The effect of tibial tuberosity advancement and meniscal release on kinematics of the cranial cruciate ligament-deficient stifle during early, middle, and late stance

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Keywords
Stifle, tibial tuberosity advancement, cranial cruciate ligament, stance, kinematics

Summary
Objectives: To evaluate the effect of tibial tuberosity advancement (TTA) and meniscal release on cranial-caudal and axial rotational displacement during early, middle, and late stance phases in the canine cranial cruciate ligament- (CCL) deficient stifle.
Study design: In vitro biomechanical study.
Methods: Eighteen pelvic limbs were evaluated for the effects of TTA on cranial-caudal displacement and axial rotation under a load equivalent to 30% bodyweight, and under the following treatment conditions: normal (intact CCL), CCL deficient, TTA-treated (CCL deficient + TTA), and meniscal release (TTA treated + meniscal release). The limbs were evaluated in the early, middle, and late stance phases using electromagnetic tracking sensors to determine cranial tibial displacement and tibial rotation relative to the femur.

Results: Transection of the CCL resulted in significant cranial tibial displacement during early, middle, and late stance (p < 0.0001) and significant internal rotation during early (p = 0.049) and middle stance (p = 0.0006). Performance of TTA successfully eliminated cranial tibial displacement in early, middle, and late stance (p < 0.0001); however, the TTA was unsuccessful in normalizing axial rotation in middle stance (p = 0.030). Meniscal release had no effect on cranial-caudal or rotational displacement when performed in conjunction with the TTA.

Clinical significance: Tibial tuberosity advancement effectively eliminates cranial tibial displacement during early, middle and late stance; however, TTA failed to provide rotational stability in mid-stance.

Introduction
Rupture of the cranial cruciate ligament (CCL) is one of the most common causes of pelvic limb lameness in dogs (1). The CCL has three main functions within the joint: preventing cranial translation of the tibia, preventing hyperextension of the stifle, and limiting internal rotation of the tibia (2, 3). Because of the complex nature of the joint, a single effective technique that restores all functions of the CCL has been elusive. The surgical techniques of tibial plateau leveling osteotomy (TPLO) and tibial tuberosity advancement (TTA) were developed with the purpose of functionally stabilizing the CCL-deficient canine stifle during weight bearing by altering the joint geometry (4, 5). However, progressive osteoarthritis following these procedures is common (6–10). While this progression is likely to be multifactorial in aetiology, potential causes include altered joint contact mechanics, abnormal joint biology, and altered joint kinematics (11).

Due to the anatomy of the femorotibial joint surface, a cranially directed shear force is created between the two bones during weight bearing (12). In a CCL-deficient stifle, this shear force is unopposed, resulting in cranial translation of the tibia (12). Functional neutralization of this femorotibial shear force is the goal of tibial osteotomy procedures.

Tibial tuberosity advancement is based on a biomechanical theory that assumes the total femorotibial joint reaction force is parallel to the patellar tendon during ambulation (13). Because of the anatomy of the canine stifle, a shear force is created in a cranial direction during weight bearing. However, the shear force reaches zero when the patellar ligament is perpendicular to the tibial plateau, and the shear force shifts caudally with further flexion of the stifle (13, 14). In the dog, the cross-over point at which the cranially directed force is neutralized and directed caudally has been identified to be 90° of stifle flexion, which is
below the normal weight-bearing angles of the canine stifle (14, 15). The TTA procedure positions the patellar ligament perpendicular to the tibial plateau at a normal weight-bearing angle, resulting in a caudal shift of shear forces during stifle joint axial loading (16). Because of the ability of TTA to neutralize the femorotibial shear force, or direct it caudally, this procedure has been used clinically to stabilize CCL deficient stifles with 90–95% of owners reporting an excellent functional outcome (10, 17, 18).

