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Original article:

A wear model to predict damage of reconstructed ACL

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Abstract

Impingement with surrounding tissues is a major cause of failure of anterior cruciate ligament reconstruction. However, the complexity of the knee kinematics and anatomical variations make it difficult to predict the occurrence of contact and the extent of the resulting damage. Here we hypothesise that a description of wear between the reconstructed ligament and adjacent structures captures the in vivo damage produced with physiological loadings. To test this, we performed an in vivo study on a sheep model and investigated the role of different sources of damage: overstretching, excessive twist, excessive compression, and wear. Seven sheep underwent cranial cruciate ligament reconstruction using a tendon autograft. Necropsy observations and pull-out force measurements performed postoperatively at three months showed variability across specimens of the extent and location of graft damage. Using 3D digital models of each stifle based on X-ray imaging and kinematics measurements, we determined the relative displacements between the graft and the surrounding bones and computed a wear index describing the work of friction forces underwent by the graft during a full flexion-extension movement. While tensile strain, angle of twist and impingement volume showed no correlation with pull-out force ($\rho = -0.321$, $p = 0.498$), the wear index showed a strong negative correlation ($r = -0.902$, $p = 0.006$). Moreover, contour maps showing the distribution of wear on the graft were consistent with observations of damage during the necropsy. These results demonstrate that wear is a good proxy of graft damage. The proposed wear index could be used in implant design and surgery planning to minimise the risk of implant failure. Its application to sheep can also provide a way to increase efficiency of preclinical testing.

1. Introduction

Impingement is a major problem in ACL reconstruction (Iriuchishima et al., 2013; Schützenberger et al., 2021; Song et al., 2021; Srinivasan et al., 2018). The contact between a reconstructed graft and adjacent tissues is known to be strongly correlated with graft failure. In a recent retrospective study on 87 patients with a 9.2% re-rupture rate, impingement was found in 80% of the re-rupture cases, and in 69% of the cases with graft degeneration, while it was absent from all the intact cases (Schützenberger et al., 2021). Ruptures caused by impingement correspond to technical errors during implantation, which represent about 21-22% (Chen et al., 2013; Vermeijden et al., 2020), and 48% when combined to other causes. In the particular case of multiple-revisions, technical errors are the main cause of failure (29% isolated, 58% combined). It is now established that the risk of graft impingement can be greatly reduced by a correct tunnel placement adjusted to the individual patient anatomy (Barnum et al., 2021; Iriuchishima et al., 2010). This can be assessed by a preoperative planning procedure consisting in imaging the knee of the patient prior to the surgery, to select a patient specific placement of the graft (Connors et al., 2019). Emerging solutions based on computer assisted surgery might also improve the accuracy of tunnel placement and ensure a better isometry (Lopomo, 2022). Nevertheless, it remains unclear how to predict the occurrence and extent of graft damage from the location and volume of the impingement zone. A more quantitative description of the mechanisms causing damage could provide more effective damage predictors. Such criteria would contribute to reduce the rate of technical failures by refining surgery planning procedures and improving the design of grafts and reconstruction methods.

Impingement implies that a contact with non-zero normal forces is created between the graft and a surrounding surface. Contact force and contact area have been proposed as proxies for graft damage: impinged areas with significant contact pressure exhibit reduced cell viability and disrupted tissue organisation (Fung et al., 2010), which was associated with ligament rupture (Gyger et al., 2007). Park et al. (Park et al., 2010) validated experimentally a finite elements (FE) model and calculated the contact pressure between the ACL and the intercondylar notch, to identify the regions in the

ligament that are the most prone to suffer damage. Using an analogous approach, Orsi et al. showed that the impingement area and force depend strongly on the size and insertion points of the graft (Orsi et al., 2017). Besides compression effects, friction and wear are also expected to occur in the contact areas where there is relative motion between the graft and adjacent tissues. Scanning electron microscopy observations of synthetic ACL substitutes showed that the damage patterns on graft fibres caused by mechanical abrasion tests were similar to those observed on failed specimens explanted from patients (Amis and Kempson, 1999; Guidoin et al., 2000; Poddevin et al., 1997). Wang et al. (Wang et al., 2020) recently proposed to use a wear model to explain graft damage around bone tunnel entrances. For that, they considered static loads and implemented a simplified Archard-Reye model (Archard, 1953; Reye, 1860) reduced to a single variable, the normal force. Their results showed that graft stiffness and circularity may influence graft wear. However, a more complete modelling accounting for the joint kinematics, is missing to assess the wear produced in impingement areas such as the intercondylar notch where large relative displacements can occur between the graft and adjacent tissues.

