All-Suture Anchor Onlay Fixation for Medial Patellofemoral Ligament Reconstruction

A Biomechanical Comparison of Fixation Constructs

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Background: The use of all-suture anchors (ASAs) for onlay patellar and femoral fixation of medial patellofemoral ligament (MPFL) grafts may provide clinical benefit, particularly in the small or pediatric knee; however, biomechanical data supporting the use of ASAs are lacking.

Purpose/Hypothesis: The purpose of this study was to compare ASAs to larger interference implants for MPFL reconstruction in a time-zero biomechanical model. It was hypothesized that ASAs would have comparable cyclic elongation to interference fixation and would exceed published biomechanical values for the native human MPFL.

Study Design: Controlled laboratory study.

Methods: Eighteen fresh-frozen porcine patellas and femurs were divided into equal groups (n = 9 per group) for MPFL reconstructions. Patellar fixation utilized two 3.9-mm interference suture anchors (ISAs) or two 2.6-mm ASAs, while femoral fixation utilized one 6×20-mm interference screw (IS) or one 2.6-mm ASA. Human gracilis tendon grafts were used. Specimens were dynamically loaded for 100 cycles each in sequential 5- to 30-N (phase 1) and 5- to 50-N (phase 2) blocks at 1 Hz followed by load-to-failure testing at 305 mm/min.

Results: No differences were found in cyclic elongation after phase 1 and phase 2 loading between ASA and interference implants on either the femoral or patellar side. On the femur, IS had significantly greater ultimate stiffness (54.2 vs 46.1 N/mm; $P < .001$) and ultimate load (366 vs 278 N; $P = .019$) compared to ASA. On the patella, ISAs had significantly greater ultimate stiffness (70.5 vs 53.1 N/mm; $P < .001$) but a significantly lower ultimate load (244 vs 307 N; $P = .014$) compared to ASAs. All groups significantly exceeded the published physiological values for native human MPFL stiffness and failure load.

Conclusion: ASA onlay fixation had comparable cyclic elongation to that of interference fixation for femoral and patellar MPFL reconstruction. Although differences in ultimate stiffness and ultimate load were noted between implants, all of the values exceeded published values for the human MPFL.

Clinical Relevance: This biomechanical study presents ASA cortical onlay fixation as a viable option for MPFL reconstruction. ASAs require less bone removal, potentially reducing the risk of patellar fracture and minimizing fixation complexity in the setting of open femoral growth plates. Future clinical studies will provide insight into successful tendon-to-bone healing, failure rates, and near- and long-term patient-reported outcomes.

Keywords: knee; ligaments; knee; patella; MPFL; patellar instability; all-suture anchor; onlay fixation

Lateral patellar instability is a common condition in young, athletic individuals, with a mean annual incidence of 6 per 100,000 individuals in the general population and 29 per 100,000 individuals in 10- to 17-year-old population.[10] Patellar instability is multifactorial and typically involves a combination of innate abnormalities of osseous and soft tissue restraints and acute injury to the medial soft tissue stabilizers, particularly the medial patellofemoral ligament (MPFL). In the presence of recurrent lateral patellar instability, surgical intervention is often

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required, with MPFL reconstruction being the most widely used technique in the absence of significant bony deformity or malrotation.[4,14]

Many different MPFL reconstruction techniques have been described with largely positive clinical outcomes, including failure rates typically <10%.[14] However, unique complications occur with this operation, and their incidence is reported to be as high as 32%.[14] The most common complications are stiffness with associated anterior knee pain and patellar fracture, the latter of which is reported in up to 8.3% of cases with the use of transosseous tunnels.[14] Another complication unique to the pediatric population undergoing MPFL reconstruction is iatrogenic physeal and articular injury during femoral fixation, given the significant anatomic constraints of the growing knee.[24] These complications can be attributed to surgical technique, in particular the use of large interference screw (IS) fixation devices, which have been shown to correlate with the risk of patellar fracture, physeal or articular injury, and patellar overconstraint due to stiffness mismatch in the face of femoral tunnel malposition.[14,24,32] Hard implants at the femoral fixation point can also be symptomatic.[33] As a result, interest has increased in the use of smaller, anchor-based fixation techniques in MPFL reconstruction[32] to reduce volumetric bone removal and create a less rigid construct that is more tolerant of small variations in graft isometry and positioning.

