Original Article

Comparison of biomechanical properties associated with the diameter and insertion depth of bone anchors in the femoral condyle of toy breed cadaver dogs

Yun-Hee An[1] Mu-Young Kim[1] Hun-Young Yoon[2*]

Abstract

The objective of the present study was to compare the biomechanical properties of implanted bone anchors based on diameter and insertion depth to yield selection criteria for bone anchor size in toy breed dogs. Twenty toy breed dog cadavers weighing < 5 kg underwent placement of five types of veterinary bone anchors (short diameter, short length [SDSL]; short diameter, medium length [SDML]; medium diameter, medium length [MDML]; medium diameter, long length [MDLL]; and long diameter, medium length [LDML]) at a predetermined femoral attachment site. Anchor screw depth/femoral condyle width (FCW) and anchor screw diameter/femoral condyle length (FCL) were measured using radiography. The yield load, Young's modulus and failure load were measured and the causes of failure for each construct were recorded. The anchor screw depth/FCW was < 50%, 50%–75%, ~50%, and 75%–100% in the SDSL, SDML, MDML, and MDLL groups, respectively. The yield load, Young's modulus and failure load were higher in the SDML and MDLL groups than in the SDSL and MDML groups. The anchor screw diameter/FCL was 12%–15% and 24%–30% in the SDML and LDML groups, respectively. No differences in biomechanical parameters were found between the SDML and LDML groups. The cause of failure in all constructs was pull-out of the bone anchor, except for distal femur fracture in five LDML constructs. In conclusion, when implanting bone anchors in toy breed dogs, the insertion depth should be > 50% of the FCW, regardless of diameter. Moreover, distal femur fracture can occur if the bone anchor diameter/FCL ratio exceeds 24%.

Keywords: dog, cranial cruciate ligament, bone anchor, biomechanical property

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Introduction

Cranial cruciate ligament (CrCL) deficiency is the most common cause of hind limb lameness in dogs (Kowaleski *et al*. 2012). This term broadly includes complete rupture or partial tearing of the ligament with a traumatic or degenerative etiology (Innes *et al*. 2000). Dogs with CrCL-deficient stifles are vulnerable to cranial translation of the tibia and are unable control the gait of the hind limb or muscle forces across the stifle (Kishi *et al*. 2013). CrCL deficiency causes pain, inflammation, stifle instability, meniscal injury and degenerative joint disease (Brown *et al*. 2017). Because conservative treatment of CrCL deficiency is usually unsuccessful, surgical stabilization of the CrCL is the preferred method of treatment (Hulse *et al*. 2010).

In toy breed dogs, extracapsular stabilization is one of the most common surgical treatment options for CrCL deficiency due to its less technically demanding nature and lower cost compared to other techniques (Comerford *et al*. 2013). The common concept underlying various extracapsular stabilization surgical techniques is that craniocaudal tibial instability during weight-bearing is restricted by the use of biological or synthetic materials that span the stifle joint on its lateral aspect and outside the joint capsule, mimicking CrCL function (Muir, 2018). The efficacy of artificial suture construct placement has been validated by demonstrating transient fixation of soft tissue to bone until sufficient periarticular fibrosis has occurred (Giles *et al*. 2008).

Despite the increasing usage of bone anchor placement in CrCL repair, research on the accurate biomechanical properties based on the diameter and insertion depth of the bone anchor in toy breed dogs is insufficient. Since no indicators have been studied to determine the ideal size of bone anchor in toy breed dogs, the size was selected according to the surgeon's subjective judgment of the weight and activity level of the patient during surgical repair of CrCL deficiency (Raske and Hulse, 2013). The objective of this study was to compare the biomechanical properties of bone anchors of different sizes in association with the diameter and insertion depth, thus providing selection criteria for bone anchor size in toy breed dogs.

Materials and Methods

Specimen preparation: Forty femurs and tibias were harvested from the pelvic limbs of 20 toy breed dogs that had been euthanized and donated for reasons unrelated to this study. Processes such as the collection and use of cadaveric pelvic limbs were approved by the Institutional Animal Care and Use Committee of Konkuk University (approval number: KU22007).

