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Original Research

Evaluation of Meniscal Load and Load Distribution in the Sound Canine Stifle at Different Angles of Flexion

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Abstract

Objectives The aim of the study was to investigate the contact mechanics and kinematic changes in the stifle in different standing angles.

Study Design We performed a biomechanical *ex vivo* study using pairs of canine cadaver hindlimbs. Motion sensors were fixed to the tibia and the femur for kinematic data acquisition. Pressure mapping sensors were placed between the femur and both menisci. Thirty percent bodyweight was applied to the limbs with the stifle in 125, 135, or 145 degrees of extension.

Results Stifle flexion angle influences femoromeniscal contact mechanics significantly. The load on both menisci was significantly higher for 125 and 135 degrees in comparison to 145 degrees. Additionally, the center of force was located significantly more caudal when comparing 125 to 145 degrees in the medial meniscus as well as in both menisci combined.

Conclusion The angle of knee flexion significantly impacts the contact mechanics between the femur and the meniscus. As the knee flexes, the load on both menisci increases.

Keywords
- stifle joint
- knee biomechanics
- menisci
- standing angle

Introduction

The articular surfaces of the stifle joint lack congruence; therefore, the medial and lateral menisci are necessary to ensure congruency. These structures play essential roles as both mechanical shock absorbers and crucial load-bearing elements in the canine stifle.[1] Damage to the menisci rapidly results in osteoarthritis.[2] Even though isolated meniscal damage is rarely seen in dogs,[3,4] secondary injury is common after cranial cruciate rupture,[5,6] one of the most common orthopaedic diseases in canines. Available evidence suggests that following cranial cruciate rupture, Labradors load their hindlimbs at higher stifle flexion than nonpredisposed individuals do.[7] Fischer and colleagues demonstrated that different dog breeds load their limbs at different angles of joint flexion during ambulation.[8] Several recent studies have addressed femorotibial and femoromeniscal biomechanics, mainly focusing on the comparison between diseased and normal conditions.[9–17] Two studies comparing

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meniscofemoral contact mechanics in 90- and 135-degree flexion showed kinetic and kinematic differences. But still 90 degrees is an extreme flexion angel that is not seen during walking, trotting, or running.[18,19]

Walker and Erkman evaluated the influence of stifle flexion on load distribution in humans.[20] Stifle angulation affects meniscal kinematics,[15] but its effects on contact mechanics in the joint have remained unclear. The aim of this study was to characterize the effect of stifle angle on meniscal load and distribution in normal canine stifle joints. Knowing that both menisci are displaced caudally during flexion,[15] we hypothesized that the center of force would move caudally in the process of flexion. The second hypothesis was that the load on the menisci increases during flexion.

Materials and Methods

Specimen Preparation

Fourteen hindlimbs were disarticulated at the coxofemoral level from seven adult Retrievers (25–40 kg body weight) that had died or had to be euthanized for unrelated reasons. To exclude stifle and tarsal joint pathology, orthogonal radiographs and orthopaedic examination of the cadavers were performed. All muscles proximal of the tarsal joint were dissected while preserving the stifle and tarsal joints. These limbs were later used in a follow-up study investigating the effects of tibial plateau leveling osteotomy (TPLO) and the modified Maquet procedure (MMP) on stifle contact mechanics. In the TPLO group, custom-made aluminum TPLO hinge plates were placed on seven selected limbs (three left and four right limbs). A radial osteotomy—as required for TPLO—was performed, using the plate as a saw guide.[21] The limbs of the MMP group were prepared to fit a custom-made aluminum MMP hinge plate. In this case, an osteotomy as required for MMP was conducted[22] (►**Fig. 1**). At these stages of preparation, the specimens could be used for the tests reported here before alterations on the tibial plateau angle or cranialization of the tibial tuberosity were performed.

