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ORIGINAL ARTICLE

A comparison of four tibial-fixation systems in hamstring-graft anterior ligament reconstruction

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Abstract The aim of this study was to evaluate at time-zero four tibial fixations on four major criteria: the elongation and cyclic stiffness of the hamstring graft construct under cyclic loading, the yield load and pullout stiffness under load at failure. Four fixation systems were tested: the Delta screw, the WasherLoc, the TightRope Reverse and the tape locking screw on 32 tibiae of adult pigs using 32 pairs of human semitendinosus and gracilis tendons. Two tests were performed: cyclic tests using loads at 70–220 N, to measure the elongation at the end of the cycles, followed by load-to-failure testing to measure the yield load and the cyclic stiffness. The mean elongation was 1.23 mm for the TLS, 3.81 mm for the Delta, 3.59 mm for the WasherLoc and 3.91 mm for the TightRope. The mean yield loads and SD were 1,015 ± 129 N for the TLS, 844 ± 394 N for the Delta, 511 ± 95 N for the WasherLoc and 567 ± 112 N for the TightRope. The results showed the significant superiority of TLS and Delta over WasherLoc and tibial TightRope in regard to yield load. The results showed the significant superiority of TLS over the other fixations in regard to slippage. The TLS system and the Delta screw provide a better quality of primary fixation to the tibia, but further in vitro studies are needed.

Keywords Tibial-fixation systems · ACL · Hamstring tendon graft · TLS

Introduction

The anterior cruciate ligament (ACL) reconstruction is mostly performed with autologous grafts: hamstring tendon grafts or bone–patellar tendon–bone grafts. The harvesting of hamstring tendons reduces the risk of pain, of flexion limitation and muscle atrophy associated with the harvest of the patellar tendon [4, 6]. Because of this lower morbidity, hamstring harvesting is preferred by many authors [18, 53].

The initial fixation of a hamstring tendon graft is one of the factors that determine successful reconstruction of the ACL [27, 36]. Insufficient rigidity of tendon fixation has been suggested as a possible cause of ultimate laxity of the ACL reconstruction [12]. Interference screw fixation of a patellar tendon reconstruction provides good initial strength, which may explain differences in clinical outcomes between these two types of grafts for reconstructions prior to 2005 [4]. The quality of the attachment of the hamstrings and not the quality of tendons seems the weak point [6, 22]. Indeed, a graft of four strand hamstrings, equally tense, has superior mechanical properties to those of a 10-mm-width patellar tendon [39]. The recent evolution of ACL reconstruction has been toward early rehabilitation, sometimes intensive, including early full weight-bearing, lack of splint, rapid return to normal life activities. All these factors load the reconstruction constructs before secondary fixation by tendon–bone integration in the tunnels occurs [6, 18, 25]. Slippage of the fixation and plastic deformation of the graft may result in a ultimate laxity by the elongation of the graft-fixation construct [36]. Secure biological fixation of tendons does not occur until a minimum of 3 months postoperatively [37, 38].

The tibial fixation is more problematic than that on the femur due to inferior bone quality of the proximal tibial metaphysics and the direction of traction on the graft in line with the tibial tunnel [5, 6, 40]. Many studies on tibial-fixation systems and devices have been published, but the conclusions have been different. For example, looking at

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the criteria of yield strength, the WasherLoc is superior in studies by Coleridge and Amis [12] and Magen et al. [29]. Kousa et al. [27] on the other hand found the Intrafix fixation to be superior and for Giurea et al. [19] the Stirrup had the best results. New hamstring fixation options have recently appeared and have not been evaluated to our knowledge: tape locking screw [13], TightRope Reverse [28]. They both have the distinctive feature of being used for both femoral and tibial fixation.

The goals of this experimental study were to evaluate, at time-zero, four tibial fixations on three major criteria: the elongation of the hamstring graft construct under cyclic loading, the yield load and the Young's modulus under load at failure [12, 27, 36].

The work tested the hypothesis that some tibial hamstring fixation devices will resist graft slippage under cyclical loads better than others and that some will have a better higher yield load than others under ultimate tensile strength tests at time-zero.

