

Effect of tibial tuberosity advancement on femoro-tibial shear in cranial cruciate-deficient stifles

An *in vitro* study

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Summary

Objectives: This *in vitro* study compares the femorotibial shear in canine stifles with intact and transected cranial cruciate ligaments (CrCL), before and after tibial tuberosity advancement (TTA) by measuring the distance between origin and insertion of the cruciate ligaments. **Methods:** Radiodense markers were inserted into bones at the attachment sites of the cruciate ligaments in sixteen cadaveric stifles of adult dogs. Each stifle was then mounted in a testing apparatus. The distances between the markers were measured on mediolateral radiographs, performed on each stifle under three different situations: intact, after CrCL transection, and after performing a TTA. Stifles were loaded to create a constant tibiofemoral reaction force by maintaining the load parallel to the patellar ligament. Radiographs were taken in a preloaded and loaded state in the intact stifle and only in a loaded state after CrCL transection, and after performing a TTA. **Results:** Loading the stifle joints after transection of the CrCL resulted in a mean lengthening of the CrCL marker distance of 22.4%. Loading the transected CrCL stifles after performing a TTA resulted in a mean shortening of the CrCL marker distance by 3.0% compared to the loaded intact condition. **Clinical significance:** This study demonstrates that, in loaded stifles with transected CrCLs, TTA causes a caudal shift in the cranial shear force, counteracting cranial subluxation of the tibia.

Keywords

In vitro study, cruciate ligament, stifle, tibial tuberosity advancement, dog

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Introduction

Partial or complete rupture of the cranial cruciate ligament (CrCL) is one of the most frequent causes of hindlimb lameness in dogs (1). An intact CrCL averts hyperextension of the stifle and limits cranial translation and internal rotation of the tibia (2–4). Rupture of the CrCL may result from acute trauma or, more frequently, from chronic degenerative processes (5, 6).

Several surgical techniques have been devised to restore the function of CrCL-deficient stifles (7). These can be divided into three categories: extra-articular techniques that augment the CrCL (8, 9), intra-articular techniques that replace the CrCL (10), and techniques that modify the geometry of the proximal tibia, altering the biomechanics of the stifle to neutralise femorotibial shear forces (11–13). The principles that underly the latter category of surgical techniques are therefore greatly influenced by biomechanical considerations.

In the human knee, the femorotibial forces that act during the weight-bearing phase of the gait are almost parallel to the patellar ligament. This joint force has a compressive and a shearing component. The angle between the patellar ligament and the tibial plateau affects both the direction and the magnitude of the femorotibial shear forces. This shear is maximal in anterior direction at full extension of the knee and shifts towards posterior during flexion (14).

In the dog it has been suggested that, as in humans, the joint force produced on the stifle by weight bearing is almost parallel to the patellar ligament (15) and produces a

shear force in a cranially oriented direction during the extension of the stifle, reaches zero at flexion when the patellar ligament is perpendicular to the tibial plateau and shifts to the caudal cruciate ligament with further flexion of the stifle (15, 16). An analysis involving the common tangent of the tibiofemoral contact point identified this cross-over point at approximately 110° of joint flexion (16).

The femorotibial shear can be eliminated by levelling the tibial plateau (17) or, as suggested, also by advancing the patellar ligament (18). This results in a caudal shift of the shear forces at a lesser angle of stifle flexion. Tibial tuberosity advancement (TTA) was developed to restore stability in CrCL-deficient stifles and has been successfully used in clinical cases (19–22).

The goal of the present study was to further investigate the biomechanical principles underlying TTA and to evaluate its effect on the femorotibial shear in CrCL-deficient canine stifle joints. Our hypothesis was that the craniocaudal stability of a stifle joint, rendered unstable by transection of the CrCL, could be restored by performing a TTA.

