# Nonanatomic Placement of Posteromedial Meniscal Root Repairs: A Finite Element Study

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*Nonanatomic placement of posteromedial meniscal root repairs alters knee mechanics; however, little is known about how the position and magnitude of misplacement affect knee mechanics. Finite element knee models were developed to assess changes in cartilage and meniscus mechanics for anatomic and various nonanatomic repairs with respect to intact. In total, 25 different repair locations were assessed at loads of 500 N and 1000 N. The two-simple-suture method was represented within the models to simulate posteromedial meniscal root repairs. Anatomic repairs nearly restored total contact area; however, meniscal hoop stress decreased, meniscal extrusion increased, and cartilage–cartilage contact area increased. Repairs positioned further posterior altered knee mechanics the most and repairs positioned further anterior restored knee mechanics for posteromedial root repairs. Despite this, repair tension increased with further anterior placement. Anterior placement of repairs results in more restorative contact mechanics than posterior placement; however, anterior placement also increased the risk of suture cut-out or failure following repairs. Anatomic placement of repairs remains the best option because of the risks involved with anterior placement; however, suture methods need to be improved to better restore the strength of repairs to that of the native insertion. Proper placement of repairs is important to consider with meniscal root repairs because misplacement may negatively affect cartilage and meniscus mechanics in patients.* [DOI: 10.1115/1.4045893]

## Introduction

Meniscal root tears, when left untreated, lead to progressive meniscal extrusion, worsening arthritis, poor clinical outcomes, and often result in knee arthroplasty [1,2]. Although insertions of both the medial and lateral menisci are susceptible to tears, the posteromedial meniscal root tear is the most common clinically [3]. To prevent progressive extrusion and degeneration, meniscal root repairs have been developed to restore meniscal function following these tears [4–7]. Regardless of the technique used, tears are typically repaired by passing sutures through the injured meniscal root and fastening to the tibia. Consequently, patient outcomes depend on the sutures within properly positioned repairs.

Recent evidence suggests that root repairs placed anatomically are ideal to restore meniscal function following meniscal root tears. For anatomic repairs, sutures are used to reduce the injured meniscus back to its original position. Depending on the technique used, the sutures are passed through transtibial tunnels or secured with a screw at the anatomic center of the meniscal root [4–7]. Anatomic repairs are designed to reduce the meniscus to its native position, which then restores meniscal load transmission and cartilage contact. In a laboratory setting, cadaveric experiments demonstrate that anatomic repairs nearly restored total contact area and contact pressures [8,9]. Despite this, in a clinical setting, the progression of meniscal extrusion and joint degeneration is not always prevented postoperatively [10]. Thus, there is a possibility that procedural complications may prevent some patients from receiving a restorative repair.

A potential cause of limited success in the clinic may be nonanatomic placement of repairs at the time of surgery [11,12]. A postoperative assessment for placement of transtibial tunnels demonstrated that root repair-specific guides created tunnels significantly closer to anatomic; however, the specific guide still resulted in an average misplacement of 9% of the tibial measurements away from anatomic [12]. Considering anthropometric measurements of the tibial plateau in the literature, this results in tunnels potentially misplaced 3–6 mm away from anatomic [13]. This study suggested that repairs may be misplaced from the anatomic position, especially when not using root repair-specific instrumentation. In a previous in vitro study, placement of repairs 5 mm posteromedial of anatomic along the cartilage line for posteromedial repairs significantly decreased meniscal load transmission [8]. This suggests that a narrow window may exist within 5 mm away from anatomic to restore knee mechanics to normal for meniscal root repairs; however, little is still known about how accurate root repairs need to be to restore mechanics or how repair position affects knee mechanics.

Therefore, the purpose of this study was to determine how the position and magnitude of misplacement from the anatomic center of the posteromedial meniscal insertion affects knee mechanics. To investigate the importance of placement accuracy, we hypothesized that placement of posteromedial meniscal root repairs at least 5 mm away from anatomic in any direction would significantly alter load transmission of the meniscus and subsequent loading of the articular cartilage with respect to the intact condition.

## Methods

**Specimens.** Two finite element tibiofemoral models were created from image datasets available through the OpenKnee(s) project [14–17]. The imaged cadaveric knee specimens used for model development included a right knee (male, age 71) and left knee (female, age 25) with no reported signs of osteoarthritis. A third model was created from an open-source dataset where subject information was not collected [18,19]. All knee specimens were imaged at full extension.

Images were imported into segmentation software (Seg3D, University of Utah, Salt Lake City, Utah) to isolate tissue components. The tibial articular cartilage, femoral articular cartilage, medial meniscus, lateral meniscus, and all meniscal insertions

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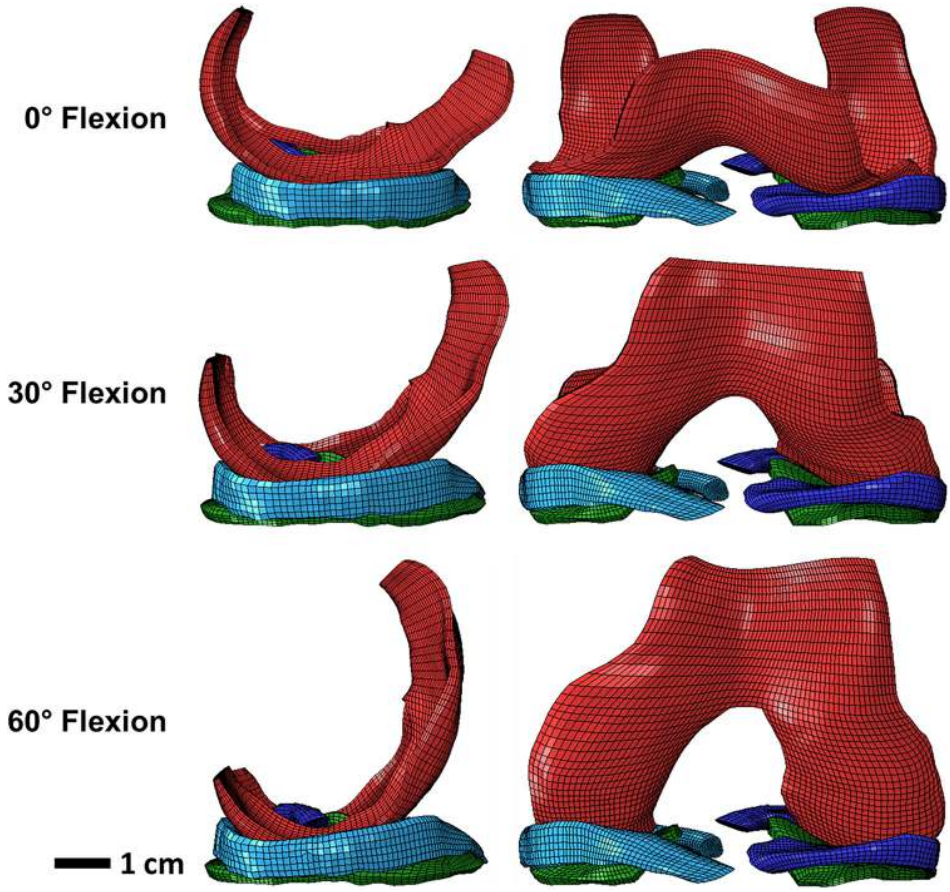


**Fig. 1 Sagittal views of a knee model in flexion angles of interest showing the continuum meshes of the femoral cartilage, medial meniscus, and tibial cartilage. Coronal views of the knee model show meshes of the femoral cartilage, tibial cartilage, lateral meniscus and insertions (right), and the medial meniscus and insertions (left).**

into the tibial plateau were isolated using a semi-automated threshold of each component in the datasets. Surface definitions of the component segmentations were then imported into the meshing software, TrueGrid (XYZ Scientific Applications, Inc., Livermore, CA) was used to generate linear, hexahedral meshes, and projected to the geometry of each segmented component (Fig. 1).

