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HANDSURGERY

Flexor tendon repair with a polytetrafluoroethylene (PTFE) suture material

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Abstract

Background There is a consensus that after a flexor tendon repair an aggressive rehabilitation protocol with early active motion can improve functional outcome, provided that the combination of material and suturing technique can meet the higher biomechanic demands. Bearing this in mind we evaluated a polytetrafluoroethylene (PTFE) suture (SERAMON®, Serag-Wiessner) as a possible material for flexor tendon repair.

Materials and methods 40 flexor tendons were harvested from fresh cadaveric upper extremities. 3–0 and 5–0 strands were used both in the polypropylene (PPL) as well as in the PTFE group. In the first phase of the study, we evaluated knotting properties and mechanical characteristics of the suture materials themselves. In the second phase, a 2-strand Kirchmayr–Kessler suture technique was applied for a core suture of a flexor tendon ($n=16$). In the third phase, we performed a tendon repair including an epitendinous running suture with 5–0 PPL or 5–0 PTFE material ($n=22$). One way ANOVA tests were performed.

Results The linear loading strength of single strand knotted PPL 3–0 was 19.87 ± 0.59 N. The linear loading strength of knotted PTFE 3–0 was 32.47 ± 1.67 N. For PPL 3–0 maximum linear strength was achieved with five knots, for PTFE 3–0 with eight knots. When a Kirchmayr–Kessler core-only repair was performed, then in the PPL group the loading strength of the repaired tendon was 30.74 ± 9.77 N. In the PTFE group the loading strength was 23.74 ± 5.6 N ($p=0.10$). However, all repairs in the PTFE group failed due to cheese wiring. When a Kirchmayr–Kessler core and epitendinous repair technique was used, then in the PPL group the loading strength of the repaired tendon was 49.90 ± 16.05 N. In the PTFE group the loading strength was 73.41 ± 19.81 N ($p=0.006$).

Conclusion PTFE demonstrates superior strength properties in comparison to PPL for flexor tendon repairs. However, standard 2 strand techniques have proved inadequate to bear the higher biomechanic demands.

Keywords Flexor tendon repair · Polytetrafluoroethylene (PTFE)

Introduction

Over 50 years have passed since Littler and Kutz reported on their excellent results regarding primary flexor tendon repair [11]. That was definitely a turning point in the way hand surgeons thought about primary flexor tendon repair and the era of Sterling Bunnels "no man's land" concept came to an end.

Few, if any, topics have received as much attention within the hand surgical science as flexor tendon repair [10]. Therefore, it is remarkable how much controversy there is still concerning repair techniques, suture materials and rehabilitation regimens. In fact Tang et al. [16–18] have frequently postulated that paradigms changing is taking place

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throughout the hand surgical landscape. For example, the significance of the epitendinous running suture is under severe reconsideration [4] in an era of stronger multistrand core sutures [18].

As an example, in a survey performed under 16 hand surgeons in Finland, almost none performed the standard Kessler type of core suture [9]. Most of the surgeons in Finland performed a core suture using the Lim-Tsai configuration [7]. Likewise, in Australia the Adelaide locking cruciate configuration, another multistrand technique has gained popularity [19].

Mechanical demands on a flexor tendon repair can be deducted from bibliographic data on in vivo measurements. In 1992 Schuind et al. [15] performed measurements during a decompression procedure of the median nerve. The FDP of the index finger could produce linear forces up to 28.4 N upon unresisted flexion whereas for the FPL tendon there were forces up to 34 N recorded. Edsfeld et al. [5] replicated this line of experiments and determined loading tensions of up to 73.8 N on the FDP tendon. It should be noted here, that peak forces were found with wrist in flexion [5]. From the above data, it can be deducted, that at least four strand techniques can provide for adequate mechanical stability in order for early active motion regimens to be performed safely and effectively. Moreover, in recent year's established techniques had to be reevaluated due to renovations in suture materials [20].

PTFE is not a new suture material, and it is widely being used in cardiovascular surgery [3], orthopedic surgery [2] and plastic surgery [6]. It has been shown to display a significantly higher tensile strength over other materials [8] and being biologically inert it displays good biocompatibility. To our knowledge, it has never been evaluated before in relation to tendon repair.

In this study, we assessed a polytetrafluoroethylene (PTFE) strand as a potential material for flexor tendon repair and compared it to a monofilamentous polypropylene (PPL) strand used in our clinic so far. Our working hypothesis was that PTFE has superior properties and offers higher mechanical stability, even with a standard two-strand modified Kirchmayr technique. We did not perform any comparisons between different suture techniques (i.e., core suture vs core suture plus epitendinous running suture) as this matter has been extensively investigated before [21] and was outside the scope of this our study.

