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The fixation strength of tibial PCL press-fit reconstructions

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Abstract
Purpose A secure tibial press-fit technique in posterior cruciate ligament reconstructions is an interesting technique because no hardware is necessary. For anterior cruciate ligament (ACL) reconstruction, a few press-fit procedures have been published. Up to the present point, no biomechanical data exist for a tibial press-fit posterior cruciate ligament (PCL) reconstruction. The purpose of this study was to characterize a press-fit procedure for PCL reconstruction that is biomechanically equivalent to an interference screw fixation.
Methods Quadriceps and hamstring tendons of 20 human cadavers (age: 49.2 ± 18.5 years) were used. A press-fit fixation with a knot in the semitendinosus tendon (K) and a quadriceps tendon bone block graft (Q) were compared to an interference screw fixation (I) in 30 porcine femora. In each group, nine constructs were cyclically stretched and then loaded until failure. Maximum load to failure, stiffness, and elongation during failure testing and cyclical loading were investigated.
Results The maximum load to failure was 518 ± 157 N (387–650 N) for the (K) group, 558 ± 119 N (466–650 N) for the (I) group, and 620 ± 102 N (541–699 N) for the (Q) group. The stiffness was 55 ± 27 N/mm (18–89 N/mm) for the (K) group, 117 ± 62 N/mm (69–165 N/mm) for the (I) group, and 65 ± 21 N/mm (49–82 N/mm) for the (Q) group. The stiffness of the (I) group was significantly larger ($P = 0.01$). The elongation during cyclical loading was significantly larger for all groups from the 1st to the 5th cycle compared to the elongation in between the 5th to the 20th cycle ($P < 0.03$).
Conclusion All techniques exhibited larger elongation during initial loading. Load to failure and stiffness was significantly different between the fixations. The Q fixation showed equal biomechanical properties compared to a pure tendon fixation (I) with an interference screw.
All three fixation techniques that were investigated exhibit comparable biomechanical properties. Preconditioning of the constructs is critical. Clinical trials have to investigate the biological effectiveness of these fixation techniques.

Keywords Knee · Posterior cruciate ligament (PCL) · Press fit · Hamstring tendon · Quadriceps tendon

Introduction

The outcome of the posterior cruciate ligament (PCL) reconstruction depends to a large extent on the graft material, the fixation technique, and the placement of the bone tunnels. Nevertheless, due to the complex anatomical

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structures of the PCL with its two bundles [1, 21], the choice of the graft is difficult.

Due to their drawbacks, synthetic grafts should be avoided [13], autografts and allografts remain the gold standard. The bone-patellar tendon-bone graft (BPTB) incorporates in the tunnel via a bone to bone healing after about 6 weeks [30]. The disadvantages of this graft are the donor site morbidity and weakening of the extensor apparatus being the most important agonist of the PCL [26, 39].

The quadriceps tendon can be harvested with a bone block from the proximal patella. The side with the bone block can be embodied in a bony tunnel. The free end of the tendon has to be fixed in the bone tunnel; tendon to bone healing is necessary for this approach. The quadriceps tendon may also be used as a split graft for a single bundle-single tunnel tibial and double bundle-double tunnel femoral fixation [9, 16].

In PCL reconstruction, the grafts have to be longer compared to an ACL reconstruction [6]. Therefore, a double semitendinosus and double gracilis graft is most commonly used.

Most surgeons today use hardware for fixation of the graft in the drill holes [4, 5, 28]. This approach is compromised by artifacts during postoperative magnetic resonance imaging and the necessity of implant removal in case of revision surgery [11, 31]. A hardware-free press-fit fixation is a potential solution for these limitations. The purpose of this study was to characterize two press-fit fixation techniques for tibial PCL reconstruction and compare it to an interference screw fixation technique. The hypothesis is that a PCL press-fit reconstruction technique is biomechanical equivalent to an interference screw fixation technique.

