# Distribution of Force in the Medial Collateral Ligament Complex During Simulated Clinical Tests of Knee Stability

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**Background:** Pivot-shift injury commonly results in combined anterior cruciate ligament (ACL)/medial collateral ligament (MCL) injury, yet the contribution of the components of the MCL complex to restraining multiplanar rotatory loads forming critical subcomponents of the pivot shift is not well understood.

**Purpose:** To quantify the role of the MCL complex in restraining multiplanar rotatory loads.

**Study Design:** Controlled laboratory study.

**Methods:** A robotic manipulator was used to apply combined valgus and internal rotation torques in a simplified model of the pivot-shift examination in 12 cadaveric knees (49 ± 11 years). Tibiofemoral kinematics were recorded with the ACL intact. Loads borne by the superficial MCL (sMCL), posterior oblique ligament (POL), deep MCL (dMCL), and ACL were determined via the principle of superposition.

**Results:** The POL bore about 50% of the load carried by the ACL in response to the combined torques at 5° and 15° of flexion. The POL bore load during the internal rotation component of the combined torques, while the sMCL carried load during the valgus and internal rotation phases of the simulated pivot. Load in the dMCL was always <10% of the ACL in response to combined valgus and internal rotation torques.

**Conclusion:** The POL provides complementary load bearing to the ACL near extension in response to combined torques, which capture key components of the pivot-shift examination. The sMCL resists the valgus component of the maneuver alone, a loading pattern unique from those of the POL and ACL. The dMCL is not loaded during clinical tests of rotational knee stability in the ACL-competent knee.

**Clinical Relevance:** Both the sMCL and POL work together with the ACL to resist combined moments, which form key components of the pivot-shift examination.

**Keywords:** pivot shift; posterior oblique ligament; superficial medial collateral ligament; deep medial collateral ligament; anterior cruciate ligament; load; superposition; robot

About 24% of all anterior cruciate ligament (ACL) injuries occur in combination with injury to the medial collateral ligament (MCL).[21] The pivot-shift phenomenon remains a primary mechanism of injury resulting in these concurrent injuries,[2,6,11] and multiple studies have documented the negative correlation between the grade of the pivot shift and clinical outcome scores after ACL injury.[15,18] Therefore, it is important to understand how the MCL and ACL act in concert to resist combined torques, such as the valgus and internal moments applied during the pivot shift.[4]

The secondary stabilizing role of the MCL complex, consisting of the superficial MCL (sMCL), deep MCL (dMCL), and posterior oblique ligament (POL,)[17,25] has been well documented during isolated, uniplanar rotational tibial loading.[8,24,27] The sMCL is the primary valgus stabilizer of the knee particularly at 25° of flexion, accounting for 80% of valgus restraint of the knee.[9] Both the ACL and POL provide secondary restraint against valgus loads with the knee near and at full extension,[8,9] while sectioning the dMCL does not produce increased valgus rotation at any flexion angle.[24] Further, the sMCL and POL are important stabilizers against isolated internal rotational moments, demonstrating reciprocal stabilization as the POL is the greater restraint near extension while the sMCL provides greater restraint in flexion.[8,24] In contrast, the dMCL does not stabilize the knee during internal rotation at any flexion angle.[24]

What is not well understood is how the MCL complex resists combined moments that are known to cause anterolateral tibial subluxation, thereby reproducing a major

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component of the pivot-shift phenomenon.[4] To better understand how the ACL and MCL complex stabilize the knee in the clinical setting, the load borne by these structures in response to not only isolated loads but also combined loading conditions should be characterized. However, these data are not available in the literature.

The purpose of this study was to quantify the load borne by the MCL complex (sMCL, POL, dMCL) as compared with the ACL in response to multiplanar rotatory loads. We asked the following questions: (1) What loads are generated in the MCL complex in response to combined valgus and internal rotation moments simulating important features of the pivot-shift maneuver? and (2) Do combined valgus and internal rotation moments generate differential loading in the structures comprising the MCL complex relative to uniplanar valgus moments?

