Material and knot properties of braided polyester (Ticron®) and bioabsorbable poly-L/D-lactide (PLDLA) 96/4 sutures

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The purpose of the present study was to investigate *in vitro* the biomechanical material and knot properties and histomorphometrical knot properties of 3–0 braided polyester suture (Ticron®) and bioabsorbable poly-L/D-lactide (PLDLA) 96/4 suture. In Ticron five throws are needed to form a secure knot, and the 1 = 1 = 1 = 1 = 1 and the 2 = 1 = 1 = 1 configurations are recommended. For PLDLA several granny and square knots formed a secure knot, but the 1 = 1 and 1 = 1 = 1 knots were the best. These PLDLA knots had lower yield force and strain at yield point, but higher stiffness than the recommended Ticron knots. The ultimate force values did not differ, but PLDLA knots had significantly higher strain at ultimate point. In the histomorphometrical analysis of the recommended knots, the PLDLA knots had a significantly smaller knot surface area than the Ticron knots. According to these results, PLDLA suture proved to be suitable for flexor tendon repair.

Introduction

Non-absorbable suture materials such as braided polyester, monofilament polypropylene, and monofilament nylon have commonly been used as core suture material in flexor tendon surgery. The modified Kessler repair is widely used with postoperative passive mobilization protocols. However, the tendency has been towards postoperative early active motion programs [1–5] which has created a need for stronger repair techniques. Hence, different multi-strand repairs have been introduced. However, multiple suture strands and knots add foreign-body to the tendon repair site. This is evident especially with coated braided polyester sutures, which, depending on the commercial product, have been reported to require from four to five throws per knot to prevent slippage [6, 7].

The advantage of bioabsorbable suture materials is their gradual absorption from the tissue [8]. Bioabsorbable suture should retain its tensile strength long enough to maintain the approximation of the tendon ends during the critical healing period. The present available bioabsorbable materials lose their tensile strength too early. In subcutaneous implantation in the rabbit the half-life tensile strength for both polyglycolic acid (Dexon®) and braided polyglactin 910 (Vicryl®) has been reported to be two weeks [9], for monofilament polyglyconate (Maxon®) three weeks [9, 10], and for monofilament polydioxanone (PDS®) from four [11] to six [9] weeks. In canine flexor tendon repair with 4-strand Kessler suture and active mobilization both the gap strength and the breaking strength were significantly weaker in monofilament polydioxanone repairs compared to braided polyester repairs from two weeks onwards [12]. Hence, not even the six week half-life tensile strength of the suture is long enough for the healing process to provide sufficient tendon strength.

Polylactide (PLA) is a bioabsorbable polymer which has the L and D isomer. The degradation rate of L/D copolymers depends on the proportion of the isomers in the polymer structure [13]. Of different copolymers

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the poly-L/D-lactide (PLDLA) 96/4 may be a suitable candidate for flexor tendon repair. *In vitro* the half-life tensile strength of PLDLA 96/4 suture fell between ten and 13 weeks, and in the rabbit subcutaneous implantation PLDLA sutures retained average 75% of the initial tensile strength at six weeks [10].

In previous biomechanical studies on coated braided polyester sutures the interest has focused mainly on the ultimate point while the biomechanical behaviour of the suture materials prior to the failure has not been investigated [6, 7]. The purpose of the present study was to investigate *in vitro* the biomechanical material and knot properties and histomorphometrical knot properties of braided polyester suture (Ticron®) and PLDLA 96/4 suture in order to evaluate their suitability for multi-strand flexor tendon repairs.

Materials and methods

Materials

The raw material used was a copolymer of L/D lactic acid (PLDLA) with an L/D monomer ratio of 96/4 and intrinsic viscosity of 4.98 dL/g (PURAC Biochem B.V., Holland). The multifilament polylactide fibres for twisting were melt-spun using an extruder (Gimac, Castronno, Italy) with a die temperature of 272 °C and oriented at elevated temperatures in a three-step process to the final draw ratio of 4.25. The final mean diameter of the filaments was 0.09 mm. The suture was made by twisting six filaments that were later folded in the middle and twisted again to form a 12-filament twine. The sutures were washed in ethanol, dried in vacuum for 16 h, and packed individually. The sutures were sterilized by gamma irradiation with a minimum dose of 2.5 Mrad. The diameter of the PLDLA suture was measured mean 0.5 mm. The 3-0 Ticron® sutures (Davies & Geck, Danbury, CT, USA) were delivered for hospital use. Both suture materials were tested as received from their individual sterile suture packages.

Biomechanical testing

Two hundred and fifty specimens were tested, ten samples in each group.

