Biomechanical evaluation of augmented and nonaugmented primary repair of the anterior cruciate ligament: an in vivo animal study

Helmut Seitz · Wolfgang Pichl · Veronika Matzi · Thomas Nau

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Abstract

Purpose The purpose of this study was to evaluate in a sheep model the biomechanical performance of augmented and nonaugmented primary repair of the anterior cruciate ligament (ACL) following transection at the femoral end during a 12-month postoperative observation.

Methods Forty sheep were randomly assigned to nonaugmented or augmented primary ACL repair using a polyethylene terephthalate (PET) band. At two, six, 16, 26 and 52 weeks postoperatively four sheep in each group were sacrificed and biomechanical testing performed.

Results Compared with nonaugmented primary ACL repair, the PET-augmented repair demonstrated superior biomechanical results from 16 weeks postoperatively onwards in terms of anteroposterior (AP) laxity, tensile strength and ligament stiffness. The augmentation device works as a stress shield

during the ligament healing process. The nonaugmented ACL repair also resulted in ligament healing, but the biomechanical properties were at a significantly lower level.

Conclusion These results support the previously reported histological findings following augmented primary ACL repair. This animal study on the healing capacity of the ACL may provide some important contributions to how primary healing in certain types of ruptures can be achieved.

Clinical relevance I

Keywords ACL · ACL repair · PET-augmented repair · Nonaugmented primary repair

Introduction

The number of anterior cruciate ligament (ACL) injuries is increasing, with a reported incidence of 38/100,000 per year [1, 2]. The ACL represents an important mechanical stabiliser during knee-joint movement, and its mechanoreceptors provide important proprioceptive input to the sensorimotor system. Acute ACL tears cause enormous changes not only in stability, but also in knee-joint kinematics, and also reduce proprioceptive function [3–5]. In addition, meniscal injuries and early osteoarthritis may occur if adequate stabilisation is not achieved [6]. Although there is still some debate about the optimal treatment for acute ACL tears, surgical management in the young and active patient is recommended [7–10]. Operative procedures include extra-articular techniques, primary repair and reconstruction using autografts, allografts, xenografts, and artificial ligaments [10–18]. Currently, reconstruction using autograft or allograft tendon material seems to be the gold standard of surgical treatment. Despite the generally accepted clinical results achieved by ACL reconstruction, there is ongoing research interest in this topic, as evident problems remain. Problems associated with the use of

H. Seitz
Department of Trauma Surgery, Medical University of Graz,
Auenbruggerplatz, Graz, Austria

W. Pichl
Department of Physics of Functional Materials, University of Vienna,
Strudlhofgasse 4, Vienna, Austria

V. Matzi
Department of Surgery, Division of Thoracic and Hyperbaric
Surgery, Medical University of Graz, Graz, Austria

T. Nau (✉)
Ludwig Boltzmann Institute of Experimental and Clinical
Traumatology, Donaueschingenstrasse 13, Vienna, Austria
e-mail: office@traumanau.com

T. Nau
The Austrian Cluster for Tissue Engineering, Vienna, Austria

T. Nau
IMSART, Institute for Musculoskeletal Analysis Research and
Therapy, Piaristengasse 2-4, Vienna, Austria

autografts are mainly linked to graft necrosis during the early postoperative period, the loss of proprioception and the inexact anatomical form of ACL reconstruction [19, 20]. Even with reconstruction, patients are known to be at risk for posttraumatic arthritis despite subjective and objective joint stability [21–23].

Although abandoned in the past, there is renewed interest in the primary healing capacity of this ligament [24–27]. In younger people in particular, ACL ruptures often occur at the femoral attachment of the ligament [28, 29]. In a previous study, we reported on histological changes that occur during the healing process [21]. Performing a primary repair offers the possibility of preserving the natural anatomical structures and sensory receptors [30–33]. Augmented and nonaugmented techniques use different materials for augmentation, but a detailed biomechanical testing and comparison has not been undertaken thus far.