To fit adjustable mounting brackets that enabled the adjustment of hip joint angles and femoral torsion, the proximal femur was embedded in polymethylmethacrylate (RENCAST FC 53, Huntsman Advanced Materials, Germany). Muscle forces were simulated using steel cables and turnbuckles. To simulate quadriceps muscle pull, a 1.5-mm braided stainless steel cable was passed through a medial-to-lateral 2-mm tunnel drilled through the widest part of the patella. To simulate the gastrocnemius muscle, a 2.0-mm cable was passed through a 2.5-mm transversal drill hole in the tip of the calcaneus and secured as a loop.

Two 3.5-mm cortical bone screws inserted into the femoral articular surface of the femorofabellar joint were used as the second attachment point for the gastrocnemius cable. The limbs were covered with physiologic saline-soaked towels and stored in vacuum bags at –20°C until testing. Prior to testing, the limbs were thawed at room temperature. Stifle kinematics were measured using the CMS20BI ultrasound system (Zebris Medical GmbH, Isny, Germany). These ultrasound motion sensors were mounted to Schanz screws with a 3.2-mm root, and inserted into the distal femoral and the proximal tibial diaphysis. A cranial and caudal arthrotomy was performed to install the pressure mapping sensors (detailed below) on top of both menisci. They were sutured and glued to the joint capsule and collateral ligaments.

Stifle kinematics (femoromeniscal loads) were continuously recorded with an I-Scan system (Tekscan Inc., South Boston, Massachusetts, United States). The sensing region of the K-Scan 4041 sensor is 31.5 mm × 12.7 mm including 90 sensels, with a thickness of 0.2 mm. The contact force ratio (CFR) was defined as the contact force divided by the applied force load acting on both menisci $\left(\frac{contact\ force}{30\%\ body\ weight}\right)$; the latter was set to 30% of the animal's body weight and applied with a material testing machine (Model Z010, Zwick & Roell GmbH & Co. KG, Ulm, Germany). Additionally, contact area, peak pressure (highest pressure measured), mean contact pressure, and peak pressure location were recorded. Pressure location was defined as the distance from the caudal meniscal border to the peak pressure-recording sensel. For each

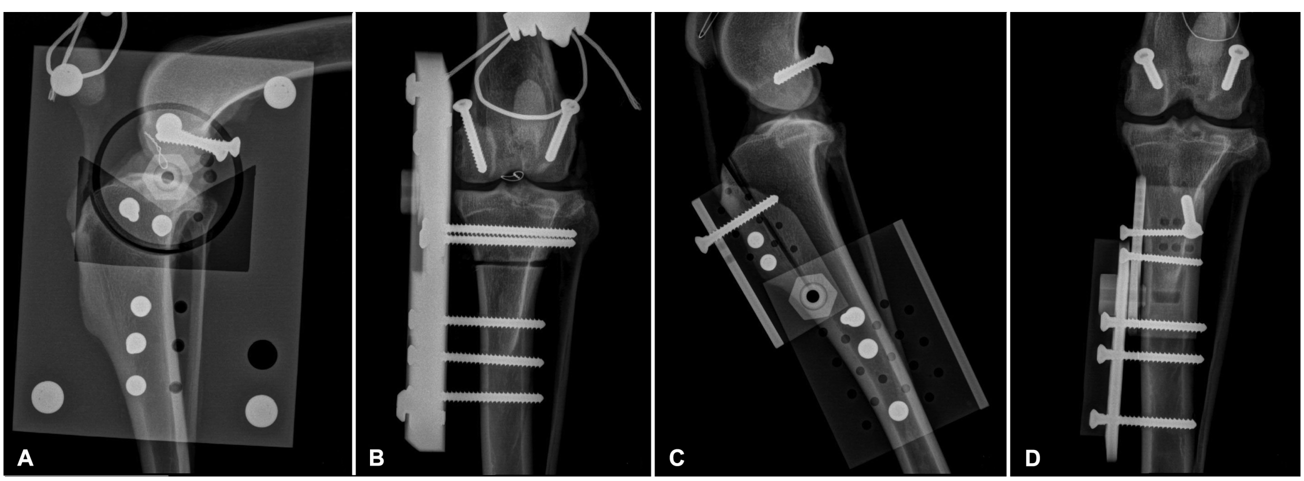

Fig. 1 Radiographs of a right stifle with TPLO-hinge plate (**A**: medio/lateral; **B**: cranio/caudal) or MMP-hinge plate (**C**: medio/lateral; **D**: cranio/caudal) in place after osteotomy in unaltered, physiologic position.