**Material Properties.** Material properties were taken from the previous studies of finite element models that were optimized and used for tibiofemoral compression [20,21]. Bone of the distal femur and proximal tibia was modeled as rigid. This assumption has a minimal effect on contact solutions when evaluating quasi-static tibiofemoral compression [22]. Tibial and femoral articular cartilage were modeled as homogeneous, linearly elastic, isotropic materials with a modulus 15 MPa and Poisson's ratio of 0.475 to maintain the nearly incompressible behavior of the articular cartilage under short loading times [23–25]. The medial and lateral menisci were modeled as homogeneous, linearly elastic, transversely isotropic materials with properties taken from the previous studies [21,26–29]. The modulus and Poisson's ratio in the collagen fiber direction of the menisci were 150 MPa and 0.3, respectively. In the plane perpendicular to the collagen fibers, the modulus and Poisson's ratio were 20 MPa and 0.2, respectively. Additionally, the shear modulus in the plane perpendicular to the collagen fibers was 57.3 MPa. Viscoelastic time dependence of the menisci was not considered due to the quasi-static model analysis. All meniscal insertions were modeled with the same properties as the meniscal bodies, with the exception that the in-plane elastic modulus was 10 MPa [30–32]. Each meniscal insertion site was represented by an insertion plane approximating the curvature of the anatomical insertion of the meniscus with the tibial plateau.

Ligament insertions into bone were identified from the image datasets and then represented as two or more tension-only, nonlinear, spring elements within the models. A piecewise function was used to represent the force–strain relationship of individual ligament bundles, as performed in the previous studies [21,33–36]. In each model, the anterior cruciate ligament was represented by anteromedial and posterolateral bundles [37]. The posterior cruciate ligament was represented by anterolateral and posteromedial bundles [38]. The medial collateral ligament was represented by anterior, middle, and posterior bundles for the superficial ligament and by anterior and posterior bundles for the deep ligament [39–42]. The lateral collateral ligament was represented by anterior, middle, and posterior bundles [42]. The anterior intermeniscal ligament, also known as the transverse ligament, was represented by superior and inferior bundles, and the ligament stiffness was taken from a previous optimization study [21]. The insertions of the anterior intermeniscal ligament were determined to be at the interface of the anterior root of the lateral and medial menisci with their anterior insertions [43]. Peripheral attachments of the menisci were not included, except a connection between the midbody of the medial meniscus and the bundles of the deep medial collateral ligament. Clinically, the posterior portion of the medial meniscus is detached from peripheral attachments prior to a meniscal root repair to reduce the meniscus back to the native position.

**Finite Element Analysis.** The general-purpose finite element code, ABAQUS (Dassault Systèmes Simulia Corp., Johnston, RI),

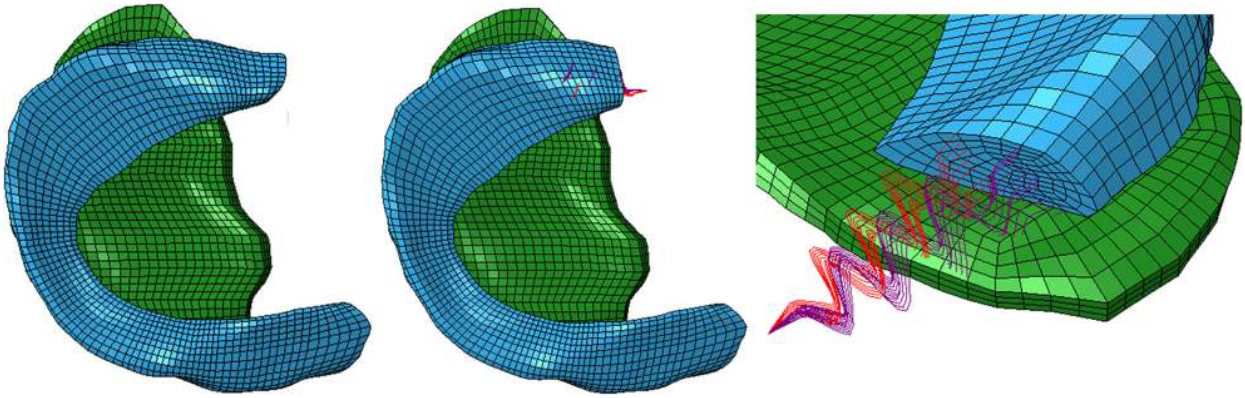

**Fig. 2 Medial compartment of a knee model with the tibial cartilage and medial meniscus for the intact condition (left) and for an anatomic meniscal root repair (middle). Elements of the medial meniscus were hidden (right) to better show the spring elements representing the interior and peripheral sutures passed through the meniscal root.**

was used to approximate contact solutions for a quasi-static analysis using the implicit solver. Hard, frictionless contact was modeled for the six contact-surface pairs in each model, which included the femoral cartilage and the meniscus body/insertions, the tibial cartilage and the meniscus body/insertions, and the femoral cartilage and the tibial cartilage for both the lateral and medial hemijoints.

Simulations of tibiofemoral compression were designed to assess rehabilitation and return-to-activity loading [44–46]. Rehabilitative loading consisted of a 500 N compressive load with knee flexion at 0 deg, 30 deg, and 60 deg flexion. This was used to represent standard range of motion and toe-touch weight-bearing protocols typical of six-week postoperative rehabilitation following meniscal root repairs. Return-to-activity loading consisted of a 1000 N compressive load with knee flexion at 0 deg, 30 deg, and 60 deg flexion. Simulations of the six loading scenarios were evaluated in the intact condition and compared to various posteromedial root repairs.

Specimen-specific kinematics for two specimens was available through the OpenKnee(s) project and used to approximate knee flexion within the models [17]. Specimen images were used to identify the medial epicondyle, lateral condyle, and the centroid of the mid-diaphysis to define the femoral coordinate systems. With the femoral coordinate system defined, the femoral articular cartilage was adjusted in the knee models based on the resultant kinematic data of the optimized passive flexion at 30 deg and 60 deg. This provided an approximation of the tibiofemoral orientation to assess joint mechanics at different flexion angles (Fig. 1). Joint kinematic data from the eight specimens available through the OpenKnee(s) project were averaged to approximate tibiofemoral flexion for the third model. This provided two models with specimen-specific joint kinematics and one model as a general case using average joint kinematics from a sample population.

