ACL Graft Compression: A method to allow reduced tunnel sizes in ACL reconstruction

Breck R Lord[1,2], Henry B Colaco[3], Chinmay M Gupte[4], Adrian J Wilson[2], Andrew A Amis[1,4]

Work performed at Imperial College London

[1] The Biomechanics Group, Department of Mechanical Engineering, Imperial College London SW7 2AZ, UK. a.amis@imperial.ac.uk (address for correspondence)

[2] Basingstoke and North Hampshire Hospital, Basingstoke, UK

[3] St Georges Hospital, London, UK

[4] Musculoskeletal Surgery Group, Department of Surgery and Cancer, Imperial College London School of Medicine, Charing Cross Hospital, London, UK

Correspondence to Prof Andrew Amis

a.amis@imperial.ac.uk

Abstract

Purpose: A common problem during ACL reconstruction (ACLR) is asymmetry of proximal-distal graft diameter leading to tunnel upsizing and graft-tunnel mismatch. Compression downsizing provides a graft of uniform size, allowing easy passage into a smaller tunnel. The purpose of this study was to quantify the graft compression technique and its effects on graft biomechanics and stability. It was hypothesized that: compression downsizing would significantly reduce cross-sectional area (CSA); that no significant changes in graft biomechanics would occur; graft fixation stability would be improved.

Method: Sixty-eight non-irradiated peroneus longus (PL) tendons were investigated. Twenty were halved and paired into ten 4-strand grafts, 20 strands were compressed by 0.5-1 mm diameter and changes in CSA recorded using an alginate mould technique. The following properties were compared with 20 control strands: cyclic strain when loaded 70-220 N for 1000 cycles; stiffness; ultimate tensile load and stress; Young's modulus. 24 PL tendons were quadrupled into grafts, 12 were compressed and all 24 were submerged in Ringer's solution at 37ºC and the CSA recorded over 12 hours. Twelve compressed and 12 control quadrupled grafts were mounted in porcine femurs, placed in Ringer's solution for 12 hours at 37ºC and graft displacement at the bone tunnel aperture recorded under cyclic loading.

Results: Mean decreases in CSA of 31% under a stress of 471 kPa and 21% under a stress of 447 kPa were observed for doubled and quadrupled grafts, respectively. Compressed grafts re-expanded by 19% over 12 hours compared to 2% for controls. No significant differences were observed between compressed and control grafts in the biomechanical properties and graft stability; mean cyclic displacements were 0.3 mm for both groups.

Conclusions: No detrimental biomechanical effects of graft compression on allograft PL tendons were observed. Following compression, the grafts significantly increased in size during in vitro joint simulation. No significant difference was observed in graft stability

between groups. Graft compression did not cause adverse mechanical effects in-vitro. Smaller tunnels for compressed grafts reduce bone loss and ease anatomical placement.

Keywords: Anterior cruciate ligament; ACL reconstruction; compression downsizing; tendon graft biomechanics.

Introduction

Bone tunnel enlargement after ACL reconstruction (ACLR) has been suggested as a factor contributing to graft failure [31]. The aetiology of this is unclear but is thought to be a combination of mechanical and biological factors [27, 28]. A common problem encountered during ACLR is asymmetry of proximal-distal graft diameter, and one way to deal with this is to drill the tunnel to match the larger end of the graft in order avoid difficulty during graft passage. The resulting graft-tunnel mismatch in diameter creates effective tunnel widening before mechanical or biological factors compound the issue. The result is reduced bone stock, which may compromise revision surgery. It has been suggested that extravasation of synovial fluid can impair graft integration in the period following surgery and thereafter cause tunnel widening, especially when cortical suspensory fixation is used, which may allow cyclic micromotion across the graft-bone interface [6, 7, 11, 15, 23]. Compression downsizing, in which the graft is squeezed to a smaller uniform diameter, could provide an ACL graft of uniform size, allowing easy passage into a smaller tunnel. It could also be hypothesised that a compressed graft would have the further advantage that it could relax and expand within the bone tunnel, effectively blocking the passage of synovia fluid. In addition, studies with up to 15 years follow up have reported ruptured ACL grafts in 4.1% to 11.2% of patients [17, 21]. Smaller tunnels would preserve bone stock to aid revision surgery, in addition to making it easier to keep the tunnel aperture within the native ACL attachment.