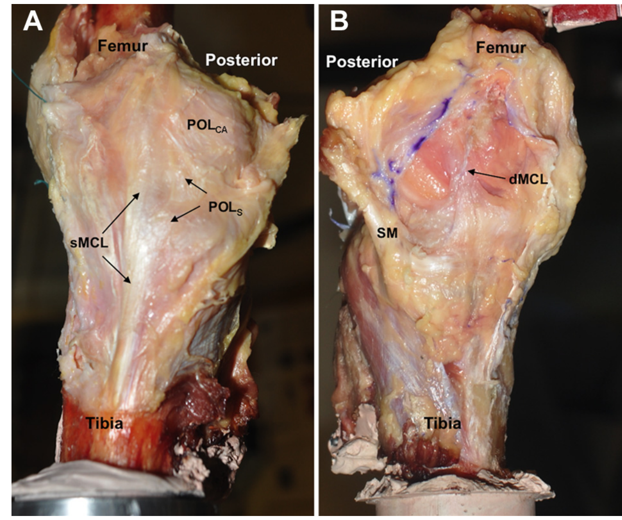


**Figure 1.** (A) Medial aspect of a right knee exposing the superficial medial collateral ligament (sMCL), superficial arm of the posterior oblique ligament ($POL_S$), and capsular arm of the posterior oblique ligament ($POL_{CA}$). (B) Medial aspect of a left knee exposing the deep medial collateral ligament (dMCL) and semimembranosus tendon (SM). The superficial arm of the POL has been dissected and removed.

## METHODS

Twelve fresh-frozen human cadaveric knees were used (7 male; mean age, 48.7 ± 11.1 years; range, 29-65 years). The fresh-frozen knees were thawed 24 hours before testing. Once thawed, soft tissue was stripped from the femur and tibia 12.5 cm proximal and distal to the tibiofemoral joint line. The fibula was secured to the tibia in its anatomic position using a steel screw 5 cm in length, placed about 4 cm distal to the tibiofibular joint. Subsequently, each bone shaft was embedded within 5-cm-diameter aluminum cylinders using bonding cement (Bondo; 3M). Before embedding each bone in the bonding cement, 2 carpenter screws were drilled orthogonally through the femoral and tibial shafts to improve fixation between bone and cement.

The potted femoral shaft was then rigidly locked to a fixture that was secured to the floor. The potted tibial shaft was rigidly attached to a fixture mounted to a universal force-moment sensor (resolution: $F_x = F_y = F_z = 0.25$ N; $T_x = T_y = 0.05$ N·m; $T_z = 0.025$ N·m) (Theta; ATI Inc) that measured forces acting across the knee joint. This universal force-moment sensor was secured to the end effector of a 6-degree-of-freedom robotic arm (ZX165U; Kawasaki) with ±0.3-mm repeatability. Each specimen was aligned in full extension, and anatomic landmarks were identified using a 3D digitizer (MicroScribe; Immersion) with manufacturer-reported accuracy of 0.23 mm. These anatomic landmarks were used to define reference frames describing tibia motion relative to the fixed femur. Tibia rotations and translations were expressed by adapting the convention described by Grood and Suntay[10] as detailed previously.[11]

Force feedback algorithms were used to identify the passive flexion path of each intact knee from full extension to 90° of flexion.[11,28] The knee was then preconditioned by determining the motion required to achieve 134 N anterior load at 30° of flexion and repeating this motion for 10 cycles. Then, the kinematic trajectory required to apply the combined moments at 15° of flexion was determined, and this motion was also repeated for 10 cycles. Knees were then flexed to 5°, 15°, and 30°, and tibiofemoral kinematics were recorded at each angle as an isolated 8-N·m valgus moment, and combined 8-N·m valgus and 4-N·m internal rotation moments were applied to the knee. Similar to the clinical pivot shift, these loads generate anterolateral subluxation in the ACL-deficient knee,[4,13,14] even though the knee was not simultaneously flexed and extended as in the actual examination.[16]

