Surgical repair options for ligamentous injury of the canine tarsal joint are limited. Current recommendations include a primary repair technique, replacement of the MCL with a prosthetic suture, use of temporary transarticular external skeletal fixation, or pantarsal arthrodesis to stabilize the joint.1–3 

The currently recommended prosthetic suture technique involves the use of cortical screws with washers and nylon suture.2,3 Inherent limitations of that technique may lessen its effectiveness or increase the risk for complications. Specifically, all elements of the repair are ipsilateral with the injury. Often the ligamentous injury is accompanied by substantial soft tissue loss in the region, which makes skin closure over the implants difficult if not impossible.4 In the authors’ experience, the implants can have a raised profile that increases the potential for trauma to the overlying skin, which can lead to chronic wounds or implant exposure. Postoperative complications involving the implants (eg, infection, irritation, or loosening) often require implant removal.3 Implant removal may damage the fibrous tissue that formed to aid in joint stability, thereby compromising any stability gained by the original surgical procedure.5 

A review of the current literature found only a few articles3–7 regarding treatment or outcomes of ligamentous tarsal joint injuries in dogs and cats. There are 2 cadaveric studies5,7 in the recent literature in which tarsal joint stability and potential surgical options were assessed, but objective force testing to determine repair strength was not included in either study. In 3 studies,5,4,6 the clinical outcome of tarsal joint repair was assessed for 51 dogs collectively, and results indicate that the repair technique was not significantly associated with clinical outcome. However,
the outcome measures for those 3 studies were subjective and relied primarily on owner satisfaction. In the study by Beever et al.,3 a prosthetic ligament was associated with a high complication rate and generally poor functional outcome regardless of the stabilizing technique or suture material used. Although a definitive difference in clinical outcome has not been identified between repairs that maintain joint function and those that do not, intuitively, it would seem that maintenance of joint function to as near normal as possible would have inherent advantages for the patient.

The purpose of the study reported here was to compare the relative stability and ultimate strength of a traditional prosthetic ligament construct to those for each of 3 recently developed prosthetic ligament constructs. We hypothesized that there would be no difference in stability between tarsal joints with an intact MCL and those stabilized with prosthetic ligaments or among the 4 prosthetic ligament constructs evaluated. We also hypothesized that implant stability following cyclic range-of-motion testing and ultimate strength would not differ among the prosthetic ligament constructs evaluated.

Materials and Methods

Tarsal joint specimens

The use of canine cadavers for the study was approved by the University of Florida Institutional Animal Care and Use Committee. Prior to study initiation, a sample size calculation was performed to determine how many limbs would be required for each construct (ie, MCL repair technique) group. For that calculation, it was estimated that the mean tarsal joint laxity following application of the repair technique would be 10° greater than that for intact (clinically normal) joints. Results indicated that at least 6 specimens would need to be included in each group.

Thirteen cadavers were random sourced; therefore, patient history was unavailable. Although the specific breed and age could not be determined for any of the cadavers, visual assessment indicated that all dogs were nonchondrodystrophic, weighed at least 15 kg, and were estimated to be skeletally mature on the basis of dentition and inspection of the tibia at the time of specimen collection. Both hind limbs were harvested from each cadaver. The limbs were disarticulated and removed at the stifle joint. All soft tissues of the tibial and tarsal regions were removed except for the structural components of the tarsal joint, such as the joint capsule, collateral ligaments, and relevant retinacular elements of the joint. The limbs were then wrapped in towels soaked with saline (0.9% NaCl) solution and stored at -20°C until analysis 1 to 3 days later.

Study design

The traditional prosthetic ligament repair technique (AN construct) consisted of 3 bone anchors connected with monofilament nylon suture. The 3 more recently developed repair techniques included the use of low-profile suture anchors made of PEEK or titanium that were connected with multifilament UHMWPE suture (AU construct) and the creation of bone tunnels and use of monofilament nylon suture (TN construct) or UHMWPE suture (TU construct) for joint stabilization.

Paired limbs (ie, both hind limbs from the same cadaver) were randomly assigned by means of a coin toss to either the anchor (AN and AU; n = 7 pairs) or tunnel (TN or TU; 6) groups. Within each of those groups, 1 limb from each pair was assigned to each repair technique such that there was a minimum of 3 left and 3 right tarsal joints repaired by each of the 4 constructs. There were 7 replicates for both the AN and AU constructs and 6 replicates for both the TN and TU constructs.

