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Computational study on extra-articular knee structures preventing the anterior cruciate ligament from re-rupturing using the GHBMC M50-PS

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Abstract – An anterior cruciate ligament (ACL) rupture is a frequent injury in athletes and may lead to a reduced activity level and subsequent joint lesions due to a non-compensable rotational instability of the human knee. In most cases, singular ACL reconstruction can restore the anterior-posterior and rotational stability. However, in some cases, rotational instability with positive pivot phenomenon persists and raises questions about additional extra-articular structures supporting the ACL in stabilizing the knee under rotational loading. The present study aims to evaluate the influence of such extra-articular structures and their impact on the rotational stability of the knee by using an isolated FE knee model extracted from the full body model of the GHBMC M50-PS. The main contributors to relieve loads on the ACL and to increase rotational stability for a given rotation of the femur will be identified.

Keywords: Finite element analysis; human knee; anterior cruciate ligament; anterolateral ligament; GHBMC.

NOTATION

ACL	Anterior Cruciate Ligament
ALL	Anterolateral Ligament
AML	Anteromedial Ligament
FE	Finite Elements
GHBMC	Global Human Body Model Consortium
LCL	Lateral Collateral Ligament
M50	Male 50th-Percentile
MCL	Medial Collateral Ligament
PCL	Posterior Cruciate Ligament
PLT	Popliteus Tendon
PMHS	Post Mortem Human Subject
POL	Posterior Oblique Ligament
PS	Pedestrian Simplified

MOTIVATION

An anterior cruciate ligament (ACL) rupture is a frequent injury in athletes and may lead to a reduced activity level and subsequent joint lesions due to a non-compensable rotational instability of the human knee [1]. In most cases, singular ACL reconstruction can restore anterior-posterior and rotational stability [2, 3]. However, in some cases, rotational instability with positive pivot phenomenon persists and raises questions about additional extra-articular knee structures supporting the ACL in stabilizing the knee under rotational motion [4-6]. The anterolateral ligament (ALL) is the current figurehead of such structures and several surgeons add the ALL reconstruction to the surgical treatment of patients with ACL re-ruptures, high-grade pivot phenomenon or persisting pivot phenomenon after ACL reconstruction [4, 6, 7].

However, in this context, a contradiction is apparent: common injury mechanisms of an ACL rupture occur (besides valgus stressing) not only due to femoral external rotations (tibia internal rotations), but also due to femoral internal rotations (tibia external rotations) [8, 9]. The ALL is known for preventing the external rotation of the femur only and it can be doubted that the ALL protects the ACL in other injury mechanisms as well. Therefore, the presented human knee FE study evaluates which extra-articular structure optimally support the ACL in excessive femoral internal and external rotations. The authors hypothesize that the ALL does not contribute to the protection of the ACL in injury mechanisms including femoral internal rotation.

METHODS

The left knee of the GHBMC M50-PS full human body finite element model with anatomical ACL, posterior cruciate ligament (PCL), lateral collateral ligament (LCL), medial collateral ligament (MCL) and an intact medial and lateral meniscus has been isolated from the full body model (Figure 1).

