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Force Measurements on the Posterior Oblique Ligament and Superficial Medial Collateral Ligament Proximal and Distal Divisions to Applied Loads

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Background: There is limited information regarding load responses of the posterior oblique and superficial medial collateral ligaments to applied loads.

Hypotheses: The degree of knee flexion affects loads experienced by the posterior oblique ligament and both divisions of the superficial medial collateral ligament. The posterior oblique ligament provides significant resistance to valgus and internal rotation forces near knee extension. Different load responses are experienced by proximal and distal divisions of the superficial medial collateral ligament.

Study Design: Descriptive laboratory study.

Methods: Twenty-four nonpaired, fresh-frozen cadaveric knees were tested. Buckle transducers were applied to the proximal and distal divisions of the superficial medial collateral and posterior oblique ligaments. Applied loads at 0°, 20°, 30°, 60°, and 90° of knee flexion consisted of 10 N·m valgus loads, 5 N·m internal and external rotation torques, and 88 N anterior and posterior drawer loads.

Results: External rotation torques produced a significantly higher load response on the distal superficial medial collateral ligament than did internal rotation torques at all flexion angles with the largest difference at 90° (96.6 vs 22.5 N). For an applied valgus load at 60° of knee flexion, loads on the superficial medial collateral ligament were significantly higher in the distal division (103.5 N) than the proximal division (71.9 N). The valgus load response of the posterior oblique ligament at 0° of flexion (19.1 N) was significantly higher than at 30° (10.6 N), 60° (7.8 N), and 90° (6.8 N) of flexion. At 0° of knee flexion, the load response to internal rotation on the posterior oblique ligament (45.8 N) was significantly larger than was the response on both divisions of the superficial medial collateral ligament (20 N for both divisions). At 90° of flexion, the load response to internal rotation torques reciprocated between these structures with a significantly higher response in the distal superficial medial collateral ligament division (22.5 N) than the posterior oblique ligament (9.1 N).

Conclusion: The superficial medial collateral ligament experienced the largest load response to applied valgus and external rotation torques; the posterior oblique ligament observed the highest load response to internal rotation near extension.

Clinical Relevance: This study provides new knowledge of the individual biomechanical function of the main medial knee structures in an intact knee and will assist in the interpretation of clinical knee motion testing and provide evidence for techniques involving repair or reconstruction of the posterior oblique ligament and both divisions of the superficial medial collateral ligament.

Keywords: superficial medial collateral ligament (sMCL); posterior oblique ligament (POL); force measurement; knee biomechanics; buckle transducers

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Compared with the information about the cruciate ligaments and the posterolateral knee, there is a paucity of quantitative biomechanical information, using the accuracy of modern testing methods, on the function of individual structures of the medial knee.[4,5,12,20,22,23,28] Because injuries to the superficial medial collateral ligament (sMCL) and its supporting structures are the most common knee injuries,[4,8,10] this information is considered necessary to assist in both the proper diagnosis and ensuing nonoperative or surgical treatment of these structures.

Although several previous studies have reported on the qualitative anatomy of medial knee structures,[2,5,8,11,17,36] it was only recently that a study quantitatively defined the courses and attachment sites of these individual structures.[14] Therefore, this study sought to quantitatively reassess the biomechanical function of the posterior oblique ligament (POL) and the sMCL, the 2 main medial knee structures, based on their quantitative anatomical descriptions.[14]

The sMCL is the largest structure over the medial aspect of the knee.[2,14,17,35] It has become commonly agreed on that the sMCL is the primary valgus stabilizer of the knee.[1,6,12,24] The sMCL has been defined to have 1 femoral and 2 tibial attachments, which effectively divides it into 2 divisions.[14] The proximal tibial attachment is primarily to soft tissues as opposed to the distal tibial attachment, which attaches directly to the bone on the medial aspect of the tibia.[14] Thus, in effect, the sMCL is split into proximal and distal divisions owing to its 3 different attachment sites over the medial knee[14] (Figure 1).

The POL is a reinforcement of the posteromedial capsule, which courses off the distal aspect of the semimembranosus tendon.[8,9,14] From a biomechanical perspective, the POL has been reported to function as an internal rotation and valgus stabilizer between 0° and 30° of flexion.[4,5,9,20,22] The POL has also been reported to function as a secondary stabilizer against posterior tibial translation in posterior cruciate and medial collateral ligament–deficient knees.[3,26,27] Despite these studies, there is still limited information regarding the function of the POL in resisting anterior and posterior tibial translation in an intact knee.

