CLINICAL ORTHOPAEDICS AND RELATED RESEARCH
Number 391, pp. 266–274

Strain in the Human Medial Collateral Ligament During Valgus Loading of the Knee

John C. Gardiner, PhD; Jeffrey A. Weiss, PhD*; and Thomas D. Rosenberg, MD*,***

The medial collateral ligament is one of the most frequently injured ligaments in the knee. Although the medial collateral ligament is known to provide a primary restraint to valgus and external rotations, details regarding its precise mechanical function are unknown. In this study, strain in the medial collateral ligament of eight knees from male cadavers was measured during valgus loading. A material testing machine was used to apply 10 cycles of varus and valgus rotation to limits of ±10.0 N-m at flexion angles of 0°, 30°, 60°, and 90°. A three-dimensional motion analysis system measured local tissue strain on the medial collateral ligament surface within 12 regions encompassing nearly the entire medial collateral ligament surface. Results indicated that strain is significantly different in different regions over the surface of the medial collateral ligament and that this distribution of strain changes with flexion angle and with the application of a valgus torque. Strain in the posterior and central portions of the medial collateral ligament generally decreased with increasing flexion angle, whereas strain in the anterior fibers remained relatively constant with changes in flexion angle. The highest strains in the medial collateral ligament were found at full extension on the posterior side of the medial collateral ligament near the femoral insertion. These data support clinical findings that suggest the femoral insertion is the most common location for medial collateral ligament injuries.

The medial collateral ligament is one of the most frequently injured ligaments in the knee.[12] Medial collateral ligament injuries often occur during sports such as football or soccer from a direct blow to the lateral side of the knee or because of external rotation of the tibia during alpine skiing. Clinical reports suggest injuries to the medial collateral ligament most commonly are located near the femoral insertion.[10,11] After an isolated medial collateral ligament injury, adequate healing often occurs without surgical intervention. However, many medial collateral ligament injuries are accompanied by disruption of the anterior cruciate ligament, resulting in a poor healing outcome

From the *Department of Bioengineering, The University of Utah; and **The Orthopedic Specialty Hospital, Salt Lake City, UT.

Supported by Whitaker Foundation Biomedical Research and Transition Grants and National Institutes of Health Grant Number AR47369.

Reprint requests to Jeffrey A. Weiss, PhD, The University of Utah, Department of Bioengineering, 50 South Central Campus Drive, Room 2480, Salt Lake City, UT 84112.

Received: September 22, 2000.
Revised: December 8, 2000; February 12, 2001.
Accepted: March 13, 2001.

for the medial collateral ligament and the anterior cruciate ligament, leading to chronic instability of the knee. A complete understanding of the normal stress and strain state in the medial collateral ligament can aid in the understanding and prevention of ligament injuries and in the formulation of treatments.

Previous studies of the mechanical function of the medial collateral ligament can be categorized broadly as studies of ligament cutting that measured changes in joint laxity after cutting specific regions of the medial collateral ligament and other structures,[5] or studies that measured overall load in the ligament[14,15] or local tissue strain[1,3,8] during external loading. Studies of ligament cutting and measurements of tissue load have shown that the medial collateral ligament provides a primary restraint to valgus rotation and a secondary restraint to external rotation and anterior and posterior translations. Measurements of local tissue strain provide detailed information regarding the relative importance of different regions of the medial collateral ligament during specific loading conditions. Measurements have indicated that the strain distribution is inhomogeneous over the medial collateral ligament surface and that this nonuniform distribution changes with flexion angle and with the application of external loads.

Computational modeling techniques such as the finite element method provide an additional means to assess medial collateral ligament function and situations that may lead to injury.[2,4] Computational models of joints have the potential to predict quantities such as tissue stress and joint contact pressure that can be difficult or impossible to measure in experimental or clinical settings. Computational models also provide a repeatable tool for evaluating multiple clinical treatments, an approach that can eliminate the large intersubject variability that often limits the sensitivity of experimental and clinical investigations. Musculoskeletal modeling methodologies and computing power have advanced significantly in recent years. Unfortunately, the effective construction and validation of complex models can be a difficult and time-consuming process. Accurate experimental input parameters such as in situ strain are essential for creating subject-specific models. The validation of computational models also is dependent on experimental data to confirm that model-predicted quantities such as local tissue strain or joint stiffness are accurate. Once validated, these models can provide a valuable tool for understanding the mechanical function of normal and diseased joints and ligaments and for assessing the effects of clinical interventions.

