THE BIOMECHANICS OF CRUCIATE LIGAMENT REPAIR

A Senior Thesis

By

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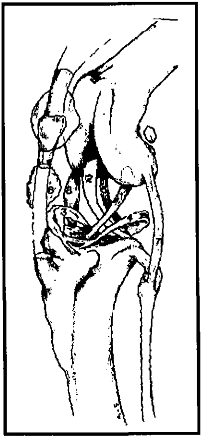
Abstract

Although numerous surgical procedures have been designed to restore stability of a ruptured cranial cruciate ligament (CrCL), the technique for reconstruction has not been perfected. Graft failure due to excessive strain is a serious complication in most reconstructive procedures. This complication has led to the development of an isometric graft placement technique in which the graft is positioned to maintain a constant length and tension throughout passive joint motion. In this project, a two-dimensional model of the stifle joint was created to explore the existence of near-isometric CrCL graft attachment sites. A lateral view of the stifle joint was illustrated using Working Model® 3.0, a two-dimensional simulation software package. The geometry of the articulating surfaces of the femur and tibia, origin and insertion points of relevant ligaments, and attachment points of relevant muscles were determined using radiographs of a canine specimen. The ligaments were represented as linear springs and the muscles were represented as non-rotating forces. Nine simulations were produced, each testing a different combination of CrCL origin and insertion points. It was determined that an attachment site inferior and proximal to the original insertion and origin points, respectively, produced the least strain variation in the ligament.

Introduction

The stifle joint in the canine represents the articulation of the femur and tibia, similar to the human knee. The primary ligamentous support of the stifle joint is provided by the femorotibial ligaments which include the collateral ligaments, the cranial cruciate ligament, and the caudal cruciate ligament (Vasseur, 1993). The anatomy and spatial arrangement of the fémorotibial ligaments are directly related to their function as primary constraints of joint motion (Butler et al., 1983). The cranial cruciate ligament (CrCL) serves to prevent excessive cranial translation of the tibia relative to the femur (Olmstead, 1995). This ligament originates from the inside (medial) surface of the lateral condyle of the femur and extends diagonally across the joint space to insert onto the craniomedial tibial plateau ***(Figure 1)***. The CrCL in canines corresponds to the anterior cruciate ligament (ACL) in humans.



1 cranial cruciate ligament

2 caudal cruciate ligament

3 medial meniscus

4 lateral meniscus

5 meniscofemoral ligament

6 infrapatellar body fat

7 lateral collateral ligament

8 medial collateral ligament

From: Adams, Donald R. (1986). *Canine Anatomy* (pp. 115). Ames: The Iowa State University Press.

Figure 1: Ligaments and Menisci of the Left Stifle

CrCL ligament ruptures are one of the most common injuries in the dog and the major cause of degenerative joint disease in the stifle joint (Vasseur, 1993). The first report of cranial cruciate ligament rupture was in 1926; since then much has been learned about this ligament yet the cause of rupture is often unknown and the optimum mode of treatment remains controversial (Brinker et al., 1990). The CrCL is most commonly injured when the stifle joint is rotated rapidly in 20-50 degrees of flexion, or when the joint is forcefully hyperextend. The ligamentous injury may be a complete rupture with obvious instability or a partial rupture with minor instability. The CrCL injury in canines is analogous to rupture of the ACL in humans and there are close parallels in both the cause and treatment of these ligament ruptures.

The management of CrCL and ACL injury has received considerable attention in both medical and veterinary literature. Numerous surgical procedures have been designed to restore the stability of a ruptured CrCL and to minimize secondary degenerative joint disease. These procedures include the following three techniques: (1) suturing the torn ends, (2) extracapsular reconstruction, and (3) intracapsular reconstruction. Extracapsular reconstruction involves the placement of an autogenous or synthetic material outside the joint capsule to stabilize the joint. Intracapsular reconstruction is performed by passing autogenous tissue, allograft tissue, or artificial materials through the joint to mimic the CrCL (Vasseur, 1996).). This reconstructive procedure allows for placement of a structure similar in anatomical position to the original ligament. None of these procedures have proven clearly superior to any other, although limb function can be improved in most dogs by using one of these techniques (Adams, 1996).

