Ex-vivo Equine Medial Tibial Plateau Contact Pressure with an Intact Medial Femoral
Condyle, with a Medial Femoral Condylar Defect, and After Placement of a
Transcondylar Screw through the Condylar Defect

THESIS

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Abstract

The stifle (knee) is a common source of equine lameness, but its anatomical complexity has limited biomechanical research. Contact pressures have been described on the equine medial tibial plateau before and after creation of a meniscal tear, but that model simplified the stifle joint by removing the patella and the medial collateral ligament, and did not allow tibial rotation during testing. One of the goals of this study was to create a stifle model that included these elements to determine contact pressures on the medial tibial plateau. Additionally, we determined contact pressures after creation of an osteochondral defect in the medial femoral condyle (MFC). This defect mimics the location and size of a subchondral cystic lesion (SCL) of the MFC, a common pathology seen in the equine medial femoro-tibial joint. Lastly, a novel technique to treat SCL based on a transcondylar screw placed in lag fashion through the MFC was also tested as a third condition.

Six cadaveric equine stifle joints were axially loaded in a servohydraulic testing machine (Bionix®) to 1800N at 155°, 145° and 130°, with an intact MFC, with a 15 mm circular osteochondral defect, 20 mm deep, on the weight-bearing surface of the MFC and with a transcondylar screw through the defect. An electronic pressure sensor (Tekscan®) on the medial tibial plateau recorded contact area (cm²), force (N), peak

pressure (MPa) and contact map images from the sensors at the 3 angles and 3 conditions (9 tests). Peak pressure and calculated stress load (N/cm^2) was analyzed on the entire medial plateau and in 3 regions of interest (ROI): cranial, caudal and central.

Flexion of the stifle joint significantly increased force, contact area and stress load on the medial tibia under all conditions and the greatest increase was on the caudal sensor. The defect significantly reduced force (145° and 130°) and contact area at 145° , and stress load on the cranial and central ROI at 130° . The transcondylar screw returned force to intact values at 130° and reduced contact area in extension. Qualitative evaluation of the combined pressure maps revealed that under flexion, contact pressure in the intact stifle was transmitted to the caudal meniscus, but peak and high pressures remained on the central cartilage area at all angles. After creation of a MFC defect, no contact pressure was registered in the defect area, and high and peak pressures increased on the mid-meniscus which migrated caudally with flexion until at 130° . The transcondylar screw pressure maps were similar to the maps with the defect.

Stifle flexion increases force, contact area and stress load on the medial tibia and is most pronounced caudally. A MFC defect alters force, contact area and pressure maps during flexion on the medial tibia. A transcondylar screw may reverse some of those changes. In the MFC with a defect, the location of peak pressure is more caudal, and at 130° , tends to be on the caudal meniscus. This suggests a biomechanical explanation for meniscal injuries associated with SCL.

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Chapter 1: Introduction

In the horse, the stifle (knee) is the location of pain in approximately 15% of horses examined for lameness¹. As in other species, the medial femoro-tibial joint is the most commonly affected compartment, and meniscal tears, subchondral cystic lesions (SCL) and/ or osteoarthritis are frequently seen in equine stifles. In contrast, cranial cruciate ligament (CrCL) injury is the most common source of hind limb lameness in dogs while primary pathology of the meniscus is rare and most commonly associated with CrCL insufficiency or deficiency.^{2,3} It is thought that after rupture of the CrCL, the biomechanics of the knee change and compressive and rotational tibio-femoral forces damage the healthy meniscus.^{2,3} These forces tend to pinch the menisci when the joint is stressed and tear it. This mechanism is thought to be responsible for the tears in the caudal pole of the medial meniscus (MM) in dogs and humans and perhaps the cranial pole of the MM in horses.²⁻⁵

The equine meniscus is the primary cause of lameness in as much as 20% of the horses undergoing stifle arthroscopy.⁶ Its escalating relevance has focused recent studies, including the development of new arthroscopic approaches to improve meniscal visualization and the development of biomechanical explanations for meniscal tears in the cranial horn of the meniscus.^{4,7-9} Two studies by Fowlie et al attempted to explain this different area of predisposition in horses. The first study showed that the cranial horn of

the medial meniscus is the least mobile and is axially compressed during full extension, presumably predisposing it to trauma.⁴ A follow-up study identified two functional areas in the cranial menisco-tibial ligament of the medial meniscus separated by a cleft which histologically was highly vascularized and had a region with a low amount of connective tissue, suggesting that it may be predisposed to tear during hyperextension.⁹ However, meniscal tears of the caudal meniscus are also seen clinically in horses and a biomechanical explanation for them is lacking.

The etiology and treatment of SCLs are controversial.¹⁰⁻²⁵ Initial reports considered SCL as a type of osteochondrosis (OC), however, current hypotheses support a traumatic or inflammatory origin.¹⁰⁻¹⁶ Analysis of the tissue lining and fluid contained in equine SCLs revealed upregulation of inflammatory mediators which may play a role in bone resorption and impair SCL healing.^{14,15} Another hypothesis is that trauma to the cartilage and/or subchondral bone creates a communication between the subchondral bone and the synovial space, allowing synovial fluid to flow under pressure in the subchondral region producing necrosis of the adjacent bone.¹² This phenomenon of bone necrosis in the subchondral plate can also occur solely after trauma without the need for a breach in the joint cartilage.¹⁶

Treatments reported for SCL include rest, osteochondral grafting, cyst injection with corticosteroids, simple debridement, and implantation of osteogenic substances and chondrocytes after debridement.¹⁷⁻²⁴ Successful treatment is usually defined as the reduction or elimination of lameness sufficient for the horse to perform at an intended task; however radiographic or arthroscopic healing of the SCL is infrequently

accomplished.^{10,17-24} Lameness resolution has been reported to be approximately 70% regardless of the treatment utilized, and there is conflicting evidence as to whether patient age, breed, size of articular communication, or pre-existing evidence of osteoarthritis (OA) impact treatment results.²⁰⁻²⁴ A novel technique placing a transcondylar screw in lag fashion through the MFC has been recently developed by one of the members of the thesis committee (Dr. Santschi) and applied successfully in clinical cases.²⁵ Because most SCL do not demonstrate radiographic healing either spontaneously or after treatment, it is probable that the SCL environment is unfavorable for trabecular bone healing²⁰⁻²⁴. It is thought that the screw may stabilize the MFC and return strength, leading to bone deposition inside the SCL. Additionally, if the MFC re-gains part of its normal shape and surface over time, this will impact joint contact in the medial femoro-tibial joint and perhaps prevent meniscal complications associated to SCL. Recently, an association was reported between SCL and meniscal damage that suggested two possible mechanisms for meniscal damage.²⁶ Firstly, debridement of the cartilage and bone overlying the SCL may result in a change in MFC contour which can damage the underlying meniscus. Secondly, the rim of the debrided SCL may physically traumatize the medial meniscus.

