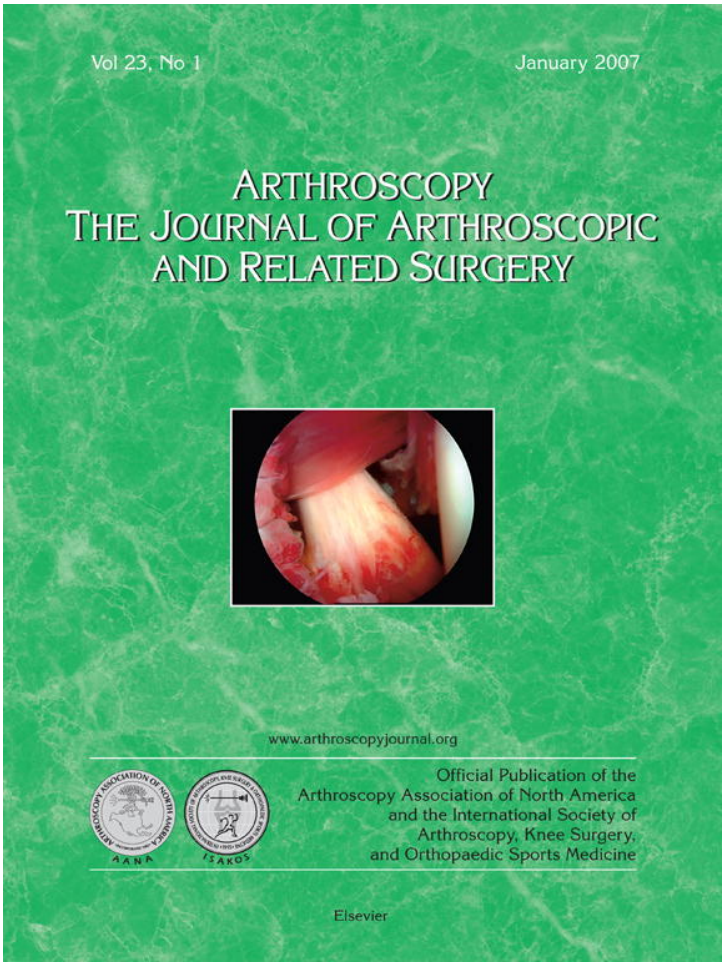

Current Concepts

Biomechanical Fixation in Arthroscopic Rotator Cuff Repair

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Abstract: Rotator cuff repair remains a challenging and rapidly evolving field. Several recent studies have shown that arthroscopic repair yields functional results similar to those of mini-open and open procedures, with all of the benefits of minimally invasive surgery. However, the "best" repair construct remains relatively unknown, with wide variations in surgeon preference and conflicting evidence in the literature. The most recent developments in basic science, suture and suture anchor technology, and innovative prospects for arthroscopic rotator cuff repair are reviewed. **Key Words:** Rotator cuff—Arthroscopic—Suture—Anchor—Biomechanics—Repair.

The goal of rotator cuff surgery is to optimize the connection between bone and soft tissue at the rotator cuff footprint. The rapid growth of arthroscopy for this purpose has been accompanied by equally rapid developments in suture and anchor technology. Over the last few years, a multitude of studies have investigated an array of sutures and anchors, as well as their respective configurations. In addition, there have been developments in basic science related to the rotator cuff to help us better understand the healing process. Newer techniques such as knotless fixation constructs and the biologic augmentation of repairs give surgeons additional choices in dealing with a range of pathologic conditions.

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The standard arthroscopic cuff repair is illustrated in Fig 1.[1] This construct must collectively withstand physiologic loads in the postoperative period while biologic healing takes place. However, it also contains several points of potential weakness. These include the stitch, the suture material, the knot, and the fixation between anchor and bone. Different studies have attempted to find the most optimal biomechanical construct to offset these potential areas of failure.