Materials and methods

Phases of the study

The study was performed in three subsequent phases. In every phase the PPL strand was compared to a PTFE strand (SERAMON®, Serag-Wiessner) In the first phase we investigated knotting properties and mechanical characteristics of the suture materials themselves. From every charge of the corresponding suture ($N = 20$) speciments were mounted and measured for linear tensile strength. Then the suture was divided and knotted with an ascending number of opposing throws. The point was verified when the strand was prone to break at the knot rather than slip ($N = 20$). In the second phase, we performed core sutures (two-strand Kirchmayr–Kessler) alone and tested for linear strength of repair ($N = 16$). In the third phase ($N = 22$), we performed a combination of a core suture (two-strand modified Kirchmayr–Kessler) along with an epitendinous running suture and tested for strength of the repair.

Measurements of linear strength

For all measurements of linear strength a universal testing device TIRAtest 28025a, (TIRA GmbH, Schalkau) was used. Testing velocity was set to 300 mm/min. For the measurements, a 100 N modular component was used. The suture material or the reconstructed tendon was fixed on both ends (Fig. 1).

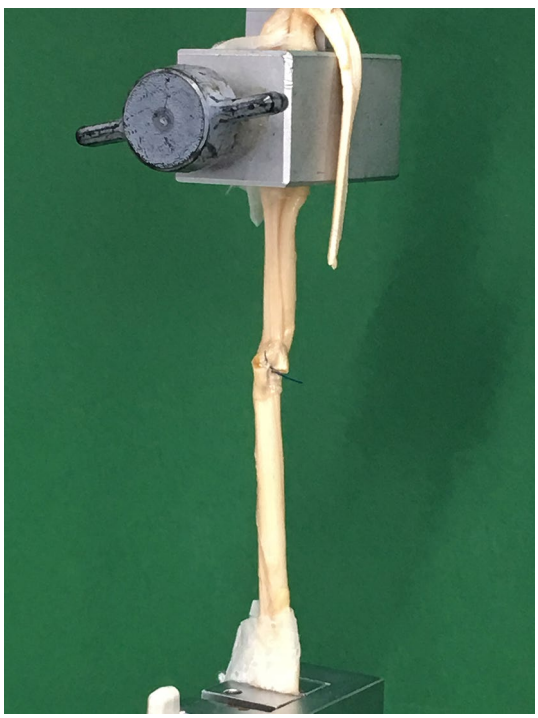

Fig. 1 For all measurements of linear strength a universal testing device TIRAtest 28025a, (TIRA GmbH, Schalkau) was used Full length of the tendinous part of the units was harvested, to provide for better anchoring onto the measuring device

Measurements of elongation

The testing device measured elongation of the material prior to breakage in millimeters (mm). The results were digitaly displayed and documented.

Cadaveric flexor tendons

For the purpose of this study, 40 flexor tendons were harvested from non fixed cadaveric upper extremities. The donor extremities were provided from the institute of anatomy, University of Erlangen. The use of the human material was in full compliance with the university policy for use of cadavers and recognizable body parts. For the study nine flexor tendons of the fingers and the thumb were utilized as well as the flexor carpi radialis tendon. Four upper extremities from four different donors were used, two female and two male ones. The tendons were obtained from geriatric cadaveric donors with an age between 65 and 80 years old. Prior to refrigerating, the cadavers were exsanguinated. No deep freezing was performed prior to harversting of the tendons. Full length of the tendinous part of the units was harvested, to provide for better anchoring onto the measuring device. The tendons were then transected at the middle point by means of a No 11 blade.

Suturing technique

For the core sutures, we used the two-strand technique introduced by Kessler–Kirchmayr with the Zechner modification. The locking suture throws were placed approximately 15 mm from the site of transection. This technique is used in our department for more than 10 years. For the core suture, we use a 3–0 strand, for the epitendinous suture 5–0 strand. For the polypropylene (PPL) sutures five opposing knots were performed. For the PTFE eight knots were performed.

Statistical analysis

One way ANOVA was used for comparison between the groups. All measurements of tensile strength (failure load) are expressed in Newton (N) with mean values and standard deviation (±). Values of elongation are expressed in percent (%) of initial length.

