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Biomechanical Evaluation of Physeal-Sparing Fixation Methods in Tibial Eminence Fractures

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Background: Tibial eminence fractures occur most commonly in skeletally immature children. Several techniques using physeal-sparing fracture fixation have been described, but their structural properties have not been evaluated.

Purpose: To determine the strength and resistance to displacement of physeal-sparing techniques used to fix tibial eminence fractures.

Study Design: Controlled laboratory study.

Methods: Skeletally immature porcine knees were randomized into 4 treatment groups: (1) ultra–high molecular weight polyethylene suture–suture button (UHMWPE/SB), (2) suture anchor, (3) polydioxanone suture–suture button (PDS/SB), and (4) screw fixation. A prospective analysis of bone mineral density using dual-energy x-ray absorptiometry was performed on all specimens. Fracture fragments were created in a standardized manner and measured for size comparison. After fracture fixation, biomechanical testing was performed with cyclical and load-to-failure protocols by loading the tibia with an anterior shear force.

Results: In load-to-failure testing, screw fixation had a significantly lower median peak failure load (186.4 N; lower quartile [LQ], 158.4 N; upper quartile [UQ], 232.6 N) than did UHMWPE/SB (465.8 N; LQ, 397.8 N; UQ, 527.8 N), suture anchors (440.5 N; LQ, 323.0 N; UQ, 562.3 N), and PDS/SB (404.3 N; LQ, 385.9 N; UQ, 415.6 N). UHMWPE/SB demonstrated a significantly higher median yield load (465.8 N; LQ, 397.8 N; UQ, 527.8 N) than did PDS/SB (306.7 N; LQ, 271.4, N; UQ, 405.7 N) and screw fixation (179.0 N; LQ, 120.2 N; UQ, 232.5 N). During cyclical testing, screw fixation demonstrated significantly lower percentage survival of specimens (0%) compared with the other groups (UHMWPE/SB, 100%; suture anchor, 78%; PDS/SB, 78%). After 1000 cycles of loading, PDS/SB fixation had significantly more median creep (6.76 mm; LQ, 6.34 mm; UQ, 8.28 mm) than did UHMWPE/SB (4.43 mm; LQ, 3.80 mm; UQ, 4.73 mm) and suture anchor fixation (3.06 mm; LQ, 2.59 mm; UQ, 4.28 mm). The lowest median stiffness was observed in the PDS/SB group (48.6 N/mm; LQ, 45.3 N/mm; UQ, 54.2 N/mm). UHMWPE/SB fixation demonstrated a significantly higher median peak failure load after cyclic testing (469.0 N; LQ, 380.6 N; UQ, 507.2 N) than did PDS/SB (237.7 N; LQ, 197.3 N; UQ, 298.3 N) and screw fixation (132.4 N; LQ, 123.7 N; UQ, 180.9 N). Suture anchor fixation had significantly more variance, as demonstrated by width of interquartile range, in peak failure load, yield load, and creep than did other techniques.

Conclusion: Physeal-sparing fixation of tibial eminence fractures with UHMWPE suture–suture button is biomechanically superior to both PDS suture–suture button and a single screw at the time of surgery and provides more consistent fixation than do suture anchors.

Clinical Relevance: Suture anchors provide inconsistent fixation for tibial eminence fractures.

Keywords: tibial eminence; tibial spine; ACL avulsion; fracture; physeal sparing; biomechanics; skeletally immature

Tibial eminence fractures, first described by Poncet[40] in 1875, are bony avulsions of the anterior cruciate ligament (ACL) from its insertion on the intercondylar eminence.[37] These injuries most commonly occur in skeletally immature children during sports participation.[24,45] Tibial eminence fractures represent 2% to 5% of knee injuries in the pediatric population[11,28] and 14% of ACL injuries.[22]

The mechanism of injury is thought to be similar to intrasubstance ACL tears[1]; however, a biomechanical study on primates showed that these injuries are more likely to occur with slower loading rates compared with intrasubstance injuries.[37] In children, weakness of the incompletely ossified tibial plateau relative to the ACL results in an avulsion fracture, as a tensile load is applied.[53]

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Meyers and McKeever[34] classified these injuries into 3 types, according to the degree of fracture displacement. A fourth type of fracture, described by Zaricznyj,[55] represents comminution of the fragment. Some authors have recommended anatomic reduction and internal fixation for displaced fractures to restore the continuity of the femur-ACL-tibia complex (FATC) and prevent a mechanical block to knee extension.[19,24,45] Despite generally good results after operative fixation, complications have been observed. The use of strong internal fixation may allow for more aggressive rehabilitation to help prevent complications such as stiffness[50] and minimize the chance of malunion, nonunion, or pathologic laxity associated with loss of fracture reduction.

