The antebrachium in dogs is prone to several conditions that can adversely affect function. These include ligamentous injury, fracture, luxation, osteoarthritis, and primary bone neoplasia. Treatment of these conditions often requires pancarpal arthrodesis. Implants used in pancarpal arthrodesis must withstand substantial loads, especially because plates are most often applied to the compression side of the radiocarpal joint.

In dogs with osteosarcoma of the distal portion of the radius, limb-sparing surgery has been successfully performed by means of primary tumor resection and placement of an allograft or metal endoprosthesis during pancarpal arthrodesis. Many currently used metal limb-sparing endoprostheses are large and composed of stainless steel. It has been hypothesized that these implants are affected by factors such as excessive weight, which may contribute to fixation failure.

Investigators have compared the metal endoprosthesis and allograft limb-sparing techniques in a prospective clinical study and a biomechanical study. There was discordance between failure-mode results for the acute biomechanical testing and for the dogs in the clinical study. The most common failure mode of these endoprostheses in clinical situations is loss of screw purchase in the proximal radius bone, which leads to loosening of the device and the need for further surgical intervention, such as revision surgery or limb amputation.

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To date, design modifications to the endoprosthesis have not been evaluated with rigorous biomechanical testing or prospective clinical trials. From the standpoint of design development of an implant, testing these implant modifications in vivo requires a substantial investment of resources (namely, time, money, and cadaveric tissue). The finite element method has been used to develop computational models of the human carpus. A finite element model of the canine radius has been developed to determine loading conditions that result in replication of in vivo strain fields. However, to our knowledge, development of a comprehensive finite element model of the canine carpal joint has not been reported.

Our laboratory group has developed a computational (finite element) model of the canine antebrachium from the elbow joint to the metacarpal bones to enable virtual evaluation of prostheses and modifications to surgical procedures used in the canine antebrachium to predict failure modes and optimize implant design. The key steps in developing an accurate finite element model include simulating the anatomic structures (geometric aspects) and assigning the appropriate material properties to the constituent components of the model. Stiffness of the ligaments in the human wrist has been reported. However, fundamental anatomic differences exist between the antebrachial joints of humans and dogs; therefore, these data cannot be directly extrapolated to canine tissues. Additionally, we are not aware of any published reports that provide descriptions of the mechanical properties of canine carpal ligaments. Therefore, the objective of the study reported here was to establish a relevant dataset of the force-displacement and stress-strain relationships of the ligaments in the canine forelimb to provide data for incorporation into a finite model of the canine antebrachium.

Materials and Methods

Sample population—Twenty-six carpal sections were collected from the cadavers of 13 dogs. Mean ± SD body weight of the dogs (7 females and 6 males) was 28.7 ± 5.69 kg. The dogs were euthanized for reasons unrelated to this study.

Sample collection and preparation—Six ligaments of interest (AMC-IV, AMC-V, PR, PU, MC, and LC) were collected (Figure 1). Because of size and anatomic limitations, all 6 ligaments were not harvested from each dog.

The forelimbs were disarticulated at the shoulder joint and frozen (−20°C) until dissection. For dissection, forelimbs were thawed; ligaments were then isolated as bone-ligament-bone preparations (with at least 10 mm of bone included on each side of each ligament), wrapped in gauze soaked in saline (0.9% NaCl) solution, and stored again at −20°C until the day of testing.

Specimens were thawed at ambient temperature for 8 hours prior to mechanical testing. The bone-ligament-bone preparations were potted in PMMA and coupled to custom-designed mechanical fixtures. All possible care was used to ensure that the long axis of the ligament was aligned with the tensile axis of a servohydraulic materials testing machine. In addition, the PMMA pot at 1 end of each ligament was attached to a universal joint that allowed orientation of the ligament along the tensile axis when tension was applied.

Measurement of cross-sectional area and length—To transform the load data into stress variables, the cross-sectional area of each ligament was measured by use of a custom-designed apparatus that included a high-resolution micrometer (Figure 2). A compressive load was applied across the cross-sectional area, which resulted in a constant pressure of 0.12 MPa. Because of the shape of the fixture, the area under consideration was modeled as a rectangle and its corresponding dimensions were the height of the specimen (as determined by the value obtained by use of the micrometer) and the width of the fixture occupied by the ligament. Cross-sectional areas for the PR, MC, and LC ligaments were measured with handheld Vernier calipers because of the small dimensions of these ligaments.

Mechanical testing—Quasistatic tensile tests were performed on all specimens (n = 8 specimens/ligament) by use of a servohydraulic MTS in conjunction with a 6-df load cell. Each specimen was preconditioned for 10 cycles by applying 2% strain by use of a Haversine waveform. Tension was subsequently applied at a strain rate of 0.5%/s to each specimen until ligament failure,
which was defined as a sharp reduction in the mono-
tonically increasing load-displacement data.

Stereophotogrammetry was used to ensure there
was no slippage between the fixtures and PMMA during
testing. Specifically, 3 reflective markers were affixed to
specimens in the experimental apparatus (1 was affixed
on each of the 2 mechanical fixtures, and the third was
sutured to the midpoint of the ligament; Figure 2).
These markers were tracked via real-time monitoring
by use of a 3-camera system. When slippage was de-
tected during any portion of the testing, the resultant
specimen data were excluded from the analysis.