Results

Phase 1

Tensile strength of PTFE 3–0 suture was 34.51 ± 2.03 N, PPL 3–0 strand 27.19 ± 0.37 N. Tensile strength of PTFE 5–0 suture was 10.69 ± 1.09 N PPL 5–0 strand 9.30 ± 0.55 N. Knot breaking load for PTFE 3–0 at knotted strand was 32.47 ± 1.67 N, for PPL 3–0 19.87 ± 0.59 N. Knot breaking load for PTFE 5–0 at knotted strand was 10.50 ± 0.80 N, for PPL 5–0 7.60 ± 0.29 N (Fig. 2). Maximal elongation of knotted strand prior to breakage was for the PTFE 3–0 strand $8.06 \pm 1.10\%$ and for the PPL 3–0 strand $21.03 \pm 0.66\%$. Maximal elongation of knotted strand prior to breakage was for the PTFE 5–0 braid $9.50 \pm 1.15\%$ and for the PPL 5–0 strand $23.83 \pm 2.05\%$. Concluding PTFE 3–0 is significantly stronger than PPL 3–0 in terms of linear loading ($p < 0.01$). PTFE 5–0 is also significantly stronger than PPL 5–0 in terms of linear loading ($p < 0.01$). Knot breaking load in both the 3–0 as well as in the 5–0 comparisons was significantly higher for PTFE than for PPL ($p < 0.01$). Maximal elongation before breakage was significantly longer in PPL in both the 3–0 and the 5–0 comparisons ($p < 0.001$). To avoid knot slippage, we had to perform eight opposing knotting throws with PTFE and five opposing throws with PPL.

A summary of the results is shown in Table 1 and Fig. 3.

Phase 2

Tensile strength of the tendon repair with a 2-strand core 3–0 suture without an epitendinous continuous suture was 23.74 ± 5.6 N for the PTFE braid and 30.74 ± 9.77 N for PPL. This effect was not significant ($p < 0.1$). All of the tendon repairs in the PTFE group failed due to cheese wiring of the suture through the tendon fibers. In the PPL group, six repairs failed due to breakage and two failed due to cheese wiring. A summary of the results is displayed in Fig. 4.

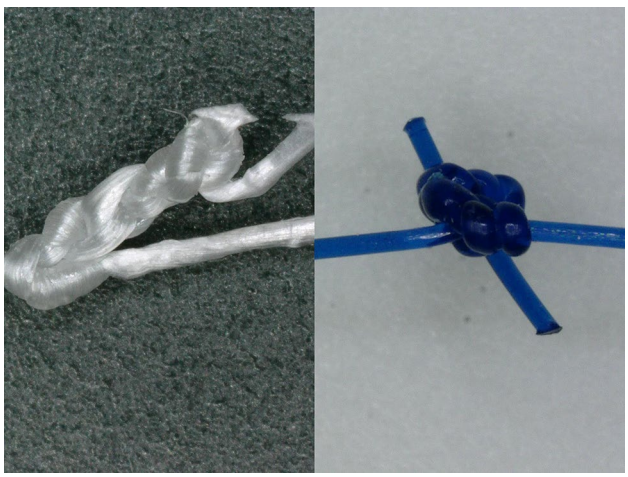

Fig. 2 For the PTFE 3–0 eight throws were necessary to achieve maximal strength of knotted material. For PPL five knots were adequate (50× magnification)

Table 1 A summary of the results is shown in Table 1 and Fig. 1

	Polypropylene (PPL)	Polytetrafuoroethylene (PTFE)	Significance
Tensile strength 3–0 strand	27.19 ± 0.37 N	34.51 ± 2.03 N	$p < 0.01$
Tensile strength 5–0 strand	9.30 ± 0.55 N	10.69 ± 1.09 N	$p < 0.01$
Knot-breaking load 3–0	19.87 ± 0.59 N	32.47 ± 1.67 N	$p < 0.01$
Knot-breaking load 5–0	7.60 ± 0.29 N	10.50 ± 0.80 N	$p < 0.01$
Maximal elongation 3–0 (% of total length)	21.03 ± 0.66%	8.06 ± 1.10%	$p < 0.001$
Maximal elongation 5–0 (% of total length)	23.83 ± 2.05%	9.50 ± 1.15%	$p < 0.001$
No of knots to prevent slippage 3–0	5	8	