All tissues, except the distal two-thirds of the femur, tibia and lateral fabellae, were removed. Each specimen was wrapped in double-layered plastic bags, stored at -20°C, and thawed to room temperature 24 h before the experiment.

Grouping: The veterinary bone anchors were classified by size into five groups (Table 1): short diameter, short length (SDSL; 2.0 × 6 mm); short diameter, medium length (SDML; 2.0 × 10 mm); medium diameter, medium length (MDML; 2.7 × 8 mm); medium diameter, long length (MDLL; 2.7 × 14 mm); and long diameter, medium length (LDML; 4.0 × 10 mm). The suture anchors were made using implant-quality 316LVM stainless steel with self-tapping cortical screws. Each group consisted of eight specimens. The bone anchors in the LDML group were manufactured by IMEX Veterinary, Inc. (Longview, TX, USA), and those in the remaining groups were manufactured by N2 (UK) Ltd. (Portsmouth, UK)

Table 1 Classification of groups based on the size of the bone anchors

Group	Anchor screw size: diameter × length
SDSL	2.0 × 6 mm (Suture anchors, N2®, Portsmouth, UK)
SDML	2.0 × 10 mm (Suture anchors, N2®, Portsmouth, UK)
MDML	2.7 × 8 mm (Suture anchors, N2®, Portsmouth, UK)
MDLL	2.7 × 14 mm (Suture anchors, N2®, Portsmouth, UK)
LDML	4.0 × 10 mm (Suture anchors, IMEX®, Longview, TX, USA)

Abbreviations: SDSL, short diameter, short length; SDML, short diameter, medium length; MDML, medium diameter, medium length; MDLL, medium diameter, long length; LDML, long diameter, medium length

Prosthetic fixation: The isometric F2 site (Hulse *et al*. 2010) was located at the caudolateral edge of the femoral condyle at the level of the distal pole of the lateral fabella (Figure 1). After setting the F2 site as the femoral anchorage site, the insertion site was marked with a slight dent using an 18-G needle to prevent the drill bit from slipping. Depending on the diameter of the bone anchor, an appropriately sized drill bit was used to drill a guide hole, with the depth corresponding to the length of the bone anchor. The bone anchor was implanted into the femoral anchoring point using a driver and the bone anchor eyelet was threaded with polyethylene suture material (75lb LigaFiba; Covetrus, Columbus, OH, USA). A hole was made in the tibia using a 1.5-mm drill bit at the bony protuberance caudal to the sulcus of the long digital extensor tendon in a straight line (Figure 2; Hulse *et al*. 2010). One of the two suture ends hanging from the eyelet of the bone anchor was passed through the tibial hole in the mediolateral direction. The free ends of the suture were knotted using five square knots.

Radiographic measurement of ratios: Radiography was used to confirm the proper placement of the inserted bone anchor and calculate ratios using the femoral condyle width (FCW) and femoral condyle length (FCL). The FCW, which was defined as a line perpendicular to the long axis of the femur at the level of the base of the intercondylar fossa, extended from the medial femoral condyle to the lateral femoral condyle (Healey *et al*. 2019) and was measured on craniocaudal radiographs (Figure 3A). The FCL was measured as a line perpendicular to the long axis of the femur extending from the most distal aspect of the lateral femoral condyle to the most proximal extent of the femoral trochlear ridge (Mostafa *et al*. 2008) on mediolateral radiographs (Figure 3B).

The depth of the anchor screw thread within the femoral condyle (Figure 3C) was expressed as a percentage of the FCW (anchor screw depth/FCW) for comparison according to the insertion depth. The external diameter of the anchor screw thread within the femoral condyle (Figure 3D) was expressed as a percentage of the FCL (anchor screw diameter/FCL) for comparison according to the diameter. Figures 3C–3D show the method of measuring these ratios using radiography.