Contact pressure on the medial tibial plateau using a pressure sensor (Tekscan®) under the meniscus has been investigated in many species and is a well-accepted method to determine joint contact.²⁷⁻³⁷ The first Tekscan® system (1987), called the T-scan, was used by dentists to visualize dynamic occlusal forces in their patients' mouths, enabling more accurate diagnosis and treatment of dental conditions.³⁸ Tekscan® technology shortly evolved and started to be used in humans and veterinary species to determine joint

contact.²⁷⁻³⁷ In early 2000, a study showed that Tekscan® sensors were more accurate than the technology used up to that date to determine joint contact, Fuji films.³⁹ Tekscan® sensors estimated contact area and pressure within 5% accuracy of a known value.

In the last 5 years, the group led by Dr. Pozzi has published multiple cadaveric biomechanical studies using this technology. They described canine femoro-tibial joint contact in normal and injured knees as well as after performing corrective surgical procedures to treat those pathologies.²⁸⁻³³ The first study evaluated changes in contact area and contact pressures in cadaveric stifles for different simulated types of medial meniscal tears (radial, vertical longitudinal, bucket handle, flap and complex).²⁸ They found no difference in contact area between intact or torn meniscus but there was significant increase in contact pressure (>45%) between intact and non-simple tears (bucket handle, flap and complex), suggesting that pronounced meniscal tissue disruption will affect joint biomechanics. A follow-up study compared the effects of suturing bucket handle tears versus performing a partial meniscectomy as a treatment.²⁹ Both groups, bucket handle tears and meniscectomy, had decreased contact area and increased pressures when compared to controls. It is possible that in this study bucket handle tears altered the contact area, in contrast to the previous study, due to a larger sample size (24 versus 12). In addition, every meniscal suture pattern (horizontal, vertical and cruciate) restored normal joint contact. Therefore, it was suggested that preservation of meniscal tissue rather than debridement is crucial to maintain joint biomechanics. The next step was to assess if more aggressive meniscectomies (75% or segmental meniscectomy)

could be done without excessive deleterious effects to the joint.³⁰ Results showed that medial and lateral joint contact decreases progressively and peak pressure increases with sequential meniscectomies. However, values were not significant enough until a segmental meniscectomy was performed. This study also revealed that both femoro-tibial compartments are affected when a medial meniscectomy is performed.

In dogs, the use of meniscal release to prevent MM damage secondary to CrCL deficiency is controversial and its biomechanical effect has also been investigated by Pozzi's group.³¹ Meniscal release has been proposed to allow the caudal pole of the medial meniscus to displace caudally and avoid impingement of the meniscal tissue between the femoral and tibial condyles following CrCL rupture and/or tibial plateau leveling osteotomy (TPLO).³ Caudal meniscal release produced a 50% decrease in contact pressure, 40% increase in stress and produced a caudal shift of pressures. These results elucidate that meniscal release alters meniscal function and thus stifles biomechanics. However, there is clinical evidence that meniscal release in CrCL deficient joints decreases the likelihood of developing a clinically significant meniscal injury in the future (5.2%).⁴⁰

Lastly, the biomechanical effects of two different surgical procedures, TPLO and tibial tuberosity advancement (TTA), to treat canine CrCL deficient knees have been tested.^{32,33} For the TPLO study, results determined that both, CrCL transection and TPLO procedure, resulted in less medial and total contact area, and contact pressures shifted caudally. CrCL transection increased joint contact pressures and TPLO returned those values to normal.³² Three-dimensional kinematics revealed cranial tibial translocation and

internal rotation after transecting the CrCL, which returned to similar levels after TPLO. The study concluded that TPLO eliminates cranio-caudal stifle instability during simulated weight bearing, but fails to concurrently restore femoro-tibial contact mechanics to normal. TPLO knees still demonstrated increased pressures caudally in the MM which it may be the reason why secondary caudal meniscal damage and progression of OA is still seen after the procedure (10.5%).⁴⁰ The TTA study revealed similar results to the previous one with one major exception.³³ TTA restored all values to normal after CrCL transection, and contact pressures shifted caudally only after CrCL transection. However, three dimensional analysis showed that 3/8 legs with TTA still had significant internal rotation as seen after CrCL transection, which may be why medial patellar luxation is seen after TTA.⁴¹ It is unknown yet if OA progresses after TTA as for TPLO but meniscal tears are seen more often than after TPLO, if meniscal release is not performed during the procedure. This may indicate that TTA is not as reliable as TPLO in preventing joint subluxation.⁴¹

To our knowledge, only one equine study has determined medial tibial plateau contact using Tekscan® technology, however, the patella and medial collateral ligament were removed and tibial rotation during axial loading was not allowed.²⁷ In that study, joint contact was determined in the naive stifle, after creation of a meniscal tear and after resection of the tear. As in human and small animal studies, it was concluded that a partial meniscectomy alters joint contact, potentially leading to damage of the articular cartilage and development of osteoarthritis.^{27,30,36} In fact, cartilage injury is identified arthroscopically in up to 76% of horses with meniscal damage.^{6,42} Our objectives were to

develop a stifle model including the three missing elements of the previous study to determine tibial contact pressure in stifles with an intact MFC and after creation of a MFC osteochondral defect of similar size and shape of a large SCL. Additionally, we placed a transcondylar screw through the defect to determine if this novel technique affected measured parameters in the medial tibial plateau. We hypothesized that our testing model with an intact MFC would detect central tibial cartilage contact pressures during flexion which was not detected in the previous model.²⁷ Our second hypothesis was that the creation of an osteochondral defect in the MFC would reduce the tibial contact pressure under the defect and transfer load caudally on the medial tibial plateau, as seen in human studies, and the transcondylar screw trials were included to investigate the potential biomechanical impact on joint contacts.⁴³

