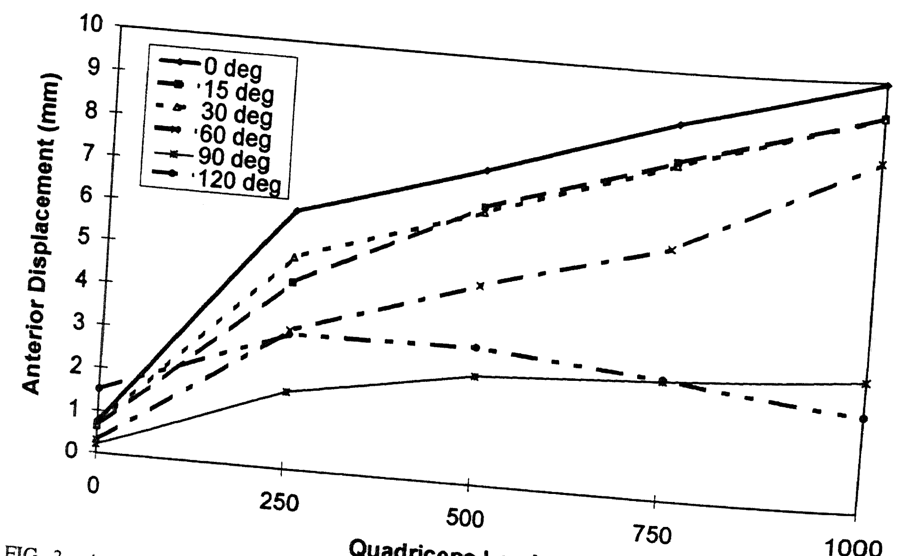

FIG. 3—*Anterior displacement of the tibia caused by quadriceps force applied at different flexion angles. Displacements are averaged over the specimens (*n = *5).*

and the other concentrated on the relationship between ACL strain and increasing quadriceps force. Because the data used to test these hypotheses were gathered experimentally, it is useful to critically examine the measurement technique 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 [*17*]. However, for length transducers such as LMSGs that have measurable stiffness [*9*] and that are installed with sufficient prestretch 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 $N\text{-}mm^{-1}$ [*9*], the method nevertheless introduces uncertainty into the reference length because the inflection point is not clearly defined [*8*].

To avoid the uncertainty mentioned above, a new method was used in this study where the reference length was set at flexion angles of 15 and 90° for the PLB and AMB, respectively. These angles corresponded to those where the total ACL tension became 5 N or less as the knee was flexed from full extension and as the knee was extended from full flexion [*32*]. In the range between the two angles of 15 and 90°, the tension remained at 5 N or less. Inasmuch as the bundles of the ACL reciprocate with the PCL carrying more load in extension and the AMB more load in flexion [2], the transition to low tension as the knee was flexed corresponded to a loss in tension and hence elongation of the posterior fibers, while the transition to low tension as the knee was extended corresponded to a loss in tension and hence elongation of the anterior fibers.

In developing the new method, the method of setting the reference length at the inflection point output voltage-load curve was also used to analyze the data. Although the strain values differed somewhat between the two methods, the conclusions of the study did not change. This dispels any concern that the new method affected the results of the study.

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 [*8*] or performing either a notchplasty or roofplasty [*13,24*]. Any of these techniques would have altered the biomechanics of the joint. In the present study, no steps were taken to prevent impingement of the AMB gage near full extension 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 on the AMB gage at 0° of flexion, these data were not included in any of the statistical 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. Each knee was visualized after gage installation to ensure that the transducer was not impinged during motion.

Because these sources of error did not influence the results, the results can be discussed meaningfully. In indicating that there was no significant difference between the two bundle stains, the results of the RANOVA did not support the first hypothesis. This behavior was unexpected based on the reciprocating behavior of the bundles during passive motion ov- the flexion/extension arc. The reciprocating phenomenon was originally identified from st ies that measured differences in bundle elongation during passive flexion/extension [2,21]. Given that the bundles' lengths are significantly different [20] with the AMB being roughly twice as long as the PLB and that the origin/insertion sites are different [2], it was reasonable to expect that the strain differences in passive motion would be maintained as the joint was subjected to a quadriceps force that would predominantly cause an anterior displacement of the tibia on the femur.

The equality of the bundle stains 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 [20]. 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 result that bundle strains are equal has potential practical importance in the design of releasable bindings that better protect against ligamentous knee injuries in Alpine skiing. One method of controlling binding release is to base the release decision on estimates of ligament strain from empirical models that relate strain to the injury-governing independent variables that affect strain. These variables include external loads, muscular loads, flexion angle, and loading rates. The equality of strain simplifies the implementation of this method because only one bundle need be considered for this specific loading case. Whether it is possible to consider only one bundle over the full spectrum of loading cases likely to cause ACL damage would be a useful question to answer.

The results of the experiments involving isolated quadriceps loading revealed that ACL strain increased almost linearly with increasing quadriceps load. Therefore the hypothesis that the rate of ACL strain increase decreases with increasing quadriceps force was rejected. The basis for this hypothesis was that a quadriceps force would cause an A/P translation of the tibia that would reduce the included angle between the tibia and the patellar tendon, thereby reducing the A/P component of the quadriceps force. In this way it was expected that quadriceps force would be self limiting in its ability to increase ACL stain. The most likely reason that this mechanism was not observed was that the anterior translations, which were bounded by 10 mm as a result of the quadriceps force, were not large enough to substantially reduce the included angle.