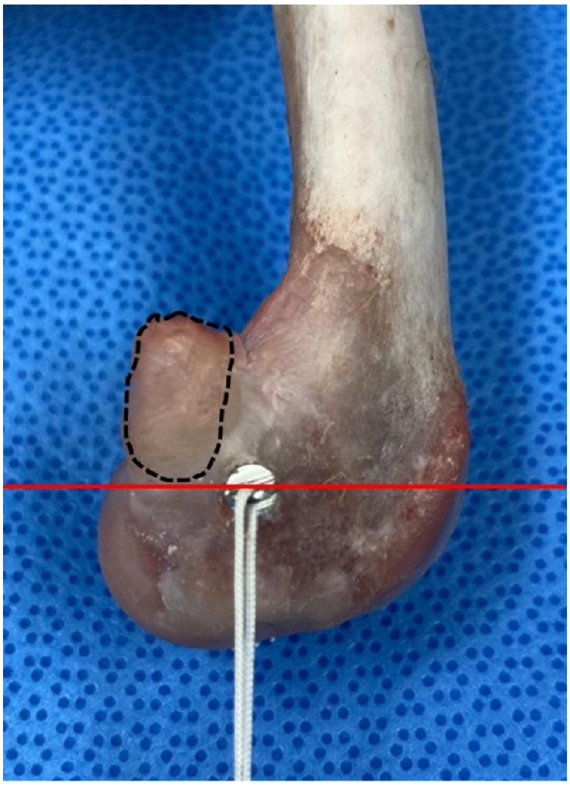

Figure 1 Photograph illustrating the isometric F2 site located along the caudolateral edge of the femoral condyle at the level of the distal pole of the lateral fabella. Dotted line – lateral fabella of the right femur. Red line – the level of the implanted site of the bone anchor

Figure 2 Photograph illustrating a hole in the tibia at the bony protuberance caudal to the sulcus of the long digital extensor tendon
Black arrow – sulcus of the long digital extensor tendon of the right tibia
Red circle – site of the tibia hole

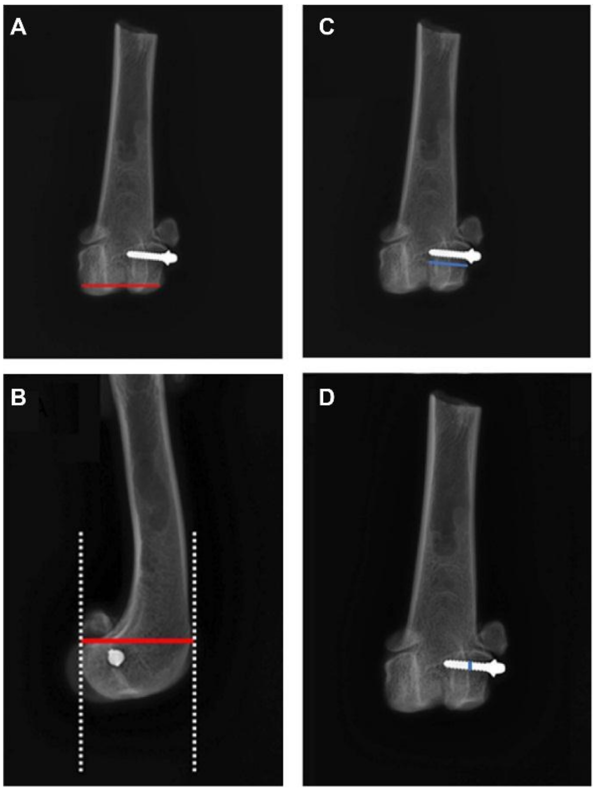

Figure 3 Illustration of radiographic measurements of the FCW, FCL, anchor screw depth, and anchor screw diameter. (A) The FCW (red solid line) is defined as a line perpendicular to the long axis of the femur at the level of the base of the intercondylar fossa, extending from the medial to the lateral femoral condyles on the craniocaudal view. (B) The FCL (red solid line) was measured by drawing a line perpendicular to the long axis of the femur, extending from the most distal aspect of the lateral femoral condyle to the most proximal extent of the femoral trochlear ridge in the mediolateral view. (C) Anchor screw depth (blue solid line) within the femoral condyle was measured on a craniocaudal radiograph. (D) Anchor screw diameter (blue solid line) within the femoral condyle was measured on a mediolateral radiograph. Abbreviations: FCW: femoral condyle width; FCL: femoral condyle length

