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Am J Sports Med 2005 33: 1030
DOI: 10.1177/0363546504271746

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Failure Mode of Suture Anchors as a Function of Insertion Depth

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Background: Surgeons can control not only the angle but also the depth of suture anchor placement during arthroscopic rotator cuff repair, although the tendency may be to place suture anchors on the deep side to avoid damage from prominent anchor eyelets. However, little information is available regarding possible effects of suture anchor depth on construct failure mechanisms.

Hypothesis: Anchor depth affects the mode of suture failure with physiologically relevant cyclic loads.

Study Design: Controlled laboratory study.

Methods: Metallic screw-in suture anchors loaded with No. 2 braided polyester sutures were inserted into the bovine infraspinatus footprint with the eyelet proud, standard, or deep. Sutures were hand tied to create a closed loop. Constructs were cyclically loaded from 10 to 90 N and, if still intact at 500 cycles, taken to ultimate failure (maximum load).

Results: When clinical failure was defined as greater than 3-mm construct elongation, anchors placed with the eyelet deep experienced statistically earlier clinical failure via cutting of the suture through the bone ($P < .02$). However, anchors placed at this level did not experience catastrophic failure during cyclic loading. The standard and proud anchors experienced 3 mm of elongation at a greater number of cycles, but the suture material degraded at the anchor eyelet, and a majority of these constructs broke during cyclic physiologic loading. At failure testing, the deep anchors had a significantly increased failure load (164 N) compared to standard (133 N) ($P < .04$) and proud (113 N) anchors ($P < .005$).

Conclusion: Varying the depth of suture anchor insertion changes the mechanical properties and mode of failure of suture anchor constructs.

Clinical Relevance: Surgeons should be aware of the effects of suture anchor depth and abrasive eyelet wear on construct failure during arthroscopic rotator cuff repair.

Keywords: rotator cuff; repair; suture anchors; insertion depth; biomechanics

A high proportion of clinically successful rotator cuff repairs are left with residual deficiencies in the cuff itself.[8,9] Suture anchor fixation for rotator cuff repairs has gained popularity in the recent past. Several studies have investigated the biomechanical characteristics of rotator cuff repairs in an effort to identify possible sites of failure. These sites have included failure via suture cutting through bone tunnels, suture breakage, knot slippage, suture anchor pullout, and soft tissue failure at the suture-tendon junction. Meyer et al[11] recently detailed that suture breakage at the anchor eyelet may in fact be a significant concern for the strength of soft tissue repairs using suture anchors. They tested 22 different metallic suture anchors loaded with No. 2 braided nonabsorbable suture material. In every case, the suture broke at the anchor eyelet at up to 73% lower loads than when the suture was tested on a smooth hook. We hypothesized that a previously unrecognized phenomenon may play a role in the failure mode of rotator cuff repairs: varying the depth of suture anchor insertion.

Two variables may play a role in exposing the suture/anchor eyelet to untoward forces. First, Roth et al have shown that anchors inserted into the glenoid settle to the cortical surface with cyclic loading.[14] This setting

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No potential conflict of interest declared.

The American Journal of Sports Medicine, Vol. 33, No. 7
DOI: 10.1177/0363546504271746
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leaves the anchor eyelet proud. This situation could possibly expose the suture to undue cyclic loading stresses at the anchor eyelet, as found by Meyer et al.[11] Second, when inserting anchors at a 45° "dead man angle" as recommended by Burkhart,[3] there is an inherent variability in the depth of insertion of the anchor owing to geometry that may leave the anchor eyelet either proud or deep. In our extensive search of the literature in the MEDLINE database, we could not find an investigation detailing the effects of the depth of suture anchor insertion on mechanisms of failure of rotator cuff repairs.

Burkhart et al[6] tested ultimate knot security and found failure strength (which was clinically defined as 3 mm of loosening/elongation or knot breakage) to be maximal with a No. 2 braided nonabsorbable suture placed in reverse-postreverse loop configuration. This failure strength was 51.6 ± 12.0 N. The study determined that the ultimate load a suture would have to withstand on a repair of a 4-cm tear with 3 anchors and 2 strands of suture per anchor (6 sutures) would be 37.7 N.[6] It did not, however, account for any possible material degradation of the suture and did not test complex sliding knots. Mishra et al[12] used No. 1 Ticron in a Duncan loop configuration and found a mean of 86.0 ± 10.1 N of force required to displace the knot 3 mm. Loutzenheiser et al[10] found a Duncan loop of braided No. 1 nonabsorbable suture tied with a standard knot pusher with alternating posts and direction of throws to withstand 89.0 ± 7.1 N of ultimate force at the 3-mm threshold. Therefore, a braided No. 2 nonabsorbable suture should be able to withstand the maximal load of 37.7 N, as defined by Burkhart et al,[6] in a 4-cm tear repaired with 3 anchors and 6 sutures.

