Over the past several years, many human spinal surgery studies have used finite element analysis (FEA) to investigate the risk of vertebral fracture and predict spinal stability after surgical treatment, demonstrating the usefulness of this technique.1–4 Several studies have also reported favorable comparisons between FEA and results from in vivo mechanical analyses, and the validity of FEA predictions is well established in human spinal surgery.5–7

More recently, the use of FEA has been gaining attention within the field of veterinary medicine, with reported uses for areas such as the cervical spine, elbow joint, femur, and ilium.8–11 However, there are few studies investigating FEA of the lumbar spine in animals. Lim et al and Yu et al described FEA of the canine lumbar spine,12,13 but these reports were only intended to verify the strength of implants used for human medical treatments and did not include biomechanical analyses of the canine lumbar spine for application to veterinary medicine. Among the biomechanical studies of the spine after surgical treatment for spinal disease conducted in dogs, most were ex vivo mechanical studies using cadavers14,15 or retrospective analyses of treatment results.
performed in clinical cases, both of which have notable limitations. In particular, although biomechanical studies performed on experimental animals or cadavers can provide important information, they are limited by the number of available samples and difficult to recommend from both an ethics and laboratory animal welfare viewpoint. Studies on clinical cases are primarily focused on treatment, and thus, investigation of experimental, high-risk surgical procedures is not acceptable.

Consequently, there is a need for a reliable method that can evaluate the mechanical properties of the canine spine without the use of live animals, and FEA is considered to be an attractive option. However, it is not clear whether FEA predictions for spinal segmental models of lumbar finite elements constructed based on canine computed tomography (CT) data are consistent with behaviors observed in ex vivo mechanical tests of real spinal segment models. Therefore, the purpose of this study was to investigate the validity of FEA predictions in the canine lumbar spine region. To this end, we compared ex vivo mechanical test results from a canine lumbar segment model with FEA predictions from a segment model constructed based on CT data from the same specimen using the lumbar finite element method. We hypothesized that, as observed for the human lumbar spine, results from FEA prediction would be consistent with those from ex vivo mechanical testing.

Methods

Animals

Six healthy adult beagle dogs were used to provide veterinary students with soft-tissue surgery training. Dogs received IV atropine sulfate (0.02 mg/kg) and midazolam (0.2 mg/kg) as a preanesthetic, followed by propofol (4–8 mg/kg) and endotracheal intubation. At the conclusion of the training session, all dogs were euthanized by intravenous pentobarbital (100 mg/kg) while the trachea was intubated. During the training session, all dogs underwent abdominal and thoracic incisions and sutures; however, the lumbar spine, including soft tissues, remained undisturbed. All dogs were untreated and showed no abnormalities in their general condition, blood tests, or neurological examinations before euthanasia. This study was conducted after obtaining approval from the Animal Experimentation Committee and Bioethics Committee of Nippon Veterinary and Life Science University (approval number: 2020K-10).

Preparation of ex vivo specimens

CT data from the lumbar spine were obtained from all experimental dogs using an 80-row/160-slice CT system (Aquilion PRIME; Toshiba Medical Systems). Immediately after euthanasia was performed, the L1–2 and L5–6 vertebrae were harvested for use in rotation and compression tests, respectively. Each specimen included the intervertebral discs, articular process cartilage, supraspinous ligaments, and interspinous ligaments, and all other soft tissues were removed. The prepared specimens were frozen at −80 °C until the day of ex vivo mechanical testing. Specimens were thawed at room temperature for 12 hours before analysis and were periodically sprayed with saline to prevent drying. Euthanasia and specimen preparation were conducted on March 26, 2020, and ex vivo mechanical tests were conducted on January 8, February 20, and March 26, 2022.

