

Biomechanical Analysis of Atlantoaxial Dorsal Fixation Using Finite Element Models

Beomju Bae

Chungbuk National University College of Veterinary Medicine

Dongwook Kim

Chungbuk National University College of Veterinary Medicine

Hyejong Oh

Chungbuk National University College of Veterinary Medicine

Gonhyung Kim

ghkim@cbu.ac.kr

Chungbuk National University College of Veterinary Medicine <https://orcid.org/0000-0002-9384-4741>

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Abstract

Background: Atlantoaxial instability can cause spinal cord compression with clinical signs ranging from cervical pain to tetraplegia and death. Although a variety of dorsal fixation techniques have been described, some of them have been related to the fracture of the dorsal arch of the atlas, leading to surgical failure. We hypothesized that the shape of the dorsal arch of the atlas and types of implants might affect these bone fractures. Thus, the objective of this study was to analyze bone stresses through simulations of the dorsal fixation using finite element models.

Results: The width between wires and the length of the bone did not affect the maximum stress on the bone. The maximum bone stress increased as the bone got thinner and the angle of the notch got steeper. The bone with band implant had lower maximum bone stress than that with wire implants. When using wire implants, wires applied beyond the notch of the dorsal arch reduced the maximum bone stress more than wires positioned within it.

Conclusions: The fracture of the dorsal arch of the atlas was related to the shape of the bone and types of implant applied. Band implant can effectively reduce the fracture of the dorsal arch compared to wire implant in atlantoaxial dorsal fixation. When considering wire implant, it is recommended to apply wires beyond the notch of the atlas.

Background

Atlantoaxial instability (AAI) occurs due to congenital or traumatic cause, or a combination of them. Congenital AAI has been commonly described in immature toy breed dogs. It develops from a loss of atlantoaxial ligamentous support, abnormalities of the dens such as aplasia, hypoplasia, dorsal angulation, or nonunion of the ossification center, and incomplete ossification of the atlas [1–3]. Traumatic AAI can occur at any age in any breed of dog. It usually results from rupture of atlantoaxial ligaments and/or fracture of the dens, although most cases are related to underlying congenital anomalies [2, 4, 5]. Regardless of the cause, AAI can cause spinal cord compression with clinical signs ranging from cervical pain to tetraplegia and death [1, 4–6]. Treatment of AAI varies from conservative management to surgical stabilization including dorsal and ventral fixation techniques. Conservative management using a cervical splint is a viable option for young dogs with mild or acute clinical signs [7], but surgical stabilization is recommended for most dogs with AAI because conservative management has a risk of relapse and deterioration [5, 8]. Although ventral stabilization by arthrodesis and fusion can provide more rigidity than dorsal stabilization by fibrous tissue formation [2, 3], it is sometimes challenging in toy breed dogs due to their small sized bones, leading to surgical failure [6]. Thus, various dorsal fixation techniques including the use of wire, dorsal cross pinning with polymethylmethacrylate, nuchal ligament, prosthetic suture material, intermuscular suture, and metallic retractor have been described in toy breed dogs [3, 6, 9–11]. However, some of them have been related to the fracture of the dorsal arch of the atlas which can cause surgical failure with subsequent relapse [11]. Finite element models (FEMs) were introduced into human biomechanics in the early 70 s. They have been developed

with continuous advancement of computer technology [12]. Although FEMs have been widely used in human spine biomechanics to analyze fractures, injuries, and surgical techniques, only a few studies have used FEMs in canine spine biomechanics known to adapt human material properties [13–21]. To the author's knowledge, there are very few biomechanical studies for the fracture of the dorsal arch of the atlas in atlantoaxial dorsal fixation. Based on our clinical experience that proper positioning of the implant could reduce fracture of the dorsal arch in atlantoaxial dorsal fixation, we hypothesized that the shape of the dorsal arch of the atlas and types of implants might affect the fracture. Thus, the objective of this study was to analyze bone stresses through simulations of the dorsal fixation using FEMs so that we could find a way to reduce bone fracture.