Materials and methods

Specimen preparation

The four fixation systems were tested on 16 paired tibiae (32 total specimens) of fresh skeletally mature porcine tibiae (aged four to 6 months) using 32 pairs of semitendinosus and gracilis human tendons. The porcine tibiae were used because they have been reported to have mineral density and biomechanical characteristics close to those of the young adult human knee [33, 36]. Twelve other semitendinosus and gracilis human tendons were harvested and submitted to cyclic loads without fixation. The 32 preparations were performed by four experienced orthopedic surgeons. The tendons and tibiae were randomly assigned to different tests. The upper extremities of the pig tibiae (about 15 cm) were obtained from a local slaughter house and cleared of all soft tissues and stored in airtight bags in a freezer at −20 °C until the day of testing. Before biomechanical testing, the tibiae were thawed at room temperature for 12 h. The tendons were harvested from human cadavers from the anatomy laboratories from two Medicine Universities. Only good macroscopic tendons were chosen and preserved, given the average age of donors was 86 years (67–101 years). The tendons were cleaned of muscle fibers, covered with gauze moistened with 0.9 % saline solution and stored at −20 °C in plastic bags. These storage conditions do not affect their mechanical properties [48]. On the day of testing, the tendons and tibiae were kept at room temperature and moistened with saline. The tendons were specifically prepared for each system according to the distal part: closed loop of semitendinosus

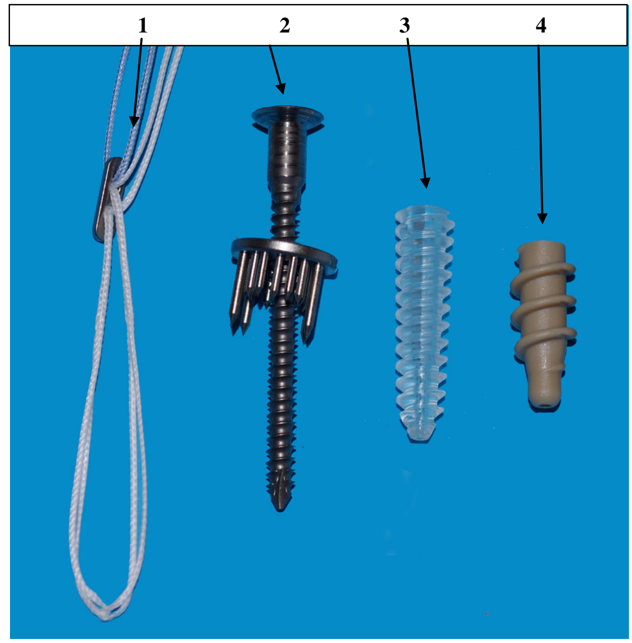

Fig. 1 Fixations from *left* to *right*: *1* TightRope Reverse, *2* WasherLoc, *3* Delta screw in poly-L-lactic acid, *4* TLS screw in PEEK

(TightRope and TLS) or open loop of semitendinosus and gracilis (Delta screw and WasherLoc). The diameters of the grafts were 8 or 9 mm. The proximal portion formed a loop around a steel cross-pin (5 mm diameter) for superior traction, simulating an identical femoral fixation for each system. In each case, the upper grip was directly connected to a 10,000 N load cell. The tendon length above the tibial tunnel was 25 mm (Fig. 1), which corresponds to the segment length of articular ACL [27, 29]. The articular surface of the tibia was cleaned, and the ACL was resected by marking out the footprint. The intraarticular tunnel exit site was centered at the footprint of the native porcine ACL. The exit point was standardized 6 cm distal to the tibial articular surface on the antero-medial tibial cortex. Tibial specimens were mounted in a custom-made jig with 6 degrees of freedom, which enabled the displacement force vector to be aligned directly with the tibial tunnel, negating the possible friction effect due to differences in tunnel and graft alignment. This creates a worst-case scenario testing condition, different from the flexion–extension motion of the knee and forces on the graft that occurs in vivo.

The fixation systems (Fig. 1)

The four systems tested covered the current options for tibial fixation at the aperture of the tibial tunnel or on the tibial cortical bone. These systems were as follows: the Delta screw, the WasherLoc the TightRope tibial and the tape locking screw (TLS). Each mounting system was tested eight times. All were prepared following the recommendations of the companies and using the specific instrumentations provided. The tibial tunnel was drilled obliquely “outside-in” with a tibial tunnel guide set at 45° from the medial metaphysics into the anatomic footprint of the ACL. The length of the tunnels was 45 mm, and the diameter was a function of the attachment system. A condition of “press-fit” was not a testing criteria.