Chapter 2: Material and Methods

Cadaveric specimens

Six hind limbs (coxofemoral joint to mid-tibia) were collected from horses that were euthanized for reasons unrelated to musculoskeletal disease. Three left and right hind limbs were used from 2 mares and 2 geldings weighing approximately 450 kg. The breeds were Thoroughbred (2), Anglo-Arab (1) and Oldenburg (1), and the age ranged from 3-10 years (mean age was 6.75 years). After euthanasia, limbs were collected and stored at -20° C. Prior to testing, limbs were thawed at room temperature for 24-36 hours and the soft tissue envelope, excluding the stifle joint ligaments, was removed while the limbs were still cold.

Testing model and sensor selection

A custom-made device connecting the femur to the patella was used to maintain and adjust stifle angles during testing (Figure 1). This device consisted on a loop of 5 mm cable placed transversely through the patella and attached to a tensometer^a to register patellar tension during testing. The other end of the tensometer was attached to a second cable that extended proximally over the femoral neck and then down the caudal femur where it was attached by a turnbuckle to the mid-femur. A second turnbuckle was attached to the femur and tibia caudal to the stifle joint to maintain the desired stifle angle when loaded (Figure 2). A 12.25 mm hole was drilled transversely from lateral to medial

through the femoral head perpendicular to the long axis of the femur to allow placement of a 10 mm stainless steel rod. The specimen was then mounted via the rod into a servohydraulic machine^b for axial load testing (Figure 1). The transverse rod connected the femur to the actuator of the testing machine and the cut end of the tibia was loosely restrained to avoid damage of the testing machine in case of construct failure. A slightly compliant pad consisting of impression material^c was placed between the tibia and the base of the testing machine to allow axial or sagittal tibial rotation during testing. An ultra-thin digital electronic pressure sensor^d was used to determine medial femoro-tibial joint contact (Figure 3A). Tekscan® sensors are comprised of an array of individual sensing elements that determine the loading force, pressure and contact area received by the sensor and provides a pressure map image. They come in a variety of shapes and can register a broad range of pressures. In fact, the company offers the possibility of making custom-made sensors upon request (www.tekscan.com). In our case, the driving factor to select the K-ScanTM System model 4000 (Figure 3A) was its rectangular shape, which allowed complete coverage of the entire medial tibial plateau while preserving the medial collateral ligament in position. The model used has two 28 mm X 33 mm sensors each with 572 sensels (single sensor element), arranged at a density of 62 sensels/cm², with a pressure range from 0-62,000 KPa (0- 9,000 psi). The sensors are surrounded by green plastic tabs which can be used to secure the sensors within the joint. Before testing, the sensors were conditioned and calibrated following manufacturer recommendations.

Briefly, the sensor needs to be conditioned or “woken-up” before testing using a uniform load exceeding 20% of the intended testing load. Calibration refers to the method

by which a known force is applied to the sensor to convert a digital output into a known engineering unit. For a two-point power method calibration process, as used in our study, the sensors are loaded to 20% and 80% of the intended testing load. In our study, we calibrated the sensors between two flat steel plates covered with 2 mm of rubber (mimicking articular cartilage) to 400N and 1400N, representing approximately 20% and 80% of the peak load applied during testing (1800N). A third technique, called equilibration, can be used to minimize individual variations between individual sensing elements on any given sensor. Equilibration is accomplished by applying a highly uniform pressure across the individual sensing elements with a commercially available air-pressure equilibration device. This technique is recommended if the pressures applied to the sensor during testing are small enough that little variations in the sensing elements output can be significant for the results. This is not the case if limbs are loaded up to 1800N.

Our two sensors were attached side by side to cover the medial tibial plateau (56mm X 33mm) with 3/0 polydioxanone^e at each end and protected from moisture with an adhesive polyester film^f. After creation of a 2 cm² bone plug at its origin, the medial collateral ligament was reflected distally to facilitate positioning of the sensors between the medial meniscus and the medial tibial plateau. The sensors were sutured in place to the adjacent soft tissues with 2 lactomer 9-1^g to prevent dislodging during testing. The bone plug with the medial collateral ligament origin was reattached using a 6.5 mm cancellous bone screw and a washer in lag fashion (Figure 3C). The sensors covered the entire medial tibial plateau (Figure3B).

Mechanical testing, defect creation and screw placement

Stifle angles were adjusted and maintained by tightening or loosening both turnbuckles; 1) femur to patella, 2) femur to tibia (caudal). Each limb was loaded to 1800N at three stifle joint angles; full extension (155°), 145° and 130°. These angles approximate the stifle position from foot strike to push-off at the gallop (S. Stover, personal communication, 2012), and is similar to the range used in previous studies.^{27,44} Full extension was the angle achieved after maximally tightening the proximal turnbuckle manually with a wrench. The caudal turnbuckle was added to prohibit femoro-tibial joint hyper-extension under load in extension, which occurred in pilot trials. Angles were measured using a goniometer at maximal load using the long axis of the femur and tibia as the arms and the vertex was centered on the MFC. The load applied was selected based on the peak vertical force produced for hind limbs at walk (1800N for a 450Kg horse).⁴⁴

Testing began with a pre-load of 100N held for 5 sec, then load was increased at 50N/s up to 1800N. Peak load was held for 5 seconds for data recording including pressure maps. Nine tests were performed on each specimen; 3 MFC conditions (intact, defect, and with lag screw) at three angles (155°, 145°, 130°). Specimen testing order by condition was necessarily 1) intact, 2) defect and 3) screw, although the angles tested were randomized within each condition. Before every test, the stifle was lubricated with saline (0.9% NaCl) and the limb was manually flexed and extended 3 times to prevent the meniscus from getting stuck in a fixed position. After completion of the 9 tests, digital photographs of the tibial plateau with the sensors in place were taken to compare to the pressure maps.