Results

1. Project A

The maximum von Mises stress related to the width (W) between wires was 21.276 MPa in model 1.1 ($W = 2$ mm), 20.798 MPa in model 1.2 ($W = 4$ mm), and 20.851 MPa in model 1.3 ($W = 6$ mm). In model 2 related to the length (L) of the bones, the maximum von Mises stress was 20.798 MPa in model 2.1 ($L = 8$ mm), 20.793 MPa in model 2.2 ($L = 9$ mm), and 20.790 MPa in model 2.3 ($L = 10$ mm). The width between wires and the length of the bone did not greatly affect the maximum bone stress. The maximum bone stress increased as the thickness (Th) of the bones decreased. The maximum von Mises stress was 20.845 MPa in model 3.1 ($Th = 1$ mm), 14.674 MPa in model 3.2 ($Th = 2$ mm), and 9.158 MPa in model 3.3 ($Th = 3$ mm). As the oblique angle (D) of the notch increased from 15° to 45° , the maximum bone stress also increased [25.954 MPa in model 4.1 ($D = 15^\circ$), 34.904 MPa in model 4.2 ($D = 30^\circ$), and 40.978 MPa in model 4.3 ($D = 45^\circ$)]. Band implant reduced the maximum bone stress (4.481 MPa in model 5.2) more than wire implant (14.674 MPa in model 5.1).

2. Project B

The maximum bone stresses were 15.207 MPa under 20 N, 29.067 MPa under 40 N, and 50.231 MPa under 80 N in model 6.1. In model 6.2, the maximum bone stresses were 13.695 MPa under 20 N, 23.584 MPa under 40 N, and 40.557 MPa under 80 N. In model 6.3 with wires applied beyond the notch, the maximum bone stresses were 10.748 MPa under 20 N, 16.333 MPa under 40 N, and 27.452 MPa under 80 N. Thus, wires applied within notches of arch-shaped bones increased the maximum bone stress more than wires positioned beyond the notch. In model 6.4 using band implant showed the maximum bone stresses of 9.591 MPa under 20 N, 13.006 MPa under 40 N, and 23.944 MPa under 80 N. When using band implant, the maximum bone stress decreased compared to that when wire implants were used. Under tension forces of 20 N, 40 N, and 80 N, differences in maximum bone stresses between model 6.3 and model 6.4 were 1.157 MPa, 3.327 MPa, and 3.508 MPa, respectively. Under the forces of 20 N, 40 N, and 80 N, differences in maximum bone stresses between model 6.1 and model 6.4 were 5.616 MPa, 16.061 MPa, and 26.287 MPa, respectively. As the force increased from 20 N to 80 N, the difference in the maximum bone stress between model 6.1 and model 6.4 (20.671 MPa) increased much more than that

between model 6.3 and model 6.4 (2.351 MPa). Under the force of 80 N, the maximum bone stresses in models 6.1 and 6.2 were relatively closer to the bone yield stress than those in models 6.3 and 6.4 (Fig. 1 and Fig. 2).

Discussion

Models for biomechanical study such as in vitro, in vivo, and finite element models have greatly improved our understanding of biomechanics and medicine [22]. FEMs can evaluate internal response such as stress-related information which is difficult to obtain in in vivo and in vitro models [23, 24]. Stress analysis is important to understand the fracture mechanism. Thus, FEMs have been widely used for studies on bone fractures [13, 14, 25, 26]. We modeled the arch-shaped bone reflecting characteristics of the dorsal arch of the canine atlas and analyzed the behavior of the spinal bone fracture in atlantoaxial dorsal fixation using FEMs. In patients with atlantoaxial instability, ligamentous lesions such as disruption of atlantoaxial ligaments, osseous lesions such as dens fracture, or a combination of them cause distraction of atlantoaxial joint related to atlantoaxial subluxation [27, 28]. Thus, we applied tension forces to implants for simulations of atlantoaxial dorsal fixation which handles the distraction force. When giving tension forces to implants applied, the maximum bone stress increased as bone got thinner and the oblique angle got steeper. Considering the shape of the dorsal arch of the canine atlas which has the thin notch with steep oblique angle, the bone will get greater maximum stress if the implant is applied to this notch during dorsal fixation. In the arch-shaped bone based on atlas, the maximum bone stress was actually higher and closer to bone yield stress when applying wires within the notch compared to that when positioning them laterally beyond the notch. In addition, as tension force to implant increased, difference in the maximum bone stress between the bone with wires applied within the notch and the bone with them positioned beyond the notch also increased. Thus, the bone with wires positioned beyond the notch is less vulnerable to bone fracture than the bone with wires applied within it. One clinical study on dorsal stabilization for dogs with AAI has reported that the fracture of the dorsal arch is reduced with the use of band implant due to an increased contact surface between the bone and the implant, leading to better repartition of forces [6]. If the bone is flat and platy, band implant can reduce the maximum bone stress effectively compared to wire implants. However, the dorsal arch of the canine atlas has protrusions that can prevent band implant from reducing the maximum bone stress as the decreased contact surface can lead to higher concentration of the stress. In our study, the reduction of the maximum bone stress by band implant in arch-shaped bone (1.157 MPa) is less than that in the platy bone (10.193 MPa) under 20 N of the force. In addition, when the tension force to implants increased from 20 N to 80 N, the difference in the maximum bone stress between the bone with wires positioned beyond the notch and the bone with band implant was increased by only 2.351 MPa while that between the bone with wires applied within the notch and the bone with band implant was increased by 20.671 MPa. Taken together, although band implant can reduce the maximum bone stress and the risk of bone fracture compared to wire implants, wire implants can also decrease the risk of bone fracture if wires are applied beyond the notch. Several limitations of our models exist. There was no consideration for individual or breed variation during the modeling of the dorsal arch. However, characteristics reflected