This situation may not be the case, however, if the suture undergoes mechanical degradation at the suture anchor eyelet with a significant decrease in its load-bearing abilities. It is unclear if the depth of anchor insertion may have an influence on the stresses applied to the suture at the suture anchor eyelet and, therefore, its failure characteristics. The mode of failure could be weakening of the suture at either the tunnel entrance or the anchor eyelet during cyclic loading. This weakening could lead to abrasion and mechanical degradation of the suture with lower breaking strengths or to failure from elongation of the construct from the suture cutting through the bone.

We therefore undertook an investigation to determine if the depth of anchor insertion may have an effect on the failure characteristics of sutures and suture anchors used in rotator cuff repairs. We isolated the bone/anchor/suture aspect of rotator cuff repairs by testing without the soft tissues. We hypothesized that cyclic loading of the suture-anchor constructs at variable depths of insertion would expose the suture at each depth to different biomechanical parameters and wear patterns and lead to different failure characteristics at varying depths.

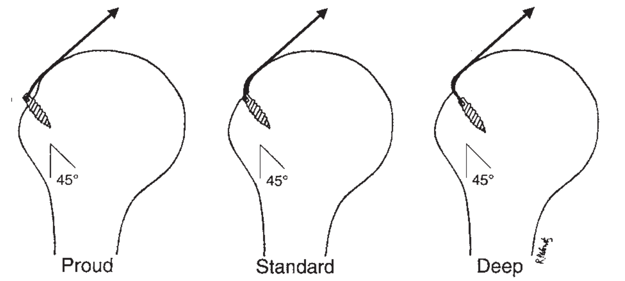

Figure 1. Proud, standard, and deep suture anchor constructs. All anchors were inserted at a 45° angle to the bone. For mechanical testing, sutures were pulled perpendicular to the angle of anchor insertion (arrow).

METHODS

Thirty metallic 5.0-mm screw-in suture anchors (Orthopaedic Biosystems Ltd, Smith & Nephew, Memphis, Tenn) loaded with a single No. 2 braided nonabsorbable polyester suture were inserted at 3 depths (proud, standard, and deep) into two 12-week-old bovine humeri. They were placed in the tuberosity footprint at the rotator cuff insertion site of the infraspinatus. The bone was previously dissected free of all soft tissues. Fifteen anchors were placed in each specimen, and each anchor starting position was placed at least 1 cm away from an adjoining anchor in a random pattern. The eyelet was aligned parallel to the direction of pull according to the laser line on the insertion device. The anchor was placed into the bone at an angle of 45° to maximize purchase according to Burkhart's dead man theory.[3] Each bovine specimen was tested with an equal number of proud, standard, and deep anchors.

The proud anchors were inserted with the shallowest part of the threaded portion of the anchor flush with the bone surface in the middle of the hole (Figure 1). This position resulted in the anchor eyelet being fully out of the bone and the suture material being exposed only to the suture anchor eyelet. This depth was chosen to mimic the settling of anchors in the glenoid bone that was observed to occur in the study by Roth et al.[14] Given that humeral tuberosity bone is less dense than that of the glenoid, we felt that extending this observed phenomenon to the humeral tuberosity was valid.

The standard anchors were inserted according to the manufacturer's guidelines, with the first laser line flush with the bone surface in the middle of the hole (threads 3-mm countersunk; Figure 1). There was, however, slight variability in the eyelet configuration after anchor insertion within this group, even while using a standardized insertion technique. Some sutures were slightly draped over the bone edge. Some sutures did not contact the bone at all, reproducing the situation seen in the proud constructs. This variability was the result of the inability to consistently place the standard anchors at precisely the same level and is similar to what is encountered surgically in vivo. This situation is due to the laser line on the insertion device being perpendicular to its long axis. When inserting the anchor at the 45° dead man angle and trying to consistently insert to the same depth, there was a slight variation to the depth of insertion.

The deep anchors were inserted at twice the depth of the standard anchors (6 mm). This location was chosen to mimic the situation in which an anchor is purposefully or inadvertently inserted too deep. All the sutures in this

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group draped over the bone edge in running from the eyelet, out the tunnel, and to the testing bar.