Small and versatile all-suture anchors (ASAs) have become prominent in orthopaedic shoulder surgery, with increasing interest in MPFL reconstruction to address the above-mentioned concerns with larger ISs. However, a paucity of biomechanical and clinical data are available for these anchors in the setting of MPFL reconstruction. A single biomechanical study evaluating patellar-sided fixation showed no significant differences between a 1.8-mm ASA and a 2.9-mm hard anchor.[30] Clinical studies using ASAs in MPFL reconstruction are limited to a handful of technique articles and case studies with short-term follow-up.[20,21,23] As a result, there is a need for further validation of ASAs for patellar fixation, as well as first-time validation of ASAs for femoral fixation of MPFL grafts, before clinical adoption can be recommended.

The purpose of this study was, therefore, to biomechanically compare ASAs to larger interference implants for MPFL reconstruction in a time-zero biomechanical model. We hypothesized that ASAs would have comparable cyclic elongation (ie, laxity) to interference fixation and would exceed published biomechanical values for the native human MPFL.

METHODS

Specimen Preparation

Eighteen fresh-frozen porcine stifles (Animal Biotech Industries) were disarticulated and dissected free of soft tissue to isolate the femurs and patellas. Two reconstruction techniques per anatomic location were investigated, totaling 4 groups and 36 constructs (9 per group). Porcine stifles were selected because of the similarity of bulk properties and anatomic landmarks to human knees and their prior use in published biomechanical studies of MPFL reconstruction.[2,6,15,22,25,39] The femurs were transversely cut 10 cm from the proximal aspect of the femoral condyles and the shafts were potted in fiberglass resin, whereas the patellas remained unaltered.

Human gracilis tendon allografts (LifeNet Health) with a diameter between 4.0 and 5.0 mm and a minimum length of 190 mm were used, consistent with common clinical practice,[5,27] and were randomly assigned to the isolated bones. Grafts were pretensioned at 20 N for 10 minutes on a graft preparation board to remove initial creep. All tissue was kept hydrated with 1X phosphate-buffered saline and frozen at −20°C until needed, whereafter the tissue was thawed at room temperature. Specimens were kept in 4°C refrigeration and acclimatized to room temperature for preparation and testing.

Femoral Fixation Techniques

Femoral fixation of the doubled graft was achieved using either a 6 × 20-mm IS (BioComposite FastThread; Arthrex) in an inlay “loop-in” fashion or a 2.6-mm ASA (Hybrid Knotless Knee FiberTak; Arthrex) in an onlay fashion (Figure 1). Implants were placed at the equivalent MPFL femoral origin, defined as the sulcus between the adductor tubercle and the medial epicondyle.[17] Grafts were sized and marked with a surgical marker to ensure a consistent 55-mm length between the femoral fixation point and the simulated patella, approximating the length of the native human MPFL.[3] Drilling and insertion were

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Ethical approval was not sought for the present study.