Materials and methods

The knees of 20 human cadavers (20 knees) were used for acquisition of the hamstring tendons and the quadriceps tendon with a bone block. The age of the cadavers from which the tissue was obtained was 49.2 ± 18.5 years (range: 23–75). The harvesting of the tendons was performed an average 1.7 ± 0.7 (range: 1–3) days postmortem. We used tendons from 15 men and 5 women with a mean body size of 175.7 ± 10.3 (range: 154–183) cm for this investigation. There were no visual signs of ligament degeneration or patellar disorders. The quadriceps tendons were harvested with a patellar bone block (30 mm in length; Fig. 1a). The knot of the semitendinosus tendon had a diameter of 10.4 ± 0.5 mm. For the tibial drill holes, we used the femurs of 30 German Landrace pigs. The pigs were 1 year old, fully grown, and had a weight between 100 and 120 lbs. The tibial neck was cut off and the shaft of the tibia cemented into an aluminum holder using cold-curing methylmethacrylate resin (Technovit 4071, Heraeus Kulzer, GmbH, Wehrheim, Germany).

Graft preparation and fixation

There were 9 constructs used in each group. Grafts and bone blocks were kept moist using saline spray during preparation and testing and refrigerated at −20°C before

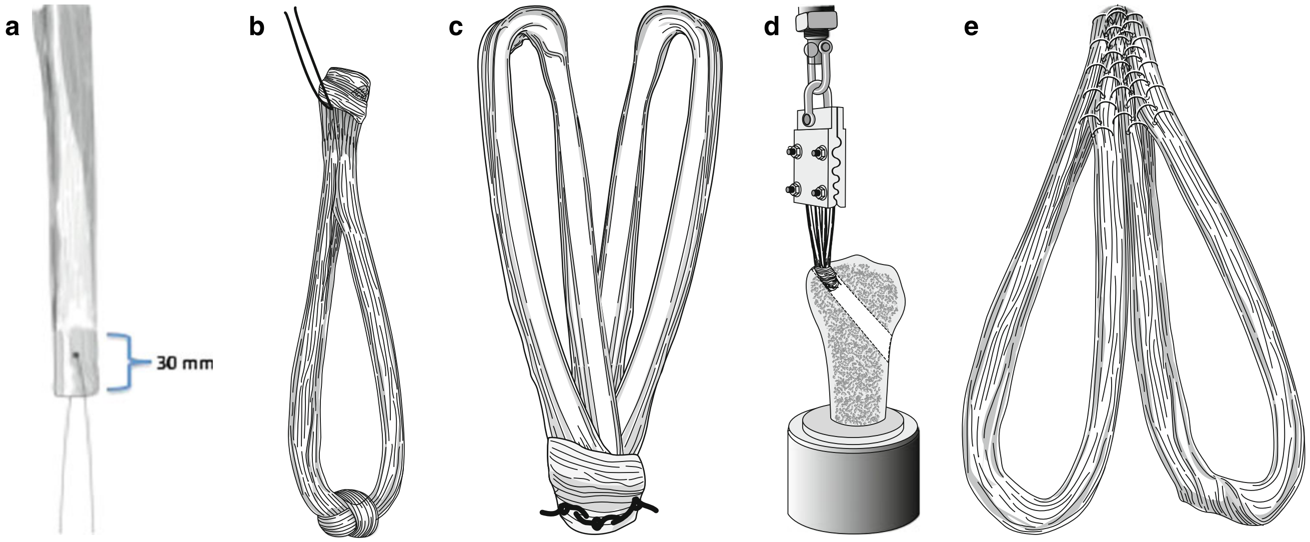

Fig. 1 Fixation techniques that were investigated: The bone blocks of the quadriceps tendon grafts were trimmed to fit into a 9–10-mm drill hole (**a**). The hamstring tendons were used as a knot graft. A knot was interlocked into the *middle* of the tendon (**b**). The free *ends* of the tendon were fixed with the interlocked knot in the *middle* of the tendon (**c**). For all groups, a *bottleneck-shaped tunnel* was created from outside in (**d**). For the I group, the semitendinosus tendons were looped and the ends sewed together with a baseball stitch (**e**)

and after preparation. For the tibial bone tunnel, a drill guide was placed on the native PCL footprint and a guide pin was drilled from outside in starting midway between the tibial tubercle and the posteromedial tibial crest.