Despite the importance of the medial collateral ligament in maintaining joint stability, many fundamental questions remain regarding its precise mechanical function. Measurements of medial collateral ligament strain reported in previous studies have been limited to few locations within the tissue, despite indications that strain is highly inhomogeneous over the entire medial collateral ligament surface.[1,3,8] In particular, strain in the region of the medial collateral ligament often referred to as the posterior oblique ligament has not been quantified previously, despite indications of its importance from ligament cutting studies.[5] In addition, strain measurements often have been made relative to a loaded reference state.[1] This procedure may grossly underestimate actual tissue strain levels by assuming a zero strain value exists in tissue that actually may be loaded at a level of 3% or more. Quantifying the initial tension in ligaments is an essential step in the construction of accurate computational models. The objective of the current study was to quantify the strain distribution in the entire medial collateral ligament during passive flexion and valgus loading. It was hypothesized that the strain distribution in the medial collateral ligament is nonuniform and that this distribution changes with flexion angle and with the application of a valgus torque. It also was hypothesized that strain would be highest near the femoral insertion in correspondence with clinically observed injury patterns and that strain in the anterior portion of the medial collateral ligament would decrease with increases in knee flexion angle.

MATERIALS AND METHODS

Sample Population and Preparation

Eight knees from male cadavers (age, 50 ± 7 years) were used in the current study. The fresh-frozen specimens were thawed at room temperature overnight before dissection and were inspected for signs of previous injury or arthritis. All periarticular soft tissue was removed until only the medial collateral, lateral collateral, anterior cruciate, and posterior cruciate ligaments and medial and lateral menisci remained intact (Fig 1). The femur, tibia, and fibula were potted in mounting tubes using a low-melt alloy. During all dissection and testing, the tissue was kept continuously moist with 0.9% buffered saline. All testing was completed within 5 hours, during which time no noticeable changes in the tissue were observed.

Kinematic Testing

The knees were mounted in custom fixtures on a biaxial material testing machine (MTS, Eden Prairie, MN) that allowed application of varus and valgus rotation at fixed flexion angles (Fig 2). A combination of linear and rotary bearings allowed joint distraction, medial and lateral translation, and tibial axial rotation to be unconstrained, whereas anterior to posterior translation was fixed in a neutral position based on an anterior to posterior test performed with the same fixtures before beginning each valgus test sequence. It has been shown that anterior or posterior translations do not occur during unconstrained varus and valgus loading,[9] suggesting that constraint of anterior and posterior translations should not affect joint kinematics. Varus and valgus torque and rotation were measured to an accuracy of ± 0.20 N-m and ± 0.10°, respectively. Ten cycles of varus and valgus rotation were applied at flexion angles of 0°, 30°, 60°, and 90°. The cyclic loading served to precondition the soft tissue structures of the knee. The varus and valgus rotation was applied at 1.0°/second to torque limits of ± 10.0 N-m. The rotation speed was chosen to provide quasistatic loading, where tissue viscoelastic effects and inertial effects of the kinematic fixtures could be minimized. The torque limit of 10 N-m is sufficiently large to enter the terminal stiffness region of the varus and valgus torque rotation curve but is much smaller than the torque required to induce tissue damage, allowing multiple tests to be done with the same specimen. All data analysis was performed using the results obtained during the loading phase of the tenth valgus cycle.

