Biomechanical comparison of a standard fabella-tibial suture and lateral sutures placed between quasi-isometric points for the treatment of cranial cruciate ligament rupture in feline stifles.

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Introduction

Cranial cruciate ligament rupture (CCLR) is detected less frequently in feline species compared to canines and humans. [1] The difficulty in detecting lameness in feline species and the spontaneous resolution without surgery in a proportion of felines with cranial cruciate ligament tears may contribute to this relatively low prevalence. [1,2,3]

Whilst some cats may return to an acceptable activity level without surgery, the instability caused by the CCLR is frequently addressed surgically. Furthermore surgical stabilization has been suggested to reduce the incidence of meniscal tears in this species.[1,4] Despite the recent advances in the surgical management of cruciate disease, lateral suture stabilization (LSS) remains one of the most common methods used to stabilize the CCLR in the feline stifle. [1,2,5,6] The kinematics of the hind limb are complex with multiple forces thought to alter the contact dynamics of the cruciate deficient stifle joint and thus contributing to the progression of osteoarthritis.[5,7] The ultimate goal of the lateral suture technique relies on the successful neutralization of these forces until secondary peri-articular fibrosis occurs. [8,9,10,11]

Critical aspects have been identified for the placement of the implant with lateral suture techniques.[9] The placement of the suture between quasi-isometric points has an important role by minimising changes in suture tension during stifle range of motion and thus maintaining joint stability. [2,11-22]

The results of a recent cadaveric study[12] evaluating the quasi-isometric points for the placement of a lateral suture in feline stifles, revealed that the most quasi-isometric points were located between the centre of the fabella and between tibial points immediately cranio-proximal to the extensor groove, and caudo-proximal to the insertion of the patellar tendon. However the authors have had some concerns from clinical cases with the laxity of the fabella-femoral ligament in cats, and have considered whether a suture anchored around this sesamoid bone would provide a suitably stable and secure attachment point.

The purpose of this study was two-fold: firstly to determine whether a suture anchored to suture screws[a] at quasi-isometric points would offer superior stabilization to the standard fabella-tibial suture when addressing the CCLR in the feline stifle; and secondly to compare surgical stabilization techniques with the intact stifle joints.

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[a] Veterinary Instrumentation Ltd., Sheffield, UK

Material and Methods

Hind Limb Specimens

Paired hind limb specimens were obtained from six skeletally mature cats of unknown breed, weighing between 3 and 6 kg, free of locomotor deficits, and euthanized for reasons unrelated to this study. Limbs were harvested by disarticulation of the coxofemoral joint. Each stifle was then palpated to confirm an intact cruciate, and manipulated through its full range of motion. Specimens were wrapped in saline (0.9% NaCl) solution soaked gauze and stored at -20°C.

Specimen Preparation

Limbs were thawed to room temperature 24 hours prior to the experimental day and tissues were kept moist by spraying isotonic saline (0.9% NaCl) solution throughout testing.

Careful dissection of the soft tissues was performed with preservation of the muscles inserting around the stifle, collateral ligaments and joint capsule. With the aid of a hypodermic needle and calipers, anatomical landmarks were identified and marked by insertion of small metal spheres (1mm diameter, chrome steel ball, Simply Bearings Ltd, Lancashire, UK) in the distal femur [proximal to the trochlear ridge (F)] and the proximal tibia [insertion of the patellar tibial ligament (T)]. (Figure 1)

Once the specimens were marked, a Steinmann pin[a] was introduced into the intra medullary (IM) canal of both the femur and tibia until the pin tip engaged the metaphyseal bone. The specimens were placed in a mounting set with the proximal end of the femoral pin firmly fixed to a wooden cube that in turn was

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secured to the mounting set with the stifle centre of motion perpendicular to the wooden board. The distal end of the tibial pin and the stifle joint were not restrained allowing cranio-caudal and proximo-distal translation and rotation around its own axis. (Figure 2)

<u>Loading specimen</u>

The tibial pin was loaded using a custom-made adapter made of stainless steel and attached, via leadscrew mechanism, to a digital force gauge[b] used to measure the axial load (Figure 2). The force gauge was fixed in a set position by placement of two screws at the base of the digital force gauge. The custom made adapter was designed with a tubular entrance to accept the tibial pin, and a leadscrew mechanism so that relative rotation of the two halves of the mechanism caused a change in length of the arrangement and hence a change in the axial load applied. Displacement of the tibia relative to the femur was assessed after application of 20 and 60 N (±2 N) of load along the tibia. The stifle constructs were loaded at three different joint angles; 75°, 130° and 160°. The joint angles were confirmed using a manual goniometer (± 0.5°)[a] as previously described.[12]

<u>Mechanical testing</u>

Loading of the tibia was performed for five different joint arrangements:

1. Intact cranial cruciate ligament (iCrCl); stifle joints were firstly tested with an intact cranial cruciate ligament.

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[b] Digitales Kraftmessgerat PCE-FM200, PCE Gmbh, Meschede, Germany

2. Transected cranial cruciate ligament (tCrCl); via medial mini-arthrotomy[10] stifle joints were subsequently explored and the cranial cruciate ligament transected.

3a. Fabella-tibial suture technique (SFT); From a proximal to distal direction the suture (monofilament nylon leader, 50lb)[a] was passed around the fabella and from a lateral to medial to lateral direction the suture was passed under the patellar tendon and then through a drill hole (1.2mm diameter) created 6 millimetres (mm) distal and caudal to the proximal insertion of the patellar tendon.[23] The suture was then secured with a metal tube crimp as described below.

3b Femoro-tibial suture technique 1 (FTS1); with the aid of a hypodermic needle and a ruler, the most caudal aspect of the bone just proximal to the joint capsule and femoral condyle was marked with the drill start point, a suture screw (cortical 2.0mm x 10mm) was then placed in this location in the lateral femoral condyle, as caudal as possible while still engaging in sufficient bone to maximise screw thread purchase and avoid breakout through the caudal cortex to avoid the potential for screw loosening and pull-out . A second suture screw (cortical 2.0mm x 10mm) was placed in the proximal tibia 6 mm distal and caudal to the proximal insertion point of the patellar ligament. The suture was then secured with a metal tube crimp as described below.