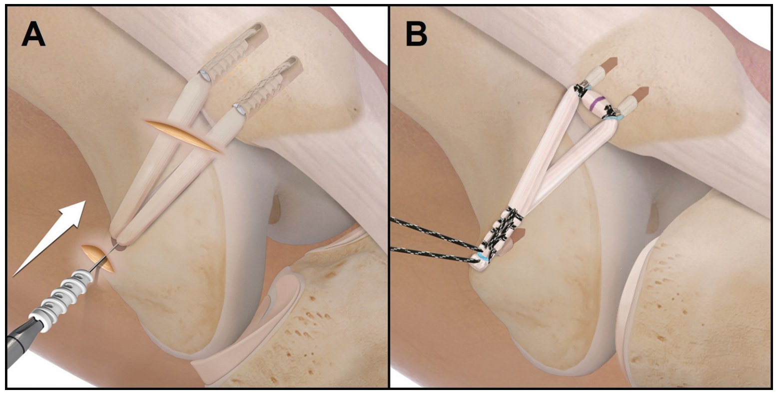

Figure 1. Investigated medial patellofemoral ligament reconstruction techniques using either (A) 2 parallel 3.9 × 17.9-mm biocomposite suture anchors in an interference fashion at the patella and a 6 × 20-mm biocomposite interference screw at the femur or (B) two 2.6-mm all-suture anchors (ASAs) at the patella and another ASA at the femur. ASAs contained a knotless suture loop mechanism (blue) and sliding suture tape (black) for graft incorporation. The white arrow depicts advancement of the interference screw. Medical illustrations provided by Arthrex.

angled slightly proximal and anterior to avoid the intercondylar notch.

For the 6 × 20-mm IS, the center of the doubled graft was pulled 25 mm into a 7-mm socket extending to the far cortex using a looped No. 2 suture. Light tension was applied to the graft while the cannulated screw was inserted over a nitinol guidewire until flush with the medial cortex (Figure 1A). The 2.6-mm ASA was implanted using the accompanying 2.6-mm drill, drill guide, and inserter. The ASA was double-loaded and contained a retensionable, preconverted, knotless suture tape loop and a sliding suture tape with swaged needles. The graft limbs were cinched by tightening the knotless loop and then stitched together using the sliding suture tape with 4 locking stitches in medial-lateral and lateral-medial directions (Figure 1B). The sliding suture tape ends were later tied superficially over the knotless loop with 5 alternating half-hitches after precycling and manual retensioning.

Patellar Fixation Techniques

Patellar fixation of the graft was achieved using either two 3.9 × 17.9-mm interference suture anchors (ISAs) (BioComposite SwiveLock; Arthrex) in an inlay fashion or two 2.6-mm ASAs (Knotless Hybrid FiberTak; Arthrex) in an onlay fashion, both spaced 15 mm apart at the superomedial aspect of the patella (Figure 1). Grafts were sized and marked with a surgical marker to ensure a consistent 55-mm length between the patellar and simulated femoral fixation points.[3] The distal 10 mm of the graft ends were whipstitched with 4 passes of looped 0.9-mm suture tape (SutureTape FiberLoop; Arthrex) in a staggered fashion to minimize the chances of vertical tear propagation.

The 3.9 × 17.9-mm ISAs were inserted 1 at a time into 4-mm reamed holes by passing the suture tails of 1 graft end through the eyelet, seating the graft and eyelet in the hole, and advancing the anchor until flush with the cortex using the provided inserter and moderate suture tension (Figure 1A). The 2.6-mm ASAs were inserted using the accompanying 2.6-mm drill, drill guide, and inserters. The center of the graft was fed through the adjacent preconverted suture tape loops and provisionally cinched. The sliding suture tapes were later tied superficially over the cinches with 5 alternating half-hitches after precycling and manual retensioning (Figure 1B).

Biomechanical Testing

Constructs were rehydrated with 1X phosphate-buffered saline and tested immediately after reconstruction on an electromechanical testing machine with a 1-kN load cell (ElectroPuls E10000; Instron). Three test setup configurations were used to simulate proper spacing and force vectors for the tested reconstructions (Figure 2). The femoral shafts were clamped in a potting holder affixed to the base of the machine, while either the graft tails were gripped 15 mm apart in a tissue clamp (IS group) (Figure 2A) or the central graft was looped over 2 hooks spaced 15 mm apart (ASA group) (Figure 2B) at the actuator. The femur was rotated such that the graft was tangential to its surface, and then the graft trajectory was aligned with the most anterior aspect of the trochlear articular margin to recreate the anatomic trajectory of the MPFL.[17] Patellas were held on the base of the machine in a custom holder containing 4 peripheral ultrasharp set screws and a central 2.4-mm drill hole for supplemental security with a 2.4-mm Steinmann pin, similar to other studies.[22,39] The graft tails (ASA group) or the looped graft (ISA group) were converged and rigidly gripped in a tissue clamp at the actuator to simulate femoral fixation (Figure 2C). The direction of pull was in line with the suture anchors, approximating the graft angle during lateral patellar translation as it bounds the medial condyle. This orientation simultaneously creates a worst-case scenario.[22,26,30,39]

