Effect of tibial tunnel diameter on the outcome of ACL reconstruction

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Summary

The number of variables that influence the success of an anterior cruciate ligament (ACL) reconstruction is so high that an in-depth analysis of the problem can only be carried out through numerical tools. Once the diameter of the substitute plasty and the interference screw has been chosen for a given patient, one of the main concerns of the surgeon is to find the most suitable diameter of the tibial tunnel for its fixation. In this work, a finite element model was developed in order to simulate both the reconstruction and the subsequent rehabilitation of the ACL at its tibial insertion. For the simulation, the chosen tendon and screw diameters were 4 mm and 7 mm, respectively, while diameters of 7, 8, 9, and 10 mm were tested for the tibial tunnel. The parameters of the behavior models of the different materials (screw, bone and tendon) were obtained through experimental tests. The results obtained show that, as the diameter of the tunnel decreases, the compressive stress over the plasty will increase (theoretical objective of the fixation), but the deformation induced on the trabecular bone also increases, which can trigger its failure. For this reason, the maximum values of the interferential pressure must be limited to those strictly necessary to ensure that the reconstruction is properly done, that is, that it prevents the tendon from slipping in the tunnel during the rehabilitation process. The simulation of the rehabilitation process was done by pulling the already fixed tendon in the femoral direction in order to extract it. It was obtained that the most suitable diameter of the tibial tunnel for the chosen plasty-screw assembly is 8 mm, since it provides a suitable subjection without high values of deformation in trabecular bone, that is, no damage in this part of the bone.

Key words:
ACL. Tibiae. Trabecular bone. Plasty. FEM.

Efecto del diámetro del túnel tibial en el resultado de la reconstrucción de LCA

Resumen

El número de variables que influyen en el éxito de una reconstrucción de ligamento cruzado anterior (LCA) es tan elevado que un análisis profundo del problema sólo puede realizarse a través de herramientas numéricas. Elegido el diámetro de la plastia sustituta y del tornillo interferencial para un determinado paciente, una de las principales preocupaciones del cirujano es dar con el diámetro de túnel tibial más adecuado para su fijación. En este trabajo se desarrolló un modelo de elementos finitos que simula tanto la reconstrucción como la posterior rehabilitación del LCA en su inserción tibial. Para la simulación, los diámetros del tendón y tornillo elegidos fueron 4 mm y 7 mm respectivamente mientras que para el túnel tibial se probó con diámetros de 7, 8, 9 y 10 mm. Los parámetros de los modelos de comportamiento de los diferentes materiales (tornillo, hueso y tendón) se obtuvieron mediante ensayos experimentales. Los resultados obtenidos muestran que, conforme disminuye el diámetro del túnel utilizado, crece la tensión de compresión ejercida sobre la plastia (objetivo teórico de la fijación), pero también crece la deformación inducida sobre el hueso trabecular, lo que puede desencadenar el fallo del mismo. Por esta razón, los valores máximos de la presión interferencial deben limitarse a los estrictamente necesarios para asegurar que la reconstrucción sea efectiva, es decir, que evite el deslizamiento del tendón en el túnel durante el proceso de rehabilitación. Simulado, también, el proceso de rehabilitación tirando del tendón en dirección femoral, se ha obtenido que el diámetro de túnel tibial más adecuado para el conjunto plastia-tornillo elegido, es el de 8 mm, ya que proporciona una fijación suficiente sin que los valores de deformación en el hueso trabecular lleguen a producir su daño.

Palabras clave:
LCA. Tibia. Hueso trabecular. Plastia. FEM.

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Introduction

ACL tears are usually treated by rebuilding the damaged ligament or by replacing it with a tendon that can perform its function. To carry out this procedure, the surgeon must drill two bone tunnels, in the tibia and femur, remove the ligament and replace it with the graft.