While subjective clinical evaluation does not present a superior surgical solution, biomechanical analysis of the ACL repair in the human has suggested that intracapsular reconstruction supports more normal motion in the knee (Lewis et al., 1990). The primary goal of intracapsular ligament replacement is to stabilize the joint against abnormal craniocaudal (head-to-tail) translation while allowing normal joint motion (Butler et al., 1983).

The technique for intracapsular reconstruction of the CrCL has not been perfected and failure due to excessive strain in the replacement graft is a serious complication. The excessive graft strain variation during joint motion results from an unfavorable position of the graft as it moves through the joint (Butler et al., 1983). To minimize dynamic strain, the graft must remain at a nearly constant length throughout the range of stifle motion.

The complication due to excessive graft strain has led to development of a graft placement technique in which the change in length of the graft is minimized throughout a range of motion (Hefzy, Grood, & Noyes, 1989). Graft position is considered one of the most important surgical variables that determine the long-term success of ACL reconstruction (Tohyama et al., 1996). Minimal elongation of the CrCL is most effectively achieved by isometric placement of the replacement graft. By definition, isometrically positioned grafts maintain a constant length and tension throughout passive knee motion. There is considerable controversy as to the existence and location of the isometric attachment sites, and the majority of human orthopedists believe near-isometry is a more realistic goal (Fleming et al., 1993).

While there is ample literature discussing the isometric and near-isometric graft positions for the ACL in humans, it does not directly correspond to information for CrCL grafts due to species variation in anatomy. There are few published studies that describe the biomechanical properties of CrCL reconstruction in canines (Hulse et al., 1996).

A two-dimensional model of the canine stifle joint was created to study the strain variation on the CrCL throughout the joint range of motion. The specific aims of this project were: to create a realistic two-dimensional model of the canine stifle joint; to explore the existence of isometric and near-isometric origin and insertion points for CrCL graft replacement; and to determine optimal graft attachment sites for intracapsular reconstruction.

Method

A realistic model of the stifle joint was produced using Knowledge Revolution's Working Model® 3.0, a two-dimensional simulation software package. Working Model's simulation engine allows the construction of complex systems and computes their motion under a variety of constraints and forces. In Working Model, a constraint is an entity that applies forces and torques to bodies only at the location of attached endpoints. There are four classes of constraints in Working Model: linear constraints, rotational constraints, joints, and forces. The stifle joint simulation in this project will use linear constraints and forces. In addition to these user-defined constraints, the engine has the capability to simulate world-level interactions such as collisions, gravity, and air-resistance. Physical properties of the simulation such as velocity, acceleration, and energy can be measured using "meters" and "vectors".

In this study, the tibia and the distal end of the femur were modeled as polygonal mass objects. The cranial cruciate ligament, caudal cruciate ligament, lateral collateral ligament, and medial collateral ligament were represented as linear springs with point origins on the femur and point insertions on the tibia. Non-rotating forces applied at a point represented the semitendinosus, semimembranosus, quadriceps, and gastrocnemius muscles. Throughout the joint's range of motion, the femur was held fixed and the motion of the tibia was constrained by the four springs and applied "muscle" forces.

Bone geometries, ligament lengths and points of attachment, and muscle points of attachment were determined from medial-to-lateral radiographs of a cadaver stifle joint specimen at 25°, 55°, 75°, and 120° of flexion. The origin and insertion points of the ligaments and muscle attachment points were marked with knots of metal suture to make them visible on the radiograph. The canine specimen and radiographs were donated by Texas A&M University's Veterinary Department. The specimen was killed for reasons unrelated to this study.