Data analysis—Force and displacement data were
synchronized by use of a light-emitting diode that was
triggered by the MTS software and recorded by means
of the stereophotogrammetry cameras during testing.
Data were recorded at the same rate (60 Hz) by use of
the MTS and camera system. Displacement data were
obtained from the MTS, and the resultant strain was
calculated by use of the equation $\varepsilon = (l_f - l_o)/l_o$, where $\varepsilon$
is the engineering strain, $l_f$ is the final ligament length,
and $l_o$ is the initial ligament length. Force data obtained
by use of the MTS were divided by the cross-sectional
area of the specimen to obtain the engineering normal
stress (ie, $\sigma$). Stress-strain data were plotted, and the
stiffness coefficient was obtained as the slope of the lin-
eral region of the associated force-displacement curve
(Figure 3). In addition, slope of the linear region was
used to obtain the elastic modulus (ie, Young's modu-
lus) of the ligament. Failure load for each specimen was
obtained at the maximum point in the load-displace-
ment plot from the MTS. The failure mode for each
ligament was also recorded.

Stiffness, modulus, ultimate strength, and cross-
sectional area were compared among ligaments by
use of a 1-way ANOVA with the Student-Newman-
Keuls post hoc test for multiple comparisons. All
data were evaluated for normality. A rank transforma-
tion was performed on data that were not normally
distributed. Ligaments were also grouped into 3
groups (AMC ligaments [AMC-IV and AMC-V], in-
tra-articular ligaments [PR and PU], and palmar car-
pal ligaments [MC and LC]). Values of $P < 0.05$ were
considered significant.

Results

Mean values of elastic modulus for each ligament
were determined (Figure 4). The mean modulus did
not differ significantly ($P = 0.142$) between AMC-IV
and AMC-V ligaments. The AMC-IV ligament had the
highest mean $\pm$ SD modulus of all ligaments tested with a value of $546.06 \pm 106.97$ MPa, followed by the AMC-
V ligament with a mean modulus of $382.38 \pm 180.50$ MPa. Elastic modulus did not differ significantly between
the PU and PR ligaments ($P = 0.856$) or between the MC and LC ligaments ($P = 0.196$). Values for elastic modulus differed significantly ($P < 0.001$) among the 3 ligament groups.

Mean stiffness coefficients for each ligament were determined. Mean $\pm$ SD stiffness did not differ significantly ($P = 0.095$ or greater for all comparisons)
among the MC ($72.65 \pm 10.86$ N/mm), LC ($61.10 \pm 30.42$ N/mm), PR ($80.20 \pm 41.21$ N/mm), PU ($94.70 \pm 14.43$ N/mm), and AMC-IV ($72.33 \pm 14.66$ N/mm) ligaments. Mean stiffness of the AMC-V ligament ($145.86 \pm 49.44$ N/mm) was significantly ($P = 0.014$ or greater for all comparisons) higher than that for all other ligaments.

Mean failure loads for each ligament
were determined. Mean $\pm$ SD failure loads did not differ significantly ($P = 0.84$ or greater for all comparisons) among the AMC-IV ($426.15 \pm 100.79$ N), MC ($392.45 \pm 132.61$ N), and PU ($414.66 \pm 72.29$ N) ligaments. Mean failure load for the AMC-IV ligament differed significantly, compared with the mean failure load for the PR ($P < 0.001$) and LC ($P = 0.013$) ligaments. The AMC-IV ligament had the highest mean failure load ($602.54 \pm 165.22$ N) and differed significantly ($P = 0.005$ or
Discussion

To our knowledge, the study reported here is the first in which mechanical properties of canine carpal ligaments have been described. Computational methods, such as finite element analysis, require accurate representation of soft tissues (including the ligaments) to reproduce the physiologic function of the joint or joints evaluated. Hence, we evaluated the mechanical properties of canine carpal ligaments for use in a finite element model of the canine antebrachium in this study.

The AMC ligaments function to prevent hyperextension of the middle carpal joint. In the study reported here, we found that they have relatively higher values for elastic modulus, compared with other ligaments evaluated in this study, which may correspond to their function and the forces they are subjected to in normal carpal extension. The PR and PU ligaments are intra-articular structures and restrict cranial and caudal instability. It has been postulated that the PU ligament is one of the principal stabilizers of the canine antebrachiocarpal joint and prevents caudal translocation. Values for the elastic modulus of these ligaments were found to be significantly (P < 0.001) higher than those of the MC and LC ligaments. The MC ligament restricts valgus deviation, and the LC ligament prevents varus deviation. These are primarily supportive ligaments and had the lowest values for elastic modulus of all the ligaments tested.