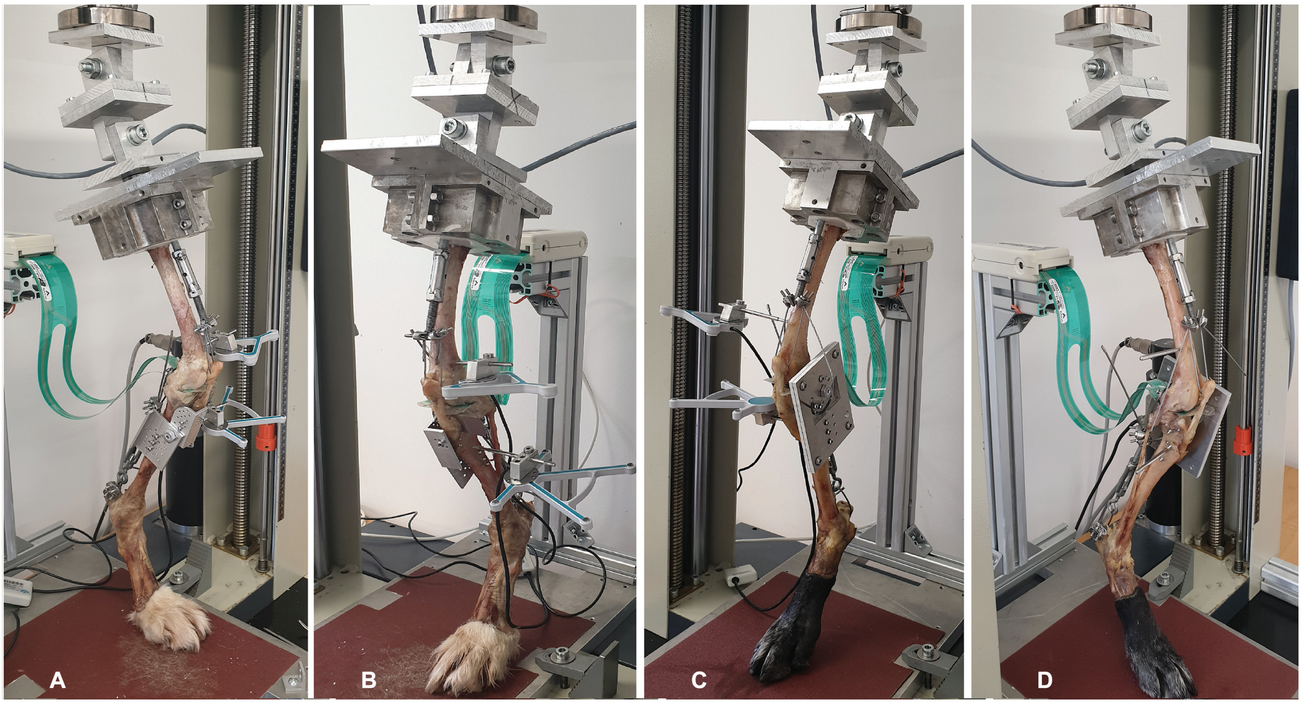

Fig. 2 Limbs ready for testing, with motion and pressure sensors in place (**A**: medial view with MMP-hinge plate in place, **B**: lateral view; **C**: medial view with TPLO-hinge plate in place, **D**: lateral view).

stifle, a new sensor was used and calibrated according to the producer's guidelines.