Strain Measurement

A noncontact three-dimensional motion analysis system (Peak Performance Technologies, Englewood, CO) was used for simultaneous measurement of strain at multiple locations on the medial collateral ligament surface. A custom calibration frame was constructed, and 18 retroreflective fiducial markers were attached to the frame. The coordinates of the markers were measured with a coordinate measuring machine (± 0.01 mm accuracy) to allow calibration of the test volume (approximately 350 cm^3) using the direct linear transformation method.[16] Before experimental varus and valgus loading, three rows of black acrylic markers (1.4 mm diameter) were attached to the medial collateral ligament surface using cyanoacrylate. The markers followed the local fiber direction between the insertions of the medial collateral ligament. A 3 × 5 grid of markers formed 12 gauge lengths of approximately 15 to 20 mm for measurement of strain along the local fiber direction (Fig 1). The marker positions on the medial collateral ligament surface were defined for universal anatomic landmarks. The anterior and middle rows were arranged along the well-defined superficial medial collateral ligament, and the posterior markers were

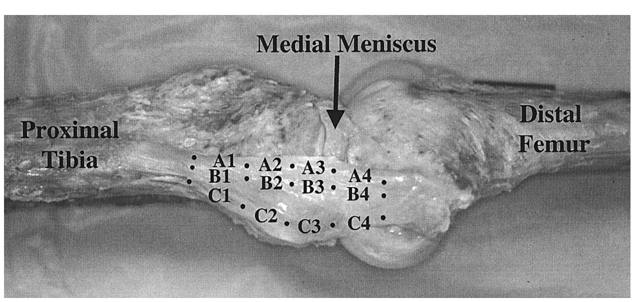

Fig 1. A photograph of a human knee that has been prepared for kinematic testing is shown. All periarticular soft tissue has been removed, and black, plastic markers have been attached to the medial collateral ligament surface in three rows (A–C) with five markers per row for video strain measurement.

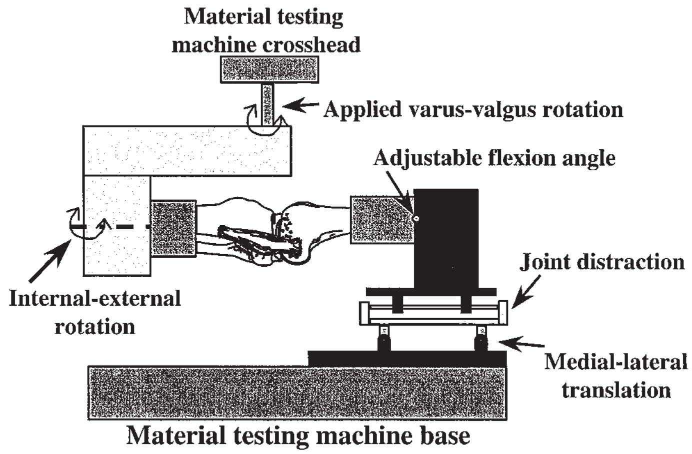

Fig 2. A schematic shows the kinematic fixtures used to apply valgus loading using a material testing machine.

attached to the area of the medial collateral ligament often referred to as the posterior oblique ligament.[7] Direct linear transformation calibration error estimates showed that the three-dimensional coordinates of each medial collateral ligament marker could be determined with a maximum error of $\pm$ 0.1 mm during all testing.

In Situ Strain

At the conclusion of kinematic testing, the medial collateral ligament was dissected free from its femoral, tibial, and meniscal attachments to measure the zero-load reference lengths between the markers.[19] The isolated ligament was placed on a saline-covered glass plate and allowed to assume its stress-free configuration. The test volume for the motion analysis system was recalibrated and used to record the stress-free position of the surface markers.

Data Analysis and Statistics

The lengths between marker pairs were determined in the reference state (l_0) and during each experimental condition (l). The tensile strain along the fiber direction between marker pairs was calculated as $\epsilon = (l - l_0)/l_0$, assuming that the deformation between marker pairs was homogenous. Although the medial collateral ligament was subjected to a complex deformation, transverse or shear strains were not measured. The effects of region, flexion angle, and loading condition (no valgus torque or 10 N-m valgus torque) on strain were assessed using statistical methods. Two separate sets of two-way analysis of variance tests with repeated measures were done using a Tukey test for multiple comparisons. In the first set of tests, the effects of region and flexion angle on medial collateral ligament strain were assessed for both loading conditions. In the second set of tests, loading condition effects were assessed for each flexion angle and region. The anterior, middle, and posterior bands of the central region (labeled A3, B3, and C3, respectively, in Figure 1) were used for statistical comparisons. Significance was set at $p \leq 0.05$ for all comparisons.