The radiographs of each rotation were traced onto graph paper and oriented in a consistent coordinate system. These graphs were then used to obtain coordinates to enter the stifle joint geometry into Working Model. The femur and its associated ligament origin points had the same coordinates in each graph since this geometry was held fixed. The relative position of the tibia, ligament and muscle insertion points, and reference points varied according to the angle of flexion. The coordinates of the ligament origin and insertion points, the muscle attachment points, and many points along the outline of the femur and tibia were recorded in an Excel spreadsheet.

The initial frame of the simulation illustrated in ***Figure 2*** was drawn by importing the coordinates of the femur and tibia at 25° of flexion into Working Model. These coordinates determined the vertices of two separate polygonal mass objects. Since the femur was anchored to the background, its mass and material related properties were not significant and Working Model default values were used. The properties of the tibia could not be disregarded because the motion of the joint was dependent on the applied forces. The tibia was assigned a mass of 1kg , default values were used for the remaining material properties.

After constructing the femur and tibia they were connected by four linear springs which represented the cruciate and collateral ligaments of the stifle joint. Each spring was placed according to the origin and insertion points obtained at 25° of flexion and assigned a spring constant, *k*, of 152 N/mm. This spring constant was determined by averaging the ligament stiffness results obtained from a recent CrCL study (Vasseur, Griffey, & Massat, 1996). Four dampers were placed in the model (***Figure 3)*** with origin and insertion points identical to the ligament springs. An arbitrary damping coefficient of 5 N•s/mm was chosen which eliminated the oscillatory motion of the springs.

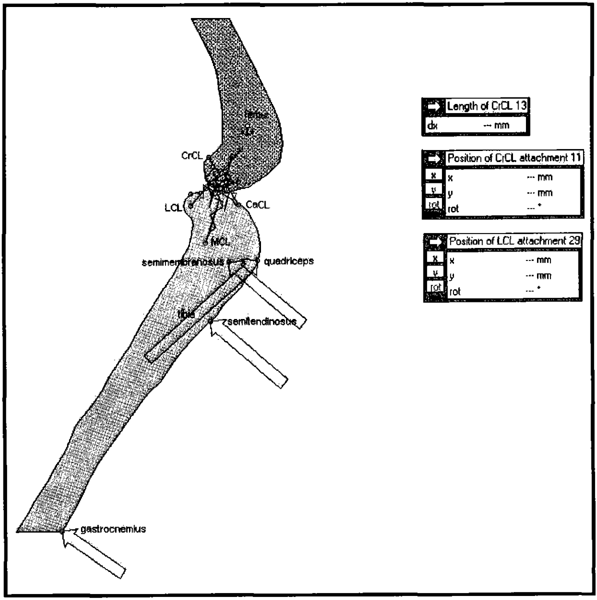


Figure 2: Screen Capture of Initial Simulation Frame

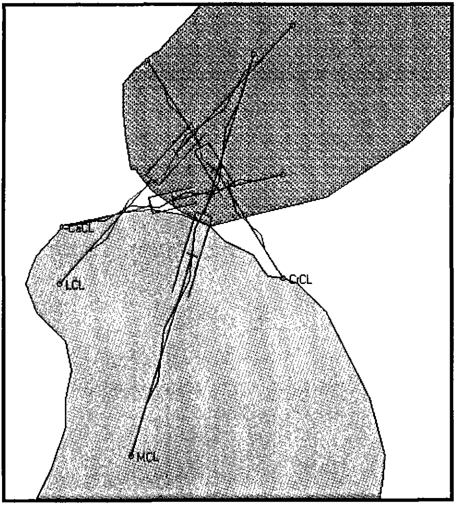


Figure 3: Magnification of Dampers and Ligament Springs

Initially, a track was used to determine the motion of the tibia relative to the femur. Entering the reference points from each of the four radiographs into Working Model created the shape of the track. A smooth curve was formed by using a third order B-spline to interpolate and connect these points. A pin element was connected to the tibia and attached to the track. A force was applied at the pin approximately tangential to the track in order to move the joint through its range of motion. The position of the CrCL and LCL insertion points from this initial simulation were compared to later untracked simulations to determine the accuracy of joint motion.