The specimen was placed between the clamps of the tensile testing machine (LR Series Material Testing Machine LR 30 K, Lloyd Instruments Limited, Hampshire, UK). The initial distance between the clamps was 35 mm corresponding to the calculated length of each suture strand including one loop at both ends in the modified Kessler suture. The strand was adjusted stretched out with a preload of 0.1 N. The specimen was distracted at a constant speed of 70 or 20 mm/min in the material testing and of 20 mm/min in the knot testing. Load-deformation data were collected with a computerised data acquisition system (R Control for Windows, Lloyd Instruments ltd., Hampshire, UK), and a load-deformation curve (Fig. 1) was produced for each specimen. The load-deformation curve typically consists of an initial non-linear toe region, a linear slope and the failure region. *The first linear point* represents the minimum point, and the *yield point* represents the maximum point of the linear slope of the curve. After this point, the slope of the curve is reduced, but the load often continues to increase to the maximum point of the curve, *the ultimate point*. The middle third of the linear slope was defined to analyse *the stiffness* of the specimen. An offset line was defined along the linear slope of the curve. *The first linear point* and *the yield point* were defined at the points of divergence of the offset line from the load-deformation curve. *The first linear force* (F_{FL}), *yield force* (F_Y), and *ultimate force* (F_U) were recorded. The repair site *strain at the first linear point* (S_{FL}), *at the yield point* (S_Y), and *at the ultimate point* (S_U) were defined as repair site deformation (the change in the distance between the tendon clamps) divided by the initial distance between the tendon clamps.

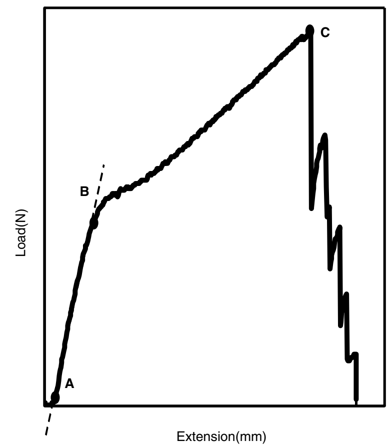

Figure 1 The load-deformation curve starts as the non-linear toe region which ends at the first linear point (A). The toe region is followed by a linear region which ends at the yield point (B). An offset line was defined along the linear slope of the curve and the first linear point and yield point were defined as the points of divergence of the offset line from the load-deformation curve. The slope of the curve reduces after the yield point and continues as the failure region. The strength increases till the ultimate point (C).

Material testing

Material testing according to the United States Pharmacopoeia (USP) [14] includes *the tensile strength* of the unknotted suture, *the knot tensile strength* of the standardized simple knot suture, and *the elongation at tensile strength* of the unknotted suture. The *tensile strength* and *knot tensile strength* are equivalent to the ultimate force (see above in 'Biomechanical testing') of the specimens. The *elongation at tensile strength* is defined as the percentage extension of the unknotted suture to the initial gauge length and is equivalent to the strain at the ultimate force (see above in 'Biomechanical testing'). To perform the knot for material testing according to the USP, a simple knot was tied by placing one end of the strand over the other and through the loop so formed and by pulling the knot tight. The suture was distracted at a constant speed of 70 mm/min defined according to the USP to equal two times the gauge length per minute.

In addition to the three USP variables, also other biomechanical variables were defined (see above in 'Biomechanical testing').

We have previously evaluated tendon repairs with different suture configurations [15] using a distraction rate of 20 mm/min, also favoured by several other investigators [16–22]. To be able to compare the testing results of the present material study to those of previous tendon repair studies and to the present knot testing results, the material testing was also performed at a distraction rate 20 mm/min.

Knot testing

The knot configurations were chosen on the basis of previous studies [6, 7, 23] and our pilot study. The knots (Tables III and IV) are presented according to the system described by Tera and Åberg [23], in which the number of wraps in each throw is indicated by an Arabic number and the relationship between each throw being either parallel (square knot indicated by =) or crossed (granny knot indicated by x).

To tie the knot, the suture material was tied around a plastic tube with an outer diameter of 6 cm. Each knot was tied by the same surgeon. The suture loops formed around the plastic tube were divided on the opposite side of the knot. These cut ends were placed between the clamps of the tensile testing machine (LR Series Material Testing Machine LR 30 K, Lloyd Instruments Limited, Hampshire, UK) with the knot midway between the clamps. The clamps were distracted at a static rate of 20 mm/min until the suture broke at the knot level or the knot failed totally by slippage.

The load-deformation curve was analysed as described previously. In addition, the *knot holding capacity (KHC)* and *strain at KHC* (S_{KHC}) were defined. KHC is the force level at which the knotted strand fails either by breakage or slippage. KHC and the corresponding *strain at KHC* (S_{KHC}) were defined by visual analysis of the specimen during testing and by analysing the load-deformation curve. The possible slippage was observed by visual analysis and the precise onset of the slippage was defined at the point where the initially smooth load-deformation curve became irregular [6]. A knot was considered as secure when the biomechanical properties did not further improve significantly from increasing the number of throws.

Histomorphometrical analysis

On the basis of the biomechanical analysis, the Ticron knots $1 = 1 = 1 = 1 = 1$ and $2 = 1 = 1 = 1$ and the PLDLA knots $1 = 1$ and $1 = 1 = 1$ were chosen for the histomorphometrical analysis. Ten samples of each knot configuration were analysed. To tie the knots, the suture material was tied around a wooden rod with an outer diameter of 7 mm. Each knot was tied by the same surgeon. The free ends of the suture were cut 2 mm from the knot. The loop formed around the rod was gently pulled off the tube, and the knot was mounted under a microscope. The knots were measured in regard to the knot surface area. A Leitz microscope was linked via a videocamera (Color View II, Soft, Soft Imagin System GmbH, Münster, Germany) to a computer (Dell Precision 340, Ireland). AnalySIS docu 3.2 (Soft-Imaging Software GmbH, Münster, Germany) was used for the image analysis. The magnification used was 25× at the screen. The error of the histomorphometric method was measured by the coefficient of variation, 1,2%.