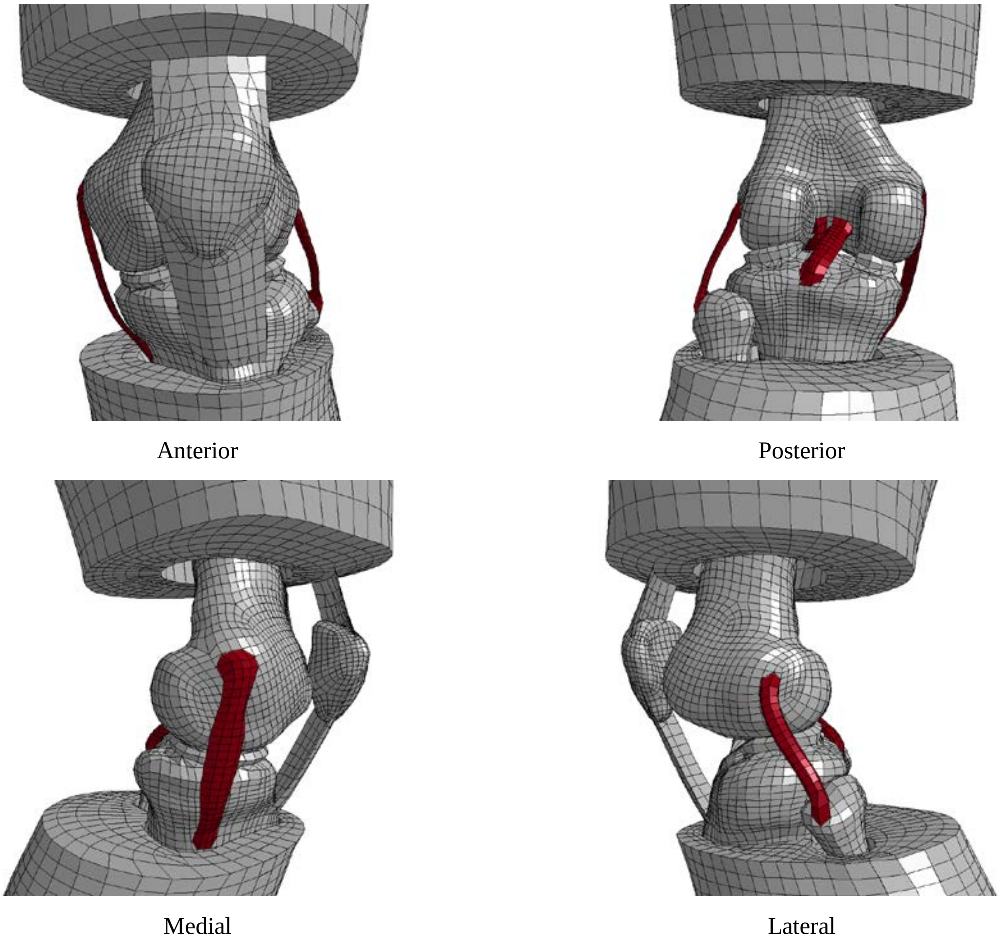


Figure 1: Base FE knee model of the GHBMC M50-PS in an undeformed configuration at an inclination of 25 degrees. The cruciate and lateral ligaments are marked in red. The soft tissue in the central region is hidden to improve visualization.

The GHBMC full human body model represents a 50th percentile male in an upright standing posture. The model has been developed by GHBMC [10] and is commonly used for traffic accident simulations involving human beings as occupants or pedestrians in combination with the explicit finite element solver LS-DYNA [11]. All properties including material definitions, element formulations, kinematical constraints and contact surfaces defined within the model have been applied to this study unchanged. The knee angle has been carefully adjusted to 25 degrees by pre-positioning in order to establish a more critical reference configuration for ACL ruptures. The material definition assigned to the existing ligaments in the base model with distinct properties for tension and compression (*MAT_PLASTICITY_COMPRESSION_TENSION) has also been used for the extra-articular structures added to the base model. In this model, Untariou et al. [12] assigned the average tensile stress-strain curve reported by Quapp et al. [13] and a less stiff curve for tension and compression, respectively.

Problem setup

The isolated FE model has been cut approximatively 170 mm above and 140 mm below the tibia plateau. The exact cutting edges are along the given spatial discretization. The model consists of roughly 50,000 elements and 37,000 nodes. All physical components (bones, muscles, ligaments, soft tissue) are organized by 32 components totally. The isolated knee model encompasses a volume of 3.7 L and a mass of 4.2 kg. The nodes on both cutting surfaces, on the femoral side and on the tibia side, are kinematically constrained so that no relative displacement of the nodes is allowed, i.e. the sectional areas are undeformable. While the rigid section of the tibia is fixed in space, the rigid section of the femur is loaded nearly instantaneously by a constant moment of 20 Nm in order to establish both loading scenarios, a femoral external rotation and a femoral internal rotation. The simulation duration has been set to 100 ms. Since the final nodal positions are almost constant after approximately 60 ms, the final deflection angle and the normal forces within the ligaments have been averaged from this point of time until the end of the simulation.