Quadriceps tendon graft (Q)

The bone blocks of the quadriceps tendon grafts were trimmed to fit into a 9–10-mm drill hole and to create a cone (Fig. 1a). A central 2-mm drill hole was placed within each bone block, and a non-resorbable suture (Ethibond 2, Ethicon GmbH, Norderstedt, Germany) was advanced through the hole in order to insert the graft inside the tibial tunnel. The soft tissue end of the graft was armed with a non-resorbable suture (Ethibond 2, Ethicon GmbH, Norderstedt, Germany). A bottleneck-shaped drill hole was created. The graft was inserted into the drill hole from outside in Fig. 1d.

Hamstring tendon knot graft (K)

The semitendinosus tendon was trimmed to 32 cm. A knot was interlocked into the middle of the tendon (Fig. 1b). The free ends of the tendon were fixed (Ethibond 2, Ethicon GmbH, Norderstedt, Germany) with the interlocked knot in the middle of the tendon (Fig. 1c). A bottleneck-shaped drill hole was created. The graft was inserted into the drill hole from outside in Fig. 1d.

Interference screw graft (I)

The semitendinosus tendons were looped and the ends sewed together (Ethibond 2, Ethicon GmbH, Norderstedt, Germany) with a baseball stitch (Fig. 1e). The grafts were fixed in the bone tunnels with an interference screw (Milagro™, Mitek GmbH, Norderstedt, Germany). The diameter of the interference screw, the bone tunnel, and the graft were of equal size in this group.

Mechanical testing

The constructs were thawed at 4°C for 24 h prior to mechanical testing and kept moist using saline spray during the entire procedure. A material testing machine (Mini Bionix 858, MTS Systems Co., Minneapolis, USA) was used for the mechanical evaluation of the constructs. The potted tibias were rigidly fixed in a base platform at 0°, setting the bone tunnel–force direction angle to 50°. There was a distance of 30 mm between the grafts and the clamp; the total length of all tendons was trimmed to 50 mm, leaving 20 mm for fixation in a custom-made s-shaped clamp (Fig. 1d).

The constructs were pretensioned with 60 N for 30 s prior to testing. Then, 500 cycles of mechanical loading in between 60 and 260 N were applied at a repetition rate of 1 Hz. The increase in construct length was recorded with a frequency of 20 Hz and a measurement accuracy of 0.1 mm. Length changes are reported between the minimum of the first (15th, 20th) and to the maximum of the 5th (20th, 500th) cycle. After a decreasing preload from 60 to 10 N and pausing for 30 s, a failure test with a ramp speed of 1 mm/s was performed. The maximum failure load, failure mode, and stiffness of the constructs were analyzed.

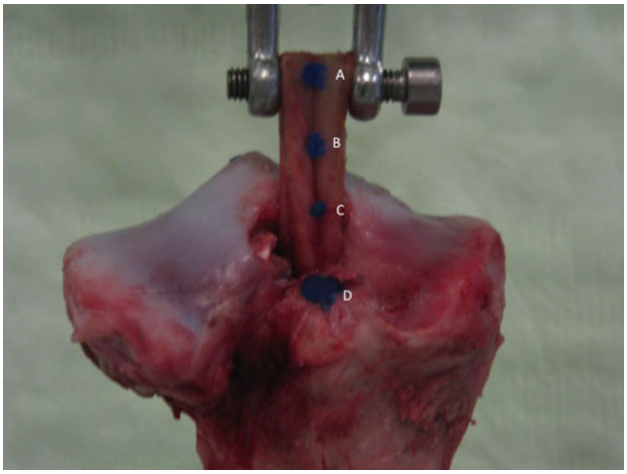

Fig. 2 Video analysis: *Markers A* and *B* were used to investigate length changes within the tendon, *markers C* and *D* to analyze changes between tendon and bone

The tests were recorded with a digital video camera (frame rate: 25 pictures/s). Constructs were photo-optically marked at intervals of 10 mm starting at the ridge of the femoral drill hole. One marker was attached to the bone and three markers within the tendons with a distance of 10 mm in between each marker. Markers A and B were used to investigate length changes within the tendon, and markers C and D to analyze changes between tendon and bone (Fig. 2).

An image analyzing software (ImageJ, NIH) was used to determine length changes. The measurements are reported in percent of the initial length. We analyzed the length change in between tendon-bone and tendon–tendon markers. The procedure was executed from the smallest length observed during the 1st (15th) (20th) loading cycle to the maximum length attained during the 5th (20th) (500th) loading cycle. Moreover, lengthening in between the beginning and end of failure testing was examined.