The effects of a clinically applicable graft compression technique on the biomechanical properties of grafts have not been reported. Much like the commonly accepted technique of pre-tensioning ACL grafts prior to insertion and fixation [8], compression downsizing is a simple concept that aids surgical technique and may benefit the patient. During surgery, graft downsizing is obtained easily by pulling the graft through progressively smaller-diameter sizing tubes. When a larger tension than that needed to simply slide the graft through the tube is felt (typically when the diameter is 0.5 or 1 mm below the nominal graft size), the graft can be left inside the tube while the ACL procedure progresses, and then may be pulled into an undersized bone tunnel..

The purposes of this study were to measure changes in clinically available human allograft CSA during and following compression downsizing. In addition, to identify any effects on graft biomechanics and fixation stability.

It was hypothesized that: compression downsizing would produce a significant reduction in CSA that would subsequently expand during in vitro joint simulation; that no significant changes in graft biomechanics would be observed; and that graft fixation stability would be improved. The clinical relevance is that smaller tunnels will reduce the volume of bone lost during ACL reconstruction, and compressed grafts – which are expected to swell when immersed in joint fluid – may also grip the wall of the bone tunnel and thereby reduce slippage under loading post-surgery and before healing occurs.

Materials and Methods

Following ethical approval, 68 peroneus longus (PL) chemically treated and non-irradiated (BioClense, RTI Surgical, Alachua, Florida, USA) allograft tendons were obtained and assigned to test groups (Figure 1); they were stored at -20^{o}C before use. These tendons were available for testing, and their mechanical properties do not differ from those of other hamstrings tendons [25].

Quantification of graft compression

This assessment was performed for each of the two common methods of 4-strand ACL graft preparation: two doubled tendons; one quadrupled tendon (Figure 1). The allograft tendons were wrapped in a surgical swab immersed in isotonic Ringer's solution at room temperature for 2 hours prior to testing to normalise any osmotic effect between specimens. Ten PL tendons were doubled over in pairs (that is: a 4-strand graft structure) and placed in a custom designed steel compression block which provided a 75 mm long cylinder of a given diameter under a compressive load (Figure 2). Starting with a 12.5 mm diameter block, progressively smaller blocks were used until a 20 N compressive load was unable to close the block completely, this was taken as the pre-compression graft diameter.

Using a non-destructive alginate moulding method [10] (Blueprint cremix, Dentsply DeTrey, Germany), moulds were taken of the central 40mm of each strand, then four 2mm axial slices were taken at equal intervals and the mean pre-compression CSA (mm^2) calculated (Image J, National Institute of Health, Bethesda, Maryland, USA) from a digital image (Canon EOS 100D). Twenty preliminary tests were carried out on a steel bar of known cross-section on 5 separate days to confirm the accuracy and repeatability of this method. The graft was then placed in a compression block 1mm less in diameter and a screw-driven Instron 5565 materials testing machine (Instron, High Wycombe, UK) used to apply a 10 N/s compressive load until a final force was recorded at closure, this was held for 10 minutes. A compression of 1 mm in diameter was chosen as this was sufficient to produce grafts of uniform diameter and the forces required to compress grafts beyond this were not easily produced in the clinical setting without specialist equipment. Each PL tendon was then divided in half and post-compression CSA measurements taken at the same positions. Thirty PL tendons (n=24 in Figure 1, plus a further 6 from the pilot testing) were quadrupled

as previously described [18], to create a clinical ACL graft structure, and underwent the same protocol (Figure 3).

ACL graft geometric response – in vitro simulation

Twenty-four quadrupled PL tendons were divided into 12 pairs consisting of one control and one graft compressed along its entire length as described above and shown in Figure 2. These were submerged in Ringer's solution (Merck, Darmstadt, Germany) at 37ºC under 20N of tension [5]. The mean CSA was measured at 2, 4, 6, 8 and 12 hours as previously described.

Tensile Testing

Forty single stranded grafts (The two groups of 20 divided grafts in Figure 1) were divided into matched pairs within control and compressed groups, and the mean CSA of each graft recorded before testing. An Instron 8874 servohydraulic materials testing machine (Instron, High Wycombe, UK) was used for the biomechanical testing. Each specimen was mounted in a pair of cryo-jaws positioned to match the central 40 mm of the tendon to approximate the intra-articular length of an ACL graft [3]. Liquid CO_2 was used to freeze the jaws before and after tightening to prevent slippage of the tendon in the clamps [26]. Each clamp had rounded edges to reduce the stress riser at the tendon interface. Two types of test were conducted: a non-destructive cyclic tensile loading test followed by an ultimate tensile stress (UTS) test.

