# Assessment of skin staples for augmentation of core tenorrhaphy in an ex vivo model of canine superficial digital flexor tendon laceration

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**OBJECTIVE**
To compare the biomechanical strength and incidence of gap formation among canine superficial digital flexor tendon (SDFT) constructs that underwent core tenorrhaphy only and those in which the core tenorrhaphy was augmented with skin staples or a continuous Silfverskiöld cross-stitch (SXS) suture pattern.

**SAMPLE**
42 cadaveric forelimb SDFTs from 21 musculoskeletally normal dogs.

**PROCEDURES**
Tendons were randomly assigned to 3 groups (14 SDTFs/group), sharply transected, and repaired with a core locking-loop suture alone (group 1) or augmented with circumferential placement of skin staples (group 2) or a continuous SXS suture pattern (group 3) in the epitenon. All constructs underwent a single load-to-failure test. Yield, peak, and failure loads, incidence of gap formation, and mode of failure were compared among the 3 groups.

**RESULTS**
Mean yield, peak, and failure loads differed significantly among experimental groups and were greatest for group 3 and lowest for group 1 constructs. The incidence of gap formation differed among the tested groups and was lowest for group 3 and highest for group 1. The most common mode of construct failure was the suture pulling through the tendon for group 1, staple deformation for group 2, and epitendinous suture breakage for group 3.

**CONCLUSIONS AND CLINICAL RELEVANCE**
Results indicated epitendinous placement of skin staples around a core SDFT tenorrhaphy site improved the biomechanical strength and resistance to gap formation for the repair but was inferior to epitendinous placement of SXS sutures. Further research is necessary before skin staples are used for tenorrhaphy augmentation in clinical patients. (*Am J Vet Res* 2020;81:681–688)

The nonlinear biomechanical properties of flexor tendons are well suited to absorbing and returning energy associated with the transmission of tensile forces across the musculoskeletal system. The resilience of such tendons means that they can serve as an effective biological spring while contributing to proprioception and maintenance of joint stability.[1] When muscular contraction occurs, fibrils at the myotendinous junction transfer kinetic energy through the musculotendinous unit to bone, thereby providing the capacity for locomotion and voluntary control over the skeleton.[2,3] Because tendons influence both structure and form, tendinous injuries can have a more dramatic effect on function, compared with muscular injuries alone.[4] In dogs, the incidence of tendon injuries is low, accounting for only 1.6% of all appendicular musculoskeletal injuries.[5] However, early surgical intervention to repair tendon injuries is often indicated for optimal return to function and a successful long-term outcome.[3,4,6]

The primary aim for surgical repair of tendons is to achieve direct-contact healing without gap formation while allowing for early loading to ensure organized and uniform collagen fibril alignment. Formation of gaps > 3 mm at a tendon repair site significantly decreases tendon strength and increases healing time and the risk for rerupture during the early phases of rehabilitation.[3,4] To prevent gap formation, tenorrhaphy techniques are designed to be functionally strong and allow gliding as well as technically easy and repeatable.[3,7,8] Various suture patterns, such as core LL, 3-loop pulley, and Krackow patterns, have been successfully used for tenorrhaphy of core tendon injuries in dogs.[4,9–14] In ex vivo studies[15,16] involving canine cadaveric tendons, placement of ESs in conjunction with core tendinous repair significantly increases the biomechanical strength and decreases the incidence

**ABBREVIATIONS**

| | |
|---|---|
| CSA | Cross-sectional area |
| ES | Epitendinous suture |
| LL | Locking loop |
| SDFT | Superficial digital flexor tendon |
| SXS | Silfverskiöld cross-stitch |

of gap formation in tendon constructs, compared with placement of core tendon sutures alone. In human medicine, simple continuous,[17] SXS,[18] and interlocking horizontal mattress[19] suture patterns are commonly used to place circumferential ESs during flexor tendon repairs, and use of ESs increases the tensile strength of the tendon repair by 10% to 50%[20] and decreases the incidence of revision surgery by 84%.[21] Collectively, these findings suggest that ESs play a crucial role in tenorrhaphy strength and loading characteristics. However, investigation of the biomechanical strength of new ES techniques for augmentation of core tendon repair is necessary before those techniques can be evaluated for in vivo applications.