Statistical analysis

The biomechanical material properties of Ticron and PLDLA suture were evaluated in regard to the influence of knotting and distraction speed. The unknotted and simple knot strands distracted at 70 mm/min (USP) or 20 mm/min were compared to analyse the influence of knotting on the materials. The results from the different distraction rates were compared to evaluate the influence of distraction rate on the biomechanical properties of the unknotted and simple knot strands. Statistical analysis was performed using one-way ANOVA, and, depending on the equality of variances either Bonferroni or Tamhane post-hoc multiple comparisons test was used.

We hypothesized that five square knots for Ticron and two square knots for PLDLA suture are sufficient to form a secure knot. Statistical comparison was performed in regard to these knot configurations. The data were analysed using the one-way between groups ANOVA with planned comparisons.

The biomechanical and histomorphometrical properties of the selected secure knots for Ticron and PLDLA were compared. Statistical analysis was performed using one-way ANOVA and Bonferroni multiple comparisons test.

The results are presented as mean value and 95% confidence interval. P values less than 0.05 were considered significant. All statistical analyses were performed with SPSS 11.0 for Windows (SPSS Inc., Chicago, IL, USA).

Results

Material properties

Ticron

The results of the tensile testing of Ticron suture material are presented in Table I.

When comparing unknotted and simple knot strands distracted at 70 mm/min there was a significant difference in the strain at the first linear point (S_{FL})

TABLE I Ticron suture. Material properties are presented as mean (95% confidence interval) first linear force (N), strain at first linear point (mm/mm), yield force (N), strain at yield point (mm/mm), ultimate force (N), strain at ultimate point (mm/mm), and stiffness (N/mm) values of biomechanical testing at distraction rates of 70 and 20 mm/min.

Ticron	S_{FL}	S_Y	S_U	F_{FL}	F_Y	F_U	Stif
70 mm/min							
unknotted	0.00	0.17	0.20	0.4	25.0	26.0	4.5
	(0.00–0.00)	(0.16–0.18)	(0.20–0.21)	(0.3–0.5)	(24.7–25.3)	(25.7–26.2)	(4.4–4.7)
simple knotted	0.01	0.15	0.15	0.2	18.6	18.6	3.8
	(0.01–0.01)	(0.14–0.15)	(0.14–0.15)	(0.2–0.3)	(18.0–19.2)	(18.0–19.2)	(3.6–4.0)
20 mm/min	0.00	0.18	0.27	0.3	24.8	28.2	4.2
unknotted	(0.00–0.00)	(0.17–0.18)	(0.26–0.27)	(0.2–0.3)	(24.6–25.1)	(27.9–28.4)	(4.1–4.4)
simple knotted	0.01	0.15	0.15	0.3	19.0	19.0	3.7
	(0.0–0.01)	(0.15–0.16)	(0.15–0.16)	(0.3–0.4)	(18.4–19.5)	(18.4–19.5)	(3.5–3.9)

S_{FL} = strain at first linear point, S_Y = strain at yield point, S_U = strain at ultimate point, F_{FL} = first linear force, F_Y = yield force, F_U = ultimate force, Stif = stiffness.

TABLE II PLDLA suture. Material properties are presented as mean (95% confidence interval) strain at first linear point (mm/mm), strain at yield point (mm/mm), strain at ultimate point (mm/mm), first linear force (N), yield force (N), ultimate force (N), and stiffness (N/mm) values of biomechanical testing at distraction rates of 70 mm/min and 20 mm/min

PLDLA	S_{FL}	S_Y	S_U	F_{FL}	F_Y	F_U	Stif
70 mm/min	0.00	0.04	0.44	0.3	11.1	28.6	8.9
unknotted	(0.00–0.00)	(0.04–0.04)	(0.41–0.46)	(0.2–0.4)	(10.6–11.6)	(27.2–29.9)	(8.5–9.4)
simple knotted	0.01	0.09	0.30	0.3	10.9	17.8	4.5
	(0.01–0.01)	(0.08–0.09)	(0.27–0.33)	(0.3–0.4)	(10.7–11.0)	(16.7–18.9)	(4.2–4.7)
20 mm/min	0.00	0.04	0.39	0.2	10.4	26.4	8.9
unkotted	(0.00–0.01)	(0.04–0.04)	(0.37–0.41)	(0.2–0.2)	(10.1–10.7)	(25.4–27.4)	(8.4–9.5)
simple knotted	0.00	0.07	0.28	0.3	10.0	16.7	4.7
	(0.00–0.01)	(0.07–0.07)	(0.25–0.30)	(0.2–0.4)	(9.6–10.5)	(15.7–17.6)	(4.4–5.0)

S_{FL} = strain at first linear point, S_Y = strain at yield point, S_U = strain at ultimate point, F_{FL} = first linear force, F_Y = yield force, F_U = ultimate force, Stif = stiffness.

