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ABSTRACT (249 words)

Medial patellofemoral ligament reconstruction (MPFLR) has emerged as the procedure of choice for recurrent patellar dislocation. This addresses soft tissue injury but does not address underlying anatomic factors, including trochlear dysplasia, that are commonly present and increase risk of dislocation. Quantification of the stability offered by other surgical interventions, namely, medializing tibial tubercle osteotomy (mTTO) and trochleoplasty, with and without MPFLR, may provide insight for surgical choices in patients with trochlear dysplasia. We developed subject-specific finite element models based on magnetic resonance scans from a cohort of 20 patients with trochlear dysplasia and recurrent patellar dislocation. The objectives of this study were (1) to compare patella stability after mTTO and trochleoplasty procedures; (2) to evaluate whether it is necessary to perform a MPFLR in combination with the mTTO or trocheoplasty procedure; and (3) to quantify the robustness of patellar stability to variability in knee kinematics. Trochleoplasty performed better than mTTO at stabilizing the patella between 5° and 30° flexion. For both mTTO and trochleoplasty procedures, it was beneficial to also perform MPFLR — inclusion of MPFLR halved the magnitude of patellar laxity predicted in the simulations. Simulations that did not include any MPFL restraint were also more sensitive to variation in tibiofemoral internal-external kinematics.

Clinical Significance: This study highlights differences in stability provided by mTTO and trochleoplasty procedures. It also highlights the importance of MPFLR in helping to stabilize the joint, regardless of other procedures that may also be performed, and the sensitivity of patellar stability outcomes to tibiofemoral kinematics.

Keywords: patella dislocation, trochlear dysplasia, medializing tibial tubercle osteotomy, finite element simulation, computational model

INTRODUCTION

Patellar dislocation is an injury that typically affects younger, active people. The biomechanics and loading of the lower limb mean that lateral dislocation is the prevalent mode of patellar dislocation, occurring in approximately 90% of patellar dislocations. Dislocation typically occurs early in flexion (<30°) with the quadriceps engaged and the femur internally rotated.[1] After an initial dislocation incident, the likelihood of developing recurrent dislocation is approximately 50%.[2-4] Recurrent dislocation can result in significant reduction in activity, or activity hesitancy due to fear of another dislocation event. Non-operative treatment through physical therapy is generally the first course of action, which progresses to surgical management if a non-operative approach fails to correct the instability.[5; 6]

Trochlear dysplasia, resulting in a flat or even convex trochlear groove is a major risk factor for patellar dislocation[7] – in a clinical study of 60 patients with recurrent dislocation and 120 controls, 68% of the dislocation group were classified with trochlear dysplasia, compared to 6% of the control group.[8] Despite its prevalence in the patient population, trochlear dysplasia is often not directly treated through surgical management. More common surgical approaches are: medial patellofemoral repair or medial patellofemoral reconstruction (MPFLR) to repair or restore, respectively, the soft-tissue ligamentous constraint that is usually ruptured with initial dislocation[5; 9; 10] and medializing tibial tubercle osteotomy (mTTO) to medialize the tibial attachment of the patellar tendon, altering lower limb biomechanics and reducing the lateral pull of the quadriceps during muscle engagement.[5; 11] Trochleoplasty to deepen the sulcus groove is a more technically complex and invasive procedure,[12-15] and so should be considered only when it will provide tangible improvements in joint stability over other, less invasive, surgical options. However, there is little guidance available to surgeons on when, or if, trochleoplasty is indicated for a particular patient.

Unlike clinical studies, which inherently include large amounts of variability and uncertainty, cadaveric and computational studies can investigate differences in surgical procedures in a more targeted fashion than is possible in vivo by directly comparing procedures on the same subject. Several cadaveric and FE studies have quantified patellar instability by applying a lateral displacement to the patella, and measured the force required to achieve this displacement.[16-19] Some have quantified the effect of trochleoplasty, either in isolation or with MPFLR, while others have quantified changes in patellar stability with tibial tubercle osteotomy, sometimes comparing between soft-tissue changes such as MPFR or lateral retinacular release.[16-18; 20; 21] However, a direct quantitative comparison of patellar stability between trochleoplasty and mTTO procedures has not been published.

We applied a computational approach to evaluate patellar stability after simulating MPFLR, mTTO and trochleoplasty procedures. The objectives of this study were (1) to compare patella stability after mTTO and trochleoplasty procedures in a cohort of 20 patients with trochlear dysplasia and recurrent lateral patellar dislocation; (2) to evaluate whether it is necessary to perform a MPFLR in combination with the mTTO or trocheoplasty procedure; and (3) to quantify the robustness of patellar stability to variability in knee kinematics.