While TTA has proven to be a successful treatment for dogs suffering from a CCL-deficient stifle, complications such as continued progression of osteoarthritis and subsequent meniscal injury have been noted (17–19). Previous biomechanical studies have investigated the effect of the procedure on cranial tibial thrust at 135° of stifle flexion, and only one study has addressed the ability of the TTA procedure to control axial rotation at this weight-bearing angle (16, 20–22). While 135° approximates the stifle angle at the middle of the stance phase of the gait, the stifle angle has been reported to range from approximately 125° to 145° during stance (15). One study evaluated the effects of TTA on eliminating cranial tibial thrust at multiple joint angles and axial loads (23). However, the effects of TTA on three-dimensional stifle kinematics throughout the entire stance phase of the gait have not been thoroughly evaluated.

The unopposed femorotibial shear force that is present in a CCL deficient stifle not only allows for cranial translation of the tibia, but also predisposes the medial meniscus to injury (4). Because of the relatively high incidence of meniscal injury subsequent to TTA, concurrent meniscal release has been recommended as standard of care (18). However, this procedure is controversial as the meniscus serves several important functions including energy absorption, stabilization of the joint by deepening the articular surfaces of the tibial plateau, joint lubrication, and improvement of joint congruity between the femur and tibia (12). Furthermore, meniscal release has been shown to result in abnormal stress concentrations within the joint, which may predispose to osteoarthritis (24). The effects of the meniscal release on stifle stability when performed in conjunction with TTA have not been evaluated.

The goals of this present study were to further investigate the biomechanical effects of TTA during the early, middle, and late stance phases of gait, specifically its effect on cranial tibial subluxation and axial rotation of the tibia when compared to normal limbs with an intact CCL. Additionally, the effects of a meniscal release on stifle stability when performed in conjunction with TTA were evaluated. Our hypotheses were: (1) cranio-caudal subluxation and axial rotation would be significantly altered in CCL-deficient limbs treated with TTA at stifle angles greater than 135°, and (2) meniscal release following the TTA procedure would lead to significant changes in joint alignment when compared to limbs treated with TTA alone.

**Materials and methods**

**Specimen preparation**

Twenty-two pelvic limbs were collected via coxofemoral disarticulation from 11 adult dogs euthanized for reasons unrelated to this study. The mass of each dog was recorded. Each limb was examined grossly and radiographed to verify skeletal maturity and absence of any signs of pathology. Four limbs were used to establish experimental methods and the data collected from the other 18 limbs were used for statistical analysis in the study. The tibial plateau angle (TPA) was measured from the medio-lateral radiographic images using previously described methods (25). The limbs were wrapped in 0.9% NaCl solution-soaked towels and stored at –20°C until testing. Immediately prior to testing, the limbs were thawed to room temperature and stripped of their musculature. The medial and lateral collateral ligaments, cranial and caudal cruciate ligaments, joint capsule, and medial and lateral menisci were carefully preserved. Throughout the remainder of the testing the limbs were kept moist with saline solution.

The proximal part of the femur was osteotomized transversely 5 cm distal to the most proximal aspect of the femoral head, and then the femoral diaphysis was potted using polymethylmethacrylate within polyvinyl chloride pipe (3.8 cm diameter). The potted limb was loaded into a custom-built mounting bracket that was designed to allow attachment to the loading frame and adjustment of the hip angle during testing (Fig. 1). The overall length of the osteotomized bone, potting material, and mounting bracket were the same as the unaltered limb. After potting, a 1.5 mm hole was drilled transversely through the widest portion of the patella. Thirty-six kg test monofilament nylon leader line was passed through the hole and tied to create a loop. A turnbuckle link extending from an eyelet attached to the most cranial and proximal aspect of the mounting bracket

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was attached to the patellar nylon loop to mimic the quadriceps mechanism. Two 3.5 mm cortical bone screws were inserted into the distal femur at the level of the origin of the gastrocnemius muscle on the caudal femur, and another nylon loop was placed through a 1.5 mm diameter hole drilled transversely through the calcaneus. A turnbuckle link that extended from the calcaneal nylon loop to a nylon loop secured to the tibial metaphysis was tightened against a plastic 'stopper plate' placed against the cut surface of the tibial metaphysis. Prior to testing, a medial mini-arthrotomy and a caudal mini-arthrotomy were performed to facilitate transsection of the CCL and the meniscotibial ligament during testing.

