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Interference screws should be shorter than the hamstring tendon graft in the bone tunnel for best fixation

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Abstract

Purpose Interference screw fixation of hamstring tendon grafts in bone has to overcome the challenges that tendons have a slippery surface and viscoelastically adapt under pressure. As the typical failure mode of the graft is to slip past the interference screw, it was hypothesized that the position and configuration of the graft end may be of influence on the fixation strength.

Methods Different configurations of the graft ending and its effect to primary fixation with interference screws after viscoelastic adaptation were tested in six groups: I: graft and the screw inserted at the same depth, II/III: the graft overlaps the tip of the screw (interference screw of 28 and 19 mm in length, respectively), IV: strengthening of the graft ending with additional suture knots, V: Endopearl, respectively, and VI: effect of partial retraction of the screw after excessive insertion. In vitro tests were performed with fresh calf tendon grafts and interference screws in bone tunnels (fresh porcine distal femur) all of 8 mm in diameter.

Results The relative position of the graft ending to the tip of the interference screw thereby was recognized as a significant factor on pullout forces. Further strengthening at the graft endings with additional suture knots or an Endopearl device could improve primary hold as well.

Conclusions Better fixation strength is achieved if the tip of interference screw does not extend past the end of a tendon graft. Enforcement of the tendon end with sutures or an implant can further improve fixation.

Keywords Hamstring graft · Viscoelastic adaptation · Interference screw · Anterior cruciate ligament

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Introduction

Tendon to bone fixation with interferential screws (IS) is considered as the clinical standard for anterior cruciate ligament reconstruction with soft tissue grafts [1, 6, 14]. Thereby, primary fixation strength was recognized an important goal to allow patients to proceed to early rehabilitation [4, 7, 20]. With insertion of the IS, which is usually of the same diameter as graft and bone tunnel, the tendon graft is viscoelastically deformed, loses up to 30 % of its volume and can enlarge the bone tunnel [18]. The tendon–bone–screw interface has been described as the weakest link in this construction [9, 21, 24], and with pure tendon grafts (e.g. hamstrings), slippage past the IS is the most likely failure mechanism at early stages [3, 15, 28]. To prevent this failure mode, hybrid fixations with additional devices (e.g. Endopearl) were brought to the market. Reinforcements of the graft end with IS fixation, similar to the bone plug (e.g. patellar tendon or quadriceps) was shown to increase pullout force [1, 25].

Prevention from graft slippage and higher pullout strength was hypothesized if the graft ending overlaps the tip of the IS and therefore remains uncompressed. In this context, mechanical behaviour of the graft end may be of underestimated importance in the primary fixation strength.

Furthermore, correct screw placing is desirable but over insertion of the IS may occur. If recognized intra-operatively, secondary adjustment is often necessary to achieve optimal placement of the IS at the tunnel entrance. It was hypothesized that this manoeuver may compromise fixation strength because the effect of an uncompressed graft ending becomes affected. Therefore, we set out to compare mechanically different relative positions of IS and tendon grafts in porcine femoral tunnels, all components (bone tunnel, tendon graft and IS) being of 8 mm in diameter.

Materials and methods

Fresh extensor tendons of 20 cm length were harvested from young bovines, folded around Ethibond sutures size 3 (Ethicon, Somerville, NJ, USA) and quadruple bundled to 8 mm in diameter. At one end, grafts were tubed at 5-mm intervals (Vicryl 0, Ethicon). The suitability of this graft material has been established earlier [12]. Thirty-six grafts were prepared so far and then divided blindly into six groups ($n = 6$). In group IV, an additional suture loop was knotted at the very end of the looped transplant or 8 mm Endopearls (Conmed, Linvatec corp. Largo, FL 33773, USA) were applied both with Ethibond sutures (group V).

Bone preparation

Thirty-six fresh porcine distal femora were randomized equally to six groups. After dissection from all soft tissues, bicortical tunnels were created at the anatomical insertion of the ACL with a conventional drill bit of 8 mm in diameter.

