Effect of configuration on the biomechanical performance of three suture materials used in combination with a metallic bone anchor

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Objective—To determine whether different suture configurations could improve the biomechanical performance of 3 suture materials used with bone anchors.

Samples—3 suture materials (60-lb test nylon leader line, size 2 polyblend polyethylene composite suture, and 150-lb test ultrahigh–molecular weight spun polyethylene).

Procedures—Each suture material was looped through the eyelet of a metallic bone anchor and constructs were evaluated by use of an acute uniaxial load. Three configurations were tested for each suture material: single stranded (SS), double stranded (DS), and single stranded plus plastic insert (SSP). Force at failure, extension at failure, force at 3 mm of extension, stiffness, and site of failure of the suture were recorded for each test.

Results—For all sutures, the DS configuration was the stiffest and yielded significantly higher forces at failure and forces at 3 mm of extension. The SS configuration had the lowest forces at failure. The SSP configuration yielded greater forces at failure for all suture materials, compared with the SS configuration, with a comparable stiffness. All sutures failed at the eyelet in the SS and DS configurations. In the SSP configuration, 60-lb test nylon leader line and 150-lb test ultrahigh–molecular weight spun polyethylene failed at the eyelet less frequently than did the polyblend composite suture.

Conclusions and Clinical Relevance—Among the tested constructs, a DS suture configuration used in combination with the metallic bone anchor gave the best biomechanical results for all suture materials. Considering that the SSP configuration yielded greater forces at failure, compared with the SS configuration, covering metallic edges in bone anchors with softer materials might protect sutures and result in increased forces at failure. (Am J Vet Res 2013;74:1487–1492)
used for ligamentous reconstructions. Nonabsorbable multifilament suture materials such as polyester have also been used extensively.\textsuperscript{11} Multifilament suture materials maintain a distinct advantage over monofilament sutures by having high tensile strength, resistance to stretch, and superior knot security. However, a potential disadvantage of multifilament suture materials is an increased risk of postoperative infection caused by the high surface area available for bacterial adherence.\textsuperscript{12} A multifilament PCS\textsuperscript{9} developed in 2000 has been reported to have superior biomechanical properties, particularly when used with bone anchors. This suture material has higher loads to failure, sustains increased cyclic loading, and has higher abrasion resistance at the suture–bone anchor interface, compared with other multifilament nonabsorbable sutures, such as polyester.\textsuperscript{13} A similar suture material, UHMWSP,\textsuperscript{10} has also become available. This suture material is claimed to have similar biomechanical properties to PCS, but it lacks the polyester braided jacket that has been reported to give PCS superior abrasion resistance in clinical situations.\textsuperscript{14} To the authors’ knowledge, only 1 recent study\textsuperscript{15} evaluating the biomechanical properties of UHMWSP has been reported in the scientific literature; results indicated that UHMWSP outperformed other polyblend sutures in acute and cyclic biomechanical testing.

Failure of suture–bone anchor constructs can occur at the anchor–bone interface, suture-anchor interface (eyelet), or soft tissue–suture interface. In metallic suture–bone anchor constructs, the most common site of failure is at the eyelet.\textsuperscript{1,3,6–8,10,14} A roughened surface associated with the eyelet (a so-called unfriendly edge) and an increased arc of contact between the eyelet and suture material contribute to failure via friction and abrasion.\textsuperscript{1,3,5,8,10} In contrast, eyelet disruption or cut-through of the eyelet by the suture material, as opposed to failure of the suture material itself, occurs more frequently in biodegradable suture–bone anchor constructs,\textsuperscript{6,8,10,14} which are uncommonly used in veterinary surgery. The purpose of the study reported here was to determine whether different suture configurations could improve the biomechanical performance of 3 suture materials used with metallic bone anchors. We hypothesized that the likelihood of sutures to fail at the eyelet would be reduced and that loads to suture failure would be increased by protecting the suture at this location with a plastic IV catheter tubing insert or by use of a DS suture configuration.