After applying a pre-conditioning loading of 20 cycles of 0-70N, the load on the tendon was increased to 145N and then 1,000 cycles between 70N and 220N were applied at a frequency of 1 Hz. This protocol represented the loads experienced by the ACL during normal walking [4]. The mean, range, minimum and maximum loads and length for each cycle were recorded.

The specimens were then tensile tested to failure at a crosshead speed of 1,000 mm/min, recording position and load. These data and the CSA were used to calculate the stiffness, UTS and Young's Modulus for each tendon.

ACL graft fixation stability - in vitro simulation

Twenty-four large porcine femurs were stripped of all soft tissue and the medial condyle resected at the level of the inter-condylar notch. The femoral shaft was mounted in a stainless steel pot of 60 mm diameter using poly methylmethacrylate (Simplex Rapid, Kemdent, UK). The CSA of 12 control and 12 compressed quadrupled PL tendons was measured and the appropriate socket created to a depth of 30mm using a FlipCutter (Arthrex Inc., Naples, Florida, USA). Cortical fixation was achieved with a 15mm RetroButton (Arthrex). Each specimen was placed in Ringer's solution at 37ºC for 8 hours with a 20 N tensile load applied to the graft. The servohydraulic Instron materials testing machine was used to conduct the same cyclic loading protocol outlined above. The specimen was mounted in a specifically designed test rig with several degrees of freedom so that the ACL graft socket was aligned on the loading axis, with an orientation of approximately to 30º in the coronal plane [2] and 60º to the long axis of the porcine femur in the sagittal plane (Figure 4). After pre-conditioning, the tensile load was held at 50 N while a black 2-0 nylon suture was passed in a 'figure of 8' fashion through all 4 graft strands 5mm from the bony aperture. Similarly, a black plastic marker mounted on the bone produced a sharply defined edge 2 mm from the bony aperture. Any increase of the distance between the tendon and bone mounted markers was taken to represent slippage of the graft out of the bone tunnel. Eighteen-megapixel digital images were recorded at a rate of 25 /s (Canon EOS 100D) and processed using Image J (NIH) and cyclic displacement was recorded over 1,000 cycles.

Research Ethics Committee Approval:

This study was approved by the Imperial College Healthcare Tissue Bank, HTA licence 12275, REC Wales approval 12/WA/0196, project R13058.

Statistical analysis

To assess the repeatability of the moulding technique to measure CSA, Intraclass Correlation Coefficients (ICC) were determined. Paired samples t-tests were used to analyse pre- and post-compression CSA for each of the two graft constructs. The independent-samples t-test was used to analyse data from the quantification of the graft compression technique, cyclic loading and ultimate load to failure testing. The CSA and cyclic motion data from the two in vitro simulations were analysed using a one way analysis of variance (ANOVA) with a *post hoc* Tukey's test. The level of significance was set at 0.05. Statistical analysis was performed using statistical software (Statistical Package for the Social Sciences (SPSS) 21.0; SPSS Chicago IL).

In addition, pilot test data from 6 specimens were used to perform relevant prospective power calculations (using software at https://www.stat.ubc.ca), to confirm that the numbers of tests would exceed the thresholds of 80% power at 95% confidence. This showed that 8 pairs of values would give a power of 96% with 95% confidence for one-tailed difference between two dependent means.

Results

Power calculation

With groups of 12 pairs of values, the retrospective power analysis found that an actual power of 0.996 was achieved with 95% confidence, based on graft compression data.

Graft compression

The CSA of the metal bar was 87.4 mm^2, and the moulded sections a mean of 87.4 mm^2 ± 1.6 mm^2. Test-retest analysis of the alginate moulding technique revealed high repeatability, with a mean difference of 3 mm^2 (that is: approximately ± 3%) on different test days, and an ICC of >0.9. The mean initial CSA of the doubled PL tendons as a 4-strand construct was 87 ± 9 mm^2 (mean ± standard deviation (SD)), equivalent to a graft diameter of 10.5 mm. A significant reduction in CSA was observed between the control and compressed groups (P < 0.001) of 16 ± 8% following a mean compressive pressure of 471 ± 196 kPa, giving a reduced diameter of 9.6 mm. In the quadrupled format, the CSA was 90 ± 9 mm^2, equivalent to 10.7 mm in diameter. The same protocol resulted in a significant reduction in CSA (P < 0.001) by 13% ± 4% following a mean compressive pressure of 447 ± 132 kPa, giving a reduced diameter of 10.0 mm..