The objective of this study was to investigate both augmented and nonaugmented primary ACL repair focusing on the biomechanical changes during a 12-month postoperative observation period. We hypothesised that augmented repair would result in significantly better biomechanical performance compared with nonaugmented repair.

Methods

Experimental animals

Following the approval by the local ethical committee, 40 female mountain sheep, with an average age of two years and mean body weight of 55 kg, were selected. Just before surgery, the sheep were randomly assigned for nonaugmented ($n=20$) or for augmented ($n=20$) primary ACL repair. Contralateral ACLs were used as controls in all animals and formed the control group.

Augmentation device

As previously described, a 3-mm polyethylene terephthalate (PET) band (Telos GmbH, Marburg, Germany), registered under the brand name Trevira® hochfest type 730 (Hoechst AG, Frankfurt/Main, Germany), was used to augment the ACL repair [21]. With the animal under general anaesthesia, a unilateral, right-sided, open ACL surgery was performed under sterile conditions. The ACL was exposed and transected with a retrograde knife (Arthrex, Naples, FL, USA) at its femoral attachment.

Group 1: nonaugmented primary repair, $n=20$

Using the previously described technique, a lateral supracondylar approach to the distal femur was performed [21]. A 2-mm-diameter canal was drilled through the lateral femoral condyle, starting in the anterior region of the femoral

ACL attachment. Similar to the technique described by Marshall et al. [34], the ACL was repaired using Ethipond (Ethicon, Norderstedt, Germany) USP 0 (3.5) sutures. The posterolateral suture ends were guided through the femoral tunnel and the anteromedial suture ends through the posterior capsule and over the top of the lateral femoral condyle. With the knee at 45° of flexion, the corresponding sutures were tied (Fig. 1).

Group 2: PET augmented primary repair, $n=20$

Two 2-mm tunnels were created—one through the lateral femoral condyle as described above, and one through the medial tibial condyle, exiting behind the medial aspect of the tibial ACL attachment. The tunnels were then enlarged to 4.5 mm. The PET band was inserted into the joint through the tibial tunnel. Then the artificial ligament and posterolateral sutures were led through the femoral condyle to the distal lateral aspect of the femur. The proximal end of the PET ligament was fixed at the femur with a 2.7-mm AO cortical screw and a washer (Mathys, Bettlach, Switzerland) and a 4-mm-wide staple (Telos). With the knee at full extension, the ligament was pretensioned to 60 N and fixed with two 4-mm staples to the tibial head using the belt-buckle technique. Then the remaining Ethibond sutures were led through the posterior capsule over the lateral femoral condyle, and the corresponding sutures were tied with the knee in 45° of flexion (Fig. 1).

Control group: intact contralateral ACL, $n=40$

The contralateral knee joints with intact ACL served as controls for both experimental groups. The anatomical preparation and experimental setup was done in the exact same way as in the experimental groups.

Postoperative management

Postoperatively, the animals received analgesia intravenously for four days. Full weight bearing and unlimited range of motion (ROM) was allowed immediately. After ten days, the animals were transferred to a farm and had weekly examinations thereafter.

Morphological investigations

At two, six, 16, 26 and 52 weeks postoperatively, four sheep from both groups were sacrificed. The femur and tibia were cut 12 cm from the joint line. All periarticular and intra-articular soft tissues except the ACL were removed. With the knee in 90° of flexion, macroscopic assessment of the ligament was performed following the PET band removal from the augmented ACL specimen. Prior to testing, the bony ends of the specimen were mounted into custom-made aluminium cylinders with polyester resin and additionally transfixed with two

