Winner of the 2009 Excellence in Research Award

# An In Vitro Analysis of an Anatomical Medial Knee Reconstruction


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**Background:** An anatomical medial knee reconstruction has not been described in the literature.

**Hypothesis:** Knee stability and ligamentous load distribution would be restored to the native state with an anatomical medial knee reconstruction.

**Study Design:** Controlled laboratory study.

**Methods:** Ten nonpaired cadaveric knees were tested in the intact, superficial medial collateral ligament and posterior oblique ligament–sectioned, and anatomically reconstructed states. Each knee was tested at 0°, 20°, 30°, 60°, and 90° of knee flexion with a 10-N·m valgus load, 5-N·m external and internal rotation torques, and 88-N anterior and posterior drawer loads. A 6 degrees of freedom electromagnetic motion tracking system measured angulation and displacement changes of the tibia with respect to the femur. Buckle transducers measured the loads on the intact and reconstructed proximal and distal divisions of the superficial medial collateral ligament and the posterior oblique ligament.

**Results:** A significant increase was found in valgus angulation and external rotation after sectioning the medial knee structures at all tested knee flexion angles. This was restored after an anatomical medial knee reconstruction. The authors also found a significant increase in internal rotation at 0°, 20°, 30°, and 60° of knee flexion after sectioning the medial knee structures, which was restored after the reconstruction. A significant increase in anterior translation was observed after sectioning the medial knee structures at 20°, 30°, 60°, and 90° of knee flexion. This increase in anterior translation was restored following the reconstruction at 20° and 30° of knee flexion, but was not restored at 60° and 90°. A small, but significant, increase in posterior translation was found after sectioning the medial knee structures at 0° and 30° of knee flexion, but this was not restored after the reconstruction. Overall, there were no clinically important differences in observed load on the ligaments when comparing the intact with the reconstructed states for valgus, external and internal rotation, and anterior and posterior drawer loads.

**Conclusion:** An anatomical medial knee reconstruction restores near-normal stability to a knee with a complete superficial medial collateral ligament and posterior oblique ligament injury, while avoiding overconstraint of the reconstructed ligament grafts.

**Clinical Significance:** This anatomical medial knee reconstruction technique provides native stability and ligament load distribution in patients with chronic or severe acute medial knee injuries.



**Keywords:** superficial medial collateral ligament; posterior oblique ligament; motion tracking system; buckle transducers; medial knee reconstruction


The fact that the superficial medial collateral ligament (sMCL) is a primary static stabilizer preventing valgus translation and assists in restraining external rotation and internal rotation about the knee is well documented in the orthopaedic literature.[9,31] The posterior oblique ligament (POL) has also been reported to be an important primary restraint to internal rotation and a secondary restraint to valgus translation and external rotation.[7,8,10,29,32] A high frequency of combined sMCL and


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Presented at the 35th annual meeting of the AOSSM, Keystone, Colorado, July 2009.

No potential conflict of interest declared.

The American Journal of Sports Medicine, Vol. 38, No. 2
DOI: 10.1177/0363546509347996

POL injuries have been reported in knees with acute or chronic valgus laxity, signifying the important role of the POL in providing static stabilization to the medial knee.[12,13] Furthermore, severe acute midsubstance isolated or combined sMCL or POL injuries initially treated nonoperatively have been reported to result in functional limitations and osteoarthritis.[17] Therefore, surgical treatment of severe injuries may be necessary in some circumstances to prevent the pathologic changes associated with chronic medial knee instability.

Historic treatment of acute medial collateral ligament injuries has focused on nonoperative therapies with early controlled motion and fairly good patient outcomes have been reported.[5,15,16,28] However, more severe acute and symptomatic chronic medial knee injuries may require operative management. Currently described techniques include direct repair of the medial structures,[13] primary repair with augmentation,[6] advancement of the sMCL tibial insertion site,[27] pes anserinus transfer,[30] advancement with pes anserinus transfer,[25] and nonbiomechanically validated reconstruction techniques.[3,33]

To our knowledge, no biomechanically validated anatomical reconstruction technique using quantitatively described anatomical attachment sites[18] to reconstruct sMCL and POL injuries has been reported. An anatomical reconstruction technique is preferable because anatomical reconstructions in other major ligaments have been demonstrated to better approximate normal knee biomechanics.[4,11,19] Our hypotheses were that an anatomical medial knee reconstruction would restore static stability and re-create the native forces experienced by each ligament to knees with grade III medial knee injuries. Therefore, the purpose of this study was to describe and biomechanically validate an anatomical reconstruction of the sMCL and POL.

## MATERIALS AND METHODS

### Specimen Preparation

A total of 10 nonpaired, fresh-frozen cadaveric knees, with an average age of 61.7 years (range, 43-81 years) and no evidence of prior injury or disease were used for this study. Knees were stored at −20°C and thawed overnight before dissection and subsequent biomechanical testing.