ACL graft geometric response – in vitro simulation

The CSAs of the grafts before they were compressed were 91 ± 10 mm^2 and 88 ± 8 mm^2, which equates to graft diameters of 10.8 mm and 10.6 mm, respectively. With immersion in Ringer's solution, significant increases of the CSA of compressed quadrupled grafts were observed at every time interval with a mean change in CSA at 12 hours of 19 ± 9% (17 mm^2; P < 0.004), while the control group changed by 2 ± 5% (2 mm^2; Non-significant (n.s.)), leading to swollen graft diameters of 11.8 and 10.7 mm, respectively (Figure 5).

Tensile Testing

No significant differences were found between the control and compressed groups in terms of cyclic strain, UFL, stiffness, UTS or Young's modulus (Table 1).

ACL graft fixation stability

No significant differences in cyclic displacement were observed between the control and compressed groups at any stage over 1,000 cycles. Cyclic displacements of 0.3 ± 0.1 mm were observed at 1,000 cycles for both the control and compressed groups.

Discussion

The most important finding in this study was that a 4-strand ACL allograft construct could be compressed so that it would fit into a bone tunnel 1 mm smaller diameter, thus reducing bone removal by approximately 20%. As had been hypothesised, there were no significant changes to the inherent biomechanical properties of the ACL allograft material following graft compression. However, contrary to our hypothesis, although compressed grafts were found to expand when soaked in Ringer's solution, this expansion in-situ did not lead to greater resistance to slipping from the bone tunnel under cyclic loading.

Compression downsizing offers the opportunity for a 21% (size 11 mm to 10 mm) to 31% (size 8 mm to 7 mm) smaller bone tunnel CSA with no compromise in the biomechanical properties of the reconstruction, enabling a smaller graft tunnel to be used; that may be advantageous if revision surgery is required. With the advent of 'anatomic' ACL reconstruction, studies have highlighted the importance of graft placement within the native ACL attachment sites both in vitro [14] and in vivo [9, 12]; a smaller tunnel size would make it easier to keep the aperture within the ACL attachment area. This is of particular importance in the context of the small female knee or in a child. ACL injury has become increasingly common in children and adolescents as the number of young athletes involved in high demand sport increases [1, 20], such cases reportedly account for 3% of all ACL injuries [19]. Recent anatomical studies have highlighted that the ACL attaches posterior to the anterior edge of the lateral intercondylar ridge [29], and a 9 mm graft may encroach beyond the ACL attachment even when correctly positioned. Some surgeons utilize a 4 tunnel (double-bundle) technique and in this context a reduction in tunnel size would make it

easier to maintain a bony bridge and avoid coalescence between pairs of tunnels [16]. Similarly, in the context of the multi-ligament knee injury, smaller tunnels decrease the chances of tunnel convergence during graft fixation.

Despite the re-expansion of the compressed grafts, which was expected to increase the pull-out strength from the bone tunnel, no significant difference was observed in graft fixation stability at the bony aperture when compared to the control group. However, extravasation of synovial fluid has been proposed as a possible cause of osteolysis at the graft-osseous interface, leading to tunnel enlargement [7, 11]. Despite not adding any mechanical stability to the graft, the press-fit created by compression downsizing and re-expansion may impedeor avoid extravasation of synovial fluid into the graft tunnels; further clinical investigation is required.

Hwang et al. [13] performed a prospective study of 35 patients in whom the femoral ACL graft tunnels were under drilled by 0.5 mm and the graft forced through. They reported no significant difference in total tunnel volume between these and the control specimens when evaluated using computer tomography at one year, no difference in International Knee Documentation Score or KT-2000 laxity. However, allografts are often oedematous and the peri-operative results presented in this study show that an allograft can be compressed and passed into a socket of 1 mm diameter smaller, representing 20-30% preservation of bone stock. Although no definitive clinical benefit has yet been identified, the clinical authors of the present study have now adopted ACL graft compression to preserve native bone stock whilst maintaining ease of graft passage in both the primary (autograft) and revision (autograft / allograft) setting. This is done easily by pulling the freshly-prepared graft into progressively smaller sizing tubes.