3c Femoro-tibial suture technique 2 (FTS2); A similar anatomical location to FTS1 was used to place the suture screw in the distal lateral femur. The tibial screw was placed cranial to the proximal aspect of the extensor groove of the long digital extensor (Figure 1). The suture was then secured with a metal tube crimp as described below.

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With the stifle held at 100° of flexion, the suture was tensioned with a force of 20N (measured using a digital force gauge[b] attached to one strand) and secured with a single crimp device[a] (10mm),[24]

For each stifle joint, the order at which the three stabilization techniques (3a,b,c above) were performed was randomly chosen. This was achieved by selecting one out of the six possible combinations previously described from an envelope.

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Biomechanical testing of suture and anchor arrangement

A preliminary test was performed to confirm that the suture-anchor arrangement would withstand the forces applied to the construct without changing the biomechanical properties. Three samples of the same monofilament nylon leader suture used in the experiment and with an initial length of 30mm were tested in a similar arrangement as used for the FTS1 and FTS2 stabilization techniques. The suture screws were held in wedge grips with their axes perpendicular to the suture and loading direction while the eyelets were aligned to lie in the same plane as the suture. Loading was performed on a load frame[c] with a strain rate of 0.08 mm/sec. The slope of the force-displacement response up to an extension of 2 mm was taken as the stiffness of the arrangement. Tensile load, elongation relative to the initial length of the suture and stiffness was measured. Data was collected using software.[c] The same test arrangement and strain rate was used to perform load-unload cyclic tests. Load cycles were performed in a test at increasing maximum extension and corresponding peak load, apply two cycles of loading at each maximum extension of 2, 3 and 4 mm,

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[c] Instron Bluehill 5584, Instron Ltd., High Wycombe, UK

with failure occurring during the cycle with a target maximum extension of 5 mm.

Radiographic and geometrical analysis

For each intact stifle joint (iCrCl), a single unloaded lateral radiograph[d] was taken with the stifle constructs positioned at 75°, 130° and 160° angles. For each loading stage, lateral radiographs were taken with stifle joints positioned at 75°, 130° and 160° angles and loaded under 20 and 60N forces.

Image analysis was used to assess tibial displacement and presumed suture elongation. Landmarks on the images were identified and located with the help of bespoke image analysis program.[e] The location of the centrelines of the femoral and tibial IM pins and positions of the markers F and T were identified and used to define the overall in-plane motion of the tibia (t') relative to the femur (f'). These points were located as the points on the centrelines of the IM pins closest to the corresponding markers F and T (so that the lines f'-F and t'-T were perpendicular to the corresponding intra medullary pin). Cranio-caudal and proximal-distal movement of the tibia (t') point relative to the fixed femur (f') point was calculated along and perpendicular to the long femur axis at 75°, 130° and 160° stifle angle, respectively (figure 1). For the stabilization techniques, two additional points were identified as the suture origin (S1) and insertion points (S2). (Figure 1) The variation in distance between S1 and S2 points for the 3 angles of joint range of motion were calculated as the length ratio relative to the length measured at 160° of joint angle. Relative movement

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[d] Celtic SMR Ltd, NOVA 30KW High Frequency Mobile, Pembrokeshire, UK

[e] Matlab version R2011b, Mathworks, Natick, MA, USA.

between points f’ and t’ and points S1 and S2 were calculated, all as projected onto the sagittal plane.

The specimens were prepared and tested by two investigators (authors RDS and NR) on different days.

Statistical analysis

Statistical analysis of the changes in suture lengths (S1-S2), movement in the proximo-distal axis, and movement in the cranio-caudal axis as it varied with joint orientation, load and ligament status (intact, transected or stabilized) were analyzed using a multi-level repeated measures ANOVA. Post-hoc pairwise comparisons, using two-sided t-tests, were undertaken with adjustments for multiple testing in order to interpret significant ANOVA results. The level of significance was set at $P<0.05$.

Results

The pre-study trial analysing the tensile strength and stiffness of the suture revealed an elongation of 0.7 and 2 mm for the 20 and 60 N applied loads, respectively with the force-displacement response linear up to an extension of 2mm. The tensile load showed a typical visco-elastic response to a series of load-unload steps up to crimp failure.

The overall pattern of movement of point t' relative to the fixed point f' was calculated from the mean distance averaged over the six specimens, illustrated in figure 3. In addition to this overall pattern of movement, changes in the mean distances between t' and f' relative to the intact joint at 0 N forces were calculated following transection of the ligament and with different stabilization types at the different angles and loads, Figure 4. (Table 1 and 2)

Effect of load on the intact cranial cruciate ligament

Analysis of the intact stifle joint showed that in the proximo-distal direction there was no statistical significance between loads applied to the construct (p=0.5). Analysis of displacement in the cranio-caudal direction showed statistically significant differences between the 0 and 20 N load cases (p<0.01) with cranial displacements of 0.34 mm (±0.08) and 0.8mm (±0.01) for 75^{0} and 160^{0} angle respectively and a caudal displacement of 0.2mm (±0.05) for 130^{0} angle. Statistical significance was also found between 0 and 60 N (p<0.01) with cranial displacement of 0.8 mm (±0.01) and 1.1mm (±0.2) for 75^{0} and 160^{0} angle respectively and caudal displacement of 0.3mm (±0.35) for 130^{0} angle. No statistical significance was found between 20 and 60 N loads (p=0.1)

Comparison between the intact and transected cranial cruciate ligament

A comparison of the changes in proximo-distal and cranio-caudal movement in the stifle joints before and after ligament transection found statistical significance associated with ligament transection (p<0.01) with the t' point moving distally and cranially relative to the f' point. No statistical significance was found between 20 and 60 N loads in the proximo-distal direction (p=0.1) but there was statistically significance between 20 and 60 N loads in the cranio-caudal direction (p<0.01) with the relative distance of t' point to f' point increasing approximately 2.5mm (±0.5) cranially for the three different angles tested.