Material and methods

Data collection

Sixteen stifles of nine dogs that had been euthanised for reasons unrelated to stifle disease and that had been free of signs of stifle disorders were collected. Only one stifle was collected from each of two dogs be-

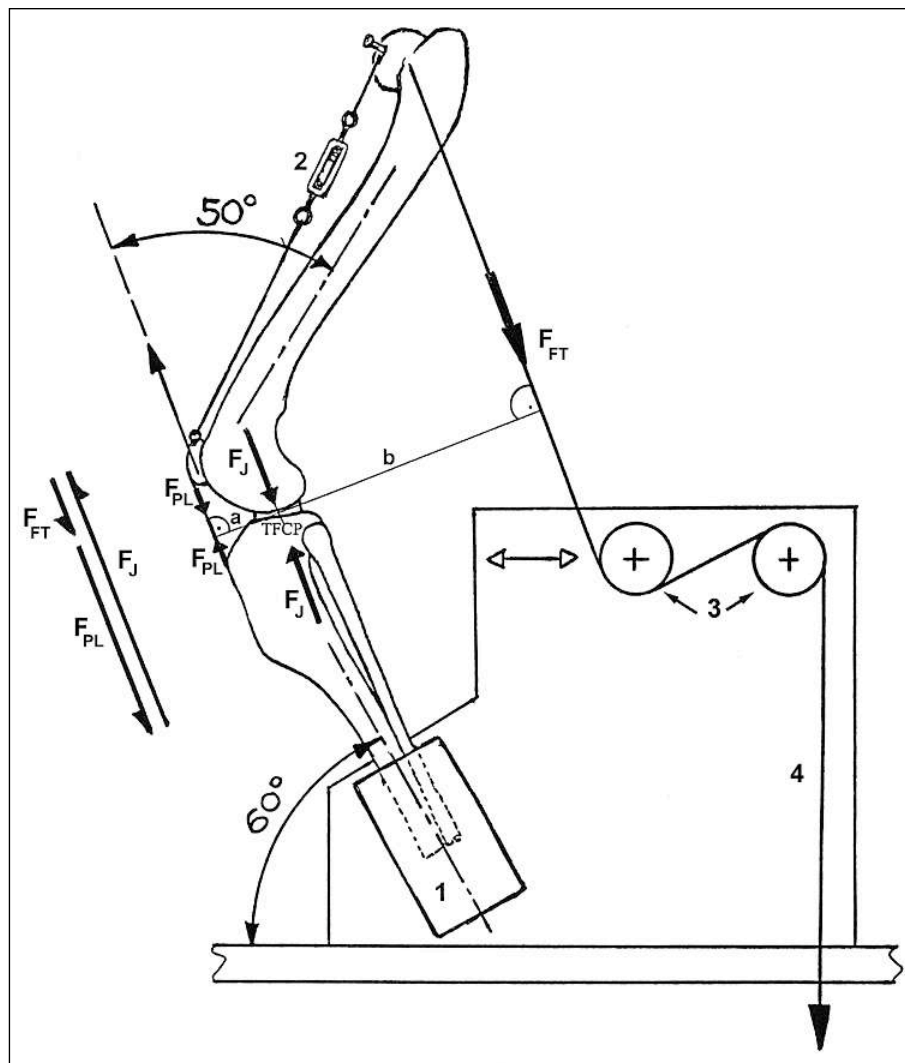


Fig. 1 Illustration of the stifle testing apparatus. The distal tibia was potted in polymethylmethacrylate and clamped into a socket (1) in the jig. The stifle joints were adjusted and locked at the desired position with the help of a turnbuckle (2). Two horizontally moveable deflection rollers (3) were tightened in a position so as to precisely direct the traction force. The tension line (4) was placed parallel to the patellar tendon and transfers over the deflection rollers to the weights. The length (a) from the patellar ligament to the tibiofemoral contact point (TFCP) in relation to the length (b) from the TFCP to the traction force was the applied lever arm. The force vectors are labelled as follows: F_{FT} Traction force, F_J Joint force; F_{PL} Patellar ligament force.

cause one limb was affected by multiple fractures caused by a road traffic accident and a regional bone tumour, respectively. The soft tissues surrounding the joints were removed, leaving in place only the peri-articular tissues, including the patella, the patellar ligament, collateral ligaments, and the joint capsule. Each limb was disarticulated at the coxofemoral and tibiotarsal joint, wrapped in saline-moistened towels, sealed in plastic bags, and stored at -24°C . Twenty-four hours prior to testing, each

specimen was thawed in a refrigerator at 3 to 5°C , and then re-warmed at room temperature (23).