The purpose of the joint model was to determine the origin and insertion points that resulted in minimal elongation of the CrCL while maintaining normal joint motion. To measure these parameters, a length meter was established to monitor the elongation of the CrCL throughout its range of motion. In addition, two position meters were established to monitor the motion of the tibia. The first meter tracked the position of the CrCL insertion point and the second tracked the position of the LCL insertion point. These meters determined the accuracy of the untracked motion as compared to the tracked motion.

In addition to the geometries of bone and surrounding cartilage, stifle joint motion is influenced by factors such as externally applied loads and active muscle forces (Torzilli, Xianghua, & Warren, 1994). To create a realistic model, muscle insertion points were placed according to the origin points determined by the radiographs at 25° of flexion. The angle of each muscle force was obtained from veterinary literature (Miller, 1964). The magnitude of each muscle force was varied until normal joint motion, as compared to the tracked motion, was reached. The magnitudes of all muscle forces relative to each other are listed in ***Table 1***. The semimembranosus was assigned a magnitude twice that of the gastrocnemius, and the semitendinosus and quadriceps were assigned a magnitude 1.3 times that of the gastrocnemius. Normal joint motion was assumed when the difference between the position of the tracked CrCL and LCL insertion points was less than 2mm from the untracked positions.

Table 1: Muscle Forces

Muscle	Relative Force
Quadriceps	1.3
Semitendinosus	1.3
Semimembranosus	2.0
Gastrocnemius	1.0

Nine simulations were performed, each with a unique pair of CrCL insertion and origin points. The test parameters used can be found in ***Table 2.*** The origin points represent the medial, proximal, and distal points on the crescent shaped band of the original CrCL ligament. The insertion points represent the original insertion point and two points along the medial face of the tibia, inferior and superior to the original point. The experimental origin and insertion points are listed in ***Table*** 3 and ***Table 4***.

Table 2: Test Parameters for the CrCL

Test	Insertion	Origin	Unstretched Length (mm)
1	inferior	medial	21.9
2	inferior	proximal	21.8
3	inferior	distal	22.2
4	superior	medial	23.8
5	superior	proximal	23.8
6	superior	distal	23.9
7	medial	medial	22.7
8	medial	proximal	22.6
9	medial	distal	22.8

Table 3: Experimental CrCL Origin Points and Change from Medial Point

Origin Points (mm)			
	x (mm)	***y (mm)***	***change***
medial	-5.59	14.37	–
proximal	-6.60	13.97	1.09
distal	-4.57	15.49	1.51

Table 4: Experimental CrCL Insertion Points and Change from Medial Point

Insertion Points (mm)			
	x	***y***	***change***
medial	6.35	-4.57	–
proximal	5.08	-4.45	1.28
distal	7.62	-5.08	1.34

Results

Raw data was imported into Microsoft Excel for processing. All test results were compared by converting CrCL lengths to values of percent elongation, a unitless quantity defined as,

$$e = \frac{length - length_{unstretched}}{length_{unstretched}} \bullet 100.$$

Both percent elongation and degree of rotation were plotted against frame number for each of the nine simulations as shown in ***Figures 4-12.*** These plots allow percent elongation to be compared to degree of rotation. This is necessary because frame numbers do not represent the same degree of rotation for each specimen.

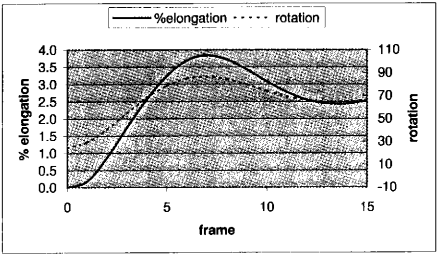


Figure 4: Test 1 (inferior insertion, medial origin)