DEVELOPMENTS IN BASIC ANATOMY

Developments have also been made in defining the rotator cuff footprint. Dugas et al.[2] found that the footprint's minimum transverse diameter was 14.7 mm, occurring across the midpoint of the supraspinatus insertion. The collective insertion occurred over an area of 6.24 cm^2 on average. Curtis et al.[3] showed that (1) the subscapularis consistently inserted in a comma-shaped pattern from 7 to 11 o'clock around the lesser tuberosity adjacent to the biceps groove at the edge of the articular surface; (2) the supraspinatus had a trapezoidal footprint that filled the sulcus between the biceps groove and the bare area of the humerus; (3) the infraspinatus interdigitated around the posterior aspect of the supraspinatus tendon and also tapered into a trapezoidal footprint, which framed the bare area; and (4) the teres minor had a triangular

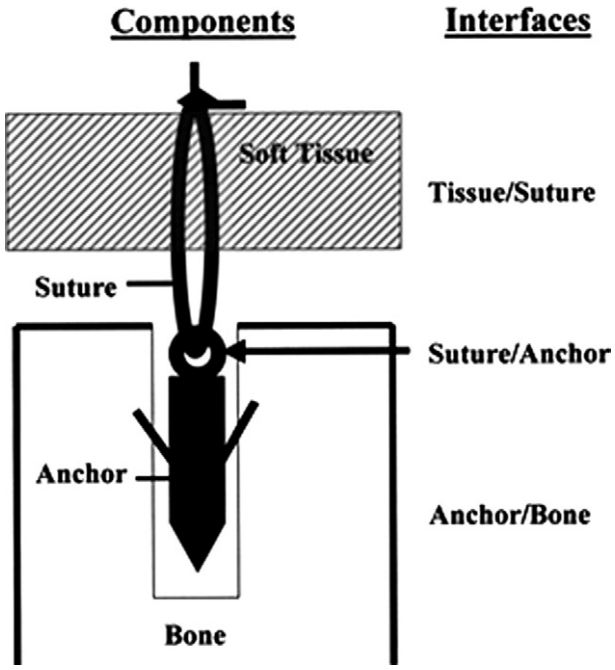

FIGURE 1. Standard construct for arthroscopic rotator cuff repair. (Reprinted with permission.[1])

insertion from 3 to 5 o'clock. Scanning electron microscopy of the supraspinatus footprint also revealed that the tendon densely adhered to the capsule and then inserted onto the very edge of the articular surface.

Recently, considerable work has gone into investigating vascular pathology at the rotator cuff because disturbances within the rotator cuff microvasculature have been hypothesized as a pathophysiologic mechanism contributing to degenerative cuff changes. Biberthaler et al.[4] used intraoperative orthogonal polarization spectral imaging to visualize the microcirculation within such diseased rotator cuffs. They found that the mean functional capillary density was 5 times lower in the diseased areas of the cuff compared with the nondiseased areas preoperatively. Fealy et al.[5] used power Doppler sonography to examine vascular changes postoperatively. They found significantly higher vascular activity at all time points compared with healthy control subjects. In addition, a significant and predictable decrease in vascular activity over a 6-month period was seen. The peritendinous region consistently had the most robust vascular response, whereas the decorticated trough had the lowest vascular response. This is surprising given that a bleeding bony bed and a vascular environment were thought necessary for sound tendon-to-bone healing.

DEVELOPMENTS IN BASIC BIOMECHANICS

The established biomechanics of the rotator cuff, as well as its repair, are less predictable with massive rotator cuff tears. These tears represent a challenge for the surgeon because the tendon is retracted and immobile, the healing response is poor, and the fixation biomechanics are poor. Burkhart et al.[6] have previously advocated using margin convergence to reduce strain at the tear site, as well as multiple stitches to distribute the load over several fixation points. The anterior interval slide, described by Tauro,[7] improves rotator cuff mobility by releasing the interval between the supraspinatus tendon and rotator interval. The technique incises the coracohumeral ligament near the coracoid base and frees up an additional 1 to 2 cm of supraspinatus tendon for lateral excursion. Although this amount of lateral excursion may suffice for adequate tendon-to-bone healing in a longitudinal massive tear, it is insufficient for adequate tendon-bone healing in a crescent-shaped tear. In such cases a double interval slide, anterior and posterior, is often necessary for satisfactory tendon-to-bone healing of both the supraspinatus and infraspinatus. As much as 5 cm of additional lateral mobility is possible with the double interval slide. The posterior interval slide improves the mobility of the infraspinatus tendon to such a degree that is usually sufficient to repair most or all of the infraspinatus to bone. When doing this, it is important to repair the inferior half of the infraspinatus to restore posterior force about the shoulder.