Measurement of biomechanical properties: For the comparison of biomechanical properties according to insertion depth, bone anchors with short diameters (SDSL and SDML groups) were used. Similarly, bone anchors of medium diameter (MDML and MDLL groups) were also compared. For the comparison of biomechanical properties according to the diameter, medium-length bone anchors (SDML and LDML groups) were used.

A micromechanical tester (Instron 3369; Instron Corporation, Norwood, MA, USA) with a 2-kN load cell attached to the crosshead was used to determine the mechanical properties of bone anchors of various sizes. Using a customized jig, the femur and tibia were fixed at 135°, which was the average angle of the stifle joint in the normal standing position. The femur and suture material were positioned at 150°, which is the angle at which the shear force is greatest (Rey *et al*. 2014). Based on the results of a preliminary study, the working length of the suture was determined to be 3 cm (Roca *et al*. 2020) (Figure 4).

The crosshead was reset to zero before each test and the construct was preloaded with a tension of 5 N and pulled to failure by the machine at a rate of 1 mm/s. The yield load, Young's modulus and failure load of the constructs were evaluated using a stress–strain curve to quantify their strengths and stiffnesses. The yield load refers to the stress at the endpoint of the elastic region that can be developed in a material without causing plastic deformation. This is the load just before the material shows a specific permanent deformation and at the point where the slope of the stress–strain curve changes rapidly. The failure load is defined as the maximum stress that a material can withstand before breaking. Young's modulus signifies the elastic properties of the constructs and is calculated as the slope of the curve in the elastic region. The yield load and failure load are proportional to the strength of the constructs, whereas Young's modulus is proportional to the stiffness of the constructs.

In this experiment, failure was defined as the pull-out of the bone anchor, anchor eyelet fracture, distal femur fracture, tibial fracture and suture material tear and the cause of failure in each construct was recorded.

Data analysis: Statistical analyses were performed using SPSS software (SPSS version 18; IBM Corp., Armonk, NY, USA). Data comparisons between the SDSL and SDML groups were performed using a paired t-test because these groups used contralateral pelvic limbs of the same dog cadaver. For the same reason, this method was applied for comparison of the MDML and MDLL groups. In contrast, data comparison between the SDML and LDML groups was performed using an independent samples t-test because pelvic limbs from different dog cadavers were assessed in these groups. A chi-square test was performed to analyze the frequency differences between the groups for the cause of failure. Significance was defined by P-values < 0.05.

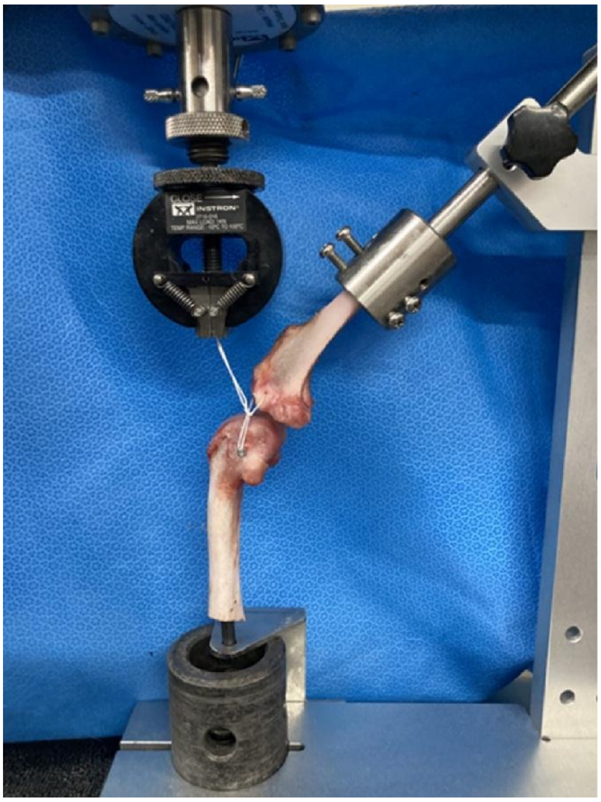

Figure 4 Photograph illustrating the fixation of the constructs to the micromechanical tester using a customized jig