Ex vivo rotation test

The ex vivo rotation tests were performed using a torsion tester (TTM-3000N-ml; Shimadzu Corporation), as described below. The cranial half of the L1 vertebra and the caudal half of the L2 vertebra were first fixed with dental resin (OSTRON II; GC Dental Corporation). While fixing vertebrae with resin, sufficient care was taken to ensure that the resin did not interfere with the articular process joints. The caudal half of the L2 vertebra was then omnidirectionally restrained, and a rotational load was applied to the cranial half of the L1 vertebra (Figure 1). A rotational load was applied to the vertebral body axis (axis of rotation) without allowing more than 6 df. Only motion in the rotational direction was analyzed. For each cross-section of the cranial and caudal L1 vertebral endplates, the intersection point between the longest transverse diameter and the shortest longitudinal diameter was defined, and the line connecting these two intersection points was designated as the axis of rotation. Each specimen was preconditioned using a rotational test using a loading speed of 1°/second in both the left and right directions, and the preconditioning values were calculated as described by de Vicente et al. Preconditioning was performed to eliminate the effects of viscoelastic properties without causing any significant biomechanical damage to the vertebral motion unit. The point at which maximum torque (Nm) was obtained was defined as the yield point, and the maximum angle of gyration (°) and maximum torque at this point were recorded. The torque value at each 5% increment of the rotation angle was also recorded until the maximum rotation angle reached 100%.

Ex vivo compression test

The ex vivo compression testing was performed using a universal testing machine (AG-100KNX; Shimadzu Corporation), as described below. A compressive load was first applied to the cranial endplate of the L5 vertebra, such that it was perpendicular to the vertebral body axis (Figure 1). Compression load was applied to the vertebral body axis (Z-axis), allowing degrees of freedom only in the forward bending direction (X-axis) of the segment model, whereas the lateral bending direction (Y-axis) and the 3 rotational axes were fixed. During the procedure, we ensured that the compression load was applied to the entire surface of the cranial endplate of the L5 vertebral body and that the pressure plate did not contact the vertebral arch or articular process. A preload of 10 N was applied continuously for
1 minute, followed by a compressive loading rate of 0.2 mm/sec. The point of maximum stress (N) was defined as the yield point, and the maximum displacement (mm) and maximum stress value of the L5 vertebral endplate at this point were recorded. The stress value at each 5% increment in displacement was also recorded until the maximum displacement reached 100%.

Creation of an FEA segment model

Lumbar vertebral segment models were created using bone strength analysis software (MECHANICAL FINDER ver.11.0; Research Center of Computational Mechanics). FEA segment models of the L1–2 and L5–6 vertebrae were created using Digital Imaging and Communications in Medicine data obtained from CT imaging of the specimen before ex vivo mechanical testing. To more accurately calculate the bone density, CT imaging was performed simultaneously with the bone mass phantom. Heterogeneous material properties can be assigned to each finite element in the bone region of the model based on the bone density obtained from the CT scan. For example, a higher elastic modulus can be assigned to cortical bone, whereas a lower value can be assigned to trabecular bone. Young's modulus, Poisson's ratio, yield stress, and critical stress in the tensile direction can be calculated using conversion formulas based on the mass density. The bone strength analysis software was able to automatically convert mass density to Young's modulus by selecting a conversion formula based on the one reported by Keyak et al. The bone geometry of the L1–2 and L5–6 vertebrae was created using a 0.3–3.0-mm tetrahedral mesh (Figure 2). Adjacent vertebrae (ie, L1 and L2, L5 and L6) were set in different regions of interest, so they could be recognized as different bones. After assembling the bone model, the intervertebral discs, articular process cartilage, supraspinous ligaments, and interspinous ligaments were inserted. The intervertebral discs were specified as annulus fibrosus and nucleus pulposus. These soft tissue structures were imported into 3-dimensional (3D) modeling software (Metasequoia 4; Tetraface) with different contrast levels for more detailed visualization. Irregularities in the margins of the intervertebral discs were smoothed and reproduced as realistically as possible and then imported as STL files into the bone strength analysis software. The material properties were the same.
as those previously reported for dog models, as all of which used human values (Supplementary Table S1). The boundaries between the vertebral endplates and intervertebral discs were set as joined to prevent slipping. Because the articular process joint cartilage is quite small, it was reproduced as a structure adhering to the articular surface of the posterior articular process of the L1 or L5 vertebra. To include the mechanical contribution of the facet joints, a contact condition was defined between the surfaces of the articular facets, with a friction coefficient of 0.01.

### FEA rotation test

A cross-shaped rotation jig created using 3D-modeling software was placed on the cranial endplate of the L1 vertebra in the FEA segment model under fixation conditions. The caudal half of the L2 vertebra was omnidirectionally restrained, and a rotational load of 22 Nm was applied to the rotation jig (Figure 3; Supplementary Video S1). We then calculated the torque load based on the yield point of the ex vivo compression test. The axis of rotation for the L1 cranial endplate was set at the intersection of the longest diameter connecting the left and right edges of the endplate cross-section, and the axis of rotation for the L2 caudal endplate was set at the shortest diameter connecting the dorsoventral edges of the endplate cross-section. The maximum torque value obtained (Nm) and the amount of displacement as the abscissa, the displacement obtained at the maximum stress (100%) was divided equally (5% increments) into 20 values, and the average stress values for the corresponding displacement values were calculated.