Each limb was mounted in a custom-designed loading frame similar to the frame used in the study performed by Warzee et al. (26). Custom hinges as described by Kim et al. were used to facilitate adjustment of the abduction-adduction angle and allow for unconstrained femoral axial rotation during testing (Fig. 1) (27). The stifle angle was adjusted using the previously placed turnbuckles to either 145°, 135°, or 125°, approximating early, middle, and late stance phase during walking (15). The hip and tarsal angles were simultaneously adjusted during testing to correspond with the phase of gait being tested (15). In the early, middle and late stance phases the hip angles used were 113°, 120° and 135° respectively, while the tarsal angles were 140°, 145° and 160°, as previously described (15). The joint angles were measured with a plastic goniometer with each arm aligned along the central axis of the bone diaphyses proximal and distal to the joint. The paw was allowed to rest on but was not fixed to the base of the loading frame. A textured surface (220 grit sandpaper) was fixed to the steel-loading frame to prevent cranio-caudal paw slippage during testing. One electromagnetic tracking sensor was attached to the lateral aspect of the distal femur and another to the proximal tibia using 4.8 mm Steinman pins and nylon spacers to prevent direct contact with the stainless steel pin. A third sensor was fixed to the cranial aspect of the proximal tibia using a custom plastic mounting bracket that attached to the proximal aspect of the hinge plate and was utilized in measurement of tibial subluxation. This electromagnetic tracking system measures six degrees of freedom in a Cartesian coordinate system (28). The system software allows three-dimensional measurement of the position of the receivers relative to a global coordinate system projected by the magnetic transmitter (28). This tracking system reportedly has a translational resolution and accuracy of 0.1 mm and 0.2 mm, respectively (29). Rotational resolution and accuracy has been reported to be 0.1° and 0.2°, respectively (29).

Testing protocol

To simulate in vivo conditions, a load of 30% of the body weight was applied to the limb during testing (27). The abduction-adduction angle was set by visual inspection prior to testing to approximate even distribution of forces across the femoral condyles. During testing, the axial rotational hinge was left unconstrained.

The limbs were tested in the following sequence of treatment groups: (1) Normal (CCL intact, sham TTA in which the osteotomy was performed but the tuberosity was not advanced) at three stifle joint angles (145°, 135°, 125°), all with the adjusted hip and tarsal angles; (2) CCL deficient (transected CCL, sham TTA) at all joint angles; (3) TTA-treated (transected CCL, TTA advanced to a PTA of 90°) at all joint angles; (4) Meniscal release (transected CCL, TTA, medial meniscal release) at all joint angles. The CCL was transected at its insertion on the proximal tibia through the previously made medial mini-arthrotomy. In the TTA treated groups, the tibial tuberosity was advanced and radiographed until the PTA was verified to be 90° at midstance when using the tibial plateau as a reference. The PTA and stifle angle were measured from a lateral radiograph as described by Dennler et al. (14). For comparison purposes, the PTA using the femorotibial tangent as a reference (PTAf) was also measured from the same lateral film as previously described (14). The advancement required to position the PTA to 90° was measured from the lateral radiographic projection along the caudal aspect of the osteotomy at a point 2 mm distal to the

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**Fig. 2** Radiograph demonstrating cadaveric limb preparation. The tibial osteotomy has been performed, stabilized using a custom hinge plate, and advanced using a metric 4 x 0.7 (4 mm diameter and 0.7 mm thread pitch) machine screw.