The prevalence of this injury estimates that 60% of them have a sporting nature, that is, they happen to the young and active population[1]. However, it also happens in people who are overweight and not very active or have motion limitations. Largely, the success of the reconstruction will depend on the chosen plasty, but also on the choice of the geometry of the tibial tunnel, the materials involved and the optimal size of each element depending on the patient. To date, the surgeon usually carries out the intervention in a standardized way, without being able to make great distinctions between patients. Through this work, it is intended that each patient can be treated individually. Moreover, this work try to find the optimal reconstruction parameters for each patient, that is, it will be performed for their type and size of tendon, as well as for their own bone characteristics.

Due to the large number of variables involved in the study, it seems clear that making the most appropriate decision for each patient is not easy, and in order to undertake the analysis, we must use numerical tools[2].

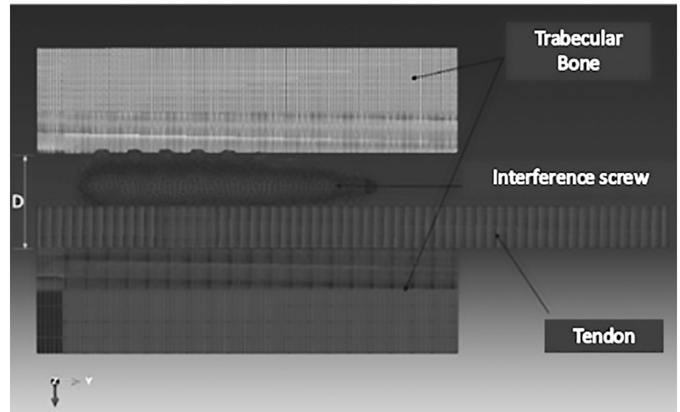
The objective of this work is, therefore, the analysis of the mechanical response of the tibia-plasty-screw assembly after the anterior cruciate ligament reconstruction, evaluating the influence of the different variables that come into play[3]. This would not be possible without the correct mechanical characterization, through laboratory tests, of the different materials involved. This need arises, in large part, due to the enormous variability in the existing bibliographic data[4-10], derived both from the use of very diverse materials (different species, screws of different materials, etc.) as well as very different test methodologies. Thus, in order to develop and subsequently validate a correct numerical model, it is necessary to describe the most appropriate mechanical behavior models for each of the intervening elements, which would not be possible without their prior experimental characterization.

Once the most appropriate behavioral models for each material involved in the reconstruction have been defined, this paper presents the finite element model used to describe the fixation process of the plasty in the tibial tunnel by using an interference screw. The states of stress and strain are also analyzed once the surgery is finished, depending on the diameter of the tibial tunnel used. Finally, the stability of the reconstruction is checked when, at the end of the plasty fixation process, the assembly must respond to normal workloads that would try to move the graft (and even the screw) along the tunnel.

Material and method

The numerical simulation of the ACL reconstruction has been carried out using the finite element method (FEM) with the commercial program ABAQUS. Hence, a two-dimensional (2D) geometric model has been made. It consisted of the trabecular bone that surrounds the tibial tunnel, inside which the substitute plasty and the interference screw are located, as shown in Figure 1. Four different diameters of the tibial tunnel (D = 7, 8, 9 and 10 mm) were analyzed, while the diameters of the plasty and the interference screw were considered invariable, with values of 4 mm and 7 mm, respectively.

Figure 1. Geometric model used in the analysis.