The skin and subcutaneous tissues were then removed from the specimen. The femur and tibia were severed 20 cm proximal and 13 cm distal to the knee joint, respectively. The tibial marrow cavity was reamed and a threaded fiberglass rod was cemented into the tibial marrow cavity, parallel with the long axis of the tibia. A hexagonal nut with an eye screw was attached to the threaded tibial rod 23 cm distal to the joint line for the application of valgus torques. A lock nut was attached at the distal end of the tibial rod to allow for the application of internal and external rotation torques. The specimen was mounted into a previously described customized knee testing apparatus[7,8,32] that firmly secured the femur at a horizontal angle and allowed uninhibited movement of the tibia. Buckle transducers were fastened to the POL, the proximal division of the sMCL, and the distal division of the sMCL similar to our previously described technique.[8,23,32] During testing of the reconstructed state, buckle transducers were applied to these same reconstructed ligament grafts.

### External Force Application

Each knee was tested at 0°, 20°, 30°, 60°, and 90° of knee flexion. The following external forces were applied at each flexion angle and testing state: 10-N·m valgus load, 5-N·m internal and external rotation torques, and 88-N anterior and posterior drawer loads. A Model SM S-type load cell (Interface, Scottsdale, Arizona), with a manufacturer-reported nonrepeatability error of ±0.01%, was used to apply valgus loads and anterior/posterior drawer loads. A Model TS12 shaft-style reaction torque transducer (Interface), with a manufacturer-reported nonrepeatability error of ±0.02%, was used to apply internal and external rotation torques. Load values were simultaneously recorded in a synchronized manner for the motion data and buckle transducer data through Motion Monitor software (Innovative Sport Training, Chicago, Illinois).

### Angulation and Displacement Measurements

Three different testing conditions were utilized involving the intact, sectioned medial knee structures (the sMCL and POL), and reconstructed medial knee states. Quantification of motion of the tibia relative to the static femur was performed using the Polhemus Liberty system (Polhemus Inc, Colchester, Vermont). Prior studies have validated the accuracy and use of alternating current 6 degrees of freedom electromagnetic tracking systems for monitoring spatial rigid body motion corresponding to human joint and limb motion.[1,2] The manufacturer-reported accuracy of this alternating current tracking device has been reported to be within 0.15° and 0.76 mm. Three receivers were bicortically attached to the specimen using threaded Kirschner wires, with 1 at the anterior midfemur and 2 on the anterior tibial crest distal to the tibial tubercle. The most medial, lateral, and anterior landmarks of the articular cartilage margins of the tibial plateau were marked with the Polhemus digitizing stylus and tracked relative to the tibial sensors to monitor movement of these landmarks throughout testing. Before any load application, the neutral position of each specimen in relationship to the fixed receivers was established for each testing angle to provide a zero reference for all future measurements of displacement or angulation. During load application, Motion Monitor software utilized the Polhemus system to detect positional changes of the tibia relative to the static femoral receiver.

### Observed Ligament Load via Buckle Transducers

Buckle transducers were used to measure the relative loads experienced in the POL and the proximal and distal divisions of the sMCL for the intact and reconstructed states after externally applied loads. The use of these devices has been previously described in detail.[8,22-24] Prior work has reported that there is a linear relationship between applied load and voltage output.[24] Therefore, the voltage outputs were converted to kilonewtons per volt according to a conversion factor determined by posttest calibration of each buckle with a known load. Buckles were zeroed before each external load application to ensure accurate load measurements. Buckle transducers have been reported to be repeatable to within 0.7% using a similar biomechanical testing protocol.[23]

## Data Point Analysis

The raw data for both the change in angulation as recorded through the Polhemus device and the ligament tension data recorded from the buckle transducers were collected by Motion Monitor software and were processed through their proprietary software. The data set from each test was imported into MATLAB R2006b analysis software (MathWorks Inc, Natick, Massachusetts) for further processing.

## Statistical Data Analysis

Software used for the statistical analysis of both the angulation and displacement data and the buckle transducer data was the R statistical computing language, version 2.8 (R Foundation for Statistical Computing, Vienna, Austria). At each applied force (external rotation, internal rotation, anterior drawer, posterior drawer, and valgus load) and knee flexion angle (0°, 20°, 30°, 60°, and 90°), we performed a 2-way analysis of variance for the model: A or D = Specimen by State, where A was angulation (in degrees), D was displacement (in mm), and State was the cut state (intact, sectioned, and reconstructed). The interaction Specimen × State was used as the error term. For the anterior drawer, posterior drawer, and valgus load data, the analysis was done on a log(10) transform of the measurements to normalize the distributions. We compared the means of the State post hoc with the Tukey honest significant difference.

Additionally, paired *t* tests were used to detect differences between the intact and reconstructed states at varying degrees of knee flexion for ligament load data using buckle transducers. In both Tukey and paired *t* test comparisons, significant differences were assumed for $P < .05$.

## Medial Knee Reconstruction Technique

A total of 11 fresh-frozen cadaveric knees were initially used to explore surgical options and refine our final surgical technique. As noted previously, an additional 10 cadaveric knees were used to biomechanically validate the final construct.