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4 Synthes Oscillating Bone Saw, Serial No. 5789; Synthes, West Chester, PA, USA

4 FASTRAK: Polhemus, Colchester, VT, USA
proximal extent of the tibia. This position correlates with correct placement of the TTA cage in clinical cases (18). The medial meniscal release was performed by transection of the meniscotibial ligament through the previously made causal mini-arthrotomy.

Data for cranio-caudal displacement and internal-external rotation of the tibia in relationship to the femur were collected for each treatment group using the electromagnetic tracking sensors. The limbs were positioned in the loading frame, the appropriate joint angles set for the phase of stance being tested, and the coordinate position of the tracking sensors was acquired and recorded for the limb while unloaded. A load equivalent to 30% of the body weight was applied to the limb using the limb-press platform and appropriately sized weights for the limb being tested (Fig. 1). The limb construct was allowed to settle under the load, and the joint angles were readjusted to account for settling of the limb. The coordinate position of the tracking sensors for the loaded limbs was acquired after adjusting these joint angles. This sequence of events was repeated three times for all joint angles evaluated and all treatment conditions, and the mean value for displacement and axial rotation calculated. The values for cranio-caudal displacement and axial rotation of the tibia in relationship to the femur were determined as the relative difference between the coordinate position of the femoral and tibial tracking sensors before and after application of the load.

### Statistical analysis

Data for cranio-caudal tibial displacement and internal-external rotation were statistically analyzed by using PROC MIXED (mixed procedure) with the REPEATED statement and the autoregressive order one (AR 1) covariance structure on the SAS/STAT software package. Efficacy analyses were performed using analysis of variance (ANOVA) with the repeated-measures design. The limbs were evaluated for differences among the four treatment groups (normal, CCL deficient, TTA treated, and meniscal release) and for differences between the stifle angles (145°, 135°, and 125°). The least square (LS) means were used when there was statistical significance to determine the differences between treatments. All statistical comparisons were two-sided using a p-value of 0.05 for the significance level.

### Results

Mean (± SD) body mass of the dogs was 22.8 ± 4.6 kg; mean tibial plateau angle was 23.0 ± 1.9°; mean PTA was 89.8 ± 0.9° at a mean stifle angle of 137.3 ± 2° following TTA; mean PTA<sub>C</sub> was 85.2 ± 1.5° following TTA; mean advancement of the tibial tuberosity required to obtain a patellar tendon-tibial plateau angle of 90° was 14.6 ± 2.4 mm.

The results for cranio-caudal tibial displacement and axial rotation are presented in Table 1. Transection of the CCL resulted in significant cranial displacement of the tibial tuberosity advancement and meniscal release throughout stance.
tibial displacement at 135° of stifle flexion and at varying stifle angles and axial loads (16, 21, 21, 23). Additionally, the results of this study revealed that the TTA failed to consistently eliminate abnormal rotational alignment secondary to a transected cranial cruciate ligament. This is the first report to identify a statistically significant difference in rotational alignment when comparing TTA treated limbs to the intact stifle.

The PTA is highly dependent upon the degree of stifle flexion and a linear relationship between the stifle angle and PTA exists (14, 30). In the normal canine stifle, the patellar tendon becomes perpendicular to the tibial plateau at 90° of stifle flexion, which is a greater degree of stifle flexion than the normal weight-bearing stifle angle (14, 15). Advancement of the tibial tuberosity effectively positions the patellar tendon perpendicular to the tibial plateau at 90° of stifle flexion. A value of 135° of stifle flexion is chosen for surgical planning of TTA because it approximates the middle of the stance phase of the gait (5). However, this fixed position does not take into consideration the stifle angle at any other point during ambulation. The effects of TTA on stifle stability at stifle angles less than 135° is supported by Dennler et al. as further flexion of the stifle beyond the point at which the patellar tendon is perpendicular to the tibial plateau results in caudal tibial thrust (14, 16). This theory was further supported by the findings reported in a recent paper that evaluated the biomechanics of TTA and TPLO at 90° of stifle flexion (31). Therefore, the fact that the stifle was stable in a cranial-caudal plane at 125° of stifle flexion in the present study is not surprising. However, the effect of TTA on stifle stability with extension of the stifle beyond 135° is less clear.