Surgical stapling devices were originally described in the mid-1800s and are widely used in veterinary practice during pulmonary, gastrointestinal, vascular, and cutaneous surgical procedures owing to the ready availability of various staple sizes, shapes, and applicators.[22–24] Benefits associated with the use of surgical stapling devices include a decrease in surgery time, less intraoperative contamination, decreased tissue trauma and manipulation, and preservation of vascular supply.[22] Skin stapling devices are preloaded with up to 35 stainless steel staples that, when deployed, interact with tissue and close to form a rectangular configuration.[25] Skin staples have been used for various applications aside from simply closing skin incisions, such as closing enterotomy sites,[25] placing skin grafts,[26] affixing surgical drains,[27] attaching surgical drapes to a patient,[28] securing mesh for hernia replair,[29] and creating end-to-end anastomoses.[30] The holding strength of surgical skin staples is equivalent to interrupted sutures for approximating functional end-to-end intestinal anastomoses in dogs.[31] Use of skin staples to augment core tendon repairs might increase the biomechanical strength of and decrease gap formation in the sutured constructs.

Theoretically, compared with placement of ESs, placement of skin staples in the epitenon to augment core tendon repair may be simplified in regard to tissue manipulation and application, thereby resulting in less tissue trauma and superior approximation of the tendon ends. The objective of the study reported here was to compare the biomechanical strength and incidence of gap formation among ex vivo canine SDFT constructs that underwent core tenorrhaphy only and those in which the core tenorrhaphy was augmented with circumferential placement of skin staples in the epitenon (ie, epitendinous skin staples) or ESs that were placed by use of a continuous SXS pattern. We hypothesized that constructs augmented with epitendinous skin staples would have a lower incidence of gap formation and be able to sustain higher loads at yield, peak, and failure, compared with constructs that underwent core tenorrhaphy only but would have a higher incidence of gap formation and lower mean loads at yield, peak, and failure, compared with constructs augmented with ESs placed by use of the SXS pattern.

# Materials and Methods

## Samples

The North Carolina State University Institutional Animal Care and Use Committee determined that the study was exempt from review because it did not involve the use of live animals. Given the failure loads reported for experimental canine SDFT constructs of other studies[15,16] and pilot tests conducted for the study reported here, a sample-size calculation determined that 14 constructs/experimental group would be necessary to detect a mean ± SD difference of 30 ± 5 N in failure load among the 3 experimental constructs with 95% confidence and at least 90% power.

Forty-two SDFTs were harvested from 21 canine cadavers that weighed between 28 and 35 kg. All dogs were euthanized for reasons unrelated to the study and were free of clinical musculoskeletal disease. Cadavers were maintained at room temperature (21°C), and tendon specimens were harvested within 2 hours after euthanasia as described.[15,16] Briefly, the superficial digital flexor muscle and tendon, including the proximal origin at the medial aspect of the humeral condyle to the distal insertion sites on the manus, were harvested intact, whereas all other soft tissues were removed and discarded.[15,16] Each specimen was labeled, wrapped in gauze soaked with saline (0.9% NaCl) solution, and stored in a sealed plastic bag[a] at –20°C in accordance with a validated technique[32] for storage of excised tendons.

Each specimen was thawed at room temperature for 8 to 10 hours prior to testing. On the day of experimental testing, a No. 10 scalpel blade was used to transect the SDFT 3 cm distal to the musculotendinous junction. The tendon was transected on a hard, durable surface to ensure that a uniform cut was achieved across the entire cross-sectional surface. The cut surface of the tendon was photographed alongside a calibrated ruler; the camera[b] was positioned 10 cm from and parallel to the cut edge of the tendon. The CSA of each tendon was measured with an imaging software program[c] by 1 investigator (Y-JC).

## Experimental groups

A random number generator[d] was used to randomly assign tendon specimens to 1 of 3 experimental groups (14 tendons/group), with care taken to ensure that tendons harvested from the same cadaver were not assigned to the same group. All transected tendons underwent core tendon repair by use of size-0 polypropylene suture[e] placed in an LL pattern as described.[13,33] During the repair, digital manipulation was used to bring the tendon ends into close apposition before the suture was knotted. The core suture was tied in a square knot followed by 4 square throws; suture ends were cut 5 mm from the knot. The knot was tied with care to ensure that the tissue did not become bunched at the repair site. Although subjectively assessed, an effort was made to ensure

that equal tension was applied during placement of all core tendon sutures.