In skeletally immature children, avoiding the physis during fixation of tibial eminence fractures has been advocated to minimize the risk of iatrogenic growth disturbance.[#] Although physeal-sparing fixation techniques have been described in the literature, including the use of all-epiphyseal screws,[6,23] transepiphyseal sutures,[30] and all-epiphyseal suture anchors,[20,27,33,51] the best method of fixation has not been determined. Biomechanical studies have determined the fixation properties of non-physeal-sparing techniques**; however, there are no published studies comparing techniques for physeal-sparing fixation.

This study determined the strength and resistance to displacement of suture anchors, screw, and suture–suture button constructs using physeal-sparing fixation techniques under cyclic and load-to-failure conditions. We hypothesized that suture anchors would have less fracture displacement during cyclic loading and superior pullout strength during single cycle load-to-failure testing.

MATERIALS AND METHODS

Biomechanical Model

Skeletally immature (mean age, 17 ± 1 weeks), fresh-frozen porcine knees were used as the model for this study. Specimens were shipped overnight (Farm to Pharm, LLC, Warren, New Jersey) at −20°C and thawed for 24 hours at room temperature before preparation and testing. Soft tissues were removed from the knees, leaving the FATCs intact. An oscillating saw was used to create standardized fracture fragments with an inverted pyramidal shape of 20 mm (length), 20 mm (width), and 5 mm (height), based on average fracture size and morphological characteristics reported in the literature.[15] Fragment dimensions were measured with a caliper for comparison.

Bone Mineral Density Analysis

Bone mineral density (BMD) was obtained using dual-energy x-ray absorptiometry (DXA) (Hologic Discovery C QDR series, Hologic Inc, Bedford, Massachusetts), with the DXA machine set on fan beam mode by use of the subregion array spine. After thawing, each specimen was placed in a custom foam holder and positioned in a standardized manner. Specimens were individually scanned with the tibial epiphysis as the region of interest.

One specimen randomly selected from the knees undergoing the cyclic protocol and a second specimen randomly selected from the knees undergoing the load-to-failure protocol were then independently rescanned 10 times to determine the measurement error of the system. Researchers were blinded to the BMD (measured in grams per square centimeter) of specimens during fracture preparation and biomechanical testing.

Study Groups

A matched-pairs design was used to randomly assign 60 knee specimens to 4 treatment groups of 15 knees each. In each group, 6 knees were allocated to the load-to-failure and 9 to the cyclic loading protocols.

Group 1. Ultra-high-molecular-weight polyethylene suture–suture button (UHMWPE/SB): 2 strands of No. 2 FiberWire suture (total cross-sectional area, 0.39 mm^2) (Arthrex, Naples, Florida) tied over a 3.5-mm titanium suture button (Arthrex)

Group 2. Suture anchor: 2 Bio-Corkscrew FT suture anchors (4.5 × 15 mm) (Arthrex), each loaded with 2 strands of No. 2 FiberWire (Arthrex)

Group 3. Polydioxanone suture–suture button (PDS/SB): 4 No. 0 PDS II sutures (total cross-sectional area, 0.39 mm^2) (Ethicon Inc, Somerville, New Jersey), tied over a 3.5-mm titanium suture button (Arthrex)

Group 4. Screw fixation: 1 fully threaded cannulated AO screw (3.5 × 16 mm) (Synthes, West Chester, Pennsylvania)

Fixation Techniques

For suture fixation (groups 1 and 3), the physeal-sparing, transepiphyseal technique described by Mah et al[30] was

[#]References 2, 3, 6, 12, 17, 21, 29, 30, 36.