At 145° of stifle flexion, the findings of our study indicated that the TTA procedure as described effectively eliminated cranial tibial displacement. This finding is counterintuitive given the effects of the stifle angle on the PTA and the results of previous studies that report the critical point for neutralization of cranio-caudal tibial thrust as a PTA of 90° at 135° of stifle flexion (14, 16, 30). One explanation for this finding may lie in a flaw in surgical planning for TTA, which uses the slope of the tibial plateau as a reference (5, 18). Recent reports have suggested determining PTA using the tangent to the femorotibial contact point rather than the tibial plateau may more accurately represent the forces acting across the stifle during ambulation and be less affected by stifle angle (14, 23, 32). The critical point for stifle neutrality has been reported to be 110° of stifle flexion when using the tangent method for determining the PTA (14). Therefore, a lesser degree of tibial tuberosity advancement would be necessary to obtain a PTA<sub>CT</sub> of 90° at any given stifle angle. The fact that the stifle remained stable at 145° of stifle flexion in the present study suggests that the advancement of the tibial tuberosity to a PTA of 90° at the stifle angle of 135° may represent over-advancement of the tibial tuberosity when using the tibial plateau as a reference. This theory has been supported by the findings of a recent study that reports cranio-caudal tibial stability to occur following TTA to a PTA of 91.1° and 98.3°; and PTA<sub>CT</sub> of 88.4° and 91.1° when the stifle is at 135° and 145° of flexion, respectively (23). The same study also found that PTA<sub>CT</sub> varied less during stifle flexion as previously suggested by Dennler, and that the PTA at 135° of stifle flexion was significantly greater that the PTA<sub>CT</sub> at the same angle (14, 23). The latter finding was supported in the current study and in the work done by Dennler (14). However, Hoffman et al. also suggest that surgical planning using the femorotibial tangent as a reference may require TTA with a target PTA of <90° (23). Further biomechanical research is warranted to determine whether using the femorotibial tangent as a reference is more reliable, and to determine the most appropriate target angle when using this method.

In the present study, the tibial tuberosity was advanced a mean of 14.6 mm in order to obtain a PTA of 90° at a stifle angle of 135°. This is a greater magnitude of advancement than is typically performed clinically in similar size dogs with similar tibial conformations (5). However, in a comparable biomechanical study by Kim and others, an advancement of 13.5 mm was necessary (20). Advancement of the tibial tuberosity to the optimal PTA in some of these cases would not be possible with the currently available implants. In some clinical cases, the TTA procedure may result in under-advancement of the tibial tuberosity, which may result in suboptimal neutralization of cranial tibial thrust. Continued cranio-caudal instability may result in complications seen with TTA, particularly the high incidence of subsequent meniscal injury (18). Alternatively, the tibial tuberosity may be advanced to a lesser degree if the femorotibial tangent was used as a reference during surgical planning, and it may explain the apparent clinical success with the TTA procedure (10, 17, 18, 20, 32). However, as noted by Hoffman et al., TTA using the femorotibial tangent as a reference may require advancement to a target PTA of <90°. Apelt and others reported a lesser degree of advancement (10.2 ± 3.7°) required to reach the ‘critical point’ in their study (16, 23). While the exact reason for the discrepancy among studies is unclear, the authors feel this is probably multifactorial and may include conformational differences amongst the dogs (such as size of the tibia, shape of the tibial crest and tibial plateau slope) as well as methodology differences between studies. In the study by Kim as well as the current study, a predetermined PTA was set (PTA of 90° at 135° of stifle flexion) and the limbs were evaluated (20). The study by Apelt and others differed slightly in that the TTA was sequentially advanced under load until the stifle reached a ‘critical point’ defined as the position one revolution before joint instability occurred. This occurred at a PTA of 90.3 ± 9° using the tibial plateau as a reference (16). Given the wide range of PTA reported and because the PTA data in relationship to the femorotibial tangent was not reported, the relationship between the ‘critical point’, tibial plateau, and femorotibial tangent is unclear (16). Further research is needed to determine the biomechanical effects of performing TTA utilizing the femorotibial tangent as a reference to define the PTA.