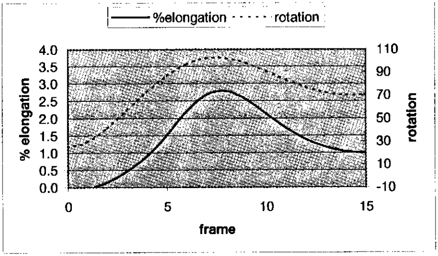


Figure 5: Test 2 (inferior insertion, proximal origin)

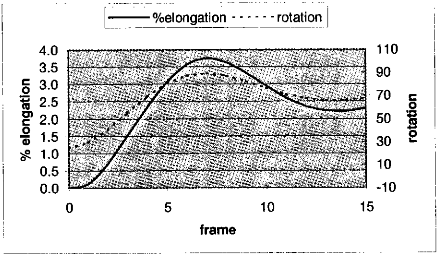


Figure 6: Test 3 (inferior insertion, distal origin)

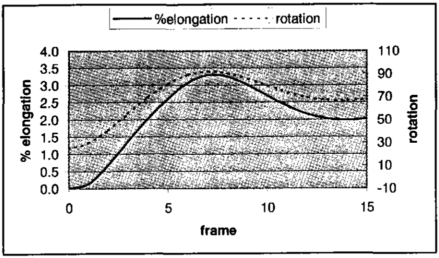


Figure 7: Test 4 (superior insertion, medial origin)

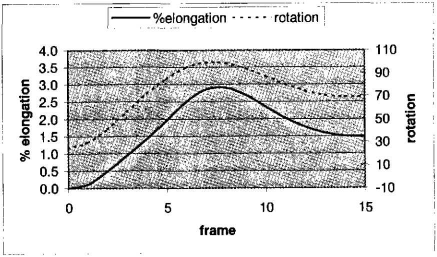


Figure 8: Test 5 (superior insertion, proximal origin)

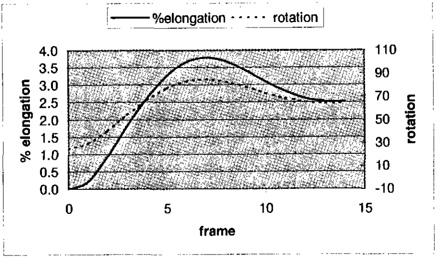


Figure 9: Test 6 (superior insertion, distal origin)

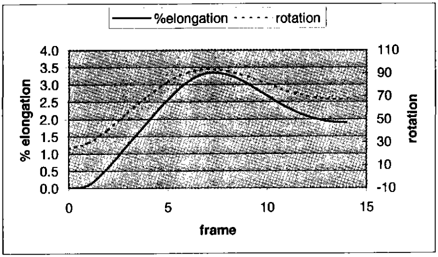


Figure 10: Test 10 (medial insertion, medial origin)

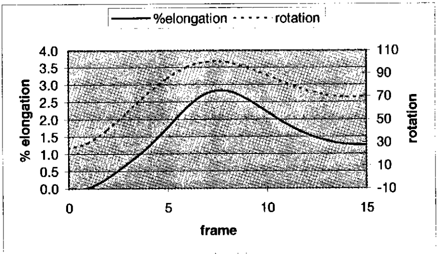


Figure 11: Test 8 (medial insertion, proximal origin)

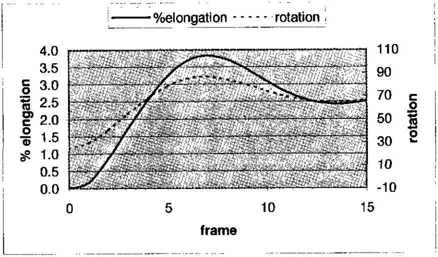


Figure 12: Test 9 (medial insertion, distal origin)

Each simulation followed a different path of motion since the path was dependent on the muscle forces present. As the initial spring length and position changed, so did the motion of the tibia. ***Table 5*** lists the average percent elongation for each of the simulations, and percent elongation at 30°, 40°, and 90° of rotation.