($p < 0.05$), yield force (F_Y), strain at the yield point (S_Y), ultimate force (F_U), strain at the ultimate point (S_U), and stiffness (Stif) ($p < 0.001$). With a distraction rate of 20 mm/min the unknotted and simple knot strands differed significantly at S_{FL} ($p < 0.01$), F_Y, S_Y, F_U, S_U , and Stif, ($p < 0.001$).

When comparing the unknotted strands distracted at 20 mm/min to those of 70 mm/min, there was a significant difference in the F_U and S_U values ($p < 0.001$). No significant differences existed between the simple knot strands distracted at 20 and at 70 mm/min.

PLDLA

The results of the tensile testing of PLDLA suture material are presented in Table II.

When comparing unknotted and simple knot strands distracted at 70 mm/min there was a significant difference between the S_{FL}, S_Y, F_U, S_U, and Stif values ($p < 0.001$). When distracted at 20 mm/min the unknotted and simple knot strands differed significantly at S_Y, Stif, F_U, and S_U ($p < 0.001$).

When comparing the unknotted strands distracted at 20 mm/min to those of 70 mm/min, there was a significant difference between the F_Y, F_U ($p < 0.05$), and S_U ($p < 0.01$) values. Comparing the simple knot strands distracted at 20 mm/min to those of 70 mm/min, there was a significant difference between the S_{FL} ($p < 0.001$), F_Y ($p < 0.01$), and S_Y ($p < 0.001$) values.

Knot properties

Ticron

The results of the biomechanical testing of the Ticron knots are summarized in Table III. The $1 = 1 = 1 = 1 = 1$ knot has been statistically compared to the other Ticron knot configurations. Significant differences are presented here.

The F_{FL} of the $1 = 1 = 1 = 1 = 1$ knot differed significantly from that of the $2 = 1 = 1$, 2×2 ($p < 0.001$) and $2 = 2$ knots ($p < 0.01$). The F_Y, F_U, and KHC of the $1 = 1 = 1 = 1 = 1$ knot differed significantly from all the four throw configurations ($p < 0.001$).

The S_{FL} of the $1 = 1 = 1 = 1 = 1$ knot differed significantly from the knots 2×2 ($p < 0.001$), $1 = 1 = 1 = 1 = 1 = 1$ and $2 = 2$ ($p < 0.01$), and $2 = 1 = 1 = 1$ ($p < 0.05$). The S_Y of the $1 = 1 = 1 = 1 = 1$ knot differed significantly from the knots

TABLE III Ticron knots. Results of biomechanical testing of different knots at a distraction rate of 20 mm/min are presented as mean (95% confidence interval) strain at first linear point (mm/mm), strain at yield point (mm/mm), strain at ultimate point (mm/mm), strain at knot holding capacity (mm/mm), first linear force (N), yield force (N), ultimate force (N), knot holding capacity (N), and stiffness (N/mm)

Tricon knots	S_{FL}	S_Y	S_U	S_{KHC}	F_{FL}	F_Y	F_U	KHC	Stiff
Four throw									
1 = 1 = 1 = 1	0.01	0.18	**0.36**	0.19	0.3	**11.2**	**13.9**	**11.8**	**1.9**
	(0.01–0.01)	(0.13–0.23)	**(0.23–0.59)**	(0.14–0.24)	(0.2–0.4)	**(9.0–13.4)**	**(12.6–15.2)**	**(9.3–14.3)**	**(1.6–2.1)**
1 = 2 = 1	0.01	**0.10**	**0.38**	**0.10**	0.3	**8.2**	**14.3**	**8.2**	2.6
	(0.01–0.01)	**(0.08–0.12)**	**(0.32–0.43)**	**(0.08–0.12)**	(0.2–0.4)	**(6.7–9.6)**	**(13.4–15.2)**	**(6.7–9.6)**	(2.4–2.8)
2 = 1 = 1	0.01	**0.12**	0.32	**0.10**	**0.2**	**7.1**	**10.3**	**7.2**	**1.8**
	(0.01–0.01)	**(0.10–0.15)**	(0.19–0.45)	**(0.06–0.14)**	**(0.1–0.2)**	**(5.5–8.8)**	**(8.3–12.3)**	**(5.6–8.8)**	**(1.6–2.1)**
2 = 2	**0.05**	**0.13**	**0.35**	**0.13**	**0.9**	**4.5**	**10.1**	**4.5**	**1.2**
	(0.03–0.08)	**(0.10–0.16)**	**(0.28–0.43)**	**(0.10–0.16)**	**(0.5–1.2)**	**(3.4–5.5)**	**(7.4–12.8)**	**(3.4–5.5)**	**(0.9–1.5)**
2 × 2	**0.00**	**0.13**	0.23	**0.11**	**0.2**	**7.6**	**9.8**	**6.7**	**1.5**
	(0.00–0.00)	**(0.09–0.18)**	(0.18–0.28)	**(0.08–0.14)**	**(0.1–0.2)**	**(4.6–10.6)**	**(7.5–12.0)**	**(4.1–9.2)**	**(1.2–1.9)**
Five throw									
1 = 1 = 1 = 1 = 1	0.01	0.22	0.23	0.22	0.3	17.8	18.1	17.8	2.5
	(0.01–0.01)	(0.21–0.24)	(0.22–0.24)	(0.21–0.24)	(0.3–0.4)	(16.7–19.0)	(17.4–18.8)	(16.7–19.0)	(2.0–3.0)
1 = 2 = 1 = 1	0.01	**0.25**	**0.25**	**0.25**	0.4	17.4	17.4	17.4	2.4
	(0.01–0.01)	**(0.23–0.27)**	**(0.23–0.27)**	**(0.23–0.27)**	(0.3–0.4)	(16.8–17.9)	(16.8–17.9)	(16.8–17.9)	(2.2–2.5)
2 = 1 = 1 = 1	**0.00**	0.23	0.24	0.24	0.3	16.5	16.7	16.5	**2.1**
	(0.00–0.01)	(0.21–0.26)	(0.22–0.26)	(0.22–0.26)	(0.2–0.3)	(15.1–17.9)	(15.4–18.0)	(15.1–17.9)	**(1.8–2.3)**
Six throw									
1 = 1 = 1 = 1 = 1 = 1	**0.00**	0.21	**0.21**	0.21	0.3	17.2	17.4	17.4	2.2
	(0.00–0.01)	(0.20–0.21)	**(0.20–0.22)**	(0.20–0.22)	(0.3–0.4)	(16.7–17.7)	(17.0–17.7)	(17.0–17.7)	(2.1–2.3)