Results

The mean body weight of the dog cadaver was 3.52 ± 0.79 kg (range, 2.2–5 kg). No significant differences were observed in the weights of the dog cadavers. Ten dogs were female and 10 were male. The mean age of the dogs was 8.2 ± 2.4 years (range, 4–15 years). The breeds of the dogs were as follows: 6 Miniature Poodles, 5 Maltese dogs, 2 Pomeranians, 2 Shih-tzus, 1 Schnauzer and 4 mixed-breed dogs.

Anchor screw depth/FCW: Table 2 shows the range (mean ± standard deviation) of anchor screw depth/FCW in the SDSL, SDML, MDML, and MDLL groups. The mean FCW was 16.34 ± 1.82 mm (range, 12.98–19.14 mm).

Anchor screw diameter/FCL: Table 3 shows the range (mean ± standard deviation) of anchor screw diameter/FCL in the SDML and LDML groups. The mean FCL was 14.46 ± 1.08 mm (range, 12.89–16.22 mm).

Biomechanical properties of the bone anchor with a diameter of 2 mm: Table 4 lists the yield load, Young's modulus and failure load of the bone anchor with a diameter of 2 mm (SDSL and SDML groups). The yield load, Young's modulus, and failure load were significantly (all $P < 0.05$) higher in the SDML group than in the SDSL group.

Table 2 Range (mean ± standard deviation) of anchor screw depth/femoral condyle width

Group	Anchor screw depth/femoral condyle width (%)
SDSL	34.05 ± 5.0 (31.34–40.22)
SDML	69.77 ± 2.98 (64.35–73.15)
MDML	49.73 ± 3.71 (47.36–54.47)
MDLL	82.76 ± 5.56 (75.84–91.62)

Abbreviations: SDSL, short diameter, short length; SDML, short diameter, medium length; MDML, medium diameter, medium length; MDLL, medium diameter, long length

Table 3 Range (mean ± standard deviation) of anchor screw diameter/femoral condyle length

Group	Anchor screw diameter/femoral condyle length (%)
SDML	12–15 (13.96 ± 1.03)
LDML	24–30 (27.67 ± 2.19)

Abbreviations: SDML, short diameter, medium length; LDML, long diameter, medium length

Table 4 Mean ± standard deviation of yield load, Young's modulus and failure load of the bone anchor with a diameter of 2 mm

	SDSL (2.0 × 6 mm)	SDML (2.0 × 10 mm)	*P*-value
Yield load (N)	30.10 ± 5.87	59.55 ± 10.49	< 0.001*
Young's modulus (N/mm)	2.75 ± 1.32	5.07 ± 1.61	0.035*
Failure load (N)	61.47 ± 12.89	87.03 ± 17.30	0.013*

* Significant difference ($P < 0.05$)
Abbreviations: SDSL, short diameter, short length; SDML, short diameter, medium length

Biomechanical properties of the bone anchor with a diameter of 2.7 mm: Table 5 shows the yield load, Young's modulus, and failure load of the bone anchor with a diameter of 2.7 mm (MDML and MDLL groups). The yield load, Young's modulus, and failure load were significantly (all $P < 0.05$) higher in the MDLL group than in the MDML group.

Biomechanical properties of the bone anchor with a length of 10 mm: Table 6 shows the yield load, Young's modulus and failure load of the bone anchor with a length of 10 mm (SDML and LDML groups). There were no differences between the SDML and LDML groups in any of the parameters tested (all $P > 0.05$).

