Published in final edited form as:

J Hand Surg Eur Vol. 2012 May ; 37(4): 316–322. doi:10.1177/1753193411422793.

The effect of core suture flexor tendon repair techniques on gliding resistance during static cycle motion and load to failure: a human cadaver study

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Abstract

The purpose of this study was to describe a modification of the Massachusetts General Hospital (MMGH) tendon repair and to compare it with three other suture techniques. Twenty human flexor digitorum profundus (FDP) tendons were randomly assigned to the modified Pennington (MP) suture and the MMGH suture. These were compared to the modified Kessler (MK) and Massachusetts General Hospital (MGH) sutures, using data from a previous study. All tendons were repaired with a similar epitendinous stitch and core sutures of 4-0 FiberWire. There was no significant difference in the normalized gliding resistance within the two-strand or four-strand core repair groups. The MP suture had significantly higher 2 mm gap force and ultimate load to failure than the MK suture. The MMGH suture had significantly higher 2 mm gap force and maximum failure ultimate load than the MGH suture. All repairs failed by knot unravelling.

Keywords

Flexor tendon; repair; core suture; gliding resistance

Introduction

The effect of suture materials and suture technique on the load to failure and gliding resistance of tendon core sutures in cadaver models has been well described (Moriya et al., 2010a; 2010b; Silva et al., 2009; Tanaka et al., 2004). Briefly, these have shown that a larger diameter suture is stronger (Taras et al., 2001); that locking loops are stronger (Xie and Tang, 2005; Xie et al., 2005); that the suture purchase has an effect (Cao et al., 2006; Tan and Tang, 2004), and that the running, finishing stitch improves gap resistance (Lin et al., 1988). These have also shown that low profile repairs, such as the modified Kessler repair, have less gliding resistance than high profile repairs such as the Massachusetts General or modified Lee repairs (Zhao et al., 2001), and that monofilament suture material, such as nylon, have less gliding resistance than braided suture material, such as Mersilene (Ethicon, Somerville, NJ, USA) (Momose et al., 2000).

FiberWire® (Arthrex, Naples, FL, USA), a new generation of polyester suture with a long chain polyethylene core, has been found to have superior strength in materials testing and other studies (Komatsu et al., 2006; Lawrence and Davis, 2005; Miller et al., 2007).

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Conflict of interests

None declared.

Previous research has demonstrated that 3-0 FiberWire has a lower frictional coefficient (0.054) than 3-0 braided nylon enclosed in a smooth nylon outer shell (Supramid, S. Jackson, Alexandria, VA, USA) (0.130) or 3-0 coated, braided polyester (Ethibond, Ethicon, Somerville, NJ, USA) (0.076) (Silva et al., 2009). FiberWire has also been reported to have lower gliding resistance and equal strength, compared with 3-0 Ethibond, when using the MGH core suture (Moriya et al., 2010b). Other studies have also shown that repairs loosen after cyclic loading (Thurman et al., 1998), and that FiberWire sutures may be particularly susceptible to knot unravelling (Waitayawinyu et al., 2008). We wished to see if a modification of the MGH repair (MMGH), with the knots inside the repair, could maintain its superior strength characteristics, while reducing its gliding resistance. A previous study had reported that the modified Kessler repair, with the knot inside, had lower gliding resistance than the Kessler repair, with the knot outside (Zhao et al., 2001), and we hypothesized a similar result for the MGH and MMGH repairs. Since the modified Pennington suture has been reported to have equal gliding resistance and higher strength than a modified Kessler suture of 4-0 Supramid (Tanaka et al., 2004), we wished to compare our new design with these repairs as well.

Previous studies measured the gliding resistance at the first cycle immediately after repair and the load to failure was also measured immediately after repair or after one cycle, not after 1000 cycles. We thought that the gliding resistance during 1000 cycles of tendon motion and the load to failure after 1000 cycles would be clinically useful information.

The purpose of this study was to describe a modification of the Massachusetts General Hospital (MMGH) flexor tendon repair and to compare its strength and gliding resistance with three other core suture techniques.