Each simulation of tibiofemoral compression began with a displacement-controlled preload step and a force-controlled step. In the preload step, the interface between femoral cartilage and bone was displaced until the femoral cartilage contacted both menisci with translations and rotations fixed to ensure initial convergence of simulations. After preloading, in the force-control step, a specified compressive load of interest was applied to the distal femur with the flexion angle fixed, all other rotations free, and all translations free. This allowed the femoral cartilage to adjust from the experimentally determined orientation at 50 N of compression from the joint kinematic data into the optimal orientation resulting from compression at the specified load. The boundary conditions were repeated for each model in all three flexion angles and with both compression load.

**Meniscal Root Repairs.** Meniscal root tears of the posteromedial meniscal insertion were evaluated because tears of this insertion are most common [3,44]. Continuum mesh elements of the posteromedial meniscal insertion that inserted into the tibial plateau were first removed to simulate the root tear. The two-simple-suture technique was represented within the models because this method has been reported to resist displacement in comparison to other suture techniques [44,47]. This suture method was represented within the model as two tension-only spring elements passed through the fibrocartilaginous region of the meniscal root approximately 7 mm from the insertion, as performed clinically (Fig. 2) [48–51]. The mean stiffness of the meniscus-suture interface using the two-simple-suture method was reported as 45 N/mm, which was implemented into the model for the repair stiffness [52].

The anatomic center of the posteromedial meniscal insertion was determined by calculating the centroid of the insertion mesh surface that interfaces with bone. To best assess different transtibial tunnel placements, anatomic coordinates were modified to account for the slope of the meniscal insertion plane on the tibial plateau. The anterior–posterior axis was defined by connecting the most anterior point of the meniscal insertion site and the most

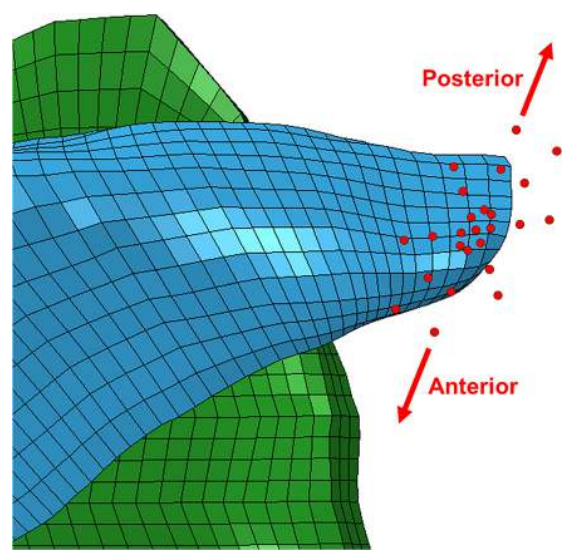


**Fig. 3 Top view of posteromedial meniscus and insertion with the tibial cartilage and locations to be used for placement of the anatomic and various nonanatomic repairs (dots) based on the slope of the meniscal insertion plane on the tibial plateau where the anterior–posterior axis was defined by the most anterior and most posterior point of the meniscal insertion site**

**Table 1 Approximate means and standard deviations from the literature used to generate material properties distributions in Monte Carlo simulation where $E$ = Young's modulus and $\nu$ = Poisson's ratio**

| | | |
|---|---|---|
| Cartilage [25,54] | $E$ (MPa) | 10 (4) |
| | $\nu$ | 0.42 (0.04) |
| Meniscus body [21,26–28,55] | $E_L$ (MPa) | 100 (35) |
| | $E_T$ (MPa) | 20 (5) |
| | $\nu_L$ | 0.2 (0.04) |
| | $\nu_T$ | 0.2 (0.04) |
| Meniscal insertion [30–32] | $E_L$ (MPa) | 180 (100) |
| | $E_T$ (MPa) | 5 (2.5) |
| | $\nu_L$ | 0.2 (0.04) |
| | $\nu_T$ | 0.2 (0.04) |

For transversely isotropic material characterization, $L$ = longitudinal and $T$ = transverse.

posterior point (Fig. 3). The medial–lateral axis was defined as being perpendicular to the anterior–posterior axis. Both axes were also defined to be coplanar with the meniscal insertion plane previously defined.

In modeling repairs, the meniscus was left in its native position to simulate reduction of the meniscus which is recommended for root repairs [7,53]. The ends of the suture elements that were not attached to the meniscus body were fixed to the anatomic center of the meniscal insertion to simulate an anatomic meniscal root repair. To simulate nonanatomic repairs, the suture elements were fixed to points away from the anatomic center. Positions evaluated include locations 1 mm anterior, posterior, medial, lateral, anteromedial, anterolateral, posteromedial, and posterolateral from the anatomic center using the modified anatomic directions. Simulated repairs were also evaluated at locations 3 mm around the anatomic center, and 5 mm around the center. This resulted in the evaluation of 25 different repair locations including the anatomic repairs and the array of locations around the anatomic center for each scenario.

**Mesh Convergence.** A mesh convergence study was performed to ensure the mesh density was appropriate for the model simulations. The convergence study was conducted with one knee model in the intact condition. The average element volume varied to create a range of knee models that ranged from coarse to fine. The root-mean-square-error was used to determine percent change in outcomes when comparing meshes with different mesh densities. The knee model was determined to have an appropriate mesh density when outcome variables of interest changed by less than 5% when compared to the most refined mesh. The average volume for the appropriate mesh density was then used to mesh the other two finite element knee models.

**Monte Carlo Simulation.** A Monte Carlo simulation was performed to produce a distribution of possible outcomes when material properties ranged from all values previously measured experimentally and implemented into the three knee meshes. MATLAB (MathWorks, Naticks, MA) was used to randomly generate possible scenarios of material properties with assumed normal distributions for all moduli and Poisson's ratios. The means and standard deviations of material properties were taken from the previous literature and listed in Table 1. Simulations were compared for the three intact knee models at 60 deg flexion and a 1000 N compressive load. A pilot analysis of one knee model demonstrated that 50 simulations were enough for means and standard deviations of outcome variables to change by less than 5% when compared to a different set of 50 simulations. Therefore, 100 simulations were enough to represent a consistent distribution of outcomes for a single knee model. Material properties were then generated to produce outcome distributions for the other two models. This resulted in 300 simulations which represented three different knee geometries each with 100 different combinations of material properties. Cumulative distribution functions were then created for all outcome variables. The three knee models using the material properties optimized for tibiofemoral compression in a previous study were compared to the 300 simulations of the three knee geometries with varying material properties. This provided a range of potential outcomes and an idea of how well the three knee models with the previously optimized material properties represented potential variance in material properties and geometry.