To aid the visualization of the results, the mean values of fiber direction strain were interpolated onto a finite element mesh constructed from computed tomography data of a human male distal femur, proximal tibia, and medial collateral ligament.[4] Areas between discrete measurement locations were assigned fiber strain values based on a least squares interpolation method to yield a continuous spatial representation of the results.

RESULTS

The application of a 10 N-m valgus torque induced a valgus rotation that increased with increasing flexion angle (3.2° $\pm$ 0.8°, 3.9° $\pm$ 1.0°, 5.3° $\pm$ 1.7°, and 5.8° $\pm$ 2.1° at flexion angles of 0°, 30°, 60°, and 90°, respectively). The valgus rotation was accompanied by internal tibial rotation, as has been observed by other investigators.[5] The cyclic varus and valgus torque versus rotation curves and strain values were repeatable by the tenth cycle because the soft tissue structures of the knee were preconditioned (Fig 3).

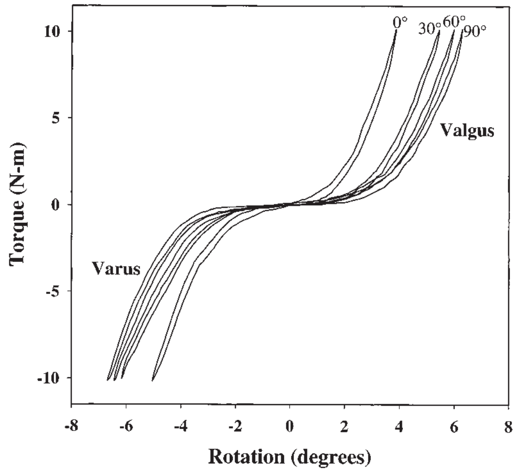

Fig 3. In this graph, varus and valgus rotation (degrees) versus torque (N-m) at flexion angles of 0°, 30°, 60°, and 90° can be seen. These data represent the tenth cycle of one knee specimen.

Effect of Knee Flexion

During passive flexion, average fiber strain ranged between 1% and 5%, depending on the medial collateral ligament region and flexion angle (Fig 4; Table 1). At full extension, the largest strain values were measured in the posterior fibers, and the smallest strain was in the anterior fibers. There was a statistically significant difference between strain in the anterior and middle regions and the anterior and posterior regions ($p < 0.05$ for both cases). Strain values in the posterior fibers decreased from their values at full extension as flexion angle was increased to 30° (not significant), 60° ($p < 0.001$), and 90° ($p < 0.001$). Similarly, the central portion of the medial collateral ligament experienced decreasing strain, with flexion angle increases to 30° (not significant), 60° ($p = 0.029$), and 90° ($p < 0.006$). Strain in the anterior border of the medial collateral ligament experienced no significant effects with changes in flexion angle from 0° to 90°. Buckling of the posterior proximal region of the medial collateral ligament was observed during testing at high flexion angles, resulting in negative measured values of strain (Fig 5).

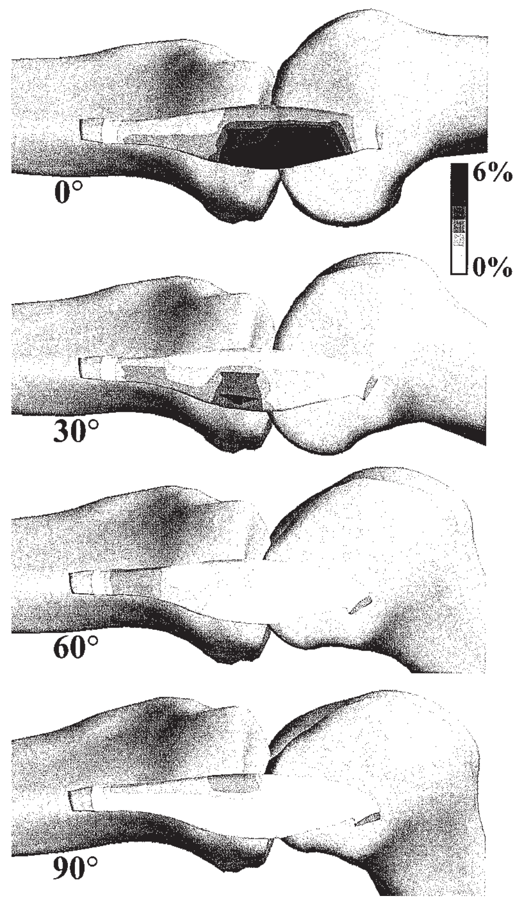

Fig 4. Regional medial collateral ligament fiber direction strain at four knee flexion angles is shown. Experimental values have been interpolated onto a finite element mesh to generate a continuous spatial representation of results. Medial collateral ligament strain varies with position within the ligament and changes with flexion angle.

