Biomechanical Evidence on Anterior Cruciate Ligament Reconstruction*

Análise biomecânica da reconstrução do ligamento cruzado anterior

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Abstract

Objective Anterior cruciate ligament (ACL) reconstruction is recommended in athletes with high physical demands. Several techniques are used in reconstruction; however, the most relevant question still is the best biomechanical positioning for the graft. The present study aimed to analyze the biomechanical effect of the position of bone tunnels on load distribution and joint kinetics, as well as the medium-term functional outcomes after ACL reconstruction.

Methods A biomechanical study using a finite element model of the original knee (without anterior cruciate ligament rupture) and reconstruction of the ACL (neoACL) was performed in four combinations of bone tunnel positions (central femoral-central tibial, anterior femoral-central tibial, posterosuperior femoral-anterior tibial, and central femoral-anterior tibial) using the same type of graft. Each neo-ACL model was compared with the original knee model regarding cartilaginous contact pressure, femoral and meniscal rotation and translation, and ligamentous deformation.

Results No neo-ACL model was able to fully replicate the original knee model. When the femoral tunnel was posteriorly positioned, cartilage pressures were 25% lower, and the mobility of the meniscus was 12 to 30% higher compared with the original knee model. When the femoral tunnel was in the anterior position, internal rotation was 50% lower than in the original knee model.

Conclusion Results show that the femoral tunnel farther from the central position appears to be more suitable for a distinct behavior regarding the intact joint. The most anterior position increases rotational instability.

Keywords
- rupture
- anterior cruciate ligament reconstruction
- anterior cruciate ligament

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Resumo

Objetivo A reconstrução do ligamento cruzado anterior é aconselhável sobretudo em atletas de alta demanda física. Diversas técnicas são usadas na reconstrução, mas a grande questão é qual o melhor posicionamento para o enxerto. Analisar o efeito biomecânico da posição dos túneis ósseos na repartição de carga e cinemática da articulação, bem como os resultados funcionais em médio prazo, após reconstrução do ligamento cruzado anterior.

Métodos Fez-se um estudo de simulação biomecânica computacional com modelos de elementos finitos do joelho original e com reconstrução do ligamento cruzado anterior (Neo-LCA) em quatro combinações de posição dos túneis ósseos (femoral central-tibial central, femoral anterior-tibial central, femoral posterossuperior-tibial anterior e femoral central-tibial anterior) com o mesmo tipo de enxerto. Para cada modelo, foram comparadas a pressão de contato na cartilagem, a rotação e translação do fêmur e dos meniscos e a deformação nos ligamentos.

Resultados Nenhum modelo de Neo-LCA foi capaz de reproduzir, na íntegra, o modelo do joelho original. Quando o túnel femoral era colocado em posição mais posterior, observaram-se pressões na cartilagem 25% mais baixas e translação dos meniscos superiores entre 12% e 30% relativamente ao modelo intacto. Quando o túnel femoral estava em posição mais anterior, observou-se uma rotação interna do fêmur 50% inferior ao modelo intacto.

Conclusão Os resultados evidenciam que uma localização do túnel femoral mais distante da posição central parece ser mais preponderante para um comportamento mais díspar relativamente à articulação intacta. Na posição mais anterior existe um aumento da instabilidade rotatória.

Palavras-chave
- ruptura
- reconstrução do ligamento cruzado anterior
- ligamento cruzado anterior

Introduction

Anterior cruciate ligament (ACL) lesions are very frequent in sports (70%)[1] However, the medium and long-term success of the reconstruction of the ACL (neoACL) is directly related to the alignment/positioning of the bony tunnels, as well as to the tension of the ligament graft. The positioning of the bony tunnels is critical to knee kinetics and biomechanics,[2] and it influences surgical outcomes. Finite elements models simulate knee biomechanical characteristics both at the ligament level and at the cartilage level; moreover, these models allow the calculation of the different tensions generated either without ACL rupture or with ligament reconstruction. In the present study, neoACL was simulated based on finite element models. The ligament was replaced by four bone-tendon-bone (BTB) grafts.[3] The positioning of the bone tunnels was reproduced from the cadaveric study developed by one of the authors of the present paper (JCN), which simulated several positional possibilities, always with the same type of reconstruction, and compared them with the original model. Some biomechanical conditions, cartilaginous contact pressures, femoral posterior translation and rotation, meniscal translation, and maximum ligamentous main strains (tension) generated by the various positions could be calculated, allowing us to predict the medium and long-term risks incurred by an operated knee.