After the kinematic trajectory of the tibia relative to the femur that achieved the defined loading conditions was identified, the sMCL, dMCL, and POL were serially dissected according to detailed anatomic descriptions (Figure 1).[17] Each structure was removed from proximal to distal via careful dissection with a No. 10 blade, fine dissection scissors,

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TABLE 1
Average Ligament Loads With Alternate Sectioning[a]

| | POL Sectioned First | sMCL Sectioned First | Difference in Load: Mean (95% CI) | Difference in Load: *P* Value |
|---|---|---|---|---|
| POL loads | | | | |
| Flexion angle, deg | | | | |
| 5 | 31.1 ± 19.0 | 27.7 ± 16.1 | 3.4 (–26.1 to 19.3) | .75 |
| 15 | 28.6 ± 18.3 | 31.6 ± 18.1 | 3.0 (–20.4 to 26.4) | .78 |
| 30 | 20.9 ± 18.5 | 19.5 ± 13.0 | 1.4 (–22.0 to 19.1) | .88 |
| | sMCL Sectioned First | POL Sectioned First | Mean (95% CI) | *P* Value |
| sMCL loads | | | | |
| Flexion angle, deg | | | | |
| 5 | 39.7 ± 10.3 | 38.1 ± 21.0 | 1.6 (–19.7 to 22.9) | .87 |
| 15 | 50.8 ± 12.6 | 42.8 ± 20.9 | 8.0 (–14.2 to 30.2) | .44 |
| 30 | 61.1 ± 18.7 | 47.7 ± 20.4 | 13.4 (–11.7 to 38.6) | .26 |

[a]Average in situ loads (in newtons) in the posterior oblique ligament (POL) and superficial collateral ligament (sMCL) in response to combined 8-N·m valgus and 4-N·m internal rotation moments at 5°, 15°, and 30° of flexion. Average load measured when the POL was sectioned first (6 specimens) and when the sMCL was sectioned first (6 specimens).

and forceps. The sMCL was identified proximally by its stout ligamentous fibers attaching immediately posterior and proximal to the medial epicondyle and distally by its bony attachment deep to the pes anserinus. The posterior margin of the sMCL was distinguished from the POL based on fiber orientation. The POL was identified proximally by its femoral attachment just anterodistal to the gastrocnemius tubercle, and the fibers of the central and superficial arms were tracked to their distal attachments at the medial meniscus, posteromedial tibia, semimembranosus tendon, and more distal tibia just posterior to the tibial attachment of the sMCL. The meniscofemoral and meniscotibial portions of the dMCL were identified as the capsular thickenings deep to the sMCL, with similar anterior-posterior dimensions to the sMCL. The dMCL was distinguished from the POL at its posterior border by tracking fiber orientation from its proximal to distal attachments (Figure 1).

Immediately before and after sectioning each structure, the previously recorded kinematics were repeated, and the resulting load across the knee was measured. The resultant of the vectoral difference in load before and after sectioning each ligament yielded the net load in each structure using the principle of superposition.[28] Like previous studies, we alternated sectioning order of the sMCL and POL (each sectioned first 6 times) to control for bias caused by potential physical interaction of the adjacent soft tissues.[1,7,26] The dMCL was always the last MCL structure sectioned since it is deep to both the sMCL and the anterior fibers of the POL. Load data for the dMCL were obtained in 10 knees. After removal of the MCL complex, the ACL was sectioned.

To assess the presence of physical interaction, the effect of sectioning order on POL and sMCL loads was compared in response to the combined torques. Any differences in load in the sMCL and POL based on sectioning order indicate the presence of physical interaction. This difference quantifies the extent of the physical interaction between

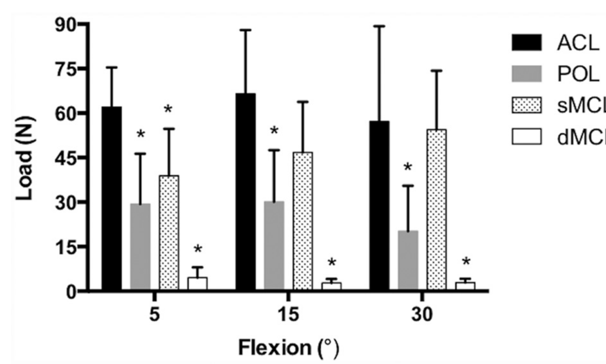


**Figure 2.** In situ forces in the anterior cruciate ligament (ACL), posterior oblique ligament (POL), superficial medial collateral ligament (sMCL), and deep medial collateral ligament (dMCL) in response to combined 8-N·m valgus and 4-N·m internal rotation moments. Whiskers indicate 1 SD. *Significant difference ($P < .05$) in ligament load compared with load in the ACL.

