

Mechanical properties of isolated loops of nylon leader material, polyethylene cord, and polyethylene tape and mechanical properties of those materials secured to cadaveric canine femurs via lateral femoral fabellae, toggles placed through bone tunnels, or bone anchors

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Objective—To determine mechanical properties of various prosthetic materials secured to cadaveric canine femurs via various methods and to compare results with those for isolated loops of prosthetic material.

Sample—80 femurs obtained from cadavers of skeletally mature large-breed dogs.

Procedures—10 femoral constructs in each of 8 groups (single circumfabellar loop of polyethylene cord, double loop of polyethylene tape secured via a bone anchor [BAPT], single or double circumfabellar loops of nylon leader material [CNL] or polyethylene tape [CPT], or single or double loops of polyethylene tape secured via a toggle placed through a bone tunnel [BTPT]) and 10 isolated loops of prosthetic material in each corresponding configuration were tested. Stress relaxation, creep, elongation, load at 3 mm of displacement, stiffness, and peak load at failure were determined.

Results—5 single CNL constructs failed before completion of testing. Double CNL and single circumfabellar polyethylene cord constructs had the lowest loads at 3 mm of displacement. Single and double CPT constructs had the highest stiffness. Double BTPT and CPT constructs had the highest peak loads at failure. Double BTPT, double CPT, and BAPT constructs were mechanically superior on the basis of lower creep and stress relaxation and higher stiffness and load at 3 mm of displacement versus other constructs. Stiffness of femoral constructs was 28% to 69% that of corresponding isolated prosthetic loops.

Conclusions and Clinical Relevance—Double BTPT, double CPT, and BAPT constructs were mechanically superior to other constructs. Mechanical properties and methods of anchorage and securing of free ends of prostheses contributed to mechanical properties of constructs. (*Am J Vet Res* 2012;73:1519–1529)

Cranial cruciate ligament insufficiency is a common cause of hind limb lameness in dogs that results in osteoarthritis of the stifle joint and often leads to meniscal injury.^{1,2} Cranial cruciate ligament insufficiency alters stifle joint and hind limb kinematics, resulting in cranial tibial translation, increased internal limb rotation, and adduction of the tibia, especially during the stance phase of the gait cycle.^{3,4} A modification of

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ABBREVIATIONS

BAPT	Bone anchor with polyethylene tape
BTPT	Bone tunnel with polyethylene tape secured via a toggle
CNL	Circumfabellar nylon leader material
CPC	Circumfabellar polyethylene cord
CPT	Circumfabellar polyethylene tape
CrCL	Cranial cruciate ligament

the procedure described by DeAngelis and Lau⁵ is one of the most widely used techniques for extracapsular stabilization of the stifle joint for CrCL insufficiency.^{6,7} This technique involves placing prosthetic material around the lateral femoral fabella (ie, sesamoid of the lateral head of the gastrocnemius muscle) and through a hole or holes drilled in the proximal aspect of the tibia.⁷ Loosening of the prosthesis in the early postopera-

tive period, prior to functional adaptation of the stifle joint, results in joint laxity and persistent instability; such instability contributes to progression of osteoarthritis and potentiates meniscal injury.⁸ Elongation of the stabilization system can develop via stretching or breakage of the prosthetic material, slippage or failure of the knot or crimp clamp securing the free ends of the prosthesis, or stretching or failure of the proximal or distal anchors.^{9,10}

Factors affecting the mechanical properties of prostheses used for extracapsular stabilization of stifle joints in dogs have been evaluated by other authors^{11–15} in an attempt to improve the clinical outcome of the procedure. Results of other studies^{11,12} indicate that loops of monofilament nylon material undergo considerable loss of tension as a knot is tied and during cyclic loading, which could lead to prosthetic construct elongation and inadequate restraint of cranial tibial drawer motion in CrCL-deficient dogs. Such elongation and loss of tension develop primarily during the first 3 cycles of loading and are dependent on configuration of the knot.¹² Crimp clamp systems were developed to avoid the problems attributable to use of knots. Loops of nylon leader material secured with such crimp clamps have higher stiffness and less elongation after cyclic loading than do knotted loops of such material.¹³ Other advantages of crimp clamp systems include maintenance of consistent loop tension during placement, elimination of knots, and limited permanent deformation after application of physiologic loads.¹³

Recently, other authors^{14,15} have advocated for use of nonabsorbable multifilament ultrahigh-molecular weight polyethylene suture materials for extracapsular stabilization of CrCL-deficient stifle joints because such materials purportedly have mechanical characteristics that are superior to other materials used for this purpose. These polyethylene cord and polyethylene tape prostheses are stronger and stiffer than nylon leader material, and elongation under cyclic loading of these materials secured with prototype crimp clamps is significantly lower than that of the materials secured with knots.¹⁵ However, to the authors' knowledge, a crimp clamp system designed for use with these materials is not commercially available.

Little research has been conducted to evaluate the biomechanical properties of the sites used to secure prostheses during performance of extracapsular stifle joint stabilization techniques for dogs. Other investigators¹¹ compared the mechanical properties of isolated loops of nylon fishing line and nylon leader material with those of loops of the same materials secured around lateral femoral fabellae. Those investigators found that constructs in which 36-kg test nylon leader material was placed around the lateral femoral fabella underwent greater elongation and loss of tension than did isolated loops of nylon leader material. This difference became greater after cyclic loading, indicating that stretch or deformation of the femorofabellar ligament contributes to progressive elongation of such constructs.¹¹ Securing of prostheses in bone by use of such methods as bone anchors and placement of polyethylene cord in bone tunnels has been investigated as a potential method for overcoming the biomechanical limi-

tations of techniques that rely on soft tissue anchorage of prostheses.^{14,16,17} Results of one of those studies¹⁴ indicate polyethylene tape secured to bone has superior elongation, stiffness, and peak load at failure, compared with the other materials and techniques tested. Because such techniques for securing of prostheses to bone for extracapsular stifle joint stabilization have been recently developed, few studies have been conducted to compare such techniques to other techniques in which prostheses are secured around lateral femoral fabellae, and further investigation is warranted.

One objective of the study reported here was to compare the mechanical characteristics of 8 constructs prepared with cadaveric canine femurs and various commercially available prostheses used for extracapsular stabilization of CrCL-deficient stifle joints in dogs. Another objective was to compare the mechanical properties of constructs in which prostheses were secured to femurs with those of isolated loops of the same prosthetic materials. We hypothesized that femoral constructs in which prostheses were secured to bone (via bone tunnels or bone anchors) would be stiffer, have less creep and stress relaxation, require higher loads to reach 3 mm of displacement, undergo less elongation, and have higher peak load at failure than femoral constructs in which prostheses were secured around the lateral femoral fabella. We also hypothesized that all femoral constructs would be less stiff, have higher creep and stress relaxation, require lower loads to reach 3 mm of displacement, undergo more elongation, and have a lower peak load at failure than corresponding isolated loops of prosthetic materials.