Materials and Methods

The original knee model was developed in a computer from the 3D Open-Knee Model, which was prepared from magnetic resonance imaging (MRI) of the left knee of a 77-year-old cadaver[4,5] and consisted of distal femur, proximal tibia, cartilage, intact menisci, collateral ligaments, cruciate ligaments, and proximal fibula (►**Fig. 1**). The tibial slope was 5° posterior.

Meanwhile, four geometric models with neoACL were developed based on the studies of Noronha.[5] These four models were prepared with the CATiA CAD software (Dassault Systèmes, Vélizy-Villacoublay, France) by replacing the ACL with a BTB graft with a cross-section equivalent to the intact ligament. Since the different positions of the tibia and femur tunnels reproduced those described in the experimental cadaveric work from Noronha,[5] which were the positions closest to the original ACL isometry, the same nomenclature was used (►**Fig. 2**). Acronyms FC and TC represent the central-natural ACL positions in the femur (FC) and in the tibia (TC), respectively; acronyms FA and TA represent the most anterior tunnels with respect to the central-natural positions of the femur (FA) and of the tibia (TA), respectively; acronym FPS represents a femoral tunnel in posterosuperior position (FPS), and acronym TAI represents a tibial tunnel in the anterointernal position (TAI). Based on the different positions of the tibial and femoral tunnels, four combinations of neoACL were analyzed: FC-TC, FA-TC, FC-TA, and FPS-TAI (►**Fig. 2**). The different geometries of

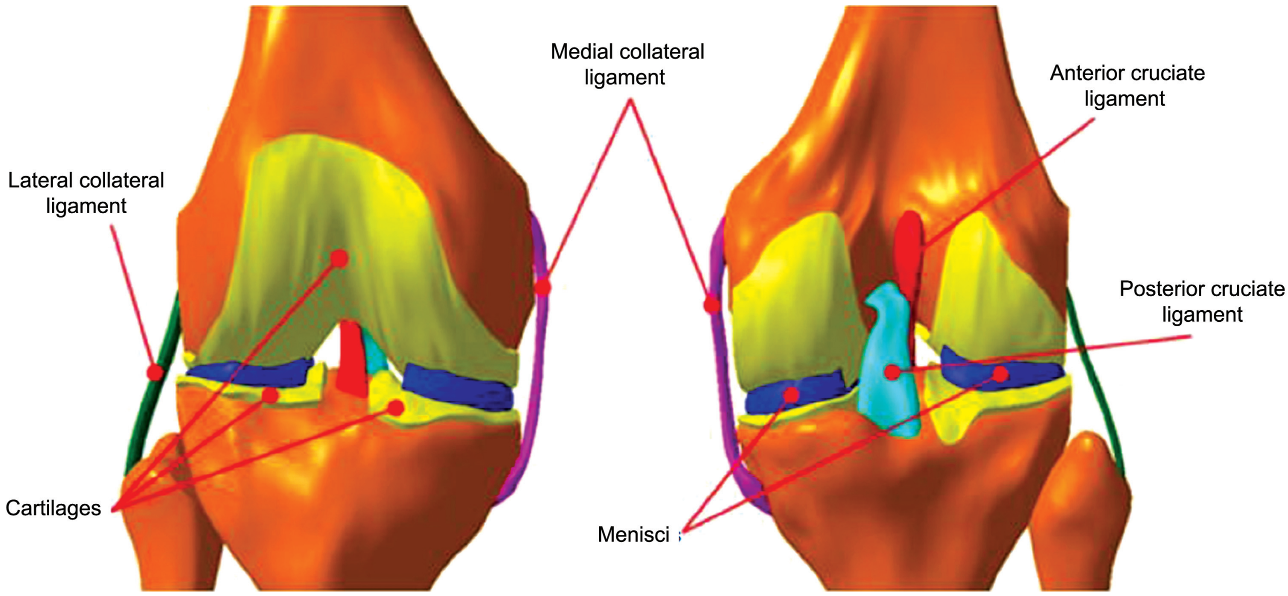

Fig. 1 Geometric model of the intact knee (Open Knee Model).