Our hypothesis is that a wear model that includes both contact forces and relative displacements provides a better prediction of graft damage than the current static assessments. In this work, we conducted an in vivo study on a sheep model of ACL reconstruction. A mini-medial arthrotomy approach implying less tissue dissection and smaller incisions was used to decrease post-operative morbidity. We used 3D imaging and motion capture to reconstruct a digital model of the joint and ACL graft with animal specific anatomy and kinematics. We computed an index of the wear underwent by the graft during a full movement of joint flexion-extension and checked if it can predict damage of the graft in-vivo at three months, as characterised by the pull-out force.

2. Materials and methods

2.1 Surgery

We used seven healthy nulliparous prealpes ewes aged 20-41 months, obtained from a licensed vendor. Before each anaesthesia procedure, a dose of penicillin–streptomycin (1 g/sheep, Bistreptine; Virbac France SAS) was administered. Anaesthesia was induced by intravenous administration of mix of ketamine (6 mg/kg, Imalgene Boehringer) and diazepam (0.5 mg/kg, Valium, Roche) and maintained by isoflurane (Isorane, Axience) in oxygen after endotracheal intubation. The sheep was positioned dorsal decubitus position and the left pelvic limb was prepared for surgery under aseptic conditions. Perioperative analgesia was provided by intravenous administration of morphine (0.2 mg/kg, as needed, Morphine, Aguettant).

The procedure was carried out as follows. A 2 cm-long mini arthrotomy of the left stifle was first performed. The fat pad was partially resected, and video-assisted resection of the ACL was performed with a 2.7 mm, 30° fore oblique vision and arthroscopic scissors, taking care to cut the two bundles of the cranial cruciate ligament (CrCL) close to its femoral and tibial attachments. The drawer sign was checked to confirm the resection of the two bundles of the CrCL. A 4 cm-long lateral supracondylar approach of the femur was performed. An Arthrex paediatric drill guide was placed on the femoral footprint of the CrCL with arthroscopic assistance and a 6 mm femoral bone tunnel was drilled outside-in. A 2 cm-long medial approach of the proximal tibia was subsequently performed. The guide was placed on the tibial attachment of the ACL caudomedially to the anterior horn of the lateral meniscus ligament on the insertion of the posterolateral bundle of the ACL; close to the base of the tibial eminences and a 6 mm bone tunnel was drilled outside-in in the tibial metaphysis. The opening of the tibial tunnel was targeted to exit on the caudal footprint of the CrCL, immediately in front of the tibial inter-eminence ridge to spare the ligamentous attachment of the anterior horn of the lateral meniscus. A flexor digitorum superficialis tendon autograft harvested from the same limb as previously described (Hunt et al., 2005). The single-bundled autograft was passed through the bone tunnels and subsequently secured with 6 mm, 25

to 30 mm-long TiAl6V interference screws, implanted outside-in in the bone tunnels. Isometry and absence of cranial drawer were controlled before conventional closure. The operated limb was held in its physiological rest position, which corresponded to a joint flexion angle of $\theta = 50° \pm 5°$ and was measured a posteriori using a protractor. The whole experiment including animal care and housing were performed in compliance with the animal experimentation legislation (24.11.1986.86/609/CEE) and approved by an ethical committee (ANSES-ENVA-UPEC, N° APAFiS: 2015122216393903, N° of approval 12/01/16-9-B).

2.2 Post operatory follow-up

Postoperative analgesia was provided by subcutaneous administration of meloxicam (0,4 mg/kg, Metacam, Boehringer) for 3 days and, if needed, buprenorphine (20 µg/kg, Buprecare, Axience). Drawer tests were performed right after the surgery on asleep animals to assess the stability of the stifle. Animals were free to move about all through the 3 months of the experiment. Clinical evaluations of the awake animals were carried out during the first week, at 1 month and at 3 months (Viateau et al., 2013). They consisted of a general examination (exploratory activity, feeding, surgical wound status) and a visual assessment of the functional recovery by the absence or presence of lameness and support on the operated limb. Joint swelling and heating were assessed by palpation and compared to the contralateral stifle. Healing was assessed based on clinical examination of the wounds and visual gait analysis.