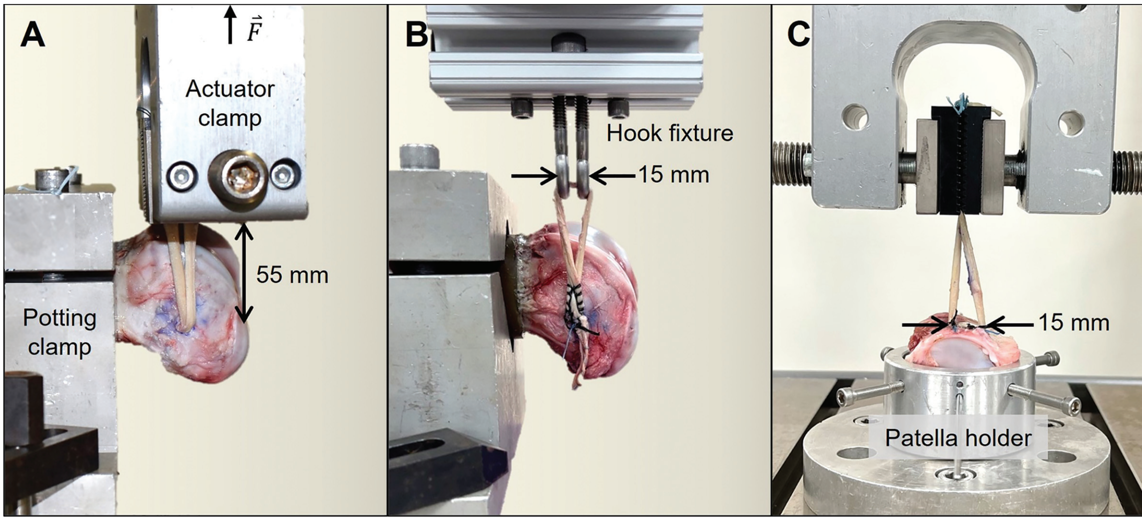

Figure 2. Three test setup configurations were used on an electromechanical testing machine to approximate anatomic pull angles for the techniques. Distal femurs were secured at the base while the lateral aspect of the graft was (A) gripped in a clamp for the interference screw group or (B) looped over hooks for the all-suture anchor group. (C) Patellas were held at the base with 4 sharp set screws and 1 bicortical 2.4-mm pin, while the medial graft was clamped for both interference suture anchor and all-suture anchor groups.

Specimens were preloaded with 5 N of tension and preconditioned from 5 to 15 N for 10 cycles at 1 Hz to reflect intraoperative knee cycling, and then the implant sutures were manually retensioned with final knot tying (ASA group only). Specimens were then dynamically loaded for 100 cycles each in sequential 5- to 30-N (phase 1) and 5- to 50-N (phase 2) blocks at 1 Hz, followed by load-to-failure testing at 305 mm/min. Retensioning of the ASA implant, a unique feature of knotless suture anchor technology, was performed using a handheld digital force gauge (Mark-10) from 50 to 75 N for patellar anchors and 100 to 125 N for femoral anchors to minimize any initial loosening during precycling. Retensioning values were standardized during piloting to reflect tactile feedback from the anchors while not incurring anchor pullout. The peak load levels were consistent with previous biomechanical studies of MPFL reconstruction and were hypothesized as physiological values based on the failure load of the native MPFL complex (upper 95% CI_{Pooled}, 174 N).[11,12,15,22,26] The 100-cycle count was selected because the majority of cyclic elongation occurs within this range, whereafter further increases in cycle count yield marginal increases in elongation (ie, steady-state behavior); this phenomenon was confirmed in pilot testing of the selected implants (data not shown).