the sMCL and POL. Load in the POL and sMCL was compared across the subgroups of 6 knees using paired *t* tests ($P < .05$). Load in each ligament was compared at each flexion angle in response to the simulated pivot shift using repeated-measures analysis of variance with Tukey post hoc test ($P < .05$). Load borne by each ligament between a pure valgus moment and combined valgus and internal rotation moments was compared at each flexion angle using paired *t* tests ($P < .05$).

## RESULTS

When combined torques were applied at 5°, 15°, and 30° of flexion, the average load in the POL differed by <3.4 N

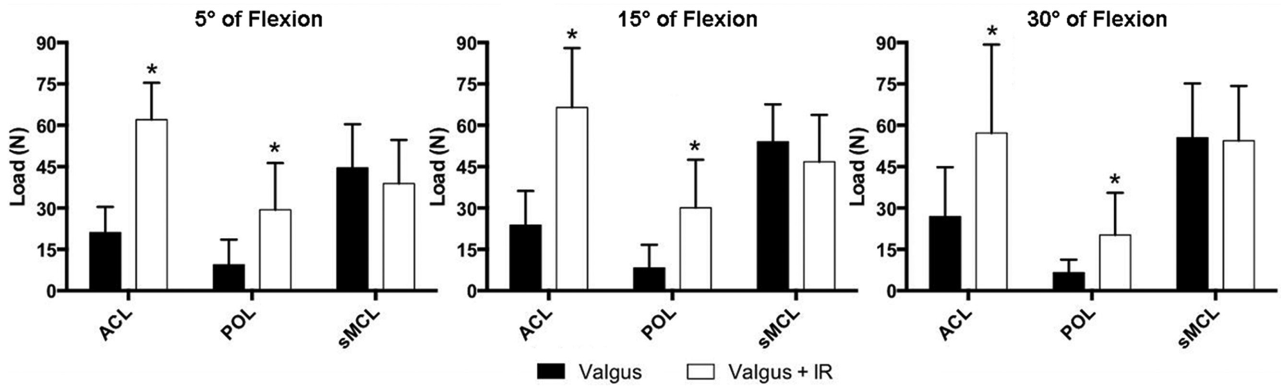


**Figure 3.** In situ forces in the anterior cruciate ligament (ACL), posterior oblique ligament (POL), and superficial medial collateral ligament (sMCL) in response to an isolated 8-N·m valgus moment and in response to combined 8-N·m valgus and 4-N·m internal rotation (IR) moments at 5°, 15°, and 30° of flexion. Whiskers indicate 1 SD. *Significant difference ($P < .05$) relative to the isolated valgus moment.

whether the POL or sMCL was sectioned first (Table 1, top). During the same testing conditions at 5°, 15°, and 30°, average loads in the sMCL differed based on sectioning order by <13.5 N (Table 1, bottom).

At 5° of flexion, the load borne by the ACL in response to combined moments simulating the pivot shift was 62.1 ± 13.3 N (Figure 2). Load in the POL was 47% of the ACL ($P < .001$), load in the sMCL was 63% of the ACL ($P = .003$), and load in the dMCL was 7% of the ACL ($P < .001$). At 15° of flexion, the ACL bore 66.5 ± 21.5 N of load, the POL carried 45% of the load in the ACL ($P < .001$), the sMCL carried 70.4% of the load in the ACL, and the dMCL carried 4% of the load in the ACL ($P < .001$). At 30° of flexion, the load in the ACL was 57.2 ± 32.1 N. Load in the POL was 35% of the ACL ($P = .002$), while the load in the sMCL was 95% of the ACL. Lastly, loads in the dMCL were 5% of the ACL ($P < .001$).