Materials and Methods

Suture types and preparation—The 3 suture materials evaluated were 60-lb test nylon LL,\textsuperscript{6} 150-lb test UHMWSP, and size 2 PCS. All sutures were supplied by the respective manufacturers. The size 2 PCS was provided prepackaged and sterilized with ethylene oxide. The 60-lb test LL and 150-lb test UHMWSP were provided unsterilized; these were plasma sterilized\textsuperscript{15} prior to use. The size of each suture material chosen for evaluation was based on the diameter of the suture itself, rather than the breaking strength. Because of the diameter of the eyelet of the bone anchor (2 mm), suture materials with diameters ranging from 0.6 to 0.8 mm were selected, thereby allowing 2 strands of suture material to pass through the eyelet without excessive friction. The reported diameter for the 60-lb test LL was 0.8 mm, for the 150-lb test UHMWSP was 0.6 mm, and for the size 2 PCS was 0.5 to 0.6 mm. The size 2 PCS has a reported breaking strength of 271 N (equivalent to 60.8 lb).\textsuperscript{13} A single 3.5-mm self-tapping metallic bone anchor\textsuperscript{14} was inserted into one end of an 11.9-cm-long, 1.2-cm-diameter solid aluminum rod following predrilling with a 3.175-mm (1/8-inch) drill bit. The aluminum rod was attached to the lower fixture of the materials testing unit; an identical aluminum rod (with no bone anchor) was attached to the upper fixture. Immediately prior to use, suture material was removed from its packaging, cut to a predetermined length (20 cm), and immersed in saline (0.9% NaCl) solution that had been heated to 37°C. Each strand of suture material was looped over the upper aluminum rod prior to passage of the suture through the eyelet of the bone anchor. Knots were tied in the same orientation for each construct, equidistant to the loading arms, by a single investigator (SMW). Knots were tied with a surgeon’s knot, followed by a number of square knots based on published data for each suture (4 additional square knots for LL\textsuperscript{16} and 7 additional square knots for PCS\textsuperscript{14,17}). Because no such scientific data relating to knot security were available for UHMWSP, a pilot study was performed to evaluate the number of throws required to create a secure knot. In accordance with a previous study\textsuperscript{18} relating to the knot security of suture materials, a secure knot was defined as a knot that resulted in breakage of the suture material when tested to failure, as opposed to knot slippage or untwisting. To determine this information for UHMWSP, a single strand of suture material was secured with a different number of square throws (4, 5, 6, and 7 throws; n = 3 tests/group) in addition to the initial surgeon’s knot, prior to being subjected to a single uniaxial load. In all instances in which 4, 5, or 6 additional throws were used, the construct failed by slippage and unraveling of the knot. When 7 additional throws were used to create the knot, the construct failed by breakage of the suture at the eyelet in each test.

Tag ends were trimmed to a uniform length of 3 to 4 mm for all sutures, in accordance with a standard clinical situation. The sutures were marked at the eyelet of the bone anchor, to allow accurate determination of the point of failure. A new bone anchor was used for each group of suture material configurations tested (total, 9 bone anchors) to minimize any potential influence on the site of failure of the suture caused by damage to the eyelet secondary to repeated use.

Each suture material was tested in 3 configurations (n = 10 samples/group; total, 90 tests): SS, DS, and SSP. In the SSP groups, a precut 1-cm length of 18-gauge IV catheter tubing (internal diameter, 1.3 mm) was used to protect the suture material at the eyelet. With each new test in each group, a new length of tubing was used because of the deformation and damage sustained by the tubing during mechanical testing (n = 30 sections of catheter tubing).

Mechanical testing—A materials testing unit with a 1-kN load cell verified to be accurate within
1%, driven by a 500-W single phase motor, was used to determine force at failure, extension at failure, force at 3 mm of extension, stiffness, and site of failure of each suture–bone anchor construct. The loading arms of the materials testing unit were positioned so that the constructs were situated directly beneath the actuator in all tests, such that the load applied to the suture material was at 90° to the angle of insertion of the anchor. The orientation of the eyelet was in the coronal plane, with an anchor rotation angle of 90°. A working distance of 3 cm between the loading arms was selected.