in our arch-shaped bone model can be applied to dorsal arches of most toy breed dogs. In performing the study, material properties for human were used in this study due to the lack of those for canine. No significant issue has been reported in studies on canine spine using human material properties [19], but further studies are needed to set proper material properties for a more realistic behavior of canine bone model and to validate FEMs in canine study. Although spine response is nonlinear, linear analysis related to distraction force was conducted in this study. It is because linear models could make analysis simpler and have been successfully used to understand bone fractures in previous studies [14, 25, 29]. In addition to distraction force, other forces such as flexion, extension, shear force, lateral bending, and axial rotation are also possible in atlantoaxial joint and should be further considered.

Conclusion

These results showed that the fracture of the dorsal arch was highly associated with bone shapes and types of implants. Band implant reduced the maximum bone stress the most. When using wire implants, consideration for the notch of the dorsal arch is necessary to decrease the maximum stress in the bone.

Methods

Geometric and finite element modeling of bones and implants for simulations of atlantoaxial dorsal fixation and finite element analyses for the bone stresses were performed using Abaqus software program (Abaqus , Version 6.10; Abaqus, Inc., Providence, RI, USA).

Geometric modeling of bones and implants

Bones used in this study were simplified to platy or arch-shaped structures. Arch-shaped bone was modeled based on a three-dimensional (3D) computed tomography (CT) image for the dorsal arch of the atlas of a toy breed dog (a 9-year-old male Maltese dog) without any cervical diseases. The 3D CT image had been taken using a four-row multidetector CT scanner (Hi Speed QX/I; GE Medical Co., Milwaukee, WI, USA) and Medicine (DICOM) viewer (OsiriX MD 4.1.2; OsiriXPixmeo, Geneva, Switzerland). Characteristics of the arch-shaped bone were as follows: 1) the bone got slightly thicker as it went laterally; 2) the bone had a notch with steep oblique angle in the craniomedial part; 3) the bone had cranial protrusions on both sides of the notch; and 4) the lateral part of the bone beyond the notch had a gentle oblique angle. Details for the arch-shaped bone are shown in Fig. 3. Two types of implants (wire implant and band implant) were modeled and applied to bones. The diameter of the wire implant was 0.8 mm. The thickness of the band implant was also 0.8 mm.

Finite Element modeling

3D FEMs for simulations of dorsal fixation were generated using hexahedral and pentahedral elements (Fig. 4). On average, 36,023 nodes and 28,557 elements were used in each simulation. Bones were defined as cortical bones. Cancellous bones in the internal matrix were ignored. Implants were made of stainless steel. Material properties of bones and implants are shown in Table 1. They were assumed to be

homogenous and linear isotropic. The friction coefficient value for bone-steel is 0.37 [30]. Material properties of bones were adapted from a previous study of human atlas [13] because of the lack of material definition for canine spine. Material properties of implants were obtained from the literature on stainless steel [31, 32].

Experimental design

Two studies were performed and referred to as project A and project B (Fig. 5 and Fig. 6). Project A using platy bones was carried out to explain stress-related results of project B. Project B using arch-shaped bones was implemented to simulate the clinical situation of dorsal fixation.

1. Project A

In project A, influences of various conditions such as bone shapes, implant types, and positioning of implants on the platy bones were investigated. Conditions were varied in five models. Model 1 was designed based on the width between wires. Platy bones with a width of 10 mm, a length of 8 mm, and a thickness of 1 mm were used and wire implants were applied to bones. The width between wires was 2 mm in model 1.1, 4 mm in model 1.2, and 6 mm in model 1.3. Model 2 was designed based on the length of the bones. Platy bones with a width of 10 mm and a thickness of 1 mm were used. Wire implants with a width of 4 mm between wires were applied to bones. The length of bones was 8 mm in model 2.1, 9 mm in model 2.2, and 10 mm in model 2.3. Model 3 was designed based on the thickness of bones. Platy bones with a width of 10 mm and a length of 10 mm were used. Wire implants with a width of 6 mm between wires were applied to bones. The thickness of bones was 1 mm in model 3.1, 2 mm in model 3.2, and 3 mm in model 3.3. Model 4 was designed based on the oblique angle of notches. Platy bones with a notch were used. They had a width of 10 mm and a thickness of 1 mm. Wire implants with a width of 4 mm between wires were applied to oblique planes of notches. The oblique angle was 15° in model 4.1, 30° in model 4.2, and 45° in model 4.3. Model 5 was designed based on types of implants. Platy bones with a width of 10 mm, a length of 10 mm, and a thickness of 2 mm were used. Wire implants with a width of 6 mm between wires and band implant with a width of 6 mm were applied in model 5.1 and model 5.2, respectively (Fig. 5).