The bioabsorbable Delta screw in poly-L-lactic acid (Arthrex, Naples, Florida, US) is slightly conical. Its length was 35 mm. Its distal end was 8.5 mm in diameter and its proximal end 10 mm. The tunnel was drilled 8 mm × 40 mm. The 4-strand tendons in the tunnel were Krackow stitch sutured with No. 1 Vicryl (Ethicon, Somerville, New Jersey, US).

The WasherLoc (Biomet, Warsaw, Indiana, US) was 20 mm in diameter, had four 11-mm long peripheral spikes that straddled the graft, and nineteen 6-mm long centrally placed spikes that penetrated the graft in multiple locations. The WasherLoc was placed on top of the four free strands of hamstring tendons, driven into the posterior wall of the tibial tunnel within the counterbore, and compressed with a 4.5-mm diameter cortical screw that purchased the posterior cortex of the tibia.

The TightRope Reverse tension (Arthrex, Naples, Florida, US) is a second generation adjustable graft loop suspensory fixation device. The TightRope tibial device consists of a loop of self-locking FiberWire and a button. The adjustable graft loop has a 4-point locking knotless mechanism to create self-reinforcing resistance to slippage under tensioning. The hamstring tendons were prepared using a closed loop of about 60 mm long before traction and placed on a traction table to 90 N. A tunnel 20 mm in length was drilled from the tibial foot print, and its diameter was adjusted to the diameter of the tendon loop. The hamstring distal loop was lowered into the tibial tunnel, the button was applied to the tibial metaphyseal cortex, with the free ends available to be tied over the button for backup tibial fixation [28].

With the tape locking screw (TLS FH Orthopedics, Mulhouse, France), the semitendinosus alone was prepared in a closed loop of four strands, 40 mm long wrapped around two tapes of polyethylene terephthalate (tensile strength: 1,400 ± 150 N, elongation at break: 27 ± 5 %) and secured together with two stitches in an X of Vicryl No. 1, at a distance of 30 mm. The loop was placed in 500 N axial tension on a specially designed tension table for less than 30 s. A retrograde tibial tunnel was drilled into the tibial footprint a length of 20 mm and a diameter of 9 mm. The tendon loop was lowered into the tunnel to a depth of 15 mm. A TLS in PEEK (polyetheretherketone), conical not broad, length 20 mm, was mounted from the distal end of the tunnel and screwed between the two tapes until flush with the metaphyseal cortex [13].

The biomechanical tests

We carefully followed the experimental protocol of Coleridge and Amis [12]. Each graft was submitted to a 30-second manual traction calibrated at 70 N by the traction machine, during the fixation. Before each test, if there was a loss of tension during placement of the fixation, the level of initial tension was increased back to 70 N [20]. Two successive tests (Fig. 2) were applied: cyclic tests then failure tests, 64 tests for the 4 fixation systems, and 12 tests on the tendons themselves.

The cyclic tests (Fig. 3) were applied from 0 to 150 N (70 to 220 N for the graft), 1,000 times at a speed of 80 mm/min (frequency of 1.33 Hz). These tests evaluated the residual elongation (mm) after each cycle. We measured the acquired permanent elongation after 20 cycles (% of maximum displacement) and the final elongation at the end of 1,000 cycles of the tendon–tibial fixation construct [9]. This protocol produced hysteresis curves that characterize the behavior of the construct in its visco-elastic phase [40]. The Phenomenon of stress relaxation is important during walking or jogging, in which the applied strains are repetitive and nearly constant [30, 49]. The more a construct leaves an elevated early and final elongation, the less

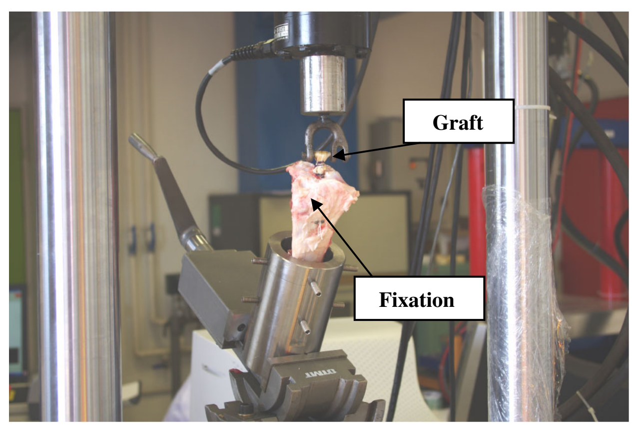

Fig. 2 The tibia was fixed to the base of the testing machine with a metallic cylinder which enabled the displacement force vector to be directly aligned with the tibial tunnel