Other techniques have recently been described. Matis et al.[8] described an arthroscopic version of the open transosseous reinsertion procedure for use in massive, contracted rotator cuff lesions. Their case series showed that mean Constant scores improved from 56 preoperatively to 80 postoperatively at a mean of 26 months' follow-up. Atkinson et al.[9] attempted to bypass poor bone quality in elderly massive tears by using fixation posts inserted into cortical bone at the surgical neck. Their case series of 32 patients showed an impressive improvement in Constant scores from 27.9 to 78.2 at a minimum 2-year follow-up. Both of these techniques show promise in augmenting treatment for a difficult problem.

SUTURE FILAMENT AND MATERIAL

The ideal suture must remain sufficiently strong over time so as to keep the construct stable under the burden of any physiologic forces in the postoperative

period. The suture should be stiff enough to resist slipping but not so stiff as to cut through tendon or bone. In addition, the operative technique for placing the suture should ideally be both reliable and relatively simple to perform.

Previous studies have established that braided sutures tend to be superior to monofilaments.[10] Current tendencies also favor the use of permanent sutures over biodegradable sutures. In recent years there has been a shift away from the use of simple braided polyester sutures, such as No. 2 Ethibond (Ethicon, Somerville, NJ), toward hybrid sutures with a core of ultrahigh–molecular weight polyethylene (UHMWPE) surrounded by braided polyester. Several studies have compared No. 2 Ethibond with No. 2 FiberWire (Arthrex, Naples, FL),[10-14] the first of these new hybrids. A recent study has also investigated another interesting hybrid, Force Fiber (Stryker Endoscopy, San Jose, CA).[10,15] When such systems are tested, the core indices of the construct's mechanics are (1) ultimate tensile load (UTL), or the pure "strength" of the suture; (2) the number of cycles that a system can withstand under physiologic forces before failure, thought to be clinically more relevant[16,17]; and (3) mode of failure.

The studies unanimously agree that FiberWire has a significantly higher UTL than Ethibond (approximately 50% to 80%, $P < .05$ in all cases).[4,10,12-14] In the single Force Fiber study the hybrid was again around 70% stronger ($P < .001$). Lo et al.[18] showed that cycles to failure were 5 to 51 times greater ($P < .05$) for FiberWire compared with Ethibond under a variety of test conditions. De Carli et al.[12] examined failure mode and observed that FiberWire constructs tended to fail by anchor slippage or eyelet rupture whereas Ethibond constructs failed by suture breakage. They suggested that the reason for this shift was a transferring of the "weak link" in the newer hybrid systems from suture breakage to other parts of the construct. Although these findings are encouraging, there is a possibility that the strength of these sutures actually causes a predisposition for the suture to cut through the anchor eyelet or the tendon.

Barber et al.[19] recently investigated several other high-strength sutures, all containing UHMWPE. They found the following sutures to be statistically similar in tensile load to No. 2 FiberWire: No. 2 MagnumWire (ArthroCare [Sunnyvale, CA] and Axya Medical [Beverly, MA]), No. 2 Ultrabraid (Smith & Nephew Endoscopy, Andover, MA), No. 2 MaxBraid PE (Arthrotek, Warsaw, IN), and No. 2 Hi-Fi (Linvatec, Largo, FL). They also tested No. 2 Orthocord (DePuy Mitek, Raynham, MA), a new suture consisting of 38% high-strength polymer and 62% polydioxanone, a biodegradable material that dissolves completely in vivo within 9 weeks.[20] They found that the UTL for No. 2 Orthocord was around 50% lower than that for the other UHMWPE sutures but was still around 1.5 times stronger than No. 2 Ethibond. One can therefore conclude that surgeons have considerable choice with regard to suitable suture materials.

KNOT CONFIGURATIONS

The choice of knot configuration, more than other aspects of operative technique, is obfuscated considerably by variations in surgeon preference. Studies have shown that closing the loop in most knots with reversing half-stitches on alternating posts gives a highly satisfactory biomechanical and clinical outcome.[14] The improvement in cyclic loading appears to plateau after 3 reversing half-stitches on alternating posts.[21] However, the "best" knot to use in the first place remains a relatively open question. Like the ideal suture, the ideal knot needs to be consistent in its properties, with a high tensile strength.