This anatomical technique consists of a reconstruction of the sMCL and POL using 2 separate grafts with 4 reconstruction tunnels. The quantitative anatomy of the medial knee has previously been described in detail.[18]

An anteromedial incision was made along the medial knee and located proximally between the medial border of the patella and the medial epicondyle. The distal end of this incision was located directly over the pes anserine tendons. Blunt dissection was performed to expose the femoral anatomical attachment points of the sMCL and POL.[18] The sMCL and POL femoral attachment sites were identified, and the soft tissues overlying the anatomical attachment points were carefully reflected by sharp dissection.

The anterior border of the distal expansion of the sartorius muscle fascia was now incised using sharp dissection, and the gracilis and semitendinosus tendons were exposed. The semitendinosus was then harvested using a hamstring stripper and sectioned into 2 parts—1 measuring 16 cm for subsequent sMCL reconstruction and the other, 12 cm for subsequent POL reconstruction. Each portion of the tendon was tubularized on both ends using No. 2 nonabsorbable sutures to fit into 7-mm tunnels. Alternatively, allograft tendon may be used for in vivo medial knee reconstructions.

Attention was returned to the distal tibial attachment of the sMCL, approximately 6 cm distal to the joint line.[18] The tibial insertion, or any remaining remnants, of the sMCL was carefully reflected off its attachment on the anteromedial proximal tibia by sharp dissection.

The tibial attachment of the POL was now identified. To protect the sartorial branch of the saphenous nerve, which usually courses posterior to the sartorius muscle belly and tendon at this level,[14] the fascia anterior to the sartorius muscle tendon was incised and the sartorius tendon was retracted distally. At this point, the attachment site of the central arm of the POL was identified at the posteromedial tibia near the direct arm of the semimembranosus tendon.[18] To expose this attachment point, a small incision was made parallel to the fibers along the posterior edge of the anterior arm of the semimembranosus tendon, in the path of the central arm of the POL. The bone was sharply cleared of soft tissues, which exposed a small bony ridge, located at the midpoint of the central arm of the POL.[18]

After the attachment locations of the sMCL and POL were isolated, attention was returned to drilling the reconstruction tunnels. An eyelet pin was drilled through the center of the femoral attachment of the sMCL anterolaterally across the femur. A 7-mm reamer was then reamed to a depth of 30 mm over the eyelet pin, and tapped with a 7-mm bioabsorbable screw tap. The 16-cm section of semitendinosus tendon that was previously tubularized was then passed into the tunnel using an eyelet pin, and was recessed 25 mm into the tunnel. Distal traction was placed on the graft along the distal course of the native sMCL and a 7-mm cannulated bioabsorbable screw was placed at the proximal aperture of the tunnel to secure the graft in place. This fixation was qualitatively verified for its fixation strength by placing medial traction on the graft.

A similar technique was then used to position the femoral POL graft reconstruction tunnel. An eyelet pin was drilled anterolaterally across the femur through the center of the femoral POL attachment point.[18] A 7-mm reamer was reamed to a depth of 30 mm and tapped with a 7-mm bioabsorbable screw tap. The previously tubularized 12-cm section of semitendinosus tendon was then recessed into the reconstruction tunnel. Manual traction was placed on the graft in the direction of the native POL and a 7-mm bioabsorbable screw was placed at the proximal aperture of the tunnel to secure the POL reconstruction graft in place. The femoral fixation was also qualitatively evaluated with manual medial traction.

The tibial fixation tunnels for the distal sMCL and POL anatomical attachment points were reamed next. The distal sMCL tibial tunnel was reamed first, and was started by drilling an eyelet pin anterolaterally, through the center of the distal sMCL anatomical attachment point, which exited along the proximal anterolateral lateral compartment of the leg. This tunnel was located 6 cm distal to the joint line, which reproduced the known anatomy of the distal sMCL tibial attachment.[18] It was imperative to ensure that the distal sMCL tibial tunnel was placed along the posterior edge of the distal sMCL footprint because grafts placed too anterior tended to result in overtightening in higher flexion angles and failure of the construct in the pilot studies. A 7-mm reamer was then reamed to a depth of 30 mm and tapped using a 7-mm bioabsorbable screw tap. Finally, the distal edge of this tunnel was notched to maintain screw positioning when securing the bioabsorbable screw to fix the graft.

Next, an eyelet pin was drilled anterolaterally through the center of the tibial attachment of the central arm of the POL, which exited just distal and medial to Gerdy's tubercle. A 7-mm reamer was reamed to a depth of 30 mm and tapped with a 7-mm bioabsorbable screw tap. The sMCL graft was then passed laterally through the distal sMCL tunnel, and recessed to a depth of 25 mm. The knee was placed in 30° of knee flexion, in neutral rotation, and a manual varus force was applied to reduce any gapping of the medial compartment. The sMCL reconstruction graft was then tensioned by placing a manual lateral traction force to tighten the graft into the tibial tunnel via the No. 2 nonabsorbable suture, and secured in place with a 7-mm bioabsorbable screw at the distal aperture of the tunnel. The knee was then placed through a full passive range of motion to verify proper positioning of the sMCL graft. Once proper positioning was verified, the proximal tibial attachment point of the sMCL, which was primarily to soft tissues and located just distal to the joint line,[18] was recreated by suturing the sMCL graft to the anterior arm of the semimembranosus muscle.