Comparison between the intact cranial cruciate ligament and the three stabilization methods

An analysis of change of measurements when comparing the three stabilization techniques to the intact stifle joint found statistically significant differences in the proximal direction between the intact cranial cruciate ligament and the SFT technique (p=0.04), with the distance between t' relative to f' decreasing approximately 1.7mm, 0.4mm and 0.2mm for 75°, 130° and 160° joint angles, respectively; and between the intact cranial cruciate ligament and FT2 technique (p=0.03) with distance between t' relative to f' decreasing approximately 1.3mm, 1.3mm and 0.5mm for 75°, 130° and 160° joint, respectively. In the cranio-caudal direction no statistical significant differences were found between stabilization techniques and intact cranial cruciate ligament (p=0.2). Comparison of the three methods of stabilization, to each other's, found no statistical significant differences in the proximo-distal and cranio-caudal directions ($p>0.05$). Comparisons between 20 and 60 N loads found statistical significance in the cranio-caudal direction but not in the proximo-distal direction with a cranial displacement of approximately 0.5mm (±0.5), 0.4mm (±0.3) and 0.2mm (±0.3) for 75°, 130° and 160° degrees, respectively ($p=0.02$).

Variation in the distance between S1 and S2 points

Results from the variation in distance between S1 and S2 (S1-S2) showed that there was no statistically significant changes in the relative length between 20 and 60 N loads but there were significant differences in length for 75° compared to 130° ($P<0.01$) and 160° ($P=0.02$) with an increase in suture length of ±0.8mm and ±0.5mm, respectively. (Figure 5).

Discussion

Image analysis was used to evaluate the stifle joint stability and change in distance between suture screws placed in quasi-isometric points. Six cadaveric feline stifles with and without CCLR and three methods of stabilization were tested. In our study it was clear that the cruciate deficient stifle joint behaved significantly differently from the normal stifle joint. The three methods of stabilization tested provided similar joint stability in the cranio-caudal saggital plane comparable to the intact cruciate ligament, whereas in the proximo-distal direction there were small but significant differences between the intact joints and SFT and FT2 techniques. No statistically significant differences were found between the different stabilization techniques.

Fabello-tibial sutures remain the most commonly accepted method of stabilization for the CCLR in the feline stifle, and several co-dependant factors have been identified that contribute to the success of this surgical technique.[9] Despite the popularity of quasi-isometric points for the placement of fabella-lateral sutures in dogs and cats,[12,13,14,25] there are few biomechanical studies that compare different anchorage points in the lateral stifle joint through the range of motion.[13,26-30] In a recent feline cadaveric study,[12] paired points located between the centre of the fabella and proximo-cranial tibia provided the most quasi-isometric points for the placement of a fabella-tibial suture. In that study no correlation was made between quasi-isometric points and stifle joint stability. In the present study, three different arrangements of lateral sutures were tested and the results showed similar behaviour in the cranio-caudal direction but not in the proximo-distal direction where the two techniques with insertion points distal to the most quasi-isometric points previously reported by the Sousa et al[12] resulted in significant differences when compared to the intact cranial cruciate ligament.

Stifle joint stability is defined as minimal and controlled degree of cranial-caudal, proximo-distal, rotational and medio-lateral motion.[31] To our knowledge very few studies have reported an objective method to evaluate cranial draw and joint stability[26-30,32] and no correlations have been made with the clinical outcome. The multiplanar motion of the stifle joint is complex and stability in a single plane does not constitute normal kinematics. During the stance phase of intact stifle joints, the cranial translation of the tibia is followed by an internal rotation of the tibia, a phenomenon also known as "screw-home mechanism". [9,10] From our study, it was clear that transection of the cranial cruciate ligament resulted in a significant cranial and distal displacement of the tibia relative to the femur, but no conclusion could be made regarding the rotation and medial-lateral translation, as those movements could not be measured with this testing method. Tension applied to the suture at the time of securing the prosthesis was based on published guidelines in which joint laxity, suture slack and draw were eliminated from the stifle joint without compromising range of motion.[33] We applied a 20 N force at the time of securing the suture. Whether a 20 N force represents the ideal tension is unknown. Further studies would be needed to correlate suture tension, stifle joint contact mechanics and clinical significance. Previous studies concluded that variations in the fabella-tibial suture tension are inherent to the individual surgeon and between surgeons.[34] Therefore suture tension in our study was standardized in all the specimens using a force gauge with the suture secured with a single crimp device. In the absence of truly isometric points for the placement of a lateral suture, the joint angle at the time of securing the suture may influence the laxity of the prosthesis through the joint range of motion[13], although the clinical consequences are not known in the feline. To the authors knowledge there is no current literature on felines stifles regarding the ideal joint angle at the time of securing the suture. Thus, results from the canine literature were extrapolated, at which 100° of flexion has been suggested as the ideal angle to secure the suture.[13]

In cats, the peak vertical force acting on the normal hind limb is reported to be around 50% of the static body weight at walk pace, whereas in dogs it seems to be slightly higher, at 60-70%.[31,35] Based on these reports specimens in this study were tested at approximately 20 and 60 N forces. While a 20 N force simulated the expected peak vertical force of an average cat of 4kg body weight, 60 N forces represented the peak vertical force of a cat with a similar body weight given unrestricted freedom. We found significant differences between 20 and 60 N in the cranio-caudal direction but not in the proximo-distal direction.

Despite the inherent risk factors associated with the use of bone anchors[26] they have the potential to minimize the other risks associated with placement of a lateral suture around the small fabella in the cat.[12] Alternatively, smaller diameter suture materials and a smaller radius and thinner needle could improve the placement of the suture around the fabella. Based on the present findings, the use of suture anchors placed in the caudal aspect of the lateral femoral condyle is comparable to the femoro-fabellar ligament as an anchor point.

While in dogs, it has been demonstrated that lateral sutures stabilised with suture anchors provide superior load-to-failure, stiffness and load-to-yield, compared to sutures anchored around the femoro-fabellar ligament,[26,36] in felines there is no literature regarding the use of bone anchors in the lateral femoral condyle and proximal tibia. In the current study, suture screws located in the proximal tibia were placed slightly distal to the most quasi-isometric points previously identified by De Sousa at al[12]. In that study, small metal spheres were used to identify the anatomical locations instead of suture screws, which accounts for some of the anatomical variation between studies. The use of suture screws of a smaller diameter could have improved the placement of the screws in a more proximal tibial location.