Material and methods

Specimen preparation

After Institutional Review Board (IRB) approval, three right hands and four left hands of seven fresh frozen human cadavers were obtained, including two men and five women, with an average age of 78 years (range 59–91). A total of 20 flexor digitorum profundus (FDP) tendons from these hands were used for this study, including two index, three middle, three ring and two small fingers for each of the two repairs studied, i.e. the modified Pennington (MP) and modified Massachusetts General Hospital (MMGH). We excluded any digits which did not have full passive motion, or evidence of previous surgery in the digit or hand. These data were then compared with data on modified Kessler (MK) and MGH repairs, reported in a previous study (Moriya et al., 2010b), again using two index, three middle, three ring and two small fingers for each repair, from cadaver hands of similar age and gender distribution, and identical digit distribution.

Both in the current study and in the previous one, each finger was disarticulated at the MCP joint level, preserving the flexor tendons at the level of wrist. The FDP tendons were accessed through a transverse incision in the flexor sheath, just distal to the A2 pulley. The FDP tendons were marked at the distal end of the A2 pulley both at full passive extension and at full flexion, with a 4.9 N load (i.e. a 500 g weight) attached to the proximal FDP tendon to maintain tendon tension. The distance between these two marks represented the FDP tendon excursion during full finger motion.

After measurement of gliding resistance in the intact tendons (see method below), a transverse laceration was made 12 mm distal to the proximal FDP tendon mark. This level of laceration was chosen in order to ensure movement of the repair site through the pulley

system during testing. The laceration level was distal to the A2 pulley with the finger extended and thus it was possible to perform the repair without incision of the pulley.

A total of 40 FDP tendons were evenly divided into the following four groups based on the different repair techniques: MK, MP, MGH and MMGH sutures (Figure 1). Twenty of the tendons (the MK and MGH sutures) were from a previous study (Moriya et al., 2010b). All the core sutures used 4-0 braided poly-blend suture (FiberWire Arthrex, Inc. Naples, FL, USA). A simple continuous epitendinous suture was performed using a 6-0 monofilament polypropylene (Prolene, Ethicon, Somerville, NJ, USA) (Kleinert et al., 1973). The simple running epitendinous throws were approximately 2 mm from the lacerated tendon ends, with approximately 1 mm between each throw. The number of epitendinous throws ranged from 12 to 16, depending upon the tendon circumference and the distance between the throws. All core and epitendinous suture knots were tied with four square knots. The tension of all the core and epitendinous sutures was set to coapt without causing bunching of the repair site. All repairs were done by the same surgeon (T.M.) to improve consistency of technique. The A2 pulleys and FDS tendons were preserved during the experiment.

Measurement of gliding resistance

The gliding resistance between tendon and pulley was measured using the method of Uchiyama et al. (1995). Briefly, the proximal and middle phalanges, A2 pulley, flexor digitorum superficialis (FDS), parietal membrane of A2 pulley and visceral membrane of FDP were preserved, while removing the remaining tendon sheath, pulleys and the distal phalanx. A Kirschner wire was used to stabilize the proximal interphalangeal joint in an extended 0° position. Each specimen was mounted on the custom testing device with the palmar side upward. To maintain tension in the FDS tendon, a 2 N load (100 g weight) was attached to its proximal end. The magnitude of this load was determined based on previous in vivo tendon research, to replicate the magnitude of load experienced by normal tendons in situ (Lieber et al., 1996; Schuind et al., 1992). The measurement system consisted of one mechanical actuator with a linear potentiometer, two tensile load transducers and a mechanical pulley. The distal transducer was connected to a 4.9 N load to simulate passive mobilization of the finger (Lieber et al., 1996; Schuind et al., 1992). The proximal load transducer was connected to the mechanical actuator.

Based on the experience of previous studies, a 50° arc of contact, 30° and 20° between the horizontal plane and the proximal and distal transducer cables, respectively, is sufficient to provide accurate measurement of gliding resistance (Uchiyama et al., 1997; Zhao et al., 2001). For each cycle of testing, the tendon was pulled proximally by the actuator against the 4.9 N load at a rate of 2 mm/second. Excursion was limited to the distance between the two FDP tendon markers, thus reproducing the tendon excursion from full finger flexion to full finger extension. The forces at the proximal and distal tendon ends and the tendon excursion were recorded. The specimens were kept moist throughout testing by immersion in a saline bath. The force differential between the proximal and distal tendon ends represents the gliding resistance. The gliding force was calculated for both flexion and extension motion. The flexion and extension means were averaged and reported as the mean gliding resistance. Data were normalized by dividing the after-repair gliding resistance by the pre-repair gliding resistance.