2.3 Necropsy

All ewes were sacrificed three months postoperatively by barbiturate overdose. Right after, both pelvic limbs were explanted by coxo-femoral disarticulation and frozen at -20ºC. Limbs were thawed at room temperature for 24 hours and subsequently dissected. All muscles were removed for the kinematic tests, except for the distal third of the quadriceps, the patella and the proximal fourth of the lateral and medial gastrocnemius muscles. The articular capsule and the ligaments of the knee were carefully preserved. For pull-out tests, these structures, the menisci and the caudal

cruciate ligament were removed to leave only the CrCL. The cartilaginous surfaces of the patella, the femoral condyles and the menisci were inspected and chondral lesions were graded using a modified OARSI score (Little et al., 2010; Pritzker et al., 2006).

2.4 Biomechanical tests and imaging

The biomechanical tests consisted of four ex-vivo experiments: three kinematic tests (flexion-extension, tibial translation, and varus-valgus) and a pull-out destructive test. Tripods with passive infrared markers were fixed to the femur and the tibia for motion capture by an optoelectronic tracking device (Polaris, NDI, Waterloo, Ontario, Canada). Anatomical frames associated to each bone were defined following a previously validated procedure (Schlatterer et al., 2009). Biplanar X-ray images of each specimen and corresponding tripods were acquired (EOS, EOS Imaging, Paris, France) and 3D digital models of femur and tibia were obtained using a reconstruction algorithm (Chaibi et al., 2012) validated for the sheep (Guerard et al., 2014). The registration between the EOS system of reference, and the motion capture system of reference was done by defining technical frames associated with the tripods (Azmy et al., 2010; Guerard et al., 2014; Rochcongar et al., 2016). The kinematic tests were performed using motorised devices previously validated (Guerard et al., 2014; Rochcongar et al., 2016). The femur was rigidly fixed, and the tibia was set in motion according to the different mobilities tested (10N force applied to the tibial plafond for flexion-extension, force applied to the tibial plafond until a torque of 5 N·m was reached for the varus-valgus, tensile force of 100 N was applied on the third part of proximal tibia perpendicular to the tibial shaft for the anterior drawer test). For each test, the 3D kinematics of the knee was computed from the instantaneous position of the tripods (femoral frame and zxy cardan sequence for the flexion-extension, tibial frame and xyz cardan sequence for the varus-valgus test). Translations were defined as the displacement of the centre of tibial frame between the initial and the final position expressed in the femoral frame. The flexion angle was defined as the angle between two 3D vectors: i) a vector passing through the middle of the line joining the centres of the condylar spheres—two spheres constructed by the least squares fit of the posterior aspect of the two condyles— and the centre of the femur head, and ii) a vector passing

through the middle of the talar dome and the middle of the peaks of the tibial intercondylar eminence (figure S1).

For the pull-out tests (figures S2 and S3), the distal femur and proximal tibia were cut 10 cm from the joint and the free bony ends were embedded in steel cylinders using a low temperature melt alloy (MCP70, Mining & Chemical Product, Wellingborough, UK). The stifles were fixed on an Instron 5566 testing machine (Instron Ltd., Buckinghamshire, England) instrumented with a 5 kN load cell and tested at room temperature. For the operated limbs, the joint was set in position aligning the axis of the graft with the axes of both tunnels, so the force was applied along the entire axis of the graft. For the contralateral limbs, the joint was positioned so the force is applied along the axis of the CrCL. This position which is not anatomically accessible, maximizes the shear forces at the implant-bone interface and was chosen as a worst-case scenario. No quantitative measurements of the flexion angle were performed. Stifles were conditioned using 10 cycles between 5 and 50 N (5 mm/min) to ensure that the structure of the specimens was at a repeatable reference state (Cheng et al., 2009). Finally, a tension load was applied to the specimen (5 mm/min) until failure. Ultimate tensile force was determined from the experimental load-elongation curve. From the results of the biomechanical tests, the following parameters were extracted: anterior tibial translation (mm) for 100 N loading during the anterior drawer test, varus valgus amplitude (°) at 5 N·m loading for the varus valgus laxity test, and the pull-out force (N) from the tensile test.