Finally, the POL graft was secured into its tunnels. The POL graft was passed into the tibial tunnel and recessed to a depth of 25 mm. The knee was held in both extension and neutral rotation with a varus force applied to reduce any medial compartment gapping. The graft was manually tensioned by placing an anterolateral traction force on the No. 2 nonabsorbable suture, and secured into position with a 7-mm bioabsorbable screw placed at the distal aperture of the tunnel. The final reconstruction (Figure 1B) was then mounted into the knee testing apparatus for further biomechanical testing.

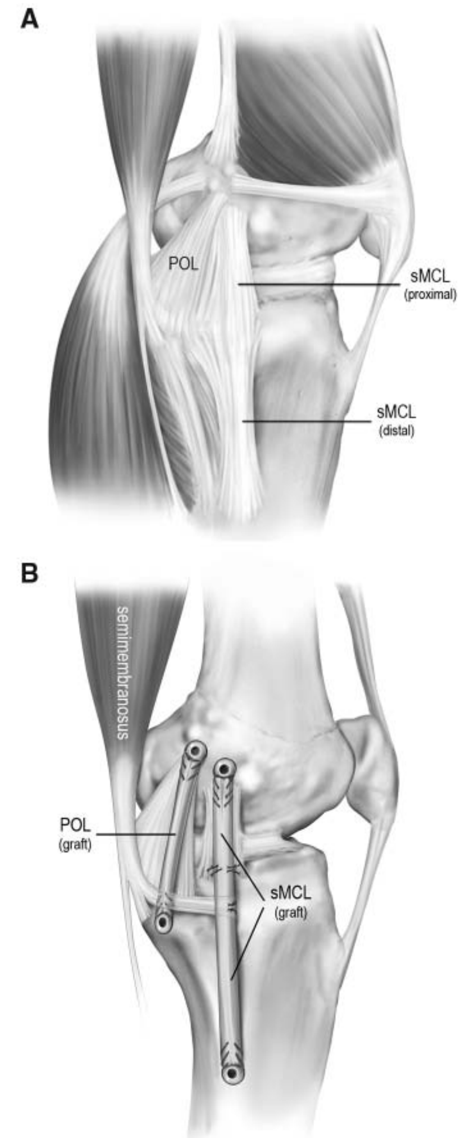


**Figure 1.** A, the superficial medial collateral ligament (sMCL) (medial aspect, left knee), and posterior oblique ligament (POL). Reprinted with permission from LaPrade RF, Engebretsen AH, Ly TV, Johansen S, Wentorf FA, Engebretsen L. The anatomy of the medial part of the knee. *J Bone Joint Surg Am*. 2007;89(9):2000-2010. B, a left medial knee reconstruction procedure demonstrating the reconstructed sMCL and POL. Note that the proximal tibial attachment point of the sMCL, which was primarily to soft tissues and located just distal to the joint line, was recreated by suturing the sMCL graft to the anterior arm of the semimembranosus muscle.

## RESULTS

### Angulation and Displacement of the Knee Joint (Motion Tracking System)

The biomechanical results for our 10 specimens are presented. No graft fixation problems or graft slippage were noted throughout testing.

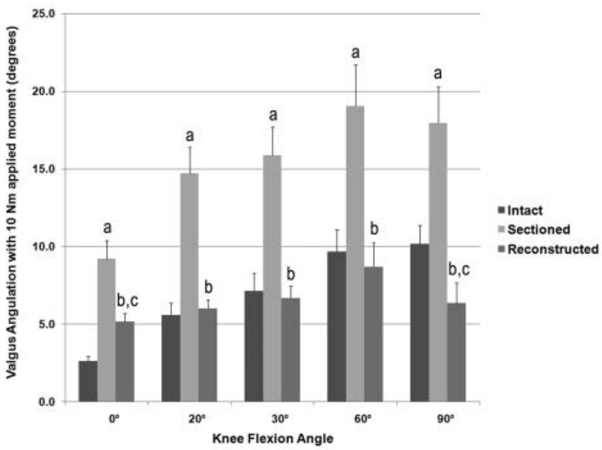


**Figure 2.** Change in valgus angulation with an applied 10-N·m load for intact, sectioned, and reconstructed medial knee structures at each flexion angle. Statistical significance is denoted in the figures as sectioned significantly different from intact (a), reconstructed significantly different from sectioned (b), and reconstructed significantly different from intact (c).