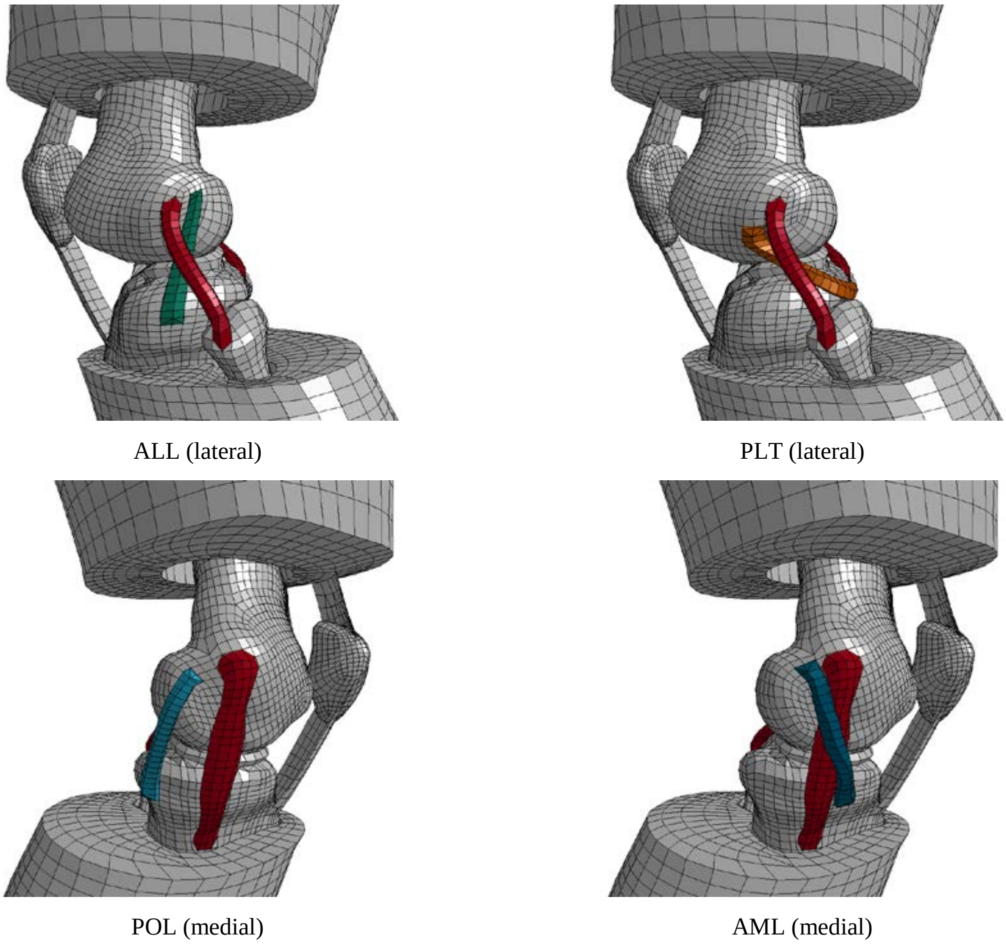


Figure 2: Model variants with various extra-articular ligaments added to the standard GHBMC M50-PS in an undeformed configuration at an inclination of 25 degrees. The soft tissue in the central region is hidden to improve visualization.

Ligaments and extra-articular structures

Four additional anatomic structures (anterolateral ligament, anteromedial ligament, popliteal tendon and posterior oblique ligament) which were believed to be able to support the ACL were added to the human knee model separately and then all together (Figure 2). Internal and external rotations were applied to the femur with and without the four additional anatomic structures. The normal force histories within the ACL, PCL, LCL and MCL and in each extra-articular structure were monitored and the rotational deflection of the rotated body part was determined for each case.

RESULTS

In the FE base model without any additional extra-articular structure, the ACL was the most loaded ligament for both, femoral internal and femoral external rotation (Table 1). The least loaded ligament was the LCL in femoral external rotation and the PCL in femoral internal rotation. When adding extra-articular structures individually to the base model, the results listed in Table 1 will undergo the subsequent changes.

Table 1: Computational results for the base FE model without any additional extra-articular structures at final equilibrium position. Listed are averaged results and standard deviations after 60 ms.

Component	Quantity	Unit	External rotation	Internal rotation
Femur	Angle	Degrees	-24.6 (± 0.0)	11.9 (± 0.0)
LCL	Force	N	-5.0 (± 0.1)	109.6 (± 0.1)
MCL	Force	N	73.3 (± 0.2)	61.4 (± 0.0)
ACL	Force	N	236.7 (± 0.4)	149.5 (± 0.2)
PCL	Force	N	177.0 (± 0.3)	56.9 (± 0.1)

Table 2: Influence of extra-articular knee structures on the femoral deflection angle. Negative values denote femoral external rotations. Positive values denote femoral internal rotations.