2.5 Statistical analysis

The sample size (n = 7) was the minimum necessary to measure a significant difference ($\alpha = 0.05$, power $1-\beta = 0.9$) in pull-out force compared to the mean value of pull-out for reconstructed CrCL using autograft as reported in the literature for sheep at 3 months: 1532±180 N (Weiler et al., 2002). Differences in necropsy and biomechanical variables among operated and control (contralateral) limbs were assessed using one-sided Wilcoxon signed rank tests (non-normal data to Shapiro-Wilk tests). A correlation study was conducted to identify relationships between the local

damage descriptors of the reconstructed CrCL (rCrCL)—maximum tensile strain, maximum angle of twist, maximum impingement volume and maximum cumulated wear index—biomechanical data (tibial translation, varus-valgus, pull-out) and necropsy data (OARSI grading). Spearman and Pearson correlations were calculated pairwise for all the studied variables. Normal data were expressed as mean ± standard deviation (angle of flexion), and non-normal data were expressed as median, [minimum, maximum]. The level of significance was 0.05 for all the tests. In the figures, p-values < 0.05, < 0.01 and < 0.001 were indicated with *, ** and ***, respectively.

3. Wear model

Simulations of reconstructed knee kinematics and modelling of damage criteria were performed using Matlab R2019b with the package geom3d (D. Legland, INRA Nantes, France).

3.1 Modelling of joint anatomy and kinematics

Subject-specific 3D surface meshes of femora and tibiæ of the 7 implanted limbs were obtained from EOS images (figure 1a). Kinematic data obtained experimentally (figure 1b) were converted into transition matrices and applied to these meshes. For each flexion angle θ from full flexion (θ_{flex} = 93.4º±4.4º) to full extension (θ_{ext} = 26.5±2.6º), the positions of femur and tibia were obtained at a sampling rate of 10 acquisitions per degree of flexion (figure 1c). When measured using a goniometric procedure (see figure S1), this range of motion (33±3º in extension, 100±5º in flexion) is comparable to the reported for sheep in vivo (34±6º in extension, 134±4º in flexion, 180º-β) (Govoni et al., 2012), not reaching hyperflexion nor hyperextension.

3.2 Modelling of the reconstructed ACL

The position of the osseous tunnels was derived from the position of the interference screws, which were visible on the roentgenograms (figure 2a). For each bone tunnel, the coordinates of the head and tip of the screw were imported into the 3D model. The insertion point of the graft was then taken as the intersection of the bone mesh with a line connecting these two points. The positions of the femoral and tibial insertion points are reproduced in figure S4 for each specimen. The graft was modelled as a cylinder joining the insertion points as portrayed in figure 2b.

3.3 Tensile strain and angle of twist

The undeformed length L_0 of the graft was defined as the euclidean distance between the femoral and tibial insertion points at $\theta = 50°$, the angle at implantation. The tensile strain along the graft axis $\varepsilon(\theta)$ was calculated as $(L(\theta)-L_0)/L_0$ where $L(\theta)$ is the distance between insertion points at each angle θ as illustrated in figure 3a. Similarly, the angle of twist $\Phi(\theta)$ was calculated as the rotation angle of the surface mesh of the cylindrical graft around its axis as compared to the reference unloaded position at $\theta = 50°$.

Figures 3b and c show the evolution of ε and Φ as a function of angle θ for specimen 1 (see Figures S5 and S6 for specimens 2-7). For all specimens, the graft stretched continuously as the knee extended. Maximum strain ε_{max} was reached at full extension θ_{ext}. The angle of twist also increased during extension. Apart from specimen 1, the maximum angle of twist Φ_{max} was reached at θ_{ext}. Both ε_{max} and Φ_{max} show significant variations among the specimens as shown in figures 3d and e.

3.4 Impingement volume

Grafts diameters were measured just before implantation. A mean value of $D_0 = 7$ mm was measured using a slotted graft sizing block. During knee flexion-extension, the graft diameter was set to vary with $\varepsilon(\theta)$ as follows:

$$D(\theta) = D_0 \cdot (1 - \nu \cdot \varepsilon(\theta)) \quad (1)$$

where D_0 is the diameter measured before implantation and ν is the Poisson coefficient set to 0.5 under the assumption of incompressibility.

In the simulations, no contact conditions were imposed: femur, tibia and graft meshes could interpenetrate each other. Impingement volumes were calculated as the intersection volumes of the graft with tibia and femur. Impinged zones are represented at each flexion angle in figure 4a and video 1 for specimen 1. For most specimens, impingement in full flexion (θ_{flex}) occurred at the tibial plateau and at the lateral condyle. As the joint extends, impingement with the intercondylar notch became more pronounced. The evolution of the impingement volume during flexion is shown in figure 4b for specimen 1 and figure S7 for specimens 2-7. We observed considerable variability across specimens with non-monotonic evolutions during extension.