Many studies have previously indirectly examined the biomechanics of the medial knee through sequential cutting studies.[4,22,23,25,28,35] However, none of the previous studies have directly measured the tensile forces on these structures during joint loading of an intact knee. Identification of the quantitative forces seen on the main static stabilizing structures of the medial knee would provide further understanding for the interpretation of clinical examinations and in understanding the clinical importance of repairing or reconstructing the main components of medial knee structures to reapproximate the native load sharing between these structures. Therefore, our hypothesis is 3-fold: (1) The degree of knee flexion affects loads experienced by the POL and both divisions of the sMCL, (2) the POL provides significant resistance to valgus and internal rotation forces near knee extension, and (3) different load responses are experienced by the proximal and distal divisions of the sMCL. Thus, the purpose of this study is to quantitatively measure the load

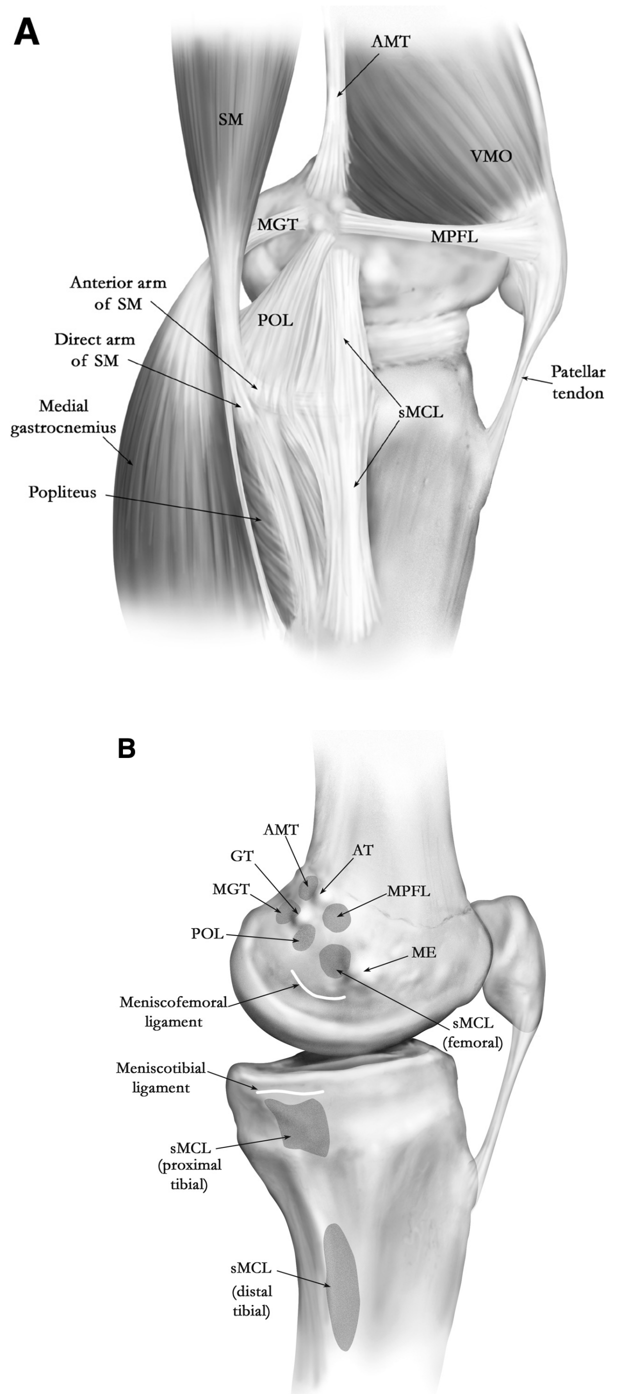

Figure 1. A, the superficial medial collateral ligament (sMCL; medial aspect, left knee). Reprinted with permission from *The Journal of Bone Joint Surgery*, 2007;89(9):2004, Figure 4. B, the medial knee bony attachment sites (medial aspect, left knee). Reprinted with permission from *The Journal of Bone Joint Surgery*, 2007;89(9):2004, Figure 2. AMT, adductor magnus tendon; AT, adductor tubercle; GT, gastrocnemius tubercle; ME, medial epicondyle; MGT, medial gastrocnemius tendon; MPFL, medial patellofemoral ligament; POL, posterior oblique ligament; SM, semimembranosus; VMO, vastus medialis obliquus.

responses of the POL and the proximal and distal divisions of the sMCL to externally applied loads in the directions of forces normally experienced by the knee at varying degrees of knee flexion.