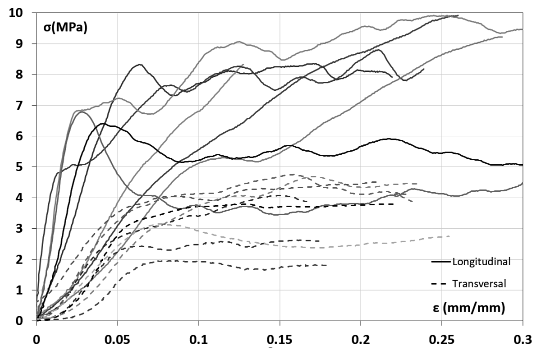
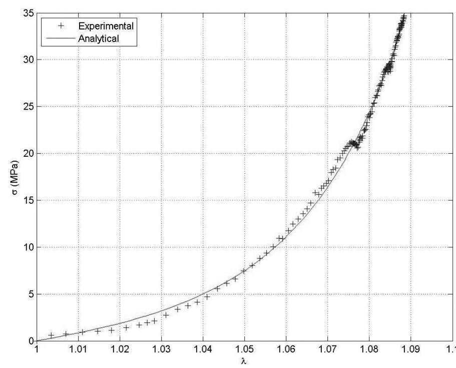
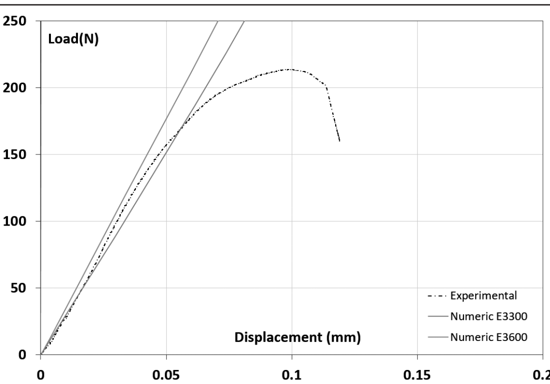
The material models used are a reflection of the results obtained after an extensive experimental work, which are collected in detail in previous publications[11,12].

Thus, trabecular bone, from porcine tibiae and characterized by compression tests[11,12], showed that once a certain level of load (σ_y) was reached, its resistant capacity remained constant until macroscopic failure due to collapse of the different trabeculae happened. It is an elastic-perfectly plastic behavior that is modeled by introducing the elastic parameters (elastic modulus, E and Poisson's ratio, v) and the yield stress, σ_y, of the material. The trabecular bone also showed greater resistance and stiffness when it was loaded in the longitudinal direction (with respect to the axis of the tibia) than when it was loaded in the transverse direction, that is, it has an anisotropic mechanical behavior. However, taking into account that it is impossible to fix the position of the tibial tunnel with respect to the axis of the tibia in an exact way numerically, the best and safest option is to consider that the behavior of the bone is isotropic using the corresponding properties to the weakest direction (transversal). This also has greatly simplified the calculus[11].

The tendons had porcine origin *(flexor digitorum)*, too. They were characterized using tensile loads and measuring the deformations with an ARAMIS 5M video correlation image equipment[12]. The behavior of the tendons was clearly hyperelastic and anisotropic, typical of this type of material. It is adjustable to a material model defined by Calvo and Peña[13], that describes the behavior of soft tissues being incorporated into the finite element program through a user subroutine (uanisohyper_inv[2,3,13]), whose constants, C_{10}, C_3 and C_4, collected in Table 1, are obtained after adjusting the model to the experimental results.

For its part, the interference screw used (with dimensions of 7 mm diameter and 25 mm length), was subjected to a compression test, loading the specimen perpendicularly to its principal axis. Its behavior was linear-elastic behavior up to very high loads. The elastic parameters obtained (E y v), shown in Table 1, were consistent with those expected for the type of material analyzed, composed of a mixture of PLLA polymer (75%) and hydroxyapatite HA (25%)[12].

Table 1. Materials involved in the simulation: Properties and constitutive model chosen.

Material	Behavior	Constitutive model	Properties
Trabecular bone		Elastic-perfectly plastic	E=73 MPa ν = 0.27 σ_Y 2.7 MPa
Tendon		uanisohyper_inv[13]	C_{10} = 7.98 C_3 = 0.374 C_4 = 19.24
Interference screw		Elastic	E = 3600 MPa ν = 0.3

To carry out the simulation, it was assumed that both the screw and the tendon are located in their final position (the one they would have after the real threading process was carried out by the surgeon) inside the tunnel. In this way, the screw is not in contact with the surrounding materials (trabecular bone and tendon) because a sufficient radial compression pressure has been applied to it. In a second step, this pressure is deleted, allowing the screw to start the contact with the other components that will be pressed, to a greater or lesser extent, depending on the size of the tunnel analyzed. The interaction properties (contacts) between the different materials have been defined based on previous studies[2,3,13,14,15]. A friction coefficient of 0.1 was used for the contact between the trabecular bone and the tendon or the screw while for the contact between the interferential screw and the tendon a much smaller coefficient (just 0.05) has been taken.