Testing Protocol

The limbs were mounted in the testing machine with the sensors in place. A special mounting bracket that allowed adjustment of hip joint angles and femoral torsion was placed between the load cell and the femur. The turnbuckles were adjusted to maintain the stifle joint angle at 125, 135, or 145 degrees, and the tarsal joint at 140 degrees under load. While flexion/extension and hip adduction/abduction were controlled, torsion of the femur was still possible. Testing was started with the most homogeneous meniscal pressure distribution possible and a preload of 10 N. Stifle angulation was changed from 125 to 135 degrees and then to 145 degrees for measurements. A load of 30% of the body weight was applied at all joint flexion settings (►**Fig. 2**).

Statistical Analysis

Homogeneity of variances was checked with Levine's test. Univariate analysis of variance (Welch's ANOVA) was performed with the SPSS statistics 26.0 software (IBM, Armonk, NY, United States) to analyze differences of contact area, contact pressure, peak pressure, center of force, and contact force for the medial and lateral menisci, and for both menisci combined at the three standing angles. To evaluate the cranial, medial, and proximal translation as well as flexion, adduction, and internal rotation of the tibia, Welch's ANOVA was also applied. Games–Howel tests were performed for paired comparison where Welch's ANOVA had indicated significant differences. Significance was assumed for $p < 0.05$.

Results

Median bodyweight of the dogs was 31.5 kg ($CI_{95\%}$: 27.6–35.3). Median TPA (±standard deviation [SD]) was 21.3 degrees ($CI_{95\%}$: 20.2–22.5). Measured stifle angles under load were 125.0 degrees ($CI_{95\%}$: 124.9–125.2) for the 125-degree setting, 134.9 degrees ($CI_{95\%}$: 134.6–135.2) for the 135-degree setting, and 145.1 degrees ($CI_{95\%}$ 144.8–145.3) for the 145-degree setting. Stifle angles were significantly different ($P^{125–135°} < 0.001$, $P^{135–145°} < 0.001$, $P^{125–145°} < 0.001$). Otherwise, no significant kinematic (e.g., cranial motion, caudal motion, or endo-rotation) differences were detected.

In the stifle, the CFR on both menisci was significantly higher for 125 and 135 degrees in comparison to 145 degrees. Additionally, the center of force was located significantly more caudal when comparing 125 to 145 degrees in the medial meniscus as well as in both menisci combined. Furthermore, the lateral meniscus contact pressure was significantly higher at 125 degrees (1.2 MPa) than at 145 degrees (1.0 MPa). This reflected the greater force in relation to the load that occurred in the lateral meniscus at 125 and 135 degrees compared with 145 degrees (►**Fig. 3** and ►**Table 1**).

Discussion

Our study aimed at determining kinematic changes in the canine stifle (femoromeniscal contact) as well as kinetic changes on both menisci brought about by changing the angle of stifle flexion. Our experimental setup allowed for continuous monitoring of stifle angles during load application. We used Retriever breeds only to reduce biological variance between specimens. We were able to demonstrate