This result has direct implications to skiing injuries. Several authors have postulated that a vigorous quadriceps contraction can result in a rupture of the ACL [15,16,23,28]. Within the quadriceps load limit of 1000 N, the linear increases in AMB and PLB strains with increasing quadriceps force found by this study support this postulate. Although care must be taken in extrapolating this result to higher quadriceps forces, if the quadriceps can develop forces on the order of 7000 N, as suggested by Herzog (1985), then ACL rupture via this mechanism may be possible. This possibility stems from the relatively high levels of strain (approximately 8%) produced by the 1000-N force at full extension, from quadriceps excitation levels during controlled skiing maneuvers that routinely exceed levels measured during maximum voluntary contraction exercises [22] and from muscle excitations measured during simulated backward falls that showed that only the vastii were active during most of the fall duration, particularly for stiff boots [6]. Further testing at higher loads will be needed to verify this possibility. If verified, then the mechanism of ACL rupture, due to quadriceps contraction, further complicates the design of releasable bindings for Alpine skiing since there may not be any loads generated between the boot and the ski with which to initiate release.

The finding in this study, that isolated quadriceps contraction may cause ACL damage, is inconsistent with the conclusion by Chiang and Mote [10] due to differences in the experimental methods between the two studies. In their study, Chiang and Mote [10] measured the

anterior displacement of the tibia under isometric quadriceps contraction and equilibrated the extensive knee moment by a posterior force applied to the tibia. In our experiments, however, the moment developed by the quadriceps force was equilibrated by a pure couple applied externally to the femur. Since the application of a posterior force would negate the effect of the anterior component of the quadriceps force, this explains why Chiang and Mote [10] reported anterior displacement of only 2 mm at 30° of knee flexion, whereas our results indicated 9 mm for comparable quadriceps forces. Since posterior force is not externally applied by the boot to the tibia simultaneously with quadriceps force in a backward fall during which the skier is trying to regain his balance [18], the experimental procedures used herein should provide a more realistic measure of the anterior tibial translation caused by quadriceps contraction during a backward falling situation.

The decreasing effect of the quadriceps load on ACL strain with increasing flexion (Table 1) requires some reinterpretation of ACL injury mechanisms in Alpine skiing. According to McConkey [23], who reported on 13 cases of ACL rupture in Alpine skiing, one of the major characteristics consistent among these cases is that the skier was out of control with both the hips and knees flexed. Injury occurred while either falling backwards or in recovery from the falling back position. Because these actions would be accompanied by eccentric and concentric contractions respectively of the quadriceps, it was concluded that a major contributor to the injury was the quadriceps force. However, the results of the current study suggest that this conclusion could only be valid with the knee near full extension. At intermediate flexion angles (45 to 90°), the dominant load causing the ACL injury would be an externally applied anterior force rather than either a combination of anterior force and quadriceps load or a dominant quadriceps load [18]. The source of the anterior force would be the reaction force between the distal shank and the back of the ski boot.

In addition to the mechanism of isolated ACL rupture proposed by McConkey [23], active contraction of the quadriceps could contribute to other mechanisms of isolated ACL rupture when the knee is near full extension. One other mechanism is the well-documented "boot induced anterior displacement (BIAD)" scenario that occurs during landing where the tail of the ski contacts the snow before the tip. In this situation the boot top is thrust forward as the ski rotates to contact the snow along its whole length, thus applying an anterior force to the tibia [5]. The development of the anterior force applied to the tibia can take as long as 125 ms depending on the boot stiffness. Since this time is more than twice the activation time constant for the quadriceps muscles [33], considerable force could be developed by the quadriceps muscles assuming that they were not already activated before landing. Also, hyperextension has been proposed as a mechanism of isolated ACL rupture [30]. Although skiing accidents during forward falls have not been studied using mathematical modeling and simulation, such an analysis has been performed for snowboarding where the falling time should be similar [14]. In this analysis, the fall duration was 125 ms, thus allowing the development of large quadriceps forces as argued above.

Although the development of isolated quadriceps forces when the knee is near full extension can load the ACL, concomitant force developed by the hamstrings would unload the ACL [12,29]. Contraction of the hamstrings and quadriceps normally occurs during controlled skiing maneuvers [22] so that the protection offered by the hamstrings to ACL damage is normally present. However, if this protection is absent, which has been demonstrated in simulated backward falls [6], then any force produced by the quadriceps muscles would exacerbate the load carried by the ACL.

To prevent knee injuries due to either isolated vigorous quadriceps contraction or quadriceps contraction in conjunction with another load such as anterior force, the development of equipment that can ameliorate the ACL loading induced by quadriceps contraction is needed. Two different directions of development may prove useful. One direction is new s bindings where release is controlled by muscle activity. Inasmuch as ACL tears due to quadriceps load occur predominantly during either a backward fall or an effort to avoid it [15,16], the release mode that would be most effective would be an upwards release at the toe. Experimental bindings that modulate the release level in twist based on quadriceps activity, as indicated by the muscle electomyogram, have been constructed [26], but these bindings have not offered the release mode previously described. Thus, binding mechanisms that enable electromechanical release of the toe upwards are needed.

To realize a binding where release is influenced by quadriceps activity, a control algorithm that processes electromyographic data in conjunction with other inputs important to the injury process is also needed. For the injury mechanism in question, other inputs of importance would include both the state of hamstrings contraction and knee flexion angle. Knee flexion angle is important because the quadriceps cause significant ACL strains, primarily near full extension (Table 1), and hamstrings contraction is important because of its protective function to the ACL in this flexion angle range [12,29]. Although bioadaptive algorithms are a topic of research [11], this research has not yet addressed this injury mechanism. Accordingly, algorithms that explicitly address ACL injury as a result of quadriceps contraction in backward falls are needed.

Another direction of equipment development that may prove useful is boot design. Tests of boots with flexibility built into the rear spoiler have shown the ability to significantly affect both knee joint flexion and muscle electromyograms [6]. Whether these effects translate into reductions in ACL strain remains an unanswered question.

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Bindings