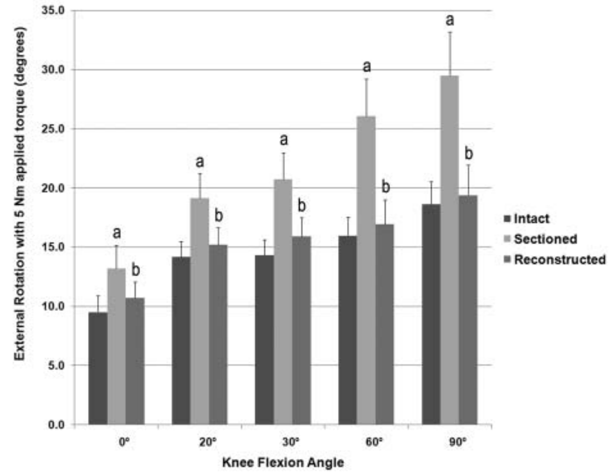


**Figure 3.** Change in external rotation with an applied 5-N·m torque for intact, sectioned, and reconstructed medial knee structures at each flexion angle. Statistical significance is denoted in the figures as sectioned significantly different from intact (a), reconstructed significantly different from sectioned (b), and reconstructed significantly different from intact (c).

*Valgus Angulation.* With an applied valgus load we found a significant increase in valgus rotation after sectioning the medial knee structures at 0° ($P < .0001$), 20° ($P < .0001$), 30° ($P < .0001$), 60° ($P < .001$), and 90° ($P < .04$) of knee flexion (Figure 2). In addition, we found a significant decrease in valgus angulation when comparing the reconstructed medial knee and the sectioned state at 0° ($P < .01$), 20° ($P < .0001$), 30° ($P < .0001$), 60° ($P < .0002$), and 90° ($P < .0001$) of knee flexion. When comparing the reconstructed and intact states, we found a small, but significant, increase in valgus rotation for the reconstructed knee of 2.5° at 0° ($P < .002$) of knee flexion. Additionally, we found a small, but significant, decrease in valgus rotation for the reconstructed knee compared to the intact state of 3.9° at 90° ($P < .02$) of knee flexion. There were no significant differences when comparing the intact and the reconstructed states at 20°, 30°, or 60° of knee flexion.

*External Rotation.* With an applied external rotation torque, we noted a significant increase in external rotation after sectioning the medial knee structures at 0° ($P < .0001$), 20° ($P < .0001$), 30° ($P < .0001$), 60° ($P < .0001$), and 90° ($P < .0001$) of knee flexion (Figure 3). In addition, we found a significant decrease in external rotation when comparing the reconstructed medial knee and the sectioned state at 0° ($P < .002$), 20° ($P < .0002$), 30° ($P < .0002$), 60° ($P < .0001$), and 90° ($P < .0001$) of knee flexion. There were no significant differences when comparing the intact knee and the reconstructed medial knee structure state.

*Internal Rotation.* With an applied internal rotation torque, we noted a significant increase in internal rotation after sectioning the medial knee structures at 0° ($P < .0001$), 20° ($P < .0001$), 30° ($P < .0001$), and 60° ($P < .002$) of knee flexion (Figure 4). In addition, we found a significant decrease in internal rotation when comparing the reconstructed medial knee and the sectioned state at 0° ($P < .0001$), 20° ($P < .0001$), 30° ($P < .0001$), and 60° ($P < .0001$) of knee flexion. There were no significant differences when comparing the intact knee and the reconstructed medial knee structure state at 0°, 20°, 30°, or 60° of knee flexion. A significant decrease in internal rotation of the reconstructed knee was found compared with the intact state of 3.2° at 90° ($P < .001$) of knee flexion.

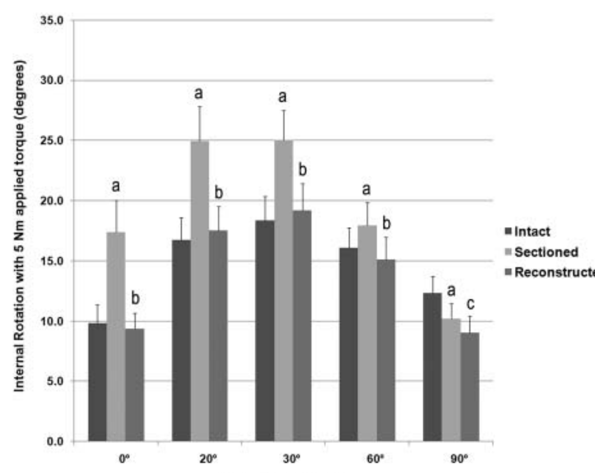


**Figure 4.** Change in internal rotation with an applied 5-N·m torque for intact, sectioned, and reconstructed medial knee structures at each flexion angle. Statistical significance is denoted in the figures as sectioned significantly different from intact (a), reconstructed significantly different from sectioned (b), and reconstructed significantly different from intact (c).