2. Project B

In project B, influences of types and positioning of implants on arch-shaped bones were investigated. Four models (referred to as model 6) were used in project B. Wire implants were applied within notches of arch-shaped bones in models 6.1 and 6.2 and beyond the notch in model 6.3. The width between wires was 2 mm in model 6.1, 4 mm in model 6.2, and 6 mm in model 6.3. Band implant with a width of 6 mm was applied in model 6.4 (Fig. 6).

Finite element analyses

Arbitrary tension forces of 20 N were applied to implants in project A while 20 N, 40 N, and 80 N of tension forces were applied to implants in project B for simulations of the dorsal fixation. The maximum

bone stresses under these forces did not exceed the yield stress of cortical bone. Considering our bone elastic modulus (10 GPa), the bone yield stress in this study was 66.4 MPa [yield stress (MPa) = 24.4 + 4.20 × elastic modulus (GPa)] [33]. If the maximum bone stress gets close to the yield stress, the bone will be vulnerable to bone fracture. The maximum von Mises stress in the bone for each model was measured and compared to each other to find out which models had lower maximum stresses.

List Of Abbreviations

AAI

Atlantoaxial instability; FEMs:Finite element models; 3D:Three-dimensional; CT:Computed tomography; W:Width; L:Length; Th:Thickness; D:oblique angle

Declarations

Ethics approval and consent to participate

As the CT image for the dorsal arch of the atlas was obtained from the client-owned dog, an informed consent was taken by the owner before using the image. This study was approved by the Chungbuk National University Institutional Animal Care and Use Committees (CBNUA-1429-20-01).

Consent for publication

Not applicable.

Availability of data and materials

The datasets used and/or analysed during the current study are available from the corresponding author on reasonable request.

Competing interests

The authors declare that they have no competing interests.

Funding

Not applicable.

Authors' contributions

The conception or design of the work: BB and GK; analysis and interpretation of data: BB, HO, and GK; have drafted the work or substantively revised it: BB, DK, and GK; All authors read and approved the final manuscript.

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Tables

Due to technical limitations, table 1 is only available as a download in the Supplemental Files section.