MATERIALS AND METHODS

Preparation

Twenty-four nonpaired, fresh-frozen cadaveric knees with no prior evidence of injury and a mean age of 69.5 years (range, 45-87 years) were used for this project. The knees were stored at –20°C and thawed overnight in a 2°C refrigerator before dissection and subsequent biomechanical testing. The femur was severed 20 cm proximal to the knee joint, and all soft tissues were removed from its proximal aspect before potting in polymethylmethacrylate. The tibia and fibula were sectioned 12.5 cm distal to the joint line, and all soft tissues were removed from the distal end. The marrow cavity of the tibia was reamed with a 13-mm drill bit, and a threaded rod was inserted into the marrow cavity, parallel with the long axis of the tibia, and fixed in position with polymethylmethacrylate to inhibit any rotation of the rod with respect to the tibia. An eye screw was placed into the tibial tubercle in anticipation of application of anterior and posterior drawer loads. A customized hexagonal nut with a statically affixed eye screw was attached to the threaded tibial rod 22.9 cm distal to the joint line in a manner in which an S-type load cell could be used to pull the specimen-rod construct medially for the application of valgus loads. A lock nut, consisting of 2 hexagonal nuts tensioned in opposing directions to inhibit rotation of the distal nut in relation to the threaded rod during the application of internal and external rotation torques, was attached to the distal end of the threaded rod. Dissection to identify the medial knee structures was then performed. The skin and subcutaneous tissues were removed initially, followed by detachment of the semitendinosus, gracilis, and sartorius muscles and tendons. Deeper dissection isolated the POL and sMCL.

The specimen was mounted into the knee testing apparatus, which firmly secured the femur at a horizontal angle while allowing uninhibited movement of the tibia and fibula at fixable flexion angles. After the knee was aligned in the testing apparatus, buckle transducers were securely fastened to the POL, the proximal division of the sMCL, and the distal division of the sMCL (Figure 2). The buckle transducers consisted of a crossbar and a rectangular stainless steel frame containing semiconductor strain gauges. The use of these devices has been previously described in detail.[15,16,18,19] Assorted depths of crossbars reduced the potential complications of surrounding bone or soft tissue impingement of the buckle frames during biomechanical testing and prevented undesirable ligament damage.[18] The distal sMCL buckle transducer was placed on the sMCL at the midpoint between the 2 tibial sMCL attachments. The proximal sMCL buckle was placed between the femoral and proximal tibial sMCL attachments (Figure 2). The buckle application consisted of placing the frame over the ligament and inserting the crossbar both below the ligament and above the frame. With this setup, loading of the specific ligament and the resultant ligamentous tension pushed the crossbar against the frame, similar to a 3-point bending, and induced a voltage response in the strain gauge inside the buckle frame, which then conveyed a proportional voltage output to a Wheatstone bridge. Finally, this voltage output was relayed to a data acquisition system. Previous studies reported that there is a linear relationship between applied load and voltage output.[19] Therefore, the final data were measured in kilonewton per volt and provided for a measured tension. Buckles were zeroed before each external load application and calibrated with a known load posttesting to ensure accurate load measurements. Buckle transducers have been reported to be repeatable to within 0.7% using a similar biomechanical testing protocol.[18]

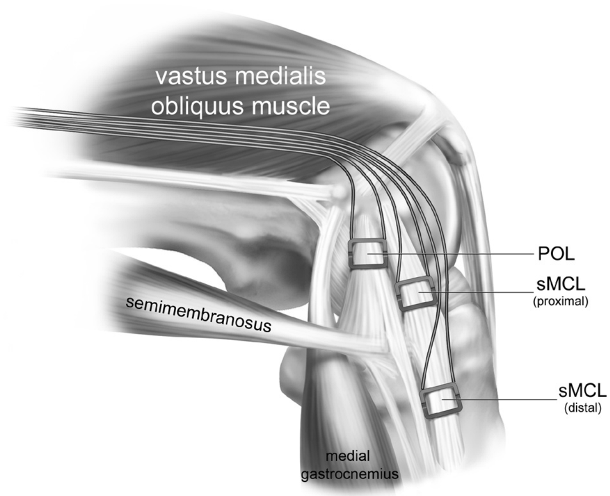

Figure 2. Approximate buckle transducer placement on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions. The sMCL buckles were placed proximal to their respective attachment sites (medial view, left knee).

Biomechanical Testing

Each knee was tested at 0°, 20°, 30°, 60°, and 90° of knee flexion, which was ensured by maintaining contact between the tibia and the flexion angle bar on the testing apparatus. A 100-N force model SM S-type load cell (Interface, Scottsdale, Arizona; manufacturer reported nonrepeatability error of ±0.01% and temperature-compensated strain gauges) was used to apply valgus loads on the tibial rod 22.9 cm from the knee joint line and anterior drawer and posterior drawer loads on the tibial tubercle. With use of a removable hook firmly attached to the S-type load cell, valgus loads were applied by directing the tibia medially through the previously described customized hexagonal nut. External and internal rotation torques were applied at the lock nut on the tibial rod through a socket firmly attached to a 15 N·m-capacity model TS12 shaft-style reaction torque transducer (Interface; manufacturer reported nonrepeatability error of ±0.02%). Specimens were loaded with the following external forces applied to the knee during testing: 10 N·m valgus loads, 88 N anterior and posterior drawer loads, and 5 N·m internal and external rotational moments. Varus loads were not applied to the knee because pilot knee testing of 3 specimens found that there were no load responses from these structures during the application of a 10 N·m varus load. The data were collected continuously while the external forces were applied to the specimen 3 times at each tested flexion angle. Before each external load application, a neutral starting position was attained with intra-articular contact between the medial and lateral tibial plateaus and femoral condyles.