Results from the pre-study trial analysing the strength and stiffness of the suture revealed that the response was linear in the range of interest up to an elongation of 2 mm, with a stiffness of 30 N/mm and without the suture implant loosing elasticity. There was relatively little change in distance between anchor points placed at the origin and insertion of the lateral suture (maximum ± 2.5%) under the action of applied loads. These changes in distance between the anchor points corresponded to a change in length of the suture smaller than 1.5mm. Further studies could be performed testing cyclic loading and load-to-failure of the bone anchors and the femoro-fabellar ligament as failure can also result from weakness caused by repetitive loads lower than those representing the maximum pull-out strength.

Various suture materials have been proposed for use in the lateral suture technique.[37,38] In the present study, monofilament nylon leader suture was used.

This material is stiffer than monofilament nylon fishing suture and carries a lower risk for infection when compared to braided materials.[39] For this reason, it is the authors' preferred implant for CCLR suture stabilisation technique.

The results from our study could not be compared to previous canine biomechanical studies as differences in testing protocols and equipment designs prevent direct comparisons between them.

Several limitations of our study are acknowledged. This biomechanical study does not account for all the "*in vivo*" musculoskeletal forces comprising complex joint motions. For example, the muscles and ligaments do not behave as they would in a living animal and the impact of the adjacent joints was not replicated, possibly affecting the overall performance of the stifle joint when stabilised by different methods.

In our study, forces were applied along the anatomical axis of the tibia and were unidirectional and uniplanar. These loads are not representative of "in vivo" loads and thus it could limit our conclusions and neglect the truly cranio-caudal and proximal-distal displacement of the tibia relative to the femur. A more sophisticated custom-made device (eg. robotic system[40] or electromagnetic tracking system[41]) would be required to control and understand the movement that occurs during the full range of stifle joint motion.

Similarly, radiographic interpretation of a three dimensional structure from a uniplanar image has limitations. Multiple orthogonal views would have been required to document multiplanar moments within the stifle joint.

Biomechanical testing of joints is complex and each step has to be carried out with care to ensure that each specimen is tested in a similar manner. “*Ex-vivo*” studies allow us to deepen our understanding as to how joints function under load and with an understanding of the limitations of the experiment the mechanical setup can be improved so that it approximates the “*in vivo*” situation.

In summary, we have demonstrated that lateral sutures placed with suture screws at quasi-isometric points performed better than SFT and FTS2 sutures in the stabilization of CCLR in cadaveric cats stifles in the proximo-distal plane. Further studies are required to test the holding strength of the bone-anchor interface in the lateral femoral condyle and elasticity of the femoro-fabellar ligament.

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Figure 1. Lateral radiographs showing the location of the centrelines of the femoral and tibial intra-medullary (IM) pins and the location of the metal spheres in the distal femur (F1) and proximal tibia (T1) from which f' and t' points were obtained to define the overall in-plane motion. A, fabella-tibial suture technique (SFT) anchored around the fabella bone and a bone tunnel created 6 mm distal and caudal to the insertion to the proximal insertion point of the patellar ligament; B, femoro-tibial suture 1 (FTS1) with cortical suture screws placed in the caudal aspect of the lateral femoral condyle and proximal to the joint capsule and 6 mm distal and caudal to the proximal insertion point of the patellar ligament; and C, femoro-tibial suture 2 (FTS2) similar to B but with tibial cortical suture screw placed cranial to the proximal aspect of the extensor groove. Note, the green dots corresponding to the suture origin (S1) and insertion (S2).

Figure 2. Mounting set (lateral view). The femoral intramedullary (IM) pin can be seen firmly attached to a wooden cube that in turn is attached to a large wooden board. The tibial IM pin is unrestrained allowing cranio-caudal, proximo-distal and rotational movement throughout the range of motion. A custom made stainless steel tube can be seen coupled to a digital force gauge through which axial forces are applied to the IM tibial pin and counteracted by the presence of two screws securing the base of the force gauge.

Figure 3. Relative movement of the tibia to the fixed femur at two axial forces (20 and 60 N); three angles of range of motion (75°,130° and 160°) and at five different joint conditions tested (iCrCl, tCrCl, SFT, FTS1 and FTS2).

Figure 4. Mean ± SD cranio-caudal (A) and proximal-distal (B) displacement of the tibia (t') to the femur (f') relative to the reference intact joint at 0 N force. Relative differences were expressed between the intact (iCrCl), deficient (tCrCl) and three stabilised techniques (SFT; FTS1 and FTS2) at 75°, 130° and 160° joint angle and tested at 20 and 60 N loading forces.

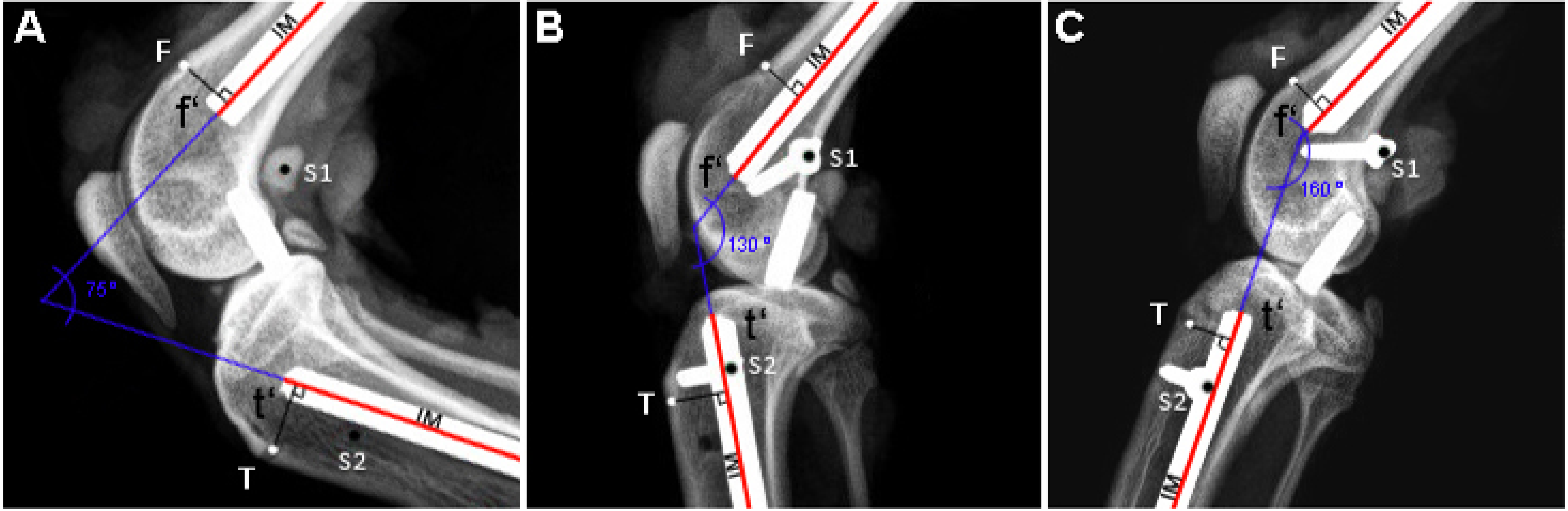
Figure 5. The mean ratio (+/- SD) of the length of S1 –S2 measured at 75 and 130 degrees relative to the length measured at 160 degrees. A change of 5% corresponds to a change in length of approximately 1.5mm.