Knot security is defined as the ability of a configuration to resist slippage as a force is applied. Uniquely of all FiberWire studies, Abbi et al.[11] found that, of the 40 No. 2 FiberWire knots that they tested against No. 2 Ethibond, 3 failed as a result of early slippage during cyclic loading and 8 failed at very low tensions during load to failure. Whereas the remaining intact FiberWire knots were significantly superior mechanically to Ethibond, no Ethibond knots exhibited slippage. Clearly, the possibility of slippage at low loads represents a weakness in a repair construct. However, without corroborating evidence, a definitive verdict on whether this is a true tendency in FiberWire systems is not possible.

STITCH CONFIGURATIONS

Open rotator cuff repair has often used a modified Mason-Allen (MMA) stitch because of its biomechanical and clinical efficacy.[17] However, the MMA stitch is difficult to perform arthroscopically, and simpler configurations with similar biomechanical properties have been sought. The massive cuff stitch is a combination of simple and horizontal stitches that has a UTL similar to that of an MMA suture (233 ± 40 N and 246 ± 40 N, respectively; $P < .05$) (Fig 2).[22] This is attractive not only because of its relative simplicity but also because its fundamental structural similarity

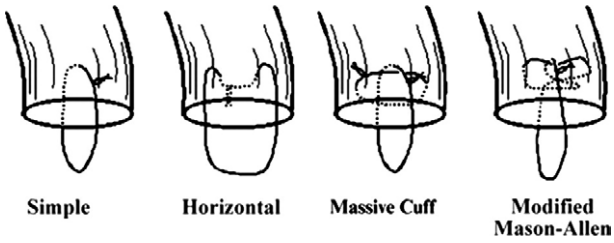

FIGURE 2. Diagram of different suture configurations. (Reprinted with permission.[22])

to the MMA suture. This study also found that both the massive cuff stitch and the MMA suture were superior to either a single simple stitch (72 ± 18 N) or horizontal stitch (77 ± 15 N) alone (Fig 2).

Koganti et al.[23] advocated the use of the locked mattress suture as an alternative to the MMA suture. They found that the mean cycles to 5-mm failure were significantly higher ($P < .001$) for locked mattress sutures (628) over locked inverted mattress sutures (197) and horizontal mattress sutures (193), followed by a single simple suture (65).

White et al.,[24] however, showed that the use of 4 simple stitches had a statistically similar UTL (155 ± 27 N) to the MMA stitch (140 ± 29 N), 2 mattress sutures (169 ± 56 N), and a single modified Kessler suture (161 ± 17 N). With stronger suture material, the "weak link" may be transferred to the stitch-tendon interface or the suture anchor eyelet, and a strong stitch configuration may be more important to prevent failure of the repair construct.

An important caveat with all of these studies remains: biomechanics do not necessarily equate with clinical studies. The properties of a system that are attractive in vitro, such as UTL, may have little relevance in vivo. Although clinical trials on this subject have generally been lacking, a recent randomized study found no statistical difference in outcomes between rotator cuff repairs via a No. 3 Ethibond with an MMA stitch and repairs with a 1.0-mm polydioxanone cord with a modified Kessler stitch.[25]

ANCHOR FIXATION: ANCHOR TYPES

The purpose of the suture anchor is to fix the suture, which itself is connected to the rotator cuff tendon, in close proximity to bone. This therefore represents a weakness at 2 major points—the interface between bone and anchor and the interface between suture and anchor. Consequently, the ideal anchor requires both an ability to withstand pullout during the physiologic loads of rehabilitation and an eyelet that protects against suture abrasion or breakage.

Most commercially available metallic anchors have a satisfactory pullout strength, although variations exist between different anchor types. Barber et al.[26] found that the pullout strengths for 11 commercially available anchors that they tested was always higher than the UTL of the sutures that they used. When examining variations between anchors, the study determined that screw-type anchors had a significantly higher failure load compared with non–screw-type anchors. De Carli et al.[12] used a human cadaveric shoulder model to compare 2 bioabsorbable anchors (Bio-Corkscrew, 5.0 and 6.5 mm; Arthrex) with a metal anchor (Corkscrew; Arthrex). Ultimate failure loads were statistically similar for all anchors. However, the bioabsorbable anchors tended to fail by eyelet rupture, whereas the metal anchors tended to fail by anchor slippage or suture breakage. Therefore the eyelet probably represents the mechanically weakest part of the newer bioabsorbable anchors.