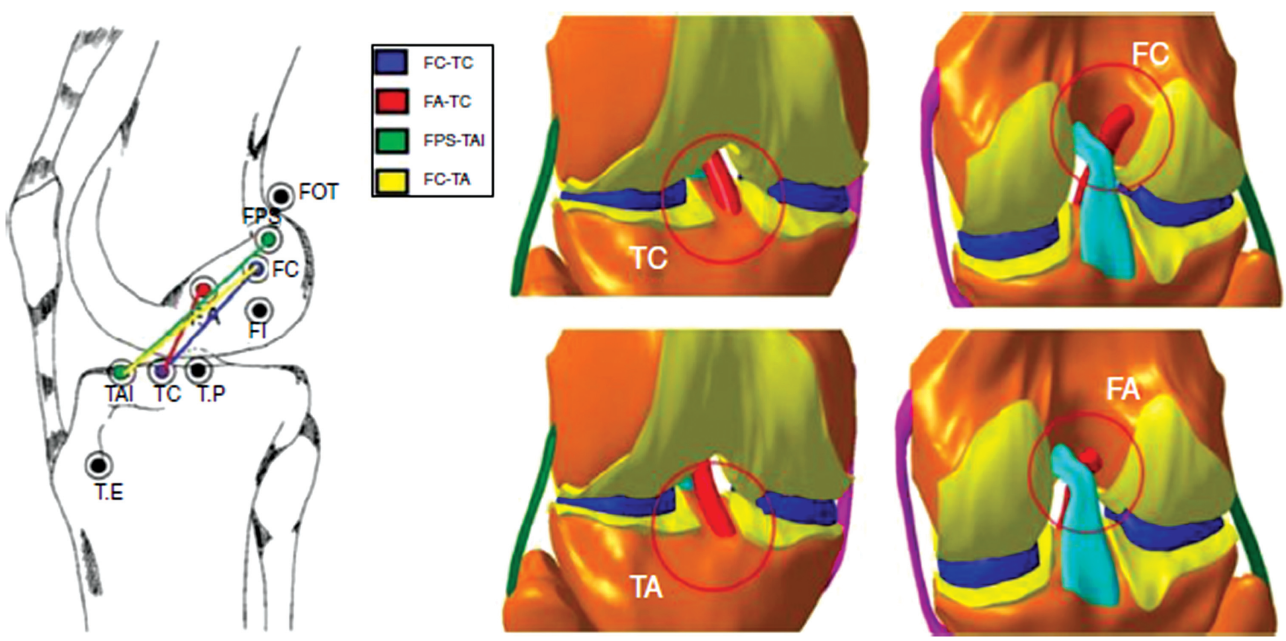

Fig. 2 Position of the bone tunnels in the analyzed tibia and femur. FC-TC, central femur and tibia; FA-TC, anterior femur and central tibia; FC-TA, central femur and anterior tibia; FPS-TAI, posterior-superior femur and anterior-internal tibia.

each model were imported to the Abaqus software, version 6.13 (Dassault Systèmes, Vélizy-Villacoublay, France), in which the finite element mesh was generated (►**Fig. 3**) and simulations were made. The type of element, the number of elements and the number of knots at each structure from the different joint models are shown in ►**Table 1**. Although all of the materials from the different joint structures present a viscoelastic behavior, the short time of articular load application during knee flexion ($t = 1$ second) approximates their behavior to linear elastic[6] with elastic moduli (E) and Poisson ratio (í),[7–12] detailed in ►**Table 2**. The interaction-attachment conditions between the different joint structures attempted to approach the physiological condition, considering that the tibia and the femur are solidary in the neoACL models reconstructed with BTB grafts. Interactions between bone surfaces and ligamentous and cartilaginous attachment zones were modeled as rigid connections. The remaining interactions between the different structures were modeled with frictionless contact.[6] The fixation of the meniscal horns was modeled with 10 springs (350 N/mm) per horn (►**Fig. 3**). Numerical models, forces, and moments developed in the knee during a 75 kg-person gait cycle were applied to the models.[13,14] The joint flexion resulted only from the application of forces and momentum in the femur, since the fibula and the tibia were fixed in the distal zone (►**Fig. 3**). The tibial-femur joint force (Fy), the patellofemoral anteroposterior joint force (Fx), and an abduction-adduction