Figures

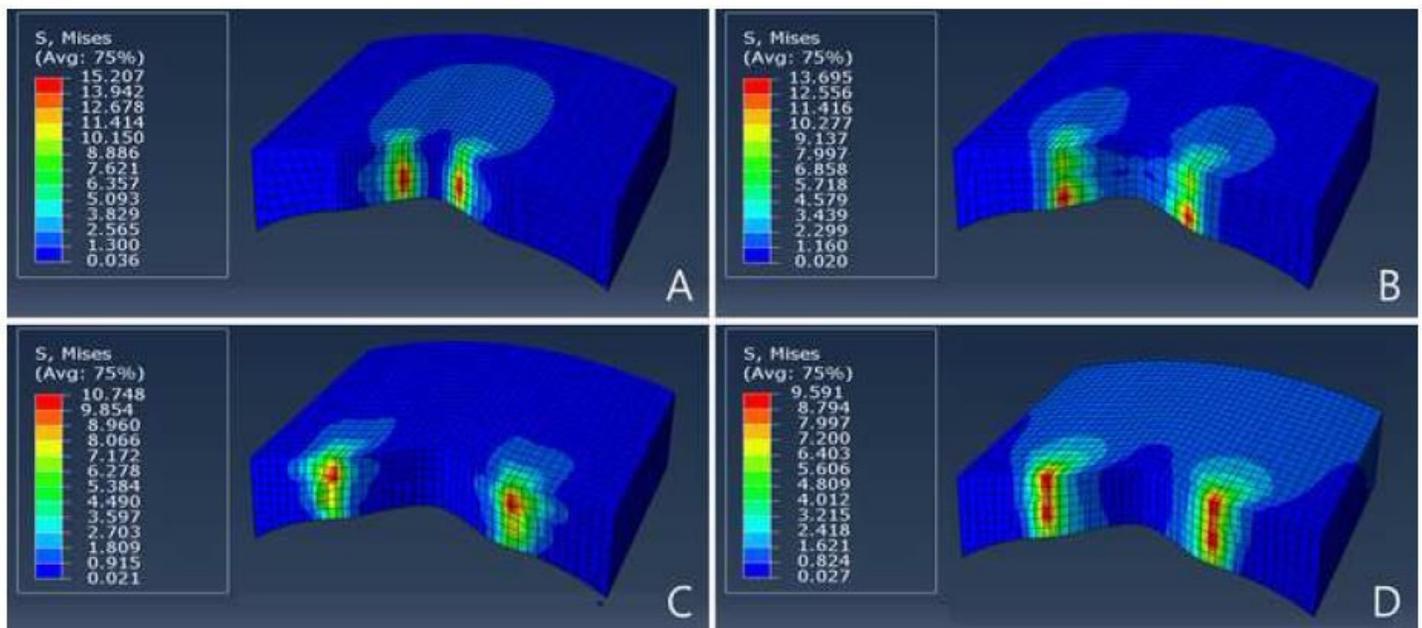


Figure 1

Von Mises stress distributions in arch-shaped bones according to the type and positioning of the implant (model 6). Wire implants were applied within notches in models 6.1 (A) and 6.2 (B). Wire implant was applied beyond the notch in model 6.3 (C). Band implant was applied in model 6.4 (D)

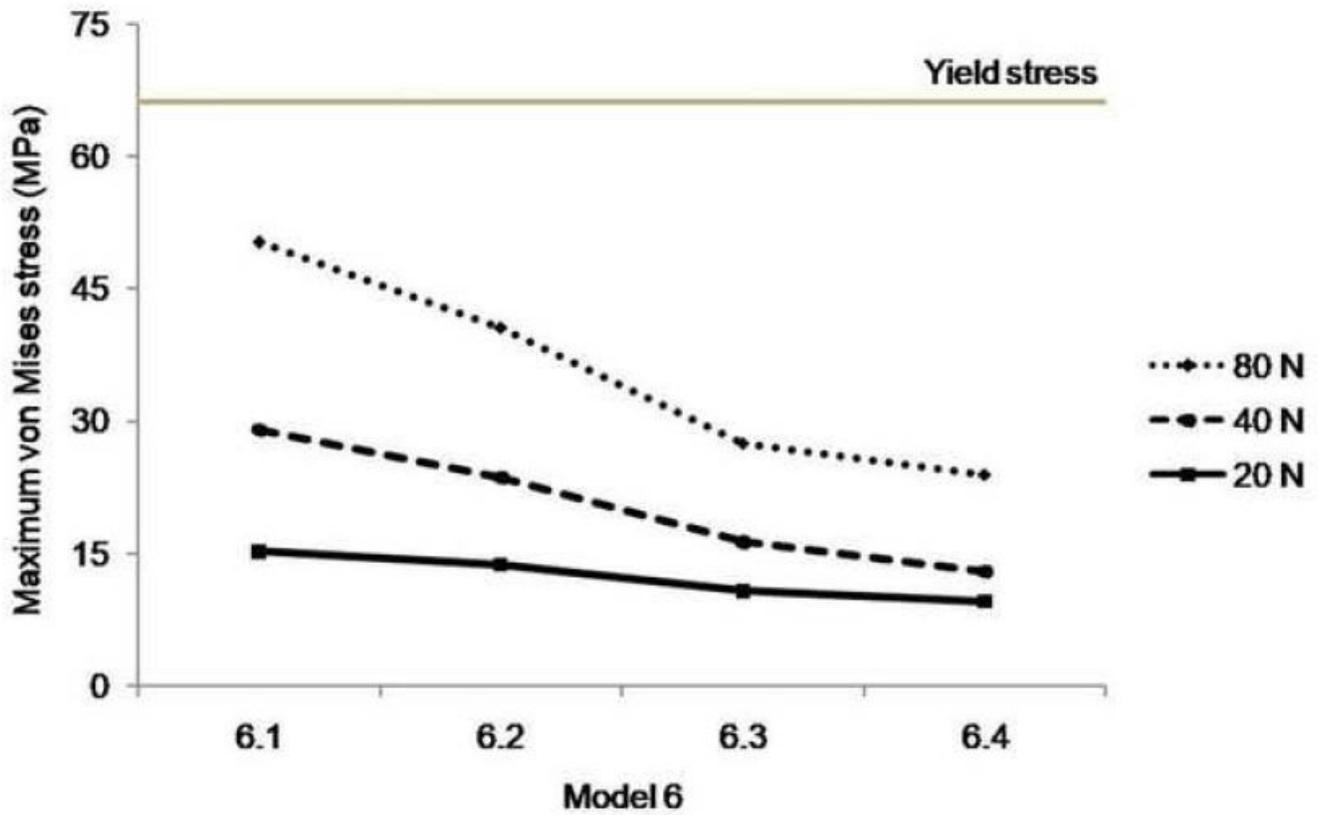


Figure 2

The maximum von Mises stresses under tension forces of 20 N, 40 N, and 80 N in model 6. Wire implants were applied within notches in models 6.1 and 6.2. Wire implant was applied beyond the notch in model 6.3. Band implant was applied in model 6.4