Variant	Unit	External rotation		Internal rotation	
Base model	deg		-24.6		11.9
ALL only	deg	(-18 %)	-20.1	(-3 %)	11.5
PLT only	deg	(-8 %)	-22.6	(-15 %)	10.0
POL only	deg	(-15 %)	-20.9	(-1 %)	11.7
AML only	deg	(-3 %)	-23.8	(-8 %)	10.8
All extra ligs	deg	(-39 %)	-14.8	(-26 %)	8.7

Table 3: Influence of extra-articular knee structures on the ACL section force.

Variant	Unit	External rotation		Internal rotation	
Base model	N		236.7		149.5
ALL only	N	(-21 %)	186.0	(-2 %)	145.5
PLT only	N	(-6 %)	220.8	(+1 %)	152.4
POL only	N	(-8 %)	215.7	(-1 %)	146.7
AML only	N	(-1 %)	232.7	(-9 %)	135.3
All extra ligs	N	(-42 %)	136.0	(-14 %)	128.5

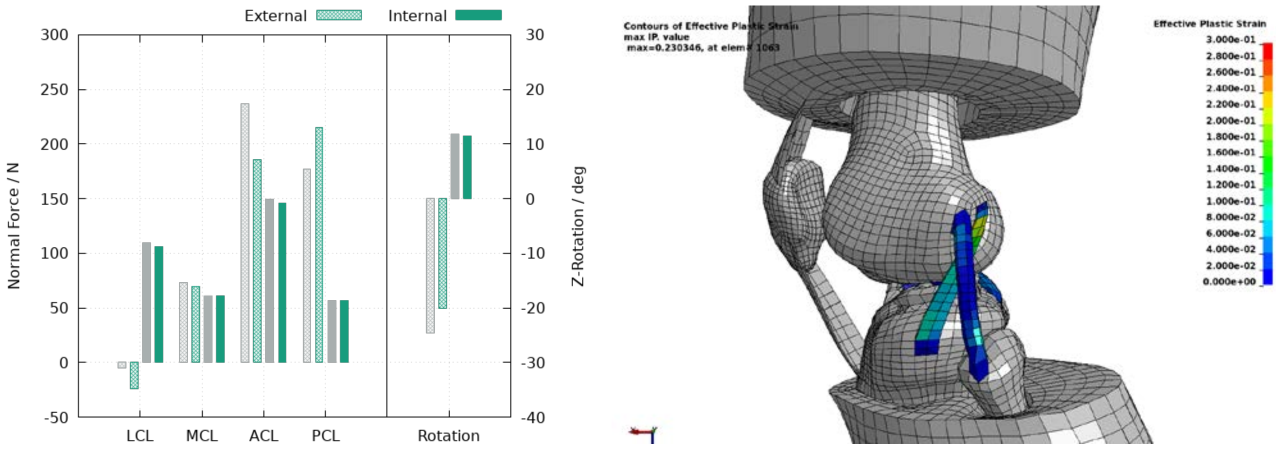
Table 4: Influence of extra-articular knee structures on the PCL section force.

Variant	Unit	External rotation		Internal rotation	
Base model	N		177.0		56.9
ALL only	N	(+21 %)	215.0	(0 %)	56.3
PLT only	N	(+1 %)	179.4	(-55 %)	25.4
POL only	N	(-34 %)	115.1	(+3 %)	59.0
AML only	N	(-6 %)	165.8	(+6 %)	60.6
All extra ligs	N	(-13 %)	152.8	(-27 %)	41.0