Graft fixation technique

Tendon graft bundles were all positioned into the tunnel in a defined depth creating a specific position of the grafts ending and the tip of the IS. In groups I, IV and V: graft and IS ending at the same level; In groups II, III and VI: graft ending overlaps the tip of the IS of 7 mm. Consequently, 6 different settings were created (Fig. 1I–VI). IS (Megafix, Karl Storz, Tuttlingen, Germany Company) all of 8 mm in diameter and 28 mm in length, except group III (19 mm), were inserted over a Nitinol guide wire (Karl Storz). In group VI, the IS was inserted 4 full turns deeper than the cortical surface (creating a flush position of the graft-/IS ending) and then trimmed back to the right position after 30 s. All specimens were tested after a delay of 60 min to achieve full viscoelastic adaptation. Specimens were kept at room temperature (20 °C) and moistened with NaCl solution 0.9 %.

Mechanical testing

All tests were performed on a materials testing machine (Zwick 1456, Zwick GmbH, Ulm, Germany). Distal

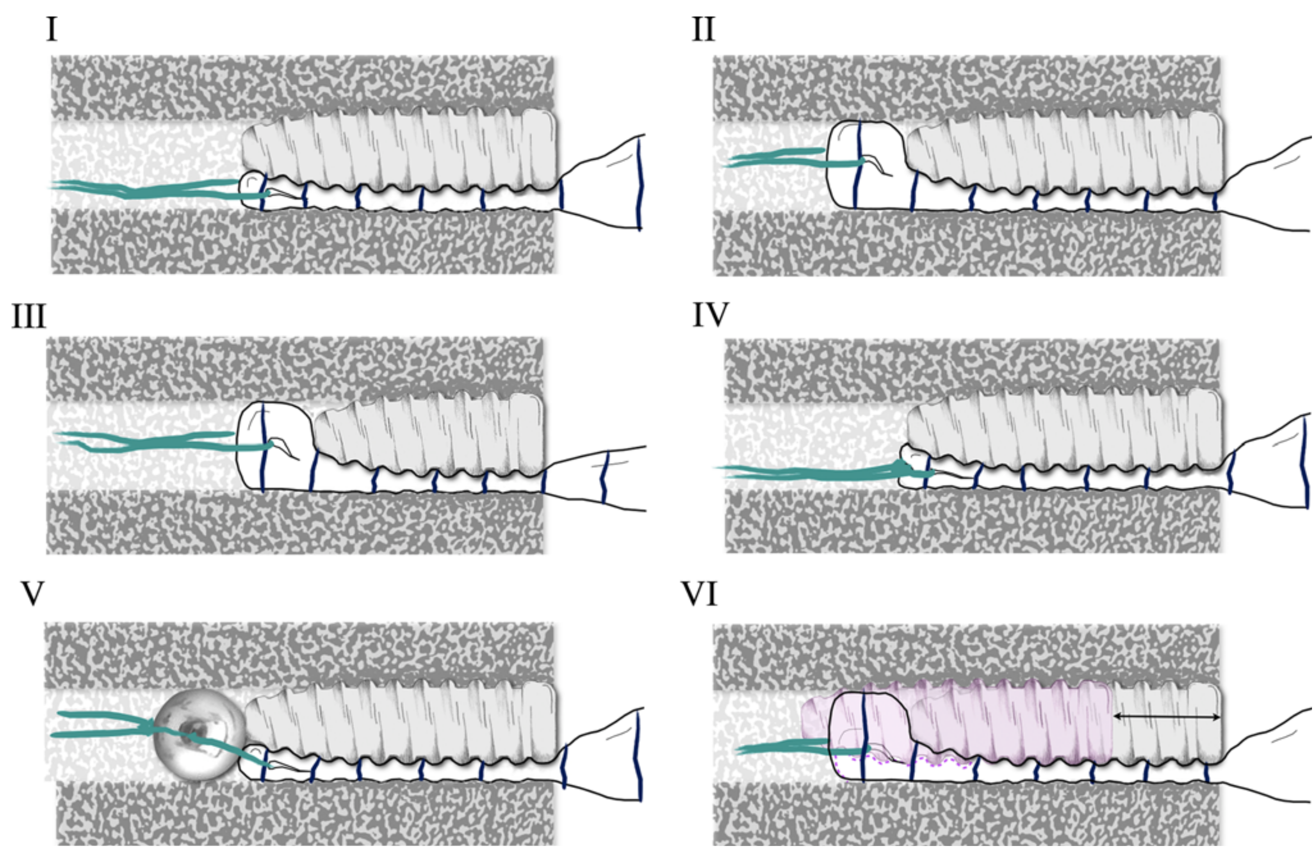


Fig. 1 The schematic concept of the graft fixation within the six groups. **I** The graft ending is at same level with the IS, **II/III** the graft relatively overlaps the IS (28 and 19 mm in length, respectively), **IV** the graft ending is enforced with a knotted suture loop, **V** graft with Endopearl attached, **VI** the IS was trimmed back after insertion 4 turns beneath tunnel aperture