At 5° and 15° of flexion, adding an internal rotation moment to the isolated valgus moment caused a respective increase in ACL load from 21.1 ± 9.3 to 62.1 ± 13.3 N and from 23.8 ± 12.4 to 66.5 ± 21.5 N ($P < .001$) (Figure 3). This represented an increase in ACL load of 194% and 179% at 5° and 15°, respectively. Adding an internal rotation moment at 30° caused a 112% increase in ACL load from 26.9 ± 17.9 to 57.2 ± 32.1 N ($P = .006$). At 5° of flexion, the load experienced by the POL was 212% higher in response to combined moments reaching 29.4 ± 16.9 N compared with an isolated valgus moment (9.4 ± 9.2 N) ($P < .001$) (Figure 3). Similarly, adding an internal rotation moment at 15° of flexion caused a 263% increase in load on the POL from 8.3 ± 8.4 to 30.1 ± 17.4 N ($P < .001$). At 30° of flexion, loads in the POL were small when an isolated valgus moment was applied (6.6 ± 4.7 N) and increased by 206% after adding the internal rotation moment (20.2 ± 15.3 N) ($P = .005$). No increases in loads in either the sMCL or the dMCL were detected in response to adding the internal rotation moment at 5° or 15° or in the sMCL at 30° (Figure 3). Loads borne by the dMCL increased minimally from 2.0 to 2.9 N at 30° ($P = .034$).

## DISCUSSION

The main findings of this study in regard to the POL were that it (1) bears up to 47% of the force borne by the ACL in resisting the internal rotation aspect of combined rotatory loads comprising critical features of the clinical pivot-shift examination, (2) played a greater role in resisting these loads in extension, and (3) has a minimal role in resisting isolated valgus loads. The sMCL played a consistent role in resisting both the isolated valgus moment and combined valgus and internal rotation moments. Finally, we identified minimal physical interaction between the sMCL and POL in response to combined torques as the difference in loads due to sectioning order was always at least 4 times less than the magnitude of average ligament load.

Our findings indicate that the ACL, sMCL, and POL act in a coordinated, flexion-dependent fashion to resist combined moments capturing key aspects of the pivot shift, while the dMCL plays a minimal role. At 5° of flexion under combined moments, the ACL carried the most load, while both the sMCL and POL were secondary contributors. At 15° near the classic pivot shift position of flexion,[5] the loads borne by the ACL remained the largest, and sMCL and POL loads were secondary load-bearing structures. At 30° of flexion, average loads in both the ACL and POL decreased relative to extension, while sMCL load increased. These results demonstrate that, in concert with the ACL, the sMCL and POL are important stabilizers against combined moments near full extension, whereas the sMCL and ACL are the principal stabilizers at 30° of flexion.

The ACL and POL act cooperatively, especially near extension, to primarily resist the axial rotation component of the moments simulating the pivot shift. Specifically, loads in the ACL and POL nearly tripled between 5° and 15° of flexion after combining a 4-N·m internal rotation moment with the existing 8-N·m valgus moment. Although it is well known that the ACL is a primary stabilizer of the lateral compartment in response to loads simulating the pivot shift,[3,4] this finding suggests that the POL acts to restrain the medial compartment.

Although loading patterns in the ACL and POL were similar in resisting the internal rotation moment of the combined valgus and internal rotation moments, the POL carried at most 47% of the load measured in the ACL. Thus, the POL is an important secondary restraint near extension in response to loads known to generate pivoting events. At 30°, the POL provides less restraint, indicating that the more posterior located POL fibers likely slacken with flexion.