A preload of 2 N was applied prior to testing to prevent suture relaxation, and the apparatus was subsequently zeroed in regard to displacement. A uniaxial force was applied at a force-to-displacement rate of 60 mm/s. The uniaxial load was applied until the point of failure of the suture, defined as an abrupt drop in tensile load. The materials testing unit was interfaced with a desktop computer for control of testing and collection of force-displacement data. The data were accumulated with an appropriate software program and tabulated in spreadsheet format. Force at failure (N), extension at failure (mm), force at 3 mm of extension (N), stiffness (N/mm), and site of failure of the suture were recorded for each sample tested.

**Statistical analysis**—Descriptive summary statistics were calculated for each of the outcome variables (force at failure, extension at failure, force at 3 mm of extension, and construct stiffness) and evaluated with box-and-whiskers plots and histograms. Normality plots were also obtained for all data to ensure that the assumptions for normality and equal variance were validated prior to proceeding with statistical analysis. Any skewed data were logarithmically transformed to improve normality. A 2-way ANOVA was used to compare the means of force at failure, extension at failure, force at 3 mm of extension, and construct stiffness among the 3 configurations of each suture material. Significance was set at P < 0.05 for all comparisons. When significant differences were found, the means were compared by means of the least significant difference approach. When the absolute differences between 2 means exceeded the calculated least significant difference value, a significant (P < 0.05) difference was observed. A Fisher exact test was used to test for difference in the site of failure for each suture. All analyses were conducted with a statistical program.

**Results**

All suture materials in all configurations had similar force versus displacement patterns. Overall, 150-lb test UHMWSP was significantly stiffer than either 60-lb test LL or size 2 PCS in all configurations, and size 2 PCS was twice as stiff as 60-lb test LL in all configurations (Table 1). There was no significant difference in stiffness between the SS and the SSP configurations of all suture materials. Results for force at failure testing (Table 2) revealed that both suture material type and configuration significantly (P < 0.001) influenced the force at failure, but the interaction between these variables was not significant. The 150-lb test UHMWSP was significantly stronger than 60-lb test LL or size 2 PCS in all configurations. No significant difference in mean force to failure was detected between 60-lb test LL and size 2 PCS. Each suture material was significantly stronger in the DS configuration, compared with the SS or SSP configuration. In addition, each suture material had increased force at failure in the SSP versus the SS configuration. Overall, the mean forces at failure observed in this study were substantially lower than the reported suture material strength; in the SS configuration, the UHMWSP and LL sutures and PCS ruptured at forces approximately 58%, 65%, and 84% of the reported material strength, respectively.

In all configurations, 60-lb test LL extended to a significantly greater degree prior to failure, compared with 150-lb test UHMWSP or size 2 PCS (Table 3). There was no significant difference in the degree of extension between 150-lb test UHMWSP and size 2 PCS in any configuration. For each suture material, the greatest degree of extension prior to failure occurred in the SSP configuration. The DS configuration had the least degree of extension prior to failure.

From the data relating to force at 3 mm of extension (Table 4), 60-lb test LL required the least amount of force to extend 3 mm in both the SS and DS configurations.
The site of failure was identified as breakage of the suture material at the bone anchor eyelet in 81 tests and breakage of the suture material immediately adjacent to the knot in 9 tests. Suture unraveling or knot slippage was not observed. A significant (P = 0.037) difference was detected in site of failure among the suture materials, regardless of the configuration tested. All sutures tested in the SS or DS configurations ruptured at 3 mm of extension, compared with the SSP configuration. For 60-lb test LL, the SSP configuration required a greater applied force to extend 3 mm, compared with the SS configuration.