Load-displacement data were continuously recorded at 200 Hz (WaveMatrix 2 software; Instron) and used to calculate total cyclic elongation (mm), ultimate stiffness (N/mm), and ultimate load (N). Total cyclic elongation quantified plastic elongation (ie, laxity) and was defined as the difference in displacement between the initial 5-N preload and the 5-N loads immediately after phase 1 and phase 2 cycle blocks. Ultimate stiffness was defined as the linear slope of the load-displacement curve in the 30- to 80-N elastic range. The ultimate stiffness and ultimate load were compared with the published data for native human MPFL complex stiffness (upper 95% CI_{Pooled}, 32.8 N/mm) and ultimate load (upper 95% CI_{Pooled}, 174 N).[12] The mode of failure was documented based on visual observation as graft pullout or anchor pullout.

Statistical Analysis

All statistical analyses were performed in SigmaPlot Version 14.0 (Systat Software) and Minitab Version 19 with α = .05 and β = 0.20. A previous a priori power analysis (analysis of variance; power, 0.8) by Johnston et al[15] was used to establish the sample size of 9 based on the detection of a 15% difference in failure load versus loop-in IS fixation (193 N).[29] Student t tests were used to compare parametric outcomes between the 2 different techniques at each anatomic reconstruction location. For nonparametric data, the Mann-Whitney rank sum test was used for data failing normality, whereas the Welch t test was used for data failing equal variance. One-sample t tests were used to compare individual group data for ultimate stiffness and ultimate load to those of the native MPFL complex. A 1-sample signed rank test was used for nonparametric data failing normality.

RESULTS

All constructs survived biphasic cyclic loading and completed load-to-failure testing for complete analysis (Table 1). No significant differences in total cyclic elongation were observed for femoral fixation with IS or ASA after phase 1 loading from 5 to 30 N (P = .368) and phase 2 loading from 5 to 50 N (P = .981) (Figure 3). The IS had

TABLE 1
Biomechanical Results of Testing Femoral and Patellar Fixation Techniques[a]

Anatomic Fixation	Implants	Total Cyclic Elongation, mm: Phase 1	Total Cyclic Elongation, mm: Phase 2	Ultimate Stiffness, N/mm	Ultimate Load, N
Femur	One 6 × 20-mm IS	0.82 (0.70-0.93)	1.58 (1.38-1.77)	54.2 (51.0-57.3)[b]	366 (333-400)[b]
	One 2.6-mm ASA	0.72 (0.52-0.93)	1.58 (1.32-1.83)	46.1 (42.9-49.3)	278 (207-348)
Patella	Two 3.9 × 17.9-mm ISAs	0.51 (0.44-0.59)	1.01 (0.89-1.13)	70.5 (64.3-76.7)[b]	244 (228-259)[b]
	Two 2.6-mm ASAs	0.42 (0.25-0.59)	1.03 (0.79-1.28)	53.1 (49.2-56.9)	307 (256-357)

[a]Data are reported as mean (95% CI). Phase 1 loading was 5 to 30 N, and phase 2 loading was 5 to 50 N. ASA, all-suture anchor; IS, interference screw; ISA, interference suture anchor.
[b]Significant difference ($P < .05$) observed versus the ASA within the same anatomic fixation.