The maximum impingement volumes reached during extension are gathered in figure 4c for all the specimens. They range from 100 mm^3 to about 450 mm^3. The smallest value was obtained for specimen 4 and is due to a slight impingement with the tibia. The largest ones (specimens 2, 3 and 7) correspond to large impingements with both femur and tibia.

3.5 Calculation of wear

We propose a model based on Archard–Reye hypothesis stating that wear—progressive loss of matter—is proportional to the work done by the friction forces. For an extension movement from θ to θ-$\delta\theta$, the local work of friction forces $\delta W_f(P,\theta)$ applied on a material point P at the surface of the graft can be expressed as:

$$\delta W_f(P,\theta) = F_f(P,\theta) \cdot \frac{\delta x}{\delta \theta} \cdot \delta\theta \qquad (2)$$

where $F_f(P,\theta)$ is the friction force applied in point P at angle θ and δx is the component of the movement along the direction $\overrightarrow{u_T}$, tangential to the surface between angle θ and θ-$\delta\theta$ (figure 5a).

Applying Coulomb's law, the friction force is proportional to the normal force through the coefficient of friction μ. Following Hertzian contact theory, we assume that the normal force $F_N(P,\theta)$ applied in point P at angle θ scales as $h(P,\theta)^{3/2}$, where $h(P,\theta)$ is the penetration depth of the implant into the bone, as depicted in figure 5b. Here $h(P,\theta)$ is defined as the distance between point P and the bone surface in the direction of the graft radius. In the absence of impingement, $h(P,\theta)$ is set to 0. As a result, $\delta W_f(P,\theta)$ is expected to scale as:

$$\delta W_{\mathrm{f}}(\mathrm{P},\theta) \sim h(P,\theta)^{\frac{3}{2}} \cdot \frac{\delta x}{\delta \theta} \cdot \delta\theta \tag{3}$$

By integrating the right-hand side of equation (3) between a given angle θ and θ_{flex}, we define a local wear index $WI(P,\theta)$ as follows:

$$WI(P,\theta) \equiv \int_{\theta}^{\theta_{\mathrm{flex}}} h(P,\theta)^{\frac{3}{2}} \cdot \frac{\mathrm{d}x}{\mathrm{d}\theta} \cdot \mathrm{d}\theta \tag{4}$$

This index $WI(P,\theta)$ is expressed in arbitrary units and quantifies the wear accumulated at each node P of the graft between θ_{flex} and θ. The values of this wear index at the surface of the graft during extension from θ_{flex} to θ_{ext} are shown in figure 5c and video 2 for specimen 1. In this case, wear was localised at the proximal half of the graft, as better shown in figure 5d using a contour map of the developed graft surface (see figure S8 for specimens 2-7).

For a given angle θ, we assess the maximum wear by calculating the maximum value of the wear index $WI_{max}(\theta)$ attained on the graft surface. Figure 5e shows $WI_{max}(\theta)$ for specimen 1 (see figure S9 for specimens 2-7). These curves show wear accumulating progressively as the stifle extends. This accumulation can be strongly non-linear. For instance, in the case of specimen 1, wear is more pronounced below 45º. This suggests that the whole flexion-extension movement must be integrated to fully characterise wear. We hypothesise that the region of maximum wear produced during full flexion-extension corresponds to the most damaged part of the graft and thus determines its strength. Accordingly, the value of $WI_{max}(\theta_{ext})$ was used as a proxy for graft damage. Figure 5f shows the values of $WI_{max}(\theta_{ext})$ for all the specimens.

4. Results of the in vivo study

4.1 Post operatory follow-up

Drawer tests performed immediately after surgery showed mild (< 2 mm) instability in 3 of the 7 ewes. Functional recovery was good for all the ewes and weight-bearing on the operated limb occurred within the first 24 hours. Joint swelling, consistent with the normal postoperative outcome, was present in all ewes for the three months of postoperative follow-up. At 3 months, donor sites had healed and were painless to palpation. No plantigrady was observed.

4.2 Necropsy

Necropsy observations revealed cartilage lesions in all the implanted stifles, as shown in figures 6 and S10. These lesions were in the lateral (3/7) and medial (4/7) femoral condyles and lateral (5/7) and medial (7/7) tibial condyles (figure 6a), on the patella (6/7) (figure S10a) and on the trochlea (6/7) (figure S10b). Reconstructed stifles had a significantly higher cartilage damage in OARSI score than the non-operated contralateral stifles (15.00, [6.00, 19.00] vs 5.00, [4.00, 9.00], $p = 0.016$) as shown in figure 6b. Specimen 7 showed erosion of the posterior circumferential rim of the lateral meniscus (figure S10c). The rest of the menisci had a normal appearance. 5/7 operated stifles had osteophytes in the patella or trochlea (figure S10d) compared to 2/7 of non-operated stifles. All grafts were in place and synovialised at the time of examination. 2/7 were partially torn at their periphery (figure S10e) and 5/7 showed adhesions to the posterior cruciate ligament.