The major limitation of this study is that it used chemically-treated PL allografts due to the number of samples required, their consistency, cost and availability. The morphology of the PL tendon is partially cylindrical and partially ribbon-like, these were allocated equally throughout to minimise any bias. This study was not intended to support allografts rather

than autografts, and the authors are unaware of evidence to suggest that the graft compression would have a different effect on autografts, other than the practical observation that fresh autografts are less oedematous and so only reduce 0.5 mm diameter. Other limitations include: the data relate only to the situation immediately post-surgery with a limited number of cycles, excluding biological factors such as ligamentisation of the ACL graft; porcine femurs were used for more consistent cancellous bone quality when compared to cadaveric specimens which are older than the patient demographic for ACL reconstruction. The structural and material properties of the PL allografts found in this study were in line with published data on ligaments and grafts: the ultimate failure load (UFL) of the ACL was found to be 2,160 N in the young adult [32] whilst Noyes et al [22], reported a UFL of 1,725 N and a UTS of 37.8 MPa. Similarly, UFL values of 1,784 -2,900 N and 2,421 N have been reported for the commonly used patellar tendon and doubled semitendinosus/gracilis hamstring grafts, respectively [22, 30]. Pearsall et al [24] tested fresh frozen PL autografts, reporting a mean UFL of 2483 N and a mean stiffness of 244 N/mm in doubled PL tendons.

Conclusion

The clinical implication of this study is that graft compression, which does not increase cost or operative time, appears to preserve bone stock whilst aiding surgical technique. No detrimental biomechanical effects on chemically treated allograft PL tendons were observed after a compression protocol that used easily applicable pressures. Following graft compression, these tendons significantly increased in size during in vitro joint simulation. Although no significant difference was observed in graft fixation stability between groups, graft compression should aid graft passage, anatomic positioning within the attachment and preservation of bone stock during ACL reconstruction.

List of abbreviations

ACL: anterior cruciate ligament

ACLR: anterior cruciate ligament reconstruction

ANOVA: analysis of variance

CSA: cross-sectional area

ICC: Intraclass Correlation Coefficients

n.s.: Non-significant

PL: peroneus longus

SD: standard deviation

SPSS: Statistical Package for the Social Sciences

USA: United States of America

UFL: ultimate failure load

UTS: ultimate tensile stress

Authors' contributions

Blinded for review

Acknowledgements

Blinded for review

Captions for illustrations

Figure 1. A total of 68 peroneus longus allograft tendons were allocated across each arm of the study.

Figure 2. A pair of peroneus longus allograft tendons doubled over to create a 4-strand soft tissue graft. The graft was placed into a custom designed compression block mounted within a materials testing machine.

Figure 3. (a) An uncompressed quadrupled peroneus longus allograft tendon. **(b)**. The same graft took on a more compact and uniform shape following compression downsizing.

Figure 4. In vitro ACL reconstruction simulation: a quadrupled peroneous longus allograft mounted within a porcine femur with the bone tunnel aligned to the loading axis of a servohydraulic Instron 8874 materials testing machine.

Figure 5. ACL graft geometric response – in vitro simulation: percentage change in quadrupled allograft cross-sectional area over a 12 hour period submerged in Ringer's solution at 37ºC under 20N of tension (Mean + SD; n=24; *=P < 0.05). Solid line: significant swelling of previously compressed grafts; interrupted line: non-significant change of CSA of uncompressed controls.

References

1. Adams AL, Schiff MA (2006) Childhood soccer injuries treated in US emergency departments. Acad Emergency Med 13:571-574
2. Ahn JH, Lee SH, Yoo JC, Ha HC (2007) Measurement of the graft angles for the anterior cruciate ligament reconstruction with transtibial technique using

postoperative magnetic resonance imaging in comparative study. Knee Surg, Sports Traumatol, Arthroscopy 15:1293-1300

3. Amis AA, Dawkins GP (1991) Functional anatomy of the anterior cruciate ligament. Fibre bundle actions related to ligament replacements and injuries. J Bone Joint Surg Br 73:260-267

4. Andersen HN, Amis AA (1994) Review on tension in the natural and reconstructed anterior cruciate ligament. Knee Surg Sports Traumatol Arthrosc 2:192-202

5. Anderson CJ, Westerhaus BD, Pietrini SD, Ziegler CG, Wijdicks CA, Johansen S, et al. (2010) Kinematic impact of anteromedial and posterolateral bundle graft fixation angles on double-bundle anterior cruciate ligament reconstructions. Am J Sports Med 38:1575-1583

6. Baumfeld JA, Diduch DR, Rubino LJ, Hart JA, Miller MD, Barr MS, et al. (2008) Tunnel widening following anterior cruciate ligament reconstruction using hamstring autograft: a comparison between double cross-pin and suspensory graft fixation. Knee Surg Sports Traumatol Arthrosc 16:1108-1113

7. Clatworthy MG, Annear P, Bulow JU, Bartlett RJ (1999) Tunnel widening in anterior cruciate ligament reconstruction: a prospective evaluation of hamstring and patella tendon grafts. Knee Surg Sports Traumatol Arthrosc 7:138-145