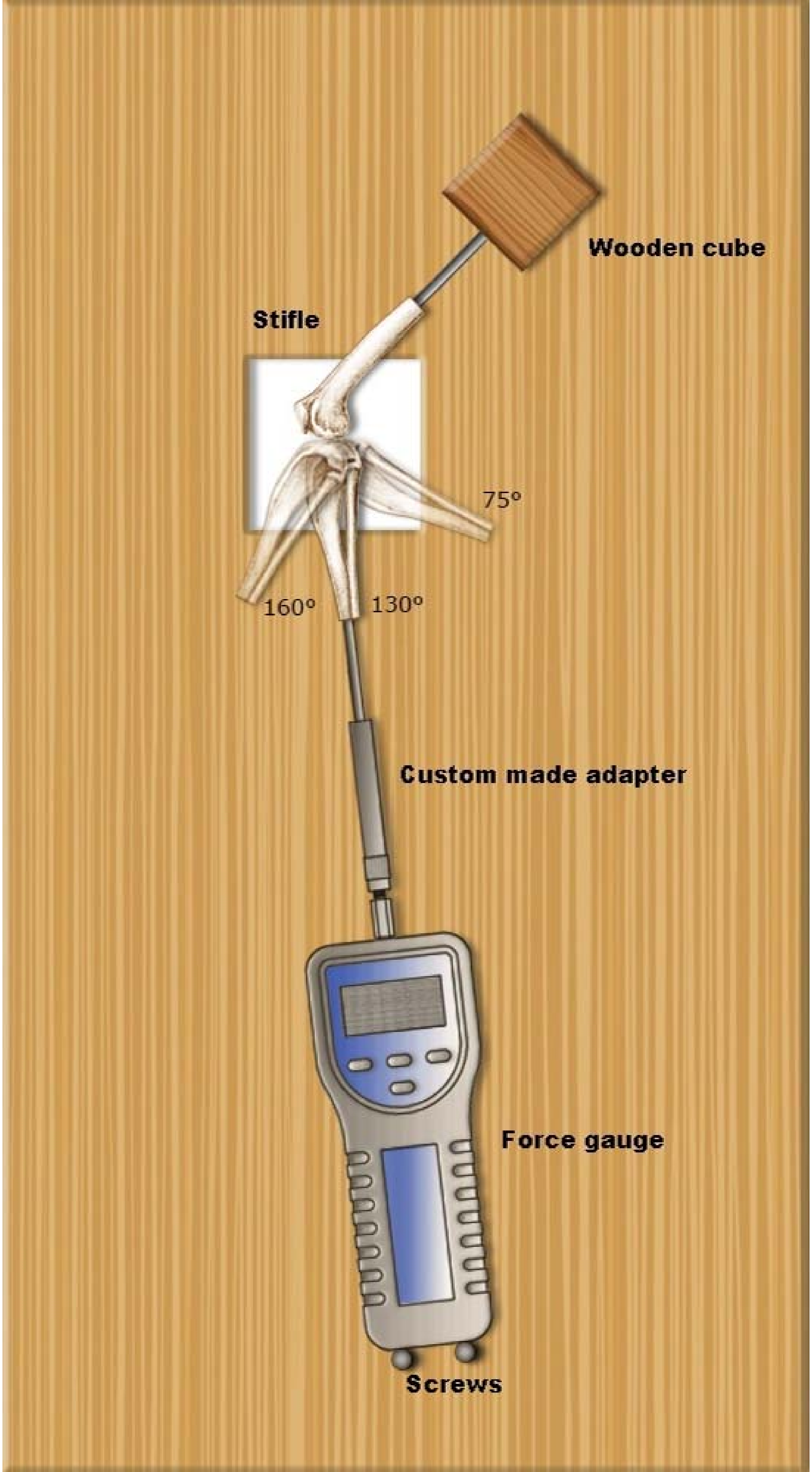
Table 1. Proximo-distal movement of the tibial t’ point relative to the femoral f’ point. All distances calculated relative to the intact joint at 0 N forces. Means ± SD during loading, stages, and angles tested. (Results expressed in millimeters).