**Outcome Variables.** The mean hoop stress, represented by the mean Cauchy stress, along the collagen fiber direction was measured at the midbody of the medial meniscus and at the midbody of the anteromedial meniscal insertion. Meniscal extrusion was measured as the distance between the outer edge of the medial meniscus and the medial edge of the tibial articular cartilage in the medial–lateral direction of the tibial coordinate system which may reflect measurements made in coronal images of the knee. Extrusion was measured at three locations on the medial meniscus body—the anterior root, meniscus midbody, and posterior root. The total contact and the cartilage–cartilage contact areas on the medial surface of the tibial plateau were evaluated as well as the total contact area of the medial meniscus with the tibial and femoral cartilage surfaces. Repair tension of nonanatomic repairs was also compared to anatomic repairs to determine how nonanatomic placement alters tension.

**Statistical Analysis.** Differences in outcomes between repairs and the intact conditions are presented as percent changes. Paired-samples $t$ tests were performed to determine if percent changes were statistically different than zero. The Benjamini–Hochberg correction method was performed to adjust the familywise error rate to ensure the significance level, $\alpha$, was 0.05. In accordance with de Winter, statistical comparisons of paired data with $N = 3$ were sufficiently powered (> 80%) if there was a strong within-pairs correlation ($r \geq 0.8$) and a large effect size (Cohen's $d \geq 2$) [56]. Therefore, significant results were only reported if $r \geq 0.8$ and Cohen's $d \geq 2$.

## Results

Table 2 provides outcome data and results of the statistical analyses for the anatomic, 5 mm posterior, and 5 mm anterior repairs with a 1000 N load. Figures were generated for each dependent variable to portray a general overview of the results in the following subsections discussing the finite element results. A template was created to visualize relative changes in outcomes with respect to repair placement (Fig. 4).

**Mesh Convergence.** The convergence analysis demonstrated that the finite element solution converged for a mesh with an average element volume of 0.88 $mm^3$. The other two models were then discretized to have a similar element volume. The average element volumes of the other models were 0.55 and 0.30 $mm^3$. The average element count of the models was approximately 32,000 elements.

**Monte Carlo Simulation.** On average, the three knee models using material properties from a previous optimization study were able to represent 42% of the 300 simulations of varying material properties with the knee geometries for all outcomes. Mean hoop stress in the meniscus midbody was bound at probabilities of 0.34 and 0.84, for a predicted outcome probability of 50% (Fig. 5). This means that the three knee geometries with previously optimized material properties may represent potential outcomes with material properties within the ranges determined experimentally. Total contact area on medial meniscus surface was bound at

**Table 2 Outcome data for the anatomic, 5 mm posterior, and 5 mm anterior repairs with a 1000 N load**

| | | Mean hoop stress | Cartilage–cartilage contact area | Total contact area | Repair tension |
|---|---|---|---|---|---|
| Anatomic | 0 deg | **−28% ± 6%** | **15% ± 3%** | −12% ± 3% | — |
| | 30 deg | **−16% ± 4%** | **30% ± 9%** | −4% ± 2% | — |
| | 60 deg | −13% ± 5% | **16% ± 5%** | −3% ± 4% | — |
| 5 mm posterior | 0 deg | **−30% ± 3%** | **19% ± 2%** | **−13% ± 1%** | 1% ± 11% |
| | 30 deg | **−29% ± 3%** | **46% ± 10%** | −7% ± 2% | −11% ± 5% |
| | 60 deg | **−25% ± 2%** | **22% ± 4%** | −5% ± 6% | −10% ± 9% |
| 5 mm anterior | 0 deg | **−31% ± 10%** | 16% ± 7% | −12% ± 6% | −2% ± 12% |
| | 30 deg | −11% ± 7% | 21% ± 11% | −4% ± 3% | 10% ± 5% |
| | 60 deg | −8% ± 9% | **12% ± 4%** | −2% ± 5% | 10% ± 9% |

Results are represented as percent of paired difference from the intact condition, except for repair tension.
Changes in tension from nonanatomic repairs are with respect to the anatomic repair. Bold text denotes statistical significance.

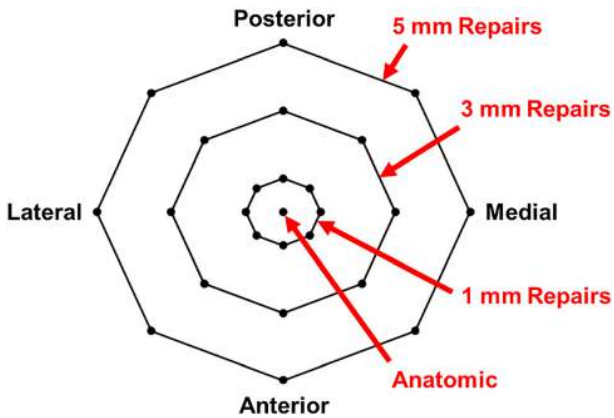


**Fig. 4 Template for subsequent figures to indicate orientation and present relative changes in the results at the location of the nonanatomic repair of posteromedial meniscal root tears**

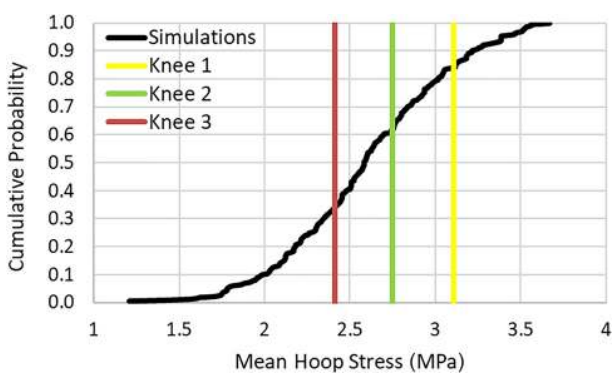


**Fig. 5 Example of cumulative density functions calculated from Monte Carlo simulations. Mean hoop stress of the medial meniscus midbody is shown to be bound at a probability of 0.34 by knee 3 and at 0.84 by knee 1, for a predicted outcome probability of 50%.**

probabilities of 0.1 and 0.24, for a predicted outcome probability of 14%. Mean hoop stress in the anteromedial meniscal insertion was bound at probabilities of 0.3 and 0.82, for a predicted outcome probability of 52%. Total contact area on the medial surface of the tibial plateau was bound at probabilities of 0.03 and 0.49, for a predicted outcome probability of 46%. The cartilage–cartilage contact area was bound at probabilities of 0.13 and 0.62, for a predicted outcome probability of 49%.

**Medial Meniscus Hoop Stress.** The mean hoop stress measured at the midbody of the medial meniscus decreased for anatomic and all nonanatomic repairs with respect to intact with varying effect by load magnitude and joint angle (Fig. 6). For anatomic repairs with a 1000 N load, meniscal hoop stress decreased at flexion angles of 0 deg and 30 deg. At 0 deg flexion, all nonanatomic repairs decreased with a 1000 N load. Mean hoop stress of all repairs except for the 5 mm anterior repair significantly decreased at 0 deg flexion with a 500 N load. Hoop stress did not significantly change at 30 deg or 60 deg flexion with the rehabilitative load of 500 N.

Mean hoop stress changes within the anteromedial meniscal insertion followed a similar trend within the meniscus midbody (Fig. 7). Repairs placed 5 mm posterior resulted in the largest decreases to meniscal hoop stress, while repairs placed 5 mm anterior best restored hoop stresses to intact values. There were no significant changes to the meniscal hoop stress with respect to intact at any of the flexion angles with a 500 N load.