porcine femora and tendon grafts were held on custom made clamps at a variable angle to achieve co-axial loading on the graft in line with the bone tunnel. Preconditioning of the graft was achieved with 5 cyclic loads between 10 and 50 N. For the pullout test, load to failure was executed with a speed of 20 mm/min until the graft was pulled out of the bone. There was registration of the failure mode, maximal load (N) and stiffness up to maximal load (N/mm).

Statistical analysis

Statistical analysis was performed using the software PASW Statistic 18.0 (IBM Corporation, Route 100, Somers, NY 10589, USA). Grouped data were tested for normal distribution using the Kolmogorov–Smirnov test. ANOVA (Bonferroni post hoc correction) was used for normal distributed data to compare intergroup differences. Correlations (stiffness) were assessed with the Pearson correlation test. Values are given in mean. Level of significance was set with a p value <0.05.

Results

In all specimens, load to failure testing resulted with an initial elongation of the tendon graft, followed by slippage at the screw–tendon–bone interface at higher loads, until the graft exited the bone tunnel. In the group with additional Endopearls, graft slippage was preceded by rupture of the Ethibond suture, leaving the Endopearl behind the screw.

Data from pullout testing are summarized in Fig. 2. Stiffness was within the same range in all groups 64 N/mm

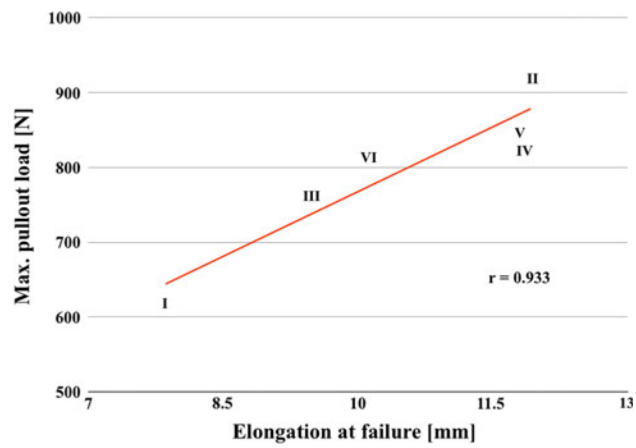
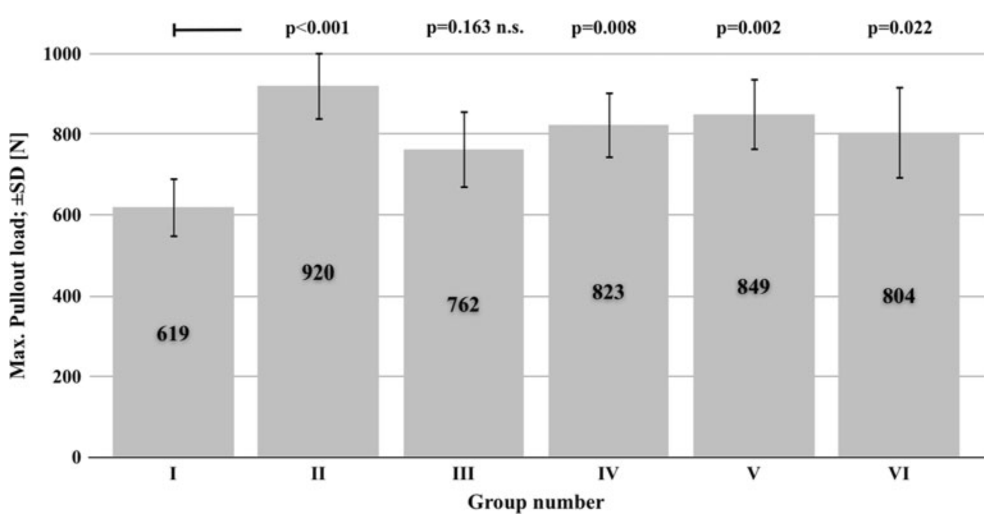


Fig. 3 Stiffness between groups. Mean values/group (load−/graft elongation at failure) are indicated with group numbers I–VI. Linear regression curve was calculated (*red*); $r =$ Pearson coefficient

and elongation at failure correlated strongly ($r = 0.933$) with the corresponding peak load (Fig. 3).