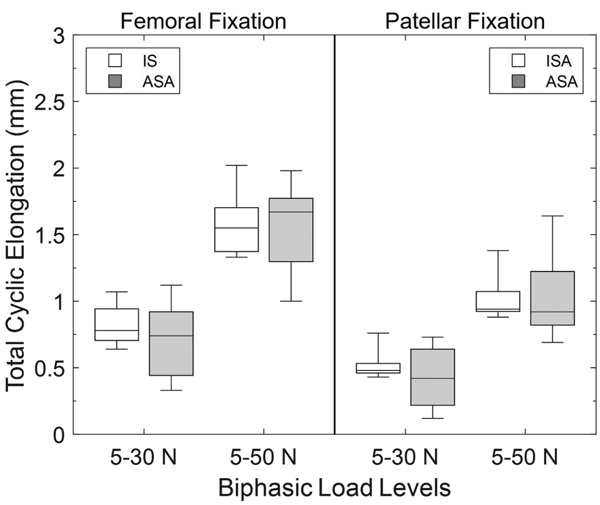

Figure 3. Total cyclic elongation measured after 100 cycles each of 5- to 30-N and 5- to 50-N loading. The box indicates the interquartile range, the central mark indicates the median, and the whiskers indicate the range. ASA, all-suture anchor; IS, interference screw; ISA, interference suture anchor.

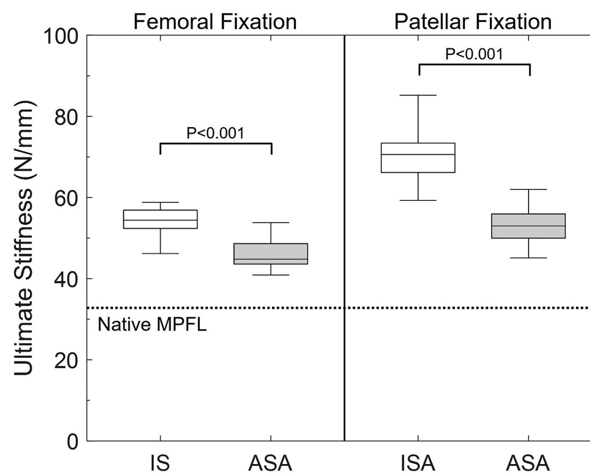

Figure 4. Ultimate stiffness of all test groups in relation to that of the native medial patellofemoral ligament (MPFL) complex (upper 95% CI_{Pooled}, 32.8 N/mm) reported by Huber et al.[12] The box indicates the interquartile range, the central mark indicates the median, and the whiskers indicate the range. ASA, all-suture anchor; IS, interference screw; ISA, interference suture anchor.

significantly greater ultimate stiffness ($P < .001$) (Figure 4) and ultimate load ($P = .019$) (Figure 5) compared to the ASA. Similarly, no significant differences in total cyclic elongation were observed for patellar fixation with ISA or ASA after phase 1 ($P = .245$) and phase 2 ($P = .596$) cyclic loading. The ISA had significantly greater ultimate stiffness ($P < .001$) (Figure 4) but a significantly lower ultimate load ($P = .014$) (Figure 5) compared to the ASA. Despite these observed differences, each group significantly exceeded the published physiological values of the native human MPFL complex[12] for stiffness ($P < .001$ for all) and failure load ($P = .004$ for femoral ASA and $P < .001$ for the rest).

All femoral IS constructs (9/9) failed by graft pullout from the socket, whereas the ASA constructs failed by either anchor pullout from the drill hole (6/9) or graft pullout from the cinched suture tape loop (3/9). All patellar ISA constructs failed by graft pullout from the drill holes (9/9), whereas all ASA constructs failed by anchor pullout (9/9).

DISCUSSION

The principal finding of the study was that ASA cortical onlay fixation had similar time-zero cyclic elongation to interference fixation for femoral and patellar MPFL reconstruction. Although the interference implants displayed greater stiffness (IS and ISA) and ultimate load to failure (IS only) in comparison to the ASA, all test groups significantly exceeded known physiologic values for human MPFL stiffness (upper 95% CI_{Pooled}, 32.8 N/mm) and ultimate load (upper 95% CI_{Pooled}, 174 N).[12] The findings suggest that the ASA technique is a biomechanically viable alternative to existing techniques, which is particularly relevant in situations where open physes (ie, in the pediatric population), patellar fracture (small patella), and articular surface violation risks are concerns.