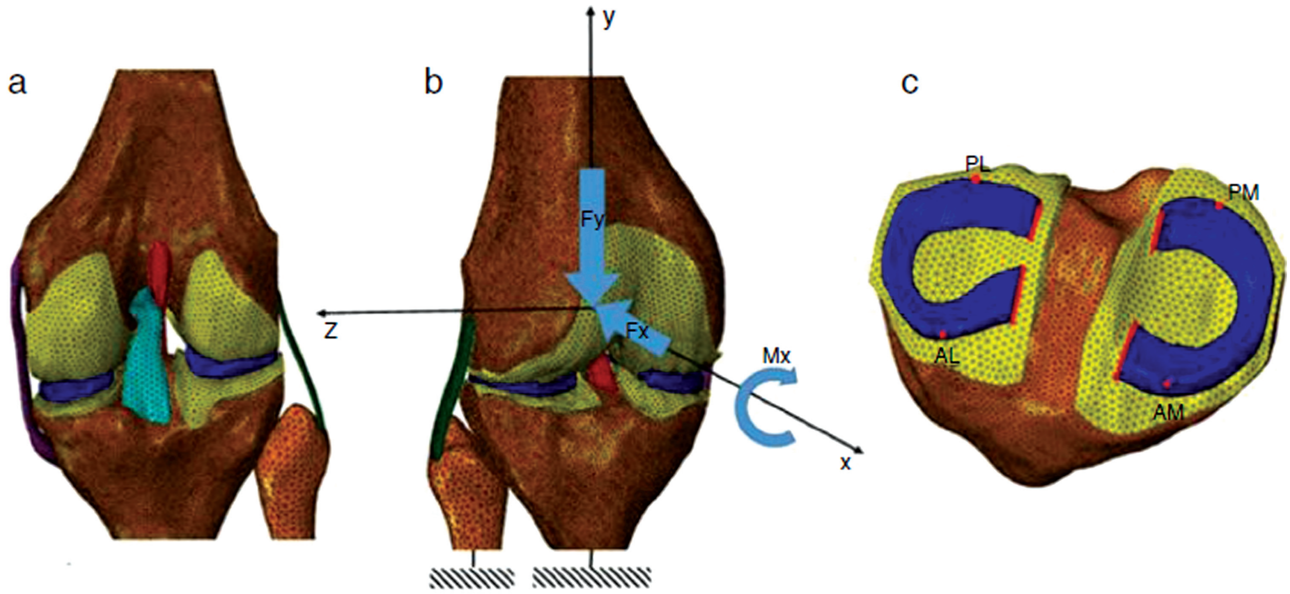

Fig. 3 A, Finite element model of the knee (posterior view); B, Schematic representation of the forces and momentum applied to the joint; C, Location of the points AL, PL, AM, PM in which menisci displacements were measured.

Table 1 Element type, elements number, and knots in each numeric model structure

Structure	Element type	Elements number	Knots number
Femur	S3R	40,628	20,316
Tibia	S3R	25,130	12,567
Fibula	S3R	1,528	766
Menisci	C3D4	25,573	5,952
Tibial cartilage	C3D10M	13,992	24,782
Femoral cartilage	C3D10M	24,094	6,405
ACL	C3D4	1,601	510
PCL	C3D4	2,381	721
MCL	C3D4	3,847	1,165
LCL	C3D4	2,453	774
NeoACL FC-TC	C3D4	6,139	1,420
NeoACL FC-TA	C3D4	5,633	1,357
NeoACL FA-TC	C3D4	3,020	734
NeoACL FPS-TAI	C3D4	5,496	1,374

Abbreviations: ACL, anterior cruciate ligament; FA-TC, anterior femur and central tibia; FC-TA, central femur and anterior tibia; FC-TC, central femur and tibia; FPS-TAI, posterior-superior femur and anterior-internal tibia; LCL, lateral collateral ligament; MCL, medial collateral ligament; PCL, posterior cruciate ligament.

momentum at the frontal plane (Mx) were applied to the femur (►**Fig. 3**). The evolution of the Fy and Fx forces and of the Mx in the joint during flexion, lasting 1 second, are shown in ►**Table 3**.[13,14] A test was performed up to a flexion angle of 100°, higher than the 60° normally developed in the gait cycle. The parameters analyzed were contact pressure in the cartilage; femoral translation and rotations; meniscal translations at AL, PL, AM and PM (►**Fig. 3**); and maximum main deformations (traction) in the ligaments and in the neoACL.

Table 2 Mechanical properties of the numeric model materials

Material	Reference	Young modulus (MPa)	Poisson ratio
Bone	[7]	17,000	0.36
Cartilage	[6]	15	0.45
Meniscus	[8]	59	0.45
ACL	[9]	280	0.42
PCL	[10]	300	0.42
MCL	[11]	372	0.42
LCL	[10]	332	0.42
NeoACL	[12]	320	0.42

Abbreviations: ACL, anterior cruciate ligament; LCL, lateral collateral ligament; MCL, medial collateral ligament; PCL, posterior cruciate ligament.