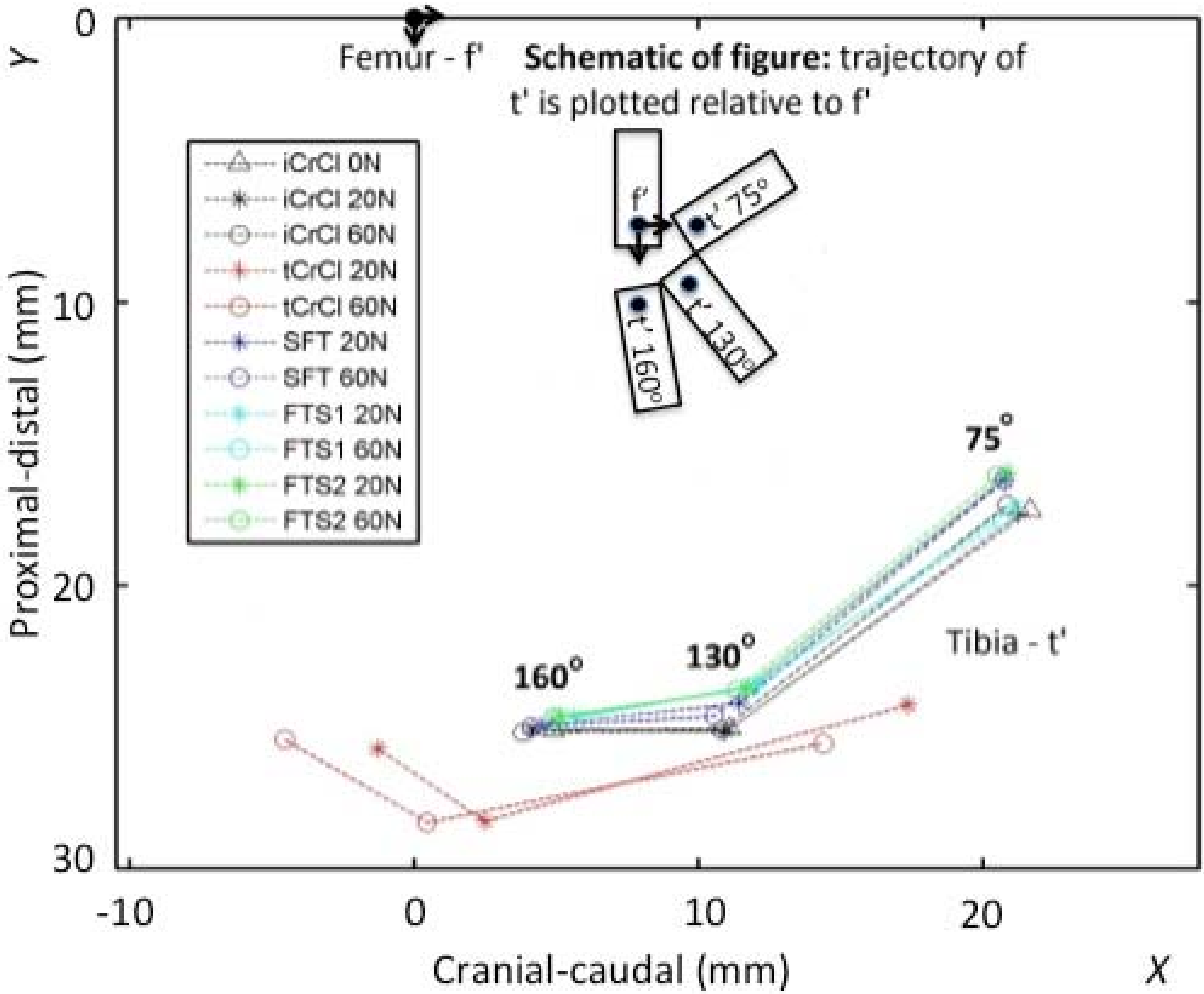
Stifle Joint Angle	iCrCl 0N	iCrCl 20N	iCrCl 60N	tCrCl 20N	tCrCl 60N	SFT 20N	SFT 60N	FT1 20N	FT1 60N	FT2 20N	FT2 60N
75 degrees	0	0.16 (0.11)	0.17 (0.19)	6.9 (1.23)	8.27 (0.55)	-1.0 (0.63)	- 1.0 (0.28)	-0.1 (0.53)	0.45 (0.43)	-1.4 (0.39)	-1.2 (0.15
130 degrees	0	0.15 (0.04)	0.0 (0.27)	3.3 (0.24)	3.3 (0.22)	-0.9 (0.1)	-0.4 (0.13)	-1.5 (0.41)	-1.4 (0.19)	-1.4 (0.26)	-1.3 (0.19
160 degrees	0	-0.05 (0.05)	0.08 (0.1)	0.6 (0.07)	0.3 (0.01)	-0.2 (0.03)	-0.12 (0.01)	-0.5 (0.21)	-0.32 (0.36)	-0.5 (0.27)	-0.44 (0.2

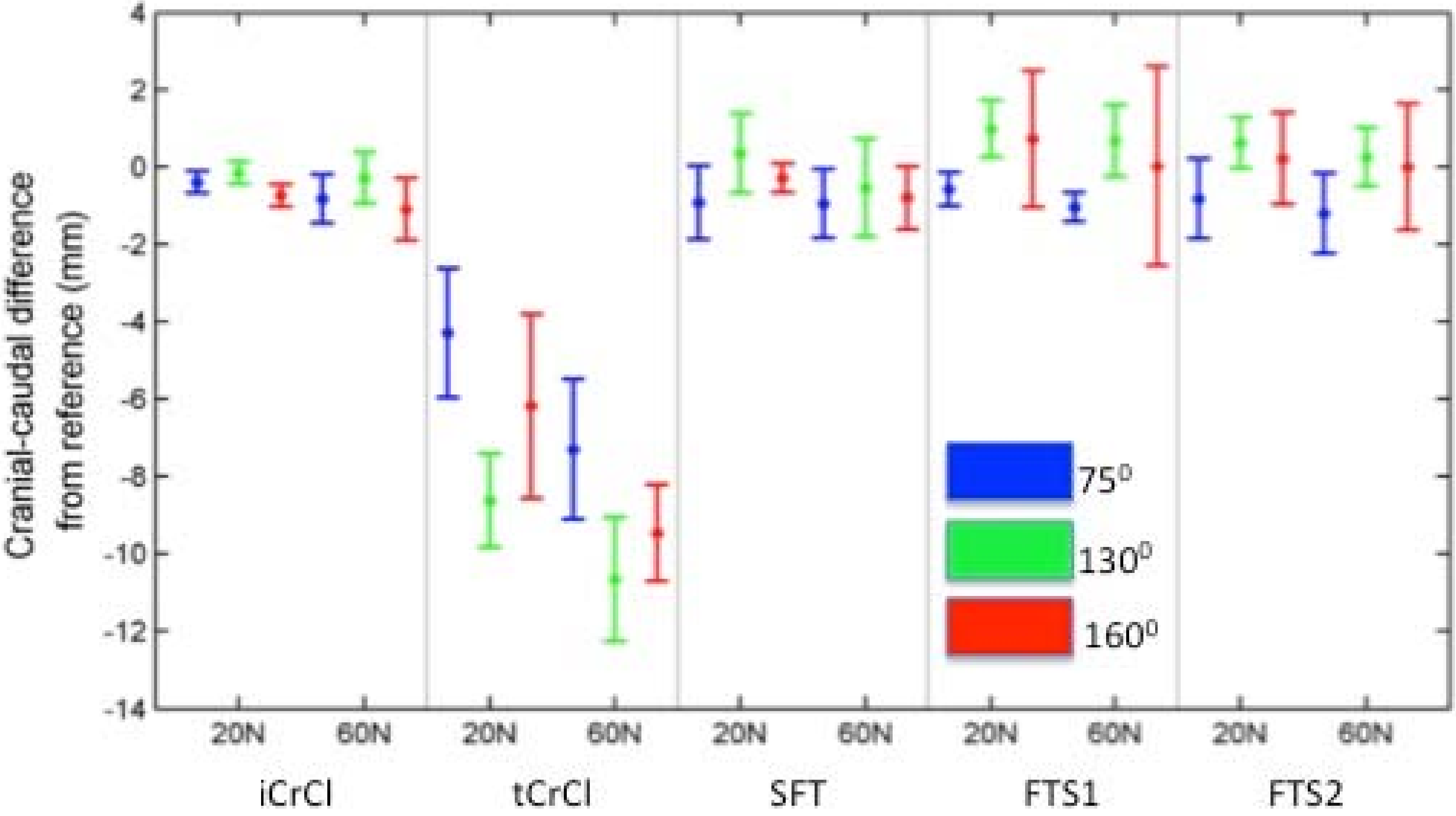
Table 2. Cranio-caudal movement of the tibial t’ point relative to the femoral f’ point. All distances calculated relative to the intact joint at 0 N forces. Means ± SD during loading, stages, and angles tested. (Results expressed in millimeters).