The gliding resistance for the intact FDP tendon was initially recorded for pre-repair. The tendon was then lacerated and repaired. After repair of the tendon, the gliding resistance was recorded after one cycle and then after every fifty cycles up to 500 cycles, and then after every 100 cycles up to 1000 cycles. These data were then normalized by dividing the gliding resistance of each repaired tendon at the various cycles by the gliding resistance of that tendon in the intact, normal state.

Tendon repair load to failure

After testing gliding resistance, the load to failure was measured in the repaired tendon using a servo-hydraulic testing machine (MTS, Minneapolis, MN, USA). The repaired FDP tendon was secured in the testing machine and distracted linearly to failure at a rate of 20 mm/min. A differential variable reluctance transducer (DVRT, Microstrain, Williston, VT, USA) was attached to the tendon spanning the repair site to measure gap formation during testing. Tensile force, grip-to-grip displacement and gap displacement measured by the DVRT transducer were collected at a rate of 20 Hz. Throughout testing the tendons were kept moist by spraying with physiological saline. Maximum breaking force and the force to produce a 2 mm gap were recorded. Additionally, a regression line was fit to the linear region of the force versus gap formation (as measured by the DVRT) to measure the resistance to gap formation. The resistance to gap formation, a measure of repair stiffness, in units of N/mm was obtained from the force versus gap relationship. Finally, the mode of failure for each test was recorded.

Gap measurement

During the 1000 cycles of tendon motion, we monitored the repairs visually for any evidence of gap formation. As the repair was distal to the A2 pulley at the beginning of each cycle and proximal to it at the end of each cycle, any gaps could be readily noted. If a visible gap formed, the gap would be measured by a caliper and the related motion cycle number would be recorded.

Statistical analysis

A previous study showed that the average gliding resistance of a human FDP tendon before and after repair using 4-0 braided polyblend suture was 0.26 N and 1.14 N, respectively, with a standard deviation of 0.11 N and 0.39 N (Moriya et al., 2010b). Using this as a guide, with a sample size of 10 we would have 80% power at a significance level of $p < 0.05$ to detect a difference in gliding resistance of 0.28 N. This difference represents a 20% decrease, which we thought could be clinically relevant.

Gliding resistance results are presented as mean (SD). Analysis of variance (ANOVA) was used to evaluate the effect of suture technique and testing cycle on gliding resistance, both one-factor ANOVA and two-factor ANOVA models were utilized. Because up to four digits from one hand of each of 18 unique cadavers were used, it was necessary to account for the within-cadaver correlation among the digits in the analysis. This was accomplished utilizing generalized estimating equations (GEE) in a generalized linear models framework. Because of the complexity of the two-factor model (suture type and cycle number), separate analyses were conducted to evaluate suture technique within each level of testing cycle and vice versa. For each model, constant statements were generated to perform pairwise testing of each level of the independent variable (suture technique or testing cycle). All statistical tests were two-sided and the threshold of statistical significance was set at $\alpha = 0.05$.

Results

There was no significant difference in the mean gliding resistance of the intact pre-repair FDP tendons (Table 1). During 1000 cycles of tendon motion after tendon repair, no gap formation was noted in any tendon in all four repair groups. One of the two core knots unravelled during cyclic testing in three of ten of the MGH sutures in the previous study, but without any gap developing (the unravelling was noted at the beginning of failure testing).

Gliding resistance

For any given cycle number, there was no significant difference in the normalized gliding resistance within the two-strand or four-strand core repair groups. Comparing all four repairs, the mean gliding resistance of the MGH suture was significantly higher than the modified Pennington suture at 1, 100 and 1000 cycles tested ($p < 0.0001$, $p = 0.0114$, and $p = 0.0040$), as well as being higher than the modified Kessler suture at all tested cycles (all $p < 0.0001$). The normalized gliding resistance of the modified MGH suture at 1 cycle and 100 cycles, respectively, was significantly higher than the modified Kessler suture at 1 cycle ($p = 0.0236$) and 100 cycles ($p = 0.0423$), as well as being higher than the modified Pennington suture at 1, 100 and 1000 cycles tested (all $p < 0.0001$).