After testing the intact MFC, the turnbuckle was disengaged, the stifle joint was flexed maximally and the dorsal joint capsule of the medial femoro-tibial joint was incised vertically to expose the weight-bearing surface of the MFC. A 15 mm round osteochondral defect was made 20 mm caudal to the cranial articular margin of the MFC at its weight-bearing center with a 15 mm drill bit (Figure 4A). The hole was drilled to a depth of 20mm. After defect testing, a 4.5 mm lag screw (glide hole 3.2 mm and thread hole 4.5 mm) 36 mm in length was placed from abaxial to axial through the defect in the condyle. The screw head was placed immediately proximal to the MFC cartilage rim, and the screw was angled distally 15-20° (Figure 4B).

Data collection and statistical analysis

For the combined sensor area (sum of cranial and caudal sensor data), the data collected consisted of total force (Newtons, N), contact area (square centimeters, cm²) and peak pressure (Megapascals, MPa) during each test, and pressure maps obtained with the Tekscan® software. Three additional regions of interest (ROI) were created in post-testing analysis in an attempt to identify pressure changes in specific regions of concern; 1) cranial sensor, 2) caudal sensors and 3) a rectangular area of 2.55 cm² in the central region of the combined sensors approximating the area of tibial cartilage uncovered by meniscus and the axial aspect of the meniscus (Figure 3B). Post testing analysis also calculated the stress load or contact pressure (force received by the sensor/contact area, N/cm²) for the 3 ROI and the combined sensors. Tension (pounds, lb) registered by the tensometer attached to the patella was recorded for every test at the preload level (100N) and at 1800 N.

Data was tested with an statistical software^h. A two-way repeated measures ANOVA with Bonferroni post-hoc tests was used to evaluate for significant differences between stifle angles and conditions (intact, defect, screw) for peak pressure, force, contact area and stress load for both sensors; for peak pressure and stress load for the 3 additional ROI. Significance was set at $P < .05$. The data obtained from the tensometer were analyzed descriptively.

Figures

Figure 1. Cranial view testing model. Cranial view of the stifle (graphic image on the left and cadaveric specimen on the right) mounted in the testing machine. The patellar tension device, including the tensometer (arrow), can be seen.

Figure 2. Caudal view testing model. Graphic caudal perspective of the stifle mounted in the testing machine.

Figure 3. Pressure sensor inside and outside the joint.

Figure 3A: Tekscan® K-Scan sensor model 4000. The sensing segments of the sensors are both rectangular areas surrounded by green tabs. Green tabs are used to secure sensors within the joint.

Figure 3B: Close-up view of the sensors placed under the medial meniscus. The sensors are attached and fixed in position; the cranial sensor is under the cranial meniscus and the caudal sensor under the caudal meniscus. The dotted line represents the area in the central region of interest.

Figure 3C: Medial aspect of the femoro-tibial joint after Tekscan placement between the medial meniscus and the medial tibial plateau and re-attachment of the medial collateral ligament using a 6.5 mm cancellous bone screw.

Figure 4. Medial femoral condyle defect and transcondylar screw through the defect.
Figure 4A: Distal aspect of the MFC after creation of the osteochondral defect in the center of the weight-bearing surface.

Figure 4B: Same view as Figure 4A with the transcondylar screw through the defect. The screw is placed from abaxial to axial immediately proximal to the medial femoral condyle cartilage rim, and angled distally 15-20°.

Chapter 3: Results

The mean stifle joint angle measured by goniometer for 6 specimens at full extension was 155°, at mid-flexion was 145° and at greatest flexion was 130°. Mean values for tibial force, contact area, peak pressure and stress load for the combined sensors (sum of cranial and caudal sensor data) are listed in Table 1.

Flexion angle effect

All comparisons (force, contact area, peak pressure and calculated stress load) for flexion angles used 155° (extension) as the baseline (Figure 6 & 7). For an intact MFC, both flexion angles demonstrated increased tibial force, contact area and stress load ($p < .001$) but no change in peak pressure. Flexion of specimens with a MFC defect also exhibited increases in force, area and stress load ($p < .05$), however the peak pressure was reduced at 130° ($p < .05$). Flexion of specimens with a MFC defect and a screw demonstrated increased force, contact area and stress load ($p < .001$), and the peak pressure at 130° was the same as at 155°.

Condition effect

The same parameters were analyzed to determine the impact of altered MFC condition (defect, and defect with transcondylar screw) at each tested angle using the intact MFC as the benchmark (Figure 8). At 155°, contact area was reduced in specimens with a defect and screw only ($p < .01$). At 145°, specimens with a defect and a defect and

screw demonstrated reduced force and contact area ($p<.05$). And at 130° , specimens with a defect demonstrated reduced force as compared to 155° ($p<.05$), with screw and intact conditions not being statistically different from each other.

Peak pressure and stress load analysis

Peak pressure and calculated stress load were analyzed in three ROI to determine the location of any significant changes (Table 2). Peak pressure decreased on the cranial sensor and central ROI with a defect at 130° only ($p<.05$). Peak pressure increased on the caudal sensor during flexion under all conditions ($p<.001$). Calculated stress load (CSL) on the cranial sensor increased with flexion in the intact and defect with screw conditions ($p<.05$) but not with a defect when angles were compared, and was decreased at 155° with a defect and screw ($p<.05$) and at 130° with a defect ($p<.05$) when conditions were compared. CSL on the caudal sensor increased with flexion ($p<.001$) and was unaffected by condition. CSL on the central ROI was increased with flexion ($p<.001$) only with intact condition, and was decreased at 130° with both a defect and a defect and a screw condition ($p<.001$).

Pressure maps analysis

Qualitative evaluation of the combined pressure map images revealed an outline of the medial meniscus and a central area corresponding to the cartilage on the medial tibial plateau uncovered by meniscus (Figure 5). In the intact stifle, when the joint was flexed, contact pressure was transmitted to the caudal meniscus, and peak and high pressures remained on the central cartilage area at all angles. After creation of a MFC defect, no contact pressure was registered in the defect area, and in extension, pressure

tended to migrate caudally. This was exacerbated progressively with flexion. At 130°, the caudal tibial plateau registered higher pressures in all limbs, and the peak pressure (indicated by the small black box in Figure 5) migrated from the central area to the caudal plateau in 3/6 specimens. The pressure maps after screw placement were not different from those of the defect alone.