On the day of testing, stainless steel beads of 1 mm diameter, acting as radiodense markers, were impacted at the femoral and tibial attachments of the CrCL and caudal cruciate ligament (CdCL). The articular capsule was incised caudally at the point where the CrCL originates, at the intercondylar surface of the lateral femoral condyle. The chosen position of drilling was

in the middle of the broad attachments of the cruciate ligaments. A 1-mm deep hole was drilled in the bone with a drill guide acting as a stopper. The stopper defined the drilling depth. Accordingly, a metal bead (1 mm) was placed into the drill guide and impacted into the bone with a punch. After preparation of the caudal joint capsule, a second bead was inserted in the same way at the insertion of the CdCL, near the popliteal notch of the tibia. A third marker was fixed at the origin of the CdCL on the intercondylar surface of the medial femoral condyle, and a fourth marker was placed at the insertion of the CrCL in the cranial intercondylar area of the tibia, after opening the cranial articular capsule with a small lateral incision. Consistency of positioning of markers was evaluated by gross visualization and radiographic imaging.

A custom-designed aluminium testing apparatus (Fig. 1) was applied to hold the specimen and to simulate the quadriceps mechanism. The distal extremity of the tibia was potted in polymethylmethacrylate and clamped at 60° to a horizontal plane into the socket of a jig that was fixed on a table. A turnbuckle was placed between the patella and the proximal femur maintaining the hindlimb in extension. The turnbuckle was fastened with a bicortical screw on the proximal femur; a Kevlar-braided line (Kevlar, AHF Leitner, Pfaffenhofen, Germany) was passed through a hole in the middle of the patella and through the distal eye of the turnbuckle. A second Kevlar line was attached to the bicortical screw in the proximal femur and lead over two deflection pulleys to a metal tray that weighed 0.8 kg. Four weights of 1.5 kg each were loaded onto the metal tray, resulting in a total load of 6.8 kg.

Mediolateral radiographs of each intact stifle were taken in preloaded ($_{i(p)}$ CrCL) and loaded ($_{l}$ CrCL) states with the x-ray beam directed perpendicular to the sagittal plane of the stifle and the femoral condyles superposed. Radiographs were again taken in a loaded state after transecting the CrCL ($_{t}$ CrCL) and after performing a TTA ($_{TTA}$ CrCL) to advance the tibial tuberosity (TTA plates, forks and cages: Kyon AG, Zurich, Switzerland) as has been described (19, 21, 22); however an older generation of cages was used in this study. This gener-