**References 4, 8, 10, 14, 16, 31, 43, 44, 49.

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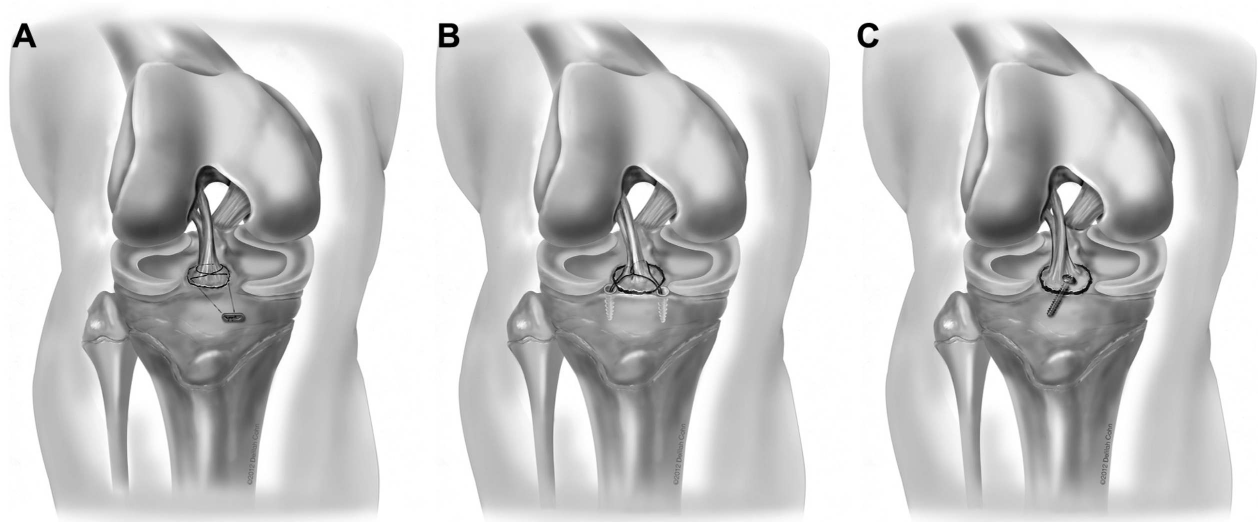

Figure 1. Physeal-sparing fixation techniques: (A) suture–suture button, (B) suture anchor, (C) screw.

used. Fracture fragments were manually reduced, and a 1.25-mm thread-tip guide wire was inserted into the anterolateral portion of the fragment for provisional fixation. The periosteum was dissected off the anteromedial aspect of the proximal tibia to determine the location of the growth plate. An ACL drill guide (Acufex, Smith & Nephew, Memphis, Tennessee) was used to make 2 parallel tunnels above the growth plate with a 2.4-mm drill-tip guide wire. Drill holes were started 1 cm apart and 1 cm distal to the articular surface of the medial tibial plateau and entered 1 cm apart on either side of the anteromedial bundle of the ACL. Sutures were inserted through the base of the ACL with a curved needle and pulled through the tunnels in a figure-of-8 fashion by use of a Hewson suture passer (Smith & Nephew). The sutures were sequentially tied over a suture button (Figure 1A).

For suture anchor fixation (group 2), we used the physeal-sparing, all-epiphyseal technique of Vega et al[51] and Lu et al.[27] Fracture fragments were manually reduced and provisionally secured with 1.25-mm thread-tip guide wire. Pilot holes for the anchors were created with a suture anchor punch (Arthrex) angled at 45° in the sagittal plane with respect to the tibial plateau. Holes were located 5 mm anterior to the fracture rim at the 5-o'clock and 7-o'clock positions. Suture anchors were placed, and sutures were passed through the base of the ACL with a free needle and tied down sequentially (Figure 1B).

For screw fixation (group 4), we used the all-epiphyseal technique of Berg[6] and Kocher et al.[23] After reduction of the bony fragment, a 1.25-mm thread-tip guide wire was placed in the central portion of the fragment at a 45° angle in the sagittal plane. A screw was then placed in lag fashion by first overreaming the guide wire with a 2.7-mm cannulated drill bit and then reaming the fracture fragment with a 3.5-mm drill bit. A cannulated 3.5-mm screw was then inserted and tightened over the guide wire until fixation was achieved (Figure 1C).

Biomechanical Testing

After fracture fixation, specimens were potted with plaster of paris (DryStone Casting Media, United States Gypsum, Chicago, Illinois) in a custom-made testing apparatus (Figure 2). The knees were placed in 60° of flexion to maximize in situ forces across the ACL during loading.[25] The testing apparatus was mounted to a material testing machine (858 Bionix, MTS Systems, Eden Prairie, Minnesota). Specimens were kept moist with a normal saline solution during preparation and testing. Tests were performed at room temperature. Graft fixation device–bone constructs were preconditioned with 100 cycles between 5 and 25 N at a rate of 60 cycles per minute. After cyclic preconditioning, a preload of 25 N was applied to each construct for 10 seconds before biomechanical testing. Specimens then underwent either a load-to-failure protocol, to simulate a traumatic incident,[7] or a cyclic loading protocol, which represented physiologic loading[48] that occurs in an early accelerated rehabilitation program.[47] Similar protocols have been previously used for biomechanical testing of tibial eminence fracture fixation.[10] After testing, specimens that underwent screw fixation were cross-sectioned with a band saw through the trajectory of the fixation construct to determine the proximity of the implant to the physis.