### FEA compression test

The FEA segment model was omnidirectionally restrained at the caudal half of the L6 vertebra, and a compressive load of 5,500 N was applied to the cranial endplate of the L5 vertebra, perpendicular to the vertebral body axis (Figure 3; Supplementary Video S2). This compressive load setting was calculated based on the yield point of the ex vivo compression test. The compression axis was set to the same conditions as in the ex vivo compression test, and the compression device was placed such that it was in contact with the entire surface of the cranial endplate of the L5 vertebral body. We then recorded the maximum stress value (N) and the amount of displacement (mm) of the ventral edge of the L5 vertebral body cranial endplate at maximum stress. In addition, using the amount of displacement as the abscissa, the displacement obtained at the maximum stress (100%) was divided equally (5% increments) into 20 values, and the average stress values of the 6 specimens for the corresponding displacement values were calculated.

### Assessment of the FEA prediction validity

To validate the predicted FEA response, the results were compared to those obtained from ex vivo mechanical testing using 2 methods. For validation at failure, we calculated Pearson’s correlation coefficients between ex vivo mechanical test results and FEA predictions for the maximum torque and angle of gyration at maximum torque in the gyration test and the maximum stress and displacement at maximum stress in the compression test. Correlation strengths were defined as follows: negligible (r = 0 to 0.2), low (r = 0.2 to 0.4), moderate (r = 0.4 to 0.7), or high (r = 0.7 to 1.0). To verify the behavior until fracture, the ex vivo mechanical test results were compared with the stress values (Nm or N) at each 5% increment when the maximum angle of gyration or maximum displacement of the FEA prediction was set to 100%. If the mean FEA prediction value was within the mean ± SD of the ex vivo mechanical test value, the model was considered to have achieved validity for a particular loading condition.

### Results

#### Animals

Of 6 total dogs, 4 were female and 2 were male. The dogs were aged 22.0 ± 5.1 (mean ± SD) months and weighed 10.9 ± 1.4 kg.

#### Rotation tests

In the ex vivo rotation test, brittle fracture, indicated by a loud rupture sound and a sudden decrease in stress, was observed when the L1–2 specimens reached the yield point. Visual inspection of the fractured L1–2 specimens revealed crack failure at the base of the L2 anterior articular process in the direction of rotation and fracture of the articular process joint structure in all 6 specimens. The FEA segment model showed increased stress in the right lateral
anterior articular process of L2 and the left lateral posterior articular process of L1, and this behavior was similar in all 6 models (Figure 4). The maximum torque values and angles of gyration at the yield point from ex vivo rotation tests and corresponding FEA predictions for the L1–2 specimens are shown (Table 1). Pearson’s correlation coefficient was 0.92 for the maximum torque value and 0.96 for the rotation angle at the yield point, both indicating strong correlations. FEA rotation test predictions for torque values obtained at each 5% increment using the angle of rotation as the abscissa were not within the SD of the ex vivo rotation test results when the rotation angle was less than 40%. However, FEA predictions were within the SD when the rotation angle was greater than 45% (Figure 5).

Ex vivo compression tests showed static fracture with a gradual decrease in stress after the specimens reached the yield point. Visual inspection of the behavior at fracture revealed compressive failure of the ventral portion of the annulus fibrosus of the intervertebral discs in all 6 specimens. In FEA, compression of the ventral portion of the annulus fibrosus of the intervertebral discs caused the segments to flex and deform in the ventral direction. As a result, the ventral edges of the L5 caudal and L6 cephalad vertebral body endplates contact each other, indicating an increase in stress at these locations (arrow).