Femoral deflection angle and cruciate ligaments’ section forces

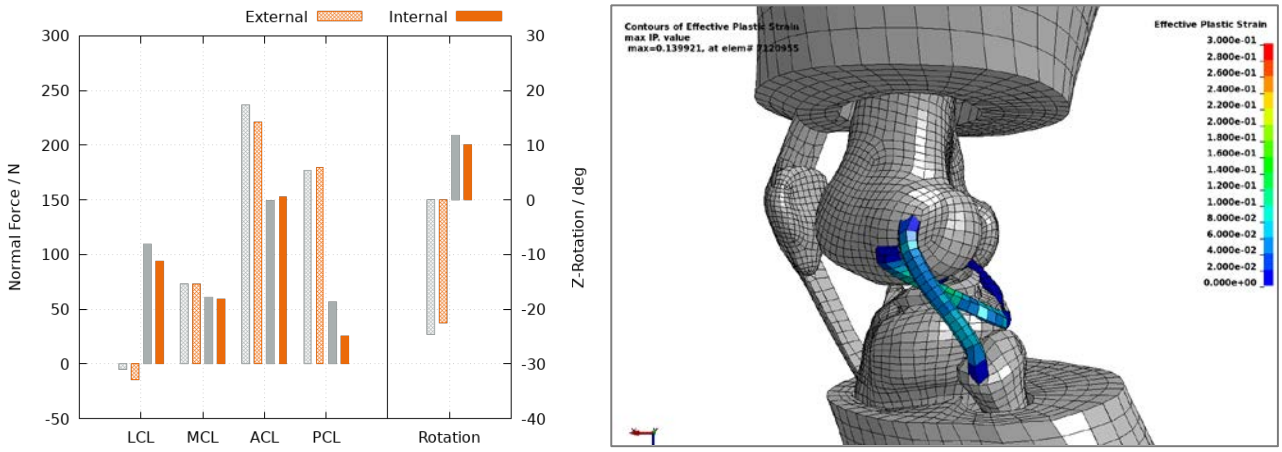
Table 2 lists the changes of the femoral deflection angle in the final position. Obviously, the ALL (-18 %) and the POL (-15 %) are the main stabilizers of the knee for femoral external rotations, whereas the PLT (-15 %) and the AML (-8 %) are the main stabilizers of the knee for femoral internal rotations. Table 3 lists the trends for the force carried by the ACL in the final position under the given load. While the ALL (-21 %) is the main contributor for reducing the load in the ACL under femoral external rotations, the PLT (-6 %) and the POL (-8 %) also take small contributions for relieving the ACL. The most interesting result is that the AML (-9 %) seems to be the only ligament that significantly contributes to unloading the ACL for femoral internal rotations: ALL, PLT and POL do not have noticeable contributions for relieving the ACL load in this case. Table 4 summarizes the trends for the resulting force carried by the PCL in the final position. Here, the POL (-34 %) is the main contributor for relieving the load in the PCL for femoral external rotations, while the PLT (-55 %) is the main contributor when being loaded by femoral internal rotations.

The computational results for each extra-articular structure are visualized in Figure 3 (ALL), Figure 4 (PLT), Figure 5 (POL) and Figure 6 (AML) on the subsequent pages. Figure 7 shows the results when all extra-articular structures are enabled simultaneously in the FE model. In these measurements, the load on the ACL was significantly reduced for both femoral external rotations (-42 %) and femoral internal rotations (-14 %), see also Table 3. With respect to these maximum values, the AML accounts for 67 % of the total ACL load reduction for femoral internal rotations, whereas 50 % of the total ACL load reduction for femoral external rotations is due to the ALL.



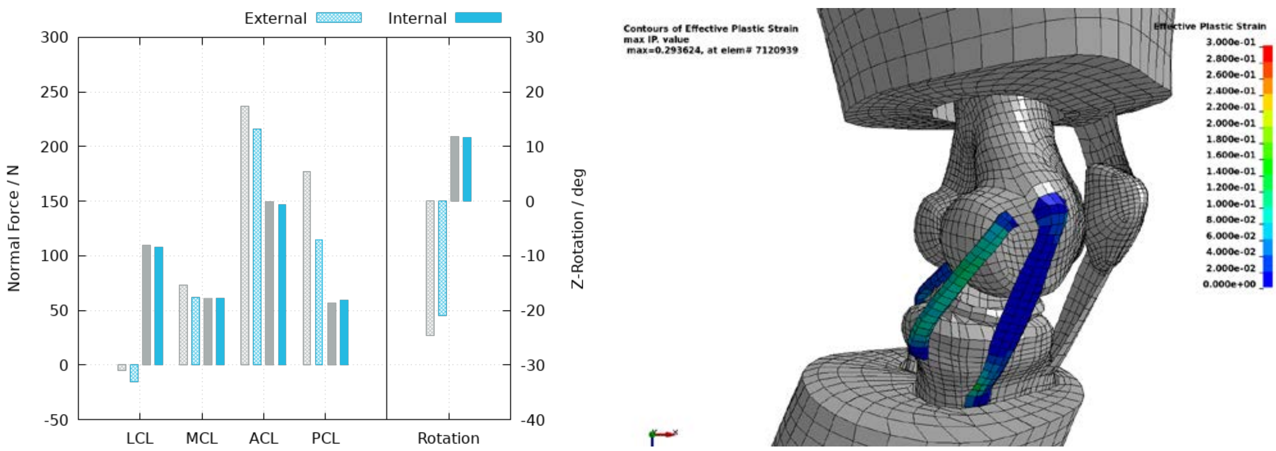
Results ALL (lateral)

