Femoral Fixation Strength as a Function of Bone Plug Length in Anterior Cruciate Ligament Reconstruction Utilizing Interference Screws

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Abstract

Purpose To determine femoral construct fixation strength as bone plug length decreases in anterior cruciate ligament reconstruction (ACLR).

Methods Sixty fresh-frozen bone–patellar tendon–bone allografts were utilized and divided into 20-, 15-, and 10-mm length bone plug groups, subdivided further so that half utilized the patella side (P) for testing and half used the tibial side (T). Ten mm diameter recipient tunnels were created within the anatomic anterior cruciate ligament footprint of 60 cadaveric femurs. All bone plugs were 10 mm in diameter; grafts were fixed using a 7×23 mm metal interference screw. An Instron was used to determine the load to failure of each group. A one-way multivariate analysis of variance (MANOVA) was conducted to test the hypothesis that there would be one or more mean differences in fixation stability between 20- or 15-mm plug lengths (P or T) versus 10 mm T plug lengths when cross-compared, with no association between other P or T subgroups.

Results The mean load to failure of the 20 mm plugs (20 P + T) was 457 ± 66N, 15 mm plugs (15 P + T) was 437 ± 74N, and 10 mm plugs (10 P + T) was 407 ± 107N. There was no significant difference between P + T groups: 20-versus 15-mm ($p = 1.000$), 15-versus 10-mm ($p = 0.798$), and 20-versus 10-mm ($p = 0.200$); P + T MANOVA ($p = 0.291$). Within groups, there was no significant difference between patella and tibial bone plug subgroups with a pullout force range between 469 ± 56N and 374 ± 116N and p-value ranging from $p = 1.000$ for longer bone plugs to $p = 0.194$ for shorter bone plugs; P versus T MANOVA ($p = 0.113$).

Conclusion In this human time zero cadaver model, there was no significant difference in construct failure between 20-,15-, and 10-mm bone plugs when fixed with an interference screw within the femoral tunnel, although fixation strength did trend down when from 20- to 15- to 10-mm bone plugs.

Clinical Relevance There is a balance between optimal bone plug length on the femoral side for achieving adequate fixation as well as minimizing donor site morbidity and facilitating graft passage in ACLR. This study reveals utilizing shorter plugs with interference screw fixation is potentially acceptable on the femoral side if shorter plugs are harvested.

Keywords
- anterior cruciate ligament reconstruction
- bone–patellar tendon–bone
- interference screw fixation
- femoral tunnel

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The anterior cruciate ligament (ACL) is the most common surgically treated knee ligament treated by orthopaedic surgeons.[1] ACL tears and reconstruction rates have increased over the past few decades, with between 100,000 and 200,000 ACL injuries every year among athletes and incidence of ACL tears of 68.6 per 100,000 person-years.[2–4] Various ACL autograft options are available to surgeons, including bone–patellar tendon–bone (BPTB), multistrand hamstring (HT), or quadriceps tendon (QT) exist, with each possessing graft-specific advantages and disadvantages related to harvest and postoperative morbidity as well as ultimate failure rates.[5–8] Although there is no consensus for the ideal graft choice in anterior cruciate ligament reconstruction (ACLR), and there are champions for each grafts' merits such as the good biomechanical characteristics of HT and QT including load to failure, the BPTB has been extremely popular for young pivoting athletes and has even been coined the "gold standard" by some.[8–10] BPTB autograft has several advantages, including decreased graft retear rate compared with HT autografts, less residual laxity in some patients relative to other options, and bone-to-bone healing, although BPTB and QT autografts are comparable regarding revision ACLR.[9,11,12] Compared with quadruple HT autograft, the BPTB autograft achieves higher-level physical activity but not without increased postoperative complications such as patella fracture and patella maltracking.[13–15]

Although there are various options and preferences for bone plug fixation for BPTB grafts within tunnels, interference screws are popular since they typically provide reliable stable fixation to enable early mobilization and accelerated rehabilitation.[10,16] Historically, the standard bone plug length used in BPTB reconstruction is 20 mm or longer to provide adequate plug fixation and maximize osteointegration.[11,17–19] However, this standard, especially when procured from short-statured individuals, creates increased stress risers in the patella and may even complicate graft delivery into the femoral tunnel.[18]