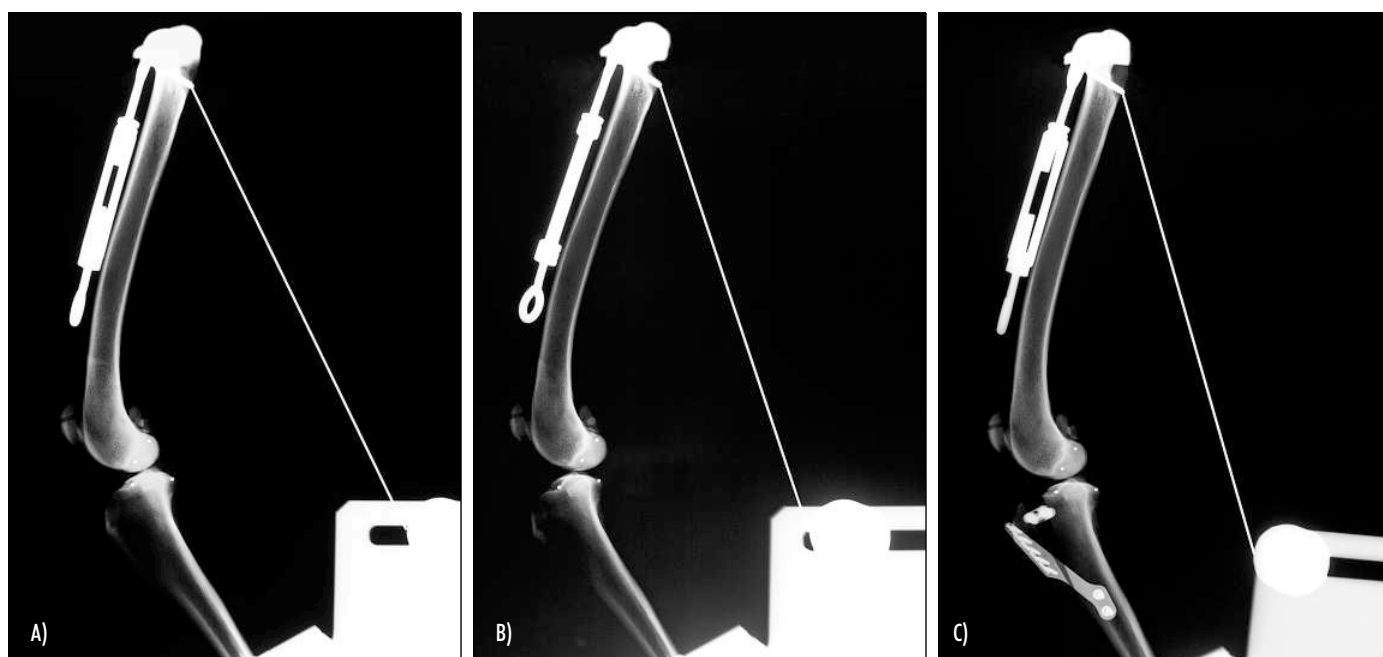


Fig. 2 A) Medirolateral radiograph of a stifle in an i CrCL situation showing the radiodense markers. Notice the distances between the markers representing the cruciate ligaments. B) Medirolateral radiograph of a stifle in a i CrCL situation. Note the displacement of the radiodense markers with the CrCL transection. C) Medirolateral radiograph of the same stifle after performing a TTA CrCL situation. The return of the radiodense markers to the initial state is to note.

ation of cages were used during the development of TTA. They are made of pure titanium with three sizes available: 6 mm, 9 mm and 12 mm. These cages were never commercially available. Only 6 or 9 mm cages were used. A preload of 0.8 kg (the weight of the tray) was used. The stifle joints were set at approximately 130° extension. This position was reset by the turnbuckle before obtaining radiographs of the stifles in each condition evaluated. The position of the pulleys was corrected to direct the load approximately parallel to the patellar ligament, so as to avoid changes in the magnitude of the force applied to the limb. The distance between the radiograph cassette, the stifle and the beam source was held constant. Fluoroscopy was used in order to obtain perfect superposition of the femoral condyles.

The distances between the beads placed at the origins and insertions of the CrCL and CdCL were measured on the radiographs (Fig. 2A-C). Measurements were made manually with a ruler, and normalised to the original lengths of the CrCLs in the preloaded, intact situation. The initial lengths of intact CrCLs and CdCLs in the preloaded

state with an applied preload of 0.8 kg were defined as 100% (Fig. 3).

To quantify the resulting force acting on the tibiofemoral joint after applying load at the proximal femur, the lever arm of the femur was calculated as follows: a segment perpendicular to the patellar ligament and to the traction force, traversing the tibiofemoral contact point (TFCP), was marked and the distances between the patellar ligament and TFCP (a), and between the TFCP and the line of the traction force were measured (b). The distance 'a' from the patellar ligament to the TFCP in relation to the distance 'b' from the TFCP to the traction force was the applied lever arm (Fig. 1). Force vectors drawn directly on the figure (F_{PL} , F_{FT} and F_J) indicate force orientations only. The magnitudes are shown on the isolated force diagram to the left of the stifle sketch and are calculated from the ratio of the distances b and a:

$$F_{PL} = (b/a) F_{FT}$$

$$F_J = F_{PL} + F_{FT} = (1+b/a) F_{FT}$$

The angle of flexion of each stifle was measured on the radiographs. The parallel-

ism between the patellar ligament and the direction of the traction force was controlled in the mediolateral radiographs.