S_{FL} = strain at first linear point, S_Y = strain at yield point, S_U = strain at ultimate point, S_{KHC} = strain at KHC, F_{FL} = first linear force, F_Y = yield force, F_U = ultimate force, KHC = knot holding capacity, Stif = stiffness. The results of the 1 = 1 = 1 = 1 = 1 knot to which other groups are compared are framed and the values differing significantly are presented with bold numbers.

1 = 2 = 1, 2 = 1 = 1, and 2 = 2 ($p < 0.001$), 2 × 2 ($p < 0.01$), and 1 = 2 = 1 = 1 ($p < 0.05$). The S_U of the 1 = 1 = 1 = 1 = 1 knot differed significantly from the knots 1 = 2 = 1 ($p < 0.001$), 1 = 1 = 1 = 1 = 1 = 1 and 2 = 2 ($p < 0.01$), and 1 = 1 = 1 = 1 and 1 = 2 = 1 = 1 ($p < 0.05$). The S_{KHC} of the 1 = 1 = 1 = 1 = 1 knot differed significantly from the knot 1 = 2 = 1 = 1 ($p < 0.05$) and all four throw knot configurations ($p < 0.001$) apart from the knot 1 = 1 = 1 = 1.

The stiffness of the 1 = 1 = 1 = 1 = 1 knot differed significantly from the knots 2 × 2, 2 = 2, 2 = 1 = 1, and 1 = 1 = 1 = 1 ($p < 0.001$), and 2 = 1 = 1 = 1 ($p < 0.05$).

PLDLA

The results of the biomechanical testing of the PLDLA knots are summarized in Table IV. The 1 = 1 knot has been statistically compared to the other PLDLA knot configurations. Significant differences are presented here.

The F_{FL} of the 1 = 1 knot differed significantly form the knots 2 × 1 and 1 × 1 ($p < 0.05$). The F_Y of the 1 = 1 knot was significantly higher than that in the knots 1 × 1, 2 × 1, and 1 × 2 ($p < 0.001$), 1 = 2 ($p < 0.01$), and 2 = 1 ($p < 0.05$). The F_U and KHC values of the 1 = 1 knot were significantly higher than in the 1 × 1 and 2 × 1 ($p < 0.001$) and 1 = 2 knots ($p < 0.05$).

The S_{FL} of the 1 = 1 knot differed significantly from all other knot configurations (for the knot 2 × 1 $p < 0.01$, for all others $p < 0.05$). The S_Y of the 1 = 1 knot was significantly higher than that in the knots 2 × 1 ($p < 0.001$), 1 = 1 = 1 and 1 × 1 ($p < 0.01$), 1 = 2 and 1 × 2 ($p < 0.05$). The S_U of the 1 = 1 knot differed significantly from that of the 1 × 1 knot ($p < 0.01$). The S_{KHC} of the 1 = 1 knot differed significantly from that of the knots 1 × 1 and 2 × 1 ($p < 0.001$).

The stiffness of the knot 1 = 1 was significantly lower than that in the knot 1 = 1 = 1 ($p < 0.05$) and higher than that in the knots 1 × 1, 2 × 1 ($p < 0.001$), and 2 = 1 ($p < 0.05$).