Recent arthroscopy techniques suggest that many surgeons are utilizing shorter patella bone plugs (bone blocks of 15 mm lengths) in ACLR with BPTB.[20,21] Bone plug length has been shown to effect fixation stability in femoral and tibial porcine models, with < 10 mm plug lengths led to significantly reduced failure loads compared with plug lengths of 10- and 15-mm.[22,23] The potential benefits of harvesting shorter plugs, likely more significant on the patellar side, include minimizing the stress riser and potential donor site morbidity, especially in smaller-statured patients.[24] Additionally, as many surgeons have gone to independent drilling of the femoral tunnel (anteromedial portal method[20,23]), the femoral bone plug from the BPTB graft needs to be delivered around a more acute angle into the femoral tunnel. Longer bone plugs may make this technically more difficult. This technical point can often be more of a problem in smaller-statured patients with a reduced intercondylar notch width and distance between the tibial and femoral tunnel apertures. In the era of transtibial femoral tunnels, this was not as much of an issue since the trajectory of femoral tunnels was in line with the tibial tunnel.

Although many surgeons currently strive to harvest a 20-mm bone plug from the patella, especially in smaller-statured patients, it is not an uncommon clinical scenario for the plug to be inadvertently harvested shorter than intended. Consequently, of clinical concern is when is fixation compromised due to the shorter bone plug? At what point should the surgeon be concerned regarding femoral fixation and alternative strategies, such as flipping the graft (longer tibial bone plug placed within the femoral tunnel) or utilizing suspensory fixation? The purpose of this study was to investigate failure loads of femoral bone plugs fixed with interference screws as a function of bone plug length within a human cadaver model. We tested bone plug lengths of 20-, 15-, and 10-mm, all of which were 10 mm in diameter, and then we also looked if there were differences between patellar and tibial bone plugs. We hypothesized that fixation strength would decrease as bone plug length decreased, with a significant difference in stability between ≥15- and 10-mm plugs within the femoral tunnel.

Methods

This was a controlled laboratory study utilizing human cadaveric specimens. G*Power software (ver. 3.1.9.7; Heinrich-Heine-Universität Düsseldorf, Düsseldorf, Germany) was used to determine that a total of 54 specimens were needed to find a statistical difference. A nonhuman subjects' determination letter was obtained by the Institutional Review Board for the study. Sixty BPTB allografts (LifeNet Health, Virginia Beach, VA) were procured as bisected patellar tendons with bone plugs from the patella and tibia; a total of 10 bone plugs per group were tested. Bone density was not documented. Sixty embalmed human cadaver distal femurs were harvested from donors and prepared so they could be clamped within the Instron machine. All soft tissue was removed from the knees to enable adequate visualization of interference screw depth, to ensure zero screw divergence, and to allow the Instron fixation clamp to engage the femur. Femurs ($n = 10$) with evidence of previous operative treatment such as total knee arthroplasty or prior ACL tear with or without reconstruction were excluded; therefore, 70 total femurs were cleaned of their soft tissue and 60 femurs subsequently harvested. The age (79 ± 12 years), relative percentage of left versus right knees utilized (53% left, 47% right), sex (47% female, 53% male), and race (7% Black, 93% White) of donors were documented.

Bone–Patellar Tendon–Bone Graft and Femur Preparation

The grafts were divided into six groups. Groups 1, 2, and 3 were designated patellar 20P-, 15P-, and 10P-mm bone plugs; Groups 4, 5, and 6, tibial 20T-, 15T-, and 10T-mm bone plugs, respectively. The BPTB grafts were stored at -20° C in individually sealed plastic bags. The fresh frozen BPTB grafts were thawed prior to testing. All plug diameters were prepared with Rongeurs to fit snugly into a 10 mm sizer. A 10 mm diameter tunnel was placed within the center of the anatomic ACL footprint of all 60 femurs. A guide pin was initially placed, followed by a reaming with a 10 mm acorn reamer to replicate the surgical setting. A 2 mm "back wall" was present on all femurs.[25] Tunnel depth matched the bone plug length.