Effect of Valgus Loading

During valgus loading, variations in medial collateral ligament surface strain between regions were observed for all flexion angles (Fig 6; Table 2). Valgus loading caused a significant increase in strain when compared with the unloaded configuration at all flexion angles and regions ($p < 0.01$ for all cases). The regional differences and trends were similar to those observed for passive flexion. At full extension, the largest strains again were found in the posterior proximal region of the medial collateral liga-

TABLE 1. Medial Collateral Ligament Strain (%) During Passive Flexion at Each Measurement Location (mean ± standard deviation)[1]

Flexion Angle	Measurement Location	1	2	3	4
0°	A	1.8 ± 1.5	1.5 ± 1.1	2.6 ± 1.5	2.4 ± 2.1
	B	2.0 ± 1.0	2.0 ± 1.1	3.9 ± 2.0	5.2 ± 3.4
	C	2.5 ± 1.6	2.7 ± 2.1	4.8 ± 2.7	5.8 ± 3.5
30°	A	1.8 ± 2.2	1.1 ± 1.0	1.7 ± 1.3	1.1 ± 1.7
	B	2.4 ± 2.3	1.7 ± 1.0	3.1 ± 2.0	1.7 ± 2.8
	C	2.3 ± 1.6	2.3 ± 2.2	3.8 ± 2.1	1.9 ± 3.3
60°	A	2.9 ± 1.9	1.4 ± 1.4	1.9 ± 1.2	*
	B	2.8 ± 1.8	1.3 ± 1.4	2.1 ± 1.2	*
	C	2.6 ± 1.5	1.7 ± 2.0	1.0 ± 1.5	*
90°	A	2.2 ± 3.3	2.0 ± 1.8	2.9 ± 2.4	*
	B	1.1 ± 2.2	1.2 ± 1.5	1.7 ± 2.2	*
	C	0.8 ± 0.9	1.2 ± 2.1	0.2 ± 2.5	*

[1]Measurement locations A1–C4 are defined in Figure 1. Ligament buckling prevented accurate strain measurement in regions indicated by *.

Fig 5. This photograph shows a human knee during flexion. Buckling can be observed in the medial collateral ligament on the posterior side of the femoral insertion.

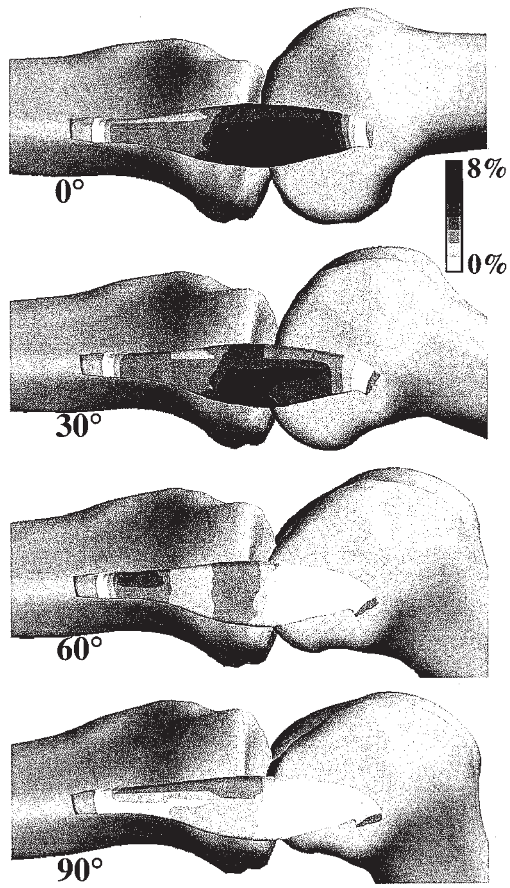

Fig 6. Medial collateral ligament fiber direction strain after application of 10 N-m valgus torque at four knee flexion angles is shown. Experimental values have been interpolated onto a finite element mesh to generate a continuous spatial representation of results. Medial collateral ligament strain varies with position within the ligament and changes with flexion angle.