The 150-lb test UHMWSP was significantly stronger than either 60-lb test LL or size 2 PCS in all configurations. This was an expected result, considering that the breaking strength of a single strand of 150-lb test UHMWSP was reported to be 27% of the breaking strength of the suture material as low as 27% of the breaking strength of the suture material were detected in a previous study and were also significantly reduced in this study for UHMWSP and LL sutures. In the present study, PCS lost the least strength when passed around a metallic edge (force at failure, 84% of breaking strength). The PCS has been reported to have a higher resistance to abrasion than other sutures. Although we only performed a single load-to-failure test and did not test abrasion resistance, which would have required repeated contact between the suture material and eyelet, it seems that PCS has a greater resistance to damage by shear forces and friction at the eyelet in metallic bone anchors, compared with the other suture types.

Because of its similar properties to PCS, UHMWSP could also be expected to have similar improved resistance to shear forces; however, PCS is coated with a polyester braided jacket, which UHMWSP does not possess, which may account for the observed superiority of PCS with regard to shear forces. It is also possible that the sterilization method used in this study may have interfered with the material properties of UHMWSP. The size 2 PCS material was obtained sterile (via treatment with ethylene oxide) from the manufacturer, whereas the 60-lb test LL and 150-lb test UHMWSP were unsterilized when acquired. To maintain consistency, we elected to sterilize the 60-lb test LL and 150-lb test UHMWSP prior to use. Steam sterilization via autoclaving has a detrimental effect on the compliance of LL, with a 2- to 4-fold increase in extension, compared with unsterilized LL. Previous studies have indicated that LL may be plasma sterilized safely, without any detrimental effect on strength or force at failure. Therefore, plasma sterilization of both the LL and UHMWSP samples was performed. To our knowledge, there are no published reports of the effect of plasma sterilization on UHMWSP; therefore, the effects of this method of sterilization on the force at failure are unclear and warrant further investigation.

An ideal suture material has a low suture diameter-to-strength ratio, thereby reducing the volume of foreign material needed for implantation in a clinical situation. Large-gauge sutures can be difficult to tie and result in bulky knots, which can cause local irritation to the adjacent soft tissues, seroma formation, and patient discomfort. In the present study, 60-lb test LL had the largest suture diameter but the lowest mean force at failure, whereas 150-lb test UHMWSP had the smallest diameter and the highest mean force at failure. However, on the basis of our findings and results of a previous study, LL requires only 4 throws in addition to a surgeon’s knot for security. In comparison, PCS and UHMWSP required 7 throws in addition to a surgeon’s knot for security, resulting in a knot that was subjectively of comparable size.

It is widely accepted that the knot is the weak point of any tied suture, particularly during distraction or tension, because a stress riser is created by shear forces at this location. An exception to this situation is that when a bone anchor is used, with passage of suture material through the eyelet, the point of contact of the suture and eyelet is considered the weak point. Protection of the suture material at the eyelet via the
addition of a plastic insert resulted in greater force at failure in the present study. This protective effect was most notable when evaluating the only monofilament suture examined (60-lb test LL) in that the site of failure changed from suture rupture at the eyelet to suture rupture at the knot in most of the tests. We obviously do not suggest use of a plastic IV catheter tubing to protect suture material when used in combination with a bone anchor in a clinical situation. However, development of a commercially available insert to protect against the effects of the edge of metallic bone anchor eyelet may be warranted. If the dimensions of such an insert were sufficient to protect against the edge of the eyelet without substantially reducing the working diameter of the eyelet, this may allow the passage of 2 strands of suture material (eg, a DS configuration) in addition to the insert. This would increase construct stiffness and force at failure, resulting in a stronger construct; this construct should be feasible when suture materials with a smaller diameter (eg, PCS or UHMWSP) are used.