Only one previous study has documented the effects of TTA on rotational stability during weight bearing (20). In that study, transection of the CCL resulted in significant internal rotation of the tibia and that rotational change was partially cor-
rected following TTA (20). In the present study, significant internal tibial rotation after transection of the CCL at 125° and 135° of stifle flexion was seen, but the values were of a small magnitude (<10°). Unlike the previous study, TTA in the current study did not consistently correct the internal rotation caused by CCL transection. Considering the small magnitude of rotation seen in this study, we question the clinical significance of this finding. Similarly small values for internal rotation during ambulation following cranial cruciate loss were noted in an in vivo study by Tashman and others (33). These studies must be compared with caution due to the multiple active stabilizers of the joint that are present in vivo that are not present in the current cadaveric study. However, the findings of the current study bring into question the clinical significance of tibial rotation as a cause of instability after CCL rupture and tibial osteotomy procedures, at least during straight-line walking.

Previous reports have documented subsequent meniscal injury rates as high 21.7% following TTA (18). For this reason, routine meniscal release at the time of initial surgery has been recommended (18). The medial meniscus has been shown to aid in stabilization of the CCL deficient stifle via a wedge-effect of the caudal pole of the medial meniscus that helps prevent tibial subluxation (34, 35). A previous study evaluated the effects of medial meniscal release when performed in conjunction with TPLO (35). In that study, medial meniscal release resulted in a significant increase in cranial tibial displacement in the CCL deficient stifle, but TPLO appeared to negate this effect by neutralization of cranial tibial thrust (35). In the present study, performance of medial meniscal release via transection of the meniscotibial ligament failed to result in a significant increase in cranial tibial displacement when performed in limbs treated with TTA. While the effects of meniscal release on rotational alignment in the canine stifle have not been evaluated, rotational alignment consequences of tears of the posterior root of the medial meniscus in humans have been studied (36). In humans, a tear of the posterior root of the medial meniscus (analogous to the meniscotibial ligament in the canine stifle) has been reported to result in significant alterations in axial rotational alignment (36). The present study is the first to report the effects of medial meniscal release on stifle rotation in the canine stifle when performed in conjunction with a tibial osteotomy procedure performed for the treatment of CCL deficiency. In the present study, medial meniscal release failed to result in significant internal rotation of the tibia when performed in conjunction with TTA. As with subluxation, this suggests that the TTA unloads the medial meniscus, and at least partially eliminates its role as a secondary restraint within the joint (35). While releasing the meniscus did not result in statistically significant alterations in axial rotational alignment, the limbs did tend to be slightly more internally rotated following meniscal release, especially during early and mid stance. These results may reflect an error such as low statistical power and further studies with additional limbs may reveal a different result. Additionally, the increased stress and friction generated between articular surfaces following meniscal release and elimination of its role as a spacer within the joint may further limit alterations in limb alignment as previously suggested by a similar study in human cadavers following meniscectomy (37, 38).

Given the in vitro nature of this study and the inherent limitations associated with the methodology and complexity of the limb-press model, these results should be interpreted with caution. While a similar limb-press model has been used in previous studies to evaluate the biomechanical effectiveness of tibial osteotomy procedures for the treatment of CCL deficient stifles, it does not accurately represent all forces acting upon the stifle during ambulation, and the effects of active stabilizers of the stifle are not evaluated (16, 20, 23, 26, 27). Furthermore, this model only takes into consideration straight-line loading of the limb. The effects of other activities such as turning, jumping, and sitting on stifle stability were not evaluated.