Failure modes for each ligament were summarized (Table 1). The highest number of midsubstance failures was for the MC and the PU ligaments, with 7 of 8 specimens failing at the midsubstance for each of these ligaments. The LC and PR ligaments each had midligament failure in 5 of 8 specimens, with the remaining failures via bony avulsions. Six of 8 specimens of the AMC-V ligament failed at the bone-ligament interface. The AMC-IV ligament had 3 specimens with midligament failure, 3 with bony avulsions, and 2 with failure at the bone-ligament interface.

Data on ultimate strength of the 6 ligaments were calculated (Figure 5). Ultimate strength was obtained by dividing the failure load by the cross-sectional area. Statistical analysis of the ultimate strength data yielded results similar to those for the elastic modulus. Specifically, significant differences were detected among the 3 ligament groups. However, ultimate strength did not differ significantly between the AMC-IV and AMC-V ligaments (P = 0.533), between the MC and LC ligaments (P = 0.892), or between the PR and PU ligaments (P = 0.197).

Mean cross-sectional area was determined for each ligament. The MC ligament had the highest mean ± SD cross-sectional area (26.68 ± 11.34 mm²); this value was significantly (P = 0.029 or greater for all comparisons) higher, compared with the values for the other ligaments. Mean ± cross-sectional area of the LC ligament (17.93 ± 11.01 mm²) differed significantly, compared with the area for the PR (6.75 ± 4.77 mm²) and AMC-IV (6.96 ± 1.98 mm²) ligaments, but did not differ significantly, compared with the area for the PU (10.82 ± 2.65 mm²) and AMC-V (12.29 ± 8.72 mm²) ligaments. Mean cross-sectional area did not differ significantly among the PR, PU, AMC-IV, and AMC-V ligaments.
not possible to obtain the instantaneous strain by use of stereophotogrammetry, which would have provided a more resolute strain measurement. Another common method for gripping soft tissue involves the use of cryogenic fixation via clamps. However, because of the small size of these ligaments, use of cryoclamps was not a viable option. Gauge lengths were approximately 15 mm for the MC, LC, and PR ligaments and approximately 25 mm for the PU and AMC-V ligaments. Cryoclamps would introduce major temperature gradients across the midsubstance of these ligaments and cause considerable changes in the mechanical properties.

The modes of failure indicated that of the 48 specimens tested, 28 (58.3%) failed at midsubstance, 11 (22.9%) had avulsion failure, and 9 (18.8%) failed at the bone-ligament interface. Most of the failures at the bone-ligament interface were for the AMC-V ligament. This may have been attributable to technical difficulties encountered in aligning this ligament along its anatomical (tensile) axis.

The high SD for the elastic modulus of the PR ligament could have been attributable to the fact that its cross-sectional area was too small to be measured with an area micrometer. Handheld Vernier calipers were used to determine the geometric area of these ligaments, which could have induced errors in measurement.

Other important soft tissue structures in the canine antebrachium aid in the stability of the joint, such as the palmar fibrocartilage and the flexor retinaculum. Harvesting intact specimens of these structures is inherently challenging; therefore, they were not included in this study. The triangular fibrocartilage and the interosseous ligament, which provide important stabilization of the antebrachial region, are being evaluated for their mechanical properties.

Analysis of the results obtained indicated a strong relationship between ligament function and elastic modulus. Because the stiffness coefficient does not take into account the geometry of a specimen, a relation between ligament function and stiffness cannot be expected. Such differences in the mechanical properties among ligaments would result in substantial improvements in the accuracy of a canine antebrachium finite element model.

A finite element model will have applications in optimizing the design of endoprosthetic implants for limb salvage after resection of primary osteosarcoma of the distal portion of the radius. It can also be used when designing focused experiments to assess modifications to surgical techniques. There are conflicting opinions on the role of preservation or resection of the distal portion of the ulna in stability of the carpal region after limb-salvage surgery and use of an endoprosthesis or allograft, as determined on the basis of biomechanical and clinical evaluation. The closest oncologic margin in many dogs with osteosarcoma of the distal portion of the radius is at the radio-ulnar articulation; hence, resection of the distal portion of the ulna is preferable from a surgical perspective because a wide surgical margin can be obtained. The distal portion of the ulna provides an insertion site for the LC, radioulnar, and PU ligaments. It also articulates with the distal portion of the radius, accessory carpal bone, and ulnar carpal bone. There is a nearly equal distribution of axial load at the proximal ends of the radius and ulna.13 Hence, from a biomechanical standpoint, the ulna appears to play a major role in load transmission. The importance of the ulna after limb-salvage surgery can be assessed by use of a finite element model of the antebrachium.

A total of 48 specimens of canine carpal ligaments were tested to obtain material properties, such as stiffness and elastic modulus, for use in development of a finite element model of the canine carpal joint. Quasi-static tensile tests were performed, and analysis of the results obtained indicated a strong function-elastic modulus relationship in the 6 ligaments evaluated. Future studies should evaluate the microstructure-function relationship by measuring total collagen content or determining the diameter of collagen fibrils by use of established protocols.

References


a. 809 Axial/Torsional Test System, MTS, Eden Prairie, NY.
c. Eagle-4 cameras, Motion Analysis Corp, Santa Rosa, Calif.
d. SigmaStat, version 3.1, Systat Software Inc, San Jose, Calif.