In this way, once the geometric model was generated (Figure 1), the materials (Table 1), contacts and the corresponding boundary conditions[12] were defined, the LCA reconstruction process was carried out in three calculation steps:

– *Step 1. Screw compression:* A uniform pressure is applied to the entire surface of the screw until its dimensions are reduced such that, once it is placed in the desired position within the tibial tunnel, there is no contact with any of the other elements of the model.
– *Step 2. Tendon pretension:* In the same way the surgeons act to ensure that the tendon does not roll or become lax during the screw entry process[12], a slight pretension is performed on it.
– *Step 3. Decompression of the screw:* Gradually, the pressure to which the screw had been subjected in step 1 is deleted. At the same time, the screw is allowed to recover little by little its initial

geometry and dimensions, starting the contact with the rest of the elements of the joint.

Once the reconstruction is complete, stress and strain values of the different elements are evaluated. However, the fundamental proof of the success of the intervention is that the substitute tendon is sufficiently fixed, so that, during the rehabilitation process, when it is subjected to a tensile load in the femoral direction, it does not experiment big displacements inside the tunnel. Thus, the following step is used.

– *Step 4. Rehabilitation:* After the surgery, the tendon is released from its initial pretension, and it is pulled from the opposite end, in the same direction, but towards the femur, trying to extract it from the tibial insertion. This simulates the natural movement of the knee. Based on the results obtained by other authors[16] when they analyzed the relative displacement of the tendons during a normal rehabilitation process, the axial displacement applied to the tendon at its femoral insertion is about 3 mm, which would be equivalent to knee flexion of 30º, position of maximum load during rehabilitation[17]. After this step, the relative displacements between the tendon and the screw teeth in different positions are analyzed.

Results

Figure 2 compares the appearance of the different elements involved in the ACL reconstruction once it is completed (step 3 is finished), for different diameters of the tibial tunnel. The configurations corresponding to the tibial tunnels with diameters of 10 mm (Figure 2a), 9 mm (Figure 2b), 8 mm (Figure 2c) y 7 mm (Figure 2d), are represented keeping constant the geometric parameters of the other components. As can be seen, as the diameter of the tibial tunnel decreases, the points of contact between the surface of the screw and the adjacent elements increase.

Figure 3 shows the distribution of minimum principal stress (radial pressure) at the points of the tendon surfaces in contact with the screw (tendon_left) and in contact with the trabecular bone (tendon_right), for the four cases analyzed. The points of contact between the teeth of the screw and the tendon are reflected in the peaks of the stress profile (Figure 3.a), showing punctual values of high compressive stresses. However, the contact between the tendon and the trabecular bone (Figure 3.b) shows a much smoother and more homogeneous stress distribution. It starts in the area corresponding to the beginning of the tibial tunnel, reaching its maximum value in the areas whose longitudinal coordinates match with the first or second tooth of the screw, the area with the largest diameter of the screw. Then, a progressively decrease of stress happens until its disappearance in an area close to the exit of the tibial tunnel towards the femur. These figures also show that as the diameter of the tunnel decreases, the stress levels on the tendon are higher as well as the area affected by these stresses, something that is beneficial in theory. However, the greatest variation in stress happens when the diameter of the tunnel decreases from 10 mm to 9 mm.

Figure 2. Numerical results of the reconstruction. Arrangement of tibial tunnels of: (a) 10 mm; (b) 9 mm; (c) 8 mm; (d) 7 mm.