**Table 1**—Rotation and compression results from ex vivo mechanical tests and FEA predictions.

<table>
<thead>
<tr>
<th>Test</th>
<th>Model</th>
<th>Mean ± SD</th>
<th>Correlation coefficient (r)</th>
</tr>
</thead>
<tbody>
<tr>
<td>L1–2 rotation test</td>
<td>Max torque value (Nm)</td>
<td>Ex vivo</td>
<td>19.1 ± 1.25</td>
</tr>
<tr>
<td>Angle of rotation at the yield point (°)</td>
<td>Ex vivo</td>
<td>11.9 ± 1.38</td>
<td>0.96</td>
</tr>
<tr>
<td>L5–6 compression test</td>
<td>Max stress value (N)</td>
<td>Ex vivo</td>
<td>4617 ± 310</td>
</tr>
<tr>
<td>Displacement of the yield point (mm)</td>
<td>Ex vivo</td>
<td>5.9 ± 0.63</td>
<td>0.94</td>
</tr>
</tbody>
</table>

FEA = Finite element analysis. SD = Standard deviation.

**Discussion**

In the present study, we compared ex vivo mechanical test results for L1–2 and L5–6 vertebrae from healthy beagle dogs with FEA predictions from models created from the same specimens. Strong correlation coefficients of 0.92 and 0.73 were observed for the maximum torque value in rotation tests with
the L1–2 vertebrae and the maximum stress value in compression tests with the L5–6 vertebrae, respectively. Similarly, we detected strong correlation coefficients of 0.96 and 0.95 for the angle of rotation at the yield point in rotation tests and the displacement in compression tests, respectively. These results suggest that the FEA predictions of the canine lumbar segment models are sufficiently reliable as well as that of the human lumbar model. However, the correlation coefficient for the maximum stress value in compression tests was 0.73, representing a weaker correlation between the ex vivo study and the FEA prediction than was observed for other measures. In rotation tests, stress is primarily generated due to bone-to-bone contact between the articular processes, whereas in compression tests, stress is primarily generated due to the repulsive force of the intervertebral disc. In this study, CT imaging data of bone tissue was used to create the segment model to reproduce the material properties as closely as possible; however, the material properties of the intervertebral discs were based on human data. Therefore, the weaker correlations observed for the compression tests may be influenced by the stress generated by arbitrarily created intervertebral discs.

To the best of our knowledge, this is the first study to directly compare ex vivo mechanical test results and predicted FEA responses for the same specimen. Although previous reports have described the utility of FEA for examining the cervical spine of Great Danes, these studies used ex vivo mechanical tests performed on different individuals (foxhounds) to examine the validity of the model.8,26 Results showed that FEA predictions were within the maximum–minimum range of the ex vivo mechanical test results, indicating some validity of the model behavior. However, there was an overall tendency for the FEA predictions to underestimate the measured values and differences between segments. This discrepancy may be due to differences in the size and shape of the vertebrae and articular processes among the breeds.27 In our analyses, differences between breeds and among individuals were not an issue, and thus, the strategy employed in the present study should be considered a more suitable method for evaluating FEA validity. Notably, we observed no significant difference between the ex vivo mechanical test results and the FEA predictions in any of the 6 samples, indicating a strong correlation between these methods.

FEA is a useful method for understanding biomechanical behavior because it can predict stresses and displacements at the model's final rupture point, as well as changes over time during the loading process.28,29 However, previous reports using canine models only examined the results at failure and not during the loading process.8,12,21 Here, to assess FEA validity during loading, we calculated stress values for each 5% increment of the rotation angle at the yield point and the displacement at the yield point, each of which was considered to be 100%, and compared ex vivo mechanical test results to FEA predictions. FEA predictions were found to be overestimated at the beginning of the load test (0–40% rotation for the gyration model and 0–45% displacement for the compression model). However, predictions were within the SD of the ex vivo mechanical test results in the latter half of the loading tests (45–100% rotation for the rotation model and 50–100% displacement for the compression model). Buttermann et al used an implanted strain gauge in a dog to analyze the gait cycle and reported that the articular process was accompanied by a 4-mm displacement between the minimum and maximum loading. In the present study, the average vertebral endplate displacement at maximal loading was 5.9 mm in the ex vivo study and 6.3 mm in the FEA predictions. Due to differences in the landmarks used to measure displacement and because our study was based on an ex vivo mechanical test, our study results cannot be compared directly with physiological walking behavior. However, these results suggest that values equal to 60–70% of our study values are within the range of clinical significance during walking. We believe that FEA predictions may have validity in the medium to high load ranges, although the weak load range may not occur during clinical gait.