We evaluated the stifle at the angles 145°, 135°, and 125°, each of which have previously been reported to represent the early, middle, and late stance phases of the gait (15). However, more recent studies have suggested a degree of stifle extension exceeding 150° at the beginning of the stance phase and postural variations among breeds (39, 40). Furthermore, a slight discrepancy exists between the methodology of determining joint angles utilized in our study and those determined from in vivo kinematic measurements (15, 39). The joint angles in the current study were measured and set as described by Dennler and others and the points of reference do not exactly coincide with those utilized by the in vivo kinematic studies (14, 15, 39). Taking this fact into consideration, there is a chance that the present study slightly underestimated the degree of joint extension for the stifle and this represents one limitation to our study. While the effect of this slight underestimation is uncertain, it is probably not significant as the difference is small and the stifle angle is not the only factor affecting limb position during ambulation (15). The hip and tarsal angles were simultaneously adjusted to approximate the forces acting across the joint during early, middle, and late stance. Future studies with the stifle in a greater degree of extension are warranted.

The current study sought to mimic the clinical scenario in which the limb was positioned at midstance and TTA performed to a PTA of 90° without regard to the effect of stifle flexion-extension on PTA. Because radiographic projections were not performed with the limb in extension and flexion beyond midstance, the actual PTA at these limb positions cannot be reported. This represents a limitation to the current study design as these values cannot be used to help explain the biomechanical results obtained. However, the clinical implications of the results remain unchanged and, as previously noted, the relationship between stifle flexion-extension and PTA has been previously described (14).

In this study, an electromagnetic tracking system was used to track the position of the tibia relative to the femur in a Cartesian coordinate system. This and similar tracking equipment has been validated for use in similar human and ovine studies (29, 41, 42). Given the high sensitivity of the tracking equipment to positional change, its use in this study is justified. Furthermore, nylon leader line was used to attach the turnbuckles to the patella, femur, and...
calcanes to mimic the quadriceps and gastrocnemius musculature, respectively. This may represent one potential limitation to the study design as nylon can ‘creep’ while under load and potentially affect the position of the limb construct. However, the significance of creep in the current model is probably insignificant given the time required for data acquisition using the electromagnetic tracking system (0.004 seconds²). Additionally, joint angles were determined using goniometry. Goniometry has been proven to be as accurate as radiography in measuring joint angles (43). However, the use of goniometry may have led to slight variation of joint positioning throughout testing which could potentially affect the results. Furthermore, because goniometry was used as the sole means to adjust the joint angles during testing, an actual value for the angles tested cannot be reported. However, given the results of the previous studies, a standard deviation of no more than ±3° was expected for joint angular adjustments using goniometry and its use in the current study is justified (43).

The effects of a medial meniscal release on cranio-caudal stifle stability when performed in the normal stifle, CCL deficient stifles, and TPLO treated stifle have been previously reported (35). In this study we only evaluated the effects of a medial meniscal release on cranio-caudal and rotational stability only in stifles previously treated with a TTA. While this limits our ability to make conclusions regarding the contribution of the medial meniscus to stifle stability in the normal and CCL deficient limbs, it does provide evidence of the effects of a meniscal release on stifle stability when performed in conjunction with TTA.

Conclusions

Tibial tuberosity advancement effectively eliminates cranial tibial displacement during early, middle, and late stance as depicted in this in vitro model. Tibial tuberosity advancement failed to consistently restore the rotational alignment of the stifle, although the magnitude of tibial rotation noted in this study is of questionable clinical significance. Release of the medial meniscus did not result in any significant cranio-caudal or rotational joint displacement when performed in conjunction with the TTA. Further in vivo and in vitro research is warranted to evaluate the biomechanical effects of TTA and meniscal release on tibial translation and axial rotational using the femorotibial tangent as a reference in surgical planning, with further extension of the stifle beyond 145°, and under loading conditions other than straight-line walking.

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Conflict of interest

None declared

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