There was about 50% variation in the loads experienced by the POL during the simulated pivot (29.4 ± 16.9 N at 5°, 30.1 ± 17.4 at 15°, 20.2 ± 15.3 at 30°), indicating that soft tissue–stabilizing mechanisms vary from knee to knee. That is, in some knees, the POL resists applied moments more strongly than in others. This may be due to variations in the collagen composition and resulting material properties of this capsular structure. Differentiating which patients rely more on their POL for rotatory stability could help to customize surgical treatment of ACL injuries and may help further guide clinicians in understanding who may benefit most from reconstruction of secondary restraints to rotatory loads. Moreover, the present findings do not indicate whether the POL can protect the ACL from increased loads in response to the combined torques. Future studies exploring ACL loads in the absence of the POL are needed.

The sMCL plays an important stabilizing role to resist valgus while the tibia is being internally rotated during application of the combined torques. Thus, the loading patterns of the sMCL were unique from those of the POL and ACL. Specifically, the sMCL did not experience increased loads when an internal rotation moment was added to the isolated valgus moment at all tested flexion angles; rather, they stayed relatively consistent. These differential roles of the sMCL and POL are likely explained by the alignment of each structure. The POL fibers have a more horizontal alignment and are better oriented to resist rotational loads compared with the vertically aligned sMCL fibers.

In isolated valgus loading, our results confirm previous studies that identified the sMCL as the primary restraint to valgus rotation and the ACL as a secondary restraint.[9] In contrast, the POL carried less load than the sMCL, which is in agreement with another study of the sMCL and POL during isolated valgus loading.[8] Our findings also agree with prior work describing the POL as a significant restraint to valgus near extension but not deeper in flexion as fibers slacken.[24] Our work is also consistent with a prior study documenting that internal rotation torques generate the greatest POL loads near extension.[8] In addition, we found that the dMCL bore minimal loads under valgus loads and during combined pivot loads. Similarly, after isolated sectioning of the dMCL, Robinson et al[24] applied isolated valgus and internal rotation moments and did not observe changes in either valgus or internal rotational laxity. Throughout our testing, loads in the dMCL were always the smallest among the 3 structures and were consistently minimal (<10 N).

Lastly, we controlled for potential physical interaction across the POL and sMCL by alternating sectioning order. Any differences in load between sectioning order indicate the presence of physical interaction. Sectioning in the same order could introduce bias since the presence of physical interaction would transfer load to the remaining unsectioned structure. However, one can account for physical interaction of adjacent soft tissues by alternating sectioning order during testing, as demonstrated in several previous studies.[1,7,26] Our findings indicate limited physical interaction between the POL and sMCL to the extent that they do not influence the conclusion of our study, as ligament loads during the application of combined torques were at least 4 times that of any load sharing. Aside from 2 measures, the average sMCL and POL loads differed at most by 3.4 N. Inspection of the data revealed that 2 knees had larger sMCL loads, which drove the average higher for comparisons at 15° and 30°. This may indicate that physical interaction can vary from knee to knee. Since physical interaction likely occurs transversely across these 2 stabilizers, our finding of limited physical interaction is not surprising given the relatively small transverse modulus of the medial collateral tissues, which is about 1/30th of that in the longitudinal direction of the sMCL fibers.[23]

This study has limitations including use of simplified loads and boundary conditions in our 2-torque model of the pivot shift. The pivot shift is performed clinically through a range of flexion and involves a complex combination of multiplanar loads.[19,22] We applied loads known to create anterolateral subluxation of the tibia,[4,12] a key characteristic of this maneuver. In addition, we did not measure the contributions of each structure to the stability of the knee because the number of specimens required for randomized testing of the MCL complex and ACL would be prohibitive.

In summary, our findings indicate that POL force parallels that generated in the ACL in response to combined multiplanar torques, although with lesser magnitudes. Given its location and orientation, the POL may act to stabilize the medial compartment, which complements the important function of the ACL in stabilizing the lateral compartment in response to the combined torques.[12,20] The sMCL plays an important stabilizing role during the valgus and internal rotation torques by resisting the valgus component of the maneuver but does not experience increased loading as the knee is internally rotated. The dMCL carries minimal force throughout application of the combined torques. These data quantify the complementary role of passive knee stabilizers beyond the ACL in resisting loads simulating important components of the pivot shift. Further work is needed to substantiate the influence of the POL in protecting the ACL or an ACL graft from increased force in response to multiplanar rotatory loads.

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