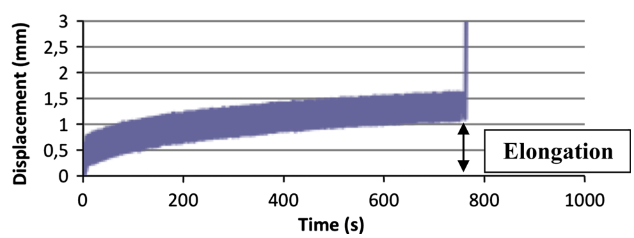

Fig. 3 Cyclic-loading test. Elongation depends on the number of preconditioning cycles. We measured the elongation at 20 and 1,000 cycles

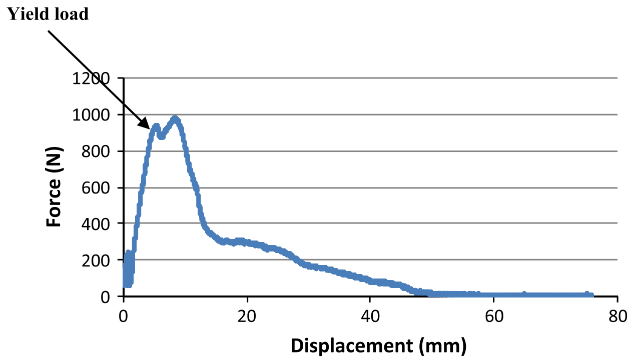

Fig. 4 Single-cycle load-to-failure test. The yield load corresponds to the inflection point of the curve. The slope of the quasi-linear curve represents the stiffness (N/mm) of the fixation–tendon construct

it will be effective in controlling anterior laxity of the knee. The cyclic linear stiffness (N/mm) was calculated at the end of 1,000 cycles (cyclical stiffness). Eight constructs were tested with each fixation device.

Single-cycle load-to-failure test (Fig. 4). After completion of submaximal-load cyclic testing, a single-cycle load-to-failure test was conducted at a displacement rate of 80 mm/min. We recorded the classic force—deformation curve, the value of the yield load (N), the pullout stiffness (N/mm) and the tendon characteristics at rupture. The yield load corresponded to the inflection point of the curve and was lower than the failure load. It characterized the elastic phase before the onset of plastic deformation and its slope was measured by the pullout stiffness. Eight constructs were tested with each fixation device.

The elongation measured under cyclic loads was the sum of the elongation of the tendons and the elongation specific to each fixation. It was essential to measure the final elongation of the tendons in isolation to deduce the global elongation recorded after 1,000 cycles. For this, we submitted 12 loops (length: 50 mm and diameter: 8–9 mm) of semitendinosus and gracilis tendons to 1,000 cycles from 70 to 220 N and measured the final average elongation. The results calculated account for the specific tendon elongation that occured independently of the fixation.

All tests were performed over 3 days at the Regional Center for Innovation and Technology Transfer on a servohydrolic testing machine (Model 858, MiniBionix II, MTS System Corp, Minneapolis, MN) by an engineer, independent of the companies that provided the fixation materials (Fig. 2).

The statistical tests

Primary endpoint was elongation. Secondary endpoints were yield load and Young's modulus after 1 and 1,000 cycles. All variables had a normal distribution (Kolmogorov–Smirnov test). Means were tested by variance analysis (ANOVA) for repeated measures. In the cyclic tests, elongation was considered as the dependant variable and devices as the independent. In the failure tests, yield load was considered as the dependant variable and devices as the independent. Means were globally tested. Then, each mean was compared and tested with the three others, taking the previous tests into account. All analyses were performed with STATA 9 software (StataCorp, TX, US). Result demonstrating $p < 0.05$ was considered statistically significant. The upper 95 % prediction intervals for the amount of graft slippage (mm) and the lower 95 % prediction intervals for the yield load (N) were calculated.

Results

The tendons alone had an average elongation after 1,000 cycles of 1.58 ± 0.48 mm and a yield load of 1,369 ± 393 N. This value was subtracted from the elongation recorded for each fixation tested, except for the TLS construct. The maximum error of measurements was 0.48 mm due to inter-specimen variation.