Figure 3: Computational results when adding the ALL to the FE base model. Left: Influence on the normal section forces of the existing ligaments under femoral external rotation (shaded boxes) and femoral internal rotation (solid boxes). Right: Contours of effective plastic strain at the final position under external rotation.



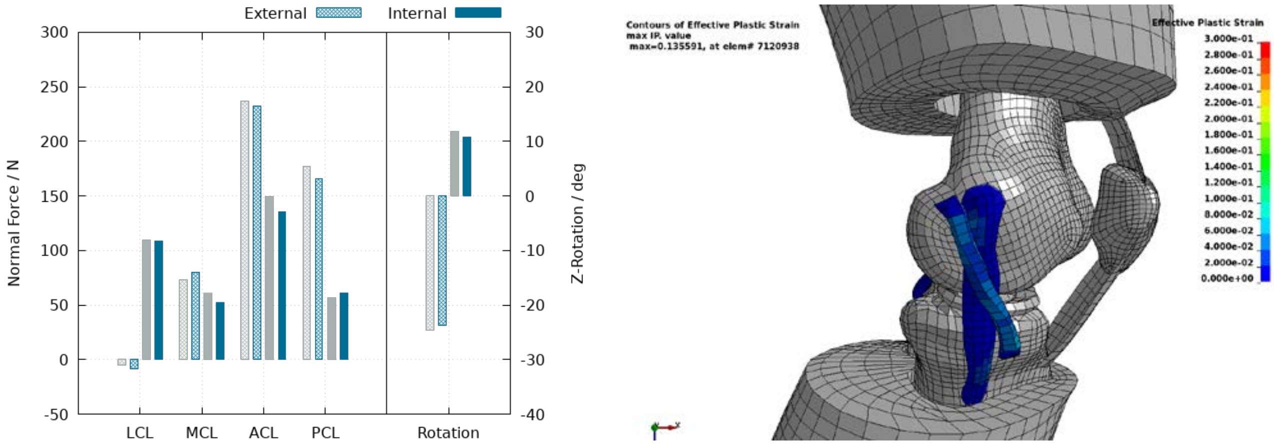
Results PLT (lateral)

Figure 4: Computational results when adding the PLT to the FE base model. Left: Influence on the normal section forces of the existing ligaments under femoral external rotation (shaded boxes) and femoral internal rotation (solid boxes). Right: Contours of effective plastic strain at the final position for the internal rotation.



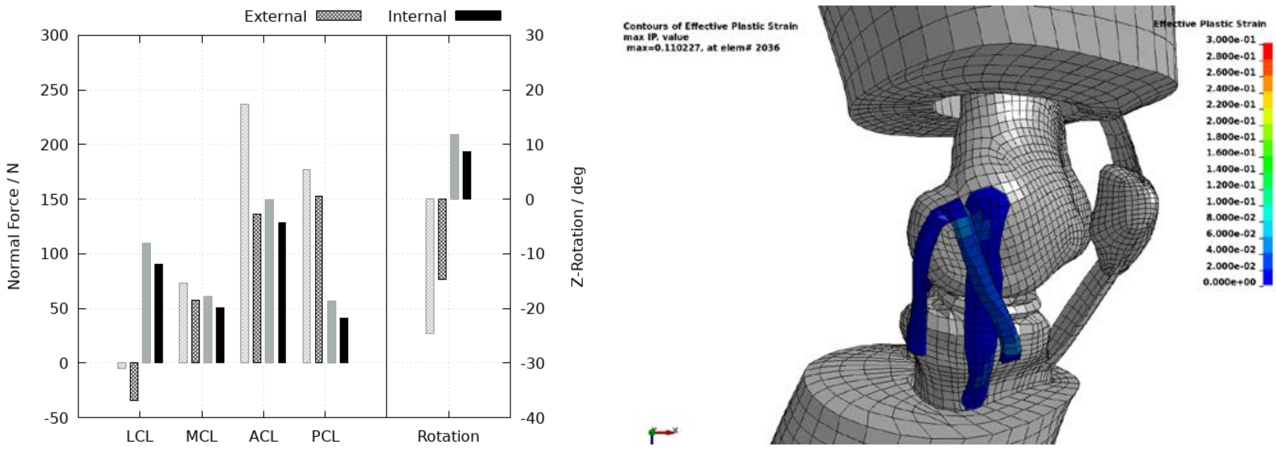
Results POL (medial)