**Meniscal Extrusion.** Tibiofemoral compression with all repairs caused the posterior root of the medial meniscus to extrude more than the anterior root or the midbody at all flexion angles. Repairs caused the meniscus to extrude less than 0.1 mm at the anterior root and the midbody of the medial meniscus in the medial direction, as if viewed from a frontal image of the tibia. For the posterior root, tibiofemoral compression following anatomic repairs caused around 0.2–0.6 mm of extrusion in the medial direction, depending on the load and flexion angle (Fig. 8). Meniscal extrusion with anatomic repairs significantly increased at 60 deg flexion with a 1000 N load ($p = 0.02$) and with a 500 N load ($p = 0.02$). The medial meniscus extruded the most following a 5 mm posterior repair with 1.1 ± 0.2 mm ($p = 0.009$) and the 5 mm anterior repair best restored the meniscus with extrusion of 0.4 ± 0.2 mm ($p = 0.11$).

**Cartilage and Meniscus Contact Areas.** Total contact area on the surface of the medial tibia was nearly restored, but cartilage–cartilage contact increased for all repairs. On average, repairs nearly restored the total contact area at flexion angles of 30 deg and 60 deg (Fig. 9). At 0 deg flexion, the average contact area decreased by at least 11% for anatomic and nonanatomic repairs. Repairs placed more posteriorly or laterally resulted in a significant decrease in total contact area ($p < 0.04$). Cartilage–cartilage contact area significantly increased with anatomic and 5 mm posterior repairs at 1000 N (Fig. 10). Repairs placed further anterior were better at restoring cartilage–cartilage contact mechanics to normal. Large variability in results for repairs at 30 deg flexion prevented any significant changes with the 500 N load.

The total contact of the medial meniscus with tibial and femoral cartilage significantly decreased for all repairs at 30 deg and 60 deg flexion (Fig. 11). As meniscal extrusion increased, contact with the tibial and femoral cartilage of the medial compartment decreased equally. Posterior repairs decreased meniscal

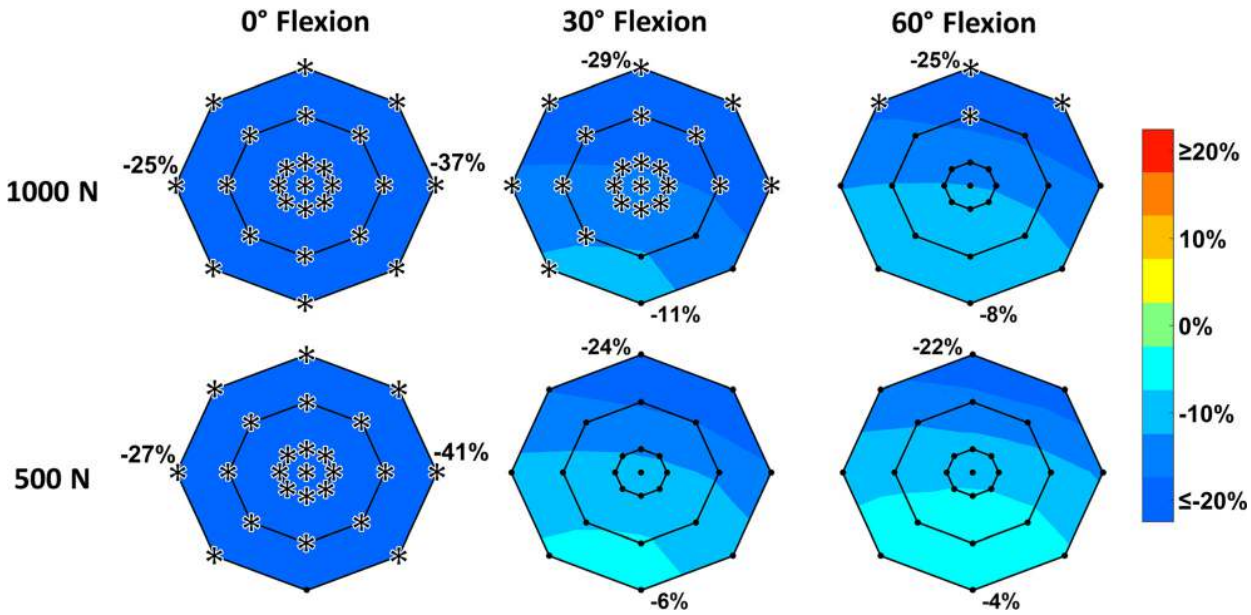


**Fig. 6 Percent changes to the mean hoop stress of the medial meniscus with respect to intact for anatomic and nonanatomic repairs. *denotes a statistically significant difference between the indicated repair location with respect to intact.**

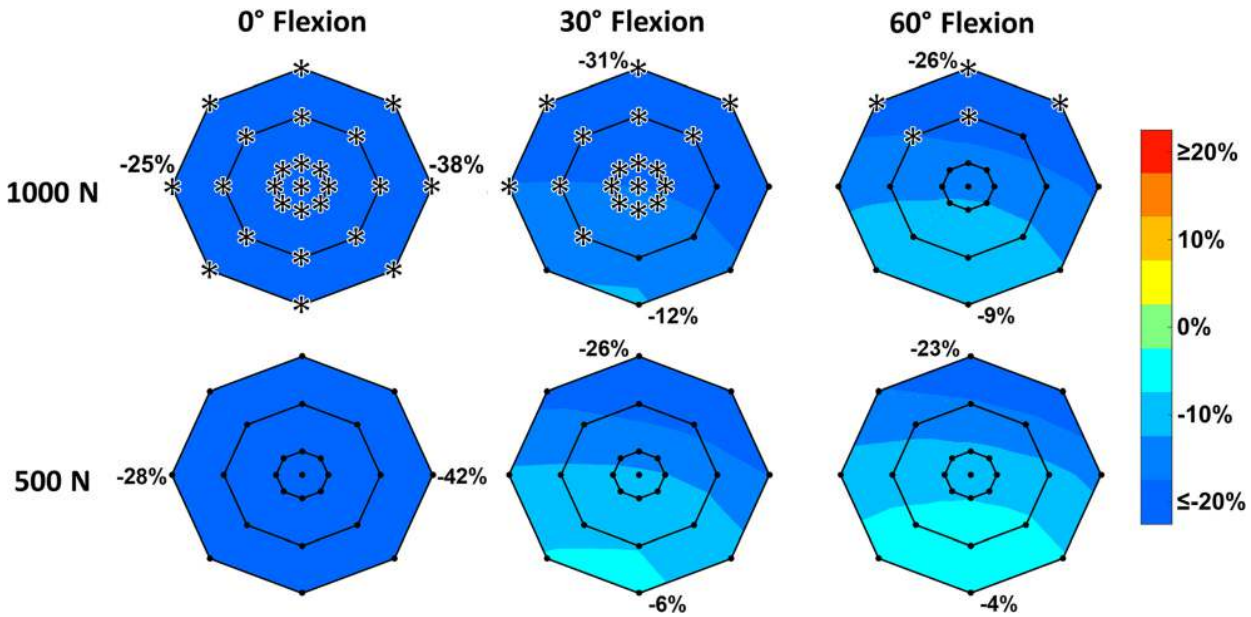


**Fig. 7 Percent changes to mean hoop stress of the anteromedial meniscal insertion with respect to intact for anatomic and nonanatomic repairs. *denotes a statistically significant difference between the indicated repair location with respect to intact.**

congruency the most for both loads at these flexion angles. Repairs placed 5 mm anteromedial were the most restorative; however, meniscal congruency still significantly decreased. At 0 deg flexion, total contact area on the medial meniscus surface decreased for all repairs but was not significant except the repair 5 mm posterior with a 500 N load.