MATERIALS AND METHODS

Patient-specific model development:

Magnetic resonance (MR) images were obtained from 20 patients with recurrent lateral patellar dislocation under Institutional Review Board approval from Mount Carmel Health System. The Oswestry-Bristol classification system[22] was used as a measure of trochlear dysplasia, as graded by an experienced orthopaedic surgeon (RNS). The femoral trochlea of nine patients were classified as moderate dysplasia (flat trochlea) while trochlea from the remaining 11 patients were classified as severe dysplasia (convex trochlea). Knee geometry for each subject was aligned to a local femoral coordinate system using an iterative closest point (ICP) algorithm implemented in MATLAB (Mathworks, MA). A dynamic three-dimensional finite element (FE) model, based on previous publications,[23; 24] was developed in Abaqus/Explicit (Simulia, RI). In brief, the model includes femur, tibia, and patella bones, femoral, tibial and patellar articular cartilage, patellar tendon and quadriceps tendon and muscle. The quadriceps muscles were differentiated into rectus femoris (RF), vastus intermedius (VI), vastus lateralis (VL) and vastus medialis (VM) bundles.[25] Similar to prior work, bony surfaces were represented with rigid triangular shell elements. Patellar, femoral and tibial articular cartilage was modeled using eight-noded hexahedral elements. First order hexahedral elements were applied as they have better convergence rates that tetrahedral meshes are so are typically more computationally efficient.[26] For computational efficiency, the patellar tendon was modeled with six non-linear springs. Quadriceps tendons were modeled as membranes with embedded springs to allow for potential contact and wrapping around the femur (**Fig. 1**).[27; 28]

A subject-specific model was developed for each of the 20 patients and a dynamic simulation was performed where the knee was extended from 50° flexion to 5° hyperextension. This range was selected as patellar dislocation typically occurs in early flexion, before the patella enters the deeper portion of femoral groove in later flexion. Tibiofemoral (TF) motions were kinematically prescribed, apart from superior-inferior translation and valgus-varus rotation, which were determined by contact between femoral and tibial articulating surfaces. The other TF motions were determined from a biplane fluoroscopy analysis of 50 healthy participants performing a weighted knee extension activity.[29] Anterior-posterior (A-P) and medial-lateral (M-L) translations were averaged across the fluoroscopy dataset and kinematically prescribed. Internal-external (I-E) rotations were varied as described below. No kinematic constraints were applied to the patella. Instead, patellar motion was determined by patellar and femoral articulation and soft-tissue constraints, namely patellar tendon, a 400 N force distributed among the quadriceps muscle, and (where applicable) the MPFL. A 400 N load was applied to represent a load in the physiological range at low flexion angles, and was also similar to the load applied at low flexion angles in the patellar kinematics validation study by Baldwin et al.[27] upon which the current model is based. A consistent load was applied across all models to eliminate quadriceps load as a confounding source of variability in our comparisons.

Simulating surgical interventions:

Each patient model was used to simulate MPFLR, mTTO, and trochleoplasty (TP) procedures (**Fig. 1).** mTTO and TP procedures were evaluated both with and without an MPFLR in order to determine if MPFLR is still necessary when another, more invasive, procedure is performed. Additionally, the pre-operative condition, without a MPFL and without any surgical intervention

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was used as a negative control (referred to here as the preoperative model). Finally, to simulate a healthy baseline condition, a combination of MPFLR, mTTO and TP procedures was used to restore sulcus angle, TT-TG distance and MPFL restraint to levels representing an anatomically normal condition (referred to here as the healthy baseline model). MPFLR force (where applicable), patella M-L shift and I-E rotation were compared between surgical groups.