Data Analysis

The amount of force applied to the ligaments during testing was determined by calibrating the measured buckle voltages against a known load. After the ligament of interest was sectioned, it was directly tensioned using the S-type load cell to calibrate the buckle of interest. The recorded voltages of both the buckle transducer and S-type load cell during this calibration step were imported into MatLab R2006b analysis software (MathWorks Inc., Natick, Massachusetts). Voltages from the load cell were converted to forces according to the load cell's force-voltage ratio to create a comparison of the applied load cell tension with the voltage recorded on the buckle transducer. A linear regression analysis was then used to define the voltage-tension relationship, which allowed for determination of the observed tensions during the testing phase of the experiment.

During biomechanical testing, voltage measurements were simultaneously collected from the buckle transducers and the previously described shaft-style reaction torque transducer or S-type load cell. Voltage outputs from the load cells were converted to forces according to the load cell's force-voltage ratio. Data were collected continuously during external load application. The data were filtered to analyze only buckle transducer outputs from moments when the target loads were applied to specimen. When the applied external force matched the target force, the correlating buckle voltage outputs were used to calculate the tensile load response in the ligament of interest according to the previously calculated linear regression outlined above.

Statistical Analysis

Statistical analysis using a paired 2-tailed Student *t* test was performed to compare the load responses of each individual ligament to internal rotation versus external rotation at each flexion angle. One-way analysis of variance (ANOVA) was performed to compare each particular ligament's load response to valgus loads, internal and external rotation torques, and anterior and posterior drawer loads for each particular flexion angle. The Tukey's honestly significant differences (HSD) test was subsequently used for post hoc comparisons to detect significant difference between particular pairs of the load responses considered in the given ANOVA. Two-way ANOVA was conducted to assess if there was a significant difference between the loading patterns of the 3 ligaments tested across all flexion angles and applied forces. A significant difference was determined to be present for $P < .05$.

RESULTS

The mean load responses of the ligaments for each applied force at a particular flexion angle are reported in Table 1.

Force on the sMCL

The maximum load on either division of the sMCL was on the distal sMCL division (103.5 N) for an applied valgus load at 60° of knee flexion. At this flexion angle, the load response to a 10 N·m valgus load was significantly higher in the distal sMCL division compared with the proximal sMCL division (103.5 vs 71.9 N, respectively; $P < .05$) (Figure 3). The distal sMCL division valgus load response at 0° of flexion was significantly less than were load responses at 30° ($P < .01$), 60° ($P < .03$), and 90° ($P < .01$) of flexion. There was also a significant increase in load on the distal sMCL division at 60° of flexion compared with 90° of knee flexion ($P < .04$). For load responses to a valgus load in the proximal sMCL division, Tukey's HSD test, performed after 1-way independent-sample ANOVA, did not detect significant differences between any of the tested flexion angles.

An applied 5 N·m external rotation torque produced its maximum overall load response on the distal sMCL (96.6 N) at 90° of knee flexion (Figure 4). Load responses were significantly higher for both the proximal and distal divisions of the sMCL compared with the POL load response at each tested flexion angle ($P < .0001$). Similar to a load response to a valgus load, the distal sMCL division external rotation torque load response at 0° of flexion was significantly less than were load responses at 30° ($P < .03$), 60° ($P < .01$), and 90° ($P < .01$). The load response of the distal sMCL division at 20° of flexion was significantly less than at both 60° and 90° of flexion ($P < .01$). In addition, compared with the load response at 30° of knee flexion, there was a significantly increased load response of the distal sMCL division at 90° of flexion ($P < .01$). The external rotation torque load response of the proximal sMCL division at 0° of flexion was significantly less than was the load response at both 60° and 90° of knee flexion ($P < .01$). Compared with the load response at 20° of knee flexion, there was also a significantly increased load on the proximal sMCL division at 60° of flexion with an applied external rotation torque ($P < .03$).