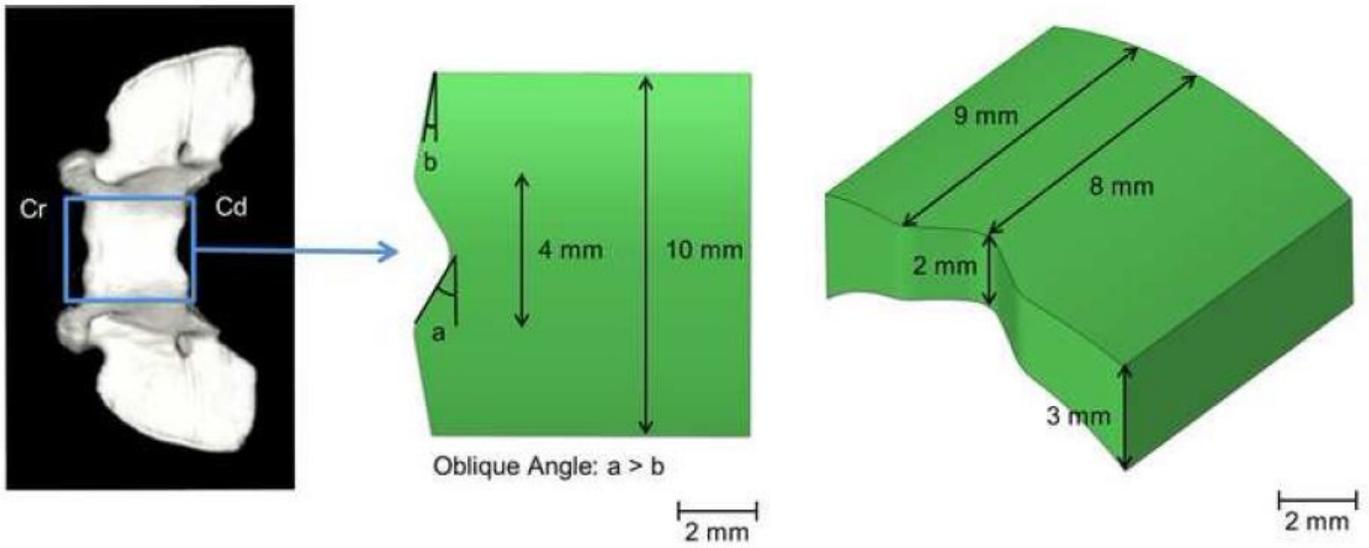


Figure 3

The geometry of an arch-shaped bone based on the dorsal arch of the atlas of a 9-year-old male Maltese dog without any cervical diseases (Cr: cranial, Cd: caudal)