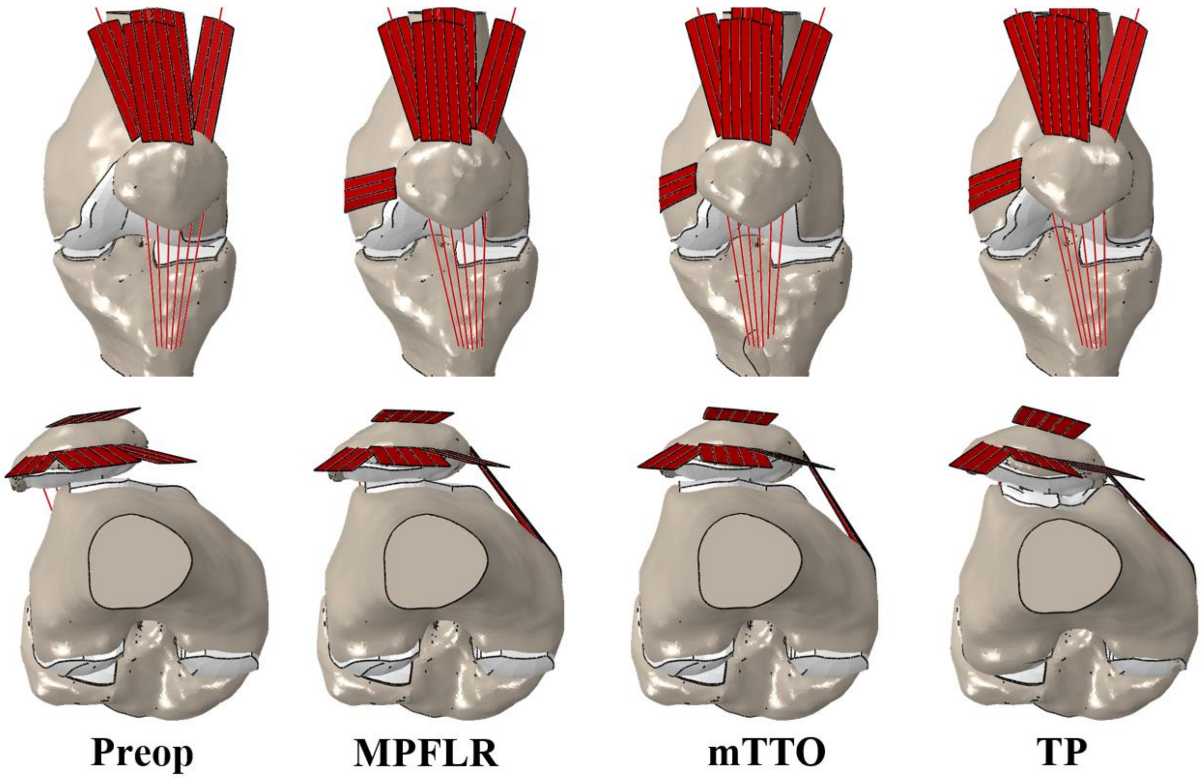

Figure 1: Finite element model of the knee including bone and cartilage, patellar tendon, quadriceps tendon and muscle and (where applicable) MFPL. Shown here (from left to right) for the preop, MPFLR, mTTO, and TP models of a representative subject in coronal (top) and axial (bottom) views.

MPFL representation and mTTO and TP simulation were implemented as described previously[24] and summarized briefly here. When present, the MPFL was modeled as a membrane with embedded fiber-reinforced springs with properties and attachment sites of a healthy intact MPFL.[30-32] The attachment sites of the MPFL were subject-specific and determined from visualization of bony landmarks on the MR images. The MPFL was initially tensioned such that ligament tension at 5° flexion was 8 N during passive loading. This ensured that the MPFL was engaged at and near full extension, but also allowed for a slack MPFL later in flexion when the patella engages the trochlear groove, minimizing unwanted PF contact stress.[33; 34] mTTO surgery was represented by medial transfer of the tibial tubercle, such that a post-operative distance of 12 mm (representative of a healthy population) was achieved.[8; 35] TP was simulated using radial basis functions (RBF) to morph the femoral trochlear geometry to deepen the sulcus.[24; 36] The sulcus angle was measured in the axial plane at the level of the most anterior prominence of the lateral ridge of the trochlear groove. The sulcus angle at this part of the groove was morphed to create a sulcus angle of 138°, the average angle reported in a control population.[37] The amount to which rest the groove was altered was linearly decreased moving down the groove to the intercondylar notch, which maintained its original sulcus angle. The femoral cartilage was morphed by the same magnitude as the underlying bone so that the original cartilage thickness of the trochlear groove was maintained. Trochlear deepening was simulated with RBF points selected with the knee flexed at 0°, 30°, 60° and 90° to allow for smooth adjustment of the femoral bone and cartilage along the entire length of the trochlear groove.

Incorporating loading and kinematic variability:

Each virtual surgery was evaluated under a variety of kinematic and external loading conditions to assess the robustness of each outcome to variability. Specifically, each subject and surgery combination were evaluated under ***12 kinematic conditions and three external loading***

conditions. The three loading conditions encompassed low, moderate and aggressive loads, represented as (a) knee extension from 50° flexion to 5° hyperextension (low load), (b) knee extension from 50° flexion to 5° hyperextension with a constant 50 N external load applied laterally to the patella (moderate load), and (c) knee fixed at 20° flexion with a ramped 200 N external load applied laterally to the patella (aggressive load). These loading levels were selected after evaluation of a range of loading conditions, such that "low" loading did not apply any additional external load; "moderate" loading resulted in dislocation in approximately half of the pre-operative simulations; and "aggressive" loading resulted in dislocation of all pre-operative simulations. To limit the number of confounding factors in our results, either the kinematics were changing (i.e. knee extension with a constant load) or the load was changing (i.e. fixed flexion angle with a ramped load), but not both simultaneously. This allowed us assess both the influence of flexion angle and external load magnitude on patellar stability in a reasonably direct manner.

Variation in TF kinematics focused on variation in TF I-E rotation, as this degree-of-freedom has the largest effect on altering the line-of-action of the quadriceps muscle. Variability in TF I-E rotation was derived from the same dataset as other TF motions used in this analysis: a biplane fluoroscopy analysis of 50 participants performing a weighted knee extension activity.[29] TF I-E profiles for the 50 fluoroscopy participants were extracted as a function of TF extension from 50° to 5° hyperextension. Principal component (PC) analysis was applied to these profiles and the PC scores were extracted for each subject. The first two PCs accounted for 98% of the total variability in the profiles, and so only these PCs were included in the instantiation of new I-E profiles. Latin hypercube sampling, constraining the PC scores to within ± 2 standard deviations, was used to randomly sample PC values to create a unique set of 12 new kinematic profiles used to represent the variability of the full dataset (**Fig. 2**). In total, this analysis resulted in simulation of 252 unique conditions for each subject-specific model (7 surgeries * 3 loading conditions * 12 kinematic conditions) (**Table 1**).

Statistical analysis:

A one-way mixed model analysis of variance (ANOVA) design was used to test the main effects of surgery (mTTO and TP, both with and without MPFLR). Separate ANOVAs were performed on patella I-E rotation, patella shift, and MPFL force during each external loading condition for each kinematic profile. For the "low" and "moderate" conditions, dependent metrics of stability were compared at 5° intervals. Patella stability during the "aggressive" condition was evaluated at the end of the application of the ramped load. Post-hoc pairwise comparisons between surgeries (TTO vs TP, TTO vs TTO without MPFLR, TP vs TP without MPFLR) were performed for each kinematic condition using the estimated marginal means. Significance level was set to alpha = 0.05 to test significance between the surgery groups, correcting for multiple comparisons using Bonferroni correction (correction factor 0.0167). R 4.1.3 was used for all statistical analyses.