Objective measures were compared among the 4 constructs. Tarsal joint stability was tested under assumed physiologic loads with the ligament under minimal direct strain. Tarsal joint strength was tested at potential maximal loads that were expected during abnormal weight bearing owing to external lateral or medial force exerting a substantial direct load on the MCL (or prosthetic ligament). Normal physiologic motion and its potential effects on the stability of the prosthetic ligament was tested by cyclically moving the joint through the complete range of motion.

Description of prosthetic ligament constructs

All prosthetic constructs were performed by the same investigator (MDJ). After the limbs were thawed, a complete tarsal MCL injury was simulated. Each limb was then repaired by use of the assigned construct.

To maximize the uniformity of bone tunnel and bone anchor placement for all constructs, an aiming device was used to place an initial stainless steel guide pin (diameter, 0.9 mm) for each tunnel and anchor. For each limb assigned to the TU and TN constructs, 3 bone tunnels were created by use of a drill with a 2.0-mm cannulated drill bit, which was placed over the guide pin. The proximal tunnel corresponded to the origin of the MCL. This tunnel was drilled beginning at the medial aspect of the distal portion of the tibia (distal tibia) and was oriented proximally in a craniodistal direction so that it emerged at the craniodistal aspect of the tibia immediately cranial to the fibula (Figure 1). One of the 2 distal tunnels corresponded to the insertion of the long branch of the MCL on the distal aspect of the talus. That tunnel was drilled beginning on the medial aspect of the talus and was oriented in a mediolateral and slightly plantar direction parallel to the proximal intertarsal joint and emerged on the lateral aspect of the calcaneus. The other distal tunnel corresponded to the insertion of the short branch of the MCL at the caudal aspect of the proximal portion of the talus. That tunnel was drilled beginning at the medial aspect of the talus and was oriented in a...
Figure 1—Representative illustrations of the craniocaudal (A and C) and lateral (B and D) aspects of the canine tarsal region that depict bone or suture anchor placement (A and B) and bone tunnel location and orientation (C and D; red lines) for 4 prosthetic ligament constructs for treatment of tarsal MCL injury. For the AN and AU constructs, 3 partial-thickness holes were drilled into the tibia and talus for placement of bone or suture anchors. One hole was drilled into the medial aspect of the distal portion of the tibia (distal tibia) at the origin of the MCL and was oriented in a craniolateral direction. Two holes were drilled into the talus. One hole was drilled at the insertion of the long branch of the MCL on the medial aspect of the distal portion of the talus and was oriented in a mediolateral and slightly plantar direction, whereas the other was drilled at the insertion of the short branch of the MCL on the caudal aspect of the proximal portion of the talus and was oriented in a distolateral-plantar direction. For the TN and TU constructs, bone tunnels were created through the tibia and talus-calcaneus. The tunnels were located and oriented as described for the holes for the AN and AU constructs except the holes were extended until the drill emerged on the lateral aspect of the tarsal region to create a tunnel through the bone or bones. The tunnel in the distal tibia emerged at the craniolateral aspect of the bone immediately cranial to the fibula. The 2 distal tunnels emerged 3 to 5 mm apart on the lateral aspect of the calcaneus. Medial is to the right in panels A and C.
distolateral-plantar direction so that it emerged from the lateral aspect of the calcaneus 3 to 5 mm from the emergent point of the other distal tunnel. Because many fibers of the short branch of the MCL attach to the talus underneath the medial malleolus of the tibia, care was taken to ensure that the tunnel did not interfere with the medial malleolus.

For each limb assigned to the AN and AU constructs, 3 holes were drilled as described for the TN and TU techniques except the holes were not extended to emerge on the lateral aspect of the limb. The medial aspect of each hole was enlarged to accommodate the bone or suture anchor. For the AN construct, the talar anchor sites were enlarged with a 2.5-mm drill bit and 3.5-mm tap to a depth sufficient to allow the anchors to be placed flush with the bone surface. The tibial anchor site was enlarged with a 3.2-mm drill bit and 4.5-mm tap; however, the shape of the eyelet on the bone anchor used for the AN technique prevented it from being countersunk so that it was flush with the bone surface. Two 4.0-mm stainless steel mini bone anchors were placed in the holes in the talus, and one 4.5-mm stainless steel bone anchor was placed in the proximal hole in the distal tibia. For the AU construct, the tibial anchor site was enlarged with a 4.0-mm spade-tip drill bit followed by a 4.75-mm push-tap as recommended by the anchor manufacturer; the holes in the talus did not need to be enlarged. One 4.75-mm PEEK suture anchor was placed in the hole in the talus, and two 2.8-mm titanium suture anchors were placed in the holes in the tarsus.