Patellar tension analysis

The tension registered by the patellar tensometer increased linearly with axial load at each angle. Tension for every condition was highest in extension (mean 70.8 ± 33.1 kg at 155°) and was similar at both flexion angles 25.4 ± 14.5 kg at 145°, and 24.9 ± 13.6 kg at 130°.

Tables

Table 1

Force (N)	Intact	Defect	Screw
155°	887 +/- 65	761 +/- 75	649 +/- 53
145°	1506 +/- 40	<u>1192 +/- 55</u>	<u>1221 +/- 67</u>
130°	1618 +/- 84	<u>1364 +/- 101</u>	1436 +/- 76
Area (cm²)	Intact	Defect	Screw
155°	11.7 +/- 0.4	10.5 +/- 0.8	<u>10.1 +/- 0.8</u>
145°	14.7 +/- 0.4	<u>13.2 +/- 0.8</u>	<u>13.3 +/- 0.5</u>
130°	14.6 +/- 0.5	13.6 +/- 0.7	13.6 +/- 0.5
Peak Pressure (MPa)	Intact	Defect	Screw
155°	2.5 +/- 0.5	2.9 +/- 0.5	2.4 +/- 0.4
145°	2.2 +/- 0.3	2.2 +/- 0.2	2.1 +/- 0.2
130°	2.2 +/- 0.3	2.1 +/- 0.3	2.3 +/- 0.2
Stress Load (N/cm²)	Intact	Defect	Screw
155°	76 +/- 4.4	72.7 +/- 4.6	64.5 +/- 2.2
145°	103 +/- 4.3	90.9 +/- 3.8	92.9 +/- 7.5
130°	111 +/- 6.1	102.3 +/- 9.8	106.4 +/- 7.4

Units are displayed in Newtons (N), square centimeters (cm²), Megapascals (MPa) and Newtons divided by square centimeters (N/cm²).

Table 1. Mean +/- SEM for force, contact area, peak pressure and stress load. Mean values +/- standard error of the mean for force registered by the sensors (Force), contact area (Area), peak pressure and calculated stress load (Stress load) are shown for the 6 stifles. Significance was set at $p \leq 0.05$. Bold values indicate a significant difference from extension (155°). Underlined values indicate a significant difference of condition as compared to an intact MFC.

Table 2

Peak Pressure (MPa)			
Cranial sensor	Intact	Defect	Screw
155°	2.5 +/- 0.5	2.9 +/- 0.6	2.4 +/- 0.4
145°	2.2 +/- 0.3	2.2 +/- 0.2	2.1 +/- 0.2
130°	2.1 +/- 0.3	1.9 +/- 0.2	2.1 +/- 0.2
Peak Pressure (MPa)			
Caudal sensor	Intact	Defect	Screw
155°	1.3 +/- 0.1	1.4 +/- 0.1	1.3 +/- 0.1
145°	1.8 +/- 0.1	1.8 +/- 0.1	1.8 +/- 0.2
130°	1.9 +/- 0.2	2 +/- 0.2	2.1 +/- 0.2
Peak Pressure (MPa)			
Central region	Intact	Defect	Screw
155°	2.5 +/- 0.5	2.8 +/- 0.6	2.4 +/- 0.4
145°	2.2 +/- 0.3	2.2 +/- 0.2	2.1 +/- 0.2
130°	2.2 +/- 0.3	1.8 +/- 0.2	2 +/- 0.2
Stress Load (N/cm ²)			
Cranial sensor	Intact	Defect	Screw
155°	88.6 +/- 3.4	84.2 +/- 8.2	<u>74.4 +/- 3</u>
145°	109.1 +/- 6.6	98.8 +/- 5.8	96.6 +/- 7.5
130°	112.9 +/- 7.5	<u>99.1 +/- 9.7</u>	108.3 +/- 8.1
Stress Load (N/cm ²)			
Caudal sensor	Intact	Defect	Screw
155°	66.3 +/- 7.6	59 +/- 5.6	54.3 +/- 5.5
145°	98.3 +/- 4.4	84.2 +/- 3.8	89.7 +/- 8.9
130°	109.5 +/- 6.6	104.9 +/- 10.1	104.9 +/- 7.5
Stress Load (N/cm ²)			
Central region	Intact	Defect	Screw
155°	109.6 +/- 6.3	110 +/- 5.5	106.5 +/- 12.3
145°	154.2 +/- 13.3	128.3 +/- 8.3	130 +/- 13.3
130°	163.4 +/- 18.4	<u>122.1 +/- 13</u>	<u>126.9 +/- 12.5</u>

Units are displayed in Newtons (N), square centimeters (cm²), Megapascals (MPa) and Newtons divided by square centimeters (N/cm²).

Table 2. Mean +/- SEM for peak pressure and stress load in the 3 ROI. Mean values +/- standard error of the mean for peak pressure and stress load values for the 3 regions of interest (cranial and caudal sensors and central region) are shown for the 6 stifles. Significance was set at $p \leq 0.05$. Bold values indicate a significant difference from extension (155°). Underlined values indicate a significant difference of condition as compared to an intact MFC.

Figures

Figure 5. Pressure maps for every angle and condition. Pressure map images of a representative stifle obtained by combining the cranial and caudal sensor maps. Angles are located from left to right (155°, 145° and 130°) and conditions from top to bottom (intact, defect and screw). The color scale on the left represents the pressure magnitude (MPa) for the Tekscan® maps, with blue being low pressures and red being maximum pressures. The white dotted circle represents the location of the 15mm round osteochondral defect in the medial femoral condyle. The black box inside every pressure map represents the location of peak pressure. Note that at full extension (155°) there is less contact area than during flexion. During flexion of intact specimens, high pressures appear in the central and caudal tibial plateau region. After creation of the osteochondral defect in the medial femoral condyle high pressures migrated even more towards the caudal aspect of the tibial plateau.

Figure 6. Mean $\pm$ SEM for force when conditions are compared. Mean $\pm$ standard error of the mean for force when the different angles were compared. An asterisk represents every group with a significant difference when compare to 155° ($p < .001$).

Figure 7. Mean +/- SEM for contact area when conditions are compared. Mean +/- standard error of the mean for contact area when the different angles were compared. An asterisk represents every group with a significant difference when compare to 155° (p<.001).