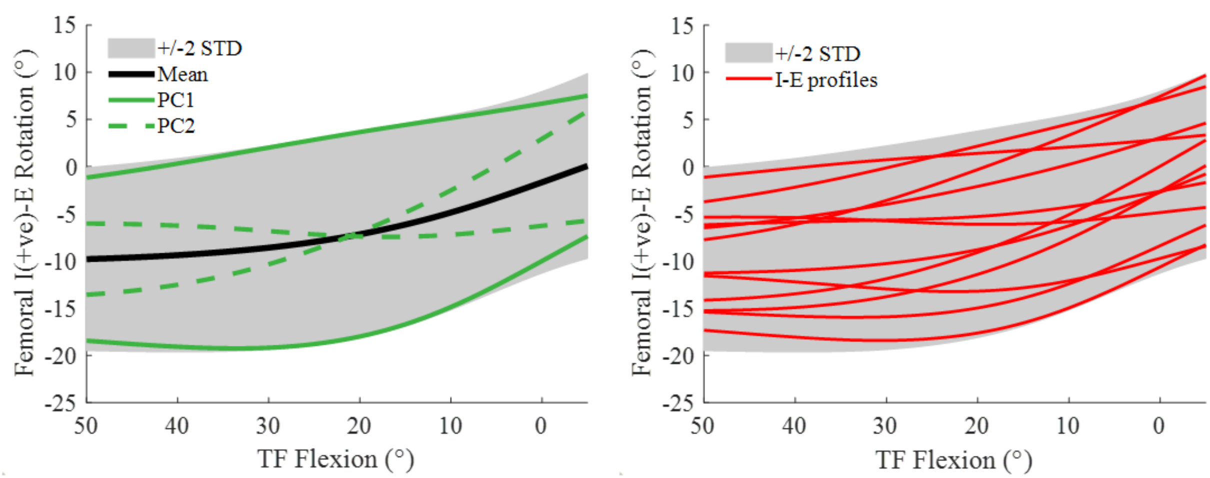

Figure 2: Mean and +/- 2 standard deviations of I-E profiles from the fluoroscopy dataset,[29] with +/- 2 standard deviations in PC1 and PC2 scores (left). +/- 2 standard deviations of I-E profiles from the fluoroscopy dataset, with 12 profiles derived from Latin hypercube sampling of PC1 and PC2 scores (right).

Table 1: Simulations for each subject comprised of a series of 7 surgeries, 3 external loading conditions and 12 TF I-E profiles.

Surgery	**Loading conditions**	**Tibiofemoral kinematics**
Preoperative MPFLR mTTO plus MPFLR mTTO w/o MPFLR TP plus MPFLR TP w/o MPFLR Healthy baseline	"**0 N**": No external load from 50° flexion to 5° hyperextension "**50 N**": Constant 50 N external load from 50° flexion to 5° hyperextension "**Ramp**": Ramped load from 0 – 200 N at 20° flexion	12 unique I-E profiles (as defined in **Fig. 2**)

RESULTS

Surgical management:

mTTO and TP surgeries (with both surgeries including MPFL representation) were compared across all 20 subjects. The primary metrics used to evaluate patellar stability in these simulations

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were force in the MPFL and M-L shift of the patella. When comparing across all kinematic and external loading conditions (36 simulations per surgery), there were statistically significant differences between mTTO and TP simulations. MPFL force averaged 102 N and 95 N for mTTO and TP simulations, respectively. When compared throughout flexion, there were consistent differences between mTTO and TP outcomes. Under low loading (0 N externally applied load during flexion), there were significant differences in MPFL force from 20° to 5° flexion (averaging 24 N and 14 N for mTTO and TP, respectively). Similarly, under moderate loading (50 N externally applied load during flexion), there were significant differences in MPFL force from 25° to 10° flexion (averaging 35 N and 18 N for mTTO and TP, respectively; **Fig. 3**). Both surgeries showed similar change in M-L shift from 50° to 5° flexion, while TP resulted in statistically significant more lateral position at full extension and 5° hyperextension as the patella left the femoral groove (**Fig. 3**). In the most aggressive loading condition (200 N externally applied load at 20° flexion), the TP simulations resulted in consistently less MPFL force generation (~20 N less) throughout the simulation, although this was not statistically significant (**Fig. 4**).