4.3 Biomechanical characterisation

Anterior tibial translation and varus-valgus displacement measurements (figure S11a and b) showed that operated joints were laxer than their respective, non-operated contralateral joints. Anterior tibial translation (mm) at 100 N loading was on average 200% higher in operated limbs (3.20, [1.80, 4.10] mm vs 0.9, [0.50, 1.20] mm, p =

0.008) as shown in figure 7a. For a 5 Nm varus-valgus load, mobility of operated stifles (º) was on average 50% higher than in non-operated ones (16.00, [12.80, 19.00] vs 9.00, [6.40, 20.90], p = 0.016) as shown in figure 7b.

For reconstructed CrCLs, pull-out tests led to rupture of the grafts in the mid-substance, within the proximal third of the graft and away from the insertion points. For the non-operated CrCLs, rupture occurred in the mid-substance except for specimens 2 and 7 for which bone avulsion occurred at the femoral insertion (see figure S3). Pull-out force (N, figures 7c and S11c) of reconstructed CrCL (rCrCL) was 80% lower in average than at the non-operated CrCL (207, [77, 300] N vs 904, [720, 1165] N, p = 0.008). The statistical dispersion of the biomechanical variables was higher in rCrCL than in non-operated CrCL. The coefficient of variation of the tibial translation was 7.5% higher in rCrCL. That of pull-out force was 25.8% higher in rCrCL. These variations in the biomechanical performance of the rCrCL may reflect variations in the host anatomy and/or in the procedure.

4.4 Model validation with in vivo data

The scatterplot matrix in figure 8 shows the pairwise comparisons between the local damage descriptors of the rCrCL (maximum tensile strain, maximum angle of twist, maximum impingement volume and maximum cumulated wear index), biomechanical data (tibial translation, varus-valgus, pull-out) and necropsy data (OARSI grading). Spearman (ρ) and Pearson (r) correlation coefficients are summarised in table ST1. Laxity variables—tibial translation and varus-valgus—were positively correlated (ρ = 0.811, p = 0.038): the looser the stifle was in one d.o.f., the looser it was in the other.

The only significant correlation between the local damage descriptors and the biomechanical variables is a strong, negative linear correlation between the wear index $WI_{\max}(\theta_{ext})$ and the pull-out force (r = -0.902, p = 0.006). This result suggests that the differences in tensile strength across specimens can be attributed to differences in wear. Other moderate but non-statistically significant correlations were found between varus-valgus and $\varepsilon_{\max}$ (r = 0.613, p = 0.144), $\varepsilon_{\max}$ and wear index (ρ = -0.571, p = 0.200), and

between impingement volume and the wear index ($\rho = 0.571$, $p = 0.200$). The last could be explained by wear occurring more likely when there is a large impingement volume.

No correlations were found between the level of chondral damage quantified by the OARSI score and biomechanical descriptors of graft damage. Yet, some necropsy observations allowed us to find a correspondence between the location of damage and the areas of maximum wear predicted by the model, as shown in figure 9 for specimens 2 and 6. The model predicted the apparition of wear on the proximal end of the grafts in both specimens. More generally, the model predicted that the maximum wear is produced in the proximal third of the graft (Figures 5d and S8), which is consistent with pull-out observations showing that the operated specimens broke in the mid-substance near the proximal attachment (Figure S3).

5. Discussion

In this study, we corroborated the hypothesis that a wear model accounting for both contact forces and relative displacements provides a better prediction of graft damage than a static assessment. We found that the impingement volume alone is not sufficient to explain the damage produced on reconstructed CrCL. The significant correlation between the wear index and the pull-out force suggests that it is necessary to consider the friction process by measuring descriptors of both the normal force and the relative displacements. This effect is well illustrated by comparing specimens 3 and 7, which both had the greatest impingement volumes (~ 420-440 mm^3) as shown in figure 4c. Specimen 3 had a fairly high pull-out force (177 N) while specimen 7 was the weakest graft (77 N). The wear assessment explains this difference. The large impingement of specimen 3 was not accompanied by large tangential displacements: the graft was mostly squashed onto the bone which resulted in a moderate wear, as reflected by the wear index (1806 arb.u.). On the contrary, large tangential displacements occurred in specimen 7, leading to high wear, as indicated by the high wear index (4379 arb.u.). These results also highlight the sensitivity of wear to the joint anatomy and to the position of the insertion points, as shown in figure S4. If we compare specimens 6 and 7, similar anatomy but different femoral insertions lead to different wear indices (189