Tendons assigned to group 1 underwent core tendon repair only **(Figure 1)**. Following completion of the core tendon repair, the tendons assigned to group 2 had 5 stainless steel skin staples (each staple had a height of 4.8 mm and two 3.4-mm short straight legs) applied equidistant to one another (ie, every 3 to 4 mm) circumferentially around the periphery of the repair site by use of a skin stapling device.[f] Digital counterpressure was used to aid in correct staple application and to maintain the staples in vertical orientation across the repair site. Each staple was applied so that when it was closed, its proximal and distal ends were each 2.4 mm from the repair site and oriented perpendicular to the tendon fibrils. For the tendons assigned to group 3, the core tendon repair was augmented by ESs consisting of 3-0 polypropylene suture[e] that was placed by use of a straight needle[g] in a continuous SXS pattern as described[18] and as commonly performed for clinical patients at our institution. Briefly, the SXS pattern is a modified horizontal mattress suture pattern. To place the pattern, a bite of the epitenon was taken parallel to and approximately 5 mm from the transected edge of the tendon. The suture was then passed obliquely to the opposite side of the repair site where another parallel bite of the epitenon was taken parallel to and approximately 5 mm from the transected edge of the tendon. When the pattern was completed, each strand of suture that crossed the core tendon repair site also crossed 2 other similar strands. Suture bites in the epitenon were placed 2 to 3 mm apart. The SXS pattern was finished with a square knot followed by 4 square throws; the ends of the suture were cut 5 mm from the knot.

All suture materials used in the experimental constructs were sterile, originated from the original manufacturer packaging, and were used before the expiration date stamped on the package. Each package of suture material was opened immediately before use. One board-certified small animal veterinary surgeon (DJD) with experience in tendon repair in both clinical and research settings performed all experimental constructs. All tendon specimens were kept moist by use of physiologic saline solution, which was applied with a spray bottle as necessary throughout creation of the experimental constructs and biomechanical testing.

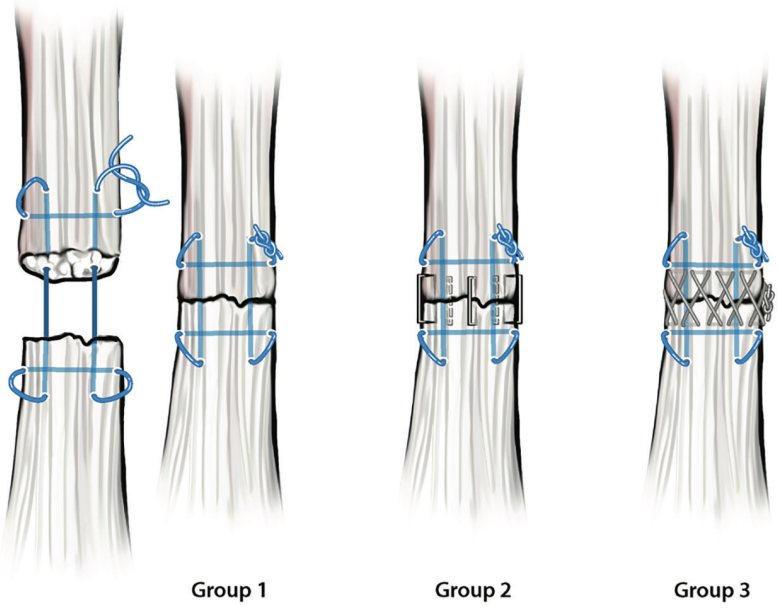


**Figure 1**—Schematic depiction of the 3 experimental groups in an ex vivo study conducted to compare the biomechanical strength and incidence of gap formation among canine SDFT constructs that underwent core tenorrhaphy only (group 1) and those in which the core tenorrhaphy was augmented with skin staples (group 2) or a continuous SXS suture pattern (group 3) placed in the epitenon. Tendons were randomly assigned to 3 groups (14 SDFTs/group) and sharply transected 3 cm distal to the musculotendinous junction. All constructs underwent a core tenorrhaphy that consisted of size-0 polypropylene suture placed in an LL pattern. For the constructs in group 2, the LL suture was augmented by the placement of 5 stainless steel skin staples (each staple had a height of 4.8 mm and two 3.4-mm short straight legs) equidistant to one another (approx every 3 to 4 mm) in the epitenon. When each staple was closed, its ends were 2.4 mm from the transected ends of the tendon. For the constructs in group 3, the LL suture was augmented by use of 3-0 polypropylene suture and a straight needle to place a continuous SXS pattern in the epitenon. The suture bites into the epitenon were positioned 5 mm from the transected ends of the tendon.