Statistical analyses

Data were analysed using statistical software (Stat View 5.1 (SAS Inc.)). The $i(p)$ CrCL was given the 100% value for each dog by measuring the length of the intact preloaded cranial or caudal cruciate ligament. i CrCL, i CrCL, and TTA CrCL were measured and compared using the same leg as a repeated measure in analysis of variance (ANOVA). Limb, sex and implants were added to different models separately since it was not possible to include all parameters in one model due to over saturation of the model. None of these parameters had a significant influence and were therefore eliminated in further evaluation. An unpaired t-test was used to compare angle of flexion and the parallelism between i CrCL and TTA CrCL.

A comparison between different groups was performed by two level repeated ANOVA and not repeated factors were com-

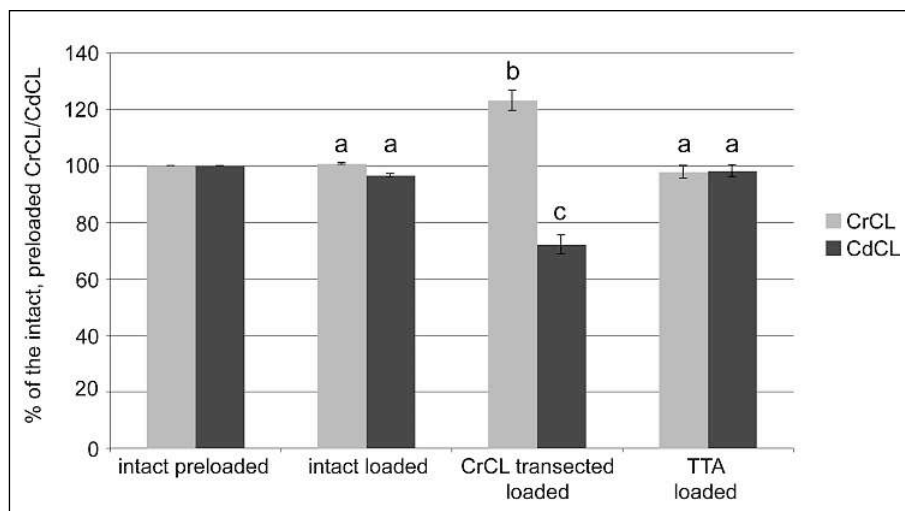


Fig. 3 Bar chart showing the distance between markers representing the CrCL and CdCL under different situations. Results are expressed as a mean \pm the standard error as percentages of the distance in a $i_{(p)}$ CrCL situation. Different indices show significant differences by means of $p < 0.05$. CrCL: cranial cruciate ligament, grey pattern; CdCL: caudal cruciate ligament, black pattern; TTA: tibial tuberosity advancement.

pared using a Bonferroni-Dunn test in a second factorial ANOVA. Linear regression was performed comparing two continuous variables, such as CrCL with age and CrCL with body weight. Significance was set at $P < 0.05$.

Results

The nine dogs ranged from three to 10 years old (mean: 6.8 ± 2.3 years). The weight of the dogs ranged from 21 to 41 kg (mean: 35.3 ± 8.3 kg). None of the stifle joints had radiographical signs of degenerative joint disease. No significant influence ($p > 0.05$) was found by means of ANOVA with length of the cruciate ligaments for sex (male, male castrated and female castrated) and implants (4-hole-plate-6 mm cage, 5-hole-plate-6 mm-cage, 6-hole-cage-9 mm-cage, 6-hole-cage-9 mm-cage) in a mixed model. There was a lack of significant relationship (regression analysis, $p > 0.05$) between age and body weight with the length of the cranial and caudal cruciate ligament between $i_{(p)}$ CrCL, i CrCL, t CrCL, TTA CrCL.