Load-to-Failure Protocol

After preconditioning, specimens were loaded to failure with an anterior tibial shear force at a rate of 200 mm/min. Load and elongation were recorded at an acquisition rate of 50 Hz. The resulting data were analyzed (MATLAB, Mathworks Inc, Natick, Massachusetts), and load-elongation curves were plotted. Primary outcome measures were peak failure load and yield load. Secondary outcome measures were stiffness and failure mode. Peak failure load was defined as maximal load observed on the load-elongation curve. Yield load

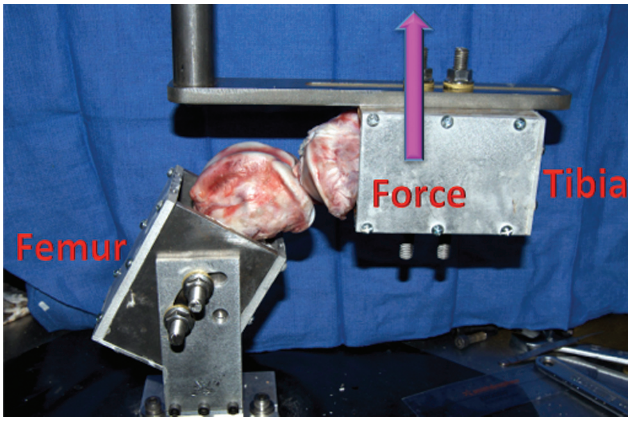

Figure 2. The test setup. Specimens were potted with plaster of paris in a custom-made testing apparatus with the knee at 60° of flexion to maximize in situ forces across the anterior cruciate ligament during loading.[25]

was the point at which the load-elongation curve is no longer accelerating linearly and represents the amount of force that irreversibly damages the fixation construct. Stiffness was defined as the slope of the linear region of the load elongation curve. Mode of failure was macroscopically documented.

Cyclic Loading Followed by Load-to-Failure Protocol

After preconditioning, fixation constructs underwent 1000 cycles between 25 and 200 N at a rate of 60 cycles per minute. The number of cycles was selected to simulate 1 week of rehabilitation with minimal biologic interaction.[7] Specimens that survived the 1000 cycles were loaded to failure at 200 mm/min. During testing, actuator displacement, cycle number, and loads were recorded at a rate of 50 Hz. Primary outcomes included creep and elongation. Creep represents the change in resting length of the FATC and was defined as the difference in anterior tibial translation with a 25-N load applied between the first and last cycle. Elongation represents the increase in anterior laxity of the FATC during physiologic loading and was defined as the change in length of the fixation construct with a 200-N force applied between the first and last cycle.

Secondary measures were average stiffness during cyclic loading, percentage of specimens in each group that survived cyclic loading without failure, peak failure load, and failure mode. Only the specimens that survived 1000 cycles were included in the analysis of creep, elongation, and average stiffness during cyclic loading. For specimens that did not survive cyclic loading, the maximum load observed during the failure cycle was used as the peak failure load in the final analysis.

Statistical Analysis

Sample sizes were determined before biomechanical testing. For load-to-failure testing, if the mean load of failure was 450, 400, 350, and 300 N for each study group, 6 knees per group gave us 80% power to detect this difference using a 1-way analysis of variance (ANOVA), assuming that the common standard deviation was 75 N and the type I error rate was 0.05. For cyclic testing, if the common standard deviation was 1.5 mm and the mean displacement for each treatment was 4, 3, 2, and 1 mm, 9 samples in each group gave us 95% power to detect this difference by use of ANOVA with the type I error rate 0.05.

The Kruskal-Wallis test was used to determine statistical differences in median BMD and avulsion fracture size between groups. The BMD difference and its 95% confidence interval between left and right knees of matched pairs were calculated and compared with measurement error of the system—determined by 10 repeated measurements of a randomly selected knee from both cyclic and load-to-failure protocols—to ensure that BMD was not different between paired knees. In addition, the Wilcoxon signed rank test was used for analysis of BMD and avulsion fracture size between right and left knees of matched pairs and for pairwise comparisons of the biomechanical outcomes between groups. The log rank test was used to determine differences in survival between groups during cyclic loading. A P value of $<.05$ was considered statistically significant for all tests. Median values with inner quartile ranges are reported for all parameters studied.

RESULTS

BMD, Eminence Fracture Size, and Proximity of Fixation to Growth Plate

No statistically significant differences were observed in the median BMD between groups in the load-to-failure ($P = .23$) (Table 1) or cyclic loading ($P = .81$) (Table 2) protocols. There were also no differences in fracture size between groups (Table 3) or between BMD or fracture size between right and left knees of matched pairs ($P > .05$). Observed differences of BMD between right and left knees of matched pairs were attributable to measurement error of the system. None of the screw fixation constructs violated the proximal tibial physis. Distances from suture anchors to the physis could not be accurately measured because of the degree of bony destruction during testing.