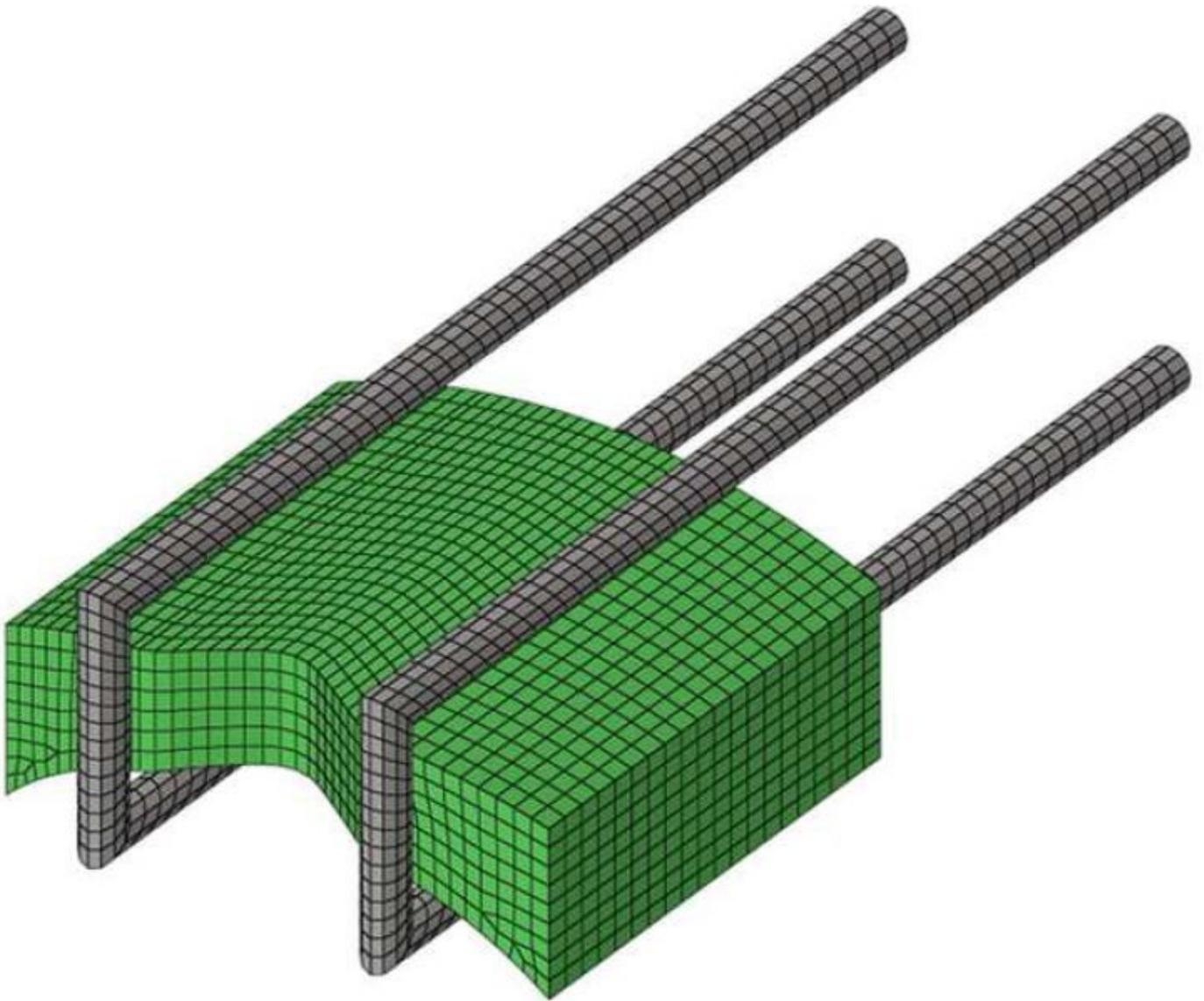


Figure 4

A finite element model representing the arch-shaped bone and wire implants with a width of 6 mm between wires applied for dorsal fixation