TABLE 2. Medial Collateral Ligament Strain (%) During 10 N-m Valgus Loading at Each Measurement Location (mean ± standard deviation)[2]

Flexion Angle	Measurement Location	1	2	3	4
0°	A	2.9 ± 1.9	2.9 ± 1.2	4.1 ± 1.6	4.8 ± 2.1
	B	4.0 ± 1.3	3.3 ± 1.0	5.8 ± 2.5	7.9 ± 3.8
	C	3.6 ± 1.9	3.7 ± 2.1	7.4 ± 2.8	9.4 ± 3.4
30°	A	3.1 ± 2.4	2.9 ± 1.2	4.1 ± 1.9	3.8 ± 1.1
	B	3.9 ± 2.8	3.4 ± 1.5	5.0 ± 2.2	5.2 ± 2.7
	C	3.8 ± 2.1	3.8 ± 2.2	6.4 ± 3.8	4.7 ± 3.2
60°	A	4.0 ± 1.8	2.7 ± 1.6	3.6 ± 1.4	1.9 ± 1.6
	B	4.3 ± 1.8	2.8 ± 1.5	3.6 ± 1.6	1.1 ± 3.3
	C	3.4 ± 2.1	2.8 ± 2.0	3.2 ± 2.2	1.0 ± 2.5
90°	A	3.1 ± 3.3	3.1 ± 1.8	3.8 ± 2.2	1.7 ± 3.3
	B	2.3 ± 2.2	1.9 ± 1.5	2.7 ± 2.5	*
	C	1.4 ± 1.2	2.1 ± 1.9	2.3 ± 3.4	*

[2]Measurement locations A1–C4 are defined in Figure 1. Ligament buckling prevented accurate strain measurement in regions indicated by *.

ment. Strain values in the posterior fibers decreased from their values at full extension as flexion angle was increased to 60° ($p = 0.007$) and 90° ($p = 0.004$). Similarly, the central portion of the medial collateral ligament experienced decreasing strain with flexion angle increases to 60° ($p = 0.015$) and 90° ($p < 0.001$). There were no significant effects of changes in flexion angle between 0° and 90° on strain in the anterior border of the medial collateral ligament.

DISCUSSION

This study determined the strain distribution in the human medial collateral ligament as a function of knee flexion angle and valgus loading using a three-dimensional motion analysis system (Peak Performance Technologies). Medial collateral ligament surface strain varied with position on the medial collateral ligament surface and with knee flexion angle, and increased substantially during valgus loading of the knee. Results indicated that the most highly strained region of the medial collateral ligament changed with flexion angle, suggesting that the localized area of the medial collateral ligament most vulnerable to injury may change with knee position.

Hull et al[8] measured strain in four regions corresponding to the four most anterior and proximal portions of the medial collateral ligament as identified in the current study. Similar trends were observed for the strain behavior under valgus loading, but the generally lower strains reported in their study can be attributed in part to the lower magnitude of joint loading, differences in strain measurement method, and location of measurement. Differences in the reported strain measurements also may be attributable to the fundamentally different techniques used for establishing the ligament reference length. In the study of Hull et al,[8] strain was measured using liquid mercury strain gauges, and the reference length was defined as the inflection point on the load versus strain curve.