Although a 5 N·m internal rotation torque produced a load response on both sMCL divisions at all flexion angles, only 1 significant difference was discovered. The load response on the distal sMCL division was significantly greater at 30° of flexion than at 90° of knee flexion ($P < .02$) (Figure 5). External rotation torques produced a significantly higher load response than did internal rotation torques for both the proximal and distal sMCL divisions

TABLE 1
Mean Load Response of Force (N) per Applied Load (mean ± standard error of the mean) for the Posterior Oblique Ligament (POL) and the Proximal and Distal Superficial Medial Collateral Ligament (sMCL)[a]

Ligament and Loading Condition	Flexion 0°	20°	30°	60°	90°
POL					
Valgus moment	19.07 ± 3.77	12.63 ± 3.15	10.61 ± 2.96	7.82 ± 2.20	6.81 ± 2.26
Internal rotation torque	45.82 ± 8.27	41.94 ± 7.71	32.39 ± 5.87	13.96 ± 3.80	9.06 ± 2.30
External rotation torque	9.82 ± 2.93	7.89 ± 2.16	9.14 ± 2.79	7.68 ± 2.48	6.45 ± 2.15
Anterior drawer moment	9.13 ± 2.36	10.28 ± 2.78	7.54 ± 2.31	6.24 ± 2.15	6.14 ± 2.22
Posterior drawer moment	6.94 ± 2.39	8.05 ± 2.57	5.86 ± 2.11	6.83 ± 2.56	5.43 ± 2.23
Proximal sMCL					
Valgus moment	55.13 ± 7.23	78.51 ± 9.15	71.83 ± 9.26	71.86 ± 8.43	60.03 ± 8.07
Internal rotation torque	20.16 ± 4.45	28.77 ± 5.47	24.36 ± 5.03	25.08 ± 5.64	20.23 ± 4.81
External rotation torque	55.66 ± 8.35	63.51 ± 10.58	78.06 ± 10.94	82.71 ± 8.16	85.96 ± 13.54
Anterior drawer moment	16.14 ± 4.72	17.59 ± 4.65	19.20 ± 5.13	20.29 ± 4.74	19.56 ± 4.86
Posterior drawer moment	17.30 ± 5.05	18.75 ± 5.33	16.55 ± 4.57	20.88 ± 4.99	21.08 ± 6.72
Distal sMCL					
Valgus moment	68.21 ± 5.54	100.05 ± 8.21	101.90 ± 9.65	103.52 ± 7.93	86.10 ± 8.01
Internal rotation torque	20.47 ± 4.03	26.02 ± 3.58	33.00 ± 8.10	33.34 ± 7.17	22.55 ± 5.33
External rotation torque	45.82 ± 6.39	50.52 ± 7.58	63.75 ± 7.42	84.64 ± 8.75	96.63 ± 9.64
Anterior drawer moment	15.04 ± 4.02	17.83 ± 3.35	17.43 ± 4.00	20.41 ± 3.29	22.00 ± 3.47
Posterior drawer moment	19.88 ± 5.42	13.99 ± 3.63	12.16 ± 3.62	16.93 ± 4.63	16.00 ± 3.08

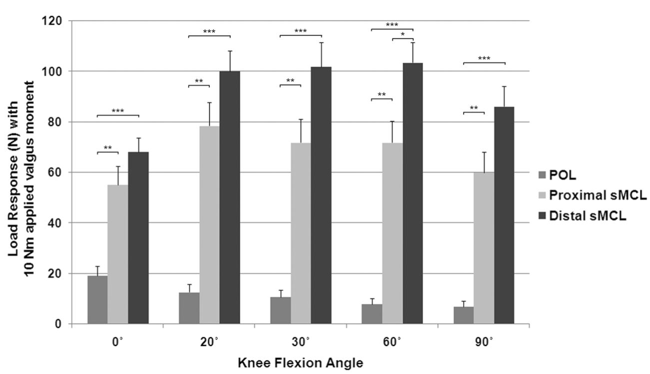

Figure 3. Forces measured on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions during application of a 10 N·m valgus load. *$P < .05$. **$P < .001$. ***$P < .0001$. Error bars indicate the standard error of the mean.

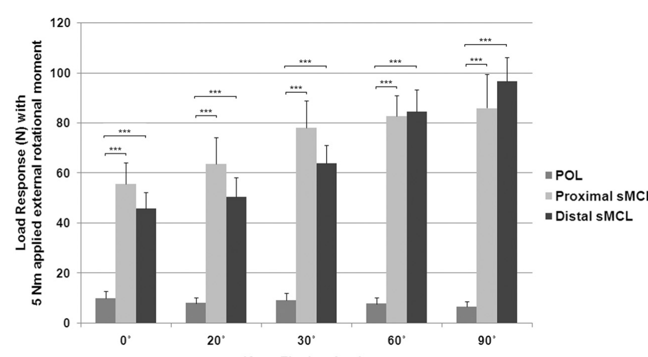

Figure 4. Forces measured on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions during application of a 5 Nm external rotation torque. ***$P < .0001$. Error bars indicate the standard error of the mean.

across all flexion angles ($P < .001$) (Table 1). The largest load response difference between applied external rotation and internal rotation torques on the proximal (86.0 vs 20.2 N) and distal (96.6 vs 22.6 N) sMCL divisions was at 90° of knee flexion ($P < .0001$).