Comparison of force for 2 mm gap, maximum failure ultimate load and stiffness

In the two-strand core repair group, the modified Pennington suture had significantly higher force for 2 mm gap and maximum failure ultimate load than the modified Kessler suture ($p < 0.0001$ and $p = 0.0043$). In the four-strand core suture repair group, the MMGH suture had significantly higher force for 2 mm gap and maximum failure ultimate load than the MGH suture ($p = 0.0003$ and $p = 0.0063$). Comparing all four repairs, the MGH suture had significantly higher force for 2 mm gap, maximum failure ultimate load and stiffness than the modified Kessler suture ($p = 0.0011$, $p = 0.0002$ and $p = 0.0029$, respectively). The MGH suture also had significantly higher stiffness than the modified Pennington suture ($p = 0.0398$). The modified MGH suture had significantly higher force for 2 mm gap, ultimate load to failure and stiffness than the modified Kessler suture ($p < 0.0001$, $p < 0.0001$ and $p = 0.0015$, respectively). The modified MGH suture had significantly higher force for 2 mm gap and ultimate load to failure than the modified Pennington suture (all $p < 0.0001$).

Mode of failure

All repaired tendons failed by knot unravelling.

Discussion

A recent study (Moriya et al., 2010b) showed that, for tendons repaired using a grasping modified Kessler or a locking MGH core suture in conjunction with a running peripheral suture, the repaired tendons with 4-0 FiberWire had lower gliding resistance and had equal strength and equal stiffness compared to similar repairs done with a 3-0 Ethibond core suture. The MGH repair had almost double the gliding resistance and about 130% of the ultimate failure load compared to the modified Kessler repair. Previous reports (Momose et al., 2000) demonstrated that the suture knot located on the tendon surface increased the repaired tendon gliding resistance. Therefore, we modified the typical MGH repair by placing the suture knots within the tendon ends to reduce its high frictional characteristics while maintaining its strong strength. However, we could not find a significant difference between MGH and MMGH in gliding resistance. A previous study had reported that a locking modified Pennington suture had equal gliding resistance and higher strength than a grasping modified Kessler suture of 4-0 Supramid (Tanaka et al., 2004), so we wished to compare our new design with these repairs as well, using a similar suture material for all four repairs.

Although the MMGH repair did not contribute to frictional reduction for a repaired tendon, it required a significantly higher force to create a 2 mm gap and maximum failure ultimate load than the MGH suture ($p = 0.0003$ and 0.0063). Pruitt et al. (1996) reported that, in a canine model, a modified Savage suture of 4-0 braided polyester, with four knots outside the repair, was 1.14 kg stronger than the same suture whose knots were inside the repair. Tying knots in any suture material has been shown to weaken the material (Herrmann, 1971).

When the knots are outside the repair, the force is transmitted across the laceration site through an intact suture strand. By contrast, locating the knots within the repair means that the knotted sutures are bearing the full tensile load. Therefore, when the knots are inside the repair, a reduced ultimate tensile strength should be expected (Pruitt et al., 1996). However, our load to failure measurements were made after 1000 cycles of motion and so are not strictly comparable to previous studies.

The knots of three of the ten MGH sutures became partially unravelled over 1000 cycles of cyclic motion. The knots themselves remained intact and no gap formed during cyclic motion, so the tendons could still be subsequently tested to failure. This unravelling might be explained by the fact that the knots were on the tendon surface and thus more subject to abrasion by the A2 pulley during cyclic motion. This effect of cyclic motion on the suture itself might explain why we found that the modified MGH suture, with no knots on the surface, required a significantly higher force to create a 2 mm gap and maximum failure ultimate load during failure testing than was observed with the MGH suture, which included these three specimens whose knots were already partially unravelled before failure testing.

In this study, all repaired tendons failed by knot unravelling. Two possible reasons for this complication, which can be devastating clinically, relate to knot tying technique and the choice of suture material. While tying a 'granny' knot when a square knot is intended can certainly occur clinically, in our laboratory setting we are quite confident that all knots were truly square. Another possibility to explain unravelling is the number of knot loops. We tied all core sutures with four square knots. We chose this number because it is what we use clinically for flexor tendon repairs; a stack of five or more throws becomes quite bulky at the repair site in our experience. However, unravelling can be a problem, even with more throws. Waitayawinyu et al. (2008) used five-throw knots of 3-0 FiberWire in a human cadaver tendon repair model. They reported that the modified Kessler suture failed by pull-out in four of seven core sutures and by knot unravelling in three of seven core sutures; the modified Pennington suture failed by pull-out in five of seven core sutures and by suture rupture in two of seven core sutures; the modified MGH suture failed by knot unravelling in 11 of 14 core sutures and by suture rupture in three of 14 core sutures. The final potential factor is the suture material itself. Both we and Waitayawinyu et al. (2008) used FiberWire, based on its better strength characteristics (Silva et al., 2009).