These studies reflect immediate postoperative strength rather than strength over time. The progressive degradation of bioabsorbable materials correlates with a clinically relevant loss of mechanical strength. This could be an advantage, because a gradual reduction in support over time may favor healing, provided that the rate of reduction does not outpace the rate of the healing itself. Other notable advantages of bioabsorbable systems include no lasting foreign body in the patient, no obstructions if revision surgery is needed, and fewer artifacts in postoperative imaging studies.

Clearly, if strength loss occurs too quickly, the repair construct may fail. The degree of this loss in bioabsorbable anchors is dependent on the nature of the material used. Polyglycolic acid (PGA) polymers tend to degrade quickly over a period of months, whereas polylactic acid (PLA) polymers tend to degrade much more slowly, over a period of years. This theory is reflected in biomechanical studies. Demirhan et al.[27] showed that a pure PGA anchor retained only 75% strength at 12 weeks ($P < .001$). PLA anchors by contrast showed no significant change in strength over the same period both in vitro[28] and in vivo.[29] Commercially available bioabsorbable anchors are often PGA-PLA hybrids, such as the Panalok (DePuy Mitek) anchor. The manufacturer of this device claims that it retains 90% strength at 3 months after implantation. However, it is important to stress that no formal studies have been conducted examining the loss of strength of bioabsorbable anchors over time.

The unique eyelets in bioabsorbable anchors discussed previously, though mechanically weak, have favorable characteristics with regard to suture abrasion. Bardana et al.[30] reported a significantly higher degree of suture abrasion with metallic anchors compared with biodegradable anchors. This was attributed to the sharper edges found in metallic anchors leading to cutting of the suture. Rupp et al.[1] found that suture abrasion was the major mechanism of failure when testing cycles to failure whereas breakage at the knot was responsible when testing UTL. Given that cycles to failure may be a clinically more relevant index of repair strength, the reduced suture abrasion of the bioabsorbable anchors may be a favorable characteristic. However, improving bioabsorbable anchors further represents a balancing act for manufacturers. One must find a way to increase the mechanical strength of the eyelet so that it breaks at higher loads but not increase the chances of abrading sutures as they pass through it.

Another feature of anchors, like any implant, is their capacity to induce a local foreign body reaction. Chow and Gu[31] reported a case in which the local reaction to a metallic anchor led to bony erosion and tissue necrosis. Burkhart[32] reported a case series of 4 patients with foreign body reactions to bioabsorbable tacks, and Kelly et al.[33] reported a case of a reaction to a bioabsorbable suture anchor. On the basis of the amount of evidence in the literature, it appears that bioabsorbable materials and, in particular, those of the more bioactive PGA are at greater risk of inducing foreign body reactions than metallic anchors. However, specific incidence rates are unclear.

ANCHOR FIXATION: CONFIGURATION

Several aspects regarding how an anchor is placed could affect a repair–the angle and depth of anchor insertion, the anchor positioning on the humerus, and the number of anchors used.

When inserting an anchor, the surgeon must choose both the depth and angle that are most biomechanically suitable for a repair. Bynum et al.[34] tested 3 different anchor depths (superficial, standard, and deep) in a bovine rotator cuff model, with mixed results. Deep anchors had significantly higher UTLs than either superficial or standard anchors ($P < .04$) but took fewer cycles to reach 3-mm elongation, the consideration for clinical failure. In a human cadaveric model, Mahar et al.[35] examined failure at standard and deep depths and also found that deep anchors exhibited significantly greater displacement under cyclic loading than standard-depth anchors. Consequently, the insertion of anchors at depths deeper than the manufacturer's standard cannot be recommended.

With regard to the angle of anchor insertion, Burkhart[36] has advocated placing the anchor at around 45° (the "deadman" angle). In vitro results are equivocal, however, with Liporace et al.[37] and Deakin et al.[13] finding no difference in pullout load at various angles between 30° and 80°. However, Deakin et al. noted that suture angles of greater than 45° cause a predisposition to abrasion and breakage but exclusively in metallic anchors. The bioabsorbable BioCorkscrew anchor was relatively insensitive to angle by virtue of its polyaxial eyelet. Therefore, although no changes in protocol for the angle of anchor insertion can be made, it can be recommended that the suture be inserted between 0° and 45° for a stronger construct in metallic anchors.