vs 4379 arb.u., respectively). Interestingly, specimen 6 is the closest to an anatomical implantation and ranks the best on both the wear index (189 arb.u) and the OARSI score (6.0). These observations suggest that surgery planning procedures using a wear assessment could be proposed to optimise implant placement and reduce the risk of graft failure.

As shown by a previous study at 52 weeks, the strength of the joint decreases between weeks 6 and 9. At week 12 the pull-out force has not reached its maximum and the graft may be still under remodelling (Weiler et al., 2002). Weiler explained this by a decrease in the fixation strength due to the degradation of the resorbable screws they used. In our study, we used non-resorbable interference screws. Moreover, the anterior tibial translation displacements we obtained already at 12 weeks (2.8±0.9 mm, 100 N load) were similar to the ones reported by Weiler at 52 weeks (2.78±0.5 mm, 50 N load), and smaller than the ones they obtained at 12 weeks (3.97 ±1 mm). Therefore, we considered a healing time of 12 weeks sufficient for stabilisation. A more recent work found no difference in pull-out force between 6 and 12 weeks and considered joints to be stable at that point (Stodolak-Zych et al., 2022). In our study, we verified the hypothesis that the differences in pull-out force observed among specimens at 3 months reflect differences in the state of damage and that this can be related to the effect of wear.

Our model has the following limitations. First, measuring joint kinematics in vivo using pedicle screws and marker tripods would have provided more physiologically accurate data. We opted for an ex-vivo procedure which validity has been demonstrated previously (Azmy et al., 2010; Chaibi et al., 2012; Guerard et al., 2014; Rochcongar et al., 2016). The specimens were carefully prepared to ensure that the capsule and all the knee ligaments were kept intact. The effect of skin and muscle dissection, and the lack of muscle activation might have an effect on the kinematics, and in particular might result in joint hyperflexion. Nevertheless, our measurements show that the range of flexion angles is within are within physiological values (Govoni et al., 2012). Hence, it is reasonable to neglect the effect of hyperflexion in this protocol. This ex vivo procedure also has the advantage of capturing the full range of motion of the joint in a

limited time. During the three post-operatory months, the stifle of the sheep not only worked within the bounded flexion extension range of the gait cycle, but also in episodes of pronounced flexion (animal laying down) and pronounced extension (animal standing on its hindquarters). Our results show that in several specimens, wear was more pronounced in positions close to full extension. Therefore, the acquisition of kinematic data through the joint's full range of motion may be more relevant for this purpose.

Second, grafts were modelled as cylinders, which is a simple approximation to the geometry of the real graft (Dabirrahmani et al., 2013; Fleute et al., 1999). The amount of wear underwent by a non-cylindrical graft may thus vary with the deviation of its section from circularity.

Third, grafts were assumed to be incompressible (Peña et al., 2006), but ligaments lose volume when stretched and hence higher Poisson's ratios may be as well justified (Lynch et al., 2003; Swedberg et al., 2014). This may result in a different wear because of a smaller diameter in extension than the one modelled. However, as far as the graft is implanted not far from the isometric points, we expect this difference to be minor.

Fourth, a source of error may arise from the method to locate the insertion points. Here, the bone tunnel axis was inferred from the orientation of the interference screws. The screws lie on the side of the graft and their axis may not be perfectly aligned with the axis of the tunnel. However, this misalignment is expected to be limited because the x-ray observations showed that the screws were completely inserted into the tunnel.