Materials and Methods

Samples—Femurs were harvested from 80 skeletally mature large-breed dogs euthanized for reasons unrelated to the study. Cadavers were weighed prior to collection of femurs. The limbs were free of gross orthopedic pathological lesions at the time of collection. All tissues were removed from each femur except the lateral fabella and its attachment to the femur. Femurs were wrapped with towels soaked in saline (0.9% NaCl) solution and stored at -20°C until they were thawed to room temperature (approx 22°C) for testing. Collection and use of the femurs were approved by the University of Florida College of Veterinary Medicine Institutional Animal Care and Use Committee.

Prosthetic construct groups—Ten femurs were allocated to each of 8 femoral construct groups by use of a randomization procedure. The right and left femurs obtained from the cadaver of a dog were removed from the storage freezer via a random selection. A coin toss was then used to determine whether the left or right femur was used in the study; the other femur was not used (ie, only 1 femur of each pair was used in the study). Forty-one left femurs and 39 right femurs were used. Femoral construct groups included single CNL, double CNL, CPC, single CPT, double CPT, single BTPT, double BTPT, and BAPT. Prosthetic materials included polyethylene cord,^a polyethylene tape,^b and nylon leader material.^c Prostheses were secured to femurs around the lateral femoral fabella (ie, circumfabellar

placement), via a toggle placed through a tunnel drilled in the distal aspect of the bone, or via a bone anchor. The order in which each type of construct was tested was determined prior to the start of the study.

The single CNL femoral constructs (Figure 1) were prepared with 1 strand of 36-kg test nylon leader material swaged onto a needle.^c The needle and nylon leader material were passed through the femorofabellar ligament and around the lateral femoral fabella in a proximal-to-distal direction. The needle was removed, and the free ends of the nylon leader material were secured with a crimp clamp^d that was crimped 3 times with a crimping tool^e designed for use with this system. The clamp was applied 8 cm from the fabella to create a 16-cm-circumference loop. The free ends of the material were trimmed 10 to 15 mm from the crimp clamp.

The double CNL femoral constructs (Figure 1) were prepared in the same manner as the single CNL constructs with 2 strands of 36-kg test nylon leader material swaged onto a single needle.^f Each strand of nylon leader material was secured separately with a crimp clamp via the same method that was used for preparation of single CNL constructs to create two 16-cm-circumference loops.

The CPC femoral constructs (Figure 1) were prepared with 1 strand of No. 5 polyethylene cord swaged onto a needle^g that was passed through the femorofabellar ligament and around the lateral femoral fabella in a proximal-to-distal direction. The needle was re-

moved, and the free ends of the material were secured with a 5-throw square knot with alternating posts that was placed 8 cm from the fabella to create a 16-cm-circumference loop.

The single CPT femoral constructs (Figure 1) were prepared with 1 strand of 2-mm-diameter polyethylene tape^b that was passed through the femorofabellar ligament and around the lateral femoral fabella in a proximal-to-distal direction with a medium cruciate needle. The free ends of the material were secured with a 5-throw square knot with alternating posts that was placed 8 cm from the fabella to create a 16-cm-circumference loop.

The double CPT femoral constructs (Figure 1) were prepared in the same manner as the single CPT constructs with 2 strands of 2-mm-diameter polyethylene tape^b placed with a medium cruciate needle. Each strand of polyethylene tape was secured separately with a 5-throw square knot with alternating posts as for the single CPT constructs to create two 16-cm-circumference loops.

The single BTPT femoral constructs (Figure 1) were prepared as described.¹⁴ Briefly, a Kirschner wire was placed in each femur 2 mm craniodistal to the lateral femorofabellar articulation, adjacent to the caudal aspect of the lateral femoral condyle. The Kirschner wire functioned as a guide wire and was advanced with a drill at a proximally directed angle so that the wire passed in a lateral to medial direction through

the distal aspect of the femur, exiting at the mediolateral aspect of the diaphysis of the femur proximal to the medial trochlear ridge. A 3.5-mm-cannulated drill bit^h attached to a drill was placed over the Kirschner wire and used to drill a tunnel in the distal aspect of the femur. The Kirschner wire and drill bit were removed. A single strand of polyethylene tape^b was threaded through a custom toggle.ⁱ The polyethylene tape and toggle were passed in a lateral-to-medial direction through the tunnel in the femur with a needle^l attached to the toggle via a pull-through suture. Tension was applied to the needle on the medial aspect of the femur and to the polyethylene tape^b on the lateral aspect of the femur to pass the toggle through the tunnel. By use of this method, the toggle was positioned on the medial aspect of the femur. The pull-through suture and needle were removed. The free ends of the polyethylene tape were secured with a hand-tied 5-throw square knot with alternating posts placed 8 cm from the lateral aspect of the femur to create a 16-cm-circumference loop.

The double BTPT femoral constructs (Figure 1) were prepared via the same method used to prepare single BTPT constructs with 2 strands of polyethylene tape^b passed through a single toggle. The toggle and polyethylene tape

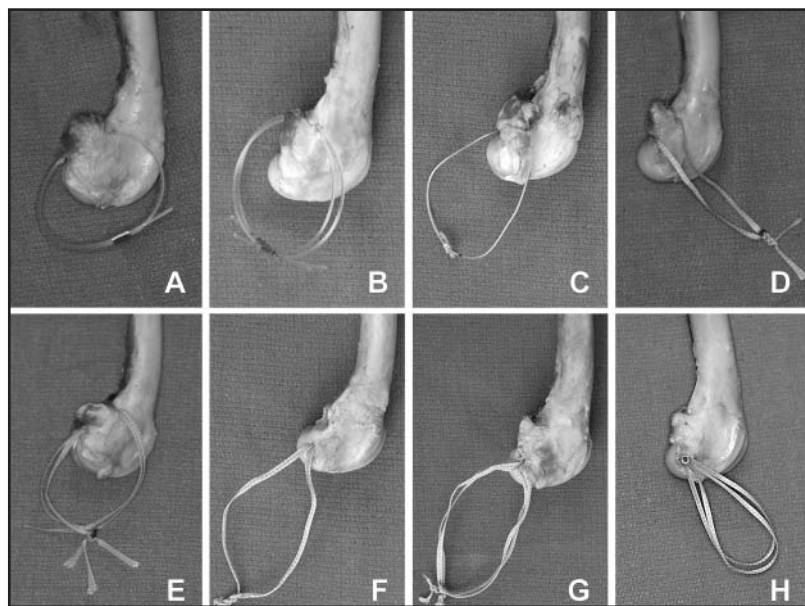


Figure 1—Photographs of representative femoral constructs with various types and configurations of prosthetic material that were prepared for mechanical testing. A—One strand of 36-kg test nylon leader material placed around the lateral femoral fabella and secured with a crimp clamp (single CNL construct). B—Two strands of 36-kg test nylon leader material placed around the lateral femoral fabella and secured with crimp clamps (double CNL construct). C—One strand of polyethylene cord placed around the lateral femoral fabella and secured with a knot (CPC construct). D—One strand of polyethylene tape placed around the lateral femoral fabella and secured with a knot (single CPT construct). E—Two strands of polyethylene tape secured around the lateral femoral fabella and secured with knots (double CPT construct). F—One strand of polyethylene tape secured to the femur via a toggle placed through a tunnel drilled in the bone (single BTPT construct). G—Two strands of polyethylene tape secured to the femur via a toggle placed through a tunnel drilled in the bone (double BTPT construct). H—Two strands of polyethylene tape secured to the femur via a 5.5-mm-diameter bone anchor (BAPT construct).