The present study expands upon a previous biomechanical study in porcine stifles by Johnston et al[15] that

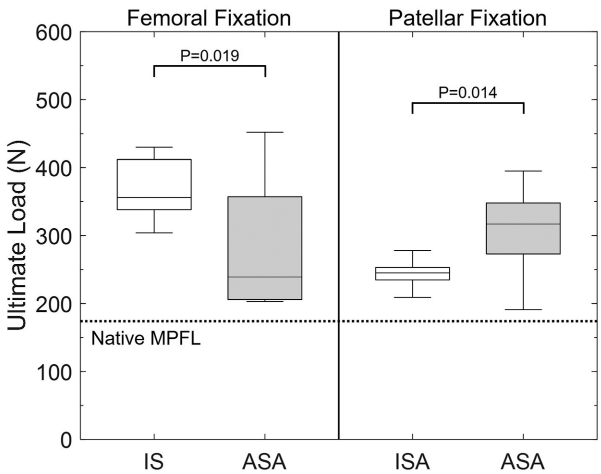

Figure 5. Ultimate load of all test groups in relation to that of the native medial patellofemoral ligament (MPFL) complex (upper 95% CI_{Pooled}, 174 N) reported by Huber et al.[12] The box indicates the interquartile range, the central mark indicates the median, and the whiskers indicate the range. ASA, all-suture anchor; IS, interference screw; ISA, interference suture anchor.

investigated 2 hard anchor onlay femoral fixation techniques for the pediatric population, namely a 4.5-mm biocomposite double-loaded threaded anchor with 2 knotted loops (DLA) and a 3.9-mm biocomposite threaded anchor with 1 knotless suture loop (KA). Johnston et al found that both anchor techniques had greater mean failure loads than the documented human MPFL (309 N for DLA, 250 N for KA) while having 3 to 6 times lower standard deviation than a 7 × 23-mm biocomposite tenodesis screw control, which can be used only in skeletally mature patients. After 100 cycles with 50-N peak loads, the DLA, KA, and control groups had approximately 2.4, 2.9, and 6.1 mm of elongation, respectively. In comparison, the present 2.6-mm ASA onlay technique also had a supraphysiological failure load (278 N) within the range of the KA and DLA.

In the current study, the authors compared the 2.6-mm ASA with a 6 × 20-mm IS (1 implant each) for femoral fixation and 3.9 × 17.9-mm ISAs (2 implants each) for patellar fixation to benchmark against current standard-of-care implants. Previous biomechanical studies have more extensively compared hard suture anchors to interference screws for femoral and patellar MPFL fixation.[||] A systematic review and meta-analysis by Sequeira et al[32] included a pooled analysis of these studies and found that the standardized mean differences favored IS over suture anchor for stiffness (P = .007), but not for ultimate load (P = .088), in patellar fixation. The opposite was true for femoral fixation, with IS favored over suture anchor for ultimate load (P = .043) but not stiffness (P = .507). The current study did not necessarily align with these pooled trends, as stiffness favored both IS and ISA over ASA, whereas ultimate load favored only IS over ASA for femoral fixation. These differences are likely a result of using different implant materials, geometry, size, and techniques.

For an MPFL reconstruction, it may be desirable to exceed the stiffness and ultimate load of the native ligament that previously failed in order to better resist lateral dislocation forces. Despite interference fixation having greater stiffness and ultimate load (femoral IS only) than ASA onlay fixation, all fixation techniques significantly exceeded published values for human MPFL stiffness and ultimate load. The ASA in the current study had noticeably greater variance for ultimate load compared with IS and ISA, an important consideration beyond a simple analysis of mean values. Although all the ranges exceeded physiological values, hard implants may have more reproducible failure mechanics; this is likely because the fixation mechanism of ASA is more sensitive to differences in bone quality than bulkier threaded, hard implants.