For all 4 constructs, suture was threaded through the anchors or bone tunnels in a manner to mimic the long and short branches of the MCL. For the AN construct, 40-lb monofilament nylon leader line was used, and the ends of the sutures were secured with 40-lb crimp tubes, which were applied with a metallic crimping system (Figure 2). For the AU construct, a double-strand of No. 2 UHMWPE suture was threaded through the anchors. For the TN construct, 40-lb monofilament nylon leader line was used in conjunction with 2 stainless steel toggle rods; the sutures were secured with 40-lb crimp tubes as described for the AN construct. For the TU construct, a 4-hole titanium button and oblong 2.7-mm titanium button were placed over the lateral aspect of the tunnels in the tibia and talus-calcaneus, respectively, through which 2 strands of No. 2 UHMWPE suture were threaded.

The tarsal joint was placed in flexion and extension for tightening of the sutures mimicking the short and long branches of the MCL, respectively, to simulate normal function of the ligament. For the TU construct, a tensiometer was used to tighten each suture to 12 Nm in accordance with the manufacturer’s recommendation for use of that suture system in other ligament reconstructions (eg, stifle and hip joints). For the AN and TN constructs, sutures were tightened with a universal tensioner, but the tension placed on the sutures could not be measured with that instrument.

### Biomechanical testing

**Construction of the biomechanical testing apparatus**—All mechanical testing was performed by use of a biaxial servohydraulic test system. To objectively measure the stability and strength of the intact tarsal MCL and the prosthetic ligament constructs, a custom-fabricated motorized apparatus was placed in the workspace of the materials testing system. The apparatus could hold the tibia in a desired position for load testing and move it through its natural range of motion as needed. The apparatus consisted of 2 plastic plates, 4 aluminum columns, a direct-current motor with a belt-driven spindle and rotational encoder, a data acquisition unit, and a tibial fixture. The motorized spindle passed through a hole in the upper plate and attached directly to the tibial fixture, which consisted of an aluminum beam with a V-shaped channel that allowed the tibial segment to be supported in an appropriate orientation and securely fixed so it could be freely moved throughout the normal physiologic range of motion (Figure 3).

Each limb was positioned in the fixture so that it was oriented parallel to the floor with the medial aspect of the limb facing downward and the tarsus positioned directly over the spindle axis of rotation. The paw was placed between guideposts that restricted flexion and extension in the sagittal plane but allowed it to move freely in the varus-valgus direction when attached by a ring clamp to the loading frame (Figure 3). Medial joint stability was assessed by the application of vertical tension (valgus stress) and measurement of the resultant vertical displacement of the distal aspect of the limb (distal limb). Angular displacement of the distal limb was calculated from the linear displacement of the paw clamp and the measured distance of the clamp from the center of the tarsal joint. The motorized fixture allowed the tibia to be moved to any desired tarsal joint angle and cycled through its range of motion without having to readjust the position of the limb.

**Pilot project to determine the maximal tension for application during biomechanical testing**—Prior to study initiation, a small pilot project was performed to determine an acceptable tension for application by the materials testing system to assess the laxity of unaltered tarsal MCLs without causing permanent deformation or impairing joint stability. Four frozen cadaveric canine hind limbs (these limbs were not assigned to any of the 4 constructs in this study) were thawed. Each limb was placed in the testing apparatus and subjected to increasing levels of static tension (25, 50, 75, and 100 N). After each predefined static tension level was achieved, the limb was reassessed at 25 N to evaluate whether permanent MCL deformation had occurred. If the extent of displacement at 25 N was greater than the extent of displacement at the immediately preceding static tension level, that level was...
Figure 2—Photographs of the medial aspect of the tarsal region of canine cadaveric hind limbs following transection of the tarsal MCL and completion of each of 4 prosthetic ligament constructs. A—The AN construct (traditional repair method) included a 4.5-mm stainless steel bone anchor that was placed at the origin of the MCL on the medial aspect of the distal tibia and two 4.0-mm stainless steel mini bone anchors that were placed at the insertions of the long and short branches of the MCL on the medial aspect of the talus. The talar anchors were placed flush with the bone surface, but the eyelet on the tibial anchor prevented it from being placed flush with the bone surface. Forty-pound monofilament nylon leader line was threaded through the bone anchors to mimic the long and short branches of the MCL. B—The AU construct included a 4.75-mm PEEK suture anchor that was placed in the distal tibia at the origin of the MCL and two 2.8-mm titanium suture anchors that were placed in the talus at the insertions of the long and short branches of the MCL. A double strand of No. 2 UHMWPE suture was threaded through the anchors in the same manner as that described for the AU construct. C—The TN construct involved the creation of 3 bone tunnels from the medial to lateral aspects of the tarsal region as described in Figure 1. Forty-pound monofilament nylon leader line was threaded through the holes to mimic the long and short branches of the MCL; the suture was also passed through stainless steel toggle rods. D—The TU construct was similar to the TN construct except No. 2 UHMWPE suture was used as the prosthetic ligament and the suture was also threaded through a 4-hole titanium button and oblong 2.7-mm titanium button that were placed over the lateral openings of the tunnels in the tibia and talus-calcaneus, respectively. All nylon sutures were secured with 40-lb crimp tubes.
considered to have caused permanent deformation. Data were reviewed to determine the maximal force applied that did not result in permanent MCL deformation; that force was 50 N.