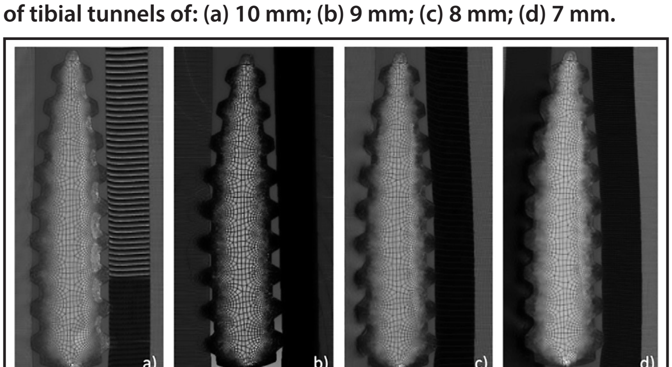
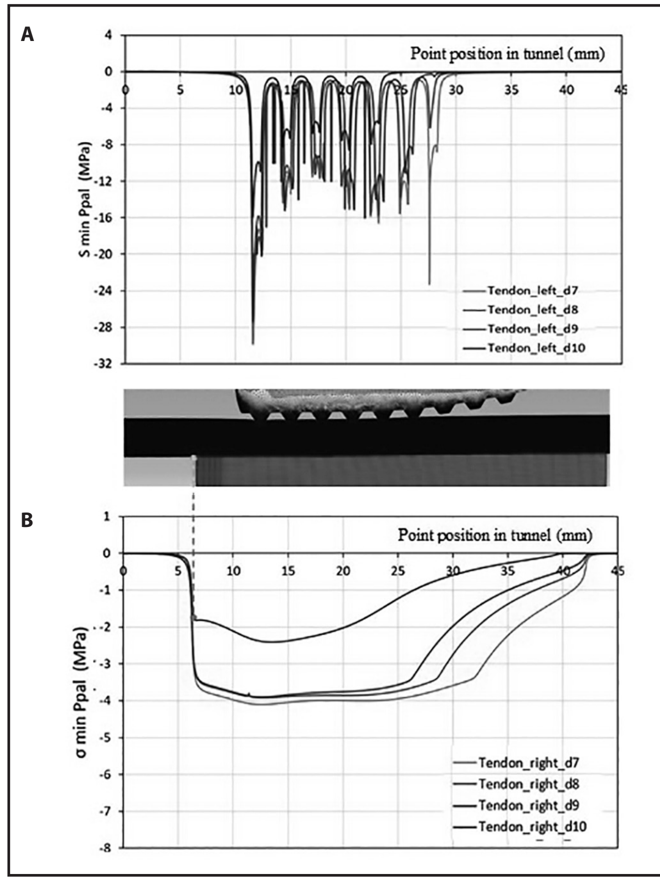


Figure 3. Influence of the tibial tunnel diameter on the Minimum Principal Stress in the tendon: (A) in contact with the screw; (B) in contact with trabecular bone.



Furthermore, as can be seen in Figure 4, the decrease in the tibial tunnel diameter also affects to the values of strain strongly, especially in the trabecular bone. The strong increase in the radial deformation of the trabecular bone is especially relevant when the diameter of the tunnel decreases from 8 mm to 7 mm.

For its part, Figure 5 shows the displacements experienced by different points of the tendon when it is subjected to the rehabilitation process once the reconstruction is completed. The zones represented are those that are in contact with the different teeth of the screw after reconstruction. *Zone 1* is the one corresponding to the tooth furthest from the femur and *Zone 8* is the one closest to the femoral part of the tunnel.

Figure 4. Influence of the tibial tunnel diameter in the strain distribution of the trabecular bone in contact with the tendon.