8. Cunningham R, West JR, Greis PE, Burks RT (2002) A survey of the tension applied to a doubled hamstring tendon graft for reconstruction of the anterior cruciate ligament. Arthroscopy: J of Arthroscopic & Rel Surg 18:983-988

9. Desai N, Björnsson H, Musahl V, Bhandari M, Petzold M, Fu FH, et al. (2014) Anatomic single-versus double-bundle ACL reconstruction: a meta-analysis. Knee Surg, Sports Traumatol, Arthroscopy 22:1009-1023

10. Goodship AE, Birch HL (2005) Cross sectional area measurement of tendon and ligament in vitro: a simple, rapid, non-destructive technique. J Biomech 38:605-608

11. Hoher J, Moller HD, Fu FH (1998) Bone tunnel enlargement after anterior cruciate ligament reconstruction: fact or fiction? Knee Surg Sports Traumatol Arthrosc 6:231-240
12. Hussein M, van Eck CF, Cretnik A, Dinevski D, Fu FH (2012) Prospective randomized clinical evaluation of conventional single-bundle, anatomic single-bundle, and anatomic double-bundle anterior cruciate ligament reconstruction: 281 cases with 3- to 5-year follow-up. Am J Sports Med 40:512-520
13. Hwang DH, Shetty GM, Kim JI, Kwon JH, Song JK, Munoz M, et al. (2013) Does press-fit technique reduce tunnel volume enlargement after anterior cruciate ligament reconstruction with autologous hamstring tendons? A prospective randomized computed tomography study. Arthroscopy 29:83-88
14. Kondo E, Merican AM, Yasuda K, Amis AA (2011) Biomechanical comparison of anatomic double-bundle, anatomic single-bundle, and nonanatomic single-bundle anterior cruciate ligament reconstructions. Am J Sports Med 39:279-288
15. Lee YS, Lee S-W, Nam SW, Oh WS, Sim JA, Kwak JH, et al. (2012) Analysis of tunnel widening after double-bundle ACL reconstruction. Knee Surg, Sports Traumatol, Arthroscopy 20:2243-2250
16. Lehmann A-K, Osada N, Zantop T, Raschke MJ, Petersen W (2009) Femoral bridge stability in double-bundle ACL reconstruction: impact of bridge width and different fixation techniques on the structural properties of the graft/femur complex. Arch Orthop Trauma Surg 129:1127-1132
17. Lind M, Menhert F, Pedersen AB (2009) The first results from the Danish ACL reconstruction registry: epidemiologic and 2 year follow-up results from 5,818 knee ligament reconstructions. Knee Surg Sports Traumatol Arthrosc 17:117-124
18. Lubowitz JH, Ahmad CS, Anderson K (2011) All-inside anterior cruciate ligament graft-link technique: second-generation, no-incision anterior cruciate ligament reconstruction. Arthroscopy 27:717-727

19. Maffulli N, Del Buono A (2013) Anterior cruciate ligament tears in children. Surgeon 11:59-62

20. Majewski M, Susanne H, Klaus S (2006) Epidemiology of athletic knee injuries: A 10-year study. The Knee 13:184-188

21. Morgan MD, Salmon LJ, Waller A, Roe JP, Pinczewski LA (2016) Fifteen-Year Survival of Endoscopic Anterior Cruciate Ligament Reconstruction in Patients Aged 18 Years and Younger. Am J Sports Med 44:384-392

22. Noyes F, Butler D, Grood E, Zernicke R, Hefzy M (1984) Biomechanical analysis of human ligament grafts used in knee-ligament repairs and reconstructions. J Bone Jjoint Surg Am 66:344-352

23. Paessler HH, Mastrokalos DS (2003) Anterior cruciate ligament reconstruction using semitendinosus and gracilis tendons, bone patellar tendon, or quadriceps tendon–graft with press-fit fixation without hardware: A new and innovative procedure. Orthop Clin N Am 34:49-64

24. Pearsall AW, Hollis JM, Russell GV, Scheer Z (2003) A biomechanical comparison of three lower extremity tendons for ligamentous reconstruction about the knee. Arthroscopy: J Arthroscopic Rel Surg 19:1091-1096

25. Palmer JE, Russell JP, Grieshober J, Iacangelo A, Ellison BA, Lease TD, Kim H, Henn RF, Hsieh AH (2017) A biomechanical comparison of allograft tendons for ligament reconstruction. Am J Sports Med 45: 701-707

26. Riemersa DJ, Schamhardt HC (1982) The cryo-jaw, a clamp designed for in vitro rheology studies of horse digital flexor tendons. J Biomech 15:619-620