There are 3 possible reasons that may account for the differences between the ex vivo mechanical test results and FEA predictions during loading. The first is the use of human material properties to set the soft tissue conditions. In previous studies using FEA to model the canine spine, we reported no substantial problems in extrapolating soft tissue conditions with reference to human data.8,12,21 However, to improve the finite element model of the dog, the material properties of each specific canine tissue need to be clarified and inserted appropriately. We note that beagles are a chondrodystrophic breed,30 and although the average age of the individuals used in this study was relatively young at 22 months, there is a risk that they had already developed disc degeneration, potentially leading to a mismatch in conditions. The second possible explanation for the observed discrepancy is that the geometry of the intervertebral discs (ie, annulus fibrosus and nucleus pulposus), cartilage of the articular process joints, and ligamentous structures were arbitrarily created using 3D-modeling software and bone strength assessment software. In particular, the shapes of the boundaries between the annulus fibrosus and nucleus pulposus were carefully extracted while adjusting the contrast level to reproduce the shapes as faithfully as possible. However, because these structures were extracted manually, some errors could have occurred. The actual annulus fibrosus is composed of dozens of concentrically layered collagen bundles,31,32 but in this finite element model, it is represented as a single inorganic structure. Additionally, the supraspinous and interspinous ligaments were also reproduced as a single linear structure, which may have caused a discrepancy in conditions. The third explanation is the unique viscoelastic properties of the spinal unit. During spinal motion, physiologic mobility exists within the elastic zone of ligamentous structures, referred to as the neutral zone.
(NZ), which experiences a relatively minor increase in stress at initial loading. Therefore, to extract only the more substantial increases in stress in the spinal unit during mechanical testing, a preload should be applied to mimic the load produced by the NZ in the range of the initial region. Although the present study applied a preload based on the predictions of de Vicente et al, the degree to which our preload was able to overcome the viscoelastic properties remains unclear. Because FEA imparts the material properties of bone tissue and ligaments onto a model generated using inorganic tetrahedral mesh and truss elements, the unique viscoelastic properties during ex vivo mechanical testing are expected to be minimal. Therefore, a more rapid increase in the initial loading stresses may have occurred in the FEA model than in ex vivo mechanical testing.

There are several limitations of this study. First, the small size of the vertebrae used in ex vivo testing may have resulted in a relatively large area of resin fixation, which may have restricted the physiological movements of the segment to some degree. However, this risk may be unavoidable when using this experimental technique, as the range of fixation was as narrow as possible without allowing the specimen to slip during testing. The important structures involved in intervertebral motion are the articular process and intervertebral disc area, and preliminary manual and visual inspection confirmed that the resin fixation did not interfere with these areas, minimizing the physiological restriction of motion. Although we based our method for comparing FEA predictions with ex vivo mechanical test results during loading from published human studies, it is possible that FEA predictions were not within the SD of mechanical strength test results due to the smaller measured values in the early loading phase and the correspondingly smaller range of SDs. To solve this problem, it may be necessary to choose a measurement and analysis method that is limited to the low load region that occurs in vivo during the dog’s normal daily life, rather than comparing a wide range of loading from the onset of loading to rupture. In addition, although we used data from the same individuals in this study, the model segment only consisted of 2 vertebrae. If a longer segment model that more closely resembles the structure of the living body is considered, even small changes in values may result in large errors. Therefore, it may be necessary to use more accurate material properties for soft tissue structures, such as muscles and other ligaments, which were not reproduced in this model. In human spine studies, finite element models have been developed that include muscle tissue, such as the erector spinae and multifidus muscle fibers, and it is hoped that this approach will also be extended to the veterinary domain for an appropriate representation of muscle strength.

In summary, we present the first report comparing results from ex vivo mechanical testing and FEA predictions using completely identical canine specimens. Our results show that FEA predictions are reliable for biomechanical studies of the canine lumbar segment model, as they correlate with the ex vivo mechanical test results. Although this study is a basic examination of FEA in the canine spine, our findings suggest that, in the future, FEA could be within the veterinary field to analyze the biomechanical characteristics of the canine spine and predict stability after surgical treatment.

### Acknowledgments

None reported.

### Disclosures

The authors have nothing to declare. No AI-assisted technologies were used in the generation of this manuscript.

### Funding

The authors have nothing to declare.

### References


---

**Supplementary Materials**

Supplementary materials are posted online at the journal website: avmajournals.avma.org.