Each BPTB allograft was utilized once, with the patella and tibial bone plugs each utilized once within a femoral tunnel. The patella or tibial bone plug was fixed within the femoral tunnel with zero divergence by direct visualization and using a guidewire parallel to the bone block during interference screw insertion. Metal (steel) 7 × 23 mm interference screws (DePuy Mitek, Raynham, MA) were utilized for every group.[26] Interference screws were placed on the cancellous side of the bone plug to replicate the clinical setting and to prevent intratendinous failure under load.[27] Clamps with serrated grips at one end held the femur to the Instron 34SC-05 (Instron, Norwood, MA) testing machine (►**Fig. 1**). A different clamp at the opposite end was fastened to an uncut bone block, creating a force vector in line with the bone tunnel. The machine mechanically pulled the fixated bone plug to determine the peak linear load to failure at a strain rate of 51 cm/min, replicating the Pomeroy et al[22] study. After failure, each bone plug was inspected to ensure uniform cutting of the screw threads into the entire length of the bone plug (confirming no screw divergence occurred). Grafts were each examined to define failure modes including mid-tendon rupture, plug fracture, tendon–plug interface failure, and construct failure due to plug pullout.

Instron Data Collection and Statistical Analysis

The failure loads of each plug length were analyzed by SPSS Statistics for Windows, Version 28.0 (SPSS Inc, Chicago, IL). A cross-analysis was performed to statistically compare tension load to failure results of the patellar versus the tibial bone plugs. A one-way multivariate analysis of variance (MANOVA) was conducted to test the hypothesis that there would be one or more mean fixation stability difference between different length patellar and tibial bone plugs within the femoral tunnel, with *p*-value set to 0.05.

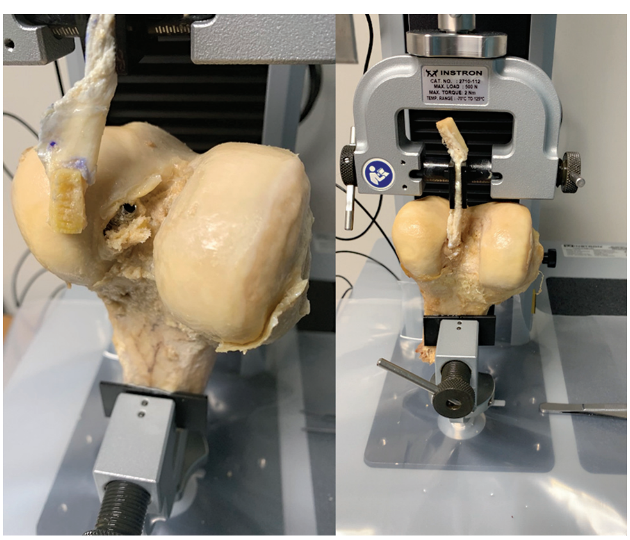

Fig. 1 Instron setup with femur fastened by clamp at one end and graft clamped at opposite end demonstrating P15 plug pullout at 469N (left) and tension loading of a different femur (right).

Results

The mean load to failure of the 20 mm plugs (20 P + T) was 457 ± 66N; the 15 mm plugs (15 P + T) was 437 ± 74N, and the 10 mm plugs (10 P + T) was 407 ± 107N. The MANOVA indicated no significant difference between P + T groups: 20-versus 15-mm ($p = 1.000$), 15-versus 10-mm ($p = 0.798$), and 20-versus 10-mm ($p = 0.200$); P + T MANOVA ($p = 0.291$). ►**Table 1** summarizes patellar and tibial combined bone plug lengths and fixation strength results.

The MANOVA indicated no significant difference in fixation stability as plug length decreased for the patellar groups P20 ($p = 1.000$) versus P15 ($p = 1.000$) versus P10 ($p = 1.000$). The tibial groups showed no statistical significance when cross-compared with each other T20 versus T15 versus T10 (T20 vs. T15, $p = 1.000$; T20 vs. T10, $p = 0.194$; T15 vs. T10, $p = 1.000$) but did show a trend that fixation strength decreased with shorter plug lengths based on trending down *p*-values (P20 vs. T10, $p = 0.924$; P15 vs. T10, $p = 0.452$). Patellar versus tibial comparisons showed no significance in fixation stability for all plug lengths, P versus T MANOVA ($p = 0.113$).