Figure 8. Mean +/- SEM for stress load when conditions are compared. Mean +/- standard error of the mean for stress load when the different angles were compared. An asterisk represents every group with a significant difference when compare to 155° (p < .05).

Figure 9. Mean +/- SEM for force when angles are compared. Mean +/- standard error of the mean for force when the different conditions were compared. An asterisk represents every group with a significant difference when compare to intact condition ($p < .05$).

Chapter 4: Discussion

Tekscan® technology is an extremely useful technique for determining joint contact, however, its use requires in-depth familiarity with the system and a number of crucial steps. The first step was to determine what type of sensor would be optimal for our purpose. After contacting the company and getting familiar with several sensors used in human and veterinary studies, numerous ex-vivo trials were performed with different sample sensors to determine the sensor to be used in our study. The sensor selected (Tekscan® K-Scan sensor model 4000. Figure 3) allowed us to obtain contact maps from the entire medial tibial plateau while preserving the medial collateral ligament. Two sensors were attached side by side and the pressure maps from every sensor were combined to obtain a graphic representation of the medial tibial plateau (Figure 5). The second step consisted in the development of a uniform technique to condition and calibrate the sensors. Based on manufacturer's recommendations, the sensors were loaded extra-articularly in contact to rubber that mimicked the articular cartilage. This process could also be done intra-articularly but uniform loading could not be assured. The third step tested sensors under experimental conditions to determine optimal sensor placement under the meniscus, sensor attachment to prevent dislodging during testing and how to prevent or identify sensor damage during testing. The sensors are extremely sensitive and

unavoidably some sensels will get damaged after few tests, so usually one sensor was used for each tested limb.

All ex-vivo models differ from in-vivo conditions, but can provide useful information without the expense and risks of live animal studies. This study demonstrates the impact of axial loading on medial tibial contact pressures in a stifle model with an intact patella and peri-articular ligaments. Furthermore, it models the impact of an osteochondral defect similar in size to naturally occurring SCL on tibial contact pressures, and the impact of a transcondylar screw through the defect on the same pressures. The contact pressure maps in the intact MFC trials revealed peak pressures on the central cartilage of the medial tibial plateau uncovered by meniscus at each angle tested (Figure 3B & 5). This differs from the previous report determining equine tibial contact pressure, which identified central cartilage peak pressures only at full extension (160°), and indicated minimal contact pressure during three flexion angles.²⁷ In humans, the tibial plateau uncovered by meniscus transmits 30-50% of the compressive load and it seems likely that the central cartilage area would also be loaded in the horse.^{36,45,46} In fact, a recent equine study using three dimensional radiostereometric analysis has determined joint contact in the central cartilage of the medial tibial plateau from 120°-155°.⁴⁷ This study also determined external tibial rotation as the joint was extended. Our model included the medial collateral ligament, allowed tibial rotation and provided tension on the patella, elements missing from the previous report using Tekscan® technology.²⁷ Inclusion of these elements may be responsible for the difference in pressure maps between studies.

Both stifle flexion angles tested resulted in an increase in contact area on the medial tibial plateau and no increase in peak pressures as previously reported.²⁷ The increase in force and stress load on the medial tibial plateau during flexion suggests the medial femoro-tibial joint plays an important role in load transfer during flexion. The lower force measured in extension may be the result of force being born by the lateral condyle, medial intercondylar eminence of the tibia, or the patella, which were not investigated in this study. The increase in contact area without change in peak pressure in flexion indicates that the medial meniscus in the horse, as in the human, plays an important role in load transfer and protects the cartilage from excessive axial loads.⁴⁸

Subchondral cystic lesions are a common cause of lameness affecting the medial femoro-tibial joint, but the mechanical impact of MFC SCL on the stifle joint is unknown. In this ex-vivo study, a MFC defect similar to a large SCL reduced force on the medial tibial plateau at two flexion angles and contact area at 145°. Stress load on the combined tibial area was not affected by a defect but was decreased in the cranial and central ROI at 130°. These alterations in the biomechanics of the medial femoro-tibial joint by the SCL may play a role in lameness and appear to have greatest effect during flexion. Although a correlation between lameness and decreases in tibial force and area is somewhat counter-intuitive, decreases in contact area may contribute to stifle OA due to overstress of other regions within the joint as shown in equine, canine and human meniscectomy studies.^{27,34,36} The question of where the applied force goes must remain unanswered as only the medial tibial plateau was tested. Areas on the tibia and patella that were not tested may be experiencing higher loads, causing lameness. Data from the

tensometer attached to the patella suggested patellar load sharing during limb loading. Also, these reductions in tibial contact area and force suggest that SCL load is greatly diminished which may be the biomechanical reason why SCL infrequently demonstrate radiographic healing after treatment.^{20,25}

Meniscal injuries associated with MFC SCL have been recently reported.²⁶ In the authors' experience, meniscal injuries associated with MFC SCL demonstrate the most severe damage in the mid to caudal meniscus, which differs from meniscal injuries not associated with SCL.^{6,49} Based on the stress load data and pressure map images, we propose that caudal meniscal damage is more likely to occur during weight-bearing flexion, and the caudal migration of high pressures during flexion suggests a focal increase in stress that may contribute to caudal meniscal damage. This finding was exacerbated after creation of the defect, with stress load decreasing in the central ROI and increasing in the caudal ROI during flexion.