Figure 5: Computational results when adding the POL to the FE base model. Left: Influence on the normal section forces of the existing ligaments under femoral external rotation (shaded boxes) and femoral internal rotation (solid boxes). Right: Contours of effective plastic strain at the final position for the external rotation.



Results AML (medial)

Figure 6: Computational results when adding the AML to the FE base model. Left: Influence on the normal section forces of the existing ligaments under femoral external rotation (shaded boxes) and femoral internal rotation (solid boxes). Right: Contours of effective plastic strain at the final position for the internal rotation.



Results All ligaments (medial)

Figure 7: Computational results when adding all extra ligaments to the FE base model. Left: Influence on the normal section forces of the existing ligaments under femoral external rotation (shaded boxes) and femoral internal rotation (solid boxes). Right: Contours of effective plastic strain at the final position for the internal rotation.

DISCUSSION

The present study showed that the ACL is a structure at risk for both, femoral internal and femoral external rotations. However, in femoral internal rotation, which represents a common and crucial injury mechanism for ACL ruptures, the ACL is mainly protected by the anterior part (AML) of the medial collateral ligament (MCL) and not by the ALL.

The ACL hinders femoral internal and external rotations. It sustains significant tensile loads in both loading directions. Therefore, ACL ruptures occur in motion patterns which consist of one of these joint positions at the time of trauma. Femoral external rotation is part of several injury mechanisms of an ACL rupture. Moreover, the ALL is known for stabilizing the femoral external rotation and thus protecting the ACL in these mechanisms. However, femoral internal rotation, often combined with valgus stressing, is also a common injury mechanism in ACL ruptures [8, 9]. Consequently, pathologically increased femoral internal rotation must exist in several patients with ACL insufficiency. Therefore, a structure that has the potential to support the ACL in restoring rotational stability or to protect the ACL from damage in these common injury mechanisms must decrease femoral internal rotation.

In the present study, the anterior part of the MCL (the authors named it anteromedial ligament) showed to have the best ability to decrease femoral internal rotation and seems to play a more important role in the prevention of ACL ruptures than expected. This relationship makes sense, because the oblique course of this ligament at the anterior medial knee is predisposing in stabilizing the femoral internal rotation, and a lesion of this structure is a common concomitant finding in patients with ACL rupture. These lesions often may not be adequately treated, which leads to residual valgus instability and possibly also to increased femoral internal rotation with rotational instability and high-grade positive pivot phenomenon. Thus, the reconstruction or augmentation of the anterior part of the MCL (AML) should be considered especially for patients with ACL re-ruptures, high-grade pivot phenomenon or persisting pivot phenomenon after ACL reconstruction.

Limitations

The present study has some computational limitations. The human knee model is merely an image of the physical knee and contains geometric, biomechanical and material simplifications so that only biomechanical tendencies may be captured. With reference to the applied FE model, the authors have identified the following limitations and deficiencies, which shall be addressed in future work:

- Dynamic effects due to forces of inertia become apparent due to the nearly instantaneous loading of the femur. These effects lead to noticeable peaks in the initial force histories in all ligaments. Applying the external load on the femur more slowly will raise the computational requirements but will reduce peak forces in the initial region of the force histories.
- The components of the FE base model of the GHBMC are enveloped in a virtual hull component that prevents the inner components from non-physical movements (Part ID 7000131) In the present study, it was not possible to model the ALL to lie completely within the given hull component.
- Pre-simulation for establishing a knee flexure of 25° induces stresses in the tissue and the ligaments that may be relevant for the evaluation of the numerical results. In the current study, it has been assumed that these pre-stresses are small compared to those induced by the rotational load, hence pre-stresses have been ignored.
- The material models of the GHBMC M50-PS have been designed for efficiency in traffic crash simulations involving the FE model as a full body model. It needs to be verified that the simplification of Untaroiu et al. [12] made to the ligaments' material definition of Quapp et al. [13] in terms of substituting an anisotropic hyper-elastic model by an isotropic elastic-plastic model with tension-compression anisotropy only is still valid for the scope of the present study. This is especially important when a detailed body region is under consideration and loading as well as unloading behavior may be of importance.