The longer patellar or tibial plugs and shorter patellar plugs demonstrated increased average pullout strengths (20P mean = 445 ± 76N; 20T mean = 469 ± 56N; 15P mean = 457 ± 58N; 15T mean = 418 ± 85N; P10 mean = 440 ± 91 N) and the shorter 10 mm tibial plug mean pullout strength was 374 ± 116N. The fixation strength results for patellar and tibial 20-, 15-, and 10-mm bone plug lengths are summarized in ►**Fig. 2**.

Each bone block and femur were inspected after pullout was achieved. In the 20 mm plugs, one failure occurred in the P20 and two in the T20 groups at the bone tendon interface due to fracture of the bone plug. Two mid-plug fractures occurred in the T20 group (Group 4) but none in the P20 group (Group 1). Two femoral condylar fractures occurred in the P20 plug group (►**Fig. 3** shows one condylar fracture). The remaining plugs (7 × P20 and 6 × T20) were pulled out at the corresponding maximum tension to failure value recorded. In the 15 mm group, one specimen failed at the plug screw interface via fracture in the P15 group. There was one bone tendon interface separation in each P15 and T15 group and 15 bone plug

Table 1 Bone plug length and fixation strength cross-comparison

Bone plug (patellar and tibial) length	Cross-comparison	*p*-Value
20 mm (Groups 1 and 4)	15 mm plugs (Groups 2 and 5)	1.000
	10 mm plugs (Groups 3 and 6)	0.200
15 mm (Groups 2 and 5)	20 mm plugs (Groups 1 and 4)	1.000
	10 mm plugs (Groups 3 and 6)	0.798
10 mm (Groups 3 and 6)	20 mm plugs (Groups 1 and 4)	0.200
	15 mm plugs (Groups 2 and 5)	0.798

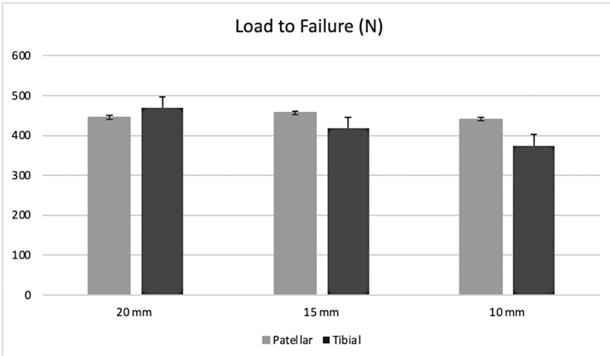

Fig. 2 Fixation strengths (N) of patellar and tibial bone plugs.

pullouts, seven in the P15 group, and eight in the T15 group. In the 10 mm plug groups, one P10 plug failed at the plug–screw interface due to a plug fracture. There were 9 patella plug pullouts and 10 tibia plug pullouts in the 10 mm groups. No mid-tendon ruptures occurred for any of the six groups. There was no observational evidence of screw divergence based on a uniform cutting of the cancellous bone by the interference screw in all 60 constructs (►**Fig. 4**).

Discussion

Our results show that femoral interference screw fixation stability was not significantly affected by decreasing bone plug lengths between 20- and 10-mm in an experimental cadaver model testing BPTB bone plug fixation on the femur only, although fixation strength did trend down when from 20- to 15- to 10-mm bone plugs. Additionally, there was no significant difference between failure loads when comparing the patella and the tibial side of the BPTB graft for 20-, 15-,

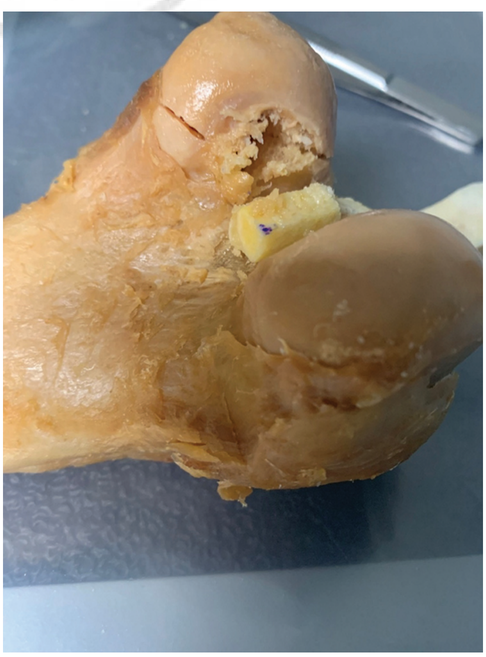

Fig. 3 Group 1 patellar plug fixated with condylar fracture at peak tension load to failure.