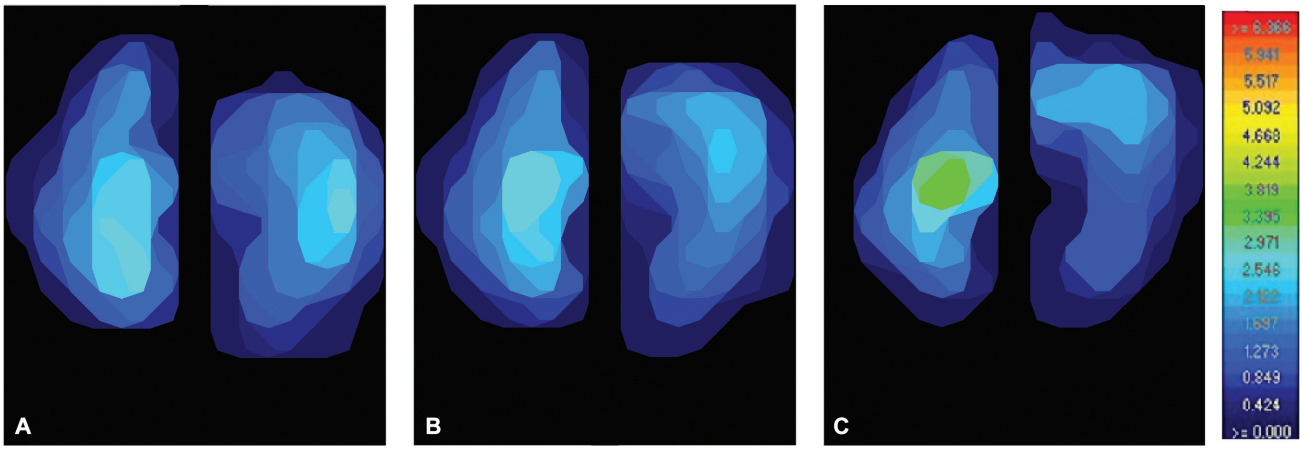

Fig. 3 Pressure distribution in the different testing positions (from left to right: 125°(**A**), 135°(**B**) and 145°(**C**) stifle angle). The medial meniscus is on the left, lateral on the right. The top of the pictures represents cranial. (Exemplary Data, Scale in MPa).

Table 1 Kinetic variables (mean (95% confidence interval)) of the stifle at three angles of flexion

Variable	125 degrees	135 degrees	145 degrees
CFR Both menisci	5.0 ($CI_{95\%} = 4.9–5.2$) $p^{125°–145°} > 0.001$	4.9 ($CI_{95\%} = 4.8–5.0$) $p^{135°–145°} > 0.001$	4.5 ($CI_{95\%} = 4.3–4.76$)
CFR Medial menisci	2.6 ($CI_{95\%} = 2.4–2.8$)	2.6 ($CI_{95\%} = 2.5–2.8$)	2.4 ($CI_{95\%} = 2.3–2.6$)
CFR Lateral meniscus	2.4 ($CI_{95\%} = 2.3–2.6$) $p^{125°–145°} = 0.001$	2.3 ($CI_{95\%} = 2.2–2.4$) $p^{135°–145°} = 0.022$	2.1 ($CI_{95\%} = 1.9–2.2$)
Center of force[a] Both menisci	4.1 ($CI_{95\%} = 3.9–4.4$) $p^{125°–145°} = 0.001$	4.5 ($CI_{95\%} = 4.2–4.9$)	4.9 ($CI_{95\%} = 4.6–5.3$)
Center of force[a] Medial meniscus	3.5 ($CI_{95\%} = 3.1–4.0$) $p^{125°–145°} = 0.003$	4.0 ($CI_{95\%} = 3.6–4.4$)	4.6 ($CI_{95\%} = 4.1–5.0$)
Center of force[a] Lateral meniscus	4.8 ($CI_{95\%} = 4.5–5.1$)	5.0 ($CI_{95\%} = 4.6–5.4$)	5.3 ($CI_{95\%} = 4.8–5.8$)
Mean pressure in MPa Both menisci	1.3 ($CI_{95\%} = 1.2–1.4$)	1.2 ($CI_{95\%} = 1.1–1.3$)	1.2 ($CI_{95\%} = 1.1–1.2$)
Mean pressure in MPa Medial meniscus	1.3 ($CI_{95\%} = 1.2–1.5$)	1.3 ($CI_{95\%} = 1.2–1.4$)	1.3 ($CI_{95\%} = 1.2–1.40$)
Mean pressure in MPa Lateral meniscus	1.2 ($CI_{95\%} = 1.1–1.3$) $p^{125°–145°} = 0.049$	1.1 ($CI_{95\%} = 1.0–1.2$)	1.0 ($CI_{95\%} = 1.0–1.0$)