CONCLUSION

The femoral internal rotation, or the tibia external rotation, respectively, representing a common injury mechanism in ACL ruptures, is pre-dominantly hindered by the anterior part (AML) of the medial collateral ligament (MCL) and not by the ALL.

REFERENCES

[1] Rothrauff, B.B., Jorge, A., de Sa, D. et al.: "Anatomic ACL Reconstruction Reduces Risk of Post-Traumatic Osteoarthritis: A Systematic Review with Minimum 10-Year Follow-Up", *Knee Surgery, Sports Traumatology, Arthroscopy*, Vol. 28, pp. 1072-1084 (2020). https://doi.org/10.1007/s00167-019-05665-2

[2] Sanders, T.L., Maradit Kremers, H., Bryan, A.J. et al.: "Incidence of Anterior Cruciate Ligament Tears and Reconstruction: A 21-Year Population-Based Study", *The American Journal of Sports Medicine*, Vol. 44(6), pp. 1502-1507 (2016). https://doi.org/10.1177/0363546516629944

[3] Musahl, V., Karlsson, J.: "Anterior Cruciate Ligament Tear", *New England Journal of Medicine*, Vol. 380(24), pp. 2341-2348 (2019). https://doi.org/10.1056/NEJMcp1805931

[4] Noyes, F.R., Huser, L.E., Levy, M.S.: "Rotational Knee Instability in ACL-Deficient Knees", *The Journal of Bone and Joint Surgery*, Vol. 99(4), pp. 305-314 (2017). https://doi.org/10.2106/JBJS.16.00199

[5] Kaplan, D.J., Jazrawi, L.M.: "Secondary Stabilizers of Tibial Rotation in the Intact and Anterior Cruciate Ligament Deficient Knee", *Clinics in Sports Medicine*, Vol. 37(1), pp. 49-59 (2018). https://doi.org/10.1016/j.csm.2017.07.007

[6] Amis, A.A.: "Anterolateral knee biomechanics", *Knee Surgery, Sports Traumatology, Arthroscopy*, Vol. 25, pp. 1015-1023 (2017). https://doi.org/10.1007/s00167-017-4494-x

[7] Sonnery-Cottet, B., Daggett, M., Fayard, J. et al.: "Anterolateral Ligament Expert Group Consensus Paper on the Management of Internal Rotation and Instability of the Anterior Cruciate Ligament – Deficient Knee", *Journal of Orthopaedics and Traumatology*, Vol. 18, pp. 91-106 (2017). https://doi.org/10.1007/s10195-017-0449-8

[8] Koga, H., Nakamae, A., Shima, Y.: et al.: "Mechanisms for Noncontact Anterior Cruciate Ligament Injuries: Knee Joint Kinematics in 10 Injury Situations from Female Team Handball and Basketball", *The American Journal of Sports Medicine*, Vol. 38(11), pp. 2218-2225 (2010). https://doi.org/10.1177/0363546510373570

[9] Musahl, V., Karlsson, J., Kuroda, R., et al.: "Rotatory Knee Instability", Springer International Publishing Switzerland 2017. ISBN 978-3-319-32069-4. https://doi.org/10.1007/978-3-319-32070-0

[10] Elemance: "Global Human Body Models Consortium-owned GHBMC M50 Pedestrian Simplified Model", Version 1.5, Elemance, LLC, Clemmons NC, 2018. https://www.elemance.com/

[11] Hallquist, J.O.: "LS-DYNA Keyword User's Manual", Release 11.0, Livermore Software and Technology Corporation. Livermore CA, 2018. http://www.lstc.com/download/manuals

[12] Untaroiu, C.D., Putnam, J.B., Schap, J. et al.: "Development and Preliminary Validation of a 50th Percentile Pedestrian Finite Element Model", *International Design Engineering Technical Conferences & Computers and Information in Engineering Conference*, 2015. https://doi.org/10.1115/DETC2015-47781

[13] Quapp, K.M., Weiss, J.A.: "Material Characterization of Human Medial Collateral Ligament", *Journal of Biomechanical Engineering*, Vol. 120, pp. 757-763 (1998). https://doi.org/10.1115/1.2834890