Results

Maximum contact pressures in the femoral and tibial cartilages are presented in ►**Fig. 4** for the intact model (without neoACL) and for the neoACL models in flexion of up to 60° (gait cycle). The highest value of contact pressure occurred in the intact model in the medial tibial cartilage (12 MPa). The neoACL FPS-TAI model was the most different from the mean pressure values of the intact model, while the remaining neoACL models presented values 25% lower than the intact model. Maximum femoral rotations in the transverse (internal rotation) and frontal planes are shown in ►**Fig. 5**. The FA-CT model was the one with the lowest rotational values in both planes, with a mean value 50% lower than the other

Table 3 Forces and momentum applied to the joint during flexion (t = 1 s)

Flexion angle	Fy (N)	Fx (N)	Mx (Nm)
0°	0	0	0
10°	950	300	7.5
20°	1,520	480	15
30°	1,330	420	10.5
40°	1,520	480	12
50°	1,900	600	13.5
60°	950	300	6
70°	760	240	4.5
80°	570	180	4.5
90°	570	180	4.5
100°	570	180	4.5

Abbreviations: Fx, patellofemoral anteroposterior joint force; Fy, Tibial-femur joint force; Mx, momentum.

models for the flexion of up to 60°. The 70° to 100° flexion interval presented nominal values of maximum rotation in the inverse direction to the other models. Regarding the posterior translation of the femur (rollback) in flexion of up to 60° (►**Fig. 6A**), all of the analyzed models presented similar values, ~ 16 mm. The movements in the anterior (AL and AM) and posterior (PL and PM) points of the meniscus (►**Fig. 6B**) presented different values among the analyzed models. The neoACL FA-CT model presented the lowest values of posterior translation, with a value 30% lower than the intact model. The neoACL FPS-TAI model presented the highest values, with translational values 12 and 30% higher than the intact model. The deformations in the different joint ligaments are presented in ►**Fig. 7**. In flexion of up to 60° (gait cycle), the posterior and anterior cruciate ligaments presented more distinct behaviors among the neoACL models. In the posterior cruciate ligament, the FA-CT model presented 40% lower deformation values than the intact model, while the neoACL FC-TC and FPS-TAI models presented 30% higher values. In the anterior cruciate ligament, the neoACL FA-CT model showed a deformation value 100% higher than the intact model, while the FPS-TAI model presented a 30% lower value. In the flexural complement between 70° and 100°, the neoACL FA-CT model showed deformation values 2 to 3 times higher than the intact model, whereas the FPS-TAI model showed 3 times lower deformation values.

Discussion

We have decided to consider only knees with intact meniscus, normal cartilage, mechanical axis of 180° and tibial inclination of 5°, and only kinematics variations and joint pressures introduced by the different bone tunnels were studied. The introduction of more variables would increase noise and difficult the interpretation of our objectives. The cartilage contact pressure gradients exhibited by the intact model (natural ACL) closely follow the normal asymmetrical load distribution on the natural knee, resulting in contact pressures in the upper medial tibial cartilage of about between 30 and 40% of those observed on the flexural lateral side during the gait cycle.[14,15] Similarly, the kinematic results of the intact model regarding femoral rotations and posterior translation (rollback), as well as the posterior meniscal movements during flexion, were in the same range obtained in the natural knee.[2,16–18] This ability of the intact model to approximate the behavior of the natural knee in terms of load distribution and of femoral and meniscal kinematics during flexion shows its validity for the comparative study of neoACL , which was the main object of the present study. In the comparison of the contact pressure in the tibial cartilage of the different models with neoACL, all of the models presented peak values within the physiological