were placed in the same manner as for the single BTPT constructs. The free ends of each strand of polyethylene tape were tied separately with a hand-tied 5-throw square knot with alternating posts placed 8 cm from the lateral aspect of the femur to create two 16-cm-circumference loops.

Knotless BAPT femoral constructs (Figure 1) were prepared in accordance with the manufacturer's directions. A spade-tipped drill bit^j designed for use with this system was used to create a pilot hole in the lateral aspect of the femur 2 mm craniodistal to the lateral fabella–femur articulation adjacent to the caudal aspect of the lateral femoral condyle. The pilot hole was tapped with a punch and tap^k designed for use with this system. Two strands of polyethylene tape,^b premeasured to each form a 16-cm-circumference loop, were threaded through the eyelet of a 5.5-mm-diameter bone anchor.^l The bone anchor was implanted into the prepared hole with an anchor driver designed for use with this system.

Ten isolated loops of each configuration of prosthetic material corresponding to configurations used in the femoral constructs (2 strands of 36-kg monofilament nylon leader material [corresponding to double CNL constructs], 1 strand of No. 5 polyethylene cord [corresponding to CPC constructs], and 1 [corresponding to single CPT and BTPT constructs] or 2 [corre-

sponding to BAPT and double CPT and BTPT constructs] strands of 2-mm-diameter polyethylene tape) were prepared; isolated loops with 1 strand of monofilament nylon leader material were not prepared. Two strands of 36-kg test nylon leader material^c were individually secured with a crimp clamp^d that was crimped 3 times with a crimping tool^e to create a 16-cm-circumference loop; free ends of the prosthetic material were trimmed 10 to 15 mm from the crimp. Polyethylene cord^d loops were created by securing 1 strand of No. 5 polyethylene cord with a hand-tied 5-throw square knot to create a 16-cm-diameter loop; free ends of the prosthetic material were trimmed 10 to 15 mm from the knot. Single polyethylene tape^b loops were created by securing a strand of polyethylene tape^b with a hand-tied 5-throw square knot to create a 16-cm-diameter loop; free ends of the prosthetic material were trimmed 10 to 15 mm from the knot. Double polyethylene tape^b loops were prepared via the same method used to prepare the single polyethylene tape loops, except loops consisted of 2 strands of polyethylene tape^b and knots were created separately for each strand.

Mechanical testing—Mechanical loading of the constructs was performed by use of an axial servohydraulic dynamic mechanical testing machine^m with a 20-kN load cell attached to the crosshead. A custom aluminum jig was used to secure femurs so that orientation of femurs and prostheses were similar to those in a standing dog with an extracapsular stabilization prosthesis (Figure 2). The mean angle (70°) at which the prostheses for each technique were oriented relative to the long axis of femurs was determined by use of measurements obtained during preliminary studies. In the preliminary studies, the extracapsular stabilization techniques were performed as recommended on cadaveric stifle joints and the angle of the portion of the prosthesis extending from the femoral anchorage site to the tibial anchorage site was measured relative to the long axis of the femur with the stifle joint at an angle of 135° (approximate stifle joint angle in a standing dog). The average angle of these measurements for all techniques was 70°. The load bar was positioned relative to femurs to achieve the appropriate construct orientation during testing. Prosthetic loops were secured around the load bar, which was oriented parallel to the base of the

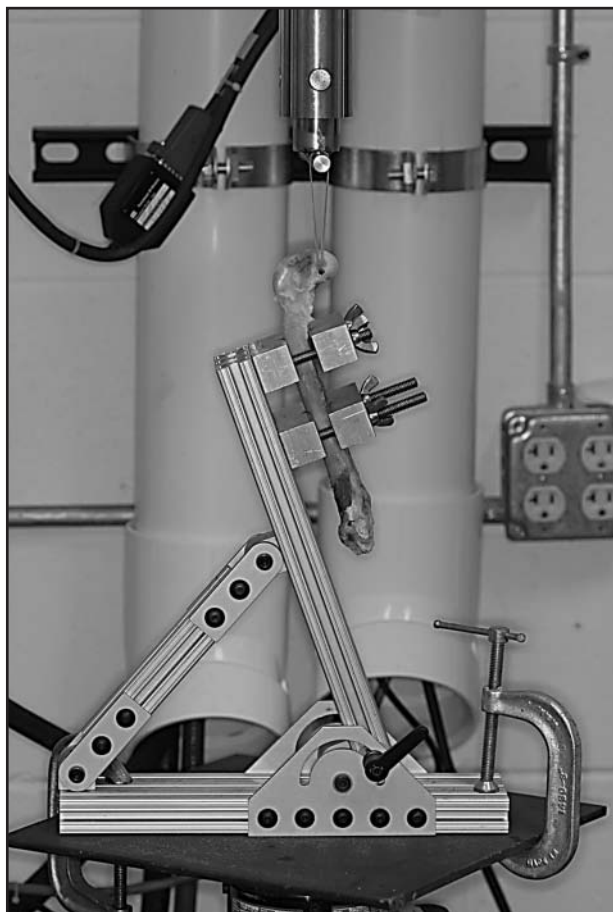


Figure 2—Photograph of a representative femoral construct secured to a custom aluminum jig and testing apparatus indicating orientation of each component during mechanical testing.