Another factor in anchor configuration is whether the anchor has a single- or double-loaded suture. In both theory and practice double loading a suture anchor doubles the number of fixation points, consequently reducing the tension at each fixation point by approximately 50%.[38] This has been shown to lead to a more secure construct biomechanically and is now considered standard practice.[39] It remains to be seen whether the difference this imparts is still important given the improved strength of the new hybrid sutures.

Because anchors insert into bone, it seems reasonable to assume that the quality of bone will affect how well the anchor is secured. Two studies by Tingart et al.[40,41] found a positive correlation between bone mineral density (BMD) and pullout strength ($P < .01$) in a human cadaveric model. They found that anchor pullout loads were consequently 62% higher in the anterior and middle parts of the greater tuberosity than in the posterior part, 53% higher in the proximal part of the tuberosity than in the distal part, and 32% higher in the lesser tuberosity than in the greater tuberosity. These findings have since been corroborated.[42] Tingart et al.[40,41] also showed that in the areas with high BMD, screw-in metal anchors were equal in pullout load to biodegradable hook anchors, but in areas in which BMD was low, the screw-in anchors were significantly superior. From these studies, two recommendations can be made: (1) The anterior and middle parts of the greater tuberosity yield improved pullout anchor strengths, and (2) osteopenic patients will benefit from the use of screw-type metal anchors and a greater number of anchors than would otherwise be used.

Another important question addressed by several recent studies is the relative merit of single-row versus double-row repairs. In vitro anatomic studies suggest that a double-row repair produces a significantly larger supraspinatus footprint than single-row repairs.[43,44] In vitro biomechanical studies on human cadaveric shoulders are also in favor of a double-row repair. Kim et al.[45] showed that there was significantly less gap formation during cyclic testing for double-row repairs over single-row repairs ($P < .05$) and that the double-row repairs had a 46% higher UTL ($P < .05$). This was corroborated by Ma et al.,[42] who found that the mean UTL for double-row repairs (287 ± 24 N) was higher than that for any of the 3 single-row repairs tested (simple suture, 191 N; MMA, 212 N; and massive cuff, 250 N) ($P < .05$).

In vivo studies of UTL in a goat rotator cuff model showed no significant biomechanical differences between single-row and double-row repairs at 4 and 8 weeks.[46] Clinically, a cohort of 80 patients treated with either a single-row or double-row repair showed no significant differences in American Shoulder and Elbow Surgeons and University of California, Los Angeles scores at 2 years' follow-up.[47] However, postoperative magnetic resonance imaging (MRI) in the double-row repair patients showed superior structural results ($P < .01$) when a subjective grading score was used. Whereas double-row repairs yield superior anatomic and biomechanical results,[48] no data currently support the theory that double-row repairs result in better clinical outcomes than single-row repairs. However, it seems reasonable to recommend repairing larger tears, as well as tears in patients with low BMD, with a double-row technique to maximize the chances of rotator cuff healing.

Despite the good functional results reported previously, Park et al.[49] showed that even double-row repairs only had 50% of the contact area and 80% of the contact pressure of transosseous repairs. Hypothesizing that a larger footprint and higher pressures favored healing, they developed a "transosseous-equivalent" technique. This technique used suture bridges between anchors to add mechanical support to the repair construct, testing 2-bridge and 4-bridge structures against standard double-row repairs. They found that the 4-bridge repairs had 2 times the contact area and 1.4 times the contact pressure of double-row repairs, suggesting that they may indeed be "equivalent" in this regard to transosseous repairs, although no direct comparisons were made. The study also found that failure load was significantly higher (50%) for the suture bridge repairs but gap formation during cyclic testing was unaffected. Further testing is currently under way, but this technique clearly shows considerable promise.

BIOLOGIC AUGMENTATION

Several new techniques have been developed to augment the conventional rotator cuff repair, as summarized in Table 1 (online only, available at www.arthroscopyjournal.org). Because nonmassive tears have generally good outcomes, the risk-benefit consideration of biologic therapy dissuades many surgeons from such relatively experimental interventions. Massive tears by contrast often have poorer long-term outcomes, and biologic therapies may be particularly valuable in such cases.