Fifth, the Archard–Reye wear model has already been used to calculate wear damage on soft tissues such as articular cartilage (Li et al., 2011; Popov, 2019). This is, to the best of our knowledge, the first time that a Archard-Reye based wear model considering forces and relative displacements is used to assess damage of ligaments. In this study, the model is based on a geometrical description of the graft-bone contact. Following Hertz theory, we approximated the normal force by a function of the penetration depth. This theory assumes elastic contact between the two bodies, but does not account for viscoelastic effects that may be present on the soft biological tissues (Machado et al.,

2011; Sverdlik and Lanir, 2001). Elastic Hertz theory was found to be robust to an intermediate viscoelastic behaviour (He and Wettlaufer, 2014) and Hertz models have been used recently to describe contact between articular cartilage (Guess et al., 2013; Purevsuren et al., 2020), which is known to be viscoelastic (Fulcher et al., 2009; Temple et al., 2016). However, when comparing an elastic and a viscoelastic model for sphere-plane contact, Sameur et al. found that the contact force is higher when viscoelasticity is taken into account (Sameur et al., 2008). This may imply that the wear index may depend on the relative speed of friction during flexion-extension. A more accurate, but computationally costly, description of contact could be obtained with a FE model including kinematic data (Limbert et al., 2004).

Sixth, the comparative study between model and necropsy images is limited by the reduced visibility of the necropsy images. A better assessment of internal damage could be done through systematic MRI observations.

Finally, the impact of fat pad resection on biomechanical characteristics of operated knees may be questioned. The partial fat pad resection, which was performed in our model to improve visualisation of the femoral CrCL footprint, may have altered joint kinematics. Modifications of the patella-femoral kinematics have indeed been observed in an ex-vivo trial in human cadaveric knees in which total fat pad resections were performed (Bohnsack et al., 2004). Whether or not these modifications apply to the sheep knee remains to be established. We did not find in the kinematic analysis, decrease of the tibial external rotation relative to the femur combined with medial translation of the patella. To the best of our knowledge, the effect of fat pad resection in sheep has not been published. A former experimental trial in goat have shown that, in CrCL central defect type injury model, resection of the fat pad did not significantly impact ATT values and of ultimate tensile loads and ultimate elongation but increased injured ligament stiffness (Karakilic et al., 2015). The question of removing or preserving the fat pad in the sheep model used here and its potential effects on joint mechanics in that species remains opened. Preservation of the fat pad in a model of ACL replacement in sheep has indeed been shown to induce an inflammatory response

which may also disturb postoperatively the knee's biomechanical characteristics (Solbak et al., 2015).

We show with this work that a wear model captures the location and intensity of graft damage. The methodology and the model (calculation method) are independent of the species to which it is applied. In particular, the simplicity of the proposed wear assessment could facilitate its implementation in existing computer assisted surgery systems to select the optimal implantation sites in human patients. Today, systems using 3D imaging techniques (CT, Optotrack™ or equivalents) provide a way to reduce both overstretching and impingement. Based on pre- or peri-surgery images, reconstructed 3D models of the joint surfaces allow the exploration of different graft placements (Dessenne et al., 1995; Fleute et al., 1999; Fung and Zhang, 2003; Julliard et al., 1998). To include wear in the predictions, joint geometry data can be acquired at different knee flexion angles, and the wear index and its localisation, as well as graft stretching, can be computed all through a knee flexion-extension. The computations of the wear index are light enough to be performed in real time and could be displayed directly on a screen to aid surgeons choose an optimal couple of femur–tibia insertion points. Certainly, such study would have to be compared with a control group using the state-of-the-art computer assisted surgery system. This calculation also gives access to the local distribution of wear on the graft and can be used to identify damage spots and optimise the design of ligament substitutes. In addition, in vitro fatigue tests under standardized conditions would be of utmost interest to gain a deeper understanding of the microscopic features governing the wear process in the impingement zone.

The current animal models in sheep, goat and pig use the cranial cruciate ligament (CrCL) as an analogue of human ACL. Even though the anatomical differences between these models and humans regarding the ACL are important (Bascuñán et al., 2019), the causes of CrCL reconstruction failure in these models have been little studied. Paradoxically, stifle anatomy is particularly prone to CrCL impingement, especially in sheep and goat, because of a smaller trochlear width and a deeper intercondylar notch (Osterhoff et al., 2011; Proffen et al., 2012). To the best of our knowledge, there is no information in the literature addressing considerations specific to each animal species

for optimal placement of ACL grafts or synthetic ligaments. Unnoticed graft misplacements can prevent discerning whether the rupture is due to the implant or the implantation procedure, and wrong conclusions might be thus drawn. Preoperative planning as proposed in our study may improve the implantation accuracy in sheep animal model and allow graft placement in a controlled and repeatable manner. This provides a straightforward way to increase efficiency of preclinical testing, by reducing the resources, costs, time, and number of animals used.

6. Conflict of interest statement

The authors have no affiliations with or involvement in any organisation or entity with any financial or non-financial interest in the subject matter or materials of this work.

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