Ticron vs. PLDLA

The biomechanical properties of the PLDLA knots 1 = 1 and 1 = 1 = 1 and Ticron knots 2 = 1 = 1 = 1 and 1 = 1 = 1 = 1 = 1 were compared. There were no significant differences in the F_{FL} or S_{FL} values compared. The F_Y and S_Y were significantly higher in the Ticron knots compared to the PLDLA knots ($p < 0.001$). There were no significant differences in the F_U or KHC values of the Ticron and PLDLA knots. The S_U and S_{KHC} of the PLDLA knots were significantly higher than those in the Ticron knots ($p < 0.01$). The stiffness

TABLE IV PLDLA knots. Results of biomechanical testing of different knots at a distraction rate of 20 mm/min are presented as mean (95% confidence interval) strain at first linear point (mm/mm), strain at yield point (mm/mm), strain at ultimate point (mm/mm), strain at knot holding capacity (mm/mm), first linear force (N), yield force (N), ultimate force (N), knot holding capacity (N), and stiffness (N/mm)

PLDLA knots	S_{FL}	S_Y	S_U	S_{KHC}	F_{FL}	F_Y	F_U	KHC	Strif
Two throw									
1 = 1	0.02 (0.01–0.04)	0.11 (0.08–0.14)	0.33 (0.30–0.36)	0.32 (0.29–0.35)	0.5 (0.2–0.9)	9.8 (9.1–10.5)	16.7 (15.5–18.0)	16.7 (15.4–18.0)	4.2 (3.2–5.2)
1 × 1	**0.01 (0.00–0.01)**	**0.05 (0.03–0.07)**	**0.17 (0.09–0.25)**	**0.06 (0.04–0.09)**	**0.2 (0.1–0.2)**	**2.0 (1.5–2.5)**	**4.6 (3.7–5.4)**	**2.7 (1.6–3.9)**	**1.5 (0.9–2.1)**
Three throw									
1 = 1 = 1	**0.01 (0.00–0.01)**	**0.06 (0.06–0.07)**	0.30 (0.27–0.33)	0.30 (0.27–0.33)	0.3 (0.2–0.3)	9.1 (8.2–9.6)	16.8 (15.2–18.3)	16.8 (15.2–18.3)	**5.3 (4.8–5.7)**
1 = 2	**0.01 (0.00–0.02)**	**0.08 (0.07–0.09)**	0.31 (0.28–0.33)	0.31 (0.28–0.33)	0.2 (0.2–0.3)	**8.4 (7.7–9.0)**	**14.9 (13.9–15.9)**	**14.9 (13.9–15.9)**	3.9 (3.5–4.3)
2 = 1	**0.01 (0.00–0.08)**	0.09 (0.07–0.11)	0.34 (0.30–0.38)	0.34 (0.30–0.38)	0.3 (0.2–0.3)	**8.7 (8.1–9.3)**	15.4 (14.3–16.6)	15.4 (14.3–16.6)	**3.3 (2.8–3.9)**
1 × 1 × 1	**0.01 (0.00–0.01)**	0.08 (0.07–0.09)	0.32 (0.27–0.34)	0.30 (0.27–0.34)	0.2 (0.2–0.2)	9.1 (8.8–9.4)	15.4 (14.5–16.4)	15.4 (14.5–16.4)	3.6 (3.1–4.1)
1 × 2	**0.01 (0.00–0.01)**	**0.07 (0.06–0.09)**	0.32 (0.28–0.36)	0.32 (0.28–0.36)	0.3 (0.2–0.3)	**7.8 (6.3–9.2)**	15.7 (15.0–16.5)	15.7 (15.0–16.5)	3.6 (3.2–3.9)
2 × 1	**0.00 (0.00–0.01)**	**0.03 (0.02–0.04)**	0.24 (0.12–0.35)	**0.04 (0.03–0.06)**	**0.2 (0.1-0.2)**	**2.0 (1.3–2.7)**	**7.3 (3.3–11.3)**	**2.6 (1.5–3.7)**	**1.5 (0.6–2.4)**

S_{FL} = strain at first linear point, S_Y = strain at yield point, S_U = strain at ultimate point, S_{KHC} = strain at KHC, F_{FL} = first linear force, F_Y =yield force, F_U = ultimate force, KHC = knot holding capacity, Stif = stiffness. The results of the 1 = 1 knot to which other groups are compared are framed and the values, differing significantly are presented with bold numbers.

of both PLDLA knots was significantly higher than that of the Ticron knots (PLDLA 1 = 1 = 1 vs. Ticron 2 = 1 = 1 = 1 and 1 = 1 = 1 = 1 = 1 $p < 0.001$; PLDLA 1 = 1 vs. Ticron 2 = 1 = 1 = 1 $p < 0.01$ and 1 = 1 = 1 = 1 = 1 $p < 0.05$).

Histomorphometrical analysis

The surface areas of the Ticron knots (1 = 1 = 1 = 1 = 1 and 2 = 1 = 1 = 1) did not differ significantly from each other; neither did the two PLDLA knots (1 = 1 and 1 = 1 = 1) differ from each other. The surface area of both PLDLA knots was significantly smaller than that of the Ticron knots ($p < 0.001$) (Table V).