## Biomechanical testing

Biomechanical testing of each experimental construct was performed at room temperature by use of a uniaxial materials tensile testing machine.[h] For each construct, the distal aspect of the humerus was rigidly attached to a custom jig[i] with a 4.5-mm-diameter stainless steel bolt that was placed in a mediolateral direction through the supratrochlear foramen. The testing jig was connected to a 500-N load cell that was mounted on the crosshead of the machine. The distal aspect of the manus was affixed securely in a custom bone clamp[i] that was modified by one of the investigators (DJD). The musculotendinous portion of the construct was oriented in the machine in a vertical position to mimic the in vivo load direction for a forelimb that was splinted immediately after surgery to repair a flexor tendon **(Figure 2)**. A preload of 2 N was applied to each construct, and the resulting elongation was measured and recorded as the zero point to ensure that biomechanical testing was initiated under consistent conditions for all constructs. Each construct was distracted at an extension rate of 20 mm/min until failure. Load and displacement data were recorded at a frequency of 100 Hz by the test system software.[j] Each biomechanical test to failure was video recorded by use of a high-definition digital camera[k] that was synchro-

nized with the test system software and positioned 15 cm from the tenorrhaphy site. A calibrated metric ruler was positioned parallel to the construct within the field of view for the camera to facilitate measurement of gap formation when present.

The testing software created a load-displacement curve for each construct from which the yield, peak, and failure loads were calculated. Yield load was defined as the greatest load observed before any initial sharp decrease in the load-displacement curve. Peak load was defined as the maximum load measured during the test. Failure load was defined as the load at which the suture or staple failed (ie, broke or pulled through the tendinous tissue). A customized program[l] was developed to assist with identification of those loads. For each construct, the mechanism of failure was visually observed and recorded at the time of testing and confirmed during review of the video of the test by 1 investigator (Y-JC). Video data were also used to assess gap formation for each construct. Briefly, a digital caliper within an imaging software program[c] was calibrated to the ruler within the video field and used to identify gaps of 1 and 3 mm between the ends of the tendon segments at the repair site. Video images were assessed sequentially to determine the exact times when gaps of 1 and 3 mm were present at the repair site. Those data were cross-referenced with the machine load data to determine the load required to create 1- and 3-mm gaps at the repair site of each construct.