The resulting force on the patellar ligament in stifles flexed at approximately 130° was an average of 3.8 times greater (range:

3.1 to 4.9) than the force applied to the proximal femur. These calculated forces were in the range of 21 to 33 kg. The adjustments on the turnbuckle resulted in a flexion angle of $126.1^\circ \pm 6.4^\circ$ in the i CrCL and $128.6^\circ \pm 6.0^\circ$ in the TTA CrCL. The adjustments of the pulley position, aimed at directing the traction force on the proximal femur parallel to the patellar ligament, resulted in a misalignment of $0.6^\circ \pm 2.3^\circ$ and $-1.3^\circ \pm 2.9^\circ$ for the i CrCL and TTA CrCL groups, respectively. Differences in the orientation of the traction force and the patellar ligament were not statistically different, thus maintaining similar force magnitudes. In addition, the stifle joint flexion angles were not statistically different from one other.

A mean lengthening of 0.8% (standard error [SE] 0.3%) was detected between the markers that represented the CrCL in the i CrCL, and a mean shortening of 3.2% (SE 0.6%) was detected between the markers representing the CdCL when compared to the $i_{(p)}$ CrCL. The distance between the markers representing the CrCL was increased by 22.4% (SE 3.6%) in the i CrCL, and the distance between the markers representing the CdCL decreased by 24.7% (SE 3.3%). Loading the stifle resulted in a mean shortening of the distance between the markers of the CrCL of 3.0% (SE

2.3%), and a lengthening of the distance between the markers of the CdCL of 1.4% (SE 2.1%) after performing the TTA CrCL (Fig. 3). In the post hoc analysis by Bonferroni, the i CrCL was significantly different from the $i_{(p)}$ CrCL, i CrCL and post- TTA CrCL. Statistically significant differences were not found between $i_{(p)}$ CrCL, i CrCL and post- TTA CrCL groups.

Discussion

The limbs used in this study were harvested from dogs of varying age and weight. These factors were not found to influence relative ligament lengths, which suggests that the changes observed under different loading states and situations were due to properties inherent in the canine stifle regardless of the size or age of the dog.

The stifles were frozen and thawed in a manner considered to preserve the structural and mechanical properties of the ligaments (23). The jig used in the apparatus assured a stable, defined and reproducible position of the tibia, allowing repeated application of loading under different test conditions. The testing apparatus enabled reproducible radiographic positioning and measurements for each stifle. The fixation of the stainless steel beads that were used as radiodense markers at insertions of the cruciate ligaments proved to be stable and comparable. The beads were visible on the radiographs at original insertion sites, thus permitting measurements on the radiographs in all of the different loading states for the CrCL in both intact and transected conditions. By displacing the pulleys in a craniocaudal direction, the traction force was directed parallel to the patellar ligament, precluding shear forces by the load itself. Additionally, this orientation maintained a constant force, which can be assumed as differences that were observed with the alignment of the forces applied relative to the patellar ligament were shown not to be statistically different. The turnbuckle, which imitated the quadriceps muscle pull, retained the hindlimbs at the desired degree of extension, was also not statistically different. The elasticity of system was adjusted as needed

in order to set up all of the stifles at the same angle of extension for the standardisation of the measurements. The target flexion angle of 130° was chosen because it is close to full functional extension, and CrCL is suggested to be critical for stability of the stifle in extension (16).

The chosen stifle angle of 130° is smaller than the angle of 135° that has been used in other studies (17, 18) and is also less than the mean femorotibial joint angle calculated by Hottinger (24), but it is still larger than the angle at which the shear forces are hypothesized to shift from a cranial towards caudal direction (16). By using the conventionally defined tibial plateau slope, this angle is suggested as being approximately 90° of flexion (16). As long as the joint flexion angle is larger than the crossover point (where there are cranially directed tibiofemoral shear forces), the cranial cruciate ligament is suggested to be under stress. Furthermore, we believe that a difference of 5° from other studies would make little or no difference in demonstrating the efficacy of this model.

The lengths of the CrCL and the CdCL in the $i_{(p)}$ CrCL were defined as 100% to compensate for any variation in the size of the dogs. A preload of 0.8 kg was used to orient the collagen fibres of the ligaments and joint capsule, thus obtaining a stable position for the radiographs. Due to the lever-arm action of the femur, the loading force applied at the proximal femur of 6.8 kg resulted in calculated forces in the tibiofemoral joint of between 21 and 33 kg. These forces may be lower than those arising from some conditions in the *in vivo* state; however we verified the expected sequence of cranial tibiofemoral subluxation after transection of the CrCL, thus confirming that the forces selected in this investigation were of sufficient magnitude to mimic that which occurs *in vivo*. As the cranial shear force was observed to shift in a caudal direction after TTA, we assumed that larger forces would only increase the caudal shear but would not alter the direction of shear.