Testing of study limbs—All 26 study limbs underwent biomechanical testing with the tarsal MCL intact, after the tarsal MCL was transected, and following completion of the assigned construct before and after cyclic range-of-motion testing. The frozen limbs were thawed in plastic bags, which were placed in warm water for 20 minutes before the initial biomechanical test. Once thawed, each limb was placed in the servohydraulic testing system, and the stability of the intact MCL was determined at each of 3 joint positions (75° [midflexion], 135° [standing angle], and 165° [full extension]) as described4,9 with a static load of 50 N (Figure 4). At each position, the extent of displacement from neutral was measured (in mm) by the servohydraulic testing apparatus 3 times.

After the stability of the intact MCL was measured, the soft tissues crossing the medial talocrural joint (ie, long and short branches of the MCL, caudal tibial muscle tendon, and medial aspect of the joint capsule) were transected to simulate a complete MCL injury. Each limb was then reassessed as previously described for the intact MCL. For many limbs, the joint became very unstable at a static load < 50 N, and testing was discontinued because of concerns about damage to other joint structures. The maximum static load applied was recorded, and the extent of ligament displacement from neutral was measured 3 times at each of the 3 joint positions.

After the assigned construct was completed on each limb, the range of motion for the prosthetic ligament was manually assessed by use of a universal goniometer for deficits that might affect range-of-motion testing. All repaired tarsal joints could be flexed to at least 75° and extended to at least 165°. Each repaired limb was then mounted in the servohydraulic test system, and the stability of the tarsal joint was assessed and recorded as previously described. Each limb then underwent approximately 1,000 cycles of the prescribed (physiologic) range of motion (75° through 165°) for a canine tarsal joint at a frequency of 1 cycle/1.25 s for a total of 20 minutes to approximate early joint mobilization with the prosthetic ligament. Following completion of cyclic range-of-motion testing, the stability of the tarsal joint was reassessed as previously described. Tissues were kept moist by external application of saline solution throughout mechanical testing and cyclic range-of-motion testing.

Following completion of tarsal joint stability testing, each limb was tested to failure. The limb was positioned so that the tarsal joint was at a standing angle (135°), then tension (axial distraction) was applied to the limb at a gradually increasing rate of 10 N/s until the prosthetic ligament construct failed or a maximum tension of 450 N was achieved (which was dictated by the physical limitations of the support materials of the testing apparatus). Failure was considered to have occurred when the construct provided < 80% of its maximal resistance. Because many of the constructs failed at a large angle of deviation (ie, an angle that was not physiologically possible), failure was assessed by another method. For each limb, the tarsal joint deviation recorded during mechanical testing of the limb before and after transection of the MCL was used to define 2 potential end points (midpoint and maximal angles of deviation). Midpoint angle of deviation was defined as clinical failure and was calculated as the angle midway between the measured joint angle deviation when the MCL was intact and that when the ligament was transected. Maximal angle of deviation was defined as the greatest deviation in joint angle that resulted in complete loss of joint stability similar to that when the
MCL was transected. The forces required to achieve the midpoint and maximal angles of deviation were recorded, as was the method of failure (implant failure or bone fracture).

**Statistical analysis**

Because tarsal joint laxity was not uniform among limbs, direct comparison of joint deviation among limbs within each group was not possible. Therefore, the change in tarsal joint laxity following repair was determined for each limb (ie, joint laxity following repair – joint laxity when the MCL was intact), and that measure was used as the dependent variable (outcome of interest). The change in tarsal joint laxity was compared among the 4 constructs with a mixed general linear model. The model included fixed effects for construct (AN, AU, TN, or TU), ligament status (transected, repaired before cyclic range-of-motion testing, or repaired after cyclic range-of-motion testing), tarsal joint angle (75°, 135°, or 165°), and all possible 2- and 3-way interactions. The model also included a random error term to account for repeated measures within each limb, and a first-order autoregressive correlation structure was used to model correlation between the repeated measures within a limb. When multiple post hoc pairwise comparisons were necessary, the Tukey-Kramer method was used to control for type I error inflation.