With an 88 N anterior drawer load, there was a maximum load of 22 N on the distal division of the sMCL at 90° of knee flexion. Although there was a load response at all tested flexion angles, there were no significant differences noted between individual flexion angles for either sMCL division. The 88 N posterior drawer loads produced similar load responses with a maximum load of 21.08 N on the proximal division of the sMCL at 90° of knee flexion. As seen with anterior drawer load responses, there were no significant differences in posterior drawer load responses between tested knee flexion angles in the proximal and distal divisions of the sMCL.

Force on the POL

The POL valgus load response at 0° was significantly greater than at 30° ($P < .01$), 60° ($P < .03$), and 90° ($P < .01$) of flexion. The maximum load (19.1 N) on the POL for an applied valgus load was at 0° of knee flexion. At this flexion

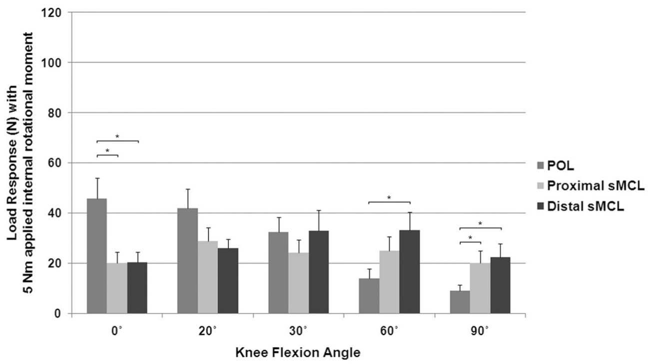

Figure 5. Forces measured on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions during application of a 5 Nm internal rotation torque. *$P < .05$. Error bars indicate the standard error of the mean.

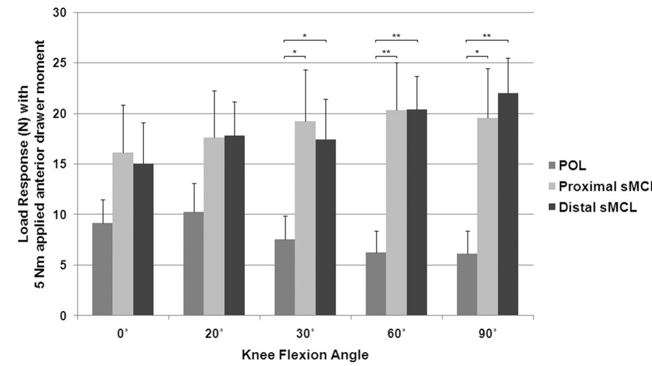

Figure 6. Forces measured on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions during application of an 88 Nm anterior drawer load. *$P < .05$. **$P < .001$. Error bars indicate the standard error of the mean.

angle, the valgus load responses for the POL (19.1 N) were significantly lower than were the load responses from both the proximal sMCL division (55.1 N, $P < .001$) and the distal sMCL division (68.2 N, $P < .0001$) (Figure 3).

The largest POL load response to an applied internal rotation torque was at 0° of knee flexion (45.8 N) (Figure 5). The POL internal rotation torque load response at 0° of flexion (45.8 N) was significantly greater than was the load response at both 60° and 90° of knee flexion (13.9 vs 9.1; $P < .01$). In addition, the load response at 20° of flexion (41.9 N) was significantly greater than was the load response at both 60° and 90° of knee flexion (13.9 N vs 9.1; $P < .01$). There was also a significantly larger internal rotation load response at 30° of flexion (32.4 N) than at 60° (13.9 N; $P < .01$) or 90° (9.1 N; $P < .012$) of knee flexion. When comparing the POL to the sMCL, we noted that at 0° of knee flexion, there was a significantly larger ($P < .05$) load on the POL (45.8 N) during internal rotation compared with the distal (20.5 N) and proximal (20.2 N) sMCL divisions. However, at 90° of knee flexion, the load on the POL (9.1 N) during internal rotation torques was significantly less ($P < .05$) than were the loads on both the proximal (20.2 N) and distal (22.6 N) sMCL divisions. Tukey's HSD test, performed after 1-way independent-sample ANOVA, did not detect a significant difference in POL load responses to an external rotation torque between any of the tested flexion angles. Internal rotation torques produced a significantly higher load on the POL compared to external rotation torques across all tested flexion angles ($P < .001$), except at 60° of knee flexion (Table 1). Furthermore, the largest POL load response difference to applied internal and external rotation torques was seen at 0° (45.8 vs 9.8; $P < .001$).