TABLE V Histomorphometrical analysis of the knots presented as mean (95% confidence interval) area (mm^2).

Knot	Area
Ticron	
2 = 1 = 1 = 1	3.5 (3.4–3.6)
1 = 1 = 1 = 1 = 1	3.7 (3.5–4.0)
PLDLA	
1 = 1	2.7 (2.5–3.0)
1 = 1 = 1	3.0 (2.8–3.1)

Discussion

As Rodeheaver and colleagues [8] stated, the calibre and type of suture material are too often left to guess or habit, and the configurations of knots to be used are often simply matters of a hand-me-down custom, with no reference to established mechanical testing. In flexor tendon surgery the aim has been towards postoperative early active rehabilitation which, in turn, sets additional requirements on the tendon repair. As the tendon repair strength depends on the properties of the suture materials, knots, and repair configuration, each component should be individually carefully evaluated for an optimal result.

In the present study, the biomechanical testing of the materials and knots was performed by using a single-strand method which has previously been commonly used [6–10]. Alternatively, a loop model has been employed [23–25]. However, it has been reported that the knot holding capacity (KHC) values of the loop method do not correspond to the values obtained by the single-strand method [6]. This may be due to friction between the suture material and the rod around which the loops are placed [6]. As seen in the present study, the stiffness, strength, and elongation properties of unknotted and knotted strands differ which may lead to unequal loading of the strands in the loop method.

In previous material testing studies [6, 8, 9, 23, 24, 26] and studies on knot properties [6, 8, 9, 23, 24], the testing conditions varied greatly from a distraction rate of 10 mm/min with a suture length of 150 mm to a distraction rate of 60 mm/min with a suture length of 30 mm. In biomechanical studies on tendon repair techniques the distraction rate has commonly been 20 mm/min [16–22]. This lower distraction rate makes it easier to observe the gradual failure of the repair composite. In order to compare the results of the present study to those of previous examinations on tendon repair techniques, we performed the material testing not only according to the USP, but also at a distraction rate of 20 mm/min and the knot testing at a distraction speed of 20 mm/min. As we noted significant differences in the biomechanical properties of the sutures due to different distraction rates, we consider that standardised distraction rates should be used when evaluating the different components of the tendon repair.

Coated braided polyesters have been commonly used as core suture in tendon surgery because of their good biomechanical properties such as high tensile strength and low extensibility. The disadvantage of poor knot security is probably due to the low friction coefficient of the coating of the polyester suture [6, 8, 24]. Previous studies have recommended from four to five square throws to achieve a secure knot [6, 7]. Also the 2×2 granny knot has been reported to reach good strength [6, 23]. In the present study with Ticron, all four throw knots slipped with increasing load. The reason for possible successful clinical use of Ticron with four throw knots may be that the forces subjected to the repair composite during passive mobilization, about 9 N [27], remain low enough not to cause significant slippage. In the present study, Ticron suture needed at least five throws to form a secure knot. However, also with all five throw knots care must be taken while knotting, because sliding knots are easily formed leading to slippage. The $1 = 1 = 1 = 1 = 1$ knot reached the best biomechanical values, but the force values did not differ significantly from the other five throw knots or of the six throw knot. Although the stiffness of the $1 = 1 = 1 = 1 = 1$ knot was significantly higher ($p < 0.05$) compared to that of the $2 = 1 = 1 = 1$ knot, the latter has an advantage when tying as the first double throw holds the tendon ends in close apposition. The $1 = 2 = 1 = 1$ knot is not recommended as it has higher strain values compared to the $1 = 1 = 1 = 1 = 1$ knot and its technical performance does not offer any advantages compared to the other five throw knots. Adding the sixth throw to the $1 = 1 = 1 = 1 = 1$ knot is unnecessary as the only achievement was the lower strain at the ultimate point.

The present results with PLDLA suture show that several analysed square knots and granny knots formed a secure knot and only the 1×1 and 2×1 configurations slipped totally. The $1 = 1$ and $1 = 1 = 1$ knots were the best. However, as the $1 = 1 = 1$ knot had higher stiffness and lower strain at yield point, it is recommended.

The biomechanical properties of the standardized simple knot represent an ideal knot and can be used as a reference when evaluating different knot configurations. The knot can be considered as "secure" when the biomechanical properties are not further improved from increasing the number of throws. In an ideal situation, the load-deformation curve of the secure knot equals that of the USP knot. In the recommended Ticron knots especially strain and stiffness values did not reach those of the simple knot. This is probably due to the tightening of the multiple throws during testing, as no slippage was seen to occur. However, the two best PLDLA knots reached the biomechanical properties of the simple knot indicating the good knot holding capacity.