Load-to-Failure Testing

The differences in peak failure load between the UHMWPE/SB group (group 1), suture anchor (group 2), and PDS/SB (group 3) were not statistically significant ($P > .1$) (Figure 3). The UHMWPE/SB group had a statistically higher yield load than did the PDS/SB group ($P = .03$). There was no difference between the yield load of the UHMWPE/SB and suture anchor groups ($P = .24$) or the PDS/SB and suture anchor groups ($P = .59$). Screw fixation (group 4) had the lowest peak failure load (Figure 3) and yield load (Table 1) compared with groups 1, 2, and 3 (peak failure load, P = .002, .002, and .026; yield load, P = .002, .043, and .026 vs groups 1, 2, and 3, respectively).

The PDS/SB fixation group had a significantly lower stiffness than did the suture anchor group ($P = .009$).

TABLE 1
Results of Load-to-Failure Testing[a]

Measurement	UHMWPE/SB (Group 1) (n = 6)	Suture Anchor (Group 2) (n = 6)	PDS/SB (Group 3) (n = 6)	Screw (Group 4) (n = 6)
Peak failure load, N	465.8 (397.8, 527.8)	440.5 (323.0, 562.3)	404.3 (385.9, 415.6)	186.4 (158.4, 232.6)
Yield load, N	465.8 (397.8, 527.8)	367.2 (272.1, 417.1)	306.7 (271.4, 405.7)	179.0 (120.2, 232.5)
Stiffness, N/mm	25.5 (21.3, 31.8)	38.1 (25.1, 43.0)	17.4 (14.6, 19.0)	25.3 (22.2, 46.0)
BMD, g/cm^2	0.70 (0.68, 0.72)	0.68 (0.66, 0.75)	0.69 (0.68, 0.73)	0.78 (0.71, 0.79)

[a]Values are expressed as the median (lower quartile, upper quartile) for continuous variables. BMD, bone mineral density; PDS/SB, polydioxanone suture–suture button; UHMWPE/SB, ultra-high-molecular-weight polyethylene suture–suture button.

TABLE 2
Results of Cyclical Testing[a]

Measurement	UHMWPE/SB (Group 1) (n = 9)	Suture Anchor (Group 2) (n = 9)[b]	PDS/SB (Group 3) (n = 9)[b]	Screw (Group 4) (n = 9)[c]
Percentage survival for 1000 cycles	100	78	78	0
Elongation, mm	3.82 (2.77, 3.89)	2.28 (1.75, 3.53)	4.54 (3.96, 5.87)	—
Creep, mm	4.43 (3.80, 4.73)	3.06 (2.59, 4.28)	6.76 (6.34, 8.28)	—
Average stiffness during cyclic loading, N/mm	88.6 (82.8, 102.4)	111.5 (104.3, 119.9)	48.6 (45.3, 54.2)	—
Peak failure load, N[d]	469.0 (380.6, 507.2)	364.4 (199.9, 471.7)	237.7 (197.3, 298.3)	132.4 (123.7, 180.9)
BMD, g/cm^2	0.72 (0.66, 0.76)	0.70 (0.67, 0.76)	0.72 (0.68, 0.76)	0.68 (0.67, 0.75)

[a]Values are expressed as the median (lower quartile, upper quartile) for continuous variables. BMD, bone mineral density; PDS/SB, polydioxanone suture–suture button; UHMWPE/SB, ultra-high-molecular-weight polyethylene suture–suture button.
[b]Only the specimens that survived 1000 cycles (n = 7 for suture anchor and PDS/SB groups) were included in the analysis of creep, elongation, and average stiffness during cyclic loading.
[c]Evaluation of elongation, creep, and stiffness in screw fixation was not possible because no specimens survived cyclic loading.
[d]For specimens that did not survive cyclic loading, the maximum load observed during the failure cycle was used as the peak failure load in the final analysis.

TABLE 3
Tibial Eminence Fracture Size[a]

Measurement	UHMWPE/SB (Group 1)	Suture Anchor (Group 2)	PDS/SB (Group 3)	Screw (Group 4)	*P* Value
Load-to-failure protocol (n = 6 for each group)					
Length	20.48 (20.18, 20.89)	19.91 (19.4, 20.36)	20.16 (19.75, 20.36)	20.42 (20.1, 20.7)	.568
Width	20.39 (18.92, 21.07)	20.47 (19.61, 20.63)	20.13 (20.06, 21.1)	20.59 (19.42, 21.17)	.947
Height	5.39 (5.17, 5.63)	5.17 (5.13, 5.27)	5.33 (5.17, 5.6)	5.62 (5.36, 5.83)	.231
Cyclic protocol (n = 9 for each group)					
Length	20.68 (20.59, 20.74)	20.57 (20.54, 20.82)	20.78 (20.71, 20.84)	20.68 (20.55, 20.78)	.270
Width	20.84 (20.67, 21.04)	20.79 (20.62, 20.9)	20.92 (20.81, 20.99)	20.6 (20.43, 20.62)	.081
Height	5.72 (5.71, 5.74)	5.8 (5.71, 5.83)	5.66 (5.57, 5.73)	5.75 (5.61, 5.77)	.14

[a]Values for the 4 groups are expressed in millimeters as the median (lower quartile, upper quartile) for continuous variables. PDS/SB, polydioxanone suture–suture button; UHMWPE/SB, ultra-high-molecular-weight polyethylene suture–suture button.