Of all the suture materials tested in this study, 60-lb test LL was the most compliant, with extension at failure approximately double that observed for either 150-lb test UHMWSP or size 2 PCS in all configurations. This was well beyond the reported 3-mm guideline for clinical failure of suture materials in soft tissue repair. Both 150-lb test UHMWSP sutures and size 2 PCS had significantly higher forces at 3 mm of extension than did 60-lb test LL sutures. Placement of all sutures in the DS configuration significantly increased the stiffness and the force required to extend 3 mm, as expected. The SS suture configuration only withstood surprisingly low forces at an extension of 3 mm, when considering that, for example, the cranial cruciate ligament in Labrador Retrievers is estimated to resist loads of 50 N at a walk, increasing to a maximum of 400 to 600 N during physical activity. The exact forces to be withstood by suture–bone anchor constructs in a clinical situation such as this are unknown but are dependent on the size of the animal, ground reaction forces, muscle and joint forces, and many other factors. However, this is clearly an indication that suture–bone anchor constructs are required to counteract forces that can be substantial. As such, use of a double strand of suture material through the eyelet of a bone anchor should be strongly considered in a clinical situation, provided that the eyelet is large enough to accommodate 2 strands of suture material without excessive friction. An unexpected finding in this study was that the SSP configuration for 150-lb test UHMWSP and size 2 PCS required significantly less force to attain 3 mm of extension, compared with the SS configuration. This may be explained by the physical properties of the plastic insert; it is possible that during application of the preload and at lower loads, the plastic insert became distorted and bent prior to the load being distributed through the suture material. This may have resulted in some give within the construct as some force was borne by the insert prior to it being applied to the suture, which may have reduced the overall force necessary to achieve 3 mm of extension.

There were several limitations to this study. The constructs were evaluated only with an acute uniaxial load to failure, rather than with cyclic loading. Although testing construct performance during a single supraphysiologic load is useful, cyclic loading would be more representative of a clinical situation, where repetitive stress on constructs during normal weight bearing, range of motion, and activity occurs. Further evaluation of these suture materials and configurations during cyclic loading is warranted to determine whether the observed findings remain consistent. In addition, the constructs were evaluated only in a single bone anchor eyelet in a single orientation. Eyelet orientation influenced force to failure in an acute load to failure study. On the basis of that report, we selected a uniform anchor rotation angle of 90° to the direction of the anchor axis to eliminate this as a contributing variable. The loading arms of the materials testing unit were positioned 3 cm apart prior to placement of each suture material and knot tying, such that each test unit maintained a uniform distance between the loading arms while the suture loop was configured. For each test, the suture loop was then pretensioned to 2 N by adjusting the distance of the loading arms, which could have resulted in minor discrepancies in loading arm starting distance among tests. It is also possible that cantilever bending of the aluminum rods could have influenced the apparent extensibility of the suture materials. However, the rods were thick (1.2 cm diameter), composed of solid aluminum, and custom made to fit the test grips of the materials testing unit. Therefore, we believe that the rods were of ample thickness to resist any deflection or bending at the loads achieved. In addition, this system has been used in other biomechanical studies involving tennis racquet strings, in which loads of up to 600 N were applied, with negligible deflection of the rods, compared with the degree of extension by the strings. Differences in sterilization methods chosen may also have influenced the results.

As expected, the DS configuration was the strongest and stiffest for all suture materials and may be the best option (of those tested) for clinical use. Protection of the suture with a plastic insert as it passed through the eyelet of the bone anchor resulted in higher forces to failure. For some sutures, the protective effect was large enough to shift the failure point of the construct to the knot of the suture loop. As expected, this effect was most striking for the monofilament suture (60-lb test LL). Improvement in bone anchor design such that problematic edges are eliminated from eyelets is also worth further consideration to reduce the risk of abrasive damage and premature catastrophic failure of suture materials.


b. FiberWire, donated by CEVA Animal Health Pty Ltd, Sydney, NSW, Australia.
c. LigaFiba, donated by St. Lucia Surgical Services, Brisbane, QLD, Australia.
d. Mason nylon leaderline, Hi-Seas American Fishing Wire, Coatesville, Pa.
e. Sterrad NX, Advanced Sterilisation Products, Johnson & Johnson, Sydney, NSW, Australia.
f. Hounsfield Model H1K-S, PCS Measurement Pty Ltd, Newcastle, NSW, Australia.
g. QMat-Pro Software, version 3.62, Tinus Olsen, Newcastle, NSW, Australia.
h. Excel 2007, Microsoft Corp, Redmond, Wash.


References