When comparing the ultimate force values of Ticron and PLDLA sutures, neither the unknotted strands nor secure knots differed significantly. Previously, the relative knot security (RKS, i.e. the percentage of KHC of the secure knot/tensile strength of the unknotted strands) for Ticron has been reported as 53% [6]. In the present study the RKS values of the secure Ticron and PLDLA knots were of the same magnitude, 59–63% and 63%, respectively. However, KHC and RKS take into account only the strength of the knot. As extensibility in a suture material has been considered a disadvantage due to possible association with gap formation [7, 28], it is of importance to evaluate the elongation properties of surgical suture materials. In the present study the strain at the ultimate point of unknotted 3–0 Ticron suture corresponds to the results reported by Holmlund *et al.* [6]. At the ultimate point the elongation values of unknotted and knotted Ticron ($2 = 1 = 1 = 1$ knot) were about 10 mm and 9 mm, respectively, and in unknotted and knotted PLDLA ($1 = 1 = 1$ knot) 14 mm and 11 mm, respectively (Fig. 2). As gap formation must

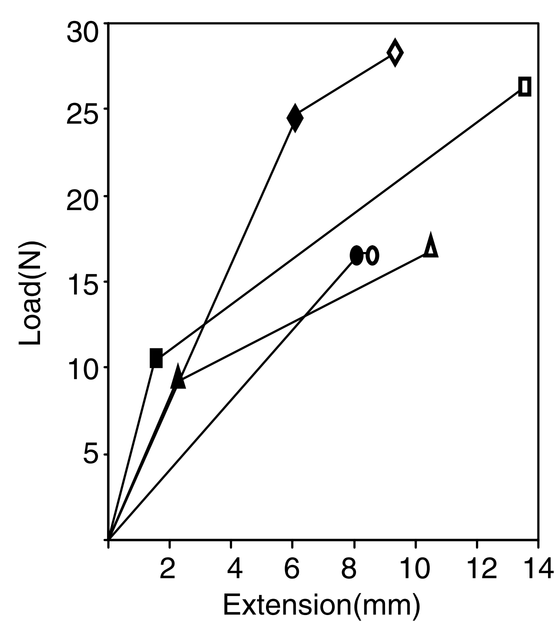

Figure 2 Schematic mean load-deformation curves of unknotted Ticron (♦) and PLDLA (■) suture strands and Ticron knot $2 = 1 = 1 = 1$ (●) and PLDLA knot $1 = 1 = 1$ (▲) distracted at 20 mm/min. Solid symbol represents the yield point and open symbol the ultimate point.

be avoided in tendon repair, in theory these elongation values are too high to maintain repair integrity, and the strands cannot be loaded till the ultimate force. However, in practise when 3–0 suture is used, the failure of the repair complex initiates before the ultimate force of the core suture is achieved. In our previous study comparing different tendon repair techniques the failure of the tendon repairs always initiated by disruption of the peripheral suture at the yield point [15]. To avoid gap formation during postoperative rehabilitation, the tendon repair should not be loaded beyond its yield force. The elongation of the tendon repairs at the yield point varied from 3 mm in the 2-strand modified Kessler to 5 mm in the 4-strand modified Kessler [15]. This total elongation includes the elongation of the suture material, tightening and sliding of the suture grips in tendon tissue, and the elongation of the tendon tissue itself. According to Lotz *et al*. [16] the peripheral suture carries approximately 64–77% of the load in the modified Kessler suture at the yield point. In our previous study the yield force of the modified Kessler suture with running peripheral suture was 25.5 N [15], and hence, estimated according to the results of Lotz *et al*. [16] the load carried by the two core suture strands together was 6–8 N. It has been noted that increasing the number of core suture strands increases the yield force of the repair composite [15, 29]. In our previous study, the yield force of 4-strand modified Kessler and Savage repairs with 3-0 Ticron was approximately 48 N [15]. As the ultimate strength of the peripheral suture was estimated as 18 N, the four core suture strands together must have carried a load of approximately 30 N at the yield point. In the locking configuration the load carried by the core sutures is not equally divided between each strand due to different stiffness and elongation properties of the unknotted and knotted strands. The estimated load per core suture strand correlates to extension values of approximately 1–2 mm (Fig. 2). In the present study, the load-deformation curves of Ticron and PLDLA suture were different. In Ticron suture the curve increased linearly close to the ultimate point, while in PLDLA suture the stiffness of the linear region was higher than in Ticron but the lower yield point was followed by a reduced, but still increasing, curve till the ultimate point. Thus, higher initial stiffness of PLDLA may decrease the proportion of the load carried by the weaker peripheral suture leading to increased yield force of the repair composite [16]. On the basis of the present results, the biomechanical properties of PLDLA suture are even better compared to those of Ticron during the critical initial period of loading.

The aim to increase tendon repair strength by using multi-strand techniques has increased the amount of suture material in tendon repair. Although the diameter of the experimental PLDLA suture is thicker than that of 3-0 Ticron, the smallest secure knots were significantly smaller. Though it is not known to what extent suture material can be added between the tendon ends without interfering with the healing process [30], we consider it an advantage that the bioabsorbable PLDLA knots are smaller as the suture can be securely tied with fewer throws than Ticron. This is emphasized in multi-strand techniques, as the amount of knot material is further increased in relation to the surface area of the cut tendon end.

In conclusion, the biomechanical properties of the bioabsorbable PLDLA suture were equal or even better compared to Ticron suture, which has commonly been used in flexor tendon repair. Furthermore, in PLDLA suture a secure knot with smaller surface area was achieved with fewer throws than in Ticron.

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