Cause of failure: Table 7 lists the frequencies of the causes of failure in each group. The cause of failure in all constructs was pull-out of the bone anchor, except for five constructs in the LDML group in which fracture of the distal femur occurred ($P = 0.027$).

Table 5 Mean ± standard deviation of the yield load, Young's modulus and failure load of the bone anchor with a diameter of 2.7 mm

	MDML (2.7 × 8 mm)	MDLL (2.7 × 14 mm)	*P*-value
Yield load (N)	49.88 ± 4.45	54.75 ± 3.25	0.044*
Young's modulus (N/mm)	4.79 ± 0.97	6.98 ± 2.21	0.048*
Failure load (N)	79.11 ± 7.16	98.09 ± 10.23	0.006*

* Significant difference ($P < 0.05$)
Abbreviations: MDML, medium diameter, medium length; MDLL, medium diameter, long length

Table 6 Mean ± standard deviation of yield load, Young's modulus and failure load of the bone anchor with a length of 10 mm

	SDML (2.0 × 10 mm)	LDML (4.0 × 10 mm)	*P*-value
Yield load (N)	59.55 ± 10.49	57.04 ± 16.50	0.722
Young's modulus (N/mm)	5.07 ± 1.61	5.55 ± 2.27	0.630
Failure load (N)	87.03 ± 17.30	82.44 ± 16.73	0.598

* Significant difference ($P < 0.05$)
Abbreviations: SDML, short diameter, medium length; LDML, long diameter, medium length

Table 7 Frequencies of causes of failure in each group

Group	Pull-out of bone anchor	Distal femur fracture	Anchor eyelet fracture	Tibial fracture	Suture material tear
SDSL	100%	0	0	0	0
SDML	100%	0	0	0	0
MDML	100%	0	0	0	0
MDLL	100%	0	0	0	0
LDML	37.5%	62.5%	0	0	0
χ^2 (P)	4.898 (0.027)				

Abbreviations: SDSL, short diameter, short length; SDML, short diameter, medium length; MDML, medium diameter, medium length; MDLL, medium diameter, long length; LDML, long diameter, medium length

Discussion

Although there are disadvantages in the clinical application of extracapsular stabilization, such as limited use in toy breed dogs, pull-out of the bone anchor, suture elongation and abrasion causing implant instability or failure (Dominic *et al*. 2020), it also has advantages, such as being less expensive, less technically demanding and having a better safety profile than osteotomy procedures (Raske and Hulse, 2013). Thus, extracapsular stabilization can be considered as a treatment option for CrCL deficiency in toy breed dogs.

However, to the best of our knowledge, there are no clear criteria for the diameter and insertion depth when selecting the size of the bone anchor in toy breed dogs. Therefore, the bone anchor size has thus far been determined subjectively by the individual surgeon according to the weight and activity level of the patient (Raske and Hulse, 2013). Without clear criteria, selection of the size of the bone anchor may be an important factor influencing the rate of postoperative complications (Fritsch *et al*. 2010). We thought that a ratio would be an appropriate criterion since the width and length of the femur are not constant in toy breed dogs.

Identification of the isometric point of the tissue can allow for more natural tight motions while reducing suture tension fluctuations caused by flexion and extension. Therefore, implants at near-isometric points reduce the risk of implant failure, possibly leading to better clinical outcomes (Raske *et al*. 2013). To the best of our knowledge, no research has been able to obtain true isometry; therefore, we consider "quasi-isometric points" to be a more accurate term. We selected the F2-located caudolateral edge of the femoral condyle at the level of the distal pole of the lateral fabella and the T3-located bony protuberance caudal to the sulcus of the long digital extensor tendon, which have previously been shown in combination to be the most isometric sites of the femur and tibia (Hulse *et al*. 2010).