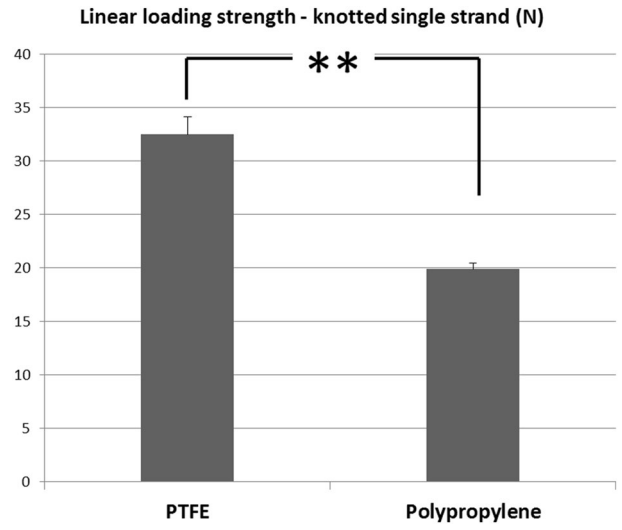

Fig. 3 Polytetrafluoroethylene (PTFE) displayed a higher linear strength than Polypropylene (PPL), when a single strand was knotted. This effect was highly significant (**$p < 0.001$). In the PTFE group eight opposing knots had to be thrown. In the PPL group, five knots were adequate to achieve maximum knotting strength

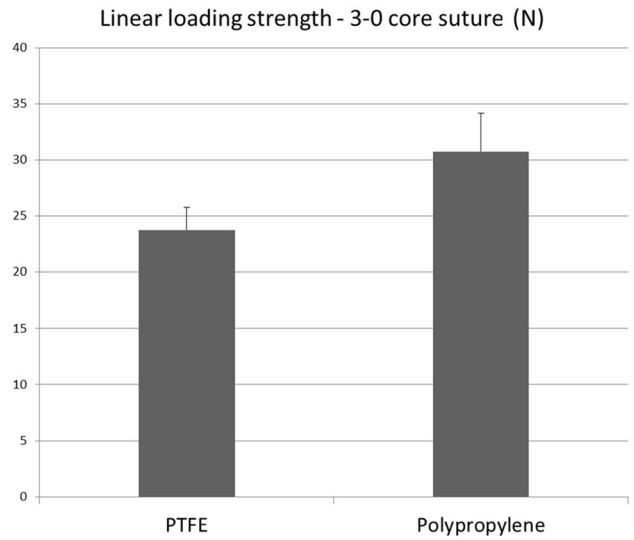

Fig. 4 Results were inconclusive ($p = 0.1$) in the experiment with a single 2-strand-core suture (3–0). All repairs with PTFE failed due to cheese wiring. PPL proved to be less amenable to cheese wiring and, therefore, performed somewhat better

Phase 3

Tensile strength of the tendon repair with a 2-strand core 3–0 suture in conjunction with an epitendinous continuous 5–0 suture was 73.41 ± 19.81 N for the PTFE braid and 49.90 ± 16.05 N for PPL. This effect was highly significant ($p < 0.006$). All of the PTFE repairs failed due to cheese wiring rather than breakage or slippage of the knot. All of the PPL repairs failed due to breakage at the knot or elsewhere (Fig. 5). A summary of the results is displayed in Fig. 6.

Discussion

In this study, we tested a new material which in this setting performed excellently in terms of loading strength. However, with the use of a standard two strand suture technique the "cheese wiring" phenomenon turned out to be a reason for the failure of the repair rather than for the breakage of the suture threads. As a consequence, other techniques of core suture should be considered to meet the higher mechanical demands. Other recent studies with fiberwire material have shown that 4–6 strand interlocking techniques, for example a four-strand interlocking Savage suture, would offer a higher primary tensile strength [14, 17, 18]. It has been proposed that under these circumstances earlier active motion exercises could be possible.

Early active and passive motion are the mainstay of postoperative therapy towards a good functional outcome [13], therefore, is primary stability of paramount importance. Towfigh [12] devised a technique for repair of the flexor tendons essentially without using a suture, though it did not gain wide popularity. Wieskoetter et al. [21] showed that slight modifications lead to significant increase in initial stability. We feel that surgeons add such modifications often in their technique as elements of ongoing experience.

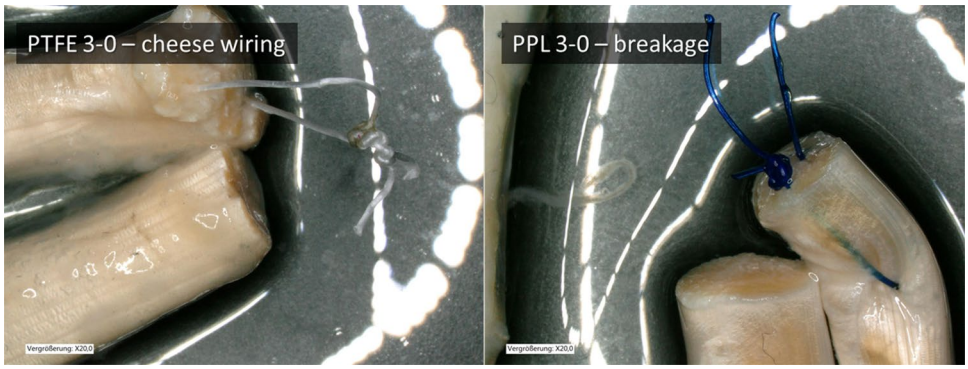

Fig. 5 Failure of the core suture on both polytetrafluoroethylene (PTFE) and polypropylene (PPL). Please note that whereas in PPL the main cause for failure was a breakage of the knot, with PTFE the main effect was cheese wiring