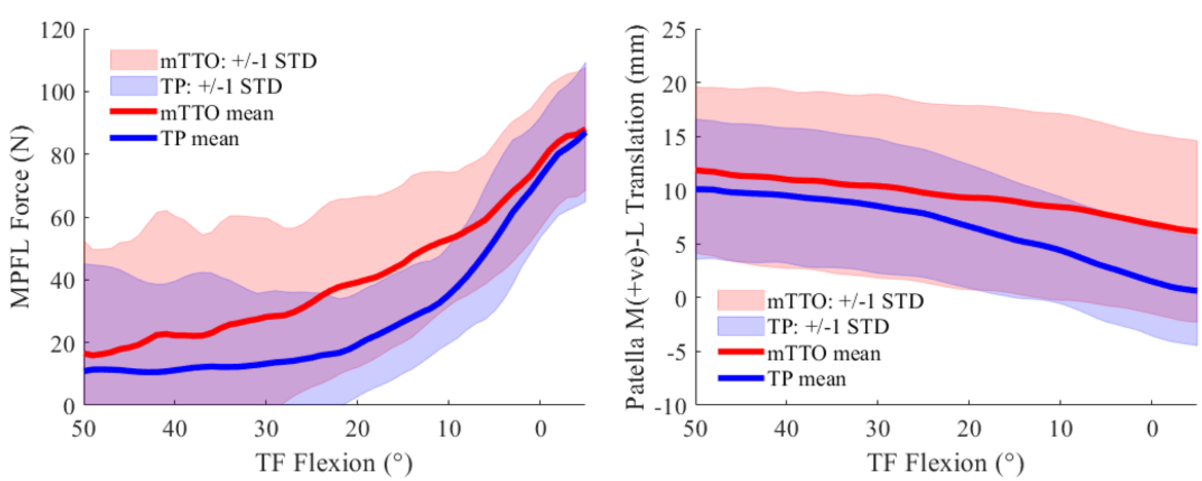

Figure 3: Mean and standard deviation of MPFL force (left) and patella M-L shift (right) during mTTO and TP simulations, averaged across all kinematic simulations with an external load of 50 N applied to the patella. Similar results were observed with an external load of 0 N.

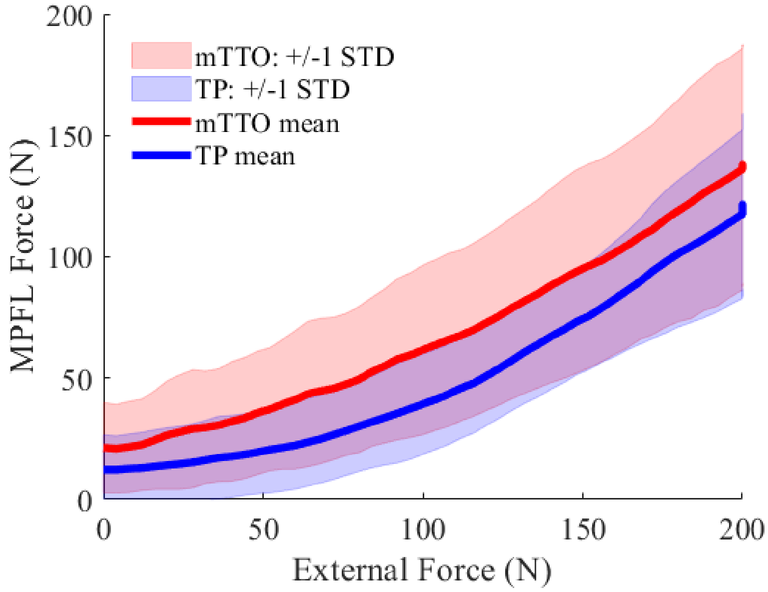

Figure 4: Mean and standard deviation of MPFL force during mTTO and TP simulations averaged across all kinematic simulations with an external load of 200 N applied to the patella and TF flexion angle held at 20°.

MPFL reconstruction:

In order to evaluate whether MPFLR is necessary when combined with a mTTO or TP procedure, PF stability after mTTO and TP surgeries were compared with and without MPFLR. For both mTTO and TP surgeries, the presence of an MPFL structure halved the laxity measurements of the PF joint. For mTTO simulations, patellar I-E rotation averaged 12.7° and 24.6° with and without MPFL, respectively, while patellar M-L shift averaged 7.4 mm and 14.7

mm, with and without MPFL, respectively. Similarly, for TP simulations patellar I-E rotation averaged 16.2° and 36.6° with and without MPFL, respectively, while patellar M-L shift averaged 10.1 mm and 19.5 mm with and without MPFL, respectively (**Fig. 5**). These differences were more apparent with increasing external load: at 0 N external load (low) differences were statistically significant from 5° flexion to 5° hyperextension; at 50 N external load (moderate), differences were statistically significant from 15° flexion to 5° hyperextension; while with a 200 N ramped load (aggressive) differences were statistically significant at 20° flexion.