TABLE 1
Comparison of Observed Force (N) With the Application of an Externally Applied Load for Each Testing State[a]

| | 0° | | | 20° | | | 30° | | | 60° | | | 90° | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | POL | Proximal sMCL | Distal sMCL | POL | Proximal sMCL | Distal sMCL | POL | Proximal sMCL | Distal sMCL | POL | Proximal sMCL | Distal sMCL | POL | Proximal sMCL | Distal sMCL |
| Valgus | | | | | | | | | | | | | | | |
| Intact | 12.3 ± 1.9 | 42.1 ± 7.1 | 62.1 ± 9.3 | 16.9 ± 2 | 57.1 ± 7.6 | 87.5 ± 12.3 | 17.4 ± 2.1 | 58.6 ± 7.7 | 90.4 ± 12.7 | 15.9 ± 2.7 | 57.8 ± 9.4 | 82.7 ± 14.7 | 16.2 ± 2.7 | 54.2 ± 9.8 | 77.7 ± 12.4 |
| Reconstructed | 10.1 ± 2.4 | 52.4 ± 12.9 | 67.7 ± 17 | 16.6 ± 3.6 | 91.6 ± 19.2 | 108.9 ± 16.2 | 18.1 ± 3.8 | 101 ± 20.6 | 120.6 ± 18.7 | 17.1 ± 3.8 | 90.1 ± 18.4 | 104.6 ± 14.5 | 12.5 ± 2.7 | 66.8 ± 14 | 79.5 ± 10.8 |
| External rotation | | | | | | | | | | | | | | | |
| Intact | 12.7 ± 3 | 43.5 ± 11.2 | 59.7 ± 13.3 | 16.2 ± 2.7 | 60.9 ± 12.7 | 79.5 ± 17 | 16 ± 2 | 59.6 ± 10.6 | 79.6 ± 14.1 | 19.1 ± 2.4 | 69.8 ± 11.2 | 93.8 ± 13.6 | 20.3 ± 3 | 82.3 ± 12.8 | 113 ± 14.4 |
| Reconstructed | 9.4 ± 2.3 | 49.6 ± 12.4 | 61.4 ± 13.9 | 12.8 ± 2.6 | 73.2 ± 15.5 | 85.8 ± 16.6 | 14 ± 3.4 | 69.5 ± 15 | 90.3 ± 19.9 | 14.6 ± 2.7 | 81.7 ± 15.4 | 99.5 ± 18.5 | 17.5 ± 3.2 | 92 ± 12.5 | 120 ± 24.9 |
| Internal rotation | | | | | | | | | | | | | | | |
| Intact | 1 ± 0.2[b] | 3.5 ± 0.6 | 5.6 ± 1.4 | 1.8 ± 0.3[b] | 6.6 ± 1.4 | 11.7 ± 4[b] | 1.8 ± 0.3[b] | 6.6 ± 1.4 | 11.5 ± 3.8 | 1.8 ± 0.6 | 6.7 ± 2.3 | 12.4 ± 5.7 | 2.2 ± 0.9 | 6.5 ± 3 | 12.4 ± 5.7 |
| Reconstructed | 0.3 ± 0.1[b] | 2.8 ± 0.8 | 2.9 ± 0.5 | 1 ± 0.4[b] | 6.2 ± 1.7 | 6.9 ± 2.2[b] | 1.2 ± 0.3[b] | 6.5 ± 1.8 | 7.3 ± 1.9 | 1.4 ± 0.5 | 12.8 ± 5.2 | 12.9 ± 4.5 | 1.6 ± 0.6 | 7 ± 1.3 | 7.2 ± 2.2 |
| Anterior drawer | | | | | | | | | | | | | | | |
| Intact | 0.6 ± 0.2 | 3 ± 0.7 | 4.6 ± 1.1 | 0.9 ± 0.5 | 3.6 ± 1.3 | 5.6 ± 2.7 | 1.1 ± 0.7 | 4.1 ± 1.8 | 7 ± 3.7 | 1.8 ± 0.8 | 6.4 ± 2.1 | 9.8 ± 4.1 | 1.9 ± 0.7 | 6.4 ± 1.9 | 7.3 ± 1.1 |
| Reconstructed | 0.4 ± 0.1 | 3.1 ± 0.8 | 2.7 ± 0.6 | 0.3 ± 0.1 | 2.3 ± 0.6 | 3.8 ± 0.8 | 0.4 ± 0.1 | 3.7 ± 0.8 | 3.4 ± 0.9 | 2.3 ± 1 | 10.9 ± 4.2 | 11.3 ± 4 | 2.7 ± 1 | 11 ± 2.3 | 14.8 ± 3.4 |
| Posterior drawer | | | | | | | | | | | | | | | |
| Intact | 2.7 ± 0.6[b] | 8.8 ± 1.3 | 13 ± 2.6 | 1.7 ± 1 | 5.1 ± 2.3 | 8.5 ± 4.4 | 1.9 ± 1.3 | 5.7 ± 2.8 | 9.4 ± 5.1 | 5.8 ± 2.2 | 17.8 ± 5 | 25.9 ± 9.1 | 9.3 ± 2.2 | 27.6 ± 4.9 | 43.5 ± 8.2 |
| Reconstructed | 1.2 ± 0.3[b] | 7.6 ± 1.2 | 8.5 ± 1.9 | 0.5 ± 0.1 | 4 ± 0.7 | 4.7 ± 1.7 | 0.7 ± 0.2 | 4.8 ± 0.8 | 5 ± 2 | 3.5 ± 1.5 | 17.1 ± 4.1 | 24.9 ± 9.2 | 5.3 ± 1.2 | 26.9 ± 3.9 | 39.3 ± 11.5 |

[a]Values are presented as mean ± standard error of the mean. sMCL, superficial medial collateral ligament; POL, posterior oblique ligament.