Data for the ultimate strength (ie, force required for construct failure) and forces required for midpoint and maximal angles of deviation were not normally distributed. Therefore, a Wilcoxon rank sum test was used to compare each of those outcomes among the 4 constructs. The Dunn method was used when post hoc pairwise comparisons were necessary. Descriptive data were generated for mode of failure within each construct group. All statistical analyses were performed with commercially available statistical software programs, and values of \( P < 0.05 \) were considered significant.

**Results**

**Tarsal joint stability**

The mixed general linear model indicated that the mean change in tarsal joint laxity following prosthetic ligament repair did not differ significantly among the 4 constructs or from the intact or transected ligament specimens when the joint was positioned at a 75° angle during biomechanical testing. Therefore, data for the limbs when the tarsal joint was positioned at a 75° angle were removed from the analysis, and all further results and discussion will concern tarsal joint angles of 135° and 165°.

The mean change in tarsal joint laxity was not significantly associated with construct \( (P = 0.17) \), the 2-way interaction between ligament status and construct \( (P = 0.99) \), the 2-way interaction between tarsal joint angle and construct \( (P = 0.41) \), or the 3-way interaction among ligament status, tarsal joint angle,
and construct \( (P = 0.22) \). However, it was significantly associated with ligament status \( (P < 0.001) \), tarsal joint angle \( (P < 0.001) \), and the interaction between ligament status and tarsal joint angle \( (P < 0.001) \).

When the tarsal joint was positioned at a 135° angle, the mean ± SE change in joint laxity relative to that for the intact MCL was 21.1 ± 0.79° for the transected ligament \( (P < 0.001) \), 7.2 ± 0.79° for the prosthetic ligament before cyclic range-of-motion testing \( (P < 0.001) \), and 9.3 ± 0.79° for the prosthetic ligament after cyclic range-of-motion testing \( (P < 0.001) \). The mean ± SE difference for the change in joint laxity between the transected ligament and prosthetic ligament was significantly different from 0 both before \( (15.3 \pm 0.75°; P < 0.001) \) and after \( (11.8 \pm 0.81°; P < 0.001) \) cyclic range-of-motion testing, as was the difference between the change in joint laxity before and after cyclic range-of-motion testing \( (2.1 \pm 0.75°; P < 0.02) \) for the prosthetic ligament.

When the tarsal joint was positioned at a 165° angle, the mean ± SE change in joint laxity relative to that for the intact MCL was 20.0 ± 0.79° for the transected ligament \( (P < 0.001) \), 4.7 ± 0.79° for the prosthetic ligament before cyclic range-of-motion testing \( (P < 0.001) \), and 4.8 ± 0.79° for the prosthetic ligament after cyclic range-of-motion testing \( (P < 0.001) \). The mean ± SE difference for the change in joint laxity between the transected ligament and prosthetic ligament was significantly different from 0 both before \( (13.9 ± 0.75°; P < 0.001) \) and after \( (11.8 ± 0.81°; P < 0.001) \) cyclic range-of-motion testing, but the difference in joint laxity between before and after cyclic range-of-motion testing \( (0.15 ± 0.75°) \) for the prosthetic ligament was not \( (P = 0.98) \).

### Tarsal joint strength

The median tarsal joint laxity and force at construct failure were greatest for the TU construct and were significantly greater than those for both the TN and AN constructs (Table 1). The median tarsal joint laxity and force at failure for the AU construct did not differ significantly from those for any of the other 3 constructs.

The median midpoint and maximal angles of deviation did not differ significantly among the 4 constructs (Table 2). All limbs repaired with UHMWPE suture (AU and TU constructs) remained intact at the midpoint and maximal angles of deviation. Among the 13 limbs repaired with nylon suture (AN and TN techniques), only 11 and 4 remained intact for calcu-

#### Table 1—Descriptive statistics for tarsal joint laxity (angle of deviation relative to that for a joint with an intact MCL) and force at failure for a traditional prosthetic MCL repair construct and each of 3 more recently developed prosthetic MCL repair constructs for dogs.