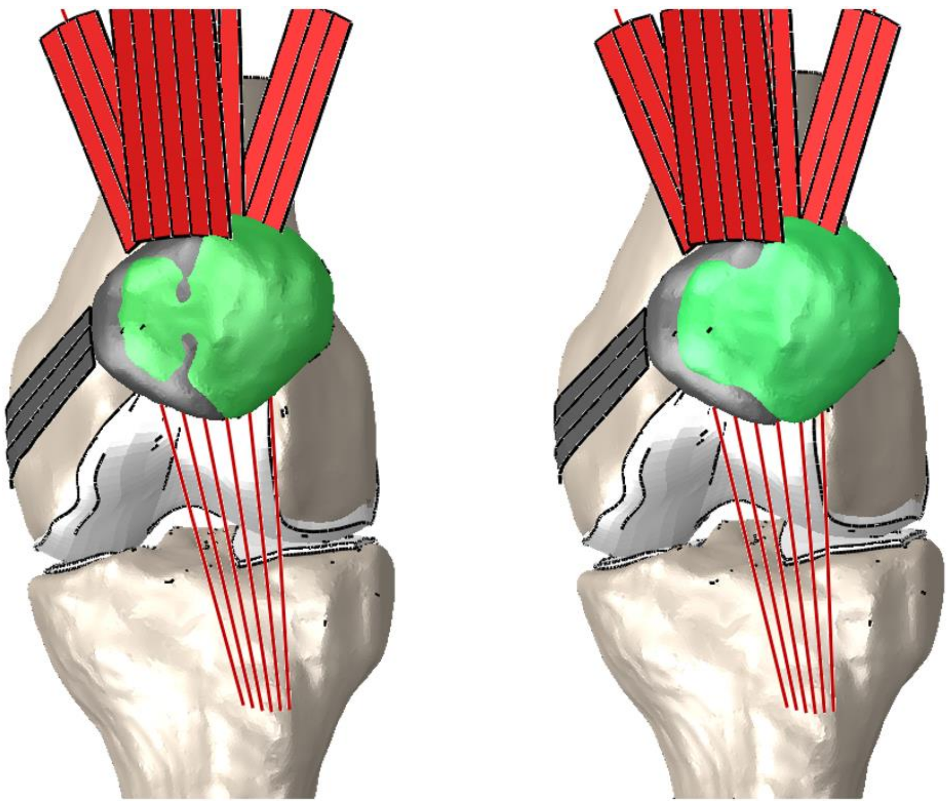

Figure 5: Patella position near full extension with (gray) and without (green) MPFL constraint under low (0 N external load – left) and moderate (50 N external load – right) loading conditions (shown for a representative subject).

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Kinematic variability:

Variation in outcomes due to TF I-E kinematic variability was evaluated by measuring the standard deviation in simulation outcomes (MPFL force, PF I-E rotation and PF M-L shift) across the 12 kinematic trials. Across simulations that did not dislocate, the average standard deviations across the 12 kinematics trials were 8.2 N MPFL force, 2.6° PF I-E rotation, and 2.1 mm PF M-L shift across all subjects and loading conditions. Simulations that did not include any MPFL restraint (preop, mTTO w/o MPFL, TP w/o MPFL) were more sensitive to kinematic variation; average standard deviations across kinematic trials were 1.9° and 4.0° PF I-E rotation, and 1.8 mm and 2.7 mm PF M-L shift, for simulations with and without MPFL, respectively (**Fig. 6**).

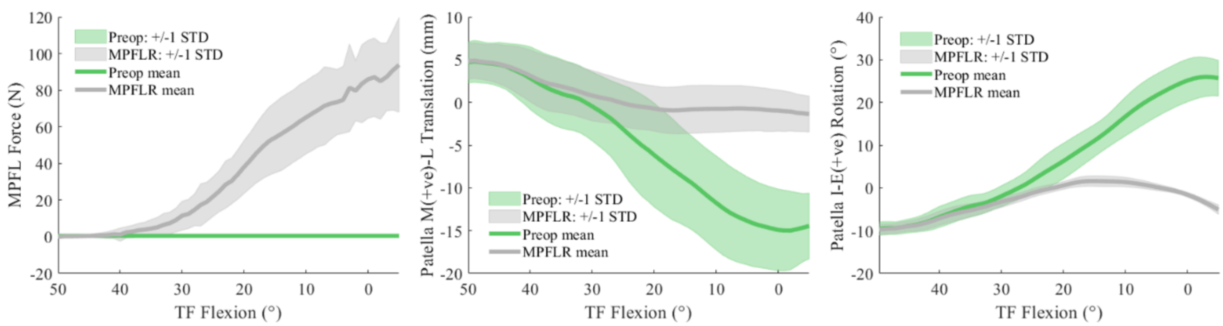

Figure 6: Variability in MPFL force (left), patellar shift (center), and patellar rotation (right) with varying TF I-E kinematics. Shown here for one representative subject during pre-operative (without MPFL) and MPFLR simulations under low (0 N external load) loading conditions.

Frequently, the TF kinematic profile was a determining factor in whether or not a dislocation occurred. In the 0 N loading condition, 10 subjects (50%) had simulations that dislocated in some, but not all, kinematic trials for a given surgical intervention (primarily occurring during the preoperative simulations). This increased to 100% of subjects with the 50 N loading condition.

Loading conditions:

Patellar dislocation was very sensitive to the external loading condition applied. The most aggressive loading condition ("ramp") resulted in dislocation in every simulation where there was no MPFL constraint (preop, mTTO w/o MPFL, TP w/o MPFL) and 10% dislocations in all other surgical simulations. With the "50 N" loading condition, dislocation occurred during 55% of preoperative simulations, 32% and 30%, respectively, in mTTO w/o MPFL and TP w/o MPFL surgeries, and 1% across all conditions with an MPFL. When no external load was applied ("0 N"), dislocation occurred during 23% of preoperative simulations, and 8% and 6%, respectively, in mTTO w/o MPFL and TP w/o MPFL surgeries, with no dislocations across all other conditions (**Table 2**).

Table 2: Percentage of dislocated simulations per loading condition and surgery type.