Cyclic-loading test

The results were presented in Table 1. For the final elongation, the difference was significant between the TLS and the three other systems, but not significant between each other. The high final elongation (31.12 mm) for Delta screw corresponded to a tendon laceration before the end of the cycles. In this work, we accepted outlying data points which may represent individual patients whose graft will tend to become loose under rehabilitation conditions. The upper 95 % prediction intervals for the amount of graft slippage (mm) was below 3 mm ("normal" on the IKDC scale) for the TLS and above 3 mm ("nearly normal" on the IKDC scale) for the three others. The elongation for three devices (WasherLoc, TLS and TightRope) occurred during the first 20 cycles and reached 50 % of the final stretch. The Delta screw had an elongation of 21.5 % at 20 cycles.

Cyclic stiffness after one cycle were 195.7 N, 138.2 N, 131.8 N and 105.2 N for the Delta screw, the TLS, the WasherLoc and TightRope, respectively, and was significant for the Delta against the TLS ($p < 0.05$), the WasherLoc ($p < 0.05$) and the TightRope ($p < 0.05$). Cyclic stiffness after 1,000 cycles was 309.7 N, 295.9 N, 232 N and 260 N for the Delta screw, the TLS, the WasherLoc and TightRope, respectively. After 1,000 cycles, cyclic stiffness was significantly higher than after a single cycle, regardless of the construct but were not significantly different between the four devices.

Table 1 Results of mean elongation (mm) of each device after cyclical loading, range (mm), upper 95 % prediction limit (mm), percentage of elongation after 20 cycles and cyclic stiffness (N/mm)

	Vis Delta	WasherLoc	TightRope Reverse	Vis TLS
Mean elongation (mm)	3.81 ± 11.25	3.59 ± 2.59	3.91 ± 1.39	1.23 ± 0.36
Range (mm)	0–31.12	0.96–7.97	2.75–6.56	0.93–1.96
Upper 95 % prediction limit (mm)	12.7	5.8	5.08	1.55
% Elongation after 20 cycles	21.5	59	54	62
Cyclic stiffness				
>1,000 cycles (N/mm)	309.7 ± 75	232 ± 58	260 ± 64	295.9 ± 78

Table 2 Results of mean yield load (*N*) of fixation devices, range (*N*), lower 95 % prediction limit and pullout stiffness (N/mm)

	Delta screw	WasherLoc	TightRope Reverse	TLS
Mean yield load (*N*)	844 ± 394	511 ± 95	567 ± 112	1,015 ± 129
Range (*N*)	324–1,437	338–607	325–697	864–1,276
Lower 95 % prediction limit (*N*)	513	432	467	907
Pullout stiffness (N/mm)	195.7 ± 59	131.8 ± 56	105.2 ± 23	138.2 ± 35

Table 3 Different types of failure under single-cycle load-to-failure test

	Delta screw	WasherLoc	TightRope Reverse	TLS
Graft failure	3	0	4	6
Graft slippage	5	8	0	2
FiberWire failure	0	0	2	0
Button migration	0	0	2	0

Single-cycle load-to-failure test

The results are presented in Table 2. The TLS system and Delta screw had the highest yield load (differences not significant), and the TLS system was significantly superior to the WasherLoc ($p < 0.01$) and to the TightRope ($p < 0.05$). The WasherLoc and the TightRope were not significantly different. The lower 95 % prediction intervals for the yield load (N) were significantly better for the TLS system (907 N), than for the 3 other devices, Delta screw (513 N), WasherLoc (432 N) and TightRope (467 N).

Pullout stiffness after one cycle was 195.7 N/mm, 138.2 N/mm, 131.8 N/mm and 105.2 N/mm for the Delta screw, the TLS, the WasherLoc and TightRope, respectively, and was significant for the Delta against the TLS, the WasherLoc and the TightRope. After one cycle, the pullout stiffness for all the devices was significantly lower than after 1,000 cycles, regardless of the construct.

The different types of rupture are reported in Table 3. Tendon rupture was the main cause of failure in ultimate load for the TLS (six cases), for the TightRope (four cases) and for the Delta (three cases). Slippage occurred in five cases for the Delta (including one case of immediate slippage of 31 mm) and two cases for the TLS screw. Laceration of the tendons under the WasherLoc occurred in all cases (eight cases). The TightRope had two particular modes of failure: migration of the button in the tunnel and FiberWire rupture.