With an 88 N anterior drawer load, there was a maximum load of 10.3 N at 20° of knee flexion on the POL (Figure 6). Despite the fact that there was an anterior drawer load response on the POL at all tested flexion angles, there were no significant differences between any of the individual flexion angles. At 30° of knee flexion, there was a significantly lower load on the POL (7.54 N) compared with the proximal (19.2 N; $P < .05$) and distal (17.4 N; $P < .01$) divisions of the sMCL. With an 88 N posterior drawer load, there was a maximum load of 6.9 N at 0° of knee flexion (Figure 7). There were no significant differences between any of the tested flexion angles. At 60° of flexion, the POL load response to a posterior drawer load was significantly less than that of the proximal sMCL division (6.83 vs 20.88 N; $P < .05$). Furthermore, the POL load response (5.4 N) to a posterior drawer load was also significantly less than those of both the proximal (21.1 N) and distal (16.0 N) divisions of the sMCL ($P < .05$) at 90° of knee flexion.

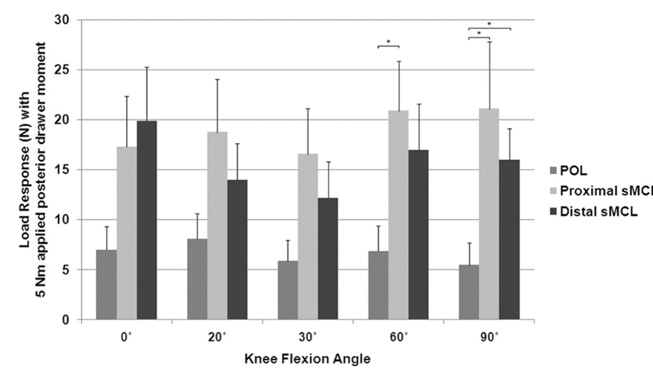

Figure 7. Forces measured on the posterior oblique ligament (POL) and the proximal and distal superficial medial collateral ligament (sMCL) divisions during application of an 88 Nm posterior drawer load. *$P < .05$. Error bars indicate the standard error of the mean.

DISCUSSION

This in vitro experiment provides new insights into the quantitative biomechanical properties of the primary stabilizing ligaments of the medial knee. Although the sequential sectioning method is useful to indirectly examine the function of a structure, it does not provide quantitative information regarding the tensile force experienced by a ligament due to an external force. Conversely, direct force measurement provides information regarding the relationships of specific structures in the native tissue without being dependent on the cutting sequence of structures. In this study, we found that the POL and both sMCL divisions were loaded at all tested knee flexion angles with the application of valgus loads, internal and external rotation torques, and anterior and posterior drawer loads. These results indicated that there is a shared loading response among these structures that was not previously well defined by the sequential sectioning technique.

Prior cutting studies have found that the sMCL is an important restraint to valgus instability of the knee.[1,6,12,24] Via the use of buckle transducers, this study quantitatively demonstrated the load response of both divisions of the sMCL. We found that with an applied valgus load, the load response for the distal division of the sMCL was dependent on the knee flexion angle, whereas the load response for the proximal sMCL division was not significantly different between any of the tested flexion angles. The distal sMCL division also experienced a significantly larger load response to a valgus load than did the proximal sMCL division at 60° of knee flexion (Figure 3). This observation of increased load on the distal sMCL division could be caused by the anatomy of the sMCL. The proximal division of the sMCL had an increased amount of soft tissue adherences,[14] which, in theory, may disperse load among both its capsular adherences as well as its attachment to the crural fascia along its proximal course.[14] In contrast, the distal portion of the sMCL is nearly devoid of any soft tissue adherences, and thus the tensile force was transferred directly to the distal tibial bony attachment.[14] Brantigan and Voshell's article in 1941 noted that the anterior fibers of the sMCL may not have attached to the proximal tibial attachment but rather coursed directly from the femur to the distal tibial attachment. Thus, the differences in force between the 2 sMCL divisions may be owing to both different fiber bundle orientation within the sMCL[2] and the fact that the proximal tibial attachment is primarily to soft tissues rather than anchored directly to bone. However, our buckle transducer enveloped the entire ligament, so that one cannot determine whether the whole width of the sMCL functions as 1 unit or if different portions of the ligament experience different loads. The implications of this observation are that although the sMCL has previously been both biomechanically tested and surgically reconstructed under the assumption that the sMCL is 1 continuous structure,[4,5,12,13,20,22,23,37] the 2 divisions of the sMCL actually function as 2 conjoined but distinct structures. Thus, this study suggests that surgical repair or reconstruction of the sMCL should strive to reconstruct the separate function of both divisions in an attempt to reproduce the overall function of the sMCL construct.