The results from the current study indicated that the largest strain in the medial collateral ligament during valgus loading occurs near the femoral insertion in the fully extended knee, suggesting that this region may be most vulnerable to injury under these loading conditions. Although previous experimental studies have not measured strain in this region,[1,3,8] computational models in the study of Bendjaballah et al[2] support this result. Clinical observations of injury patterns and locations also confirm that this region of the medial collateral ligament is the most common location for medial collateral ligament injuries,[10,11] and injuries in this area generally do not heal as well as injuries located distal to the joint line.[13]

Measurement of ligament strain was limited to uniaxial stretch along the local collagen fiber direction. Transverse and shear strains in the medial collateral ligament also are present during the complex deformations experienced by the medial collateral ligament. Fiber direction strain has been the focus of other studies of ligament strain[1,3,8] attributable to ligament's primary function of resisting tensile loads. In addition, transverse and shear strains are small in magnitude relative to the fiber direction strains and were not possible to quantify accurately using the current techniques. The fiber direction strain values also are essential for the approach the laboratory has used for providing an initial tension to computational models of the medial collateral ligament.[4,18]

When interpreting the results from kinematic measurements taken in the cadaveric knee, one must consider that normal kinematics may have been altered because of the dissection necessary for strain measurement or because of the lack of stabilizing muscle or joint compressive forces. These changes may increase joint laxity from normal physiologic conditions and may contribute to an overestimation of medial collateral ligament strain. In addition, viscoelastic effects that may be important in impact loading situations were neglected in the quasistatic loading of the current study. Under high-rate loading, a stiffer joint response may be observed because of viscoelastic effects resulting in less medial collateral ligament strain for a given loading level.

The calculation of strain in the current study was based on changes in gauge length between discrete markers. Inherent to this technique is the assumption of a homogenous deformation between marker pairs. Any bending or buckling of the tissue between markers will cause the measured value of strain to differ from the actual local tissue state. To ensure a homogenous deformation, markers should be placed as close together as possible. Conversely, accuracy limitations of the optical system imply that markers should be separated as far as possible for the most accurate measurement of length changes between markers. In the current study, markers were attached to form gauge lengths of 15 to 20 mm. This distance was chosen to provide a compromise between the two conflicting accuracy limitations and allowed measurements to be made to within an accuracy of 0.5% strain. Buckling was observed in the medial collateral ligament in the posterior proximal region during knee flexion (Fig 5). Under these conditions, the measured strain in the corresponding regions was negative (Tables 1,2). It is unlikely that strain in these regions is in as severe a compressive state as predicted by the data. More likely, the strain measurement technique is incapable of distinguishing the differences between compressive strain and an inhomogeneous strain field.

Additional characteristics of the strain measurement method may introduce some inaccuracy and variability between specimens. The use of cyanoacrylate adhesive for marker attachment may cause local hardening of the tissue, altering natural tissue deformation. However, the area of application for the adhesive was small relative to the marker size and ligament cross section, so this effect likely was minimal. In addition, strain measurement on the medial collateral ligament surface may not accurately predict the state of tissue deeper within the ligament substance. However, previous work has suggested that strain is uniform throughout the thickness of relatively thin soft tissue structures such as the medial collateral ligament.[17] Similarly, the continuous strain values shown in Figures 4 and 6 were calculated assuming that fiber direction strain is distributed uniformly in the cross-fiber direction between rows of markers. These interpolated values may be inaccurate if inhomogeneities exist in the cross-fiber direction.

The strain variability between specimens shown in Table 2 is a result of many factors, but most notably, the relative stiffness of the joints in valgus loading will affect the strain measured between specimens. Although each knee was loaded with the same 10 N-m valgus torque, geometric and material differences between specimens caused a wide range in the subsequent amount of valgus rotation. Interspecimen strain variability likely would be less if rotation limits were used, rather than torque limits. For the current study, a torque-controlled experiment allowed assessment of the variation in strains between specimens that was produced by the differences in joint stiffness of the studied knee specimens. This information allowed the current authors to assess the variance in the population. Rotation limits may provide more consistent strain measurements between specimens, but they would not be at comparable levels of joint loading. It has been suggested that ligaments function to resist repetitive loads, rather than repeated deformations during cyclic activities such as walking.[6] In addition, it can be argued that clinically observed injuries generally are caused by exceeding a critical torque or force limit, rather than a rotation limit, although torque and rotation certainly are interrelated, and it seems likely the quantity that causes a particular injury will depend on the specific injury mechanism.