Fig. 4 Group 6 tibial 10 mm bone plug post tension load to failure demonstrating entire length of bone block was pulled out of the socket without fracture. Also, demonstrating zero divergence of screw based on uniform screw threads throughout the entirety of the tibia 10 mm bone plug (top). Patella side untrimmed.

and 10-mm bone plugs, which aligns with the null hypothesis that there was no difference between bone plug lengths in terms of fixation stability within the femoral tunnel, tibia, or patella plug side. As far as we know, patella versus tibia plug subgroup cross-analysis has not been reported in the literature (►**Table 2**).[22,23,25,27–31]

The results disprove our hypothesis, however, based on the shorter 10 mm tibia bone plugs achieving rigid fixation even though bone density of the tibia plug is generally softer than the patella side. Our hypothesis was based on the idea that tibia porosity would weaken plug stability, not render it resistant to tension loading, the latter of which appears to have been possible in our experiment.

Our human cadaveric study shows an average pullout strength of 434 ± 85N for all bone plug lengths between 20- and 10-mm, somewhat more than the Meuffel et al's human cadaver femoral construct (410 ± 171N), which utilized 20- and 10-mm plug length only.[31] The average pullout strength of our study is less than a study by Posner who utilized porcine bone. They found a pullout force of 658 ± 92N for 15 mm plugs and 540 ± 203N for 20 mm plugs compared with our 437 ± 74N for15 mm P + T plugs and 457 ± 66N for 20 mm P + T plugs.[23]

Variation in bone density and biomechanics between porcine and human bone based on quantitative computerized

Table 2 Femoral and tibial interference screw fixation construct stability: in vitro reports

First author (year)	Femoral or tibial tunnel	Porcine or human model	Bone plug length (side)	Interference screw	Fixation stability
Baydoun et al[25] (2021)	Femoral	Porcine	35 mm (patellar)	8 × 23 mm biocomposite and 12 × 30 mm biocomposite filler	626 ± 145N (8 mm screw); 653 ± 152N, and 720 ± 125N (12-mm filler screw)
Caborn et al[28] (1997)	Femoral	Human	25 mm (NR)	7 × 25 mm metallic and biocomposite	558 ± 68N (metallic) and 553 ± 56N (bioabsorbable)
Lee et al[29] (2003)	Femoral	Porcine	20 mm (patellar)	7 × 20 mm metallic and 7 × 20 mm biocomposite	691 ± 146N (metallic) and 707 ± 169N (biocomposite)
Marsh et al[27] (2018)	Tibial	Porcine	20 mm (NR)	9 × 25 mm metallic	493 ± 245N (cancellous surface of bone plug) and 304 ± 145N (cortical surface)
Meuffels et al[31] (2009)	Femoral	Human	20- and 10-mm	7 × 25 mm metallic	403N median (10 mm); 456N median (20 mm)
Pomeroy et al[22] (1998)	Tibial	Porcine	20-, 10-, and 5-mm (NR)	7 × 20 and 9 × 20 mm metallic	7 mm screw: <200N (5 mm plugs); < 400N (10 mm plug); < 350N (20 mm plug) 9 mm screw: 200N (5 mm plug); < 600N (10 mm plug); <700N (20 mm plug)
Posner et al[23] (2014)	Femoral	Porcine	20-, 15-, and 10-mm	7 × 20 mm metallic	Average all plug lengths = 573 ± 171N; 614 ± 110N (10 mm); 658 ± 92N (15 mm); 540 ± 203N (20 mm)
Rupp et al[30] (1999)	Femoral and tibial	Human	Unspecified bone plug lengths	Unspecified interference screw material and diameter	Passive extension peak of 128 ± 25N; quadriceps pull peak of 219 ± 25N; varus and valgus torque peak of 127 ± 23N and 98 ± 29N, respectively

NR, not reported.