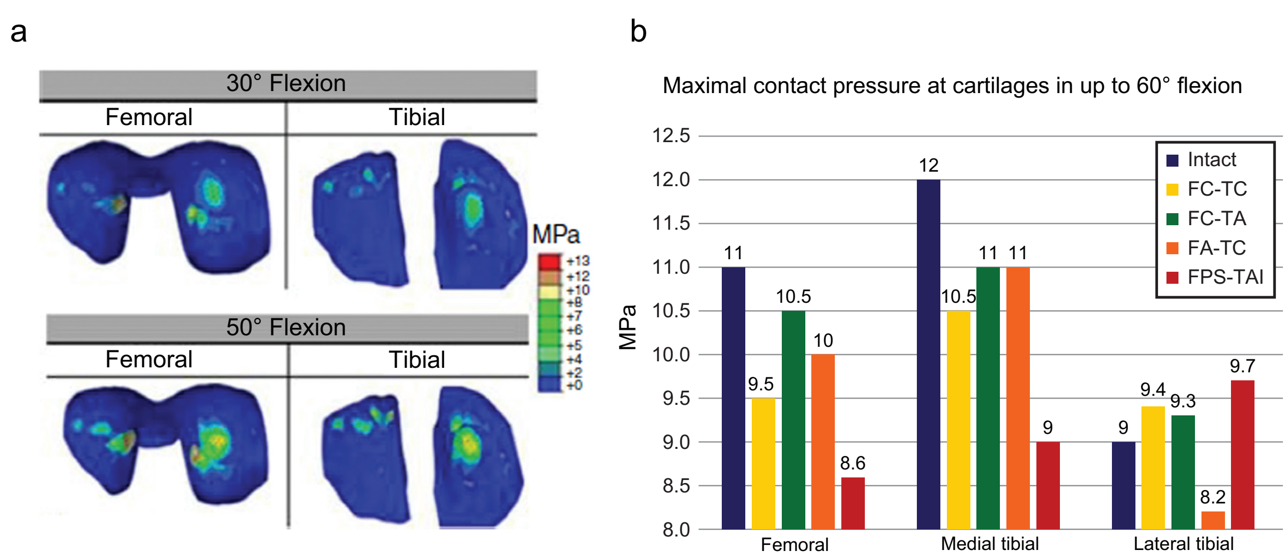

Fig. 4 A, Contact pressure gradients at the femoral and tibial cartilage; B, Maximum contact pressure at the femoral and tibial cartilage (0-60° flexion).

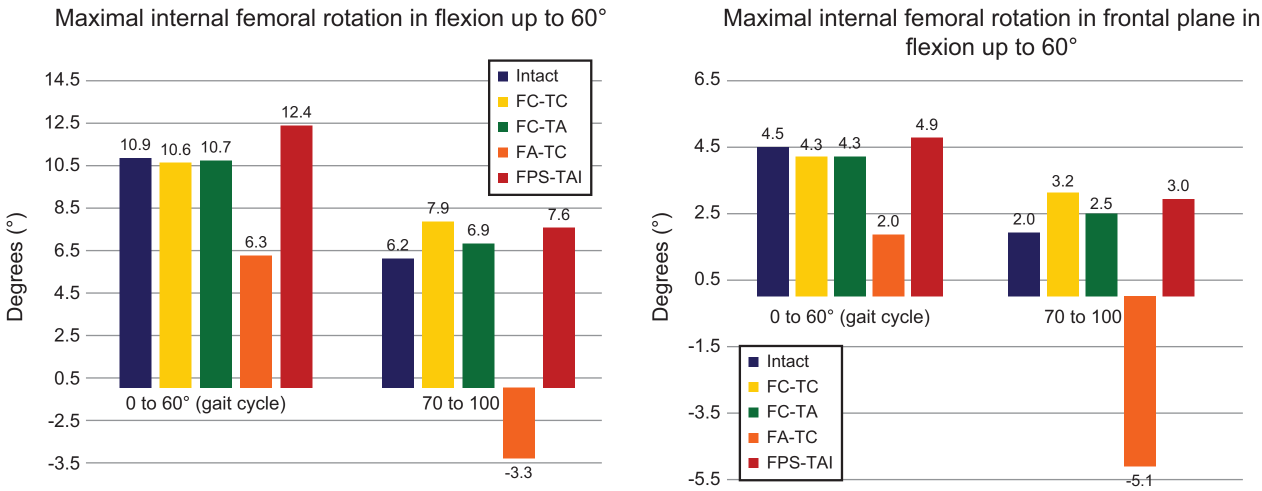

Fig. 5 Maximal femoral rotations in cross-sectional and frontal planes during a movement in flexion up to 60°