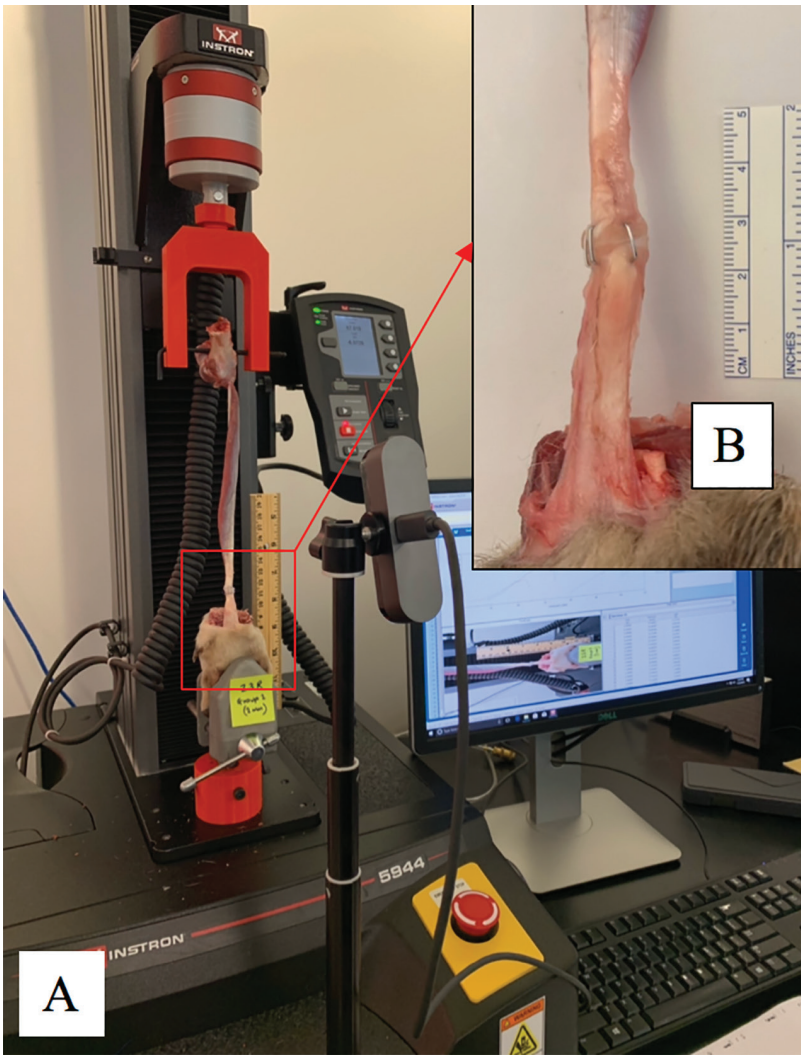


**Figure 2**—Photograph of the uniaxial materials tensile testing machine used to conduct a single load-to-failure test on each construct described in Figure 1 (A) and a close-up photographic image of a group 2 construct mounted in the machine (B). For each construct, the distal aspect of the humerus was rigidly attached to a custom jig with a 4.5-diameter stainless steel bolt that was placed in a mediolateral direction through the supratrochlear foramen. The testing jig was connected to a 500-N load cell that was mounted on the crosshead of the machine. The distal aspect of the manus was affixed securely in a custom bone clamp. The musculotendinous portion of the construct was oriented in a vertical position to mimic the load direction for a forelimb in vivo. A preload of 2 N was applied to each construct, and the resulting elongation was measured and recorded as the zero point to ensure that biomechanical testing was initiated under consistent conditions for all constructs. Each construct was distracted at a rate of 20 mm/min until failure. Load and displacement data were recorded at a frequency of 100 Hz by the test system software. Each test was video recorded by use of a high-definition digital camera that was synchronized with the test system software and positioned 15 cm from the tenorrhaphy site. A calibrated metric ruler was positioned parallel to the construct within the field of view for the camera to facilitate measurement of gap formation.

## Statistical analysis

Data were assessed for normality by use of the Shapiro-Wilk test. All continuous variables were normally distributed; therefore, results were summarized as the mean ± SD. Mixed linear models were used to assess factors associated with yield, peak, and failure loads and the loads required to create 1- and 3-mm gaps. Each model included fixed effects for experimental group and tendon CSA and a random effect to account for evaluation of SDFTs from both forelimbs of each cadaver (ie, repeated measures). When necessary, post hoc pairwise comparisons were performed with the Bonferroni adjustment for multiple comparisons. The mode of failure (suture breakage, staple deformation, or tissue failure [sutures or staples pulled through the tissue]) and proportion of constructs that developed 1- and 3-mm gaps at the tenorrhaphy site prior to failure were compared among the experimental groups by Pearson $\chi^2$ tests of association. Values of $P < 0.05$ were considered significant for all analyses. All analyses were performed by use of a commercial statistical software program.[m]

# Results

## Experimental constructs

All SDFTs were successfully transected and sutured. Biomechanical testing and collection of load-displacement data were performed without technical error, and no constructs were excluded from the analysis because of visible musculotendinous or bone defects. The distribution of left and right forelimbs did not differ significantly ($P$ = 0.909) among the 3 experimental groups. The mean ± SD CSA for all SDFTs was 0.26 ± 0.06 $cm^2$ and did not differ significantly between SDFTs harvested from the same cadaver ($P$ = 0.798) or among experimental groups ($P$ = 0.421).

## Biomechanical testing

Mean ± SD yield, peak, and failure loads differed significantly among the 3 groups and were consistently greatest for group 3 and lowest for group 1 **(Table 1)**. The mean ± SD loads required to create 1- and 3-mm gaps at the tenorrhaphy site also differed significantly among the 3 groups and were greatest for group 3 and lowest for group 1 **(Table 2)**. The proportion of constructs that developed 1-mm gaps prior to failure did not differ significantly ($P$ = 0.317) among the 3 groups; however, the proportion of constructs that developed 3-mm gaps prior to failure in group 1 (14/14) was significantly greater ($P$ = 0.024) than that in groups 2 (9/14) and 3 (6/14).