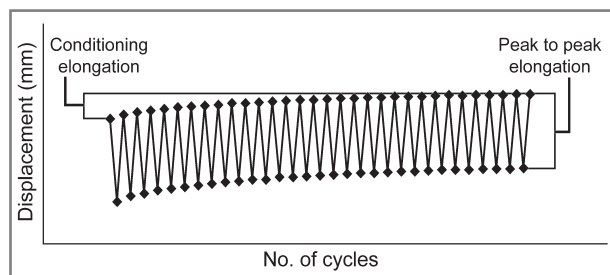


Figure 3—Illustration of a displacement versus cycle curve indicating the method of determination of conditioning elongation and peak-to-peak elongation for femoral constructs and isolated loops of prosthetic materials of various types. Conditioning elongation was defined as the difference in displacement between the first peak and the 100th peak during cyclic loading. Peak-to-peak elongation was defined as the difference in displacement for the construct as load increased from 25 to 250 N during the 100th cycle. Notice that fewer than 100 cycles are indicated.

machine. Loading of femoral constructs was controlled with a commercially available software program.¹¹ Prostheses were preloaded to a tension of 5 to 10 N, and the relative location of the mechanical testing machine crosshead at that load was established as 0 mm of axial displacement for each construct. Each construct was tested for stress relaxation by increasing tension to 200 N at a rate of 50 N/s and maintaining the resultant displacement for 2 minutes. The crosshead was then returned to 0 mm of displacement, which then corresponded with a 0-N load on the construct. Creep was measured by loading the sample to 200 N at a rate of 50 N/s and maintaining a load of 200 N for 2 minutes. The crosshead was returned to 0 mm of displacement, and the construct was cyclically loaded under sinusoidal load control from 25 to 250 N at 1 Hz for 100 cycles. After cyclic loading, the crosshead was returned to 0 mm of displacement, and the stress relaxation and

creep tests were repeated. The crosshead was returned to a displacement of 0 mm, and the construct was loaded to failure at 1 mm/s. Data were recorded at 100 Hz.

The isolated loops of suture material were secured to the mechanical testing machine with a hook and clevis apparatus. The knots or crimps were positioned approximately halfway between the hook and the clevis. Each isolated loop was subjected to the same mechanical testing procedure as were the femoral constructs.

The mechanical testing outcome measures evaluated were stress relaxation, creep, conditioning elongation, peak-to-peak elongation, load at 3 mm of displacement, stiffness, and peak load at failure. Stress relaxation and creep were measured before and after cyclic loading. Stress relaxation was defined as the decrease in stress that a viscoelastic material undergoes when loaded and maintained at a constant length.¹⁸

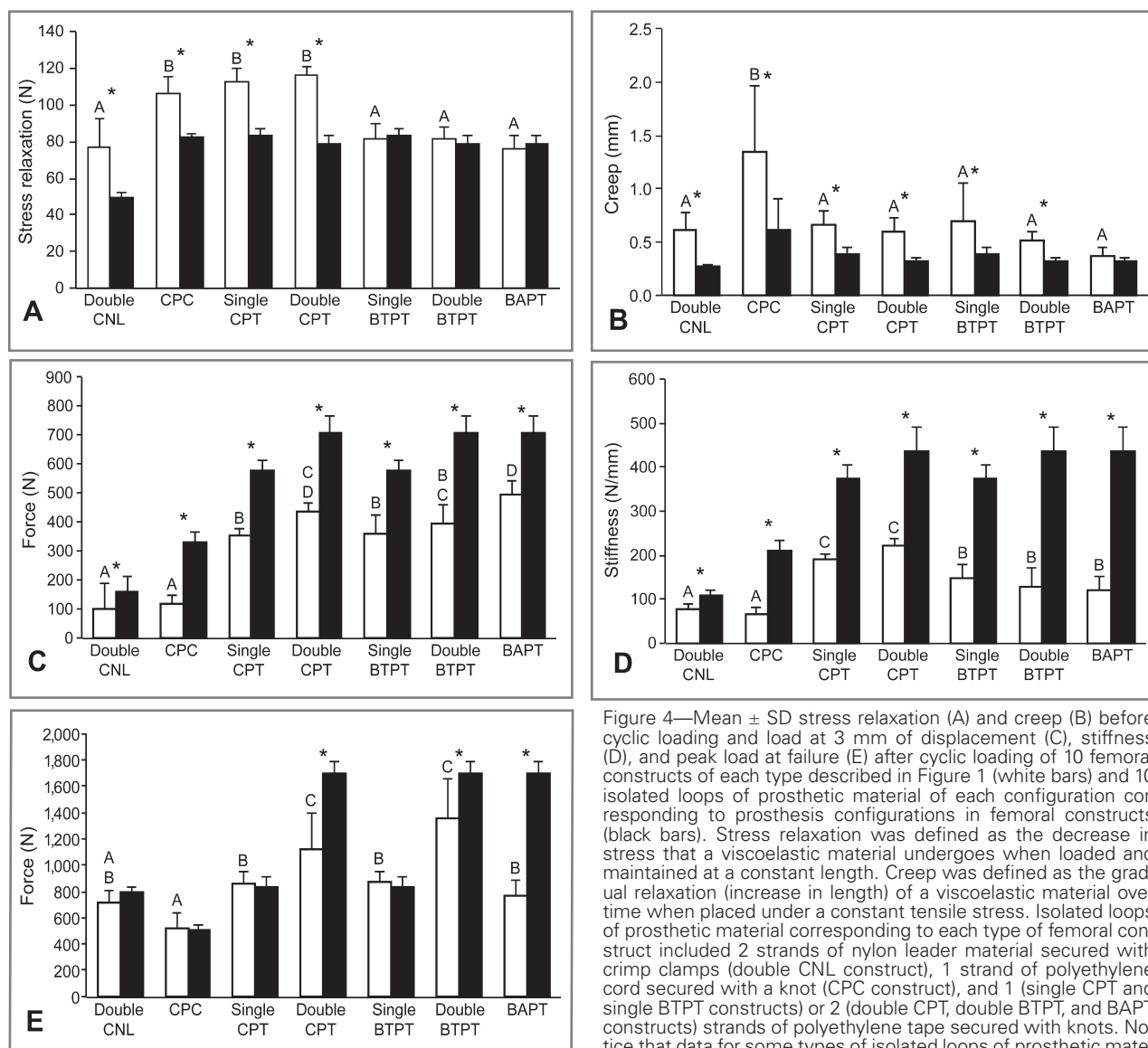


Figure 4—Mean \pm SD stress relaxation (A) and creep (B) before cyclic loading and load at 3 mm of displacement (C), stiffness (D), and peak load at failure (E) after cyclic loading of 10 femoral constructs of each type described in Figure 1 (white bars) and 10 isolated loops of prosthetic material of each configuration corresponding to prosthesis configurations in femoral constructs (black bars). Stress relaxation was defined as the decrease in stress that a viscoelastic material undergoes when loaded and maintained at a constant length. Creep was defined as the gradual relaxation (increase in length) of a viscoelastic material over time when placed under a constant tensile stress. Isolated loops of prosthetic material corresponding to each type of femoral construct included 2 strands of nylon leader material secured with crimp clamps (double CNL construct), 1 strand of polyethylene cord secured with a knot (CPC construct), and 1 (single CPT and single BTPT constructs) or 2 (double CPT, double BTPT, and BAPT constructs) strands of polyethylene tape secured with knots. Notice that data for some types of isolated loops of prosthetic material are repeated because these types correspond to prosthesis types and configurations in more than 1 type of femoral construct. Results for single CNL femoral constructs are not reported because 5 of those constructs failed before mechanical testing was completed. *Values are significantly ($P < 0.05$) different between the femoral construct and the corresponding isolated prosthetic loops. A-D Values for femoral constructs with different letters are significantly ($P < 0.05$) different.

material are repeated because these types correspond to prosthesis types and configurations in more than 1 type of femoral construct. Results for single CNL femoral constructs are not reported because 5 of those constructs failed before mechanical testing was completed. *Values are significantly ($P < 0.05$) different between the femoral construct and the corresponding isolated prosthetic loops. A-D Values for femoral constructs with different letters are significantly ($P < 0.05$) different.