To our knowledge, only 1 previous ASA study exists, in which two 1.8-mm knotted ASAs were compared with 2 knotted 2.9-mm bioabsorbable hard suture anchors for onlay patellar fixation.[30] The stiffness of these anchors fell below native MPFL stiffness levels (21.3 and 20.9 N/mm, respectively), and only the ASA group exceeded the native MPFL failure load (229 and 156 N, respectively). In contrast, the 2.6-mm knotless hybrid ASAs in the current study had substantially greater stiffness (53.1 N/mm) and ultimate load (307 N) for patellar fixation. Both ASAs had similar variance for stiffness and ultimate load, which establishes a degree of consistency for ASA and may help predict performance in future studies.

MPFL reconstruction, with bony correction as necessary, is largely the standard of care for recurrent patellofemoral instability and dislocation in skeletally immature and adult patients.[7,13,18,27,37,38] In adolescents, the proximity of the MPFL insertion described by Schöttle[31] to the open physis of the medial distal femur has been shown to range from roughly 2.7 mm to 8.5 mm.[9,34] Consequently, the drill size, implant size, and drilling trajectory are all important considerations in reducing the risk of growth plate damage while avoiding the intercondylar notch during femoral fixation. Patellar fixation also warrants consideration based on patellar fracture and articular surface violation concerns in patients undergoing MPFL reconstruction.[21,30,33] In a systematic review of 28 studies, Jackson et al[14] found a patellar fracture risk of up to 8.3%, where all fractures occurred with a full-length transverse tunnel or 2-tunnel technique ranging from 4.5 mm to 6.0 mm in diameter. Improper tunnel placement and larger diameter drill bits may violate the articular surface, especially for smaller geometries. The 2.6-mm ASA technique in the current study may minimize the risk of physeal or articular violation and patellar complications because it requires less bone drilling than many larger commercial implants, using 2.6-mm drill holes of 25-mm length. However, further in vivo studies should confirm the absence of significant tunnel lysis, which could increase the risk of patellar fracture, and assess whether complications are reduced in practice.

Overconstraint of the patella is an additional concern after MPFL reconstruction based on the documented

[||]References 8, 11, 15, 16, 19, 22, 26, 28, 29.

frequency of postoperative knee stiffness.[14,33] The MPFL serves as a passive restraint to lateral patellar translation rather than bearing a constant tensile load. Overconstraint can occur when rigid femoral fixation is suboptimally positioned with respect to the isometric point or when the graft is overtensioned.[35,36] The 2.6-mm ASA in the current study may minimize the risk of overtensioning by allowing the surgeon to titrate the initial graft tension and retension as needed to achieve optimal patellar tracking during intraoperative knee cycling. In contrast, an IS may overgenerate tension upon insertion.[1] Regardless, clinical data are necessary to understand whether the ASA technique has less overconstraint than the IS technique.

Limitations

The present study has several limitations. This was a time-zero, cadaveric, biomechanical study that cannot recreate the in vivo effects of graft healing and dynamic muscle stabilization during rehabilitation that may protect the graft. The techniques were tested in a hemi-construct model where the femurs and patellas were isolated from each other for simplification and to reduce additional variables. Other secondary patellar stabilizers, such as the medial quadriceps tendon femoral ligament, medial patellotibial ligament, and medial patellomeniscal ligament, were not included. Different multiplanar knee movements (flexion/extension, rotation, etc) were not evaluated because the testing was uniplanar and favored extension. The orientation of loading simulated worst-case anatomic angles, particularly for the patella, and may not reflect true performance under various in vivo conditions. Porcine bone was used instead of human cadaveric bone due to anatomic and bulk property similarity to human bone.[2,6,25] However, porcine bone may not adequately reflect the full range of bone quality in humans, particularly poor bone quality. Other methods of fixation, including transosseous tunnels, were not investigated in this study. Larger implant sizes, such as a 7-mm IS, were not investigated but could further increase construct stiffness and failure load. Comparisons of biomechanical performance to the native human MPFL used published data for human knees, with the assumption that error sources and magnitude are consistent between studies.

CONCLUSION

ASA onlay fixation had comparable cyclic elongation to interference fixation for femoral and patellar MPFL reconstruction. Although differences in ultimate stiffness and ultimate load were noted between implants, all of the values exceeded published values for the human MPFL.

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