**Table 1**—Mean ± SD load at yield, peak, and failure for ex vivo canine SDFT constructs that underwent core tenorrhaphy only (group 1; n = 14) and those in which the core tenorrhaphy was augmented with skin staples (group 2; 14) or a continuous SXS suture pattern (group 3; 14) placed in the epitenon.

| Group | Yield load (N) | Peak load (N) | Failure load (N) |
|---|---|---|---|
| 1 | 53.4 ± 15.7 | 69.2 ± 9.4 | 69.1 ± 9.4 |
| 2 | 94.6 ± 22.4 | 109.0 ± 20.3 | 106.9 ± 20.0 |
| 3 | 138.8 ± 31.0 | 152.8 ± 32.2 | 150.2 ± 29.6 |

Each construct underwent a single load-to-failure test at a distraction rate of 20 mm/min, and load and displacement data were recorded at a frequency of 100 Hz. Yield load was defined as the greatest load observed before any initial sharp decrease in the load-displacement curve. Peak load was defined as the maximum load measured during the test. Failure load was defined as the load at which the construct failed (ie, the suture broke, staples became deformed, or the suture or staple was pulled through the tendinous tissue). The mean ± SD differed significantly ($P$ < 0.001) among the 3 groups for all loads.

The number of constructs that failed because of suture breakage (groups 1 and 3) or staple deformation (group 2) was 3, 10, and 12 for groups 1, 2, and 3, respectively. The number of constructs that failed because the suture or staple pulled through the tissue (ie, tissue failure) was 11, 4, and 2 for groups 1, 2, and 3, respectively.

# Discussion

Results of the present ex vivo study supported our a priori hypotheses that, for experimentally transected canine SDFTs, augmentation of core tendon repair by the placement of staples in the epitenon circumferentially across the tenorrhaphy site (epitendinous staples) significantly increased the biomechanical strength of the repair and decreased the incidence of gap formation between the transected tendon ends. Compared with core tendon repair by placement of an LL suture alone, augmentation of the repair by placement of epitendinous staples increased the yield, peak, and failure loads of the repair by 77%, 57%, and 55%, respectively. However, augmentation of the core tendon repair by placement of epitendinous staples was inferior to augmentation of the repair by ESs placed in a continuous SXS pattern in terms of both biomechanical strength and resistance to gap formation.

For the group 2 constructs of the present study, 5 epitendinous staples were placed equidistant around the tenorrhaphy site. The finding that those staples increased the strength of the tenorrhaphy by > 50%, compared with core suture placement alone, was consistent with results of other studies[18,34–36] in which the core tendon repair was augmented with ESs. To our knowledge, the present study was the first to describe the use of epitendinous staples for augmentation of core tendon repair. Although we achieved our primary goal of demonstrating the application and potential benefits of epitendinous staples for augmentation of core tendon repair, it is evident that further research is necessary to determine the optimal staple size, height, number, and configuration to achieve biomechanical properties similar to those achieved with ESs before epitendinous staples are used for in vivo applications.

Although epitendinous staple placement significantly improved the biomechanical strength and decreased the incidence of gap formation relative to

**Table 2**—Mean ± SD force required to induce 1- and 3-mm gaps at the tenorrhaphy site and the incidence rate of such gaps for the 3 groups of experimental constructs described in Table 1.

| | 1-mm gap | | 3-mm gap | |
|---|---|---|---|---|
| Group | Force (N) | Proportion of constructs that developed gaps | Force (N) | Proportion of constructs that developed gaps |
| 1 | 37.6 ± 12.1 | 14/14 | 46.7 ± 11.9 | 13/14 |
| 2 | 89.5 ± 15.9* | 14/14 | 100.2 ± 18.7* | 9/14* |
| 3 | 120.4 ± 25.5*† | 12/14 | 148.8 ± 21.4*† | 6/14* |

*Within a column, value differs significantly ($P$ < 0.05) from that for group 1. †Within a column, value differs significantly ($P$ < 0.05) from that for group 2.

core tendon repair alone, it was inferior to placement of ESs in an SXS pattern. Results of a computerized analytic model[37] suggest that the peripheral component of a tenorrhaphy sustains 64% to 77% of the load applied to the repair. The surgical staples used in the group 2 constructs of the present study had a height of 4.8 mm and two 3.4-mm short straight legs. When the staples were closed, the ends were 2.4 mm from the repair site in accordance with the manufacturer's recommendations for staple placement in cutaneous tissues.[22] In comparison, the SXS sutures in the constructs of group 3 were placed 5 mm from the tenorrhaphy site. Results of another study[38] indicate that the tensile strength of a tenorrhaphy increases as the placement of ESs relative to the repair site increases from 5 to 15 mm. Thus, differences in the placement of staples and SXS sutures relative to the transected ends of the tendon may have contributed to the differences in biomechanical strength and gap formation observed between groups 2 and 3.