<table>
<thead>
<tr>
<th>Repair construct</th>
<th>No. of limbs</th>
<th>Tarsal joint laxity at failure (°)</th>
<th>Force at construct failure (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Median (IQR)</td>
<td>Range</td>
<td>Median (IQR)</td>
</tr>
<tr>
<td>AN</td>
<td>7</td>
<td>27.5 (25.4–30.8)(^{ab})</td>
<td>24.2–34.8</td>
</tr>
<tr>
<td>AU</td>
<td>7</td>
<td>33.8 (32.2–39.8)</td>
<td>29.9–43.2</td>
</tr>
<tr>
<td>TN</td>
<td>6</td>
<td>28.1 (25.2–31.0)(^a)</td>
<td>23.4–33.0</td>
</tr>
<tr>
<td>TU</td>
<td>6</td>
<td>42.7 (38.6–44.8)</td>
<td>35.8–49.2</td>
</tr>
</tbody>
</table>

- *All repair constructs were performed in cadaveric canine hind limbs after the long and short branches of the MCL were transected. The AN construct (traditional repair method) consisted of 3 bone anchors connected with monofilament nylon suture. The 3 more recently developed repair constructs included the use of low-profile suture anchors made of PEEK or titanium that were connected with multifilament UHMWPE suture (AU construct) and the creation of bone tunnels and use of monofilament nylon (TN construct) or UHMWPE (TU construct) suture for tarsal joint stabilization. Failure was considered to have occurred when the construct provided < 80% of its maximal resistance. The force applied to each limb during failure testing could not exceed 450 N owing to physical limitations of the testing jig, and a default force of 450 N was assigned to any construct that did not fail.
- *Value differs significantly \( (P < 0.05) \) from that for the TU construct.
- IQR = Interquartile (25th to 75th percentile) range.

#### Table 2—Descriptive statistics for tarsal joint laxity and force at the midpoint and maximal angles of deviation for the canine hind limbs of Table 1.

<table>
<thead>
<tr>
<th>Repair construct</th>
<th>Midpoint angle of deviation</th>
<th>Maximal angle of deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Laxity (°)</td>
<td>Force (N)</td>
</tr>
<tr>
<td>AN</td>
<td>23.7 (22.9–24.6)</td>
<td>65.2 (52.5–75.9)</td>
</tr>
<tr>
<td>AU</td>
<td>24.0 (23.3–25.5)</td>
<td>71.6 (64.3–89.1)</td>
</tr>
<tr>
<td>TN</td>
<td>21.6 (20.3–22.5)</td>
<td>52.2 (37.0–82.8)</td>
</tr>
<tr>
<td>TU</td>
<td>24.5 (22.2–24.5)</td>
<td>64.9 (55.9–85.2)</td>
</tr>
</tbody>
</table>

- Values represent the median (IQR) unless otherwise indicated. The midpoint angle of deviation (ie, clinical failure) was defined as the angle midway between the measured tarsal joint angle deviation (laxity) when the MCL was intact and that when the ligament was transected. The maximal angle of deviation was defined as the greatest deviation in tarsal joint angle after the MCL was transected.
- *Only 2 replicates from each of the nylon groups (AN and TN) reached the maximal angle of deviation without breaking; therefore, the force for those 2 replicates was provided instead of the IQR.
- See Table 1 for remainder of key.
ulation of the midpoint and maximal angles of deviation, respectively.

**Mode of failure**

The prosthetic ligament failed because of suture failure for all 7 limbs repaired with the AN construct; 1 limb in that group also had bone anchor failure (ie, the talar anchor for the suture mimicking the short branch of the MCL was pulled out of the bone during failure testing). The suture mimicking the long branch of the MCL was the suture that failed most frequently in the AN-repaired limbs (6/7), and it failed at the distal talar anchor, tibial anchor, or crimp for 3 limbs, 2 limbs, and 1 limb, respectively. The suture mimicking the short branch of the MCL failed at the tibial anchor in 1 limb. Both sutures failed in 1 limb.

For the 7 limbs repaired with the AU construct, the prosthetic ligament most frequently failed owing to suture slippage at the tibial anchor \((n = 3)\) and suture failure at a talar anchor \((2)\). No visible laxity was present following failure testing for the remaining 2 limbs.

Among the 6 limbs repaired with the TN construct, the suture mimicking the long branch of the MCL failed at the crimp for 5 and elongated for 1. Among the 6 limbs repaired with the TU construct, the suture mimicking the long branch of the MCL loosened in 5, presumably subsequent to knot slippage because no obvious alteration in the suture was observed. For the remaining TU-repaired limb, the suture mimicking the long branch of the MCL loosened owing to fragmentation of the talar bone tunnel medially and the suture mimicking the short branch of the MCL also loosened. Overall, only 2 of the 13 constructs involving UHMWPE suture (AU and TU constructs) failed because of suture breakage.