tomography data reveal significant cross-sectional differences that likely affect graft fixation in ACLR.[32] Furthermore, the mean bone mineral content and bone mineral density (BMD) in humans compared with pigs in a study by Aerssens et al is 76.3 mg and 178 mg/cm^3 versus 173 mg and 373 mg/cm^3, respectively.[33] Lenz et al reported a significant BMD difference in porcine bone was observed compared with human non-osteoporotic ($p < 0.01$) and osteoporotic ($p < 0.01$) human bone; however, the difference between human nonosteoporotic and osteoporotic bone was not statistically significant.[34] Additionally, porcine bone is plexiform with characteristic osteonal banding, unlike human bone.[34] This bone density and architectural difference can potentially lead to much higher load failure values in ACL. Nurmi et al demonstrated porcine knee failure loads to be two times higher than human cadaveric tissues (however, tibia).[35] Thus, although our pullout strength may be compromised due to cadaver ages and bone quality,[36] we feel this may represent a more conservative evaluation of fixation stability relative to a porcine model. Although the tibial side is typically considered the "weak link" in terms of initial interference screw fixation in BPTB due to a decreased relative bone density compared with the femoral side as well as the vector of force being in line with the tibial tunnel, there were potential drawbacks of testing the tibia side.[32,37,38] We focused on the femoral side since from a clinical application standpoint many will have a low threshold to place backup fixation on the tibia if the fixation is felt to be compromised. This is not the case on the femoral side where although you can perform secondary backup fixation on the lateral cortex of the femur, this is relatively rarely performed compared with tibial-sided bone plug fixation. Lastly, previous studies showed no difference in biomechanics between fresh frozen and embalmed human femurs as used in our study.[39]

Rupp et al[30] previously looked at ACL graft forces in normal joint passive extension, quadriceps pull, and varus and valgus torque in a human cadaveric study using BPTB autograft and a complete lower extremity. The authors found a peak force in passive extension to range between 92 and 162N, quadriceps pull force to range between 175 and 247N, and varus and valgus forces to range between around 40 and 127N, utilizing a 10-mm diameter plug with unknown lengths and interference screw with an unspecified diameter and material (metallic vs. bio composite).[30] Our results indicate much higher failure loads of 445 ± 76N, 457 ± 59N, 441 ± 91N, 469 ± 56N, 418 ± 85N, and 374 ± 116N for Groups 1, 2, 3, 4, 5, and 6, respectively, which therefore exceed normal ACL forces of the knee joint as did Posner et al's[23] porcine model.

While several fixation methods may be utilized on the femoral side, interference screw fixation of the BPTB graft within the femoral tunnel has been the most widely used for decades in ACL surgery due to reliability and rigid fixation.[40] Interference screw fixation strength varies depending on several factors: screw divergence, bone quality, plug compression within the tunnel, screw–thread–length contact with bone, and bone plug length.[22,37] Interference screws have been proven to provide rigid and adequate fixation strength during early rehabilitation with standard 20+ mm bone plugs.[41] Metallic screw diameter was a nonsignificant variable in Pomeroy et al's study.[22] Both 7- and 9-mm metallic screw diameters did not significantly affect pullout strength in the tibial tunnel.[22] We used 7 mm screws on the femoral side, which is more clinically applicable, whereas 9 mm screws are typical for the tibial side. Some surgeons utilize 7- or 8-mm femoral screws. Additionally, the vector pullout angle in our study was in line with the femoral tunnel; clinically, it is not. Therefore, this is the worst-case scenario, including the use of cadaveric femurs, and clinical failure loads may be higher (unlike the tibial tunnel).

At least one study has reported clinical results utilizing shorter plugs in ACL with BPTB.[17] Tsuda et al had a "minimum-sized" plug group, P18.9 ± 2.3 and T17.0 ± 2.9 mm.[17] The authors found no fixation failure and added that too short a plug might compromise fixation stability; however, minimizing plug length at harvest facilitates packing the defect.[17] Additional studies highlight that too long a plug harvested for ACLR using BPTB causes donor site morbidity, and too small a plug may not achieve the requisite fixation strength for early rehabilitation.[14] However, we could not find recommendations for or against shorter plugs as they approach lengths of 15 mm or perhaps even shorter. The only biomechanics studies to test shorter plug lengths were in porcine models or lacked 15 mm plug lengths.