In the present study, constructs in both groups 2 and 3 were superior to constructs of group 1 (core tendon repair by an LL suture only) in terms of biomechanical strength and resistance to gap formation. For tendons repaired with ESs, the strength of the tenorrhaphy is proportional to the gauge of the suture material used and number of suture strands that cross the repair site.[18,20] Results of a review[20] of the human medical literature likewise suggest that the overall strength of a tenorrhaphy increases and gap formation decreases as the number of suture strands crossing the repair site increases. In the present study, the number of ES strands that crossed the tenorrhaphy site for each construct in group 3 was greater than the number of epitendinous staples that crossed the tenorrhaphy site for each construct in group 2. We believe that increasing the number of epitendinous staples placed around the tenorrhaphy site will increase the biomechanical strength of the repair, and further research is necessary to test that hypothesis.

Resistance to gap formation is an important consideration when selecting the suture material and suture pattern used for tendon repair in dogs. The formation of gaps > 3 mm at the tenorrhaphy site is associated with a decrease in the ultimate strength and rigidity of the repair and an increase in the risk for failure to achieve normal repair strength and stiffness between 10 and 42 days after surgery.[3] In human patients with flexor tendon injuries of the distal extremities, gap formation is a major cause of adhesion formation and a predisposing factor for re-rupture of the tendon at the surgical site within the first 6 weeks following repair.[3,14] In the present study, the mean load required to cause formation of a 1-mm gap at the tenorrhaphy site for constructs in groups 2 and 3 was 2.4 and 3.2 times that for the constructs in group 1, respectively. Similarly, the mean load required to cause formation of a 3-mm gap at the tenorrhaphy site for constructs in groups 2 and 3 was 2.1 and 3.2 times that for the constructs in group 1, respectively. Results of other studies indicate that the overall tensile strength and resistance to gap formation increase as the gauge of suture material used for ESs increases[39] and when deeper bites into the tendon are taken during placement of the ESs.[40] For the constructs of groups 2 and 3 of the present study, care was taken to keep the depth of staple and SXS suture placement into the epitenon as constant as subjectively possible. Nevertheless, the bite depth for the epitendinous staples in the group 2 constructs was less than the bite depth for the ESs in the group 3 constructs, which may also have contributed to the differences in biomechanical strength and resistance to gap formation observed between the 2 groups. Regardless, the findings of the present study should be interpreted with caution because the resistance to gap formation for a tendon repair method is dependent on many factors such as the suture material type, gauge, and pattern used for the tenorrhaphy. Repair site strength is also reliant on suture-tissue interactions and the resulting biomechanical properties of the construct.[18] Focused research is warranted regarding the association between depth of penetrance into tendinous tissue by ESs or epitendinous staples and the resulting biomechanical and healing properties of repaired tendon constructs.

Epitendinous suture constructs fail because the suture breaks or pulls through the tissue.[18,36,39] In another ex vivo study[15] of various tenorrhaphy constructs for repair of SDFT injuries in dogs, the suture pulling through the tissue was the primary mode of construct failure. In the present study, the majority (10/14) of group 2 constructs failed as a result of staple deformation. Staple deformation likely occurred because the strength required to overcome the staple-tissue interaction exceeded the material properties of the closed stainless steel staples. The use of staples made from stronger and less deformable material than stainless steel might have resulted in different results and warrants further research.

Knots are the weakest part of suture lines placed in the skin.[41] However, in the present study, the mode of failure for the majority (12/14) of group 3 constructs was ES breakage, rather than knot failure or the suture pulling through tissue. That finding was likely associated with the distribution of tension within the suture and the interaction between the suture material and tissue. Augmentation of a core tendon repair by placement of ESs around the periphery of the repair results in a greater proportion of the CSA of the tendon being used, including a larger number of collagen fibrils being incorporated into the repair, which allows the tension and load to be distributed and shared among the tendon substance, core suture, and ESs.[16,18] Augmentation of a core tendon repair by placement of epitendinous staples results in load sharing in a similar manner. However, when the force required to overcome the staple-tissue interaction is exceeded, there is asymmetric loading of the repair construct that does not occur with the use of a con-

tinuous ES pattern. This results in a greater proportion of the load being transferred to the core suture, which in turn leads to the core suture being pulled through the tissue or breakage of the construct at the stress-concentration point of the staple-tendon interface.