**Discussion**

In the present study, the stability and strength of 3 recently developed prosthetic ligament constructs (AU, TN, and TU) for repair of the tarsal MCL of dogs were compared with those for a more traditional prosthetic ligament construct (AN). Results indicated that construct type was not significantly associated with tarsal joint stability. All 4 constructs provided adequate initial stability to the tarsal joint. When the tarsal joint was positioned at 75° (flexion) for biomechanical testing, it was inherently stable (intact, construct, and transected), so much so that those data could not be used to compare joint stability. When the tarsal joint was positioned at angles of 135° (standing position) and 165° (full extension) for biomechanical testing, tarsal joint stability for all 4 constructs was significantly greater than that when the MCL was transected, and although the repaired joints were significantly more lax than joints with an intact MCL, that increase in laxity was < 10°, which is unlikely to be clinically relevant. Collectively, these results suggested that all 4 prosthetic ligament constructs were equally effective in providing tarsal joint stability following an MCL injury.

For all 4 prosthetic ligament constructs, joint laxity increased significantly following cyclic loading when the tarsal joint was positioned at 135° for biomechanical testing but not when it was positioned at 165°. However, that increase was small (mean ± SE, 2.1 ± 0.75°) and might not have a clinical effect. Nevertheless, that increase in laxity would appear to support the practice of joint immobilization in the early postoperative period to maintain joint stability while supportive fibrous tissue formation is occurring. In the present study, each limb underwent only 1,000 loading cycles, which represented only a small fraction of the amount of movement that would be expected with normal use of the limb during the postoperative period. Further research regarding the proper duration for postoperative tarsal joint immobilization or use of a controlled range-of-motion supportive device on the hind limb is warranted.

The recently developed tarsal MCL repair techniques evaluated in the present study may provide certain advantages, compared with the traditional repair technique (AN construct). The bone tunnel techniques (TN and TU constructs) allow the prosthetic ligaments to be anchored on the lateral side of the limb, which might permit implant removal from the lateral aspect of the limb without the risk of damage to potentially joint-stabilizing scar tissue on the medial aspect of the joint. The use of transverse bone tunnels also allows for a low-profile ligament reconstruction, which may facilitate primary closure in patients with limited soft tissue in the tarsal region. The use of low-profile bone anchors on the medial aspect of the tarsal area (AU construct) might also facilitate primary closure over the implants. The AU and TU constructs involved UHMWPE suture, which is used clinically for stabilization of joints other than the tarsal joint and has been proven to have greater ultimate strength than nylon suture in the present study and other types of suture in other cadaveric studies. Bacterial adherence to braided suture material is generally greater than that to smooth or monofilament suture material; consequently, use of braided suture in contaminated wounds is not advised. In a retrospective study of dogs with tarsocrural instability, prosthetic ligaments were significantly associated with the development of major complications, and the prosthetic ligament had to be removed because of major complications in 5 of 8 repaired joints. Interestingly, the use of braided multifilament suture for prosthetic ligaments was not associated with an increased risk, compared with that of monofilament suture, for the development of postoperative complications in that study. Newer prosthetic ligament implant materials, such as UHMWPE suture, suture anchors, and titanium buttons, cost more than traditional materials (ie, nylon suture and bone anchors), and their use in a clinical setting will undoubtedly increase procedure costs for clients.

In the present study, the type of suture material used for the prosthetic ligament appeared to be associated with the overall strength of the re-
pair construct. The median force at failure was greatest for the TU construct followed by the AU construct (constructs that used UHMWPE suture). Moreover, the median force at failure for the TU construct was significantly greater than that for the TN and AN constructs (constructs that used nylon suture). Only 2 of the 13 limbs that were repaired with UHMWPE suture failed because of suture breakage, and biomechanical testing for several of those limbs had to be discontinued prior to construct failure because the force at failure appeared to exceed the force capable of being sustained by the testing apparatus (approx 450 N). Apparent suture strength and force at failure for the AN and TN constructs assessed in the present study were consistent with those reported in other studies. Likewise, differences in the ultimate strength between UHMWPE and nylon prosthetic ligament constructs observed in the present study were similar to those observed in other studies in which those 2 types of suture material were compared. Median joint laxity at the time of failure was greater for the TU and AU constructs than for the TN and AN constructs. However, the force necessary to achieve the midpoint angle of deviation (ie, clinical failure) did not differ significantly among the 4 constructs; hence, it seems reasonable to expect that the 4 repair techniques will behave similarly in patients.