Creep was defined as the gradual relaxation (increase in length) of a viscoelastic material over time when placed under a constant tensile stress.¹⁸ Conditioning elongation was defined as the difference in displacement between the first peak and the 100th peak during cyclic loading (Figure 3). Peak-to-peak elongation was defined as the difference in displacement for the construct as load increased from 25 to 250 N during the 100th cycle. Load at 3 mm of displacement was defined as the load that produced 3 mm of axial displacement in the construct once the construct was under tension during load to failure testing. To eliminate previously accrued slack in the constructs from calculations, constructs were considered to be under tension once axial load exceeded 5 N. Stiffness was defined as the resistance of an elastic body to deformation by an applied load and was calculated as the slope of the linear portion of the load-displacement curve generated during load to failure.¹⁹ Peak load at failure was defined as the load at which the integrity of any component of the construct was lost. The mode of failure was recorded for each construct.

Statistical analysis—Statistical analysis was performed with statistical software.⁹ Multivariate ANOVA was used to evaluate differences in mean body weight, stress relaxation and creep before cyclic loading, conditioning elongation, peak-to-peak elongation, load at 3 mm of displacement, stiffness, and peak load at failure. When significant differences between groups were detected, a Tukey pairwise multiple comparisons procedure was performed. Data for the femoral constructs and isolated prosthetic loops were analyzed separately via the same statistical protocol. Comparison between data for femoral constructs and corresponding isolated prosthetic loops was performed with a paired *t* test. Comparison between creep and stress relaxation values before and after cyclic loading was performed with a paired *t* test. For all statistical analyses, values of *P* < 0.05 were considered significant.

Results

Mean ± SD weight of the cadavers of dogs from which femurs were obtained was 30.1 ± 8.1 kg. No significant differences were detected among femoral construct groups regarding weight of cadavers of dogs from

which femurs had been obtained. Implantation of bone anchors created a grossly visible fracture in the caudal aspect of the lateral femoral condyle in 2 femurs. These femurs were discarded, and 2 additional femurs that did not fracture during implantation of bone anchors were used in the study.

The testing protocol was completed for only 5 of the single CNL constructs. The other 5 constructs failed via slippage of the nylon leader material through the crimp clamp during cyclic loading (*n* = 3), during the stress relaxation test after cyclic loading (1), or during the loading creep test before cyclic loading (1). Mean ± SD stiffness and peak load at failure for the samples for which the testing protocol was completed were 60.5 ± 3.8 N/mm and 276.8 ± 21.2 N, respectively. Data for single CNL constructs were excluded from statistical comparisons because of the bias (attributable to differing numbers of constructs among groups) that would potentially have been introduced during analysis.

Stress relaxation before cyclic loading was significantly higher for CPC, single CPT, and double CPT constructs than it was for any other type of femoral construct, and values for construct groups other than CPC, single CPT, and double CPT constructs were not significantly different (Figure 4). All 4 construct types with circumfabellar prosthesis placement had significantly higher stress relaxation than the corresponding isolated prosthetic loops, but stress relaxation for constructs in which prostheses were secured to bone was not significantly different from that for corresponding isolated prosthetic loops. Creep before cyclic loading was significantly higher for CPC constructs than it was for any other type of femoral construct, and values for construct groups other than CPC constructs were not significantly different. All femoral construct groups except the BAPT construct group had higher creep than the corresponding isolated prosthetic loops. For all femoral construct groups, stress relaxation and creep were significantly (*P* < 0.002 for all comparisons) lower after limited cyclic loading than they were before cyclic loading (Table 1).

The BAPT and double CPT constructs required the highest loads to reach 3 mm of displacement, and the CPC and double CNL constructs required the lowest loads to reach 3 mm of displacement (Figure 4). All femoral constructs reached 3 mm of displacement at

Table 1—Mean ± SD creep and stress relaxation before and after cyclic loading of 10 samples of each type of femoral construct with various types and configurations of prosthetic material.

Femoral construct	Creep before cyclic loading (mm) *	Creep after cyclic loading (mm)	Stress relaxation before cyclic loading (N)*	Stress relaxation after cyclic loading (N)
Double CNL	0.61 ± 0.16	0.27 ± 0.10	77.3 ± 15.0	24.1 ± 4.4
CPC	1.35 ± 0.61	0.32 ± 0.05	106.1 ± 9.4	29.9 ± 5.5
Single CPT	0.67 ± 0.12	0.21 ± 0.02	112.8 ± 7.7	37.3 ± 2.7
Double CPT	0.60 ± 0.12	0.18 ± 0.02	115.6 ± 5.0	37.7 ± 3.2
Single BTPT	0.70 ± 0.36	0.22 ± 0.05	81.8 ± 7.6	33.1 ± 1.9
Double BTPT	0.51 ± 0.08	0.19 ± 0.03	81.5 ± 6.6	31.7 ± 3.3
BAPT	0.37 ± 0.08	0.14 ± 0.01	75.9 ± 7.8	30.1 ± 2.1

Creep was defined as the gradual relaxation (increase in length) of a viscoelastic material over time when placed under a constant tensile stress. Stress relaxation was defined as the decrease in stress that a viscoelastic material undergoes when loaded and maintained at a constant length. Results for single CNL femoral constructs are not reported because 5 of those constructs failed before mechanical testing was completed.

*For all femoral construct groups, values for creep and stress relaxation before cyclic loading are significantly (*P* < 0.002 for all comparisons) greater than values for those variables after cyclic loading.

Table 2—Mean \pm SD conditioning and peak-to-peak elongation values for 10 samples of each type of femoral construct with various types and configurations of prosthetic material and 10 samples of each configuration of isolated loops of prosthetic material corresponding to prosthesis configurations in femoral constructs.