[b]Values with significant differences between the intact and reconstructed states.

There was no significant difference between the sectioned and reconstructed states at 90° of knee flexion.

*Anterior Drawer.* For an applied anterior drawer load, we found a small, but significant, increase in anterior translation after sectioning of the medial knee structures of 2.4 mm at 20° ($P < .01$), 2.4 mm at 30° ($P < .03$), 3.0 mm at 60° ($P < .0006$), and 4.9 mm at 90° ($P < .0002$) of knee flexion. No significant changes were observed after sectioning at 0° of knee flexion. There was no significant difference when comparing the reconstructed to the sectioned states at any knee flexion angle. A small, but significant, increase in anterior translation was found when comparing the reconstructed to intact states of 2.0 mm at 60° ($P < .01$) and 4.1 mm at 90° ($P < .0009$) of knee flexion. There were no significant differences between the intact and reconstructed state at 0°, 20°, or 30° of knee flexion.

*Posterior Drawer.* For an applied posterior drawer load, we found a small, but significant, increase in posterior translation after sectioning of the medial knee structures of 1.9 mm at 0° ($P < .03$) and 1.3 mm at 30° ($P < .01$) of knee flexion. No significant differences between the intact and sectioned states were found at 20°, 60°, or 90° of knee flexion. There were no significant differences observed between the sectioned and reconstructed medial knee structure states at any knee flexion angle. Additionally, there were no significant differences found between the intact and reconstructed states at any knee flexion angle.

## Ligament Force and Load Sharing (Buckle Force Transducers)

The mean force responses for each ligament after an applied load at each flexion angle for the intact and reconstructed states were recorded in newtons (N) and are presented in Table 1. No bony or soft tissue impingement of the buckle transducers or undesirable ligament damage was noted throughout testing.

*Proximal Superficial Medial Collateral Ligament.* No significant difference in the mean force response was noted between the intact and reconstructed states at any flexion angle for the proximal sMCL for applied valgus loads, external and internal rotational torques, or anterior and posterior drawer loads.

*Distal Superficial Medial Collateral Ligament.* No significant difference in the mean force response was noted between the intact and reconstructed states at any flexion angle for the distal sMCL after applied valgus loads, external rotational torques, or anterior and posterior drawer loads. Additionally, no significant difference in the mean force response was found between the intact and reconstructed states after an applied internal rotational torque at 0°, 30°, 60°, and 90° of knee flexion. A significant decrease in the force experienced by the reconstructed distal sMCL compared with the intact ligament was found after an applied internal rotational torque at 20° of knee flexion ($P < .05$).

*Posterior Oblique Ligament.* No significant difference in the mean force response was noted between the intact and reconstructed states at any flexion angle for the POL after applied valgus loads, external rotational torques, or anterior drawer loads. Additionally, no significant difference in the force response between the intact and reconstructed states for the POL was noted after an applied internal rotational torque at 60° and 90° of knee flexion. A significant decrease in the mean force experienced by the reconstructed POL compared with the intact ligament after an applied internal rotational torque was found at 0° ($P < .002$), 20° ($P < .05$), and 30° ($P < .04$) of knee flexion. Furthermore, no significant difference force response between the intact and reconstructed states for the POL was noted after an applied posterior drawer load at 20°, 30°, 60°, or 90° of knee flexion. A significant decrease in the mean force experienced by the reconstructed POL compared with the intact ligament after an applied posterior drawer load was found at 0° ($P < .03$) of knee flexion.

## DISCUSSION

Our results validate and support our hypothesis that an anatomical medial knee reconstruction technique can restore near-normal stability to a knee after complete sectioning of the sMCL and POL. Additionally, by use of buckle transducers, the mean force response of the intact and reconstructed ligaments after an applied load were measured and we validated that the reconstructed ligaments did not have a greater force response when compared with the intact ligaments at any point during testing. This suggests that overconstraint of the knee and overloading of the reconstruction grafts, which could lead to graft failure, were prevented in our construct.