In group I (graft and IS inserted at same level), the lowest failure loads were achieved 619 N. In comparison with this group, significantly enhanced peak loads were measured in group II (the graft ending overlaps the screw tip, IS of 28 mm in length) 920 N; $p < 0.001$, group IV (screw and graft at same level with suture knot) 823 N; $p = 0.008$, group V (graft with additional Endopearl) 849 N; $p = 0.002$ and group VI (screw readjusted after deep insertion) 804 N; $p = 0.022$. Fixation with a shorter IS and overlapping graft ending (group III) increased hold by 19 % compared with group I (ns). With group VI (secondary readjustment of the IS), mean pullout force was reduced by 14 % (ns) in comparison with group II.

Fig. 2 Pullout forces per group. All values are mean ± SD. p values show significance in comparison with group I



Discussion

The most important finding in the present study was the confirmation of the hypothesis that any feature preventing the graft from slippage may improve its fixation with an IS. With the graft overlapping the IS, its ending is protected from viscoelastic adaptation, and improved hold was registered.

Fixation of tendon grafts in a bone tunnel using an IS is a straightforward and effective method but seemingly small details, such as the insertion angle of the IS relative to the bone tunnel [13], graft position relative to the centre of the bone tunnel [10] or the number of graft bundles (single vs. double bundle) [17] may significantly influence the primary stability of the construct. The inherent problem of the method is that tendons, which are designed to glide in the body, are difficult to grasp by compression, particularly because they tend to dissipate hydrostatic pressure through viscoelastic adaptation [18], and consequently, the preferred failure mechanism is slippage of the graft from the tunnel past the IS [2]. A graft end overlapping the inserted IS remains in its physiological state (no viscoelastic adaptation) and therefore counters the graft from slippage.

In the literature, the enhancement of the IS length was shown as an important factor to increase the pullout force for the fixation of pure tendon grafts [23, 26]. Unexpectedly, we found also increased pullout loads with a shorter screw, not overlapping the tendon end, which achieved higher pullout strength than a longer, overlapping screw. Particularly with the use of bioabsorbable screws, it appears favourable to implant as little material as possible, which also leads to favouring shorter screws. Of course, as long as there is no overlap, fixation strength will still increase with a longer implant.

The observed favourable effect could be further augmented if any knob-like characteristics (e.g. Endopearl) at the graft ending were implemented and could show increased failure loads and stiffness [22, 27]. Additional loops around the tendon and suture knots are easy to apply and may give the graft end resistance against deformation during pullout. Compared with the same fixation configuration without knot (group I) significant improvement of failure loads were achieved by 25 % ($p = 0.008$). The Endopearl could show a similar effect (+27 %, $p = 0.002$). However, it must be considered a major limitation of this study that we only used a single loop of Ethibond USP no. 3, which represented the weakest link in this configuration. It is highly likely that additional sutures would achieve higher failure loads. Both features (suture knots and Endopearl) seem to prevent the graft from slippage. The same effect was shown with bone tendon grafts (e.g. patellar or quadriceps tendon) where the effect is achieved from the natural bone plug. Here, different length of IS had no effect on pullout strength [5, 19].

Misplacement (over insertion) of the IS and secondary adjustment by trimming back was investigated in group VI. Is hypothesized a viscoelastic adaptation of the graft end, despite its position overlapping the IS. As hypothesized, this appeared to impair hold by 14 % in comparison with group II, and it is likely that longer remaining of the screw in the deep inserted position (30 s in this study) would show further decreased hold. Therefore, it appears preferable to trim back a screw immediately until the end does not overlap the graft rather than to leave it over inserted. However, further investigations are needed.

A limitation of this study is the use of an animal test surrogate for human bone, and future experiments in human bone could better account for variations in bone quality and anatomy. However, the porcine bone model is well established in this context and was proven for its similar characteristics compared with young human knee bone [8, 11, 16].

Conclusion

Fixation strength of IS may be significantly improved by approximately 30 %, if the screw is shorter than the graft end in the bone for pure tendon grafts (e.g. hamstring). Therefore, it seems of clinical relevance to correlate screw length with the insertion depth of the graft. Further fixation strength is achieved if there is a suture loop tied and knotted around the tendon graft ending.

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