Because the constructs with UHMWPE suture generally failed at a large angle of deviation, we decided to evaluate whether the force required to achieve either the midpoint or maximum angle of deviation differed among the 4 constructs. As previously mentioned, the midpoint angle of deviation was believed to be consistent with clinical failure of the implant, and the maximal angle of deviation represented complete loss of tarsal joint stability similar to that observed when the MCL was transected. Neither the midpoint nor maximal angle of deviation differed significantly among the 4 constructs. Interestingly, although the forces required to achieve those angles also did not differ among the 4 constructs, only nylon-suture constructs failed prior to the midpoint (n = 2) or maximal (9) angle of deviation being achieved. That finding might be reflective of an inability to objectively measure the tension generated when the nylon suture was tightened. It is possible that the constructs that failed prior to achieving the midpoint or maximal angle were specimens that were placed under unnecessarily high tension during prosthetic ligament placement.

The mode of failure also appeared to differ between constructs with UHMWPE suture and those with nylon suture. Eleven of 13 nylon-suture constructs failed because of breakage of the suture mimicking the long branch of the MCL, whereas only 2 of 13 UHMWPE-suture constructs failed in that manner. The ultimate strength for many of the UHMWPE-suture constructs exceeded the strength limits of the custom-fabricated testing apparatus, and most of those specimens failed because of knot slippage or suture slippage at an anchor. The inherent differences in strength for nylon and UHMWPE suture may offer different clinical advantages. Although nylon suture may be weaker than UHMWPE suture, that feature may result in more overt clinical signs (eg, palpable joint laxity) at the time of implant failure, which might facilitate detection and early intervention. Conversely, the ability of UHMWPE suture to withstand high forces and remain at least partially intact may allow patients to be treated conservatively, albeit for a longer duration, without the need for further surgical intervention. It is important to note that, for the AU construct, the PEEK suture anchor was used in a manner not intended by the manufacturer. Specifically, a total of 8 strands of UHMWPE suture were secured by the PEEK anchor, which is twice the number of strands the anchor was designed to secure. To increase the stability of the AU construct, the suture ends could be drawn through the initial small tunnel to emerge on the lateral side of the tibia and then cut flush with the lateral tibial cortex such that only 4 strands of suture would be secured by the anchor.

The number of limbs repaired by each construct (n = 6 or 7) was determined by an a priori sample size calculation and power analysis for which it was assumed a 10° increase in joint laxity from that for joints with an intact MCL was clinically relevant. In actuality, the median change in tarsal joint laxity relative to that for joints with an intact MCL was < 10° for all 4 constructs evaluated and did not differ significantly among the constructs. This suggested that any differences in the change in joint laxity among the repair techniques were very small and statistical significance was not achieved owing to the small sample size and type II error. Results of post hoc power analyses indicated that 11 limbs/construct would have been necessary to detect a statistically significant difference in joint laxity between limbs with an intact MCL and those with a prosthetic ligament as well as between construct and joint angle; 415 limbs/construct would have been necessary to detect a difference between construct and ligament status (ie, before and after cyclic range-of-motion testing); and 17 limbs/construct would have been necessary to detect a difference between ligament status and joint angle. Although evaluation of a greater number of limbs may have allowed us to detect significant differences among various variables, it was felt those differences would have been too small to be clinically relevant and therefore did not justify the use of additional cadaveric specimens.

All soft tissues except those directly associated with the tarsal joint were removed from the limbs prior to biomechanical testing. This was done to improve visibility of repairs and decrease potential variables. It remains unknown whether removal of those tissues affected joint stability during biomechanical testing or cyclic range-of-motion testing. Additionally, because cadaveric specimens were used in this study, physiologic changes associated with MCL transaction and postoperative healing could not be assessed.
Results of the present study indicated that all 4 prosthetic ligament constructs evaluated for repair of the tarsal MCL of dogs provided satisfactory initial stability for the joint and substantially improved stability relative to that for unrepaired MCL-deficient joints. All 4 constructs performed well following cyclic range-of-motion testing and were equivalent in terms of strength at physiologically relevant forces, with the TU construct having the greatest ultimate strength. Relative to the traditional repair technique (AN construct), the 3 more recently developed techniques (AU, TN, and TU constructs) have lower-profile implants, and the tunnel techniques (TN and TU) have anchor elements on the lateral aspect of the joint, which may facilitate primary closure of the surgical incision. Thus, it was concluded that the 3 recently developed MCL repair techniques may be superior to the traditional technique for treatment of tarsal MCL injury in dogs; however, further research with live dogs is necessary before any of those techniques can be recommended for patients in clinical practice.

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Footnotes

b. Securos Surgical, Fiskdale, Mass.
c. 858 Mini Bionix, MTS Systems Corp, Eden Prairie, Minn.
d. USB-6009, National Instruments, Austin, Tex.
e. SAS, version 9.4, SAS Institute Inc, Cary, NC.

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