A strong, hand-tied knot was used to concentrate the bone-anchor-suture interface as the variable of failure. The suture was tied with 7 square knots backing up a surgeon's knot over a 7-cm-diameter dowel to create a closed loop. The loop was then attached to an MTS machine (MTS, Eden Prairie, Minn) over a smooth metallic rod.

The humerus was potted in a 2-part epoxy resin (Bondo-Marhyde, Atlanta, Ga) and held in place with a custom-designed fixation rig. The sutures were then placed under 10 N of preload, and the apparatus was zeroed in regard to displacement. The knot was placed equidistant from the bone hole exit and the smooth metallic rod on the MTS machine during load application. The location of the 2 suture strands touching the bone hole exit was recorded via an ink mark. The specimens were cyclically loaded from 10 to 90 N at 0.5 Hz to a maximum of 500 cycles. The cycles to clinical failure, as defined by Burkhart et al[6] as 3 mm of elongation in the construct, were recorded. If the sutures broke during the cyclic loading, their ultimate load strength was defined as the applied load at the time of failure. If the constructs were still intact after 500 cycles, they were loaded at 0.5 mm/s to failure to determine ultimate load strength. The direction of suture pull was perpendicular to the angle of insertion of the suture anchor; this angle was achieved by appropriate orientation of the humerus within the MTS machine. The number of cycles, mode, magnitude, and location of failure were recorded for each specimen. When the suture broke, the suture failure site was identified by the previously placed ink marks on the suture. All data were averaged within groups for descriptive statistics and compared with a 1-way analysis of variance ($P < .05$). When the analysis of variance reported a statistical difference, a Tukey honestly significant difference post hoc test was used for multiple comparisons.

RESULTS

When clinical failure was defined as greater than 3-mm construct elongation,[6,10,12] the sutures in the deep and standard anchor groups experienced clinical failure at fewer cycles than did the proud anchor group ($P < .02$) (Table 1). In 9 of the 10 deep specimens, failure via the suture cutting through the cortical margin of the bone was observed and occurred at a mean of 7.9 ± 5.6 cycles. The standard anchor constructs failed at the 3-mm level at a mean of 11.7 ± 1.9 cycles with a variable degree of suture cutting through the bone. Those sutures draped over the edge of the bone cut through the bone early and led to the 3 mm of failure. The proud anchor sutures failed at the 3-mm clinical level at a mean of 30 ± 30.6 cycles. There was no statistical difference between the deep and standard anchors in the number of cycles to 3 mm of failure. None of the anchors failed by translation or rotation within the humeral head.

There was a significant difference in the number of cycles to catastrophic failure (suture breakage) between the deep (500 ± 0 cycles) and both the standard (377 ± 197.2 cycles) and proud sutures (259.4 ± 206.8 cycles) ($P < .0004$). No deep construct failed early via suture breakage. None of the deep anchors settled to the cortical surface. Four of 11 standard-depth and 6 of 9 proud-depth anchors failed early during cyclic testing. All early failures in these groups occurred by suture breaking at the suture anchor eyelet. There was no statistical difference in cyclic failure between the proud and standard anchors (Table 1).

TABLE 1
Suture Anchor Cyclic Loading Data

Depth	Cycles to Catastrophic Failure[a]	Failure Mode	Cycles to 3-mm Failure[b]	Maximum Load, N[c]
Deep				
1	500.0	Knot	17	191.0
2	500.0	Knot	13	184.1
3	500.0	Knot	8	173.5
4	500.0	Knot	17	169.1
5	500.0	Knot	3	137.5
6	500.0	Knot	5	132.9
7	500.0	Knot	3	169.7
8	500.0	Knot	5	163.3
9	500.0	Knot	4	168.2
10	500.0	Knot	4	156.6
Mean	500.0		7.9	164.6
SD	0.0		5.6	18.3
Standard				
1	500.0	Anchor head	10	118.0
2	30.0	Anchor head	8	94.5
3	500.0	Knot	9	169.6
4	500.0	Anchor head	13	177.1
5	55.0	Anchor head	37	85.3
6	500.0	Knot	5	141.9
7	139.0	Anchor head	11	89.4
8	500.0	Knot	5	171.6
9	500.0	Knot	6	161.4
10	423.0	Anchor head	4	90.5
11	500.0	Knot	10	161.5
Mean	377.0		11.7	132.8
SD	197.2		1.9	37.6
Proud				
1	500.0	Anchor head	7	157.6
2	307.0	Anchor head	102	97.0
3	500.0	Knot	31	180.3
4	500.0	Anchor head	24	137.8
5	38.0	Anchor head	36	85.4
6	52.0	Anchor head	46	89.0
7	40.0	Anchor head	8	87.6
8	102.0	Anchor head	11	86.5
9	296.0	Anchor head	5	91.1
Mean	259.4		30	112.5
SD	206.8		30.6	36.3