Femoral constructs and corresponding isolated loops of prosthetic material	Conditioning elongation (mm)	Peak-to-peak elongation (mm)
Double CNL 2 strands of nylon leader material	0.28 \pm 0.13 ^{*a} 0.15 \pm 0.04	1.63 \pm 0.22 ^{*d} 1.28 \pm 0.05
CPC 1 strand of polyethylene cord	1.68 \pm 0.62 ^c 1.22 \pm 0.47	1.77 \pm 0.22 ^{*d} 0.54 \pm 0.11
Single CPT 1 strand of polyethylene tape	0.64 \pm 0.14 ^{a,b} 0.68 \pm 0.12	1.00 \pm 0.07 ^{*c} 0.39 \pm 0.06
Double CPT 2 strands of polyethylene tape	0.51 \pm 0.14 ^{a,b} 0.47 \pm 0.10	0.87 \pm 0.05 ^{*b,c} 0.33 \pm 0.04
Single BTPT 1 strand of polyethylene tape	0.82 \pm 0.28 ^b 0.68 \pm 0.12	0.77 \pm 0.12 ^{*b,c} 0.39 \pm 0.06
Double BTPT 2 strands of polyethylene tape	0.84 \pm 0.28 ^{*b} 0.47 \pm 0.10	0.69 \pm 0.09 ^{*a,b} 0.33 \pm 0.04
BAPT 2 strands of polyethylene tape	0.36 \pm 0.12 ^{*a} 0.47 \pm 0.10	0.54 \pm 0.06 ^{*a} 0.33 \pm 0.04

Data for some types of isolated loops of prosthetic material are repeated because these types correspond to prosthesis types and configurations in more than 1 type of femoral construct.
^{*}Within a column, values for femoral constructs and corresponding isolated loops of prosthetic material are significantly ($P < 0.05$) different.
^{a-d}Within a column, values for femoral constructs with different superscript letters are significantly ($P < 0.05$) different.
 See Table 1 for remainder of key.

significantly lower loads than the corresponding isolated prosthetic loops. The single and double CPT constructs had the highest stiffness, and the CPC and double CNL constructs had the lowest stiffness. All femoral construct groups had significantly lower stiffness than the corresponding isolated prosthetic loops.

The CPC constructs had the highest conditioning elongation, whereas the double CNL and BAPT constructs had the lowest conditioning elongation (Table 2). The double CNL and double BTPT constructs had significantly higher conditioning elongation than the corresponding isolated prosthetic loops, and the BAPT constructs had significantly lower conditioning elongation than the corresponding isolated prosthetic loops. The CPC and double CNL constructs had the highest peak-to-peak elongation, and the BAPT and double BTPT constructs had the lowest peak-to-peak elongation. All femoral constructs had higher peak-to-peak elongation than the corresponding isolated prosthetic loops.

The double BTPT and double CPT construct groups had the highest peak load at failure, and the CPC and double CNL construct groups had the lowest peak load at failure (Figure 4). The femoral construct groups in which 2 strands of polyethylene tape were used (double CPT, double BTPT, and BAPT) were the only construct groups that had a significantly lower peak load at failure than the corresponding isolated prosthetic loops. The mode of failure differed among the femoral construct groups. All of the constructs in the double CNL

group failed via breakage or slippage of prostheses at the crimp. Nine CPC constructs and 8 single CPT constructs failed via prosthesis breakage at the knot. Prostheses pulled through the femorofabellar ligament in 1 of the CPC constructs, 2 of the single CPT constructs, and all 10 of the double CPT constructs. The single BTPT constructs all failed via prosthesis breakage at the knot. Only 1 double BTPT construct failed via breakage of the prosthesis, and the other 9 constructs in that group failed via breakage or deformation of the toggle, at which time the toggle was pulled through the bone tunnel. The BAPT constructs failed via pullout of the anchor ($n = 5$), polyethylene tape cutting through the femoral condyle (4), or condylar fracture (similar to the fractures that developed during bone anchor implantation in the 2 femurs that were discarded; 1). The isolated prosthetic loops failed in manners similar to those detected in the corresponding femoral constructs, except all isolated double polyethylene tape loops failed via breakage of prosthetic material at the knot of each strand.

Discussion

Results of the present study did not support our hypothesis that all femoral constructs in which prostheses were secured to bone would have better mechanical characteristics than construct groups in which prostheses were secured around the lateral femoral fabella. Results indicated that multiple factors, including mechanical properties of prostheses, method of securing free ends of prostheses, and method of securing prostheses to femurs (bone vs circumfabellar), seemed to contribute to the relative strengths and weaknesses of each type of construct. However, the BAPT, double BTPT, and double CPT constructs were considered mechanically superior to the other femoral constructs tested on the basis of their lower creep and stress relaxation, greater stiffness, and higher loads required to reach 3 mm of displacement.

Stress relaxation and creep are mechanical properties of viscoelastic materials, including tissues such as ligaments and bone.^{18,19} Such properties are rarely evaluated in mechanical studies of prosthetic materials (which are typically regarded as elastic materials), but the inclusion of viscoelastic sites for securing prostheses in the femoral constructs tested in the present study warranted evaluation of these variables. Consideration should be given to the method of attachment of prostheses to femurs when comparing results of the present study with those of other studies.

Stress relaxation was the only variable that seemed to be primarily affected by the method used to secure prostheses to femurs. All 4 femoral construct groups in which prostheses were secured around the lateral femoral fabella had significantly higher stress relaxation than the corresponding isolated prosthetic loops, but there was no difference in stress relaxation between femoral constructs in which prostheses were secured to bone and the corresponding isolated prosthetic loops. The double CNL constructs were the only circumfabellar femoral constructs that did not have significantly higher stress relaxation than constructs in which prostheses were secured to bone. This finding was likely at-

tributable to the use of crimp clamps in the double CNL constructs; such crimp clamps aid retention of tension in nylon leader material loops.¹³ The variation in methods used to secure free ends of prostheses in the present study may have confounded interpretation of some results. However, the primary goal of this study was to compare mechanical properties among femoral constructs in which prostheses were applied in manners similar to those used for clinical application of commercially available products; no commercially available crimp clamp system for the polyethylene prosthetic materials used in the present study currently exists.

Unlike for stress relaxation, the method used to secure prostheses did not seem to be the primary factor contributing to creep. The CPC constructs had significantly higher creep than any other type of construct, and isolated loops of polyethylene cord had significantly higher creep than any other type of isolated prosthetic loop. This finding indicated that the mechanical properties of prostheses had a greater effect on creep than did other properties of the constructs.

During cyclic loading, polyethylene cord material has 2 phases of creep²⁰; this finding is similar to findings for polyethylene cord and polyethylene tape in the present study. Initially during cyclic loading, there is a change in creep attributable to conformance of prosthetic material to bone and tightening of the knot (settling phase). This is followed by true creep, which includes damage to bone at the prosthesis interface, slippage of the fastening method, and stretching of the prosthesis.²⁰ The findings of that other study²⁰ were supported by results of the present study that indicated creep and stress relaxation for each femoral construct group were significantly different before and after cyclic loading. Although it would be difficult to extrapolate findings of this study to the use of prostheses in a clinical setting, intraoperative preconditioning and retensioning (ie, tensioning of an extracapsular prosthesis, repeatedly moving the limb through a full range of motion, then retensioning the prosthesis before securing it) of prostheses may decrease postoperative creep and stress-relaxation. Therefore, such intraoperative preconditioning and retensioning of prostheses may be indicated during performance of any extracapsular stabilization technique.