**Repair Tension.** Repair tension did not significantly change for nonanatomic repairs with respect to anatomic repairs at any flexion angle or tibiofemoral load assessed. While not statistically significant in this pilot study, the repair tension increased when repairs were placed further anterior or medial and decreased when repairs were placed further posterior or lateral with flexion (Fig. 12). At these flexion angles, there were average increases of at least 16% when repairs were placed 5 mm anteromedial and average decreases of at least 10% when repairs were placed 5 mm posterior and 5 mm posterolateral. At 0 deg flexion, the nonanatomic repairs were all within 3% of the anatomic repair tension.

## Discussion

This study demonstrated that repair placement was important to restore knee mechanics following meniscal root repairs. Anatomic repairs were able to nearly restore the total contact area, as has been demonstrated in cadaveric studies; however, this study revealed that anatomic repairs do not completely restore other

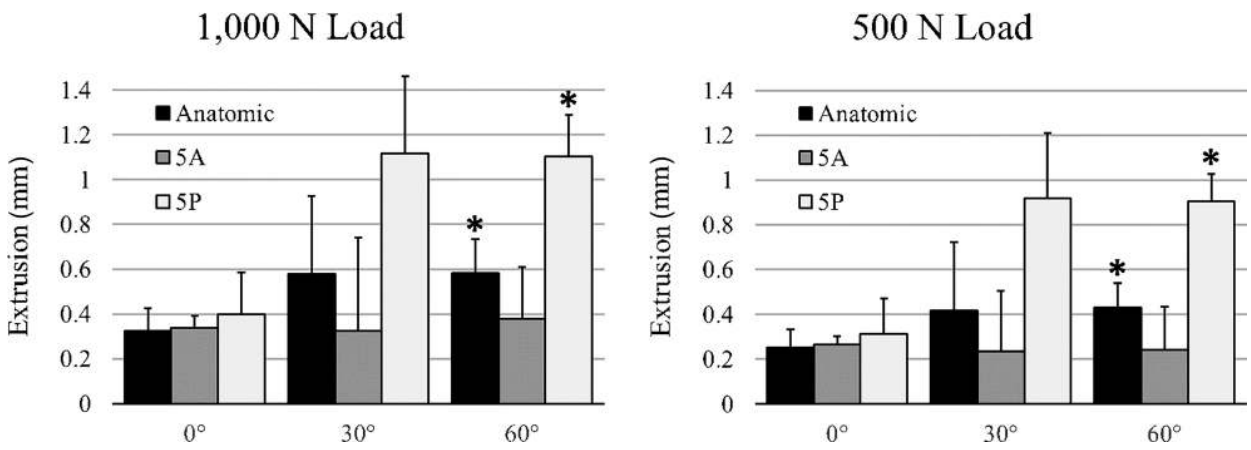


**Fig. 8 Increases in meniscal extrusion measured at the posterior portion of the medial meniscus. * denotes a statistically significant difference between the indicated repair location with respect to intact. 5*A* = 5 mm anterior repair; 5*P* = 5 mm posterior repair.**

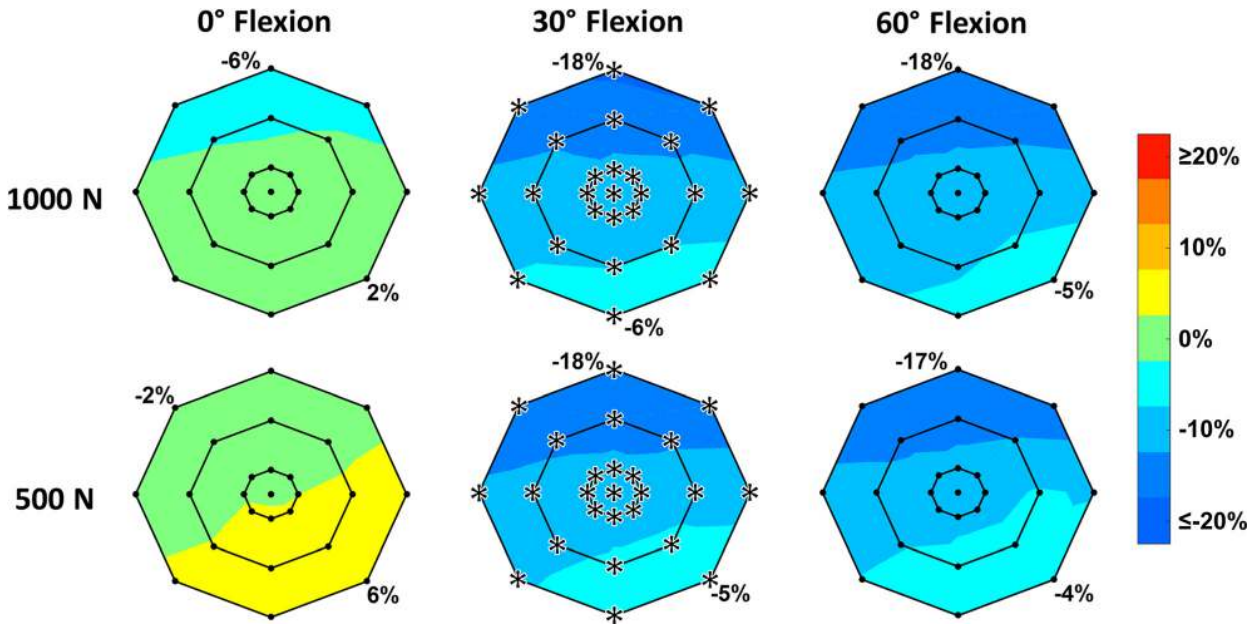


**Fig. 9 Percent changes of the total contact area on the medial surface of tibial articular cartilage for anatomic and nonanatomic repairs with respect to the intact condition. *denotes a statistically significant difference between the indicated repair location with respect to intact.**

important changes to the tibiofemoral joint that have not been previously evaluated. Repairs misplaced further anterior were better at restoring knee mechanics to normal than anatomic or repairs misplaced further posterior. Repairs misplaced further posterior resulted in the greatest changes to cartilage and meniscus mechanics.