Fig. 1 Surgical technique

stainless steel nails perpendicular to the longitudinal axis. The knees were rigidly attached to a servoguided closed-loop testing machine (Instron, Model 1253, High Wycomb, UK) using a 5,000-N load cell. During the tests, the specimens were moisturised with normal saline. Anterior–posterior (AP) laxity testing was performed with the knee joint in 45° of flexion and the tibia in a horizontal position. The force was directed in a vertical direction, perpendicular to the tibial axis, creating a linear anterior translation of the tibia. All other degrees of freedom of motion were locked in the neutral position. To avoid any irregularities in the starting point, a pretension of 0.5 N was used. Five testing cycles of anteriorly directed loads of 50 N, each with a velocity of 5 mm/min, followed by immediate relaxation, were performed. Displacement data of the fifth cycle were acquired to depict the load–displacement curve using standard computer software (Excel, Microsoft Corp, Redmond, WA, USA). Following the cyclic testing for AP laxity, the specimens were prepared for tensile testing. Tibia and femur were both rigidly fixed using a different set of custom-made devices and positioned so the mechanical axis of the ACL was co-linear with the load axis of the testing device. Without pretension, a tensile load with a ramp at 100 mm/min was applied until failure. Displacement to failure, maximum failure load and linear stiffness were determined.

Statistical analysis

Statistical evaluation of the biomechanical parameters, AP laxity, tensile strength and ligament stiffness was performed using analysis of variance (ANOVA) and covariance components. A p value ≤ 0.05 was considered statistically significant.

Results

Macroscopic assessment

There was no significant difference with respect to ACL length between experimental groups at any time. Joint surfaces showed no differences between operated and control joints during the follow-up period. PET devices were attached to ligament tissue but in all cases were separate enough to be removed without damaging the ligament. Macroscopically, no PET augmentation device exhibited substantial defect; in particular, at their femoral entry point, no damage or degradation was observed.

AP laxity

Both experimental groups demonstrated increased AP laxity compared with controls up to 16 weeks postoperatively. AP laxity also revealed significant differences between groups throughout the entire observation period. In weeks two and six, the augmented group showed higher AP laxity, whereas this was the case for the nonaugmented group at weeks 16, 26 and 52. These differences were significant (Fig. 2).

Tensile strength

Tensile strength in both experimental groups was very low after two and six weeks postoperatively. Primary ACL-repair sutures were all cutting through during testing at week two. Tensile strength increased in both groups at week 16 and remained constant until week 52 but never reached the

Fig. 2 Anteroposterior laxity (millimetres) at 50 N (mean $\pm$ standard deviation) following nonaugmented and polyethylene terephthalate (PET)-augmented primary anterior cruciate ligament (ACL) repair compared with the control group

strength of the intact ACL. Intergroup comparison revealed significantly higher tensile strength for the augmented repair at week 16 to week 52 (Fig. 3).

Regarding rupture, at week two, sutures were cutting through the repaired ACL in all cases in both experimental groups. Rupture mode at the later testing times showed one predominant type in each experimental group. Fifty-two weeks postoperatively, an intra-ligamentous rupture of the repaired ACL was seen predominantly in both experimental groups.

Stiffness

In both experimental groups, ligament stiffness constantly increased during the entire observation period but never reached the level of intact ACLs. Stiffness was significantly lower in the augmented repair group at weeks two and six. At weeks 16–52, the augmented group revealed significantly higher ligament stiffness than the nonaugmented group (Fig. 4).

Discussion

In this study, we aimed to compare the biomechanical performance of nonaugmented and PET-augmented primary ACL repair during a 52-week investigation period. We found that several biomechanical parameters—AP laxity, tensile strength and ligament stiffness—revealed significant differences between the two groups. In the initial period (weeks two and

six), the nonaugmented ACL revealed superior results, whereas from week 16 onwards, the augmented ACL constantly demonstrated better biomechanical performance. In a previous series using the same experimental approach, histological processes during healing were demonstrated [21]. It was obvious that the augmentation device worked as a stress shield for the healing ligament in contrast to the nonaugmented repair, in which the ligament underwent a process of necrotisation and ligamentisation similar to an autograft or allograft ligament reconstruction. Histologically, healing was achieved after 16 weeks in the augmented repair group and after 26 weeks in the nonaugmented repair group. We believe this stress shielding by the augmentation device may also be responsible for the different biomechanical performances, as shown in our series. That healing has taken place is also supported by our findings regarding ligament rupture mode. As expected, sutures were cutting through after two weeks. In contrast, after 52 weeks, a predominantly intraligamentous rupture was observed, indicating that the primary site of repair was healed.