Bravman et al.[50] recently reported on the placement of a biodegradable button to augment a suture anchor repair. A poly-L-lactid acid suture button (Arthrex) was placed on the bursal surface of the torn tendons before knot tying. In theory, this distributed the fixation tension over a wider area. The cycles to 7.5-cm gap were higher for the augmented group than for the standard repairs (420 cycles versus 135 cycles, $P < .05$). Given that several studies now point to the suture-tendon interface as a major point of weakness within a construct, such a result warrants consideration for further studies.

Several different materials have been proposed as grafts in massive tears. These include porcine small intestinal submucosa (PSIS) grafts, human/animal skin grafts, and muscle autografts/allografts, as well as several synthetic materials.

PSIS is available as the Restore (DePuy Orthopaedics, Warsaw, IN) and CuffPatch (Arthrotek) grafts. It has been used previously in dog models for the repair of ruptured Achilles tendons, with good new tendon formation and minimal residual tissue, adhesions, or chronic inflammation.[51] PSIS acts as a 3-dimensional scaffold attracting host cells and promoting regeneration. Several animal and human in vivo studies are now available to evaluate the role of PSIS in rotator cuff repair.

Barber et al.[52] recently published a comparison of the in vitro properties of 7 different grafts. Although this study represented the UTL of a repair, it clearly did not reflect any properties of the graft in augmenting healing, and thus its relevance is relatively limited. The authors found that the strongest repairs were with human skin (157 N, 182 N, or 229 N for GraftJacket [Wright Medical Technology, Arlington, TN] depending on the thickness used), followed by porcine skin

TABLE 2. *Clinical Studies of Arthroscopic Rotator Cuff Repair With Clinical and Radiographic Outcomes*

Study	Tear Size	Cohort Size	Suture Material	Stitch	Anchor Type	Anchor Configuration	Clinical Scoring	Preop Score (Mean)	Postop Score (Mean)	MRI/Ultrasound Findings
Galatz et al.,[60] 2004	All >2 cm	18 (all arthroscopic)	No. 2, nonabsorbable	Not specified	5-mm PLLA corkscrew (Arthrex)	2-5 anchors used, variable configuration	ASES	48.3	84.6	94% showed retear on US
Boileau et al.,[61] 2005	97% stage I/II	65 (all arthroscopic)	No. 1 PDA	Inverted horizontal mattress	Panalok RC (DePuy Mitek)	2-4 anchors used, tension-band suture technique	Constant	51.6	83.8	29% showed retear on MRI
Verma et al.,[62] 2006	Mean size, 2.0 cm for arthroscopic group and 2.8 cm for mini-open group	38 arthroscopic and 33 mini-open	No consistent style	No consistent style	No consistent style	No consistent style	ASES	Not specified	94.6 for arthroscopic and 95.1 for mini-open	24% showed retear on US in arthroscopic group; 27% in mini-open group
Lichtenberg et al.,[63] 2006	Not specified	53 (all arthroscopic)	Not specified	MMA	Bioabsorbable	Single row, 1-3 anchors	Constant	53.6	86.1	24.5% retear on MRI

(128 N for Permacol [Tissue Science Laboratories, Covington, GA; licensed to Zimmer, Warsaw, IN]), bovine skin (76 N for TissueMend [TEI Biosciences, Boston, MA; licensed to Stryker Howmedica Osteonics, Kalamazoo, MI]), and lastly, PSIS (38 N for Restore and 32 N for CuffPatch).

DeJardin et al.[53] repaired infraspinatus defects in a canine model with and without PSIS grafts. They reported no gross and histologic differences at 6 months after operation, and although the reconstructed tendons were weaker than the preoperative tendons ($P < .001$), the PSIS repairs were no weaker than the tendon-reinsertion sham operation. Zalavras et al.[54] found that at 16 weeks after rotator cuff repair in a rat model, the rats with PSIS had a UTL 78% of normal compared with 36% of normal in the unaugmented repairs ($P < .008$). However, in a sheep model Schlegel et al.[55] found no significant differences in UTL between PSIS and unaugmented repairs.

Concerns remain about any xenograft, and PSIS is no exception. Zheng et al.[56] reported that the Restore acellular graft still contained porcine deoxyribonucleic acid and thus recipients were exposed to the theoretic risk of xenograft retroviruses and immunologic rejection at the graft site. Malcarney et al.[57] reported a series of 25 massive cuff tear patients treated with PSIS grafts (Restore), in which 4 patients had an early (mean, 13 days), nonspecific inflammatory reaction at the graft site requiring a second operation for debridement and graft removal. They did not confirm the cause of the inflammation, although infection and graft rejection are certainly possibilities.