A major finding was that anatomic repairs were unable to restore meniscal hoop stresses to normal for all loading conditions. Repairs misplaced anteriorly were better at restoring load transmission through the menisci; however, these repairs were still unable to completely restore meniscal hoop stresses. This means that the clinical standard of placement for meniscal root repairs was unable to restore normal function of the meniscus. Although only one suturing method was considered in the models, this result is important because it highlights the insufficiency of current meniscal root repair techniques, which has also been demonstrated in previous studies for more than just the two-simple-suture method [52,57]. Kopf et al. demonstrated that suture repairs were unable to replicate the ultimate tensile strength of intact meniscal insertions [57]. Rosslenbroich et al. demonstrated that the stiffness of the native meniscal insertion was not significantly different than the two-simple-suture technique; however, the suture repair on average was less stiff [52]. Other suture methods need to be evaluated; however, the presented results and the previous studies have demonstrated the inability to mimic the mechanics of the native insertion with sutures and should motivate more robust fixation for repairs.

Another interesting finding was that all repairs, to some degree, increased extrusion from the joint with respect to the native meniscus. Tibiofemoral compression of anatomic repairs created up to 0.6 mm of extrusion due purely from creation of the repair. While there was only a significant increase with an anatomic repair at 60 deg flexion, this data still suggests that anatomic repairs are unable to eliminate extrusion entirely. This result corresponds well with recent experimental measurements of extrusion where Daney et al. demonstrated that anatomic repairs were

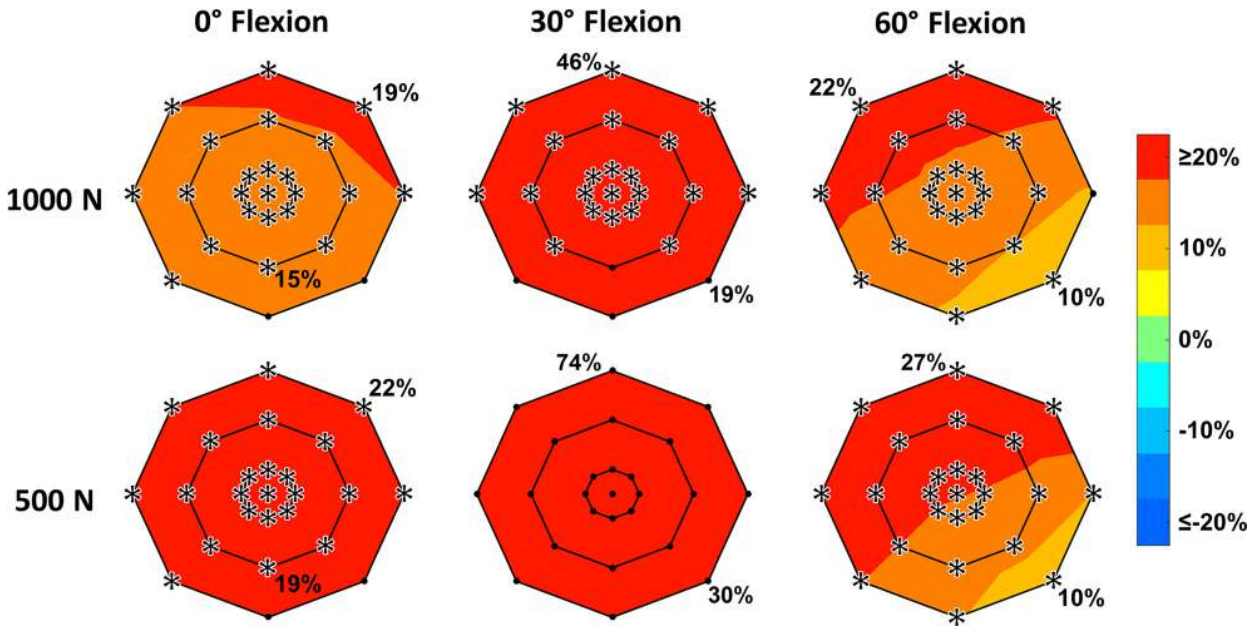


**Fig. 10 Percent changes of the cartilage–cartilage contact area on the medial surface of tibial articular cartilage for anatomic and nonanatomic repairs with respect to the intact condition. *denotes a statistically significant difference between the indicated repair location with respect to intact.**

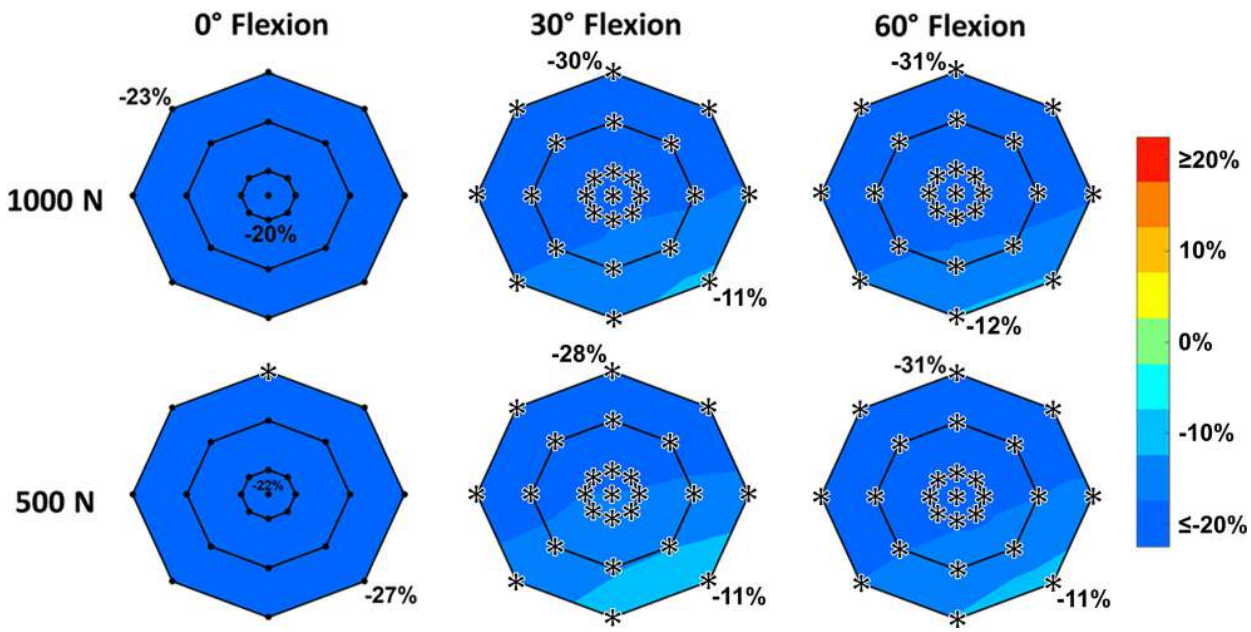


**Fig. 11 Percent changes to area of contact between medial meniscus and both tibial and femoral cartilage with respect to intact for anatomic and nonanatomic repairs. * denotes a statistically significant difference between the indicated repair location with respect to intact.**

unable to restore extrusion without help from peripheral stabilization [58]. In addition, repairs misplaced 5 mm posterior from the anatomic center of the meniscal insertion extruded around 1 mm purely from creation of the repair. Repairs misplaced further anterior were able to reduce the amount of meniscal extrusion below 0.5 mm. Although the magnitudes of extrusion measured in this study are relatively small compared to the clinical threshold of 3 mm that defines major meniscal extrusion [59], these results still demonstrate that suture placement is important to prevent unnecessary extrusion from occurring. A previous study demonstrated that repairs created using root repair-specific guides may be misplaced away from the anatomic center by approximately 3–6 mm [12,13]. The results of this study suggest that patients may be susceptible to meniscal extrusion, at least in part, due to the malposition of the root repairs.