	Preop	**MPFLR**	**mTTO**	**mTTO w/o MPFL**	**TP**	**TP w/o MPFL**	**Healthy baseline**
0 N	23.3	0	0	7.9	0	5.8	0
50 N	54.6	4.2	0	31.7	0	30.4	0.4
Ramp	100	20.8	4.6	100	3.3	100	1.7

DISCUSSION

mTTO simulations consistently resulted in greater MPFL force generation than TP simulations, indicating that TP simulations provide additional anatomic restraint and less reliance on soft-tissue constraint to prevent dislocation. These differences were most significant between 25° and 5° flexion; in later flexion the trochlear groove provided similar constraint between surgical groups, while in very early flexion, the patella often sits above the femoral groove so there is little anatomic trochlear constraint close to full extension. For both mTTO and TP surgeries, it was necessary to also perform MPFLR — inclusion of MPFLR halved the magnitude of kinematic laxity (M-L and I-E motions) predicted in the models with low and moderate external loads. With a ramped 200 N external loading condition, all simulations without an MPFL resulted in dislocation.

Simulations that did not include an MPFL structure were not robust to variation in TF I-E kinematics. Without inclusion of an MPFL structure, patellar dislocation at both low and moderate loading conditions was dependent on the TF I-E kinematic profile applied, while with an aggressive loading condition, dislocation occurred in all simulations. Most clinical studies reported in the literature include some type of MPFLR procedure, regardless of whether another restorative procedure is also performed.[17; 38; 39] The results from this analysis support that approach – patellar stability after both mTTO and TP simulations benefited significantly from the inclusion of the MPFL. Most computational evaluations of patellar stability have used a single kinematic condition per knee[18; 24; 40] — however, this study assesses the robustness of patellar joint stability to kinematic variability. When performing cadaveric or computational analyses to quantify the effectiveness of a particular procedure on patellar stability, it is important to evaluate this procedure over a variety of kinematic and loading conditions, rather than a single deterministic set of conditions, to ensure that the procedure is robust to variation encountered across the patient population.

MPFL reconstruction is commonly studied when quantifying patellar stability and recurrent patellar dislocation. Several researchers have performed cadaveric studies to quantify the role of medial stabilizers on lateral patellar stability in healthy knees.[41; 42] In computational analyses based on patients treated for recurrent lateral patellar instability, Elias et al.[43] and Tanaka et al.[44] assessed the influence of MPFLR and the graft tension, concluding that MPFLR reduces but does not eliminate patellar instability when trochlear dysplasia and high TT-TG distance are present. The same research group also performed extensive computational analyses to quantify the role of tibial tubercle position and osteotomy on patellar stability.[45; 46] Biomechanical analyses focusing on trochlear dysplasia and trochleoplasty have become increasingly prevalent in recent years. Given the difficulty of obtaining cadaveric specimens with trochlear dysplasia, several studies have artificially simulated trochlear flattening.[16-18; 47] Experimentally, Vinod et al.[17] performed a series of trochlear flattening procedures inserting prosthetic wedges varying from 0° to 40° to incrementally flatten the sulcus angle, finding a reasonably linear decrease in lateral force required to displace the patella with increasing sulcus angle. Computationally, Kaiser et al.[18] applied finite element analysis to quantify patellofemoral biomechanics in models with artificially generated trochlear dysplasia and subsequent trochleoplasty, concluding that trochleoplasty restored patellar stability to levels found in healthy knees from 10° to 45° flexion, but were similar to knees with trochlear dysplasia at full extension. However, the current study is the first to directly compare between mTTO and trochleoplasty procedures in a patient cohort with trochlear dysplasia and recurrent lateral patella instability.

There is little consensus on which surgery is best to perform for a patient population with trochlear dysplasia and recurrent patellar dislocation.[48] The comparative analysis presented in this study shows clear differences in the anatomic restraint between the mTTO and TP procedures. While a trochleoplasty provides improved patellar stability in the 25° to 5° flexion range where patellar dislocation typically occurs, surgical decision making should also account for other subject-specific anatomical and activity preferences. A patient with a longer patellar tendon (patella alta) would likely get less benefit from a TP procedure, as their patella would exit the trochlear groove at an earlier point in the extension activity. However, a patient whose sport or activity frequently involved the participant planting and pivoting their leg (i.e. knee flexed 20-30°, quadriceps engaged, I-E rotation of the TF joint) would likely see additional benefit from a TP over a mTTO during this vulnerable kinematic and loading condition.

There are a number of limitations and simplifications associated with this work. The MPFL structure does not include a rupture criterion. In an experiment, or in practice, an aggressive loading condition may ultimately cause rupture and failure of the MPFL, which would likely result in an abrupt dislocation of the patella. While we do not directly simulate these failures, generation of high MPFL force in our models is an indicator that these simulations are vulnerable to dislocation. In this analysis, we represented the MPFLR using force-length characteristics calibrated to native MPFL parameters. In prior work, we perturbed MPFL stiffness and found consistent, but relatively small changes in kinematic results as due to changes in MPFL stiffness.[24] Additionally, in a cadaveric study, Vinod et al.[17] compared knees with native MPFL and reconstructed MPFL and found no statistically significant difference in the force required to laterally displace the patella by 10 mm. Hence, we do not expect that our choice of native MPFL parameters should have a significant impact on the outcomes of this study. We simulated only a simplified set of three external loading conditions (extension; extension with a 50 N load applied directly to the patella; and 20° flexion with a 200 N load applied directly to the patella). These activities do not directly represent the pivoting motion at low flexion (quadriceps engaged, internal femoral rotation) that has been associated with patellar dislocation.[1] A more complex loading condition may be required to simulate a high-risk condition for dislocation in a more physiological manner. We applied a single quadriceps load across all of our simulations. We expect that changing the magnitude of the quadriceps load will certainly alter the magnitude of patellar displacement under a consistent laterally applied load. However, we expect that the trends in our results and the relative comparison between simulations will not be altered. Although the combination of muscle load, soft-tissue restraint and joint motion will vary from patient to patient and the loading conditions resulting in dislocation may be generated in numerous and complex ways, this comparative study allows us to predict the *relative differences* in outcomes and estimate the relative amount of patellar restraint provided by anatomic and soft-tissue structures between mTTO and TP simulations.