Discussion over the last four decades regarding the optimal surgical treatment of the injured ACL includes numerous strategies and procedures. Early suture repair in the 1970s and early 1980s were abandoned because of poor functional results and recurrent instability. In the same decade, augmented primary ACL repair showed superior results compared with simple suturing. Artificial devices and biological autograft augmentation was performed, and autologous augmentation consistently revealed better results [35]. On the other hand, a biological

Fig. 3 Tensile strength in Newton (mean $\pm$ standard deviation) following nonaugmented and polyethylene terephthalate (PET)-augmented primary anterior cruciate ligament (ACL) repair compared with the control group

Fig. 4 Ligament stiffness in Newton per millimetre (mean $\pm$ standard deviation) following nonaugmented and polyethylene terephthalate (PET)-augmented primary anterior cruciate ligament (ACL) repair compared with the control group

augmentation using the patellar or hamstring tendon could already be considered as a form of autograft reconstruction rather than a primary ACL repair. However, even the so-called gold standard of ACL reconstruction has a relatively high failure rate [36–38]. In addition, posttraumatic arthritis following ACL reconstruction remains an ongoing problem despite subjective and objective joint stability [21, 23]. Only recently has primary suture repair of the torn ACL experienced a kind of renaissance, as tissue engineering techniques combined with suture repair demonstrated promising results in some studies [39, 40]. Vavken et al. reported on the biomechanical outcome of bioenhanced ACL repair [22]. In a large animal model, they found the results of their bioenhanced primary repair to be comparable with a common ACL reconstruction. As bioenhancement, a collagen scaffold and platelet concentrate was used. In comparison, in our series, we used a PET augmentation device, the biocompatibility of which was reported previously [41]. The PET device is inserted in anatomical position and works as a stress shield during ligament healing. Although we aimed to use a biocompatible device that provides mechanical stability during ligament healing, it was not our initial intent to perform a bioenhanced primary ACL repair. It was also obvious that the augmentation device could easily be removed without damaging the ligament. However, some recent studies suggest that PET fibres may play a positive role in proliferation and differentiation of mesenchymal stem cells and may be used in several tissue engineering applications [42, 43].

It should be mentioned that the animals in this study were skeletally mature, whereas in the majority of work on this topic, skeletally immature animals are used: cellular migration and proliferation is increased in young animals and may contribute to better biomechanical outcomes [44, 45]. However, our study suggests that the augmented primary ACL repair also leads to ligament healing and to acceptable biomechanical performance in the adult setting as well. Nevertheless, the role of skeletal maturity and age in the healing capacity of the ACL needs to be explored further and may reveal interesting aspects in terms of eventual clinical feasibility or limitations.

Limitations

Our study has some important drawbacks. Animal models are never able to fully reproduce the clinical situation in humans. We used a standardised ACL injury from sharp transection at the femoral origin of the ligament. Although ACL tears at the femoral origin are common, a sharp transection will never fully simulate an ACL injury, and a frayed rupture may heal differently. Furthermore, it should be noted that despite several mentioned advantages of an augmented repair, biomechanical properties of the repaired ligaments in this study never reached the level of the intact ACLs. Although there

were no significant differences in AP laxity after 16 weeks, tensile strength and ligament stiffness both remained at a significantly lower level compared with the intact status. Similar results were found in the study by Vavken et al., in which a bioenhanced ACL repair demonstrated that tensile strength and linear stiffness remained considerably different compared with that of the intact ligament [22]. Given the fact that overall knee homeostasis should be the ultimate goal, the fact that biomechanical properties of the natural ACL were not fully restored remains a somehow critical point.

Conclusion

In summary, this study demonstrates that augmented primary repair of proximal ACL ruptures in sheep leads to superior biomechanical results compared with nonaugmented primary repair and supports the previously reported histological results. However, biomechanical properties of the natural ACL are not fully restored. In the future, bioenhanced ACL repair and improved tissue engineering approaches may open additional strategies to improve these results.

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Conflict of interest None.

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