27. Siebold R, Cafaltzis K (2010) Differentiation between intraoperative and postoperative bone tunnel widening and communication in double-bundle anterior cruciate ligament reconstruction: a prospective study. Arthroscopy 26:1066-1073

28. Silva A, Sampaio R, Pinto E (2010) Femoral tunnel enlargement after anatomic ACL reconstruction: a biological problem? Knee Surg, Sports Traumatol, Arthroscopy 18:1189-1194

29. Śmigielski R, Zdanowicz U, Drwięga M, Ciszek B, Ciszkowska-Łysoń B, Siebold R (2014) Ribbon like appearance of the midsubstance fibres of the anterior cruciate ligament close to its femoral insertion site: a cadaveric study including 111 knees. Knee Surg, Sports Traumatol, Arthroscopy 22:1-8

30. Wilson TW, Zafuta MP, Zobitz M (1999) A biomechanical analysis of matched bone-patellar tendon-bone and double-looped semitendinosus and gracilis tendon grafts. Am J Sports Med 27:202-207

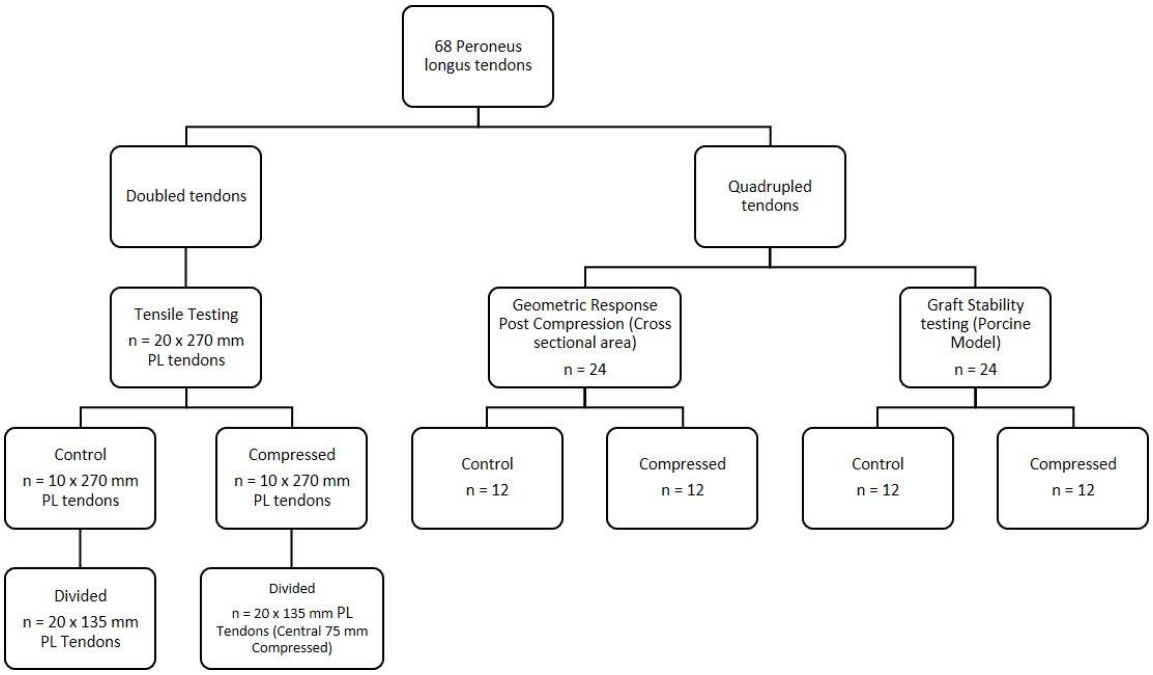
31. Wolf RS, Lemak LJ (2002) Revision anterior cruciate ligament reconstruction surgery. J South Orthop Assoc 11:25-32

32. Woo SL, Hollis JM, Adams DJ, Lyon RM, Takai S (1991) Tensile properties of the human femur-anterior cruciate ligament-tibia complex. The effects of specimen age and orientation. Am J Sports Med 19: 217-225.