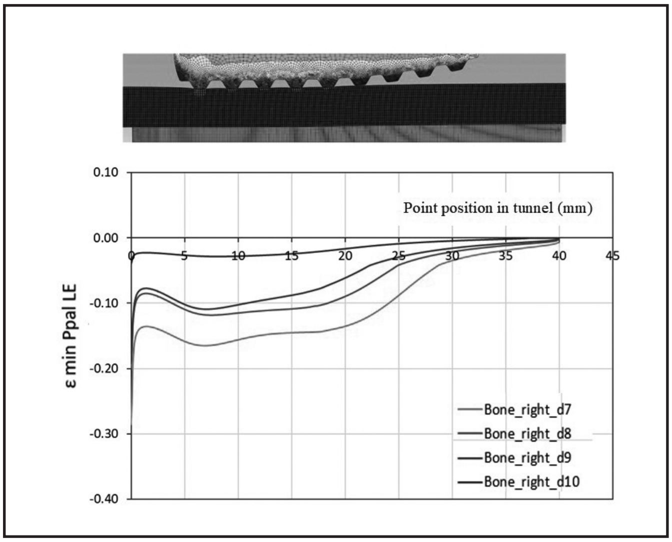
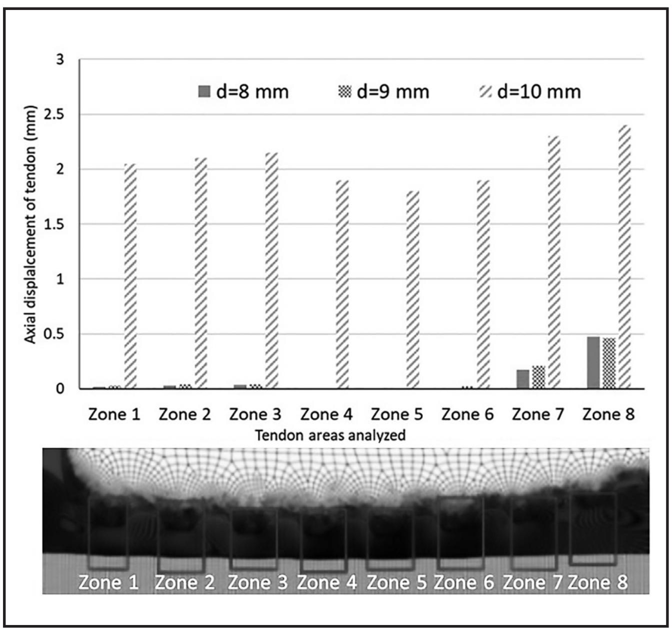


Figure 5. Displacement of the different points of the tendon along the tibial tunnel for different diameters of tunnel (d=10, 9 y 8 mm).



As can be seen, when the tunnel diameter used is 10 mm, all points of the tendon suffer longitudinal displacements close to or greater than 2 mm, which indicates that with this tunnel diameter, the tendon has not been sufficiently fixed. Thus, the reconstruction cannot be validate. However, when the diameter of the tibial tunnel becomes 9 mm, the displacements in the areas furthest from the femur are very small, being almost zero the one corresponding to the first tooth. With an 8 mm tibial tunnel diameter, all tendon displacement is zero, except to the ones that are near the femur.

Discussion and Conclusions

First, it should be noted that this work collects the results obtained after developing and applying a finite element model that simulates ACL reconstruction in the tibial area when a single interference screw with a diameter of 7 mm and a length of 25 mm is used to fix the position of a single 4 mm diameter substitute plasty. The variable that will change its value is the diameter of the tibial tunnel (between 7 mm and 10 mm). The geometric model, made in two dimensions, has been modeled in the most reliable way possible, and the behavior models of the materials involved (trabecular bone, tendon and interference screw) come from results obtained from an extensive experimental work collected in previous publications[11,12]. Due to the large number of contacts established between the elements involved, the problem to solve is very complex in numerical terms, so a strategy consisting of using three steps to simulate the actual reconstruction and one more step to simulate the rehabilitation process has been used. This early rehabilitation therapy lies in the application of a longitudinal displacement of 3 mm to the tendon, already located in the proper position inside the tibial tunnel, but trying to extract it in the femoral direction.