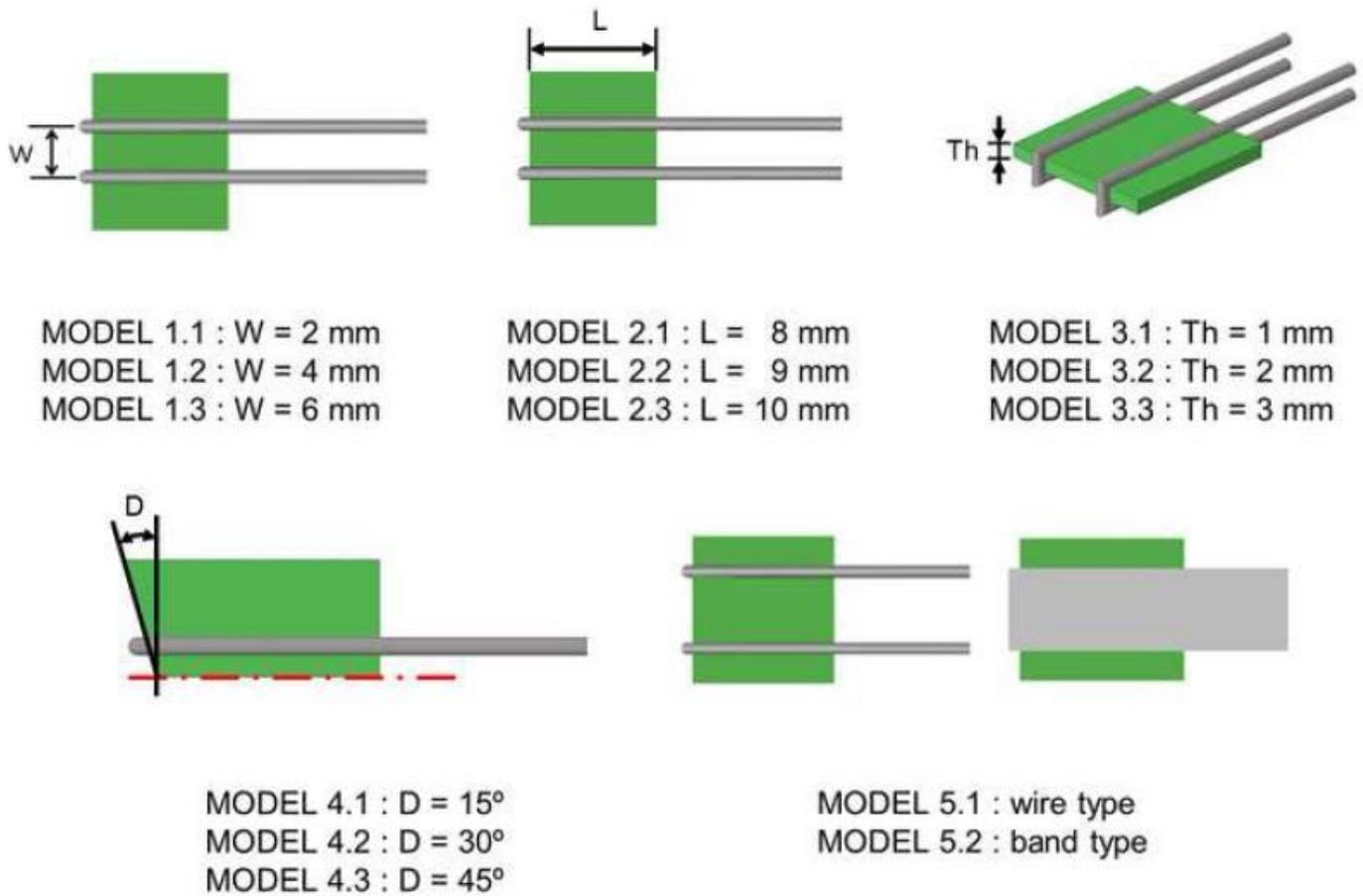
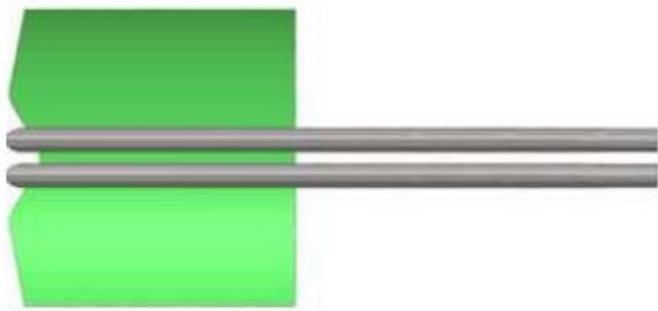
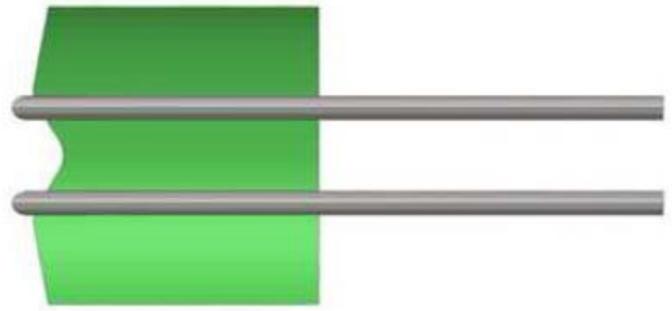


Figure 5

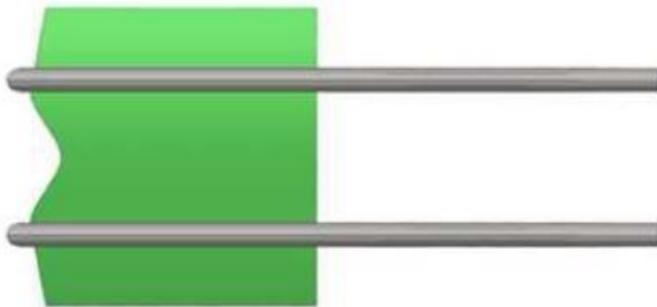
Project A (models 1-5) using platy bones under various conditions. Models were designed based on the width between wires in model 1, the length of bones in model 2, the thickness of bones in model 3, the oblique angle of notches in model 4, and types of implants in model 5 (W: width, L: length, Th: thickness, D: oblique angle)



MODEL 6.1



MODEL 6.2



MODEL 6.3



MODEL 6.4

Figure 6

Project B (model 6) using arch-shaped bones designed based on types and positioning of implants. Wire implants were applied within notches of arch-shaped bones in models 6.1 and 6.2. They were applied beyond the notch in model 6.3. The band implant was applied in model 6.4

Supplementary Files

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