These data were compared with the length changes that were recorded by the mechanical testing machine.

Statistical analysis

All mean values are reported with standard deviations. The three groups were compared using a one-way ANOVA.

Normality and equal variance tests were conducted. If normality test failed, a Kruskal–Wallis ANOVA on ranks was executed with a post hoc Scheffe test. If normality tests were passed, an equal variance test was conducted. Comparison of two groups was conducted using a non-parametric *t* test. All operations were performed using Sigma Stat 15.0 (SPSS-company, Chicago, IL 60606, USA). A significance level of $P < 0.05$ was assumed.

Results

Surgical data

The maximum diameter of the drill hole was 10.4 ± 0.5 mm in the K group, 8.0 ± 0 mm in the I group, and 10.2 ± 0.4 mm in the Q group. There was one (11%) pull out of a graft during cyclic loading in the K group, none in the I group, and none in the Q group.

The failure mode in the K group was a pull out of the knot in 8/9 (88%) and a rupture of the tendon in 1/9 (11%) of the cases. I group fixations failed as a result of pull out of the semitendinosus tendon in 6/9 (66%) and a rupture of the tendon in 3/10 (30%). In the Q group, a rupture of the tendon at the insertion to the bone block was the mode of failure in 6/9 (66%), a pull out of the bone block in 1/9 (11%) and a fracture of the tibia in 2/9 (22%) of the cases.

These results are mirrored in the data obtained from video analysis of the load to failure experiments. Distance between photo optical markers attached to bone and tendon at the moment of failure increased 35.4 ± 30.5% in the K, 53.5 ± 30.8% in the I, and 42.1 ± 27.9% in the Q group. In contrast, the markers on the tendons were stretched 3.5 ± 3.3% in the K, 1.8 ± 1.0% in the I, and 15.8 ± 29.2% in the Q group. The elongation during failure testing between tendon and bone was significantly larger for all fixations than between markers placed on the tendons ($P = 0.026$).

Biomechanical data

Maximum failure loads observed were 518 ± 157 N (range: 859–350 N) for the K fixation, 558 ± 119 N (range: 788–390 N) for the I group, and 620 ± 102 N (range: 793–455 N) for the Q technique. The loads between these groups were not significantly different (Fig. 3).

The stiffness was 55 ± 27 N/mm (18–89 N/mm) for the K group, 117 ± 62 N/mm (69–165 N/mm) for the I group, and 65 ± 21 N/mm (49–82 N/mm) for the Q group. The stiffness of the I group was significantly larger ($P = 0.01$).

The cyclical loading elongation, determined by optical tendon-bone markers from the smallest length observed during the 1st (15th) (20th) loading cycle to the maximum length attained during the 5th (20th) (500th) loading cycle, was significantly larger for all fixations than between markers placed on the tendons ($P < 0.001$; Fig. 4; Table 1).

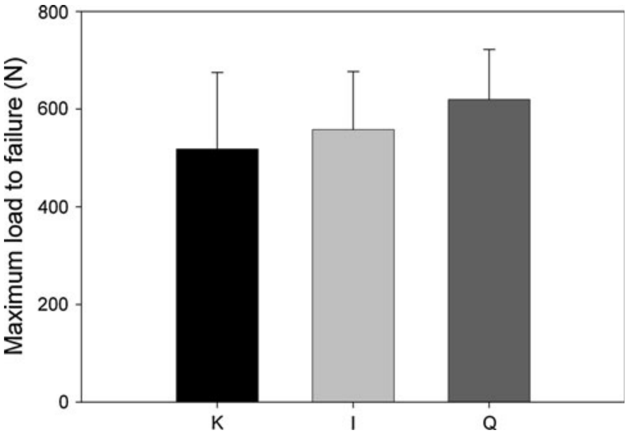

Fig. 3 The maximum failure loads were not significantly different between the groups (n.s)

The cyclical loading elongation determined by the mechanical testing machine in between the 1st and 5th loading cycle showed significantly smaller elongation for the I group compared to the Q group ($P = 0.022$). From the 15th to the 20th loading cycle, the elongation was significantly smaller for the I and the K group compared to the Q group ($P < 0.01$). Length changes from 15th to 500th loading cycle were significantly different for the K group compared to the I group ($P = 0.01$; Table 2).