Table 5: Percent Elongation of CrCL Compared to Angle of Rotation

Test	**% elongation**				
	30°	40°	90°	average	Standard deviation
1	0.04	0.44	2.93	1.84	1.05
2	0.00	0.15	2.13	1.49	0.79
3	0.12	0.71	3.75	2.29	1.18
4	0.13	0.68	3.25	2.31	0.69
5	0.11	0.49	2.36	1.93	0.57
6	0.21	0.94	3.78	2.46	1.21
7	0.10	0.60	3.08	2.01	1.09
8	0.02	0.31	2.22	1.67	0.72
9	0.18	0.86	3.83	2.44	1.19

Discussion

By definition, a graft that is placed isometrically will allow constant graft forces while maintaining normal joint motion (Fleming, 1993). The results of these nine simulations show a trend towards isometry when using a graft origin point proximal to the midpoint of the original ligament and a graft insertion point inferior to that of the original ligament.

The range of motion of the joint model was limited by the applied "muscle" forces and ligament springs. The maximum angle of rotation for the model was never greater than

102°, this is less than the angle of flexion of a true stifle joint. Although the degree of flexion was not as large, the path of motion was precise during each test. All simulations remained within 2 mm of the original tracked motion, thereby satisfying the maintenance of normal joint motion.

Since normal ligaments can not support compression, accurate representation of the stifle joint requires maintaining tension of all ligaments through full range of motion (Miller, 1964). The joint model gave positive values of percent elongation for each simulation. This proves that the CrCL ligament remained in tension, satisfying Miller's constraint.

Test two produced an average percent elongation of 1.49 %, a value of 0.18 % smaller than the other origin/insertion combinations. The percent elongation, or strain variation, for every angle of rotation in this test remained smaller than the other tests through the entire range of motion. This can be easily seen when comparing ***Figure 5*** to the other eight plots. The largest strain variation was seen in test six, where the CrCL placement was opposite (distal and superior) from test two. The average percent elongation for test six is 2.46 %.

Studies in human literature indicate that the origin site on the femur has greater influence on replacement graft tension than the insertion site on the tibia (Hefzy,1989). Following the nine initial simulations, a number of tests were run to further examine the isometric trend. Origin and insertion points continuing in a more proximal and inferior direction, respectively, than test two were chosen to determine if the trend towards isometry continued. Strain variation did not continue to decrease as more inferior and proximal points were chosen, although it did remain less than the other eight original tests. These results lead to the conclusion that near-isometric points exist but they are randomized and do not follow a specific path along the femur or tibia.

In future studies additions to this two-dimensional model could more closely predict isometric graft placement technique. Recommendations for future study with this model include the use of non-linear springs and additional anatomical features within the joint such as menisci. This two-dimensional model is limited because the CrCL attachment points are restricted to a single plane. To eliminate this restriction, the possibilities of a three dimensional model should be explored. Suggestions for three-dimensional modeling include the use of a computer aided design (CAD) package such as Pro/Engineer. The challenge of this model would be the computer programming required for processing the simulations. Although this would be an involved project, it would allow for an even closer approximation for true isometric points.

Conclusion

All objectives of this project were satisfied. The two-dimensional model of the canine stifle joint produced in this project provided an accurate representation of the anatomy. Attaching the CrCL replacement graft at a position slightly proximal to the femoral origin point and slightly inferior to the tibial insertion point gives the closest approximation to isometry. Small changes in graft position of 1 – 2 mm have a significant impact on strain variation of the CrCL. This should be taken into account when placing a intracapsular replacement graft in the stifle joint. The results of this study, in agreement with other studies, support the hypothesis that points of near-isometry exist.

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Biography

Kathryn Ferry is from The Woodlands, Texas. Currently, she is a senior Biomedical Engineering student at Texas A&M University. Kathryn is a member of Tau Beta Pi Engineering Honor Society and an associate member of Sigma Xi Scientific Research Society. She plans to complete her Bachelor of Science degree in December of 1998 and attend medical school in the fall of 1999. Her plans after medical school include involvement in a pediatric medical practice and in the research of genetic disorders.