Stifle Joint Angle	iCrCl 0N	iCrCl 20N	iCrCl 60N	tCrCl 20N	tCrCl 60N	SFT 20N	SFT 60N	FT1 20N	FT1 60N	FT2 20N	FT2 60N
75 degrees	0	-0.34 (0.08)	- 0.8 (0.01)	-4.3 (0.98)	-7.23 (0.7)	- 0.03 (0.99)	- 0.03 (1.01)	- 0.5 (0.24)	-1 (0.22)	- 0.8 (0.09)	- 1.14 (0
130 degrees	0	0.2 (0.05)	0.3 (0.35)	-8.7 (0.15)	-10. 6 (0.52)	0.3 (0.24)	-0.6 (0.39)	0.9 (0.04)	0.6 (0.52)	0.6 (0.28)	0.2 (0.0
160 degrees	0	-0.8 (0.01)	-1.1 (0.2)	-6.17 (0.4)	-9.4 (0.69)	-0.3 (0.1)	-0.8 (0.63)	0.7 (1.37)	0.02 (2.16)	- 0.2 (0.97)	0 (1.4)

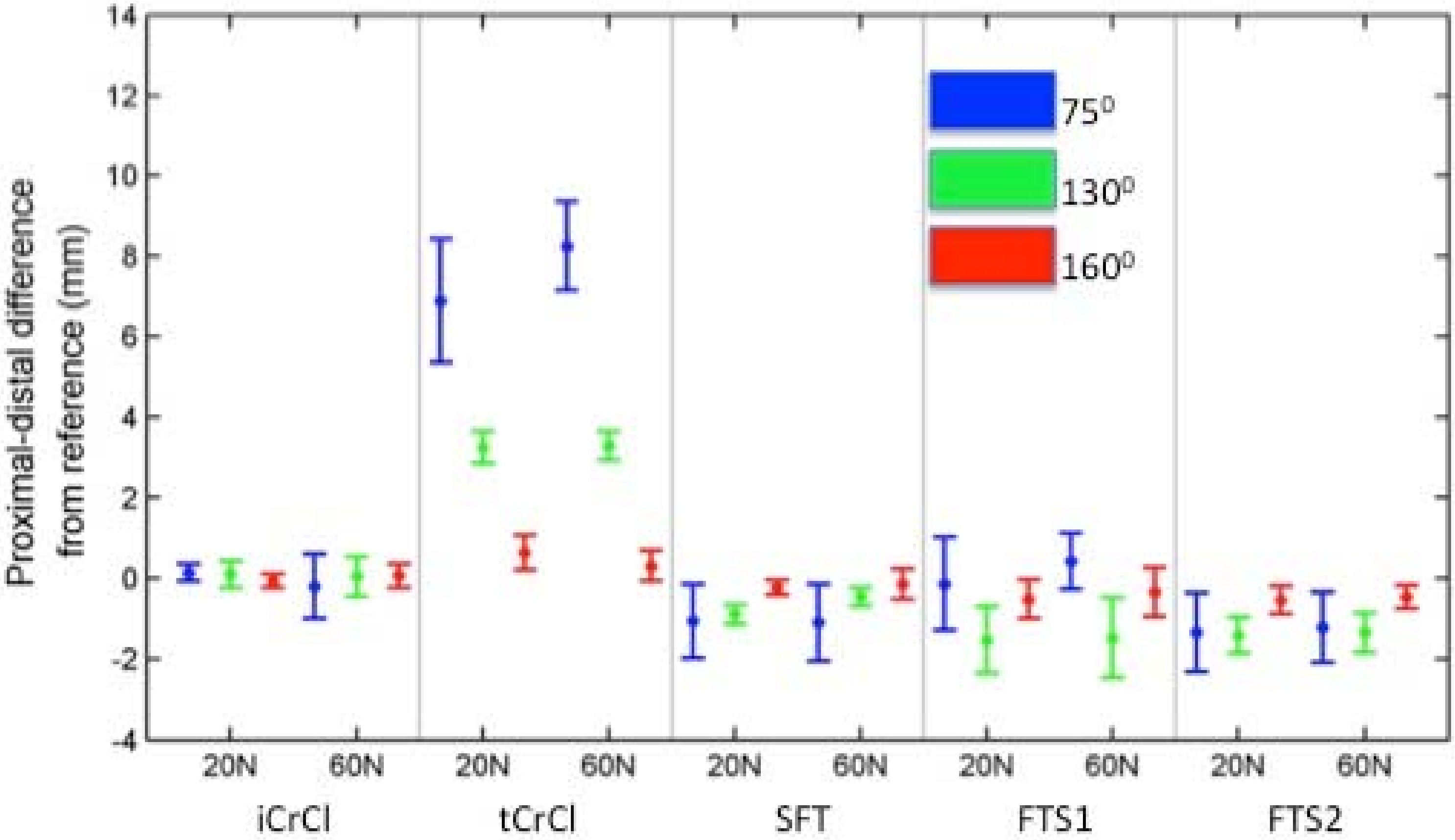

A
F
IM
f'
S1
75 °
t'
T
S2
IM
B
F
IM
f'
S1
130 °
t'
S2
T
IM
C
F
IM
f'
S1
160 °
T
t'
S2
IM


Wooden cube
Stifle
75°
160°
130°
Custom made adapter
Force gauge
Screws