In vitro studies have been used to investigate the effectiveness of tibial plateau leveling osteotomy (TPLO) for the prevention of subluxation of the stifle with a transected CrCL (17, 25). A similar protocol has re-

cently been used to study the efficacy of TTA (18). The loading scheme used in the present study emulates these studies with a simple, tabletop jig, using only the tibia and the femur (Fig. 1). The aim of this scheme is to load the stifle joint with a pair of forces: the patellar ligament force, denoted by F_{PL} ; and the total joint reaction force, denoted by F_J . Those should be parallel, and they will be, if and only if the traction applied to the proximal femur, denoted by F_{FT} , is adjusted to be parallel to the patellar ligament (if two forces acting on a body are parallel, the third one must be as well, or else the sum of forces would not equal zero, as required for the static equilibrium). The traction force (F_{FT}) on the proximal femur does not simulate a physiological load on the femur; it is simply used to generate an internal stifle reaction, known to exist under loading conditions (25–27) (Fig. 1).

In regard to this experimental design, caution must be exercised when extrapolating the results of an *in vitro* study to *in vivo* situations. In the present study, stifles were loaded and the positions of the insertions of the ligaments recorded by radiography for a single position of flexion and under a single load, arranged to simulate the joint forces presumed to occur *in vivo* under certain conditions of activity. This represents only a fraction of the extremes of a normal range of motion and loading forces. By superposing the medial and lateral femoral condyles, marker positions and displacements were only measured in the sagittal plane, precluding observation of any effects of rotational forces, which are also considered relevant to the function of CrCL. An *in vitro* study of the type presented here has other limitations, most importantly relating to the mechanism of joint movement, which is fundamentally based on co-contraction of flexor and extensor muscles. These are the quadriceps via the patellar ligament, the hamstrings, and the caudal muscles of the crus pulling on the calcaneal tendon. However, even sophisticated dynamic gait analyses do not account for co-contraction (26, 27).

The intact CrCL experienced a distension of only 0.8% under load in this study, which may be due to the fact that the ratio of compression to shear is very high in in-

tact stifles under physiological function (28). Intact stifles in extension experienced a small shortening of the CdCL of 3% under load. This demonstrates that the shear force was directed cranially with respect to the tibia at the degree of stifle extension used, which corroborates suggestions of a radiographic study investigating the angle between the tibial plateau and the patellar ligament (16). In the transected, loaded condition the distance between the CrCL markers was significantly longer and the distance between CdCL markers was significantly shorter than in the loaded intact condition, which is consistent with a subluxation of the femur in relation to the tibia. This finding indicates that the model used replicates the clinical situation, at least in the craniocaudal plane.

A slight shortening of the distance between the CrCL markers after TTA (TTA CrCL) was observed and was attributed to overcorrection. The articular surfaces of the distal femur and proximal tibia are both convex, which renders neutralisation of the cranial shear without creating caudal shear difficult, and probably impossible, in all situations and flexion angles of the stifle. Further studies and long-term postoperative follow-up investigation are required in order to evaluate the extent to which TTA-induced caudal shear in the CdCL is of clinical relevance. A thorough preoperative planning aimed at avoiding overcorrection may be important in averting excessive loads on the CdCL and its possible subsequent degeneration (29).

The experimental method employed here to investigate cruciate ligaments with radiodense markers cannot be reliably used to measure deformation of the intact ligaments. However, we believe that it was adequate for the purpose of comparing the effect of transecting the CrCL and the efficacy of TTA in eliminating craniocaudal shear. The present study demonstrates that, under the selected loading condition and angle of flexion, TTA effectively neutralises cranial subluxation of the tibia and shifts the cranial shear into a caudal direction in CrCL-deficient stifles. Further studies are necessary to access the effect of TTA in relation to internal rotation of the tibia and hyperextension of the stifle.

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