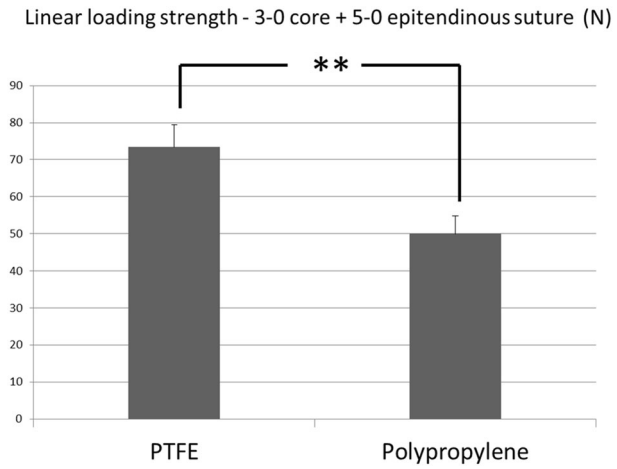

Fig. 6 When a combination of a 2-strand-core suture and an epitendinous running suture was performed, then the PTFE repairs showed a clear superiority in comparison to PPL (**$p < 0.006$)

Although in principal PTFE is a promising technical material in various fields of surgery [1] we encountered some difficulties concerning the handling of the PTFE sutures. Since the material turned out to be extremely slippery it needs thorough and repeated knotting. Depending on the material of the surgical gloves, it might pose a challenge to the surgeon when compared to standard PPL or polyester materials. On the other hand, one can close the knotting throws with any possible traction, without worrying about breakage of the suture. In addition, due to very low friction the surgeon can smoothly approximate the tendon stumps by tightening up the knot after the first or second throw applying as much traction as necessary. From our findings, it is obvious that clinical studies need to be performed to define how the extremely slippery material will perform under repetitive loading stress.

As expected, tensile strength of non knotted threads was superior to those after knotting. We believe that mechanical distortion at the knot produces a breaking point. This findings are in concordance with the literature [1].

The role of epitendinous running suture has been extensively investigated in the past [21]. In some studies, an epitendinous suture could as much as double the initial strength of the tendon repair. However, in recent years, with the evolution of 4–6 strand interlocking techniques in combination with new materials, the epitendinous suture seems to lose significance [4]. In our PTFE groups, with the addition of an epitendinous suture the tendon repair achieved markedly higher tension strength (23.74 N vs 73.41 N). This owes to two factors. First, all repairs in phase 2 failed due to cheese wiring rather than breakage. So practically, we were not able to measure the real breakage point of the tendon repair under 2-stranded Kirchmayr–Kessler repair conditions. Second, the epitendinous suture might have produced an additional interlocking effect. As we commended in the introduction, no intergroup, interphase comparison was performed, because that would be outside the scope of this study.

Our finding, that at least eight throws are needed to achieve adequate knotting strength warrants some comment. A bulky knot might interfere with tendon movements if placed near the A2 pulley. However, the material is extremely pliable in comparison to other monofilament strands. We are currently performing tests with 4–0 strands and loop configurations to overcome this problem. Tang et al. have proposed new guidelines for tethering the A2 pulleys to make some room for the adequate amplitude of flexion [16].

The problem of cheese wiring in the second phase of our experiments intrigued us. We think that the phenomenon was partially due to the very low elongation properties of the material, as we found out in the first phase, as reported by the manufacturer and according to the literature (Table 1) [1]. With PPL the strand can accommodate some of the cutting- through by elongation prior to breakage. Another reason, of course, is the very high tensile strength of the PTFE strand which proves stronger than the tendon and, therefore, at the point of failure it is eventually the tendon or the knotting that fails and not the material.

Conclusion

PTFE is a promising material for use in the flexor tendon surgery. A primary loading strength of a flexor tendon repair of 70 N or more can be achieved, provided an interlocking suture technique is used. Handling and knot-slippage could be a challenge due to extremely low friction properties. Further studies should be performed with 4 or 6 strand techniques before we can incorporate PTFE into our clinical routine.

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Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

Ethical approval This article does not contain any studies with human participants or animals performed by any of the authors.

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