Table 1: Biomechanical Data (Mean ± SD)

	Δ Length (mm) - 1,000 cycles	Cyclic Strain (%)	Ultimate Failure Load (N)	Stiffness (N/mm)	Ultimate Tensile Stress (MPa)	Young's Modulus (MPa)
Control	1.6 (± 1.1)	4 (± 3)	1629 (± 430)	329 (± 98)	82 (± 24)	671 (± 185)
Compressed	1.8 (± 1.0)	5 (± 2)	1664 (± 332)	343 (± 71)	93 (± 21)	764 (± 164)

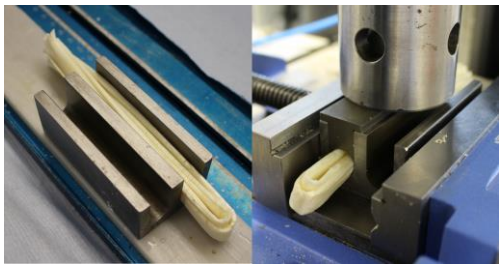
440 Fig 1



441

442

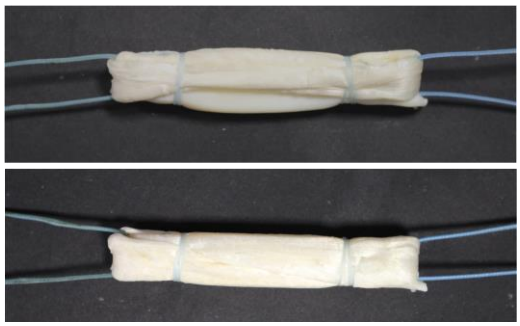
443 Fig 2



444

445

446 Fig 3



447

448

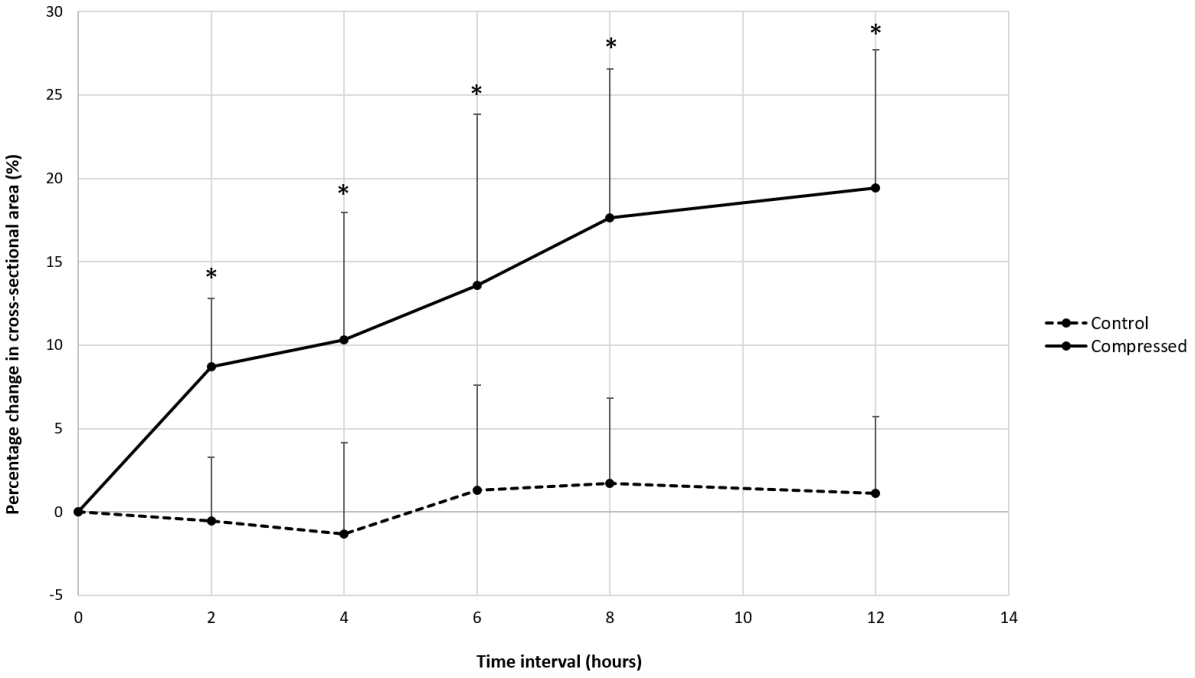
449 Fig 4



450

451

452 Fig 5



453

454

455 Table 1

456

Table 1: Biomechanical Data (Mean ± SD)

	Δ Length (mm) - 1,000 cycles	Cyclic Strain (%)	Ultimate Failure Load (N)	Stiffness (N/mm)	Ultimate Tensile Stress (MPa)	Young's Modulus (MPa)
Control	1.6 (± 1.1)	4 (± 3)	1629 (± 430)	329 (± 98)	82 (± 24)	671 (± 185)
Compressed	1.8 (± 1.0)	5 (± 2)	1664 (± 332)	343 (± 71)	93 (± 21)	764 (± 164)

457
458

459