The medial collateral ligament is one of the most commonly injured ligaments of the knee, and combined medial collateral ligament and POL injuries are frequently found in patients with valgus laxity about the knee joint.[12,13] Despite the presence of a high frequency of medial knee injuries, an extensive literature review yielded no biomechanically validated anatomical reconstruction techniques, based on quantitative anatomical studies, which reconstruct sMCL or POL injuries. Our study describes and biomechanically validates an anatomical medial knee reconstruction technique using previously described quantitative medial knee landmarks (Figure 1A).[18]

We demonstrated that an anatomical medial knee reconstruction technique significantly improved knee stability compared with a proximal and distal sMCL and POL sectioned state. For valgus translation, we found significant increases in knee instability after sectioning of the medial knee structures at all tested knee flexion angles. An anatomical medial knee reconstruction technique significantly decreased the instability created with sectioning the medial knee structures, and furthermore, provided a full recovery to native stability at 20°, 30°, and 60° of knee flexion (Figure 2). Although we did find small differences in valgus motion at 0° and 90° of knee flexion, we do not believe these differences were of clinical importance because of the large amount of valgus instability restored after reconstruction at these 2 flexion angles. Moreover, the success of the reconstruction at other degrees of knee flexion for valgus rotation, and the absence of any evidence of overconstraint at 90° of knee flexion, further demonstrate the restoration (Table 1). We also found significant increases in external rotation at all degrees of knee flexion after sectioning the medial knee structures, which was restored after an anatomical medial knee reconstruction (Figure 3). Furthermore, sectioning of the medial knee structures resulted in significant increases in internal rotation at 0°, 20°, 30°, and 60° of knee flexion, which was restored following an anatomical medial knee reconstruction at these knee flexion angles (Figure 4).

Changes in anterior and posterior translation after an applied anterior or posterior drawer load were less predictable (Table 1). For applied anterior and posterior drawer loads, we found small increases in anterior translation after sectioning of the medial knee structures at 20°, 30°, 60°, and 90° of knee flexion and small increases in posterior translation after sectioning the medial knee structures at 0° and 30° of knee flexion. Although our reconstruction showed a trend toward recovering this small amount of anterior and posterior instability, we did not find a significant decrease in the anterior or posterior translation of the reconstructed state compared with the sectioned state at any flexion angle.

Buckle transducers have previously been used to measure the mean force response experienced by native ligaments following an externally applied load.[8,22,24] However, to our knowledge, these methods have never been applied to directly compare the relative forces experienced by native intact ligaments to that of reconstructed ligaments. The importance of using buckle transducers to examine these mean force responses would be to validate that overconstraint in the reconstruction grafts did not occur, because overconstraint could still result in improved static stability in vitro, yet potentially result in failure of the reconstruction because of overloading of the grafts if applied in vivo.[20,21] Our results show that the mean force responses of the intact and reconstructed proximal and distal divisions of the sMCL and POL to applied loads were very similar (Table 1). Furthermore, at no point was the mean force response of the reconstructed ligaments significantly greater than that of the intact ligaments.

Our study reports on an anatomical medial knee reconstruction technique that places the reconstruction grafts for both the sMCL and the POL at their anatomical attachment points (Figure 1).[18] We chose to study an anatomical reconstruction technique because previous studies have demonstrated that anatomical reconstructions better approximate normal knee biomechanics.[4,11,19] Furthermore, the historical understanding of ligament function in the knee led to approaching the sMCL as 1 continuous structure and reconstructing it as such.[9,10,18,26] However, further detailed studies of these structures revealed that both divisions of the sMCL, as well as the POL, had specific functions in creating the native load distribution of the medial knee.[8] To address these findings, an accurate anatomical reconstruction technique was developed and validated through direct force measurements to compare the reconstructed load distributions with those present in the native knee.

The sMCL was tightened at 30° of knee flexion because our study and previous biomechanical studies demonstrate that sectioning the medial structures at this flexion angle creates the greatest change in valgus laxity.[9,10,29] Furthermore, the POL was tightened at 0° of knee flexion based on previous biomechanical studies that reported that the POL has the greatest role in primary restraint of internal rotation at 0° of knee flexion.[8]

We recognize that our study has limitations typical to an in vitro study because it was a static biomechanical study of ligaments that have dynamic interactions in vivo. Because this study did not reproduce the dynamic stability provided by the semimembranosus muscle and its attachments to the medial knee,[18] it is possible that the magnitude of the forces seen on these structures may be different compared with a dynamic in vivo condition. Therefore, we recommend prospective clinical studies to fully evaluate the utility of this reconstruction technique in patients. In addition, we recognize that one of the limitations of buckle transducers is impingement during loading, which can affect the measured forces on the ligaments.[23] However, we believe that impingement was minimal because there were no abnormal voltage spikes from the buckle transducers' output during testing, which would have occurred with an impingement event. Moreover, buckle forces were comparable among intact versus reconstructed states, demonstrating that impingement did not occur, despite differences between the intact and reconstructed structures.

This reconstruction technique restores native stability, without concerns of producing undesirable overconstraint, in patients with chronic or severe acute medial knee injuries. Prospective clinical outcome studies are currently in progress to evaluate the use of this reconstruction technique in vivo.

## ACKNOWLEDGMENT

This research study was supported by grants from the Research Council of Norway (grant #175047/D15) and Health East Norway (grant #10703604). The authors acknowledge Conrad Lindquist for his help with biomechanical testing and Paul Lender for his assistance with the statistical analysis.

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