Suture anchors had the highest stiffness, but no statistical differences were observed compared with UHMWPE/SB (P = .24) or screw fixation (P = .70). No statistical differences in stiffness were noted between the UHMWPE/SB, screw, and PDS/SB groups. Suture anchor fixation had significantly more variance in peak failure load (Figure 3) and yield load compared with groups 1, 3, and 4.

In the UHMWPE/SB group, the suture material and button construct remained intact in all 6 specimens and failed by pullout through the tibial epiphysis. Suture anchor failure modes included anchor breakage (n = 2), simultaneous breakage and pullout of the epiphysis (n = 3), and simultaneous breakage and ACL rupture (n = 1). In the PDS/SB group, 5 constructs failed by rupture of the suture and 1 failed by pullout through the epiphysis. In the screw fixation group, 1 specimen failed by screw pullout from the fracture fragment and the other 5 by screw pullout from the epiphysis.

Cyclic Loading Followed by Load-to-Failure Testing

Screw fixation had significantly lower survival (0%) than the other groups (P < .001). There were no statistical differences in survival between groups 1, 2, or 3 (UHMWPE/SB vs PDS/SB, P = .15; suture anchors vs

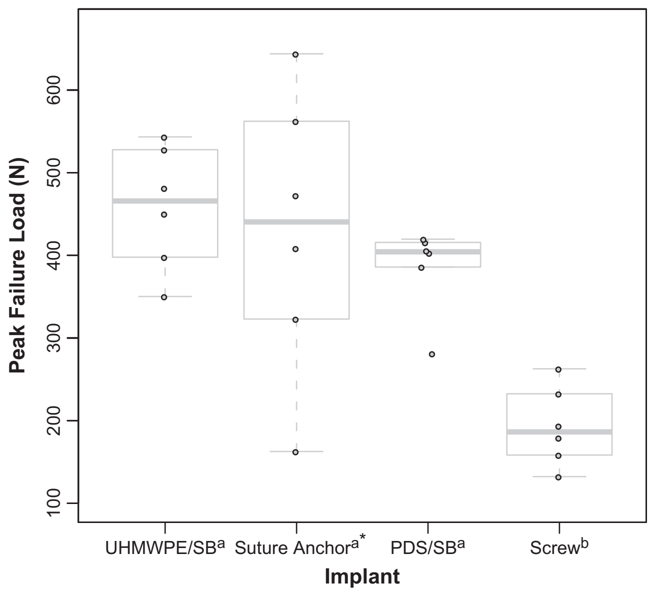

Figure 3. Peak failure load of implants during load-to-failure test. [a]No significant difference in median peak failure load ($P > .1$). *Significantly more variance compared with other groups. [b]Significantly lower median peak failure load compared with other groups ($P < .05$).

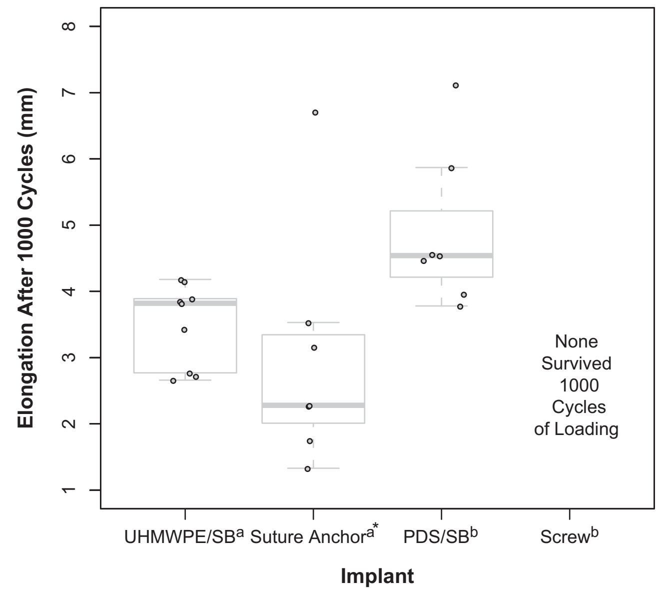

Figure 4. Elongation of specimens that survived 1000 cycles of loading (UHMWPE/SB, n = 9; suture anchor and PDS/SB, n = 7). [a]No significant difference in elongation ($P > .05$). *Significantly more variance compared with other groups. [b]Significantly more elongation compared with other groups ($P < .05$).