Although augmentation of a core tendon repair by placement of epitendinous staples was not as effective as placement of ESs in a continuous SXS pattern, the epitendinous staples did significantly increase the biomechanical strength of and decrease gap formation at the repair site, relative to placement of a suture in the core tendon without any epitendinous augmentation. Subjectively, placement of epitendinous staples also improved apposition of the transected tendon ends. Further investigation is necessary to determine whether use of an interlocking staple design or interdigitation of the staple ends in the epitenon may further enhance construct strength and resistance to gap formation at tenorrhaphy sites.

Limitations of the present study included its ex vivo nature. The effects of important factors associated with in vivo surgical repair of tendon injuries, such as inflammatory processes, degenerative changes within the tendon, and damage to the tendon blood supply, on the biomechanical properties of the repair constructs could not be evaluated in this study. Also, although a concerted effort was made to ensure that the depth of tendon penetration by epitendinous staples and ESs was consistent during construct creation, it was not formally evaluated. Variation in bite depth of ESs and epitendinous staples among constructs might have contributed to some of the differences observed because results of other studies indicate that suture bite depth is positively correlated with construct strength and resistance to gap formation.[39,40] All constructs underwent a single load-to-failure test at a distraction rate of 20 mm/min to simulate the application of sufficient acute stress to cause clinical failure of the repaired tendon. Cyclical testing of specimens at physiologic loads would have resulted in more accurate representation of in vivo conditions.[42] All SDFT specimens used in this study were fairly homogenous in size and were obtained from dogs that were free of clinical musculoskeletal disease and weighed between 28 to 35 kg. Therefore, the results may not be applicable to small dogs with smaller SDFTs or SDFTs with degenerative changes that might impair the suture holding capacity of the tissue. Finally, we did not evaluate glide function in this study. Materials (eg, sutures or staples) can protrude from a surgical repair and interact with adjacent soft tissue and impede its movement. Glide function is a measure of how easily the construct moves relative to adjacent tissues. It is an important concept in the repair of tendons of the human hand but is likely of less clinical concern for tenorrhaphy of the SDFT in dogs because fine motor control of the manus is less refined in dogs than it is in humans.

Findings of the present ex vivo study indicated that, for experimentally transected canine SDFTs, augmentation of a core tendon repair by placement of epitendinous staples significantly increased the biomechanical strength and resistance to gap formation for the resulting construct, relative to constructs consisting of core tendon repair by placement of an LL suture alone. However, augmentation of the core tendon repair by placement of epitendinous staples was inferior to augmentation of the repair by ESs placed in a continuous SXS pattern. Further research is necessary to determine the optimal staple composition, size, number, design, and configuration for use as epitendinous augmentation of core SDFT repairs. In vivo studies are necessary to assess the effect of epitendinous staple placement on tendon healing, blood supply, glide function, and peritendinous adhesion formation prior to implementation in clinical patients.

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The authors declare that there were no conflicts of interest.

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## Footnotes

a. Ziplock 1-gallon bags, SC Johnson and Son Inc, Racine, Wis.
b. iPhone SE, Apple Inc, Cupertino, Calif.
c. ImageJ, National Institutes of Health, Bethesda, Md.
d. Research Randomizer, version 4.0, Geoffrey C. Urbaniak and Scott Plous. Available at: www.randomizer.org. Accessed Oct 4, 2019.
e. Covidien Ltd, Dublin, Ireland.
f. Appose ULC Auto SutureTM Slim body skin stapler 35, Covidien Ltd, Dublin, Ireland.
g. Monoject 20-gauge needle, Kendall Healthcare, Mansfield, Mass.
h. 5944 Universal Testing System, Instron Inc, Norwood, Mass.
i. Sawbones, Vashon Island, Wash.
j. Bluehill 3, Instron Inc, Norwood, Mass.
k. Brio 4k Webcam, Logitech Inc, Newark, Calif.
l. Matlab R2018b, Mathworks, Natick Mass.
m. SAS, version 9.4, SAS Institute Inc, Cary, NC.

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