[a] $P < .0004$ between the deep and the standard anchors and between the deep and the proud anchors (analysis of variance).
[b] $P < .02$ between the proud and the deep and standard anchors (analysis of variance).
[c] $P < .04$ between the deep and the standard anchors (Tukey honestly significant difference post hoc test) and $P < .005$ between the deep and the proud anchors (Tukey honestly significant difference post hoc test).

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In evaluating ultimate failure load, the deep anchors had a statistically greater failure load than either the standard anchors ($P < .04$) or the proud anchors ($P < .005$) (Tukey honestly significant difference post hoc test; Table 1). In all 10 specimens in the deep construct, failure occurred at the knot at a mean load of 164.6 ± 18.3 N. For the standard constructs, the mean ultimate failure load was 132.8 ± 37.6 N. For the standard anchors remaining for testing to failure, 66% of samples failed by suture rupture at the suture eyelet. For the proud anchors, failure occurred at 112.5 ± 36.3 N by suture rupture, and 88% of failures were located at the anchor eyelet.

DISCUSSION

Several mechanisms for the possible failure of rotator cuff repairs have been postulated. In Barber and Herbert's study of the ultimate strength of suture anchors, they found that all anchors were stronger than the suture they were designed to accept.[2] In an in vivo study, Barber et al[1] consistently found suture breakage as the mode of failure after retrieval of Mitek G2 and Statak anchors placed in sheep. In cyclically loaded repairs through bone tunnels versus the use of suture anchors, Burkhart et al found that the weak link in the system moved from the bone to the suture-tendon construct.[4,5] In cyclically loaded repairs with the anchor on the lateral cortex of the humerus and the sutures draped over the greater tuberosity, Rossouw et al[13] found that failure occurred via suture cutting through bone and sawing through tendon, with ultimate failure by suture breakage. In a cyclically loaded model, Goradia et al[7] found suture breakage as the mode of failure in 1 of 8 rotator cuff repairs with anchors. One anchor pulled out, and the other 6 failed by suture cutting through the cuff tissue. In cyclically loaded screw-in and barbed anchors for the repair of glenoid labrum defects, Roth et al[14] found that settling of the anchors occurred within the first 100 cycles until the barbs or screws came to rest against the cortical margin. This setting resulted in the anchors being between 1- and 2-mm proud. We did not observe this phenomenon. This occurence may have been due to the differences in bone density in our bovine model. Young bovine bone has been found to have a density of 0.76 gm/cm^3, which is similar to data reported for young human bone of 0.8 gm/cm^3.[3] Thus, anchor purchase in the current study may actually be better than the purchase acquired during previous human cadaveric experimentation. In cyclically loaded suture anchors in a porcine model, Rupp et al[15] found that Mitek G2 and Statak 3.5-mm anchors predominantly failed at the suture-anchor interface, whereas other anchors generally failed via suture breakage at the knot. Other studies[6,10,12] have detailed clinical failure modes, which they defined as 3 mm of knot slippage, during cyclic loading of suture loop constructs.

Our study showed that by varying the depth of insertion of a suture anchor, the mode of failure of the construct changed from catastrophic abrasive wear of the suture at the eyelet to earlier 3 mm of elongation in the construct, defined as clinical failure by Burkhart et al.[6] For the proud anchors, early catastrophic failure due to mechanical weakening of the suture at the eyelet was the predominant mode of failure. However, these sutures did not clinically fail via elongation to 3 mm as early during cyclic loading. They did, however, fail to this threshold within a mean of 30 cycles. The deep anchors failed earlier by the clinical criteria to 3 mm, but they were stronger in ultimate strength. The 3-mm clinical failure occurred in the deep construct by the suture that was draped over the bone tunnel edge cutting through the bone, resulting in elongation. This bone tunnel, however, protected the suture from cyclic abrasion and degradation at the eyelet and eventual catastrophic early failure. The standard anchors showed a mix of both modes of failure. This failure was due to a slight variation in the depth of insertion achieved when inserting the anchors at a 45° dead man angle. A slightly proud insertion resulted in a situation mimicking that of the proud anchors, and a slightly countersunk anchor resulted in biomechanical behavior such as that seen with the deep anchors.