Unfortunately, it was not possible to obtain patient-specific kinematics for these patients to directly validate kinematic predictions from the model. In an ongoing study with data from a separate patient cohort, we are working to obtain follow-up data on clinical outcomes of their surgery, however, this was beyond the scope of the current study. Instead, the model is based on a previously published isolated-PF model (i.e. TF soft-tissues are not included and TF kinematics are prescribed). The isolated-PF model[27] has demonstrated fidelity in reproducing patellar kinematics with an average root-mean-square difference of 3.1° and 1.7 mm for all rotations and translations when compared to kinematics from cadaveric specimens performing gait and deep knee bend activities in an experimental knee simulator. The current model was updated to include multiple quadriceps bundles for more physiological distribution of quadriceps load across the tendon-patella interface.[34] A limitation of this work is that these cadaveric knees did not have signs of patella instability or trochlear dysplasia, and so we have yet to validate the model kinematics with dysplastic knees or the specific lateral patellar loading condition applied in these simulations.

Joint mechanics and cartilage stress are also important considerations for the long-term clinical outcome of a joint stabilizing procedure. A recent computational study evaluated the effect of trochleoplasty on both lateral patellar tracking and contact mechanics, showing decreased lateral tracking but 13-23% increase in peak contact pressure.[49] Previous cadaveric and computational studies have reported increases in mean medial trochlear contact pressure with tibial tubercle anteriomedialization,[45; 50] while in a cadaveric study, Stephen et al.[51] reported attachment location and tensioning during MPFLR altered medial joint contact pressure. Our study has specifically focused on evaluation of joint stability and patellar kinematics. Additional work is needed to account for other joint mechanics factors that should also be included to assess long-term efficacy or risk of patellofemoral osteoarthritis as a result of these interventions. This will assist in determining if the decision on optimal treatment is altered by these longer-term considerations, particularly in patients with medial chondral defects who may be more vulnerable to increased medial forces.

In this cohort of patients with trochlear dysplasia and recurrent patellar dislocation, trochleoplasty provided greater patellar stability than mTTO between 5° to 25° flexion but did not show an advantage over mTTO at full extension. This aligns with Kaiser et al.[18] who reported that post-trocheoplasty knees had similar patellar kinematics to knees with trochlear dysplasia at full extension and similar patellar kinematics to healthy knees in later flexion. Regardless of whether an mTTO or trochleoplasty procedure was performed, the addition of an MPFLR consistently improved patellar stability, halving the amount of M-L movement at the patellofemoral joint. Finally, patellar stability was very sensitive to tibiofemoral kinematics; whether or not a simulation resulted in dislocation was affected by the TF kinematic profile. Simulation outcomes were increasingly sensitive to TF kinematics with more aggressive external loading conditions. Although further work in a longitudinal clinical study is required to validate computational predictions of if and when a particular knee may suffer a patellar dislocation, this work illustrates the trends in anatomic restraint, soft-tissue restraint, and joint mobility that we may expect between mTTO and TP procedures.

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FIGURE LEGENDS:

Figure 1: Finite element model of the knee including bone and cartilage, patellar tendon, quadriceps tendon and muscle and (where applicable) MFPL. Shown here (from left to right) for the preop, MPFLR, mTTO, and TP models of a representative subject in coronal (top) and axial (bottom) views.

Figure 2: Mean and +/- 2 standard deviations of I-E profiles from the fluoroscopy dataset,[29] with +/- 2 standard deviations in PC1 and PC2 scores (left). +/- 2 standard deviations of I-E profiles from the fluoroscopy dataset, with 12 profiles derived from Latin hypercube sampling of PC1 and PC2 scores (right).

Figure 3: Mean and standard deviation of MPFL force (left) and patella M-L shift (right) during mTTO and TP simulations, averaged across all kinematic simulations with an external load of 50 N applied to the patella. Similar results were observed with an external load of 0 N.

Figure 4: Mean and standard deviation of MPFL force during mTTO and TP simulations averaged across all kinematic simulations with an external load of 200 N applied to the patella and TF flexion angle held at 20°.

Figure 5: Patella position near full extension with (gray) and without (green) MPFL constraint under low (0 N external load – left) and moderate (50 N external load – right) loading conditions (shown for a representative subject).

Figure 6: Variability in MPFL force (left), patellar shift (center), and patellar rotation (right) with varying TF I-E kinematics. Shown here for one representative subject during pre-operative (without MPFL) and MPFLR simulations under low (0 N external load) loading conditions.