Simulating the ACL reconstruction process with four different diameters of the tibial tunnel (10, 9, 8 and 7 mm), the stress and strain values in the contact surfaces of the tendon with the trabecular bone and with the screw have been presented (Figures 3 and 4). The results have revealed that the diametrical pressure exerted by the screw against the tendon (Figure 3.a), with peaks and valleys -depending on whether the contact area is the teeth or the area between them-, increases as the diameter of the tunnel decreases. Practically, the maximum value of diametrical pressure is reached when the tunnel passes from a 10 to 9 mm diameter. It also happens with the contact pressure between the tendon and the trabecular bone (Figure 3.b), which, although with a much smoother profile, goes from a value of about 2.5 MPa to 4 MPa when the diameter of the tunnel is reduced from 10 to 9 mm.

Subsequent reductions in diameter of the tunnel (8 mm and 7 mm) have more influence in expanding the area subjected to maximum stress (by increasing the area of contact between the screw and the rest of the materials surrounding it) than in increasing the value of this stress distribution. These results are not far from those obtained by other authors[18], who used a much simpler model, without the presence of a tendon. This author used an interference screw with a diameter of 8 mm and a length of 20 mm that perfectly adjusted to the tibial tunnel and in which the material bone was simplified to only cortical bone behavior, as linear elastic material (E=13.4 GPa, v=0.28). They obtained that the pressures exerted on the bone tunnel did not exceed 3 MPa, a value well below the yield limit of the cortical bone used in said study (182 MPa).

Although the decrease in the diameter of the tunnel barely affects the stress values from 9 mm diameter of tunnel, it does not occur with the strain in the trabecular bone (Figure 4), which grows very noticeably, especially when going from a tunnel diameter of 8 mm to one of 7 mm. When the results of these two tunnel diameters (8 and 7 mm) are compared, although the stresses hardly change, the diametrical deformation in the trabecular bone increases by almost 50% in the contact areas affected transversely by the pressure of some of the screw teeth. The high values of strain reached at certain points could imply a localized deterioration of the trabecular bone and the consequent loss of pressure on the reconstruction.

Based on these results, it is conceivable that the ideal tunnel will be the one that manages to fix the plasty without extremely high values of strain in the trabecular bone, regardless of the pressure reached. Thus, it is necessary to analyze the displacements suffered by the plasty after the rehabilitation process, which are summarized in Figure 5.

Analyzing the results showed in Figure 5, it could be inferred that the use of a 10 mm diameter tibial tunnel is unable to fix a 4 mm plasty with a 7 mm diameter interference screw, so the choice of this combination of parameters would be completely wrong. It should be noted that the verification of screw slippage is usually carried out one or two months after the reconstruction, subjecting the joint to cyclic loads[19]. The fact that there could be tendon slippage in such an early and non-aggressive rehabilitation is a reason to dismiss the tibial tunnel configuration studied with the selected tendon and screw.

However, the use of a 9 mm diameter tunnel already ensures that the tendon is totally fixed in zones 4 and 5 (displacement 0 mm) and that in the rest of the zones the displacement is very small, so it could be considered that the choice of this tunnel diameter would already be sufficiently valid. Nevertheless, if the diameter tunnel used is 8 mm, the displacements suffered by the plasty during rehabilitation are even lower, without the implication of a sudden change in the values of stress and strain in the different elements involved, something that happens when the diameter of the tunnel decrease (Figure 4.b). Then, it seems that the optimal choice for a successful reconstruction of a 4 mm tendon (assuming a double arrangement), using a screw with the analyzed geometry (7 mm in diameter and 25 mm in length) would be a tibial tunnel diameter of 8 mm or 9 mm, which also coincides with that obtained by other authors[20-23] for a similar arrangement. Going below that diameter would mean an excessive increase of the strain in the trabecular bone, without providing any improvement in terms of tendon support.

Having presented the results and discussed them, it is worth noting that the model which has been developed is capable of simulating ACL reconstruction in a reliable and relatively simple manner. Thus, it is emerging as a very useful tool in making decisions about the most appropriate tibial tunnel diameter to use depending on the interference screw and the plasty chosen for the patient.

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Conflict of interest

The authors do not declare a conflict of interest.

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