Results regarding contact area in this study demonstrated how information previously gathered about root repairs may be misleading. In this study, anatomic repairs nearly restored the total contact area at all flexion angles and both loads assessed. In the previous cadaveric studies, anatomic repairs have also been shown to restore the total contact area on the tibial plateau [8,9,60]; however, these studies were unable to separately distinguish cartilage–

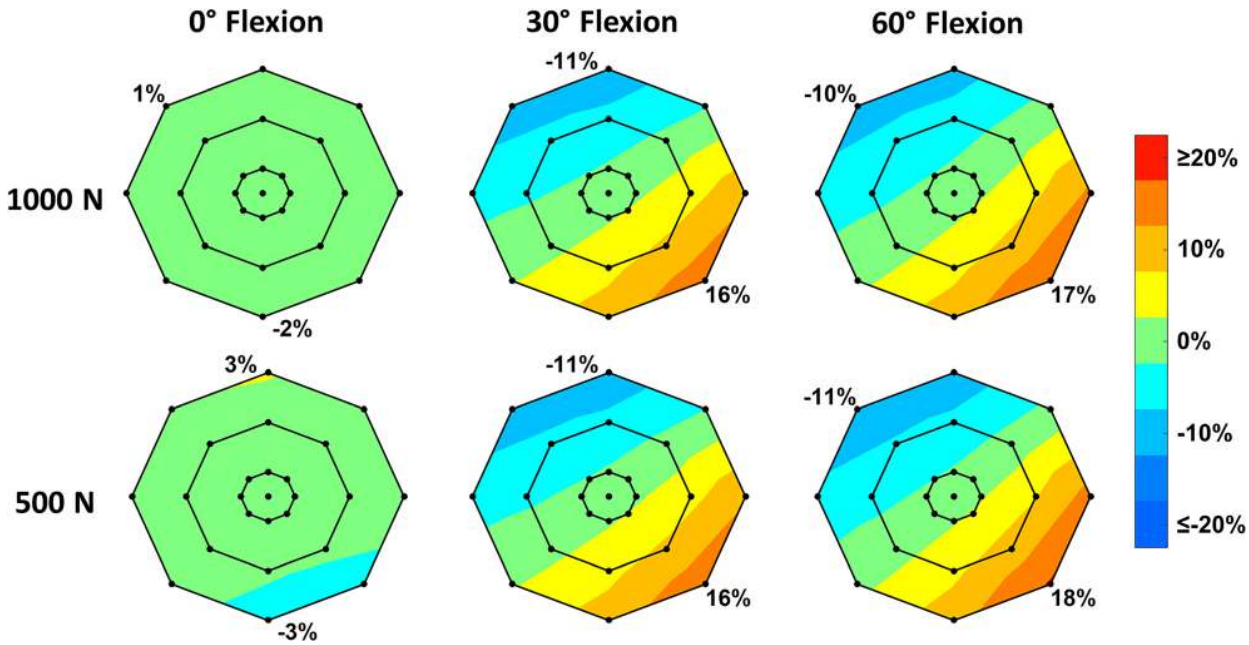


**Fig. 12 Percent changes to nonanatomic root repair tension with respect to anatomic root repairs. * denotes a statistically significant difference between the indicated repair location with respect to anatomic root repairs.**

cartilage contact. Using finite element analysis, this study reaffirmed the minimal differences in total contact area and distinguished that the contact between the tibial and femoral cartilage significantly increased for most repairs. The results of this study suggest that current root repairs may cause the menisci to be less congruent within the articular surfaces. This decrease in congruency then leads to the increase in cartilage–cartilage contact to compensate for the meniscus. The restoration of total contact may mislead others into thinking that loads are also being properly distributed between the meniscus and cartilage. Instead, the results suggest that the articular cartilage may be overloaded due to the meniscal repair and insufficient load transmission.

At higher degrees of flexion, repairs misplaced further anterior and medial increased the tension and repairs misplaced further posterior and lateral decreased tension with respect to the anatomic repair. These results suggest that posteromedial repairs placed further anterior and medial may increase the risk for suture cut-out through the meniscal root or potential failure of the repair. Clinically, anterior repairs have been observed to be insufficient where the knee is flexed, and the sutures pull out of the root repair because of the increased tension. The ultimate failure load of repairs has also been demonstrated to be much lower than the native meniscal insertion [57]. By increasing the tension seen in repairs due to placement, the repairs may reach the ultimate failure load from lower tibiofemoral loads than with repairs placed further posterior. Additionally, loosening of repairs has been demonstrated to occur because of suture cut-out from repetitive loads that simulate rehabilitation [61]. Ultimately, an increase in repair tension may lead to a progression of meniscal extrusion postoperatively, which has been demonstrated to occur clinically [10].

This study is not without limitations that need to be considered when interpreting the results. The results of this study are limited to three knee models using material properties from a previous optimization study. The Monte Carlo simulation of varying material properties from previous experimental studies provides a better understanding of how representative the results may be; however, the probabilistic simulations are still limited because they do not account for many other variables. Despite the small sample size and limitations on material properties used, the results presented were enough to detect significance for large, relative changes that occurred with different placement of meniscal root repairs compared to intact and provide a better idea of how changes in knee mechanics vary with position. In addition, this study did not consider any effects that may change results due to healing or any effects that may occur with repair loosening. Therefore, the results of this study only considered effects due to misplacement of root repairs. Suture placement in this study was also idealized with respect to clinical root repairs. When sutures are passed through tunnels, or secured to a bone screw, at the anatomic center of the injured meniscal insertion clinically, the sutures may not exactly be in the most anatomic position. For example, the diameter of tunnels used for transtibial pull-out root repairs may range from 2.4 mm to 4.5 mm in diameter [12,62]. Therefore, if the center of the created tunnel matches the anatomic center of the meniscal insertion, the sutures will be pulled further away from the center with a larger tunnel during joint loading. Therefore, the anatomic placement in this study may not be directly compared to an anatomic repair created clinically and should be considered when interpreting the results.

In summary, posterior placement of posteromedial root repairs should be avoided to prevent any additional, induced extrusion and altered contact mechanics. Additionally, anterior placement of posteromedial repairs better restores cartilage and meniscus mechanics to intact, but also increases tension and thus increases the risk of suture cut-out or repair failure. Anterior placement may also not always be possible without damaging other soft tissue structures and further compromising the joint. Anatomic placement of repairs remains the best option because of the risk involved with further anterior placement; however, suture methods needs to be improved to better restore strength of repairs to that of the native insertion. This study only investigated the two-simple-suture method, so other repair methods may be better at restoring mechanics with anatomic repairs.

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