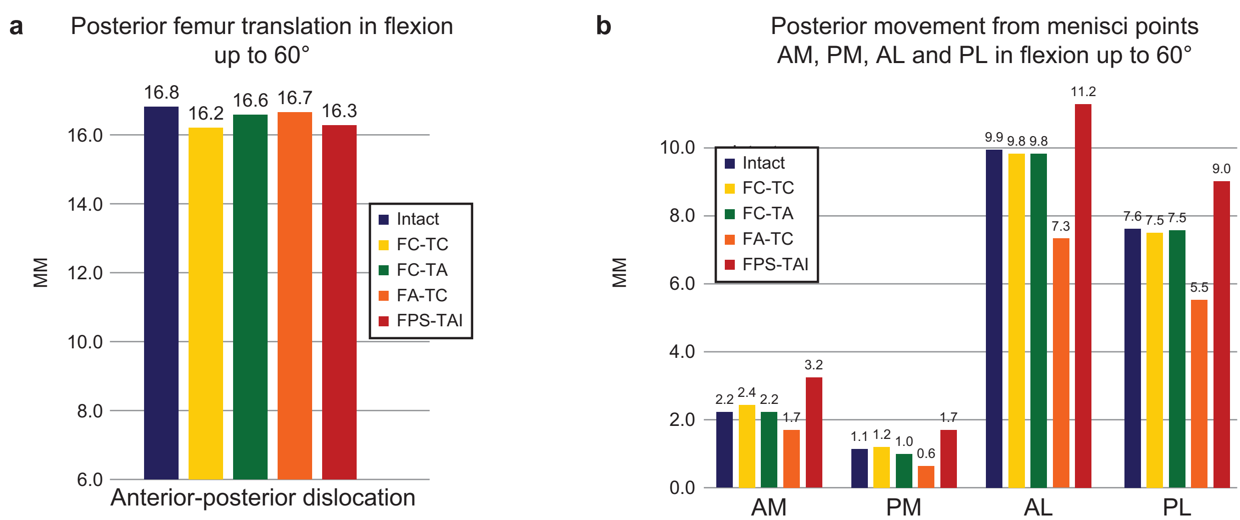

Fig. 6 A, Posterior femoral translation in up to 60° flexion; B, Posterior meniscal translation at points AM, PM, AL and PL (Fig. 3) in up to 60° flexion

range, between 8.2 and 12 MPa.[15] However, the FPS-TAI model was the most distant from the behavior of the intact model and from the normal load distribution in the joint, since it presented higher pressure values in the lateral tibial cartilage than in the medial one. Apparently, the combination of the posterosuperior femoral tunnel with an anterointernal tibial tunnel alters the load distribution in the joint in a more significant way. Regarding the maximum femoral rotations, the FA-CT reconstruction model, with a more anterior femoral tunnel and a central-natural positioned tibial tunnel, showed the lowest values of femoral rotation in the transverse plane (internal rotation) and in the frontal plane rotation in up to 60° flexion, with values 40% lower than the intact model and other neoACL models. This same FA-CT model presented peak rotational values in the opposite direction to the other models at 70° to 100° flexion, indicating that the most anterior position of the femoral tunnel (FA) changes more significantly the femoral rotational kinematics in this range of joint flexion. Regarding posterior femur translation during flexion, all of the neoACL reconstruction models presented values identical to the intact model; apparently, the different locations of femoral and tibial tunnels did not alter the femoral rollback effect in the range of flexion of the gait cycle. Regarding the movement of the menisci in their anterior and posterior regions, the neoACL reconstruction models that presented values more distinct from the intact model were the FPS-TAI, which showed a tendency for a greater posterior displacement of both menisci, and the FA-TC, which exhibited the smallest displacement of the menisci of all of the analyzed models. In this case, the removal of the tunnels from their natural central positions in the femur, either anteriorly (AF) or posteriorly (FPS), appears to have the greatest influence on meniscal mobility. As for the state of deformation of the ligament and neoACL traction, in up to 60° flexion (gait cycle), it was verified that the models FA-TC and FPS-TAI presented the most different values of deformation compared with the intact model, especially in the cruciate ligaments. The model with the most anterior femoral tunnel, FA-CT, showed the lowest deformation in the posterior