Two other clinical series are of note. Sclamberg et al.[58] conducted a retrospective review of 11 patients with MRI and clinical follow-up at 6 months. They found no significant improvement in American Shoulder and Elbow Surgeons scores, and MRI showed retears in 10 of 11 patients. Iannotti et al.[59] recently published a clinical trial of 30 patients divided evenly into PSIS-augmented and PSIS-unaugmented groups. Contrary to their hopes, they found that 9 of 15 unaugmented repairs showed healing compared with 4 of 15 augmented repairs ($P = .11$). The PENN score was 83 postoperatively in the PSIS group compared with 91 in the control group ($P = .07$). On the basis of these data, it does not appear as if PSIS currently has a role in rotator cuff repair.

CONCLUSIONS

The orthopaedic surgery literature is inundated with new devices and techniques for arthroscopic rotator

cuff repair. The biomechanics of the rotator cuff repair should be optimized to increase the likelihood of tendon-to-bone healing. The biomechanical construct of the repair of a torn rotator cuff can be broken down into 3 potential areas of failure: tissue-suture interface, suture-anchor interface, and anchor-bone interface.[1] With the development of reliable suture anchors and synthetic hybrid suture materials, the "weak link" has been shifted to the tissue-suture interface. Studies have shown that small rotator cuff tears have a higher rate of healing compared with larger tears, which may reflect further tendon degeneration (Table 2).[60-63] Although the quality of the tissue cannot be altered, surgical repair should be performed when the tear is small, while the tendon is amenable to repair. Particularly with the synthetic hybrid suture materials, a more robust stitch configuration is critical to prevent tendon pullout and diminish the possibility of failure. At present, the correlation between biomechanical strength and clinical failure is not yet known, and in vitro studies may not necessarily apply to conditions in vivo. Until additional studies are performed to clearly define the mechanical strength of repair that is required for biologic healing, surgeons should aim to produce the strongest possible biomechanical repair construct.

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TABLE 1. *Summary of Biomechanical Strengths of Rotator Cuff Repair Constructs*

Component	Study	Relative Biomechanical Strength	Control Group (n)	Experimental Group (n)
Suture material		Load to failure (ratio)	Ethibond	Test
No. 2 Ethibond (control)		1.00		
No. 2 ForceFiber	Mahar et al.,[15] 2006	1.72	148	254
No. 2 FiberWire	Barber et al.,[19] 2006	2.04	92	188
No. 5 Ethibond	Barber et al.,[19] 2006	2.10	92	193
No. 5 FiberWire	Barber et al.,[19] 2006	5.25	92	483
Stitch	Ma et al.,[22] 2004		Simple	Test
1 Simple (control)		1.00	72	72
1 Horizontal		1.07	72	77
Massive cuff		3.24	72	233
MMA		3.42	72	246
	White et al.,[24] 2006		MMA	Test
MMA (control)		1.00	140	140
4 Simple sutures		1.11	140	155
2 Mattress sutures		1.15	140	161
Modified Kessler		1.21	140	169
Anchor material				
Several acceptable	Barber et al.,[19] 2006			
	Chhabra et al.,[39] 2005	Cycles to 5-mm failure (ratio)	Single loaded	Test
Loading				
Single loaded		1.00	24	24
Double loaded		1.27	24	31
	Tingart et al.,[41] 2003	Pullout strength (ratio)	Anterior greater tuberosity (proximal)	Test
Positioning				
Anterior greater tuberosity (distal)		0.60	275	165
Posterior greater tuberosity		0.63	275	173
Anterior greater tuberosity (proximal)		1.00	275	275
Middle greater tuberosity		1.03	275	284
Lesser tuberosity		1.21	275	333
	Ma et al.,[42] 2006	Load to failure (ratio)	Single row	Test
Rows				
Single row		1.00	212	212
Double row		1.35	212	287
Angle				
Anchor angles		No differences reported		
Suture angles		<45° recommended		
Depth		Standard depth recommended		

The ratios are determined from results within a single study, and the ratios across several studies are presented. The relevant figures and their original studies are also given.