Based on our findings and those of Waitayawinyu et al. (2008) we believe that the choice of suture material may be a factor in the unravelling. FiberWire is twice as stiff as Ethibond (Miller et al., 2007). FiberWire also has a lower frictional coefficient (0.054) than Supramid (0.130) and Ethibond (0.076) (Silva et al., 2009). These characteristics of more stiffness and lower friction, combined with the repetitive loading, may be more likely to loosen knots of FiberWire. To take advantage of the superior strength of FiberWire in situations where suture bulk can have an adverse effect, as in tendon repair in the fingers (Momose et al., 2000), it may be necessary to design a new type of knot, that is both more secure and with little bulk. Based on the results of this study we plan to develop such a knot.

This study has several limitations. First and most obviously, it was done in vitro and results may be different in vivo. The set up was also non-physiological, in that the tendon glided against a segment of an annular pulley, not the intact digital sheath, and the angles of loading the tendon were fixed. These are very different from in vivo human conditions. We measured the gliding resistance during 1000 cycles of motion to simulate rehabilitation therapy, but in vitro no component of healing or postoperative oedema can be simulated. Second, the DVRT device that we used to measure gap formation might affect load to failure. However, the device was inserted into an intact part of the tendon above and below the suture site and beyond the bite of the sutures themselves. Thus, although the device

might damage some tendon fascicles, it was unlikely that it would have weakened the repair construct in any way. The DVRT also might have displaced during testing, affecting the gap recordings. However, we observed no such displacement during testing. Third, tendon diameter varied somewhat from specimen to specimen, as did, consequently, the number of epitendinous suture throws. The choice to vary throws in this way was based on and designed to be consistent with, previously published descriptions (Moriya et al., 2010a; 2010b; Silva et al., 2009), but could have affected the result of load to failure. Fourth, we monitored the repairs visually for any evidence of gap formation. We could not see a gap if it was only present when the repair site was under the pulley. Finally, although every attempt was made to repair the tendons with similar tension, ultimately the repairs were hand made and could vary slightly from tendon to tendon. However, we believe that these effects should be random and not affect one repair type more than another.

In summary, we found that the modified Pennington suture had similar gliding resistance and higher load to failure after 1000 cycles of motion compared with the modified Kessler suture. The modified MGH suture had similar gliding resistance and higher load to failure after 1000 cycles of motion compared with the unmodified MGH suture. All repaired tendons failed by knot unravelling. To take the most advantage of the material properties of FiberWire, new knot configurations that are both more secure and with little bulk may be necessary. Repaired tendon strength measured after cyclic motion is more clinically relevant. It should be cautioned that placing suture knots outside the tendon surface weakens the strength of the repaired tendon by cyclic motion.

Acknowledgments

Funding

This study was supported by a grant from the Mayo Clinic.

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Figure 1. Schematic drawings of the four core suture techniques. A. The modified Kessler suture. B. The modified Pennington suture. C. The Massachusetts General Hospital (MGH) suture. D. The modified MGH suture.

Table 1

Mean gliding resistance, force for 2 mm gap, maximum failure ultimate load and stiffness, mean (SD)

Suture method	Mean Gliding Resistance (N)				Force for 2 mm Gap (N)	Maximum Failure Ultimate Load (N)	Stiffness (N/mm)
	Pre-repair	At 1 cycle	At 100 cycles	At 1000 cycles			
Modified Kessler	0.26 (SD 0.14)	0.86 (SD 0.35)	0.82 (SD 0.24)	0.88 (SD 0.30)	29.5 (SD 4.5)	39.6 (SD 7.7)	29.4 (SD 9.8)
Modified Pennington	0.30 (SD 0.06)	1.13 (SD 0.21)	1.10 (SD 0.17)	1.08 (SD 0.23)	41.0 (SD 5.3)	48.7 (SD 7.7)	35.8 (SD 15.3)
MGH	0.27 (SD 0.08)	1.46 (SD 0.13)	1.34 (SD 0.25)	1.39 (SD 0.27)	38.5 (SD 8.4)	52.0 (SD 8.9)	50.9 (SD 20.7)
Modified MGH	0.27 (SD 0.11)	1.53 (SD 0.30)	1.51 (SD 0.34)	1.50 (SD 0.26)	55.4 (SD 12.34)	65.4 (SD 11.3)	42.0 (SD 12.6)