Our results support that shorter 15 mm patella bone plugs harvested in ACLR may achieve femoral rigid fixation using BPTB in vivo. Although BPTB in ACLR is very popular in young pivoting athletes, there are associated complications. These include anterior knee pain, patellar fracture, patellar tendon rupture, and patellofemoral crepitation related to donor site morbidity from BPTB graft harvest.[42–44] Donor site pain and difficulty kneeling have been documented in 40 to 60% of patients in ACL using BPTB autograft.[13,44,45] While complications following donor harvest at the patella and tibia may be reduced by bone grafting,[14] even small efficiencies gained potentially by minimizing donor plug length may be optimal in some cases,[11] especially in smaller statured individuals.

Conventional patellar bone harvest for BPTB used in ACL has been shown to significantly change patella axial strain, which may predispose to transpatellar fracture during submaximal quadriceps contraction.[46] These data are based on removing a rectangular bone plug from the inferior pole extending proximally to the horizontal midline of the patella and mostly rectangular in shape, with an average patellar defect height, width and base of 19.3-, 9.5-, and 5.7-mm, respectively.[46] Therefore, decreasing the length of the patella bone plug during harvest may mitigate patella fracture risk, albeit this is uncommon with a 0.1 to 3% complication rate, by preserving mechanical force distributions via minimizing the potential stress riser effect.[47–49] It has also been reported that disrupting the mid-patella and polar blood supplies, causing devascularization, may increase the risk of postoperative patella fracture following patella plug harvest.[47] There may be a clinical benefit if shorter bone plugs can achieve similar stability and healing. The option of a smaller diameter or triangular patellar plug are further considerations for reducing patella fracture risk, since length of defect is not the only variable to reduce fracture risk.

It is not uncommon to intentionally or inadvertently harvest a bone plug shorter than 20 mm during surgery, especially with a small patient. It can be of clinical concern how short surgeons can go with the bone plugs before fixation becomes compromised to the point that modification of the fixation technique is needed. Additionally, in the past two decades, surgeons have trended toward shorter femoral bone plugs since this facilitates passage in independently drilled femoral tunnels. Graft delivery with longer-length plugs may be more problematic for shorter-statured patients or when a narrow notch constricts graft passage. Schmidt-Wiethoff et al addressed the central problem in finding the "perfect size" bone plug, saying there should be a balance to optimize fixation stability while minimizing donor site morbidity.[50]

Limitations

In this human time zero cadaver biomechanical model, there was no cyclic loading tested, which would be important information to know from the clinical perspective. Furthermore, inferences on actual bone plug healing with shorter bone plugs, or the validity of using smaller bone plug length for ACLR cannot be made from the results of this study. Additionally, the Instron testing machine had a peak tension load to failure of 500N, and values achieved above this threshold were out of the instrument's calibration range. With an instrument calibrated up to 1,000N, there may be some values far greater than 500N, which theoretically could change the results; however, few failure loads exceeded 500N in our study. Therefore, a statistically significant association between incrementally shorter bone plug lengths and fixation stability is plausible, with the 10 mm tibial bone plug still being the most likely to achieve early failure based on the trends of our study. Some additional variables introduced in the study were the differences in cadaveric specimens and utilizing embalmed elderly cadaveric knees rather than younger, fresh-frozen knees. Younger knees in the clinical setting are expected to provide much higher fixation strength values based on increased BMD around the ACL femoral tunnel. Since the Instron pulled in line with the femoral tunnel, this theoretically reduces fixation stability as the graft is being pulled in line with the tunnel (similar to a tibial tunnel).[51] Thus, the clinical scenario when the graft is fixed around a more acute angle is likely to provide greater biomechanical stability. The plugs did not have suture holes to minimize variables, which could have created a stress riser and could cause earlier failure in some situations. Lastly, the authors could not obtain detailed demographics of the graft donors.

Conclusions

In this human time zero cadaver biomechanical model, there was no significant difference in construct failure between 20-,15-, and 10-mm bone plugs when fixed with an interference screw within the femoral tunnel, although fixation strength did trend down when from 20- to 15- to 10-mm bone plugs.

Ethical Approval
Institutional Review Board approval at Eastern Virginia Medical School was obtained for this study (21-04-NH-0119-EVMS).

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Conflict of Interest
None declared.

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