Abbreviations: CFR, contact force ratio; CI, confidence interval.
Note: Significant differences according to Welch's ANOVA (analysis of variance) are indicated.
[a]The center of force was defined as the distance from the caudal border of the meniscus to the sensor element that recorded the highest load.

significant changes in meniscal load, load distribution, and contact pressure with changing standing angles. A limitation of our study lies in its *in vitro* nature, which represents a simplification of the *in vivo* function of the stifle. Following the measurements reported here, osteotomy procedures were performed in all specimens for unrelated purposes, but no relevant anatomical changes had been introduced at the time of our testing. We tested a range of motion of 20 degrees (i.e., 125- to 145-degree stifle flexion), which covers the majority of the stance phase in dogs.[8,10,23] A larger range of motion (40–57 degrees) has been reported for dogs in walk and trot.[10] Changes occurring at high flexion angles of 90 degrees were investigated before.[18,19] This angle of flexion will most likely only occur during standing up or stair accent. This is very interesting, but it was not the focus of our study. Additionally, no comparisons between different states of cranial cruciate ligament integrity were made. Unfortunately, it was *impractical* to test kinetics and kinematics during the swing phase in our experimental setup.

The stabilizing effect of the joint capsule was compromised due to its transection for the intra-articular placement of the I-Scan sensor.[24] Generally, joint mechanics may be affected by sensors inserted into the joint space.[25,26] However, as these alterations were the same for all tests, the comparison between different settings should still provide meaningful information.

We placed new sensors in every specimen to reduce the risk of sensor damage or kinking. The types of sensors we used had previously been applied successfully in other studies. Other investigators were able to demonstrate changes in femorotibial contact mechanics after meniscal surgery or damage, and characterized the influence of TPLO and tibial tuberosity advancement (TTA) on stifle kinetics and kinematics.[9,12,13,18,19,27] Nevertheless, this type of sensor has a reduced accuracy of 1 to 4% for peak pressure and 3 to 9% for average pressure and contact area on spherical surfaces.[28]

In humans, the load on the menisci rises from 50 to 70% of body weight when standing fully upright to as much as 85 to 90% during knee flexion.[29–31] In our study, the contact force significantly decreased in setups with 125- and 135- in comparison to the 145-degree setup. This can be explained by the increase in quadriceps force,[30] which causes an increase of total joint force (sum of muscle and ground reaction force).[32]

We observed a caudal shift of the center of force when flexing the stifle joint, especially on the lateral meniscus; an analogous effect has been described in humans.[29,33] In dogs, this may be due to the anatomical shape of the femoral condyles, which causes a rollback effect when the stifle joint flexes.[33–35] On the other hand, others demonstrated that canine menisci are highly mobile. Their location on the tibia changes during extension and flexion,[15] which probably contributes to shifting the position of the center of force. The mean pressure on the lateral but not on the medial meniscus decreased in extension, possibly a result of the different shapes of the two menisci.

Our findings should be taken into consideration for further studies and the development and evaluation of existing and new treatment options for pathologies of the canine stifle. For example, performing TPLO rotates the tibial plateau and therefore changes the position of the menisci in relation to the femur to different degrees, depending on joint flexion angle. The influence of this effect should be the subject of future investigations.

Conclusion

Both our hypotheses can be accepted. Stifle flexion angle influences femoromeniscal contact mechanics significantly, but no changes in femorotibial kinematics were found. Further investigations of these kinetic changes might help understand the risks of cranial cruciate rupture and secondary meniscal injury.

Authors' Contribution

J.S., A.M-L., and P.A. contributed to conception of the study, study design, data analysis, and interpretation. M.G. and J.S. additionally contributed to data acquisition and data analysis. All the authors also drafted, revised, and approved the submitted manuscript.

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Conflict of Interest

None declared.

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