The sMCL has also been reported to have an important role in restraint to both external rotation[32] and internal rotation.[23,31,34] Our study demonstrated a significant increase in load response on the sMCL to applied external rotation torque as the amount of knee flexion increased (Figure 4). Internal rotation load responses in the sMCL were also highest at increased degrees of knee flexion, although there was significantly more load response from both sMCL divisions to external rotation compared to internal rotation (Figure 5 and Table 1). Previous sequential sectioning studies have concluded that the sMCL is a secondary stabilizer in resisting both anterior and posterior tibial translation in anterior and posterior cruciate ligament–deficient knees, respectively.[3,7,21,29] However, it has not been demonstrated to be an anterior tibial translation stabilizer in knees with an intact ACL.[21,30] Although there were no significant differences between flexion angles, the sMCL did produce a load response to anterior and posterior drawer loads at all knee flexion angles. This sMCL load-sharing function against anterior and posterior drawer loads in an intact knee has not been previously described.

In combination with the sMCL, the POL has been previously reported to contribute to stabilization of external rotation, internal rotation, and valgus movement of the tibia.[5,28,35] Previous work has reported that the POL does not have a functional stabilizing role with increased knee flexion and that it is most influential to resisting abnormal knee laxity between 0° and 30° of flexion.[4,5,9,20,22,28] Results from our study support the previous literature in the POL's role near 0° of knee flexion but, for the first time, adds new quantitative data about the distinct importance of the POL in resisting valgus loads, internal and external rotation torques, and anterior and posterior drawer loads (Table 1). At 0° of knee flexion, the POL produced a significantly higher load response to internal rotation torques than did either sMCL division. Interestingly, there was a reciprocal load response to internal rotation torque between the POL and sMCL as the degree of knee flexion increased, with a higher load response on the sMCL at 90° of knee flexion. This observation exhibits a complementary relationship between the POL and the sMCL in resisting internal rotation torque, which has not been previously appreciated. In addition, the POL displayed a significant load response to valgus loads near 0° of knee flexion (Figure 4) but did not exhibit any significant stabilizing role against external rotation when compared to the internal rotation load responses (Table 1). Previous sequential sectioning studies have indirectly demonstrated that the POL has a secondary stabilizing effect against posterior tibial translation in posterior cruciate and medial collateral ligament–deficient knees.[3,25-27] A study by Sullivan et al[33] also reported that the POL did not contribute to increased anterior tibial translation in a knee with an intact ACL. Our results demonstrated for the first time that the POL shares the load response against both anterior and posterior tibial translation in an intact knee.

One of the limitations of this study was that it was a static study of ligaments that have dynamic interactions in vivo. Because this study did not replicate the dynamic stability provided by the semimembranosus muscle and its attachments to the medial knee,[14] it is possible that the magnitude of the forces seen on these structures may be different than in a dynamic in vivo condition. However, we believe that direct measurement of force on both the POL and the sMCL provides useful new information in assessing the role of these structures in knee stability that was not previously appreciated using indirect measurement methods. In addition, it is recognized that one of the limitations of buckle transducers is impingement during loading, which can affect the measured forces on the ligaments.[18] However, we believe that impingement was minimal in this study as indicated by a lack of any abnormal voltage spikes from the buckle transducers output during testing, which would have occurred with an impingement event.

In conclusion, direct force measurements found that (1) the degree of knee flexion affected the load response of the POL and both divisions of the sMCL, (2) there were different load responses for the proximal and distal divisions of the sMCL, and (3) the POL provided significant resistance to valgus and internal rotation forces near knee extension. This information supplements and improves the understanding of previously obtained indirect information from sequential cutting studies regarding the clinical importance of these structures. In combination with a recent study that quantitatively redefined the anatomy of the medial aspect of the knee,[14] this biomechanical study further defines the individualized function of the proximal and distal divisions of the sMCL. This study is clinically significant because it demonstrated that both divisions of the sMCL should be repaired or reconstructed to best reproduce the native function of the sMCL. Furthermore, this study also demonstrated the importance of repairing or reconstructing the POL in isolated or combined medial knee injuries, especially in knees that demonstrate internal rotation and valgus motion instability, to best reproduce the normal load distribution of the medial knee. We believe that the results presented in this study provide clinically valuable information in regards to the load distribution of the primary medial knee structures. These results, in turn, will provide new insight into the necessary interactions of these ligaments through clinically applied loading forces at respective flexion angles. Future studies will be essential to determine the biomechanical effects and load distributions on medial knee structures in the face of medial knee injuries, as seen clinically with varying grades of knee injury.

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