The BAPT constructs were the only constructs for which creep was not significantly different from that of the corresponding isolated prosthetic loops. Because a knot was not used to secure the BAPT constructs and the corresponding isolated double loops of polyethylene tape were secured with knots, the initial creep for the BAPT constructs during the settling phase was attributed to movement of prostheses and conformance of material to the bone. We also found that both single and double BTPT constructs had higher creep than corresponding isolated single and double polyethylene tape loops. This may have been attributable to multiple factors, including viscoelastic properties of the anchorage points in bone, deformation of the metal toggle, and the greater length of polyethylene tape used in the femoral constructs; the length of polyethylene tape in the tunnel in the femur was not included in measurements for the 16-cm-circumference loops of polyethylene tape

in BTPT constructs. The 16-cm-circumference prosthesis loop size was intended to maintain consistency in size of prosthetic loops between femoral constructs and isolated prosthetic loops and to facilitate attachment of prostheses to the mechanical testing machine.

The femoral constructs had only 28% to 69% of the stiffness of the corresponding isolated prosthetic loops. To the authors' knowledge, the optimal construct stiffness for a clinically useful extracapsular stabilization technique is unknown. A stiff construct is likely to have low elongation and effectively mitigate cranial tibial drawer motion, and a stiff construct secured at nonisometric locations may cause alterations in kinematics or contact mechanics of a stifle joint.²¹

Both single and double CPT constructs were stiffer than the other circumfabellar constructs and the constructs in which prostheses were secured to bone in the present study. This finding was attributed to the viscoelastic properties of ligaments.²² Ligaments have highly nonlinear stress-strain properties. At rest, collagen fibrils in ligaments are crimped so that an initial load primarily straightens these fibrils. At higher strains, the straightened collagen fibrils are lengthened, which requires greater stress. As a result, ligaments and tendons are more compliant at low loads and less compliant at high loads.²³ We suspected that with a stiff and strong prosthesis (polyethylene tape), the femorofabellar ligament functioned in the stiffer, more linear region of the viscoelastic stress-strain curve.

Authors of another study¹⁵ found that polyethylene tape loops secured with a crimp clamp are significantly stiffer than nylon leader material loops secured via that same method. Those authors also found that polyethylene cord loops secured with knots are stiffer than nylon leader material loops secured via that same method. One of the findings of that study¹⁵ was that polyethylene suture is stronger, stiffer, and has less elongation than nylon leader material. Findings of another study¹⁴ were similar regarding comparison of mechanical properties of polyethylene tape or cord secured via a toggle in a bone tunnel versus mechanical properties of CNL secured via a crimp clamp. Results of the present study were similar to results of those other studies^{14,15}; femoral constructs with polyethylene tape were stiffer than those with nylon leader material. The only finding of the present study that differed from findings of one of the other studies¹⁴ was that knotted polyethylene cord was less stiff than nylon, although results were not significantly different. This difference in results between studies was attributed to the fact that statistical analysis in the present study was used to compare data for constructs with 2 strands of nylon leader material with data for constructs with a single loop of polyethylene cord, but analysis in that other study¹⁴ was used to compare data for constructs with a single loop of nylon leader material with data for constructs with a single loop of polyethylene tape or polyethylene cord.

Minimizing postoperative prosthesis elongation should be a primary goal for any extracapsular joint stabilization technique. Although a decrease in prosthesis tension may increase joint laxity, some conditioning elongation and peak-to-peak elongation may be acceptable and may compensate for lack of isometry of points

to which extracapsular stifle joint stabilization prostheses are secured. Authors of another study²⁴ reported that prostheses secured around the lateral femoral fabella and those secured in bone do not have consistent tension during movement of stifle joints through a full range of motion. The lack of isometric points for securing of prostheses (particularly stiff prostheses²¹) may increase the risks of overconstraining joint motion and increasing pressure in tissues of the lateral aspect of a stifle joint. On the basis of the findings of the present study, prosthesis tension that is initially too high may be mitigated by stress relaxation, creep, conditioning elongation, and peak-to-peak elongation that develop during cyclic loading.

We selected 25 to 250 N forces for cyclic loading because those forces are similar to forces in CrCLs in dogs during activities ranging from resting to a slow run.^{11,25} The mean amount of conditioning elongation induced by 100 cycles of loading in each femoral construct group ranged from 0.28 to 1.68 mm, which is a clinically acceptable amount of elongation in a CrCL repair system.²⁵ Most conditioning elongation developed during the first few cycles of loading in constructs in the present study, which was consistent with results of another study.¹² This finding supported our recommendation for intraoperative preconditioning and re-tensioning of prostheses to decrease postoperative loss of tension.

Peak-to-peak elongation was measured in the present study as an estimate of the elastic deformation that would develop during cyclic loading in an early postoperative convalescent period. Isolated loops of nylon leader material and polyethylene cord had greater peak-to-peak elongation than isolated double or single loops of polyethylene tape in this study, which is likely the reason that CPC and double CNL constructs had significantly greater peak-to-peak elongation than single CPT and double CPT constructs in which prostheses were secured in the same manner. Although values were not significantly different, greater peak-to-peak elongation developed in constructs in which prostheses were secured around the lateral femoral fabella versus constructs in which the same prosthetic materials were secured to bone. In addition, peak-to-peak elongation was significantly lower for prostheses secured with a bone anchor than it was for prostheses secured around the lateral femoral fabella. These results supported our hypothesis that securing of prostheses to bone would cause less peak-to-peak elongation than securing of prostheses around the lateral femoral fabella and indicated that the femorofabellar ligament undergoes elastic deformation when subjected to physiologic loads.

The amount of elongation at failure exceeded 1 cm for the femoral constructs and isolated prosthetic loops in this study. Although this finding was interesting, the peak load and peak elongation that develop during a single load to failure have little clinical relevance. Because of the effects of secondary restraints of CrCL-deficient stifle joint motion (eg, collateral ligaments) and physiologic forces in stifle joints,²⁶ it is unlikely that extracapsular stabilization constructs in dogs would develop the amount of elongation detected at the time of prosthesis failure in cadaveric femurs in the present study.

Comparison between monotonic load to failure test results for femoral constructs in which prostheses are secured around the lateral femoral fabella and results for isolated prosthetic loops has not previously been reported, to the authors' knowledge. On the basis of results of the present study, the strength of femorofabellar ligaments seemed to exceed that of nylon leader material, polyethylene cord, and a single strand of polyethylene tape. This finding was supported by the finding of a high incidence of prosthesis breakage or knot or crimp slippage in the circumfabellar femoral constructs and the finding that only 3 femoral constructs failed via polyethylene cord or polyethylene tape pulling through perifabellar soft tissues. These findings and the finding that CPT constructs had high stiffness suggested that early construct elongation and failure attributed by other authors^{8,9} to properties of femorofabellar ligaments were more likely attributable to mechanical properties of prostheses or failure to properly secure prostheses around fabellae.