UHMWPE/SB, $P = .15$; PDS/SB vs suture anchors, $P = .90$) (Table 2).

After 1000 cycles of loading, PDS/SB fixation had significantly more elongation (Figure 4) and creep than did UHMWPE/SB ($P = .01$ and $< .001$, respectively) and suture anchors ($P = .02$ and .01, respectively). There was no difference in elongation (Figure 4) or creep between the suture anchor and UHMWPE/SB groups ($P = .11$ and .11, respectively). Suture anchors had higher average stiffness than did UHMWPE/SB ($P = .002$) and PDS/SB ($P < .001$). Average stiffness was higher with UHMWPE/SB fixation than with PDS/SB ($P < .001$). Evaluation of elongation, creep, and stiffness in screw fixation was not possible because no specimen survived cyclic loading.

In load-to-failure testing after cyclic loading, UHMWPE/SB and suture anchors had higher peak failure loads than did PDS/SB or screw fixation (UHMWPE/SB vs PDS/SB, $P = .001$; suture anchor vs PDS/SB, $P = .04$; UHMWPE/SB vs screw, $P < .001$; suture anchor vs screw, $P = .003$). PDS/SB had a higher peak failure load than screw fixation ($P = .006$). Suture anchor fixation had significantly more variance in elongation (Figure 4), creep, and peak failure load than the other groups, as measured by width of the interquartile range.

Failure mode in the UHMWPE/SB group was from pullout through the epiphysis in 8 specimens and suture rupture in 1 specimen. In the suture anchor group, 8 specimens failed because of simultaneous breakage and pullout of the anchor and 1 specimen because of simultaneous anchor breakage and ACL rupture. All specimens in the PDS/SB fixation group failed by suture rupture. In the screw group, 7 specimens failed by pullout of the screw from the epiphysis and 2 specimens by screw pullout from the eminence fragment.

DISCUSSION

The biomechanical integrity of implants used for repair of tibial eminence fractures is an important factor in optimizing fracture fixation. The results of this controlled laboratory study indicate that the security of the fragment at the time of surgery depends on the type of the implant used for fixation. In load-to-failure testing, screw fixation provided significantly lower pullout strength than did the other groups. In addition, although high survivorship (78%-100%) was observed with the other implants during repeated cycles of loading, all of the screw fixation constructs failed. After cyclic loading, both UHMWPE/SB and suture anchors had less displacement and higher peak failure loads than did PDS/SB. A comparison of UHMWPE/SB and suture anchor demonstrated that UHMWPE/SB had a significantly higher yield load than did PDS/SB in single cycle testing; this difference was not observed when suture anchors were compared with PDS/SB. Additionally, suture anchors had more variance in outcome measurements, suggesting less consistent fixation compared with UHMWPE/SB. Last, peak failure loads were decreased after cyclic testing compared with single-cycle loading in all groups except the UHMWPE/SB group. Therefore, the advantages of the UHMWPE/SB construct found in this study do not prove our original hypothesis that suture anchors provide more fracture stability than other types of fixation.

Although the results of this study indicate that screw fixation has inferior biomechanical properties compared with the other implants, clinically inferior results with metal screw fixation have not been observed in the literature. These clinical results may be in part attributable to management of patients with prolonged postoperative immobilization in an effort to prevent fixation failure. However, prolonged postoperative immobilization is known to increase the risk of arthrofibrosis and loss of knee motion, the most common complications associated with this procedure.[5,19,35,39,50,52] Early postoperative range of motion exercise has been shown to decrease the chance of knee stiffness and improve outcomes.[35,39] The strength of implant fixation, however, must exceed the forces applied to the ACL to prevent fracture displacement. Studies in adults have found in situ forces of 303 N[48] on the native ACL during activities performed in early rehabilitation. Although forces on the ACL are probably less in adolescents than in adults, implant fixation strength of 300 N may be necessary to prevent fixation failure during early postoperative rehabilitation. In the current study, only the UHMWPE/SB and suture anchor groups exceeded this threshold during a single destructive test and after cyclical loading. However, differences in bone morphological features, trabecular architecture, and mineral density between porcine and human knees[38] limit direct clinical translation of the results of this study to children. Even so, these data suggest that slower rehabilitation may be necessary when screw or PDS/SB fixation is used for tibial eminence fracture fixation.