Discussion

This study demonstrated the superiority of TLS and Delta fixations over the TightRope tibial and WasherLoc for the yield load (higher yield load) and the superiority of TLS for the cyclic-loading tests (less elongation) over the three others fixations. If we take 500 N as a minimum yield load to avoid exposing the construct to deformation during the stress of everyday life and intensive rehabilitation, only the Delta screw and TLS screw are above (513 and 907 N, respectively) [29, 30, 32]. If we take 3 mm as a threshold minimum elongation compatible with a successful final result, only the TLS construct was below (1.55 mm), the three others were above 3 mm. A stable mechanical environment is required for graft maturation and ligamentization [42, 52]. If the graft elongates but still heals, patients may experience instability when they return to sport without a rerupture [36]. The stiffness was high enough for all fixations to not have clinical significance.

The TLS system and Delta screw gripped on the tibial cortex at the tunnel entrance and cancellous tunnel bone and they may be considered as a semiaperture fixation or intratunnel fixation [23, 27, 41], while the TightRope and the WasherLoc gripped only on the cortical and are considered as a non-anatomical fixation or extratunnel fixation [23]. Our results for elongation and yield load of the Delta screw were similar with those obtained by Coleridge et al. [12]. The density of the cancellous bone closely conditions the torque of the tightening of the screws [5, 10], and the density of porcine cancellous bone is very high and higher than that found in humans [27]. The TLS screw derived benefits from its conical shape and its large diameter (10 mm) allowing the penetration of polyethylene terephthalate tapes into the metaphyseal cancellous bone [25]. In addition, Delta and TLS screws gripped partly on the tibial cortex, which increased resistance. The length of 35 mm and the conical shape of the Delta screw were an advantage for the stabilization of the grafts compared to a shorter (20 and 25 mm) and cylindrical interference screw, which was used previously [10–12, 46, 55]. The WasherLoc had poor results in this study, due to the laceration of the tendons (8 out of 8), compared to the results of Colerige et al. [12] or Kousa et al. [27], but similar to those of Giura et al. [19] and those of Fabbriciani et al. [17]. The quality of the preparation of the tendons with a running suture tendon and the tendon density are essential for obtaining good results with this fixation [27]. Our results of the two tests were lower than those of other series, probably as a result of the preparation and the age of the donor tendon [12, 27]. The TightRope system consists of a loop of FiberWire and a button applied to the metaphyseal cortex. The lower mechanical performance is related to progressive tightening of the knots, the deformation of the button and the lack of close contact between the button and the tibial cortex ("stretch out" of suture/tape materials). The total elongations on the TightRope are superior to those reported by Petre et al. [36] (1.10 ± 0.20 mm), but close than those reported by Barrow et al. [1] (3.22 ± 1.33 mm after 1,000 cycles). Our macroscopic findings showed poor contact of the button on the bone. For some authors, the growing number of interfaces reduces the overall stiffness of the construct [9, 40].

Graft failure was the main failure mode for TLS, which reflected the good quality of the primary tape and screw fixation. The graft slippage was more a feature of Delta and WasherLoc and corresponded to a laceration of the tendons under the screws or the spikes [55]. The TightRope had two specific modes of failure: FiberWire rupture and button migration into the tunnel.

We have shown that the cyclic traction of grafts enhances the stiffness of the construct for all fixations and approach that of the native ACL [36]. Before traction, the stiffness was lower than that of the native ACL [51]. This effect, obtained in the first cycles, is related to the visco-elastic behavior of tendons and confirms previous work on the hamstrings, the patellar tendon and quadriceps tendon [8, 19, 27, 40, 44]. The preconditioning of grafts can be achieved in clinical practice by a series of the cyclically applied loads (30–50 times) of the graft prior to tibial fixation or by an axial traction table [8, 13, 28]. The high pretransplant conditioning by the TLS was probably one factor explaining the good mechanical performances registered in this study. The axial pretensioning on a specific table up to 300 N for 30 s was much stronger in the TLS than the TightRope system (90 N). Neither the Delta screw nor the WasherLoc currently come with an instrumented graft-tensioning device. This preconditioning had a positive effect on the stiffness of the tendon loop and the tendon-tape junction. In the TLS system, the closed loop tendon lengthened moderately, remaining in an elastic phase and without compromising the micro architecture of collagen fibers [22]. Like other authors, we assume cyclic stiffness did not change because of alterations of the collagen fibers but because of progressive fiber recruitment and interstitial fluid loss [40]. Preconditioning is "implant specific" but remains controversial for several authors. Some recommend not exceeding 40 N to avoid problems with revascularization. Yoshiya et al. show over constraining lesions from 30 Newton. Elias et al. prefer a high pretension of 160 N to a low pretension of 80 N for reducing the postoperative loss of tension [16, 54]. Other authors found no significant difference between the preconditionned group and the non-preconditioned group at similar prestress levels in mechanical and clinical studies incorporating laximeters and clinical and functional scores [15, 45, 54].