In humans, an ex-vivo study using Tekscan® technology has shown that the size of an osteochondral defect in the MFC will determine the location of stress concentration around the rim of the defect.³⁷ As in our study, size defects small in comparison to MFC size produced re-distribution of the stress on the defect periphery that was transferred to the meniscus. Force alterations, combined with the irregularity of the cartilage surface of the MFC SCL before or after surgical debridement, may result in meniscal damage and contribute to OA and lameness in some horses. This has been already proposed by Hendrix et al in clinical cases with concomitant SCL and medial meniscal damage.²⁶

The authors have developed a MFC transcondylar lag screw technique that in clinical patients promotes SCL healing.²⁵ In this study, we were obligated to test the limbs in a sequential mode: intact, defect and screw. Significant effects of a transcondylar screw should reverse the alterations caused by the defect or differ from intact values when the defect condition had no significant effect. In this study, a MFC defect reduced measured force on the medial tibia at both flexion angles, and decreased the contact area at 145°. The transcondylar lag screw resulted in reversals of the force reduction caused by the defect at 130°, and a decrease in contact area at 155°. The force reduction reversal suggests that a transcondylar screw may reverse some biomechanical alterations at some angles caused by a MFC defect. The calculated stress load data indicates that this effect also occurs on the cranial ROI of the medial tibial plateau. The decrease in contact area in extension suggests that the screw placed in lag fashion across the defect may have resulted in compression of the MFC. We suggest this data supports the hypothesis that a transcondylar screw placed through a defect may affect MFC biomechanics. Further research is ongoing to determine changes in the MFC biomechanics after defect creation and screw placement as well as the optimal placement and number of implants to normalize the MFC biomechanics.

There are many potential shortcomings in this model. Any model can only approximate in vivo conditions, and this model applies load uniformly and simplistically unlike dynamic living conditions, and does not account for the effects of muscle work. Therefore, while these contact maps approximate in vivo loading, they are probably best suited for comparisons between each other. We were unable to achieve the 160°

extension angle that has been reported in a previous ex-vivo study, but were closer to the angles reported in an in-vivo kinetic study and an ex-vivo kinematic study.^{27,44,47} This is probably due to differences in the preparation of ex-vivo specimens as the limbs in this study retained some caudal soft tissues of the stifle joint and also the patella, which prevents cranial translocation of the femur. As used in other studies, we temporarily detached the MCL to allow intra-articular sensor placement.^{34,36,48} The technique disrupted the attachment between the medial meniscus and MCL but avoided extrusion of the medial meniscus during loading, which it could happen if the MCL is removed.

Future directions using Tekscan® technology includes testing of stifles with naturally occurring MFC SCL although SCL size variability may affect results. We would also like to investigate lateral femoro-tibial joint contact and the changes on it after creation of a MFC defect. As seen in other studies, pathology in the medial compartment may affect the lateral compartment.^{30,32,33} Lastly, investigation of different MFC defect sizes impact on joint contact may help determine when surfacing chondrocyte transplantation techniques may be advocated in adjunction to transcondylar screw technique or other therapies.³⁷

In conclusion, in the intact stifle the peak load is on the central cartilage area on the medial tibial plateau between 155°-130°. As the stifle flexes, the medial tibial plateau experiences more force, and the engagement and radial deformation of the meniscus increases contact area and stress load. A simulated MFC defect reduces the force and contact area on the medial tibial plateau during flexion and alters the pressure maps by transferring high pressures to the caudal meniscus. Placement of a transcondylar screw

through the defect reverses some of the alterations in force and stress load caused by the defect. These results add to our understanding of equine stifle biomechanics. The alterations in tibial force, contact area and map pressures in the MFC with a defect improve our understanding of SCL pathology and its potential repercussion in the medial meniscus.

Footnotes

- a. LCR-500 S-Beam load sensor and DP25B-S strain meter Omega®, Omega Engineering INC., Stamford, CT.
- b. 858 Bionix®, MTS Systems Corporation, Eden Prairie, MN.
- c. Reprosil®, Dentsply® Caulk, Milford, DE.
- d. K-Scan™ System, Model 4000, Tekscan Inc, South Boston, MA.
- e. Ethicon®, Somerville, NJ.
- f. Ioban®, 3M™, St. Paul, MN.
- g. Syneture®, Covidien, Mansfield, MA.
- h. Graph Pad Prism 5®, La Jolla, CA.

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Appendix A: Sensor calibration and raw data

The sensor used in our study is a K-ScanTM System model 4000. This model has two rectangular 28 mm X 33 mm sensors each with 572 sensels (single sensor element), arranged at a density of 62 sensels/cm², with a pressure range from 0-62,000 KPa (0-9,000 psi). Before testing, the sensors were conditioned and calibrated following manufacturer recommendations. Briefly, the sensor needs to be conditioned or “woken-up” before testing using a uniform load exceeding 20% of the intended testing load. Calibration refers to the method by which a known force is applied to the sensor to convert a digital output into a known engineering unit. For a two-point power method calibration process, as used in our study, the sensors are loaded to 20% and 80% of the intended testing load. In our study, we calibrated the sensors between two flat steel plates covered with 2 mm of rubber (mimicking articular cartilage) to 400N and 1400N, representing approximately 20% and 80% of the peak load applied during testing (1800N).

Table 3

Specimen	Force for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	806	1530	1765	989	1069	1523	630	1464	1579
Leg 2	852	1392	1523	730	1148	1347	564	1181	1375
Leg 3	1195	1604	1597	995	1242	948	769	964	1190
Leg 4	857.1	1506	1842	637.7	1393.5	1432.2	525.4	1180.9	1331.4
Leg 5	881.1	1613.5	1711.5	612.7	1031	1265.2	558.4	1308.8	1426.2
Leg 6	733.5	1391.6	1267.8	603	1266.6	1670.7	848.2	1230.1	1714.8

Table 3. Raw data for force registered by the sensor. Force (Newton) registered by the sensor for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 4

Specimen	Contact area for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	11.5	14.21	13.68	10.74	11.76	11.89	9.74	12.77	12.53
Leg 2	11.82	15.08	15.21	10.92	14.05	14.23	9.74	13.69	13.92
Leg 3	13.1	16.37	16.34	13.76	15.69	15.69	12.82	15.19	15.42
Leg 4	12	14.97	15.66	9.21	14.92	14.55	8.37	12.63	13.37
Leg 5	10.11	13.24	13.58	7.9	10.65	11.24	7.73	11.94	12.19
Leg 6	11.4	14.23	13.24	10.31	12.21	13.97	12.26	13.63	14.27

Table 4. Raw data for contact area registered by the sensor. Contact area (cm²) registered by the sensor for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 5