The importance of implant strength is affirmed by the number of studies published on biomechanical evaluation of non-physeal-sparing techniques for tibial eminence fracture fixation.[††] Similar to the results of the current study, previous research has determined that suture fixation strength depends on the type of suture material used.[10,16] Furthermore, studies performed by Bong et al,[8] Sawyer et al,[43] and Eggers et al[10] demonstrated that FiberWire suture is superior to single screw fixation in biomechanical testing. The current study shows that FiberWire suture tied over a suture button also offers improved fixation compared with a single screw; however, in contrast to Sawyer and colleagues,[43] we did not find suture anchors to be structurally superior to FiberWire suture. Despite some methodological similarities, the differences in fixation techniques, biomechanical models, testing protocols, and implant types limit comparability between previous research and the current study.

Iatrogenic growth disturbance is another rare but serious complication of tibial eminence fracture fixation[3,12,36]; however, none of the previously published studies compared growth plate–sparing fixation techniques. Mylle et al[36] and Fabricant et al[12] reported cases of transphyseal screw fixation resulting in tethering of the growth plate and subsequent limb length discrepancy and angular deformity. Ahn and Yoo[3] reported a series of displaced fractures treated with PDS suture fixation through parallel tunnels crossing the tibial physis, with 2 cases of growth disturbance. The investigators recommended avoiding the physis by drilling in a transepiphyseal manner or by using the smallest possible guide wire with transphyseal drilling. Studies on animal models have concluded that the threshold for growth disturbance is destruction of 1% to 7% of the cross-sectional area of the physis.[18,32] The physeal area destroyed by drill holes during transphyseal suture fixation has not been quantified, and recommendations on appropriate guide wire size to avoid growth disturbance are lacking. Given this uncertainty, and the consequences of iatrogenic growth disturbance using transphyseal fixation, many researchers advocate the physeal-sparing techniques.[‡‡]

In addition to biomechanical differences, clinical factors also influence the choice of fixation technique for tibial eminence fractures. Screw fixation has the advantage of direct reduction and compression of the fracture fragment, but screw head impingement on the intercondylar notch can decrease terminal knee extension, which may necessitate screw removal after fracture healing.[19,23] If the fragment is small or comminuted, screw fixation may not allow adequate purchase of the fracture fragment, and suture anchors or sutures are recommended for stabilization. The PDS suture and Bio-Corkscrew FT suture anchors are bioabsorbable and obviate problems associated with residual hardware, but significant soft tissue reactions have been seen with biodegradable implants.[13] Suture anchors also do not provide a direct reduction of the fracture fragment but rather rely on a tension band effect from force applied to the ACL for bony compression. Nonetheless, healing times and functional outcomes appear comparable with those of other treatment techniques.[27]

The current study had some limitations. First, a porcine model was used for biomechanical testing because young human specimens could not be obtained. Nevertheless, the porcine knee is the most commonly used model for biomechanical studies evaluating fixation of ACL tendon grafts, given its similarities with the human knee,[54] and has been used by other studies evaluating tibial eminence fracture fixation.[4,10,14,43] Second, the ACL is known to experience both rotational and shear forces with normal movement.[42] Unidirectional biomechanical testing may not accurately reflect forces observed in vivo. Furthermore, removal of soft tissues surrounding the knee may alter forces normally experienced by the ACL. However, removal of soft tissue is necessary to create accurate and reproducible fracture fragments, and unidirectional testing is an accepted way to evaluate eminence fracture fixation.[††] Third, eminence fractures have been shown to have some variability in size and presence of comminution in the clinical setting.[15] The fractures created in this study do not reflect this variability, but fracture consistency

††References 4, 8, 10, 14, 16, 31, 43, 44, 49.

‡‡References 2, 6, 9, 17, 20, 21, 23, 26, 27, 30, 33, 41, 51, 55.

prevented heterogeneity between test specimens. Furthermore, although arthroscopy has become the most popular method of eminence fracture fixation, open fixation was used in the current study. It is unknown whether biomechanical differences exist between arthroscopic and open placement of implants in eminence fracture fixation. However, open fixation allows ideal and reproducible hardware positioning for testing and comparison purposes. Additionally, although the growth plate was avoided in both the screw and suture groups, damage to the suture anchor specimens during failure prevented accurate measurements of implant distance to the physis. Even so, implant sizes were selected based on epiphyseal dimensions reported in the literature[46] and are appropriate for avoiding the growth plate in skeletally immature patients.

CONCLUSION

Under cyclic and load-to-failure conditions, growth plate–sparing repair of tibial eminence fractures with UHMWPE/SB was biomechanically superior to both repair with PDS/SB and repair with screw and provided more consistent fixation than suture anchors.

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