Burkhart et al[6] stated that in cyclically loaded bone-suture-tendon systems, failure typically occurred at a biologic location. The biologic materials protect the suture because of energy absorption via elongation or failure of the biologic tissue. The suture does not fail even though loads are well above the typical strength of suture material and knots. In ultimate load situations as would occur with a fall or reflexive contraction of the cuff, the suture is not shielded and thus is the weak link in the system. Burkhart et al[6] suggested that the ultimate load a suture would have to withstand in a 4-cm tear with 3 anchors and 2 strands of suture per anchor would be 37.7 N. In their study of knot security, they found a No. 2 braided nonabsorbable suture could withstand approximately 52 N and concluded that knot security and suture strength are adequate to withstand the theoretical applied forces.[6] This finding represents an ideal situation, however. In an unbalanced repair or if an anchor has only 1 suture, loads could be greater than the 37.7-N load as predicted by Burkhart et al,[6] and we therefore tested the construct at roughly double the predicted loads of that study. We did not test the complete tendon-suture-anchor construct and therefore were limited by the effect the soft tissues may have had on the construct. Also, if the suture were to undergo mechanical degradation through cyclic tensioning and abrasive wear at the suture eyelet, the suture could be weakened and prone to early failure. This concern was displayed in our study, as well as those by Meyer et al[11] and Rupp et al.[15]

In our study, we found that a suture attached to an anchor placed in a deep position was prone to failure by a different mechanism. This mechanism was clinical failure and occurred via the suture, which was draped over the bone edge of the tunnel, cutting through the bone. Although a suture cut-through might not cause immediate catastrophic failure of the construct, gap formation at the bone-tendon interface may interfere with the quality and speed of tissue healing.

To our surprise, our data also suggested that the deep anchor construct was stronger in ultimate load and was

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less prone to mechanical degradation and catastrophic breakage of the suture at the anchor eyelet. The standard and proud anchors had significantly decreased ultimate load strength and had a significantly greater occurrence of early suture breakage in cyclic loading. This situation occurred via mechanical degradation and breakage of the suture at the anchor eyelet. We postulate that in the deep anchor configuration, early failure occurred by suture cutting through bone, but then the suture tethered at the bone margin. This region of the suture protected the suture-eyelet interface from experiencing early suture abrasion with cyclic load (all constructs in this position survived to 500 cycles without suture breakage). In the standard anchor position, the anchor head was partially buried, with the suture slightly draped over the bone margin. This position still allowed the suture to cut through bone, with early 3-mm failure. However, in this configuration, the suture did not become tethered against the bone margin. Instead, the force of cyclic loading was transferred to the anchor eyelet, and the suture was exposed to early mechanical degradation and failure. For the proud anchors, early suture cut-through was not possible because the anchor head was outside of the bone. This construct exposed the suture to early abrasive wear at the anchor eyelet (only 3 of 9 constructs in this position survived to 500 cycles without suture breakage). This situation may be encountered clinically if the anchor is placed proud or if it settles back to the cortical margin, as was found in the previously cited study by Roth et al.[14]

In the clinical situation of rotator cuff repairs, patients' rehabilitation and recovery will no doubt involve more than 30 cycles of motion during their recovery. Cycling past this limit would place all of our constructs under the definition of a clinical failure with more than 3 mm of elongation. What is interesting to note, however, is that the sutures in the deeply placed anchors were protected from abrasive suture wear, and none had early catastrophic failure. In a double-row rotator cuff repair, the medial anchors may be exposed to tensile forces before the lateral anchors and may thus protect the lateral anchor sutures from early abrasive wear.

Our findings are specific to a metallic anchor with a metal eyelet using a No. 2 Ethibond suture (Ethicon Inc, Johnson & Johnson, Somerville, NJ). The findings may not be directly related to other anchor designs, eyelet configurations, or suture materials. However, based on these data, we cannot make a recommendation against burying or leaving proud anchors to avoid early 3-mm clinical failure, as all constructs failed at this threshold within a mean of 30 cycles. All patients would no doubt cycle their shoulders more than 30 times during their recovery. It is clear, however, that suture abrasion at the anchor eyelet may result in catastrophic failure of the repair construct, as seen in the proud and standard anchors but in none of the deep anchors. There are at least 2 logical solutions for this phenomenon. Anchor eyelets should be designed to lessen the possibility of cyclic abrasive degradation of the suture, and abrasion-resistant materials should be developed to decrease the risk of suture fretting.

ACKNOWLEDGMENT

We thank Smith & Nephew for supplying the anchors and sutures for this study.

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