Specimen	Peak pressure for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	3.49	2.98	3.22	4.6	2.51	3.11	3.43	2.72	3.08
Leg 2	1.69	1.72	1.69	1.86	1.59	1.75	1.65	1.55	1.83
Leg 3	1.66	1.39	1.43	2.74	2.7	1.53	2.13	2.27	1.93
Leg 4	4.25	3	2.72	4.16	2.66	2.03	3.66	2.3	2.33
Leg 5	1.41	1.83	1.79	1.22	1.55	1.7	1.1	1.61	1.69
Leg 6	2.79	2.1	2.06	2.74	2.29	2.62	2.72	1.94	2.68

Table 5. Raw data for peak pressure registered by the sensor. Peak Pressure (MPa) registered by the sensor for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 6

Specimen	Peak pressure in cranial ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	3.49	2.98	3.22	4.6	2.51	2.53	3.43	2.72	2.85
Leg 2	1.69	1.65	1.52	1.86	1.58	1.4	1.65	1.55	1.28
Leg 3	1.66	1.39	1.43	2.74	2.7	1.53	2.13	2.27	1.93
Leg 4	4.25	3	2.72	4.16	2.66	1.88	3.66	2.3	2.23
Leg 5	1.41	1.83	1.79	1.03	1.55	1.7	0.98	1.61	1.69
Leg 6	2.79	2.1	2.06	2.74	2.29	2.62	2.72	1.94	2.68

Table 6. Raw data for peak pressure registered by the sensor in cranial ROI. Peak Pressure (MPa) registered by the sensor in the cranial ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 7

Specimen	Peak pressure in central ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	3.49	2.98	3.22	4.6	2.51	2.53	3.43	2.72	2.85
Leg 2	1.51	1.72	1.69	1.57	1.58	1.49	1.5	1.55	1.32
Leg 3	1.31	1.39	1.43	2.74	2.7	1.53	2.08	2.27	1.93
Leg 4	4.25	3	2.72	4.16	2.66	1.83	3.66	2.3	2.23
Leg 5	1.41	1.83	1.79	1.22	1.55	1.7	1.09	1.61	1.69
Leg 6	2.79	2	2.06	2.74	2.13	1.93	2.72	1.94	1.77

Table 7. Raw data for peak pressure registered by the sensor in central ROI. Peak Pressure (MPa) registered by the sensor in the central ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 8

Specimen	Peak pressure in caudal ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	1.36	2.36	2.99	2.02	2.21	3.11	1.66	2.51	3.08
Leg 2	1.51	1.72	1.69	1.41	1.59	1.75	1.24	1.55	1.83
Leg 3	1.22	1.27	1.28	1.32	1.89	1.53	1.13	1.42	1.83
Leg 4	0.92	1.81	2.18	1.09	2.02	2.03	1.22	2.06	2.16
Leg 5	1.33	1.71	1.72	1.22	1.43	1.58	1.1	1.59	1.6
Leg 6	1.38	1.81	1.64	1.36	1.94	2.17	1.55	1.94	1.84

Table 8. Raw data for peak pressure registered by the sensor in caudal ROI. Peak Pressure (MPa) registered by the sensor in the caudal ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 9

Specimen	Stress load for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	70.08	107.67	129.02	92.08	90.9	128.09	64.68	114.64	126.01
Leg 2	72.08	92.3	100.13	66.84	81.7	94.65	57.9	86.26	98.77
Leg 3	91.22	97.98	97.73	72.31	79.15	60.42	59.98	63.46	77.17
Leg 4	71.42	100.6	117.62	69.23	93.39	98.43	62.77	93.49	99.58
Leg 5	87.15	121.86	126.03	77.55	96.8	112.56	72.23	109.61	116.99
Leg 6	64.11	97.79	95.75	58.48	103.73	119.59	69.23	90.24	120.16

Table 9. Raw data for calculated stress load. Calculated stress load (N/cm^2) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155° , 145° , 130°) tested.

Table 10

Specimen	Stress load for cranial ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	93.68	115.93	125.38	114.35	107.16	123.25	82.5	119	124.8
Leg 2	76.02	84.63	92.92	73.22	76.2	82.52	64.73	75.19	91.67
Leg 3	94.48	97.14	89.48	90.49	89.66	61.3	78.7	74.49	77.38
Leg 4	97.79	120.86	128.96	97.26	105.58	96.09	80.89	104.56	110.31
Leg 5	88.31	129.04	130.37	69.71	97.84	114	68.34	109.55	117.78
Leg 6	81.59	107.14	110.3	59.97	116.32	117.33	71.47	96.96	127.71

Table 10. Raw data for calculated stress load in cranial ROI. Calculated stress load (N/cm^2) in the cranial ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 11

Specimen	Stress load for central ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	134.04	193.06	235.04	112.6	159.75	179.2	155.25	190.12	185.62
Leg 2	102.39	133.3	135.7	92.89	107.66	111.46	82.01	111.59	100
Leg 3	97.6	106.04	105.14	126.33	140.44	98.65	82.74	104.68	106.5
Leg 4	98.52	186.12	191.45	120.78	113.73	90.55	118.97	143.4	119.03
Leg 5	101.02	148.58	158.1	95.63	112.82	134.25	80.48	122.08	126.19
Leg 6	123.96	158.22	155.2	111.6	135.35	118.63	119.52	108.34	123.93

Table 11. Raw data for calculated stress load in central ROI. Calculated stress load (N/cm^2) in the central ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.

Table 12

Specimen	Stress load for caudal ROI for every condition and angle								
	I-155°	I-145°	I-130°	D-155°	D-145°	D-130°	S-155°	S-145°	S-130°
Leg 1	51.37	101.72	131.55	62.55	80.61	130.94	48.17	112	127.16
Leg 2	71.2	98.7	106.34	61.34	86.66	105.54	52.42	94.95	104.77
Leg 3	88.56	98.86	105.18	53.48	69.49	59.67	39.37	52.58	77.04
Leg 4	45.23	84.08	107.21	38.49	81.24	100.46	45.11	84.89	90.87
Leg 5	86.31	116.03	122.48	80.46	96.23	111.39	73.81	109.79	116.28
Leg 6	54.89	90.38	84.25	57.45	91.13	121.31	67.23	84.07	113.46

Table 12. Raw data for calculated stress load in caudal ROI. Calculated stress load (N/cm^2) in the caudal ROI (region of interest) for every stifle under every condition (I: Intact, D: Defect, S: Screw) and angle (155°, 145°, 130°) tested.