The structure, function and motion of the mechanical features of biological systems describe their biomechanical properties. Among several factors affecting biomechanical properties, we focused on analyzing strength and stiffness. Strength refers to the amount of stress that can be applied to a material until it breaks, while stiffness refers to the degree to which the material resists external deformation. The stress increases and reaches a certain point called the yield point; beyond this, the strain increases faster because the stress increases more slowly and the material loses its elasticity. The point that demarcates the elastic region from the plastic region is the yield point. Young's modulus is defined as the slope (gradient) of the curve in the elastic region; when the stress is gradually increased beyond the elastic limit, the material undergoes plastic deformation. At stresses beyond the elastic limit, the material exhibits plastic behavior, i.e., it deforms irreversibly. The failure load is the maximum stress that a material can withstand prior to breaking. In other words, the yield load and failure load are proportional to the strength of the constructs, whereas Young's modulus is proportional to the stiffness of the constructs.

In the present study, we found that the constructs with an anchor screw depth/FCW of 50%–75% (SDML group) or 75%–100% (MDLL group) had more strength and stiffness than those with an anchor depth/FCW of less than 50% (SDSL group) or approximately 50% (MDML group), respectively. The physiologic in vivo load applied to the CrCL of canines has not been accurately defined (Rose *et al*. 2012). Because the bone anchor must resist cranial translation in the CrCL-deficient stifle, it is assumed that the extracapsular bone anchor would be subject to loads similar to the native CrCL. Roca *et al*. (2020), extrapolated the load on the native CrCL of a 4.5-kg dog to be 10–50 N throughout their experiment. Implantation of the bone anchors used in the SDML and MDLL groups would be suitable for toy breed dogs because the failure loads of constructs in these groups (87.03 ± 17.30 N and 98.09 ± 10.23 N, respectively) were higher than the aforementioned threshold of 50 N.

In the experiment comparing diameters, a difference of twice the diameter of the bone anchor was observed between the SDML and LDML groups but there were no statistically significant differences in terms of strength and stiffness. Based on these results, we focused on the causes of failure in the LDML group. The cause of failure of all constructs in the SDSL, SDML, MDML, and MDLL groups was pull-out of the bone anchor, suggesting that the bonding force between the bone anchor and the bone tissue of the femur might dictate the performance of the bone anchor. In contrast, 62.5% of the failures in the LDML group were caused by fracture of the distal femur. Factors to be considered when implanting the bone anchor in toy breed dogs include not only the diameter and insertion depth of the bone anchor but also the available bone stock of the distal femoral condyles (Comerford *et al*. 2013). Thus, when the bone anchor was pulled by the micromechanical tester, the counteraction to the force received by the bone anchor was also applied to the distal femur. In addition, the volume removed from the distal femur was approximately four times larger in the construct with a 4.0 × 10-mm bone anchor than in the construct with a 2.0 × 10-mm bone anchor. As a result, there was not enough femoral bone volume to endure the force applied to the distal femur in toy breed dogs, resulting in fracture of the distal femur in the constructs with a 4.0 × 10-mm bone anchor. Therefore, additional future experiments of bone anchor success associated with the diameter and insertion depth are recommended in larger-breed dogs with sufficient femoral bone volumes.

According to a piece of research proposed by Hsu *et al*. (2007), bone quality, bone density, frictional force, contact area and thread type determine the bonding force between the anchor and bone. Therefore, when implanting bone anchors in toy breed dogs, these factors should be considered to determine whether the bone anchor can be bonded well to the cortex of the femur.

The most important limitation of this study was the non-physiological ex vivo study design. Although the experiment was conducted at an angle known to receive the greatest shear force, this did not represent the condition in vivo. Manufacturer-related differences in the bone anchors used in the LDML group (IMEX) and the other groups (N2) are also a limitation of this study. As an alternative, we aimed to use a self-made customized bone anchor or not to insert a bone anchor to the end. However, even if the there are differences among anchors from different manufacturers, the use of commercialized bone anchors provided the most realistic conditions when applied clinically to toy breed dogs.

In conclusion, when implanting bone anchors in toy breed dogs, the insertion depth of the bone anchor should be more than 50% of the FCW, regardless of diameter. Moreover, fractures of the distal femur could be a postoperative complication if the ratio of the diameter of the bone anchor to the FCL exceeds 24%.

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