Discussion

To our knowledge, the current study is the first to examine the properties of two different types of PCL press-fit fixation techniques. The most important findings of this study are that the press-fit fixations showed equal properties compared to the interference screw fixation and other hardware fixations in the literature.

Other authors have focused on tibial inlay or tibial inlay versus transtibial techniques [8, 15, 27, 34, 36] or examined graft types and hardware fixation techniques [3, 8, 14, 23, 25, 37]. Few biomechanical data exist in the context of anterior cruciate ligament reconstruction and press-fit techniques [17, 18, 35], in contrast to that, no biomechanical results are available for the PCL.

Several study limitations should be noted. This study explores the biomechanical properties of tibial PCL press-fit fixations using human tendons and porcine femora. Care was taken to harvest the tendons shortly postmortem (1.7 ± 0.7 days). The mechanical properties of the tendons

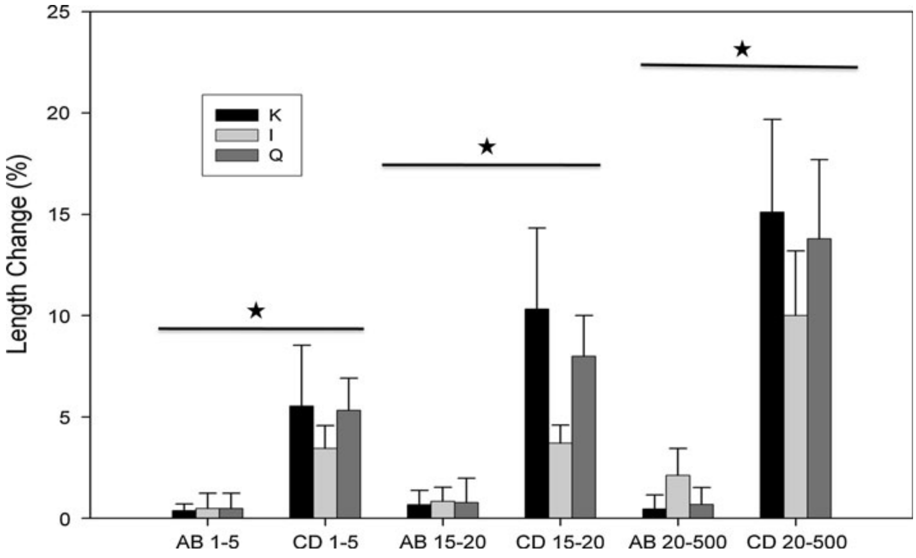

Fig. 4 The cyclical loading elongation, determined by optical tendon-bone markers from the smallest length observed during the 1st (15th) (20th) loading cycle to the maximum length attained during the 5th (20th) (500th) loading cycle, was significantly larger for all fixations than between markers placed on the tendons

Table 1 Optical data computed from video analysis

	Markers A–B; cycles 1–5$^+$	Markers A–B; cycles 15–20$^+$ (%)	Markers A–B; cycles 15–500$^+$ (%)	Markers C–D; cycles 1–5$^+$ (%)	Markers C–D; cycles 15–20$^+$ (%)	Markers C–D; cycles 15–500$^+$ (%)
K	0.8 ± 0.8	0.7 ± 0.7	0.5 ± 0.7	5.5 ± 3.0	10.3 ± 4.0	15.1 ± 4.6
I	0.8 ± 1.0	0.8 ± 0.7	2.1 ± 1.3	3.4 ± 1.1	3.7 ± 0.9	10 ± 3.2
Q	0.5 ± 0.8	0.8 ± 1.2	0.7 ± 0.8	5.3 ± 1.6	8.0 ± 2.0	13.9 ± 4.0