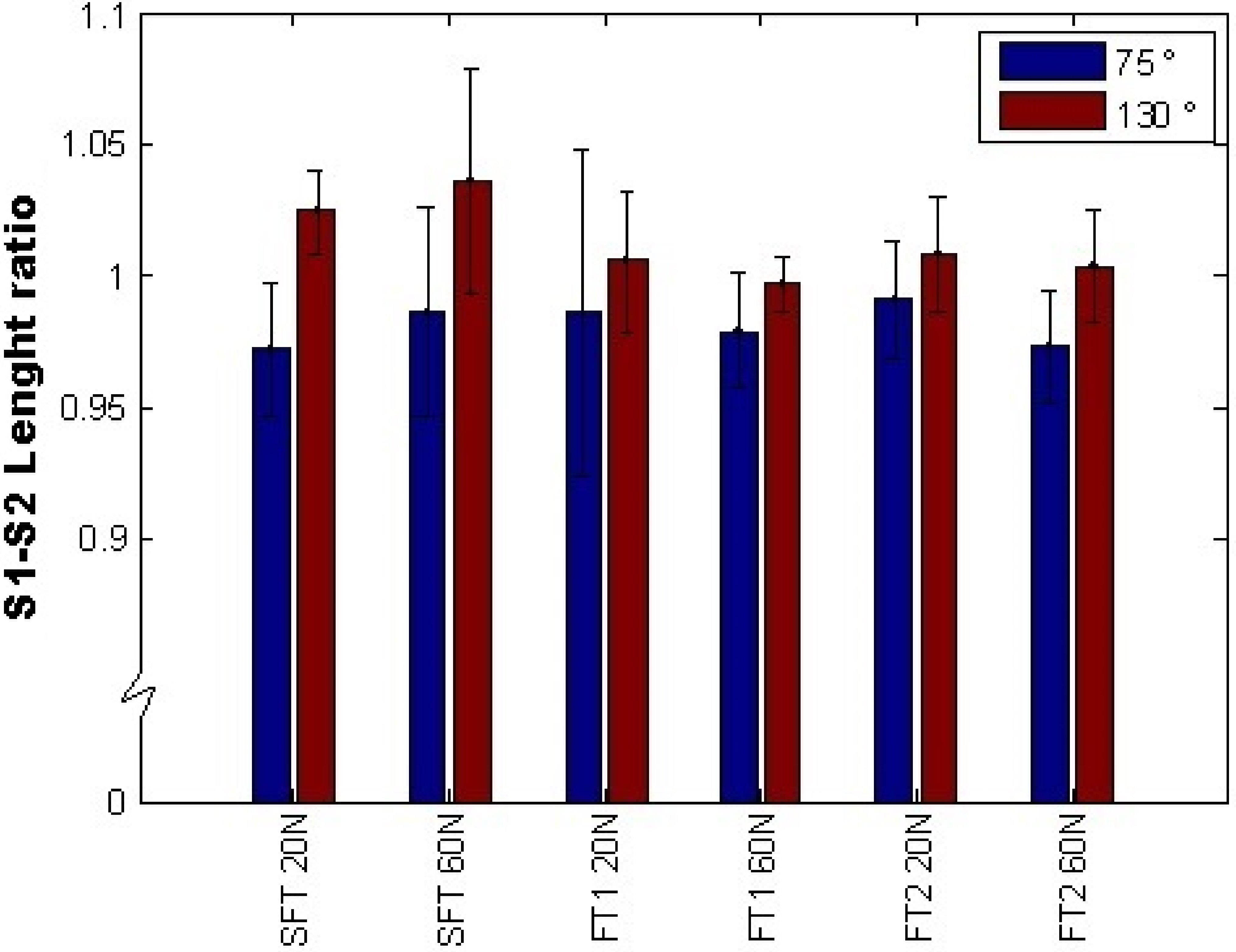

Femur - f'
Schematic of figure: trajectory of t' is plotted relative to f'
f'
t' 75°
t' 130°
t' 160°
iCrCl 0N
iCrCl 20N
iCrCl 60N
tCrCl 20N
tCrCl 60N
SFT 20N
SFT 60N
FTS1 20N
FTS1 60N
FTS2 20N
FTS2 60N
75°
130°
160°
Tibia - t'
Y
Proximal-distal (mm)
0
10
20
30
-10
0
10
20
Cranial-caudal (mm)
X



Cranial-caudal difference from reference (mm)
4
2
0
-2
-4
-6
-8
-10
-12
-14
20N
60N
20N
60N
20N
60N
20N
60N
20N
60N
iCrCl
tCrCl
SFT
FTS1
FTS2
🟦 75⁰
🟩 130⁰
🟥 160⁰



Proximal-distal difference from reference (mm)
14
12
10
8
6
4
2
0
-2
-4
20N
60N
iCrCl
tCrCl
SFT
FTS1
FTS2
75^0
130^0
160^0



1.1
1.05
1
0.95
0.9
0
S1-S2 Lenght ratio
75 °
130 °
SFT 20N
SFT 60N
FT1 20N
FT1 60N
FT2 20N
FT2 60N