This study showed that medial collateral ligament strain varies dramatically with region and flexion angle. The medial collateral ligament is most highly strained near the femoral insertion while the knee is at full extension, indicating that the femoral insertion is likely to be most vulnerable to injury at this flexion angle. Valgus rotation causes an increase in medial collateral ligament strain in all regions at all flexion angles. These results are useful for understanding the relative roles of different regions of the medial collateral ligament in various knee orientations. The strain distribution data obtained in this study currently are being used for the development and validation of subject-specific finite element models for each knee of this study.[4] The regionally measured in situ stretch is applied to each model,[18] and local strain measurements during valgus loading are used for validation. Validated models of the medial compartment will be used to simulate the effects of injury and surgical treatments.

References

1. Arms S, Boyle J, Johnson R, et al: Strain measurement in the medial collateral ligament of the human knee: An autopsy study. J Biomech 16:491–496, 1983.
2. Bendjaballah MZ, Shirazi-Adl A, Zukor DJ: Finite element analysis of human knee joint in varus-valgus. Clin Biomech 12:139–148, 1997.
3. Fischer RA, Arms SW, Johnson RJ, et al: The functional relationship of the posterior oblique ligament to the medial collateral ligament of the human knee. Am J Sports Med 13:390–397, 1985.
4. Gardiner JC, Weiss JA: Experimental testing and computational modeling to determine the stress-strain distribution in the human medial collateral ligament. Trans Orthop Res Soc 23:1027, 1998.
5. Haimes JL, Wroble RR, Grood ES, et al: Role of the medial structures in the intact and anterior cruciate ligament-deficient knee: Limits of motion in the human knee. Am J Sports Med 22:402–409, 1994.
6. Holden JP, Grood ES, Korvick DL, et al: In vivo forces in the anterior cruciate ligament: Direct measurements during walking and trotting in a quadruped. J Biomech 27:517–526, 1993.
7. Hughston JC, Eilers AF: The role of the posterior oblique ligament in repairs of acute medial (collateral) ligament tears of the knee. J Bone Joint Surg 55A:923–940, 1973.
8. Hull ML, Berns GS, Varma H, et al: Strain in the medial collateral ligament of the human knee under single and combined loads. J Biomech 29:199–206, 1996.
9. Inoue M, McGurk-Burleson E, Hollis JM, et al: Treatment of the medial collateral ligament injury: I. The importance of anterior cruciate ligament on the varus-valgus knee laxity. Am J Sports Med 15:15–21, 1987.
10. Kawada T, Abe T, Yamamoto K, et al: Analysis of strain distribution in the medial collateral ligament using a photoelastic method. Med Eng Phys 21:279–291, 1999.
11. Lee JI, Song IS, Junk YB, et al: Medial collateral ligament injuries of the knee: Ultrasonographic findings. J Ultrasound Med 15:621–625, 1996.
12. Miyasaka KC, Daniel DM, Stone ML, et al: The incidence of knee ligament injuries in the general population. Am J Knee Surg 4:3–8, 1991.
13. Robins AJ, Newman AP, Burks RT: Postoperative return of motion in anterior cruciate ligament and medial collateral ligament injuries: The effect of medial collateral ligament rupture location. Am J Sports Med 21:20–25, 1993.
14. Sakane M, Livesay GA, Fox RJ, et al: Relative contribution of the ACL, MCL, and bony contact to the anterior stability of the knee. Knee Surg Sports Traumatol Arthrosc 7:93–97, 1999.
15. Shapiro MS, Markolf KL, Finerman GA, et al: The effect of section of the medial collateral ligament on force generated in the anterior cruciate ligament. J Bone Joint Surg 73A:248–256, 1991.
16. Shapiro R: Direct linear transformation method for three-dimensional cinematography. Res Q 49:197–205, 1978.
17. To YC, Kwan ML, Woo SL-Y: Simultaneous measurement of strains on two surfaces of tendons and ligaments. J Biomech 21:511–514, 1988.
18. Weiss JA, Maker BN, Schauer DA: Treatment of initial stress in hyperelastic models of soft tissues. Proc ASME Bioeng Con 29:105–106, 1995.
19. Woo SL-Y, Weiss JA, Gomez MA, et al: Measurement of changes in ligament tension with knee motion and skeletal maturation. J Biomech Eng 111:46–51, 1990.