Table 2 Biomechanical data computed from the testing machine

	Cycles 1–5$^+$ (mm)	Cycles 15–20$^+$ (mm)	Cycles 20–500$^+$ (mm)
K	2.7 ± 0.6	1.5 ± 0.2	6.7 ± 4.5
I	2.0 ± 0.9	1.2 ± 0.4	2.5 ± 1.4
Q	3.9 ± 1.8	2.3 ± 1.1	4.6 ± 1.7

of this collective were different from young human tendons; however, most failures were observed as a result of fixation failure rather than from tendon rupture. The video analysis indicated that 21/27 (77%) failures in this study were fixation failures. The first limitation of this study is the fact that one component of our setup was animal tissue. Porcine femora are smaller in size than human femora. Creation of a 10-mm bone tunnel creates a greater stress riser than it does in human bone. This may explain the incidence of bone fractures in the Q group. Although the use of porcine bones has been criticized [20, 32], we chose porcine femora because of their availability and the fact that due to the same age of the donors we have a more uniform bone quality [20]. Porcine bones have been used in many recent studies [17, 20, 24, 28]. The overestimation of failure loads and graft slippage criticized by Nurmi et al. [32] may have influenced interference screw fixations more than press-fit fixations, because grafts are pressed against the spongiosa of the tunnel by an interference screw.

The second limitation is, due to the fact that we have focused on a single tunnel-force angle, we are not able to evaluate the flexion–extension motion. Moreover, this controlled laboratory study just reflects the mechanical properties of tibial PCL fixations without any biological healing or remodeling responses.

Higher strain rates have been used by other authors (30 mm/s) [2, 22]. The strain rate used in this study (1 mm/s) has most commonly used in recent investigations [12, 14, 17, 25, 28, 40]. Various different numbers of cycles, recovery periods, and strain magnitudes have been used in order to determine the biomechanical properties of graft fixations [24, 33]. In our view, several thousand cycles would have biased the measurements obtained for SG, because tendon knots are affected more by dehydration than bone blocks. In vivo, dehydration is less likely to occur. Thus, we decided to limit the number of cycles to 500.

Providing a sufficient primary stability is the basic challenge for a fixation technique to allow adequate postoperative rehabilitation [10]. The maximum failure load of the K group (518 ± 157 N) is similar compared to the I group (558 ± 119 N) and interference screw fixations; Weiler et al. [41] showed 507 ± 93 N for an interference screw fixation; Steiner et al. [38] indicated 821 ± 219 N for a screw/washer fixation and 573 ± 109 N for a suture/post fixation. Lee et al. [23] showed comparable results for the Bio-TransFix device (570 ± 96.9 N). The Q group shows superior fixation strength compared to Brand et al. [5] 293 ± 137 N (interference screw) and similar failure loads compared to a BPTB graft (691 ± 77 N [10]). This study examined the "time 0" properties of three PCL fixation techniques. The 250 N peak cyclic load in this study is indeed lower than the estimated 350 N in vivo force in normal walking [29], but after successful graft incorporation and insertion side healing, the structural properties (maximum load to failure) are expected to increase. Therefore, the focus of this study is on the direct postoperative time where lower graft forces would be expected.

The stiffness of the press-fit techniques K (55 ± 27 N/mm) and Q (65 ± 21 N/mm) archived comparable results to common used fixation techniques: Chen et al. [10] showed 33.9 ± 5.6 N/mm for a BPTB graft and 25.7 ± 6.2 for a four-strand hamstring tendon graft. Brand et al. [5] showed 45 ± 15 N/mm for an interference screw fixation with the quadriceps tendon and 58 ± 14 N/mm for an interference screw fixation with hamstring tendons. The interference screw fixation of this study (117 ± 62 N/mm) shows superior stiffness compared all named studies above.

The sample size of $n = 9$ per group that was tested is of comparable size to similar studies throughout the literature [7, 14, 19, 34, 35]. The variability of the biomechanical data examined in this study might be due to the differences in tissue properties of the human donors.

The present study compared the "time 0" properties of two press-fit fixations and one interference screw fixation. The press-fit fixations showed equal properties compared to the interference screw fixation and other hardware fixations in the literature. Clinical trials are necessary to evaluate the performance and incorporation of a tibial press-fit fixation.

Conclusion

The results of this study suggest that a tibial press-fit fixation does not hamper the biomechanical properties and such techniques can be applied similar to hardware fixation techniques. Hence, the results of this study indicate that a tibial press-fit fixation is an alternative for the tibial PCL reconstruction fixation.

Conflict of interest The authors declare that they do not have any conflict of interest.

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