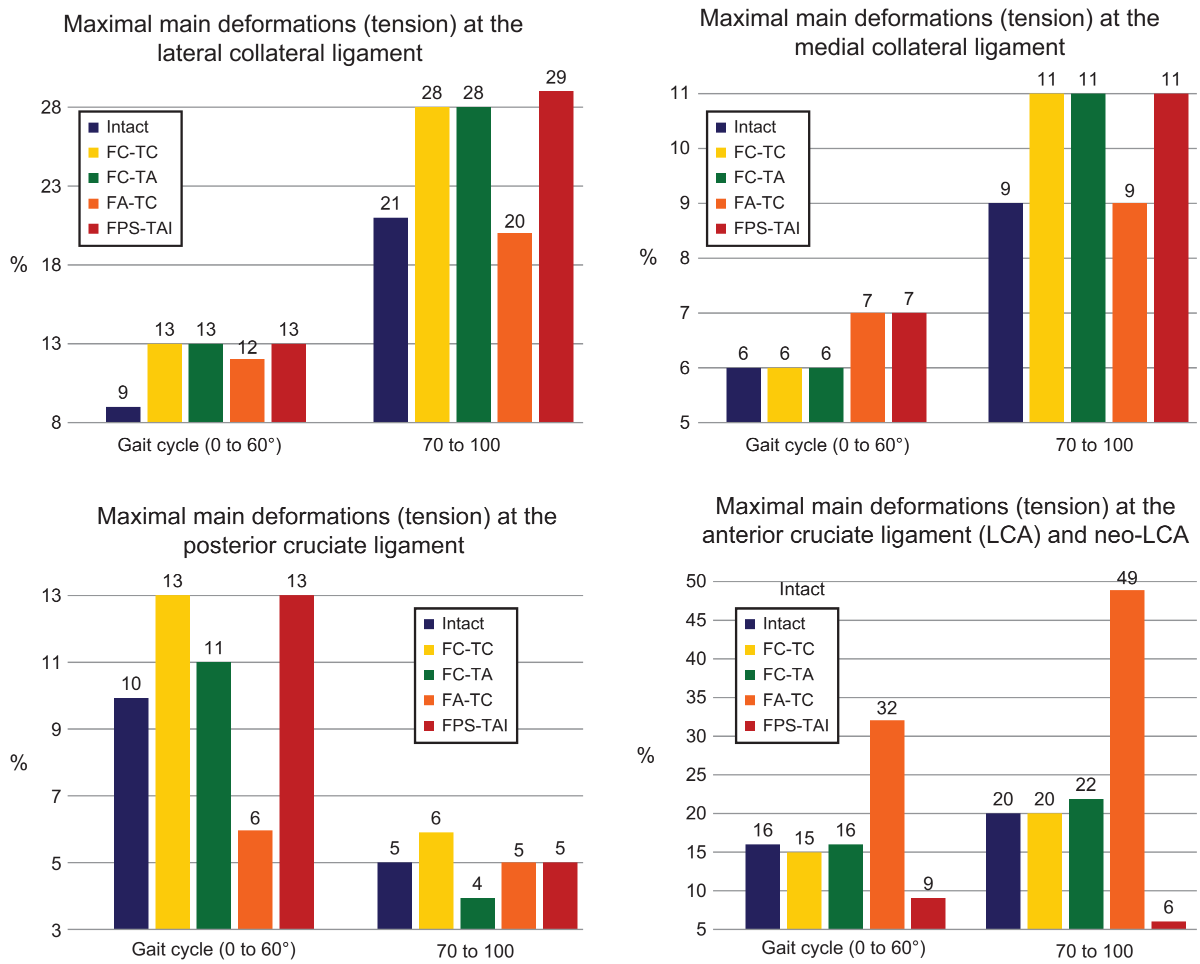

Fig. 7 Maximal main deformity (tension) on knee ligaments and neoACL in up to 60° and 70° to 100° flexion

cruciate ligament. On the other hand, the femoral tunnel model in the most posterior position, FPS-TAI, showed the lowest deformation value in the neoACL between all of the analyzed models, whereas the model with the most anterior femoral tunnel, FA-CT, presented the highest deformation values, two times higher than in the intact model. This confirms that the positioning of bone tunnels during neoACL affects both the load distribution at the joint and the kinematics of its structures. The neoACL models closer to the structural and kinematic behavior of the intact model were those with more central-natural positioned femoral tunnels, namely FC-CT and FC-TA. Both models with femoral tunnel farthest from the center, either in the anterior direction, FA-CT, or in the posterior direction, FPS-TAI, presented the most distinct behaviors from the intact model for most of the analyzed parameters.

In agreement with the literature reports,[19] the positioning of the femoral tunnel is important for joint mobility and the clinical outcome. However, we know that after neoACL, there is still the possibility of developing arthrosis, even without meniscectomy associated with the procedure. In the long-term, which corresponds to 10 years, this development is associated with loss of full extension and joint mobility.[20] In 20 years of follow-up, the described risk factors for developing arthrosis were loss of full extension, meniscectomy (medial or lateral), cartilage disease, and aging of the patient.[21] The present study shows that after neoACL, there is no return to the biomechanical state prior to the rupture of the ACL and, that by positioning the femoral tunnel more posteriorly, the surgeon contributes to a change in the load exerted at the cartilage level of about 25% compared with the knee without rupture of the ACL; in the medium/long-term, this can lead to degenerative cartilage changes. These experimental data compel us to reflect and try to find a femoral tunnel position that does not significantly change cartilage pressures, but that allows good knee stability after neoACL.

There are limitations associated with the present study. One of them is related to the simplification of the load state in the joint. However, the most preponderant joint forces during the gait cycle were considered. In addition, the viscoelastic behavior of different structures was not considered. Nevertheless, due to the short time of force application ($t = 1$ second), it is reasonable to consider an elastic behavior of these structures. Moreover, all of the structures were considered homogeneous, a situation different from the real one. However, due to the comparative nature of the present study, in which only the positioning of the bone tunnels was distinct between the models, it is assumed that this simplification does not alter the relative outcomes from different models.

Conclusion

The present study illustrates that the structural and kinetic behavior of the knee joint structures with neoACL varies according to the positioning of the bone tunnels. The best position seems to be central, that is, anatomical. The location of the femoral tunnel farthest from the central-neutral position is more predisposing to an unbalanced structural and kinematic behavior with altered cartilage load, and it may be the cause of the development of arthrosis in the long term.

Conflicts of Interest

The authors have no conflicts of interest to declare.

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