Full-thickness fractures of the caudal aspects of femoral condyles occurred during implantation of bone anchors in 2 femurs. In the femurs that fractured, bone anchors appeared to have been placed 1 to 2 mm more caudal than bone anchors in femurs that did not fracture. No other appreciable differences in size or pilot hole orientation were detected among these femurs. Those fractures were attributed to technical errors by investigators before proper implantation techniques were learned because the fractures occurred in 2 of the first 3 BAPT constructs prepared. The bone anchor in the femur for which the testing protocol was completed and for which failure occurred because of fracture of the lateral condyle appeared to have been placed correctly; no fracture had been grossly evident in that femur after anchor implantation and before testing.

Failure of femoral constructs and isolated prosthetic loops in this study was defined as the point at which the integrity of any component of the system was lost. Peak load at failure was easily measured during mechanical testing, but the clinical relevance of results for this variable should be interpreted with caution. Clinical failure has been reported to develop in dogs once prosthesis elongation allows > 2 mm of cranial tibial drawer motion¹⁴; therefore, resistance to elongation may be a more useful variable for comparing clinical extracapsular stabilization techniques than is peak load at failure because prosthesis failure may occur at supraphysiologic loads.^{11,13,27} For this reason, we also chose to evaluate the load required to cause 3 mm of elongation in constructs after constructs were under tension during load to failure testing. The viscoelastic properties of femorofabellar ligaments contributed to the results of this study. The low-stiffness prostheses used in the double CNL and CPC constructs did not prevent elongation at low loads; therefore, the femorofabellar ligament was not loaded in the linear portion of the stress-strain curve, which would have provided increased resistance to deformation. Thus, the double CNL and CPC constructs had 3 mm of elongation at significantly lower loads than the constructs in which prostheses were secured to bone and those in which polyethylene tape (which has high stiffness) was se-

cured to the fabella. The load on CrCLs in dogs at a walk is approximately 50 N, and the load on CrCLs during vigorous activity ranges from 400 to 600 N.¹¹ None of the constructs in this study had 3 mm of elongation at loads generated in stifle joints of dogs at a walk, which would be an acceptable amount of activity during convalescent periods after extracapsular stabilization surgery. All constructs in this study had 3 mm of elongation at a single monotonic load consistent with forces generated during vigorous activity of dogs; this finding supported the clinical recommendation for restriction of activity during a postoperative convalescent period.

In vivo conditions cannot be fully replicated in studies in which cadaveric tissues are used. In the present study, we chose to investigate the effects of the method used to secure prostheses to femurs on the mechanical properties of several types of constructs used for extracapsular stabilization of stifle joints of dogs. We did not investigate effects of securing of prostheses to tibias or effects of extra-articular soft tissues. The effects of drill holes in femurs on mechanical properties of constructs were indicated by the finding that creep was higher for single and double BTPT constructs than it was for isolated polyethylene tape and double polyethylene tape loops. Extra-articular soft tissues, such as collateral ligaments, can restrain motion of stifle joints in dogs and contribute to stability of CrCL-deficient stifle joints.²⁶ Such soft tissue structures might improve the in vivo mechanical performance of the stifle joint stabilization methods evaluated in the present study.

Comparison of data determined in the present study with data determined in other studies would be difficult. We chose to investigate the mechanical properties of constructs incorporating femoral attachments. To the authors' knowledge, results for such constructs incorporating polyethylene prostheses have not been previously reported. Comparison of data determined in the present study regarding isolated loops of prosthetic material to data determined in other studies^{14,15} regarding similar prostheses would be hampered by different sizes of prostheses, numbers of strands of material in prostheses, lengths of prosthetic material, methods of securing prostheses to testing machines, and loading protocols used among these studies. All of these factors can affect the mechanical properties of constructs, which would make comparisons among data inaccurate.

Our hypotheses that femoral constructs in which prostheses were secured to bone would have better mechanical properties than those in which prostheses were secured around lateral femoral fabellae and that femoral constructs would have worse mechanical properties than corresponding isolated loops of prosthetic material were not supported. None of the femoral construct groups had superior mechanical characteristics for every variable. However, on the basis of lower creep and stress relaxation, greater stiffness, and higher loads required for 3 mm of displacement, the BAPT, double BTPT, and double CPT constructs were considered mechanically superior to the other femoral constructs tested. Multiple factors, such as mechanical properties of prostheses, viscoelastic properties of femoral and tibial attachments, method of securing prostheses to

tissues, method of securing loops of prostheses (knot vs crimp clamp), and biological responses to prostheses, contribute to the mechanical characteristics of constructs for extracapsular stabilization of stifle joints in dogs. Further studies are warranted in which mechanical properties of the extracapsular joint stabilization techniques used in the present study are investigated in conditions that are similar to in vivo conditions. Such conditions may include incorporation of both femoral and tibial anchorage points and more extensive testing of constructs after cyclic loading, which may allow more accurate determination of the in vivo mechanical properties of constructs and better prediction of the performance of extracapsular stabilization prostheses in dogs with CrCL deficiency.

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- a. FiberWire, AR-7210, Arthrex Vet Systems, Naples, Fla.
 - b. FiberTape, AR-7237, Arthrex Vet Systems, Naples, Fla.
 - c. CRMS80, Securos, Fiskdale, Mass.
 - d. CR12520P, Securos, Fiskdale, Mass.
 - e. CRSC080, Securos, Fiskdale, Mass.
 - f. CRMS806, Securos, Fiskdale, Mass.
 - g. FiberWire canine cruciate kit, VAR-2000, Arthrex Vet Systems, Naples, Fla.
 - h. 3.5-mm cannulated drill bit, AR-8920DC, Arthrex Vet Systems, Naples, Fla.
 - i. TightRope CCL, VAR-2800, Arthrex Vet Systems, Naples, Fla.
 - j. Spade-tipped drill bit, AR 1927D, Arthrex Vet Systems, Naples, Fla.
 - k. SwiveLock punch/tap, AR 1927CTB, Arthrex Vet Systems, Naples, Fla.
 - l. 5.5-mm SwiveLock C, VAR-2323 PSLK, Arthrex Vet Systems, Naples, Fla.
 - m. 858 MiniBionix II, MTS Systems Corp, Eden Prairie, Minn.
 - n. Multipurpose testware, MTS Systems Corp, Eden Prairie, Minn.
 - o. SPSS, version 15.0 for Windows, SPSS, Chicago, Ill.
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