For reasons of availability, all authors have used bone and tendon samples of animals or aged human cadavers. It is difficult to compare the results of these studies because of different experimental conditions utilized: the origin of the tendons and bone, number of traction cycles and the speed and levels of traction. Our experimental conditions (human tendons and porcine bone) are close to those of Kousa et al. [27].

Table 4 demonstrates the testing conditions from several publications studying tibial fixations. It appears that the density of porcine tendons provides better mechanical conditions compared to human tendons [27]. By necessity, we used the tendons of elderly humans (average age 86 years), which reduced the performance of all systems.

Table 4 Different experimental conditions in the literature

	Tendon origin	Bone origin	Cyclic conditions	Speed
Kousa et al. [27]	Human	Porcine	1,500 cycles from 50 to 200 N	50 mm/min
Magen et al. [29]	Bovine and human	Porcine	1,100 cycles from 50 to 1,000 N	Steps of 50 N
Giurea et al. [19]	Bovine	Bovine	1,100 cycles from 0 to 150 N	250 mm/min
Coleridge et al. [12]	Bovine	Veal	1,000 cycles from 70 to 220 N	100 mm/min
Our study	Human	Porcine	1,000 cycles from 70 to 220 N	80 mm/min

The cancellous porcine bone has a slightly higher density than that found in young adult human bone, which can improve the performance of systems such as the TLS and Delta screw [5]. The bovine bone has a similar density to that of young human bone, but better than that of aged human bone [2, 7].

There are several limitations in this study. For reasons of availability, we used aged human tendons which penalize all systems. The failure values would likely have been better if the hamstring tendons from young patients, more representative of those in the population who undergo an ACL reconstruction, had been available. We used porcine bone, which has a slightly higher density than that found in young adult human bone, and can improve the performance of systems such as the TLS and Delta screw. However, the donor age and bone quality are more uniform in pig specimens than those obtained from human donors [29, 33]. In addition, the allocation was random and the animals were the same age (4–6 months) at slaughter. We did the tests in the axis of the tibial tunnel, which is different from the loading conditions for an ACL and represents a worst-case loading scenario. The loading of constructs at a higher level of traction and cycles would most likely have led to more laxity. Six weeks of performing the activities of daily living corresponds to approximately 220,000 cycles for the ACL at a tension of 170 N [32, 43]. We strictly applied the Coleridge and Amis [12] protocol, without attempting to reproduce the mechanical conditions of the first postoperative weeks. It is a time-zero study which does not take into account the evolution of the elongation in the first weeks after the implantation.

The hypothesis of this work has been confirmed; some devices have better mechanical behavior than others under cyclic and tensile loading conditions. The yield load and elongation under cyclic loads were more discriminating than the pullout or cyclic stiffness to differentiate these four methods of fixation. Further work is needed to confirm these results.

While we must be careful to transfer these results to clinical practice, some fixations would not support an early aggressive rehabilitation or running, where the level of stress on the ACL is estimated at 500 N [31, 32]. Although cyclic displacement does not correspond directly to anterior displacement, cyclic displacement greater than 3 mm does cause concern that anterior laxity may occur postoperatively. The upper 95 % prediction intervals for cyclic displacement were above 3 mm for three devices. All these mechanical tests done in the laboratory at time-zero should take into account osseous integration in tunnels and ligamentization phenomena experienced by the graft. In humans, the tendons have a reliable secondary fixation at the entrance to the tunnels after 3 months minimum, and they undergo a process of ligamentization for a minimum of 2 years [38, 42, 48]. In addition, the insertion of grafts in anatomic and not isometric sites is essential in order to respect the mechanics of the knee, ligamentization and to ensure long-term clinical success [14, 26, 27]. For Han et al. [23] at a minimum of 2 years' follow-up, patients who underwent hamstring autograft ACL reconstruction with intratunnel or extratunnel fixation displayed comparable outcomes based on objective IKDC, Lysholm score and Tegner activity level survey results.

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Conflict of interest Two authors (MC and XC) are consultants for FH Orthopedics.

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